Department of Biomedical Engineering and Physics
University of Vienna

International Society for Artificial Organs

International FES Society (IFESS)

7th VIENNA INTERNATIONAL WORKSHOP ON FUNCTIONAL ELECTRICAL STIMULATION

Basics, Technology, Application

Vienna, September 12 - 15, 2001

Parkhotel Schönbrunn, Vienna

PROCEEDINGS

ISBN 3-900928-05-3

Proceedings

7th Vienna International Workshop on Functional Electrical Stimulation

Vienna, Austria, September 12-15, 2001

Edited by

Winfried Mayr
Manfred Bijak
Christine Jancik

Published by

Department of Biomedical Engineering and Physics
University of Vienna, Vienna Medical School
AKH 4L
Währinger Gürtel 18-20
A-1090 Vienna
Austria

Tel.: +43-1-40400-1984
Fax : +43-1-40400-3988
http://www.bmtp.akh-wien.ac.at

ISBN 3-900928-05-3

TABLE OF CONTENTS

TABLE OF CONTENTS

TABLE OF CONTENTS

TABLE OF CONTENTS

TABLE OF CONTENTS

TABLE OF CONTENTS

TABLE OF CONTENTS

POSTER PRESENTATIONS

Session 1

DENERVATED MUSCLES

FES TRAINING OF DENERVATED MUSCLES IN HUMAN

H. Kern[*], C. Hofer[*], M. Mödlin, C. Forstner, D. Raschka-Högler, W. Mayr[**], H. Stöhr[***]

[*]Ludwig Boltzmann Institute of Electrostimulation and Physical Rehabilitation,
Department of Physical Medicine, Wilhelminenspital Wien, Austria
[**]Department of Biomedical Engineering and Physics, University of Vienna, Austria
[***]Department of Biomedical Research, University of Vienna, Austria

SUMMARY

Prior clinical work showed that electrical stimulation therapy with exponential current is able to slow down atrophy and maintaining the muscle during non permanent flaccid paralysis. But exponential currents are not sufficient for long-term therapy of denervated degenerated muscles (DDM).
We initiated a European research project investigating the rehabilitation strategies in human but also studying the underlying basic scientific knowledge of muscle regeneration from satellite cells or myoblast activity in animal experiments.
The necessary stimulation parameters used in the prior studies that are sufficient for long-term treatment of DDM are forbidden by current EU regulations. In spite of this fact we were able to show that stimulation training of denervated degenerated muscles is possible. At beginning of training only single muscle twitches can be elicited by biphasic pulses with durations of 120 – 150ms. Later with an improved structural and metabolic muscle state the tetanic contraction of the muscle with special stimulation parameters (pulse duration of 30 – 50ms, stimulation frequency of 16 – 25Hz, pulse amplitudes of up to 250mA) can be achieved /14/. Since there are no nerve endings for conduction of stimuli large size anatomically shaped electrodes are used for stimulation. The electrodes made of soft conducting rubber are applied with a wet sponge cloth or gel directly to the skin. This ensures an evenly contraction of the whole muscle.

PRIOR RESEARCH

In the last 15 years we examined the possibility of training denervated human muscles by means of functional electrical stimulation (FES) in complete denervated paraplegics.
In patients with complete conus cauda lesion (Frankel A) was examined if by using appropriate electrical stimulation muscular degeneration could be prevented and and already degenerated muscles could be improved to an extent that tetanic contractions and functional movements could be triggered using FES. These patients started training with electrical stimulation within the first 2 years after the lesion. Tetanic muscle contractions with knee extension moments of about 20Nm could be obtained after one and a half year.
Muscle training was begun in the sitting position with single twitches (biphasic rectangular pulse, average pulse width 120 ms). Later on training with tetanic contraction cycles (on: 2 seconds, off: 2-4 seconds) was performed. Initially without weight until nearly complete knee extension was achieved and later with weight cuffs 0,5 – 1 kg on the ankle (8 – 12 repetitions, 4 - 6 series, once a day).
Improvement in muscle force (+60%) and muscle cross-section area (about 35%) within 1 year of tetanic muscle training was documented by means of torque measurement and computer tomography respectively.
As knee stabilization in the standing position requires relatively low muscle force and

standing up is supported by the upper body, standing training can be started early on. In our patients FES started 1 – 2 years after lesion (complete conus cauda). Standing training could be carried out 9 – 12 months after commencement of FES.

CLINICAL PROBLEMS OF FES OF DDM IN HUMAN

The prior study showed that denervated muscle can be treated with electrical stimulation, but at the same time highlighted problems that remain unsolved:

1. Maximum time span between injury and onset of stimulation training where successful restoration of DDM is still achievable
2. Slow increasing contraction force
3. Daily time expenditure for training is very high
4. High variation in developed contraction force (knee extension torque varies 20-25%) due missing warning signal in case of overstraining the muscle like pain or co-contraction that are reducing the knee extension torque
5. Muscle endurance is very low despite daily training of up to 2 x 30min.

Maximum time span elapsed after denervation

Stimulation therapy for denervated musculature should start within 5 years after denervation. After 5 years degeneration of muscle with connective tissue and fat is progressed so far that restoration of muscle tissue is hardly achievable. But to date 2 patients with a complete conus cauda lesion started stimulation after about 25 years and 1 patient with a plexus lesion started stimulation after 12 years are participating in our program.
The underlying mechanism of regeneration of denervated degenerated muscle and the factors triggering this mechanisms are at present unknown. Especially if there is a spontaneous regeneration of myoblasts (autonomous regeneration) or an electrically induced activity of satellite cells is to be answered still.

Long time span for hyperplasia and hypertrophy

If training with FES is commenced late after injury and the muscle is already degenerated, restoration of muscle takes at least two years. Within this period about 1 – 2 (4) Mio. new muscle fibers have to be built. That equals a synthesis rate of minimal 1000 new fibers per day over a period of 3 years.
This explains the slow increasing muscle force and combined with it the slow functional improvement of the muscle.
Intramuscular structural changes are much bigger than measurable gain of muscle force because of connective tissue and fat as well as broken sarcomeres have to be transformed and rebuilt.

Daily training time expenditure

Due to the fact of missing nerves and thus no distribution of the stimulation pulses each muscle or muscle group respectively has to be trained individually. Treatment of multiple muscles is impossible because of technical limitations regarding the maximum stimulation current. Connective tissue and fat acting as an electrical shunt are reducing the difference of electrical potential that is needed to excite the muscle cell membrane. The decreased membrane resting potential and the reduced metabolism of the muscle fiber seem to be the reason for the long pulse duration and the slow twitch character and fatigue of the muscle.

High variations in developed knee extension torque

In spite of continuous FES training with constant pulse parameters and treatment protocols we observed variations of up to 25% in developed knee extension torque.
Different co-activation of the hamstrings by the electrical field distributed over the whole thigh could be one of the effects causing this variations. Another reason for this effect could be a local overstrain of the muscle that is not prevented because of missing warning signals like pain. An accompanying monitoring of CPK acting as an indicator for muscle overstrain would be desirable in the future. We did not perform in our outpatients this monitoring of CPK because of its short half-life period and the necessary frequent measurements.

Endurance of muscle is very low

In the experimental work already done the concentration of aerobe muscle enzymes recovered up to the normal range. Nevertheless muscle endurance during standing exercises with switching stimulation alternating left and right (simulation of gait) was not very high. With this intermitting stimulation (1s on- 1s off-time) sufficient blood circulation was ensured, but the stimulated musculature fatigued very fast.
This behavior might be caused by the long diffusion distance between the capillaries and the muscle fibers because of the intrafibrillary located mitochondria in electrically trained muscles. Contrary to voluntary trained muscle where the mitochondria are subsarcolemmal located. These observations and the hypothesis regarding the long diffusion distance are to be proved in the proposed European project. Additionally should be examined if the site of mitochondria formation could be influenced by FES in animal experiments.

REFERENCES

/1/ Boonstra A.M., Va Weerden T.W., Eisma W.H., Pahlplatz V.B.M. and Oosterhuis H.J.G.H.: The effect of low-frequency electrical stimulation on denervation atrophy in man. Scand J Rehab Med 19: 127-134, 1987.

/2/ Carraro U, Catani C, Belluco S, Cantini M, Marchioro L: Slow-like electrostimulation switches on slow myosin in denervated fast muscle. Exp Neurol 94: 537-553, 1986.

/3/ Carraro U, Catani C, Saggin L, Zrunek M, Szabolcs M, Gruber H, Streinzer W., Mayr W, Thoma H: Isomyosin changes after functional electrostimulation of denervated sheep muscle. Muscle Nerve 11: 1016-1028, 1988.

/4/ Duchateau J., Hainaut K.: Effects of immobilization on contractile properties, recruitment and firing rates of human motor units. J.Physiol.442: 55-65, 1990.

/5/ Galvani A.: De viribus electricitatis in motu musculari; Commentarius de Bononiensi Scientarium et Artium Instituto adque Academia Commentarii 7: 363, 1791.

/6/ Gordon T., Stein R.B., Martin T.: Physiological and histochemical changes in human muscle produced by electrical stimulation after spinal cord injury, J Neurol Sci 98, Suppl. 141, 1990.

/7/ Greve J.M., Muszkat R., Schmidt B., Chiovatto J., Barros T.E., Batisttella L.R.: Functional electrical stimulation (FES): muscle histochemical analysis. Paraplegia 31: 764-770, 1993.

/8/ Holle J., Frey M., Gruber H., Kern H., Stöhr H., Thoma H.: Functional electrostimulation of paraplegics (experimental investigations and first clinical experience with an implantable stimulation device). J. Orthopedics 7: 1145-1155, 1984.

/9/ Kern H.: Functional Electrical Stimulation in Paraplegic Spastic Patients. Artificial Organs 21(3):195-196, 1997.

/10/ Kern H., Kainz A., Lechner J., Tausch F., Mayr W., Franke H., Schmutterer R., Schwanda G., Stöhr H., Kumpan W., Schurawitzky J., Mostbeck A., Gruber H.: Auswirkung elektrisch induzierter Bewegungstherapie. Z.Phys.Med.Baln.Med.Klim. 15: 317-318, 1986.
/11/ Kern H., Frey M., Holle J., Schwanda G., Stöhr H., Thoma H.: Funktionelle Elektrostimulation querschnittgelähmter Patienten 1 Jahr praktische Anwendung, Erfolge der Patienten und Erfahrungen. Z.f. Orthopädie 1: 123, 1-12, 1985.
/12/ Kern H.: Elektrostimulation im Sport und Rehabilitation, Dissertation, Universität Wien, 1994.
/13/ Kern H.: Funktionelle Elektrostimulation paraplegischer Patienten. Österr. Z. Phys. Med. 5 , Heft 1 Supplementum, 1995.
/14/ Kern H., Hofer C., Strohhofer M., Mayr W., Richter W., Stöhr H.: Standing up with denervated muscles in humans using functional electrical stimulation. Artif Organs., 23(5):447-52, 1999.
/15/ Lake D.A.: Neuromuscular Electrical Stimulation. An Overview and its Application in the Treatment of Sports Injuries. Sports Medicine 13(5): 320-336, 1992.
/16/ Lomo T., Westgaard R.H., Engebretsen L.: Different stimulation patterns affect contractile properties of denervated rat soleus muscles. In: Pette D. (ed) Plasticity of muscle. De Gruyter, Berlin, pp. 297-309, 1980.
/17/ Neumayer C., Happak W., Kern H., Gruber H.: Hypertrophy and Transformation of Muscle Fibers in Paraplegic Patients. Artificial Organs 21(3): 188-190, 1997.
/18/ Nix W.A: The plasticity of motor units in regard to electrically imposed activity patterns- electrical stimulation and its possible clinical application. Fortschr.Neurol.Psychiat. 57: 94-106, 1989.
/19/ Reichmann H., Hoppeler H., Mathieu-Costello L., Von Bergen F., Pette D.: Biochemical and ultrastructural changes of skeletal muscle mitochondria after chronic electrical stimulation in rabbits. Pflügers Arch 404: 1-9, 1985.
/20/ Salmons S., Henriksson J.: The adaptative response of skeletal muscle to increased use. Muscle and Nerve 4: 94-105, 1981.
/21/ Salmons S, Jarvis JC, Mayne CN, Chi MM, Manchester JK, McDougal DB Jr, Lowry OH: Cahnges in ATP, phosphocreatine, and 16 metabolites in muscle stimulated up to 96 hours. Am J Physiol, 271(4 Pt 1):C1167-71, 1996.
/22/ Schubert W.: Funktionelles Training schlaff gelähmter Muskulatur. Biomed. Technik 30, 115, 1985.
/23/ Zrunek M, Bigenzahn W, Mayr W, Unger E, Feldner-Busztin H. A laryngeal pacemaker for inspiration controlled direct electrical stimulation of denervated posterior cricoarytaenoid muscle in sheep. Eur Arch Otorhinolaryngol 1991;248(8):445-8.

AUTHOR'S ADDRESS

Univ. Doz. Dr. Dr.
Helmut Kern
Ludwig Boltzmann Institute of Electrostimulation and Physical Rehabilitation,
Department of Physical Medicine and Rehabilitation,
Wilhelminenspital Wien
Montleartstrasse 37
A-1171 Wien helmut.kern@phys.wil.magwien.gv.at

THE EU-PROJECT RISE: USE OF ELECTRICAL STIMULATION TO RESTORE STANDING IN PARAPLEGICSWITH LONG-TERM DENERVATED DEGENERATED MUSCLES (DDM)

W. Mayr[1], H. Kern[2], C. Hofer[2], J. Jarvis[3], S. Salmons[3]

[1] Department of Biomedical Engineering and Physics, University of Vienna, Austria
[2] L.-Boltzmann-Inst. of Electrical Stimulation and Physical Rehabilitation, Dept. of Physical Medicine, Wilhelminenspital, Vienna, Austria
[3] Department of Human Anatomy and Cell Biology, The University of Liverpool, UK

SUMMARY

The article describes the project RISE that was selected for funding by the European Community FP5 program. The project start is expected for October/November 2001.
A novel clinical rehabilitation method for patients suffering from long-term flaccid paraplegia (denervated degenerated muscles - DDM) with no chance of recovery of the nervous system, will be developed. It will restore their muscle fibres (and mass), muscle function (tetanic contractions, weight bearing) and thus their ability to rise ('standing up') and maintain a standing posture ('standing'). Based on the results of animal experiments on rabbit and pig and initial clinical trials the associated technology will be developed and an application for modification of EU-standards is planned. It will provide European industry with a novel product. The method addresses the needs of about 100 patients per million EU inhabitants.

STATE OF THE ART

Current practice in the clinical treatment of patients suffering from muscle denervation and degeneration is unsatisfactory (Fig. 1). There is no adequate rehabilitation method for patients with long-term denervated degenerated muscles (DDM), caused by peripheral nerve lesions (plexus paresis, conus cauda lesion, poliomyelitis, multiple sclerosis, etc.).

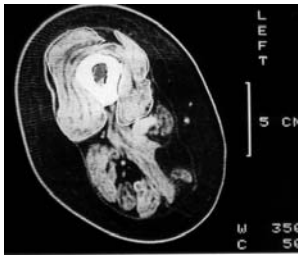
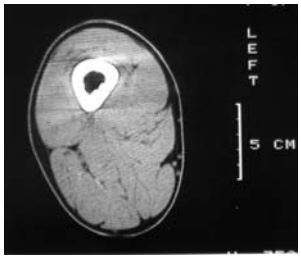


Fig. 1: CT-scans through the mid-quadriceps region of the thigh of two different paraplegic patients, about 4 years post-injury.
Left - of a patient with a conus cauda lesion,
right - of a patient with a spastic lesion



The use of electrical stimulation for denervated muscles is still considered to be controversial by many rehabilitation facilities and medical professionals. This is largely because current teaching and training for therapists is still based on the scientific and technological knowledge of the fifties and sixties.
There are few basic scientific studies relevant to FES treatment of DDM in the literature and the potential for this technique is not generally appreciated within rehabilitation facilities.
The literature contains a limited number of studies of the effects of long-term stimulation on denervated muscles in the rat and rabbit. These have been designed mainly to address basic scientific issues [Lit. 1, Lit. 2]. In virtually every case the muscles were denervated only for a short time before stimulation is commenced, and the results therefore have limited relevance to this human condition, in which long-standing denervation has already resulted in severe atrophy and degeneration of muscle fibres. Other experimental work has been concerned mainly with strategies for reinnervation. A clinical study by Eichhorn et al. has, however, shown that degeneration after denervation could be slowed down, even with less than optimal equipment [Lit. 4].
Although there is a substantial literature concerning the electrical stimulation of paralysed muscle, there are only few published studies that are directly relevant to muscles that are also denervated.

One study showed that FES delivered via implanted electrodes could maintain the action of the denervated sheep cricoarytenoid muscle for up to 18 months [Lit. 3]. Only two functional clinical studies have been published. Valencic et al. have demonstrated correction of dropped foot by FES of the denervated tibialis anterior muscle [Lit. 5]. The other study, conducted by Kern et al., is the basis for the project. Preliminary experiments were conducted on 3 patients with complete injury of conus and cauda equina at spinal level T12 / L1, and motor and sensory loss in both legs. The study showed that the problems of muscle training and restoration of basic lower extremity movements in cases of flaccid paraplegia could be solved in principle, but also highlighted some of the problems that must be overcome before the technique could be used in clinical practice [Lit. 6]. The training regime was based on trial and error. It took more than 2 years to achieve muscle strength and endurance and 2 hours stimulation, delivered twice a day. This burden is unacceptable to most patients.
To create more acceptable protocols, we need to acquire a better understanding of the features of the training regime that induce muscle regeneration and that improve and maintain endurance and force. We also need to establish safe limits for stimulation. These and other open questions are to be addressed systematically in the project.
Technical equipment for FES treatment of DDM is not available commercially, and the current EU standards for medical stimulators are not appropriate for this application. We believe that it is possible to use stimulators safely at higher current densities than those specified within the EU regulations. Safety in use will be a paramount consideration in the development of the new equipment. The work will be used as a basis for revising existing regulatory standards, so that effective stimulators can be made available commercially for clinical use.

OBJECTIVES AND EXPECTED ACHIEVEMENTS

Objective 1: To create a systematic body of basic scientific knowledge about the restorative effects of electrical stimulation on muscles that have undergone degeneration through long-standing denervation.
Objective 2: To use this knowledge base to design an optimised protocol for the therapeutic use of electrical stimulation to treat denervated degenerated muscles (DDM) in patients. The protocol for home use must be effective but must not intrude excessively into the patient's normal daily activities.
Objective 3: To assemble a scientific basis for revising existing EU Regulations governing the use of electrical stimulation, which currently exclude the possibility of therapeutic use in patients with DDM.
Objective 4: To design, construct, and bring to the point of commercial adoption, equipment that will enable patients to use the new therapeutic stimulation protocol safely and effectively in their own homes.
Objective 5: To design, construct, and assess the value of new diagnostic and measurement equipment (dynamometry, surface accelerometry, recording of M-wave) for monitoring the progress of the therapy during the periodic assessment of patients in the clinic.
Expected achievements:
To illustrate the way in which this study may be expected to lead to patient benefits it will be convenient to examine a typical case history.
A patient suffered a conus cauda lesion with peripheral denervation in the gluteal region and lower limbs. After the accident he was operated for stabilization of the spine at T11 – L 1. After one week he was transferred to the rehabilitation centre for spinal cord injury. Following wound healing he was instructed in bladder and bowel management, given training in the use of a wheel chair and in prevention of pressure sores. After 4-6 months he was released from hospital to his home and family. There was little adaptation of his home and no professional re-integration. He was left to his own resources to manage daily living. Information was available to him about training for the muscles of his upper body, but not for his denervated gluteus and limb muscles.
After 2 months he developed a pressure sore and had to be undergo plastic surgery with transfer of a musculocutaneous flap and resection of the ischial tuberosity. Post-operatively he was required to lay on his stomach for 4 – 6 months.
He then heard about electrical stimulation and began to seek information at a specialized clinic. He was anxious to prevent a recurrence of pressure sores and to acquire a more acceptable cosmetic appearance

of his legs during swimming exercise. He was accepted into the stimulation program.
Now, 3 years later, he stimulates his muscles for about 30 min a day, stands up 20 times a day by means of electrical stimulation, and is satisfied with the normal appearance of his leg muscles. He feels fit, does not get breathless when propelling his wheelchair uphill, and although he fell from the wheelchair during a transfer he did not break any bones. His skin is in good condition and he is not troubled by pressure sores during his daily and professional activities.

PROJECT WORKPLAN

The objectives will be achieved through an experimental programme conducted in rabbits, pig and man. The rabbit is the animal of choice for establishing most quickly and economically the conditions for safe and effective stimulation, protocols for training and maintenance, and the most appropriate outcome measures. The results will be used to refine protocols for testing in the pig, an animal whose musculature is physically more similar to that of man. Equipment, such as the stimulator devices and electrodes, will be evaluated and modified in the course of the animal experiments. The equipment and protocols will be transferred to the clinic for trials to be conducted in patients. Any problems that arise at this stage will be addressed through further animal experiments.
The Project is organised as 5 interdisciplinary workpackages which will interact closely. Each workpackage has its own director. Within a given workpackage specialists are responsible for important modules such as developing special technical equipment, performing an experiment, or conducting laboratory investigations. The Gantt chart (Fig. 2) shows the time plan of the whole project, divided into workpackages, and the approximate timing within the workpackages.

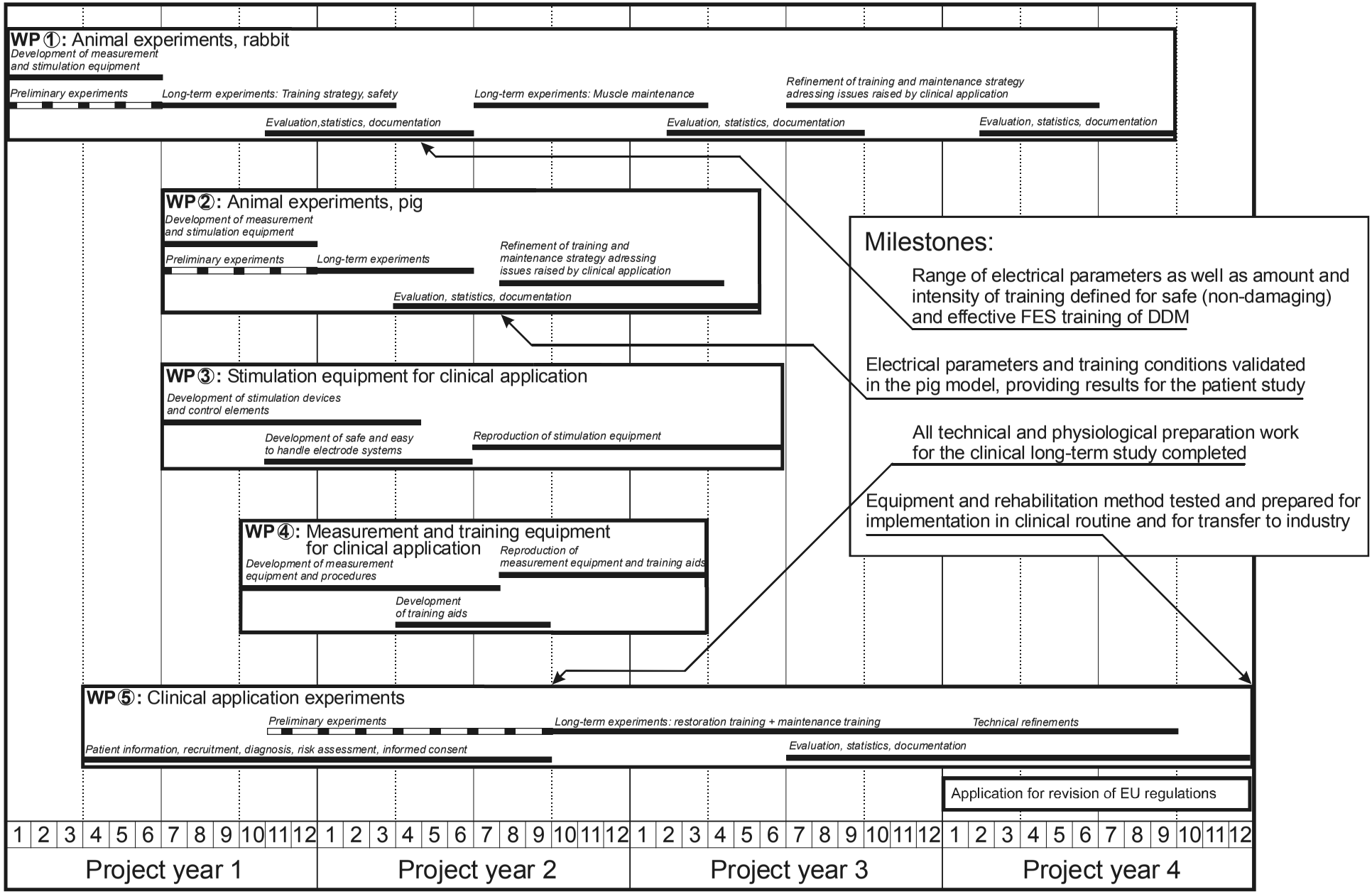
REFERENCES

1. Gundersen, K. Determination of muscle contractile properties: the importance of the nerve. Acta Physiol Scand 162: 333-341, 1998.
2. Mokrusch, T., A. Engelhardt, K. Eichhorn, G. Prischenk, H. Prischenk, G. Sack, and B. Neundorfer. Effects of long-impulse electrical stimulation on atrophy and fibre type composition of chronically denervated fast rabbit muscle. J Neurol 237: 29-34, 1990.
3. U.Carraro, C. Catani, L. Saggin, M. Zrunek, M. Scabolcs, H. Gruber, W. Streinzer, W. Mayr, H. Thoma: Isomyosin changes after functional electrostimulation of denervated sheep muscle. Muscle Nerve 11: 1016-1028, 1988.
4. Eichhorn K., Schubert W., David E.: Maintenance, training and functional use of denervated muscles. J Biomed Eng 6: 205-211, 1984.
5. Valencic V., Vodovnik L., Stefancic M., Jelnikar T.: Improved motor response due to chronic electrical stimulation of denervated tibialis anterior muscle in humans. Muscle Nerve 9: 612-617, 1986.
6. H. Kern, C. Hofer, M. Strohhofer, W. Mayr, W. Richter and H. Stöhr: Standing up with denervated muscles in humans using functional electrical stimulation. Artif Organs 23(5):447-452, 1999.

AUTHOR´S ADDRESS

Dr. Winfried Mayr
Department of Biomedical Engineering and Physics
AKH, Ebene 4/L, Waehringer Guertel 18-20
A-1090 Vienna, Austria
E-mail: w.mayr@bmtp.akh.wien.ac.at

Fig. 2: Projekt Workplan



OUR EXPERIENCE WITH SLOW PULSE STIMULATION IN PATIENTS WITH POST-TRAUMATIC NEUROPATHIES

Terzis, J. K. and Liberson, W. T. [deceased]

Eastern Virginia Medical School

Objective: To share our experience with slow purse stimulation of denervated skeletal muscle in patients that underwent microsurgical repair following devastating lesions of the brachial plexus, in obstetrical paralysis and facial paralysis.

Methodology: A "slow pulse stimulator" was developed by one of the authors [W. L. Liberson] that permits a rational technique for treating denervated muscle. First, slow pulses are calibrated as to their duration so that the latter may be chosen for each individual patient according to his needs. The pulses are delivered automatically at a rate that may be harmless. We are now satisfied with the fact that stimuli succeeding each other at a rate of one during each 10 seconds do not harm the skin if applied for a period of 20 to 30 seconds. The stimulator has a timer limiting sessions to 20 minutes.
In areas where denervated fibers may be mixed with normal muscle fibers, to avoid stimulation of the normal muscles, the use of progressive onset of the stimulating pulses has been successfully employed. A time of onset on the order of 100 or 200 msec indeed suppresses the contraction of the innervated muscles. We limit the total time of stimulation to 5 hours for adults and 3 hours for children. The treatment sessions are 20 minutes each, with an interval of 1 hour in between. The patient is warned never to restimulate an area that remains red following the previous session of stimulation.
We have used this type of "slow pulse stimulation" for the past 15 years in all our patients that have undergone microsurgical reconstruction of the injured brachial plexus, in cases of obstetrical brachial plexus palsy and in selected cases of facial paralysis.

Results: Outcomes of microreconstruction all combined with this form of "slow pulse" electrotherapy will be presented and benefits outlined.

Conclusion: Our technique allows simultaneous stimulation of practically all upper extremity muscles by placing one electrode on the shoulder and the other on the palmar aspect of the fingers.
Although the wisdom of stimulation of denervated muscles has often been challenged, we believe strongly that this does not apply to human denervated muscle. Until new therapeutic procedures are developed to compensate for the loss of the nutrient axon, "slow pulse" electrotherapy is an effective alternative for the treatment of denervated muscle.

Dr. Julia K. Terzis, MRC
Eastern Virginia Medical School
700 Olney Road, Room 2055
Norfolk, VA 23501 USA

FUNCTION OF SKELETAL MUSCLE TISSUE FORMED AFTER MYOBLAST TRANSPLANTATION INTO IRRADIATED MOUSE MUSCLES

A. Wernig, M. Zweyer, A. Irintchev

Department of Physiology, University of Bonn

Pretreatment of host muscles with ionising radiation enhances the formation of donor-derived tissue after myoblast transplantation in the mouse but there is little evidence for improvement of muscle function. To investigate this, we implanted myoblasts from an expanded, male-donor derived, primary culture (i28) into X-ray irradiated (16 Gy) or irradiated and damaged soleus muscles of female syngenic mice (Balb/c). Three to 6 months later the isometric contractile properties of transplanted and control muscles were studied *in vitro*, and the Y chromosome positive progeny of the implanted cells was visualised on muscle cross-sections.

Irradiated and vehicle-injected muscles had significantly smaller weights than untreated solei and produced less twitch and tetanic tension (all about 18%). Such deficits were not found in irradiated solei implanted with 10^6 myoblasts. Increase of muscle mass and strength was due to the integration of donor-derived cells. No deficits in nerve-evoked tension were found.

Repeated freezing/thawing *in situ* of irradiated muscles led to formation of soleus remnants devoid of or containing only small amounts of contractile tissue (1-50 muscle fibres). Myoblasts (10^6) implanted into such destructed muscles generated numerous muscle fibres (1200-5000 per muscle). Upon direct electrical stimulation these fibres produced considerable twitch (53% of normal) and tetanic tensions (35%). The newly-formed muscles were, however, insufficiently innervated presumably due to radiation mediated arrest of Schwann cells. Separate studies on nerve regeneration following x-irradiation and nerve crush or botulinum application will be demonstrated. In any case, even after complete muscle destruction the disorganised suspension of donor cells will produce new organised contractile tissue thus replacing the host muscle. Limiting factor is the reduced capacity of the nerve to regenerate after x-irradiation.

Apart from these results, the role of the satellite cell in muslce fiber repair as well as a possible source for pluripotent stem cells will be discussed. The deviating behaviour of human myogenic cells in their limited proliferative capacity and telomere lenght related senescence will be mentioned.

Department of Physiology, University of Bonn, Wilhelmstraße 31, D-53111 Bonn, Germany

Session 2

PARAPLEGIA, UPPER MOTOR NEURON LESION

A TASK ORIENTED APPROACH BY MEANS OF A TRANSCUTANEOUS NEUROPROSTHESIS: A 7 YEAR CLINICAL EXPERIENCE FOR EXERCISE, STANDING AND WALKING ON SCI SUBJECTS

H.A. Cerrel-Bazo[1]., A. Rizzetto[1]., E. Corà[1]., R. Bogoni[1]., A. Bolner[1]., S.W.Brown[2]

Centro di Neuroriabilitazione e Ricerca di Villa Margherita, Arcugnano, Vicenza, Italy[1]
Metropolitan State Hospital, Norwalk California, USA[2]

Aims of the Study: Train spinal cord individuals to perform a task by means of FES and analyse metabolic and physiological responses, energy efficiency and performance of SCI standing and walking by means of a transcutaneous FES system (TFES).

Methods: 60 motor complete SCI are able to stand and walk following a computer-controlled FES exercise program (CCE) and a gait training program (GTP) by TFES. Under standard conditions the above mention variables were recorded while standing (Tilt-Table and by TFES) for 30 min and walking (on a treadmill) for 6 min (at 5,10,15,18,20 m/min speed) and for 45 minutes (at self preferred maximum speed [SPMS]). All data were entered into a Statview SE+ for computer. ANOVA was used to assess energy cost between groups (the TFES group was compare to a control group [able body and ARGO SCI users]).

Results: CCE-GTP and ability to stand-walk independently with TFES average 5-6 months. Base on standing and walking endurance TFES walkers were classified: Good walkers: Speed (m/min) 14, walking distance (m) 580 m, walking time (min) 42, standing time (min) 30. Moderate walkers: Speed (m/min) 10, walking distance (m) 160 m, walking time (min) 12.5, standing time (min) 23. Poor walkers: Speed (m/min) 8, walking distance (m) 42 m, walking time (min) 3, standing time (min) 3. The TFES group showed higher statiscally significance ($p<.001$) values for most variables analysed when compare to the control group. Oxygen cost increase one time fold from sitting to TFES-standing and 3 to 4 times fold when TFES-walking. At SPMS 45 min walking post-hoc analysis showed a higher metabolic efficiency for the able body group compare to the TFES and ARGO group. During standing-walking metabolic and physiological responses were significantly higher for the TFES group. Statiscally significant differences were reported for lactic acid, O2-Pulse, ventilation and catecholamines, higher for the TFES group.The TFES mean total sweat rate was higher and statistical significant compare to the able body group.

Discussions: Well selected and trained SCI can perform a task oriented approach by means of TFES. Functional achievment was not the goal. Treadmill-Walking at a costant speed is metabolic demanding. Good TFES walkers adapt their cardiovascular and polmonary responses to the physical effort. Good standing-walking performances appear to be correlated to a higher metabolic efficiency, daily changes of the SCI system need to be study to better address the problem of efficient continuative performance. TFES standing and walking is a good active physical therapy on well selected SCI subjects.

MUSCLE ACTIVITY DURING NORMAL WALKING AND ITS RELEVANCE FOR THE FUNCTIONAL ELECTRICAL STIMULATION APPLICATIONS

Sabine Mangold*, Thierry Keller**, and Milos R. Popovic***

* Automatic Control Lab., Swiss Federal Institute of Technology - ETH Zürich, Switzerland
** ParaCare, University Hospital Balgrist, Switzerland
*** Institute of Biomaterials and Biomedical Engineering, University of Toronto, Canada

SUMMARY

Exact knowledge of the temporal pattern of muscle activity during walking is important for a proper sequencing in Functional Electrical Stimulation (FES) supported locomotion. The objective of this study was to give an overview of the EMG activity of functional muscle groups that control movements of the hip, knee, ankle, and subtalar / transverse tarsal joints during normal walking. In literature about 25 important muscles were identified as the muscles that actuate these joints during walking on a plain surface. From the mean EMG profiles of these muscles three ranges of muscle activity 25%-50%, 50%-75%, and 75%-100% of the maximum EMG activity during the stride were extracted. Agonistic muscles were identified and analyzed with respect to their function and estimated activity level during the single gait phase cycle. Seven characteristic muscles for the articulation of the lower limb joints were selected for this presentation. We believe that this information is essential in identifying which muscles and muscle groups have to be stimulated, to which extent, and in which sequence to facilitate dynamic walking.

STATE OF THE ART

Currently existing FES systems that support or facilitate walking generate gait patterns that look like sets of static leg positions with abrupt and jerky transitions between these static positions. Since such "static" walking patterns are slow and physically demanding it is essential that new generations of FES systems for walking facilitate gait cycles which resemble as much as possible the able body subject's dynamics during walking. Muscular activity during locomotion was already quantified by EMG profiles of normal subjects /1,2,4-6/. On-off patterns, i.e. the temporal distribution of each muscle's activity (on) and inactivity (off) during the gait cycle, were derived from these EMG profiles /1,2/ and used for FES-supported walking /1/. These on-off patterns make no difference between higher and lower activity levels during the gait cycle, although it might be an important factor for the creation of stimulation patterns.
The objective of this study was to deliver more sophisticated stimulation patterns serving as a model for FES-supported walking.

MATERIAL AND METHODS

EMG profiles from two standard publications on gait pattern /2,4/ that presented data of 25 important muscles for locomotion were analyzed and compared. Normalized average EMG profiles of normal subjects during free level walking were used. In /4/ the normalization was carried out by adopting the mean value of each subject's EMG over the stride period as the reference value (100%). In /2/ EMG profiles were normalized by an additional manual muscle test (MMT) that was used to measure the maximum voluntary EMG activity as the reference value (100%).
The different activity ranges were extracted in the following way: For the EMG profiles of /4/ it was chosen to cut off a basic activity of 0%-25% of maximum activity (Fig. 1). For EMG activity below 25% the activity could not be clearly attributed to the measured muscle, since the measurements were

performed with surface electrodes that might also record activities of neighboring muscles. Low (25% - 50%), middle (50% - 75%), and high (75% - 100%) activity ranges normalized to the maximum EMG activity during the stride were defined as illustrated in Fig. 1. In /2/, where wire electrodes were used, the basic activity was already cut off so that the profiles were subdivided into three activity ranges (low, middle, high) comparable to those of /4/.
Furthermore, the maximum activity during the gait cycle [% MMT] was singled out from the average EMG profiles from /2/. These values indicate the percentage of the maximally voluntary activity that a muscle needs for level walking.

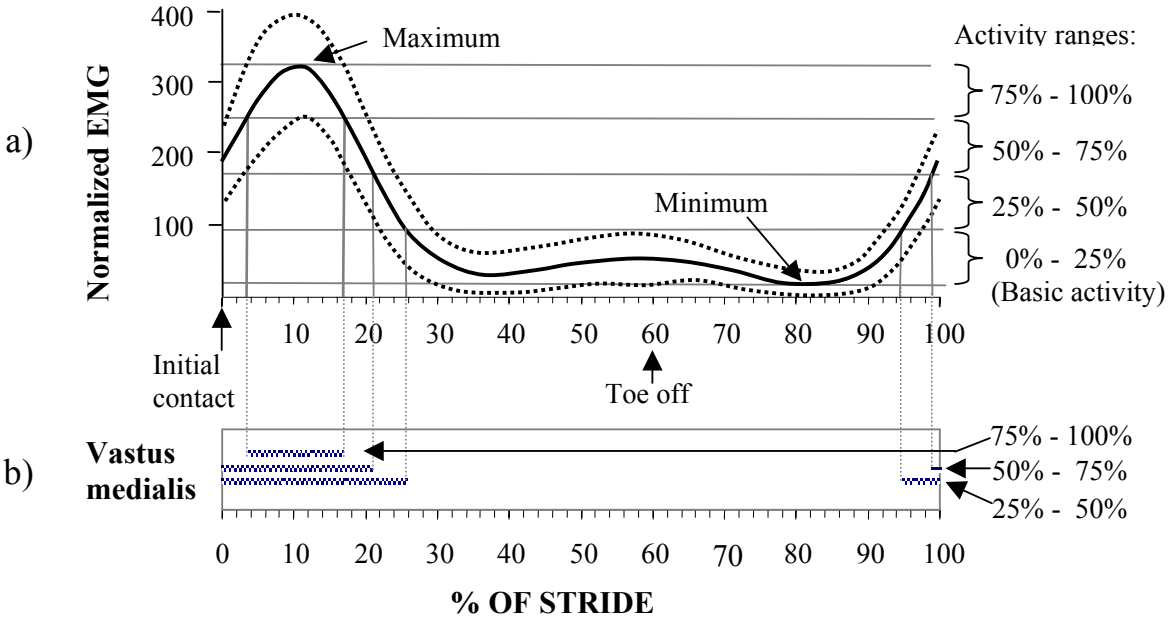


Fig. 1: The method that was used to extract different activity ranges from average EMG profiles in /4/. M. vastus medialis is shown as an example.
a) Average EMG profile (modified from /4/) and the definition of the different activity ranges. — Average EMG profile; ····· standard deviation.
b) Extracted graduated EMG activity.

RESULTS

In Fig. 2 the activity ranges of the muscles 1. for foot eversion and ankle dorsiflexion (extensor digitorum longus); 2. for foot inversion and ankle dorsiflexion (tibialis anterior); 3. for ankle plantar flexion (soleus); 4. for knee flexion (biceps femoris short head); 5. for knee extension (vastus medialis); 6. for hip flexion (adductor longus); and 7. for hip extension (gluteus maximus) are presented. The number of subjects (N) varied for the different muscles. In /4/ the median was N=17 (range: 12-26). In /2/ the median was N=25 (range: 8–51).
The results of both authors are not completely congruent. Three major differences could be observed: 1. In some cases EMG activity was measured in only one of the two references as for example the first 20% of the gait cycle in the adductor longus. 2. EMG activity occurred sometimes in significantly different activity ranges, e.g. the tibialis anterior shows during the swing phase in /4/ a low activity, but in /2/ a middle to high activity. 3. A temporal shift between the data of /2/ and /4/ was observed: For most of the muscles the onset as well as the cessation of the activity was measured earlier in /2/ compared to /4/.

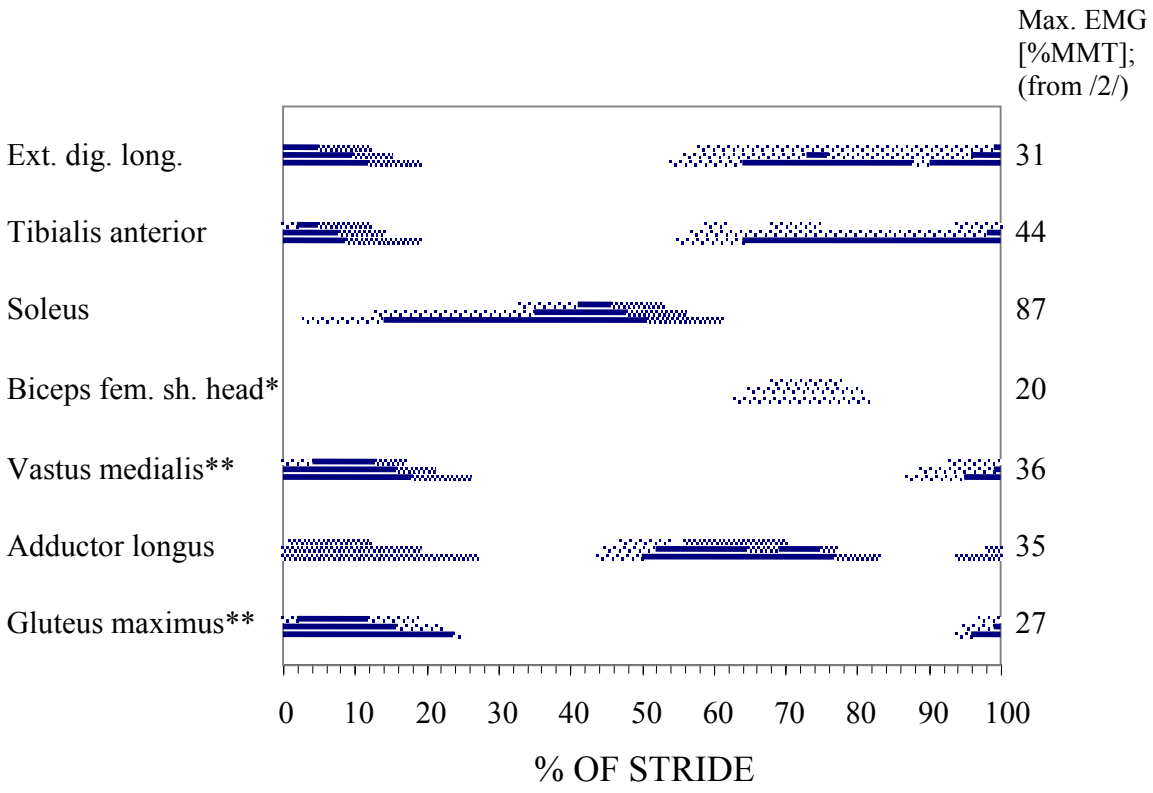


Fig. 2: Activity ranges that were obtained using the method explained in Fig. 1.
EMG Activity reported in /4/
EMG Activity reported in /2/
Overlapping EMG activity reported both in /2/ and /4/
* *This muscle part (i.e. short head of the biceps femoris) was only reported in /2/*
** *The upper and lower part of the gluteus maximus were examined separately in /2/. For presentation the profiles were taken together to be comparable with the data of /4/.*

DISCUSSION

The presented study compared EMG profiles obtained from standardized data published in /2/ and /4/. Three different ranges of EMG activity were extracted with the aim to retrieve stimulation patterns that can be used in neuroprosthesis for walking.

When applying the presented patterns for FES supported walking, two main sources of uncertainty have to be considered: I. Delays due to EMG signal processing used in /2/ and /4/ and muscle excitation-contraction delays; and II. Muscular-skeleton system redundancies.

I. It was observed that the EMG data published in reference /4/ were delayed compared to reference /2/. Different signal processing methods might be one of the main reasons for this shift: In /4/ a temporal delay between real muscle activity and its corresponding EMG profile was estimated to be about 90 ms (8% of the gait cycle) /5/. Thus, in Fig. 2 the data of /4/ (〰) should be shifted about 8% to the left. Furthermore, because of the excitation-contraction delays of the muscle of about 20-50 ms /3/ the stimulation profiles have to start 2-4% earlier in the gait cycle than the presented EMG profiles of both references.
II. Redundancies in mono- and biarticular lower limb muscles facilitate different combinations of muscle activity patterns that lead to similar lower limb kinematics. This might be one of the reasons for the incongruity between the data of /2/ and /4/. Specifically, the activity of proximal muscles during locomotion has a higher intersubject variability than of distal muscles /5/ and the activity of biarticular muscles have a higher intersubject variability than of monoarticular muscles /5,6/.

Thus, the presented data in Fig. 2 have to be considered as a scheme that can be used as a starting point for the synthesis of stimulation patterns for FES supported walking. We suggest to apply the proposed

and time shifted activity patterns in a first step and, if necessary, to modify the timing and amplitudes individually to the subject's anatomical and muscular conditions.
Using time-corrected EMG patterns with different activity ranges has the potential to make FES supported walking smoother and to facilitate a more dynamic locomotion. Furthermore, it accelerates the creation of an appropriate stimulation patterns for the subjects.
Our next step will be the implementation of such delay compensated muscle activity profiles in our neuroprosthesis for walking with the aim to generate smoother and more dynamic walking patterns.

REFERENCES

/1/ Kobetic R., Synthesis of paraplegic gait with multichannel functional neuromuscular stimulation, IEEE Transactions on Rehabilitation Engineering 2(2), 1994, 66-79
/2/ Perry J., Normal Gait, in: Perry J. (ed.), Gait Analysis. Normal and Pathological Function, SLACK incorporated, Thorofare NJ, 1992, 49-167
/3/ Popović D.B., Sinkjær T. (eds.), Control of Movement for the Physically Disabled. Springer Verlag London Berlin Heidelberg, 2000, p. 329
/4/ Winter D., Electromyography in Human Gait, in: Winter D. (ed.), The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological. University of Waterloo Press, 1991, 53-73
/5/ Winter D.A., Yack H.J., EMG profiles during normal human walking: stride-to-stride and inter-subject variability, Electroencephalogr Clin Neurophysiol 67(5), 1987, 402-11
/6/ Wootten M.E., Kadaba M.P., Cochran G.V.B., Dynamic electromyography. II. Normal patterns during gait, J Orthop Res 8(2), 1990, 259-65

ACKNOWLEDGEMENTS

This project was supported by a grant from the Swiss National Science Foundation, Project-No. 5002-057811

AUTHOR'S ADDRESS

Dr. sc. nat. Sabine Mangold
ParaCare, University Hospital Balgrist
Balgriststrasse 340, CH-8008 Switzerland
e-mail: smangold@balgrist.unizh.ch
home page: http://www.aut.ee.ethz.ch/~fes/

FUNCTIONAL ELECTRICAL STIMULATOR "COMPEX MOTION"

T. Keller*, M. R. Popovic**, I. P.I. Pappas**, and P.-Y. Müller***

*ParaCare - Institute for Rehabilitation and Research, University Hospital Balgrist Zurich, Switzerland
**Automatic Control Laboratory, Swiss Federal Institute of Technology Zurich, Switzerland
***Compex SA, Ecublens, Switzerland

SUMMARY

Research groups in the field of Functional Electrical Stimulation (FES) are often confronted with the fact that existing and commercially available FES stimulators do not provide sufficient flexibility and cannot be used to perform different FES tasks. The lack of flexibility of the commercial systems until now forced various FES research teams to develop their own stimulators.
This paper presents a newly developed firmware and programming software for the commercial Compex 2 stimulator that enhances the stimulator's versatility and capabilities from a medical and therapeutic device to a neuroprosthesis and research tool. With the new firmware and a new graphical user interface software the stimulator can now be used to develop various custom-made neuroprostheses, neurological assessment devices, muscle exercise systems, and experimental setups for physiological studies. The new stimulator, called "Compex Motion", can be programmed to generate any arbitrary stimulation sequence that can be controlled or regulated by various external sensors, sensory systems or laboratory equipment. By interconnecting two or more Compex Motion stimulators the number of stimulation channels can be increased to multiples of four channels (8,12,16,20,…). The stimulation sequences and the control strategies are stored on exchangeable credit-card sized memory chip-cards read by the stimulator. The chip-cards are programmed using a standard PC. The function of the stimulator is changed instantaneously by simply inserting a different chip-card. The stimulator has four biphasic current regulated stimulation channels and two general purpose analog input channels that can be configured to measure the output voltage of a variety of sensor systems like goniometers, inclinometers, gyroscopes, or EMG sensors. For real-time EMG control of the stimulation patterns an EMG processing algorithm with software stimulation artifact blanking was implemented. The Compex Motion stimulator is manufactured by the Swiss company Compex SA and is currently undergoing clinical trials.

STATE OF THE ART

Several portable microprocessor or microcontroller FES stimulators for transcutaneous stimulation have been developed to improve upper and lower limb functions in spinal cord injured (SCI) and stroke subjects /1-4/. Most of these systems were built for one specific application and did not have an open architecture. In general the setup options were limited, device dependent and the control options were fixed. The preprogrammed stimulation patterns were stored internally. A fixed set of sensors combined with a control algorithm triggered the preprogrammed stimulation sequences. Some systems allowed changes of the stimulation intensity either during the initialization phase or during stimulation on-line. In some cases, a separate PC software allowed one to download new settings for trigger levels and stimulation sequences.
In this article we are describing a new generation of transcutaneous electric stimulators called "Compex Motion". This stimulator represents further evolution and expansion of the already existing ETHZ-ParaCare FES system /5,6/. The new portable stimulator "Compex Motion" exceeds the limitation of other systems by providing to the user a high flexibility in the programming of the stimulation sequences, the control schemes, and the choice of the man-machine interfaces.

MATERIAL AND METHODS

Hardware

"Compex Motion" is a microcontroller based electric stimulator with four stimulation channels (see Table 1) used for transcutaneous electrical stimulation of selected muscles or muscle groups. The stimulator can deliver current regulated stimulation pulses of maximal 100 mA with a rise time of 3 µs. It has two input channels A and B, and a special purpose port C. A and B can be configured either as analog or digital input channels with a voltage range of 0-5 V. The special purpose port C is used to interconnect two or more stimulators, to serially communicate with a PC, or to trigger the stimulator using a push button. By interconnecting two or more stimulators the number of stimulation channels can be increased from four to multiples of four channels (8,12,16,…). In such an arrangement, one of the stimulators operates as a *master stimulator* and all other stimulators operate as *slave stimulators*. The master stimulator ensures that all interconnected stimulators operate synchronously.

Figure 1: *Compex Motion electric stimulator: 1) stimulator; 2) keypad with 9 push buttons; 3) three memory chip-cards; 4) two EMG sensors; and 5) two stimulation electrodes*

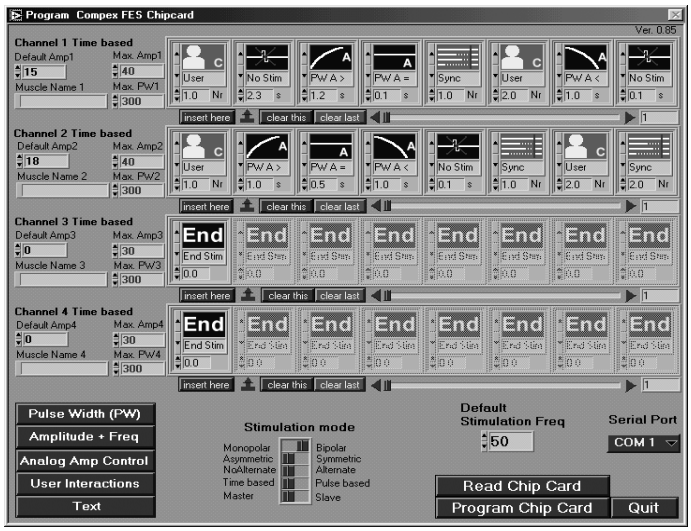


Figure 2: *Main window of the Compex Motion GUI software. It shows four horizontal timelines associated with each stimulation channel (center and right), pulse amplitude and pulse width safety limits (left), pulse type settings (center bottom), memory chip-card functions (right bottom), and setup functions (left bottom).*

The Compex Motion stimulator has a rechargeable NiMH battery that provides eight hours of continuous stimulation and is recharged in less than two hours. The stimulator also has a dot matrix LED display that provides a visual interface between the user and the stimulator. The user can interact with the stimulator via nine push buttons on the stimulator (see Figure 1) or via any other user interfaces that are connected to port A, B and C. Currently, EMG sensors and a push button can be purchased with the stimulator. These interfaces can be used to control the stimulation sequences and to regulate the stimulation intensities. Additional stimulator features, accessories, and hardware data are provided in Table 1.

Software

The Compex Motion stimulator is programmed with a Graphical User Interface (GUI) software that is installed on a PC (see Figure 2). The GUI software used a “drag-and-drop” technique to program the stimulation sequences. This is done by sequentially placing icons called *primitives* on a timeline that describes the chronological sequence of the tasks that will be carried out by a stimulation channel. There are four such time lines, one for each stimulation channel. A total of 56 primitives are available in order to take the full advantage of the full flexibility of the system (see Table 2). There are two classes of primitives: *global* and *local*. Global primitives represent tasks that affect all active stimulation channels while local primitives affect only the specified channel. The drag-and-drop technique make it easy to compose rapidly precisely timed stimulation sequences including customized pulse width ramps, loops, branches, pauses, user interaction rules, and text display.

The primitives used for user interactions define how a subject must interact with the stimulator and can be customized to individual needs. For example the user can initiate or terminate a stimulation sequence via a predetermined analog or digital sensor signal curve profile detected at the input port A or B. Two different sensor signal curve profiles can be used to select between two different stimulation sequences. Sensors such as EMG sensors, force sensitive resistors, gyroscopes, foot switches, and push buttons have already been successfully applied with the user interaction primitives.
Continuous regulation of the stimulation intensity can be achieved in real-time using an analog input signal, i.e. the pulse amplitude depends on the voltage level of the input signal. This dependence can be arbitrarily defined by a lookup table that can be imported as an ASCII file and can be edited both graphically and numerically. Each stimulation channel has its own lookup table. Thus far sensors such as EMG sensors, sliding resistors, and potentiometers have already been successfully used with this control strategy.

RESULTS

Table 1 shows the specification of the Compex Motion hardware. Table 2 lists the available primitive types. Using the drag-and-drop technique these primitives are stacked by the programmer in any desired chronological order in a timeline, which form the stimulation sequences.

Feature	**Characteristics**		
4 stimulation channels	Current regulated		
	Pulse amplitude	*Range:* 0-100 mA	*resolution:* 1 mA (8 Bit)
	Pulse width	*Range:* 0-16 ms	*resolution:* 500 ηs (14 Bit)
	Stimulation frequency	*Range:* 1-100 Hz	*resolution:* 1 Hz (8 Bit)
2 digital input ch. (A & B)		*Range:* 0-5 V TTL	
2 analog input ch. (A & B)	*Max sampl. freq.:* 8 kHz	*Range:* 0-5 V	*resolution:* 20 mV (8 Bit)
1 special purpose port (C)	Push button, serial port communication, and stimulator interconnection		
Working regimes	Master/slave		
Stimulation pulses	Monophasic/biphasic; monopolar/bipolar; and alternating/non-alternating		
Microcontroller	Motorola HC11		
dot matrix LED display	*No. pixels:* 165 x 64		*dimensions:* 72 mm x 30 mm
chip-card	*Can store up to 255 primitives per channel and all relevant stimulation parameters*		
NiMH battery	Rechargeable, 8 h of continuous stimulation		
Stimulator dimensions	148 mm x 80 mm x 30 mm & 420 g		
Accessories	AC/DC adapter, push button, 4 cables, self-adhesive electrodes, EMG sensor, ...		

Table 1: *Compex Motion data sheet specifications*

Pulse Width Primitives:		
Icon	**Name**	**Description**
	constant pulse width	Generates a pulse with a constant width (4 different values are available per channel)
	pulse width ramp-up	Profile for changing the pulse width (2 different profiles are available per channel; profiles are described with 16 values)
	pulse width ramp-down	Profile for changing the pulse width (2 different profiles are available per channel; profiles are described with 16 values)
	no stim	Pulse width equal to 0
	delay	Keeps the actual pulse width at the previous level for the given time interval
Pulse Amplitude Primitives:		
Icon	**Name**	**Description**
	change amplitude	Changes the amplitude from previous to new value in a specified time period (change is linear)
Pulse Frequency Primitives:		
Icon	**Name**	**Description**
	change frequency	Changes stimulation frequency (4 different values are available and they apply to all stimulation channels)

Primitive Sequence Control:		
Icon	**Name**	**Description**
	jump back	Program jumps back *n* times in the sequence to the marker primitive, where *n=1-255,* or *n* is infinite *(n=0).*
	synchronize	Synchronizes otherwise independent stimulation sequences in all 4 stimulation channels
Human Interaction Primitives:		
Icon	**Name**	**Description**
	user interaction	This primitive waits for a specific user action to trigger a stimulation sequence. One can use an arbitrary triggering profile and a sensor.
	user branch	Two trigger criteria set with the *user interaction* primitive are used to generate branching. If criterion 1 is fulfilled the program proceeds with the next primitive in the line. If criterion 2 is fulfilled the program jumps to a predefined marker and proceeds with the next primitive after the marker.
	user interrupt	One trigger criterion set in *user interaction* primitive is used to generate interrupt. If this criterion is fulfilled between markers ON and OFF the program jumps to a predefined marker and proceeds with the next primitive after the marker.

General Primitives:		
Icon	**Name**	**Description**
End	end	Terminates stimulation in the specified channel – time line
	turn off	Turns off the stimulator
Text A	display text	Displays two text lines with 8 characters in each text line
A	generate sound	Generates a melody (2 different short melodies are available)
Special Primitives:		
Icon	**Name**	**Description**
Freq	random frequency	Activates the stochastic variation of the frequency. The frequency varies randomly about the nominal value (± 0-100%), following a uniform probability distribution function.
	random pulse width	Activates the stochastic variation of the pulse width in the specified channel(s). The pulse width varies randomly about the nominal value, within a specified range (± 0-100%), following a uniform probability distribution function
Amp	random amplitude	Activates the stochastic variation of the pulse amplitude in the specified channel(s). The actual amplitude varies randomly about the nominal value, within a specified range (± 0-100%), following a uniform probability distribution function

Table 2: *List of GUI primitives*

DISCUSSION

The new portable and programmable electrical stimulator "Compex Motion" that can be used for a wide range of transcutaneous Functional Electrical Stimulation (FES) applications has been presented. The stimulator can be used to develop various custom-made neuroprostheses, neurological assessment devices, muscle exercise systems, and experimental setups for physiological studies. The Compex Motion stimulator can be programmed to generate any arbitrary stimulation sequence, which can be controlled or regulated using any external sensor, sensory system or laboratory equipment.
Currently, a number of patients are using the Compex Motion stimulator as a neuroprosthesis for grasping or walking at our facilities. Besides walking and grasping the system was used to treat shoulder subluxation and to strengthen muscles in some patients. Multi-center trials are expected to start by the end of this year.

REFERENCES

/1/ A. Prochazka, M. Gauthier, M. Wieler, and Z. Kanwell, “The bionic glove: an electrical stimulator garment that provides controlled grasp and hand opening in quadriplegia,” *Arch Phys Med Rehabil*, vol. 78, pp. 1-7, 1997.

/2/ M. J. Ijezerman, S. T.S., F. A. C. G. in 't Groen, M. A. P. Klatte, G. J. Snoeck, J. H. C. Vorsteveld, R. H. Nathan, and H. J. Hermens, “The NESS Handmaster orthosis: restoration of hand function in C5 and stroke patients by means of electrical stimulation,” *J. of Rehabilitation Sciences*, vol. 9, pp. 86 - 89, 1996.

/3/ W. Mayr, M. Bijak, W. Girsch, C. Hofer, H. Lanmuller, D. Rafolt, M. Rakos, S. Sauermann, C. Schmutterer, G. Schnetz, E. Unger, and G. Freilinger, “MYOSTIM-FES to prevent muscle atrophy in microgravity and bed rest: preliminary report,” *Artif Organs*, vol. 23, pp. 428-31., 1999.

/4/ P. N. Taylor, J. H. Burridge, A. L. Dunkerley, D. E. Wood, J. A. Norton, C. Singleton, and I. D. Swain, “Clinical use of the Odstock dropped foot stimulator: its effect on the speed and effort of walking,” *Arch Phys Med Rehabil*, vol. 80, pp. 1577-83., 1999.

/5/ T. Keller, A. Curt, M. R. Popovic, V. Dietz, and A. Signer, “Grasping in high lesioned tetraplegic subjects using the emg controlled neuroprosthesis,” *The J. of NeuroRehabilitation*, vol.10, pp.251-255, 1998.

/6/ M.R. Popovic, T. Keller, I.P.I. Pappas, V. Dietz, and M. Morari, “Surface-stimulation technology for grasping and walking neuroprostheses”, *IEEE Eng. in Medicine and Biology*, January/February 2001.

ACKNOWLEDGEMENTS

This project was supported by a grant from the Federal Commission for Technology and Innovation, Switzerland - Project No. 4891.1

AUTHOR’S ADDRESS

Dipl. Ing. Thierry Keller
ParaCare, University Hospital Balgrist — e-mail: kellert@balgrist.unizh.ch
Forchstrasse 340, CH-8008 Zurich, Switzerland — home page: http://www.aut.ee.ethz.ch/~fes/

THE VIENNA FES SYSTEM FOR RESTORATION OF WALKING FUNCTIONS IN SPASTIC PARAPLEGIA

M. Bijak W. Mayr, M. Rakos**, C. Hofer*, H. Lanmüller, D. Rafolt, M. Reichel, S. Sauermann, C. Schmutterer, E. Unger, M. Russold, H. Kern*

Department of Biomedical Engineering and Physics, University of Vienna, Austria
*Department of Physical Medicine and Rehabilitation, Wilhelminenspital, Vienna, Austria
**Otto Bock, Austria

SUMMARY

An eight-channel stimulation system, currently intended for stimulation of lower extremities was developed and is introduced. Major development goals were easy handling, modularity to make the system easy adaptable for other FES applications and a wide stimulation parameter range for application specific parameter optimisation.
For paraplegic stepping the system worn by the patient consists of two four channel stimulation modules, a central unit holding the battery and circuitry for power management and communication control, a wireless remote control unit and a palmtop computer as main control and input device.
A software package for MS-Windows supports the design and optimisation of stimulation sequences in the rehabilitation centre.
First tests with patients used to FES showed smoother movements during stepping and acceptable good handling. In combination with the PC software the required stimulation sequences could be created in a very short time.

STATE OF THE ART

Stimulation of leg muscles in spastic paraplegia to restore standing up from the wheelchair, stepping and sitting down into the wheelchair is a well-known FES application /1/. Muscle activation with implanted stimulators via nerve and motor point attached electrodes /2,3,4/ or with external stimulators via surface electrodes /5/ are state of the art.
Surface stimulation needs careful system setup before usage, has a limit in the amount of useful channels and a worse muscle selectivity in comparison to implanted systems but does not require any surgical intervention.

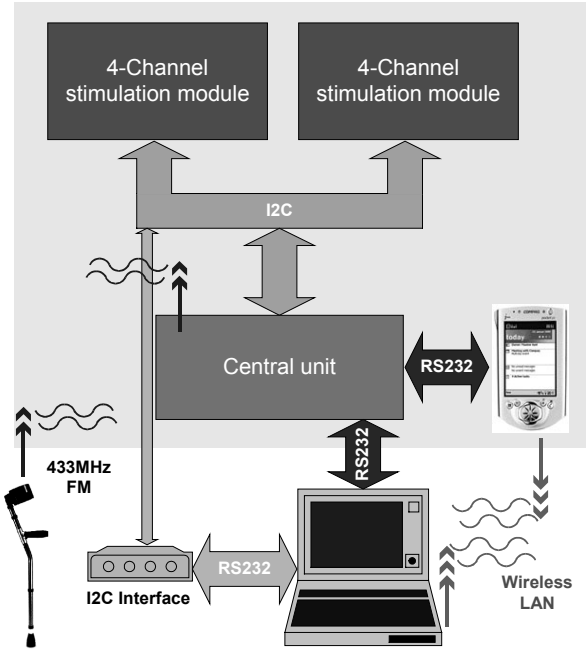


Fig. 1: Block diagram of the eight channel stimulation system

MATERIAL AND METHODS

The introduced stimulation system (Fig. 1) consists of two four-channel stimulators (one for each leg), a control device (Central unit) and a Windows CE based palmtop computer. An additional battery powered Command Unit can be mounted on a crutch or a walker for manual wireless stimulator control by pressing push buttons.
A special software package, installed on a

standard PC or Laptop computer supports stimulation pattern design and interactive testing.

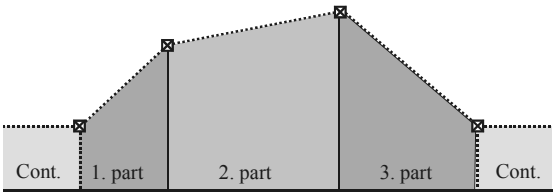


Fig. 2: Envelop of a stimulation burst

Stimulation module:

The stimulation module houses four independent stimulation stages (channels). Each channel is controlled by its own microprocessor (PIC 16F876, Microchip, Chandler, AZ, USA) and can deliver any stimulation sequence that can be defined by moving the corner points of the burst envelope shown in Fig. 2. Stimulation frequency and pulse duration of the positive and negative part of the biphasic rectangular impulses can be set individually for each greyed region.

The micro controller drives the impulse generating output stage. For safety reasons and to keep the electrodes potential free an output transformer is used.

Electrode impedance measurement and m-wave acquisition are implemented even though the m-wave is currently not evaluated.

All channels are linked together and are controlled via Inter-Integrated Circuit bus (I2C bus).

The chosen microprocessor is flashable and allows in circuit firmware update.

Central unit:

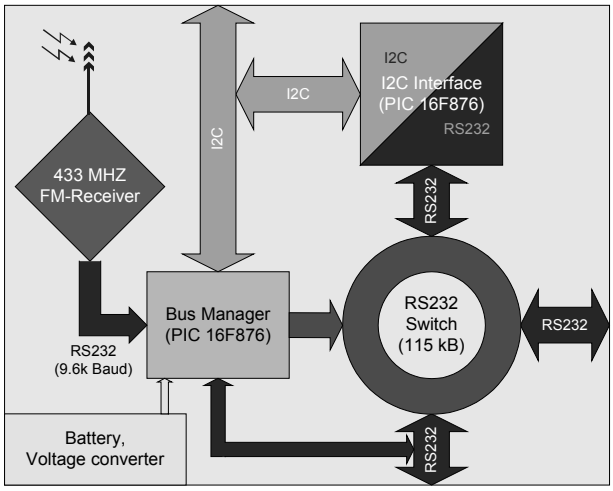


Fig. 3: Block diagram of the Central Unit

The central unit holds the batteries, power supply and battery charging circuitry, bus management circuitry, the RS232 / I2C translation unit and the 433 MHz receiver (RX2-433, Radiometrix Ltd, Hertfordshire, England) for the Command Unit.

Two RS232 ports are available to connect a Palmtop Computer and a PC. During the initialisation phase the Bus Manger can be programmed to connect the PC to the RS232 / I2C Interface or palmtop computer to the RS232 / I2C Interface giving them access to the stimulation devices or to connect PC and palmtop for direct data transfer via RS232.

Beside this the Bus Manager is also responsible for the decoding of the control signals sent by the command unit (standard RS232 protocol, 9600 baud). The state of the Command Unit can be polled either from the Palmtop or the PC. Further more the Bus Manger can be configured (via I2C) to trigger autonomously the stimulation devices on Command Units request. Therefore multimaster management had to be implemented in both, I2C interface and Bus Manger.

Command Unit

The Command Unit is based on a commercially available 433MHz FM transmitter (TX2-433, Radiometrix Ltd, Hertfordshire, England) and a microprocessor (16F876) with integrated A/D converter. The state of up to four pushbuttons and the value of up to 4 analog signals are continuously scanned, encoded and sent. To get a data transfer rate of 9600 Baud a coding strategy is necessary that keeps the digital signal DC free. This is simply done by consecutive sending one byte and the inverted byte, a technique also useable for data validation.

The described data transmission procedure results in a sample rate of the pushbuttons of 250/s, in 250/s for one analog channel and 62.5/s each for four analog channels.

After an idle time of 10 minutes the command unit enters sleep mode to reduce power consumption. Wakeup is performed when any button is pressed.
The radio link can be bypassed by a cable connection in the case when more Command Units are operating in a close distance and interfere with each other.

Palmtop Computer

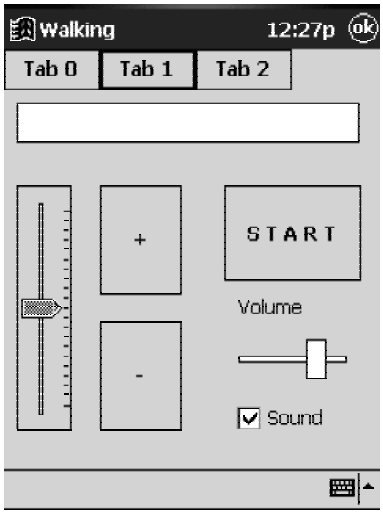


Fig. 4: Screenshot from the Palmtop Computer software

A MS-Windows CE based device with touch screen (Ipaq 3130, Compaq, Houston, Texas, USA) is used as portable main control unit. The advantage is that the operating system takes over all the interface handling, offers convenient methods for database access and synchronisation between Plamtop and PC and also network access.
As development environment MS Embedded Visual C++ is used. ADOCE, also a MS product, supports data exchange with MS Access databases while ActiveSync provides all drivers for automatic database synchronisation.
The graphical user interface is designed as simple as possible. The often used buttons are sized to be easily activated with the fingertip. After turning on the device, the patient can choose a stimulation sequence from the database for standing up, walking and sitting down. In the next step each single channel can be tested with the possibility to adapt the stimulation amplitude within a 20% range. Finally the stimulation is activated and the Palmtop can be stowed away. Then the control is handed over to the Command Unit.
When the stimulation is running the screen shown in Fig. 4 offers the opportunity to adjust the stimulation amplitude of all channels at once, again in a 20% range.
Changes of the device's state like switching from standing up to walking are confirmed by either playing a wav-file or activating the beeper. Obviously the sound volume can be adjusted or switched off.
A wireless LAN connection to the PC, also supported by the operating system and ActiveSync, can be used for data exchange but requires additional hardware (a PCMCIA wireless LAN card and a PC card jacket) to be mounted on the Palmtop Computer.

PC Software Package

The software package for Laptop Computer or PC is designed to devise and optimise stimulation sequences.
Different visual and non-visual software components were created with Delphi (Borland, Scotts Valley, CA, USA) and provide the basic functions like communication with the hardware, graphical stimulation pattern editing and data handling. These components can be easily integrated in any user specific Delphi application or can be converted to ActiveX components and then implemented in most other MS-Windows development environments. After compiling the source code with Kylix (Borland) Linux applications are also supported. For a more detailed description please refer to the poster presentation of M. Russold at this conference.
In this particular case the described components are used to build a comfortable Graphical User Interface (GUI) to ease the patient individual stimulation parameter optimisation. During the testing phase the PC controls the stimulators via cable connection directly or with the wireless LAN connection via the Palmtop Computer.
After the stimulation pattern works satisfying the patient related data is extracted from the underlying MS-Access database and transferred to the Palmtop Computer for patients personal use.

RESULTS

The strictly kept modular concept (in hardware as well as in software design) allowed to break down the whole functionality to specific tasks and to distribute them to eleven microprocessors that cooperate among each other and with one PC and one Palmtop Computer reliably.
First tests showed that the whole system is easy to use although some software improvements are still ongoing.
The previous version had nearly the same features like the described system, but was more bulky and was not intended to be used outside the clinic or rehabilitation centre. Two T6 patients, familiar with FES supported stepping with a six channel surface stimulator, used the prototype and participated in first trials. Both patients agreed with the therapists that smoother and better coordinated movements could be achieved by exploiting the available parameter range during the optimisation phase /5/.
The prototype was also successfully adapted for a project where paraplegic patients rode a newly constructed tricycle by means of FES /6,7/.

DISCUSSION

The introduced eight-channel stimulation system for lower extremities is now subject to an industrial transfer project with Otto Bock Austria. After first testing clinical trials will start soon.

REFERENCES

/1/ Strojnik P, Kralj A, Ursic I: Programmed six-channel electrical stimulator for complex stimulation of leg muscles during walking. IEEE Trans Biomed Eng 1979;26:112-116
/2/ Davis R, Houdayer T, Andrews B, Emmons S, Patrick J: Paraplegia: prolonged closed-loop standing with implanted nucleus FES- 22 stimulator and Andrews' foot-ankle orthosis. Stereotact Funct Neurosurg 1997;69:281-287.
/3/ Holle J, Frey M, Gruber H, Kern H, Stöhr H, Thoma H: Functional Electrostimulation of Paraplegics; Experimental Investigations and First Clinical Experience with an Implantable Stimulation Device. Orthopedics 1984;7:1145-1155.
/4/ Kobetic R, Triolo R J, Uhlir J P, Bieri C, Wibowo M, Polando G, Marsolais E B, Davis J A Jr, Ferguson K A: Implanted functional electrical stimulation system for mobility in paraplegia: a follow-up case report. IEEE Trans Rehabil Eng 1999;7:390-398.
/5/ Bijak M, Hofer C, Lanmuller H, Mayr W, Sauermann S, Unger E, Kern H: Personal computer supported eight channel surface stimulator for paraplegic walking:first results. Artif Org 1999;23:424-427
/6/ Angeli T, Gföhler M, Eberharter T, and Rinder L: Tricycle for paraplegics using functional electrostimulation. Med&Biol. Eng&Comput 1999, 37:326-327.
/7/ Bijak M, Reichel M, Hofer C, Gföhler M, Mayr W, Eberharter T, Angeli T, Lugner P, Rinder L, Kern H: Tricycle for Paraplegic's: Stimulation Equipment. 3rd Int. Conference on Bioelectromagnetism Proceedings (ISBN 961-6210-95-5), 2000, 217-218

ACKNOWLEDGEMENTS

This project is supported by Otto Bock Austria.

AUTHOR'S ADDRESS

Manfred BIJAK, PH.D.
Department of Biomed. Eng. an Physics
Waehringer Guertel 18-20/4L
1090 Vienna, Austria

Tel.: +43-1-40400/1992 (1983)
Fax.: +43-1-40400/3988
E-Mail: m.bijak@bmtp.akh-wien.ac.at

Session 3

FES CYCLING

A COMPARISON OF THE PEAK PHYSICAL WORK CAPACITY DURING ARM ERGOMETRY, FES CYCLING, AND TWO HYBRID EXERCISE CONDITIONS IN SPINAL CORD INJURED

Verellen, J.[1,2], Kerby, B.[1], Olijnyk, B.[1], Saab, S.[1], Smith, G.[1], Jeon, J.[1], Steadward, R.D.[1], Vanlandewijck, Y.[2], Andrews, B.J.[3], Wheeler, G.D.[1]

[1]The Steadward Centre, Faculty of Physical Education and Recreation, W1-67, Van Vliet Centre, University of Alberta, Edmonton, Alberta, Canada. T6G 2H9
[2]Faculty of Physical Education and Physiotherapy, Catholic University of Leuven, Tervuursevest 101, B-3001 Heverlee, Belgium
[3].University of Oxford, England, and The Stoke Mandeville Hospital, National Spinal Injuries Unit.

SUMMARY

The purpose of this study was to compare a newly developed functional electrical stimulation (FES) assisted rowing machine (ROWSTIM) with arm ergometry (ACE), FES cycling (CFES LE), and hybrid cranking and cycling (CFES LE + ACE). Five SCI participants (C7-T12) underwent a progressive maximal peak oxygen assessment to ascertain peak physical work capacity across 4 conditions. Three trials per exercise modality were completed to examine peak physical work capacity and to establish reliability of measurement. A one-way ANOVA with condition as the main factor showed peak absolute and relative functional aerobic capacity and heart rate to be significantly lower for CFES LE versus ACE, CFES LE + ACE and FES ROW measures ($p < 0.05$). However there were no significant differences between both hybrid exercise conditions or between hybrid exercise and arm ergometry. Preliminary results suggest that the ROWSTIM is as effective as an exercise device or training tool for SCI as FES cycling or hybrid cycling and cranking. However, a larger sample size and further technological developments of the ROWSTIM are needed to demonstrate the efficacy of rowing over upper extremity exercise and hybrid cycling and cranking.

STATE OF THE ART

Interest in enhancing exercise opportunities for persons with SCI has been mediated by evidence of increased risk for heart disease in this population./1/ One possible solution to enhance physical activity of persons with SCI is through functional electrical stimulation (FES)-assisted exercise technology. Demonstrated fitness and health related benefits of FES assisted exercise include 1) increased cardiac function, such as oxygen uptake, cardiac output and myocardial function; 2) increased venous return by activating the muscle pump below the lesion level; 3) (theoretically) reduced risk for deep vein thrombosis; 4) increased bone mineral density; 5) increased muscle area and joint range of motion; 6) increased muscle strength and endurance; 7) improved pulmonary function; and 8) increased self esteem and enjoyment of the social contact./2/ More recently research has been focused on combining FES leg ergometry with voluntary arm ergometry, referred to as hybrid exercise (HE). This theoretically augments the cardiovascular stress during exercise and consequently increases cardiovascular training effects./3,4/ Most commonly used in this hybrid approach is the addition of FES cycling to upper arm ergometry. Several studies have already demonstrated increased oxygen consumption levels and cardiovascular training effects during submaximal and maximal work compared to upper or lower extremity exercise alone./3,5,6,7,8/ Others have focused on a new form of hybrid exercise, FES rowing./5/ An FES assisted rowing machine (ROWSTM) was developed at the Robert Steadward Centre, Edmonton, Alberta by Dr Garry Wheeler and associates, to examine the potential for extending hybrid training opportunities for persons with SCI./4,9/ Studies by Laskin (1993) and later by Wheeler et al. (in press) suggest that the peak oxygen consumption measures achieved using the ROWSTIM I and II systems are comparable or superior to those reported in literature for other hybrid exercise systems. In addition, it has been demonstrated that up to 36 sessions of progressive FES rowing produces significant changes in cardiovascular fitness in spinal cord injured participants./5/ Such data are important since the FES rowing device is a much less expensive option for training and is very well tolerated and accepted by SCI

participants who have used it (personal communications during the Laskin et al. (1993), the Wheeler et al. (in press) studies and this investigation).

The purpose of this study was therefore to conduct a within subject comparison of peak physical work capacity as defined by peak functional aerobic power (VO_2 peak) across 4 different types of exercise: arm crank ergometry (ACE), FES leg cycling (CFES LE), hybrid cycling and arm cranking (CFES LE + ACE) and electrical stimulation assisted rowing (FESROW).

MATERIALS AND METHODS

5 participants, 4 with complete and 1 with incomplete SCI underwent a progressive maximal peak oxygen assessment to ascertain peak physical work capacity across 4 conditions. Conditions were arm ergometry (ACE), FES cycling (CFES LE), FES cycling combined with arm ergometry (CFES LE + ACE) and FES rowing (FESROW) with 3 maximal exercise trials per exercise modality to establish reliability of measurement. Each test was preceded by a 2 minute rest. The protocols for each exercise modality were the following:

ACE: Participants were instructed to crank an arm crank ergometer (Monark Model 881, Varberg, Sweden) at 15 Watts for 2 minutes. Every 2 minutes thereafter, power output was increased by 15 Watts until voluntary fatigue. Peak functional aerobic power was defined as VO_2 at the point of failure to maintain 50 rpm arm cranking for at least 15 seconds or until the participants voluntarily stopped cranking.
CFES LE: CFES LE was performed on the ERGYS II cycle ergometer. The test protocol consisted of a 5 minute warm up with assisted pedalling. Participants were then instructed to cycle at 0 Watts for 2 minutes. Every 2 minutes thereafter cycle load was increased by 3 Watts. Peak functional aerobic power was defined as VO_2 at the point of failure to maintain 35 rpm at maximum stimulation.
CFES LE + ACE: The test protocol consisted of a combination of the ACE and the CFES LE protocol. Power output for cycling and cranking were increased every 2 minutes with 3 and 15 Watts respectively in a way that participants were both cycling and cranking simultaneously at their maximum as defined by their previous isolated ACE and CFES LE tests until fatigue. Peak functional aerobic power was defined as VO_2 at the point of failure to maintain 35 rpm cycling at maximum stimulation, failure to maintain 50 rpm cranking or until the participants voluntarily stopped cranking.
FESROW: Participants were instructed to start arm rowing at a heart rate equivalent to 40 % of their ACE functional maximum aerobic capacity. After 2 minutes, participants started arm rowing with assisted passive leg rowing at 50 % of their VO_2 max for 2 minutes. Thereafter, participants rowed (with leg stimulation) for 2 minutes at 60 % of their VO_2 max. After a 1 minute resting period, participants rowed for 2 minutes at 80% of their VO_2 max, and after another 1 minute resting period participants were instructed to row at their maximum until fatigue as defined by collapsing legs during the pull phase /5/ or if the subjects reached exhaustion.

RESULTS

VO_2 peak and HR were significantly lower for CFES LE versus ACE, CFES LE + ACE and FES ROW measures (all $p < 0.05$). VO_2 peak was consistently lower for ACE across all trials versus CFES LE + ACE and FESROW, but the reported differences were not significant ($p>0.05$). (Table 1)

	ACE	CFES LE	ACE + CFES LE	FESROW
VO_2 (l/min)	1.72	1.02	2.03	2.06
VO_2 (ml/min/kg)	20.47	12.23	24.28	23.79
HR (bpm)	160	103	161	162

Table 1: Mean peak absolute (l/min) and relative (ml/min/kg) VO_2, and HR (bpm) during ACE, CFES LE, CFES LE + ACE and FESROW

DISCUSSION

The need to develop exercise opportunities for persons with SCI has already been demonstrated. Ashley et al. (1993) previously suggested the ROWSTIM may be an alternative exercise solution for persons with SCI. Wheeler et al. (in press) demonstrated a 11.2 % increase in peak O_2 consumption after a 3 month progressive rowing training program. However, the only available comparison with other electrical stimulation (ES) assisted exercise modes to date was through literature data. Clearly the next step in the development of the ROWSTIM was to conduct a within subject comparison to determine peak work output from 4 different exercise modalities.

Interestingly, the data indicate how limited the potential CV training effect of cycling is. In agreement with the findings of previous studies, data show that values for hybrid training are significantly higher in comparison with the values of cycling only and clearly suggest that hybrid exercise is superior to simple leg exercise./6,10/ In addition, Hooker (1992) and Laskin (1993) demonstrated a significant difference of hybrid exercise over ACE.

ES assisted and hybrid exercise has already been associated with a number of physiological and psychological benefits. However, additional benefits of FES rowing in terms of client appreciation have also been reported. All participants preferred FES rowing over hybrid cycling and cranking as rowing was considered to be a more natural movement. In addition, since rowing utilises the shoulder retractors and muscles of lower and upper back and forearms, this exercise could have additional benefits reducing the risk of overuse and other wheelchair use related shoulder problems./5/ As well, the ROWSTIM is a minimally adapted version of an exercise tool for able-bodied people and is therefore easier and cheaper to manufacture.

However, there remain a number of considerations that have to be taken into account. Only a small sample size was used, reducing the statistical power of this investigation. Clearly a larger sample size is necessary to further demonstrate the superiority of hybrid rowing over upper extremity exercise and other hybrid training modalities. In addition, there's a number of modifications to the ROWSTIM that have to be considered. Regarding electrode placement, an attempt was made to stimulate the common peroneal nerve to facilitate the forward movement during the return phase. The authors suggest that by stimulating the common peroneal nerve a less powerful spring could be used, increasing the muscle work during training. As well, power output during rowing can not be controlled. Clearly a controllable power output would facilitate comparing hybrid and other training modalities in terms of mechanical efficiency. However, the authors suggest that with further technological developments, rowing represents a better hybrid training activity than ACE or CFES LE + ACE. A brake that would sufficiently prevent the legs from collapsing during the arm pull phase would neutralise the forward momentum originated in this phase. Consequently, leg fatigue would be delayed and exercise time and arm power output would further increase peak functional aerobic capacity. As well, a combination of hybrid exercise and upper extremity exercise alone would allow new participants to continue training while the legs are recovering. As a consequence of these technological developments, an increased arm power output and therefore a further increase in VO_2 peak by neutralising the forward momentum during the pull phase, can be expected.

CONCLUSION

Previous studies have already demonstrated the safety and efficacy of the ROWSTIM II system and suggested that this device represents a great potential for cardiovascular training for persons with SCI. This study clearly demonstrated that the aerobic demand during hybrid exercise and ACE were significantly higher than elicited during electrical stimulation assisted cycling. However, no significant differences were found between both hybrid exercise modalities and ACE. Further technological developments to the ROWSTIM II system are now necessary to further increase the metabolic demand during rowing. At present the team is working on a new break that would allow persons to train longer by minimising the force necessary to control for the forward momentum generated in the legs during the pull

phase. In addition, we are confident that this break will also allow persons to train at a higher power output.

REFERENCES

1. Mohr T, Andersen JT, Biering-Sorensen F, Galbo H, Bangsbo J, Wagner A, Kjaer M. Long-term adaptation to electrically induced cycle training in severe spinal cord injured individuals. Spinal Cord 1997; 35 (1): 1-16.
2. Andrews BJ, Wheeler GD. Functional and therapeutic benefits of electrical stimulation after spinal cord injury. Current Opinion in Neurology 1995; 8: 461-466.
3. Hooker SP, Figoni SF, Rodgers MM, Glaser RM, Mathews T, Suryaprasad AM, Gupta SC. Metabolic and hemodynamic responses to concurrent voluntary arm crank and electrical stimulation leg cycle exercise in quadriplegics. Journal of Rehabilitation Research and Development 1992; 29 (3): 1-11.
4. Laskin JJ, Ashley EA, Olenik LM, Burnham R, Cumming DC, Steadward RD, Wheeler GD. Electrical stimulation-assisted rowing exercise in spinal cord injured people. A pilot study Paraplegia 1993; 31 (8): 534-541.
5. Mutton DL, Scremin AM, Barstow TJ, Scott MD, Kunkel CF, Cagle TG. Physiologic responses during functional electrical stimulation leg cycling and hybrid exercise in spinal cord injured subjects. Archives of Physical Medicine and Rehabilitation 1997; 78 (7): 712-718.
6. Raymond J, Davis GM, Climstein M, Sutton JR. Cardiorespiratory responses to arm cranking and electrical stimulation leg cycling in people with paraplegia. Medicine and Science in Sports and Exercise 1999; 31 (6): 822-828.
7. Gurney AB, Robergs RA, Aisenbrey J, Cordova JC, McClanahan L. Detraining from total body exercise ergometry in individuals with spinal cord injury. Spinal Cord 1998; 36 (11): 782-789.
8. Krauss JC, Robergs RA, Depaepe JL, Kopriva LM, Aisenbury JA, Anderson MA, Lange EK. Effects of electrical stimulation and upper body training after spinal cord injury. Medicine and Science in Sports and Exercise 1993; 25 (9): 1054-1061.
9. In press:
   Wheeler GD, Andrews BJ, Lederer R, Davoodi R, Natho K, Weiss C, Jeon J, Bhambhani Y, Steadward RD. Functional electrical stimulation assisted rowing: improvements in hybrid exercise technologies for persons with spinal cord injury. Archives of Physical Medicine and Rehabilitation. In press.
10. Gurney AB, Robergs RA, Aisenbrey J, Cordova JC, McClanahan L. Detraining from total body exercise ergometry in individuals with spinal cord injury. Spinal Cord 1998; 36 (11): 782-789. Phillips W, Burkett LN. Augmented upper body contribution to oxygen uptake during upper body exercise with concurrent leg functional electrical stimulation in persons with spinal cord injury. Spinal Cord 1998; 36 (11): 750-755

ACKNOWLEGEMENTS

The authors wish to acknowledge the Alberta Heritage Foundation for Medical Research for providing a Phase I technology commercialisation grant in support of this investigation.

AUTHOR'S ADRESS

Joeri Verellen
Katersberg 25
2440 Geel, Belgium
joeriverellen@hotmail.com

LEG POWERED PARAPLEGIC CYCLING SYSTEM USING SURFACE FUNCTIONAL ELECTRICAL STIMULATION

T.A. Perkins*, N. Donaldson*, R. Fitzwater**, G.F.Phillips**, D.E. Wood***

*UCL Department of Medical Physics, London, UK

**Royal National Orthopaedic Hospital, Stanmore, UK

***Salisbury District Hospital, Salisbury, UK

SUMMARY

We have established a simple system for Spinal Cord Injured (SCI) patients to achieve leg propelled cycling using surface Functional Electrical Stimulation (FES). We initially intended our partial lesion T11/12 male paraplegic subject to achieve outdoor leg powered cycle rides that would be useful for both exercise and mobility. We hoped this added function would encourage the patient to greater effort, particularly as more intense exercise /1/ has reportedly given some recovery in bone density as well as the other health benefits expected from more moderate exercise levels /2/. Our first surface stimulation volunteer has repeatedly cycled about 12km at a time (both indoors and outside). After 20 months of FES exercise (at almost 150 minutes per week), bone density in his more paralysed left leg had increased in several sites, that in the tibial tuberosity by 44+/-2%. Other health benefits, including increased muscle bulk and voluntary function /3/ were also noted.

We are currently extending this English pilot study to complete-lesion paraplegics. So far, two out of our three complete paraplegic subjects have already attained kilometre plus leg powered cycling capability.

STATE OF THE ART

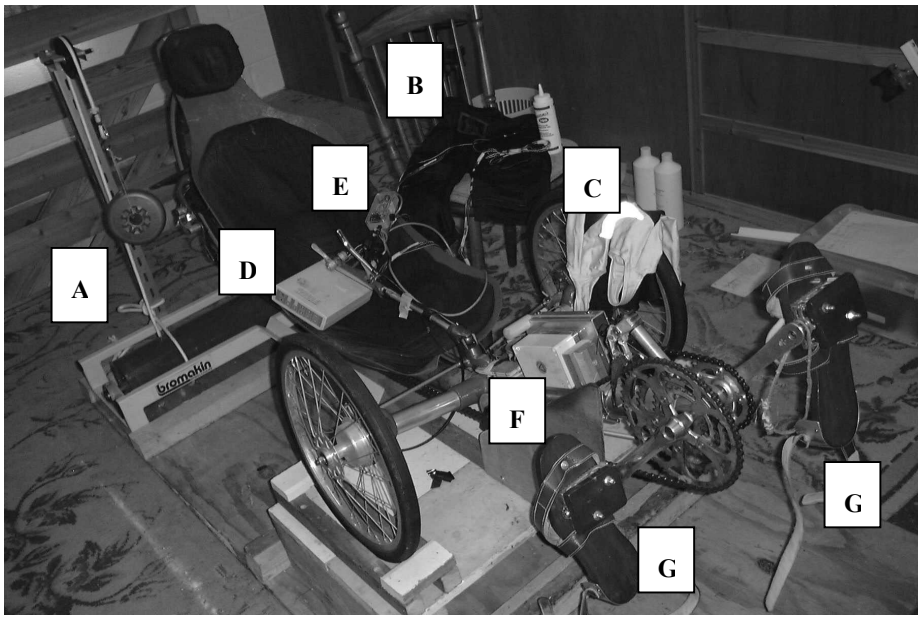
Leg propelled cycling was first demonstrated in the 1980s, both with surface stimulation /4/ and with implanted electrodes /5/. Since then, most FES cycling exercise for SCI patients has been on static ergometers, often in a clinic or laboratory for nominally three 30 minute sessions per week /6/, /7/ and /8/. With compliance around 70%, this amounts to less than 10 minutes per day. Following the demonstration of kilometre plus capability in two complete lesion paraplegics in 1998, /9/ and /10/, we felt it was time FES cycling exercise moved out of the laboratory and into patients' everyday lives, where we expected more frequent and prolonged exercise would be possible. A reliable system for home use was therefore required.

MATERIAL AND METHODS

Equipment

Commercially available recumbent tricycles (Trice and Windcheater) were chosen because of their inherent stability and low seating pressure. Switches were attached to the handlebars for starting and stopping the cycling program. A 7-bit shaft encoder was driven synchronously by the cranks. Purpose-built foot-plates were attached to the pedals to help secure the feet as shown in Fig 1. For the paralysed limbs, Ankle Foot Orthoses (AFOs) were attached to the foot-plates to hold the knees close to the sagittal plane. To control pedalling speed, a throttle potentiometer adjusted stimulation strength (linearly

interpolated between threshold and maximal). A resistance trainer stand was provided so that the tricycle was turned into a static ergometer for indoor cycle exercise at home. For outdoor use, the cycle was removed from the trainer, keeping the vital mechanical set-up for the patient exactly the same. An 8 channel surface stimulator, giving pulses of up to 500µs and 150mA, was used.

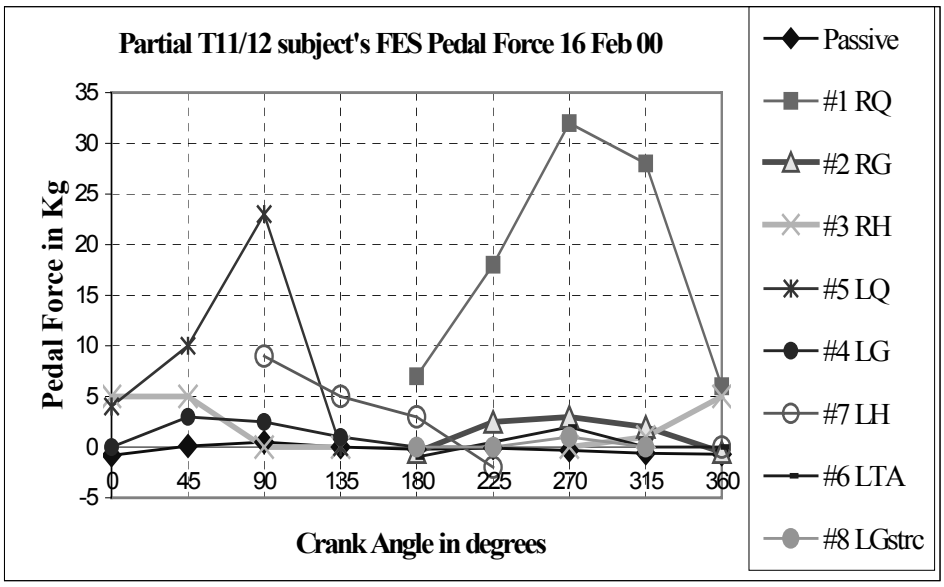


Key to Fig 1:- A: Trainer. B: Electrode shorts. C: Tights to retain cables & electrodes. D: Stimulator. E: Throttle & switches. F: Shaft encoder. G: foot-plates.

Fig 1. Partial lesion subject's Windcheater tricycle at home.

Methods

Seat to pedal distance was adjusted for minimum passive resistance to crank rotation without excessive knee extension. Two of our complete lesion patients required special cranks for this to be possible. For our partial lesion subject, quadriceps, hamstring and gluteal muscles both sides were activated, with gastrocnemius and tibialis anterior on the more paralysed left side. The complete paraplegics used gastrocnemius both sides, leaving out tibialis anterior. In each case, stimulation frequency was 20Hz.



Initially, static pedal force readings were taken with a spring balance for all likely crank angles, as shown in Fig 2, in order to determine favourable crank angles for stimulating each muscle group.

Fig 2. Maximal stimulation pedal forces. Crank angle from Left hip most flexed Top Dead Centre (LTDC).

For all our surface stimulated patients, control was derived from the crank angle shaft encoder, having set start and stop angles for each muscle group as shown in Table 1. A time advance compensated for muscle response delay. This time advance was initially varied in 50ms increments and the resulting pedalling speed against fixed resistance measured with a stopwatch. We found 150ms was broadly optimal for these subjects. Changes in cadence from about 25 to 55 rpm could then be accommodated on load.

Mode 3, program 4 (commutator) **Cycling program for partial T11/12**

Channel	1	2	3	4	5	6	7	8	
Muscle group	**RQ**uad	**RG**lut	**RH**am	**LG**lut	**LQ**uad	**LTibA**	**LH**am	**LGstrc**	
Pulse Current	110	60	60	60	80	40	80	40	**mA**
Pulse Period	50	50	50	50	50	50	50	50	**ms**
Threshold p.w.	40	40	40	40	100	40	40	40	**µs**
Maximal p.w.	304	300	300	300	296	300	302	300	**µs**
Start Angle	183	205	318	45	3	225	115	180	
Stop Angle	312	318	68	139	129	315	219	270	
LTDC offset°	**315**	150	Time Advance in ms						
Shaft sense	**255**		**+ve=0/-ve=255**						

Table 1: Parameter block (from stimulator print out). Start and stop angles in degrees past LTDC.

RESULTS

Fig 3. Partial lesion subject cycling with FES. Note AFO on left.

Over the first 16 months, our partial lesion patient averaged 21 minutes of daily FES training with a simpler system /3/, /9/. He has exercised for a further 16 months with the system described here. At 20 months, bone density in his more paralysed limb segments had increased significantly, that in his left tibial tuberosity by 44+/-2%. With this new system, our subject has repeatedly cycled ca. 12km at a time either in a sports hall or on near level ground outdoors (Fig 3).

In this English pilot study, we now have three complete lesion paraplegics (one implant and two surface stimulated), all using Trice recumbent tricycles (with moulded cushions to reduce seating pressure). So far, two out of these three complete paraplegics have already cycled more than one kilometre at a time while powered by their own FES activated leg muscles.

DISCUSSION

We have clearly established a practicable system for FES leg powered cycling. This may be a valuable function following spinal cord injury, giving considerable health benefits, especially as our patients, training at home, are able to maintain a high level of exercise. We hope to design a new stimulator with

more channels, so that tibialis anterior can be included for the complete paraplegics. Readily replaceable battery packs and improved electrode suits are also under consideration. The high proportion of our patients (3 out of 4) who have already achieved a kilometre plus cycling capability is most encouraging.

REFERENCES

/1/ Mohr T., Podenphant J., Biering-Sorensen F., Galbo H., Thamsborg G. and Kjaer M., Increased bone mineral density after prolonged electrically induced cycle training of paralyzed limbs in spinal cord injured man, *Calcified Tissue International*, **61**, 1997, 22-25.

/2/ Janssen T.W.J., Glaser R.M. and Shuster D.B., Clinical efficacy of electrical stimulation exercise training: effects on health, fitness, and function, *Topics in Spinal Cord Injury Rehabilitation*, **3**, 1998, 33-49.

/3/ Donaldson N., Perkins T.A., Fitzwater R., Wood D.E. & Middleton F., FES cycling may promote recovery of leg function after incomplete spinal cord injury, Spinal Cord, **38**, 2000, 680-682.

/4/ Petrofsky J.S., Heaton H.H. and Phillips C.A., Outdoor bicycle for exercise in paraplegics and quadriplegics, *J.Biomed.Eng.*, **5**, 1983, 292-296.

/5/ Kern H., Frey M., Holle J., Mayr W., Schwanda G., Stohr H. and Thoma H., Functional electrostimulation of paraplegic patients – 1 year's practical application. Results in patients and experiences, *Z-Orthop.*, **123**, 1985, 1-12.

/6/ Petrofsky J.S., Phillips C.A., Heaton H.H. and Glaser R.M, Bicycle ergometer for paralyzed muscle, *J. Clin. Eng.*, **9**, 1984, 13-19.

/7/ Ragnarsson K.T., Pollack S., O'Daniel W., Edgar R., Petrofsky J. and Nash M.S., Clinical evaluation of computerized functional electrical stimulation after spinal cord injury: a multicenter pilot study, *Arch. Phys. Med. Rehab.,* **69**, 1988, 672-677.

/8/ Arnold P.B., McVey P.P., Farrell W.J., Deurloo T.M. and Grasso A.R., Functional electric stimulation: its efficacy and safety in improving pulmonary function and musculoskeletal fitness, *Arch. Phys. Med. Rehab*, **73**(7), 1992, 665-668.

/9/ Carmen Bruck cycling*, Tomorrow's World*, BBC1 TV program, 4 November, 1998.

/10/ Perkins T.A., Donaldson N.deN., Harper V.J., Tromans A.M., Wood D.E. and Rushton D.N., Provision of standing, stepping and cycling for a T9 paraplegic with a lumbo-sacral anterior root stimulator implant (LARSI), in *IFESS Conf. Proc.: 1996-1998 CDROM*, Ed P. Meadows, Pub IFESS, Internat.Funct. Elec., 5030 N. Hill Street, La Canada Flintridge, CA 91011-2335, USA, INSIFESS98 Proceedings, Lucerne, 20-26 August1998, 1999, \Full Papers\Perkins.htm.

ACKNOWLEDGEMENTS

This work was sponsored by INSPIRE, the Wellcome Trust and Royal National Orthopaedic Hospital (Stanmore). We also thank Mr. Tromans & Professor Swain (Salisbury District Hospital), Drs. Middleton & Mathew (Stanmore Spinal Injury Unit) and Professor Hunt of Glasgow University for their support.

AUTHOR'S ADDRESS

Mr T. Perkins, Implanted Devices Group,
UCL Dept. Medical Physics & Bioeng,
11-20 Capper St., LONDON, WC1E 6JA, UK.

e-mail: tap@medphys.ucl.ac.uk.
home page:
http://www.medphys.ucl.ac.uk/research/idg/

FES-CYCLING - MEASUREMENTS AND INDIVIDUAL ADAPTATION OF STIMULATION PATTERNS ON A TEST BED AND A MOBILE TRICYCLE

M. Gföhler[1], T. Angeli[1], P. Lugner[2], C. Hofer[4], M. Reichel[3], W. Mayr[3], M. Bijak[3]

[1]Institute of Machine Elements, Vienna University of Technology, Vienna
[2]Department of Mechanics, Vienna University of Technology, Vienna
[3]Department of Biomedical Engineering and Physics, Vienna University Medical School, Vienna
[4]Department of Physical Medicine, Wilhelminenspital, Vienna

SUMMARY

In this study static and dynamic measurements with paraplegic test persons on a freely adjustable test bed were performed to determine the individual FES-cycling performance and to individually optimize the stimulation patterns. Hydrogel surface electrodes were used to activate quadriceps, hamstrings, and gluteus maximus and to elicit the peronaeus reflex. The optimized stimulation patterns were then applied to cycling on a moving tricycle.

STATE OF THE ART

An important motivation in studying FES (functional electrical stimulation) - cycling is to combine the physiological benefits of FES with the psychological incentive of independent locomotion. As paralyzed muscles are mostly atrophied and only a limited number of effective leg muscles can be reached by surface electrodes it is important to convert the generated muscle forces into drive power with the highest possible efficiency.

Angeli /1/ has shown that the drive power output of neurologically intact subjects can be raised by using an optimized pedal path, which is realized by a 4-bar linkage pedal drive, instead of the common circular pedal path. In a simulation study this pedal path was also optimized for FES pedaling with quadriceps, hamstrings, gluteus maximus and peroneal reflex /2/.

MATERIAL AND METHODS

The measurements with paraplegic test persons were carried out on a freely adjustable test bed (Fig. 1) /3/. The paraplegic test person is seated on a specially adapted wheelchair. It was decided to use only one leg for the measurements to make sure that the results are not influenced by spasms or other forces generated by the other leg. The inclinations of seat and backrest are adjustable in 10° steps. The chair is horizontally moveable along two guide rails, and its position can be fixed in 2.5 cm steps by an alignment pin. For getting onto the test bed the chair is moved to the horizontal end position E to provide enough space between chair and crank bearing. The crank bearing is mounted on an electrical vertical lift. The crank axis is pivoted at the crank bearing, and the force measuring crank is mounted on the measuring side. A chain connects the chain wheel to the torque measuring shaft which is coupled with a gear motor by a torsional stiff coupling. Because the vertical movement of the crank axis the distance between crank axis and torque measuring shaft is changed, but the length of the connecting chain is constant. Therefore the measuring shaft and motor are mounted on a pivoting lever which changes its inclination according to the vertical position of the crank axis. The lift with crank axis, along with the lever-carrying motor and measuring shaft, are mounted on a ground plate. This plate is, like the seat, moveable along two horizontal guide rails, and its horizontal position can be fixed by an alignment pin in 5 cm steps. The pedal is either mounted directly to the crank in PA, as usual moving on a circular path, or to the coupler of a 4-bar linkage in P. This linkage consists of wing, coupler and crank and makes the pedal move along a non-circular pedal path. The wing bearing W is mounted on the same vertical lift as the crank axis C, the bars are connected by pin joints in B and PA. The foot is fixed to the pedal by Velcro fastenings, which means that both tensile and compressive forces are transferred. For paraplegic subjects it is

necessary to fix the ankle joint rigidly because they are not able to stabilize the joint. This is done by an orthosis which also supports side-to-side stability of the leg. The right foot is placed on the ground plate of the test bed in a resting position.

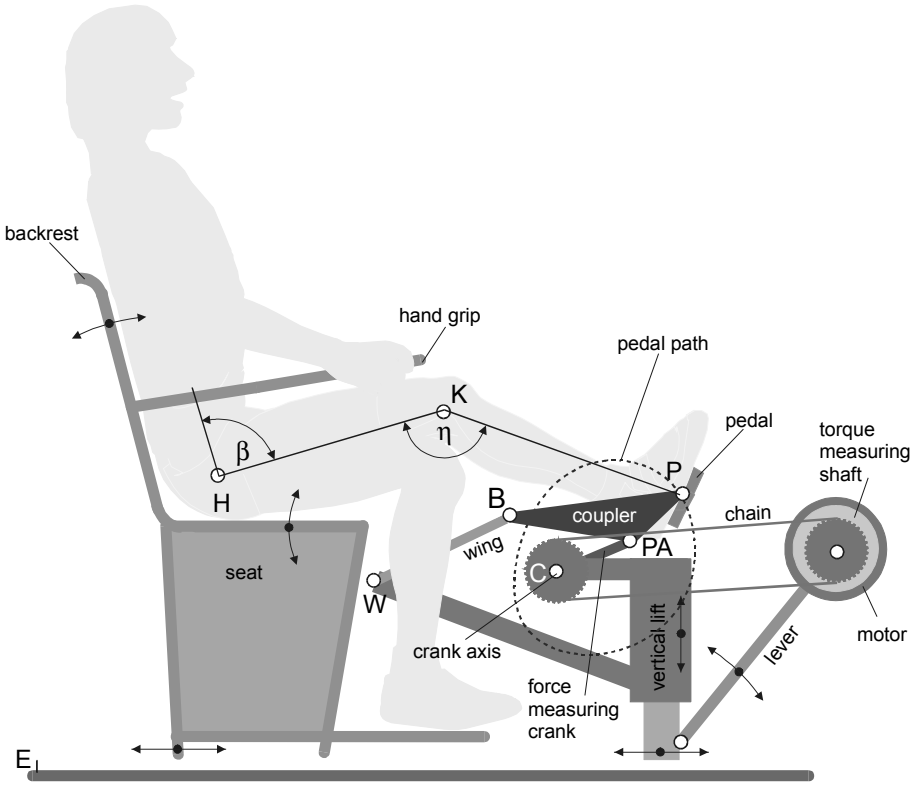
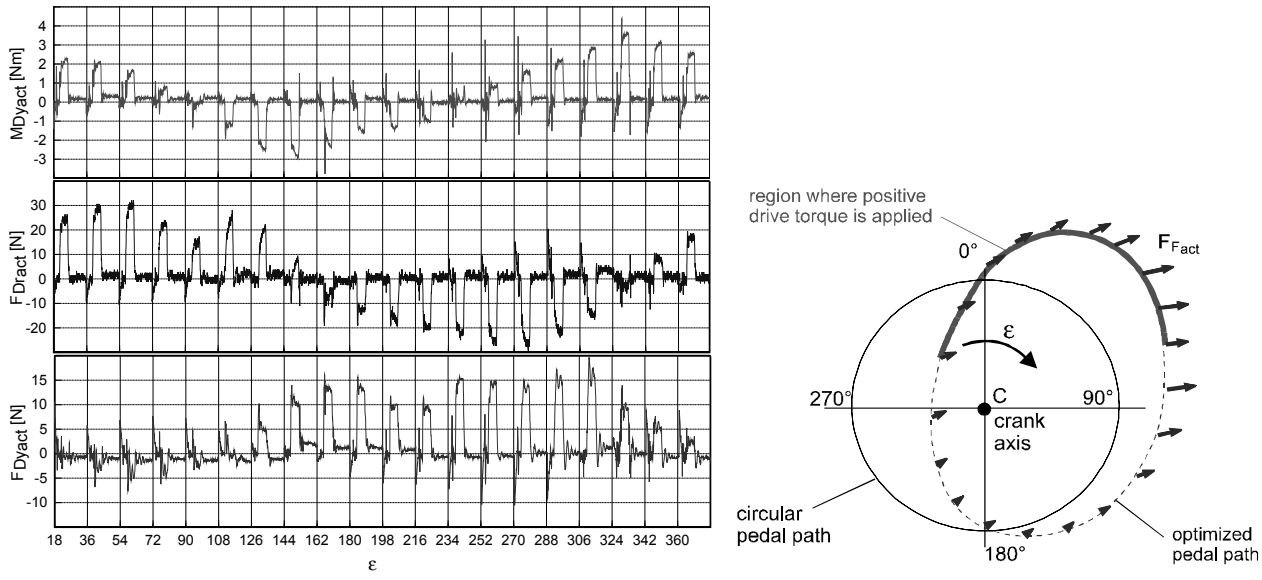


Figure 1: Schematic of the test bed. K and H indicate knee and hip. The arrows show how the elements of the test bed can be adjusted.

RESULTS



a) b)

Figure 2: Results of isometric force measurements at 20 points along the optimized pedal path, stimulation of quadriceps of test person ZA. (a) Active loads (directly caused by active muscle forces) at point D of the crank (passive loads have been eliminated). (b) Resulting pedal force vectors in the parasagittal pedaling plane and region where positive drive torque is applied**.**

For the static measurements 20 equiangular points were defined along the pedal path. In every point the muscle was stimulated for 0.75 sec, the pedal forces were measured by a force measuring crank. Orientation and magnitude of the active forces (directly caused by active muscle forces) applied to the crank in the parasagittal pedaling plane were calculated for a number of geometrical positions of the rider and variations of the stimulation parameters. Out of this data it could be seen in which part of the pedal path positive drive torque was applied. Figure 2 shows results of static measurements.

In the dynamic measurements (Fig. 3) the muscle was stimulated in its concentric range, starting with an estimated interval derived out of the static measurements. Then the start and end points of the stimulation interval were varied to find out in which interval maximum positive drive torque could be applied. Tests with all muscles stimulated together were performed to show how much the muscles influenced each other (Fig. 3d).

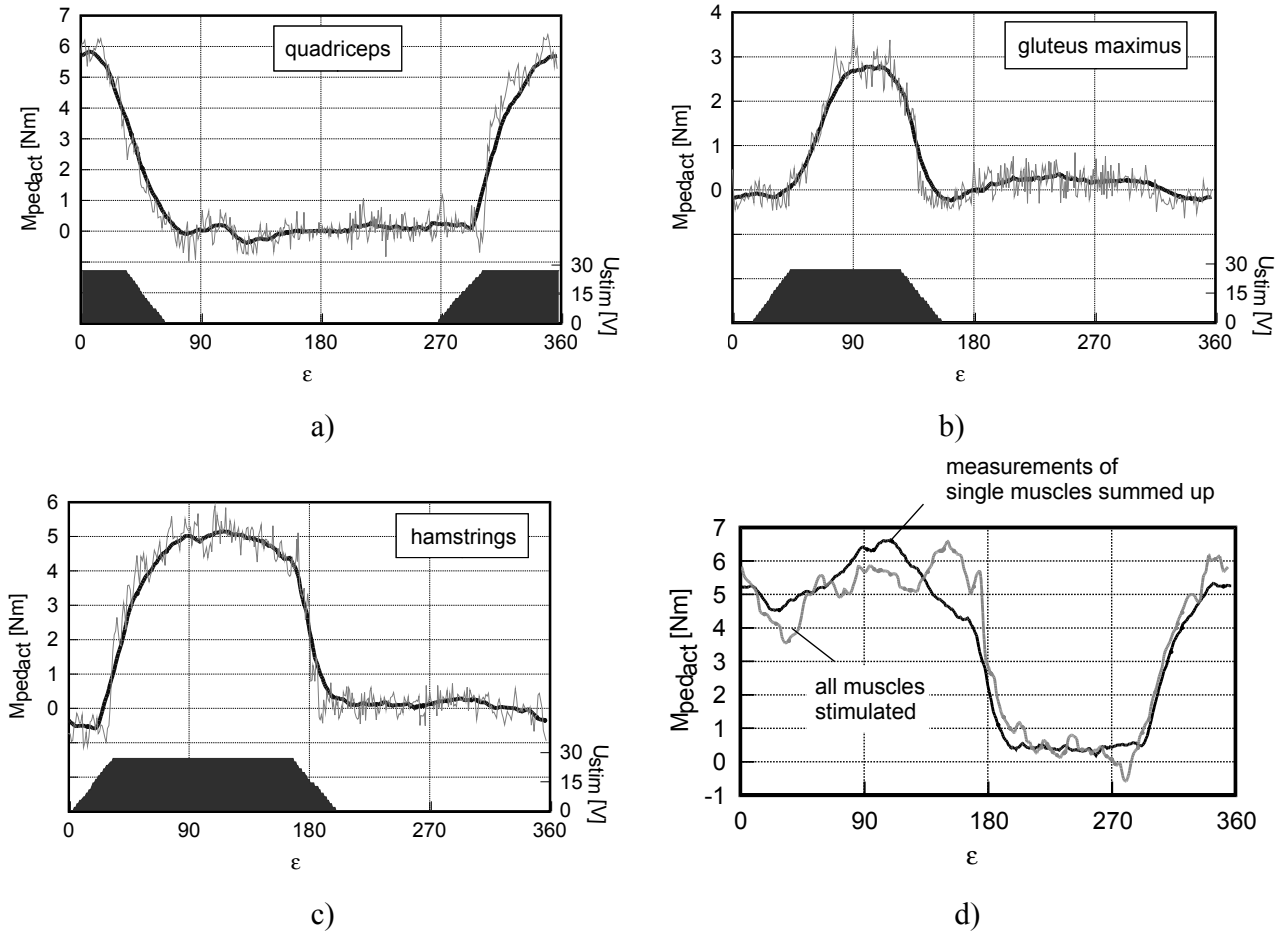


Figure 3: Results of dynamic measurements with test person ZA during one full rotation at 25 rpm: (a) - (c) show the resulting active drive torque from stimulation of (a) quadriceps, (b) gluteus maximus and (c) hamstrings (with optimized stimulation interval (stimulation voltage U_{stim})). (d) comparison between the summation of the results for the single muscles and the measured results of the stimulation with all muscles during one full rotation of the crank

Finally a set of optimized parameters for stimulating all muscles together while pedaling was derived for each individual test person and tested on a specially developed mobile tricycle for paraplegics. Figure 4 shows a paraplegic test person on the moving tricycle /4/ and results of measurements while cycling average crank angular velocity 33 rpm.

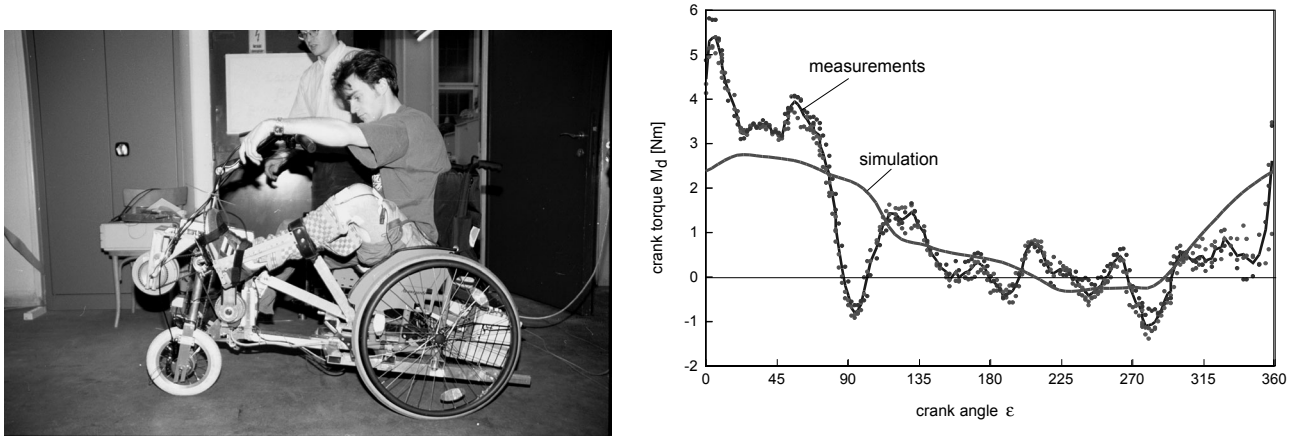


Figure 4: Paraplegic test person on the moving tricycle and results of measurements during steady-state cycling at crank angular velocity 33rpm. The results are compared to results of a simulation study /5/.

DISCUSSION

Leg muscles are activated by surface stimulation and the measurements show which forces and torques are applied to the crank. Thus, stimulation patterns and geometrical position of the rider may be optimized for the development of an optimized cycling movement, influenced by individual parameters, for paraplegic subjects. It has been shown that a moving tricycle can be powered by FES in the optimized stimulation patterns.

REFERENCES

/1/ Angeli T. (1996): 'Propulsion units of bicycles - optimization of capacity', *PhD thesis*, Vienna University of Technology, Vienna, 1996.
/2/ Angeli T., Gföhler M., Eberharter T., Lugner P., Rinder L., and Kern H. (2001): 'Optimization of the pedal path for cycling powered by lower extremity muscles activated by Functional Electrical Stimulation', in Middleton J., Jones M.L., Shrive N.G., Pande G.N. (Ed): 'Computer Methods in Biomechanics and Biomedical Engineering-3', pp.263-268 (Gordon and Breach Science Publishers)
/3/ Gföhler M., Angeli T., Eberharter T., Lugner P., Mayr W., and Hofer C. (2001): 'Test bed with force measuring crank for static and dynamic investigations on cycling by means of functional electrical stimulation', *IEEE Trans. neural Systems and Rehabilitation Engineering TRE*, 9, pp. 169-180.
/4/ Angeli T., Gföhler M., Eberharter T. and Rinder L. (1999): 'Tricycle for paraplegics using functional electrostimulation', *Med. & Biol. Eng. & Comp.*, 37, Supp. 2, pp. 326-327.
/5/ Gföhler M., Angeli T. and Lugner P., (2001): 'Optimal control of cycling by means of functional electrical stimulation - a dynamic simulation study', VIIIth International Symposion on Computer Simulation in Biomechanics, Milan, Italy, accepted.

ACKNOWLEDGEMENT

This work was sponsored by the Austrian Science Foundation - FWF and Otto Bock Austria.

AUTHOR'S ADDRESS

Margit Gföhler
Institute of Machine Elements
Vienna University of Technology

Getreidemarkt 9/306
A-1060 Vienna
margit.gfoehler@tuwien.ac.at

LEG-PROPELLED WHEELCHAIR USING FES

Richard B. Stein*, Su Ling Chong*, Kelvin B. James*, Gordon J. Bell, Ph.D.**

*Division of Neuroscience, University of Alberta, Edmonton, AB, Canada
**Faculty of Physical Education, University of Alberta, Edmonton, AB, Canada.

SUMMARY

A wheelchair has been developed that can be propelled by movement of the legs around the knee joint. Knee movement is generated by FES of the quadriceps (knee extensors) or hamstring (knee flexor) muscles or by voluntary movement. The movement is mechanically coupled to the rear wheel through a combination of levers, gears and chain. Automatic control of the FES is enabled using a rule base software program with knee angle as the input control. Efficiency of using the legs to propel the wheelchair was tested using voluntary activation and FES. PCI measurements were made while subjects wheeled with their arms and legs and where possible walked. Subjects consisted of complete spinal cord injury (9), other motor disorders (13) and control subjects (13). Results show that arm wheeling took significantly more effort (mean PCI = 0.52 beats/m) than walking (0.33) and leg wheeling required the least effort (0.23) in the control sbjects. For SCI subjects leg wheeling with FES required less than 1/2 the effort (0.18) compared to arm wheeling (0.40). The subjects with other motor disorders could walk with substantial effort (1.81) compared to arm (0.76) or leg wheeling (0.64). Control of the wheelchair using FES of the lower extremities is more effecient than wheeling with the arms and may offer superior whole-body fitness and mobility for the SCI subject.

STATE OF THE ART

Many disabled individuals use a wheelchair because they can walk only with great effort, if at all, using either voluntary control or FES. Cerny *et al*. /1/ showed that the decision of persons with SCI to walk or use a wheelchair depended on the energy costs. Subjects who could not walk aerobically with a velocity of at least 54 m/min. tended to use wheelchairs, whereas those who could walk faster within aerobic limits tended not to use wheelchairs. A number of FES systems have been developed that allow people with complete SCI to walk with or without additional bracing /2/3/4/5/. Use of suitable bracing reduces the energy cost somewhat, but the cost is still high and the speed low compared to use of a wheelchair /6/7/. Thus, most individuals use FES systems only for exercise and rely on a wheelchair for mobility in daily tasks.

For such individuals there is no good alternative to the wheelchair at present, but with a lack of exercise the legs become progressively weaker. Bones also atrophy (osteoporosis and osteopenia) with the lack of loading /8/9/. As the muscles atrophy, the wheelchair becomes increasingly necessary, which leads to further atrophy in a vicious cycle. Glaser *et al.* /10/ proposed using FES of leg muscles to propel a wheelchair or a tricycle. These authors built and commercialized a stationary, FES-powered exercise bicycle /11/ with success although it did not offer the user mobility.

Therefore, we built devices that allow individuals to propel a wheelchair with their legs either by voluntary activity or electrical stimulation. The stationary foot rests of a conventional wheelchair were replaced by movable foot rests that allowed movement of the legs about the knee center (Figure 1). This rotation is coupled to movement of the wheels by a sprocket, a chain and a gearbox that converts bidirectional rotation to unidirectional rotation and increases the magnitude by a ratio of 1:3. Thus, FES or voluntary movement of the legs can produce forward movement of the chair. A steering lever is used to control the direction of the wheelchair.

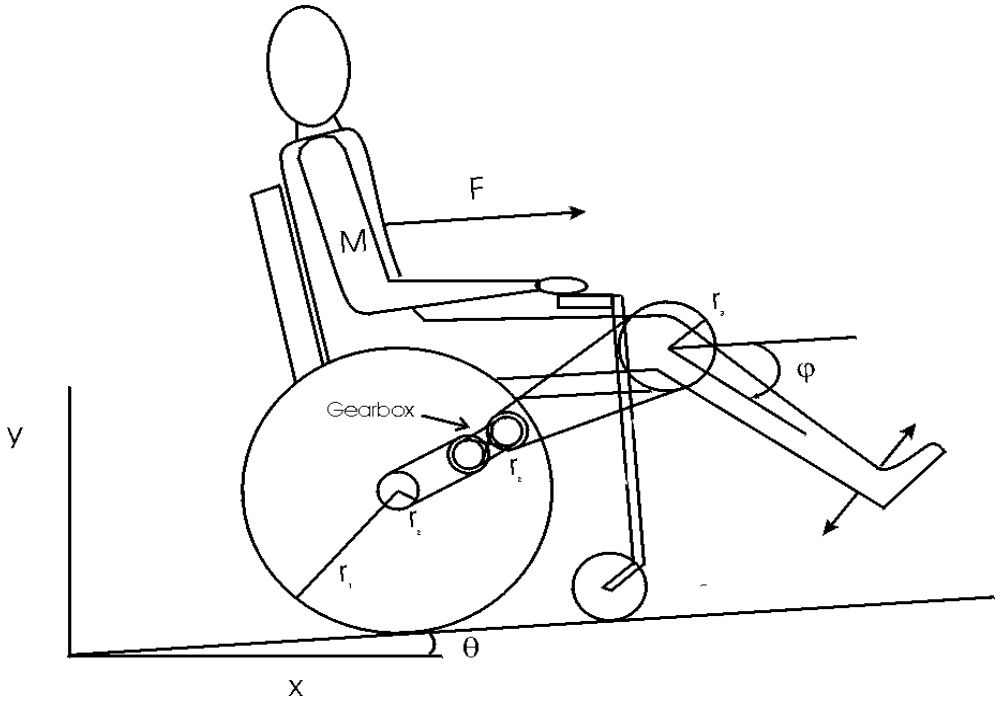


Figure 1; Schamatic diagram of leg powered wheelchair.

MATERIAL AND METHODS

Three groups were studied: control subjects (13) who had no known neural or muscular disease, subjects with complete spinal cord injury (9) who required FES for leg movements and those with a variety of motor disorders (13) who still retained some voluntary control of the legs. The third group included people with incomplete spinal cord injury (4), hip problems (3), head injury (1), spina bifida (1), Parkinson's Disease (1), balance problems (1), stroke (1) and chronic obstructive pulmonary disorder (1). We included such a variety of disorders in this group to see which types of disabilities would benefit most from this new device. In later studies we will concentrate on more homogeneous populations, but some general principles emerged by studying a diverse group initially.

The physiological cost index (PCI) was studied in all subjects. The PCI measures the change in heart rate (beats/min.) during exercise, divided by the velocity of movement (m/min.) /15/. Therefore, the PCI has units of heart beats per meter and is a measure of the effort expended to cover a given distance.

Resting heart rate was measured for 2 min.; then subjects wheeled or walked for 4 min. around a 200 m track. Finally, another 4 min. was recorded, while their heart rate returned toward the resting value. The heart rate during the last 2 min. of exercise was compared to that in the final 2 min. before and after exercise. For subjects who could walk, the resting heart rate was measured while they were seated before and after the 4 min. of exercise for comparison with the other conditions. The order of measurement was randomized between arm wheeling, leg wheeling and walking.

For subjects with a motor-complete SCI, electrical stimulation was provided by an eight channel, portable stimulator. Three electrodes were placed over the quadriceps muscles close to the motor points of vastus medialis, vastus lateralis and rectus femoris muscles. Two or 3 electrodes were also placed over the hamstring muscles, as required for effective stimulation. The stimulus pulses were typically 200-300 μs in duration and were applied at a rate between 20 and 25 Hz. The pulses were partially charge-balanced (an active charge recovery phase followed a monophasic stimulus pulse). Maximal torques produced by stimulation were quite reproducible ($\pm$ 5%) in subjects who were measured on more than one occasion.

Knee angle was measured using a potentiometer mounted on the wheelchair coaxially to the knee centre. The shaft rotated with the foot rest and produced changes in the resistance of the potentiometer that were proportional to changes in the knee angle. When the knee angle exceeded an extension threshold, stimulation of the flexors was turned on and stimulation of the extensors was turned off. Later, when the knee exceeded a flexion threshold, the opposite pattern of stimulation began. In this way

the movement was controlled automatically, but the subject could adjust the stimulus intensity and the upper threshold to make the movements most effective.

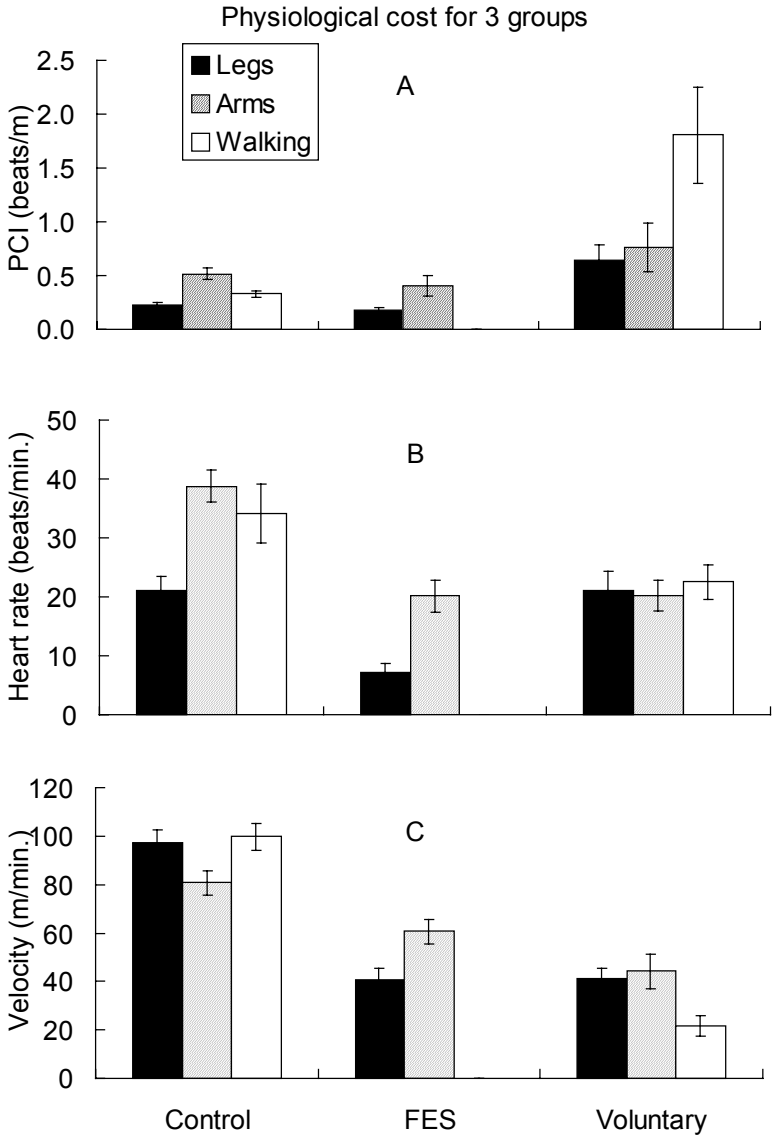


Figure 3. A) Mean (± S.E.) of the physiological cost index (PCI) for the three groups of subjects. Note that wheeling with the legs (left bars in each set) had the lowest cost in all three groups. The mean change (± S.E.) in the heart rate (B) and the velocity (C) that the subjects used during the exercise are also shown, since these are the component elements that make up the PCI.

RESULTS

Figure 3A shows the average data (with the standard error, S.E.) for all three groups studied. For the control group arm wheeling took significantly more effort (mean PCI=0.52 beats/m) than walking (0.33; Students t-test, $p < 0.02$). Leg wheeling was the most efficient (0.23, $p<0.05$), requiring less than half the effort of arm wheeling and 30% less effort even than walking. For complete SCI subjects leg wheeling with FES (0.18) was also significantly more efficient ($p<0.05$), again requiring less than half the effort of arm wheeling (0.40). Subjects in the complete SCI group could not walk. Subjects with other disabilities could walk under voluntary control, but with much more effort (1.81) compared to arm wheeling (0.76) or leg wheeling (0.64). These last two values were not significantly different for reasons that will be considered later.

Since PCI is the change in heart rate divided by velocity, the differences described above could be due to differences in heart rate or velocity or a combination of the two. Figure 3B shows that the voluntary group produced a similar change in heart rate (about 20 beats/min.) under all three conditions. However, this effort produced a much lower velocity when walking (Figure 3C) so the physiological cost was much greater. In contrast, the control subjects matched velocity quite well under the three conditions, so the difference in PCI was mainly due to the change in heart rate required to produce this velocity. The SCI group had a somewhat higher velocity, but a much higher heart rate when wheeling with the arms than with the legs (using FES), which explains the greater PCI during arm wheeling.

DISCUSSION

The major result of this paper is that most people require significantly less effort, in terms of heart rate changes or oxygen consumption, to wheel a chair with the legs than with the arms. This was true for 1) control subjects, 2) those with complete SCI who used FES to move their legs and 3) many subjects with incomplete lesions who retain some voluntary movement of the legs. This is the first time such a dramatic decrease in effort has been produced to our knowledge.

REFERENCES

1. Cerny K, Waters R, Hislop H, Perry J. Walking and wheelchair energetics in persons with paraplegia. Phys Ther 1980;60:1133-1139.
2. Marsolais EB, Edwards BG. Energy costs of walking and standing with functional neuromuscular stimulation and long leg braces. Arch Phys Med Rehab 1988;69:243-249.
3. Kralj A, Bajd T. Functional Electrical Stimulation, Standing and Walking after Spinal Cord Injury. Boca Raton, FL: CRC Press, 1989.
4. Graupe D, Kohn KH. Functional electrical stimulation for ambulation by paraplegics: twelve years of clinical observations and system studies. Malabar FL: Krieger, 1994.
5. Solomonow M,. Aguilar E, Reisin E, Baratta RV, Best R, Coetzee T, *et al.* Reciprogating gait orthosis powered with electrical muscle stimulation (RGO II). Part I: performance evaluationof 70 paraplegic patients. Orthopedics 1997;20:315-324.
6. Hirokawa S, Grimm M, Le T, Solomonow M, Baratta RV, Shoji H, *et al.* Energy consumption in paraplegic ambulation using the reciprocating gait orthosis and electrical stimulation of the thigh muscles. Arch Phys Med Rehab 1990;71:687-694.
7. Nene AV, Patrick JH. Energy cost of paraplegic locomotion using the ParaWalker - electrical stimulation "hybrid" orthosis. Arch Phys Med Rehab 1990;71:116-120.
8. Rodgers MM, Glaser RM, Figoni SF Hooker SP, Ezenwa BN, Collins SR *et al.* Musculoskeletal responses of spinal cord injured individuals to functional neuromuscular stimulation-induced knee extension exercise training. J Rehab Res Devel 1991;28:19-26.
9. Hangartner TN, Rodgers MM, Glaser RM, Barre PS. Tibial bone density loss in spinal cord injured patients: effects of FES exercise. J Rehab Res Dev 1994;31:50-61.
10. Glaser RM, Petrofsky JS, DuFour HR. Wheelchair and drive system therefor. U.S. Patent 1985;# 4,523,769.
11. Petrofsky JS, Phillips CA, Heaton HH. Bicycle ergometer for paralysed muscle. J. Clin. Eng. 984;9:13-19.
12. Petrofsky J, Glaser RM. Vehicle for the paralyzed. U.S. Patent 1983;#4,421,336.
13. Nichols PJR, Norman PA, Ennis JR. Wheelchair user's shoulder? Shoulder pain in patients with spinal cord lesions. Scand J Rehab Med 1979;11:29-32.
14. Sie IH, Waters RL, Adkins RH, Gellman H. Upper extremity pain in the postrehabilitation spinal cord injured patient. Arch Phys Med Rehab 1992;73:44-48.
15. Subbarao JV, Klopstein J, Turpin R. Prevalence and impact of wrist and shoulder pain in patients with spinal cord injury. J Spinal Cord Med 1995;18:9-13.
16. MacGregor J. The evaluation of patient performance using long-term ambulatory monitoring technique in the domiciliary environment. Physiotherapy 1981;67:30-33.

A PERSONAL USERS VIEW OF FES CYCLING

Dr Roger Fitzwater

UK Spinal Injuries Association & Royal National Orthopaedic Hospital, London.

SUMMARY

Two years of FESC (functional electrical stimulation cycling) as a researcher and subject have given me an insight into the direction that future FESC should take as well as providing me with significant health benefits and an enjoyable and functional ability to cycle. If FESC is to benefit SCIPs (spinal cord injured persons) researchers must turn their attention to making the activity convenient and enjoyable.

What follows is a personal view and will be less scientifically rigorous than other presentations but hopefully still of value. It calls upon my experience as a general medical practitioner with a special interest in the value of exercise, a human powered vehicle enthusiast, an amateur FES researcher but most importantly a SCIP and FES cyclist.

STATE OF THE ART

A number of researchers have developed systems of FESCE (functional electrical stimulation cycling exercise) and shown that most of the probable potential health benefits, both general and specific, do occur. Why then, when FESC is possible and the potential benefits enormous, are not a significant number of SCIPs cycling around using their legs, not even in Holland?

Many studies failed to demonstrate sufficient power and endurance to make a significant cycle ride a possibility. However most of the studies had what I regarded as an inadequate training schedule. Twenty minutes three times per week for three months being fairly typical. None of the papers I read used a cycle that was suitable for an open road cycle ride. All the protocols I read were laboratory based making it extremely inconvenient for the subjects to commit themselves to regular and long-term training. In fact none of these researchers had creating a take-home open road cycling function as an objective.

MATERIAL AND METHOD

Before my accident I kept fit by recreational running and hill walking. Since then I have attempted to maintain health and fitness by playing wheelchair basketball (very badly), swimming, wheeling in the countryside and hand cycling. Perhaps this is the reason that I have two golfers elbows, one tennis elbow and intermittently painful shoulders, neck and wrists. These musculoskeletal problems have resulted in me giving up wheelchair basketball and they are beginning to cause me problems when hand cycling.

After my recovery I became interested in the potential of FESC to benefit health and create an additional sport, recreation and transport option. I could see no reason why FESC should not confer all the health benefits on SCIPs that cycling confers on the AB (able-bodied). Such benefits as a decrease in all cause morbidity and mortality, protection against cardiovascular disease, improved fitness, maintenance of function, help in preventing obesity and diabetes, elevation of mood and prevention of depression. As the benefits of exercise obey the law of diminishing returns SCIPs should benefit more from any given amount of exercise than the AB.

Much of the ill health and disability suffered by SCIPs is directly related to the inactivity of their large lower limb muscles. Lack of muscle padding together with poor circulation predisposes to pressure sores, gross muscle size decreases leading to thin unattractive limbs, blood circulation decreases and coagulability increases leading to vascular problems, osteoporosis and joint contractures develop. FESC has the potential to minimise these problems.

In October 1998 I started surface FESC by the method described in the conference manuscript by Perkins et al.

RESULTS

In 1988 I received an injury which left me with an incomplete paraplegia at T11. Before the FESC my standard neurological classification motor score left leg was zero (except for knee extensors 1, long toe extensors 3). My right leg scored 4 for all measurements. My sensory score was a uniform 1. Since the accident I have been a wheelchair user but able to slowly walk short distances with two crutches.
Before I started my FESC training I was only able to rotate the pedals of my recumbent tricycle less than one complete revolution, despite the fact that it was in its lowest gear on a flat, smooth, hard indoor surface. I was unable to rotate the pedals beyond right bottom dead centre.
I have cycled over 10 km on the flat and continuously for over two hours on my trainer. I have even used my FES cycle to accompany my family on a countryside walk. I know from my personal experience that with the assistance of hand cranking, a tandem companion or a motor assist I will be able to cycle as far as I would wish with my able-bodied friends and family. As well as benefiting my health this has and will bring me great pleasure.

Muscle bulk, body image and pressure sores

Muscle bulk has increased benefiting body image and decreasing the likelihood of developing pressure sores. Ultrasound measurements have shown increases in the thickness of the lower limbs. My thighs showed an average increase in depth of 14.5% and my calves showed an average increase of 6.5%. What was even more satisfying than this increase in size was that this increase was almost entirely of muscle bulk. The average thigh muscle depth increased by 21% but the fat depth only increased by 3.75%. In the calves the average muscle depth increased by 8.8% while the average fat depth decreased by 1.75%.
Left gastrocnemius and anterior tibialis were stimulated but due to insufficient stimulator channels right lower leg muscles were not. Notably the non-stimulated calf muscles depth increased by only 3.5% but the stimulated calf muscles by 14%.
The buttock musculature was not scanned but the circumference measured across the anterior superior iliac spines increased from 93cms to 96cms and the circumference across the greater trochanters from 96cms to 98cms. An average increase of 2.7%, perhaps not significant.

Osteoporosis

Significant changes have occurred in my bone density. Lumbar spine density decreased by 6.1% from 110% of ASAAN (age and sex adjusted average normal). Left hip increased by 4.3% (from 87% ASAAN) but the right hip (the good leg) showed no significant change (from 91% ASAAN). The left knee region showed an increase in bone density of 26%. The greatest increases in bone density were seen in the areas of insertion of the muscles stimulated, there was an increase of 44% around the insertion of the left patella ligament. I was pleased to see that the least dense areas of my scanned bones showed an increase and that the indications were that bone density increases were as a result of FESC muscle activity.

Cardiovascular health, fitness

I believe the FESCE is benefiting my cardiovascular health and fitness. My resting pulse rate is normally in the low 60s. During FESCE I become warm and maintain a pulse rate of around 100. I believe that since my FESCE my legs have been less cold.

Mood

I always feel better after my FESCE even when it is indoors on the trainer, which I find boring. It is the same mood improvement that I used to experience after running.

Function

I have had significant improvements in my voluntary function. It was expected that FES generated knee extension forces would increase after FESC and they did, the right by 48% from 69Nm to102Nm and the left by 32% from 72Nm to 95Nm. However what was not expected was the increase in voluntary strength. The right increased by 45% from 59Nm to 85.5Nm while the left increased from 0 to 33.5Nm. This measured quantitative increase in voluntary strength has been mirrored in a qualitative increase in functional ability. I can now pick things up from the floor more easily when standing, walk short distances with one crutch and I use my crutches more and my wheelchair less. I can now voluntarily move my left leg, something I find very useful to relieve cramp when sitting.
I no longer get any clonus in my left leg nor inversion spasm about the left ankle, something that used to be very troublesome and would on occasions prevent me crutch walking.

Problems and disadvantages

The two major problems with FESC are the relative boredom of indoor FESCE and the time taken. Typically a one hour FESCE session at home takes me 2.5 hours mostly because of the time taken to apply and remove electrodes.

DISCUSSION

Relative merits of FES walking and FES cycling

FES walking programmes often have a high profile and get reported by the media. A typical response from most non-SCIPS to such reports is very positive, the researchers and research are "brilliant" and the subject "brave". Many SCIPs respond differently; typically the research might be described as "a waste of resources that could be better spent" and the subject as "mad or sad". Indeed high profile FES walking research is one of the reasons responsible for the negative and cynical attitude that many UK SCIPs have about SCI (spinal cord injury) research. They see researchers not as having the aim of improving the quality of life for SCIPs but rather attempting to make them as much like their able bodied selves as possible.

I do not share these views but I empathise with them. At our present level of technology it is difficult to see how FES walking can be more functional than a wheelchair and thus it is difficult to see what benefits it can offer. It is highly inconvenient, demands enormous energy expenditure and looks grotesque. It is more disabling than enabling. FES walking and the training required does of course offer some health benefits. However because FES walking is so functionally useless and so extremely inconvenient it is doubtful if it will ever significantly benefit the health of SICPs in the foreseeable future.

By contrast FES cycling can be made functional, enjoyable and health promoting with today's technology. It has been very well received by the UK Spinal Injuries Association and many SCIPs have expressed a strong desire to participate.
SCIPs want to enjoy a good quality of life and the means by which that is achieved is not of prime importance. Many if not most are not overly concerned with being "able bodied" or "being able to walk". What they desire is to be able to conveniently access and do those things that are important to them.

Conclusion

Despite the enormous health benefits of exercise SCIPs, like most other people, will not exercise unless that exercise is convenient and enjoyable. Thus research and development's prime objective should be to create a FESC function that is convenient and enjoyable enough to attractive a significant numbers of SCIPs.

The FESC function should be capable of being used on the open road with or without friends and family and be easily usable without any more assistance than that already required for the activities of daily living. To achieve this objective it is necessary that-

- the training period is of sufficient duration, frequency and intensity to allow the development of adequate muscle strength and endurance. This requires that the bulk of the training be done at home.
- the application of electrodes and connecting to the cycle is reasonably quick and easy.
- the cycle is attractive, efficient and looks like a piece of recreational or sports equipment not like a piece of disability or hospital equipment.
- the FES cycle system is capable of the required speed and distance. This will often require an alternative power source such as an ancillary motor, a hand cranking arrangement, a tandem arrangement or any combination of the three.

For many years we have had, through FES, the ability to exercise muscles paralysed through SCI, and through FESC we have had the potential to develop a convenient and enjoyable method of doing so. When FESC becomes convenient, attractive and readily available there will soon be a significant population of SCIPs whose lower limbs will not be wasted. This population will provide the material for much useful research and we will be able to accurately and reliably assess the long-term health benefits of frequent and continued FES exercise.
The equipment developed and expertise gained in creating user-friendly FESC will surely be of benefit in other FES applications. Perhaps it will even assist in the eventual development of a useful and functional FES walking capacity although I suspect a wheeled orthosis will be needed!
The limbs of SCIPs continue to wither. The development and widespread provision of a functional, convenient, and enjoyable FES exercise is long overdue.

ACKNOWLEDGMENTS

I should like to thank Tim Perkins and Nick Donaldson for making it possible for me to cycle with my legs. I should like to acknowledge the achievements of Carmen Bruck in being the first "proper open road" FES cyclist. Carmen is neither a scientist nor doctor but a SCIP who developed her own system.

AUTHORS ADDRESS

Dr Roger Fitzwater, Scarborough House, Pilgrims Way, Wouldham, Kent ME1 3RB.
e-mail roger@rfitzwater.freeserve.co.uk

Session 4

IMPLANT TECHNOLOGY AND APPLICATION 1

27 YEARS OF CLINICAL EXPERIENCE WITH IMPLANTABLE NEUROPROSTHESES FOR VARIOUS APPLICATIONS

Ross Davis

Neural Engineering Clinic, 76 Eastern Av., Maine, USA.

SUMMARY

Since 1973, Dr. Davis has been implanting neural stimulators and later drug pumps to restore or improve motor function in spinal cord and brain injury, stroke and multiple sclerosis. Also these neural implants have been used to modulate and decrease pain (chronic pain syndromes), spasticity (cerebral palsy, spinal cord injury, stroke and multiple sclerosis) and reduce chronic intractable seizures.
During these 27 years, many worthwhile developments in the implantable neural equipment have been realized, as well as many lessons learnt to improve operative techniques to ensure safer and improved results for the Patients.
With the advent of regulatory Boards and increasing restrictions by Health Insurance Companies, more challenges have arisen to allow new equipment and applications to be started, tested and accepted for clinical use and hopefully for payment.
The relationships between Manufacturers and 'Concept Originators' have varied over the years. Often, personalities can make or break the developmental process. The increasing needs to patent ideas/intellectual properties, which sometimes have been around for years and even previously published (missed by the patent officer), have lead to vigorous legal battles, consuming money with time delays, or resulting in surrendering worth-while projects. There is a need for a responsible independent Appeals Board, made up of senior experienced Researcher, to review these disputed patent claims. Then their findings should be admissible in Court, if the case should reach this stage.
Much has been accomplished in the Neuroprosthesis area by the expanding presence of Biomedical Engineers, more scientifically trained Physician and Therapists, and successful Biomedical Manufacturers. Because of the previous and continuing investments by Private Individuals, Universities, Research Foundations, Private and Government Granting Bodies, the future for Neuroprosthetic device development and use continues to look very bright.

STATE OF THE ART UP TO 1973.

Chronically implantable neural stimulators evolved from cardiac pacemaker technology, and began in 1963 with the activation of the phrenic nerve for long-term artificial respiration. The larger field of pain control started in 1965, with a single channel stimulator activating a peripheral nerve, and in 1967 with spinal cord stimulators. In 1972 chronic cerebellar stimulation (CCS) was used to activate the human anterior lobe cortex in patients with disorders of posture and movements secondary to vascular stroke and to cerebral palsy (CP), and achieved a reduction in muscular hypertonus, so allowing functional improvements. In 1973, chronic deep brain stimulation for pain control was started.

CLINICAL EXPERIENCE WITH IMPLANTABLE NEUROPROSTHESES

In 1973, Dr. Davis started implanting stimulators chronically on peripheral nerve (PNS), spinal cord (SCS) and cerebellar cortex (CCS). PNS and SCS were used to modulate and decrease pain (chronic pain syndromes), and spasticity (spinal cord injury: SCI; stroke and multiple sclerosis: MS). CCS was used to reduce spasticity in CP and brain injury, and chronic intractable seizures.
During these last 27 years, many physicians and biomedical engineers have contributed worthwhile developments in the field of implantable neural prosetheses. Many lessons were learnt to improve operative techniques to ensure safer and improved results for the Patients, and to improve the equipment the implanted.

A. Surgical lessons: The most important problem with implanting neural prosthetic devices is the appearance of an infection in the tissues surrounding the implant. After a needle specimen of the fluid around the device has been taken and cultured, antibiotics must be started immediately. Then in the next 1-3 days, the device and electrodes must be removed; and reimplanted in 6-8 weeks. The question always is how did this infection occur and how could it be prevented. Most infections occurring in the tissues around implanted devices are from the skin organisms, principally Staphylococcus epidermitus, which is low on the pathogenic scale and takes months to show its presence as a swelling and slight color change around the implanted and leads. Rarely, Staphylococcus aureus is found in the tissues around the device and usually appears in the tissues as a swelling with considerable redness usually in the first 2-4 weeks following implantation. Very rarely does a blood-borne infection occurs in the tissues around at the implant; these can occur from infected teeth, septicemia or other distant sources, especially in diabetic patients.

Preventative techniques should be followed: 1. Do a thorough skin preparation with a penetrating antiseptic solution. 2. Careful draping of the surgical area particularly with a clear plastic adhesive covering to prevent touching of the skin. 3. Irrigate all wounds with antibiotic solution regularly; leave antibiotic solution around the device and electrodes, and all wounds. 4. When closing the wounds, make sure there is no blood oozing from the tissues, and that the subcutaneous tissues are sutured tightly as a barrier, before closing the skin. 5. Skin sutures should be inserted carefully so that the edges come together evenly; this is the only barrier to stop infection. Leave the skin sutures in for 14 days as a precaution against possible skin opening. 6. Start I.V. antibiotics prior to anesthesia; continue for 10 days by mouth. 7. For future prevention of blood-borne infection: give oral antibiotics when dental surgery is undertaken, and when there are other sites of infection, particularly in diabetic patients.

B. The surgeon and implant devices: 1. It is assumed by the surgeon in the operating room that the device has been fully tested before removing it from its sterile package. 2. The Manufacturing company's "Field Engineer" should be present during the surgery to assist with any testing or calibration that has to be done to insure that the device does work. 3. The engineer should return to the surgeon's office to instruct the nurse/ therapists / surgeon as well as the patient as to how the external controller works and how the engineer can be contacted for future follow-up. Although the above appears obvious, it must also be understood that at this time only a few surgeons are interested or even skilled with the implantable devices. They are generally busy with little time to be dealing with these adjustments. In larger clinics, nurse clinicians and therapists are trained to make adjustments to the patient external controller and accessories. Only in the past 4 years has mutual interests

developed between the physicians and biomedical engineers, resulting in a sharing of each other's society meetings and journal ('Neuromodulation').

C. The relationship between the physician and the manufacturer: Following the experience of a physician implanting devices over time, new concepts of how the equipment could be improved are discussed with the manufacturer. Depending upon the manufacturer, the physician will find a combination of different responses, from one of indifference to one of a partnership, which can benefit both and eventually the patients. The physician, who is the advocate of the patient, may find a conflict of interest and then must seek legal assistance. The physician, usually intent upon publishing his results and concepts, now faces secrecy, patent applications and future royalties. The physician's time and effort now has to be redirected in order that the device development and its animal and clinical testing are carried out with the approval of the medical center's institutional review board. U.S.FDA/ CE approval for early clinical investigations must be sort and carried out prior to further approval to market the device. With a new device comes new surgical procedures; therefore applications must be made for new codings so that the insurance companies will pay for the devices and the procedures, this last step may take up to two years. The above entire process requires considerable time, collaborative effort and understanding by all parties, plus the infusion of sufficient money.

D. Patents: From most physicians' experiences, little is known of patents, the process of drafting and obtaining a patent, and the protections and problems that arise from patents in industry. Many physicians, who have examined patents related to their specialty are often surprised by a claim, for example, made in 1996, which was known by the physician to be published in the medical literature some years before. From a physician's point of view, there appears to be a basic lack of knowledge of the medical literature either by the person who is making the claim or the patent examiner. Since industry is dedicated to producing safe and useful products for the benefit of patients at considerable cost and time, it is recommended that an independent advisory board be made up of experienced medical scientists to examine and determine whether the claim is original or not, and so notified the patent examiners and, if necessary, the court.

E. Patients: Above all, the patient and the medical problem are at the center of our interest and endeavors. The devices that are conjointly developed, manufactured and tested should be of the highest quality with the best support given for its continued care and performance after implantation.

CONCLUSION

Much has been accomplished in the Neuroprosthesis area by the expanding presence of Biomedical Engineers, more scientifically trained Physician and Therapists, and successful Biomedical Manufacturers. Because of the previous and continuing investments by Manufacturers, Private Individuals, Universities, Research Foundations, Private and Government Granting Bodies, the future for Neuroprosthetic implantable device development and use continues to look very bright.

AUTHOR'S ADDRESS

Ross Davis M.D.
Neural Engineering Clinic
76 Eastern Av.
Augusta, ME, 04330
USA
email: rosdavis@qsilver.net

SECOND GENERATION MICROSTIMULATOR

I. Arcos, R. Davis, K. Fey, D. Mishler, D. Sanderson, C. Tanacs,
M.J. Vogel, R. Wolf, Y. Zilberman, J. Schulman

Alfred Mann Foundation, Valencia, CA

SUMMARY

The first generation injectable microstimulator was glass encased with an external tantalum capacitor electrode. This second generation device utilizes a hermetically sealed ceramic case, with platinum electrodes. Zener diodes protect the electronics from defibrillation shocks and from electrostatic discharge. The capacitor is sealed inside the case so that it cannot be inadvertently damaged by surgical instruments. This microstimulator referred to as BION®, is the main component of a 255-channel wireless stimulating system.

BION® devices have been implanted in rats for periods of up to 5 months. Results show benign tissue reactions resulting in identical encapsulation around BION® and controls. Stimulation threshold levels did not change significantly over time and ranged between 0.81 to 1.35 mA for all the animals at 60 µsec pulse width. All of the tests performed to date indicate that the BION® is safe and effective for long-term human implant.

We have elected to develop BION® applications by seeking collaboration with the research community through our BION® Technology Partnership.

INTRODUCTION

The BION® system is a wireless network of up to 255 single-channel stimulators controlled and powered by an RF link from a central external controller. Each stimulator consists of a ceramic and titanium cylinder capped at each end with anode and cathode, both platinum. The cylinder is 15.6 mm long and 2.5 mm in diameter. Each stimulator produces asymmetric biphasic capacitively-coupled constant-current pulses. Pulse width (0 to 500 µsec), pulse amplitude (0 to 40 mA) and pulse frequency (0 to 3,472 pps shared among all active BION® stimulators) are controlled digitally by the external controller via a 2 MHz AC magnetic link. [1-6]

The second generation BION® stimulator includes additional characteristics. a) Bipolar zener diodes were installed around the electronics module to protect the BION® from defibrillator shocks and from electrostatic discharge. These diodes can protect against 25,000 volts, 10 µsec electrostatic discharge and against a 0.3 amp. defibrillator pulse. b) The tantalum capacitor was moved into the package to protect it from accidental damage during handling and implantation. c) The packaging was changed to a 250-micron walled ceramic cylinder hermetically sealed to titanium annular rings, to which platinum electrodes are attached at either end.

MATERIALS AND METHODS

In-vitro Tests

BION® stimulators underwent accelerated life and mechanical stress tests. The accelerate test was performed with 31 BION® internal electronics activated for 1,000 hours each at an elevated temperature of 125ºC while being operated at a maximum energy level (burn-in test). For the mechanical stress tests, the case was subjected to a three-point bend test and a tensile test.

In-vivo Tests

A. Surgical Preparation

Experiments were carried out on 11 rats (4-5 months old, 280 - 300 g females). Animals in the Study Group (n=6) were implanted with four devices: one BION® in each thigh, and the two passive controls were implanted sub-fascially on both sides of the thoracic spine area. To insert the implants in the hindlimbs, the sciatic nerve was exposed at the thigh level by separating the fascia and dissecting deeper between the vastus lateralis muscle and the biceps muscle. A 1-2 cm long tunnel was dissected along the mid-thigh exposing the sciatic nerve. The BION® was oriented so that the cathode was inserted towards the knee. Implants were not sutured. The fascial layer and the skin layers were then closed. The 2 passive controls, a BION® without electronics and a similar sized silicon rod (NuSil Technology, Carpenteria, CA), were implanted in the posterior thoracic area. A mid line skin incision was made midway along the thoracic spine, then about 1 cm off the midline, the underlying muscle fascial layer was separated. A 1-2 cm long tunnel was dissected at 45 degrees outwards on each side so that the 2 controls could be inserted. The fascial layer and skin were then sutured closed. The Control Group consisted of 5 rats implanted with a BION® casing (left thigh) and a silicone rod (right thigh) adjacent to the sciatic nerve.

B. Stimulation

After surgery, animals were allowed to recover for 2 weeks. Animals in the Study Group were placed into rodent restrainers (Harvard Apparatus, Holliston, MA) and then in groups of three into the transmitting coil that sends data and power to the BION® stimulators. In all but one animal, the BION® implanted in the left leg was set as inactive and the stimulation settings were programmed to zero. Stimulation settings (Pulse Width, Pulse Amplitude) required to produce threshold twitches of the muscles were determined for each active BION®. The Current amplitude used during daily stimulation was set in order to produce a strong visible twitch (usually twice the threshold). Three animals were stimulated with a stimulation pattern of 25-s OFF period and 5-s ON period at 20 Hz. The other three animals were stimulated with a stimulation pattern of 5-s OFF period and 5-s ON period at 20 Hz. In all the animals, stimulus trains were applied starting at two 20-min sessions a day and finishing with three 1-h sessions a day, four to five days a week, for a period of up to five months [7].

C. Measurements

In order to verify the implant location X-rays were taken after the 2-week recovery period and again before the explantation. To take x-rays, animals were anesthetized with isofluorane gas, 1.5 - 2.5 % via facemask. Thresholds to produce a palpable muscle were measured twice a month for every implanted BION®. The daily stimulation was video recorded.

D. Tissue Processing

Animals were euthanized using pentobarbital S (100 mg/kg, i.p.). Passive implants and their surrounding tissues were dissected from the mid thoracic spine area. Each rear leg containing a BION® was also dissected. All the samples were preserved in 10% neutral buffered formalin and sent to a commercial lab (Pathology Research Laboratory, Berkeley, CA).

Implants were removed and each muscle sample containing fibrous tissue capsule was divided in three segments including anterior, middle and posterior levels. The corresponding segment of Sciatic nerve at the various regions was also included. The sections were sent to HistoTec Laboratory (Hayward, CA) for tissue processing and hematoxylin and eosin slide preparations. Selected slides were further stained with Siever-Munger and Cresyl Echt Violet stains to evaluate the nerve fibers.

RESULTS

In-vitro Tests

All 31 units successfully passed the 1,000-hour burn-in test. When comparing the device parameters before and after the burn-in, only the 10 microamps recharge current was altered. The average 10 microamps recharge current increased by about 4 microamps and the maximum increase in recharge

current was 6.7 microamps. The changes in the 0, 100, and 500 microamps recharge current were negligible.

The mechanical tests showed that the ceramic case can sustain approximately 40 lbs of force in a three-point bend test while the glass BION® fails at around 5 lbs. The ceramic-metal brazed case pulls apart at 40 pounds of applied tension, which is equivalent to over 10,000 psi of internal pressure.

In-vivo Tests

All BION® devices stimulated and produced muscle contractions in the legs during the total duration of the study. The twitches were well tolerated by the animals, which usually slept during the stimulation period. During the study period, thresholds were measured for each implanted device, including the inactive BION® devices. These thresholds did not change significantly over the study period and rather decreased when compared to those values measured right after the implantation (Figure 1). Figure 2 shows the cumulative charge density applied to each animal. No device migration was found.

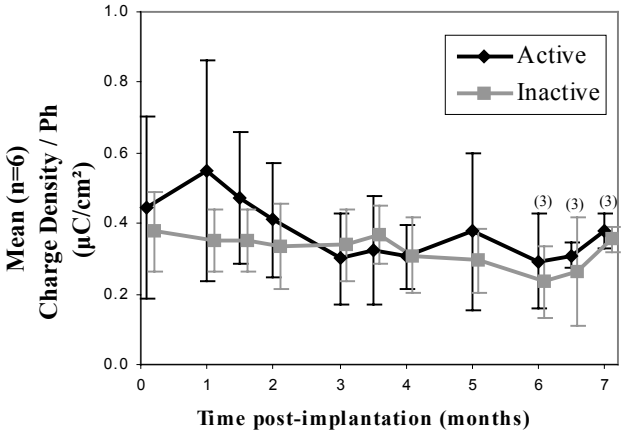


Figure 1. Threshold level mean for active and inactive BION® stimulators

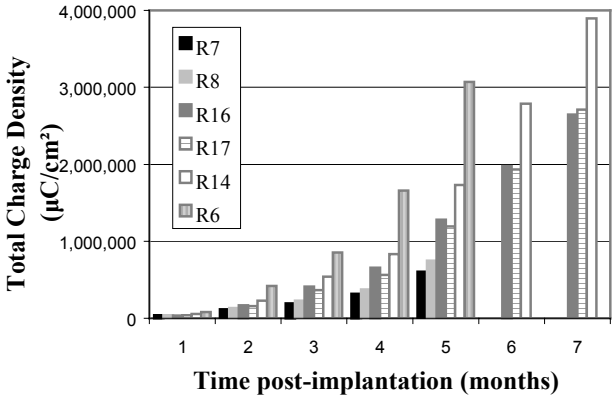


Figure 2. Cumulative charge density applied to each individual

All sections of the leg muscles implanted with BION® or silicone rods had similar fibrous tissue reaction (encapsulation). The round capsules consisted of compact laminated layers of fibro-collagenous tissue varying from 50 to 100 microns in thickness. There were varying degrees of mononuclear cell infiltration, primarily small macrophages and lymphocytes, within the capsular walls. Polymorphonuclear cells (neutrophils) were rare. Sciatic nerves and/or nerve branches were evident in each of the sections. Microscopically, all nerve bundles and muscle fibers were within normal limits. The two special stains confirmed cellular integrity of the nerve sections (figure 3).

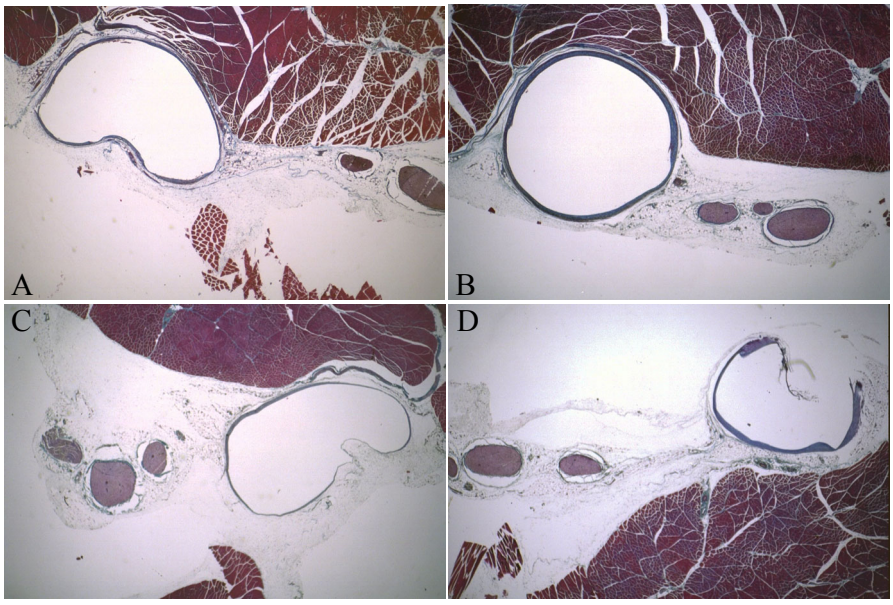


Figure 3. Photomicrographs of the implantation site of (A) an active BION®; (B) inactive BION®; (C) casing; and, (D) silicone rod. Sciatic nerve bundles, capsule and muscle tissue are shown.

DISCUSSION

All of the tests performed to date indicate that the BION® is safe and effective for long-term human implant. BION® technology could enable many FES applications. We considered developing applications in house vs. the alternative of developing an infrastructure that would enable collaborators to develop applications with our support. In coordination with Advanced Bionics – the licensor of this technology, we opted to do the latter. Our decision was primarily based on our desire to make BION® technology as widely available as possible. This infrastructure is the BION® Technology Partnership or BionTech™.

Through BionTech™ we offer technical and regulatory support, an arena for communication and exchange of information through our web site: www.biontech.org and of course, BION® devices available for a nominal price.

Also, to support this project we have been developing a control system containing external hardware and fitting software both designed for many purposes. The external controller is easily wearable; it can control up to eight BION®devices and can be attached to a wide range of coils. The fitting software is also designed for general purpose and for ease of use in a clinical setting.

REFERENCES

[1] Loeb GE, Richmond FJR, Olney S, Cameron T. Bionic Neurons for Functional and Theurapeutic Electrical Stimulation. 20th Annual IEEE-EMBS, Oct.29-Nov. 1, 1998, Hong Kong.
[2] Cameron T, Loeb GE, Peck RA, Schulman JH. Micromodular implants to provide electrical stimulation of paralyzed muscles and limbs. IEEE trans Biomed Eng 44:781-790, 1997.
[3] Cameron T, Richmond FJR, Loeb GE. Effects of regional stimulation using a miniature stimulator implanted in feline posterior biceps femoris. IEEE trans Biomed Eng 45:1036-1043, 1998.
[4] Cameron T, Liinamaa TL, Loeb GE, Richmond FJR. Long-term biocompatibility of a miniature stimulator implanted in feline hind limb muscles. IEEE Trans Biomed Eng 45:1024-1035, 1998.
[5] Loeb GE, Zamin CJ, Schulman JH, Troyk PR. Injectable microstimulator for functional electrical stimulation. Med Biol Eng Comput 29:NS13-NS19, 1991.
[6] Troyk PR, Schwan MA. Close-loop class E transcutaneous power and data link for microimplants. IEEE Trans Biomed Eng 39:589-599.
[7] Guajardo A, Sutherland H, Jarvis JC, and Salmons S: "Conditioning muscles for fatigue resistance: the effect of on/off pattern," Proc. 5 th Annual IFESS Conference, pg. 119. June, 2000.

ACKNOWLEDGEMENTS

The authors would like to thank:
- A. Grinnell, R. Edgerton, R. Roy, and M. Herrera (UCLA, CA) for assistance with the surgery and preparation of the animal tests.
- Previous funding of National Institute of Health.

BION® is a registered mark of Advanced Bionics Corp., Sylmar, California.

AUTHOR'S ADDRESS

Isabel Arcos
Alfred Mann Foundation
e-mail: Isabela@amf.org
home page: www.aemf.org

BILATERAL COCHLEAR IMPLANTATION

W.Baumgartner, J.Hamzavi, C.Vasak, B.Schenk, S.M.Pok

ENT University Department Vienna, Austria
Währinger Gürtel 18-20 A-1097 Wien

wolf-dieter.baumgartner@akh-wien.ac.at
www.univie.ac.at/cochlear

SUMMARY

Since 1995 22 deaf patients (8 adults, 14 children) underwent bilateral cochlear implantation in a two step surgery. In the adult group we could identify, that bilateral implantation only in selected cases is useful. After two years experience in bilateral adult cochlear implantation, we started in children in 1998. So far all bilaterally implanted school children attend mainstream primary school.

STATE OF THE ART

In the last decades around 35.000 patients (out of them around 15.000 children) were cochlear implanted world wide. Cochlear implantation became a routine procedure in the treatment of postlingual deafness in adults and deafness (in all etiologies) in children. Cochlear implantation still is no indication in prelingually deafend adults, or adults integrated in the deaf community by using sign language. Generally only one ear will be implanted . For decades bilateral hearing aid support for hearing impaired patients (especially in children !) is out of discussion. The importance of bilateral hearing for the development in children is quite clear, but bilateral cochlear implantation is still rare. The new approach of this study, out of the experience of more than 340 cochlear implantations at the ENT university department Vienna, is the bilateral cochlear implantation. Right now there are only very view ENT centers woldwide, performing bilateral implantation (for example ENT University department Würzburg, Germany- and the Vienna ENT University department, Austria). Right now bilateral cochlear implantation happened only incidentially. For example up to date, there are only 12 bilateral CI recipients in the US and none in Japan. Aim of this study and sharing this experience with international colluegues, is to define bilateral implantation as a new state of the art in future, when indicated in a patient.

MATERIAL AND METHODS

Since 1995 eight adults underwent bilateral cochlear implantation in a two step surgery. Six patients are wearing bilaterally implants from the Combi 40 - Combi 40 + series (Med El, Innsbruck, Austria) or older analog implants (Med El, Innsbruck, Austria), one patient wears two implants from different manufacturers- Nucleus mini 22 (Cochlear corp., Melbourne, Australia) / Combi 40+ (Med El) and one patient wears Nucleus mini 22 / Nucleus contour (Cochlear corp., Melbourne, Australia). Analysing the patients data , ethiology, surgical situs and implant type it is useful to form four different groups.

Group A: older analog device versus Med El Combi 40 / Combi 40+
Group B: Nucleus mini 22 versus Med El Combi 40+
Group C: same implanttype on both ears, but different fitting (= programming) parameter, because of etiology or surgical circumstances

Group D: same cochlear implant bilaterally (for example two Med El, or two Nucleus) and symetric fitting parameter

After two years experience in adults we started bilateral implantation 1997 in children. So far 14 children underwent cochlear implantation in a two step surgery. Prior to bilateral implantation all children were good unilateral cochlear implant performers. The decision in favour of bilateral implantation was done after careful audiological reevaluation, together with parents and children. If necessary, the radiological evaluation through CT and MRI scans was repeated, for example in case of meningitis. The timeperiod in between the first unilateral and the following contralateral (bilateral) cochlear implantation, was within two months, up to seven years (mean 20 months). The childrens age at the time of bilateral implantation was between 24 months and 14 years old (mean 80 months).

RESULTS

As a result, significant superior speech understanding- especially in noise, and acoustic orientation, can be achieved through bilateral cochlear implantation, in a “group D situation” only !
To obtain a subjective and in speech understanding performance evident benefit of a bilateral cochlear implantation, two multichannel fast stimulators, are mandatory. Otherwise the “better” ear remains dominant. For the intended and prooved benefit of bilateral implantation we recommend bilateral cochlear implantation in a “group D” situation only.
We could not observe any surgical or psychological problem. All “group D” children and adults (out of them seven school children), wear both implants with a behind the ear speech processor (Med El Tempo+®, Nucleus esprit®) all over the day. Seven school aged children (=100%) attend regular mainstream school successfully. Children, parents and teacher report improved life quality, which is correlated to enhanced speech understanding and directional hearing. According to this data and the cumulative bilateral cochlear implant experience in children and adults of over 36 years, bilateral cochlear implantation should be golden standard in all meningitis cases and should be at least considered- especially in children, whenever audiologically, anatomically and financially possible.

REFERENCES

1. W.Baumgartner, W.Gstöttner, J.Hamzavi, P.Franz
   RESULTATE BILATERALER COCHLEAIMPLANTATIONEN
   Acta Oto-Rhino-Laryngologica Nova, 1999: 9/1-2; 25

2. J.Helms, C. Ilberg, W.Gstöttner, J.Müller, J.Kiefer, W.Baumgartner, M.Zwicknagel, C.Zierhofer, B.Stöbich
   COMPARISON OF THE MED EL BTE PROCESSOR TEMPO+ AND THE BODY WORN PROCESSOR CISPRO+
   Proceedings of the 6th International Cochlear Implant Conference Miami 2000; 139/18

3. Baumgartner W.D., Gstöttner W., Hamzavi J., Franz P.
   RESULTS OF BILATERAL COCHLEAR IMPLANTATION
   Abstractbook 3rd International Symposion on Electronic Implants in Otology & Conventional Hearing Aids, Birmingham 2000: 31-32

4. W.Baumgartner, J.Hamzavi, B.Egelierler, W.Gstöttner
   RESULTS OF BILATERAL COCHLEAR IMPLANTATION IN 10 CHILDREN
   Abstract book of 5th European Symposion of Paediatric Cochlear Implantation, Antwerpen 2000: 100

5. Baumgartner W.D., J.Hamzavi, P.Franz, W.Gstöttner
   MODERNE OHRIMPLANTOLOGIE
   Acta Oto-Rhino-Laryngologica Nova, 2000: 10/2; 56

6. K.Strohmayer, W.D.Baumgartner
   COCHLEAR IMPLANTKINDER IM REGELSCHULSYSTEM EIN SCHULVERSUCH
   Acta Oto-Rhino-Laryngologica Nova, 2000: 10/2; 59

7. W.Baumgartner, J.S.Hamzavi, B.Egelierler, C.Vasak, W.Gstöttner
   RESULTS OF BILATERAL COCHLEAR IMPLANTATION IN CHILDREN
   Abstractbook 8th Symposium "Cochlear Implants in Children"
   House Ear Institute 2001: 47

8. Baumgartner W., Hamzavi J., Vasak C., Schenk B., Jappel A.
   RESULTS OF BILATERAL COCHLEAR IMPLANTATION IN 13 CHILDREN
   Abstractbook 3rd Asia Pacific Symposium on Cochlear Implant and related Sciences 2001: 54

9. W.Baumgartner, J.Hamzavi, C.Vasak, P.Franz
   BILATERALE COCHLEA IMPLANTATION BEI KINDERN
   HNO Informationen 2/2001: 58

10. Baumgartner WD, Hamzavi JS, Vasak C, Schenk B, Pok SM
    RESULTS OF BILATERAL COCHLEAR IMPLANTATION IN 14 CHILDREN
    Abstractbook ESPORL-Conference 2001: 14

ACKNOWLEDGEMENT

The authors thank Mrs.cand.med. Alexandra Jappel , collecting all the patients data.

THE 'MIVIP' VISUAL PROSTHESIS FOR OPTIC NERVE STIMULATION

J. Delbeke*, M.C. Wanet-Defalque*, B. Gérard*, M. Troosters**, G. Michaux*, and C. Veraart*

Université catholique de Louvain, Neural Rehabilitation Engineering Lab, Brussels *
& NeuroTech S.A., Louvain-la-Neuve, Belgium**

SUMMARY

The 'MIVIP' visual prosthesis generates visual perceptions well below safety and stimulator saturation limits. These phosphenes are of reasonably small size and broadly distributed in the visual field. They can thus be used to convey useful visual information. Psychophysical evaluations are being performed in order to assess the implantee's benefits in the use of the «MiViP» optic nerve visual prosthesis. In a pattern recognition task, the performance improved regularly with practice with an increasing score and a decreasing delay to recognition. These observations open the way towards an evaluation of general mobility improvement with the portable system. In conclusion, the results obtained so far still support the potential usefulness of the optic nerve visual prosthesis. A low-resolution artificial vision can be expected from the prosthesis after extensive training.

STATE OF THE ART

The visual cortex has been the stimulation target for the very first visual prosthesis implanted in a blind person /1/. More recent attempts /2/ rely on high-tech electrodes and improved stimulators but have not yet reached the level of chronic human implantation. In industrial countries, a significant number of the late blind patients who might benefit from a visual prosthesis are suffering from retinitis pigmentosa. It has been shown /3/ that in this disease, a large number of ganglion cells do survive even when the photosensitive layer of the retina has dyed out completely, thus leading to complete blindness. This fact has led several teams to explore the possibility of a direct retinal stimulation, either with subretinal electrodes, between the retina and the choroids, or epiretinal devices placed between vitreous and retina /4/. The main advantage of leaving the brain intact is counterbalanced however by huge technological hurdles that have not been passed yet. As a simpler and more immediately available technique, a self sizing cuff electrode /5/ was implanted around the optic nerve of a blind volunteer /6/. In this implementation, control of the potential field generated by four contacts is used to selectively stimulate a fraction of the nerve /7/. The purpose of this study is to provide an update account of the potentialities of this approach as can be deduced from results obtained in the first implanted human volunteer.

MATERIALS AND METHODS

A 59 years old lady with retinitis pigmentosa has been implanted with a self-sizing spiral cuff electrode around her right optic nerve on February 1998. Later, on August 20 2000, the percutaneous lead was replaced by an implanted stimulator and antenna for telemetry. The silicone rubber cuff electrode includes 4 platinum contacts of 0.2 mm² area. These are driven by the independent current sources of the implanted stimulator. Biphasic pulses with charge recuperation have a time resolution of 21.3 μs, and a current intensity ranging from 10 μA to 3 mA with a non-linear amplitude resolution. Each stimulator has an output span of ± 8.5 Volts, which thus provides a range of 17 Volts when one of the contacts is used as an anode in a bipolar montage. Only half that voltage is obtained with a monopolar montage whereby the stimulator case is used as reference.

Based on data from the literature /8/, /9/, the stimulation strength was kept below 340 nC for single pulses and 100 nC for 300 Hz pulse trains, which corresponds to 170 and 50 µC/cm².phase respectively.

The optic nerve activation is achieved under control of external equipment using radio-frequency transmission with a 3 Mbit/s data rate. This external equipment includes a dedicated head-worn artificial retina and a digital signal processor. This system extracts significant pixels from camera images and defines appropriate stimulating conditions of the nerve susceptible to elicit corresponding phosphenes in real time. Psychophysical experiments included a pattern recognition task /10/. Fifty simple patterns were used during a ten-session program with feedback from the instructor. Four evaluation sessions were embedded in the training program to assess learning improvement.

This project fully complies with the Declaration of Helsinki, and was approved by the Ethics committee of the School of medicine and University Hospital of the University of Louvain, Brussels.

RESULTS

Measured electrode impedance values /11/ have been used to calculate the stimulator saturation levels as plotted for different pulse durations in figure 1. Another trace of the figure describes the safety limits complied with in this study. As can be seen from the left part of the figure, the safety level corresponds to the 200 µs monopolar single pulse saturation point, but falls clearly below that level for longer pulses. The right part of figure 1 gives the corresponding values for a burst stimulation: 300 Hz safety limit.

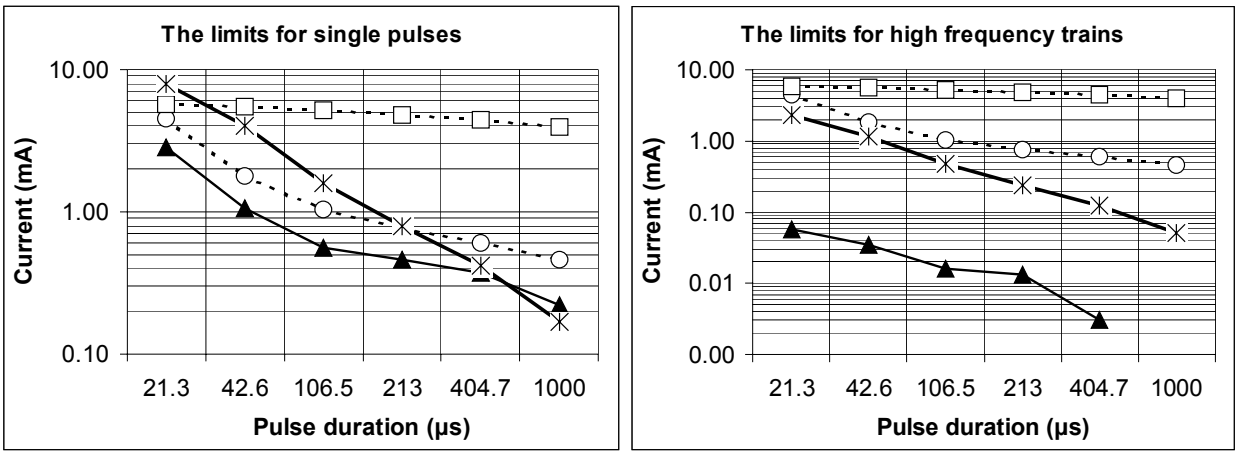


Figure 1.
Left panel: *Current ranges for single pulses of various durations. Open squares represent the saturation level of the stimulator output in a bipolar montage (17 Volts range), open circles represent the same limit when the stimulator case is used as a reference (8.5 Volts limit). Asterisks stand for the safety limit as gathered from the literature. Filled triangles represent the typical phosphene perception thresholds.*
Right panel: *Current ranges for pulse train stimulations. All symbols are similar to those of figure 1. The safety limit and typical threshold traces are obtained here for high frequency pulse trains.*

The perception threshold bounds the lower side of the operational range. Again, the single pulse results are plotted on the left while the right panel of the figure holds the values for 17 pulse trains at 160 Hz. Although the single pulse thresholds are close to the corresponding monopolar saturation

level, a useful range is still available for pulse durations below 200 µs. A much broader stimulation parameter choice is given with pulse trains but in this case, safety considerations will have to limit the current values used.

The phosphenes described have areas from 1 to 50 square degrees. Their central position in the visual field can reach from 35° upwards to 50° downward and from 30° to the right to 30° left of the vertical meridian. For repeated stimuli, flicker fusion is observed between 8 and 10 Hz but perception fades out between 1 and 3 s. A few phosphenes can be produced simultaneously, be it through the interlacing of the pulses of individual trains. The phosphene position is stable enough for these summated stimuli to be perceived as reproducible shapes.

Phosphenes generated by stimuli selected according to the processed signal from the head-worn camera have allowed successful pattern recognition. In this task, the volunteer scans a projection screen with head movements. Between 4 and 24phosphenes have been used in this test with somewhat better results for the larger number of phosphenes. There is an obvious learning curve, which tends towards stabilisation over the 12 sessions. When the system exploits 24 phosphenes, recognition times including the scanning fall below 1 minute with a score of 60%.

DISCUSSION

The thresholds to single pulses are markedly higher than for burst stimulations. This is perfectly in line with the already described link between perception threshold currents and other stimulation parameters /12/. Safety limits abound in the literature /8/, /9/, but they have been obtained in different nervous structures, using other electrodes and incompatible stimulation regimen. These must thus be interpreted with caution because all those parameters could make a significant difference. However, the stability of our long-term results over more than two years /12/ do suggest that the maximal values selected here are unlikely to have damaged the optic nerve in any way. In addition to the given limits, single pulses durations should be kept at 200 µs or less. Saturation can occur with monopolar single pulse stimulation using an 8.5 volts stimulator output range. Pulse trains are not likely to lead to such a limitation.

The phosphenes can be assembled in simple patterns /10/. Even very few phosphenes generated according to the output from a forehead worn camera can lead to satisfactory recognition of simple shapes, using scanning head movements. Accepting that a visual prosthesis can by far not be compared with normal vision, the results obtained still hold much promise when considering the level required to reach usefulness for the totally blind /13/. Evaluation of general mobility improvement with the portable system is in preparation.

The results thus obtained demonstrate the potentials of the optic nerve visual prosthesis. A low-resolution artificial vision can be expected but will require extensive training. The final objective of the project to allow the volunteer to better cope with her visual environment during mobility and grasping remains realistic.

REFERENCES

/1/ Brindley GS, Lewin WS. : The sensations produced by electrical stimulation of the visual cortex. Journal of Physiology (London), 1968,196(2):479-493.
/2/ Sterling TD, Bering EA, Pollack SV, Vaughan HG. Visual prosthesis: the interdisciplinary dialogue. Academic Press, New York, London. 1971;1-382.
/3/ Santos A, Humayun MS, de Juan E, Jr. et al. : Preservation of the inner retina in retinitis pigmentosa. A morphometric analysis. Arch Ophthalmol, 1997,115(4):511-515.
/4/ Rizzo JF, Wyatt J, Humayun M et al. : Retinal prosthesis: an encouraging first decade with major challenges ahead. Ophthalmology, 2001,108(1):13-14.
/5/ Naples GG, Mortimer JT, Scheiner A, Sweeney JD. : A spiral nerve cuff electrode for peripheral nerve stimulation. IEEE Trans Biomed Eng, 1988,35(11):905-916.
/6/ Veraart C, Raftopoulos C, Mortimer JT et al. : Visual sensations produced by optic nerve stimulation using an implanted self-sizing spiral cuff electrode. Brain Res, 1998,813(1):181-186.
/7/ Veraart C, Grill WM, Mortimer JT. : Selective control of muscle activation with a multipolar nerve cuff electrode. IEEE Trans Biomed Eng, 1993,40(7):640-653.
/8/ McCreery DB, Agnew WF, Yuen TG, Bullara LA. : Comparison of neural damage induced by electrical stimulation with faradaic and capacitor electrodes. Ann Biomed Eng, 1988,16(5):463-481.
/9/ Agnew WF, McCreery DB, Yuen TG, Bullara LA. : Histologic and physiologic evaluation of electrically stimulated peripheral nerve: considerations for the selection of parameters. Ann Biomed Eng, 1989,17(1):39-60.
/10/ Wanet-Defalque MC, Delbeke J, Michaux G et al: A visual prosthesis based on electrical stimulation of the optic nerve. Proceedings of the 5th Annual Conderence on the International Functional Electrical Stimulation Society - IFESS2000, Aalborg, Denmark, June 18-21th 2000, 146-148.
/11/ Delbeke J, Gérard B, Veraart C: The electrical behavior of a cuff electrode implanted on a human optic nerve. The 6th Annual Conderence on the International Functional Electrical Stimulation Society (IFESS), Cleveland, Ohio, USA, June 17-20th 2001, 323-325.
/12/ Delbeke J, Parrini S, Michaux G, Vanlierde A, Veraart C: Perception threshold changes in phosphenes generated by direct stimulation of a human optic nerve. Proceedings of the 5th Annual Conderence on the International Functional Electrical Stimulation Society (IFESS), Aalborg, Denmark, June 18-21th 2000, 152-155.
/13/ Terasawa Y, Yagi T, Uchikawa Y: Quantitative evaluation of reading ability using visual prosthesis simulator. Invest Ophthalmol Vis Sci, 2001, 42(4): S813.

ACKNOWLEDGEMENTS

CEU grants # 22 527 (MiViP) and IST-2000-25145 (Optivip); FMSR grant # 3.4584.98.

AUTHOR'S ADDRESS

Dr. J. Delbeke
Neural Rehabilitation Engineering Laboratory
Avenue Hippocrate, 54, B-1200 Brussels, Belgium
e-mail: delbeke@gren.ucl.ac.be
home page: www.gren.ucl.ac.be

Session 5

IMPLANT TECHNOLOGY AND APPLICATION 2

THE HAEMODYNAMIC FUNCTION OF INTRATHORACIC SKELETAL MUSCLE VENTRICLES AFTER RECOVERY FROM SURGERY IN PIGS

Massimo Capoccia, Hazel Sutherland, Stanley Salmons, Jonathan C. Jarvis

Department of Human Anatomy and Cell Biology, The University of Liverpool, U.K.

SUMMARY AND STATE OF THE ART

The shortage of donor organs for heart transplantation and the high cost of mechanical assist devices highlight the need for alternatives for the treatment of end-stage heart failure. A promising experimental approach is the use of skeletal muscle ventricles (SMVs), which can be connected to the circulation in various ways to provide both left and right ventricular assistance (1-5). A subcutaneous SMV has pumped in the circulation of the dog as an aortic diastolic counterpulsator for more than 4 years (5). Here we present a new recovery model in pigs. The aim of the study was to form an intrathoracic pumping chamber from the latissimus dorsi muscle (LDM) and to connect it to the descending thoracic aorta in a single surgical procedure without compromising the host circulation.

MATERIALS AND METHODS

Female domestic pigs were used according to the Animals (Scientific Procedures) Act 1986, which governs animal experimentation in the United Kingdom.
Animals were premedicated with azaperone (Stresnil) 2-4 mg/kg i.m., 30-45 minutes prior to transfer to the anaesthetic room. Antibiotic prophylaxis was provided with lincomycin 4.5-11 mg/kg. Induction was achieved with the administration of an i.v. bolus of propofol (Diprivan) 1-2 mg/kg followed by endotracheal intubation. Respiratory assistance was provided to maintain oxygen saturation above 95%. General anaesthesia was maintained by a continuous i.v. infusion of propofol 8-12 mg/kg. Analgesia and respiratory depression were provided by the administration of alfentanil 0.5-1.5 micrograms/kg/hr or diamorphine 0.062-0.09 mg/kg/hr.

Conditioning
The anterior border of the left latissimus dorsi muscle was exposed through a limited flank incision with minimal disturbance of its blood supply. The muscle was partially dissected free from subcutaneous tissue and the chest wall. An epimysial monopolar electrode was placed between the muscle and the costal surface and secured with non-absorbable suture. The pulse generator (Itrel, Medtronic, Inc.) was placed beneath the left rectus sheath muscle. During the operation one unit of blood was harvested and stored under controlled temperature for transfusion purposes in the next, more major surgical procedure. After one week, the stimulator was programmed to deliver stimulation to the LDM at a frequency of 1 Hz with a pulse duration of 190 msec and an amplitude of 5V. Four weeks of prestimulation was used to transform the muscle to a fatigue-resistant type.

Construction of SMVs
The animals were anaesthetized according to the same protocol. The stimulator was removed from the abdominal pocket and a further unit of blood was harvested and stored. The left latissimus dorsi muscle was dissected free from subcutaneous tissue

and the chest wall, leaving its humeral insertion and the neurovascular bundle intact. In order to make the SMV blood-tight during this procedure, the muscle was fashioned into a pumping chamber around a pre-formed lining. The lining was a composite homograft constructed from the pulmonary artery with a complete ring of right ventricular muscle and part of the descending thoracic aorta of a donor pig. Two new electrodes (Model 6500; Medtronic, Inc.) were secured in place, one on the deep and one on the superficial surface of the proximal insertion of the LDM. The SMV was transposed into the left hemithorax through a window in the chest wall, created by partial resection of the anterior portion of the third rib.

Connection to the circulation

The chest was entered through the fifth intercostal space and the pericardium was opened. Two epicardial unipolar pacing leads were placed on the surface of the left ventricle. The muscular and epicardial leads were tunnelled to the original subcutaneous abdominal pocket and connected to an R-wave synchronous pulse train stimulator (LD Pace II, CCC Uruguay). The stimulator was programmed to deliver a burst of impulses at 33 Hz every third left ventricular diastole (1:3 assist ratio).

The SMV conduit was connected to the descending aorta by an end-to-side anastomosis. Blood was then allowed to flow in the SMV, which had previously been filled with heparinised normal saline. Haemostasis was ensured and two chest drains were placed. The wounds were closed in layers and the animal transferred to a recovery area, where two members of the team monitored it over night.

Haemodynamic data collection and analysis

Haemodynamic data were recorded at the time of connection of the SMV. A more complete analysis of SMV function was made at an elective terminal procedure after one week. The previous left thoracotomy was opened and the SMV assessed visually. The heart was then approached through a midline sternotomy. Transducers were placed to measure left ventricular pressure and volume, SMV pressure and volume, SMV and aortic arch pressure, and flow in the aortic root and in the conduit connecting the SMV to the host aorta. The transducers were secured in place with non-absorbable purse-string sutures. Data, including pressure-volume loops, were collected with the SMV off and with the SMV activated at a range of delays and durations in relation to the cardiac cycle.

RESULTS

Three animals were recovered successfully. In two, the SMV was still pumping but the aortic homograft was kinked, resulting in thrombus formation within the SMV. Haemodynamic measurements were made from one of these SMVs after the removal of thrombus and reconfiguration of the conduit. In the third the problem of kinking was avoided by constructing the SMV closer to the aortic arch with a shorter conduit. The SMV was contractile and free from thrombus after pumping in circulation for one week. Activation of the SMV from the end of systole for 80% of diastole increased mean diastolic blood pressure above control levels (obtained with the SMV off) by 11.2±1.6% in one animal and by 15.8±0.3% in the other. Peak diastolic aortic pressure was increased by 16.9±1.4 % and 20.1±1.2%. The left ventricular stroke work in the post-assisted beat was decreased by 8.7±5.8% and 10.1±2.2% and the overall stroke work by 7.4±1.2% and 9.4±0.8%.

DISCUSSION

Permanent cardiac assistance from skeletal muscle is an attractive prospect. So far only dynamic cardiomyoplasty and dynamic aortomyoplasty have been introduced into clinical practice. Although the protocol widely used for cardiomyoplasty is far from ideal (6, 7), follow-up studies show that 80% of the patients experience symptomatic improvement. It has, however, been difficult to demonstrate any consistent haemodynamic change (8). The benefit probably derives from a reduction of wall stress, which interrupts the progressive cycle of dilatation and overload. Under these conditions, the muscle is heavily loaded and operates far from the peak of its power curve (9). The active assistance available from skeletal muscle can be optimized by configuring it as a separate auxiliary pump, or skeletal muscle ventricle. The effectiveness of this approach has been clearly demonstrated by work in Detroit, where subcutaneous skeletal muscle ventricles (SMVs) have pumped as diastolic counterpulsators in dogs for up to 4 years (5), longer than any other cardiac assist device, mechanical or biological, in man or animal. The surgical technique for constructing SMVs and evaluating them in circulation has undergone progressive refinement and problems such as thromboembolism and rupture have been largely overcome (10, 11). However, some issues need to be addressed. Subcutaneous placement of the SMV is clearly not feasible in man. The current two-stage surgical procedure is unacceptable for patients in end-stage heart failure. The ligation of the descending thoracic aorta would not be accepted by the majority of surgeons.

We chose the pig as our animal model because it has been agreed that the SMV approach must be shown to work in species other than the dog before clinical application could be considered. Pigs are more sedentary animals than dogs and their muscular and cardiovascular physiology is more similar to that in the human. Our objective was to place an SMV intrathoracically in a single-stage surgical procedure without compromisisng the host circulation. Only one end-to-side anastomosis was required to connect the SMV in circulation. Our preliminary results have shown that these targets have been achieved. The SMV was able to reduce the workload of the heart at the same time as increasing the mean diastolic pressure, and therefore the pressure available to drive coronary flow. The benefits of stimulating the LDM prior to raising it as a graft have been discussed previously (12-14). The main drawback of this approach is the need for a preliminary invasive procedure under general anaesthesia, but this problem could be overcome by the use of non-invasive or minimally invasive techniques of electrode and device placement or stimulation. The use of the composite homograft was the key to forming and connecting the SMV in a single procedure. It has the further beneficial effect of reducing the effective preload for the SMV, and thus decreases the risk of ischaemia of the inner wall. The functional role of the homograft lining could be taken on in clinical practice by a synthetic composite or by a tissue lining produced in culture.

REFERENCES

1. Hooper TL, Stephenson LW. Using skeletal muscle to assist the heart. Br Heart J 1991;66:261-3.
2. Niinami H, Hooper TL, Hammond RL, Ruggiero R, Lu H, Spanta AD, Pochettino A, Colson M, Stephenson LW. Skeletal muscle ventricles in the pulmonary circulation: up to 16 weeks' experience. Ann Thorac Surg 1992;53:750-7.
3. Lu H, Fietsam R Jr, Hammond RL, Nakajima H, Mocek FW, Thomas GA, Ruggiero R, Nakajima H, Colson M, Stephenson LW. Skeletal muscle ventricles: left ventricular apex to aorta configuration. Ann Thorac Surg 1993;55:78-85.
4. Hooper TL, Niinami H, Hammond RL, Lu H, Ruggiero R, Pochettino A, Stephenson LW. Skeletal muscle ventricles as left atrial-aortic pumps: short-term studies. Ann Thorac Surg 1992;54:316-22.
5. Thomas GA, Hammond RL, Greer K, Lu H, Jarvis JC, Shortland AP, Pullan DM, Salmons S, Stephenson LW. Functional assessment of skeletal muscle ventricles after pumping for up to four years in circulation. Ann Thorac Surg 2000;70:1281-90.
6. Gealow KK. Latissimus dorsi stimulation in dynamic cardiomyoplasty: how should we proceed ? Basic Appl Myol 1998;8:41-50.
7. Salmons S. Permanent cardiac assistance from skeletal muscle: a prospect for the new millennium. Artif Organs. 1999 May;23(5):380-7
8. El Oakley RM, Jarvis JC. Cardiomyoplasty: a critical review of experimental and clinical results. Circulation 1994;90:2085-90.
9. Salmons S, Jarvis JC. Cardiomyoplasty: the basic issues. Card Chron 1990;4:1-7.
10. Thomas GA, Lu H, Isoda S, Hammond RL, Nakajima H, Nakajima HO, Colson M, Stephenson LW. Pericardium-lined skeletal muscle ventricles in circulation up to 589 days. Ann Thorac Surg 1994;58:978-88.
11. Thomas GA, Lu H, Isoda S, Hammond RL, Nakajima H, Nakajima HO, Stephenson LW. Skeletal muscle ventricles in circulation: decreased incidence of rupture. Ann Thorac Surg 1996;61:430-6.
12. Mannion JD, Velchik M, Hammond R, Alavi A, Mackler T, Duckett S, Staum M, Hurwitz S, Brown W, Stephenson LW. Effects of collateral blood vessels ligation and electrical conditioning on blood flow in dog latissimus dorsi muscle. J Surg Res 1989;47:332-40.
13. Jones J, Emmanuel J, Sutherland H, Jackson MJ, Jarvis JC, Salmons S. Stimulation-induced skeletal muscle damage: cytoprotective effect of prestimulation. Basic Appl Myol 1997;7:39-44.
14. Tang ATM, Jarvis JC, Hooper TL, Salmons S. Cardiomyoplasty: the benefits of electrical prestimulation of the latissimus dorsi muscle in situ. Ann Thorac Surg 1999;68:46-51.

ACKNOWLEDGEMENTS

The financial support of the British Heart Foundation is gratefully acknowleged.
We thank Mr. Robert Galvin and Miss Elaine Sawyer for their technical assistance.

CONDITIONING PROTOCOL AND STIMULATION DEVICE OF THE VIENNA SMV PROJECT.

H. Lanmüller, U. Windberger*, S. Sauermann, M. Rab**, E. Unger, W. Girsch***

Department of Biomedical Engineering and Physics, University of Vienna
*Department of Biomedical Investigation, University of Vienna
**Department for Plastic and Reconstructive Surgery, University of Vienna,
*** Orthopaedic Hospital Speising, Vienna, Austria

SUMMARY

The Vienna SMV is designed to perform counterpulsation parallel to the descending aorta. The neo-ventricle is created by enlarging a segment of a thoracic aorta with pericardium, both from a donor, driven from the latissimus dorsi muscle (LDM). This configuration has been proofed in acute experiments and now in a ongoing chronic animal experiment.
In the chronic study the LDM is conditioned in situ before the SMV is constructed. At the beginning of the study the clinically approved stimulation protocol for cardiomyoplasty was used bilaterally in four sheep. Conditioning is completed after 12 weeks and results in a completely transformed muscle with slow-twitching type I fibres. In two of these sheep a special concept of muscle prefabrication was applied for the left LDM. The left LDM was detached from the thoracic wall, divided longitudinally and reattached in situ to achieve vascular delay. This concept of muscle prefabrication proved to be a failure. The histomorphological analysis showed a dramatic increase of perimysial and endomysial connective tissue at the end of conditioning.
A shorted protocol for conditioning in situ has been used up till now in three goats. Conditioning is started 1 week after electrode implantation with 30 contractions per minute. The number of pulses per contraction is incremented biweekly from 2 pulses up to 5 pulses. Conditioning is completed after 8 weeks. The histomorphological analysis of the first animal showed a completely transformed muscle with slow-twitching type I fibres.
At the beginning of the study an implantable nerve stimulator was applied in four adult sheep. This device was developed at our own department, powered from a Lithium Thionyl Chloride battery and hermetically sealed in a titanium case. To achieve a reduction in costs this device was modified into a semi-implantable version. In this version only the electrode leads are implanted chronically. Three sensing leads and four epineural stimulation leads are connected to a single skin button with an integrated connector. The external ECG triggered pulse generator is fixed on the animal by an elastic belt, its functionality is equal to the implantable version. Up to now the semi-implantable stimulator was used in 4 adult goats. An implantation period of 4, 7 and 6 month was reached without complications. An infection because of the percutaneous link appears in one goat after 7 month. Overall, this resolution turned out as a well useful and cost-saving device for animal experiments.

STATE OF THE ART

In the Vienna SMV project the skeletal muscle ventricle (SMV) consists of hemocompatible biomaterials only and is designed to perform counterpulsation parallel to the descending aorta. The neo-ventricle is created by enlarging a segment of a thoracic aorta with pericardium, both from a donor. The LDM is divided longitudinally and the two resulting branches are wrapped around the neo-ventricle in counterrotating direction. This concept was applied in a acute series in sheep /1/ and in a chronic animal experiment. The chronic study starts in 1997 and is still ongoing. This paper is focused on the protocol for preconditioning the LDM and the stimulation device of the chronic SMV study.

MATERIAL AND METHODS

Two different protocols had been used for preconditioning of the LDM. In 4 adult female sheep (group 1) the conditioning protocol was set according to the clinically approved stimulation protocol for cardiomyoplasty by Chachques et al /2/. In this protocol the number of pulses (1,2,3,5) in one burst is incremented biweekly, afterwards the number of bursts is incremented from 35 to 70 contractions per minute. Overall, conditioning is completed after 12 weeks.
The surgical procedure was performed under general anesthesia. After surgery the animals were kept in cages. The thoracodorsal nerve was exposed under careful preservation of the vascular pedicle emerging from the thoracodorsal artery. Four epineural electrodes were sutured to the epineurium of the nerve in helical manner. This procedure was done bilaterally and the electrode leads from the right and left side was connected to an implantable stimulator /3/.
The stimulator and the epineural electrodes were designed and manufactured at our Department. The nerve pacing leads are made from stainless-steel stranded wire (Eticon 612P, Ethicon Co., Norderstedt, Germany), coiled and embedded in silicon (Selastic, Dow Corning, S.A., France). The battery-powered pulse generator is hermetically sealed in a titanium case. Implant dimensions are 65 x 17 mm (diameter x height), the device weighs 88 g. This pulse generator can be used for activating two skeletal muscles via the motor nerves, using constant-current impulses with a maximum current of 4 mA at a pulse duration of 0,2 to 1 ms. Stimulation can be achieved by either single channel or multichannel methods (with up to 4 electrodes for each nerve), i.e. carousel stimulation and sequential stimulation. A stimulation burst can be activated unsynchronized or synchronized to the ECG. Additionally, the system allows a dynamic adaptation of the synchronization delay and the burst duration to the heart rate. Both parameters are calculated by an internal microcontroller. The current values are the sum of a constant and a percentage of the RR interval. The value of the constant, the percentage of the RR interval and all other parameters can be set by an external PC, which is used as a programmer unit.
In the first two sheep out of this series a special concept of muscle prefabrication was applied for the left LDM. The left LDM was detached from the thoracic wall, with all perforating vessels deriving from the intercostal vascular bundles being ligated, while the LDM insertion at the humeral bone was kept unaffected. After the intramuscular vascular architecture of the left LDM had been identified by means of translumination, the muscle was divided longitudinally from its distal end up to the entry of the neurovascular bundle in order to create two muscle branches of equal size. The LDM was reattached to the thoracic wall with absorbable sutures in original position.
A shorted protocol for conditioning in situ was applied in 3 adult female goats (group 2). Conditioning is started 1 week after electrode implantation with 30 contractions per minute. The number of pulses per contraction is incremented biweekly from 2 pulses up to 5 pulses. Conditioning is completed after 8 weeks. In this group only the left the thoracodorsal nerve was exposed and four epineural electrodes were sutured to the epineurium of the nerve. The left LDM was left untouched, without mobilization or splitting.

Table 1. Shorted protocol for conditioning (group 2).

Weeks	Burst type	Muscle contractions per minute
1	no stimulation	no stimulation
2-3	double pulses	30/min
4-5	triple pulses	30/min
6-7	four pulses	30/min
8-9	five pulses	30/min

Used stimulation parameters: frequency 30Hz, pulse duration 0.6ms

The leads terminate in one skin button with an integrated multipolar connector, which was placed in a subcutaneous pocket (see figure 1). The connector (Redel Lemo Group, Ecublens, Switzerland) is screwed into a disc with an axial flange covered entirely by a woven double velour patch (Meadox Medical, Ratingen, Germany) which provides an infection barrier. The electrode wires are crimped to the connector contacts and embedded in silicone rubber (Silastic, Dow Corning, S.A., France). Additionally, three temporary pacing leads (Medtronic, Inc., Minneapolis, MN, USA) embedded in a velour pouch are connected to the skin button. The pouch is opened by building the SMV and the electrodes are placed on the heart and used for ECG sensing. An external ECG triggered pulse generator was fixed on the animal by an elastic belt. The functionality of this semi implanted device is equal to the fully implantable version.

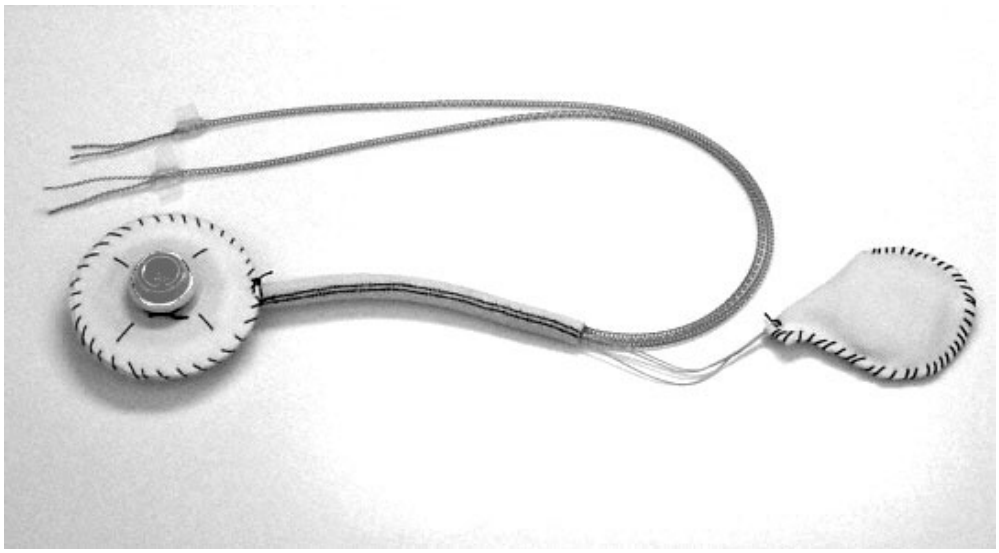


Figure 1. Electrodes used for conditioning (group 2) and ECG triggered stimulation of the SMV. Four epineural leads and three sensing leads are connected to an integrated connector in one skin button.

RESULTS

The histomorphological analysis was done for all sheep of group 1. After conditioning the LDM was completely transformed to slow twitching Type I fibers. The equivalent diameter of the Type I fibers was decreased and the percentage of the connective tissue was slightly increased. The fully implantable stimulation device performed without complications. The electrodes of one nerve had to be exchanged in one instance after they were dislocated.

The concept of muscle prefabrication, which was applied for the left LDM in the first two sheep out of this series proved to be a failure. A high increase in the perimysial-and endomysial connective tissue together with a complete loss of muscle fiber architecture was evident. Beside the high increase of the connective and fatty tissue the muscle fibers showed typical signs of degeneration or necrosis (for details see /4/).

The histomorphological analysis of the first animal in group 2 showed a completely transformed muscle of the left LDM with slow-twitching type I fibres. The percentage of the connective tissue was slightly increased. Further analysis had not be done up till now.

An infection due to the skin button of the electrode leads appears in one goat after a implantation time of 7 month. An implantation period of 4, 7 and 6 month was reached without complications.

DISCUSSION

Two clear massages can be drawn from the histomorphological results. The concept of muscle prefabrication caused a disastrous outcome. As a consequence, muscle splitting and mobilization followed by vascular delay and in situ conditioning as a concept of muscle prefabrication should be strictly avoided. The stimulation protocol applied in group 1 without mobilization of the muscle showed the expected outcome. The muscle was in a good condition and completely transformed to slow-twitching type I fibres.
The first histomorphological analysis of the shortened protocol (group 2) did not show the result we had hoped for. The protocol ends with a stimulation frequency of 2.5 Hz in average per day. Nevertheless, we could not found a relevant part of type 2A fibres, as it was found in rabbit by Jarvis JC et. al. /5/ by continuous stimulation at 2.5Hz.
The stimulation device, particularly the functionality turned out to be useful. The dynamic adaptation of the synchronization delay and the burst duration to the heart rate enables a fast and simple adjustment. This feature was very helpful during the chronic employment of the SMV, which was achieved in three goats for 4, 5 and 4 month up till now.
The semi implantable device turned out as a useful and cost-saving device for this animal experiment. The skin button seems to be resistant again an infection for an implant period of six month. However, this setup is not resistant again the animal itself, which leads to interruptions of the stimulation for some hours. The cable which was used to connect the skin button with the stimulation device was often found in the cages used as a chewing gum. By that reason and additionally to prolong the implant period we are now working on a full implantable and cost-saving device.

REFERENCES

/1/ Girsch W, Koller R, Lanmüller H, Rab M, Avanessian R, Schima H, Wolner E, Seitelberger R. Experimental development of an electrically stimulated biological skeletal muscle ventricle for chronic aortic counterpulsation . Eur J Cardio Thorac. JAN 1998; 13 (1) : 78-83
/2/ Chachques JC, Carpentier A. Postoperative Managment. In A Carpentier, JC Chachques, P Grandjean (eds.): Cardiomyoplasty. Mount Kisco, NY, Futura Publishing Co., Inc., 1991, pp 131-138
/3/ Lanmüller H., Sauermann S., Unger E., Schnetz G., Mayr W., Bijak M., Girsch W. Multifunctional implantable nerve stimulator for cardiac assistance by skeletal muscle. Artif-Organs Apr 1999; 23 (4): 352-359
/4/ Lanmüller H. ,Girsch W., Rab M., Sauermann S., Kamolz LP, Seitelberger R, Wolner E.: Preparation of a skeletal muscle ventricle in sheep: severe damage to the latissimus dorsi muscle due to mobilization before preconditioning. European Surgical Research 2000;32:129-134
/5/ Jarvis JC; Sutherland H, Mayne CN, Gilroy SJ, Salmons S. Induction of a fast-oxidative phenotype by chronic muscle stimulation: mechanical and biochemical studies. Am J Physiol. 1996; 270(1 Pt 1):C306-12.

AUTHOR'S ADDRESS

Hermann Lanmüller Ph.D.
Department of Biomedical Engineering and Physics, AKH 04L, Währinger Gürtel 18-20, A1090 Vienna, Austria,
e-mail: Hermannn.Lanmueller@univie.ac.at
home page: www.bmtp.akh-wien.ac.at

THE VIENNA SMV-PROJECT
FIRST RESULTS FROM CHRONIC EXPERIMENTS IN GOAT

W. Girsch, H. Lanmüller, M. Rab, LP. Kamolz, U. Windberger, H.Schima,
R. Avanessian, M. Greher, M. Ulrich, H. Beck, W. Haslik and R. Seitelberger
Medical School, University of Vienna, Austria

SUMMARY

A new type of skeletal muscle ventricle (SMV) constructed of biological materials only was designed as an aortic counterpulsation device: The SMV consists of an aortic homograft, enlarged to a pouch with pericardium and is anastomosed in parallel to the descending aorta. The latissimus dorsi muscle (LDM) is wrapped around the pouch. Stimulation electrodes are applicated to the thoracodorsal nerve and R-wave triggered burst stimulation is applied during cardiac diastole. In a series of acute experiments in sheep the hemodynamic efficacy was demonstrated and a standardised surgical procedure was established. A series of chronic experiments in goat was scheduled in order to evaluate the reliability and patency of this new type of SMV and its overall influence to the circulation.
Methods: In one goat an unconditioned LDM was used for SMV construction. In two goats the LDM was preconditioned by means of a newly developed fully implantable ECG-triggered multi-channel-stimulator and the SMV was constructed.
Results: The goats survived the procedure without any complications. The goats survived the procedure without any complications. The SMV was in circulation for four, five and six months in these first animals, stimulated chonically at a rate of 1:5 compared with the native heart rate. Reconstitution of the resected third rib caused compression of muscle and nerve in the first animal. Consecutive loss of muscle force terminated this experiment. Second and third goat were sacrificed as scheduled. In all three animals the SMV was found patent at the end of the experiment.
Conclusion: Goat turned out to be appropriate for this kind of experiments. The configuration produced distinct hemodynamic changes required for aortic counterpulsation. Several problems seen in the first and second experiment necessitated modifications of the surgical procedure. The observed results still are promising and the experiments will be continued.

INTRODUCTION

A new type of skeletal muscle ventricle (SMV) constructed of biological materials only was designed as an aortic counterpulsation device: The SMV consists of an aortic homograft, enlarged to a pouch with pericardium and is anastomosed in parallel to the descending aorta. The latissimus dorsi muscle (LDM) is wrapped around the pouch (fig. 1). Stimulation electrodes are applicated to the thoracodorsal nerve and R-wave triggered burst stimulation is applied during cardiac diastole. In a series of acute experiments in sheep the hemodynamic efficacy was demonstrated and a standardised surgical procedure was established [1]. A series of chronic experiments in goat was scheduled in order to evaluate the reliability and patency of this new type of SMV and its overall influence to the circulation.

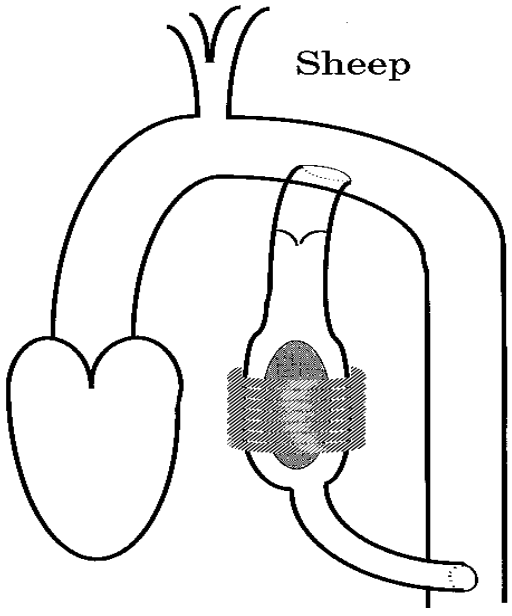


Fig. 1: Scheme of the SMV model

MATERIALS AND METHODS

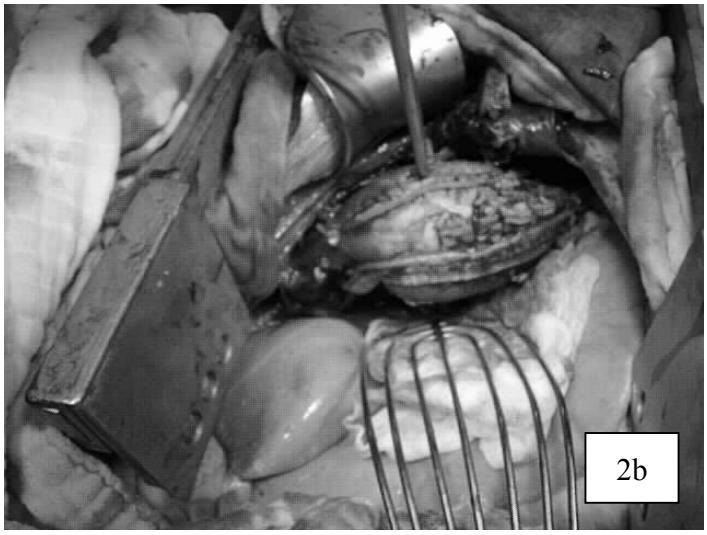
3 female adult alpine goats, weighing about 50kg, underwent the experimental procedure. In one goat an unconditioned LDM was used for SMV construction. In one goat an unconditioned LDM was used for SMV construction. In two goats the LDM was preconditioned by means of a newly developed fully implantable ECG-triggered multi-channel-stimulator and a new stimulation protocol. 4 ring-shaped stimulation electrodes were applied to the epineurium of the thoracodorsal nerve while the LDM was left untouched [2,3,4]. During 8 weeks the LDM was conditioned to perform 30 contractions per minute fatigue-free around the clock (the protocol is presented in detail by H. Lanmüller)

Prior to the main surgical procedure the pericardium and the entire thoracic aorta had been excised from fresh goat cadavers. These ”homografts” were cryopreserved according to approved techniques [5,6].

The SMV construction was performed as a standardized procedure using a left side lateral flank incision. The serratus muscle was detached from the thoracic wall. A segment of the third rib was resected subperiostally and the fifth and sixth rib were removed and preserved as pedicled vascularized grafts. Enlarging the circumference of the homograft with strips of homologous pericardium created a neoventricle of about 50ml. This biological conduit was anastomosed in parallel to the descending aorta (fig 2a, b). The LDM was wrapped around the ne oventricle, applying near physiologic resting tension to the muscle and the free end of the muscle was fixed to the sixth rib. Two ECG-sensing electrodes were fixed directly to the heart and the electrode leads were connected with the implanted stimulation unit. Finally the thorax was reconstructed by means of the rib grafts and the mobilized serratus muscle.

At the end of the procedure functional electrical stimulation (FES) was started. R-wave triggered burst stimulation at a rate of 1:5 with the native heart rate and limited to maximum 30 contractions per minute was applied during cardiac diastole to simulate aortic counterpulsation.

Fig. 2: Intraoperative situs
- after completion of proximal anastomosis (2a)
- and connected to the circulation on both ends (2b)



RESULTS

The goats survived the procedure without any complications. The SMV was in circulation for four, five and six months in these first animals, stimulated chronically at a rate of 1:5 compared with the native heart rate.
Activation of the SMV during diastole augmented the diastole to systolic pressure levels. Ultrasound investigation revealed a reduction of the SMV-diameter of about 30% during LDM contraction, causing arterial back-flow in the most proximal part of the aorta.

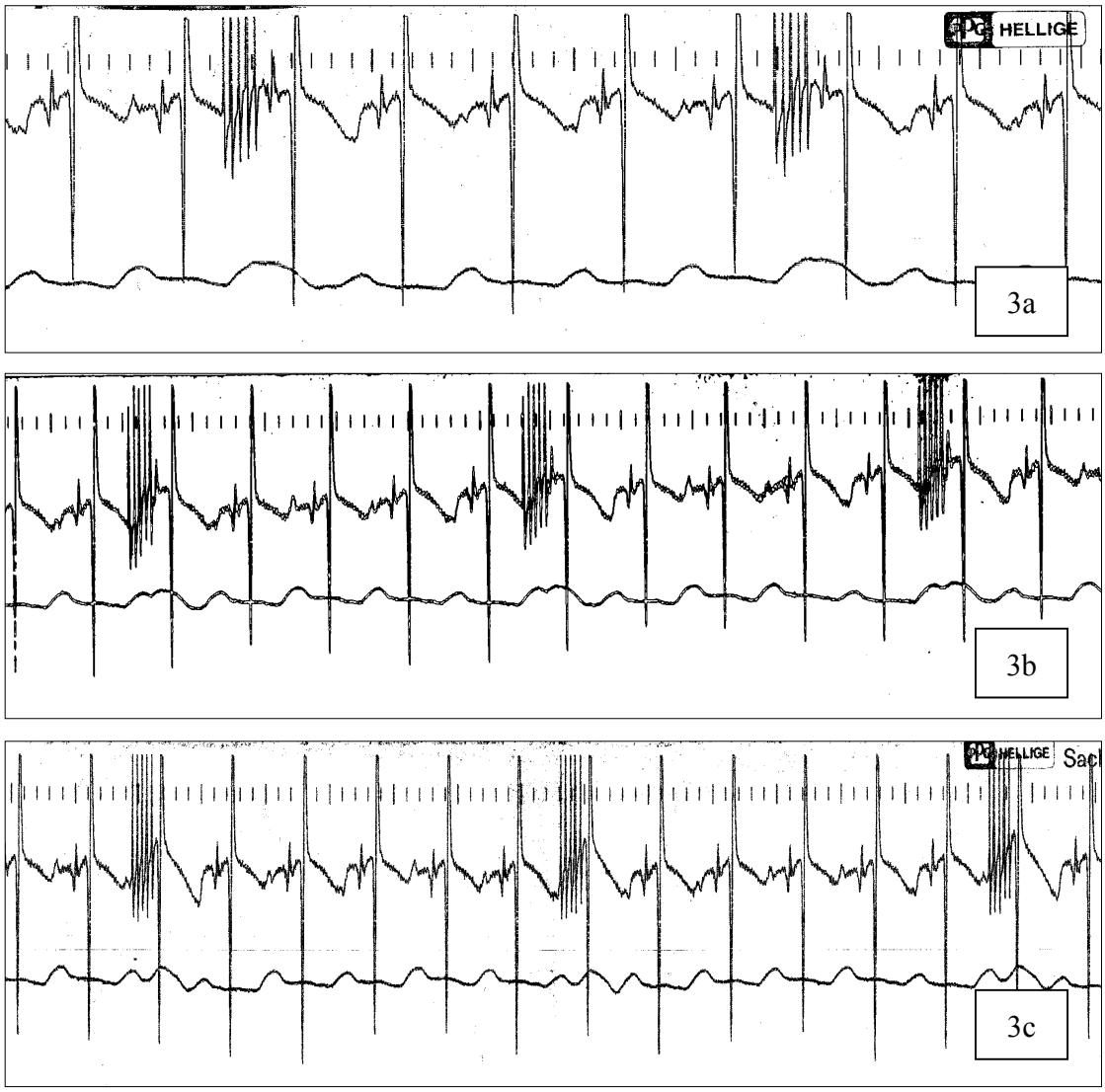


Fig. 3: Hemodynamic measurements (arterial pressure curve derived from the brachiocephaliv trunk) during ECG-triggered activation of the SMV (ECG curve interrupted by FES-pulse trains) at different R-R intervals (3a -35%, 3b - 45%, 3c -60%)

Reconstitution of the resected third rib caused compression of muscle and nerve in the first animal. Consecutive loss of muscle force terminated this experiment. Second and third goat were sacrificed as scheduled. In all three animals the SMV was found patent at the end of the experiment.

CONCLUSIONS

Goat turned out to be appropriate for this kind of experiments. The configuration produced distinct hemodynamic changes required for aortic counterpulsation. Several problems seen in these first experiments necessitated modifications of the surgical procedure. The observed results still are promising and the experiments will be continued.

REFERENCES

1. Girsch, W. Koller, R. Lanmüller, H. Rab, M. Avanessian, R. Schima, H. Wolner, E. Seitelberger, R. Experimental development of an electrically stimulated biological skeletal muscle ventricle for chronic aortic counterpulsation. Eur J Cardiothorac Surg 13(1):78-83, 1998.
2. Koller, R. Girsch, W. Liegl, Ch. Gruber, H. Holle, J. Losert, U. Mayr, W,. Thoma, H. Long term results of nervous tissue alterations caused by epineurial electrode application: An experimental study in rat sciatic nerve. PACE 15(1):108-115, 1992.
3. Mayr, W. Bijak, M. Girsch, W. Holle, J. Lanmüller, H. Thoma, H. Zrunek, M. Multichannel stimulation of phrenic nerves by epineurial electrodes: Clinical experience and future developements. ASAIO J 39(3):729-735, 1993.
4. Lanmüller, H. Sauermann, S. Unger, E. Schnetz, G. Mayr, W. Bijak, M. Girsch, W. Multifunctional implantable nerve stimulator for cardiac assistance by skeletal muscle. Artif Organs 23(4):352-359, 1999.
5. Grabenwöger, M. Grimm, M. Eybl, E. Moritz, A. Müller, MM. Bock, P. Wolner, E. Endothelial cell lining of bioprosthetic heart valve material. J Card Surg 7(1):79-84, 1992.
6. Leukauf, C. Szeles, C. Salaymeh, L. Grimm, M. Grabenwöger, M. Losert, U. Moritz, A. Wolner, E. In vitro and in vivo endothelialization of glutaraldehyde treated bovine pericardium. J Heart Valve Dis 2(2):230-235, 1993.

AUTHOR´S ADDRESS

Univ.Doz.Dr. Werner Girsch
Orthopädisches Spital Speising
Speisingerstr. 109
A-1134 Vienna, Austria
Tel +43 1 80182 509
Fax +43 1 80182 285
eMail werner.girsch@univie.ac.at

DEVELOPMENT OF MUSCULAR BLOODPUMPS, PERFORMED IN A ONE-STEP OPERATION

N. W. GULDNER[1], P. KLAPPROTH[1]M. GROßHERR[2], H.-H. SIEVERS[1]
Klinik für Herzchirurgie[1], Klinik für Anästhesiologie[2]Medizinische Universität zu Lübeck,
D-23538 Lübeck Germany

SUMMARY

Up to now skeletal muscle ventricles(SMVs) have required a 2-step procedure i.e. the construction, followed by a vascular delay and electrical conditioning and the integration into circulation by a second operation. As shown previously clenbuterol increased power of electrical conditioned SMVs wrapped around a mock system (1,2) as shown in Fig.1

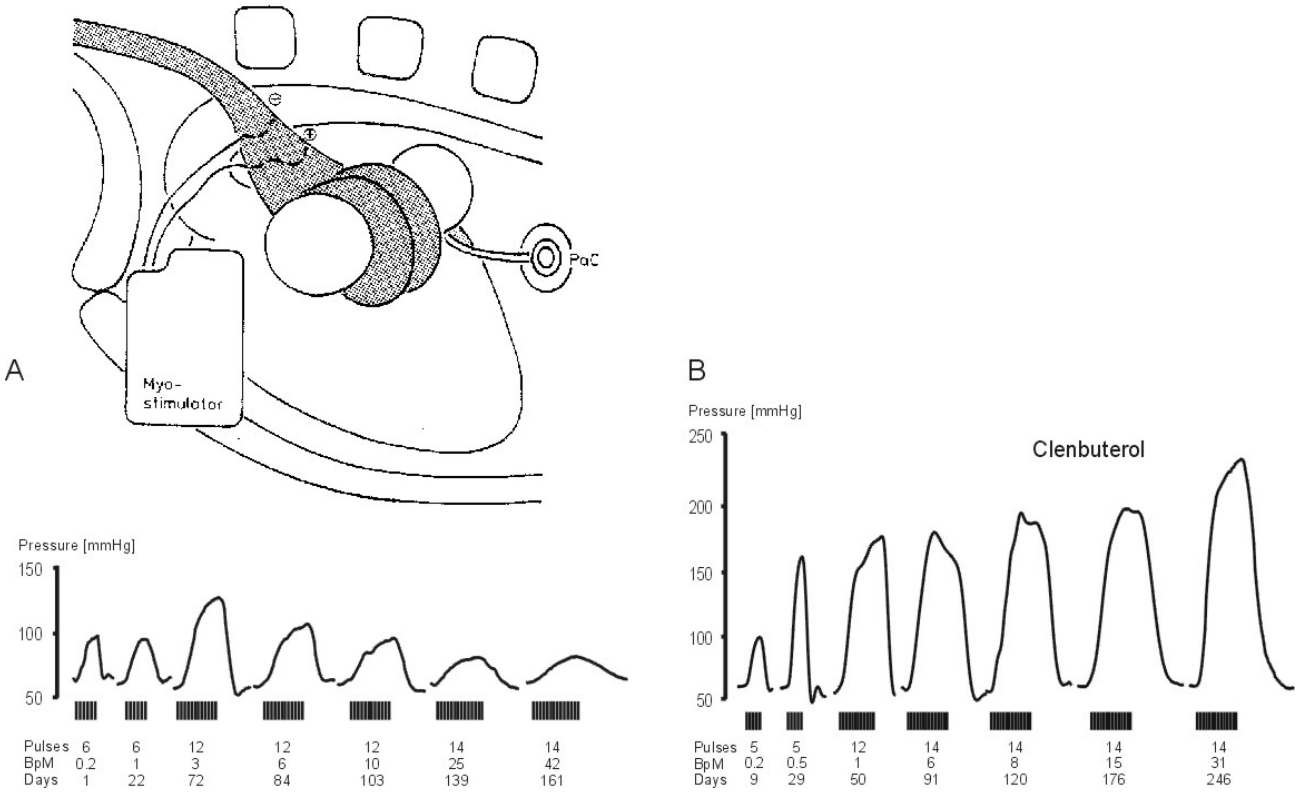


Fig. 1: Top: A SMV around an intrathoracic elastic training device. A muscular contraction induces a pressure increase within the cavity of the elastic device. Bottom: Original pressure curves during a dynamic training from a SMV of group I without β-2-stimulation (A) and group II supported by Clenbuterol (B). Clenbuterol supported SMVs of group II maintained pressure (function) at a high level over time. Stimulation pattern is shown with an increasing number of pulses per burst..

They pumped successfully from construction to several months against a pressure of 60-70mmHg (3).
Thus a muscular blood pump was performed in a one-step procedure and trained within the circulation under support of clenbuterol. It showed to be hemodynamically relevant and is expected to become clinic practicable for the treatment of end-stage heart failure.

STATE OF THE ART

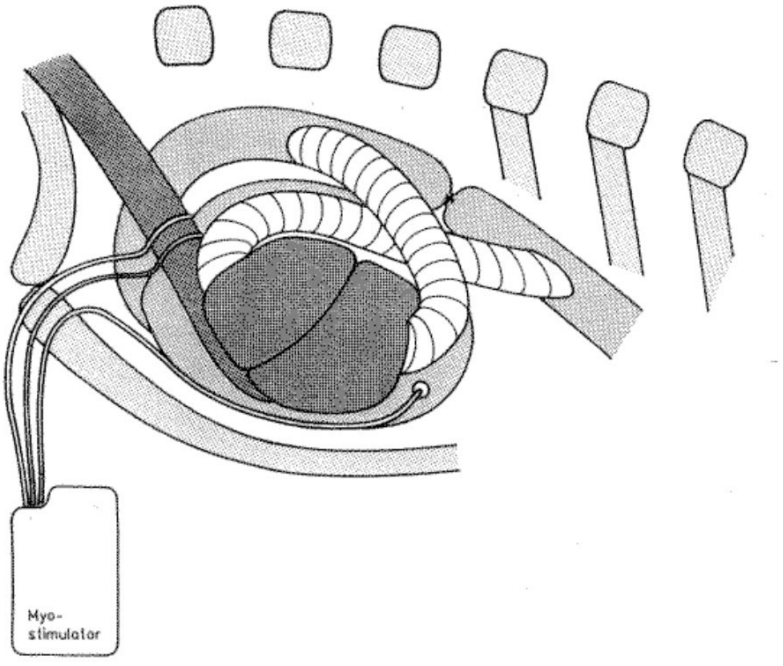
The prevalence of heart failure is increasing in industrial countries and evolving a major problem of health care in the next decades. To meet this challenge intensified research on the disease and especially its treatment alternatives is mandatory since the standard therapeutical option of heart transplantation is limited. Muscle powered cardiac assistance might become a real alternative to heart transplantation.

Desirable properties of skeletal muscle for cardiac assistance include fatigue resistance and powerful mechanical performance. Current clinical application of non fatigable muscle was limited due to the profound power loss after electrical conditioning. In long-term investigations, skeletal muscle ventricles (SMVs) were performed around an elastic intrathoracic training device (1-4). SMVs were treated over several month by a threefold approach in order to achieve non fatigable and powerful muscle pumps by a combination of electrical conditioning (4), dynamic training against systemic load, and pharmacological support with clenbuterol. SMVs treated in that way became powerful, as demonstrated elsewhere (3).

These SMVs were expected to become effective as muscular blood pumps performed in a one-step operation and trained within circulation. This autologous muscular blood pump with a stabilizing inlay, performed in a one-step operation and trained within the circulation, is defined as a Biomechanical Heart (BMH).

MATERIAL and METHODS

The operative procedure was performed under general anaesthesia in adult 5 Boer goats of a mean weight of 79±6.6 kg. Left latissimus dorsi muscle was dissected free and Myoelectrodes were placed wavelike around the branches of nervus thoracodorsalis. The threshold of the muscle was determined by electrical stimulation. From a double layered muscular tube of latissimus dorsi muscle, a skeletal muscle ventricle was placed around a silicone-polyurethane inlay and inserted into the thorax after partial resection of the 3rd and 4th rib. Through an inferior thoracotomy, the vascular prostheses from the ventricular inlay of the BMH were anastomosed with the descending aorta end-to-side (Fig.2)



2).

Fig. 2: Topography of a biomechanical heart in aorta-descendens position in a goat. The aorta is ligated between the two anastomoses

The aorta was ligated totally between the two anastomoses. A myocardial sensing electrode was placed and connected with a myostimulator to trigger muscle contractions in the sense of counterpulsation. Pump function of the BMH after several weeks of a dynamic training was demonstrated by arterial blood pressure, ventriculography and by means of a conductance catheter resulting in pressure-volume loops (5) (Fig. 3).

RESULTS

Intra-operatively, mean stroke volume of BMHs was 53.8±22.4ml. One month post-operatively in peripherial arterial pressure, mean-diastolic(P_{MD}) and minimal diastolic pressure(P_{min}) of BMH supported heart cycles differed significantly from non supported ones as described elsewhere. One BMH, catheterized 132 days postoperatively, shifted 34.8ml per beat and 1.4L/min using a LDM of 330g (Fig.3).

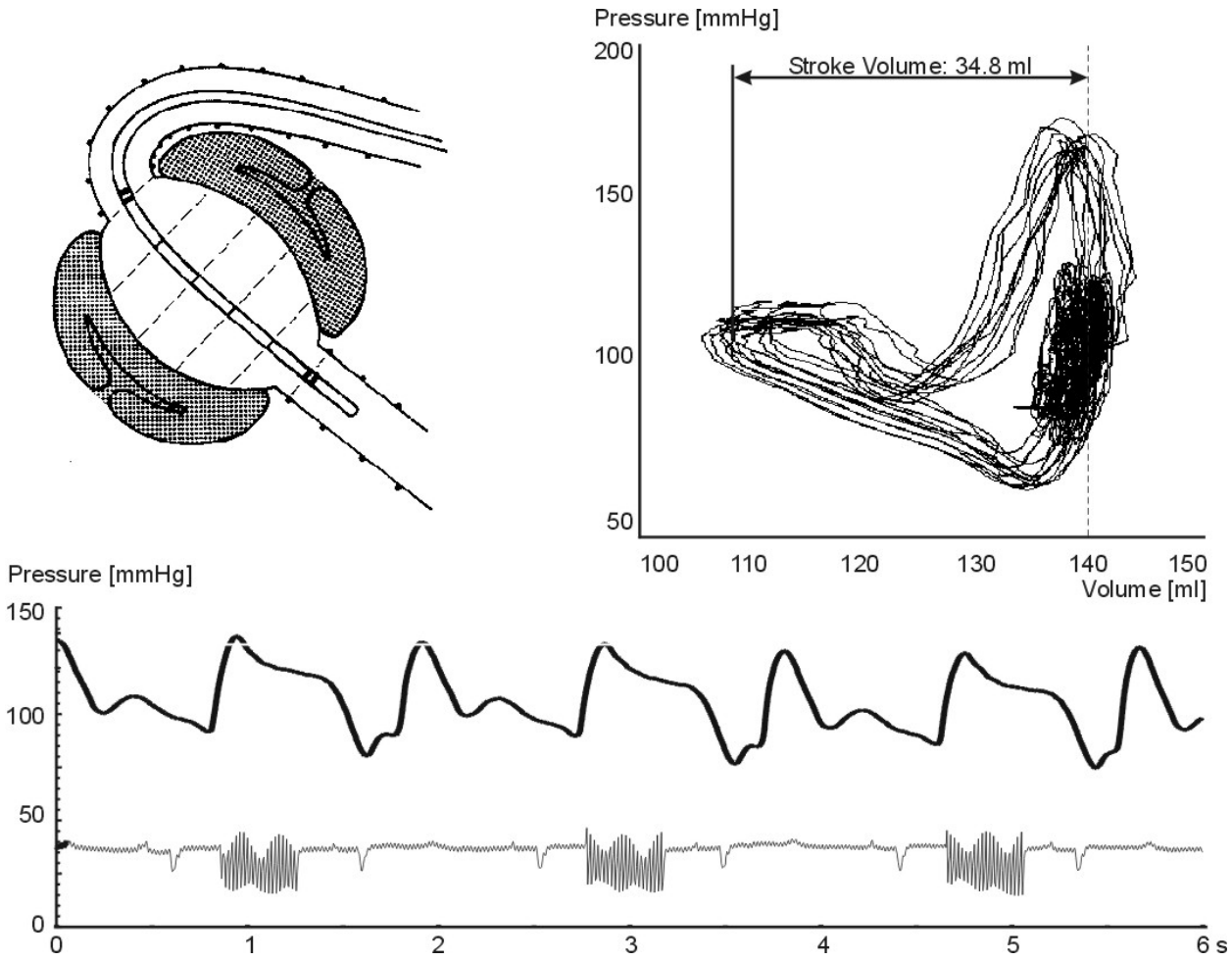


Fig. 3: Volumetry within a BMH by the conductance catheter method (top). ECG with stimulation bursts and pressure trace from a peripherial artery with BMH contractions in 1:2 mode (bottom):

CONCLUSION

Under support of clenbuterol, Biomechanical Hearts of a clinical relevant size can be trained effectively in systemic circulation after a one-step-operation and offer the prospect of a sufficient volume shift and probably unloading of the left ventricle.

REFERENCES

1. GULDNER NW, EICHSTAEDT HC, KLAPPROTH P, TILMANS MHJ, THUAUDET S, UMBRAIN V, RUCK K, WYFFELS E, BRUYLAND M, SIGMUND M, MESSMER BJ, BARDOS P. Dynamic training of skeletal muscle ventricles. A method to increase muscular power for cardiac assistance . Circulation 89 (3):1032-1040 (1994)

2. KLAPPROTH P, GULDNER NW, SIEVERS HH. Stroke volume validation and energy evaluation for the dynamic training of skeletal muscle ventricles. Int J Artif Organs 20:313-321 (1997)

3. GULDNER NW, KLAPPROTH P, GROßHERR M., RUMPEL E, NOEL R, SIEVERS HH. Clenbuterol supported Dynamic Training of Skeletal Muscle Ventricles Against Systemic Load- A Key for Powerful Circulatory Assist ? Circulation 101:2213-2219 (2000)

4. GULDNER NW, KLAPPROTH P, FISCHER T, RUMPEL E, BÜCHNER I, KELLER R, KLEMPIEN R, KRISCHER H, THUAUDET S, NOEL R, KUPPE H, SIEVERS HH . Functionally adapted stimulation patterns for a dynamic training of skeletal muscle ventricles in adult goats . BAM;8(1):67-72 (1997)

5. GULDNER NW, KLAPPROTH P, GROßHERR M., BRÜGGE A; SHEIKHZADEH A; TÖLG R; RUMPEL E, NOEL R, SIEVERS HH Biomechanical Hearts, Muscular Blood Pumps, Performed in a 1-Step Operation, and Trained Under Support of Clenbuterol Circulation 104 (2001)

.

Priv. Doz. Dr. med. Norbert W. Guldner, Klinik für Herzchirurgie,
Medizinische Universität zu Lübeck, Ratzeburger Allee 160,
D-23538 Lübeck, Germany.
Tel.: 0049-451-500-2108 Fax: 0049-451-500-6035
e-mail: guldner@medinf.mu-luebeck.de

A COMBINED STIMULATION AND TELEMETRIC MONITORING SYSTEM: CURRENT DEVELOPMENT FOR BIOMECHANICAL HEARTS

Klapproth P[1], Guldner NW[1], Otten J[2], Maczinowski P[2],
Nestler B[2], Ryschka M[2], Büter T[2], Schirmer R[2]

(1) Clinic of Cardiac Surgery, Medical University of Lübeck, Germany
(2) University of Applied Sciences, Lübeck, Germany

SUMMARY

Myostimulators were used to evoke muscle contraction synchronously or asynchronously in order to assist failing heart. Many interesting features of these implanted devices were dispersed and might be combined into a new stimulation system. Additionally, our approach includes a fully independent measurement unit for electrical and non-electrical biosignals (ECG, pressure). Via a radio-telemetric link, new and user adapted stimulation protocols as well as measured data could transferred bi-directionally.

STATE OF THE ART

Skeletal muscle is clinically and experimentally used to assist failing heart in several manners. Clinical applications are the cardio- and aortomyoplasty, experimentally the skeletal muscle ventricles and the Biomechanical Hearts. Independently of the type of application, skeletal muscle contractions must be evoked synchronously to subjects heart beat. Therefore, an ECG-triggered myostimulator has to be used, which delivers an electrical pulse train via stimulation electrodes to the muscle respectively to the muscle's nerve. Common myostimulator systems allow via a nearfield telemetry the adjustment of the electrical stimulation pattern deliver to skeletal muscle. Thus, the stimulation parameter could be adapted to a user defined stimulation protocol, varying the pulse amplitude, the pulse width, the number of pulses per burst, the interpulse interval (constant, incremental, decremental), the number of bursts per minute respectively heart beat in R-wave - triggered mode, the R-wave-distance of the burst, etc. Furthermore, actual devices allow /1/ a multichannel stimulation, a dynamic adaptation of the synchronization delay and the burst duration to the heart rate, an optimized pulse train definition (N-lets), /2/ a work-rest regime and a day - and night setting with time-dependent cardiac assist ratios.
Although the current stimulation systems allow a precise setting of the stimulation parameter, some of the interesting features are dispersed over several devices and might be combined in one. Additionally some new features for experimental and later clinical purpose are desirable.

Beside the standard stimulation techniques, a new myostimulator might be improved, if it is able to:

- adjust within a burst the pulse amplitude, the pulse width and within the pulse its polarity (biphasic pulse).
- deliver a burst with free defined interpulse distance between each pulse in order to achieve the "doublet" or "N-let" effect. This effect was described in /3, 4 and others/ as a method to increase muscular power by initiate an electrical stimulus with an double, triple, etc. pulse sequence with smaller interpulse distance than the following pulses.
- provide a low to very low constant frequency prestimulation (below 1Hz) over weeks with work-and rest regime. In /5, 6/ a prestimulation frequency of 2.5 Hz is described as useful to improve blood supply in the distal region of the latissimus dorsi muscle in sheep.
- assist the cardiac cycle with an assist ratio of 1 muscle contraction per 255 heart beats up to a 1:1 mode, rather than a common assist mode of 1:20 up to 1:1.
- allow a work-rest regime and a days and night rhythm with a reduced assist mode.
- synchronize muscle's contraction into the "late distolic" phase of the heart beat, in order to reduce the loading of the following heart beat in a counterpulsator setup /7/. This might be performed by a

"backward" definable R-wave delay of the stimulation burst and a dynamic, heart rate dependent burst duration adaptation.
- stimulate two skeletal muscle independently.

Additionally to the stimulation task, a new to design device might be used to register the effect of muscle's contraction on circulation. Thus, the effect of different stimulation patterns but also the effect of a drug administration might become valuable /7, 8/. Therefore, the myostimulator must be extended with the ability of measuring, storing, transmitting and evaluating electrical and non-electrical biosignals as the ECG but also the blood pressure. In /9/, a new implantable MEMS based pressure sensor was introduced which might become implanted f.e. into the left atrium, the aorta or the pulmonary artery, if long term stability of the sensor is shown.

MATERIAL AND METHODS

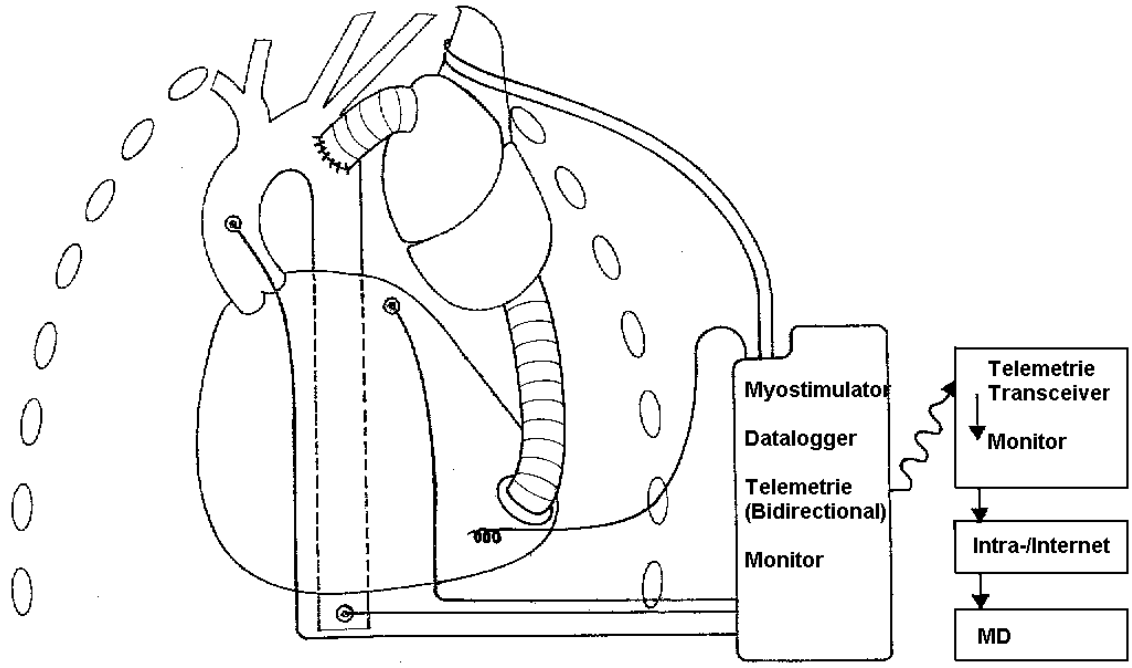


Fig. 1: Schema of a Biomechanical Heart in apico-aortal position. The new device needs a cardiac sensing electrode and two muscle electrodes. Additionally, three pressure sensors might become integrated f.e. into the left atrium, the aorto or the pulmonary artery.

Our design consists of two independent working microcontrollers (COP8, National Semiconductor), which could be programmed in the language C. Main features of these devices are the 32 k of flash memory, the In System Programmability (ISP), the low power consumption with power saving and idle modes, the three timers/counters and a powerful buildin serial communication (microwire and RS232). For muscular as well as cardiac stimulation, there is an 8 channel, 8 bit digital-analog converter (AD8803, Analog Devices) with following current amplifier. Additionally, the system incorporates an 8 channel, 12 bit analog-digital converter (AD 7888, Analog Devices) and a programmable logic circuit (XCR3032, Xilinx), which cascades the pressure sensors. For communication purpose, a bi-directional telemetric unit is integrated in order to transfer data from the device and to the device.

The microcontroller 1 (μC1) is the myostimulator controller. Its main work is to evaluate the current heart rate (sensed via a cardiac sensing electrode) and to decide, which heart rate dependent stimulation patterns should be used. The stimulation patterns contain the information, when and how an electrical burst should be delivered f.e. which mode should be used (R-wave-triggered or untriggered mode), after what time delay should the electrical burst occurs respectively what is the cycle time of an untriggered

burst, and how many single pulses form the burst. Via the 8 channel, 8-bit digital-analog converter and a following current amplification, the predefined burst is stimulate the muscle and if necessary the heart.
The µC2 is the measurement controller. Its main task is to cyclic control via the telemetric system, if a message f.e. request of measurement, reprogramming of a burst definition table, a new software for one of the controller, occurs from an outer system. If a measure request is received, the device turns into the measuring mode. Otherwise it goes back to the sleep mode, which is cyclic interrupted via the idle timer.
In the measuring mode, the device generates a clock of 125 kHz in order to drive the pressure sensors. On each logic event of one of the three sensors cascaded via the logic circuit , an interrupt is generated. At the interrupt handle, the complete register of 8 bit containing all relevant information (which sensor do what event) is stored. Additional the time of the event as well as up to 4 analog digital converter channels are read. All data are stored in a buffer. Via the telemetric link, the data within the buffer are send over the serial line to an outer system. After a predefined amount of time, the device leaves the measuring mode.

RESULTS

The myostimulator unit (µC1) can generate impulse pattern corresponding to the burst definition table under respect of a R-wave event. In general, there are 4 muscle stimulation leads and 2 cardiac stimulation leads. Each lead could be defined as active or ground. The following table demonstrates the electrical output of the device:

Table I: Features of the myostimulator unit

Pulse
Amplitude: 0 - 12 V (in steps of 0.05 V)
Polarity: biphasic
Width: 170µs, 240µs + n x 2.5µs ($1 < n < 65535$) (= $Width_{min}$: 170µs, $Width_{max}$ = 164 ms)
Interpulse distance: 700 µs + n x 2.5µs ($1 < n < 65535$) ($Distance_{min}$: 0.7 ms; $Distance_{max}$: 164 ms)

Burst up to 20 pulses each pulse with its own amplitude, polarity, width, interpulse distance

Stimulation modes
Synchronized: R-wave triggered, 1:1 - 1:255 muscle:cardiac contraction
Heart rate boundaries : 6 between 50 - 150 bpm
R-Wave delay: absolute -> 50 ms+n x 10.24 ms ($Delay_{min}$: 50ms; $Delay_{max}$ >1200 ms)
relative (RR-Interval), positive , negative if sinudial rhythms
Dynamic burst duration: not yet

Untriggered Cycle Time = Burst duration + n x 10.24 ms ($1 < n < 65535$)
(Cycle Time $_{min}$: Burst duration + 10 ms; Cycle Time $_{max}$: 669 s)

Prestimulation Untriggered, single pulse < 1Hz
Work-Rest Mode (work cycle: 1s - 254d; rest cycle: 1s - 254d)

Day - Night Rhythm controlled via real time clock

Electrodes Muscle: 4 x Stimulation, Heart: 2 x Stimulation, 2 x Sensing

The µC2 controls the measurement unit. It works independently from the myostimulator unit (µC1). The main features of this unit are:

- cyclic telemetric check, if a measurement request is ask from the outer system.
- In measuring mode: acquire up to 3 pressure sensor signals, digitise the ECG
- determine the resistances of each single electrode refer to electrical ground

- establish a serial data transfer (up to 19.2 kBaud) via the telemetry circuit
- store long term data like heart rate, minimal and maximal pressure

DISCUSSION

On of the major problems of implanted active devices are the energy management. In our approach, we decide to use standard 3 lithium batteries à 3.6V (1.1 Wh). With that energy source, the theoretical durability of the device was calculated to be one year. Using patient safe batteries, the durability must be recalculated. Up to now, this system is an experimental design. It must be shown in animal experiments, if the functionality and the reliability is given. Especially the housing of the device as well as the coating of the pressure sensors must be revised.

REFERENCES

/1/ Lanmüller H, Sauermann S, Unger E, Schnetz G, Mayr W, Bijak M, Girsch W. Multifunctional Implantable Nerve Stimulator for Cardiac Assistance by Skeletal Muscle. Artif. Organs 1999; 23 (4):352-359

/2/ Chekanov V, Chachques JC, Brum F, Arzuga J, Arzuaga P, Krum DP, Hare JW, Maternowski MA, Tchekanov GV, Fiandra O, Hammond R, Melamed V, Chiu RCJ, Stephenson LW. LD-Pace II: A New Cardiomyostimulator for Cardiac Assist. ASAIO Journal 2001; 47:50-55

/3/ Wakabayashi T, Kuroda T. Response of crayfish muscle preparations to nerve stimulation with varoius patterns of impulse sequence. Tohoku J Exp Med 1977; 121(3):207-18

/4/ Kebaetse MB, Lee SC, Binder-Macleod SA. A novel stimulation pattern improves performance during repetetive dynamic contractions. Muscle Nerve 2001; 24(6):744-52

/5/ Tang AT, Jarvis JC, Hooper TL, Salmons S. Observation and basis of improved blood flow to the distal latissimus dorsi muscle: a case for electrical stimulation prior to grafting. Cardiovasc Res 1998;40(1):131-7

/6/ Tang AT, Jarvis JC, Hooper TL, Salmons S . Cardiomyoplasty: the benefits of electrical prestimulation of the latissimus dorsi muscle in situ. Ann Thorac Surg 1999; 68(1):46-51

/7/ Guldner NW, Klapproth P, Großherr M, Brügge A, Sheikhzadeh A, Tölg R, Rumpel E, Noel R, Sievers HH. Biomechanical Hearts - Muscular Blood Pumps, performed in a 1-Step Operation, and Trained Under Support of Clenbuterol. Circulation 2001; 104

/8/ Guldner NW, Klapproth P, Großherr M, Stephan M, Rumpel E, Noel R, Sievers HH. Clenbuterol Supported Dynamic Training of Skeletal Muscle Ventricles Against Systemic Load - A Key for Powerful Circulatory Assist ? Circulation 2000;101:2213-2219

/9/ Eggers T, Binder J, Marschner C, Laur R. Application of Miniaturized (Wireless) MEMS Based Hybrid Pressure Sensors in Blood Pumps. 1st Congress of Cardiac Bioassist Association 2001; Abstract

AUTHOR'S ADDRESS

Dr. Peter Klapproth
Clinic of Cardiac Surgery, Research Laboratory
Medical University of Lübeck, Ratzeburger Allee 160, D-23538 Lübeck
e-mail: klapprot@medinf.mu-luebeck.de

STIMULATION OF NERVES INNERVATING THE DOG'S PANCREAS

J. Rozman, [1]B. Zorko, [2]M. Bunc, [3] M. Žitko

ITIS d. o. o. Ljubljana, Centre for Implantable Technology and Sensors, Lepi pot 11, University of Ljubljana, [1]Veterinary Faculty, Gerbičeva 60, [2] School of Medicine, Institute of Pathophysiology, Zaloška 4, [3]University Medical Centre Ljubljana, Department for Nuclear Medicinea, Zaloška 7, 1000 Ljubljana, Republic of Slovenia

SUMMARY

Our aim was to modulate secretion of insulin and glucagon into the blood of a healthy and diabetic dog with stimulation of nerves innervating the pancreas. The 33-electrode spiral cuffs were implanted in an adult Beagle canine. The first cuff was installed on the vagus nerve, the second one on the splanchnic nerve and the last one on the pancreatic nerve. To cause insulin-dependent Type I diabetes, partial dysfunction of the pancreas, was induced. Nerves were stimulated using biphasic, rectangular current pulses (10mA, 200μs, 20Hz). Samples from the femoral artery were drawn before the experiment, after 5min and 5min after the stimulation stopped. Results of radioimmunological assay (RIA) of blood samples showed that, in intact pancreas, stimulation of the vagus nerve caused a considerable increase in insulin secretion, not significant change in glucagon secretion. and decrease in C-peptide secretion. Splanchnic nerve stimulation did not cause considerable change in insulin and C-peptide secretion while considerable increse in glucagon secretion was noticed. Pancreatic nerve stimulation did not change considerably the secretion in any of the three hormones. In dysfunctioned pancreas, vagal nerve stimulation caused an increase in insulin and glucagon secretion and minor increase in C-peptide secretion. Splanchnic nerve stimulation caused a minor decrease in insulin secretion, a considerable increase in glucagon secretion. and a small increase in C-peptide secretion. Pancreatic nerve stimulation did not cause considerable change in insulin secretion while minor increase in glucagon and C-peptide secretion was observed.

STATE OF THE ART

In the field of research considering the possibilities in the application of functional electrical stimulation (FES) of the autonomic nervous system, interest is relatively large. This is evident from the few actually very good but not numerous publications in literature (4, 5). Even less has been done in research involving the selective stimulation of the autonomic nerves as they are, for instance, sympathetic and parasympathetic nerves innervating the pancreas. Since the discovery of insulin in the early twenties (6), research has been concentrated on elucidation of the mechanisms controlling the secretion of hormones by the pancreas (1, 2, 8, 9). Most recent reviews have dealt with the myriad of chemical factors known to affect these hormones, but few have dealt with the great deal of possibilities enabled by the FES of the autonomic peripheral nerves.

MATERIALS AND METHODS

A cuff was made by bonding two 0.1mm thick silicone sheets together. One sheet stretched and fixed in that position was covered by a layer of adhesive. A second unstretched sheet was placed on the adhesive and the composite was compressed. When released, the composite curled into a spiral tube. 33 rectangular platinum electrodes (0.6mm X 1.5mm) connected to the lead wires were mounted on the third silicone

sheet. They were arranged in 3 parallel groups each containing 11 electrodes at a distance of 0.5mm. The distance between the spiral groups was 6mm. Accordingly, 11 groups of 3 electrodes in the same line in a longitudinal direction were formed. All electrodes of the central and 2 outer groups were then connected to the lead wires. The sheet with electrodes was then bonded on the inner side of the cuff. The length of the cuff was optimized so that the surface of the nerve covered by the cuff would be as small as possible as shown in Fig. 1.

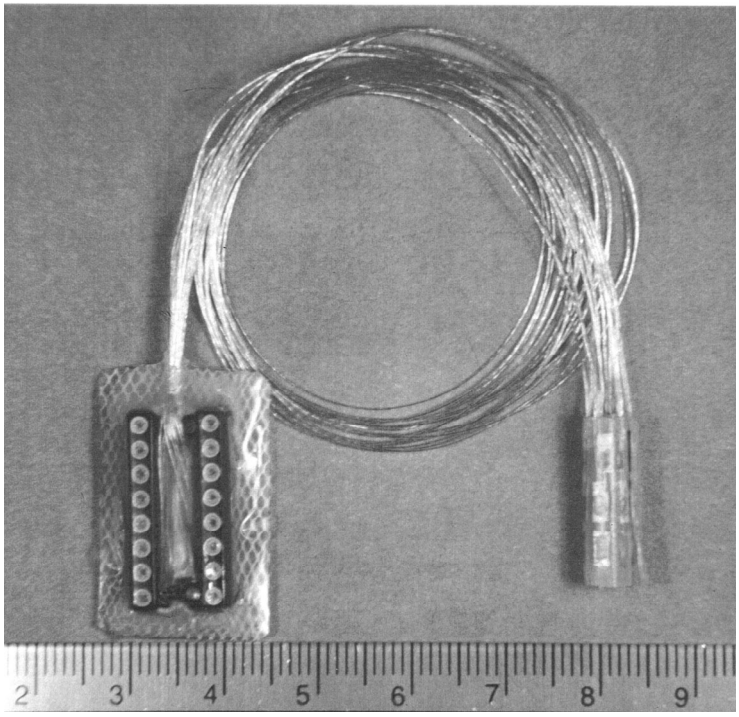


Fig. 1. 33-electrode spiral cuff for stimulation of nerves innervating the pancreas of a dog.

Lead wires were connected to the common connector to be implanted within the lateral subcutaneous tissue for the time between experiments. Gas-sterilized cuffs were implanted according to the protocol approved by a local ethics committee. The animal was premedicated with medetomidine 40μg/kg i.m. (Domitor, Orion Corp.) and methadone 0.2mg/kg s.c. (Heptanon, Pliva). Induction was performed with propofol 1.0 to 2.0mg/kg i.v. (Diprivan, Zeneca Pharamaceuticals Ltd.). General anaesthesia was maintained with isoflurane 0.8 to 1.5 vol.% (Forane, Abbott) in 100% O_2 (7). Analgesia during surgery was sustained with ketamine 0.5 to 2.0mg/kg i.v. (Ketamine, Veyx-Pharma GmbH) when necessary. Antibiotics (cefazolin 20mg/kg i.v.; Cefamezin, Krka) were administered perioperatively. Analgesia during the early recovery period was provided with methadone 0.3 to 0.5mg/kg s.c. TID. Tramadol 8.0mg/kg s.c. TID (Tramal, Grünenthal GmbH) was administered for an additional two days. To allow the devices to stabilize, the first experiment was performed 30 days after the implantation. The first cuff was installed on the vagus nerve at the neck as shown in upper part of the Fig. 2. In the splanchnic nerve the cuff was installed on the nerve before the celiac ganglion as shown in the lower part of Fig. 2. In the pancreatic nerve the cuff was installed on the nerve at the site before it enters the pancreas as also shown in the lower part of Fig. 2. To incompletely dysfunction a pancreas, thus causing Type I diabetes, the death of a certain portion of islet β-cells in the pancreas was induced (3). An Alloxan (Sigma Chemical Co., St. Louis, Mo.) was dissolved in a physiological solution (0.9% NaCl) (200mg/ml). A freshly prepared dissolution (50mg/kg) was then intravenously injected into the blood. After 24 hours the pancreas was irreversibly, incompletely dysfunctioned and permanent hyperglycemia was induced. Since

the protocols for curing naturally or experimentally induced Type I diabetes are the same insulin (Homofan 100, Pliva) therapy was applied (1IE/kg).

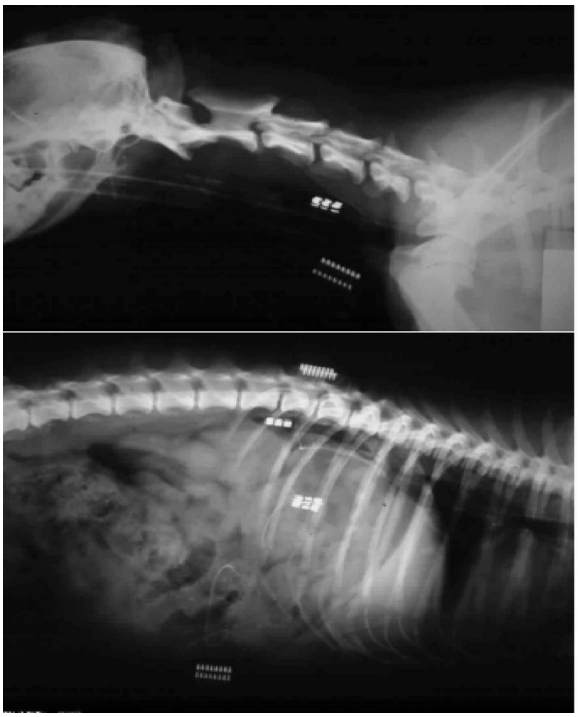


Fig. 2. (Upper part) X-ray of implanted 33-electrode spiral cuff showing it's position on the vagus nerve of a dog; (lower part) X-ray of implanted 33-electrode spiral cuffs showing their position on the splanchnic and pancreatic nerves of a dog.

To stabilize the blood glucose at a level approximately four times greater than normal a time period of 7 days prior to the stimulation was introduced. Blood samples from the femoral artery were drawn each time before experimental activity to measure the basal level of glucagon, insulin and C-peptide in the blood. They were drawn also at the beginning, after 5min, and 5min after the stimulation stopped.

RESULTS

Results of RIA confirmed the hypothesis that, in intact pancreas, stimulation of the vagus nerve caused a considerable increase in insulin secretion (from 13 to almost 110μunits/ml). However, vagal nerve stimulation did not cause significant changes in glucagon secretion (from 106 to 125pg/ml). Vagal nerve stimulation caused also some decrease in C-peptide secretion (from 0.74 to 0.34ng/ml). Result also showed that splanchnic nerve stimulation did not cause considerable change in insulin secretion (from 17 to 19μunits/ml) while considerable increse in glucagon secretion was noticed (from 125 to 175pg/ml). Splanchnic nerve stimulation did not cause considerable change in C-peptide secretion (from 0.48 to 0.49ng/ml). Pancreatic nerve stimulation did not change considerably the secretion in any of the three hormones. In severed pancreas, vagal nerve stimulation caused an increase in insulin secretion (from 12.6 to 28μunits/ml). Moreover, vagal nerve stimulation also increased glucagon secretion (from 119.3 to 130pg/ml). Besides, vagal nerve stimulation caused minor increase in C-peptide secretion (from 0.55 to 0.58ng/l). Splanchnic nerve stimulation caused a minor decrease in insulin secretion (from 15.8 to

13.3µunits/ml) and a considerable increase in glucagon secretion was observed (from 74.4 to 133.1 pg/ml). Splanchnic nerve stimulation caused a small increase in C-peptide secretion (from 0.29 to 0.68ng/ml). Pancreatic nerve stimulation did not cause considerable change in insulin secretion (from 6.7 to 6.9µunits/ml). while minor increase in glucagon and C-peptide secretion was observed (from 90.9 to 96.1pg/ml for glucagon and from 0.49 to 0.57ng/ml for C-peptide).

DISCUSSION

The results could be used in various animal and human basic studies concerning neurophysiology of endocrine glands and internal organs and their relation to bodily changes and disease. Ultimately, the method of FES of peripheral autonomic nerves with multi-electrode spiral cuffs could be used for both stimulation and recording in different combinations. A methodology as well as developed accompanying technological solutions could be used in the transfer of this model to the human alternative model of curing diabetes mellitus. Future studies will focus on chronic, selective stimulation of different superficial regions of nerves innervating the intact and damaged pancreas of a dog. Therefore, the long-range goal of our research will be to understand how the various branches of the utonomic nervous system regulate pancreatic endocrine and exocrine secretion.

REFERENCES

1) Ahrén B. "Autonomic regulation of islet hormone secretion--implications for health and disease", Diabetologia 2000 Apr;43(4):393-410.
2) Ahrén B. "Regulation of insulin secretion by nerves and neuropeptides," Ann Acad Med Singapore 1999 Jan;28(1):99-104
3) Ahrén B, Sundkvist G. "Long-term effects of alloxan in mice", Int J Pancreatol 1995 Apr;17(2):197-201.
4) B. Ahrén and G. J. Taborsky, “The mechanism of vagal nerve stimulation of Glucagon and Insulin secretion in the dog”, Endocrinology, vol. 118, No. 4, pp. 1551-1557, 1986.
5) Ahrén B, Veith RC, Taborsky GJ Jr. "Sympathetic nerve stimulation versus pancreatic norepinephrine infusion in the dog: 1). Effects on basal release of insulin and glucagon", Endocrinology 1987 Jul;121(1):323-31
6) F. G. Banting and S. Gairns, “Factors influencing the production of insulin”, Am. J. Physiol. vol. 68, pp. 24-39, 1924.
7) Havel PJ, Paquette TL, Taborsky GJ Jr. "Halothane is less suppressive than pentobarbital on reflex and neural activation of pancreatic F-cell", Am. J. Physiol. 1986 Jul;251(1 Pt 1):E111-6.
8) M. W. Roy, K. C. Lee, M. S. Jones and R. E. Miller, “Neural control of pancreatic Insulin and somatostain secretion”, Endocrinology, vol. 115, No.2, pp. 770-775, 1984.
9) S. C. Woods and D. Porte, Jr., "Neural control of the endocrine pancreas", Physiological Reviews, vol. 54, No. 3, pp. 596-619, 1987.

ACKNOWKEDGEMENT-This work was financed by Research Grants: J2-0542 from the Ministry of Education, Science and Sport, Ljubljana, Republic of Slovenia and HPRN-CT-2000-00030 from the European Commission.

AUTHOR'S ADDRESS

Dr. Janez Rozman, Centre for Implantable Technology and Sensors, Lepi pot 11, 1000 Ljubljana, Republic of Slovena

THE VIENNA PHRENIC PACEMAKER, LONGTERM DATA OF FAILURES

H. Thoma*, P.Darrington**, E.Unger*

* University Vienna, Austria, Department of Biomedical Engineering and Physics
** Börgel Inc., Limburg, Germany

INTRODUCTION

The Vienna Phrenic Pacemaker System consists of extracorporal (control unit, RF transmission coil) and implanted components (electronic 8-channel implant with stimulation electrodes/1/). To avoid fatigue of the diaphragm muscle because of continous electrostimulation – 24 h per day – the so-called "roundabout-stimulation" was developed and patented /2/. The implant is powered and controlled by RF-impulse transmission from the control unit. The control unit is powered by rechargable batteries. This development was used 1982/83 for stimulation of lower extremities in 4 paraplegic patients for standing up and walking using forearm crutches. The same system was implanted in the first quadroplegic patient (level: C1/2) with paralyzed breathing in 1983, but new software for stimulation of inspiration cycle was used. Since 1980, the development of this system has not undergone relevant changes despite of some technological improvements – e.g. total metal sealing of all electronic components in 1993 /3/. Therefore it is of interest to do an analisis of relevant failures and to develop an update control unit in order to avoid such failures in future.

In the following we will focus on a statistic of failures of the electronic implant and the control unit. We have nearly no failures in electrodes, RF coil and 220 V power supply. Otherwise batteries and cables were exchanged routinely about once a year.

METHODS

For this paper we collected material from different sources. Patient's data and implant failures were found in former publications. In addition, we could prove this data by auditing accounts of Medimplant Inc. Vienna – producer of the pacemaker system. The yearly check of the patients´ pacemaker is an order by law. Therefore we got excellent data concerning the functions of the pacemaker via the patient's medical record keeping device. This documentation is done by Börgel Inc., Limburg, German representative of the Vienna phrenic pacemaker. A new control unit was developed this year.

In the following section detailed results of patients, the lifetime of implants and specific failures of the control unit as well as the features of the new development are presented.

RESULTS

1. The implant

An overview of our phrenic pacemaker patients is summarized in fig. 1. Despite of these 26 registered patients one male patient was operated on in 1982 and one 6 year old boy with a lesion in the level C 0 was not registrated /4/.

Reimplantation was necessary 18 times. In 3 patients reimplantation was necessary 3 or 4 times, in 17 patients there was no reimplantation. In 4 patients lifetime of the implant is between 9 and 11 years. The main reason of the frequent early failures was due to the implant technology: some electronic parts were sealed only in Hysol. Implants after 1993 were totally wrapped in metal housing, the receiver RF coil protected by a ceramic ring. Up to now, there are not any failures with

this updated implants. As our implants have connectors to the 8 electrodes, is the change quite similar to the cardiac pacemaker.

In 1990 we started a development of a 20 channel implant encapsulated by niobium. This implants were tested in animal studies (calfs) for a period of cumulative 4,5 years in Vienna and Ljubljana. No failures were observed during these tests.

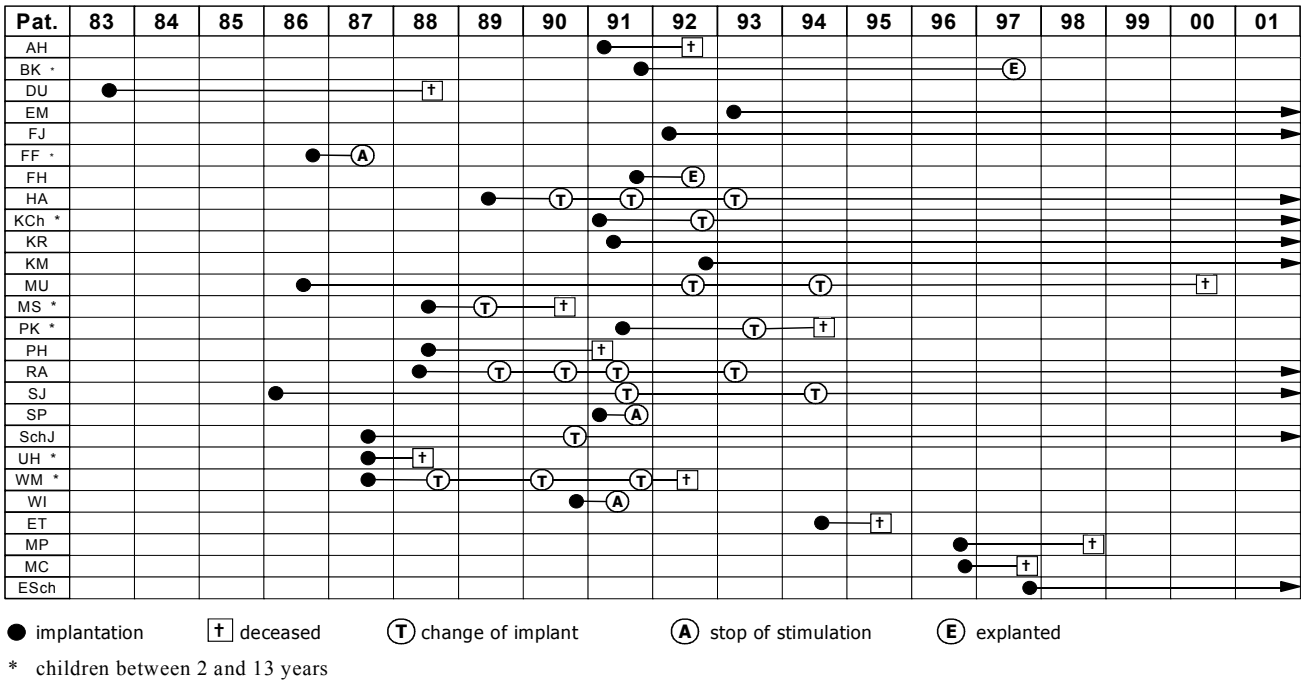


Fig 1: patient overview, starting 1983 up to 2001

2. The control unit T154S

Specific failures of the battery driven control unit T154S were analysed in 5 patients, lasting for cumulative 60 patient years. In fig. 2 patients were listed according to "years after implantation" (7 up to 15 years). In the process of time 3 different groups may be discussed. During the first 5 years few failures were only due to mechanic components. The most frequent failures were registered during the period of the years 6, 7 and 8. The remaining years 9 to 15 have to be discussed: on the one side the number of evaluated patients is going down, on the other, the yearly failure rate decreases unproportionally to the number of remaining patients, e.g. no registered failures in the year 11 after implantation. We may get additional informations calculating the failure rate of each patient:

	years	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	total
	number of patients	5	5	5	5	5	5	5	4	4	4	4	3	3	2	1	
mechanic	housing			I		II		II	I						I		7
	cable connectors			I		I	I	I	II	I	I			I	I		10
	cable break			I			I	I	II	I				I	I		8
	plug connection break						II			I	I			I			5
hardware	display																0
	switch print			I			I	II	I	I	I				I		8
	power print					I	II	I					I		I		6
	main print						I	I							I		3
software	eprom update			I			I		II								4
	microprocessor						I		I								2
	total	0	0	5	0	4	10	8	9	4	3	0	1	3	6	0	53

Fig 2: 5 patients with phrenic pacemaker, two control units per patient, statistic of failures of the control unit T154S, years after implantation

No.	Patient's code	Failure	Years
1	MU	3	15
2	RA	21	14
3	HA	7	13
4	KC	14	11
5	BK	8	7

Finally we may analyse the mechanic stress. According to fig. 2 30 mechanic failures have been registered. This number increases to 38 if we add the failures of the switch print, which are usually caused by mechanic stress. This calculation and the different failure rate of our patients (3 – 21) lead to the conclusion, that the activity and mobility of our patients – handling of the control unit – influences the failure rate essentially.

3. The control unit T161S

The control unit T161S (fig. 3) is an update of the 20 year old T154S unit. According to the main results of long term failure analysis, the 161S is much better protected against mechanical stress. The housing is milled from solid aluminium and additionally protected by two aluminium plates. For the front plate foil keys are used to avoid a breaking of conventional switches which were used in the T154S unit. The plugs for cables of the RF coil and the charger were sunked in the milled aluminium housing (see fig. 3) for optimal protection against mechanic stress. The electronic is manufactured in Surface Mounted Technology which is more stable compared to the technology used in the T154S.
The 161S control unit is smaller, of less weight and has twice battery capacity, compared to the 20 year old T154S development.

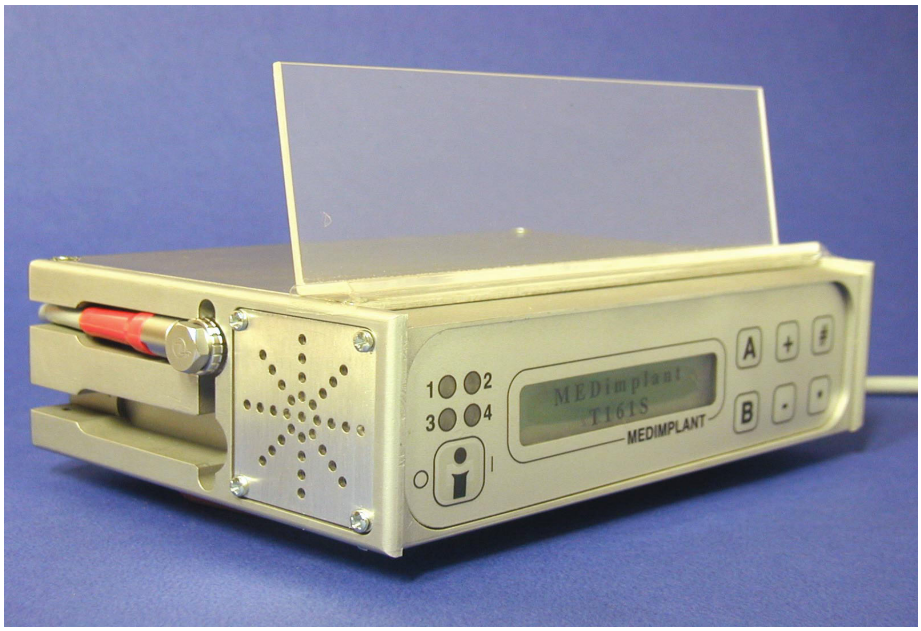


Fig. 3: The new phrenic pacemaker control unit T161S

FINAL STATEMENT

It is unusual that a working group publishes their own failures in such details. But in 1993 we launched a much better implant and put on the market and we developed a much better control unit for our patients. We hope that in future other groups will also publish their failure analysis as we did.

The application of a phrenic pacemaker is internationally seen very seldom. But with good rehabilitation and continous service we increase our patients´ quality of life enormously. Patients with closed tracheostoma living at home equipped with our very stable epineural multielectrodes have good chances for a long life time.

REFERENCES

1. H.Thoma, H.Gerner, J.Holle, P.Kluger, W.Mayr, B.Meister, G.Schwanda, H.Stöhr: The Phrenic Pacemaker: Substitution of paralysed functions in tetraplegia. ASAIO 1987, New York, 33rd Annual Meeting, Proceedings.
2. H.Thoma: Verfahren und Vorrichtung zur Langzeitstimulation von Nerven und Muskeln. Österr. Patent 330 342, 1975
3. W.Mayr, M.Bijak, W.Girsch, J.Holle, H.Lanmüller, H.Thoma, M.Zrunek:
   Multichannel stimulation of phrenic nerves by epineural electrodes.
   ASAIO Journal, 93, Vol.39, No.3, p.729-735
4. W.Girsch, R.Koller, J.Holle, M.Bijak, H.Lanmüller, W.Mayr, H.Thoma: Vienna Phrenic Pacemaker – experience with diaphragm pacing in children
   European Journal of Pediatric Surgery 6 (1996), p.140-143, ISSN 0939-7248, Hippokrates Verlag Stuttgart

Authors´s address
Univ.Prof.Dr.Herwig Thoma, Department of Biomedical Engineering and Physics, University of Vienna, General Hospital, Waehringer Guertel 18-20, A 1090 Vienna, Austria.Phone +43-1-40400-1986, fax +43-1-40400-3988

Session 6

COMMAND AND FEEDBACK SIGNALS, STIMULATION PARAMETERS

FATIGUE PROCESS OF TYPE I AND TYPE II MUSCLE FIBERS

K. Kerševan, V. Valenčič and N. Knez

Laboratory of Biomedical Visualization and Muscle Biomechanics
University of Ljubljana, Faculty of Electrical Engineering, Slovenia

SUMMARY

Skeletal muscles' fatigue process was examined in muscles biceps brachii and tibialis anterior in 9 male subjects, aged 21 – 24 years. Measurements were performed under isometric conditions. Pulse trains (0.1 ms) of different stimulation frequencies and voltage (40–50 V above threshold) were used for 10 s bipolar electrical stimulation. The non-invasive tensiomyographic (TMG) method was used for detection of muscle responses to electrical stimulation. Using pulse trains of different frequencies enabled separate observation of fatigue and tetanization process for both muscle fibre types:

- Type I (fatigue resistant) muscle fibres exerted and retained force throughout the 10 s electrical stimulation. Fatigue process was observed at stimulation frequency of 2 Hz but the type I muscle fibre fatigue did not occur due to their fatigue resistant character. Complete tetanus of type I muscle fibres was noticed at stimulation frequencies of 4 – 6 Hz .
- Fatigue process of type II (fast fatigue) muscle fibres ended before the electrical stimulation was over and was noticed at stimulation frequencies of 7.5 – 9 Hz. Complete tetanus of type II muscle fibres was noticed at stimulation frequencies higher than 18 Hz.

Separate observation of type II muscle fibres enables more efficient treatment and observation of pathological changes in dystrophic, atrophic and denervated patients and also helps professional athletes and their trainers to improve their training technique.

STATE OF THE ART

The diversity of skeletal muscles, which is reflected by the heterogeneity and spatial arrangement of their individual fibres, enables numerous movements of different velocities, forces and endurances, as well as adaptation to altered demands of usage. Both clinical and sport physiologists have always been interested in studying muscle adaptation process as a result of either pathological changes or targeted training process.

Type II muscle fibre fatigue process was observed in unfused tetanus. When a series of stimuli were given, each stimulus elicited a single twitch response so that a series of twitch responses was produced. As the stimulating frequency increased, the twitches began to sum and the response profile changed from unfused to fused tetanus (Fig. 1). For this purpose two muscles were selected with respect to percentage of type I muscle fibres they comprise: muscle biceps brachii

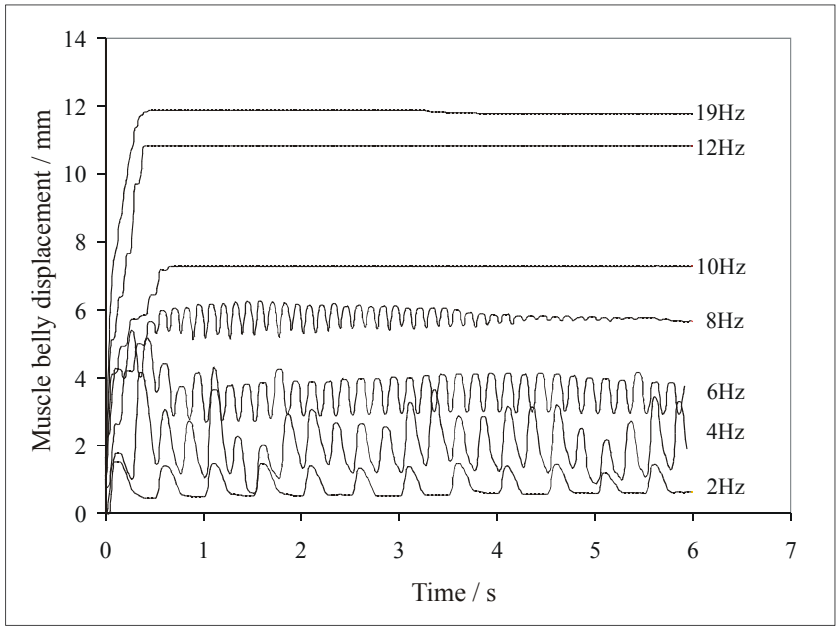


Fig. 1: Twitch summation and tetanization in human biceps brachii

(BB): approx. 52 % and muscle tibialis anterior (TA): approx. 77 % /1/; according to dissimilar percentage of type I muscle fibres we expected dissimilarities in type II muscle fibre fatigue process as well.

MATERIAL AND METHODS

TMG recordings

In order to record unfused tetanus, tensiomyographic (TMG) measuring method was used. This is a non-invasive, selective method for detection of skeletal muscles' contractile properties /2/. In previous studies it has been established that the radial displacement of muscle belly is proportional to muscle force /3/ as well as to percentage of type I muscle fibers /1/.

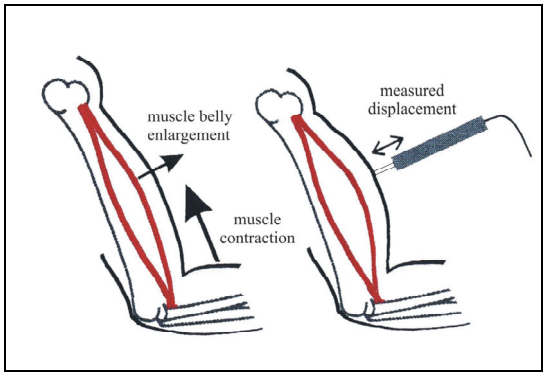


Fig. 2: Principle of TMG measuring method: when muscle contracts, its belly enlarges. Radial enlargements can be measured by displacement sensor.

TMG method is based on muscle contracting principle under isometric conditions: when the muscle is contracted it increases the force between insertions and its middle part – muscle belly – is thickened. Displacement of a muscle belly is measured with the displacement sensor, positioned radial to the skin above the observed muscle (Fig. 2).

For the presented study, the measured subject was sitting on a measuring chair with his measured arm/leg fastened to the frame with one or two bands in order to achieve isometric condition. Measurements were performed with an inductive sensor incorporating a spring of 0.17 N/mm, which provides an initial pressure of approximately 1.5 x 10^{-2} N/mm^{-2} on a tip area of 113 mm^2.

Measuring point for each muscle was determined anatomically on the basis of the anatomic guide for electromyographers /4/ – BB (right side): midpoint of the line between lateral head of calvicula and head of radius; TA (left side): four fingerbreadths below tibial tuberosity and one fingerbreadth lateral to tibial crest.

Each muscle was stimulated with pulse trains of 0.1 ms duration, stimulation frequencies ranging from 5–25 Hz and voltage ranging from 40–50 V above threshold. For 10 s stimulation, two self-adhesive electrodes were used, placed symmetrically to the sensor: the anode was placed distally and the cathode proximally, 20–50 mm from the measuring point. The stimulator used was a Grass 8800 stimulator, with voltage output through an insulation unit. The measured muscle responses were stored and analysed using a PC.

Data analysis

In unfused tetanus, the dynamics of type II fibre fatigue process (oscillating part of TMG response) was observed (Fig. 3) and described with the following parameters: τ – fatigue process time constant, t_F – fatigue process duration, f_F – frequency, at which type II fibres fatigue process was observed,

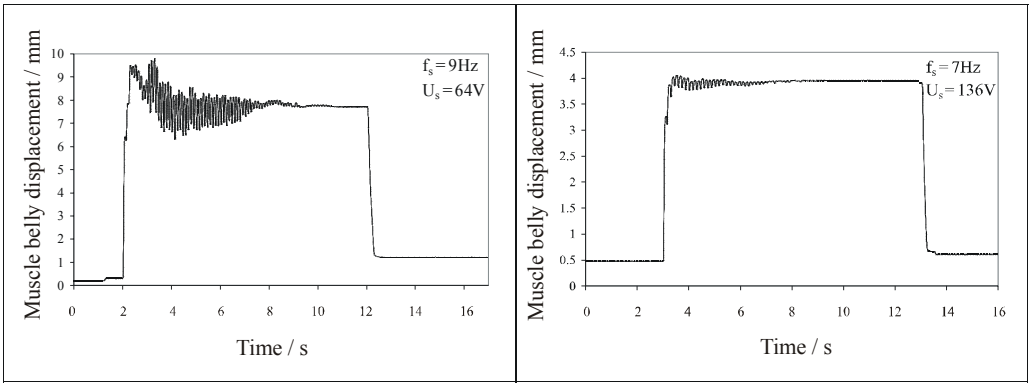


Fig. 3: Time course of muscle response to pulse train in human biceps brachii (left) and tibialis anterior (right)

f_T – fused tetanus frequency and d_{PPmax} – maximum value of peak to peak oscillation. According to Thomas, Johansson & Bigland-Ritchie (1991) /5/, fast-twitch motor units exhibit a sag profile in the unfused tetanus. In the presented study, the sag profile was eliminated from data analysis (Fig. 4).
Before the data was analysed, the oscillating part of TMG response was rectified and course of amplitudes was fitted by exponent curve (Fig. 5):

$$y = a \cdot e^{-\frac{t}{\tau}} + c$$

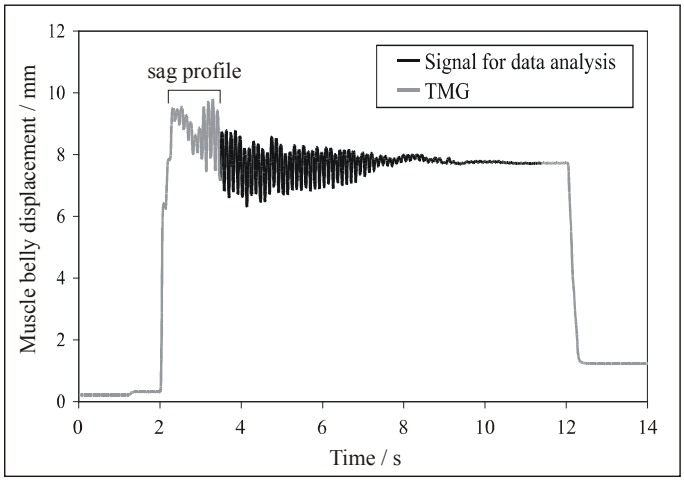


Fig. 4: The sag profile was eliminated from data analysis

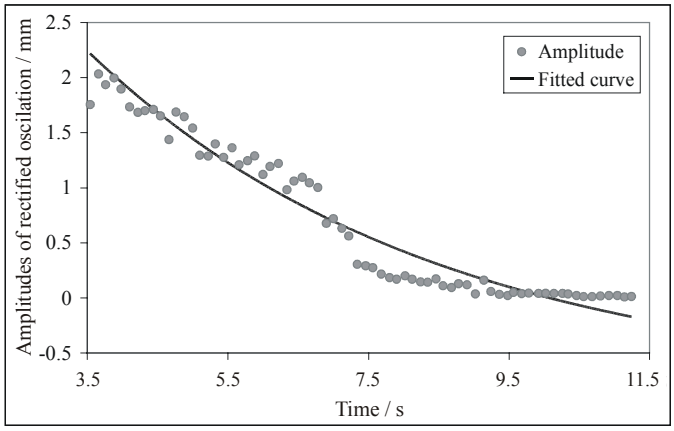


Fig. 5: Time course of amplitudes in rectified oscillation

RESULTS

The dynamics of type II fibre fatigue process in human BB and TA was observed and fatigue process parameters were compared (Table 1):

Muscle	$\bar{\tau}$	$\bar{t}_F$	$\overline{f}_F$	$\overline{f}_T$	$\overline{d}_{PP\max}$
Biceps brachii (right side)	3.94 s	8.1 s	8.6 Hz	19.6 Hz	1.9 mm
Tibialis anterior (left side)	1.02 s	6.4 s	7.7 Hz	18.3 Hz	0.2 mm

Table 1:Aaverage values of type II fibres fatigue process parameters for two different human skeletal muscles

DISCUSSION

The main objective of this study was to find out whether the TMG method is suitable for monitoring the unfused tetanus and whether it provides any information on skeletal muscles' structural or functional changes. According to both muscles' sizes and to the percentage of type I muscle fibres comprised in them, the above listed results were expected: (1) the greater the amount of type II muscle fibres (in BB), the longer the duration and the greater the time constant of the fatigue process; (2) in TA, both frequency-dependent phenomena occurred at lower stimulation frequencies – again, the reason might be the percentage of type II fibres. In case of a different protocol (e.g. supra-maximum stimulation and fixed stimulation frequency), with respect to their diameter, maximum value of peak-to-peak oscillation could be used as an indicator for the type II fibres quantity (Fig. 5) – however, the physiological properties of the measured muscle have to be taken into account (muscle's volume, fascia's thickness, etc.). Type I muscle fibres fatigue was not noticed during the 10 s stimulation.

This procedure (supra-maximum stimulation (amplitude), variable stimulating frequency) has already been evaluated in *clinical environment* with patients after poliomyelitis in muscle rectus femoris. When compared to EMG output, TMG

Muscle	$\bar{t}_F$	$\overline{f}_F$
Rectus femoris (after poliomyelitis)	35.5 s	8.3 Hz
Rectus fempris (healthy subjects)	73.3s	12.3 Hz
Statistical significance (P)	<0.05	0.02

Table 2: Comparison of average values of type II fibres fatigue process parameters with patients after poliomyelitis and control group

method provided useful data on muscle dysfunction as well (Table 2). Apart from that, TMG method is easy to apply, the same set of equipment is suitable for measurements of all surface skeletal muscles and measuring results are available immediately after the measurement.

In the *sport field*, the described procedure (supra-maximum stimulation, fixed stimulating frequency) is used to monitor the adaptation of the motor system to specific training process. These adaptations can be extensive and have been shown to affect most aspects of the system, both morphological and functional. TMG method's selectivity enables observation of a single muscle within a given muscle group (Fig. 6). In Fig. 6 the effect of specific training process on muscle vastus lateralis is presented: initial muscular status (upper) and muscular status after specific 10 day training process (lower).

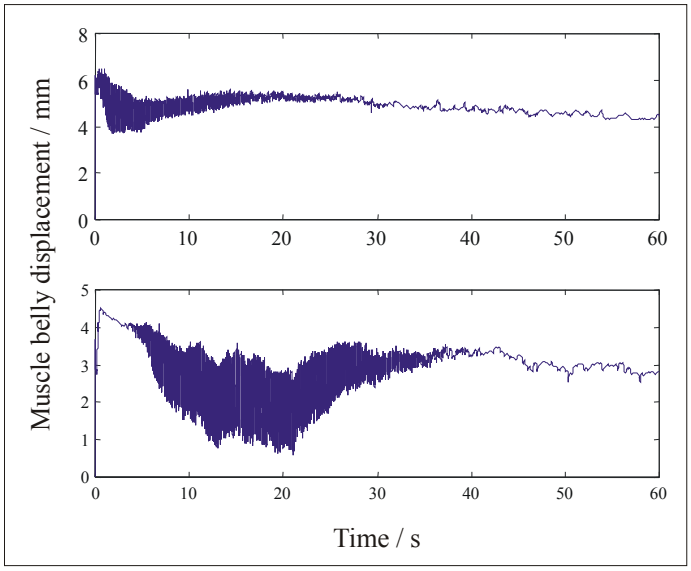


Fig.6: Responses to electrical stimulation (supra-maximum, 10 Hz) before (upper) and after (lower) targeted training process, m. VL

REFERENCES

/1/ Dahmane R., Valenčič V., Knez N. and Eržen I., Evaluation of the ability for non-invasive estimation of the muscle contractile properties on the basis of the muscle belly response, Medical & Biological Engineering & Computing, Vol. 39, 2001, 51-55

/2/ Kogovšek N. and Valenčič V., Measuring of skeletal muscles' dynamic properties, Artificial organs, Vol. 33, No. 3, 1997, 240-242

/3/ Valenčič V., Direct measurement of the skeletal muscle tonus, Advances in external control of human extremities, Vol. 10, 1990, 575-584

/4/ Delagi E. F., Perotto A., Iazzetti J. and Morrison D., Anatomic guide for the electromyographer: the limbs, Charles C. Thomas, Springfield, Illinois, USA, 1975

/5/ Enoka R. M., Neuromechanical Basis of Kinesiology, Second edition, Human Kinetics, The Cleveland Clinic Foundation, 1994

/6/ Bottinelli R. and Reggiani C., Human skeletal muscle fibres: molecular and functional diversity, Progress in Biophysics & Molecular Biology, Vol. 73, 2000, 195-262

/7/ Salmons S. and Vrbova G., The influence of activity on some contractile characteristics of mammalian fast and slow muscles, Journal of Physiology, Vol. 201, No. 3, 1969, 535-549

ACKNOWLEDGEMENTS

The presented study was supported by the Slovenian Ministry of Education, Science & Sport, Foundation for financing sport organizations in Republic of Slovenia and Our Space Ltd.

AUTHOR'S ADDRESS

Katarina Kerševan
Laboratory of Biomedical Visualization and
Muscle Biomechanics
University of Ljubljana
Faculty of Electrical Engineering
Tržaška 25, 1000 Ljubljana
Slovenia
e-mail: katarina.kersevan@fe.uni-lj.si

Twitch summation with double stimulation

Rainoldi A.[1,2], Durfee W.K[3], Merletti R.[2]

[1]*Centro di Bioingegneria, Dip. di Elettronica, Politecnico di Torino, Torino, Italy*
[2]*Dept. of Physical Medicine and Rehabilitation, Università di Tor Vergata and Fondazione Don Gnocchi, Roma, Italy*
[3]*Dept. of Mechanical Engineering, University of Minnesota, Minneapolis, USA*

SUMMARY

Both the mechanical and the EMG responses to a pair of electrical impulses depend upon the interpulse interval (PI) and the structure of the muscle. The latter can therefore be investigated using the paired stimuli technique and information concerning twitch summation and refractory period may be obtained. We documented how the force twitch and the electrically elicited response are affected by the PI and the pulse amplitude. Our findings showed that the maximum synergic effect of the doublet, that is the ratio between the peak force produced by a doublet and that produced by a single pulse, is between 3 and 4 and is obtained for PI between 3.5 ms and 30 ms. The absolute refractory period average value was 2.2±0.1 ms. Muscle fiber conduction velocity (CV) is higher for the second response. Repeatability of mechanical and electrical responses is poor and methodology must be improved.

STATE OF THE ART

The mechanical response of muscle to multiple pulses has been shown to be a non linear summation of the evoked twitches. The use of N-let impulse train in FES applications, following the terminology introduced by Karu et al /1/, is a well known technique aimed to increase force with a concomitant reduction of muscular fatigue /1/2/. Most of the published works focused on the time course of the force signal as the available information to monitor fatigue /3/4/5/6/7/. It is well known that mechanical manifestations of fatigue are related to muscular endurance and that they are anticipated by changes in the electromyographic signal /8/. The effect of N-let impulse train on myoelectric signal is not discussed in the literature. For this reason the aim of the present work is to evaluate the effect of double impulses on both force and EMG signals and to assess the repeatability of mechanical and electrical responses in different days.

MATERIALS AND METHODS

We define *doublets* as the stimulus pattern made by two rectangular pulses separated in time by the pulse interval (PI) and *singlets* the stimulus made by one single rectangular pulse. Five subjects (27.8±3.8 years) were involved in this experiment. The protocol consists of one session made of five different trials in the following order: 25 singlets, 200 doublets with PI randomly chosen in the range 1-5 ms, 200 doublets with PI in the range 1-20 ms, 200 doublets with PI in the range 1-100 ms and, finally, 25 singlets. Each stimulus was supramaximal and was generated once a second; trails were separated by 10 minutes of rest. The whole session was repeated in three different days on the biceps brachii muscle to test the repeatability of the measures. The force twitches generated by these impulse trains were recorded by means of a load cell mounted on an aluminum arm designed to avoid resonance frequencies in the force signal bandwidth. The EMG signals were recorded by means of a four electrode bar array placed distally on the muscle between the innervation zone and the tendon region. The proper electrode position was determined for each subject by means of a multi-channel linear electrode array according to the findings described elsewhere /9/10/. The variables under study were: the peak of force (PF) the time to reach the peak (TTP), the half relaxation time (HRT), the twitch area (TA), the conduction velocity (CV) and the correlation coefficient (CC) between two double differential signals used to estimate CV.

Supramaximal constant current stimulation was applied in monopolar mode with one small electrode (Ø = 3 cm) on the motor point of the biceps and one large electrode (10x12 cm) on the triceps.

RESULTS

Comparison between the 25 singlets stimuli at the beginning and the end of each session (included in the protocol to check the *stability* of the experiment) show great differences, indicating instability of the response during the experiment. Hence, if fatigue or, more probably, movements of the stimulation and/or of the recording electrodes occur, the three trials within each session may be different not only because of the PI variation but also because of uncontrolled factors. As a consequence also the day to day repeatability may be affected and was indeed poor. For such reasons the best session for each subject was selected by means of a multi-criteria method based on CC value, twitch and double differential signals morphology, trends of CV and PF and uniformity and superimposition of the force plots in the three PI ranges (see Figure 1).

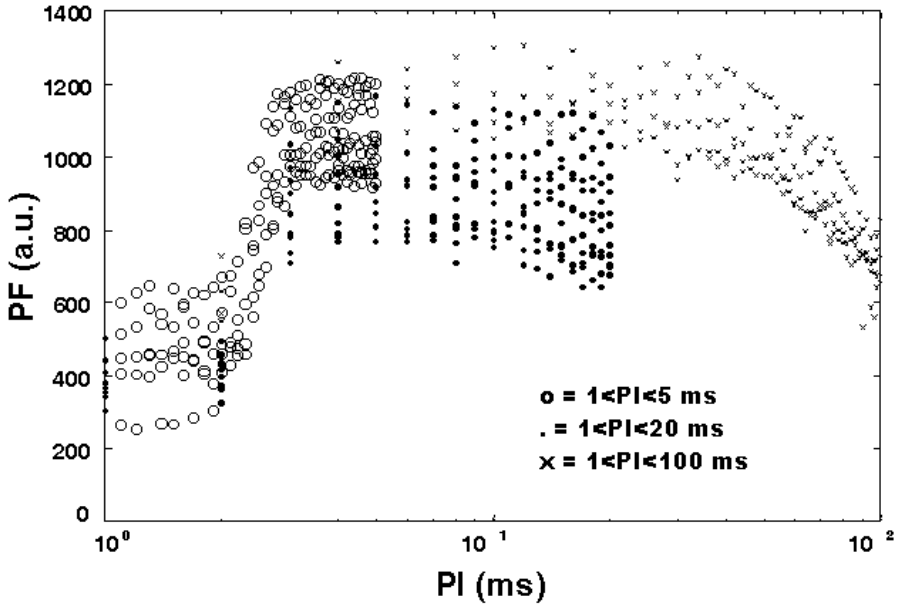


Figure 1. Example of a selected session showing PF in the three trials as a function of the PI (log X axis).

The pattern of force plotted with respect to the PI (for PI in the range 1:5 ms) allows to evaluate the average value, among five subjects, of the Absolute Refractory Period (ARP) as 2.2±0.1 ms. This result is in agreement with those found by Buchthal and Engbaek extrapolated for the temperature of 35 °C /11/.
As well known, the use of N-let pulse trains allows to increase, in a non-linear fashion, the force induced by the stimulus. The average maximal ratio between PF obtained with doublets and with singlets was 2.79±0.93 (range 2.12 - 3.96), quite constant within subjects (the average standard deviation between days was 0.28) and was obtained for PI = 3.5±0.1 ms (see Figure 2).

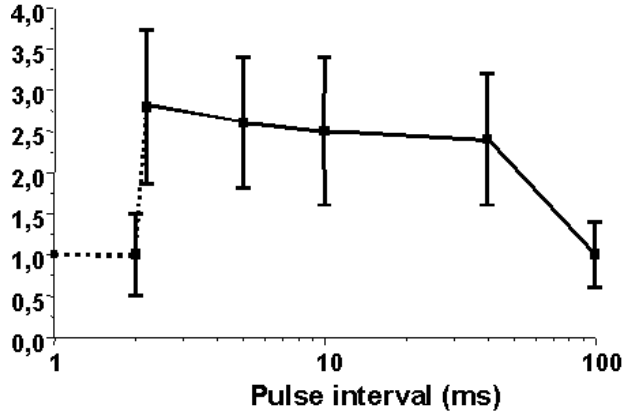


Figure 2. Ratios between the PFs obtained with a doublet and a reference force obtained from a single stimulus (mean and std. dev, N=5)

The TTP is always greater in double response with respect to singlets for PI>ARP. The mechanical summative effect due to the second stimulus always occurs for PI>ARP. Up to values of 70-80 ms the two twitches are fused together and the second peak is higher: this explain the TTP increase. The minimum values of TA correspond to the singlets. The maximum values is observed immediately after the ARP and decreases slightly in the range ARP-30 ms.
For PI>30 ms the two stimuli in each doublet generate two M-waves. The second M-wave has CV significantly larger (Wilcoxon paired test, $p < 0.01$) than the first for 30<PI<100 ms in four subjects out of five (see for example the behavior of one subjects in Figure 3).

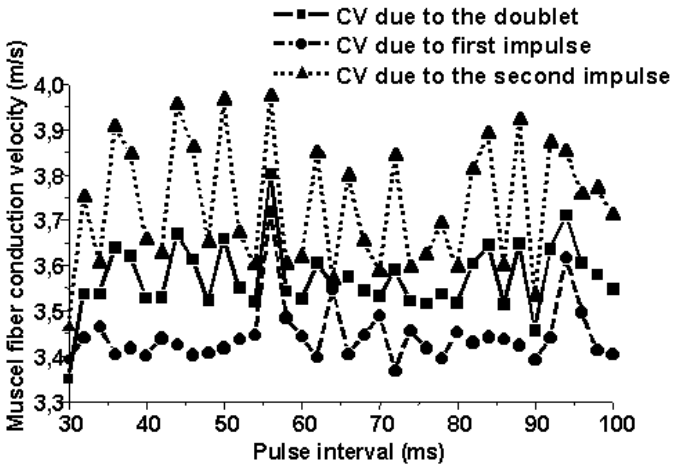


Figure 3. Comparison of CV values of the myoelectrical responses due to the doublet as a whole and due to the two separate stimuli in the range 30-100 ms.

Referring to the work of Hopf et al /12/ where a clear correlation between CV and TTP was found (although by using invasive techniques), we plotted the first variable versus the second one. TTP, in fact, could be considered as a good estimate of the contractions time when the stimuls is a singlet and in the case of doublet with PI<ARP. The two variables have shown large variability and poor correlation, hence the method seems not useful to properly estimate the contraction time.

DISCUSSION

The values of PI intervals corresponding to the ARP and to the maximal increase of PF were found to be highly stable and repeatable. On the contrary, other parameters, such as PF, TTP, TA, where subject to large fluctuations when electrodes were repositioned, when the subject slightly changed position or straps were adjusted. It is clear that the technique for investigating this phenomenon must be improved in order to study fatigue and any other long term effect. Nevertheless, it is unquestionable that the motor axons (or their terminal branches) have a refractory period of 2.2 ± 0.1 ms and the maximum PF is obtained for PI=3.5±0.1 ms but longer pulse intervals have similar effects. The first pulse has a conditioning effect upon the muscle fiber membrane which presents a higher CV value at the second pulse. This phenomenon has been previously reported /13/ and seems to be unaffected by PI within the range 30-100 ms.

REFERENCES

/1/ Karu Z. Z., Durfee W. K., Barzilai A. M., Reducing muscle fatigue in FES applications by stimulating with N-let pulse trains, *IEEE Trans. Biomed. Eng*., 1995,42;(8);809-817.
/2/ Dowling J.J. and Kennedy P., Non linear twitch summation of the human tibialis anterior, in Proceedings of the XVIth International Society of Biomechanics Congress, Tokyo 1997.
/3/ Levy M, Mizrahi J, Susak Z, Recruitment, force and fatigue characteristics of quadriceps muscles of paraplegics isometrically activated by surface functional electrical stimulation, *J Biomed Eng* 1990;12(2):150-6.

/4/ Houston ME, Grange RW, Myosin phosphorylation, twitch potentiation and fatigue in human skeletal muscle, *Can J Physiol Pharmacol* 1990;68(7):908-13.
/5/ Binder-Macleod SA, Lee SC, Baadte SA, Reduction of the fatigue-induced force decline in human skeletal muscle by optimized stimulation trains, *Arch Phys Med Rehabil* 1997;78(10);1129-1137.
/6/ Binder-Macleod SA, Lee SC, Russ DW, Kucharski LJ, Effects of activation pattern on human skeletal muscle fatigue, *Muscle Nerve* 1998;21(9):1145-1152.
/7/ Mourselas N and Granat MH, Evaluation of patterned stimulation for use in surface functional electrical stimulation systems, *Medical Engineering & Physics* 1998,20:319-324.
/8/ Merletti R., Roy S., "Myoelectric and Mechanical Manifestations of Muscle Fatigue in Voluntary Contractions", *JOSPT*, vol. 24; 6; 342-353, 1996.
/9/ Rainoldi A., Nazzaro M., Merletti R., Farina D., Caruso I., Gaudenti S., Geometrical factors in surface EMG of the vastus medialis and lateralis, *J Electrom Kinesiol* 2000;10(5):327-336.
/10/ Merletti R, Rainoldi A, Farina D. Surface electromyography for noninvasive characterization of muscle. *Exerc Sport Sci Rev*. 2001;29(1):20-5.
/11/ Buchthal F, Engbaek L, Refractory period and conduction velocity of the striated muscle fibre, *Acta Physiol Scand*. 1963;59:199-220.
/12/ Hopf H. C., Herbort R. L., Gnass M., Günther H., Lowitzsch K., Fast and slow contraction times associated with fast and slow spike conduction of skeletal muscle fibers in normal subjects and in spastic hemiparesis, *Z. Neurol*.,1974, 206:193-202.
/13/ Nishizono H, Kurata H, Miyashita M. Muscle fiber conduction velocity related to stimulation rate. *Electroencephalogr Clin Neurophysiol*. 1989 Jun;72(6):529-34.

AUTHORS' ADDRESSES

Alberto Rainoldi
LISiN Centro di Bioingegneria
Politecnico di Torino
Via Cavalli 22/g
10138 Torino, Italy
rainoldi@athena.polito.it
www.lisin.polito.it

William K Durfee
Dept. of Mechanical Engineering
University of Minnesota
111 Church Street S.E.
Minneapolis, MN 5545-0111
wkdurfee@tc.umn.edu
www.me.umn.edu/~wkdurfee

Roberto Merletti
LISiN Centro di Bioingegneria
Politecnico di Torino
C.so Duca degli Abruzzi 24
10129 Torino, Italy
merletti@polito.it
www.lisin.polito.it

FES INDUCED SURFACE MUSCLE STIFFNESS CAPTURED BY COMPUTER CONTROLLED TONOMETRY

D.Rafolt*, E.Gallasch**, M.Fend**, M.Bijak*, H.Lanmüller*, W.Mayr*

* University of Vienna, Department of Biomedical Engineering and Physics
** University of Graz, Department of Physiology

SUMMARY

A new tonometric test system to assess surface stiffness over relaxed and activated calf muscles was developed. The mechanical arrangement consists of a skin indentor driven by a force controlled galvo-drive which is rigidly connected to an ankle dynamometer. Software routines for cyclic indentation (recording of stiffness curves), static indentation (sensing of twitch responses), and vibration (skin admittance) were implemented. A visual interface is used to capture surface stiffness during defined voluntary calf contraction/ relaxation. For FES-applications the software includes an impulse synthesizer, to generate arbitrary stimulation test patterns. The system's performance was tested in FES and voluntary contraction procedures.

STATE OF THE ART

Surface stiffness over a muscle belly basically consists of passive and active components. In the relaxed belly passive components such as skin and fiber network viscoelasticity as well as interstitial fluid pressure play a dominant role /1/. During contraction the fiber network shortens which is accompanied by belly shape changes and an increase of intramuscular pressure. If surface stiffness is sensed by local indentation (in vivo) further the shape of the indentor and bone architectural factors have to be considered. Due to these multifactorial influences on surface stiffness, rather reliable relative than absolute results are expected from such measurements.

Various techniques and test schemes for surface stiffness are described in the literature. The first indentation apparatus was described by Schade /2/ to study the creep properties of skin and subcutaneous tissues. Recently Veldi et al /3/ described a hand-held device to produce short force impacts, with an accelerometer to study the excitation response. Here the surface stiffness is expressed by the oscillation frequency estimated from the response. Another approach is based on static preloading and the measurement of muscle belly enlargement. Typically such a system is based on a inductive displacement sensor with an internal spring to provide contact loading during isometric contractions /4/.

The tonometric system described here was projected to assess FES induced surface stiffness over calf muscles. To obtain smooth stiffness curves, the stimulation frequency has to be set high enough (< 30 Hz, fused tetani). In contrast to voluntary contractions, FES causes an inverse recruitment, and therefore big and fast fatigable motor units are activated first. For low stimulation frequencies the contraction responses of these units cause skin surface motions which are also detectable by indentation. To precisely track these muscle responses, a fast responding, force programmable indentor was developed.

MATERIALS AND METHODS

The tonometric system is build around a leg attached isometric ankle dynamometer /5/ consisting of footplate and telescopic stand, see Fig.1. During plantar flexion the produced force against the stand is sensed by a load cell (0-3000 N). The dynamometer further serves as mechanical reference for the indentor. The indentor consists of the rotary drive with lever and an exchangeable skin interface. To

achieve a high system dynamics, a force controlled galvo-drive was used for indentation (Type M3, General Scanning Inc, USA). The torque produced by this drive (± 20 Ncm) is linear to the input current. The lever length is 5 cm resulting in a 1N/A indentation force (10N max). Further the drive includes a rotary displacement sensor (± 15°) to measure the indentation depth. For the skin interface a 7 mm ball element is used.

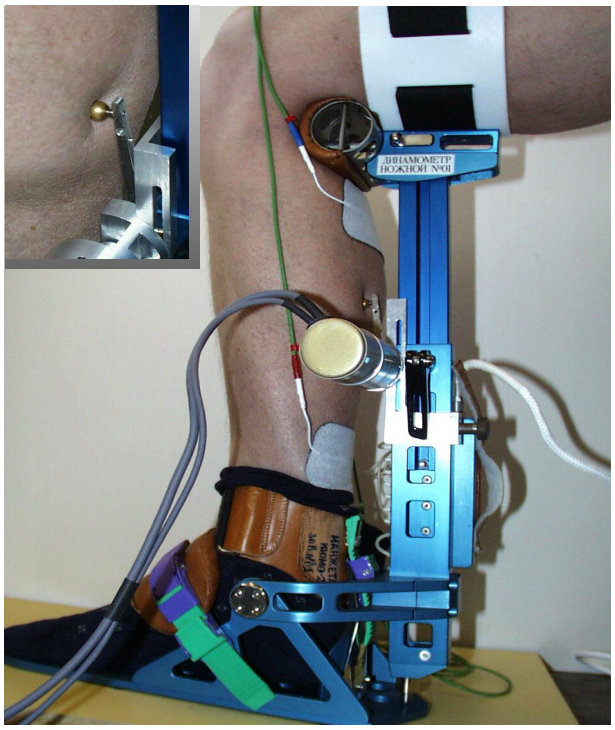


Fig.1: Mechanical arrangement of the dynamometer with tonometer

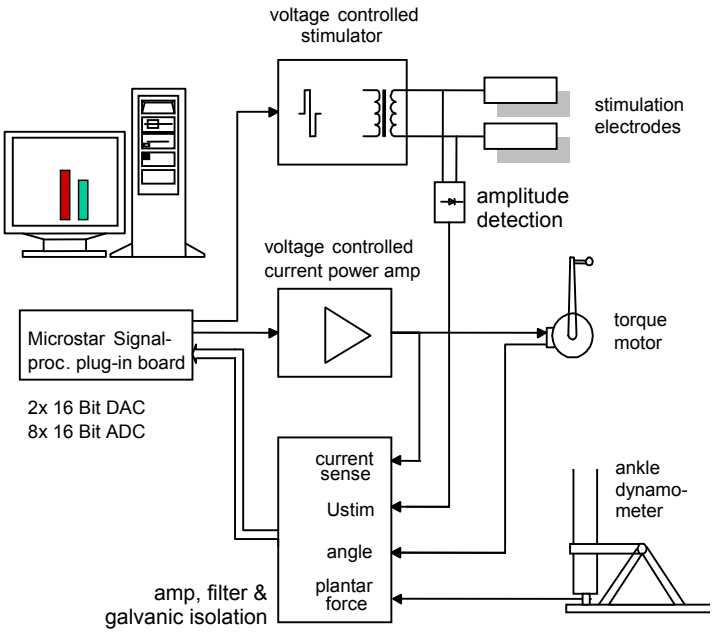


Fig. 2:
Control and signal flow in the myotonometric system

The hardware to operate the tonometric system is shown in Fig. 2. The rotary drive is powered by voltage controlled current source (drive amplifier) in order to avoid friction from back-EMV. The stimulator with internal galvanic isolation (MYOSTIM) converts an unipolar input signal into biphasic voltage waveform. Both the amplifier and stimulator input signals are generated by realtime DSP processes (DAP1216a/6, Microstar Laboratories, Inc.). Four measurement channels (indentation force and deflection, contraction force and stimulator output) are sampled at 1 kHz. Experiment guidance, visual feedback of contraction force and data storage runs on a standard PC, with the Matlab software package for data analysis. The system is calibrated with elastic specimens.

RESULTS

Three test procedures are implemented until now: (1) cyclic indentation for obtaining stiffness curves during FES induced or voluntary contraction, (2) static indentation for sensing twitch response patterns, and (3) vibration to estimate the skin's surface impedance. Fig. 3 shows a signal section during cyclic indentation, which was taken from the left medial gastrocnemius. The stimulation frequency was set to 30 Hz and the amplitude was stepwise increased (upper trace). The force signal (second trace) gives a measure of the isometric contraction in the ankle joint, which increases with stimulation time. The indentation input force (third trace) has a constant triangular waveform and the responding indentation displacement magnitude (fourth trace) decreases with contraction level.

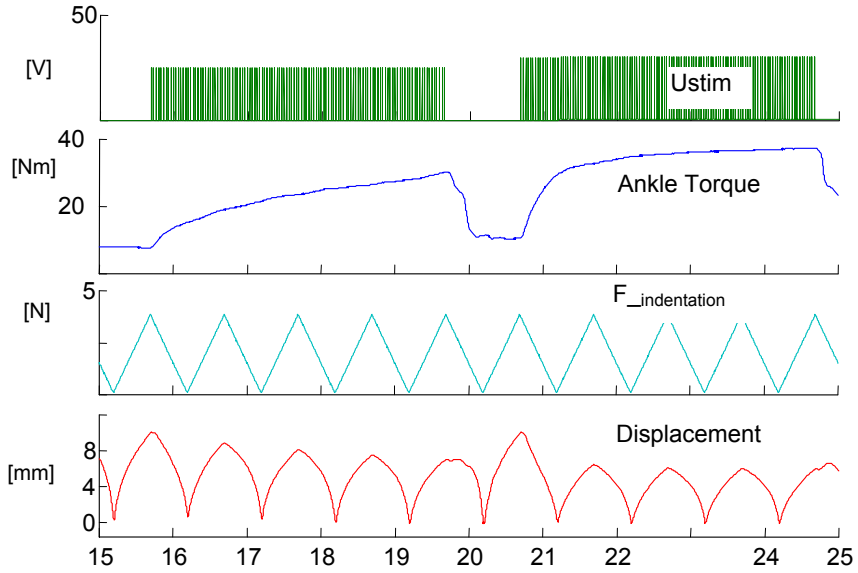


Fig.3:
Cyclic indentation signals during FES induced contraction. (Offsetchanges of the position of the muscle belly due to contraction are removed)

From the measured raw signals the indentation force/displacement curves were extracted. In Fig.4 the curves 1, 3 and 5 are from voluntary contractions and curves 2, 4 and 6 are from FES induced contractions at 500N and 750N (50Nm/75Nm torque) of plantar flexion force level. In all curves the slope increases with indentation force, which reflects a common feature of loaded biotissues /6/. With contraction level the curves became steeper at both contraction schemes. For subject #3 the stiffness values estimated for the relaxed muscle are 255N/m at a indentation force of 1N and 550N/m at 3N. At voluntary contraction these values increases to 320N/m and 805N/m respectively (ankle torque =50Nm) and to 543N/m and 898N/m respectively (torque=75Nm). At FES induced contractions stiffness increase to 588N/m and 964N/m respectively (torque= 50Nm) and to 722N/m and 1250N/m respectively (for 75Nm).

Table 1 shows the calculated stiffness data from 3 subjects. These values are comparable to those found with the oscillation method /3/.

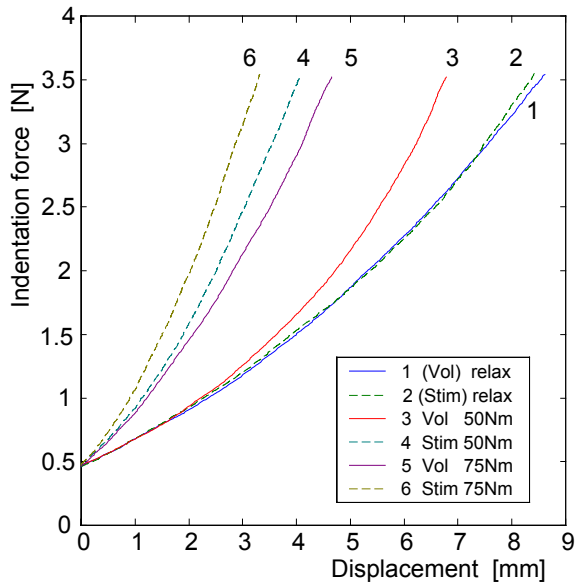
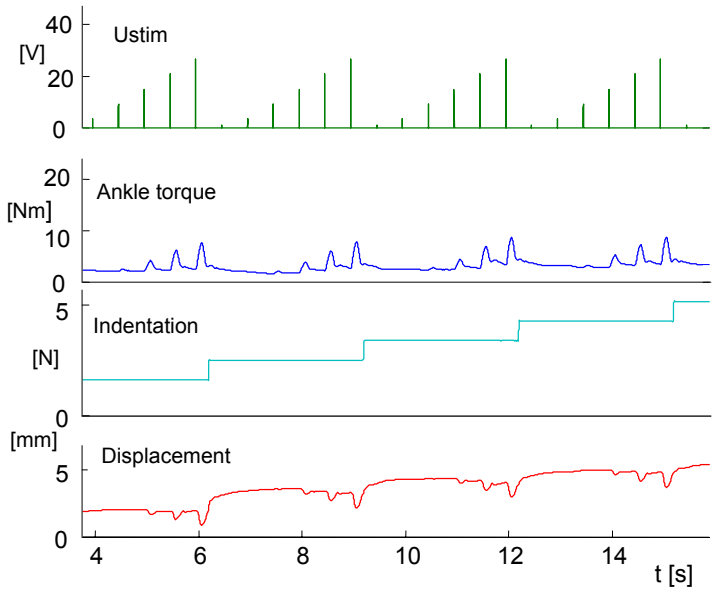


Fig.4: indentation force / displacement curves with contraction as parameter

Ankel torque	relaxed	relaxed	50Nm	50Nm	75Nm	75Nm
Indentation level	1N	3N	1N	3N	1N	3N
Sub #1 voluntary	253	564	354	833		
Sub #1 FES	290	625	625	1071		
Sub #2 voluntary	217	421	315	789		
Sub #2 FES	217	407	625	1071		
Sub #3 voluntary	255	540	305	805	543	898
Sub #3 FES	268	580	588	984	722	1250

Tab.1
Stiffness dF/dD in [N/m] at 1N and 3N identation level

In Fig. 5 belly displacement responses to single stimulation impulses are shown. Series of six impulses with increasing amplitude were applied to the calf muscles (upper trace) while they voluntarily were kept in the relaxed state. Indentation force loading was stepwise increased from 0 to 5N. There are minor



changes of the displacement response with indentation force level (third trace). The responses were additionally captured by the dynamometer load cell (second trace).
Compared with the belly displacement responses the isometric force responses show a better resolution of the amplitude contour, as still the small stimulation pulses are detected.

Fig.5: Single twitch responses with indentation force as parameter

DISCUSSION

The configuration of an isometric ankle dynamometer serving as a mechanical reference for a tonometer proved very advantageous, as the lower leg is kept in a defined position during the measurements. This is very important when repeated measurements have to be performed on the same subject, for example during neuromuscular retraining in rehabilitation. Further the documentation of both the ankle contraction and the surface stiffness values would allow a more detailed analysis of FES or fatigue induced contractile changes /7/. One big advantage of the force controlled indentor is its independence of the zero position. Other systems using spring loads or positional actuators have a defined zero position, and therefore have to be adjusted precisely at the beginning of a measurement series.

REFERENCES

/1/ CWJ Oomens, DH Campen, HJ Grootenboer: In vitro compression of a tissue layer on a rigid foundation. J Biomech 20: 923-935, 1987
/2/ H. Schade: Untersuchungen zur Organfunktion des Bindegewebes. Zeitschr f Exp Path und Therapie 11: 369-399, 1912
/3/ M. Veldi, V. Vasar, A.Vain, T.Hoin, M.Kull: Computerized endopharyngeal myotometry. J Sleep Res 9: 279-284, 2000
/4/ Dahmane R, Valencic V, Knez N, Erzen I., Evaluation of the ability to make non-invasive estimation of muscle contractile properties on the basis of the muscle belly response.
Medical & Biological Eng. & Comp 39: 51-55, 2000
/5/ D. Rafolt, E. Gallasch: Body fixed ankle dynamometer for isometric tests in space flights. Biomedizinische Technik 41: 91-97, 1996
/6/ E.Gallasch, HW. Weizsäcker: An automated tensile test system for biotissues. Biomed. Technik 31: 224-229, 1986
/7/ D. Rafolt, E. Gallasch, W. Mayr, H. Lanmüller: Dynamic force responses in electrical stimulated triceps surae muscles: effects of fatigue and temperature. Artificial Organs 23: 436-439, 1999

AUTHOR'S ADDRESS

Dietmar Rafolt Ph.D
e-mail: d.rafolt@bmtp.akh-wien.ac.at
Dep. of Biomedical Engineering and Physics
General Hospital, Univ. Vienna,
Währinger Gürtel 18-20, A-1090 Vienna

STIMULATION ARTIFACT REMOVAL ALGORITHM FOR REAL-TIME SURFACE EMG APPLICATIONS

T. Keller* and M. R. Popovic**

*ParaCare - Institute for Rehabilitation and Research, University Hospital Balgrist Zurich, Switzerland
**Automatic Control Laboratory, Swiss Federal Institute of Technology Zurich, Switzerland

SUMMARY

A direct and intuitive method to control a neuroprosthesis for grasping is to use surface EMG (SEMG) activity of muscles that subjects can voluntary control, e.g. in the case of C5 or C6 SCI subjects the deltoid muscles. The measured voluntary SEMG activity in such applications is contaminated with stimulation artifacts (SA) that are much higher in amplitude compared to raw SEMG signals. Hence, to be able to use SEMG signals for control purposes one has to remove the SA from the measured SEMG signal. In closed-loop applications the SA can produce a positive feedback, which further stresses the importance of removing the SA from the measured signal in close-loop SEMG control applications.
Well-established SA removal techniques are artifact blanking and filtering methods. Real-time SA blanking methods, either hardware built sample-hold circuits or software blanking routines in digital processed SEMG signals loose all EMG information during the blanking period. Especially with current controlled stimulators, which have a very high output impedance, the electrode-tissue impedance can cause a SA of several milliseconds. Most of the SEMG SA filtering techniques are not viable in case of current stimulators using surface stimulation electrodes, since the long lasting SA tail overlaps in frequency and time domain with the voluntary SEMG activity.
A new method that encounters the randomness and stationarity of voluntarily generated EMG is presented. An ensemble averaged SA with exponential forgetting was subtracted from the recorded SEMG and an almost artifact free SEMG signal was obtained. Measurements with multi-channel stimulation patterns showed fast convergence of the algorithm. The algorithm was significantly less sensitive to changes of the stimulation pulse amplitude than to changes of the stimulation pulse width. The method can be implemented in real-time applications and requires a low computational power.

STATE OF THE ART

SEMG signals that are recorded during surface functional electrical stimulation (FES) from muscles close to the stimulation site are always contaminated with SAs. If a current regulated stimulator is used its high output impedance produces a slowly decaying SA that can last longer than 10 ms. The methods that were proposed in the past to eliminate the SA can be divided into three main groups: SA blanking, SA filtering, and SA subtraction methods.
Hardware /2, 3/ and software /1, 4/ artifact blanking or sample-and-hold blanking methods blanked or sampled-and-held the SEMG during the SA while loosing all signal information during that time.
SA filtering methods /5-8/ reduced the SA using linear, non-linear, or/and adaptive filtering, gain switching, slew rate limiting, or constant current/voltage switching techniques. Because the SEMG signal and the SA overlapped in time and frequency domain, all applied filters influenced the quality of the SEMG signal.
Software artifact subtraction methods /9-11/ subtracted a more or less pure SA from the mixed SEMG. The presented methods differed in the way the pure SA was obtained. For the control of neuroprostheses the proposed SA subtraction algorithms cannot be used, because the produced SAs changed with the action (e.g. grasping or releasing) over time and differed from a priori extracted SAs.
To overcome the above problems an enhanced ensemble averaged SA subtraction method with real-time capabilities was developed.

MATERIAL AND METHODS

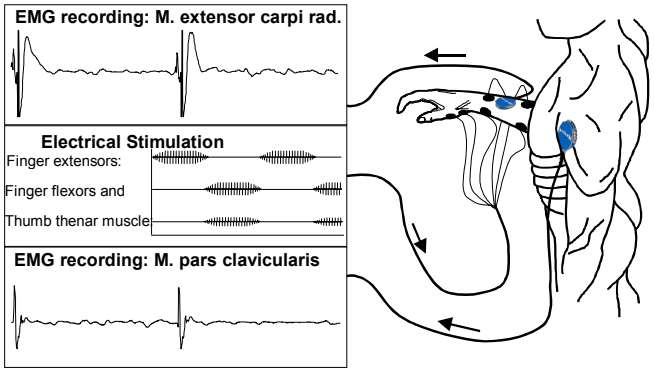
Algorithm

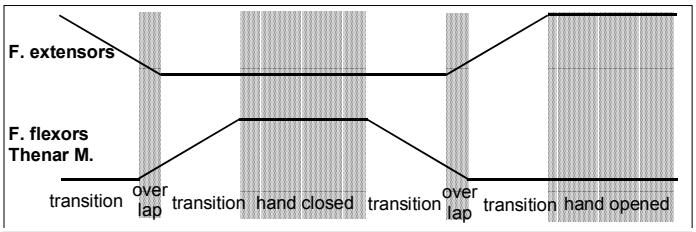
The SA was extracted from the first 125 samples (12.5 ms) post stimuli of the recorded SEMG signal that lasted 500 samples (50 ms) between two artifacts. A moving ensemble averaging algorithm with exponential forgetting was used to extract the SA and the direct muscle responses. The algorithm was deliberately kept very simple by applying a first order infinite impulse response (IIR) filter for the exponential forgetting. For each sample *n* the following recursive filter output was calculated:

$y(n|t) = \frac{x(n|t) + p \cdot y(n|t-1)}{p+1}$, where *p* is the weight that controls the forgetting. $x(n|t)$ is the n^{th} sample of the SEMG curve $X(t) = (x(1|t), x(2|t), \cdots x(125|t))$ measured at time t and $Y(t) = (y(1|t), y(2|t), \cdots y(125|t))$ is the extracted SA. Small *p* values stand for fast forgetting. The moving ensemble averaged SA (*Y(t)*) then was subtracted from the SEMG (*X(t)*). The algorithm did not process the SEMG from samples 126 to 499 post stimuli since it was always SA free.

Experiment

A COMPEX MOTION constant current stimulator provided a three channel stimulation sequence that alternating opened and closed the subjects' hand. COMPEX (5050MED) self-adhesive electrodes were used to stimulate the finger extensors (channel 1) during hand opening, the finger flexors (channel 2), and the thenar muscle (channel 3) during hand closing. The stimulation frequency was 20 Hz.
Two COMPEX biofeedback sensors (gain: 1400, bandwidth: 100-4000 Hz) were taped on the skin surface: one between the finger extensor stimulation electrodes over the M. extensor carpi radialis, and one on the M. pars clavicularis of the contralateral deltoid muscle. The sampling frequency was 10 kHz.





sequence	transition	overlap	hand close 10 mA / 8 mA F. flexors / thenar M	transition	overlap	hand open 12 mA F. extensors
1			start	1 s (PW)	0 s	5 s
2	5.4 s (PW)	0.6 s	2 s	5.4 s (PW)	0.6 s	2 s
3	2.7 s (PW)	0.4 s	2 s	2.7 s (PW)	0.4 s	2 s
4	1.8 s(PW)	0.2 s	2 s	1.8 s(PW)	0.2 s	2 s
5	0.9 s (PW)	0.1 s	2 s	0.9 s (PW)	0.1 s	2 s
6	2.7 s (AMP)	0.3 s	2 s	2.7 s (AMP)	0.4 s	2 s
7	0.9 s (AMP)	0.1 s	2 s	0.9 s (AMP)	0.1 s	2 s
8	1 s (PW)	0.2 s	5 s	end		

and recording **Table 1:** *show the performed stimulation sequences with different transition times.*

Stimulation sequences similar to the one used by our neuroprosthesis for grasping were applied to produce the time variant SAs. In each sequence the hand was closed for 2 s and then opened for 2 s.
A trial consisted of eight concatenated stimulation sequences that represented a typical grasping task with different transition times (see Table 1). When during the transitions the pulse width was changed (between 0 and 250 µs) it is marked with (PW) and when the pulse amplitude was changed (between 0 and 12 mA / 8 mA) it is marked with (AMP). Two such trials were conduced, one without and one with voluntary muscle contraction.

Signal Processing

The raw SEMG recording between two stimuli was divided into two parts (see Figure 2):

A the SA, 12.5 ms long
B the remaining SA free part curve that was not processed

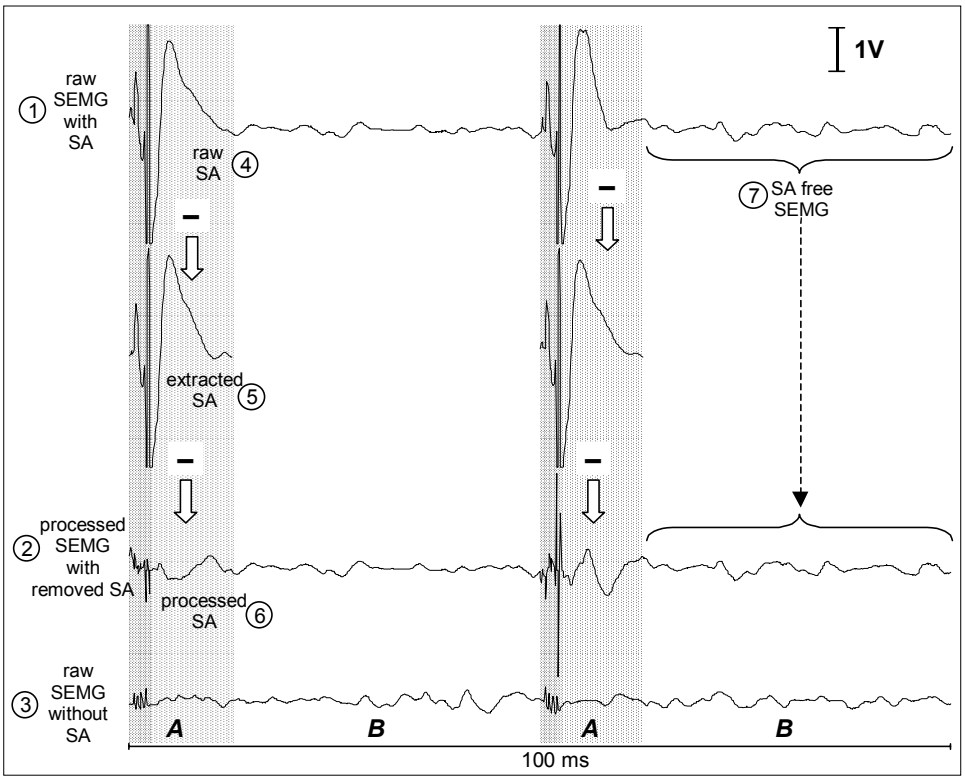


Figure 2: *The data processing steps are shown for the SA removal method and the result is compared to normal voluntary contraction for SEMG signals recorded over the wrist extensor muscles.*

The recorded signals were processed as follows (see Figure 2):

1. Part ***A*** (first 125 samples) (curve ④) were cut from the raw SEMG signal ① for each stimuli
2. The moving ensemble average algorithm provided curve ⑤ that was
3. subtracted from ④ and resulted in curve ⑥.
4. The result from step 3 was concatenated with the SA free part ***B*** (curve ⑦) and the first 3 ms after stimulus containing residual SAs were blanked (darker shaded in Figure 2).

RESULTS

During the constant stimulation phases of 2 s, the SAs were almost completely eliminated from the recorded SEMG signals for both electrode locations. In the processed wrist extensor SEMG signal only a few residual SA spikes during stimuli were left (see Figure 2, curve ②). The rest of the curve was SA free. The SA recorded on the deltoid muscle occurred only during the stimuli. No SA tail was produced. It has to be mentioned that in Figures 3 and 4 only the first 12.5 ms post stimuli are concatenated. The SA free part **B** is not shown.

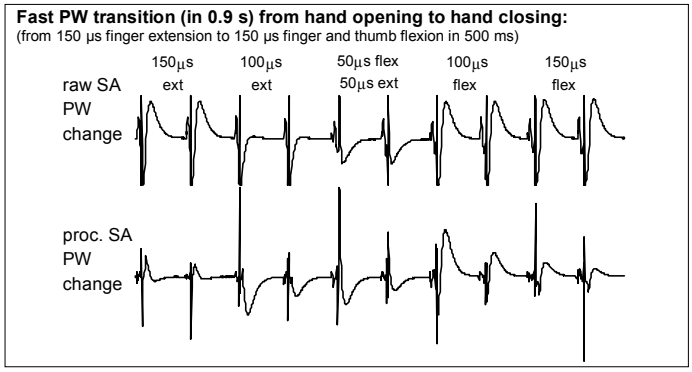


Figure 3: *The SA that was recorded between the stimulation electrodes on the M. ext. carpi radialis changed strongly during transitions for changing PWs.*

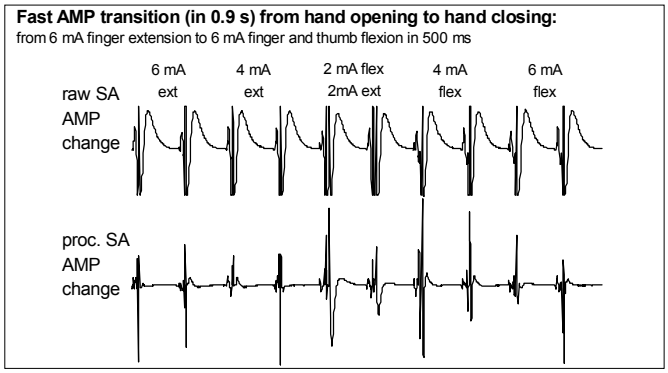


Figure 4: *Even very low stimulation amplitudes produced similar SA between the stimulation electrodes for constant (250 µs) PWs*

The transitions from hand opening to hand closing or vice versa were more problematic. If the pulse widths were rapidly changed and shorter than 100 µs, then the SA changed dramatically from pulse to pulse (see Figure 3) and caused errors in the estimation of the SA. Trials with different forgetting weights p could not reduce this effect. Optimal results were obtained with a forgetting weight $p = 1$.
If the pulse amplitude was changed the SA remained almost the same, even for very low amplitudes. Here also a forgetting weight $p = 1$ was optimal.

DISCUSSION

A novel SA removal method for real-time applications was presented. The algorithm subtracted a moving ensemble averaged SA with exponential forgetting from the SA contaminated SEMG of a voluntary activated muscle. The algorithm was capable of eliminating SA tails in presence of voluntary SEMG activity, even if the SA shapes were changing due to changing stimuli. The stimulation spikes could not be eliminated. We suggest blanking the signal during that saturated period (see dark shaded region in Figure 2). For fast transitions with pulse amplitude modulated stimulation pattern (PW 250 µs) it could be shown that the SA removal performance remained good. Fast changing stimulation pulse widths during transitions produced SA tails that could not be removed, because the SA changed strongly from pulse to pulse.

REFERENCES

/1/ T. Keller, A. Curt, M. R. Popovic, V. Dietz, and A. Signer, "Grasping in High Lesioned Tetraplegic Subjects Using the EMG Controlled Neuroprosthesis," *J. NeuroRehab.*, vol. 10, pp. 251-255, 1998.
/2/ J. A. Freeman, "An electronic stimulus artifact suppressor," *EEG and Clinic. Neurophysiol.*, vol. 31, pp. 170-2, 1971.
/3/ J. Minzly, J. Mizrahi, N. Hakim, and A. Liberson, "Stimulus artefact suppressor for EMG recording during FES by a constant-current stimulator," *Med. Biol. Eng. Comput*, vol. 31, pp. 72-5, 1993.
/4/ T. Handa, H. Takahashi, C. Saito, Y. Handa, M. Ichie, J. Kameyama, and N. Hoshimiya, "Development of an FES system controlled by EMG signals," Proc. of Int. Conf. IEEE EMBS, 1990.
/5/ M. Solomonow, R. Baratta, T. Miwa, H. Shoji, and R. D. Ambrosia, "A technique for recording the EMG of electrically stimulated skeletal muscle," *Orthopedics*, vol. 8, pp. 492-5, 1985.
/6/ C. M. Epstein, "A simple artifact-rejection preamplifier for clinical neurophysiology," *Am. J. EEG Techn.*, vol. 35, pp. 64-71, 1995.
/7/ F. Del Pozo and J. M. R. Delgado, "Hybrid stimulator for chronic experiments," *IEEE Trans. Biomed. Eng* , vol. BME-25, pp. 92-4, 1978.
/8/ V. Parsa, P. Parker, and R. Scott, "Convergence characteristics of two algorithms in non-linear stimulus artefact cancellation for electrically evoked potential enhancement," *Med. Biol. Eng. Comput.*, vol. 36, pp. 202-14, 1998.
/9/ K. C. McGill, K. L. Cummins, L. J. Dorfman, and B. B. Berlizot, "On the nature and elimination of stimulus artifact in nerve signals evoked and recorded using surface electrodes," *IEEE Trans. Biomed. Eng.*, vol. 29, pp. 129-37, 1982.
/10/ T. Blogg and W. D. Reid, "A digital technique for stimulus artifact reduction," *EEG and Clinic. Neurophysiol.*, vol. 76, pp. 557-61, 1990.
/11/ T. Wichmann, "A digital averaging method for removal of stimulus artifacts in neurophysiologic experiments," *J. Neurosci. Method.*, vol. 98, pp. 57-62, 2000.

ACKNOWLEDGEMENTS

This project was supported by grants from the Federal Commission for Technology and Innovation, Switzerland - Project No. 4891.1 and the Swiss National Science Foundation - Project No. 5002-057811

AUTHOR'S ADDRESS

Dipl. Ing. Thierry Keller
ParaCare, University Hospital Balgrist
Forchstrasse 340, CH-8008 Zurich, Switzerland
e-mail: kellert@balgrist.unizh.ch
home page: http://www.aut.ee.ethz.ch/~fes/

BIPHASIC STIMULATION: AN ALTERNATIVE APPROACH TO MINIMIZE THE STIMULUS ARTIFACT FOR DIAGNOSTIC AND FOR CONTROL APPLICATIONS

M. Reichel, M. Bijak, W. Mayr, H. Lanmüller, D. Rafolt, S. Sauermann, E. Unger, E. Turkof*

Department of Biomedical Engineering & Physics, University of Vienna
* Department of Plastic and Reconstructive Surgery, University of Vienna

SUMMARY

Diagnostic methods, where the electrical response to an electrical stimulus is recorded, provoke charging processes in the electrode and tissue interface. In the recorded signal the stimulation artifact normally strongly dominates in relation to the electrical stimulation response. If the response signal and the artifact overlap, the response signal is lost due to overdrive of the recording amplifier. This especially occurs during Intraoperative Motor Evoked Potentials in brachial plexus surgery and scoliosis surgery, where the stimulation amplitude may reach several hundreds volts: the ratio of the stimulation amplitude and the evoked, efferent motor Electroneurogram (ENG) is in the range of 10^8 together with a short latency ranging between 2 and 4ms. Similar problems occur during intraoperative localization of chronic facial nerve lesions, where the stimulation site (skull) and the recording site (facial nerve trunk) are especially close to each other, with latency down to 1ms.

Considering the fact that every rising edge of the stimulation impulse leads to a charging current and that every falling edge leads to a discharging current in the capacitive part of the load impedance, the two edges of a monophasic stimulation impulse produce a slowly decreasing discharge current through the tissue after the second (falling) edge. In order to minimize the remaining charge at the end of the impulse, we used a biphasic stimulation impulse with asymmetric length of the phases. By adjusting the ratio of the durations of the first and second phase, the remaining discharge current can theoretically be driven towards zero.

Reducing the electrode-tissue interface to an equivalent network of a resistor, the phase duration ratio can be calculated as described in the methods part. The resistor represents the tissue which is connected serially to a parallel circuit of a capacitor and a resistor, both representing the electrode-skin junction. This ratio depends on the time constant τ of the equivalent network, which is in the range of 0.5 to 2ms, and the duration of the first stimulation phase T_1 that is normally used in the range of 50µs to 500µs.

For example: with a pulse width of the first phase of 50µs, the percentage of the second phase T_2 of the stimulation impulse needs to be 95% to optimally minimize the stimulus artifact at a time constant of 1ms and with a pulse width of 500µs this percentage needs to be 65%. This shows that longer time constants need shorter second phases for optimal compensation and vice versa. The theoretical findings were confirmed by series of animal and clinical experiments.

STATE OF THE ART

Intraoperative Electroneurodiagnostic (IOE) /1/ with high voltage (above 100V) stimulus often lead to technical problems if latencies of the recorded ENG are very small, i.e. between 1 and 4ms. This is the case with intraoperative MEP in brachial plexus surgery, scoliosis surgery and transcranial electrical stimulation (TES) of the facial nerve. Following electrical stimulation of the motoric cortex, the evoked efferent motor nerve action potential is recorded from the surface of the exposed peripheral nerve in order to conclude from the potential's shape to the function of the nerve.

Electrical stimulation is applied by transcutaneous needle electrodes. The stimulus is a single, constant voltage, rectangular monophasic pulse up to 750V in amplitude and 100µs in duration (Digitimer 180) or up to 1000V and 50µs (Digitimer 185), respectively /2/. In spite of this short stimulus duration, the stimulation artifact will overdrive the recording amplifier and the evoked Electroneurogram (ENG) with a delay down to 1ms cannot be measured. One reason of the above mentioned is the recovery time of the instrumentation amplifier, and the second is the discharging process for charge balance of the load impedance. If fast recovery amplifiers are used, only discharging dominates overdriving the amplifier after the second (falling) edge of the monophasic pulse.

Biphasic stimulation pulses are well known in Functional Electric Stimulation (FES) /3/ for charge balancing over trains of pulses to avoid electrode and tissue damage. Usually rectangular biphasic stimulation pulses with up to 100V in amplitude and up to equally 1ms in duration of first and second phase are used to activate peripheral nerve structures via surface electrodes for causing muscle contraction. For IOE it is necessary to activate the central nervous system like the motoric cortex or the spinal root, respectively. Charge balance is not optimally provided by equally spaced phase duration, especially for higher stimulation amplitudes, which are needed to reach the nervous structures under bone tissue. The goal is to find the ratio between the first and the second phase of the biphasic rectangular stimulation pulse to get an optimally charge balanced situation immediately after the stimulus.

MATERIAL AND METHODS

Amplifier and Stimulator

The IOE-System consists of two computer controlled modules, a single-channel electrical stimulator and a six-channel ENG/EMG-recording unit /4/. The maximum amplitude of a single biphasic stimulation pulse is 1000V per phase at 50µs duration. The duration of the first phase is adjustable from 50µs to 1ms and the second phase can be tuned from 0% to 100% of the first (i.e. from monophasic to symmetric biphasic) due to optimal compensation of the stimulus artifact.

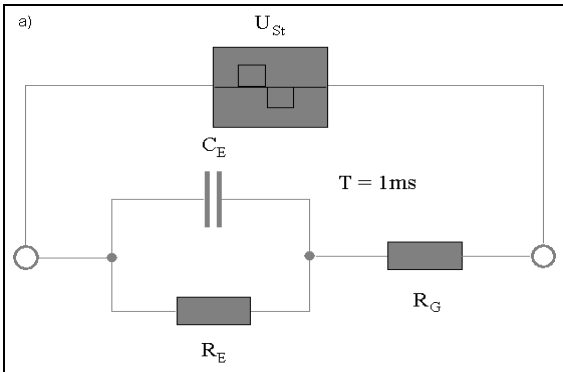
The amplifier consists of three parts. The first is a shielded part (i.e. the preamplifier) and has an amplification of 20. The second part consists of an isolation amplifier with two fixed amplifications, one for EMG- and the other for ENG-signals, 5 and 500, respectively. The third part is a multifunction I/O-PCMCIA-card for A/D-conversation of 12bit and an amplification of 1 to 100. Thus the amplification range is adjustable from 10^2 up to 10^6.

Theory of Compensation

The load impedance, including tissue, skin and electrodes, can simply approximated by an equivalent network of a resistor R_G, representing the tissue, serially connected to a parallel circuit of a capacitor C_E and a resistor R_E, both representing the electrode-skin junction *(Fig. 1a)*. Rectangular, constant voltage stimulation pulses produce a current flow over the electrodes and the tissue. In *Fig. 1b* three different pulses applied to the equivalent circuit are compared, where the amplitude of the current can be scaled because of independency regarding to minimize the artifact. Thus the monophasic pulse produces 20% of the maximum amplitude after its falling edge and has an exponential decay with a time constant of 1ms afterwards. The symmetric (100%) biphasic pulse leads to amplitude of 8% after its final rising edge, which is lower than the 12% of the monophasic at this point of time (1ms). The biphasic pulse with different length of phase 1 (100%) and phase 2 (67%) results in amplitude at almost 0% of the maximum current amplitude. The optimized ratio of the second phase T_2/T_1 depends on the pulse width of phase 1 T_1 and the time constant τ, which results from the equivalent circuit in *Fig. 1a* and can be expressed as

$$\frac{T_2}{T_1} = \frac{\tau}{T_1} \cdot \ln\left(2 - e^{-\frac{T_1}{\tau}}\right). \qquad \textit{Equ. 1}$$

This means that for a pulse width of the first phase of 500µs a pulse width of the second phase of 335µs (67% of the first) is the optimal duration for minimizing stimulation artifact.



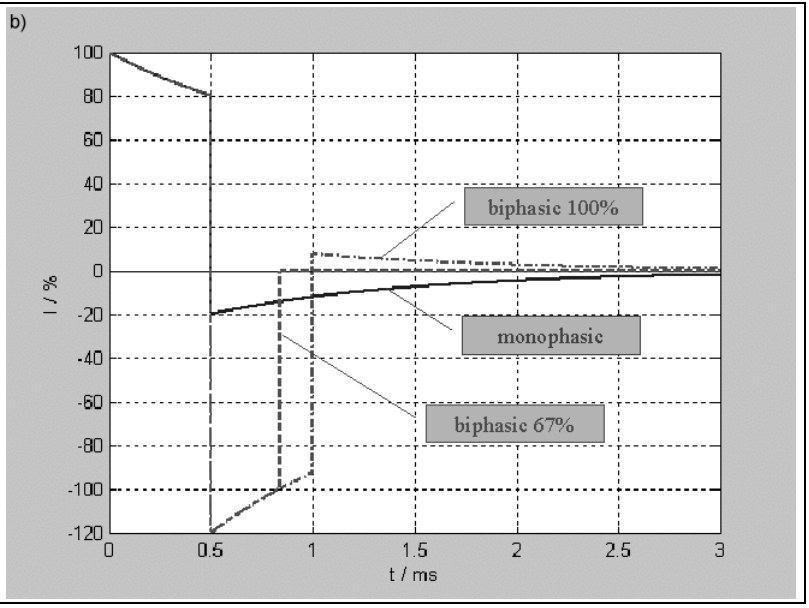
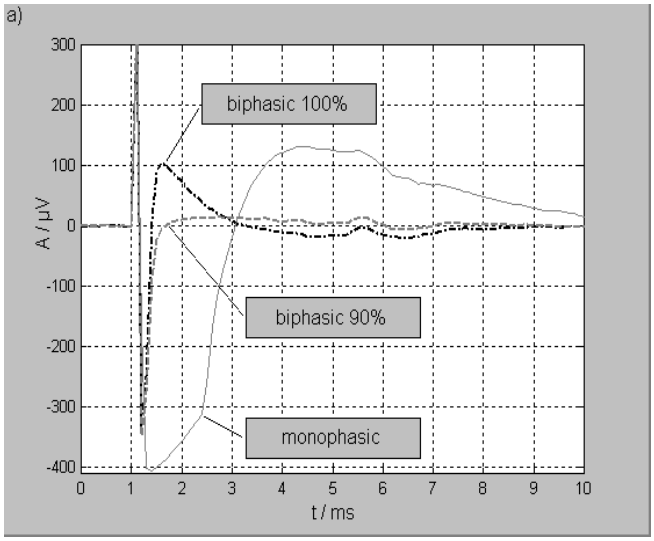


Fig. 1: a) Equivalent stimulation circuit with a time constant of 1ms and b) corresponding scaled stimulation current, calculated for rectangular (0.5ms), monophasic as well as 67% and 100% biphasic stimulation pulse of constant voltage U_{ST}.

RESULTS

An example in *Fig. 2a* points out the difference between three stimulation signals with amplitude of 500V and a pulse duration of 100µs, where first is monophasic, second and third are biphasic symmetric (100%) as well as asymmetric (90%). Due to a high pass filter, implemented in the recording amplifier, the recorded signals do not have a DC-part and the shape look different to the calculated shape in *Fig. 1b*.



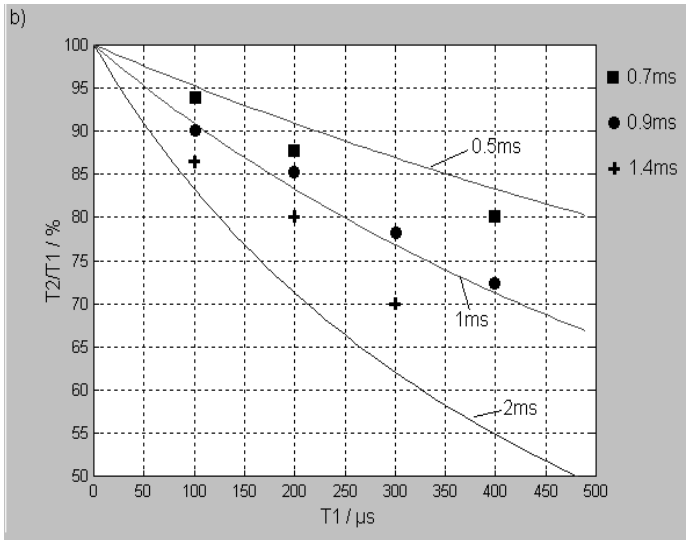


Fig. 2: a) ENG-recordings, corresponding to a time constant of about 0.9ms and a stimulation with an amplitude of 500V and a pulse duration of 100µs for monophasic, biphasic 100% and biphasic 90% pulses, where biphasic 90% has minimum artifact within 0.5ms after stimulation onset; b) percentage of T_2 over T_1 for minimized artifact depending on time constant τ; calculated for 0.5ms, 1ms and 2ms (solid lines) and recorded for 0.7ms (squares), 0.9ms (dots) and 1.4ms (crosses).

Also the monophasic stimulation pulse overdrives the amplifier for about 1.5ms, which means that no ENG-signal can be recorded in-between 1.5ms after the onset of the stimulation pulse. The recorded signal after the 100% biphasic pulse shows no overdriving of the amplifier, but it takes about 2ms of discharging exponential decay to reach an amplitude in the range of the amplitude of the ENG-signal

itself (20µV). Minimized artifact, even at 0.5ms after stimulation onset, occurs after a asymmetric biphasic stimulation pulse with duration of the second phase of 90% from the first. Recordings confirm to the theory of *Equ. 1*, which is shown in *Fig. 2b*. The continuous lines are drawn for the calculated ratio of T_2/T_1 over T_1 to minimize stimulation artifact at time constants of 0.5, 1 and 2ms. The squares, dots and crosses represent the percentage of T_2 at time constants of about 0.7, 0.9 and 1.4ms for recordings with minimized stimulation artifact.

DISCUSSION

In the course of IOE, where the electrical response to an electrical stimulus is recorded, charging processes in the electrode and tissue interface are provoked. The results show (*Fig. 2a*), that the discharging processes in the recorded signal can be shortened in time due to different pulse shape of the stimulation signal. A monophasic rectangular pulse of constant voltage has two disadvantages. First the recording amplifier is overdriven for several miliseconds and second the discharging exponential decay starts at high amplitude. This means, that lower amplification is required and the ENG-signal is very small or even cannot be detected. In comparison, biphasic stimulation signals do not (or only for some microseconds) overdrive the recording amplifier and the exponential decay starts at lower amplitude. This starting point of the discharging exponential decay can be adjusted by appropriate discharging due to vary the duration of the second phase of the biphasic stimulation pulse. The duration of the second phase for minimizing stimulation artifact is dependent on the duration of the first phase. For this reason, longer pulse width require shorter width of second phase for minimizing stimulation artifact and vice versa.

The required duration of the second phase can be predicted by *Equ. 1*, if the time constant of stimulation circuit is known. But due to noise and the essential filtering of the recording system, the time constant cannot be determined exactly. In some cases, where optimally minimized artifact is required caused by very short delay of ENG-signal, the duration of the second phase has to be adjusted manually. In most other cases, minimizing of stimulation artifact can be done automatically after determining of the time constant of the stimulation circuit.

REFERENCES

/1/ Turkof E., Millesi H., Pfundner P., Mayr N., Intraoperative Electroneurodiagnostics (Transkranial Electrical Motor Evoked Potentials) to Evaluate the Functional Status of Anterior Spinal Roots and Spinal Nerves During Plexus Surgery, Plastic and Reconstruction Surgery, 1997, 99(6): 1632-1641

/2/ Turkof E., Tambwekar S., Kamal S., El-Dahrawi M., Mansukhani K., Soliman H., Ciovica R., Mayr N., Leprosy Affects Facial Nerves at the Main Trunk and Neurolysis Can Possibly Avoid Transfere Procedure, Plastic and Reconstruction Surgery, 1998, 102(5):1565-1573

/3/ Kralj A. and Bajd T., *Functional Electrical Stimulation: Standing and Walking after Spinal Cord Injury*, 1989, CRC Press, Inc.

/4/ Reichel M., Bijak M., Mayr W., Lanmüller H., Rafolt D., Rakos M., Sauermann S. , Unger E., Turkof E., Mobile PC-System for Intraoperative Electroneurodiagnostics, 7th Vienna International Workshop on Functional Electrical Stimulation, Vienna 2001, Proceedings, 2001, currently in press

ACKNOWLEDGEMENTS

Supported by the Austrian National Bank. Projects 6946, 7937 and 8661.

AUTHOR'S ADDRESS

Dr. Martin Reichel
Dep. of Biomedical Engineering & Physics
Währinger Gürtel 18-20, A-1090 Vienna

e-mail: m.reichel@bmtp.akh-wien.ac.at
home page: http://www.bmtp.akh-wien.ac.at/bmt/alllanguages/people/reicma1

USABILITY OF EEG SIGNALS FOR FES CONTROL
-through Event Related Potentials Caused by Obscure Pictures-

Mariko F. Funada*, Satoshi Suzuki**, Keiko Kasamastu**,
Miki Shibukawa**, Satoki P. Ninomija**

*Hakuoh University
**Aoyama Gakuin Univerisity

SUMMARY

The goal of this report is to consider the usability of change of human event related potentials characterize when they watch some obscure pictures and judge the normality of the pictures. To reach the goal, a model is considered and simulations are done on the model, the characters are clarified by the results of the simulation and we discuss whether the signals of changes of ERPs are useful for FES control or not.

STATE OF THE ART

Event related potentials (ERPs) are special potentials measured from electroencephalograms (EEGs) and caused by special events like watching some pictures, listening to some sounds, or feeling some pain, etc. ERPs are one of very objective signals obtained from human beings when they are working. Farther, since the brains of human beings control their recognitions, judgments and activities and EEGs or ERPs explain directly the states of brains, EEGs or ERPs are one of the most excellent indicators that show the dynamic states of brains. If it is possible to use these potentials for functional electrical stimulation (FES), it becomes very useful by the reason of their characters.
The goal of this report is to consider the usability of ERPs for FES. And our approaches to the goal is to use the character of changes of actual ERPs when subjects are doing some task, to make a model to cause the same changes of ERPs, to do several simulations on the model, to clarify the characters of ERPs estimated by the results of simulation, and finally to discuss the usability for FES.

MATERIALS AND METHODS

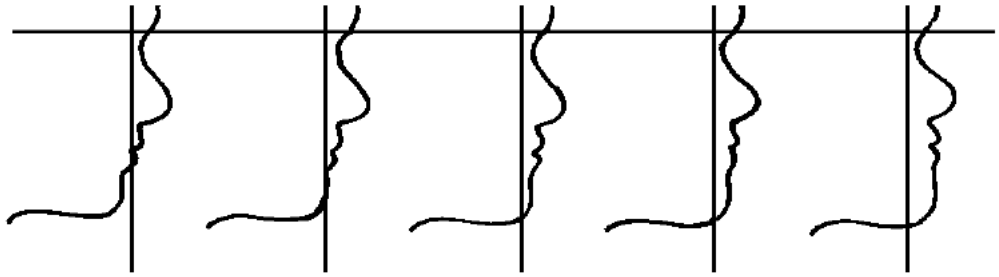
1. The Experiments that obtained the changes of ERPs
The experiment is followings;

(1) The displayed obscure pictures: five kinds of different facial outlines of human being called “1”, “2”, “3”,”4”, and “5” in Fig.1. “1” is the most normal, “5” is the most abnormal, and “2”, “3” and “4” are among them. The picture “3” is the most obscure one in these pictures. We call these pictures stimuli because they cause several ERPs on a brain.
(2) The frequency of displaying: 1 time 4 seconds.
(3) The displaying length of time: for 2 seconds.
(4) The number of displayed pictures : more than 250 times in total. The five kinds of stimuli are at least included in 50 times each.
(5) The order of five kinds of pictures: at random.
(6) The task: the subjects input the key "1" when they think the displayed picture is normal, and the key "2" when they think the picture is abnormal from the viewpoint of engaging.
(7) EEGs: single polar 19 channels by the 10-20 international method.
(8) The sampling frequency from analogue EEGs to digital one: 1KHz.

The number of subjects is eight. They are normal male students and an expert of teeth correction. The subjects without the expert are all 21-22 years old.

The analytical methods are followings;

(1) The measured EEGs were filtered by a kind of adaptive filtering methods[1].
(2) The he filtered data were smoothed by a moving average method and obtaining single recording ERPs.
(3) Three kinds of event related potentials called P100, N200, and P300 were detected automatically from the single stimulated ERPs which are calculated in (2), and getting the latencies and amplitudes of these potentials.
(4) The regressive lines were calculated for each potential and for the case of each picture.



stimulus 1 stimulus 2 stimulus 3 stimulus 4 stimulus 5

Fig.1 The obscure pictures used for the experiments

2. The changes of ERPs

Fig.2 is an example of the filtered data of measured EEGs3,4,5,7,8. The bottom of the waves is the first ERP and next is the second one. The potentials of P100, N200 and P300 appear clearly in the figure (The positive and negative direction is inverse by course of EEG recording.) We detected automatically the latencies of single recorded ERPs like waves in Fig.2, and calculated the regressive line of the latencies 7. Fig.3 is an example of the regressive lines of N200. The regressive line of stimulus"3" is very different from others. The stimulus "3" in Fig.1 is the most obscure to judge abnormal. That is, the change of latencies is the smallest in the most obscure stimulus.

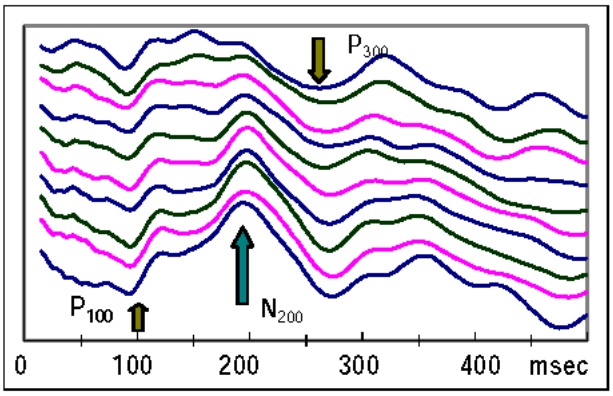


Fig.2 An example of single recorded ERPs

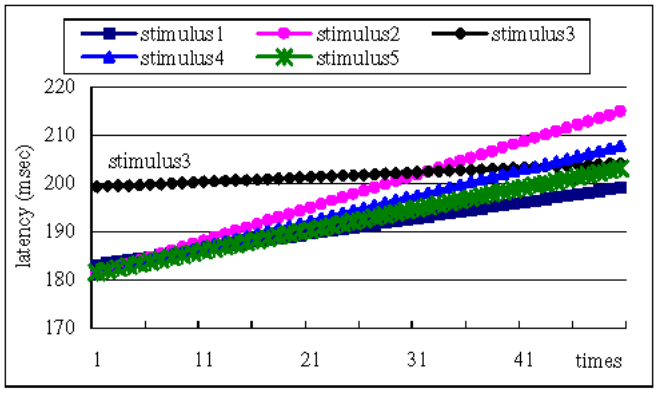


Fig.3 An example of regressive line of N200

3. A model to explain the changes of ERPs

EEG is the potential measured on the scalp and the potential consists of the sum of bursting neurons. Therefore, when high amplitude is observed, it means that a lot of neurons burst together in a short time. On the other hand, when low amplitude is observed, it means that a little number of neurons burst or the time lag of bursting is relatively large. This phenomenon matches the next formula[8];

$$ERPj(t) = \sum_{i=1}^{n(t)} Ti(j) \cdot wi(j) \cdot Ei \qquad \text{t=1,2, …. ,500msec, j=1,2, ….. ,250 times, wi(j)=)1,0 ….. ….. (1)}$$

1<=T(j)<=500

Ti(j) is the position which the i-th neuron bursts. wi(j) is the function which determines the i-th neuron bursts or not. Ei is the potential of the i-th neuron. On this model, we are able to explain that degree of obscurity to some subject determined the value of the function wi(j), Ti(j) and n(t). The most obscure picture causes neurons bursting in a short period.

3. Methods of simulations

We determined the parameters on the model (1) are followings;

Ei = sin(2*Pi/10*k) k=1, 2, …, 10, Pi=3.141592,

n(t) = (int) (sin(2*Pi/500*t - Pi/2)*50+51) t=1,2,…,500,

Ti(j)*wi(j) = (int) 50*sin(2*Pi/200*k + dd(j)) k=1,2,…,500, -2*Pi/100 <= dd(j) <= 2*Pi/100

Fig.4 shows the parameters for the simulation for stimuli without stimulus "3", and Fig.5 is the parameters for stimulus "3". n(t) is the same in these figures, and Ti(j)wi(j) is slightly different each other. We did some simulations using these values.

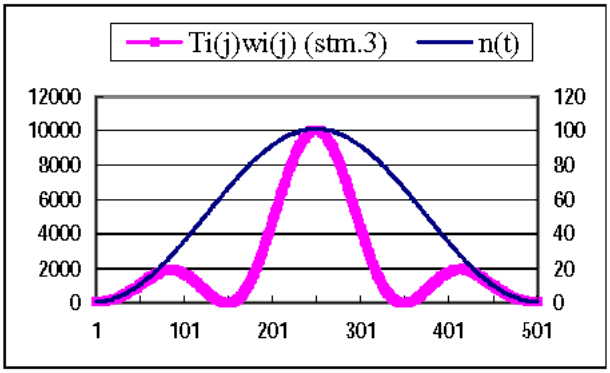


Fig.4 An example of parameters for the simulations in the case of stimulus "3"

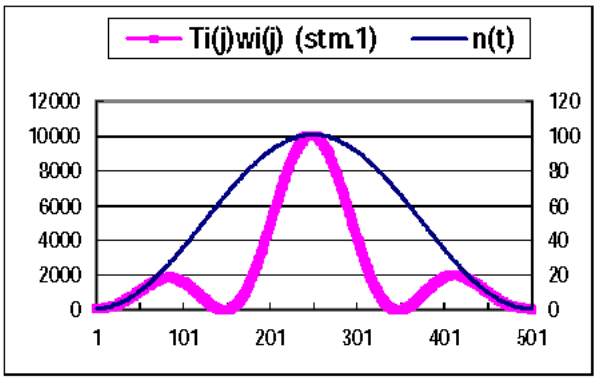


Fig.5 An example of parameters for simulations in the cases of others stimuli

RESULTS

Fig.6 and 7 are the results of simulations. Fig. 6 is a simulation for stimulus "3", and Fig.7 is a simulation for other stimuli. Each curve in these figures have peaks 100, 200 and 300msec, and they are very resembled the actual ERPs. Ei is too small to be recognized in the summed up data. These results of the simulations show that the model (1) can explain the changes of latencies of the actual ERPs.

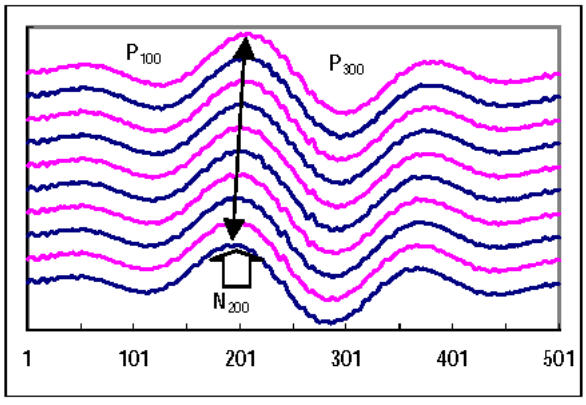


Fig.6 An example of data simulating the stimulus "3"

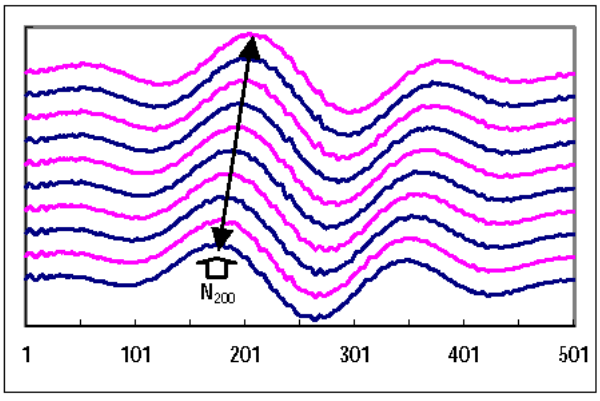


Fig.7 An example of data simulating other stimuli

DISCUSSIONS

Parameter Ti(j)wi(j) in Fig.4 and Fig.5 seems almost all the same, however, they are slightly different. Fig.8 shows the difference, and the maximum value of the differnce is about 300. The value 300 is not small, but the ratio to the maximum value of Ti(j)wi(j) is only 3%. This result of simulation shows that slight change of values of parameters, which can not be recognized by eyes, cause large changes that are observed by eyes.

Fig.9 shows the changes of latencies of N200 in the simulations, and two lines are similar to the stimulus "3" and "others". These results of simulations suggest that relatively large changes in ERPs or EEGs appear when the small changes occur in a brain. The signal for FES requires objectivity and sensitivity to detect small change of state. Special EEGs like ERPs have these required properties for FES. Therefore it is possible to use the changes of ERPs as a signal to useful FES.

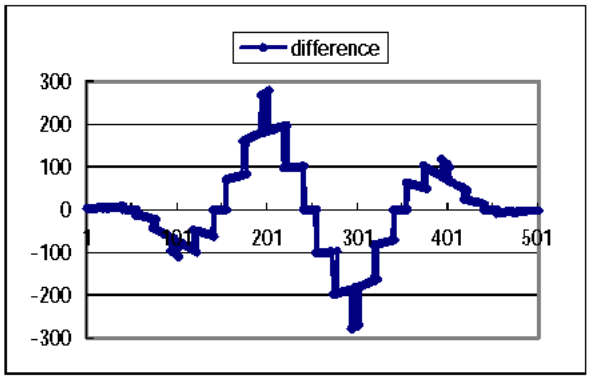


Fig.8 The difference of Ti(j)wi(j) in Fig.4 and 5; the difference of parameters between stm1 and stim3 in the simulations

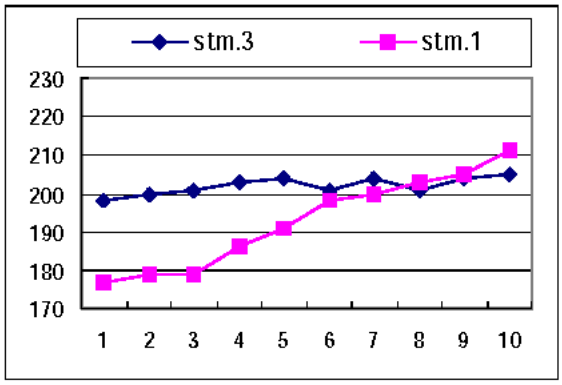


Fig.9 The change of latencies of data simulating stimulus "3" and "others"

REFERENCES

1 Defayolle, M., et al. Averaged evoked potential in relation to attitude, mental load and intelligence, Measurement of man at work, 1971: 141-152.

2 Donchin, E. et al. Application of brain event related potentials to problems in engineering psychology. Psycholophysiology, 1986: 702-718.

3 Mariko Fujikake, Satoki P. Ninomija, Hideki Fujita. A Coded Summation Method to Measure Evoked Responses of Human Beings, Journal of Medical Systems, 1989, 5(5): 275-292.

4 Mariko Fujikake, Satoki P. Ninomija, Hideki Fujita. On Considerations of Coded Summation Method to measure a Slow Vertex Response, Proceedings of the 11th Annual Conference of the IEEE Engineering in Medical and Biology Society, 1989.

5 Satoki P. Ninomija, Mariko Fujikake..A Coded Summation Method for Measuring a Visual Evoked Response, Proceedings of 11th Annual Conference of the IEEE Engineering in Medical and Biology Society, 1989.

6 M. Fujikake Funada, and Satoki P. Ninomija; Analysis of brain Activity for HCI, Symbiosis of Human and Artifact, 20B, 1995: 839-844.

7 Mariko F. Funada, Satoki P. Ninomija, et al. Difference between Changes of single stimulated Experts' ERPs and Those of Non-experts ERPs, Proceedings of the 1998 International Symposium on Noise Reduction for Imaging and Communication Systems, 1998: 91-95.

8 Mariko F. Fnada, Satoki P. Ninomija, et.al. Objective Evaliation of Human Fatigue under Computer Works through Event Related Potentials, APIEMS-2000, 42-46, 2000.

9 Wickens, C.D. Applications of event related potential research to problems in human factors. Basic issues and applications, 1990: 301-309 .

10 Williams, R. Westerkamp, J.J. Single Epoch Adaptive Filtering of Evoked Responses without using A Saparate training Signal. Proceeding of IEEE 1994 (15): 434-435.

AUTHOR'S ADDRESS

Prof. Dr. Mariko FUNADA
Graduate School MBA, Hakuoh University
1117 Daigyouji, Oyamashi, Tochigi, 323-8585, JAPAN
e-mail: funada@hakuoh.ac.jp

DEVICES FOR NON-INVASIVE TRANSCRANIAL ELECTROSTIMULATION OF THE BRAIN ENDORPHINERGIC SYSTEM: APPLICATION FOR IMPROVEMENT OF HUMAN PSYCHO-PHYSIOLOGICAL STATUS

Lebedev V.P.[1], Malygin A.V.[1], Kovalevski A.V.[1], Rychkova S.V.[1], Sisoev V.N.[2], Kropotov S.P.[2], Krupitski E.M.[3], Gerasimova L.I.[4], Glukhov D.V.[5], Kozlowski G.P.[6]

[1]Pavlov Institute of Physiology, [2]Military Medical Academy, [3]Regional Narcological Dispensary, Saint-Petersburg; [4]Sklifasovsky Research Institute of Emergency, [5]Center of Extreme Medical Situations, Moscow, Russia; [6]University of Texas Southwestern Medical Center, Dallas, TX, USA

SUMMARY

Here, we describe the clinical application of devices for non-invasive and selective transcranial electrostimulation (TES) of the antinociceptive system; and, their endorphinergic and serotoninergic neurotransmitter components. Our data is based on a large number and variety of experimental and clinical studies. The process of development and a brief description of these devices are presented. We also demonstrate the high efficacy of TES treatment for reduction of psycho-physiological disturbances elicited by stress of different intensities and a variety of other factors.

STATE OF THE ART

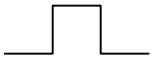
During the last century, transcranial electrostimulation (TES) was proposed as a method to elicit electronarcosis /1/, electrosleep /2/ and electroanalgesia /3/. Many attempts in several countries were made to introduce these methods into clinical practice /4/. Despite the breathless expectations based on TES methodology (i.e., non-pharmacalogic, easily controlled, with few side effects); the practical results of TES applications were quite negative. In hindsight, it is understandable because there were: no suitable experimental models, little knowledge of the optimal parameters for treating specific abnormalities, or well-controlled studies; that, acceptance of TES into the clinical area was slow, if at all. Our general aim was to develop the method of non-invasive, selective activation of the brain antinociceptive system with its endorphinergic and serotoninergic mechanisms by means of TES. Some data were presented previously /5, 6/. The specific aim of this paper is to describe the process of development of TES devices and its applications for reduction of psycho-physiological disturbances elicited by stress of different intensity and some other factors. This study was accomplished according to the established order of GMP, GLP and GCP.

MATERIALS AND METHODS

Development of devices

In the long history of TES, several different electrical regimens were introduced rather arbitrarily. To make a choice

Table 1. Characteristics of impulses studied in screening experiments to elucidate of its optimal parameters for transcranial electroanalgesia

Impulse shape	Frequency (Hz)		Width (msec)		Combination with DC
	Range	Steps	Range	Steps	
	40 - 100	1 – 5	0.1 – 5.0	0.05	Yes
	100 - 3500	10 -50	0.1-1.0	0.1	No
	40 -250	10	0.1 –5.0	0.2	Yes / No

of optimal regimen for stimulation of antinociceptive system, broad screening experiments were performed with non-traumatic pain models in several species of animals and volunteers with registrations of emotional, motor and autonomic pain-related events. The shape of impulses, range of frequencies, impulse width and the steps of its changes are presented in Table 1. All parameters studied were withinin the limits of ones of previously described devices for electrosleep and electroanalgesia /2, 3/. It was found that optimal stimulation antinociceptive system to elicit analgesic effect and maximal β-endorphin release are produced by rectangular impulses only of narrow band parameters of frequency and width slightly different for different species. The optimal parameters for humans were respectively 77.5 Hz and 3.5-4.0 msec. The relationship between the TES analgesic and other effects and impulse frequencies was very sharp and had rather "quasiresonance" shape. For example: ± 2 Hz frequency deviations from resonance point reduced effects at about 50 %, ± 4 Hz deviations abolished TES effects /6/. These data gave the basis to exclude an individual adjustment of frequency and impulse width for concrete patient in devices..

It was also found that constant voltage impulses are effective in combination with DC in ratio 1:2 only. Contrary constant current impulses of the equal amplitude had the same efficacy without additional DC. This result gave an opportunity to reduce significantly the level of current applied. In comparison between the analgesic efficacy of present TES regimen and regimen described by A.Limoge

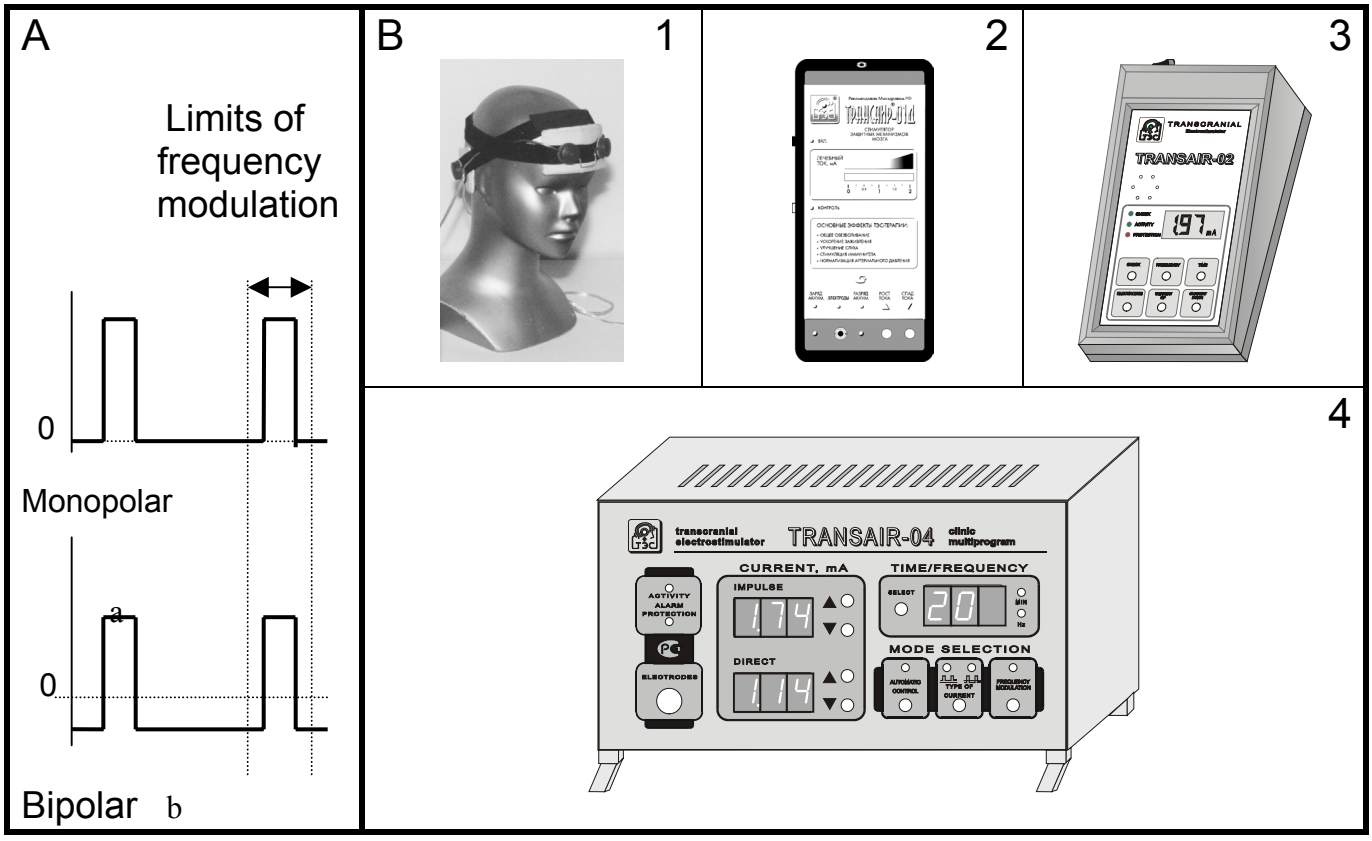


Fig. 1. Output current impulses (A) of TES devices (B).
A. Shape of impulses and limits of frequency modulation.
B. Different TRANSAIR models. 1- Headset of electrodes. 2- TRANSAIR – 02, the simpliest for outdoor usage, monopolar output impulse, rechargeable battery. 3- TRANSAIR – 01, for practitioners, mono- and bipolar output impulses, LCD, timer, frequency control, alarm and protection systems, plug in. 4- TRANSAIR – 04, for hospitals and outpatients clinics, mono- and bipolar output impulses with or without frequency modulation, LD indicators, timer, automatic control, alarm and protection systems, verbal dialogue with user in process of adjustment of parameters, plug in.

/7/ it was elucidated that the first one is much more effective. Further improvements of electrical regimen were directed to diminution of local irritating action on the skin under electrodes and increase the TES efficacy in population of patients. To realize the first aim the bipolar impulse with zero net charge (Fig.1A, phase "a" is equal phase "b") was used. For the second aim the stochastic frequency

modulation was introduced in the limits of the width of "quasiresonance" curve at the 50% level of its height (Fig.1,A).
On the basis of these results the development of some models of devices named later as TRANSAIR (abbreviation - **TRAN**scranial **S**timulator for **A**nalgesia, **I**mmunity and **R**epair) and adjusted for out- and indoor usage were developed and manufactured (Fig.1,B).

Experimental studies

The immobilization and cold stress of different intensity in rats were used as a model estimation of possible TES antistress effect. The intensity of stress-related events before and after TES were estimated immunocytochemically in neurons of brain cortex and several brainstem nuclei activation by immediately early gene (C-Fos) expression /8/ and morphologically by measurement of number and shape of gastric ulcers /9/. TES was performed by regimen specially adjusted for rats /10/.

Human studies

The blind and placebo-controlled (passive and active placebo) studies were produced to estimate the TES effects on stress-related events, affective disorders, and accompanied psycho-physiological and autonomic disturbances of different intensity in several groups of volunteers and patients. Some subjective verbal and non-verbal tests and objective tests were used for estimation of initial level psycho-physiological status and it changes after TES sessions (Table 2).

Table 2. Groups of volunteers and patients with stress-related events, affective disorders and accompanied psycho-physiological disturbances and tests its estimation.

No	Groups of volunteers and patients	Types of stress and fatigue	n	Tests and number of indexes (n)
1	Workers	Everyday stress and fatigue	141	*Subjective tests* – non-verbal: VAS, Color Lusher's test (4). *Subjective tests* – verbal: Self-estim. test (3), Spilberger's test (2), Quality of life. *Objective tests*: Correction test with Landolth's rings (8), Critical frequency of flashing merger, Reaction on moving object, Circulation tests (4), Breathing tests (3), Heart rate variability – two tests (9)
2	Soldiers	Stress and fatigue during the 1st year of military service	24	
3	Servicemen	Stress and fatigue in real field battle conditions	65	
4	Rescue workers	Professional stress and fatigue	12	
5	Relatives of the losts	Stress – syndrome of "terrible bereavement"	67	
6	Patients	Fatigue during depression	18	
7	Patients	"Chronic fatigue" syndrome	27	
8	Patients	Stress in postabstinence syndrome	247	
9	Patients	Posttraumatic stress (heavy thermal burns)	207	

RESULTS

Experimental studies

It was demonstrated that after even one TES session (30-60 min, current 1.0-1.2 mA) substantially reduced the number of neurons activated after immobilization and marked by C-Fos staining. This effect was found in 9 cortex areas of 12 studied especially in deep cortex layers. The reduction of stress-related C-Fos expression was also observed subcortical structures: in 4 of 6 thalamic nuclei and in 6 of 10 hypothalamic nuclei.

One TES session had curative and preventive effects on gastric ulcers elicited by immobilization in cold environment stress. Numbers and severity of ulcers in treated animals were substantially lower in comparison with untreated ones. TES effects were naloxone reversible that support of its endorphinergic nature. Thus experimental data presented gave the basis for clinical application of TES antistress effect.

Human studies

All groups of volunteers and patients included are into Table 2. In the members of the groups 1-4 stress of different level was elicited by the conditions of professional activity including groups 3-4, which had some level of danger of death. Members of 5th group had un-escapable stress as a relatives of lost in mass disaster. Members of group 6-7 have mainly high level of fatigue. In group 8 patients after treatment alcohol and heroin withdrawal were included. Patients of group 9 had posttraumatic stress disorders. In all cases it was demonstrated that fatigue, stress and other accompanied psycho-physiological disturbances were significantly improved or abolished after 2-5 TES sessions. The TES effects were more pronounced in cases of heavier disturbances.

DISCUSSION

It is well known that deficit of endorphins play important role in stress and affective disturbances of human psycho-physiological status. TES devices are effective for activation of the brain endorphinergic structures and its practical application is the effective homeostatic FES method for reduction of stress-related event and other psycho-physiological disturbances. Therefore, TES is an effective tool to greatly improve the quality of life.

REFERENCES

/1/ Leduc S. Production de sommeil et de l'anesthesie general et local par le courants electriques. C.R. Acad.Sci., 1902, 135 : 199-200.
/2/ Giliarovsky VA, Liventsev NM, Segal YE, Kirillova EA. Electrosleep. Moscow, Medgiz, 1958, 172 p.(in Russian)
/3/ Persianinov LS, Kastrubin EM, Rasstrigin LS. Electroanalgesia in obstetrics and gynecology. Moscow, Meditsina, 1978, 240 p. (in Russian)
/4/ Sances A.,Larson SJ. Electroanesthesia. Biomedical and biophysical studies, N.Y., Academic Press, 1975, 366 p.
/5/ Transcranial electrostimulation. Ed. Dvoretsky DP. Saint-Petersburg, Art of Russia, 1998, 528 p. (in Russian).
/6/ Lebedev V.P. Non-invasive transcranial electrostimulation of the brain antinociceptive system as a method of TES: an overview. 5th Annual Conference of the International Electrical Stimulation Society, Aalborg, Denmark, 2000 :123-126.
/7/ Limoge A. An introduction to electroanesthesia. Baltimore, University Park Press, 1975, 121 p.
/8/ Imaki T, Shibasaki T, Hotta M, Demura H. Early induction of C-Fos precedes increased expressiom of corticotropin-releasing factor messenger ribonucleinic acid in the paraventricular nucleus after immobilization stress. Endocrinology, 1992, 131 : 240 – 246.
/9/ Ferry S, Arrigo-Reina R, Candeletti S et al. Central and peripheral sites of action for protective effect of opioids on the rat stomach. Pharmacol. Res. Commun., 1983, 15 : 409-418.
/10/ Lebedev VP, Savchenko AB, Fan AB, Zhiliaev SV. Transcranial electroanalgesia in rats: optimal electric parameters. Fiziol. Zhurn.SSSR, 1988, 74 : 1094-1101 (in Russian).

Valery P. Lebedev. Transcranial Electrostimulation Center, Pavlov Institute of Physiology, nab. Makarova, 6, Saint-Petersburg 199034, Russia, E-mail: tes@infran.ru; lebedev@infran.ru, lvp@infran.ru; www.tes.spb.ru.

Session 7

STIMULATION AND CLOSED LOOP CONTROL

RIGOROUS GREEN'S FUNCTION FORMULATION FOR TRANSMEMBRANE POTENTIAL INDUCED ALONG 3-D INFINITE CYLINDRICAL CELL

L.M. Livshitz[*], P.D. Einziger[**], and J. Mizrahi[*]

Departments of Biomedical[*] and Electrical[**] Engineering, Technion, Israel Institute of Technology, Haifa 32000, Israel

SUMMARY

The quasistatic electromagnetic field interaction with 3-D infinite cylindrical cell is investigated for both intra and extracellular current point source excitations (IPS and EPS, respectively). The induced transmembrane potential (TMP), expressed conventionally via Green's function, is expanded alternatively into a faster converging representation using a complex contour integration, consisting of an infinite discrete set of exponentially decaying oscillating modes (corresponding to complex eigenvalues) and a continuous source-mode convolution integral. The TMP is insensitive to large variations of the internal/external conductivity ratio as long as the membrane/external (or membrane/internal) conductivity ratio is maintained very low. The dominant contribution for both the IPS and EPS problems are obtained in simple closed-form expressions, including well documented special functions.

STATE OF THE ART

In the IPS case the dominant modal contribution (of order zero), an exact solution of the well-known cable equation, is explicitly and analytically corrected by the imaginary part of its eigenvalue and the source-mode convolution contribution, both of the order of the membrane/external conductivity ratio. The limit where the external conductivity approaches infinity is well documented in the literature /1/. In this limit the convolution integral contribution vanishes and complex eigenvalues become real. In the EPS case the dominant contribution is expressed as a source-mode convolution integral. However, for a long EPS distance (> 100 fiber diameters) the mode (of order one) involved in the convolution is not a solution of the cable equation. Only for shorter EPS distance should the cable equation solution (i.e., mode of order zero) be included in addition to the dominant mode (of order one). For on-membrane EPS location additional modes should be included as well. In view of our EPS result, we suggest that the cable equation analysis and modeling already contained in the literature and related to functional electrical stimulation for EPS problems /2/, /3/ should be critically reviewed and corrected.

METHODS

The physical configuration of our problem, depicted in Fig.1, consists of source-point S, an observation point P, and two cylindrical regions, the axoplasmic core and the surrounding fluid, separated by a thin membrane of radius a. Assuming that the core, the outer fluid and the membrane are homogeneous, isotropic, ohmic conductors, their electrical parameters are denoted by σ_i, σ_e and G_m, respectively.

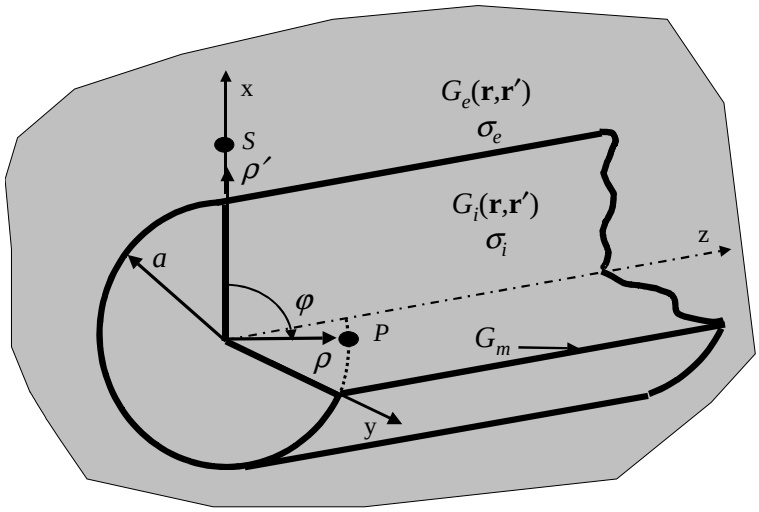


Figure 1. Physical configuration

The TMP, i.e., the difference between the internal and external potentials on the membrane surface, is given for a point source excitation as

$$V(\varphi, z, \mathbf{r}') = \frac{I}{\sigma(\rho')}[G(\mathbf{r}^-, \mathbf{r}') - G(\mathbf{r}^+, \mathbf{r}')] = \frac{I}{2\pi\sigma(\rho')}\sum_{n=0}^{\infty}\varepsilon_n V_n(z, \rho')\cos(n\phi), \quad \textbf{(1)}$$

where $G(\mathbf{r},\mathbf{r}')$ denotes the point source response (Green's function), ε_0=1, and ε_n=2, for n>0. The coordinates $\mathbf{r}'$=(ρ',0,0) and $\mathbf{r}$=(a,φ,z) corresponds to locations of the source-point S and the observation point P, respectively. The superscripts '+' and '-' represent quantities evaluated at the outer and inner surface of the membrane, ρ=a, respectively. The point source current is I, and $\sigma(\rho')$ is either σ_i for ρ'<a or σ_e for ρ'>a. The Green's function in (1) can be expressed in terms of cylindrical harmonics /1/ leading to

$$G(\mathbf{r},\mathbf{r}') = \frac{1}{2\pi^2}\sum_{n=0}^{\infty}\varepsilon_n\int_0^{\infty} g_n(\rho,\rho',k)\cos(kz)\cos(n\phi)dk. \quad \textbf{(2)}$$

Both $G(\mathbf{r},\mathbf{r}')$ and $g_n(\rho,\rho',k)$ in (2) satisfy, 3-D and 1-D Laplace equations and appropriate constraints. For the internal source problem, the Fourier coefficients $V_n(z,\rho')$ in (1), obtained after a straightforward but somewhat tedious calculation, are given via

$$V_n^i(z,\rho') = \frac{1}{\pi}\int_0^{\infty}\left\{\frac{I_n(k\rho')}{I_n(ka)}\Big/\left[\frac{G_m a}{\sigma_i} + ka\frac{I_n'(ka)}{I_n(ka)} - \frac{G_m a K_n(ka) I_n'(ka)}{\sigma_e K_n'(ka) I_n(ka)}\right]\right\}\cos(kz)dk, \quad \textbf{(3)}$$

where $I_n(ka)$ and $K_n(ka)$ is the modified Bessel functions of the first and second kind, respectively. Similarly, the expression for the external source problem (Fig.1) is given as

$$V_n^e(z,\rho') = \frac{1}{\pi}\int_0^{\infty}\left\{\frac{K_n(k\rho')I_n'(ka)}{K_n'(ka)I_n(ka)}\Big/\left[\frac{G_m a}{\sigma_i} + ka\frac{I_n'(ka)}{I_n(ka)} - \frac{G_m a K_n(ka) I_n'(ka)}{\sigma_e K_n'(ka) I_n(ka)}\right]\right\}\cos(kz)dk. \quad \textbf{(4)}$$

The integral representation in (2)-(4), representing continuous slowly converging summation over the spectral component of the transmembrane spectrum (the slow convergence is of O(1/ka) as $ka\to\infty$), can be expressed alternatively via equivalent discrete spectrum and an O($G_m a/\sigma_i$) continuous spectrum contribution, which converges much faster (i.e., exponentially for $z>0$ and $G_m a/\sigma_e\to 0$). The alternative representation utilizes φ-ρ eigenfunctions expansion, guided along z direction, rather then φ-z expansion guided along ρ direction which is definitely more appropriate for axon propagation problems.

RESULTS

Calculations of $V(\varphi, z, \mathbf{r}')$ and its Fourier coefficients $V_n(z, \rho')$ for both the EPS (Figs.2a-2d) and IPS (Figs.3a-3b) problems are carried out, utilizing (1), (3)-(4) (or better, its alternative representation), with the parameters σ_i/σ_e =1 and $\sigma_i/G_m a$=200. The normalized source location ρ'/a are specified for the indicated values.

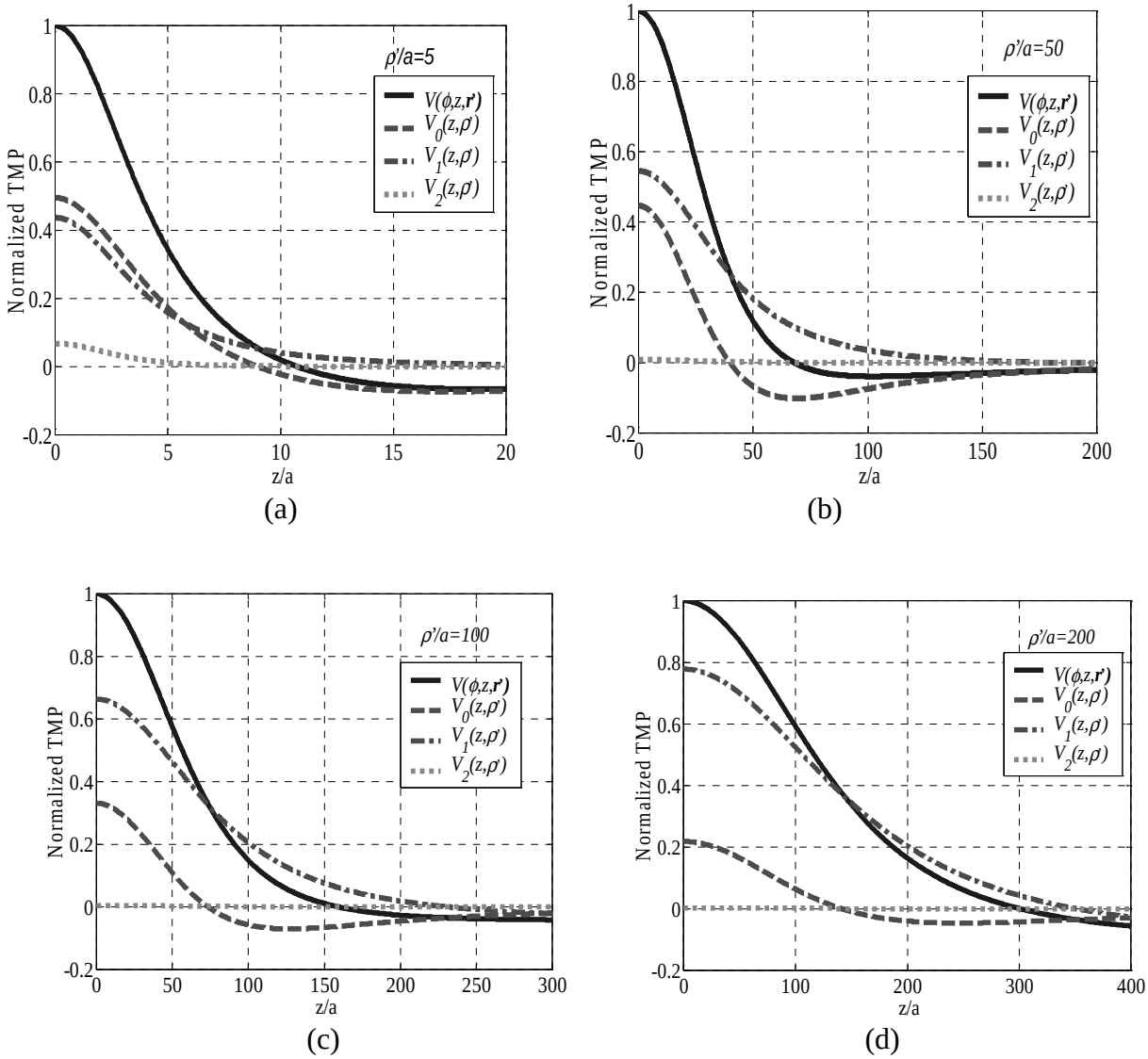
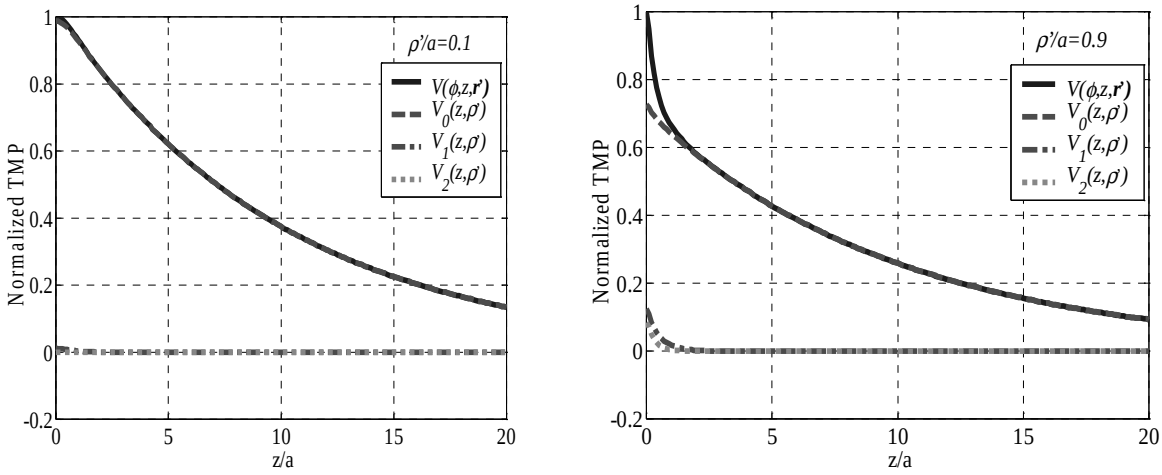


Figure 2 Normalized TMP distribution for the EPS problem.



(a) (b)

Figure 3 Normalized TMP distribution for the IPS problem.

DISCUSSION

The EPS problem signifies the importance of the n=1 mode (Figs.2a-2d). The n=0 mode, solution of the cable equation, is significant only for source-point locations which are relatively close to the axon. Thus, the conventional approach which accounts for the zero order mode only /2/, /3/ should be critically reviewed and corrected.

The IPS calculation highlights the dominance of the n=0 mode (Figs.3a-3b). Note that for large values z/a, the TMP decays algebraically for finite σ_e values and exponentially in the limit $\sigma_e \to \infty$ (Eq.(3)). The exponential decay is due to the largest current leakage from the axoplasmic core into the surrounding fluid.

REFERENCES

/1/ Adrian, R.H., Electrical Properties of Striated Muscle in Handbook of Physiology Section X *Skeletal muscle* Ch.10. ed. L., Peachy, Bethesda, Maryland, American Physiological Society, pp. 275-300, 1983.

/2/ McNeal, D.R., "Analysis of model for excitation of myelinated nerve", IEEE Trans.Biomed.Eng., vol. BME-23,329-337, 1976.

/3/ Rattay, F., *Electrical Nerve Stimulation* Wien, Springer-Verlag, 1990.

/4/ Jackson J.D., Classical Electrodynamics, 3rd edition, Wiley, New York, 1999.

ACKNOWLEDGEMENTS

This work was supported by the ISLER Foundation.

AUTHOR'S ADDRESS

Prof. J.Mizrahi, Dept. Biomedical Engineering, Technion, IIT, Haifa 32000, Israel, Fax +972 4 8234131, e-mail: jm@biomed.technion.ac.il

SIMULATION OF THE THREE DIMENSIONAL ELECTRIC FIELD IN THE COURSE OF FUNCTIONAL ELECTRICAL STIMULATION

M. Reichel, T. Breyer, W. Mayr, F. Rattay*

Department of Biomedical Engineering & Physics, University of Vienna
* Department of Analysis and Technical Mathematics, Vienna University of Technology

SUMMARY

Optimization of stimulation parameters as well as shape and positioning of electrodes are important questions in Functional Electrical Stimulation (FES) of paraplegic patients. For that reason a MATLAB tool, named FES-FIELD, modeling the three dimensional electric field in human body has been developed to calculate the electric field in a region of interest. The simulation tool provides a Graphic User Interface (GUI).

In case of denervation of the lower extremities an important target muscle is the m. quadriceps femoris. The electrical potential distribution along its fibers is representative for its functional activation. For this special application the human thigh stimulated by skin electrodes has been modeled.

The simulation process has been done in five steps:
1) Reading the geometric information of the thigh from 50 CT-slices (256 by 256 pixels).
2) Segmentation in tissue types by pixel value and definition of each conductivity.
3) Selection of electrode geometry and positioning.
4) Calculating the electric field iteratively by solving the system of approx. 1.5mio. linear equations.
5) Visualization of the solution by equipotential lines in either cross- or length-sections of the thigh.

The simulation requires 5-6 hours time for approx. 6000 iterations computed with a standard PC (800MHz CPU, 512MB RAM). The graphical information and the electrical solution can be exported to binary files for further investigations like calculation of the activating function of representative fibers in several muscle regions.

STATE OF THE ART

Most of the knowledge on Functional Electrical Stimulation (FES) is the product of more than 30 years of experimental use /1/ /2/, i.e. it is empirical and subjective. For quantitatively observation a 2D-model of the distribution of the extracellular electric field in a length section of the human thigh has been established in 1999 /3/. It focuses on denervation of the lower extremities.

The major target muscle for FES that causes knee extension is the m. quadriceps femoris. Distribution of the electrical potential along its fibers is representative for its functional contraction caused by its electrical activation. In this 2D-model, where the femur is of low conductivity (approx. less than 1/10 of the conductivity of muscle), the current - applied through surface electrodes covering the m. quadriceps femoris - cannot be passed on to the hamstrings below the femur.

However, in a 3D-simulation the stimulation current would be passed on to the hamstrings through the muscles enclosing the femur. Voltage distribution would change considerably compared to the 2D-model, i.e. a stronger electric field would be observed in the region of the hamstrings. Hence, functional tetanic contraction of the quadriceps influenced by the co-contractions of the hamstrings would be investigated much better.

MATERIAL AND METHODS

Geometric model

The geometric information is available as a number of either frozen cuts or CT cross-sections each stored in a - 256 by 256 pixel - graphic file format of *.jpg or *.tif, respectively. The MATLAB application FES-FIELD provides an input filter for both formats, which first converts the files into gray scaled 8bit images (*Fig. 1*), then finds the outermost contour (skin) and last colors the contour-pixels white and all outlying pixels black (air). In this example, pixel size is 1 by 1mm (Δx, Δy) and the distance between each cross-section is 10mm (Δz). This determines a 3D-grid equidistant in each direction (x, y, z).

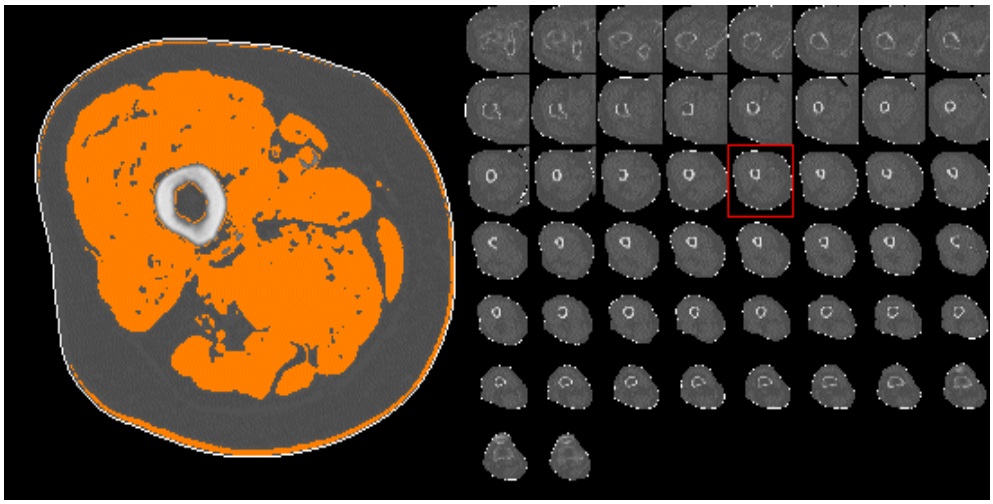


Fig. 1: 8bit gray scaled cross-section (256 x 256pixel ≙ 1x1mm) in the upper third of the left human thigh (left) corresponding to the framed cross-section and further cross-sections taken lengthwise every 10mm from the hip down to the knee (right).

relative conductivities

material	γ (S/m)
muscle	0.1
fat	0.03
bone	0.016
skin	0.06
conn.-tissue	0.06
electrode	1

Tab. 1: Conductivities of several types of material.

Depending on their gray values, the pixels inside the contour (skin) represent basic types of tissue like fat, muscle, bone and connecting tissue. The entire 8bit gray scale can be divided into several groups (i.e. segments), where each group stands for a type of tissue. For each group a value for the conductivity γ (shown in *Tab. 1*) has to be defined. For example the segmentation of muscle tissue is shown in *Fig. 1*.

Electrode positioning

To apply electrical stimulation (i.e. the electric field) via surface electrodes, shape and position of electrodes have to be selected. The geometric information of the electrodes must be added to the 2D-superficial surface (skin) and the junction points for anode and cathode of the stimulator have to be defined. This is shown in *Fig. 2*.

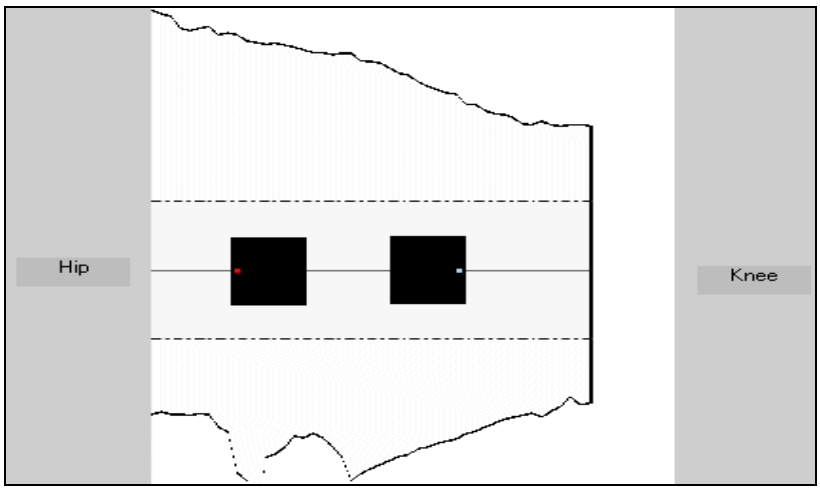


Fig. 2: Skin of human thigh sliced at the bottom side representing a 2D-superficies surface with two rectangular electrodes on top (black). Electrodes should be placed within the gray shaded area bounded by the dash-dotted lines, because adding electrodes outside of this area increases the risk of them loosing their shape when the 2D configuration is attached onto the 3D thigh. The two small squares, placed onto the electrodes represents the junctions to the anode and cathode of the stimulator.

The electric field

After segmentation of tissue has been performed and electrodes and voltage sources have been added into the geometric information of the thigh, all data required for calculating the stationary electric field, i.e. a 3D-matrix of the thigh including surface electrodes, and the correlation between pixel values and conductivities, is available. Describing the voltage distribution in a medium with variable conductivity leads to an elliptic boundary value problem denoting a current density

$$-\nabla[\gamma(x)\cdot\nabla V(x)]=0 \quad \text{with boundary conditions} \quad \begin{matrix} D: V(x)=\varphi(x) \\ N: \dfrac{\partial}{\partial n}V(x)=\gamma(x) \end{matrix}. \qquad \textit{Equ. 1}$$

The coefficient γ is a function, which gives the conductivity depending on the location x, and the function V determines the voltage at x. Dirichlet (D) boundary conditions define voltage sources and Neumann (N) boundary conditions describe the behavior of the field at the crossover from skin to air and at the hip and knee, where the 3D-matrix-representation ends. n denotes the outward pointing normal vector, i.e. the vector orthogonal to the boundary curve. A large system of linear equations (e.g. approx 1.5 million linear equations for 50 cross sections) can be obtained via discretization of *Equ. 1*.

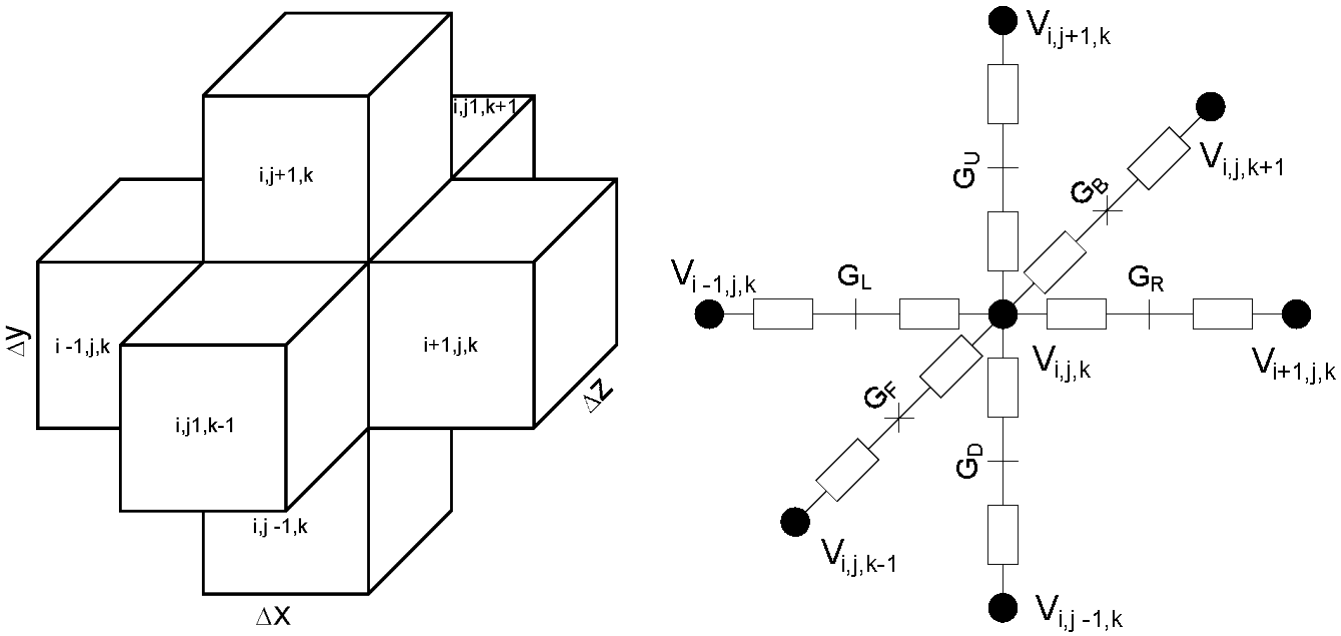


Fig. 3: a) Interdependent voxels in a 3D-grid, equidistant in each direction (Δx, Δy, Δz), neighboring a central voxel-i,j,k and b) corresponding voltages V of each voxel-center connected to the neighboring voxel by the conductance G depending on direction (L…left, R…right, D…Down, U…up, F…front, B…back).

Physical interpretation of *Equ. 1* (i.e. Ohm's and Kirchhoff's law) leads to the system of linear equations

$$\begin{aligned} &G_{Li,j,k}\cdot(V_{i-1,j,k}-V_{i,j,k})+G_{Ri,j,k}\cdot(V_{i+1,j,k}-V_{i,j,k})+G_{Di,j,k}\cdot(V_{i,j-1,k}-V_{i,j,k})+ \\ &G_{Ui,j,k}\cdot(V_{i,j+1,k}-V_{i,j,k})+G_{Fi,j,k}\cdot(V_{i,j,k-1}-V_{i,j,k})+G_{Bi,j,k}\cdot(V_{i,j,k+1}-V_{i,j,k})=0 \end{aligned}, \qquad \textit{Equ. 2}$$

by summing the currents coming from neighboring voxels (*Fig. 3*). Current flows over the conductance G depending on direction (L…left, R…right, D…Down, U…up, F…front, B…back) because of the voltage difference. This system can be solved iteratively using the MATLAB's numerical method of conjugate gradients (CG).

RESULTS

In *Fig. 4* an example shows the stationary electric field calculated by FES-FIELD in one of the cross-sections and in one length-section. The length section was extracted from all cross-sections using the data at the corresponding vertical line.

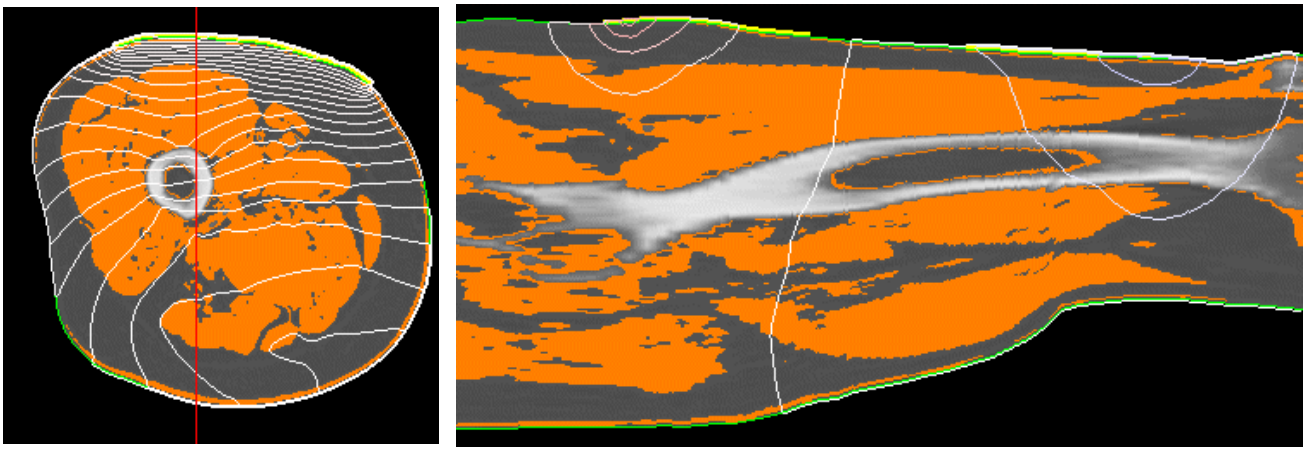


Fig. 4:Cross-section from Fig. 1 (left) and length-section with attached electrodes corresponding to the vertical line in the cross-section (right) both displaying the stationary electric field using equipotential lines.

DISCUSSION

In this research the tool FES-FIELD was implemented to simulate the stationary electric field in the human thigh while applying functional electrical stimulation to paraplegic patients. This tool could be used for other purposes as well. It could be applied directly to the lower leg or the upper extremities and with minor modification also to other regions or organs of the body. One of the initial motivations for developing FES-FIELD was being able to determine voltage distribution along muscle fibers. Using the 3D-matrix giving a voltage value for each pixel, the voltage distribution along any fiber or path can be determined. This process could be automated in a future version.

REFERENCES

/1/ Kern H., Funktionelle Elektrostimulation paraplegischer Patienten, Österr. Z. Phys. Med., 1995, 1, Supplementum

/2/ Kralj A. and Bajd T., *Functional Electrical Stimulation: Standing and Walking after Spinal Cord Injury*, 1989, CRC Press, Inc.

/3/ Reichel M., Mayr W., Rattay F., Computer Simulation of Field Distribution and Excitation of Denervated Muscle Fibers caused by Surface Electrodes, Artificial Organs, 1999, 23(5): 453-456

ACKNOWLEDGEMENTS

Many thanks to Prof. W. Auzinger from the Dep. of Applied and Numerical Mathematics at the Technical University of Vienna for support in numerical mathematics and to Dr. D. Fleischmann from the Dep. of Diagnostic Radiology at the University of Vienna for providing the CT-dataset.

AUTHOR'S ADDRESS

Dr. Martin Reichel
Dep. of Biomedical Engineering & Physics
Währinger Gürtel 18-20, A-1090 Vienna

e-mail: m.reichel@bmtp.akh-wien.ac.at
home page: http://www.bmtp.akh-wien.ac.at/bmt/alllanguages/people/reicma1

GENERATION OF NEW WAVEFORMS TO ACHIEVE ANODAL BLOCK: A COMPUTER STUDY

A. Uranga*, N. Rijkhoff**

* Dept. Enginyeria electrònica, Universitat Autònoma de Barcelona, Spain.
** Center of Sensory-Motor Interaction, Aalborg University, Denmark.

SUMMARY

The main aim of this work is to study different waveforms for selective small fiber activation in order to minimize charge injection when using anodal blocking. In order to performe this study, computer simulations have been done.

STATE OF THE ART

Electrical nerve stimulation is used to restore control of different organs. However, electrical stimulation results in an inverse recruitment order of nerve fibers since large fibers are activated before smaller ones. As a consequence, fast onset of muscle fatigue and poor force gradation are some drawbacks. Moreover, some applications, such as electromicturition and electrodefecation, require activation of small fibers without activation of the larger ones /1/.

Different methods have been proposed to achieve a more natural recruitment order: selective anodal block /2,3/, slowly rising pulses /4/ and high frequency block /5/. We have focussed in this study on the anodal block technique. This technique takes advantage of the different excitation and blocking thresholds for small and large fibers. It is possible with this technique, using a combination of excitation and selective blocking, to activate primarily the small diameter nerve fibers.

In order to obtain anodal blocking, relative large pulse widths and high amplitudes are used compared with a traditional stimulation pattern. This induces a larger charge injection, which could be harmful not only to the tissue but also to the electrode /6/. In the present study new pulse shapes are developed so that injected charge could be reduced. Because the action potential needs time to travel from the cathode to the anode, strong hyperpolarization is not needed at the beginning of the pulse. Pulse shapes with two different amplitude levels (a smaller amplitude at the beginning of the pulse) and with an increasing slope, have been simulated in a computer model and the reduction in the amount of charge injected has been evaluated. In addition, the effect of the use of a hyperpolarizing prepulse on the excitation and blocking thresholds has been analysed. Parameters such as hyperpolarizing pulse amplitude, pulse width and delay have been optimised to obtain maximum charge reduction.

MATERIAL AND METHODS

The electrical potential field generated by a cuff electrode (2 mm inner diameter) with metal ring contacts, spaced 3 mm, has been calculated using a volume conductor model described by Rijkhoff et al./7/. A nerve bundle with a radius of 0.7 mm has been used.

A nerve fiber model used to analyze the responses of the membrane to the extracellular electrical potential field is described by McNeal /8/. The Frankenhaeuser-Huxley equations, adapted for a rabbit according to Chiu et al. /9/, have been used to describe the membrane kinetics. All temperature-dependent parameters were scaled to 37 ºC /10/. Two different fiber diameters have been analyzed to study the

selectivity (12 µm and 4 µm). The 12 µm fiber has been placed on the axis of the bundle and the 4 µm has been placed at the border. The cathode was always situated right above the central node of Ranvier of the fiber.

Parametric simulations have been done in order to investigate the different effects. The pulse shape was optimized so that it would result in a minimum charge injection, allowing a faster recovery of charge (with the subsequent possibility of increasing the signal frequency) and a safer stimulation pattern, able to generate fiber diameter selectivity.

RESULTS

A study of the influence of the waveform on the charge per pulse needed to block a 12 µm fiber situated on the axis has been done. Two different waveform shapes have been used (see Fig. 1).

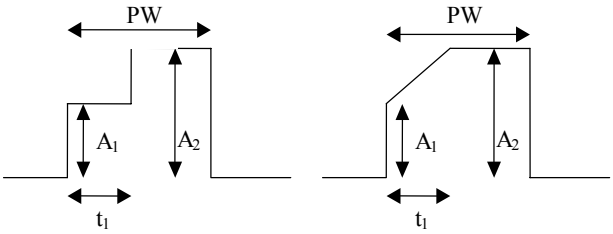
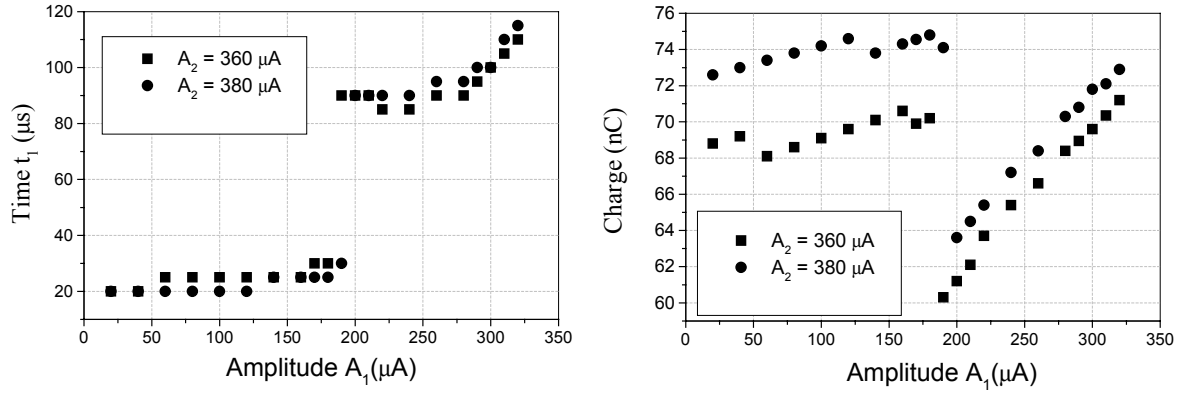


Fig. 1 *: Stimulus waveforms for anodal blocking (PW = 210* µs) .

A rectangular pulse with an A_1 amplitude during t_1 followed by an amplitude A_2 during the rest of the pulse has been applied. Fig. 2 shows the relationship between the initial amplitude (A_1) and the maximum time (t_1), for two fixed amplitudes A_2, in order to block the action potential generated under the cathode, along with the charge injected. It is shown how an increase in the amplitude A_1 allows an increase in the time t_1 since a bigger hyperpolarization is generated under the anode. If a rectangular pulse with the same total width (210 µs) is used, an amplitude of 336 µA is needed to block the 12 µm fiber situated on the axis. This pulse injects a charge of 70.5 nQ. By analyzing and comparing the charge injected between the pulses, it is shown how up to a 13% charge reduction can be achieved by using this new shape.



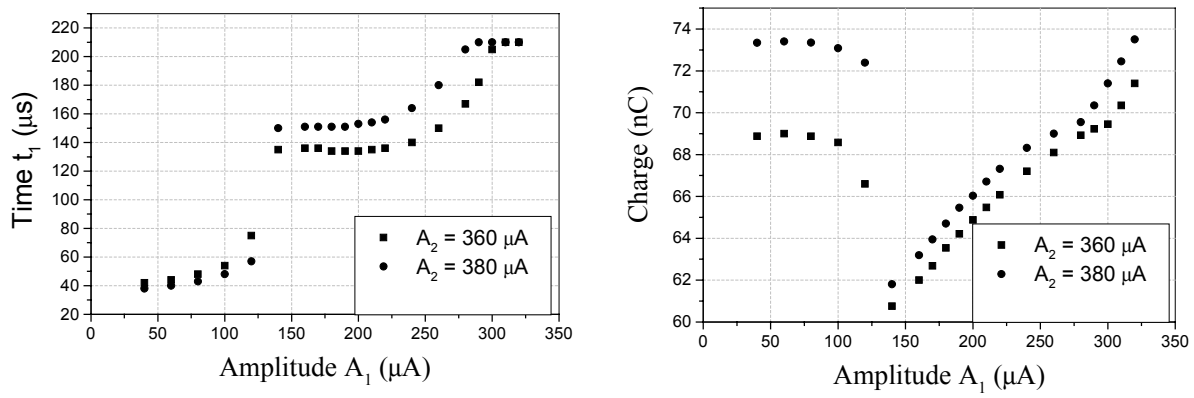
(a) (b)

Fig. 2: (a) *Time t_1 as a function of the amplitude A_1 and (b) total charge injected when stimulus waveform of fig. 1 (left) is being used to block the 12 µm fiber.*

In order to increase the time t_1 and thus, decrease charge injection, a bigger hyperpolarization is needed. The first part of the waveform has been replaced with a linear increasing amplitude starting at A_1 which reaches amplitude A_2 after t_1 µs (see Fig. 1 right). The value A_2 is maintained during the rest of the pulse.

Figure 3 (a) shows the results of the performed simulations. Again, an increase in the initial amplitude allows to have a bigger time t_1. However, comparing the total amount of charge injected (Fig. 3 b),

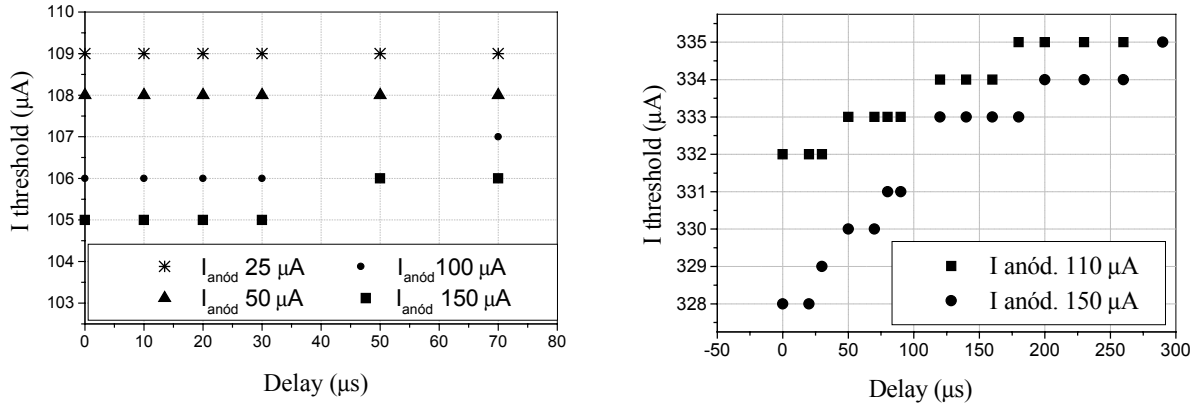
similar values to the ones achieved by means of the use of the previous waveform are found; additionally, a reduction of charge, related to the rectangular pulse, is obtained.



(a) (b)

Fig. 3: (a) *Time t_1 as a function of the amplitude A_1 and (b) total charge injected when stimulus waveform of fig. 1 (right) is being used to block the 12 µm fiber.*

The influence of a hyperpolarizing prepulse on the excitation and blocking threshold of a 12 um fiber situated on the axes of the bundle has been investigated. Fig. 4 (a) represents the effect of a square anodic pulse (pulsewidth: 210 µs) on the excitation threshold versus the delay between the anodic pulse and the excitation pulse, for different anodic pulse amplitudes. Threshold excitation amplitude without prepulse is 110 µA. It shows that an increase in amplitude of the hyperpolarizing pulse results in a decrease in the activation threshold. However, the amplitude of the anodic pulse is limited because during hyperpolarizing prepulses, the anodes become cathodes and, a large enough current excites the fiber. On the other hand, an increase in the delay will bring the threshold to the initial value (threshold without prepulse), indicating that the membrane has recovered to its initial state.



(a) (b)

Fig. 4: *(a) Excitation threshold amplitude and (b) blocking threshold amplitude versus delay between the anodic pulse and the normal pulse of a 12 µm fiber situated on the axes of the bundle(perpulse and excitation pulse witdh: 210 µs)*

The effect of the hyperpolarizing prepulses on the blocking threshold of a 12 µm fiber is shown in fig. 4b. Two different anodic currents pulses, both with a pulse width of 210 us, have been analysed. In both cases, a decrease in the threshold, in relation to the waveform without prepulse (336 µA), has been observed. It is shown how an increase in the amplitude of the cathodic pulse decreases the threshold of blocking. However, an increase in the delay will lead the membrane to its original state, producing an increase in the blocking threshold.

From the results of the previous analysis, a hyperpolarizing prepulse can be used to reduce the excitation and blocking threshold. As a consequence, the prepulse reduces the charge injected during the cathodical

pulse and the total amount of charge injected because of the charge extraction that is produced during the anodical pulse. This charge reduction will allow us to work with higher stimulation frequencies since less charge is needed to extract.

DISCUSSION

Based on this study we conclude that anodal blocking can be obtained with new waveforms, which allow for a reduced charge injection. In addition to this pulse modification, the influence of the hyperpolarizing prepulse in the activation and blocking thresholds has been studied. Besides the threshold modification, the hyperpolarizing prepulse generates a previous charge extraction just before the injection, reducing the charge that needs to be extracted after the stimulation. The use of these shapes will be safer in chronic applications of anodal blocking.

REFERENCES

/1/ N. J. M. Rijkhoff, H. Wijkstra, P. E. V. Van Kerrebroeck, F. M. J. Debruyne, Urinary bladder control by electrical stimulation. Review of electrical stimulation techniques in spinal cord injury, Neurourol. & Urodyn., vol. 16, pp. 39-53, 1996.
/2/ G. S. Brindley and M.D. Craggs, A technique for anodally blocking large nerve fibers through chronically implanted electrodes, J. of Neurol. Neurosurg. and Psychiatry, vol. 43, pp. 1083-1090, 1980
/3/ N. J. M. Rijkhoff, E. L. Koldewijn, P. E. V. van Kerrebroeck, F. M. J. Debruyne, and H. Wijkstra, Acute animal studies on the use of anodal block to reduce uretral resistance in sacral root stimulation, IEEE Tran. Biomed. Eng., vol. 41, pp. 413-424, 1994.
/4/ Warren M. Grill and J. Thomas Mortimer, Inversion of the current-distance relationship by transient depolarization, IEEE Tran. on BME., vol. 44, no. 1, pp. 1-9, 1997.
/5/ Bruce R. Bowman, Donald R. McNeal, Response of single alpha motoneurons to high-frequency pulse trains. Firing behavior and conduction block phenomenon, Appl. Neurophysiol, 49, pp. 121-138, 1986.
/6/ S. B. Brummer and M. J. Turner, Electrochemical considerations for safe electrical stimulation of the nervous system with platinum electrodes, IEEE Tran. BME., pp. 59-63, 1997.
/7/ N. J. M. Rijkhoff, J. Holsheimer, E. L. Koldewijn, J. J. Struijk, P. E. V. van Kerrebroeck, F. M. J. Debruyne and H. Wijkstra, Selective stimulation of sacral nerve roots for bladder control: A study by computer modeling, IEEE Tran. BME, vol. 41, n 5, pp. 413-424, 1994.
/8/ D. R. McNeal, Analysis of a model for excitation of myelinated nerve, IEEE Tran. BME, vol. BME-23, pp. 329-337, 1976.
/9/ S. Y. Chiu, J, M. Ritchie, R. B. Rogart and D. Stagg, A quantitative description of membrane currents in rabbit myelinated nerve, J. of Physiol., vol 292, pp. 149-166, 1979
/10/ J. J. Struijk, J. Holsheimer, G.G. van der Heide and H. B. K. Boom, Recruitment of dorsal column fibers in spinal cord stimulation: Influence of collateral branching, IEEE Tran. BME, vol. 39, n 9, pp. 903-912, 1992.

ACKNOWLEDGEMENTS

This work has been realized in the SMI center (Aalborg University, Denmark), supported in part by the CICYT (under project number TIC 2000-1398) and funded by a grant from the Comissionat per a Universitats i Recerca (Generalitat de Catalunya, Spain)

AUTHOR'S ADDRESS

Arantxa Uranga del Monte
Dept. Enginyeria Electrònica.
Escola Tècnica Superior d'Enginyeria Electrònica.
Universitat Autònoma de Barcelona.
08193 Bellaterra, Barcelona, Spain.
e-mail: arantxa.uranga@uab.es

AUTOMATED ESTIMATION OF STIMULATION THRESHOLDS USING SINGLE TWICHES AND ACCELERATION SENSORS

S. Sauermann, M. Bijak, M. Reichel, D. Rafolt, E. Unger, W. Mayr, H. Lanmüller

Department of Biomedical Engineering & Physics
University of Vienna, Vienna, Austria

SUMMARY

In Functional Electrostimulation (FES) the relation between stimulation input and muscle force output is often called the “recruitment curve”. Especially with surface stimulation systems the properties and position of stimulation electrodes have strong influence on the recruitment curve. In the clinical application it has therefore to be established at every session and even within sessions. This estimation process is of clinical relevance as it takes time and effort which would be better invested into optimisation of the stimulation outcome.

This work was therefore aimed at a method for automated estimation of the stimulation threshold in the clinical application of FES systems. It uses low frequency ramp-like bursts stimulation with increasing amplitude. An acceleration sensor is used to measure the resulting muscle twitches. A signal analysis algorithm then evaluates the resulting spikes within the acceleration signal. A test on the quadriceps femoris muscle of 21 healthy testpersons was performed.

The algorithm estimated the number of spikes correctly in 15 of 21 trials. It missed a single spikes in two cases and detected one wrongly in one case. In the remaining three cases the number of spikes was wrong by more than one. The automated measurement equipment proved very useful in speeding up the measurement process.

STATE OF THE ART

There have been numerous approaches to measure the recruitment characteristic of FES Systems with the help of automated procedures in order to ease and shorten the process of setting up stimulation parameters.

Kilgore and Peckham [Kilgore et.al. 1993] have used the "External Moments Grasp Synthesis Procedure" which is based on measurements of the joint moments under varying joint angles and stimulation parameters, and measurements of the passive joint moments without stimulation [Kilgore et.al. 1993]. This results in a "stimulus map" which describes the stimulation parameters necessary for a certain hand grip movement task.

Other studies used the simpler "rule based" approach for control of FES. In a rule based procedure the stimulation thresholds and saturation values are established, and the necessary stimulation strength is expressed in relation to these values [Kilgore et.al. 1989].

Muscular dynamics has been described by a so called Hammerstein structure which consists of a static nonlinearity followed by a linear dynamic system [Durfee et.al. 1989]. This concept has been used in many studies of open- and closed-loop control of electrically stimulated muscle. The static nonlinearity

is the model for the "recruitment curve" that describes the relation between strength of a stimulus activation and muscle output. This model was also used in this work.

Valencic has described a measurement system for muscular dynamics based on a displacement sensor placed above the belly of the muscle [Valencic et.al. 1997]. This mechanical approach has the advantage that stimulation artefacts are avoided. Nevertheless sensors for force and displacement need an external reference, which complicates the setup. In this work therefore an acceleration sensor was used in order to make the system smaller and easy to use.

The aim was to develop a fast and simple procedure for estimating the stimulation threshold in clinical practice as a first step for assessment of the complete recruitment curve. The following criteria were to be met: (1) The estimation procedure must be based on a single measurement in order to reduce the time needed for measurements, to keep muscle fatigue low and to reduce discomfort for the patient. Single impulses are preferred. (2) The stimulation and measurement procedures have to be performed synchronously and automatically without any need of inputs from the operators. (3) All data must be stored automatically. The output of the estimation is to be fed back to the stimulation control scaled in stimulation parameters. (4) The estimated stimulation thresholds are intended as an initial starting point for optimisation by the operator.

MATERIAL AND METHODS

Experimental Setup

Acceleration signals were measured during surface electric stimulation of the quadriceps femoris muscle. Measurements were performed using the FESDaq measurement system [Sauermann 1999]. Stimulation was delivered by the stimulation system developed in Vienna [Bijak et.al. 1999]. Each stimulation burst consisted of 10 consecutive biphasic impulses of 150+150 µs impulse duration delivered at a frequency of 3 Hz. The amplitude increased linearly from a 5 to 40 V.

The acceleration sensor (EGAS-FS-5, Entran Sensoren GmbH, Ludwigshafen, Germany) was attached to the skin above the bulk of the muscle with double stick tape. The acceleration was measured perpendicular to the skin. The number of muscle twitches that were observed visually by the operators and the number of stimulation impulses that could be perceived sensually by the test person were noted in each measurement.

Signal Analysis

The analysis algorithm basically identifies and counts spikes within the measured signal. Spikes below a given signal-to-noise level are discarded, as well as spikes out of a given time window around the stimulation period. Together with the known number of impulses and the stimulation amplitude at the start and the end of the stimulation ramp an amplitude interval is then calculated, that contains the stimulation threshold. The number of spikes found by the algorithm was compared to that found in the acceleration signal by an experienced observer in order to estimate the accuracy of the procedure.

RESULTS

Fig. 1 shown an acceleration signal resulting from the measurements. The algorithm estimated the number of spikes correctly in 15 of 21 trials. It missed a single spike in two cases and detected one wrongly in one case. In the remaining three cases the number of spikes was wrong by more than one. There are three causes of error: (1) spikes are lost in noise as marked in Fig. 2. (2) The decay curve of the last spike is mistakenly classified as an additional spike, as marked in Fig. 2. (3) Due to non-linear signal dynamics the shape of the spikes changes, as in Fig. 1. Therefore peaks appear outside the allowed time window.

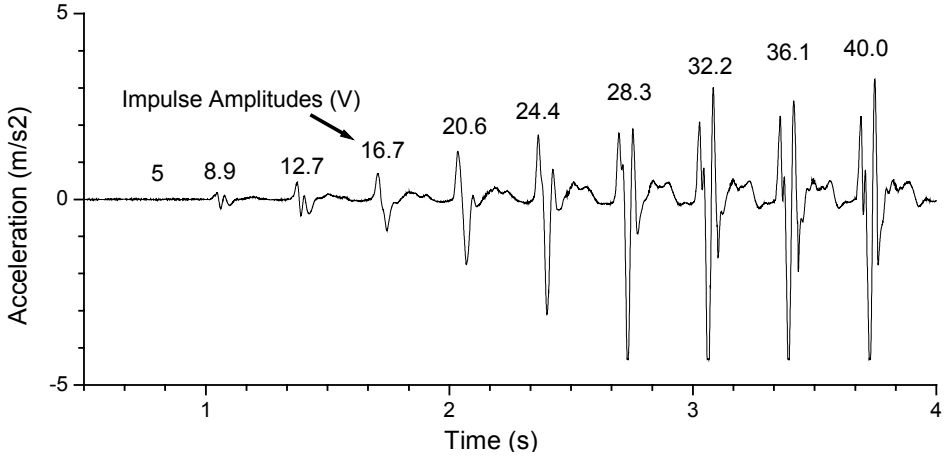


Fig. 1: Acceleration of the skin above the bulk of the quadriceps femoris muscle during FES (10 biphasic impulses of 150+150 µs duration at 3 Hz, linear amplitude ramp from 5 to 40 V). The first impulse (5 V) did not cause a detectable spike, therefore the stimulation threshold was estimated to be between 5 and 8.5 V. Due to a marked non-linear behaviour of the signal dynamics, the second peak of the spike becomes higher than the first one at stimulation amplitudes >24.4 V.

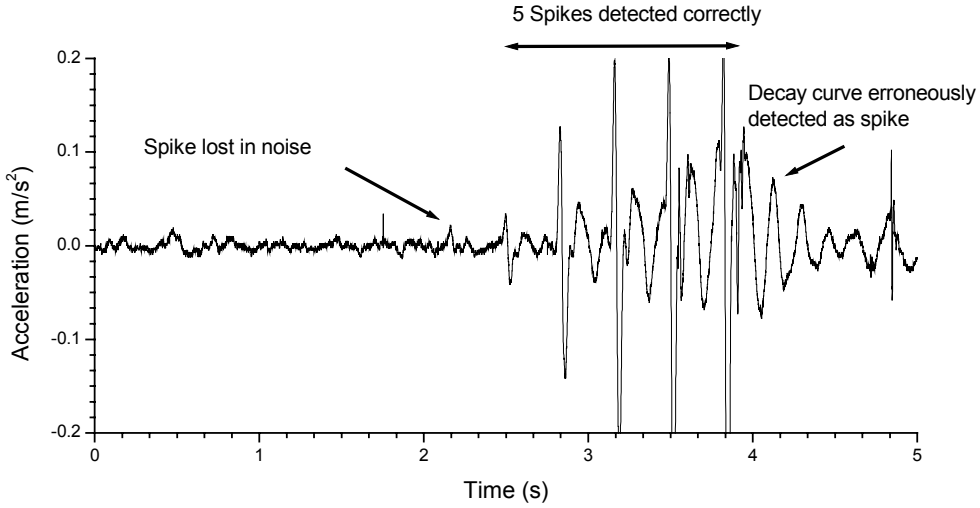


Fig. 2: Acceleration measured as in Fig. 1. Erroneous detections of spikes: one spike is lost in signal noise, one peak of the decay curve of the last spike is mistakenly classified as a separate spike.

All testpersons perceived all stimulation impulses as a tingling sensation on the skin beneath the stimulation electrodes, even at the minimum stimulation amplitudes of 5 V. The number of impulses that were observed visually was not assessed properly in all cases because the operators were not able to count as fast as would be necessary. Repeated stimulation or a video camera would have been necessary

to get reliable results. The number of visually perceived muscle twitches was therefore not included in the analysis.

DISCUSSION

During the experiments the automated measurement equipment proved very useful in speeding up the measurement process. The acceleration sensors provided signals of sufficient quality at a reasonable effort for setting them up. The graphical display of the acceleration signal was useful for detecting the stimulation threshold, and more reliable than visual perception: The signal may be reviewed in detail on the screen, with much more time and ease than during the very brief period of the stimulation.
Further work will now be directed at more sophisticated signal analysis based on the dataset of signals harvested in this trial. Especially the non-linear behaviour of the signal dynamics of the acceleration signal has to be dealt with. As regards sensing methods, the necessary amount of cabling has to be reduced significantly and the process of setting up the sensor system has to be eased. It will also be of primary concern to establish the accuracy of the recruitment curve which is necessary to achieve satisfying stimulation outcome with the intended applications.

REFERENCES

/1/ Kilgore, K. L. and Peckham, P. H.: Grasp synthesis for upper-extremity FNS. Part 1. Automated method for synthesising the stimulus map. Med.Biol.Eng Comput., 31, (6), 1993, 607-614.
/2/ Kilgore, K. L., Peckham, P. H., Thrope, G. B., Keith, M. W., and Gallaher-Stone, K. A.: Synthesis of hand grasp using functional neuromuscular stimulation. IEEE Trans.Biomed.Eng, 36, (7), 1989, 761-770.
/3/ Durfee, W. K. and MacLean, K. E.: Methods for estimating isometric recruitment curves of electrically stimulated muscle. IEEE Trans.Biomed.Eng, 36, (7), 1989, 654-667.
/4/ Valencic, V. and Knez, N.: Measuring of skeletal muscles' dynamic properties. Artif.Organs, 21, (3), 1997, 240-242 .
/5/ Sauermann, S.: Computer Based Data Acquisition and Online Analysis in Functional Electrostimulation. Pages 902-903, in Vol.37, Suppl.2, Proceedings of the European Medical & Biological Engineering Conference, Part II, held on November 4-7,1999, Vienna, Austria, 1999.
/6/ Bijak, M., Hofer, C., Lanmuller, H., Mayr, W., Sauermann, S., Unger, E., and Kern, H.: Personal computer supported eight channel surface stimulator for paraplegic walking: first results. Artif.Organs, 23, (5), 1999, 424-427.

AUTHOR'S ADDRESS

Stefan Sauermann
Department of Biomedical Engineering & Physics
General Hospital of Vienna
Waehringer Gürtel 18-20 / 4L
A-1090 Vienna

e-mail: st.sauermann@bmtp.akh-wien.ac.at
http://www.bmtp.akh-wien.ac.at/bmt/e/home.html

FIRST INSIGHTS ON MUSCLE AFFERENT NERVE SIGNALS FOR CLOSED-LOOP CONTROL OF FES-GENERATED RABBIT ANKLE MOVEMENTS

F. Sepulveda, W. Jensen, T. Sinkjaer

Center for Sensory-Motor Interaction (SMI), Aalborg University, Denmark

SUMMARY

Control of FES-based movements can be enhanced by proper estimation of the current dynamic state of the motion. Some of this dynamic information can be obtained from muscle afferent fibers. To this end a simple approach was recently taken by our group whereby rectified and bin-integrated (RBIN) electroneurographic (ENG) signals were previously found to allow for a reasonable mapping onto angular and torque data by means of neural and fuzzy models. This paper presents our first findings from using the extracted angular information in a closed-loop scheme. Acute experiments were conducted with rabbits. For extraction of the ENG signals, tripolar cuff electrodes were implanted onto the peroneal and tibial nerves in the left hind limb. A neural network was used for extraction of joint angles from the ENGs. For stimulation purposes, percutaneous stainless steel wires were placed intramuscularly into the tibialis anterior and lateral gastrocnemius muscles, respectively. FES intensity was varied by changing the applied pulse width (PW). Step and sinusoidal tracking tasks were performed using a standard PID controller. Best results were obtained when starting with the ankle joint at a neutral, rest angle. Angles estimated from the ENG were found to lose correlation with measured angles as a given experiment progressed. Improvements were seen when the neural network was allowed to learn intermittently during an experimental session. Further, a standard PID controller also required frequent retuning during a session, which, not surprisingly, suggests that an adaptive controller should be used instead.

STATE OF THE ART

Control of FES-based movements can be greatly enhanced by real-time estimation of the generated dynamics. For example, in the case of a subject walking with the aid of FES, it is important that we use reliable sensory information pertaining to the generated angular trajectories. It is possible to obtain this information from muscle afferent fibers. This is due to the fact that muscle afferent fibers carry signals proportional to muscle length, length changes, and tension. The relationship between a muscle's length and an associated joint's angle is not linear, but a proper combination of feature extraction and nonlinear modeling may allow us to extract the desired angular data. To this end, our research group has been using implanted cuff electrodes to extract joint angular information by simple analysis of the recorded nerve signals /1//2//3/: rectified and bin integrated (RBIN) electroneurographic (ENG) signals have been recorded and have been found to allow for a reasonable mapping onto angular data by means of neural and fuzzy models /1//2/. The present paper discusses our findings from the first attempts at using the extracted angular information in a closed-loop FES controller scheme.

MATERIALS AND METHODS

Experimental Setup

Acute experiments were conducted with 4 female New Zealand rabbits. The rabbits were pre-anesthetized with an injection of Midazolam (2.0 mg/kg; Dormicum™, Alpharma, Norway). Then, after 15 to 20 min, anesthesia was initiated by an injection of Hypnorm™ (0.095 mg/kg Fetanyl +

3.0 mg/kg Fluranison; Janssen Pharmaceutica, Belgium). The anesthesia was maintained by applying intramuscular injections of Dormicum™ (0.15 mg/kg Midazolam), and Hypnorm™ (0.03 mg/kg Fetanyl + 1.0 mg/kg Fluranison) every 20 min. All procedures were previously approved by the Danish Committee for the Ethical Use of Animals in Research. During the experiments the rabbits were placed onto a mechanical device for fixating the knee and ankle joints in place /3/.

For extracting the ENG signals, tripolar cuff electrodes were implanted onto the Peroneal and Tibial branches of the sciatic nerve in the left hind limb. The Tibial and Peroneal nerves were transected just above the ankle joint to minimize sensory inputs from the foot. Further, to minimize cutaneous inputs, the sural nerve was transected distal to its origin in the Tibial nerve. The internal diameters for the cuff electrodes were 2 mm for the Tibial nerve and 1.8 mm for the Peroneal nerve. The cuff length was 22 mm in both cases (10 mm between each contact zone). The cuff electrodes were manufactured according to the procedure described in /4/ but, in this case, a longitudinal cut was made to open the cuff. The ENG signals were sampled at 10kHz after being submitted to a bandpass analog filter (1st order Butterworth, 500Hz to 2kHz bandpass). To eliminate stimulation artifacts when FES was applied, ENG was recorded only for 4ms before each stimulation pulse /5/. A neural network (trained off-line) was used for extraction of joint angles from the ENGs. Inputs for the neural network were the mean ENG values over the above 4ms period. The neural network for angle extraction had 4 input neurons (1 RBIN tibial ENG channel, 1 RBIN peroneal ENG channel, plus one delay for each RBIN channel), 20 hidden layer neurons, and one output neuron whose output was proportional to the predicted angle. The neural network also had an Elman-type recursive connection for each hidden and output layer neuron.

For stimulation purposes, percutaneous stainless steel wires were placed intramuscularly into the tibialis anterior and lateral gastrocnemius muscles, respectively, in the same hind limb as the cuff electrodes. A constant current stimulator was used with the output set to 5mA. Stimulation intensity was varied by the controller through changes in the applied pulse width (PW) (from 0 to 500 μs, in 10μs steps).

Closed-Loop Control Test

Step and sinusoidal tracking tasks were performed using a standard PID controller. An overview of the closed-lop system using ENG is shown in Fig. 1. Further, an optical angle sensor was used to measure the real ankle joint angles as a function of time. The readings from the optical sensors were used for monitoring both the controllers performance and the adequacy of the angle predictions from the neural network. The optical angle transducer was also used as feedback for testing the controller in isolation from the ENG feedback.

RESULTS AND DISCUSSION

Isolated Controller Test

First we ran several tests with an optical angle sensor that bypassed the ENG and angle extraction routines in order to verify the best possible control performance to be expected from the system shown in Fig. 1. Due to the errors in angle predictions inherent in the ENG-plus-neural network approach, the use of ENG for extraction of the angles is expected to yield worse performance than when real angular data are used. Thus, real angular data were used as feedback for tuning the PID controller and for estimating its best possible performance. The controller was tuned using the Ziegler-Nichols method.

The controller's performance was found to rapidly degrade due mainly to the time-variance of the muscles' response to stimulation as a given controller test progressed. Thus, continuos retuning of the controller was required during all the experiments. However, the process of retuning required the application of FES to the muscles, which accelerated the onset of muscle fatigue. Thus, it is suggested that an adaptive controller be used in the future.

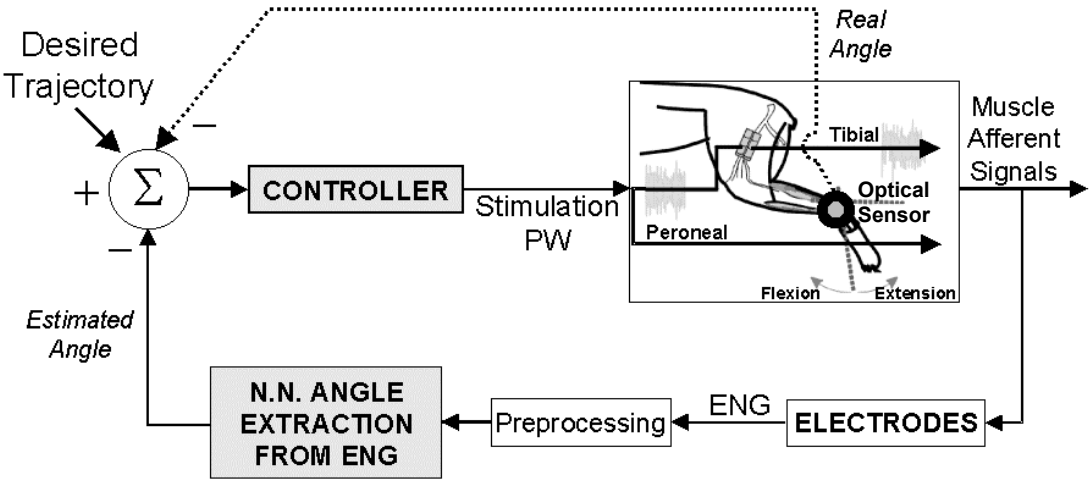


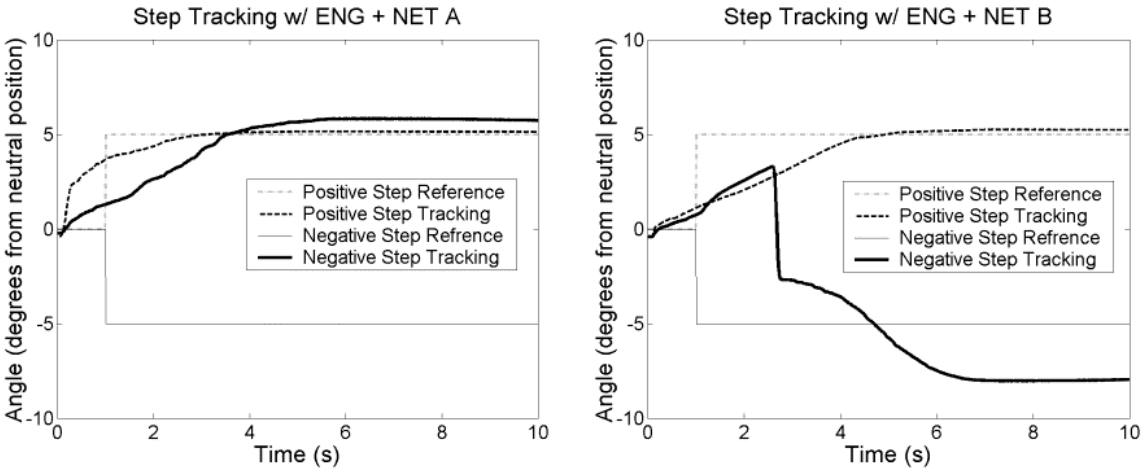
Fig.1 – Feedback control of FES-based rabbit ankle motion using ENGs from muscle afferent fibers. The figure also includes an optic angle transducer that was used for obtaining real-time angular data both for later comparison with values extracted from the ENGs and for testing the controller isolated from the ENG feedback loop.

Closed-Loop Control with ENG Feedback

According to preliminary results, the system's performance was highly sensitive to the initial joint angle. The best results were obtained when starting with the ankle at a neutral, rest angle. This value was found to be between 100^{o} and 120^{o} for all rabbits after they were placed onto the mechanical support unit. Control was first done with a neural network trained only with data from previous experiments (see /1/). Closed-loop performance with this network (NET A, Fig. 2a) was found to be very poor. The poor performance was found to be due mostly to very noisy angle predictions from NET A. Significant improvements were found when a sliding 8-point average window was applied to the network's output. Further, angles estimated from the ENG by the neural network were found to lose correlation with measured angles as a given experimental session progressed. Improvements were seen when the neural network was allowed to learn intermittently during an experimental session (leading to NET B). In this case, the closed-loop process was stopped for 15 min. During this pause the neural network was allowed to learn data from the rabbit being used in the specific experiment. Although the system's performance was still far from acceptable (long tracking delays and large final offsets are still observed, Fig. 2b), this neural network tuning process was found to yield substantial improvements.

CONCLUSIONS

A standard PID controller required frequent retuning during an experimental session. Thus, it is suggested that an adaptive controller be used in the future. Best results were obtained when starting with the ankle joint at a neutral, rest angle. Control with a neural network trained solely on data from other rabbits yielded poor performance. Improvements were seen when neural network outputs were smoothened and the network was allowed to learn intermittently during an experimental session. Finally, angular predictions from ENG were still found to be unreliable for closed-loop when the procedures described here are used.



(a) (b)

Fig. 2 – Improvement in control performance using ENG. (a) NET A: original neural network trained solely on data from previous experiments. (b) NET B: NET A submitted to output smoothing and rabbit-specific tuning. Data shown are for a single rabbit.

ACKNOWLEDGMENTS

This work has been supported by the Danish National Research Foundation. The authors would like to thank the staff at the Animal Laboratory in the Pathology Institute, Aalborg North Hospital, for their important help in this work.

REFERENCES

/1/ W. Jensen, R. Riso, and F. Sepulveda 'On-line joint angle estimation based on nerve cuff recordings from muscle afferents'. Proceedings of the *2000 Congress of The International FES Society*, Aalborg, Denmark, pp. 376-377, 2000.

/2/ S. Micera, W. Jensen, F. Sepulveda, R. Riso, and T. Sinkjaer 'Neurofuzzy extraction of angular information from muscle afferents for ankle control during standing is paraplegic subjects: an animal model'. *IEEE Trans. Biom. Eng*, pp. 787-794, July, 2001.

/3/ W. Jensen, R. Riso, and T. Sinkjaer 'Effect of intertrial delay on whole nerve cuff recordings of muscle afferents in rabbits'. *Neuromodulation*, vol. 3(1) pp. 43-53, 2000.

/4/ M. Haugland 'A flexible method for fabrication of nerve cuff electrodes.' Proceedings of the *18th Annual Conference of the IEEE Engineering in Medicine and Biology Society*, Amsterdam, pp. 359-360, 1997.

/5/ M. Haugland and A.J. Hoffer 'Artifact-free sensory nerve signals obtained from cuff electrodes during functional electrical stimulation of nearby muscles'. *IEEE Trans. Rehab. Eng.*, vol. 2(1), pp. 37-40, 1994.

ADDRESS FOR CORRESPONDANCE

F. Sepulveda, Ph.D. e-mail: *frs@smi.auc.dk*
Center for Sensory-Motor Interaction (SMI)
Aalborg University
Fredrik Bajers Vej 7-D3
DK 9220, Aalborg
DENMARK

SELF-TUNING REGULATION OF MUSCLE GENERATED MOMENT INDUCED BY ELECTRICAL STIMULATION

M. Ponikvar, M. Munih

Faculty of Electrical Engineering,
University of Ljubljana, Slovenia

SUMMARY

This paper presents the development, implementation, and experimental evaluation of adaptive feedback system for control of the muscle generated moment. The control system, which utilizes self-tuning regulator (STR) and real-time estimation of local muscle model parameters triggered the stimulator and muscle contractions. Design of such moment controller represents the base stage leading toward the construction of a position controller. In order to simplify in vivo experiments with FES, we focused only on gastrosoleus muscles under isometric conditions.

The stimulated subject stood in mechanical rotating frame (MRF), which prevented rotation of knee, hip and lumbosacral joints. MRF is locked, meaning that also ankle joint is stiff. The moment in the ankle as output of the standing human was measured separately for left and right foot with two force plates. The STR was then used to automatize the muscle model identification and the calculation of controller parameters. The fidelity of muscle models was estimated on-line with least square recursive method, which enables the implementation of adaptive control. Adaptive control of electrically stimulated muscle can in real-time embrace the changes caused by time-varying muscle behavior such as fatigue. On-line identification does not require any advance identification procedures and stimulation. The control law was based on the pole placement design that gives desired closed-loop poles. The STR was realized with program Matlab Simulink by using program blocks for identification part, controller, force plates and computer controlled electrical stimulator. The linear controller operation was then tested in three activation regions between stimulation threshold and saturation to exclude nonlinear activation effects. For all three activation regions were utilized different controller configurations with advance adjustment of tuning parameters that indirectly reflect the desired closed-loop transfer function. Sinusoidally shaped moment trajectories included oscillation frequencies between 0.1 Hz and 1.5 Hz. The muscles were stimulated with controlled repetition pulses at 20 Hz. The controller tracking demonstrated to be satisfactory, however due to simple adaptive constitution not enough robust to attenuate well larger disturbances.

STATE OF THE ART

We are interested in studying of the control of paraplegic standing, including standing up and sitting down, by using closed-loop functional electrical stimulation (FES). The method used by nature to enable precise positioning of body parts and achieve body stability is to simultaneously control many synergistic muscles with feedforward and feedback commands. To efficiently investigate the artificial control we want to simplify the demanding body stabilisation conditions and avoid the problem of muscle redundancy. For this reason we try to restrict the arbitrary body motion with the MRF and focus on ankle plantarflexor muscles only. In this study is presented simple single-input single-output controller of the FES generated muscle moment that accounts for slow changes of muscle properties.

MATERIAL AND METHODS

Development of a control system involves many tasks such as modeling, design of a control law, implementation, and validation. The STR attempts to automate several of these tasks /1/. This is illustrated in Fig. 1, which shows a Matlab Simulink block diagram of a process with a STR.

There are many possible choices of model and controller structures. In our study was the process, electrically stimulated muscle, presented with a linear second order discrete transfer function with a pure time delay z^{-1} /2/:

$$G_p(z^{-1}) = \frac{b_1 z^{-1} + b_2 z^{-2}}{1 + a_1 z^{-1} + a_2 z^{-2}} \tag{1}$$

The model sampling time was fixed at 0.05 s, which was also the stimulation rate. The linear model is only valid for a limited region of stimulation levels and was estimated with recursive least squares identification method (RARX) /3/. A simple pole placement method was selected to define a discrete-time domain controller that gives desired closed-loop poles /1/. In addition it is required that the system follows the reference signal u_c in a specified manner.

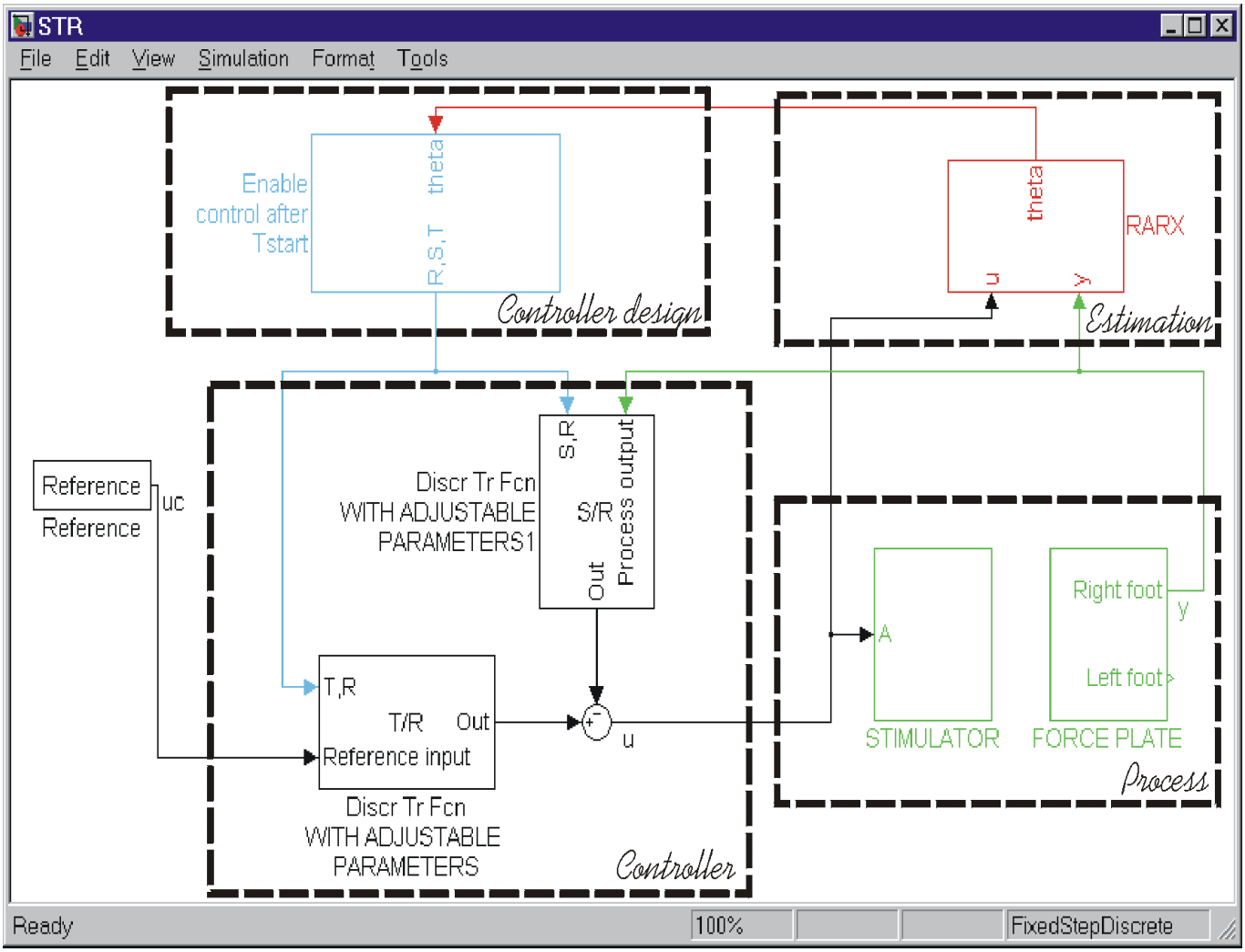


Fig. 1. Matlab Simulink diagram of an adaptive control scheme for reference moment tracking of electrically stimulated plantarflexors. u_c is the reference moment, u is the control signal and y denotes the measured moment. theta is the vector of identified process parameters and R, S, T are the controller polynomials.

The controller was described by

$$Ru(t) = Tu_c(t) - Sy(t) \tag{2}$$

where R, S, and T are polynomials, u is the muscle stimulation level and y is generated moment. This control law represents a negative feedback with the transfer operator $-S/R$ and a feedforward with the transfer operator T/R. The controller was utilized by using two custom discrete transfer functions with adjustable parameters as is shown in Fig. 1, *Controller*. After the model (1) parameters $theta = [a_1, a_2, b_1, b_2, mean\_moment]$ were identified, was the control law defined according to pole placement controller specifications.

The pole placement procedure for reference model-following was designed to operate without cancellation of the process model zero to avoid unstable operation in the case of biased model parameter estimates. Since the process model is of second order, the minimum-degree solution has polynomials *R, S,* and *T* of first order and the closed-loop system is of third order /1/. The reference model, which implements the controller specifications thus needs to be a third order transfer function.

RESULTS

The described controller was tested in moment tracking experiments with intact subjects. The moment reference signals u_c were sinusoidally shaped and included oscillation frequencies between 0.1 Hz and 1.5 Hz, Fig. 2. The effect of varying the parameters of reference model was studied for signals around three output levels: 37.5 %, 62.5 % and 87.5 % where the 100 % stands for the generated moment at maximal (saturation) stimulation level. The sine amplitude was app. 10 % of the maximal generated moment measured in advance tests. At all stimulation levels were achieved the best tracking results if the selected reference model (controller specifications) was a second order transfer function with equivalent damping factor of 0.8 and natural frequency 10 rad^{-1}s. The initial values of the process parameters were $theta = [0,0,0,0, anticipated\_mean\_moment]$ and reached the stationary values after app. 1.5 s, Fig 3. The self-tuning of the controller was therefore started after 1 s of stimulation control with nonadaptive poleplacement controller. The control signal (stimulation level) oscillations arose due to the identified parameter oscillations and due to the initial commutation between constant and adaptive controller parameters.

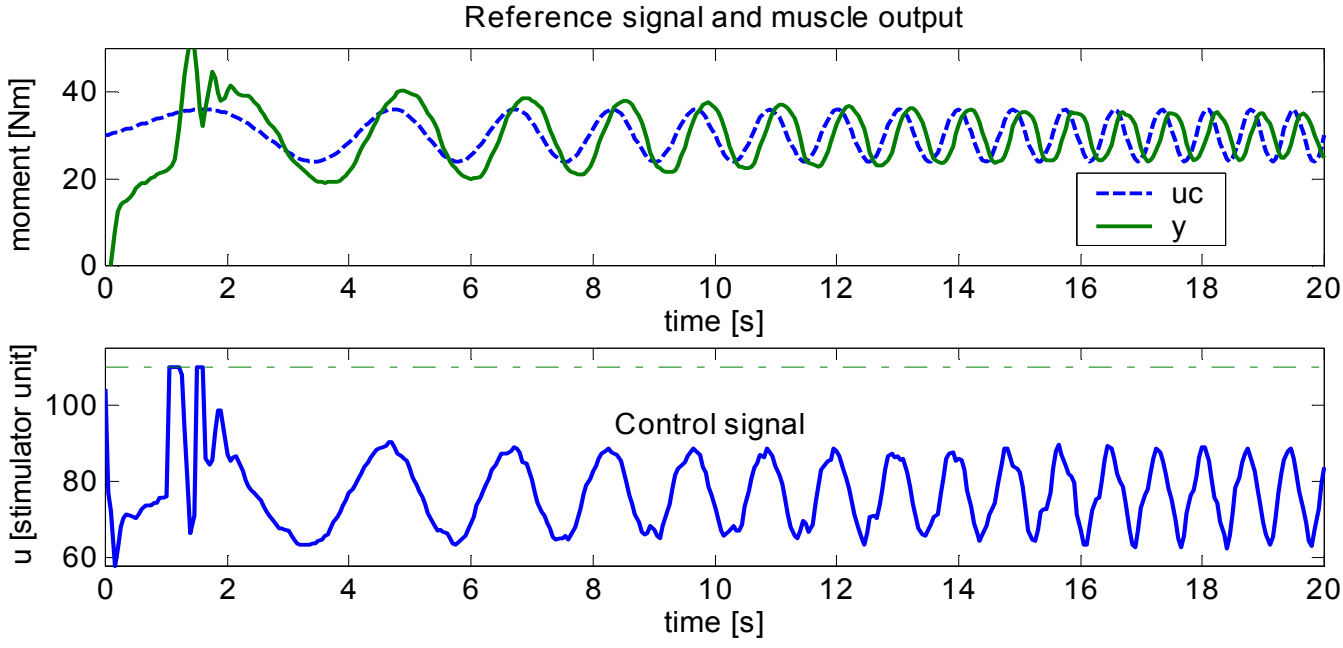


Fig. 2. Moment tracking of electrically stimulated muscle with sinusoidal reference signal u_c. The mean reference level is 62.5 % of the moment at saturation level. 1 stimulator unit equals 0.5 mA. The pulse width was constant at 400 μs.

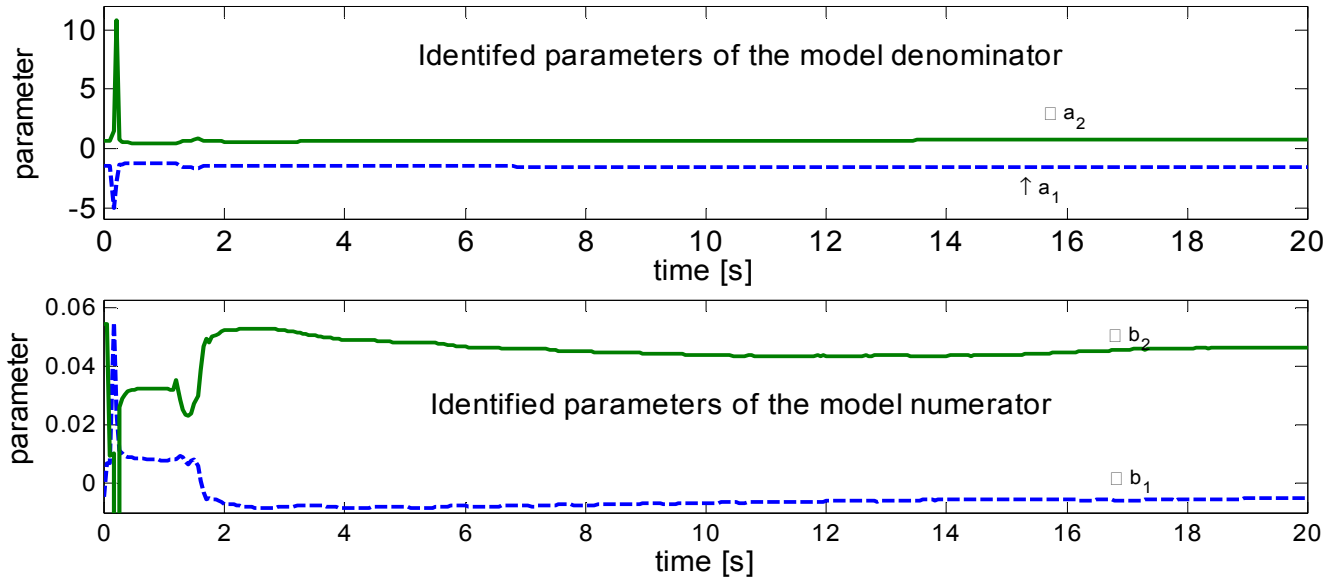


Fig. 3. Denominator (a_1, a_2) and numerator (b_1, b_2) parameter courses of discrete linear transfer function.

DISCUSSION

The controllers used in this work are linear and can only adjust to relatively slow changes in muscle properties. The local model approach used here is a possible stage in a process of nonlinear controller design. By using a linear controller we can simplify the validation of adaptive properties of the closed-loop system. Nonlinear control can be afterwards designed by switching between the linear controllers, where the switching can be such as gain scheduling between controllers or fuzzy switching between single controller parameters.

Moment tracking results show time delays of controlled moment after the reference signal. The delays were induced by the slow closed-loop program and will be reduced in our future work. Such delays additionally reduce the system robustness.

REFERENCES

/1/ Astrom K. J., Bjorn W., Adaptive Control, Addison-Wesley Publishing Company, Inc., 1995, 90 – 137

/2/ Hunt K. J., Munih M., Donaldson N., Barr F., Investigation of the Hammerstein Hypothesis in the Modeling of Electrically Stimulated Muscle, IEEE Trans Rehab Eng, Vol. 5, No. 4, Dec. 1997, 998 – 1009

/3/ Ponikvar M., Munih M., Setup and Procedure for On-line Identification of Electrically Stimulated Muscle with Matlab Simulink, accepted for IEEE Trans Neural Rehab Eng, Vol. 9, No. 3, Sep. 2001

ACKNOWLEDGEMENTS

The authors acknowledge the financial support of the Ministry of Science and Technology of the Republic of Slovenia.

AUTHOR'S ADDRESS

Matija Ponikvar, M. Sc. e-mail: matijap@robo.fe.uni-lj.si
Faculty of Electrical Engineering
Tržaška 25, Ljubljana

SLIDING MODE CONTROL OF FUNCTIONAL ELECTRICAL STIMULATION FOR KNEE JOINT ANGLE TRACKING

Sašo Jezernik [1], Philipp Inderbitzin [1], Thierry Keller [1], and Robert Riener [2]

[1] ETH Zürich, Automatic Control Laboratory, Physikstrasse 3, 8092 Zürich, Switzerland
[2] Technical University of Munich, Institute of Automatic Control Engineering, Arcisstr. 21, 80333 Munich, Germany

SUMMARY

An inherently robust closed-loop control strategy called sliding mode (SM) was applied to the problem of knee joint angle tracking by stimulation of the knee extensor and flexor muscles. The controller was synthesized using a state-space model of a human knee and muscles (composed of activation dynamics, muscle dynamics, and biomechanics) /1/. The model was experimentally validated in our earlier studies. The derived SM control law provided asymptotic stability of the knee joint angle and velocity, and was robust to the muscle fatigue. It was derived in a closed-form by considering only a subsystem of the muscle-knee model and using a joint angle-joint velocity sliding surface stabilized by appropriately generated muscle activation that followed from the sliding mode reachability and stabilizability conditions. In our earlier studies, instead of using a closed-form expression for the sliding mode controller, we have only used an approximate sliding mode control law, and have additionally necessitated a cascaded PI controller, which degraded the performance and sometimes caused instability /2/. Simulations with the new controller were carried out in SIMULINK with different knee angle trajectories and different controller parameters, and the results were analyzed. The controller with best parameters was able to track ramps and real knee joint angle walking trajectories with a root-mean-square (RMS) error of about 2 degrees at physiological (fast) speeds.

STATE OF THE ART

Functional Electrical Stimulation (FES) can be used to restore movement in certain paralyzed individuals by stimulation of intact peripheral nerves. Stimulation produces muscle contractions and generates joint movements. The underlying physiological/biomechanical system is highly nonlinear and time-variant, and a feedback control strategy is necessary for satisfactory control of the joint angles /3/. However, many classical closed-loop control algorithms were found unable to provide an adequate movement control. Some newer approaches gave good results though, but with no guarantees on stability /4/. Additional reasons for inadequate performance of closed-loop control strategies for FES are input constraints and bandwidth limitations due to low stimulation frequencies (<50 Hz). Our research goal is to develop new FES control strategies that insure stable angle tracking with an improved performance, and that could be applied to control several joints in order to control different movements like grasping, reaching, and walking.

In this study we have evaluated the performance of the nonlinear closed-loop control law called sliding mode /5,6/ that was used to control the knee joint angle by FES of suitable muscles. The SM control strategy was chosen as it was earlier shown to be robust to parameter variations and was already successful in controlling different complex nonlinear plants. Two variants of the sliding mode control were tested (discontinuous and continuous sliding mode control), both with the control of: (1) knee extensor muscles only, and (2), concurrent knee extensor and knee flexor muscle stimulation.

MATERIAL AND METHODS

Plant Model

The human leg model for one degree of freedom movement at the knee joint consisted of a simple leg biomechanical model and of a model describing muscular properties. The biomechanical part was

basically described by the equation of motion at the knee joint (parameters: leg mass, leg inertia, leg geometry, damping, passive knee torques, active muscle torques). The muscle dynamics subject to FES excitation was described by a static recruitment curve, nervous and muscular delay, linear second order Calcium (Ca) release and reuptake model, muscle fatigue model, and a modified Hill-type model of the active muscle contraction/force (containing force-length and force-velocity relationships). The total plant model can be written in the following state-space form:

$$\begin{bmatrix} \dot{x}_1 = x_2 \\ \dot{x}_2 = \dfrac{ma(x_1)\cdot fit\cdot x_3\cdot g_1(x_1)\cdot g_2(x_1,x_2)+\tau_{gravity}(x_1)+\tau_{passive}(x_1)-d\cdot x_2}{I_{tot}} \\ \dot{x}_3 = x_4 \\ \dot{x}_4 = \dfrac{1}{T^2}\cdot(-2T\cdot x_4 - x_3)+\dfrac{1}{T^2}u(t-t_d) \end{bmatrix}$$

whereby the state space variables x_1-x_4 stand for the knee angle, knee angular velocity, concentration of the Ca ions, and the derivative of the Ca ion concentration respectively. In this set of differential equations, *ma* stands for the muscle moment arm, *fit* describes the muscle fitness, g_1 and g_2 stand for the force-length and force-velocity relationships, τ_g for gravity, τ_p for passive knee joint torques, *d* for linear damping, *T* is the Ca release-reuptake time constant, and t_d is the delay.
Note that this model can be split into two submodels in the following way:

Subsystem 1 $\begin{bmatrix} \dot{x}_1 = f_1(x_2) \\ \dot{x}_2 = f_2(x_1,x_2,x_3) \end{bmatrix}$ Subsystem 2 $\begin{bmatrix} \dot{x}_3 = f_3(x_4) \\ \dot{x}_4 = f_4(x_3,x_4,u) \end{bmatrix}$

The state variable x_3 can now be regarded as an input to the subsystem 1 and as an output of the subsystem 2, which is driven by the input *u*. It is important to note that x_1 and x_2 can be measured, but x_3 and x_4 can not.

Sliding Mode Control

The basic idea behind the sliding mode control is to define a sliding surface (submanifold) in the $\mathfrak{R}^k$ vectorspace and to generate a control law (sliding mode control) that will force the system trajectory to reach the sliding surface in a finite time, and that will ensure that the subsequent evolution of the system trajectory will remain on the sliding surface (this latter mode is called a *sliding mode*). The sliding mode usually describes the error dynamics and forces the error to asymptotically decay to zero.
Let us define the sliding surface to be $s(x,x_{reference})=0$. If we select the control law to guarantee condition $ds/dt\cdot s<0$ (condition (1)) then the sliding mode surface will necessarily be reached (because (1) implies $ds/dt<0$ for $s>0$ and vice versa). The control that will furthermore ensure that *s* will remain 0 (sliding mode, condition (2)) can be calculated from the formula $ds/dt=0$. This control is called the equivalent control, u_{eq}. The discontinuous control law that combines (1) and (2) for the first subsystem is given by (3):

$$u_{SM\_SS1} = (x_3) = u_{eq} - k\cdot\mathrm{sgn}(s) = u_{eq} - k\cdot\mathrm{sgn}(x_2+\lambda\cdot x_1) =$$
$$= \frac{1}{ma(x_1)\cdot fit\cdot g_1(x_1)\cdot g_2(x_1,x_2)}(-\lambda\cdot x_2\cdot I_{tot} - \tau_g - \tau_p + d\cdot x_2) - k\cdot\mathrm{sgn}(x_2+\lambda\cdot x_1)$$

In the continuous case, the discontinuous term $k\cdot sgn(s)$ is replaced by a continuous one $k\cdot s$.
Both laws guarantee reachability of the sliding mode AND its stability. In our two dimensional case, the sliding surface represents a line through the origin of the phase plane that has the slope of $-\lambda$.
By substitution of (3) into the state-space subsystem 1 one can easily check that the conditions (1) and (2) are satisfied.

Simulations

Simulations were performed with the sliding mode control of the first subsystem only, and with the combined sliding mode control of the first subsystem and a corresponding feed-forward (FF) control of

the second subsystem. The control input was always constrained and its bandwidth limited with a zero-order-hold block (f_{stim}=20 Hz). First we have stimulated knee extensors only (quadriceps muscles) and then we have repeated the simulations with the concurrent stimulation of the knee extensors and knee flexors (biceps femoris muscles). The control task was to track a ramp reference with a p-p amplitude of 60 degrees and a period of 2 s, and a real knee joint trajectory reference measured during walking that had a period of 2 s. Extensive simulations with the ramp reference were performed in order to obtain the optimal controller parameters λ and k. The optimal values were then used in the subsequent simulations.

RESULTS

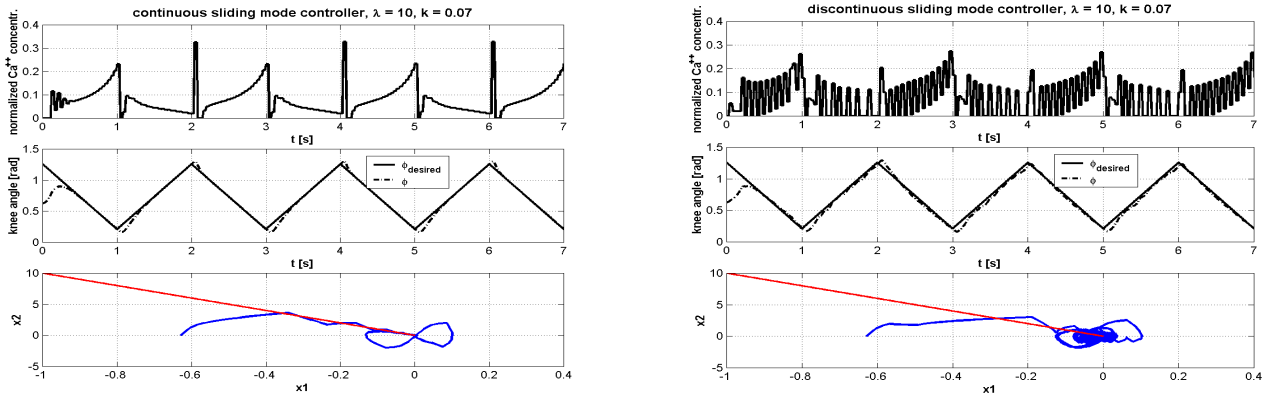


Fig.1 A (left) and B (right).

Control of Knee Extensors

The results of tracking ramps in the knee joint angle (sliding mode control, subsystem 1, $k=0.07$ and $\lambda=10$) are shown in Fig.1A for the continuous, and in Fig.1B for the discontinuous sliding mode control. The top panels show the control input x_3, the second panels the knee angle reference and the actual knee angle, and the bottom panels show reaching of the sliding surface and the convergence to the origin (asymptotic stability). Table 1 lists the RMS tracking errors for the continuous sliding mode for different controller parameters k and λ. Best results were obtained for $k=0.05$ and for $\lambda=20$.

k	e_{RMS} [deg] (λ=10)	e_{RMS} [deg] (λ=20)	e_{RMS} [deg] (λ=30)
0.01	4.24	3.41	2.95
0.05	2.48	**2.10**	2.45
0.1	2.34	3.06	4.07
0.2	5.02	8.06	9.07

Table 1

Fig.2A shows simulations where a real walking trajectory was tracked (cont. SM control, subsystem 1 only). Fig.2B in turn demonstrates the performance of the complete controller (FF 1st order control of the second subsystem combined with the SM control of the first subsystem). The control input (pulse width) was limited to 100-500 µs.
Higher values of k should provide more robustness (compensate for model uncertainties), but limits on x_3 (0 and 1) lead to control signal saturation that causes performance deterioration. This can be avoided if two muscles are used so that positive and negative torques can actively be generated.

Control of Knee Extensors and Flexors

In the case of stimulation of an agonist/antagonist muscle pair, we have split the control law (3) in such a way that the extensor muscles were activated for positive s and that the flexor muscles were activated when s was negative (only the second term in (3) was split). The tracking performance improved only slightly as the flexor muscles got activated mainly at the corners of the knee joint angle ramps

(Fig.3A+B). The top panels (Fig.3) show the x_3 /pulse width of the extensor muscles, and the middle panels the x_3 /pulse width of the flexor muscles for the subsystem 1/combined controller respectively.

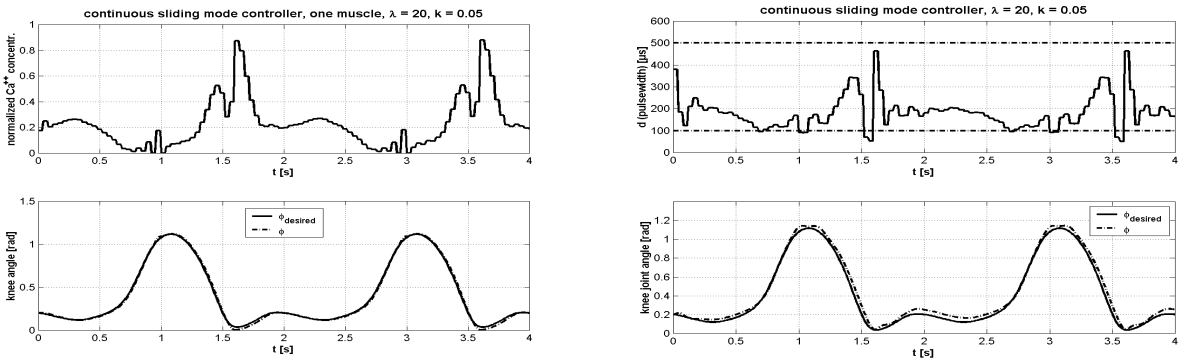


Fig.2 A (left) and B (right).

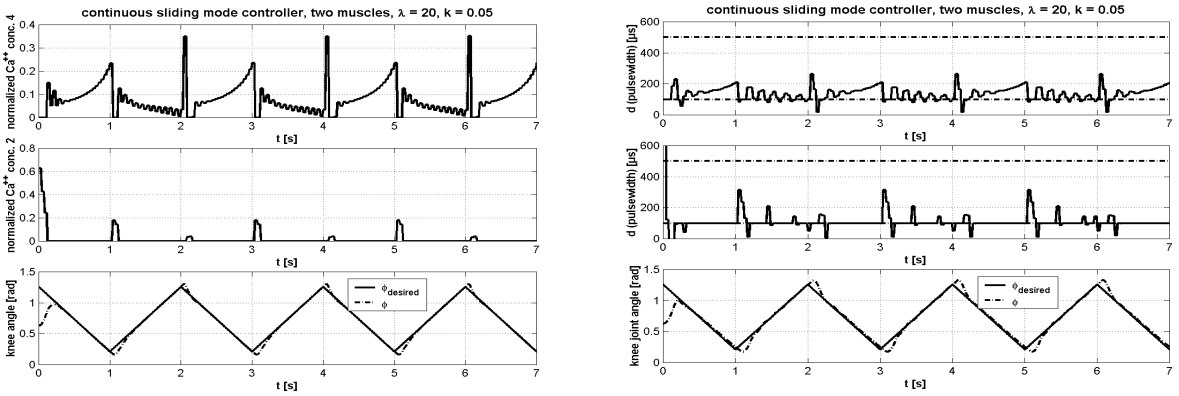


Fig.3 A (left) and B (right).

DISCUSSION

A nonlinear SM control law was derived for the knee joint angle control by stimulation of a single muscle group, and of an agonist/antagonist muscle group pair. This law guaranteed stability and achieved good results in computer simulations. Human experiments are currently being performed in order to test the developed controllers in real experiments. Next task will be to augment the SM control to two and subsequently three joints. Major limitations in the performance in simulations were due to low stimulation frequency and due to constraints on the control input signal.

REFERENCES

/1/ Riener R, Fuhr T: Patient-driven control of FES supported standing-up: A simulation study. IEEE TRE, Vol. 6, No. 2, pp. 113-124, 1998.
/2/ Jezernik S, Riener R: A computer simulation of tuned PID and continuous sliding mode FES control. In Proceedings of the International Biomechatronics Workshop, Enschede, Netherlands, pp.37-41, 1999.
/3/ Crago PE, Lan N, Veltink PH, Abbas JJ, Kantor C: New control strategies for neuroprosthetic systems. J. of Reh. Res. and Dev., Vol.33, No.2, pp.158-172, 1996.
/4/ Chang GC, Luh JJ, Liao GD, Lai JS, Cheng CK, Kuo BL, Kuo TS: Neuro-Control System for the Knee Joint Position Control with Quadriceps Stimulation. IEEE TRE, Vol.5, No.1, pp. 2-11, 1997.
/5/ Utkin VU: *Sliding Modes in Control and Optimization*. Springer-Verlag, Berlin, 1992.
/6/ Slotine JJE, Coetsee JA: Adaptive sliding controller synthesis for nonlinear systems. Int. J.Control, Vol.43, No.6, pp.1631-1651, 1986.

CORRESPONDING AUTHOR'S ADDRESS

Sašo Jezernik, Ph.D.
ETH Zürich, Automatic Control Laboratory
Physikstrasse 3, 8092 Zürich, Switzerland
Email address: jezernik@aut.ee.ethz.ch
Homepage: www.aut.ethz.ch/~jezernik

Session 8

DROPPED FOOT, FUNCTIONAL RESTORATION

CLINICAL ELEMENTS FOR THE NMS AND FES PROTOCOLS IN THE PRACTICE OF NEUROREHABILITATION

Meta M. Dimitrijevic, M.D. and Milan R. Dimitrijevic, M.D., D.Sc.

Dept of Physical Medicine and Rehabilitation, Baylor College of Medicine, Houston, Texas. USA.
University Institute of Clinical Neurophysiology, Ljubljana, Slovenia.

INTRODUCTION

We are witnessing a very fruitful period in the development of progress of biomedical engineering of devices, new methods for external control of paralyzed extremities due to neurological disorders. If we give just a look to the program of the 7th Vienna International Workshop on Functional Electrical Stimulation, it is easy to support this statement. During this meeting there are presentations and discussions on

1. electrical stimulation of denervated muscles,
2. NMS and FES in the paraplegia due to upper motor neuron lesion,
3. FES cycling,
4. implant technology,
5. command and feedback signals, stimulation parameters, up to
6. stimulation and closed loop control,
7. functional restoration of the drop foot,
8. stimulators for upper extremity and their functional restoration.

Therefore, it is opportunity to advance further practice of the NMS and FES within clinical programs of neurorehabilitation.

In the past, it can be that we didn't promote enough presentation and demonstration of clinical protocols of NMS and FES for treatment of neurological conditions and applications of electrophysiological bracing, orthosis. Therefore, to bring to your attention of necessity to discuss clinical protocols and their elements, in order to facilitate interest of clinician for NMS and FES in the clinical practice of the neurorehabilitation, we shall present in this lecture, some of the elements of such protocols for percutaneous stimulation.

First steps before we initiate programs of NMS and FES

After a patient with paralysis meet recruitment criteria for treatment with NMS or FES procedure or for fitting with electrophysiological brace, orthosis and thus become potential candidate for clinical program and defined functional goal, following steps we should perform:

1. Practicing professional must be fully informed by patient attending physician about patient condition. After this we shall describe to the patient what he will experience while we are applying electrodes for application of stimulation of peripheral nerves structures. Furthermore, we shall explain operation of stimulation unit and then whenever possible apply at first at the site of the skin where we expect normal sensation (preferably over the skin with underlying muscle and skin intact innervation) as well as under visual control of the subject while we perform initial testing.

   We should provide opportunity to the subject to experience tingling sensation evoked by train of stimuli and progressively and carefully by controlling stimulus parameters to inform subject about

different threshold of stimulation. Thresholds for

a) sensation and just below the sensation
b) definite and comfortable sensation
c) evoked movement
d) maximal tolerance of discomfort.

Moreover all this threshold can be changed by repetition of stimulation and patient accommodation to externally induced sensation and thresholds above evoked motor responses can be decreased by stimulus features and duration of stimulation. We shall avoid application of NMS or FES within areas with altered, decreased threshold for electrical stimulation, hyperesthesia, hyperalgesia.
After such training session we shall be ready to proceed with examination and evaluation of functional responses to neuromuscular stimulation in order to learn of neuromuscular physiological conditions and effect of electrical stimulation on initiation, external control and modification of movement performances.

2. The second step consist of evaluation procedure of muscle capability to respond to percutaneous stimulation to repetitive single stimuli with visible or by palpation recorded muscle twitches. If we measure time for how long muscle respond with muscle twitch we shall learn about endurance and fatigability of muscle contraction and force production capabilities.
   Afterwards depending from patient condition and our tentative goals, we should examine response of stimulation of cutaneous nerves for modulation of muscle tone, modification of volitional movements as well when necessary what will be effect of stimulation of mixed nerve trunk, or motor point for eliciting functional movements.
   In short we should not start treatment program of NMS and FES unless we examine and prove that we have available expected properties of the electrical stimulation of the nerve structures and there are responding to our electrical parameters within comfort to the stimulated person. We should carry examination of physiological properties of neuromuscular system and responses of altered motor control to stimulation parameters to the different sites and by use of different size of electrodes.
   Such clinical evaluation sessions should be performed in several sessions.
   After evaluation is completed, we shall be ready to start to develop clinical goals of NMS and FES protocols by being aware that they can be:

   a) Conditioning protocol,
   b) Neuroaugmentive and
   c) Learning one.

   None of these protocols exclude each other and their sequence or application will depend from our findings.

Neuromuscular stimulation (trophic state of the muscle)

During upper motor neuron dysfunction, either partial or complete, muscles with impaired innervation (after weeks and months) will suffer from progressive muscle disuse atrophy and alter capacity for muscle force and fatigue resistance. Muscle bulk decreases, and when muscle contraction are induced electrically there can be sometime in the decondition muscle only a few contractions and their amplitude will progressively diminish. Thus it is essential to develop a rigorous daily NMS program before use of the muscles with impaired upper motor neuron innervation by means of externally controlled electrical stimulation for the generation of the muscle force.
After the muscle force has become stronger, we then add a program of physical therapy with subject active movements, so that we can increase the endurance of the whole body and not only of the stimulated muscles.

Nerve electrical and neuromuscular stimulation and modification of muscle hypertonia

Muscle hypertonia usually results from chronic upper motor neuron dysfunction. When it is not to severe and its distribution is not generalized, but moderate and restricted to several muscle groups, stimulation of cutaneous nerves or spastic muscle groups can be effectively applied to diminish increased muscle tone (Lit. 1, Lit. 2).

The train of stimuli (20-50 Hz) adjusted to strength below threshold for sensation is the appropriate electrical stimulation strength for the control of spasticity. (In case patient has absent sensation we can use threshold for minimal motor response and then to adjust amplitude to be reasonable below threshold for motor response and within the range we shall expect to be if subject has preserved sensation). Stimulation should be applied for 30 minutes, twice a day. It is important to take care of the skin and to build skin tolerance for long-lasting electrical stimulation. Once skin tolerance is developed, and if muscle hypertonia is persistent, it is possible to stimulate one or several cutaneous nerves or the skin above the spastic muscle groups for several hours and several times per day.

Overall, electrical nerve and neuromuscular stimulation for the modification of muscle hypertonia has the advantage of being simple, whereas some difficulties lies in the proper placement of the electrodes over cutaneous nerves of sural,saphenus, lateral cutaneous femoral nerve, and cutaneous branches of radialis, musculocutaneous,ulnaris, radialis, axillaris nerves Another requirement is careful tuning of the strength of train of stimuli. This procedure is useful in those cases when spontaneous recovery will diminish spasticity.

Neuromuscular stimulation for the modification of patterns of movement

Another feature of upper motor neuron dysfunction is the presence of multi-joint patterned flexion-extension movements instead of the fine coordination control of different joints during the same motor sequence. A predominant extensor trust pattern with weakened flexor pattern is usually well recognized as the circumduction movement of the ambulatory hemiparetic patient after stroke, or the ambulatory SCI patient who can develop functional or non-functional slow gait when using crutches or other devices. In these patients, it is first necessary to use the electrical muscle conditioning (described above) and when muscle resistance improves, then proceed simultaneously with electrical stimulation and volitional movement. This is always a need for multi-site stimulation since motor deficits are present in several muscle groups.

There is not general recipe indicating how and when to stimulate different muscle groups: for example, in ambulatory spinal cord subjects, there are rarely identical motor patterns for both limbs or similarity between patients even when they have similar spinal cord lesions. Therefore, we have found that it is beneficial to use multi-site stimulation in the laboratory environment to assess the responsiveness of motor pattern, and one- to two- channels unit at home for daily training.

Functional electrical stimulation for impaired functional movement of the single muscle group

Isolated drop-foot and drop wrist are rare motor deficits in chronic neurological patients after stroke, brain or spinal cord injury, usually there are part of impaired pattern movements. However, these conditions are more likely to respond positively to the use of FES, since it is a very simple technological task to substitute the loss of control of single muscle group in the presence of volitional activity of all other muscle groups. This external approach is also very effective for the correction of more deficits and the improvement f motor activity.

However, while working with upper motor neuron drop-foot or drop-wrist, we have found that on eliciting functional movement of single muscle groups, the presence of subclincal impairment of other leg and thigh muscle group become more noticeable. Therefore, even when applying FES to one muscle group, it is imperative to incorporate in the program exercise and gait correction of motor activity of the other muscles groups not obviously affected at the beginning.

Externally electrically induced modification of altered neurocontrol

The application of external electrical control in patients with paralyzed extremities has given rise to two basic questions:

1. how effective is this approach to overcoming motor neuron dysfunction and
2. is it possible to accomplish the long term modification of motor control even in the absence of electrical stimulation.

The answer these question depend from the degree and the pattern of upper motor neuron dysfunction and also on the level of the lesion (spinal cord, brain stem, or brain).
In ambulatory SCI patients there have been reported observations that after longer period of NMS or FES the patient can achieve new features of motor activity which persist without any further stimulation (Lit. 3). Similar finding were reported by Kralj and Bajd (Lit. 4) and Boucher and Pepin (Lit. 5).

The electrically and externally induced modification of altered neurocontrol in this population of patients requires that stimulation should be tailored according to the patient's residual motor control. Therefore, it is essential to have a multi-site stimulation system with a variety of controls for the amplitude and duration of the train of stimuli from different channels. The patient's understanding of this approach and his commitment to the relatively modest functional outcome is also factor to be considered.

The topic of improvement of locomotor recovery after sensorimotor stimulation in the spinal cord injury has been revived from the neuroscience's point of view by Muir and Steeves, (Lit. 6) and from Clinical point of view recently by P.H. Gorman (Lit.7).

SUMMARY AND CONCLUSIONS

Significant contemporary progress in the designs, production of devices and methods for external control of impaired motor control in humans is asking for promotion of the NMS and FES practice of neurorehabilitation. Our recommendation is to make an effort to report together with description of clinically applicable devices also detailed protocols how application supposed to be conducted when a clinical device is introduced to NMS and FES therapy as well as when FES orthotic device is applicable.

In this lecture, we have illustrate some of the elements of such clinical protocols in order to promote discussion of how to facilitate wider application of NMS and FES neurophysiological procedures to the clinical programs of neurorehabilitation.
Neurophysiological features of NMS as an external substitution and control of peripheral inflow from paralyzed parts of the body can bypass alter connections by the injury of the CNS between processing nuclei of the brains and spinal cord.
This additional input to the CNS can prevent secondary neurogenic lesions by prevention of the effects of disuse in the early stages of CNS injuries. In addition, during recovery of function after acute phase by NMS we can facilitate and maintain nonspecific and generalized "central state of excitability of the CNS" which is critical to be operational on the appropriate higher functional level during recovery of impaired specific sensory-motor functions.
All this above listed effects of application of NMS and FES as a procedure for prevention of effects of disuse and to provide externally controlled substitution for activity-dependent processes of recovery are at present known neurobiological bases for repair processes. However, we should not neglect active role of NMS and FES procedures for maintaining optimal biological conditions in patients with chronic lesion and established incomplete recovery. We should use those mentioned procedures for clinical protocols to support maintenance of the optimal patient brain and neuromuscular functional condition even when neurological deficits can't be ameliorate.

REFERENCES

[1] Bajd T, Gregoric M, Vodovnik L, Benko H. Electrical stimulation in treating spasticity due to spinal cord injury. Arch Phys Med Rehabil. 1985 Aug;66(8):515-7.

[2] Dimitrijevic MM, Dimitrijevic MR, Verhagen-Mrtman L, Partritge M. Modification of muscle tone in patients with upper motor neuron dysfunction's by electrical stimulation of sural nerve. American Academy of Clinical Neurophysiology, 1987, Abstracts 2, 9.

[3] Dimitrijevic MM, Dimitrijevic MR, Partridge M, Verhagen-Metman L. Alteration of neurocontrol in chronic ambulatory spinal cord injury patients after long-term peripheral nerve stimulation. American Spinal Injury Association, 1988, Abstract Digest, p. 32.

[4] Kralj AR, Bajd T. Functional electrical stimulation: standing and walking after spinal cord injury. 1989. CRC Press, Boca Raton, Florida.

[5] Boucher JP Pepin A. Effects of patterned electrical stimulation in recent and chronic quadriplegia. In Neuromuscular stimulation basic concepts and clinical implications (eds. Rose FC, Jones R and Vrbova G) 1989, Demos, New York

[6] Muir GD, Steeves JD. Sensorimotor stimulation to improve locomotor recovery after spinal cord injury. Trends Neurosci. 1997 Feb;20(2):72-7. Review.

[7] Gorman PH. An update on functional electrical stimulation after spinal cord injury. Neurorehabil Neural Repair. 2000;14(4):251-63.

AUTHOR´S ADDRESS

Prof. Dr. Milan Dimitrijevic
Institute of Clinical Neurophysiology at the Division of Neurology,
University Medical Center Ljubljana
SI-1526 Ljublijana
Slovenia
E-mail: naisus@cs.com

A PILOT STUDY TO INVESTIGATE THE COMBINED USE OF BOTULINUM NEUROTOXIN A (BoNTA) AND FUNCTIONAL ELECTRICAL STIMULATION (FES), WITH PHYSIOTHERAPY, IN THE TREATMENT OF SPASTIC DROPPED FOOT IN SUBACUTE STROKE.

C.A. Johnson*, A.M. Tromans**, D.E. Wood*, D. O'Keefe*, D. Buhrs*, I.D. Swain*, J.H. Burridge***,

Department of Medical Physics and Biomedical Engineering* and Duke of Cornwall Spinal Injuries Unit**, Salisbury District Hospital; University of Southampton***.

SUMMARY

The purpose of the study was to investigate the feasibility of combining BoNTA and Functional Electrical Stimulation (FES) using a single channel drop-foot stimulator, with physiotherapy, in the treatment of spastic dropped foot in sub-acute stroke. Results of the study will be used to design a randomised, controlled trial (RCT). Subjects were randomised into two groups; both groups received physiotherapy, the treatment group also received BoNTA and FES. Patients recruited were within one year following stroke, and a baseline period (weeks -4 to 0) was incorporated to allow change in rate of recovery to be measured. BoNTA injections were given and FES treatment commenced at week 0, and follow-up assessments were made at weeks 2, 4, 8 and 12. The primary outcome measures were walking speed and the Physiological Cost Index (PCI) of gait. Measurements were taken of stimulated and non-stimulated walking. Analysis of the results of the primary outcome measures suggest that physiotherapy alone and the combined use of BoNTA and FES with physiotherapy has a beneficial effect on walking speed and PCI. The study has provided evidence of an additional effect of BoNTA and FES, and with sufficient information on effect variability to enable sample size calculations for a follow up RCT, to estimate the effect size with a useful degree of precision.

STATE OF THE ART

There are approximately 100,000 hospital admissions for stroke in the United Kingdom annually; 20% are fatal, 20% make a full recovery and 60% make a partial recovery but have a restricted lifestyle /1/. Physiotherapists working with these patients aim to maximize the recovery of movement and function through the re-education of postural control and normal activity. There are no statistics on how many patients might benefit from the combined use of FES and BoNTA, but 20% /2/ would be a conservative estimate of the proportion of patients suffering from a spastic drop-foot. This group of patients experience difficulty when walking because they are unable to effectively dorsiflex their ankle during the swing phase of walking. The problem arises partly through an inability to activate the anterior tibial muscles, and partly through restraint from the calf muscles. In some, but not all cases, the restraint is due mainly to inappropriate (spastic) calf activity; in others the mechanical resistance offered by the calf is the predominant factor. Knuttson and Richards /3/ described premature calf activity in one third of the study group of patients with spastic hemiparesis. EMG activity in the triceps surae began and peaked earlier in the gait cycle, compared with normal; peak activity was lower in triceps surae, and in tibialis anterior.

FES applied to the common peroneal nerve, and timed to the swing phase of walking has been shown to be effective in patients with poor active ankle dorsiflexion and calf spasticity /4/5/. FES is thought to have an inhibitory effect on antagonist activity, and thus reduce calf spasticity; it may also improve the mechanical component of calf restraint, through stretching of the triceps surae muscle group. BoNTA has also been shown to reduce inappropriate calf activity /6/ particularly in patients with premature calf activation during walking. The presence of spastic dropped-foot frequently causes stroke patients to adopt an abnormal gait pattern that may exacerbate extensor spasticity. Once established, such abnormal patterns are difficult to correct. By applying these combined treatments during the recovery phase, the long-term outcome of the mobility of stroke patients may be improved.

MATERIALS AND METHODS

19 patients were recruited from the stroke services of 4 NHS Trusts in England. All participants fulfilled the following selection criteria: 1) single stroke of vascular origin with hemiplegia during the previous 12 months; 2) inability to achieve a heel strike, correctable by FES; 3) between 3 and 6 inclusive on the Hauser Ambulation Index; 4) an increased calf stretch response on examination; 5) premature calf activation during gait identified by EMG. Patients were excluded who had: 1) any additional medical condition that might influence response to treatment; 2) prescribed anti-spastic medication; 3) prescribed medication that may have influenced heart rate measurements, either at the time of recruitment or during the 4 weeks prior to recruitment; 4) severe psychological problems; 5) any patient unable to give informed consent. Participants were randomised into 2 groups; each selected a sealed envelope containing an explanatory letter allocating them into one of the groups. The control group (CG) received physiotherapy, the treatment group (TG) received physiotherapy and the combined use of FES and BoNTA. Both groups continued with their physiotherapy programme throughout the study period. In-patients received a minimum of 3 sessions per week, out-patients 2 sessions per week.

Subjects in the TG received BoNTA injections at week 0. In most cases the medial and lateral heads of gastrocnemius were each injected with 200 units of Dysport, and tibialis posterior with 400 units. This dose was modified for less spastic muscles or smaller patients. Injections were given under electromyographical (EMG) guidance. FES was used by the treatment group patients immediately following BoNTA injections. An Odstock Dropped Foot Stimulator mark III (ODFS III) was set up by the Research Physiotherapist, following the standard protocol, with a full explanation and a copy of the user instruction manual. The device delivered a train of electrical impulses, at a frequency of 40 Hz, via surface electrodes. The common peroneal nerve was stimulated, and timed to the person's walking speed by a footswitch. Patients were asked to use the system on a daily basis to assist their walking, and for most of each day. Cyclic exercise stimulation was used in patients with higher levels of calf spasticity, and where mechanical constraint of the triceps surae was present, to maximize patient's response to the use of the ODFSIII.

Assessments for the study were carried out by the Research Physiotherapist in the patient's local Physiotherapy Department. The primary outcome measures were non-stimulated (CG and TG) and stimulated (TG) walking speed and PCI of gait. Secondary outcome measures included: a neurological examination, consisting of tendon and cutaneous reflexes in the lower limb, the Modified Ashworth Scale to measure spasticity in the quadriceps, hamstrings, dorsiflexors and plantarflexors, and scoring of clonus; the Rivermead Motor Assessment scale; the SF 36 Health Survey; a semi-structured interview; kinetic EMG for tibialis anterior and the lateral head of gastrocnemius during walking.

RESULTS

Descriptive statistics based on the demographic data of the study participants are tabulated below.

	Control	Treatment
Number of subjects	9	10
Age mean (SD)	59.33 (12.46)	58.2 (12.72)
Age range	44 – 78	41 – 78
Sex	4 female, 5 male	2 female, 8 male
Side of hemiplegia	5 right, 4 left	7 right, 3 left
Time since stroke 0 - 6 months	3	6
Time since stroke 6 – 12 months	6	4

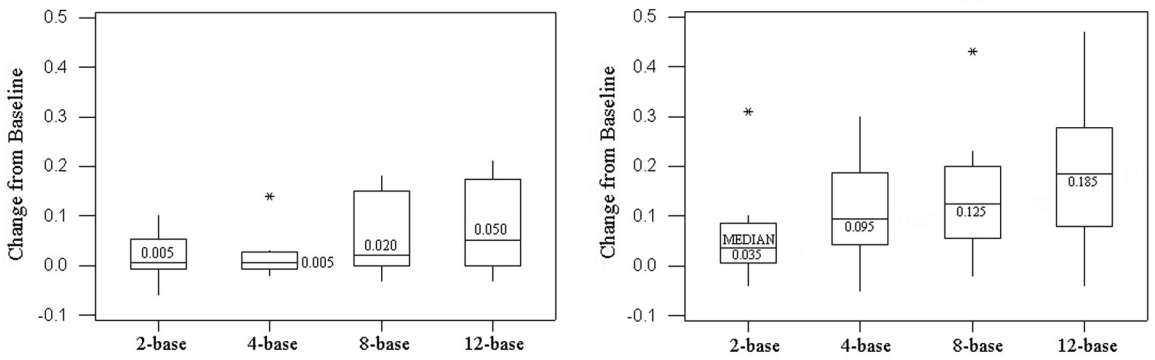
Results of the primary outcome measures of walking speed and PCI are presented. Statistical methods included graphical plots of individual patient response curves, non-parametric test statistics (Mann-

Whitney and Wilcoxon paired-sample tests), and a summary measures approach to assessing between group differences using regressions of median walking speed and PCI, and median change in walking speed and PCI on time.

A similar pattern of response for walking speeds and PCI was seen under both non-stimulated and stimulated test conditions. In the non-stimulated walking tests, a significant upward trend in median walking speed for both the CG (P=0.020) and TG (P=0.004) was seen, the trend lines being significantly different in location (P=0.040). Comparison of stimulated walking tests show a significant upward trend in median walking speed for both the CG (P=0.020) and the TG (P=0.042), the trend lines being significantly different in location (P=0.009). In non-stimulated walking tests a significant downward trend was demonstrated in median PCI for the treatment group (P=0.007), but not for the control group (P=0.292), the trend lines being significantly different in location (P=0.038). In stimulated walking tests a significant downward trend in median PCI was evident for the treatment group (P=0.020), but not for the control group (P=0.292), the trend lines being significantly different in location (P=0.016).

Regression of median change in walking speed was significant for the CG (P=0.043) and TG (P=0.025); the trend lines being in a sufficiently different location (P=0.0194). The following graph illustrates the changes from baseline Week 0, in median walking speed (m/s) for both CG and TG (stimulated).

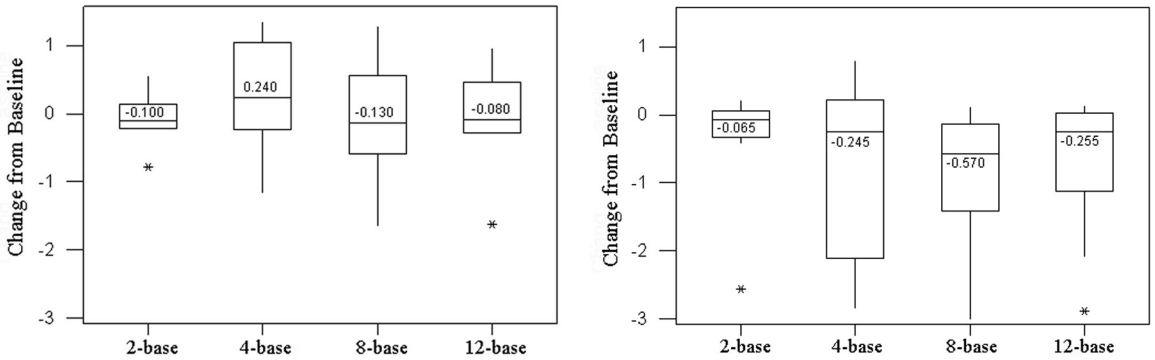
WALKING SPEED CHANGES FROM BASELINE



CONTROL TREATMENT

The following box and whisker plots demonstrate the changes from baseline Week 0 in median PCI (heart beats/m) for both CG and TG (stimulated). No trends emerged as significant.

PCI CHANGES FROM BASELINE



CONTROL TREATMENT

DISCUSSION

This pilot study has provided evidence of a real treatment effect. It has also given sufficient information on the variability of the outcome measure to facilitate sample size calculations for a subsequent study, to clarify the magnitude of the treatment effect with a meaningful degree of precision. Given a baseline median walking speed of 0.2 m/s, with a pooled standard deviation for median change in walking speed of 0.106 m/s, a test significance level of 5% (assuming the use of a non-parametric test statistic) and test power set at 80%, 470 subjects per group would be required to detect a 10% difference (0.02 m/s) between the treatment and control groups (as defined above) in respect of median change in walking speed from baseline (at 12 weeks), 120 subjects per group to detect a 20% difference (0.04 m/s), 55 subjects per group to detect a 30% difference (0.06 m/s) and 32 subjects per group to detect a 40% difference (0.08 m/s).

REFERENCES

/1/ Royal College of Physicians (1989) 'Stroke: towards better management. Summary & recommendations of a report of the Royal College of Physicians.' *Journal of the Royal College of Physicians* **24**, 1, 15-17

/2/ Gracanin F. (1984) 'Functional electrical stimulation in external control of motor activity and movements of paralysed extremities.' *International Rehabilitation Medicine* **6**, 25–30

/3/ Knutsson E., Richards C., (1979) 'Different types of disturbed motor control in gait of hemiparetic patients.' *Brain* **102**, 405-430

/4/ Burridge J.H., Taylor P.N., Hagan S.A., Wood D.E., Swain, I.D., (1997) 'The effects of common peroneal stimulation on the effort and speed of walking', *Clinical Rehabilitation* **11**, 201-210.

/5/ Burridge J.H., Taylor P.N., Wood D.E., McLellan D.E., (1998) 'The effect of different patterns of abnormal muscle activation during walking on the response to common peroneal stimulation', presented at the Human Performance Meeting, University of East London, July.

/6/ Hesse S., Ct at (1996) 'Ankle muscle activity before and after botulinum toxin therapy for lower limb extensor spasticity in chronic hemiparetic patients.' *Stroke* **27**, 3.

ACKNOWLEDGEMENTS

This work was supported by a grant from Ipsen Ltd, including the supply of BoNTA (Dysport), and Salisbury Area Healthcare NHS Trust. The author would also like to thank the team at the Department of Medical Physics and Biomedical Engineering, Salisbury District Hospital; the research team, including consultants, pharmacists and physiotherapists in the participating NHS Trusts; the Research and Development Support Unit at Salisbury District Hospital; and especially the participants themselves and their families.

AUTHOR'S ADDRESS

CA Johnson, Clinical Research Physiotherapist, Department of Medical Physics and Biomedical Engineering, Salisbury District Hospital, Wiltshire, SP2 8BJ. Telephone: 01722 336262 x 4060, Fax 01722 425263, e-mail calj@mpbe-sdh.demon.co.uk

USING FUNCTIONAL ELECTRICAL STIMULATION (FES) IN PARKINSON'S DISEASE

Finn S.M., Mann G.E., Taylor P.N.

Department of Medical Physics and Biomedical Engineering, Salisbury District Hospital, Salisbury, Wilts, SP2 8BJ England

SUMMARY

Parkinson's disease (PD) is a neurological impairment resulting in motor and functional problems including reduced stride length (shuffling) and akinesia (freezing). The deprivation in function of the basal ganglia in the brain results in diminished internal cueing associated with the planning of complex movements. There is evidence that external cues – visual, auditory, cognitive or sensory, may be able to compensate for the defective internal 'cueing system', improving step initiation and quality of walking in PD patients.
It is hypothesised that FES may act as a sensory cue to help maintain stride length and reduce akinesia in PD. 2 subjects diagnosed with PD were set up with FES devices. Foot switch triggered stimulation was applied via surface electrodes to the common peroneal nerve and the motor point of tibialis anterior. Although 1 subject could achieve active dorsiflexion, bilateral FES was effective in producing improved heel strike and reduced shuffling. Video footage from the subjects home showed an average increase in stride length of 19% and more importantly, appeared to have a safer gait. Subject 2 who has a dropped foot gained a beneficial effect from the dorsiflexion as well as the sensory cue provided by FES. By using a single channel FES device an increase in stride length of 24% was achieved.
This new application of FES has increased the mobility, stability and confidence of walking in the two subjects.

STATE OF THE ART

Introduction

PD is a progressive neurological condition resulting in motor and functional disability. It affects 1% of people over 50 years of age and is a leading cause of disability in those over 60 years of age /1/. The progressive nature of the gait disturbance in PD can lead to an eventual loss of mobility, an increased incidence of falls, and a loss of independence /2/. It is associated with reduced production of the neurotransmitter, dopamine, through degeneration of the basal ganglia and the substantia nigra. Characteristic neurological impairments in PD include resting tremor, rigidity, slowness of movement (bradykinesia) and postural instability. Walking becomes slower, with reduced stride length and cadence giving rise to a shuffling gait. As the disease progresses there are difficulties in initiating and maintaining movement resulting in freezing episodes, which occurs particularly in confined spaces. An increasingly flexed posture often develops over time. Symptoms may be relieved by levodopa - based drugs, which increase the amount of dopamine in the body. These drugs become less effective as the disease progresses and maximum dosage is reached /3/.

Functions of the basal ganglia

The basal ganglia are thought to have two main functions in movement. Firstly, the provision of internal motor 'cues' to the supplementary motor area of the cerebral cortex, to enable the release of each sub movement of a movement sequence to be correctly timed. Secondly, they contribute to the cortical 'motor set', which maintains movement sequences ready for execution /4/.

External Cueing

The effect of basal ganglia dysfunction in PD leads to defective cueing, which is manifested in early fatigue and reduction in speed and amplitude of movement. There is evidence that external cues – visual, auditory, cognitive or sensory, may be able to compensate for the defective internal 'cueing /5/. Bagley et al., /1/ showed that visual cues in the form of lines at regular intervals on the floor or as a target above eye level, improved the step length, velocity and heel strike of gait. Cutaneous cueing has also been shown to improve step initiation, force production and walking speed in Parkinson's Disease patients /6/. The concept of using FES to aid walking in PD is that stimulation of the common peroneal nerve produces dorsiflexion, providing a cutaneous and proprioceptive sensory cue and assisting the patient to step, thus avoiding shuffling and helping to overcome freezing.

MATERIALS AND METHODS

Devices Used

Two clinically used, CE marked devices were used. The Odstock 2-Channel Stimulator (02CHS) is a dual channel, foot switch triggered stimulator designed to elicit dorsiflexion of the foot by stimulation of the common peroneal nerve, (maximum amplitude 80mA, 300μs pulses, 40Hz). Surface electrodes are placed over the head of the fibula bone and the motor point of tibialis anterior.

The Odstock Dropped Foot Stimulator (ODFS) is a single channel device, currently being used by well over 1000 stroke and multiple sclerosis victims suffering from dropped foot. A randomised-controlled trial with hemiplegic patients showed a statistically significant increase in walking speed of 16% and a reduction in the Physiological Cost Index (PCI) of 29% /7/.

Subject 1

This is a 68-year-old man with a 14-year history of PD. He is currently prescribed levodopa – based medication, which is effective in relieving most of his symptoms. His main problems are initiating and maintaining movement, which cause him to adopt a shuffling gait, with reduced heel strike and to 'freeze' in confined spaces, especially in doorways and when turning. He often falls, having a detrimental effect on his confidence and therefore, mobility.

He was issued with the 02CHS in order to provide a bilateral sensory and motor cue. A footswitch placed in the right shoe triggers the stimulation to the right side on heel-rise and delivers stimulation to the left side on heel-strike. The stimulation envelope is ramped for comfort. The patient was video recorded walking with and without stimulation. The "Get-up" and go test /8/ that measures balance on a 5-point scale was conducted. This requires the patient to stand up from a chair, walk a short distance, turn around, return, and sit down again. This test was used to assess the risk of falling.

Subject 2

This is a 55-year-old woman with a 5-year history of PD. She is currently prescribed levodopa – based medication, which is effective in relieving most of her symptoms. Her main problem is an associated dropped foot. This results in a reduced stride length, tripping, hip 'hitching', large energy expenditure and again, loss in confidence when walking.

She was issued with the single channel ODFS for her affected (left) side. The foot switch was placed in the affected side's shoe so that the stimulation adapts to the speed of walking. The stimulation current was delivered on heel-rise, with ramping for comfort and to prevent 'foot-flap' on heel strike.

RESULTS

The "Get-up" and go test carried out for subject 1 suggests that by using FES as a sensory cue a more 'normal' (therefor safer) gait could be achieved in PD victims with substantial gait problems. Videos were taken of the subject performing a typical task at home with and without FES. An average increase in stride length of 19% was achieved using FES. This increase in stride length led to a safer gait pattern, agreeing with the findings of Morris et al. /2/, who states "regulation of stride length is the key deficit in gait hypokinesia".

Subject 2 achieved an increase in stride length of 24% over a 10m walkway. This large increase is probably due to the orthotic effect of FES (ankle dorsiflexion) as well as providing a sensory cue.

Both patients felt more confident when walking using FES.

DISCUSSION

From the preliminary work described above, both PD subjects appeared to benefit from the use of FES. This may be due to the sensation of stimulation acting as a triggering external sensory cue, to increased proprioceptive input from the contraction of the anterior tibial muscles, or a combination of the two. There is also probably some placebo effect in the knowledge that the stimulator will reliably lift the foot. It is hoped that there may be an application for FES for patients with PD in overcoming akinesia and maintaining their gait pattern, particularly when their drug treatment is becoming less effective after a number of years. We attempted to improve the quality and safety of PD walking so parameters such as walking speed and cadence were not measured. We also found it necessary to take measurements in the patient's home as the doorways and corridors are generally large and spacious in hospital clinic areas, and are not typical of the more confined spaces in the everyday environment of most patients.

The advantage of using FES over other cueing systems is the adaptability to the environment. The system will work around the home and outside – in any lighting conditions. It also adapts to the walking speed of the user as it is triggered on heel rise – overcoming audio cue systems with a metronome or music, where the user has to maintain a pre-set speed.

REFERENCES

/1/ Bagley S, Kelly B, Tunnicliffe N, Turnbull G.I., Walker G.M. (1991). The effect of visual cues on the gait of independently mobile Parkinson's disease. *Physiotherapy*. **77(6)**. 415-420.

/2/ Morris M.E., Iansek R, Matyas T.A, Summers J.J., (1994). Ability to modulate walking cadence remains intact in Parkinson's disease. *Journal of Neurology, Neurosurgery, and Psychiatry*. **57**. 1532-1534.

/3/ Bloem B.R., Beckley D.J., Van Dijk J.G. (1993). Pathophysiology of balance impairment Parkinson's disease. *Focus on Parkinson's disease*. **April**. 11-18.

/4/ Iansek R, Bradshaw J.L., Phillips J.G., Cunnington R. & Morris M.E. (1995). Interaction of the Basal Ganglia and Supplementary Motor Area in the elaboration of movement. *Motor Control and Sensory Motor Integration*: Issues & Directions, Glencross,D.J. & Piek, J.P. (Eds.). Elsevier Science B.V.

/5/ McIntosh G.C., Brown S.H., Rice R.R., Thaut M.H. (1997). Rhythmic auditory – motor facilitation of gait patterns in patient's with Parkinson's Disease. *Journal of Neurology, Neurosurgery and Psychiatry*. **62(1)**. 22–26.

/6/ Jacobs A.B., Horak F.B., Nutt J.G., Obeso J.A. (1997). Step initiation in Parkinson's Disease: Influence of Levodopa and External Sensory Triggers. *Movement Disorders*. **12(2).** 206–215.

/7/ Burridge J, Taylor P, Hagan S, Wood D, Swain I. (1997). The effects of common peroneal nerve stimulation on the effort and speed of walking: A randomised controlled clinical trial with chronic hemiplegic patients. *Clinical Rehabilitation*. **11**. 201-210.

/8/ Mathias S, Nayak U.S.L., Isaacs B. (1986). Balance in Elderly Patients: The "Get-up and Go" Test. *Arch Phys Med Rehabil*. **67**. 387-389.

ACKNOWLEDGEMENTS

The Authors would like to thank the Department of Medical Physics and Biomedical Engineering, Salisbury District Hospital, Salisbury, Wilts, SP2 8BJ England, for the supply of the FES devices used.

AUTHOR'S ADDRESS

Mr. Stacey M Finn Department of Medical Physics and Biomedical Engineering, Salisbury District Hospital, Salisbury, Wilts, SP2 8BJ England

e-mail: s.finn@salisburyfes.com
homepage: www.salisburyfes.com

DERMATOME ELECTRICAL STIMULATION AS A THERAPEUTIC AMBULATORY AID FOR INCOMPLETE SCI PATIENTS

T. Bajd, M. Munih, R. Šavrin*, H. Benko*, and I. Cikajlo

Faculty of Electrical Engineering, University of Ljubljana, Slovenia
*Rehabilitation Institute, Ljubljana, Slovenia

SUMMARY

Electrical stimulation of the L-3,4 dermatome during treadmill walking is proposed as a gait training modality in incomplete spinal cord injured (SCI) patients. The dermatome stimulation proved to be efficient in diminishing the extensor tone occuring after loading of the paralyzed limb during the stance phase of walking and resulting in improved flexion of the leg during the swing phase.

STATE OF THE ART

In the last decades more incomplete than complete SCI patients are arriving to the spinal units. One of the primary goals of the rehabilitative program for the incompletely paralyzed subjects is not only returning them into standing position, but also restoring their walking patterns. There are several gait training modalities available for this group of patients. In Table 1 they are divided into the methods based on mechanical activation or electrical stimulation of the partially paralyzed lower extremities. Some of these approaches are only provoking passive movement of the leg, while others are eliciting reflex responses.

Table 1. Gait training in partially paralyzed patients

	Providing movement of leg	Eliciting reflex response
Mechanical activation	robot manipulators active exoskeletons	treadmill active joint orthosis tendon vibration
Electrical stimulation	multichannel surface multichannel percutaneous	afferent nerve spinal cord dermatome

It is our belief that the approaches which are merely providing the movement of the leg are less efficient in re-learning of patient's walking. Robot manipulators combined with treadmill are used to lift the leg and bring the foot forward. In this case only the afferent input from the joint receptors may promote the gait re-learning process. Similar movements can be accomplished by active exoskeleton systems. Gait pattern can be restored in paralyzed persons also by surface or percutaneous multichannel electrical stimulation. Strong electrical stimuli delivered to the efferent nerves may represent unwanted noise in the afferent nerves, thus hindering the re-learning process.

The gait training modalities eliciting reflex responses result in more complex and natural like movements which are provoking afferent signals in joints, tendons, and muscles. Treadmill is producing hip extension at the end of the stance phase which is inducing reflex hip flexion and thus initiating the swing phase of walking /1/. A powerful motor and gear system attached to a mechanical ankle joint orthosis by means of flexible bowden cables, can elicit stretch reflexes by displacing rapidly the ankle joint /2/. Vibration of muscles and tendons activates muscle spindle afferents and produces illusory changes in joint position /3/. Illusion of the altered position may play important role in gait training. It has been demonstrated already in 1973 /4/ that electrical pulses applied to the sural or tibial nerves result in reflex hip and knee flexion with a simultaneous reflex ankle dorsiflexion. The swing phase obtained by eliciting a synergistic flexion response through electrical stimulation of the common peroneal nerve was extensively used by our group /5/. The spinal cord stimulation has a predominantly afferent influence /6/. Due to this stimulation the supraspinal structures exert their influence through the descending pathways and segmental reflexes, and thus at least partially restore the brain control over the locomotor system. In this paper we are proposing the dermatome stimulation combined with treadmill walking as a modality for gait training in incomplete SCI persons /7/.

METHODS

The swing phase of walking can be influenced through cutaneous stimulation of the selected dermatomes. In the investigation a 63y. old patient with C 2-6 spinal cord lesion resulting from an accident has been selected. The electrodes were placed over the L-3,4 dermatome, one medially below the knee and the other laterally above it, with the aim to decrease the extensor spasticity of the knee extensors, innervated from the same spinal cord level as the dermatome. Schematic representation of electrodes positioning is shown in Figure 1. A stimulation frequency of 100 Hz and a pulse duration of 0.3 ms were used without interruption during the gait cycles. The electrical stimulation was not causing any muscle contraction. It is our belief that the sensory electrical stimulation was delivered predominantly through the large diameter afferent fibers.

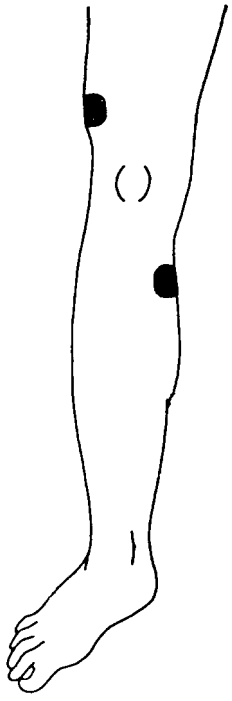


Fig. 1 Positioning of the surface electrodes over the L-3,4 dermatome.

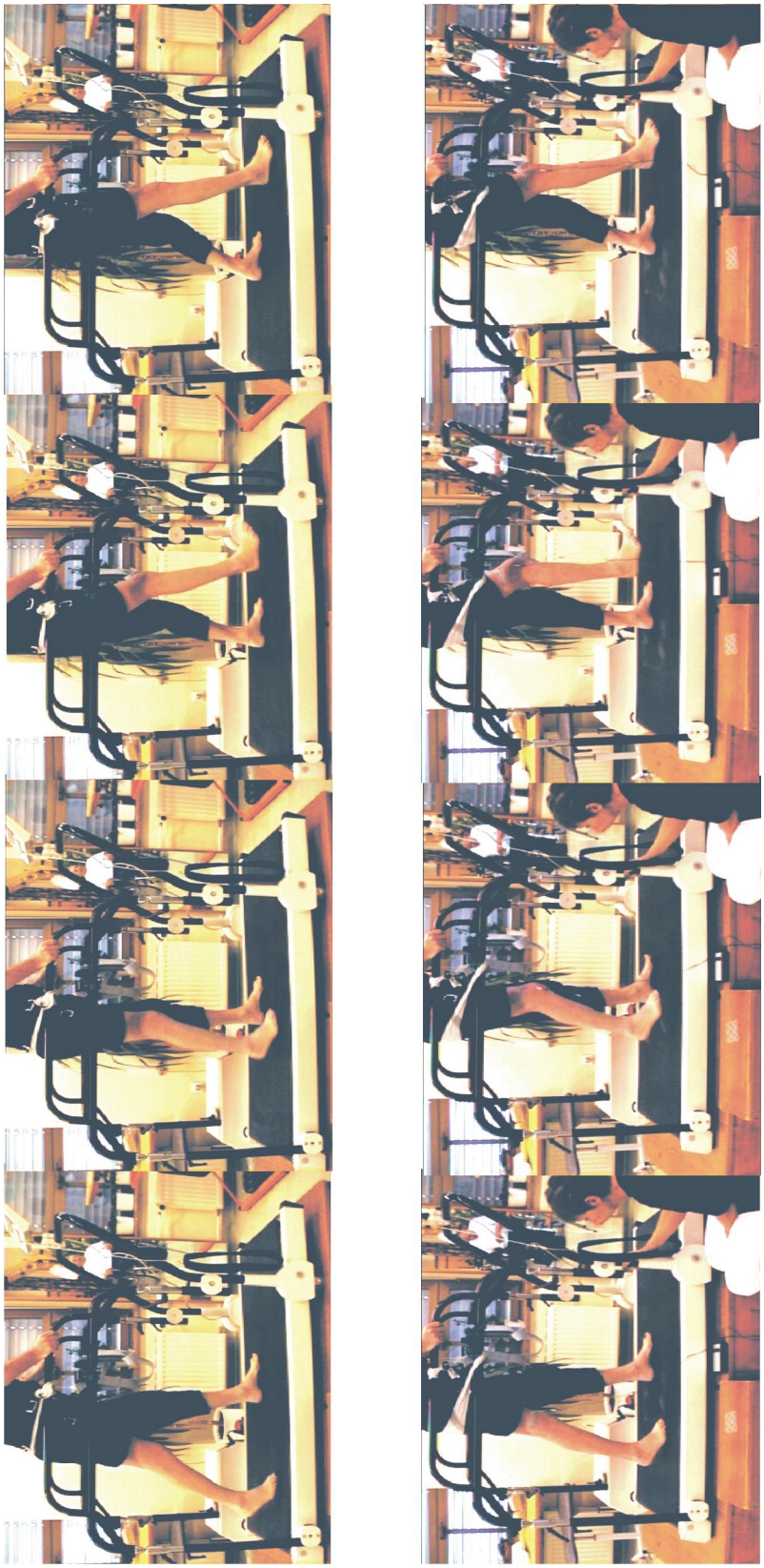


Fig. 2. Record of an incomplete SCI patient's walking without (upper row) and with the dermatome electrical stimulation (lower row).

RESULTS

Strong extensor spasticity is often observed in the lower extremities of the incomplete SCI patients. After loading the paralyzed limb during the stance phase of walking, the patients have difficulty to break this extension tone and cannot initiate a step. The stimulation of the L-3,4 dermatome proved to be efficient in diminishing this extensor activity. Hip and knee flexion and ankle dorsiflexion were significantly increased during the swing phase of walking. Also, the eversion of the foot was noticeably improved when delivering the dermatome stimulation. The upper series of the photographs in Figure 2 belongs to walking without stimulation. The lower photographs show improved swing phase of walking during L-3,4 dermatome stimulation. The dermatome stimulation proved to be specially efficient when combined with treadmill walking.

DISCUSSION

In spite of several decades of investigations of FES for lower extremities we cannot claim that FES of lower limbs is widely used in clinical environment. The FES synthesis of walking requires complex multijoint movements which further require large number of surface or implanted electrodes together with special algorthms providing coordination of many channels of electrical stimulation. It is our belief that only simple FES systems are perspective from the clinical point of view. Electrical stimulation of spinal neural circuits, rather than direct activation of motoneurons, will simplify generation of complex motor behaviours /8/. Electrical stimulation of the dermatomes, described in this paper, is just one possible access to the spinal neural circuitry from the periphery.

REFERENCES

/1/ Wernig A., Müller S., Laufband locomotion with body weight support improved walking in persons with severe spinal cord injuries, *Paraplegia*, Vol. 30, 1992, 229-238.
/2/ Andersen J.B., Sinkjaer T., An actuator system for investigating electrophysiological and biomechanical features around the human ankle joint during gait, *IEEE Trans. Rehab. Eng*., Vol. 3, 1995, 299-306.
/3/ Gandevia S.C., Kinesthesia: Roles for afferent signals and motor commands, In: Rowell L.B. and Shepherd J.T. Eds. *Handbook of Physiology*, Oxford University Press, 1996, 128-172.
/4/ Liberson W.T., Functional electrical stimulation in paraplegics and »Reflex Walking«, *Arch. Phys. Med. Rehabil.*, Vol. 54, 1973, 588.
/5/ Bajd T., Kralj A., Turk R., Benko H., Šega J., The use of a four-channel electrical stimulator as an ambulatory aid for paraplegic patients, *Phys. Ther*., Vol. 63, 1983, 1116-1120.
/6/ Cook A.W., Taylor J.K., Nidzgorski F., Functional stimulation of spinal cord in multiple sclerosis, *J. Med. Eng. Technol*., Vol. 3, 1979, 18-23.
/7/ Bajd T., Gregorič M., Vodovnik L., Benko H., Electrical stimulation in treating spasticity resulting from spinal cord injury, *Arch. Phys. Med. Rehabil.*, Vol. 66, 1985, 515-517.
/8/ Grill W.M., Electrical activation of spinal circuits: Application to motor-system neural prostheses, *Neuromodulation*, Vol. 3, 2000, 97-106.

AUTHOR'S ADDRESS

Prof. Dr. Tadej Bajd,
Faculty of Electrical Engineering
University of Ljubljana
1000 Ljubljana, Tržaška 25, Slovenia

e-mail: tadej.bajd@fe.uni-lj.si
home page: www.fe.uni-lj.si

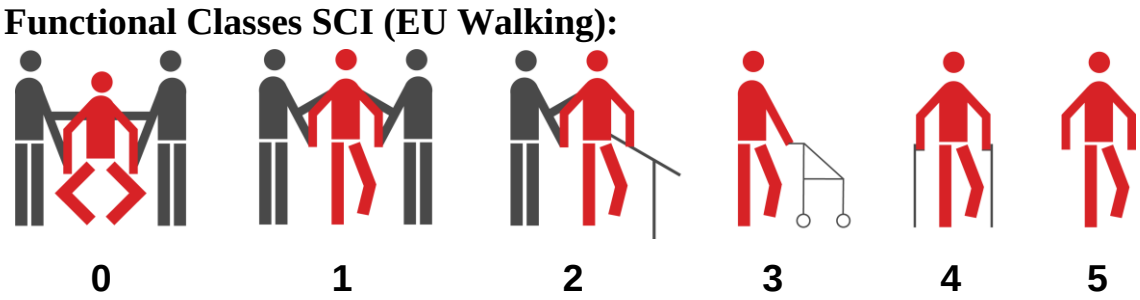
Principles of Laufband (LB) therapy for spinal cord damaged persons.

A. Wernig, S. Müller

Department of Physiology, University of Bonn and Klinikum Karlsbad-Langensteinbach

Laufband therapy was developed in the late 80ies independently by two groups of researchers (1,2). The principles had previously been worked out in the spinal cat, some decades ago, others only recently. „Spinal locomotion“ was already noticed by Freusberg 1874. Why its significance for the human was not noticed before, is not clear; possibly the fact that “spinal man” **does not** walk as a spinal cat does, blurred the view at **severely though incompletely** paralyzed patients. Today we interpret the surplus in evokable muscle activity during locomotion versus resting positions as instrumented by spinal locomotor centers fed with proprioceptive information. This view is supported by the finding that “rules of spinal locomotion” facilitate stepping in the (incompletely paralyzed) human as well Recent EMG recordings verify that muscles not evokable during single joint attempts may become active during air stepping (i.e.multijoint) and much more so during locomotion. Patients will be demonstrated on video films who learned to perform independent stepping over ground with little voluntary muscle activities in resting positions.

In order to better evaluate such phenomena a **cumulated muscle score** (**EU-Muscle**) is suggested (3) which summarizes functionally important rather than spinal segments-characterizing muscles (as the ASIA muscle score does). To asses locomotor capability, a **functional classification** (0 to 5, **EU Walking**) of SCI persons with a finer grading than the ASIA score is proposed. **Dependent:** 0: not capable of walking even with help of two therapists; 1: capable; 2: walking at the railing with one therapist. **Independent:** 3: rolator or reciprocal frame; 4: Regular crutches; 5: without devices. The classes graphically:



First, results of a 5-year study are reported in which 89 **incompletely** paralysed (44 chronic and 45 acute) patients who underwent LB-therapy, are compared with a total of 64 patients treated conventionally for comparable periods of time (median 10.5 weeks). LB-therapy achieved significantly better results in all comparisons (Wernig et al., Europ J Neurosci. 7, 823-829, 1995). A number of **chronically** wheelchair-bound patients (not capable of raising from the wheelchair or walking without help from other persons) became independent and walked with help of a rollator or two canes for distances of at least 100 meters. Most chronic patients not capable of stair case walking learned to do so either by themselves or with help

from another person following Laufband therapy. Also **acute** patients treated on the Laufband achieved better results than conventionally treated patients.
The results of a follow-up evaluation are reported, in which walking capability of Laufband treated patients, immediately following therapy, is compared with that after 1-4 years in domestic surrounding (Wernig et al., Spinal Cord 36, 744 – 749, 1998).

Summary: Locomotor training on the Laufband focuses on intensive walking in upright position, facilitated by body weight support via a harness, the moving band of the treadmill and initial limb setting by two therapists if necessary. These principles are derived from observations in spinal animals on activity-related "learning" of the isolated spinal cord and on the "rules of spinal locomotion" in lower vertebrates. The noval approach to intensely train precisely that skill which needs to be achieved (upright walking) allows to reach therapeutic goals far beyond those possible by conventional physiotherapy. As a rule in motor rehabilitation, therefore, each patient with some if little voluntary muscle activity remaining, needs to be trained on the Laufband to find and approach his individual limits of locomotion. Practical obstacles for a successful application of Laufband therapy on a broader scale at the time are the poor education of the therapists in practical handling of the patients and in their imagination of the possible therapeutic goals to be reached.

Session 9

DROPPED FOOT, STIMULATORS

A PILOT STUDY OF AUTO-MYO-ELECTRIC CONTROL OF FES ON TIBIALIS ANTERIOR IN CVA AND SCI SUBJECTS

R. Thorsen[1,2,4,5], J. Burridge[3], N. Donaldson[4], M. Ferrarin[1], J. Norton[4] and P. Veltink[5]

[1]Centro di Bioingegneria, Fond. Don Gnocchi-Politecnico di Milano, Italy.
[2]Salisbury District Hospital, Salisbury, England (SDH)
[3]University of Southampton, England
[4]University College London, London, England
[5]University of Twente, Enschede, The Netherlands

SUMMARY

Myo-electric signals (MES) from the paretic anterior tibial muscle (TA) have been used to control functional electrical stimulation (FES) of the same muscle. The technique has been termed Auto-Myo-electric Control of FES (AutoMeCFES). The MES was recorded using surface electrodes and processed digitally to allow continuous control of the amplitude of FES applied to the common peroneal nerve. Dorsi-flexion torque was measured, in isometric condition, with the subject sitting and the leg hanging freely. A tracking test was used to compare the amplitude of the ankle torque with and without stimulation. Seven of the 9 stroke and 2 of the 3 SCI subjects showed increased torque amplitude with the system. An immediate carryover effect was seen in one subject. It is concluded that, for selected subjects, AutoMeCFES can increase the muscle force of TA.

STATE OF THE ART

In some cases, a dropped foot may be corrected by use of Functional Electrical Stimulation (FES) /1/. Existing dropped-foot stimulators are most commonly controlled by a switch /1, 2, 3/. Although this method is satisfactory for many subjects it tends to over-ride the remaining muscle activity whereas myo-electric control may enhance it, which may therefore facilitate motor re-learning /4,5/. The feasibility of proportional myo-electrically controlled stimulation (MCS) for the upper extremity has been described /6/ and demonstrated /7/ although the system is still confined to laboratory tests.
In a previous paper /8/ we described the use of a tracking test for evaluation of the torque enhancement by comparing the root mean square (RMS) values of the subject generated tracks with and without MCS. In this paper we have modified this method and included the results from 3 SCI subjects with paretic TA.

MATERIAL AND METHODS

Subjects

Subjects were selected from the database and medical records of >600 dropped-foot stimulator users at the Medical Physics Department, Salisbury District Hospital (SDH) using the following criteria: ≥1 year after stroke; medically stable; living within 60km from SDH; 18-60 years of age; near normal passive range of movement of the ankle joint; some residual voluntary TA activity, but unable to dorsiflex the ankle in standing; able to give informed consent and comply with the test protocol; good response to FES and no history of adverse effects due to stimulation; no serious medical or heart/respiratory problems or using electrical life support devices. All subjects were users of dropped foot stimulators. Approval from the ethics committee and the subjects' informed consent were obtained. The subjects are listed in Table 1.

System

The AutoMeCFES system was built in Centro di Bioingegneria and programmed at University Twente, where preliminary tests were performed. It amplifies and filters the MES (20Hz-500Hz) using a dedicated amplifier /9/ and converts it to a 2.5kHz digital signal. Signal processing is as follows: reduction of stimulation artefacts by comb-filtering and elimination of initial 10ms post stimulation signal; root mean

square (RMS) calculation; smoothing by low-pass filtering; a noise offset is subtracted and a gain determines the stimulation amplitude, which is limited between zero and an upper limit individually defined for each subject. The low-pass filter has a cut-off frequency of 1Hz. The stimulation current is amplitude modulated with rectangular biphasic balanced waveform. Further information can be found in /7, 8/

Setup

Stimulation electrodes were placed, one just under the head of fibula and the other 2/3 down on the belly of TA, to induce dorsiflexion of the paretic foot. In all but one case (Subj: B) this placement coincided with the one the subjects used for their usual dropped foot stimulator. To minimise stimulation artefacts the recording electrodes were located between and aligned perpendicular to the stimulation electrodes.
A light-weight footrig was designed and built to hold the ankle in 10° plantarflexion (from normal posture) and to measure isometric dorsiflexion torque. Subjects were placed comfortably on a couch with a free swinging shank and the foot clear of the ground, thus avoiding accessory forces.

The Sessions

Each session was limited to two hours to avoid fatigue. Maximal stimulation amplitude (*IMax*) was established as the upper comfortable limit (all subjects had skin sensation) and the recruitment curve was measured. The subject was then asked to produce a strong volitional contraction during maximum stimulation. The torque produced during this pre-test was used as the target amplitude in all the tracking tests (with and without stimulation). This target amplitude was, for simplicity of the experiment, limited to 6, 13, 19 or 25Nm, whichever was just above the pre-test torque.
In the tracking test, a circle and a cross were displayed to the subject on a computer screen. The circle (target signal) oscillated vertically in a sinusoid (period = 5sec) for 30 seconds. The subject's task was to keep the cross, representing dorsiflexion torque (tracking signal), within, or as close to the target as possible, by contracting the dorsiflexors. The test was performed first without stimulation (natural) and then with AutoMeCFES.

Evaluation

We define the peak torque (PT) as the difference between maximum and minimum torque calculated as the mean value of, respectively, the 16 (1 sec) largest and smallest sample values in the subject generated tracking. This gives us the absolute range in which the subject can work. As a measure of the dynamic range, we have chosen the standard deviation (STD) of the target as a score for the variation around the mean value. This has the advantage over the root mean square (as used in /7/) that it will not be influenced by eventual offset that could be caused by increased muscle tone.
Thus is defined the torque enhancement in two ways.

$$\text{PT Enhancement} = (PT_{MCS} - PT_{Nat.}) / PT_{Nat.}$$

$$\text{SD Enhancement} = (SD_{MCS} - SD_{Nat.}) / SD_{Nat.}$$

RESULTS&DISCUSSION

Subjects are denoted by ID number and the result of the test is shown in Table 1.

ID	A	B	C	D	E	F	G	H	I	X	Y	Z
Lesion type	CVA	CVA	CVA	CVA	CVA	CVA	CVA	CVA	CVA	SCI	SCI	SCI
Age [Years]	62	50	38	58	27	48	53	65	63	47	25	30
Injury [Years]	6	7	2	4	13	7	12	9	8	5	8	2
IMax [mA]	45	63	40	40	15	10	55	21	50	45	33	25
Target [Nm]	25	19	12	19	11	6.2	6.2	12	25	37	50	37
$PT_{Nat.}$ [Nm]	13	11	11	10	5.6	3.5	0.62	2.6	8.4	24	21	22
PT_{MCS} [Nm]	23	17	13	9.5	8.7	3.5	4.5	9.3	12	26	23	25
$STD_{Nat.}$ [Nm]	4.8	2.5	3.7	3.8	1.6	0.94	0.21	0.96	3	7.8	6.5	7.1
STD_{MCS} [Nm]	8.7	5.6	4.5	3.5	2.9	0.88	1.4	2.6	3.9	9.3	6.1	8.2

Table 1. Shows, for each subject, the maximum values for stimulation (IMax), target torque, Peak torque (PT) and the standard deviation (STD) of the natural track and the MCS track.

As it can be seen in Figure 1, showing the torque enhancements, the trend of both PT and SD enhancement are the same. Figure 2 shows selected tracking tests from the stroke subjects. Subject B gained an offset during natural tracking immediately after the first period in the test which reflects in a low PT enhancement but a high SD enhancement.
In 7 of the 9 stroke and 2 of the 3 SCI subjects the AutoMeCFES increased the torque and thereby muscle force. In 3 subjects (D, F and Y) the torque was not augmented; the enhancement scores are less than zero. The *IMax* was very low for F (see Table 1) because this person found the sensation of FES very uncomfortable. In the pre-test for D, it was revealed that stimulation did not increase force above maximum voluntary contraction.
None of the subjects had been trained in the tracking test and they only tried 2-4 tracking tests before the measurements were taken. This part of the study, therefore, does not take into account the effect of an eventual learning, which might improve the control.
Case story: Subject G declared that the foot was 'dead'; he was not aware of any volitional contraction whatsoever. During pre-tests of the session, the level of myoelectric activity was displayed instead of the torque without application of stimulation and the subject regained a volitional torque of 2Nm and 6Nm with MCS.
Immediately after the session, the subject was thus able to perform slight dorsiflexion against gravity. This can be interpreted as a short-term carry-over effect from the biofeedback that the AutoMeCFES provided.
The SCI subjects had curves morphologically similar to subject A but with less torque enhancement.

CONCLUSION

We conclude that, in some cases of paretic TA due to stroke or SCL, AutoMeCFES can be used to increase isometric force and that the control can be continuous rather than on/off.

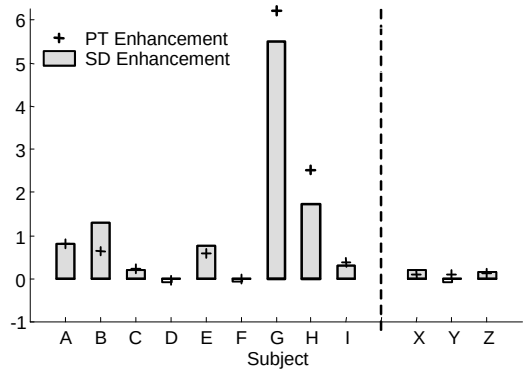


Figure 1. Torque enhancement calculated as both PT enhancement and SD enhancement. The data to the left of the vertical dividing line are the stroke subjects and to the right are the subjects with a spinal cord injury.

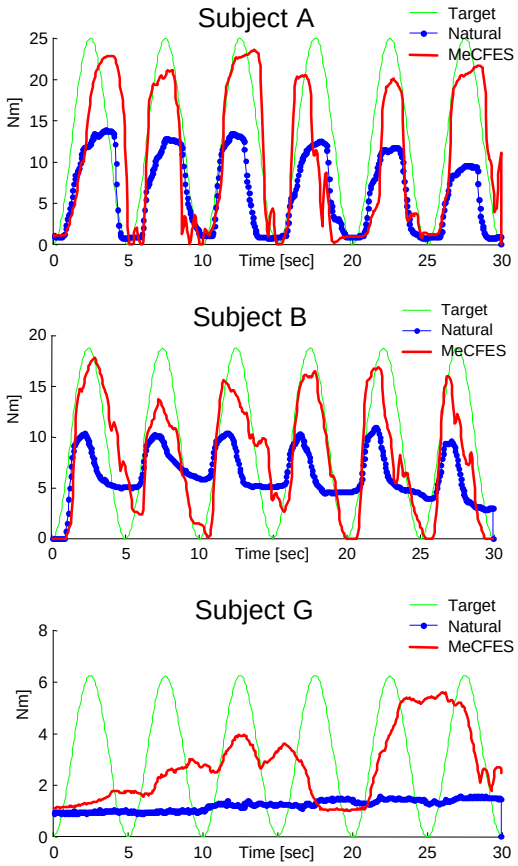


Figure 2. Three selected tracking tests. Top: An example of a good tracking test. Middle: A subject that was unable to return to the baseline between the periods, due to increase muscle tension. Bottom: A poor tracking but high torque enhancement.

REFERENCES

/1/ P. Taylor, J. Burridge, et. al., “Clinical audit of 5 years provision of the Odstock dropped foot stimulator”, *Artif Organs*, vol. 23, pp. 440-2, 1999.

/2/ W. T. Liberson, H. J. Holmquest, et. al., “Functional Electrotherapy: Stimulation of the Peroneal Nerve Synchronized with the Swing Phase of the Gait of Hemiplegic Patients”, *Archives of Phys. Med. Rehab.*, vol. 42: 101-105, 1961.

/3/ R. Dai, R. B. Stein, B. J. Andrews, K. B. James, and M. Wieler, “Application of tilt sensors in functional electrical stimulation,” *IEEE Trans Rehabil Eng*, vol. 4, pp. 63-72, 1996.

/4/ J. Cauraurgh, K. Light, et.al., “Chronic Motor Dysfunction After Stroke, Recovering Wrist and Finger Extension by Electromyography-Triggered Neuromuscular Stimulation”, *Stroke*, vol. 31, pp. 1360-64, 2000.

/5/ R. A. Schmidt and T. D. Lee, “Motor Control and Learning”, Human Kinetics ISBN 0-88011-484-3, pp. 24, 1999.

/6/ M. Popovic;, T. Keller;, I. P. I. Papas;, V. Dietz;, and M. Morari;, “Surface-stimulation technology for grasping and walking neuroprostheses,” *IEEE Engineering in Medicine and Biology Magazine* ,, vol. 20, pp. 82 -93, 2001.

/7/ R. Thorsen, R. Spadone, and M. Ferrarin, “A pilot study of myoelectrically controlled FES of upper extremity,” *IEEE Trans. Neural Syst. Rehab. Eng.*, vol. 9, pp. 161 -168, 2001.

/8/ R. Thorsen, J. Burridge, N. Donaldson, M. Ferrarin, J. Norton, and P. Veltink, “Auto-Myo-Electric Control of FES on Tibialis Anterior. A Pilot Study With Stroke Patients.,” *Proceedings of the 6 th Annual Conference of the International Functional Electrical Stimulation Society*, pp. 147-149, 2001.

/9/ R. Thorsen, “An artefact suppressing fast-recovery myoelectric amplifier”, *IEEE Trans Biomed Eng.*, vol. 46, pp. 764-6, 1999.

ACKNOWLEDGEMENTS

This study was only possible due to the helpful co-operation from stroke patients and medical staff from Het Roessingh Rehabilitation Institute, The Netherlands, during the preliminary investigation and Salisbury District Hospital, England during these experiments. This work has been supported by the European Union TMR programme NEUROS

AUTHOR’S ADDRESS

NeuroPro Post.Doc.
Rune Thorsen, Ph.D.
Centro di Bioingegneria
Fond. Don Gnocchi-Onlus
via Capecelatro, 66
I-20148 Milano, ITALIA

Tel:+39 0240308305
Fax:+39 024048919
EMail: Thorsen@mail.cbi.polimi.it

INITIAL RESULTS FROM TWO TRIALS OF AN IMPLANTABLE TWO CHANNEL DROP FOOT STIMULATOR

Laurence Kenney[1], Gerrit Bultstra[3] , Rik Buschman[1,2], Paul Taylor[4], Geraldine Mann[4],Hermie Hermens[1], Jan Holsheimer[3], Anand Nene[1], Martin Tenniglo[1], Hans van der Aa [2], John Hobby [4].

1. Roessingh Research and Development bv, Enschede, The Netherlands; 2. Twente Institute for Neuromodulation, Enschede, The Netherlands; 3. University of Twente, Enschede, The Netherlands; 4. Salisbury District Hospital, Salisbury, UK.

SUMMARY

This paper reports preliminary results of pilot studies of a new implantable two channel drop foot stimulator. The two subjects use the stimulator on a daily basis and have shown increases in walking speed between 10% and 44% when compared to their baseline measurements. Isometric tests have demonstrated that the stimulator allows repeatable and selective stimulation of ankle joint muscles.

STATE OF THE ART

There is a growing body of evidence, including one RCT, supporting the orthotic benefits of using single channel, surface-mounted drop foot stimulator for Cardio Vascular Accident (CVA) patients and others [1],[2]. The clinical benefits are now being seen by growing numbers of patients in the UK and mainland Europe, not just at the centres with a direct research interest in the technology but also elsewhere. This move away from the research centres into more widespread clinical usage can be seen as a coming of age for the technology. However, this has been a very slow and patchy process and even in the UK where clinical usage is now relatively widespread, the total number of patients being treated remains relatively small [2]. While less than desired funding is undoubtedly a factor, technical limitations have also caused problems, chiefly those inherent with the use of surface electrodes

In the past, surface stimulators have suffered from a number of practical problems, particularly associated with the foot switch and leads. Despite considerable effort on the part of engineers to replace the footswitch with an alternative, for now it remains the sensor of choice in clinical drop foot systems. The traditional problem with the footswitch and leads was a poor robustness, with fatigue failures commonplace. In recent years, this low-tech, practical but important problem has been tackled and footswitches and leads are now available that typically last in excess of 6 months daily use.

Nevertheless, the problems inherent to stimulating the Common Peroneal nerve using surface electrodes remain. These include a lack of selectivity over the muscles and nerves recruited, sensitivity of muscle recruitment to electrode placement and pain and tissue irritation associated with passage of current through the skin. Taylor [3] identified problems with locating the electrodes as the most common non-physiological reason for discontinuing use of the surface stimulator. These issues have long been recognised and various attempts have been made to implant a single channel drop foot stimulator system on the Common Peroneal nerve [4],[5]. However, this approach failed to solve the selectivity problem as it was not possible to control the relative proportions of dorsiflexion and eversion. This project attempts to resolve this problem by stimulating the two branches of the Common Peroneal nerve separately. The Deep Peroneal nerve innervates muscles that primarily dorsiflex and invert the foot, whilst the Superficial innervates everting muscles.

MATERIAL AND METHODS

Stimulator development

The stimulator that is based on transcutaneous RF-coupling was developed over several years at the University of Twente and Roessingh Research and Development [6]. It is based on a very simple receiver design, using basic passive components, encapsulated in silicon rubber. The novel aspect of the design lies in the type and location of the electrodes. The electrodes are sub-epineural type, developed for this application, but similar in design concept to certain electrodes used in pain relief applications. The location of the receiver distal to the knee avoids the need for the cabling to cross a joint, a common cause of failure in similar applications. The transmitter is located over the site of the receiver, and is triggered in the same manner as the conventional surface stimulator, using a foot switch.

Clinical protocol

The pilot study was intended to investigate the following questions: does the stimulator function as predicted, is it safe for use in humans and, are there any side-effects. The predicted functions were that its use would result in an improvement in gait, that stimulation response would be relatively insensitive to minor (1-2cm) changes in transmitter positioning and that selective stimulation of the two branches of the Common Peroneal nerve could be achieved. Ethics and regulatory approval for both trials were granted from the appropriate authorities.

The first implant took place in The Netherlands in July 2000 [7]. Since then, a further 3 implants have taken place, one in The Netherlands and two in the UK. The subjects are all CVA patients with a stable neurology, at least 3 years post stroke and aged between 31 and 48 years. Baseline data on walking speed and endurance were gathered on at least 3 separate occasions, both without and, with their normal walking aid (if any). The UK group also measured Physiological Cost Index (PCI) data. Prior to the implant operation nerve conduction measures were taken to check the integrity of the Deep (DPN) and Superficial Peroneal (SPN) nerves [8]. Following implantation all measurements were repeated. Furthermore, isometric torque measures were taken following a period of recovery and at regular time intervals at follow up. Isometric measures of ankle moment were taken using custom-built devices described elsewhere [9],[10].

RESULTS

The results of the first two implanted subjects in the Netherlands and the UK are presented. Figures 1 and 2 show the results of the walking speed and 6 minutes endurance measurements pre and post implant, respectively for the Dutch (NL) and English (UK) patient. The Dutch patient was an occasional walker with an AFO and therefore measurements with and without the orthosis were taken. When using the implanted system in both patients the walking speed and distance were increased by respectively 10% (UK) and 44% (NL) from mean baseline values. Figures 3 and 4 show typical graphs of the isometric response to stimulation at ‘optimal’ setting for the 2 subjects. The patients themselves defined the optimal setting. The stimulation times and ramping

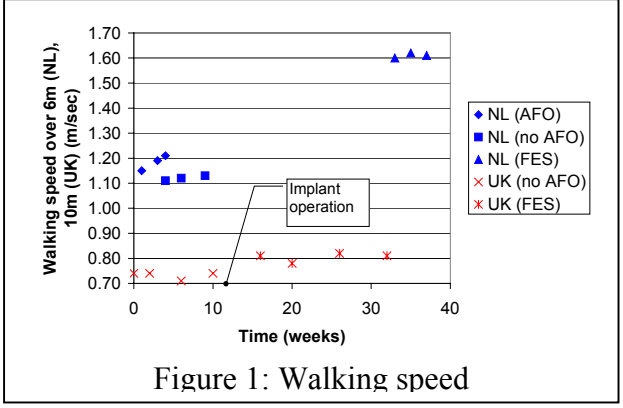


Figure 1: Walking speed

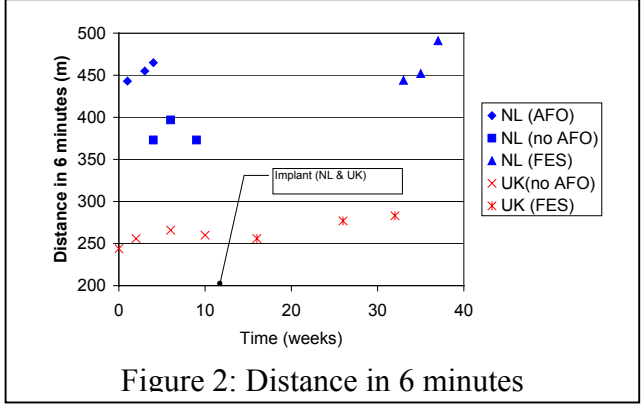


Figure 2: Distance in 6 minutes

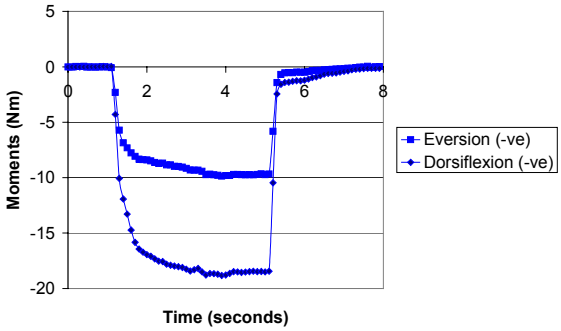


Figure 3: Isometric moments from stimulation of Deep and Superficial Peroneal nerves (NL)

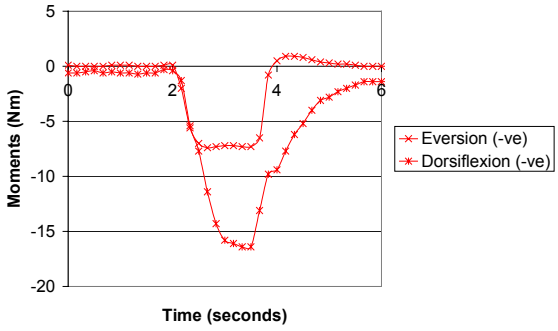


Figure 4: Isometric moments from stimulation of Deep and Superficial Peroneal nerves (UK)

varied between the Dutch and UK transmitter, due to minor changes in the settings for the two patients. These graphs show that at the onset of stimulation the force produced increased rapidly and was maintained at a stable level. After termination of stimulation the force rapidly declined.

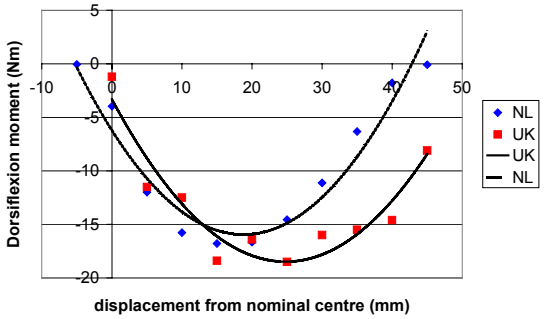


Figure 5: Sensitivity of dorsiflexion response to transmitter placement

The sensitivity of isometric response to transmitter movement was also investigated. In the case presented here, sensitivity was defined as change in dorsiflexion moment with proximal/distal displacement. Results from the UK and Dutch patients are shown in figure 5 (2nd order polynomial curve fitted to data). This graph shows that a displacement of about 1cm would not significantly affect the moment produced, indicating that the implantable system is relatively insensitive to minor positioning errors.

DISCUSSION

As may be seen in figures 1 and 2, both patients gained orthotic benefit using the stimulator. Taylor [11] showed a mean change in walking speed amongst CVA patients of 12% (0.07 m/sec) at 6 weeks and 27% (0.16 m/sec) at 4 ½ months use of the surface stimulator. The results obtained in the present study are similar to those reported by Taylor [11]. Figures 3 and 4 show that stimulation on user-defined optimal settings resulted in a response characterised by smooth dorsiflexion with moderate eversion. These results were repeatable both within, and between test sessions [10].

The sensitivity to transmitter placement was determined to be relatively low. There are no reports in the literature quantifying typical sensitivity to surface stimulation of the CPN, but experience suggests that it is significantly higher. This ease of positioning may be of specific benefit when we consider that typical CVA patients also have less control over their upper extremities. This new system may therefore also help these subjects to gain more independence.

The implants in two other patients have shown failures after having functioned properly for periods of months. An investigation of one of the explanted systems has shown that the system failure was caused by a fault in the receiver manufacturing process. Prior to failure, both of these patients also showed orthotic benefit from the device and similar isometric results to those reported here. The manufacturing process has now been adapted and a new receiver version is currently in production. Subject to regulatory approval, the clinical trials will continue in the near future.

REFERENCES

1. Burridge JH, Taylor PN, Hagan SA, et al. The effects of common peroneal stimulation on the effort and speed of walking: a randomized controlled trial with chronic hemiplegic patients. Clin Rehabil 1997;11: 201-210

2. Burridge J. Does the drop-foot stimulator improve walking in hemiplegia? Neuromodulation 2001;4: 77-83

3. Taylor PN, Burridge JH, Dunkerley AL, et al. Patients' perceptions of the Odstock Dropped Foot Stimulator (ODFS). Clin Rehabil 1999;13: 439-446

4. Waters RL, McNeal DR, Faloon W, et al. Functional electrical stimulation of the peroneal nerve for hemiplegia. Long-term clinical follow-up. J Bone Joint Surg Am 1985;67: 792-793

5. Strojnik P, Acimovic R, Vavken E, et al. Treatment of drop foot using an implantable peroneal underknee stimulator. Scand J Rehabil Med 1987;19: 37-43

6. Holsheimer, J., Bultstra , G., Verloop, A. J., van de Aa, H. E., and Hermens, H. J. Implantable dual channel peroneal nerve stimulator. Proc Ljubljana FES Conf (ed. Jaeger, R. and Bajd, T.) 1993: 43 -44.

7. van der Aa HE, Bultstra G, Verloop AJ, Kenney L, Holsheimer J, Nene A, Hermens HJ, Zilvold G, Buschman HPJ. Application of a dual channel peroneal nerve stimulator in a patient with „central“ drop foot. Proceedings of the World Federation of Neurosurgical Societies. 2001 (in press).

8. Lee HJ et al. Peroneal nerve conduction to the proximal muscles – an alternative to conventional methods. Am J Phys Med Rehab 1997;76:197-199.

9. Wood DE, Donaldson NN, Perkins TA. Apparatus to measure simultaneously 14 isometric leg joint moments. Part 2: Multi-moment chair system. Med Biol Eng Comput 1999;37: 148-154

10. Buschman HPJ, Kenney LJ, Nene AV, Bultstra G, Tenniglo M, Hermens HJ, van der Aa HE. Development and performance of an implantable 2 channel peroneal nerve stimulator for dynamic equinovarus foot in stroke. Proceedings IEEE-student branch. Eindhoven, The Netherlands May 2001

11. Taylor PN, Burridge JH, Dunkerley AL, et al. Clinical use of the Odstock dropped foot stimulator: its effect on the speed and effort of walking. Arch Phys Med Rehabil 1999;80: 1577-1583

ACKNOWLEDGEMENTS

The authors gratefully acknowledge the support of the EU (NEUROS TMR network) and the UK Medlink programme (IMPULSE project). Thanks also to Duncan Woods of Salisbury District Hospital.

AUTHOR'S ADDRESS

Dr Laurence Kenney,
Centre for Rehabilitation and Human Performance Research,
Brian Blatchford Building,
University of Salford,
Salford. M6 6PU. UK
e-mail: l.p.j.kenney@salford.ac.uk
home page: http://www.ihr.salford.ac.uk/

TOWARDS AUTOMATIC OPTIMIZATION OF GAIT SUPPORTED BY A TWO CHANNEL IMPLANTABLE DROP FOOT STIMULATOR

Peter H. Veltink*, Per Slycke***, Edwin Morsink***, Johan Hemssems*,
Gerrit Bultstra*, Hermie Hermens**

* Institute for Biomedical Technology (BMTI), University of Twente
** Roessingh Research and Development
*** Xsens Sports Technology b.v.
Enschede, the Netherlands

SUMMARY

An automated system is developed to optimize the timing and magnitudes of stimulation for a two-channel drop foot stimulator on the basis of criteria for the quality of the movement of the foot during the swing phase of gait. These criteria encompass the timing of the start of dorsiflexion movement after heel-off, foot clearance during the swing phase, and balance between inversion and eversion and heel-first foot placement at the end of the swing phase. These criteria are evaluated using a six degrees of freedom inertial sensor system placed on the foot. On the basis of this evaluation, the timing and magnitude of the stimulation are adapted from cycle-to-cycle.
The observability of the formulated movement criteria from the inertial sensor measurements have been preliminary evaluated in a patient experiment with and without stimulation of the peroneal nerve. The approach is intended to result in an evaluation and tuning system for clinical use.

STATE OF THE ART

Stroke may have many physical and cognitive consequences. Physically, the motor control of one side of the body may be deteriorated. Often, such a deteriorated motor function improves during the recovery period and by training, although in many cases motor control problems remain. Among others, a frequent consequence is the inability to voluntarily lift the foot at the affected side (drop foot). Since Liberson /1/, FES systems have been developed to artificially activate the dorsiflexor muscles during the swing phase of gait. More recently, a two channel implantable drop foot stimulator has been developed /2/, allowing electronic balancing of inversion and eversion during foot lift. In the past year, this system has been implanted in four stroke patients in Twente and Salisbury.
For effective application of the electronic balancing feature of this two channel implant, criteria for optimal movement patterns need to be stated, a sensor system for measuring relevant movement parameters of gait and procedures for the adjustment of the stimulation parameters both in time and amplitude of both channels need to be developed.

MATERIAL AND METHODS

Criteria for optimal gait improvement by drop foot stimulation

The potential improvement of gait through the use of a drop foot stimulator is not limited to the foot movement. It may prevent the lifting or circumduction of the affected leg due to drop foot gait. Foot landing after swing is also of primary importance. Heel first landing without excessive inversion or eversion contributes to stable gait and optimal loading transition from swing to stance. Heel-first landing also avoids excessive stretch reflexes in the calf muscle which may result from fast dorsiflexion when the forefoot first hits the ground /3/. Also at the onset of stimulation, fast dorsiflexion may elicit stretch reflexes in the calf muscles.

On the basis of these considerations criteria were formulated, concerning three phases of the walking cycle: the transfer from stance to swing phase, the swing phase and the transfer from swing to stance phase:

1. transfer from stance to swing:
   1a. dorsiflexion starts after heel-off, allowing for push-off, if available
   1b. dorsiflexion with limited angular velocity, avoiding excessive calf muscle stretch reflexes.
2. mid-swing:
   2a. foot clearance during whole swing phase without excessive heel lift or circumduction
   2b. limited inversion/eversion of the foot.
3. transfer from swing to stance:
   3a. heel-first landing
   3b. limited inversion/eversion of the foot

Assessment of foot movement during the swing phase of gait

Each of these criteria can be conceived as the reference value for related movement parameters (e.g. time of start of dorsiflexion movement after heel-off, foot clearance during mid-swing, foot orientation at foot contact). These movement parameters can be assessed for each walking cycle using a six degrees of freedom inertial measurement unit (IMU) attached to the foot. The IMU consists of 2 dual-axis accelerometers (Analog Devices ADXL210) and 3 uni-axial angular rate sensors (Murata ENC-03J). Advantages of micro-machined inertial motion sensors are that they do not need external sensors or references to operate, their small size and low cost. When interpreted correctly an IMU will not only provide angular velocity and linear acceleration but also position and orientation with respect to an inertial frame of reference. Orientation is optimally estimated using sensor fusion and Kalman filtering techniques /4/. Here, however, straightforward integration and coordinate system transformation is used.

Adequate application of initial conditions during the stance phase of gait (zero linear and angular velocity of the foot) and adequate subtraction of the gravity acceleration component result in accurate calculation of the 3D orientation and position of the foot during the swing phase without integration drift. This has first been used in a step-to-step velocity meter for runners (RunnersWatch™ by Xsens). It should be noted that the orientation and position of the foot are calculated with respect to the inertial reference frame, not with respect to the shank, since for most movement criteria the orientation and movement of the foot with respect to the floor is of interest. If an additional inertial sensor system is integrated with the transmitter or implant of the stimulator on the shank, additional information about the movement of the foot relative to the shank is available.

Automatic tuning of a two-channel drop foot stimulator

The timing and amplitude of the stimulation patterns generated by the two-channel implantable drop-foot stimulator can be adjusted automatically by a controller consisting of two levels:

1. finite state detection of gait phases, especially transitions from stance to swing and from swing to stance, including heel- and toe-off moments. Control of the timing of the stimulation (onset, ramp-up, ramp-down, offset time) is relative to the timing of the states assessed using this finite state control scheme.
2. A cycle-to-cycle control scheme /5/ adjusts the timing and amplitude parameters of the stimulation. This discrete-time control scheme determines the amplitude and timing parameters of stimulation for all cycles of gait on the basis of the evaluation of the movement parameters of all cycles in relation to the criteria. The timing parameters are relative to the phases determined by the finite state detection scheme.

RESULTS

Assessment of foot movement during the swing phase of gait

The feasibility of the assessment of the movement parameters is illustrated in figure 1. 3D foot orientation and position during the swing phase are shown for one step of a stroke patient walking with and without the application of a external drop foot stimulator. Without stimulation (figures 1a,b), this example patient walks with extreme eversion of the foot (positive x-rotation in fig. 1a), which, among others, results in non-optimal foot contact during the transfer from swing to stance phase. This is improved with surface stimulation (figure 1c,d).

without stimulation *with stimulation*

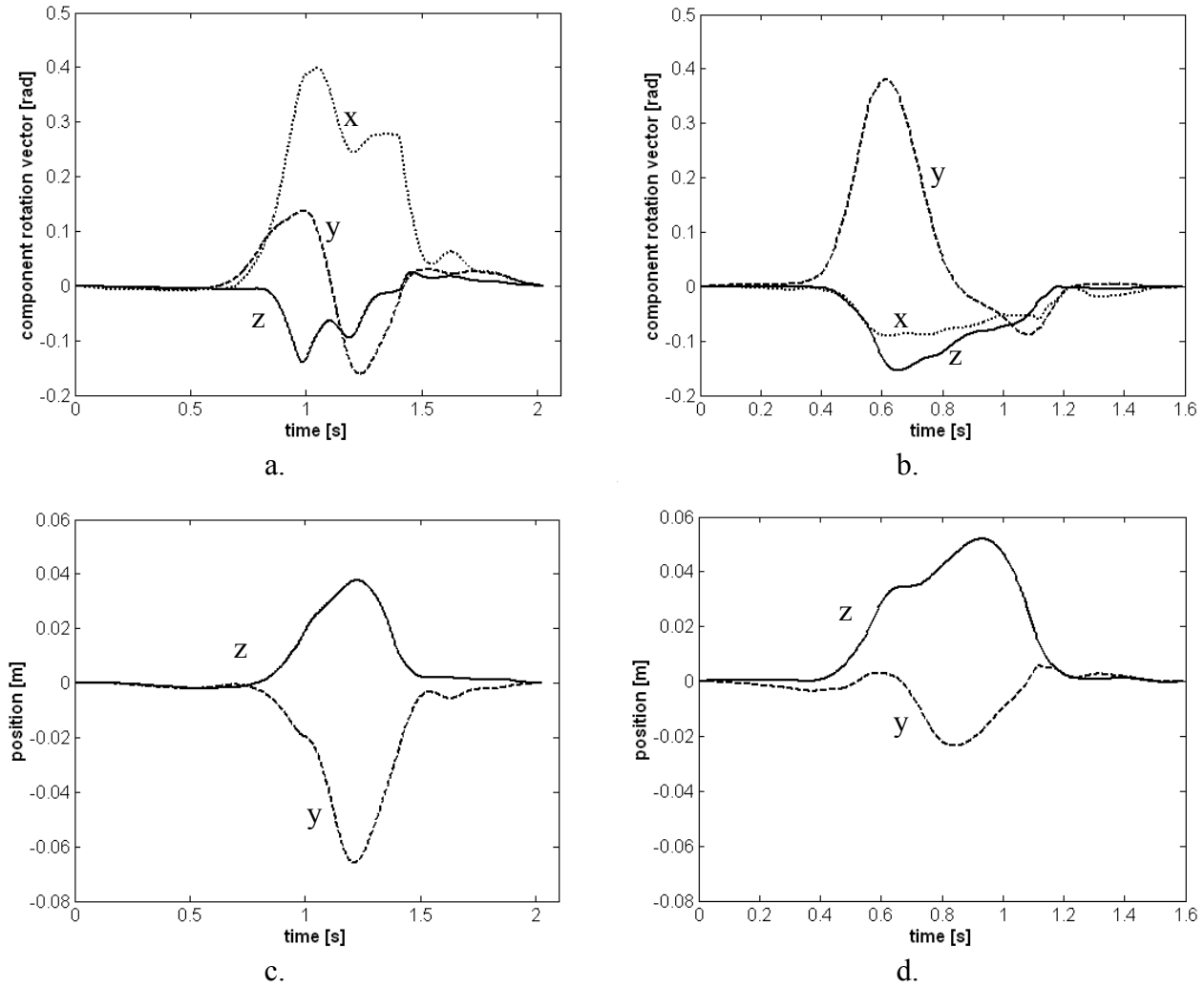


Figure 1: Example foot orientation (a,b) and position (c,d) with respect to the inertial reference frame during swing phase of gait for a stroke patient with (b,d) and without (a,c) surface stimulation of the peroneal nerve, as measured by a 3D inertial measurement unit on the foot. Dotted (x) is walking direction, dashed (y) is medial/lateral direction, solid (z) is vertical direction

Relevant movement parameters can be derived from these measurements. As an example, figure 2 shows one of the movement parameters, the inversion/eversion angle at the instance of foot contact, as a function of cycle number without and with stimulation.

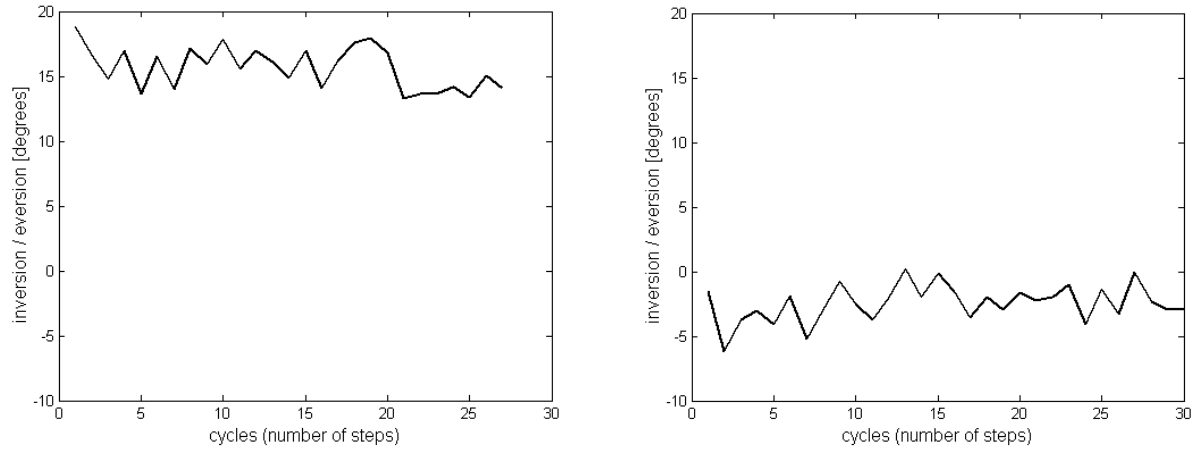


Figure 2. Example registration of one of the movement parameters, the inversion/eversion foot angle (orientation around x-axis) at the instance of foot contact, as a function of cycle number without (a) and with stimulation (b) (same stroke patient as in figure 1)

DISCUSSION

The preliminary results of movement measurement and cycle-to-cycle movement parameter extraction are the basis for the automatic adjustment of stimulation patterns for the two-channel implantable drop foot stimulator. In first instance, the automatic adjustment may only be performed during clinical sessions. In this case, the foot sensor module is only applied during these clinical sessions and not during daily use. An inertial sensor module integrated with the transmitter or implant stimulator on the shank is used for phase detection during daily use /6/. This avoids the daily use of sensors on the foot, which has been a major practical drawback of dropfoot stimulators using footswitches. If, eventually, the foot inertial sensor unit can be integrated in the sole of the shoe and telemetrically linked to the stimulator, the continuous use of the cycle-to-cycle controller during daily application of the stimulator is an option. This would enable continuous adjustment of stimulation patterns when muscles fatigue or when the walking surface, walking speed or other circumstances change.

REFERENCES

/1/ Liberson W.T., Holmquest H.J., Scott D., Dow A., Functional Electrotherapy: stimulation of the peroneal nerve synchronized with the swing phase of the gait of hemiplegic patients, Arch. Phys. Med. Rehab., 1961, 42, 101-105.
/2/ Holsheimer J., Bultstra G., Verloop A.J., van der Aa H.E., Hermens H.J., Implantable dual channel peroneal nerve stimulator, Ljubljana FES Conference, 1993, 42-44.
/3/ Veltink P.H., Ladouceur M., Sinkjær T., Stretch reflex contribution to soleus activation during spastic gait, Proc. 20th Annual Int. Conf. IEEE-EMBS, Hong Kong, 1998, 2328-2331.
/4/ Luinge H.J., Veltink P.H., Baten, C.T., Estimating orientation with gyroscopes and accelerometers, Technology and Health Care, 1999, 7,455-459.
/5/ Veltink P.H., Control of FES-induced cyclical movements of the lower leg, med. Biol. Eng. Comput, 1991, 29, NS8-12.
/6/ Willemsen A.Th.M., Bloemhof F., Boom H.B., Automatic stance-swing phase detection from accelerometer dataa for peroneal nerve stimulation, IEEE Trans. Biomed. Eng., 1990, 37, 1201-1208.

ACKNOWLEDGEMENTS

Financial support from the European Union (project neuralPRO) and the Dutch ministry of economic affairs (IMPULSE) and the help of Dr. Rik Buschman are gratefully acknowledged.

AUTHOR'S ADDRESS

Prof. Peter H. Veltink
Faculty of Electrical Engineering /BMTI
P.O. Box 217, 7500 AE Enschede,
the Netherlands

tel. x31-4892765; fax x31-4892287
e-mail: P.H.Veltink@el.utwente.nl
home page: http://www.utwente.nl/bmti/
http://www.rrd.nl ; http://www.xsens.com

AN EXPERIMENTAL EVALUATION OF THE GYROSCOPE AS A SENSOR IN FES FOOT-DROP CORRECTION SYSTEMS

SN Ghoussayni, JR Henty, DE Wood*, DJ Ewins

Biomedical Engineering Group, University of Surrey, Guildford, UK
*Dept. of Medical Physics and Biomedical Eng., Salisbury District Hospital, Salisbury, UK

SUMMARY

This paper presents the work done on the use of a Gyroscope (Gyro) as a sensor for foot-drop correction systems using Functional Electrical Stimulation (FES). These systems usually employ foot switches to control the timing of stimulation. It is believed that the replacement of the heel switch with the gyroscopic sensor would offer several advantages. The performance of the new sensor and associated software in gait event detection have been tested and compared to foot switch and kinematic data. The tests were carried out with both able-bodied and hemiplegic patients walking over different terrains. The results indicate that the Gyro can be used to detect the necessary gait events for controlling the stimulator timing.

STATE OF THE ART

Since the work of Liberson in 1961 /1/, many researchers have been interested in the application of electrical stimulation for the restoration of lost or impaired function. The correction of foot-drop is one common application with increasing clinical use. FES foot-drop stimulators typically employ a physical sensor (usually a foot switch) as a feedback source controlling the timing of stimulation. These sensors are reported – by more than one study /2/, /3/, /4/ – to suffer from a number of limitations, including detection errors, in particular with pathological gait, and a relatively short mean time to failure with some patients, for example children with cerebral palsy. Previous work at the University of Surrey /5/, /6/, has investigated the use of a gyroscopic sensor as an alternative sensor to be used with FES foot-drop correction systems. The use of a vibratory gyroscope in combination with other sensors has also been investigated by other researchers /7/, /8/. A wide variety of other sensors, both artificial and natural, has been investigated in detail to be used with FES systems. Accelerometers, goniometers, and inclinometers are some of the artificial sensors that have been researched for use with FES systems /2/, /4/. Recordings of both electroneurogram (ENG) and electromyogram (EMG) have also been studied as possible natural sensors for FES systems /9/, /10/.

The replacement of the foot switch by the Gyro offers several advantages, which could improve system reliability and function. The Gyro is a small and lightweight sensor (with potential for further miniaturisation and implantation), which can be easily donned and doffed (positioning is not very critical) with minimal encumbrance to the patient. The nature of the Gyro contributes to its high reliability and long lifetime during which there is little or no deterioration in its performance. When compared to the information provided by foot switches, which is essentially of a binary nature, the Gyro based sensor system has the advantage that it could predict a gait event - and therefore begin, end, ramp up or ramp down stimulation - before its occurrence.

As a replacement for foot switches in FES foot-drop correction systems, the sensor output (voltage proportional to angular rate of moving limb segment) has to be adapted to be suitable for timing the stimulation. In our work this has been done by developing detection software for the Gyro signal to give four gait events (Heel Contact HC, Foot Flat FF, Heel Rise HR, and Toe Off TO). A portable microcontroller based unit with data logging capability was developed. This unit runs the gait event detection software and provides real time detection. This system was used for the evaluation of the sensor

and associated software by comparing the detection times to those obtained from simultaneously recorded foot switch and kinematic data (kinematic data was not available for evaluating the system when used by the patients, as these tests were not performed in a lab setting). The evaluation of the system involved its use by both able-bodied subjects and patients with foot-drop, and the system performance was tested over different terrains including leveled and inclined walking, and staircase climbing.

MATERIAL AND METHODS

A Murata[a] ENC-05E piezoelectric vibrating gyroscope was used to capture the angular velocity of the foot in one plane (worn on the anterior side of foot). The output of the Gyro and two foot switches (located under the heel and first metatarsal head) was sampled by the data logger at 164 Hz. The tests were split into two sets.

In the first set of tests, the system was used by five able-bodied subjects (average age 38 years), and simultaneous kinematic data was captured using a Qualysis ProReflex motion capture system sampling at 240 Hz. Three retroreflective markers were attached to the foot at the lateral malleolus, the lateral side of calcaneus, and above the first metatarsal head. The positions of these makers were visually inspected afterwards to determine the times of the gait events. Each subject performed six trials walking on leveled floor, up and down a 7º ramp, and a 7 step staircase, at two speeds (self-selected normal and slow). The performance of the sensor was analysed by comparing its predicted times to those given by the foot switches, and then by comparing both methods against the kinematically determined gait events.

In the second set of tests, the system was used by four patients with foot-drop condition (average age 53 years, 3 females and 1 male, 2 MS and 2 CVA patients, average time since diagnosis 4 years, average time since stimulator was first used 5 months). The patients were asked to perform two trials: with and without using their Odstock Dropped Foot Stimulator (ODFS, Salisbury District Hospital, Salisbury, UK). The data from the Gyro and two foot switches was collected using the data logger while the patients walked around the rooms of the hospital over both inclined and leveled floor. The data was then analysed by comparing the Gyro and foot switches detection times for each gait event (foot switch detection times were determined by thresholding). Two of the patients also ascended a stairway, however this data is still being processed.

RESULTS

For the first part of the study, the differences in time between the three methods are presented in Tables 1-3. These show the average difference between each of the methods used for different events against the three terrains.[b] The absolute difference was also calculated and averaged to avoid any misleading conclusions from considering the average difference alone.

Table 1: The results of the comparison of detection times (in ms) between the Gyro-Kinematic method. (D = Average Difference; |D| = Average of Absolute Difference)

Event\Terrain	Level Floor		Ramp Up/Down		Stairs Up/Down	
	D	|D|	D	|D|	D	|D|
HC	-7	28	-11	25	-47	95
FF	22	25	28	30	37	47
HR	4	29	25	68	110	136
TO	56	58	69	69	80	98

[a] Murata Electronics (United Kingdom) Ltd., Hampshire, GU13 8UN, UK.

[b] Negative values indicate that the predicted time was earlier than the time given by the second method. For Table 1 this would be the kinematic method.

Table 2: The results of the comparison of detection times (in ms) between the Foot switch-Kinematic method.

Event\Terrain	Level Floor		Ramp Up/Down		Stairs Up/Down	
	D	|D|	D	|D|	D	|D|
HC	31	37	42	63	55	111
FF	43	79	15	64	11	58
HR	-44	81	-65	112	-73	108
TO	-29	90	23	73	-47	78

Table 3: The results of the comparison of detection times (in ms) between the Gyro-Foot switch.

Event\Terrain	Level Floor		Ramp Up/Down		Stairs Up/Down	
	D	|D|	D	|D|	D	|D|
HC	-38	38	-53	57	-102	160
FF	-21	78	14	66	-4	64
HR	48	99	90	112	179	179
TO	85	118	45	101	128	138

The results from the second part of the study are presented in Table 4. As the trials with the 4 hemiplegic subjects were performed in a non-lab setting, the data collected is from the Gyro and 2 foot switches only.

Table 4: The results of the comparison of detection times (in ms) between the Gyro-Foot switch when used by 4 hemiplegic patients walking on leveled and inclined floor.

Event\Trial	No Stimulation		With Stimulation	
	D	|D|	D	|D|
HC	-7	48	-3	32
FF	-3	16	42	51
HR	118	135	86	156
TO	98	98	-3	144

DISCUSSION

The results from the first part of the study show that when compared to the foot switches detection times the Gyro had a similar performance and was generally closer to the kinematically determined times. As the trials with the 4 hemiplegic subjects were performed in a non-lab setting, the data collected was from the Gyro and 2 foot switches only. This meant that the Gyro and foot switch times could not be compared against a third method. However, one observation is that the differences between the Gyro and foot switch detection times (Table 4) are comparable to those from the first set of results (Table 3). It is suggested that this is because the sensor system performed equally well as in the first part of the study. It can be concluded from the presented results that the Gyro sensor system is capable of detecting four gait events in both able-bodied and hemiplegic patients. The sensor also appears to be appropriate for staircase and slope walking.

After some necessary modifications to the software, the sensor was also tested when worn on the anterior aspect of the shank. The preliminary results for this are promising. The use of the sensor when worn on the shank may offer a more convenient use with potential for developing a miniature unit encompassing the sensor, stimulator, electrodes and microcontroller. Future work includes completion of analysis of patient data, and further testing of the system when used on the shank. This will be followed by real time use with the stimulator, evaluation of patient feedback and the development of a take home system.

REFERENCES

/1/ Liberson WT, Holmquest HJ, Scot D, Margot D. (1961). Functional Electrotherapy: Stimulation of the Peroneal Nerve Synchronised with the Swing Phase of the Gait of Hemiplegic Patients. Archives of Physical Medicine and Rehabilitation, 42, pp 101-105

/2/ Dai R, Stein RB, Andrews B. (1996). Application of Tilt sensors in Functional Electrical Stimulation. IEEE Transactions on Rehabilitation Engineering, 4, pp 63-71

/3/ Taylor PN, Burridge J, Ewins DJ, Swain ID. (1995). A Two Channel Stimulator for Gait Assistance. Proceedings of the BES Symposium on Electrical Stimulation – Clinical Systems, University of Strathclyde, pp 41-42

/4/ Willemsen ATM, Bloemhof F, Boom HBK. (1990). Automatic Stance-Swing Phase Detection from Accelerometer Data for Peroneal Nerve Stimulation. IEEE Transactions on Biomedical Engineering, 37, pp 1201-1208

/5/ Henty JR, Ewins DJ. (1998). Applications of Gyroscopic Angular Velocity Sensors in FES Systems. Proceedings of the 6th Vienna International Workshop on Functional Electrical Stimulation, Vienna, pp 157-160

/6/ Henty JR, Wood DE, Ewins DJ. (1999). Detection of Gait Events using a Vibratory Gyroscope. Proceedings of the 4th Annual conference of the international FES Society, Japan, pp 73-76

/7/ Popovic MR, Keller T, Ibrahim S, Bueren GV, Morari M. (1998). Gait Identification and Recognition Sensor. Proceedings of the 6th Vienna International Workshop on Functional Electrical Stimulation, Vienna, pp 153-156

/8/ Williamson R, Andrews B. (1997). Sensors For FES Control. Proceedings of the Second Annual IFESS Conference and Neural Prosthesis: Motor Systems V, Canada, pp 213-215

/9/ Strange KD, Hoffer JA. (1999). Gait Phase Information Provided by Sensory Nerve Activity DuringWalking: Applicability as State Controller Feedback for FES. IEEE Transactions on Biomedical Engineering, 46, pp 797-809

/10/ Kershaw RA, Jones R, Bateman A. (1993). The Use of EMG for Real Time Closed Loop Control of Functional Electrical Muscle Stimulation. Proceedings of The Ljubljana FES Conference, Ljubljana, pp 123-125

ACKNOWLEDGEMENTS

The authors would like to thank the University of Surrey, the Engineering and Physical Sciences Research Council, and the Lebanese National Council for Scientific Research, and the staff and patients at Salisbury District Hospital for supporting this project.

AUTHOR'S ADDRESS

Salim Ghoussayni
Biomedical Engineering Group
School of Engineering
University of Surrey
Guildford, Surrey
GU2 7XH
United Kingdom

e-mail: s.ghoussayni@surrey.ac.uk
home page: http://www.bmesurrey.org

QUALITY PRODUCTS AND QUALITY FOR PATIENTS -- ISO9002 FOR MANUFACTURE AND CLINICAL SERVICE.

S.E. Crook, I.D. Swain, P.N. Taylor, S. Finn.

Department of Medical Physics and Biomedical Engineering,

Salisbury District Hospital, Salisbury, UK

SUMMARY

In 1998 it became illegal to supply a Medical Device within the EC without a CE mark. The purpose of the mark was to harmonise the various safety requirements of the Member States which were otherwise becoming a barrier to trade. Companies or Institutes manufacturing medical devices now need to show documented compliance with the Medical Devices Directive (MDD) /1/. It was no longer sufficient to have engineers taking professional responsibility for their work. To demonstrate consistency and good quality the MDD required that a Quality Management System (QMS) be in place as part of the manufacture of medical devices.

The Odstock Dropped Foot Stimulator (ODFS) is a body worn, battery operated stimulator which was being increasingly widely used across the UK. Review of the MDD showed that an externally verified QMS would be required if we wished to continue to supply the ODFS outside of our Institution. Since the clinical service was increasing as well, the decision was made to include both the manufacture and patient treatment in our QMS assessment. The route to this successful implementation is described along with the advantages and drawbacks encountered.

STATE OF THE ART

At the time of the June 1998 deadline for CE marking commercial companies had moved to comply with the MDD for products they wished to continue to sell. Most hospital departments viewed the work involved as excessive but the ODFS was a successful product and in Salisbury preparations were started over a year before the deadline.

In the UK the National Health Service (NHS) was in a period where internal (between hospital) competition was encouraged. It was considered that an externally recognised validation of the quality of our patient treatment would give us an advantage in this environment. It is important to note that external assessment is only concerned with how well we deal with patients or how consistently we can make stimulators. It is not an endorsement of the effectiveness of our treatment.

METHOD

Choosing the Assessment route through the MDD

The first step is to confirm that your product is a Medical Device within the definition of the MDD. The ODFS is a Medical Device because it's purpose is "alleviation of or compensation for an injury or handicap" (MDD, Art 1, para 2(a)). By reference to Annex IX of the MDD a muscle stimulator that uses skin surface electrodes falls into Class IIa. This was confirmed with the Medical Devices Agency which is the UK Competent Authority overseeing implementation of the MDD.

Once the class of your device has been established it is necessary to choose a Conformity Assessment procedure (Art. 11, MDD). For the Department in Salisbury it was decided to comply with Annex V (Production quality assurance) which would require a QMS verified by a Notified Body. A Notified

Body is simply a company that has been approved by the Competent Authority to award a certificate showing compliance with the MDD. The MDD does not specify how a QMS should be designed, only that it should be externally verified by a Notified Body. In practice this means that the international standard ISO9000:1994 /2/ should be used since Notified Bodies are familiar with systems based on this document so accreditation would be more straight forward.

(N.B The 1994 edition of ISO9000 is being replaced by ISO9000:2000 with a phased assessment program for existing certificate holders.)

Starting up a Quality System

Upon reading the ISO9000:1994 set of documents a decision had to be made as to which individual standard would be used. Since we were not offering a design service for our customers we would not generate appropriate records for ISO9001 which included design. ISO9002 covers production from start to finish of an existing design and ISO9003 covers testing of a finished product. Since we would be developing our own products we wanted a system that covered their manufacture so ISO9002:1994 was chosen. In the case of medical devices a supplemental standard, BS EN 46002:1994 /3/, is also required and this was included when the quality system was being developed.

Stimulator Manufacture

This activity was judged to be a priority since non-achievement of the ability to CE mark would prevent us supplying stimulators outside of our hospital. A Quality Manager was designated to oversee the introduction and running of the system. In order to reduce the time taken to compile the procedures required by ISO9002 a management consultant was employed. In consultation with the staff doing a particular job he was able to look at the way we were working and write a procedure that would satisfy the requirements of the standard. In almost every area the requirement was for traceable records to be made showing that a specific check or activity had been carried out. Therefore record books had to be modified or introduced to hold these records.

In order to affix a CE mark a Technical File for each medical device needs to be kept. This will contain manufacturing and design drawings, risk assessments, product compliance with other standards, instruction manuals and so on. Approximately half of the required documents were available before the Technical File was put together for the ODFS. The remainder had to be generated specifically.

An external company was used to verify compliance with EMC standards and the pass certificate was inserted in the Technical File. Since this is a costly process some pre-testing was done at a local EMC facility and slight modifications made in order to guarantee compliance. For compliance with the applicable electrical safety standard (BS EN 60601-1 /4/) a checklist was completed by the designer and signed off before insertion into the Technical File.

Once the procedures had been in use for several months an audit of the QMS was carried out to determine how well it was working.. The audit is itself a requirement of ISO9002 and any shortcomings need to be addressed by staff training or modification of the procedures. Staff not involved in the area being audited should be used to carry out the audit. This prevents assumptions being made about the activity being carried out and ensures the auditor follows the procedures and records applicable for that area.

At audit or during normal operation a failure in the QMS needs to receive attention. Such failures are typically called non-conformances and they can arise from a manufactured item not reaching the required specification or a delivery containing the wrong goods etc. Non conformances are logged on

a form together with the measures taken to correct any problems. These forms are then reviewed at management meetings where measures to prevent any recurrence are discussed and implemented. When an external assessor reviews the operation of the QMS these non-conformances and the actions taken will be examined so minutes should be kept in sufficient detail to permit this.

A Quality System for Patient Treatment

Our method of treating patients with the ODFS had been carefully worked out over several years /5/ with input from engineers and physiotherapists. Several forms were in use in the clinic to collect data about patient performance and equipment settings. This data was then used to demonstrate the effectiveness of the device and highlight any perceived problems. The results were published in the scientific press.

Introduction of the QMS only required slight modification of these clinic forms to record the person completing them, some equipment details and confirmation that the patient had received instructions. A procedure was written to cover the operation of the clinic which then specified the use of the new forms.

As patient numbers increased the administration for the appointments and funding also required a procedure to be written. The ISO9002:1994 standard uses language that is very focused on 'the customer' and the contract between the customer and supplier. Whilst the obvious conclusion would be that the patient was our customer, in practice our contract is with the doctor or health authority. Contract review is a central ISO9000 requirement so it is these people that need to be informed about our service before we can agree to see their patient(s). This required information on the treatment to be sent to the referring doctors before their patients could be seen.

The Assessment

The Notified Body selected to carry out the external assessment of our QMS was the British Standards Institute (BSI). A Client Manager was assigned to us on the basis of our operation, i.e. electrical engineering for medical device production. A pre-assessment visit was made to ensure that the system we had was ready for assessment both on ISO9002 and the MDD. Since our Client Manager felt that the patient treatment aspect of our work fell outside his expertise a second BSI Assessor was brought in to do this assessment at later date.

Assessment of the QMS for stimulator CE marking took two days initially and a number of shortcomings or non-conformances were discovered. However, these were not sufficiently serious to prevent granting of the EC Certificate and a Certificate showing compliance with ISO9002. A plan showing how these non-conformances would be rectified had to be submitted to the assessor and approved before the certificates were awarded. After six months an assessor for the patient treatment came to look at this aspect of our operation in addition to a follow up visit for the manufacture. Our patient treatment actives were then added to the ISO9002 certificate.

Assessments then take place at six monthly intervals to ensure the QMS is being maintained. In addition the assessor will wish to see how the QMS system is being used to improve quality by monitoring such activities as repair, customer complaints and other non-conformances.

RESULTS AND DISCUSSION

Benefits

The major beneficial result of achieving the ISO9002 registration is that we can continue to make stimulators for supply outside our hospital. This extends the benefit of the treatment and generates

some income to continue development of the ODFS and other devices. We now supply to several countries in Europe. For this we need to translate the instruction manuals and other safety information (e.g. labelling)

Records that are kept as part of the QMS have improved the way we run the manufacture and especially the patient treatment. Any other devices we design can be brought into our system and CE marked. Similarly we can take on the role of manufacturer for products designed elsewhere and CE mark those.

Drawbacks

There is some extra work to do and there was staff resistance to overcome. However, the day to day running of the QMS is very smooth and the kind of records we now keep should probably be kept by a responsible department in any case. When audits, reviews and assessments take place there is a high workload on the Quality Manager which should not be underestimated. In addition there are costs for the Notified Body amounting to £2500/year.

Conclusions

Whenever some question arises as to the value of the QMS and the other documentation it is helpful to consider what would happen if a product we had manufactured were involved in a serious incident that harmed a patient. In a court room or other inquiry questions would be asked about the quality of our products. In such a case there is ample documentation to demonstrate that our products have been properly designed and made. This was probably not the case before the MDD requirements. There are also continuous benefits from the improvement in record keeping and the documentation of procedures.

REFERENCES

/1/ Official Journal of the European Communities L169, vol. 36, 12 July 1993: Medical Devices Directive 93/42/EEC HMSO ISBN 0119122138

/2/ ISO9002:1994(E) Quality Systems - Model for quality assurance in production, installation and servicing. ISO Switzerland.

/3/ BS EN 46002:1994, Application of EN29002 to the manufacture of medical devices. BSI 1994. ISBN 0 580 21813 9

/4/ BS EN 60601-1:1990 Medical Electrical equipment Part 1 General requirements for safety 1997 BSI, London

/5/ Burridge, J. Taylor, P. Swain, I. (1997) Clinical Experience of the Odstock Drop Foot Stimulator. Artificial Organs 21 (3): 254-260

AUTHORS ADDRESS

Dr. Steven Crook
Department of Medical Physics
Salisbury SP2 8BJ, UK

email: s.crook@salisburyfes.com
www.salisburyfes.com

Session 10

UPPER EXTREMITY, FUNCTIONAL RESTORATION

RESTITUTION OF REACHING AND GRASPING PROMOTED BY FUNCTIONAL ELECTRICAL THERAPY

M.B. Popovic[1,3], D.B. Popovic[1], T. Sinkjær[1], A. Stefanovic[2], L. Schwirtlich[2]

[1]Center for Sensory Motor Interaction, Aalborg University, Denmark
[2]Institute for Rehabilitation "Dr Miroslav Zotovic", Belgrade, Yugoslavia
[3]Institute for Medical Research, Belgrade, Yugoslavia

SUMMARY

The Functional Electrical Therapy (FET) is a new term describing a combination of functional electrical stimulation that generates life-like movement and intensive exercise in humans with central nervous system lesions. We hypothesized that the FET can promote a significant recovery of functioning if applied in sub-acute stroke subjects. The study included 16 stroke subjects divided in a low functioning group (LFG) and a high functioning group (HFG) based on their ability to control wrist and fingers, and randomly associated into the FET and controls. The FET consisted of 30-minutes daily sessions during three weeks. The exercise comprised functional use of daily necessities (e.g., writing, using a telephone receiver, drinking from a can). The outcome presented in this paper is the upper-extremity function test performed before and after the therapy. The change in performance of the HFG group was significant: 1) the number of successful repetitive movements in two minutes was doubled, and 1.6 times increased for controls; and 2) the time to perform the movement was decreased by 71 percent, and by 36 percent in controls. In the LFG FET group the difference in performance was the following: 1) the number of tasks was increased from zero to six tasks (total of 11 tasks); and 2) the averaged number of successful repetitive movements from zero to three. The functional improvement in FET LFG is probably not sufficient to make the more affected arm/hand effective for daily necessities; thus, the FET effects could deteriorate at longer time. The subjects from a control LFG made only a marginal improvement. The study will continue for the next 18 month.

INTRODUCTION

In one of the earliest studies Merletti *et al.* /1/ applied a two channel functional electrical stimulation (FES) in order to augment the elbow and fingers/wrist extensions. The conclusions were that FES contributed greatly to recovery of hand and elbow movements in five stroke subjects, yet in the remaining three the improvement was significant only at the elbow joint. Electrical therapy has been applied as a therapy in humans with CNS injuries although there are no conclusive directions which technique works the best for a given indication. Kraft *et al.* /2/ reported that subjects assigned to electrical therapy improved their aggregated Fugl-Meyer score significantly from pre-treatment to post-treatment, and the improvement was maintained at three- and nine-month follow-ups. Feys *et al.* /3/ used FES for six weeks in stroke subjects. Subjects performed better on the Fugl-Mayer test compared to the control group throughout the study period, but differences were significant only at follow-up. 26 subjects were randomly assigned to receive either neuromuscular stimulation or placebo /4/. The treatment group received surface neuromuscular stimulation to produce wrist and finger extension exercises, and the controls received placebo stimulation over the paretic forearm one hour per day, for a total of 15 sessions. Parametric analyses revealed gains in Fugl-Meyer scores for the treatment group immediately, and after four and twelve weeks of treatment. Francisco *et al.* /5/ and Cauraugh *et al.* /6/ reported results that suggest that the EMG-triggered FES is effective for rehabilitating wrist and finger extension. The Handmaster NMS-1 system is becoming widely used for the therapy in stroke subjects /7/. The Handmaster NMS-1 uses three-channel FES to control opening and closing of the fingers and the thumb opposition. The common conclusions from all studies are that combined electrical stimulation and extensive physical exercise with enhanced feedback contribute to the recovery, and that the contribution is more exposed if the treatment is applied timely, *i.e.,* shortly after the stroke. Evaluation of the Bionic Glove /8/, and Belgrade Grasping System /9/ in chronic tetraplegic subjects showed that FES

improves the reach and grasp. The results indicate important short- and/or long-term carry-over effects. This paper presents the results from a study in progress where FET is applied to the more affected arm in sub-acute stroke subjects. The FET uses a neuroprosthesis that mimics life-like movements.

METHODS

Subjects. Sixteen volunteer stroke subjects (Table I) have been included in the study. The subjects were associated to a higher functioning group (HFG) or lower functioning group (LFG) prior to the study upon their active range of motion at the wrist and fingers. The determinations were made with the subject sitting, the forearm resting on a supported surface, and the forearm in pronation. The hand was hanging over the edge of the supporting surface (e.g., the arm of a chair) to allow for maximum wrist flexion with gravity. The subjects were identified as belonging to a HFG if they were able to actively extend the paretic wrist further than 20 degrees and actively extend the MP and IP joints of all digits at least 10 degrees. The subjects were assigned to a LFG if they could actively extend the paretic wrist at least 10 degrees ($<20^0$), and extend the MP and IP joints of the thumb and at minimum two additional digits at least 10 degrees.

The study included eight subjects from the HFG subjects, and eight subjects form the LFG.

N°	Age	Start after stroke	Ashworth	Diagnosis	Group
1	62	4 weeks	1+	Hemi. l. sin.	HFG-FET
2	61	6 weeks	1+	Hemi. l. sin.	HFG-FET
3	50	7 weeks	2-3	Hemi. l. sin.	HFG-FET
4	52	11 weeks	4	Hemi l. sin.	HFG-FET.
5	62	8 weeks	1+	Hemi l. dex.	HGF-control
6	65	4 weeks	2	Hemi l. sin.	HGF-control
7	61	9 weeks	3	Hemi l. sin.	HGF-control
8	72	5 weeks	2-3	Hemi. l. sin.	HGF-control
9	41	4 weeks	2-3	Hemi. l. dex.	LFG-FET
10	27	3 weeks	2-3	Hemi. l. dex.	LFG-FET
11	72	7 weeks	3	Hemi. l. dex.	LFG-FET
12	65	8 weeks	2	Hemi. l. sin.	LFG-FET
12	72	6 weeks	2	Hemi. l. sin.	LFG-control
14	45	4 weeks	2-3	Hemi. l. dex.	LFG-control
15	51	7 weeks	2	Hemi. l. dex.	LFG-control
16	67	10 weeks	2-3	Hemi. l. sin.	LFG-control

Table I Basic data for sub-acute stroke subjects included in the FET). The HFG stays for high functioning group and LFG for low functioning group.

Procedure. The stimulation was applied for three weeks, for 30 minutes daily. The FET was applied with the four channel stimulator and surface disposable electrodes. Two channels were used to stimulate the finger flexors and finger extensors. The cathodes were placed over the *extensor digitorum communis m*. and *flexor digitorum profundus* and *superficialis m.* A common anode for these two channels covered the major arm nerves. The other two channels were applied to control the thumb. Subjects used the switch to start a stimulation pattern to grasp or release an object. The pulse duration, frequency and pulse amplitude (current) were set to minimize discomfort during stimulation, yet provide externally assisted grasp. The typical values for stimulation were the following: f = 50 Hz, T = 300 ìs, I = 15 - 45 mA.

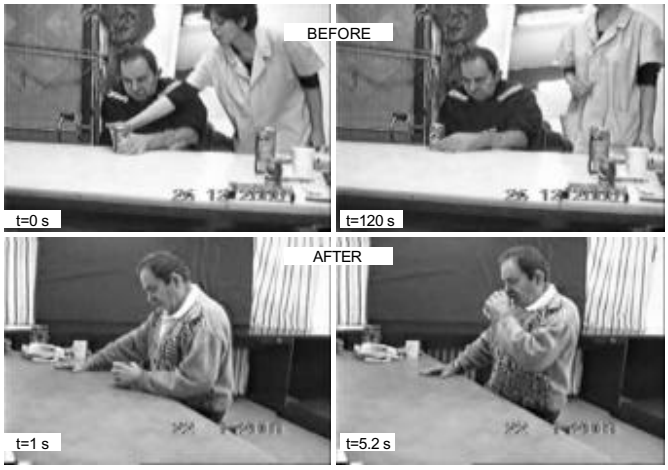


Fig. 1: An acute stroke subject from HFG before FET (upper panels) and after three weeks of FET (bottom panels) exercising to drink from a can. At the beginning he was not able to hold his left arm, neither to grasp a can (upper panels); after three weeks of the FET he could repeat the task of drinking from a can for 20 times during a two minutes session.

The FET tasks were to actively reach and functionally use selected objects typical for daily activities (e.g., juice can, telephone receiver, pen, comb, and toothbrush). The control group was required to exercise the same tasks as the FET group, yet without electrical stimulation for 30 minutes daily. Fig. 1 shows an example of the effective functional outcomes captured from the movies recorded before and after the FET in one of the subjects from HFG.

Outcome measures. Evaluations were done at the beginning and the end of the FET. Evaluations are planned during every two months for the next 18

months. Here we summarize only one of the several measured outcomes: the Upper Extremity Function Test (UEFT). The purpose of UEFT was to determine the differences in the performance of certain activities of daily living before and after the FET without the use of the stimulator. The following tasks were tested: 1) combing hair; 2) using a fork; 3) handling a VHS tape; 4) picking up a beer can; 4) picking up a small (pop/soda) can; 6) writing with a pen; 7) handling the telephone receiver; 8) brushing teeth; 9) pouring from a one litter container; 10) drinking from a mug; and 11) handling finger food. The performance of the tasks was graded as Success (YES) and Failure (NO), and if "YES" the time for accomplishing the task was estimated. The time was determined by dividing the time interval (120 seconds) with the number of successful trials during this interval.

RESULTS AND DISCUSSION

The results of the UEFT are summarized in Tables II and III. The data includes the number of trials that the subjects were able to accomplish before and after the therapy for HFG and LFG. There is a substantial difference between the performance of the LFG and HFG both before and after the FET.

		FET GROUP				CONTROL CROUP			
HFG - Subject N°		1	2	3	4	5	6	7	8
Number of successful tasks	after	11	10	10	6	11	9	11	7
	before	11	3	7	6	11	8	11	7
Averaged number of successful trials	after	31.54	10.8	24.9	17.83	8.27	2.11	18.34	12.3
	before	19.36	2	4.1	15.5	5.54	1.63	10.36	8.8
Averaged time (s)	after	3.8	11.11	4.81	6.73	14.51	28.43	6.47	9.75
	before	6.19	60	19.26	7.74	21.66	73.62	11.58	13.83
LFG - Subject N°		9	10	11	12	13	14	15	16
Number of successful tasks	after	9	0	10	8	0	0	3	3
	before	0	0	0	0	0	0	0	0
Averaged number of successful trials	after	3.8	0	3.3	1.75	0	0	1	1.33
	before	0	0	0	0	0	0	0	0
Averaged time (s)	after	31.58	N/A	36.36	68.57	N/A	N/A	120	90
	before	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A

Table II: The averaged number of tasks and successful trials. Sub-acute stroke subjects were randomly assigned to FET group (1-4, and 9-12) and controls (5-8, and 13-16). All subjects use medications against high blood pressure, anticoagulants, and medications for spasticity reduction. First eight subjects (1-8) were classified to the high functioning group (HFG), while the subjects 9-16 to the low functioning group (LFG).

The results for the HFG show that total number of tasks in both groups (FET and controls) at the end of three weeks are comparable. The only task that was not accomplished by almost all subjects is handling of the one litter container being too heavy and big. The FET group comprised subjects whose performance was initially worse compared to controls: in average they accomplished 6.75 tasks, compared to 9.25 tasks performed by controls. The improvements in UEFT scores in the HFG are substantial: the average number of trials was increased for 10.88±1.01 in FET group, and 3.67±0.88 in controls. This difference can be even better appreciated by analyzing the time to perform a task. The time in average decreased 71% (from 23.29 to 6.61 seconds) in FET group and 51% (from 30.17 to 14.79 seconds) in controls.

The control LFG showed marginal improvement. The FET LFG functioning improved: in average they could accomplish 6.75±2.28 tasks (zero at the beginning), and in average they could repeat almost three times every task during two minutes (zero at the beginning). Results in Table III show that the improvement in the functioning is probably not good enough to make the affected arm useful for daily activities. We speculate that the period of three weeks is not long enough for the therapy to show effects, and that the stimulation of more proximal muscles (e.g. elbow flexion and extension) could contribute significantly.

The use of electrical stimulation has been shown to produce therapeutic effects: decreased spasticity, increased movement range and speed, and increased muscle strength. The mechanisms by which those changes are occurring are still controversial. We hypothesize that FET generates activity-dependent changes

		FET GROUP	CONTROL CROUP
HIGH FUNCTIONING GROUP			
Number of successful tasks	after	9.25±1.1	9.5±0.95
	before	6.75±3.3	9.25±1.03
Averaged number of successful trials	after	21.26±4.47	10.25±3.41
	before	10.24±4.14	6.58±1.93
Averaged time (s)	after	6.61±1.61	14.79±4.83
	before	23.29±12.58	30.17±14.64
LOW FUNCTIONING GROUP			
Number of successful tasks	after	6.75±2.28	1.5±1.5
	before	0	0
Averaged number of successful trials	after	2.95±1.7	0.58±0.5
	before	0	0
Averaged time (s)	after	45.50±11	105±15
	before	N/A	N/A

Table III: The averaged number of successful tasks, and trials, and the averaged time required accomplishing a task for the FET and control group. The acronym N/A is used because the time could not be calculated for the tasks that have not been accomplished.

within the CNS when applied during appropriate motor tasks. This follows the findings that the brain possesses the capability to reorganize itself in such a way to allow neighboring cortical regions to expand into territories normally occupied by input from other organs. The FET most likely manipulates with the sensory input; thus, modulate the magnitude of cortical response and modulate motor pathway excitability, thereby produces a mixture of excitation and inhibition at supraspinal levels.

REFERENCES

/1/ Cauraugh J., Light K., Kim S., Thigpen M., Behrman A., Chronic motor dysfunction after stroke: recovering wrist and finger extension by electromyography-triggered neuromuscular stimulation. *Stroke* 31:6 1360-4, 2000.

/2/ Chae J., Bethoux F., Bohine T., Dobos L., Davis T., Friedl A., Neuromuscular stimulation for upper extremity motor and functional recovery in acute hemiplegia. *Stroke* 29:5 975-9, 1998.

/3/ Feys H.M., De Weerdt W.J., Selz B.E., Cox Steck G.A., Spichiger R., Vereeck L.E., *et al.,* Functional magnetic resonance imaging of the human motor cortex before and after whole-hand afferent electrical stimulation. *Scand J Rehabil Med* 31:3 165-73, 1999.

/4/ Francisco G., Chae J., Chawla H., Kirshblum S., Zorowitz R., Lewis G., Pang S., Electromyogram-triggered neuromuscular stimulation for improving the arm function of acute stroke survivors: a randomized pilot study. *Arch Phys Med Rehabil* 79:5 570-5, 1998.

/5/ Kraft G.H., Fitts S.S., Hammond M.C., Techniques to improve function of the arm and hand in chronic hemiplegia. *Arch Phys Med Rehabil* 73:3 220-7, 1992.

/6/ Merletti R., Acimovic R., Grobelnik S., Cvilak G., Electrophysiological orthosis for the upper extremity in hemiplegia: feasibility study. *Arch Phys Med Rehabil* 56:12 507-513, 1975.

/7/ Nathan R., (1997) Handmaster NMS - present technology and the next generation. In: Popovic D. (Ed*.) Proc 2nd Int Symp IFESS*, Burnaby, pp 139-140.

/8/ Popovic D.B., Popovic M.B., Stojanovic A., Pjanovic A., Radosavljevic S., Vulovic D., Clinical Evaluation of the Belgrade grasping system, *Proc V Vienna FES Workshop*, 1998. 247-250.

/9/ Popovic D.B., Stojanovic A., Pjanovic A., Radosavljevic S., Popovic M.B., Jovic S., Vulovic D., Clinical evaluation of the bionic glove. *Arch Phys Med Rehabil* 80:299-304, 1999.

ACKNOWLEDGEMENTS

This project was partly supported by the Danish National Research Foundation, Denmark.

AUTHOR'S ADDRESS

Mirjana B. Popovic, Visiting Associate Professor
Aalborg University, SMI
Fredrik Bajers Vej 7, D-3, 9220 Aalborg, Denmark

e-mail: mpo@smi.auc.dk
home page: http://www.imi.bg.ac.yu

A PILOT STUDY IN PREPARATION FOR AN INVESTIGATION INTO THE EFFECTS OF ELECTRICAL STIMULATION ON RECOVERY OF HAND SENSATION AND FUNCTION IN STROKE PATIENTS

Mann G.E.,[1] Malone L.[1] Taylor P.N.[1] Burridge J.H.,[2]

[1]Department of Medical Physics and Biomedical Engineering, Salisbury District Hospital, Salisbury, Wiltshire, SP28BJ
[2]Department of Rehabilitation and Health Sciences, University of Southampton, Southampton, SO17 1BJ

SUMMARY

An important aim of rehabilitation therapy for the upper limb following stroke is the recovery of motor function. Recovery of sensation is also thought to have an effect on functional recovery.
The aim of this study is to investigate the effect of electrical stimulation on motor and sensory recovery in the upper limb and to establish their relative contributions to functional ability.
Subjects, within one year following a first stroke, were randomised into treatment and control groups. The treatment group received electrical stimulation of elbow and wrist extensors and the control group, elbow, wrist and finger extension exercises. Upper limb function was assessed using the Action Research Arm Test. In addition the Jebsen Taylor Hand Function Test was performed and hand sensation was assessed using the static Two Point Discrimination Test. Twelve subjects have completed the treatment stage of the study, 6 in the stimulation group and 6 in the control group.
This paper presents the results of the Action Research Arm Test for 12 subjects who have completed the treatment phase of the study.

STATE OF THE ART

There are nearly 150,000 first strokes a year in the United Kingdom. About half of these have impaired function of one upper limb and of these only 14% will regain any useful function.
The main focus of therapy for most patients following stroke is on recovery of motor function rather than any associated sensory deficits. However, it is well documented that these deficits have an adverse effect on functional outcome although the degree of functional impairment is not necessarily related to the extent of sensory loss.
Some physiotherapy treatment includes specific re-education of sensation in addition to motor retraining to improve function but few attempts have been made to promote recovery of sensation directly following stroke.
There has been little work with electrical stimulation to re-educate sensation in the hand of the hemiplegic patient with identified sensory deficit. Baker et al /1/ stimulated the wrist and finger extensors of 16 subjects over a 4 week period but recorded no changes in sensation using the 2 point discrimination test.
Prada /2/ found that use of 'contingency stimulation', that is, electrical stimulation to the hemiplegic forearm triggered by movement of the unaffected arm, resulted in significant improvement in awareness of the affected arm measured by the Rivermead Perceptual Assessment Battery and that this improvement was maintained.
Taylor et al /3/ conducted a retrospective analysis of 20 patients who had suffered a stroke at least 6 months prior to commencement of electrical stimulation treatment for the upper limb. These patients had restricted hand function but were able to take the hand to the mouth. They received electrical stimulation to the wrist and finger extensors reciprocally with either the lumbricals, the finger flexors or a rest period. Exercises were carried out twice daily for up to an hour a session. Tests of hand function using the Jebsen-Taylor hand function test, grip strength and the 2 point discrimination test for sensation were carried out.

Improvements in function were recorded and sensation tested in 11of the original 20 subjects showed improvement in 7 and no change in 4 subjects.
Few studies have quantified functional changes following therapy. Kraft et al /4/ compared three groups who received proprioceptive neuromuscular facilitation therapy, bias balance electrical stimulation therapy or EMG initiated stimulation to wrist extensors, with a control group. Assessments using the Fugl-Meyer (FM) post stroke recovery test showed significant improvements in all treatment groups, with the greatest gain in the EMG triggered stimulation group. Those in the treatment groups who could complete a Jebsen –Taylor hand function test also showed improvement. Controls showed no improvement.

MATERIALS AND METHODS

Subjects within a year of recovery from a first stroke leading to hemiplegia were recruited following admission to an acute inpatient rehabilitation unit, a neurological outpatient service or an Elderly Care Day Hospital. They were at least 18 years old with no upper age limit, medically stable, able to give informed consent and to comply with assessment procedures. Subjects who had evidence of sensory impairment and who were able to carry the hemiplegic hand towards the mouth were included.
Subjects were excluded if they had cognitive or psychiatric problems affecting their ability to understand or comply with treatment procedures, had a history of cardiac problems or wore a cardiac pacemaker.
Ethical approval for the study was obtained and subjects signed informed consent. Subjects were randomly assigned to the electrical stimulation group or the control group, using computer generated sealed allocation codes. All subjects continued with their standard physiotherapy, occupational and speech therapy.
Subjects in the electrical stimulation group received stimulation to the triceps and extensor carpi radialis and extensor digitorum communis muscles simultaneously to achieve a quasi functional movement. Stimulation was applied initially for 10 minutes twice a day, increasing to two 30 minute sessions a day.
Stimulation was delivered by a Microstim 2 neuromuscular stimulator, powered by a PP3 9 volt battery and producing a train of pulses of 300 microseconds duration at a frequency of 40Hz. Stimulation is applied for 8 seconds alternately with a rest period of 8 seconds with ramps of 2 seconds. The stimulation was applied using Pals skin surface self-adhesive electrodes. Current amplitude was adjusted to achieve full elbow, wrist and finger extension at a comfortable level.
Subjects in the control group were instructed in self-administered passive stretches of the elbow, wrist and fingers. These were carried out for the same length of time as the electrical stimulation exercises.
All inpatient treatments were supervised by a trained therapist. On discharge, treatment was supervised by a carer or carried out independently. Subjects carried out treatment for 12 weeks.
Demographic data was recorded for all subjects (Tables1 & 2).
Upper limb motor function was assessed using the Action Research Arm Test /5/ and hand function using the Jebsen-Taylor Hand Function Test /6/. The 2 point discrimination test was used to assess sensation in the hand /7/.
Assessments were carried out by a trained therapist at week 0 (start of treatment), week 6 (treatment) week 12 (end of treatment) and week 24, (12 weeks post treatment). Results of the assessments at from weeks 0-12 were analysed using paired t tests.

RESULTS

Results are presented for the Action Research Arm Test (ARAT) /6/only.This test consists of sections for Grasp, Grip, Pinch and Gross Movements. Each section is scored separately and the scores added. The maximum possible total score is 57.
Twelve subjects, 8 female and 4 male, mean age 71.9 years (+/-10.1) have completed the treatment phase of the study (seeTable1). Six subjects received electrical stimulation and 6 were in the control group.

	FES group	**Control group**
Mean Age (SD) yrs.	68.2 (9.1)	75.7 (10.3)
Age range yrs.	58 - 80	63 - 88
Sex	3 Female – 3 Male	5 Female – 1 Male
Side of hemiplegia	3 Right – 3 Left	1 Right – 5 Left
Mean time since stroke (SD)	7.2 (5.0) months	8.8 (2.8)
Cause of stroke	5 Infarct – 1Haemorrhage	5 Infarct – 1 Haemorrhage

Table1: Demographic data

Following randomisation it was observed that initial Action Research Arm Test scores for the stimulation group were higher than those for the controls. All subjects showed improved scores over the treatment period from weeks 0-12. Changes in weeks 0-6 of treatment were significant in both groups – FES: $p<0.010$, :Controls $p<0.042$. Improvement continued in both groups during weeks 6-12 but was only significant in the stimulation group FES $p<0.001$, Controls $p< 0.094$ (see Table 2).

	Mean ARAT Scores FES			**Mean ARAT Scores Controls**		
Week	0	6	12	0	6	12
Mean score(SD)	23.0(17.2)	32.7(15.1)	40.0(14.0)	8.8(7.7)	13.7(9.2)	16.5(11.0)
p-value at 95% C.I		0-6weeks p<0.010 0-12weeks p<0.001			0-6weeks p<0.042 0-12weeks p<0.094	
Confidence interval		-15.88, -3.45	-23.74, -10.26		-9.77,-0.023	-14.02,+1.45

Table 2 SD = Standard Deviation

DISCUSSION

These preliminary results indicate that although improvement in Action Research Arm Test scores is achieved by both stimulation and control groups between the first 2 assessments between weeks 0 and 6 of treatment , there is greater improvement in the stimulation group Between the second and third assessments at weeks 6 and 12 there is continued improvement in both groups but this is significant only in the stimulation group. There are a number of possible reasons for this. Despite randomisation the initial Action Research Arm Test scores for the FES group at week 0 were lower for the control group than for the stimulation group. The mean age of the control subjects was greater and they had a longer time interval since onset of stroke. These factors need to be examined further for their possible contribution to the differences in level of improvement in the 2 groups.
Clinically, subjects require at least some active movement at the shoulder to be able to perform the tasks in each section of the Action Research Arm Test successfully. It was evident that some subjects who were able to perform the hand movement to initiate a task were unable to complete it because of lack of active movement at the shoulder. The criteria for inclusion in the study did not identify poor shoulder movement

as a limiting factor and this could be a reason for the poorer performance of some subjects who were able to achieve the required hand movements.
Although the title of this study states that it is concerned with hand function and sensation, it will be necessary in a future trial to include assessment of the shoulder as an essential component of the upper limb in performance of function.
Therefore, in addition to informing sample-size calculations for a larger trial in the future, this study is also important to identify the most appropriate outcome measures for recovery of upper limb function and sensation following stroke.

REFERENCES

Baker LL, Yeh C, Wilson D, Waters RL. Electrical stimulation of wrist and fingers for hemiplegic patients. Physical Therapy 1979; 59 (12):1495 – 1499.
Prada G, Tallis R. Treatment of the neglect syndrome in stroke patients using a contingency stimulator. Clinical Rehabil 1995; 9:304 – 313.
Taylor PN, Burridge JH, Hagan SA, Chapple P & Swain ID. Improvement in hand function and sensation in chronic stroke patients following electrical stimulation exercises. A retrospective clinical audit.Pro Vienna 5th. International Workshop on Functional Electrostimulation ISBN3-900928-03-7 pp359-362.
Kraft GH, Fitts SS & Hammond, MC. Techniques to improve function of the arm and hand in chronic hemiplegia. Arch. Phys. Med. Rehabil. 1992; 73: 220-227.
Carroll, D. A Quantitative Test of Upper Extremity Function, Journal of Chronic Diseases,1965 Vol.18: 479-491.
Jebsen RH, Taylor N et al. An objective and standardised test of hand function. Arch. Phys. Med. and Rehabil. 1969; June: 311 – 319.
Lee Dellon, A. The Sensational Contributions of Eric Moberg. Journal of Hand Surgery, 1990 15B: 14-24.

ACKNOWLEDGEMENTS

The authors would like to acknowledge the contribution of the Physiotherapy Research Foundation in part funding this study.

AUTHOR'S ADDRESS

Geraldine E. Mann
Dept. Medical Physics
Salisbury District
Wiltshire SP2 8BJ
England

e-mail: G.Mann@mpbe-sdh.demon.co.uk
homepage: www.salisburyfes.com

PRELIMINARY ECONOMIC EVALUATION OF ELECTRICAL STIMULATION TREATMENT OF THE UPPER EXTREMITY IN POST-STROKE HEMIPLEGIA

M.J. IJzerman[1], J.R. de Kroon[1], J.J.M. Jannink-Nijlant[1], G.J. Renzenbrink[1] and J.L. Severens[2]

1. Roessingh Research and Development, Enschede, the Netherlands
2. Dept. Medical Technology Assessment, University Medical Centre, Nijmegen, the Netherlands

SUMMARY

Cost-effectiveness analysis (CEA) increasingly receives attention as reimbursement and social justification of health care expenses is considered. A preliminary economic evaluation was conducted on electrical stimulation (ES) treatment in post-stroke hemiplegia. Costs of ES treatment were estimated using a survey amongst professionals with experience with ES therapy. Benefits of ES treatment were assessed using a willingness to pay approach. Net health benefit (NHB>0: cost-saving) was estimated EURO -89.- and EURO -997.- for two commercially available ES devices respectively. Using a more progressive scenario it is concluded that ES treatment may be cost-saving if success of the treatment can be guaranteed.

STATE OF THE ART

Worldwide, the experience with technology assessment and CEAs on neuro-prosthetics in the field of physicial medicine and rehabilitation is limited. Regarding neuro-prostheses (NP) probably most experience is available on the (VOCARE) bladder stimulator in SCI (Wielink *et al*, 1997; Creasey *et al*, 2000). Both authors performed a cost-minimization analysis, i.e. they assessed the additional costs and savings after introducing the stimulator. The studies are conducted using the data collected in an uncontrolled trial in the Netherlands and USA respectively. Only direct medical costs (surgery, hospital visits and reduction in self-care and medication) were considered. In addition to the medical costs, Wielink *et al* also assessed health related quality of life (HRQoL) using the Nottingham Health Profile. The average pay back time of implanting the system was estimated to be approx. 8 years for the Netherlands (discounted at 5 %) and 5 years in the USA. In both studies it is concluded that the stimulator is economically viable.

Since the early seventies many studies have been published on the use of electrical stimulation to enhance motor control in post-stroke hemiplegia. The majority of clinical research on ES treatment of the upper extremity in post-stroke hemiplegia is still in an experimental stage. Nevertheless, a few years ago some commercial systems became available, i.e. the automove AM800 and the NESS Handmaster. The AM800 provides EMG triggered stimulation and the Handmaster is an orthosis with integrated electrical stimulation electrodes (Chae and Yu, 1999; Hendricks *et al*, 2001). Cost-effectiveness of the treatment is expected to be useful while negotiating with insurance companies about reimbursement. Sculpher *et al* (1997) argued that an economic evaluation should be designed according to the stage of clinical investigation and each of these stages requires a particular focus on economic evaluation. In the first stage emphasis is given to systematic collection of evidence on costs and effects and to collect informal clinical opinions. The second stage is more characterised by a modelling approach, whereas the third and fourth stage usually collect costing data alongside the trial. Most of the trials on electrical stimulation devices can be considered as stage II trials. Therefore, the current study was carried out to gain insight into the economic potential of electrical stimulation treatment in post-stroke hemiplegia. This paper intends to explore whether electrical stimulation (ES) treatment of the upper extremity in stroke patients can be economically viable. Although we studied both devices it is not the intention to compare them. We will focus on the best available evidence on effectiveness and on the relevant costs that are associated with electrical stimulation.

MATERIAL AND METHODS

Preliminary economic evaluation

A societal perspective was chosen for this study. The time horizon was set at one year, because it is not known what the long term effects of the treatment are. Also, it is unknown whether the subjects still use the device after a year. Because the time horizon was only one year, costs were not discounted.

Study population

Estimates of potential cost savings were obtained retrospectively from a group of experts that was expected to be able to judge the effects of electrical stimulation treatment on resources consumed. The group comprised of 7 physicians for PM&R and 7 therapists who participated in a multicentre trial on electrical stimulation therapy in stroke (response was 57 %). Willingness to pay values (benefit of treatment) were obtained from a non-experienced reference population. In this study we selected 25 workers in a rehabilitation institute (response was 76 %).

Assessment of effectiveness (willingness to pay)

A systematic literature was carried out in order to assess the effectiveness of electrical stimulation therapy in post-stroke hemiplegia (Kroon de *et al*, 2001). The review led to the conclusion that ES appeared to have a positive effect on motor control, whereas no conclusions could be drawn on improvement in functional abilities. Although EMG triggered stimulation theoretically would be more effective (Chae and Yu, 1999) this could not be established in the review. There is a moderate evidence of a larger treatment effect in a subgroup of patients with higher initial motor scores. The intermediate outcome measures that were used in the reviewed studies, typically can not be used in an economic evaluation. On the basis of the systematic review and other case studies it is expected that the effects of electrical stimulation therapy can be described as slight to moderate improvement in motor control, reduced oedema, shoulder complaints and spasticity. Assuming these effects it is possible to value the ES treatment. Valuation of the ES treatment was performed in monetary terms, i.e. cost-benefit analysis. A contingent valuation (willingness to pay) approach was used to obtain estimates of the monetary value of the treatment outcome (O'Brien and Gafni, 1996). WTP was obtained for three different probabilities (10, 50 and 100 %) of a successful outcome. Success was defined by using the results from the systematic review. The following effects were shown to the respondents: 1) reduced arm spasticity, 2) slightly less oedema, 3) less shoulder pain and a 4) moderate improvement in hand function. WTP estimates were obtained from experienced clinicians (n=8) as well as a non-experienced reference population (n=19).

Assessment of costs and resource effects

Relevant costs associated with stroke (direct and indirect medical costs and productivity loss) were determined using a database of the Health Care Insurance Board of the Netherlands (1997). Costs of the electrical stimulation treatment (therapist time and incidental and yearly cost of the equipment) were incorporated. The ES devices have an incidental cost (device) as well as yearly costs due to replacement of electrodes. Dutch market prices were used. The effects of ES on resource use (e.g reduction in medical consumption) were estimated using a standardised questionnaire. The questionnaire was send to the group of experienced clinicians. The clinicians were asked to mark the resources in which they expected a change in consumption, e.g. prescription of spasmolytics. Finally, they were asked to estimate the magnitude of the changes in resource use (percentage). Macro budgetary impact was calculated assuming that 15 % of the stroke population would be suitable for the ES treatment (mild spasticity, some voluntary activity in the hand) and that 80 % of that population actually receives the stimulation.

Data analysis

A cost model was made in EXCEL using the costing data and resource effects. Main assumption in the model is the equal distribution of costs in the population of stroke patients of interest. The estimated resource effects of each clinician were averaged and confidence intervals were calculated. Net health benefit (NHB) is calculated by subtracting the costs from the benefits (WTP) of ES treatment. A NHB>0 implies a cost-saving treatment. Sensitivity analysis is performed on the main resource parameters by using the lower and upper boundary of the confidence interval. Three scenario's were calculated: the

average scenario was calculated using the point estimates for costs and WTP, the conservative scenario was calculated using the lower and upper limit of the WTP values and costs respectively and the third (progressive) scenario was calculated in order to maximise the NHB (maximum value of WTP and most reduction in health care expenses).

RESULTS

The experienced clinicians expected that ES treatment would not reduce health care expenses, except for the use of spasmolytics (n=8) and physical therapy (n=6). Three clinicans expected a decrease in occupational therapy. Two clinicans expected a reduction in specialist consultations, the other expected that there would be more consultations.

Resource	**N^{o} hits (total: n=8)**	**Δ_{costs}**		
		Avg.	95% LL	95% UL
Spasmolytics	8	-15	-21.1	-8.9
Physiotherapy	6	-12	-16.2	7.96
Specialist consultation	3	+3.5	-26.0	32.6
Occupational therapy	3	-13	-20.7	6.0
Aids for self-care	2	-12.5	-18.0	-7.0

Table 1 Estimated resource effects (negative is reduction in costs). 95% confidence intervals are presented (UL: upper limit and LL: lower limit).

Taking into account the resource effects it is estimated that use of ES devices will increase the total costs of treatment of a stroke patient between EURO 2540.- (AM800) and EURO 3448.- (Handmaster) assuming one year of ES treatment. It appeared that the reduction in medical consumption is only marginal compared to the costs of treatment. The overall increase in cost of ES treatment is caused by electrode supplies. Yearly costs of electrodes is approx. EURO 408.- for both devices. Reduction in medical consumption does not outweigh the increase in costs due to electrodes. Costs were also calculated using the upper and lower limits of the 95% confidence intervals (table 1). Using a conservative estimation of the resource effects (upper limit, table 1) cost of ES treatment was estimated to be EURO 3967.- and EURO 3117.- for the Handmaster and AM800 respectively.

	WTP	**95% LL**	**95% UL**
100%	2451	1382	3520
50%	1708	655	2762
10%	759	151	1367

Table 2 Willingness to pay (EUROs) for ES treatment at three different probability levels.

WTP of a reference population was approx. EURO 2450.-, assuming a 100% probability of success. WTP assuming a 50 % and 10% probability was EURO 1708.- and EURO 759.- respectively (table 2).

Net health benefit was calculated using the estimates of the cost of ES treatment and the WTP values if a 100% probability of success is assumed ($WTP_{100\%}$). NHB was EURO -997.- and EURO -89.- for the Handmaster and AM800 respectively (average scenario). Using a progressive scenario ES treatment may be cost-saving (table 3).

Scenario	**$WTP_{100\%}$**	**Costs**	**$NHB_{Handmaster}$**	**NHB_{AM800}**
Average	Point estimate	Point estimate	-997	-89
Conservative	95% CI LL	95% CI UL	-2585	-1735
Progressive	95% CI UL	95% CI LL	228	1079

Table 3 Net health benefit calculated using three different scenarios.

DISCUSSION

The aim of this study was to gain insight in economic potential of ES treatment in post-stroke hemiplegia and the present study can be considered a pilot investigation in order to judge whether a full economic evaluation would be useful. In contrast to other neuroprostheses, e.g. the VOCARE bladder stimulator, it is not very likely that health care expenses will be reduced after introducing ES treatment. It is expected that only a small reduction in medical consumption (spasmolytics and physiotherapy) will be achieved whereas costs outside healthcare (absence from work) will not be relevant. Also, ES treatment requires the use of electrodes that contribute to a rise in annual treatment cost. Both ES devices use electrodes that are relative expensive compared to the market price of the device. A critical aspect in the study is that ES treatment usually requires a substantial time (e.g. 3 hours/day) investment of patients. There is an ongoing

debate about how to take the time investment into account. It is obvious however, that time investment of the patient should be incorporated in the analyses. Another critical issue is the long term use of ES devices. It is not clear if patients are long term users of these devices and abandoning ES devices may result in large costs for society without any beneficial effect.

The effects of ES treatment were derived from a systematic review, but in an economic evaluation one needs outcome measures on an aggregrate level. The most common approach is the measurement of a utility, which allows the researchers to calculate a quality adjusted life year (QALY). It is questioned whether these instruments are able to measure the effects of ES treatment. The present WTP approach is, at least in theory, more responsive to detect treatment effects. However, the approach is relatively new and not without -methodologic- discussion. For instance, WTP depends on the wages of the respondents and it is required to select a „representative sample from the general population". The amount of information on treatment effects provided to the respondents influences the outcome. For instance, the experienced clinicians had lower WTP values compared to the reference population, which may be caused by a difference in a priori knowledge about the effects. The WTP survey is difficult to complete and to overcome that problem we have chosen to include highly educated people. Although they may not represent society, it is expected that they are better suited to capture the cognitive task.

Almost 50% of the reference population values ES treatment at EURO 2451.- or higher and for those respondents, the value of ES treatment outweighs the costs. This would suggest that ES treatment may be cost-saving for a part of the respondents if the probability of success is nearly 100 %. In order to establish a high success rate it is required to critically examine the patient before an ES device is prescribed. In the present pilot it was decided to study ES treatment as a general entity and it was not the intention to compare two devices. Recently, different reviews have concluded that more fundamental research is required in order to underpin the theoretical foundations of ES treatment in general (Chae and Yu, 1999; Kroon, *et al* 2001). From that perspective it may be encouraged to improve existing knowledge on treatment mechanisms and approaches before conducting a full economic analysis.

REFERENCES

- Chae J and Yu D., Neuromuscular stimulation for motor relearning in hemiplegia. Critical reviews in PM&R, 1999; 11: 279-297
- Creasey GH, Kilgore KL, Brown-Triolo DL, Dahlberg JE, Peckham PH, Keith MW. Reduction of costs of disability using neuroprostheses. Assistive Technology 2000; 12: 67-75
- Hendricks HT, IJzerman MJ, Kroon JR de, Zilvold G., Functional electrical stimulation by means of the "NESS Handmaster" in chronic stroke patients. Clinical Rehab, 2001;15: 207-210
- Kroon JR de, IJzerman MJ, Lee JH van der and Lankhorst GJ: Therapeutic electrical stimulation to improve motor control and functional abilities of the upper extremity after stroke: a systematic review. Proc. 6th annual conference of IFESS 2001, p.155-157, Cleveland, USA
- O'Brien B, Gafni A., When do the dollars make sense? Toward a conceptual framework for contingent valuation studies in health care. Medical Decision Making 1996; 16: 288-299
- Sculpher M, Drummond MF, Buxton M., The iterative use of economic evaluation as part of the process of health technology assessment. J. Health Serv Res Policy, 1997; 2: 26-30
- Wielink G, Essink-Bot ML, Kerrebroeck PhEV van, Rutten FFH: Sacral rhizotomies and electrical bladder stimulation in spinal cord injury. Cost-effectiveness and quality of life analysis. Eur Urol, 1997; 31: 441-446
- Ziekenfondsraad. Treatment after stroke (Dutch text). Report. Amstelveen. 1997

AUTHOR'S ADDRESS

Maarten J. IJzerman, PhD, PT
Roessingh Research and Development
PO Box 310
7500 AH Enschede
the Netherlands

e-mail: m.ijzerman@rrd.nl
home page: www.rrd.nl

POSTER PRESENTATIONS

EFFECTS OF TRAINING WITH ELECTRICAL STIMULATION ON KNEE JOINT TORQUE

T. Angeli, T. Denkmeier

Institute of Machine Elements and Machine Design, University of Technology, Vienna, Austria

SUMMARY

The aim of this study was to investigate the influence of an eight-week Electrical Stimulation (ES) training program on the knee joint torque of the knee joint extensor muscles of six test persons (three trained hobby sportsmen 'T ES' and three untrained test persons 'UT ES'). ES sessions were carried out with simultaneous maximum voluntary isometric contraction five times weekly. A control group of six test persons (three 'T NoES' and three 'UT NoES') completed the same training program without ES. Every two weeks measurements on a dynamometer were done before, during and after training sessions. The averaged strength increase of the 'UT ES'-group amounted to 9.5% and of the 'T ES'-group to 5.7% at the knee joint flexion angle of 60°. The 'UT NoES'-group achieved 9.1% at the same knee joint flexion angle.

STATE OF THE ART

The physiological experiments of Galvani who made frog muscles contract with the aid of contact current around 1791 are generally known. However in the year of 420 before Christ Hippocrates already treated asthma with electrical strokes (ES!) of the torpedo-fish. Functional Electrical Stimulation (FES) has been employed with spinal cord injury subjects for a long time. Kern /4/ reports on clinical and physiological effects of eight months FES of the m. quadriceps femoris on 16 paraplegic patients. After eight months of FES training muscle perfusion was augmented by 80% and the muscle fibre diameters showed an average increase of 50% .

In recent years attempts have been increased to also use ES in training of healthy test persons (sportsmen). Studies were conducted about ES for weight lifting /3/, for basketball players /6/, for swimming /7/, for cycling /10/ and in the explosive strength training /11/. ES training proved to be effective in all these studies. Very different strength increases of between 0 % and 44 % were reported /2/, /5/, /8/, /9/.

In Austrian high-performance sports ES is used in skiing for strength training. In rowing it is used only for regeneration, because the coach doubts the effectiveness of strength training with ES. These doubts inspired us to investigate whether or not strength increase is attainable by training with ES. Tests were done on untrained and trained test persons.

MATERIAL AND METHODS

12 healthy persons (two female and ten male, average age 22.4 years) who never had trained with ES before were randomly assigned to the ES group or the control (NoES-) group. Both legs were trained in this study. Surface electrodes (rectangular self adhesive electrodes, 50 mm x 50 mm and 100 mm x 50 mm) were placed above the motor points of the m. quadriceps (fig. 1). The muscle was stimulated by a commercial stimulation unit (Compex Sport). The stimulation data are shown in table 1. The leg was fixed at a knee joint flexion angle of approximately 90°. Both training periods I and II (tab. 1) lasted three weeks and the training period III 2 weeks. At first the test person warmed up his m. quadriceps voluntarily and then the muscle was warmed up for five minutes with ES (low impulse frequency and amplitude). The amplitudes of the stimulation currents (for warming up, strength training and regeneration) were chosen by the test persons themselves depending on their pain sensitivity. The

strength training, that lasted between 20 and 25 minutes depending on training period, followed. The test persons had to contract the m. quadriceps isometrically (at a knee joint flexion angle of 90°) with maximal force while being stimulated. The training cycle was completed by the regeneration part which helped the test persons to cool down. The control group had to train, after warming up for 20 – 25 minutes, with maximal voluntary isometric contraction at the same knee joint flexion angle.

training period	warm up	strength training	impulse frequency	contraction time	relaxation time	regeneration	total time
	(min)	(min)	(Hz)	(s)	(s)	(min)	(min)
I	5	20	83	4	23	10	35
II	5	22	90	4	27	10	37
III	5	25	96	4	31	10	40

Table 1. Stimulation data

The measurements of the knee joint torque during contraction of the knee joint extensor (m. quadriceps) were performed on a dynamometer, that had been developed at our institute /1/. For the calculation of the active knee joint torque two measuring steps were necessary. First the test person's shank was moved passively (i.e., without muscle contraction) by the dynamometer and the passive knee joint torque was measured (step one). This value includes gravity forces and mass moment of inertia of the lower extremity (plus that of the measuring arm), passive muscle forces and loss of power due to friction. Secondly the test person was requested to extend the knee joint with maximum force at a fixed knee joint flexion angle of 110° (0° knee joint flexion angle equals fully extended lower extremity). This isometric contraction (between 0 ms and 1800 ms in fig. 2) allows the muscle to be prepared for the following concentric contraction. The isometric and concentric contractions are referred to as step two. These steps are performed at the knee joint angular velocities of 15°/s, 30°/s (fig. 2 and fig. 3), 60°/s, 90°/s, 120°/s and 180°/s. The test person tries to accelerate the measuring arm during the concentric movement as hard as possible. The active knee joint torque is calculated as the measured knee joint torque (step two) minus the passive knee joint torque (step one). For example: To hold the leg at the knee joint flexion angle of 5° the m. quadriceps must overcome the weight of the shank and foot and the passive torque of the antagonist muscles (hamstrings, m. gastrocnemius and others) which means that it is actually performing a higher force than measured. The knee joint angle course as a function of time is depicted in figure 2. Additionally isometric measurements were done at 30°, 60° and 90° knee joint flexion angles.

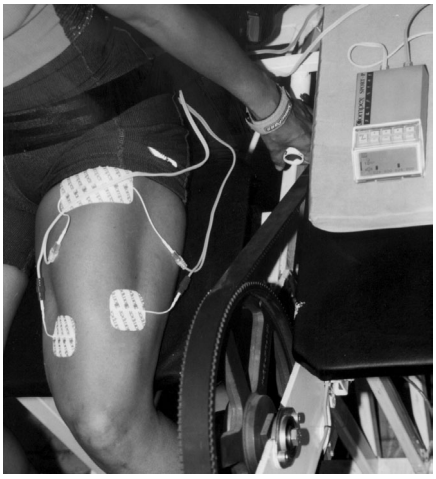


Fig. 1: Surface-electrodes and stimulation unit

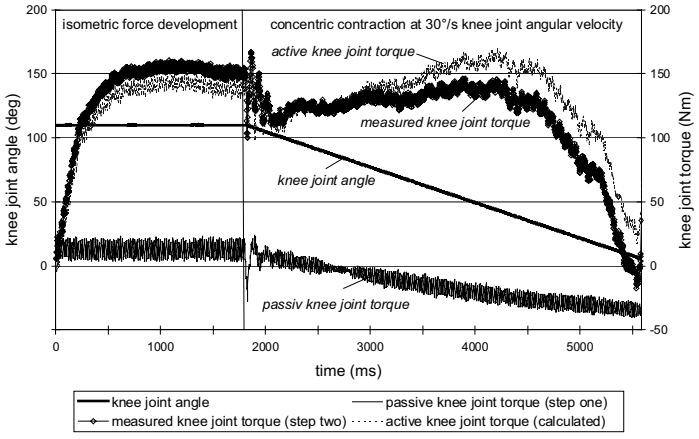


Fig. 2: Results of knee joint torque measurements and knee joint flexion angle as a function of time of one measurement cycle on one test person time at 30°/s knee joint angular velocity.

RESULTS AND DISCUSSION

Figure 3 depicts the knee joint torque increase for the 'UT ES'-group measured at the angular velocity of 30°/s. After 8 weeks of training the torque of the isometric contraction at the start angle of 110° was 27% higher than before training. This enormous strength increase can be explained by the fact that these tests persons almost never contracted their m. quadriceps maximally. A strength increase would have also been accomplished by the training effect of the six measurement cycles alone without the actual training. The lower the knee joint flexion angle was, the smaller the strength increase was.

Due to the small number of test persons and the deviation of their daily conditions the measurement results varied. The force velocity relation diagrams (fig. 4 - 6) were therefore smoothed. This was done by averaging the values of both legs of two consecutive measurement cycles and depicting the results in one curve (for example ○ in fig. 4 – 6). The force velocity relation values of the untrained test persons are shown in figure 4 ('UT ES') and figure 5 ('UT NoES') for the knee joint flexion angle of 60°. The comparison of the results was surprising. After eight weeks of training 9.5 ± 6.4% strength increase was measured for the 'UT ES'-group and 9.1 ± 5.5% for the 'UT NoES'-group. All test persons who had trained with ES complained about strong muscle aches that were stronger than after training without ES. These aches also persisted longer than usual. It was presumably for this reason, that the ES group probably chose smaller stimulation intensities and as a result, stimulated fewer motor units. This could also be the reason for the low values of the averaged results of the second and third measurements of the 'T ES'-group (fig. 6). In addition, it seems that the value of the first two

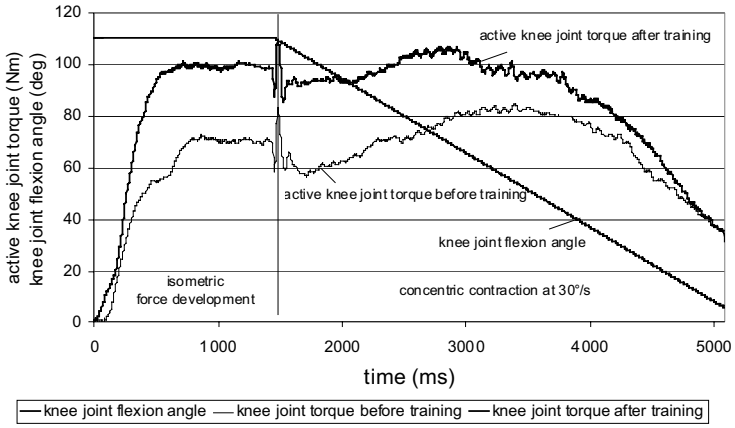


Fig. 3: Active knee joint torque before and after training and knee flexion angle as function of time at 30°/s knee joint angular velocity

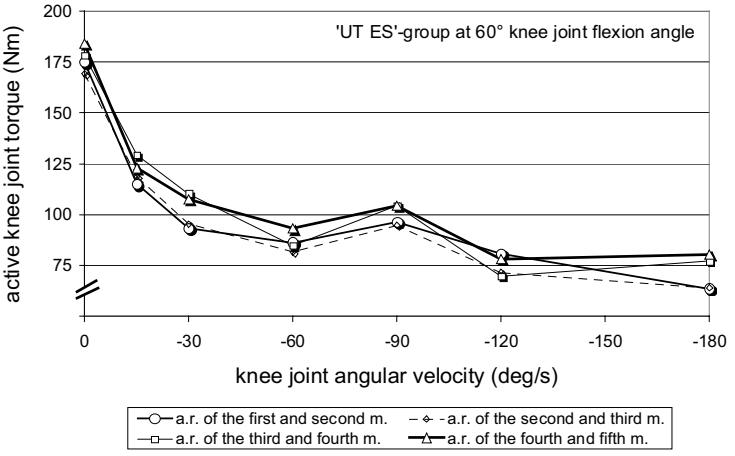


Fig. 4. Averaged results (a.r.) of the different measurements (m.) of the 'UT ES'-group at the knee joint angle of 60°. For details see text.

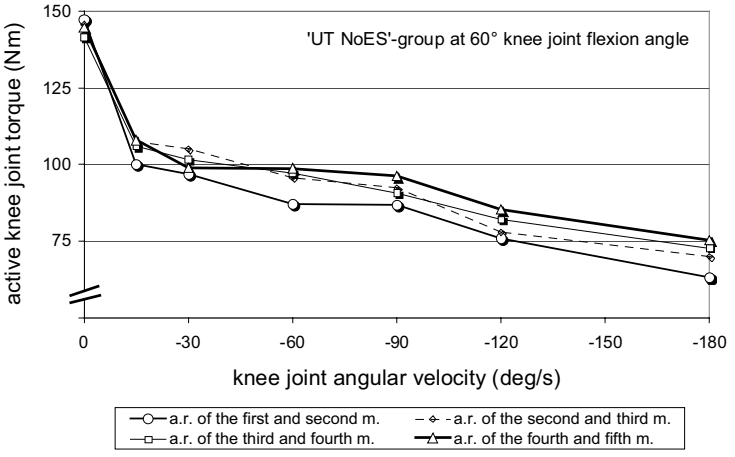


Fig. 5. Averaged results (a.r.) of the different measurements (m.) of the 'UT NoES'-group at the knee joint angle 60°

measurements at the knee joint angular velocity of 60°/s in figure 5 is a deviation and therefore the averaged strength increase for the control group rose.

It was no surprise that the strength increase of the untrained test persons achieved by this training was higher than that of the hobby sportsmen. The 'T ES'-group achieved 5.7 ± 3.3% strength increase at the knee joint flexion angle of 60° (fig. 6). The 'T NoES'-group achieved almost no strength increase (0.1 ± 3.5%) without ES. This surprising result of the control group might have come about because of lack of motivation. Moreover, one of the test persons complained about muscle aches caused by strain.

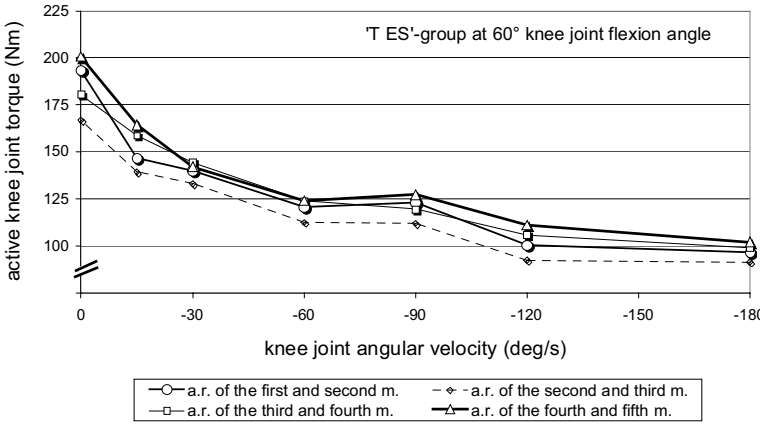


Fig. 6. Averaged results (a.r.) of the different measurements (m.) of the 'T ES'-group at the knee angle 60°

CONCLUSION

To support the informative value of this test, more persons would have to be tested (there were only 6 stimulation units available for this study). It was surprising that such a great strength increase could be measured with the untrained test persons without ES. It is to be presumed, that with top athletes the strength increase is even smaller than with hobby sportsmen.

REFERENCES

/1/ Angeli, T. (2000). Dynamometer for the measurement of torques on human joints. In: XVI IMEKO World Congress 2000: Proceedings Volume VII. Vienna. 167-172.

/2/ Currier, D. P.; and Mann, R. (1983). Muscular strength development by electrical stimulation in helthy individuals. Phys. Ther., 63, 915-921.

/3/ Delitto, A., Brown, M., Strube, M.J., Rose, S.J. and Lehman, R. C. (1989). Electrical stimulation of quadriceps femoris in an elite weight lifter: a single subject experiment. Int.J.Sports Med., 10, 187-191.

/4/ Kern, H. (1995). Funktionelle Elektrostimulation paraplegischer Patienten. Österr.Z.Phys.Med, 1, Suppl.

/5/ Laughman, R.K., Youdas, J.W., Garret, T.R. and Chao, E.Y.S. (1983). Strength changes in the normal quadriceps muscle as a result of electrical stimulation. Phys. Ther., 63, 494-499.

/6/ Maffiuletti, N.A., Cometti, G., Amiridis, I.G., Martin, A., Pousson, M. and Chatard, J.C. (2000). The effects of electromyostimulation training and basketball practice on muscle strength and jumping ability. Int. J.Sports Med., 21, 437-443.

/7/ Pichon, F., Chatard, J.C., Martin, A. and Cometti, G. (1995). Electrical stimulation and swimming performance. Med. Science in Sport Exerc., 27, 1671-1676.

/8/ Selkowitz, D.M. (1985). Improvement in isometric strength of the quadriceps femoris muscle after training with electrical stimulation. Phys. Ther., 65, 186-196.

/9/ Stefanovska, A. and Vodovnik. (1985). Change in muscle force following electrical stimulation. Scand.J.Rehab.Med., 17, 141-146.

/10/ Theriault, R., Boulay, M. R., Theriault, G. and Simoneau J.A. (1996). Electrical stimulation induced changes in performance and fiber type proportion of human knee extensor muscles. Eur.J.Appl.Physiol., 74, 311-317.

/11/ Witt, M., and Voß, G. (1996). Muskelstimulation im Schnellkrafttraining von Sportlern. In: Bochdansky T., Kollmitzer J., Krösel P. & Lugner P. (Ed.) Österr.Z.Phys.Med.Rehabil., Suppl. 2. Vienna, 98-100.

AUTHOR'S ADDRESS

Dr. Thomas Angeli
Inst. of Machine Elements and Machine Design
Vienna University of Technology
Getreidemarkt 9/306, 1060 Vienna

e-mail: thomas.angeli@tuwien.ac.at
home page: www.imel.tuwien.ac.at

VARIABLE FUNCTIONAL ELECTRICAL STIMULATION DEPENDING ON KNEE FLEXION ANGLE

T. Angeli, M. Gföhler, A. Karg

Institute of Machine Elements and Machine Design, University of Technology, Vienna, Austria

SUMMARY

To clarify the different results of our simulation and FES-cycling tests, measurements on a knee dynamometer were made. The m. quadriceps of 16 healthy test persons was activated both by FES and voluntary contraction. Stimulated with the same level of intensity in a knee flexion angle range from 5° to 105°, the diagrams showed a very unusual course. The knee torque shows its maximum at the knee flexion angle of approx. 30°. Additional isometric measurements using stimulation intensity on constant on-verge-to pain levels for different knee angles were made. The measured courses of the resulting knee torque as a function of the knee angle are much closer to the results of physiologically activated muscle. These measurements show that for optimum power release, the stimulation intensity must be regulated depending on the knee flexion angle.

STATE OF THE ART

The loss of voluntary muscle control below the spinal cord lesion limits mobility of the spinal cord injury (SCI) subjects. Additionally the patients get atrophy of the muscle, bone demineralisation, decubitus ulceres, heart disease and a general loss of cardiopulmonary fitness. Paraplegics are able to use a bicycle by means of Functional Electrical Stimulation (FES). The aim of this study was to find the cause of the discrepancy (deviation) between the simulation /5/ and the measurements /2/ of cycling for paraplegics.
Schutte et al. /7/ published the muscle joint torques dependent on the knee flexion angle for the isometric contraction by FES for paraplegics. But the isometric muscle joint torques are only relevant to start cycling. Stein et. al. /8/ measured knee torques on both SCI subjects and healthy persons in motion. They described that the results of the voluntary contraction were quite reproducible, while the amplitudes resulting from stimulated contraction decayed substantially between trials. To get parameters for the simulation of the cycle movement we measured the knee torque at different angular velocities. As the m. quadriceps is basically responsible for the power output (/1/, /6/), we chose this muscle for the measurements.

MATERIAL AND METHODS

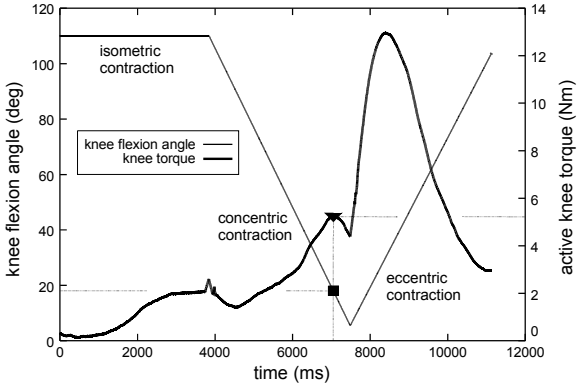
16 healthy persons (eight female and eight male, the average age of our test group was 32,4 years) who never had been treated with FES before have been tested. The measurements were made on a knee dynamometer, that was developed at our institute /3/. Four surface electrodes (rectangular self adhesive electrodes, 50x130 mm) have been placed on the motor points of the m. quadriceps and the muscle was stimulated by the programmable stimulation unit ‘Compex Sport p’. The stimulation frequency was 30 Hz. This low stimulation frequency was chosen to reduce muscle fatigue. The amplitude of the stimulation current gradually ramps up to and down from a plateau level (compare /7/). Without the ramps the spasms increased. At first the test person’s quadriceps is warmed up for ten minutes. The stimulation intensity was chosen individually for each test person depending on their pain sensitivity at a knee flexion angle of 15°. It is necessary to extend the leg almost completely, as in this position the pain sensitivity is at its maximum. The individual stimulation intensity levels attained in this manner had the effect that none of the test persons suffered from pain during tests. After getting familiar with the test procedure, measurement cycles (Table 1) were done on two different days. These measurement cycles

consisted of FES and voluntary physiological stimulation. For both stimulation types measurements were made for isometric, concentric and eccentric contraction. For the calculation of the active knee torque the passive knee torques (inertia force and mass moment of inertia, joint torques from passive muscle forces) had to be subtracted from the total measured torque. For example: to hold the leg at the knee flexion angle of 5° the m. quadriceps must overcome the weight of the shank and foot and the passive torque of the antagonist muscles (hamstrings, m. gastrocnemius and others).

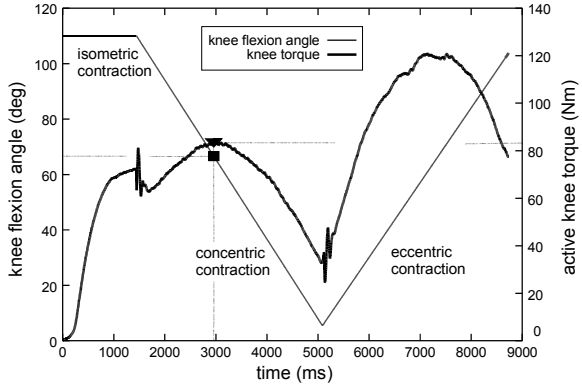
cycle	1	2	3	4	5	6	7	8	9	10	11	12	13
start angle	110°	110°	110°	110°	110°	110°	105°	91°	75°	61°	45°	31°	15°
end angle	5°	5°	5°	5°	5°	5°	104°	90°	74°	60°	44°	30°	14°
angular velocity	15°/s	30°/s	60°/s	90°/s	120°/s	180°/s	1°/s	1°/s	1°/s	1°/s	1°/s	1°/s	1°/s

Table 1. Measurement cycles

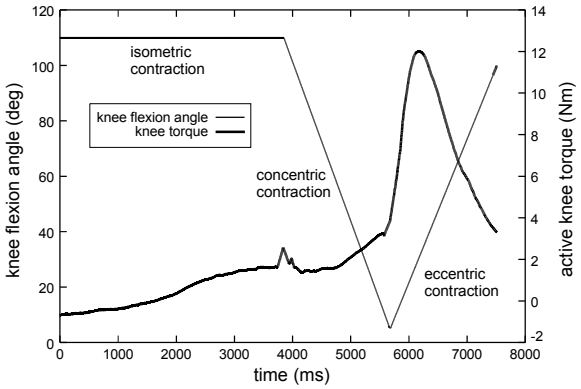
RESULTS AND DISCUSSION



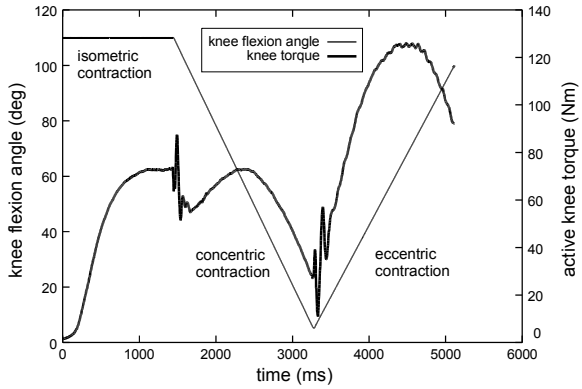
a) FES at 30°/s



c) voluntary contraction at 30°/s



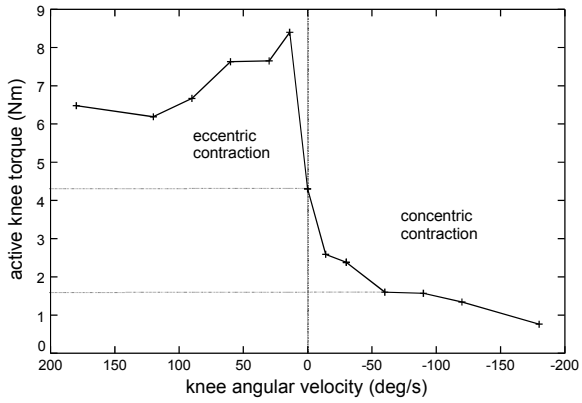
b) FES at 60°/s



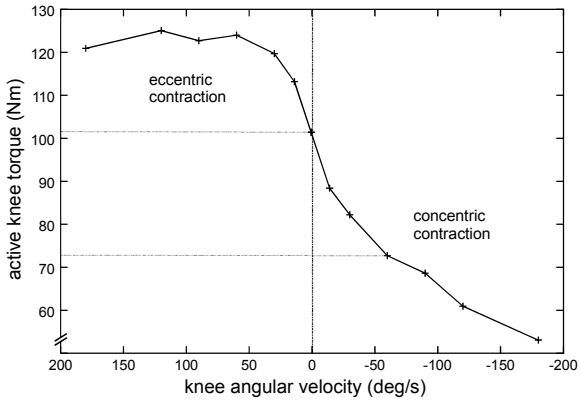
d) voluntary contraction at 60°/s

Fig. 1 a) to d). Active knee torque and knee flexion angle as functions of time; averaged results of all test persons (both legs, both test days); maximum knee torque ▼ in concentric contraction at the knee flexion angle (marked with ■) at 30°/s

When comparing the results, it becomes evident that the knee torques of the FES contraction measurements (fig. 1 a – b) are clearly lower than the knee torques measured with voluntary contraction (fig. 1 c – d). This applies for both isometric and concentric contractions. The peak knee torque obtained by FES is reached at clearly lower knee flexion angles when contracted concentrically. This could have to do with the adjustment of the stimulation intensity at this angle range (see Methods).



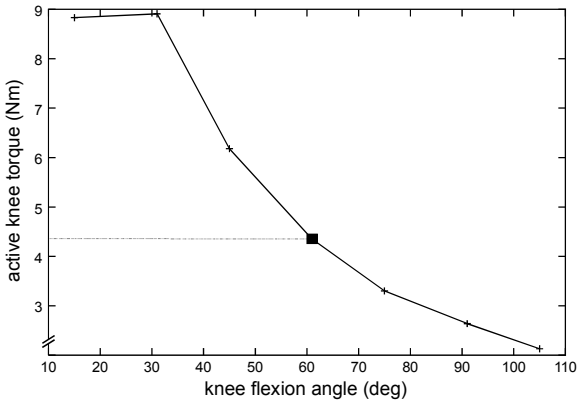
a) Contraction by FES



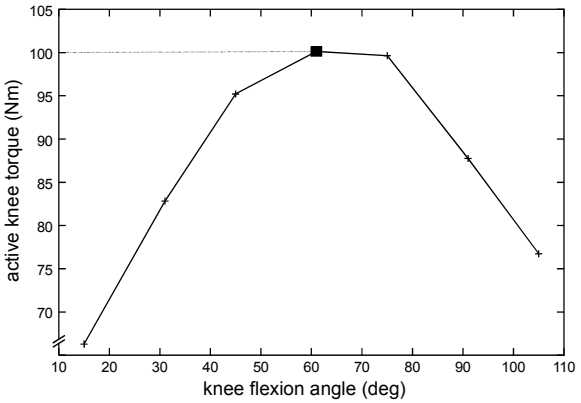
b) Voluntary contraction

Fig. 2. Knee torque as function of knee angular velocity at a knee flexion angle of 60°; averaged results of all test persons

The diagram (fig. 2 a; FES) showing knee torque over contraction velocity differs substantially to the force velocity relation of activated muscle tissue /9/. Voluntary contraction shows a similar course to the force velocity relation of activated muscle tissue. It is striking that the measured torque declines during rise of the flexion angular velocity in the eccentric contraction by means of FES. At the knee extension velocity of 60°/s during voluntary concentric movement over 70% of the isometric knee torque is reached (see fig. 2 b). Less than 40 % of the isometric knee torque achieved by FES is obtained at the same knee angular velocity.



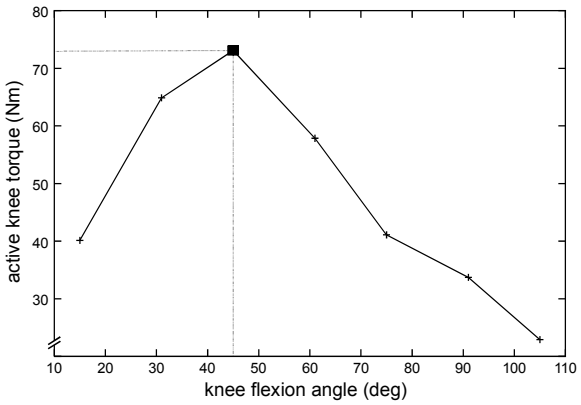
a) Contraction by FES



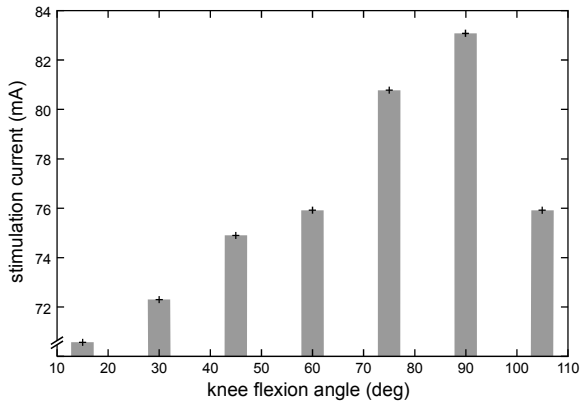
b) Voluntary contraction

Fig. 3. Isometric active knee torque as function of the knee flexion angle with constant stimulation intensity; averaged results of all test persons; knee torque at the knee flexion angle 60° marked with (■)

The course of torque in fig. 3 a) is not explainable with the current available muscle models. The EMG activity over the knee angle in fig. 3 b) is not constant /4/. These two observations prompted us to repeat the isometric measurements alternatively in a second test cycle with six test persons (two females and four males; the average age was 28,0 years). This time knee torques were measured isometrically at 7 different knee flexion angles. The other difference to the first test cycle was, that the stimulation intensity was adjusted to the maximum individually for each test person and at each knee flexion angle as done at 15° in the first test cycle (fig. 4 b). From the group of 16 the 6 test persons with the lowest pain sensitivity were chosen so that high stimulation intensity levels could be used. The resulting course of active knee torque over knee flexion angles (fig. 4 a) has a greater resemblance to that of voluntary contraction. The difference between the two is their maxima; being at 45° knee flexion angle with FES (second cycle) in comparison to 60° with voluntary contraction.



a) Contraction by FES



b) Stimulation current

Fig. 4 a. Isometric knee torque as a function of knee flexion angle with subjectively adjusted stimulation intensity; averaged results of 6 test persons; maximum knee torque at the knee flexion angle of 45° (marked with ■); fig. 4 b. Subjectively adjusted stimulation current as a function of the knee torque; averaged results of six test persons

CONCLUSION

In present applications of FES the knee flexion angle is read by the control unit in order to protect the lower extremity against hyperextension of the knee joint. The measurement results presume that it is necessary to vary the stimulation intensity according to the knee flexion angle to obtain optimal power output with FES. To achieve this, it will be necessary to develop stimulation units that are able to read the knee flexion angle and adapt the stimulation intensity accordingly.

REFERENCES

/1/ Angeli T. (1996). Leistungssteigerung bei Fahrradantrieben (Improvements of performance in bicycle drive units). Ph.D. thesis, Vienna Univ. of Techn., Vienna, Austria.

/2/ Angeli T., Gföhler M., Eberharter T., Lugner P. and Rinder L. (1998). Testbed for measurements on pedaling by Functional Electrical Stimulation. In: 15th Symp. „Danubia-Adria“ on experimental methods in solid mechanics, Bertinoro, 23-24.

/3/ Angeli T. (2000). Dynamometer for the measurement of torques on human joints. In: XVI IMEKO World Congress 2000: Proceedings Volume VII. Vienna. 167-172.

/4/ Bochdansky T., Lechner H. & Krista K. (1990). The EMMG (Electro-Mechano-Myogramme) - A new approach towards testing muscle elasticity. In: Proc. XXIV World Congress of Sports Medizine. Amsterdam. 300-304.

/5/ Gföhler M., Angeli T., Eberharter T. & Lugner P. (1999). Dynamic simulation of cycling powered by lower extremity muscles activated by functional electrical stimulation. In: Proc. XIIth Biomechanics Seminar (ISSN 1100-2247). 101-122.

/6/ Pawlik R. (1995). Biomechanik des Radfahrens (Biomechanics of cycling). Ph.D. thesis, Vienna Univ. of Techn., Österr. Kunst und Kulturverlag, Vienna.

/7/ Schutte L.M., Rodgers M.M., Zajac F.E. and Glaser R.M. (1993). Improving the efficacy of electrical stimulation-induced leg cycle ergometry. an analysis based on a dynamic musculoskeletal model. IEEE Trans. Rehab. Eng., 1, 109-125.

/8/ Stein, R.B., Momose, K. and Bobet, J. (1999). Biomechanics of human quadriceps muscles during electrical stimulation. J. Biomechanics, 32, 347-357.

/9/ Zajac F. (1989). Muscle and tendon: Properties, models, scaling and application to biomechanics and motor control. Biomedical Engineering, 17, 359-411.

AUTHOR'S ADDRESS

Dr. Thomas Angeli
Inst. of Machine Elements and Machine Design
Vienna University of Technology
Getreidemarkt 9/306, 1060 Vienna

e-mail: thomas.angeli@tuwien.ac.at
home page: www.imel.tuwien.ac.at

EFFECTS OF NMES IN PATIENTS WITH REFRACTORY HEART FAILURE

M. Bijak [(1)], W. Mayr[(1)], M. Quittan[(2)], A. Sochor[(2)], R. Pacher[(3)], E. Unger[(1)], D. Rafolt[(1)], H. Lanmueller[(1)], S. Sauermann[(1)], G. Freilinger[(4)]

[(1)] Department of Biomedical Engineering and Physics,
[(2)] Department of Physical Medicine and Rehabilitation
[(3)] Department of Radiology, [(4)] Department of Plastic and Reconstructive Surgery
all University of Vienna, University Hospital (AKH), Vienna, Austria

SUMMARY

Skeletal muscle strength and mass are severely impaired in patients with chronic heart failure.
The aim of the study was to determine the impact of Neuromuscular Electrical Stimulation (NMES) on the thigh muscles of this group of patients. Forty-two subjects with stable disease course were assigned randomly to a stimulation group (SG, mean age: 59y) and a control group (CG, mean age: 57y). A modified version of a space approved eight-channel electrical stimulation device was used to exercise SG for eight weeks.
Control parameters were isometric and isokinetic thigh muscle strength and muscle cross sectional area (CSA). An increase in muscle strength of 22.7% for knee extensors and 35.4% for knee flexors could be demonstrated in SG while CG remained unchanged or decreased by 8.4% in extensor strength. CSA increased in SG by 15.5% and in CG by 1.7%.
NMES of thigh muscles in patients with refractory heart failure is effective in increasing muscle strength and bulk.

INTRODUCTION

In long-term space flights reduced gravity causes muscle atrophy, orthostatic hypotension and reduction in blood volume. Physical activity and daily training are used as countermeasure /1/. As alternative a stimulation system for leg muscles was developed and transferred to space station MIR where two cosmonauts used it on a regular basis for 6 hours per day. After returning to the earth both reported a better muscle status and a shorter rehabilitation time in comparison to previous space flights /2,3/.
The introduced study is based on these findings and investigates the effects of NMES on non-healthy subjects with reduced physical abilities.

MATERIAL AND METHODS

Forty-two patients with an established diagnosis of severe chronic heart failure, making them listed for heart transplantation, were included in the 17 month lasting study at the Department of Physical Medicine and Rehabilitation, University hospital Vienna.
Randomization to the NMES group (SG) or the control group (CG) was done after baseline measurements in accordance with a block-wise randomization list. The personnel performing all measurements was not aware of the patients group assignment. All patients were reviewed once a week to check the skin and the equipment and to encourage adherence to the stimulation program. The CG patients were asked about their medical condition and encouraged to continue their usual activities of daily living. The overall duration of follow-ups was 8 weeks.

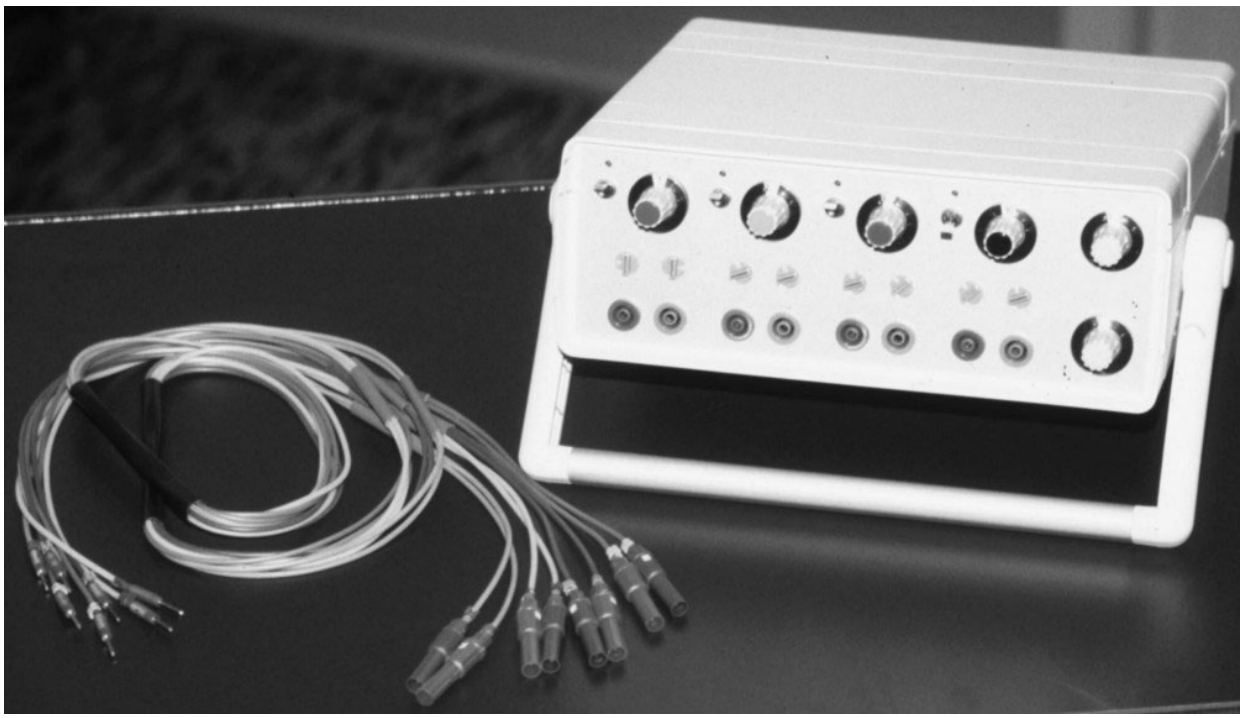


Fig. 1: Four channel surface stimulator

The stimulation device offers four independent channels with transformer isolated constant voltage outputs. Each channel is controlled by a micro-controller (PIC 16C57, Microchip, Chandler, AZ, USA) allowing independent adjustment of pulse width, stimulation frequency and stimulation amplitude.
Variation of the amplitude can be easily done with a rotary knob while alteration of pulse width and frequency required a special tool. All control elements and connection cables are consequently color coded to ease the handling. For safety reasons the device is powered with a rechargeable battery that is charged externally in a commercially available device (Fig. 1).
Multi-usable hydrogel surface electrodes (130cm^2, Bentronic, Munich, Germany) were placed bilaterally on the thighs to stimulate quadriceps and hamstring muscles.
Biphasic impulses with duration of 0.7ms+0.7ms and a frequency of 50Hz were applied in a 2s on and 6s off regime. The amplitude was set to achieve a strong tetanic contraction, corresponding to 25 to 30% of maximum voluntary contraction. After the patient was introduced to the handling of the equipment stimulation started at home with 30 minutes per day, 5 days a week and increased to 60 minutes per day after 2 weeks.
Muscle strength was evaluated with the leg extension apparatus of Cybex 6000 dynamometer (Cybex, Henley, USA). After a warm up with increasing effort consisting of 3 sets of sub-maximal isometric and isokinetic repetitions, 3 reciprocal knee extension and flexion movements with an angular speed of 60 degrees per second were performed with maximal effort. The highest value achieved was regarded as peak torque. After 4 minutes rest the maximal isometric strength of extensor and flexor muscles was measured at a knee angle of 60° by pushing and pulling as hard as possible against the fixed lever arm for 3 seconds.
The CSA of the mid-thigh was evaluated by computer tomography using a single slice technique and a semiautomatic segmentation with automatic fat tissue and bone exclusion.

As secondary outcome measures we used a muscle fatigue protocol, the NYHA functional classification, the functional assessment of activities of daily living related to leg muscle strength (ADL score), and parts of the Medical Outcome Study Short Form 36 questionnaire.
The detailed results are published in /4/.

	NMES (n=17)		**Control group (n=16)**	
Age	59 (±6)		57 (±8)	
BMI	22.7 (±3.2)		25.7 (±3.9)	
	Baseline	8 weeks	Baseline	8 weeks
Knee ext. IMPT [Nm]	109.5 (±37.9)	130.9 (±40.3)*	124.7 (±41.6)	115.8 (±42.3)
Knee ext. IKPT [Nm]	85.9 (±27.8)	103 (±28.9)*	103.8 (±39.4)	94.0 (±37.6)*
Knee flex. IMPT [Nm]	57.5 (±25.0)	69.2 (±26.4)*	60.1 (±18.9)	56.4 (±18.4)
Knee flex. IKPT [Nm]	44.3 (±19.0)	55.2 (±18.7)*	52.5 (±20.5)	49.5 (±18.3)
CSA [cm2]	98.5 (±27.6)	111.3 (±24.2)*	104.4(±21.6)	106.4 (±22.8)

Tab 1: Isometric peak torque (IMPT) and isokinetic peak torque (IKPT) of knee extensor and knee flexors, cross sectional area (CSA) of mid-thigh muscles for NMES group and control group. Body mass index (BMI). All values mean (±SD); * $p<0.001$

RESULTS

33 patients completed the study, nine dropped out for several non NMES- related reasons.
Baseline measurements of both isometric and isokinetic muscle strength did not differ significantly between SG and CG. After eight weeks SG showed increased isometric and isokinetic muscle strength of both extensors and flexors ($p<0.001$) while CG remained unchanged. Increase of CSA was in SG significant ($p<0.001$) as well (Tab. 1). Isometric muscle strength of both muscle groups adjusted for total muscle CSA did not show significant differences in the observed groups.

DISCUSSION

NMES proved to be an effective and safe therapy to revert leg muscle wasting, enhance leg muscle strength and improve activities of daily living in patients with severe heart failure awaiting transplantation. NMES has no side effects, can be administered by the patients at home and is economical. Therefore, the present study leads us to conclude that NMES of thigh muscles should be a promising adjunct to drug therapy in patients with severe heart failure.

REFERENCES

/1/ V.A.Convertino, Exercise as a countermeasure for physiological adaption to prolonged spaceflight, Med Sci Sports Exerc, 28/8, 999-1014, 1996

/2/ W. Mayr et al, FES as a Countermeasure against Muscular Atrophy in long-term Space Flights – First Application on Board of MIR-Station, „5th Annual Conference of the IFESS“, Aalborg, Proceedings, 27-30; 2000

/3/ W. Mayr et al, Myostim - FES to prevent from muscle atrophy in microgravity and bed-rest: Preliminary report, Artif. Org., 23/5, 428-431, 1999

/4/ M. Quittan et. al., Improvement of thigh muscles by neuromuscular electrical stimulation in patients with refractory heart failure, Am J Phys Med Rehabil, 2001

ACKNOWLEDGEMENTS

We thank the Austrian Federal Ministry of Education, Science and Culture for funding this study.

AUTHOR'S ADDRESS

Manfred BIJAK, PH.D.
Department of Biomed. Eng. an Physics
Waehringer Guertel 18-20/4L
1090 Vienna, Austria

Tel.: +43-1-40400/1992 (1983)
Fax.: +43-1-40400/3988
E-Mail: m.bijak@bmtp.akh-wien.ac.at

PARAPLEGIA: THE IMPLANTABLE PRAXIS FES-24B SYSTEM FOR MULTI-FUNCTIONAL RESTORATION.

Ross Davis, Thierry Houdayer, Andrew Barriskill*, Zoran Milijasevic*, Brian Andrews**.

Neural Engineering Clinic, Augusta, Maine, U.S.A.

*Neopraxis Pty. Ltd., Lane Cove, N.S.W., Australia.

** National Spinal Injury Center, Stoke Mandeville Hospital, U.K.

SUMMARY

Neopraxis Pty. Ltd. (Lane Cove, N.S.W., Australia) is manufacturing the new implantable Praxis FES24-B System to provide multiple benefits to suitable paraplegic users: Bladder Control, Upright Functional Mobility, Pressure Relief and Lower Extremity Exercise. The implant achieves these functions through epineural stimulation of nerves in the legs, the medial lumbar region, the sacrum and the conus medullaris. A body worn controller, the Navigator, transcutaneously powers and controls an implanted Stimulator via a magnetically held Transmit Coil/Antenna. The Stimulator is connected to 22 Electrodes by flexible and stretchable insulated leads. Sensor Packs attached to each thigh, shank and the trunk, provide the Navigator's software Strategies with real time information on the position of the lower extremities and the trunk. To assist simple locomotor functions, we will focus on how the system can complement the use of a wheelchair and be helpful in overcoming obstacles to wheelchair access especially doorsteps and unadapted bathroom facilities. In addition, being able to stand up to reach objects and perform prolonged manual tasks would be convenient for many workplace and home situations.
The FES24-B Stimulator provides up to 8mA of charge-balanced stimulation current via a biphasic waveform. Pulse widths can be varied from 25μsec up to 500μsec. The total pulse rate can range up to 14,400 pulses per second. The Stimulator provides real time data telemetry functions including the ability to measure the impedance of the current path through each electrode and the ability to transmit voltage measurements from each electrode. Since 1991, two paraplegic subjects have been implanted with earlier models. A Multi-centre Clinical Trial the system will commence in August 2001.

STATE OF THE ART

The authors aim has been to develop a generic FES implant for the restoration of function in spinal cord injured (SCI) paraplegic individuals, the functions or modes of which can be matched to an individual's requirements: Upright Functional Mobility, Pressure Relief and Lower Extremity Exercise, Bladder Control (1-4). In addition, for bladder control, less invasive surgical procedures were proposed to avoid posterior conus rhizotomy, and sacral laminotomy in order to access the sacral nerve roots for stimulation (3,4). It is hoped that this system will offer more functions and less surgery to patients with a cost- benefit ratio. We call this new approach "Multi-Functional".
To assist simple locomotor functions, we will focus on how the system can complement the use of a wheelchair and be helpful in overcoming obstacles to wheelchair access especially doorsteps and unadapted bathroom facilities. In addition, being able to stand up to reach objects and perform prolonged manual tasks would be convenient for many workplace and home situations (2-4).

METHOD

A] **Controller** ('Navigator'): A body-worn controller transcutaneously powers and controls the implanted Stimulator via a Transmit Coil (Fig. 1). The Coil is held in place on the skin surface over the Stimulator by magnetic force. This battery-powered Controller runs software Strategies designed to provide the user with a variety of functions. A touch sensitive LCD and remote control units provide simple menu driven operation of the FES system.

Fig. 1

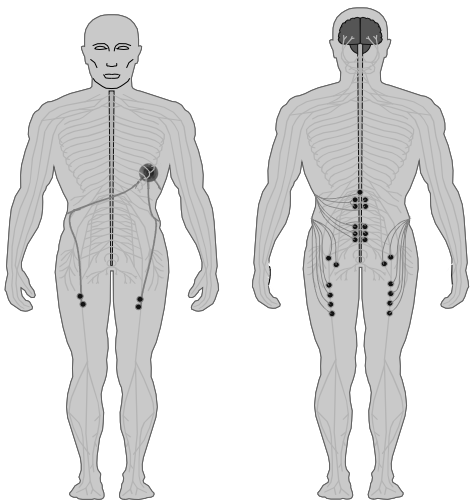


Fig. 2

B] The **FES24-B Stimulator** is the primary implanted component and is placed subcutaneously above the left costal margin. The FES24-B provides 22 stimulation channels and two-monopolar case Electrodes. The FES24-B utilizes a CIC3 integrated circuit (which is used in Cochlear Ltd.'s Nucleus 24 implants). The "CIC3" integrated circuit implements a high-speed radio-frequency data protocol. It can operate in common ground stimulation mode (all non-stimulating Electrodes are placed in parallel to provide the current return path) or monopolar mode (a case mounted electrode provides the current return path). The Stimulator provides real time data telemetry functions including the ability to measure the impedance of the current path through each electrode and the ability to transmit voltage measurements from each electrode. The FES24-B Stimulator provides up to 8mA of charge-balanced stimulation current via a biphasic waveform. Pulse widths can be varied from 25μsec up to 500μsec. The total pulse rate can range up to 14,400 pulses per second.

D] **Leads and Electrodes:** The implanted Stimulator is connected to 22 Electrodes (Fig.2) by highly flexible and stretchable insulated leads. Eight Electrodes are implanted in each lower extremity adjacent to motor nerves. Two Electrodes activate the quadriceps, two the hamstrings, two the ankle and two the gluteal muscles. These Electrodes facilitate the system's mobility, exercise and pressure relief functions. Two electrodes are implanted unilaterally in the medial lumbar region to activate the psoas muscle for hip flexion. Three additional Electrodes are implanted bilaterally adjacent to three sacral roots (S2, S3 and S4). These Electrodes facilitate the system's bladder control function. Pulsatile voiding is produced by periodically stimulating the sacral roots causing the bladder's detrusor muscle and sphincter to contract and then allowing the sphincter to relax at its faster rate. The final Electrode is implanted epidurally directly over

the conus medullaris at the T12-L1 level. This Electrode provides a neuromodulation function to assist with a reflexive bladder. It is anticipated that this technique may reduce the need to perform a posterior rhizotomy.

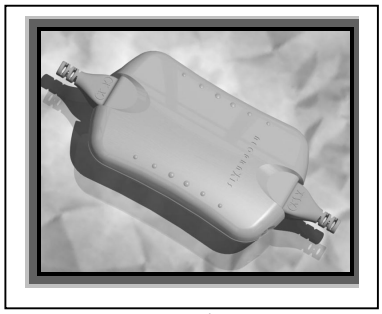


Fig. 3

C] **Sensor Packs,** (Fig.3) attached to each thigh, shank and the trunk, provide the Navigator's software Strategies with real time information on the position of the lower extremities and the trunk. Each microprocessor controlled Sensor Pack contains a miniature rate gyroscope and two 2-dimensional accelerometers. The Sensor Packs derive their power from the Navigator avoiding the need for additional batteries.

E] **Pre-surgical Subject Training**: The Praxis system includes a FES surface stimulator (Fig. 4), the *ExoStim*, which is intended to provide pre-surgical training of subjects in order to condition muscles and familiarize patients with FES training. The *ExoStim* is controlled by the Navigator and allows simple lower extremity software Strategies to be utilized prior to surgical implantation. The *ExoStim* is powered by 'AA' batteries and can deliver up to 200mA of biphasic stimulation to 8 channels with pulse-by-pulse control of the stimulation parameters. The *ExoStim* has extensive built in status detection capabilities including over voltage detection, pulse-by-pulse compliance checking, battery status monitoring and stimulation pad impedance.

RESULTS

Subject F.R, a 35 yr. old paraplegic male (F.R.: T10; ASIA: A), was implanted with an initial version of the Praxis system (FES24-A stimulator) in August 1998. This earlier implanted stimulator suffered breakage of its internal antenna coil after 8 months of operation. Redesign work has overcome this problem by inclusion of the stimulator's antenna coil within the FES24-B stimulator's rigid polymer housing.

A] Exercise and standing: For the year prior to his implantation, F.R. was able to stand without knee bracing using a combination of the Andrews' Anterior Floor Reaction Orthosis (5) and closed-loop skin surface FES applied directly over the femoral nerves. With closed-loop control of stimulation, he would typically stand uninterrupted for 30 minutes, and up to 70 minutes. With training, F.R. has achieved the 'C' posture and can stand with the stimulation 'OFF' for more than 50% of the standing time (3,4). After implantation of the Praxis FES 24-A system in August, 1998, subject F.R., carried out an FES exercise routine, FR found that daily stimulation decreased his muscle spasms and spasticity.

B] **Bladder Results:** On September 4th 1998, with urodynamic testing, F.R.'s sacral roots (S3 & 4) were bilaterally stimulated intermittently. On 3 occasions F.R.'s bladder contracted with

recorded pressures of between 45 and 50 cm of water. On December 14th 1998, urodynamic testing again showed consistent results from S3 & 4 sacral root stimulation producing 3 sustained bladder contractions with pressures of 40-55 cm water and urination with each stimulation pattern (5 sec on / 5 sec off, 20 Hz, 8 bursts). On April 2nd 1999, urodynamic testing was repeated with 2 bladder reflex activations from each pattern of stimulation (5 sec on / 5 sec off, 20 Hz, 8-14 bursts). Pressures of 50-70 cm water were recorded (4).

A FDA approved Clinical Trial of the Praxis24-B system is planned to start August 2001.

DISCUSSION

In the developing field of FES and implantable neural prosthetic devices, there has been a need for reliable and safe, multi-channel implantable stimulating systems to restore function in neurologically impaired patients. In paraplegic individuals, the stimulating systems' functions should be designed to modulate spasticity and precisely activate individual muscles for joint movement and control of bladder and bowel functions. The more channels available, the more nerves can be activated and the more modes of functionality can be restored. Our contribution to this aim has been continuous since 1983, and the Praxis FES System (3,4), its FES24-B stimulator, Navigator body worn controller and the connected Sensor Packs for sensing of joint and body position in the paraplegic subjects (4) provides the hope for a new rehabilitation aid for restoration of function in spinal cord injury paraplegia.

REFERENCES

1. Davis R, MacFarland W, Emmons S. Initial Results of the Nucleus FES-22-Implanted Stimulator for Limb Movement in Paraplegia. Stereotact. Funct. Neurosurg. 1994; 63:192-197.
2. Davis R, Houdayer T, Andrews B, Emmons S, Patrick P. Paraplegia: Prolonged Closed-Loop Standing with Implanted Nucleus FES-22 Stimulator and Andrews Foot-Ankle Orthosis. Stereotact. Funct. Neurosurg. 1997; 69:281-287.
3 Davis R, Houdayer T, Andrews B, Barriskill A. Prolonged Closed-Loop Functional Electrical Stimulation and Andrews Ankle-Foot Orthosis. Artif. Organs, 1999; 23: 418-420.
4. R. Davis, J. Patrick, A. Barriskill. Development of Functional Electrical Stimulators utilizing Cochlear Implant Technology. Medical Electronic and Physics, 2001, 23: 61-68.
5. Andrews et al. Hybrid FES Orthosis Incorporating Closed Loop Control and Sensory Feedback. J.Biomed Eng, 1988; 10: 189-195.

ACKNOWLEDGMENTS

This work was in part funded by Cochlear Ltd. and Neopraxis Pty. Ltd., Lane Cove, NSW, Australia, Neural Engineering Clinic Research Foundation, by the Paralyzed Veteran of America SCRF grant #1246, Veterans Administration Research & Rehabilitation Grant, and NIH-SBIR Phase 1 grant (1R43 HD38494-01).

AUTHOR'S ADDRESS

Ross Davis M.D., Neural Engineering Clinic, 76 Eastern Av., Augusta, ME, 04330, USA.

[email: rosdavis@qsilver.net].

A STIMULATOR FOR FUNCTIONAL ACTIVATION OF DENERVATED MUSCLES

C. Hofer[*], W. Mayr[**], H. Stöhr[***], E. Unger[**], H. Kern[*]

[*]Ludwig Boltzmann Institute of Electrostimulation and Physical Rehabilitation,
Department of Physical Medicine, Wilhelminenspital Wien, Austria
[**]Department of Biomedical Engineering and Physics, University of Vienna, Austria
[***]Department of Biomedical Research, University of Vienna, Austria

SUMMARY

In the last years various studies proved that electrical stimulation can improve contractile capability and restore muscle function in long-term denervated degenerated muscles. The low excitability of the muscle cells at the initial stage of training and surrounding connective tissue - acting as an electrical shunt - requires special stimulation parameters.
Until now no appropriate devices (stimulators) are commercially available, therefore we were forced to design our own stimulators.

The control unit of the stimulators is based on a microprocessor for maximum flexibility regarding the generation of the parameters such as pulse amplitudes, pulse width, frequency, stimulation times, ramps etc. In addition the microprocessor design allows recording of training data such as stimulation date, time, duration and used programs.

The voltage constant output stage of the stimulator is able to generate biphasic charge balanced stimulation pulse with pulse width of 1-300ms, stimulation voltages up to ±80V (160 V_{ss}) and 250mA maximum stimulation current. To prevent direct current due to inexact charge compensation, the stimulation pulses are coupled capacitively. Simultaneous two-channel stimulation with independent intensity is possible.

The stimulators are programmed with Notebook or Personal Digital Assistants (PDA's) via infrared serial interface. This concept avoids stimulation with wrong parameters because the patient can only use the stimulation parameters preprogrammed for him in the outpatient clinic. For training at home only changes in stimulation intensity within given limits are possible. The portable units are powered by an internal rechargeable battery ensuring mains isolation. Highly efficient switched voltage regulators are used for power supply of all circuits to increase operating time of the stimulator.

INTRODUCTION

Direct electrical muscle stimulation with long pulse width – up to 300ms – has been the common treatment for denervated muscle. But this therapy is still being considered to be controversial by many rehabilitation centers and medical professionals. This is largely because current teaching and training for therapists is still based on scientific and technological knowledge of the fifties and sixties. Other sources of criticism are the contradictory statements regarding its effect on nerve growth and re-innervation /1,2/. Commercially available stimulation devices are only able to slow down atrophy or maintain the muscle during recovery after non permanent denervation. The stimulation parameters cannot be changed in the scope necessary for efficient therapy of permanent denervated muscles. Most of the stimulators are only able to generate pulse widths of 300, 200 or 100ms

and much shorter pulses with about 1 ms pulse width but nothing between. Another limitation is induced by current EU standards for medical devices which are limiting the single pulse energy to 300mJ per impulse. This is not sufficient for stimulating denervated degenerated muscles (DDM) because of the connective tissue surrounding the muscle acting as an electrical shunt and the long pulse duration needed for excitation of the denervated muscle fibers.
In more recent studies investigating the effects of electrical stimulation on denervated muscles especially build research prototypes or inappropriate commercially available devices adapted to the specific requirements were applied. /3,4,6,7,8/.

CONCEPT OF A SUITABLE STIMULATOR

Based on experiences from previous experiments in the past years the following concept for the design of the stimulator was made /6/.
Stimulation pulse width should be changeable continuously in a range between 1 and 300ms depending on the excitability of the muscle fibers. The output stage of the stimulator has to be capable of generating pulse energies high enough for effective treatment of long-time denervated and degenerated muscles at an initial stage of training were extremely long pulse durations and high pulse intensities are required for training.
The necessity of different biphasic pulse shapes for treatment of patients with incomplete denervation has two main reasons. First for specific treatment of the denervated muscle fibers without eliciting contractions in the innervated parts of the muscle – altered accommodation in the denervated fibers. Second in case there are any sensory nerve fibers in the denervated area intact the use of triangular or trapezoidal shaped stimulation pulses is less distressing for the patient.
Multi channel simultaneous stimulation of different muscle groups should be possible in order to reduce daily training time.
The device should be controlled by a microprocessor for:

- highest flexibility regarding the generation of stimulation parameters,
- storing different stimulation programs that are pre-programmed at the outpatient clinic,
- easy handling of the stimulator by the patient (only a few control elements: start/stop, therapy program, intensity) and
- permanent recording of stimulation activities.

For safety the stimulator should be battery powered to avoid direct line connection and during recharging the battery the unit has to be inactivated.
Since there are no nerves distributing the stimulation impulse the use of large size electrodes is essential for ensuring a contraction of the whole muscle /6/. Electrodes made of soft flexible conducting rubber, that are applied with a wet sponge cloth or gel directly to the skin, already used in our previous trials, proved to be best suited for stimulation of denervated muscles.

RESEARCH PROTOTYPE

Based on the above concept a stimulation system specially adapted to the needs of patients with degenerated denervated muscles was designed. Main parts of the stimulator are the control unit for generating pulse parameters, storing stimulation programs and training records, the power supply with the rechargeable battery supplying all components and generating the necessary high voltages for stimulation and the output stage with two independent channels for simultaneous stimulation of two different muscle groups.

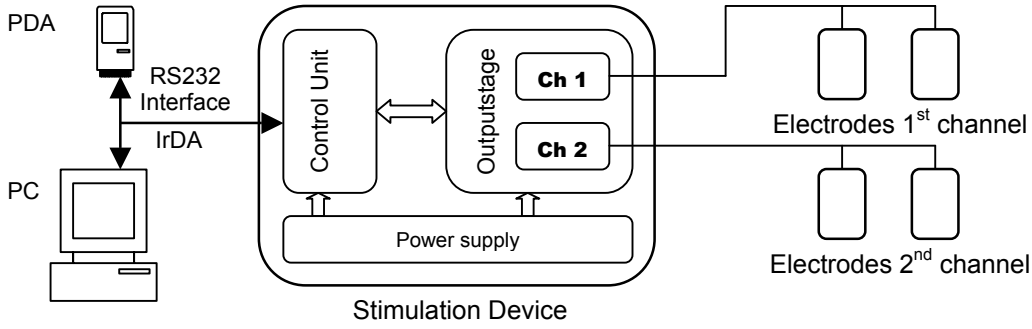


Fig. 1: Stimulation system overview

Power supply:
For powering the stimulation device special NiMH battery packs made from single high capacity cells with additional thermistors monitoring the temperature were chosen. All circuits in the control unit and the output stage as well as the stimulation circuits are supplied by switching regulators in order to increase efficiency. This allows long training sessions without the necessity of recharging the stimulator. When running the stimulator with both stimulation channels driven at maximum power the overall power consumption is about 35W. To prevent failure of the stimulator due to battery undervoltage an acoustic signal is warning the patient if battery low voltage occurs .

Output stage:
The output stage delivers current limited constant voltage impulses to avoid damage of tissue in case of electrode peeling or in case of electrode short circuits damage of the stimulator. Two separate stimulation channels with independent adjustable stimulation amplitude are realized. This allows simultaneous stimulation of two different muscle groups reducing the overall training time for the patient and thus increasing the compliance. To prevent direct current the stimulation pulses are capacitively coupled.

Fig. 2:PDA with developed software for programming the stimulator

Control unit:
Based on a 80C517 microprocessor the control unit is generating all stimulation parameters such as pulse width, pulse shapes (4 different), stimulation frequency, stimulation on- and off-time, ramps, amplitude, program duration etc. Via infrared serial interface the stimulator is programmed at the outpatient clinic according to actual status of the patient. This interface is activated by a magnetic switch to prevent unintentionally programming of the stimulator. The control unit records training program, date and time of activation and stores this information in an onboard memory. This information is kept till the next visit to the outpatient clinic where the data is downloaded to the Personal Computer and added to the patient record for evaluation.
A software for Personal Digital Assistants (PDA) comprising all features of the PC based application was developed allowing on-site programming with these small and cheap devices instead of the need to carry around a notebook. The control panel of the stimulator consists of 2 panel coders and 4 push buttons. For safe handling of the device the user access is restricted to selecting different programs and varying the intensity within a preset range.

DISCUSSION

With the developed stimulator training of DDM is possible increasing contractile capability and muscle bulk. Prolonged treatment of denervated degenerated muscle with electrical stimulation specially adapted to the current state of the muscle tissue improves the metabolism of the muscle cells and decreases the pulse width to get a muscle contraction. Shorter stimulation pulse widths sufficient for training due to the increasing excitability of the muscle fibers are allowing higher stimulation frequencies eliciting tetanic contractions that are necessary to achieve the desired muscle fiber tension, constituting a hypertrophic stimulus.
For further reduction of daily training time and functional training like standing up with electrical stimulation the development of devices with 4 stimulation channels is necessary. This requires new high capacity accumulators and a very efficient power supply for the devices to ensure a sufficient long operating time.
Electrode garments with integrated electrodes and cables to simplify donning and doffing will also reduce training time and improve safety preventing inappropriate connections and electrode placement.

REFERENCES

/1/ Boonstra A.M., Va Weerden T.W., Eisma W.H., Pahlplatz V.B.M., Oosterhuis H.J.G.H.: The effect of low-frequency electrical stimulation on denervation atrophy in man. Scand J Rehab Med. 19(3): 127-134, 1987.

/2/ Eberstein A., Eberstein B.: Electrical stimulation of denervated muscle: is it worthwhile? Med Sci Sports Exerc., 28(12): 1463-9, 1996.

/3/ Eichhorn K.F., Schubert W., David E.: Maintenance, training and functional use of denervated muscles. J Biomed Eng., 6(3):205-11, 1984.

/4/ Hofer C., Kern H., Mayr W., Stöhr H., Abou-Zahra S.: Funktionelle Elektrostimulation denervierter Muskulatur. Österr. Z. Phys. Med. 7 ,Supplement 2, 1997.

/5/ Kern H.: Funktionelle Elektrostimulation paraplegischer Patienten. Österr. Z. Phys. Med. 5 , Heft 1 Supplementum, 1995.

/6/ Kern H., Hofer C., Strohhofer M., Mayr W., Richter W., Stöhr H.: Standing up with denervated muscles in humans using functional electrical stimulation. Artif Organs., 23(5):447-52, 1999.

/7/ Mokrusch T., Neundorfer B.: Electrotherapy of permanently denervated muscle - long term experiment. Eur J Phys Med & Reha. 4(5):166-173, 1994.

/8/ Woodcock A.H., Taylor P.N., Ewins D.J.: Long pulse biphasic electrical stimulation of denervated muscle. Artif Organs. 23(5):457-9, 1999.

ACKNOWLEDGEMENTS

The authors would like to acknowledge the support of the “Jubiläumsfonds der Österreichischen Nationalbank”.

AUTHOR’S ADDRESS

Christian Hofer
Ludwig Boltzmann Institute of Electrostimulation and Physical Rehabilitation,
Department of Physical Medicine and Rehabilitation,
Wilhelminenspital Wien
Montleartstrasse 37,
A-1171 Wien christian.hofer@phys.wil.magwien.gv.at

FES AS A COUNTERMEASURE AGAINST MUSCULAR ATROPHY IN LONG-TERM SPACE FLIGHTS

W. Mayr[1], G. Freilinger[2], D. Rafolt[1], M. Bijak[1], W. Girsch[2], C. Hofer[1], H. Lanmüller[1], M. Reichel[1], S. Sauermann[1], C. Schmutterer[1], E. Unger[1], V. Gratchev[3], Y. Koryak [3], B. Shenkman [3], I. Belozerova [3],I. Kozlovskaya [3], A. Gregoriev [3]

[1] Department of Biomedical Engineering and Physics, University of Vienna, Austria
[2] Department of Plastic and Reconstructive Surgery, University of Vienna, Austria
[3] Institute for Biomedical Problems (IBMP), Moscow, Russia

SUMMARY

Long-term flights in microgravity cause atrophy and morphological changes of skeletal muscles. Conventional training is insufficient regarding time to exercise and space for equipment.
The objective of the project MYOSTIM is to develop a training method, based on FES to preserve muscle mass and fiber composition with minimal impairment to the cosmonaut.
For a pilot experiment on board of the MIR station a suitable 8-channel FES equipment was developed. It consists of electrode trousers, that carry surface electrodes and cables, two interconnected 4-channel stimulators, and a laptop-PC for stimulator programming and processing compliance data. An automatic extensive training of 4 muscle groups of the lower extremities is performed for 6 hours per day, with 1 s on / 2 s off tetanic contractions at 20 to 30% of maximum tetanic muscle force. Synchronous activation of antagonists of thigh and lower leg prevents from uncoordinated movements.
The first successful test on board of MIR was performed by 2 cosmonauts between December 98 and February 99 respectively March 99 and August 99.

INTRODUCTION

Long-term flights in microgravity cause atrophy and morphological changes of skeletal muscles. Extensive daily physical training using mechanical devices raises the caloric intake, shortens the operational activities and requires extreme motivation of the crew members. The limitation for an active muscle training during a long-term space mission in terms of time and space needs the consideration of an automatic support.
Functional Electrical Stimulation (FES) is well established in terrestrial rehabilitation and sport training since years [1] [2] [3]. It has a high potential to serve as an efficient countermeasure that avoids most of the cited impairments, as long as the equipment is comfortable and easy to handle under space conditions.
To investigate the effectiveness and practicability of FES as a countermeasure mean, a co-operation with IBPM in Moscow was established and led to the recent first two successful applications on board of MIR space station (Fig.1).

Fig. 1: Commander of MIR crew 26 using MYOSTIM on board of the station

MATERIALS AND METHODS

Principle of training and parameters: FES is applied to 4 muscle groups of both lower extremities. Electrodes are placed on the skin above the quadriceps femoris muscles, the hamstrings, the tibialis anterior- and peroneal muscles, and the triceps surae muscles. Synchronous stimulation of antagonistic muscle groups prevents from unwanted joint movements.
FES is applied additionally to the Russian routine physical training program, that comprises predominantly intermittent treadmill exercises and resistance exercises with bungee-cords for 1-2h/day, organised according to a 4-day cycle (Tab. 1).

Day	Goal	Work load	Intensity of load	Energy expenditures
1	Maintenance of high velocity muscle characteristics and orthostatic tolerance	small	submax./maximal	380-420 kcal
2	Maintenance of muscle strenghth-velocity properties	middle	middle	450-500 kcal
3	Maintenance of endurance and of movement co-ordination	high	small	550-600 kcal
4	Active rest, physical exercices of cosmonaut´s own choice	small	ad lib.	150 kcal

Tab. 1: Russian routine countermeasure training scheme

The FES training is performed during 6 hours per day with 1 sec “on“ and 2 sec “off“ trains at intensity levels of 20 - 30% of maximum tetanic force (MTF) and a frequency of 25 Hz.

Equipment: The technical equipment consists of electrode trousers carrying stimulation electrodes for the 8 channels, and 2 interconnected 4-channel stimulators carried on a belt.
The electrode trousers simplify handling and placement of the electrodes by a patented construction of two flexible flaps, carrying the electrodes and corresponding protection foils, that are alternatively exposed to the skin. All cables are integrated in the garment (Fig. 2).
The 8-channel stimulation system is divided into two 4-channel modules interconnected by an I^2C-bus (Fig. 3). The 4 channel module contains circuitry for M-wave and impedance-recording, the stimulation output stage, micro-controllers for impulse generation and measurement purposes (one for each channel), a co-ordinating micro-controller, the power supply, the graphical display, control elements and a bus-interface. The circuitry is miniaturised in SMD (surface mounted device) technology and integrated in a robust metal case. All stimulation and training parameters can be set by a personal computer (PC) via an RS232 link. The training protocol is transferred weekly to the PC via the RS232 link and stored in a database.
All technical solutions were tested in-vivo and underwent the standardized Russian qualification and certification test procedures.

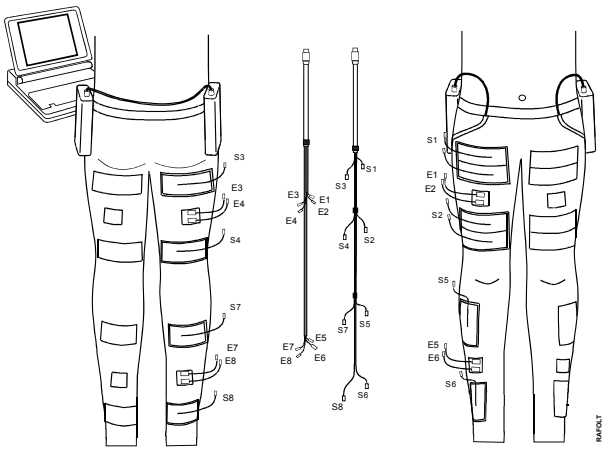


Fig. 2: Principle of electrode configuration and equipment design

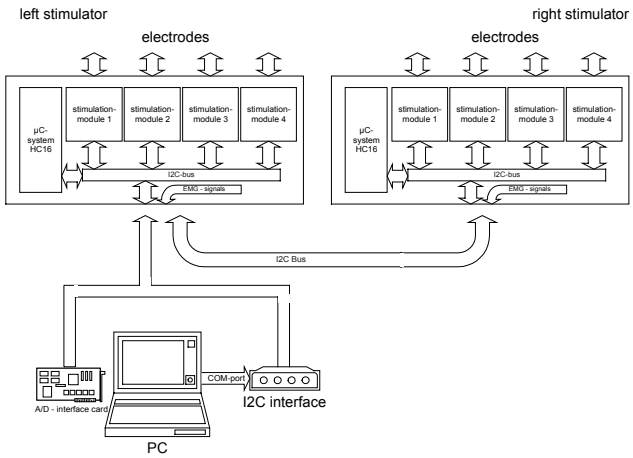


Fig. 3: System configuration with two 4-channel stimulators and laptop

Evaluation: Evaluation is done in accordance with the standard protocol of the Moscow IBMP, which is routinely used to investigate the effectiveness of various countermeasure means. The protocol contains physiological as well as morphological pre-flight and post-flight examinations, and ergometric tests additionally during flight. In addition the stimulation device records and stores all stimulation activities, M-wave and impedance data (compliance data).

RESULTS

The equipment was transported to the MIR space station for the first applications on October 27, 1998. It was successfully applied by one cosmonaut between December 98 and February 99 (Fig. 1) and by a second cosmonaut between March 99 and August 99. Both cosmonauts went to MIR station in August 98 and used the equipment in a later phase of their flight till their landing. Up to now only data of the first of the two applications are available.
During this first pilot application MYOSTIM was used for 3 hours per day additionally to the routine physical training program. 4 control cosmonauts from different missions with similar flight duration performed only the routine countermeasure training and no FES (Tab. 1). The compliance of training in volume and intensity was between 80 and 100 %.
The first data showed promising results. It has provided us with data of both handling of the equipment in microgravity and effectiveness of FES muscle training in space. However, we have only pilot results and further systematic investigations are absolutely required.
In comparison to the control group, who performed only the routine physical exercise program, the commander of crew 26, who used in addition MYOSTIM, was in much better condition during flight and after landing. This was especially confirmed by the pre-, in- and post-flight locomotion tests where he developed much lower heart rate and lactate levels.
Muscle contraction dynamics, investigated with tendometry (Tab. 2) and dynamometry, showed clearly better values for the FES user in post-flight investigations.

		Cosmonaut, routine physical training + FES	4 cosmonauts, routine physical training only
Single twitch	Peak amplitude	+134,5 %	-8,3 %
	Time to peak	+12,4 %	+4,0 %
	Half relaxation time	-2,4 %	-23,4 %
Max. voluntary contraction	Peak amplitude	-27,1 %	-43,5 %
Max. stimulated contraction	Peak amplitude	-12,9 %	-27,1 %

Tab. 2: Tendometry, triceps sure, changes in relation to pre-flight values

The histo-morphological investigations did show a similar reduction of fiber cross sectional area, when compared with the control results, but the typical atrophy-related increase of interfascicular connective tissue did not appear in the FES trained muscle. In contrary a decrease from 11% to 6% was observed. Cytochrome-C-oxidase indicated a substantial increase of aerobic metabolism of both type 1 (+152%) and type 2 fibers (+131%), an effect that was emphasized by an increase of capillary density by +174,5 %.
The positive influence of additional FES training was further confirmed in the posture stability tests and in reflex tests that showed significantly better results and a much earlier recovery.
The subjective judgement of both cosmonauts was extremely positive: There were no complains concerning daily handling of the equipment over months and practicability of the training during work, except seldom extremely fine-motorial tasks, when they had to switch of the stimulation temporarily. The reported improved fitness and well-being, the feeling of “complete muscle integrity” and the lack of previously experienced muscle pain.

DISCUSSION

To substitute the terrestrial muscular load during long-term space flights exercise and training programs

are required. The training and the devices consume extensively time and space, an alternative would be helpful. The objective of our project was to provide an alternative method for avoiding or at least reducing the changes in the neuromuscular system with minimal impairment to the cosmonaut.
Morphologically microgravity causes a loss of muscle mass and a reduction of type I muscle fibers, which are responsible for muscle tone and posture above all. It is common knowledge that extensive FES training tends to transform type II to type I fibers, an effect that seams to be useful to compensate the type I fiber loss in microgravity [3].
The level of 20 to 30% of MTF was chosen to achieve substantial training at minimal sensible inconvenience. Simultaneous activation of antagonistic muscle groups prevents unpleasant movements. The first test under space conditions showed, that this isometric low level training is comfortable and does not interfere with daily operational activities.
An exact simulation of terrestrial muscular activity cannot be expected from stimulation via surface electrodes. Distribution of fiber types in an FES trained muscle will always differ from a normal muscle. This effect is known to be totally reversible within several weeks after end of stimulation training under normal muscular activity, i.e. under terrestrial conditions [4].
The results of the functional tests, the faster functional recovery and the subjective judgement of the cosmonaut, who had experienced a long-term flight previously, lead to the assumption, that - besides pure muscle preservation - a major benefit of applied FES training lies in the stimulation of the proprioceptive system and induction of afferent activity.
The first applications in space have provided us with data of both handling of the equipment in microgravity and effectiveness of FES muscle training in space. However, we have only pilot results, that call for further systematic investigations. Provided that the technique further proves to be effective, the application should be extended to the trunk and neck muscles.
Profit for terrestrial applications in medicine can be expected, as long-term immobilization causes similar degenerative changes in the neuromuscular system. A first successful clinical study on patients with severe chronic heart insufficiency confirms this assumption [5].

ACKNOWLEDGMENTS

The work on this project was funded by the Austrian Ministry of Education, Science and Culture.

REFERENCES

[1] Gould N, Donnermeyer D, Pope M, Ashigaka I. Transcutaneous muscle stimulation as a methode to retard disuse atrophy. Clin Orthop 1982;164:215-220.
[2] Cabric M, Appell HJ. Effect of electrical stimulation of high and low frequency on maximum isometric force and some morphological characteristics in men. Int J Sports Med 1987;8:256-260
[3] Appel HJ, Cabric M, Resic A. Fine structural changes in electrostimulated human skeletal muscle-Evidenz for predominant effects on fast muscle fibers. Eur J Appl Physiol 1987
[4] Kirschbaum BJ, Pette D. Low-frequency stimulation of rat fast-twitch muscle induces rapid, reversible changes in myosin heavy chain expression. Sarcomeric and non-sarcomeric muscles: Basic and applied research prospects for the 90´s Carraro U, ed. Unipress Padova, 1988;337-342
[5] Quittan M, Wiesinger G, Sturm B, Puig S, Mayr W,et al. Improvement of thigh muscles by neuromuscular electrical stimulation in patients with refractory heart failure. A single-blind, randomized controlled trial. Am J Phys Med Rehabil 2001 Mar;80(3):206-14

AUTHOR´S ADDRESS

Dr. Winfried Mayr
Department of Biomedical Engineering and Physics
AKH, Ebene 4/L, Waehringer Guertel 18-20
A-1090 Vienna, Austria
E-mail: w.mayr@bmtp.akh.wien.ac.at

MOBILE PC-SYSTEM FOR INTRAOPERATIVE ELECTRONEURODIAGNOSTICS

M. Reichel, M. Bijak, W. Mayr, H. Lanmüller, D. Rafolt, M. Rakos, S. Sauermann, E. Unger, E. Turkof*

Department of Biomedical Engineering & Physics, University of Vienna
* Department of Plastic and Reconstructive Surgery, University of Vienna

SUMMARY

Intraoperative Electroneurodiagnostic (IOE) is a diagnostic tool used during surgery of peripheral nerves to precisely localize the site of lesions or to determine the functional status of the exposed nerve segments. In principle, the nervous structure to be investigate is stimulated at its most proximal (healthy) site, hereby evoking an efferent nerve compound action potential which is recorded from the surface of the nerve at the exposed site. Stimulation sites vary according to the structure to be investigate: for brachial plexus lesions, we perform Motor Evoked Potentials (transcranial electrical stimulation of the motor cortex), as we do for scoliosis surgery. For peripheral nerve lesions, we stimulate the spinal roots of the damaged nerve with surface electrodes placed paravertebrally above the respective nerve segments, and for unknown sites of facial nerve lesions, we stimulate the facial nerve transcrianally at its exit from the pons. Subsequent to the electrical stimulus, the evoked nerve action potential's shape we conclude to the function of the nerve: the amplitude of the Electroneurogramm is proportional to the number of functioning nerve fibers

Former measurements have been done using commonly used 2-channel electromyograph with self constructed, special prototype electrodes in combination with a commercially available stimulator (Digitimer 185). The Digitimer can deliver monophasic rectangular impulses with a fixed pulse width of 50µs amplitudes of up to 1000V. The findings using this equipment, like mains dependency, fixed pulse width, only 2-channel recording and impossibility of short latency ENG recordings due to overdriving the recording amplifier, led to the development of a 6-channel mobile (Laptop-PC), flexible and save (battery powered) system with newly designed multi channel recording electrodes for reliably reproducible measurements combined with an integrated adjustable biphasic electrical stimulator.

The preamplifier is equipped with a crowbar input protection circuit caused by the high stimulation amplitude (up to 1000V) and it recovers after overdriving within 1ms. To amplify signals in the range of 10µV with a bandwidth of 10kHz a high signal to noise ratio (SNR) is required. The stimulator is variable in amplitude up to 1000V and pulse width from 50µs to 1ms. Furthermore the discharge phase of the impulse is separately adjustable in duration to minimize stimulus artifact. The recording is triggered by the stimulation pulse and is active for 100ms, thus it can - together with parameters - be comparable stored in a database. Without important loss of time the system is rapid applicable in the operating room because of its save and easy handling.

STATE OF THE ART

In the course of Intraoperative Electroneurodiagnostic (IOE) electrical stimulation is applied by transcutaneous needle electrodes. The stimulus is a single, constant voltage, rectangular monophasic pulse up to 750V in amplitude and 100µs in duration (Digitimer 180) or up to 1000V and 50µs (Digitimer 185), respectively /1/ /2/ /3/.

Biphasic stimulation pulses are well known in Functional Electric Stimulation (FES) for charge balancing over trains of pulses to avoid electrode and tissue damage /4/. Usually rectangular biphasic stimulation

pulses with up to 100V in amplitude and up to 1ms in duration are used to activate peripheral nerve structures via surface electrodes for causing muscle contraction. For IOE, it is necessary to stimulate a variety of nervous structures like the motor cortex, the spinal root, the spinal cord or the facial nerves. For that purpose, higher stimulation amplitudes are necessary to reach the structures to be stimulated which lay under bony tissue. However, to the desired activation of the nervous structure may be accomplished by increased pulse width. This was defined by Lapique in 1926 and thus the intensity *I* of the stimulus can be expressed by

$$I = B \cdot \left(1 + \frac{C}{T}\right), \qquad \textit{Equ. 1}$$

where *B* represents the rheobase and *C/T* stands for the ratio of chronaxie *C* and pulse width *T*. The log-log plot of the strength duration curve can be seen in *Fig. 1*.

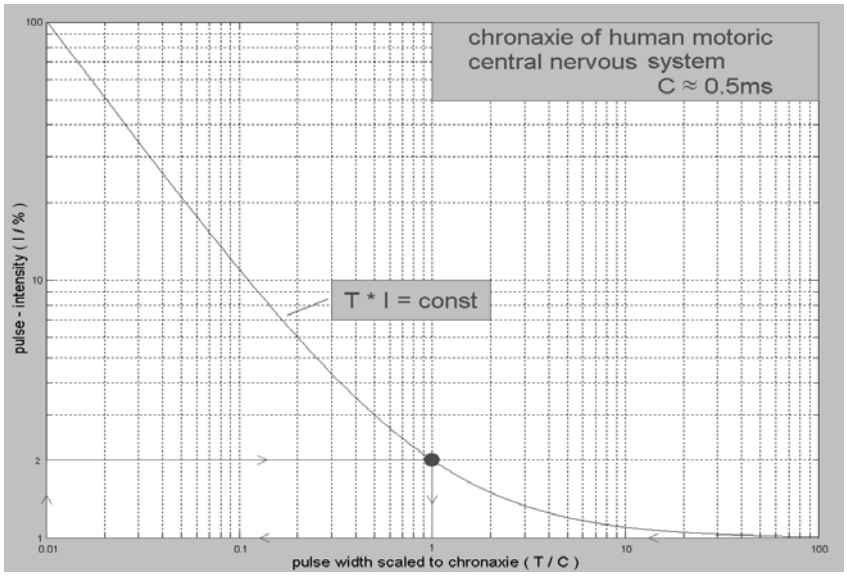


Fig. 1: Strength duration curve, scaled to chronaxie C and pulse-intensity I. Rheobase is chosen as 1% of maximum intensity and chronaxie of human motoric central nervous system is in the range of 0.5ms.

But pulse duration is limited as well, caused by the already mentioned short delay of the evoked potential. For that reason a proper ratio of amplitude and pulse width of the stimulation signal has to be found to minimize artifact and therefore optimize possible amplification of ENG recording.

MATERIAL AND METHODS

In the course of the project a newly developed IOE-System was used. It consists of two computer-controlled modules, a single-channel electrical stimulator and a six-channel ENG/EMG-recording unit. The maximum amplitude of a single biphasic stimulation pulse is 1000V per phase at 50µs duration. The duration of the first phase is adjustable from 50µs to 1ms and due to minimizing the stimulus artifact /5/ the second phase can be tuned from 0% to 100% of the first (i.e. from monophasic to symmetric biphasic).

The amplifier consists itself of tree parts. The first is a shielded part (i.e. the preamplifier) and has an amplification of 20. The second part consists of an isolation amplifier with two fixed amplifications, one for EMG- and the other for ENG-signals, 5 and 500, respectively. The third part is a multifunction I/O-PCMCIA-card for multi-channel A/D-conversation with a sampling rate of 10kSamples for duration of 100ms, a resolution of 12bit and an amplification of 1 to 100. Thus the amplification range is adjustable from 10^2 up to 10^6.

A Laptop-PC running a specially developed Microsoft-Win9x application controls the stimulator and the recording unit. The communication to the stimulation module is realized via the serial interface (RS-232) and the recording module is controlled by a multifunction-card (DAQ-Card 1200, National Instruments). The whole system (*Fig. 2*) is - due to safeness of patients - battery powered and fits into a suitcase to be easily portable and available within a few minutes in the operation room.

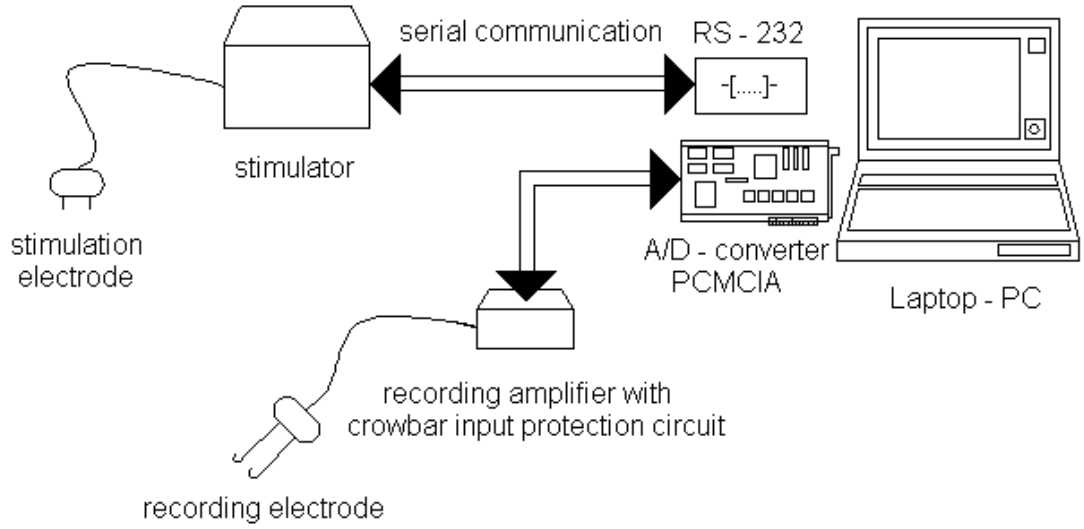
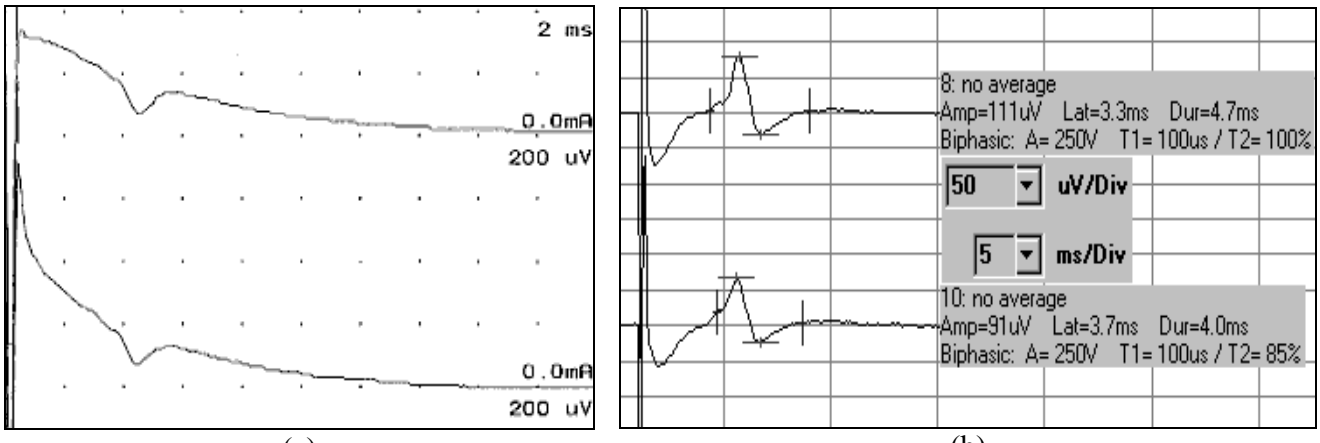


Fig. 2: ENG-recording system with computer controlled (Laptop-PC) stimulation and recording module. Communication to the modules is realized via either serial interface (RS-232) or multifunction I/O-card (PCMCIA).

The data acquisition and hardware controlling software was developed with Borland Delphi Professional. The recording screen has got 12 tracks divided by 1, 2, 3, 4 or 6 channels and a parameter panel holds the recording parameters for each track such as number of averaged recordings, 4 cursor positions as well as amplitude, pulse width and shape (mono/biphasic) of stimulation signal (*Fig. 3*). Referring to the patient all recording parameters together with recording data are editable stored in a Paradox database.

RESULTS

ENG-recordings in *Fig. 3* show the difference between the old (EMG-system with Digitimer 185) system (*Fig. 3a*) and the new IOE-system (*Fig. 3b*).



(a) (b)

Fig. 3: ENG-recordings from a) EMG-system whereas stimulation was done by a monophasic pulse with amplitude of 600V and duration of 50µs (Digitimer 185) and b) IOE-system stimulated by a biphasic pulse with amplitude of 250V and duration of 100µs per phase.

The old system requires monophasic stimulation amplitude of 500V with duration of 50µs to perform an ENG-signal of about 100µV in amplitude and the new system needs only 250V with biphasic stimulus to get the same result. This validates the theory of *Equ. 1* shown in *Fig. 1*, that double pulse width requires only half amplitude. Also the effect of the biphasic stimulation pulse for charge compensation compared to the monophasic stimulation signal can be observed in *Fig. 3*.

DISCUSSION

The new mobile IOE-system can be handled much saver and easier than the old EMG-system caused by the battery power supply and the all in one suitcase technology. The stimulation and recording module, controlled by the Laptop-PC running a Microsoft Win9x application, can easy be adjusted to the ENG recording conditions. Caused by the indirect proportionality of amplitude and pulse duration lower stimulation amplitudes (>100V) with longer pulse widths (<500µs) can be used to elicit proper ENG-signals. Asymmetric biphasic pulses are capable to minimize stimulation artifact within 1ms to detect small ENG-signals (10µV) with low latency as well.

REFERENCES

/1/ Turkof E., Millesi H., Pfundner P., Mayr N., Intraoperative Electroneurodiagnostics (Transkranial Electrical Motor Evoked Potentials) to Evaluate the Functional Status of Anterior Spinal Roots and Spinal Nerves During Plexus Surgery, Plastic and Reconstruction Surgery, 1997, 99(6): 1632-1641

/2/ Turkof E., Tambwekar S., Mansukhani K., Millesi H., Mayr N., Intraoperative Spinal Root Stimulation to Detect Most Proximal Site of Leprous Ulnar Neuritis, The Lancet, 1994, 343:1604-1605

/3/ Turkof E., Tambwekar S., Kamal S., El-Dahrawi M., Mansukhani K., Soliman H., Ciovica R., Mayr N., Leprosy Affects Facial Nerves at the Main Trunk and Neurolysis Can Possibly Avoid Transfere Procedure, Plastic and Reconstruction Surgery, 1998, 102(5):1565-1573

/4/ Kralj A. and Bajd T., *Functional Electrical Stimulation: Standing and Walking after Spinal Cord Injury*, 1989, CRC Press, Inc.

/5/ Reichel M., Bijak M., Mayr W., Lanmüller H., Rafolt D., Sauermann S., Unger E., Turkof E., Biphasic Stimulation: An Alternative Approach to Minimize the Stimulus Artifact for Diagnostic and for Control Applications, 7th Vienna International Workshop on Functional Electrical Stimulation, Vienna 2001, Proceedings, 2001, currently in press

ACKNOWLEDGEMENTS

Supported by the Austrian National Bank. Projects 6946, 7937 and 8661.

AUTHOR'S ADDRESS

Dr. Martin Reichel
Dep. of Biomedical Engineering & Physics
Währinger Gürtel 18-20, A-1090 Vienna

e-mail: m.reichel@bmtp.akh-wien.ac.at
home page: http://www.bmtp.akh-wien.ac.at/bmt/alllanguages/people/reicma1

OBJECT ORIENTED SOFTWARE COMPONENTS FOR MULTICHANNEL-STIMULATOR CONTROL

M. Russold, M. Bijak, S. Sauermann, M. Reichel, M. Rakos*, W. Mayr

Department of Biomedical Engineering and Physics, University of Vienna, Austria
* Otto Bock, Austria

SUMMARY

During the bygone years, a PC-based software package was used to set-up the stimulation parameters for an eight channel surface stimulator, mainly used for paraplegic subjects reactivation of lower extremities.
The growing complexity of the system, as well as system changes, required to design a new toolset to simplify further programming. A set of flexible object oriented software components was created. These components allow the manipulation of stimulation parameters in various ways.
Additionally, these tools can be easily adapted and extended according to the project specific needs. Usage in different development environments (e. g. MS Visual C++, MS Visual Basic etc.) is possible in a straightforward way.

STATE OF THE ART

Systems, either using surface- or implanted- stimulation electrodes are state of the art. The latter once provide better muscle selectivity and more useable channels, but require surgical intervention. Most groups, dealing with reactivation of lower extremities use six to eight-channel (surface) stimulators. Some of these groups try to achieve better results with an increased number of channels, while others try to optimise the stimulation parameters. The Viennese group takes the second chance to gain better results.

MATERIAL AND METHODES

The used system (Figure 1) consists of two modular designed four channel stimulators, a Bus Manager for communication-control, a I2C-Interface to communicate directly with the stimulators, using their internal I2C bus and finally a PC-based software package to ease the set-up of stimulation parameters. /1/

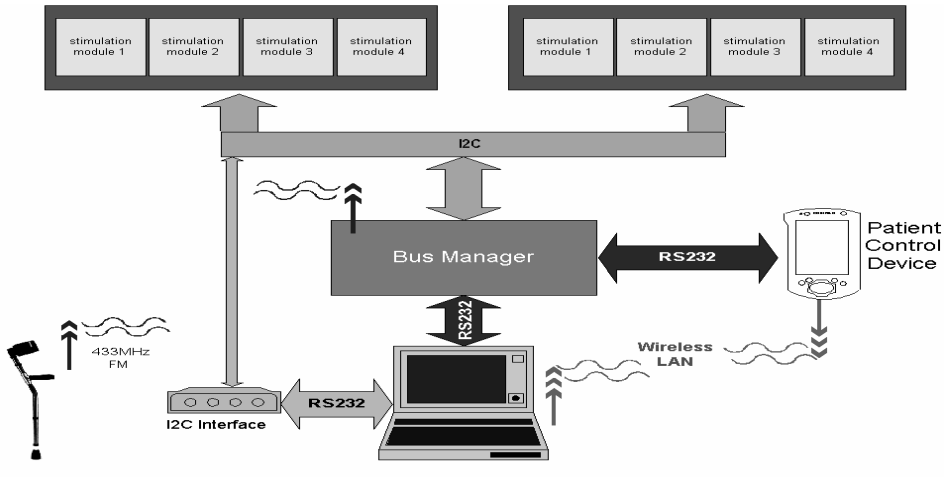


Figure 1: Complete system

The new toolset may be used to simplify the design of the PC-based software package. As most people are familiar with the MS Windows' interface, a similar - stimulation related - one was designed, using Delphi (Borland, Scotts Valley, CA USA) as programming environment.

After a detailed functional analysis eight components, for the following tasks, were realized: burst envelope setup, data handling and synchronization, user management and communication with the stimulators.
Each of these components contains a set of properties, methodes and events to fulfil their required tasks.

Figure 2 shows the interaction and relations between these components in general. The components communicate with their properties and methods and allow a subtle control of the main implementation.

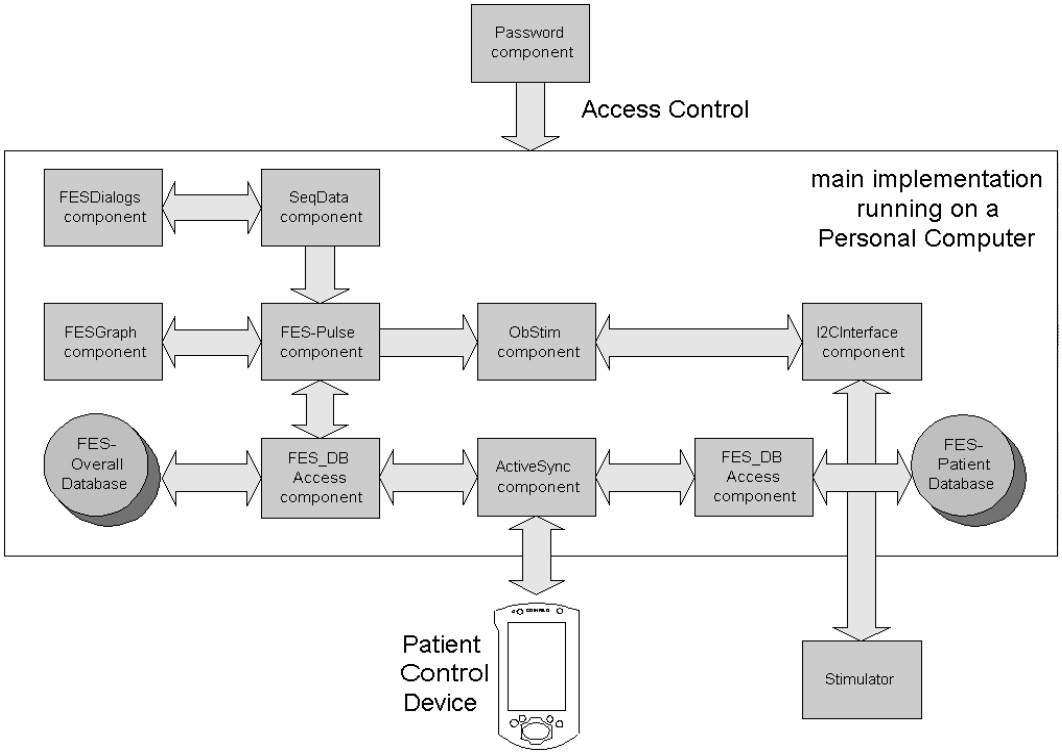


Figure 2: Components' relationship

The Security-component supports a simple, hierarchically user management. Another task of this component is to control user access to the PC based software package or main implementation.

Once the implementation is successfully launched, the Dialogs-component – together with the SeqData-component – loads some basic dataset information from the database. This information is used to set-up the Pulse-component and to fetch the complete selected dataset from the database.

Afterwards the Graph-component can be initialised. The Graph-component allows a mouse-controlled manipulation of the burst envelope (points 1-4, Figure 3, left side) in an easy and intuitive way. The points can be manipulated using drag'n drop, observing the specific ancillary conditions. Special features, like showing of time and amplitude, as well as safety borders during drag'n drop, can be activated according to the implementation needs. These features provide additional information and may increase security.

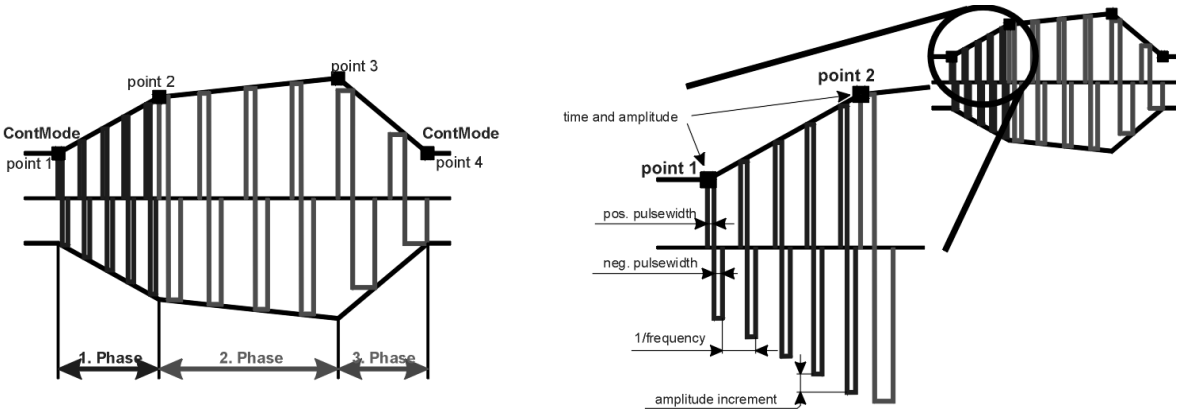


Figure 3: Burst envelope and burst details

All further values, like the positive- and negative pulse-width and the frequency (see Figure 3, right side) must be set outside the Graph-component. All values may be set independently for all phases. The remaining amplitude increment will be calculated, using the time and amplitude values of two neighbouring points.

Communication with the stimulators happens by means of the ObStim-component. This component also transforms all parameters into the correct format for the stimulators and allows to control the BusManager, too. Communication between PC and the Patient Control Device happens via RS232, USB or a wireless LAN, using the MS ActiveSync-software package.

For storage of stimulation parameters and measurement results, as well as for later on data analysis, a relational database structure is implemented. Several databases are used with this implementation. One database – the *general* database - stores all datasets of all patients, while another database – the *patient* database - stores only the synchronized datasets from one patient. Synchronized means that these datasets are available for stimulation with the Patient Control Device. Obviously a *patient* database has to be available for every patient.

Communication with these databases and all data handling is established by means of the Database-component. To access the databases the PC's **O**pen **D**ata**B**ase **C**onnectivity (ODBC)-service is used. This service supports connection to various databases. Additionaly more than one database-connection may be used at a time; necessary during synchronization, where two databases are manipulated simultaneously.

The Patient Control Device, based on a standard handheld computer running MS Windows CE is used to increase patient's mobility and can store an adjustable selection of datasets. To use these datasets later on, this handheld computer is equipped with special software. Communication with this device can be obtained with the ActiveSync-component.
This component allows to transfer the *patient* database between the PC and the Patient Control Device. Additional synchronization between the *patient*- and the *general* database is supported.

All these components were combined in a sample implementation (Figure 4). Necessary additional forms are called at runtime using the Dialogs-component. These forms may be used for: selection of datasets prior to loading, setting dataset-names prior to storage, dataset selection for use with the Patient Control Device.

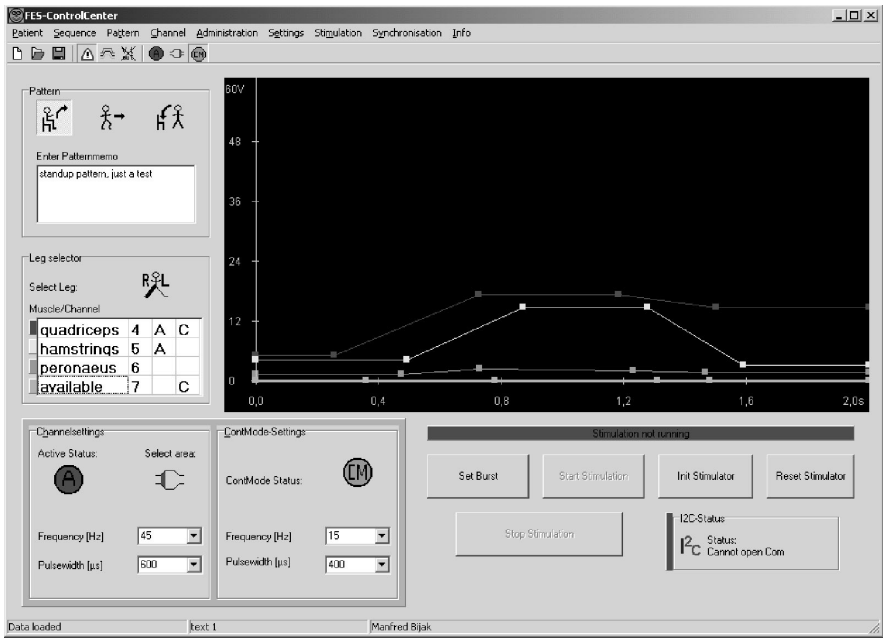


Figure 4: Implementation at runtime

RESULTS

The sample implementation (Figure 4) demonstrated their proper usage. After an extensive testing phase, a first patient approach was done at the Department of Physical Medicine and Rehabilitation[1]. Final modifications are considered and will be finished by the time of this workshop.
The sample implemenation, together with some further implementations for testing purposes, showed a notedly decreased development time. The components allowed reduction of programming effort for the main implementation. This allows rapid prototyping and a shorter development time, resulting in earlier patient test, for the future.

REFERENCES

/1/ Personal Computer supported eight channel surface stimulator for paraplegic walking: first results
M. Bijak, C. Hofer, H. Lanmüller, W. Mayr, S. Sauermann, E. Unger, H. Kern
Artificial Organs, 23(5): 424-427, © 1999 International Society for Artificial Organs

ACKNOWLEDGEMENTS

This project is supportd by Otto Bock, Austria.

AUTHOR'S ADDRESS

Michael RUSSOLD
Department of Biomedical Engineering and Physics, AKH 4/L
Währinger Gürtel 18-20
A-1090 Vienna, Austria
email: friedrich@russold.at

[1] Wilhelminenspital Vienna

WHEN CAN AUTOCORRELATION BE USED FOR THE CLASSIFICATION OF ENG SIGNALS?

L.Zhou, N.de.N.Donaldson

Implanted Devices Group
University College London, United Kingdom

SUMMARY

The use of naturally-occurring sensory nerve signals, recorded from sacral nerves and used for the control of neuroprostheses to inhibit hyperreflexic bladder contractions, has been the aim of recent studies /1/. One of the problems is that, the sacral nerves, which carry bladder afferent signals, also carry signals that originate in the cutaneous receptors, or the rectum. Autocorrelation is a possible method to identify the signal source. However, the ENG signals, recorded from cuffs on the sacral nerve roots, are very small compared to the noise. How does the noise affect the signal's autocorrelation function? At what signal-to-noise ratio (SNR) can we distinguish the ENG signal when it is buried in noise? Our simulation shows that, in the ideal situation, where noise is white (wide bandwidth), its ACF is large for zero lag and then random for other lags. These random values are added to the ACF of the signal. In a real situation (narrow bandwidth), as the noise increases, the autocorrelation function of the signal plus noise will increasingly resemble noise alone. So, both the SNR and the filter passband determine the detection error rate for ENG signals mixed with wide-band noise from the source resistance. We simulated classification by autocorrelation in MATLAB, using synthesized ENG signals and white noise. We found that, with the best passband, the error rate for detecting bladder signals falls below 50% when SNR is –8dB. The best passband is significantly wider than the signal spectrum. This result will allow us to assess the suitability of autocorrelation for classification when we know more about the SNR of actual bladder signals. It also suggests that, when recording signals that will be autocorrelated, the amplifier passband should be wider than the signal spectrum.

STATE OF THE ART

Brindley pioneered the use of the sacral anterior root stimulator to empty the bladder after complete spinal cord lesions. However, his method also uses dorsal root rhizotomy to prevent hyperreflexic incontinence and dyssynergia (inhibited voiding). Kirkham *et al.* /2/ have shown that stimulation of the sacral posterior roots without rhizotomy can reduce the hyperreflexia. Craggs *et al.* /3/ and Sinkjaer *et al.* /4/ have suggested that posterior root stimulation might be activated automatically from the natural nerve signal (*conditional neuromodulation*). Jezernik /5/ showed that neural signals, picked up by cuff electrodes on the extradural sacral roots in pigs, increases slightly with bladder pressure, from 1.2μV to 1.3μV ($SNR=20*\log_{10}(0.1/1.2)=-22dB$). He suggested that this bladder signal might be distinguished from the cutaneous and rectal signals by autocorrelation (ACF), exploiting the fact that their tripole action potentials are different because they are carried on fibres of the different diameters. However, although this would be true if the ENG signals had high SNR (and distinct ranges of fibre diameter), the SNR of actual ENG signals, recorded from cuffs on the sacral nerve roots, is poor.

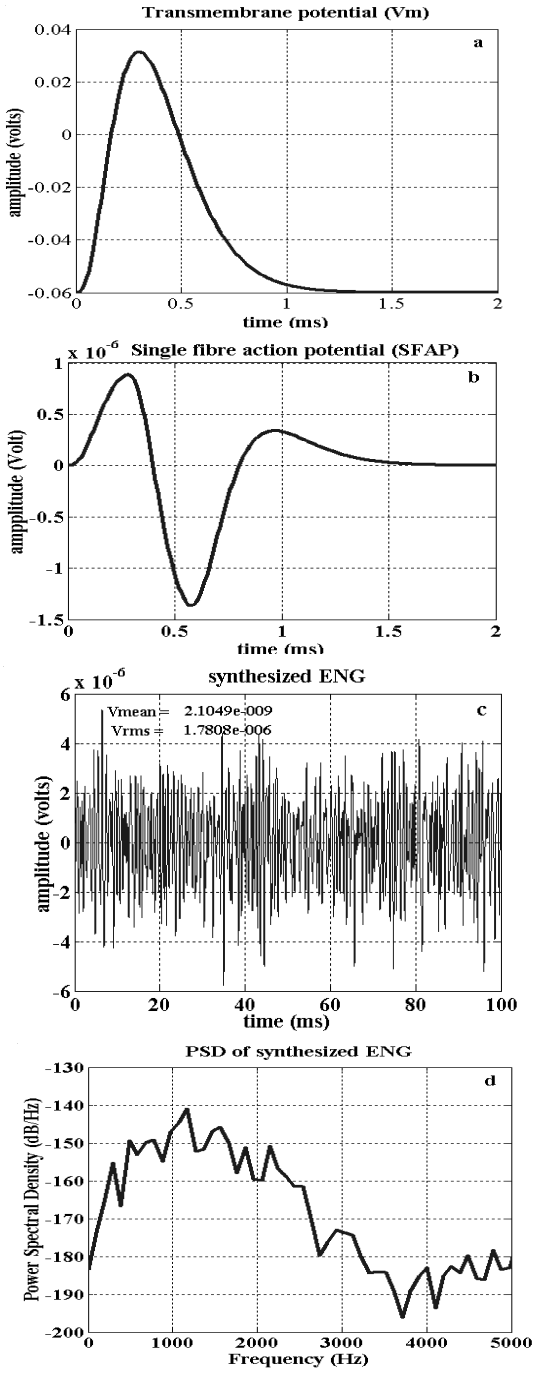
This paper examines the limitation of this classification method for poor SNR signals. It shows that to minimise the effect of noise on the signals autocorrelation function, the amplifier passband should be optimised.

MATERIAL AND METHODS

The investigation was conducted by simulation in MATLAB using artificially-generated bladder ENG and noise. The autocorrelation functions (ACF) of pure ENG, and ENG with added white noise, which had been filtered to one of three passbands, were compared. The ACFs are classified as bladder signals according to one criterion. The detection rate for each combination of SNR and passband was calculated from 50 ACFs.

ENG generation The artificial transmembrane potential (V_m) was generated using the formula $V_m=At^3e^{-Bt}-C$, and compared with standard V_m to adjust A,B,C (figure 1, a). It was used with bladder-associated human nerve parameters from Schalow /6/ (fibre diameter = 10.5µm, myelin thickness = 1.55µm, conduction velocity = 37m/s, average diameter of extradural sacral root S3 = 1.75mm) and tripolar electrode parameters (cuff diameter = 2mm, cuff length = 20mm, electrode spacing = 10mm) to calculate a single fibre action potential (SFAP) /7/ (figure 1, b). After that, many SFAPs, occuring at random times but uniformly distributed over 100ms, were added together. The number of SFAPs (680) was estimated from the number of fibres in the nerve root and by the observed firing frequency /6/. The sample rate was 100kHz. The sums had nearly zero means, 1.8µVrms amplitude, and half power frequencies of the spectrum at 0.3 to 2kHz (figure 1 c, d).

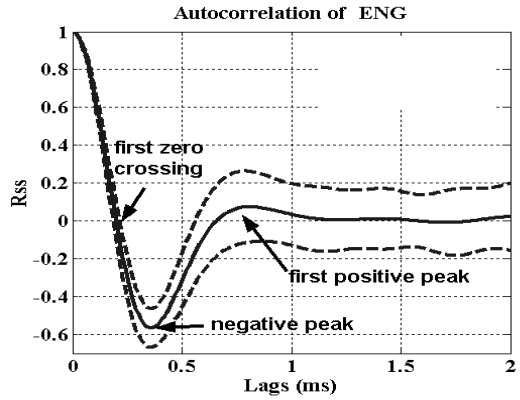
Figure 1

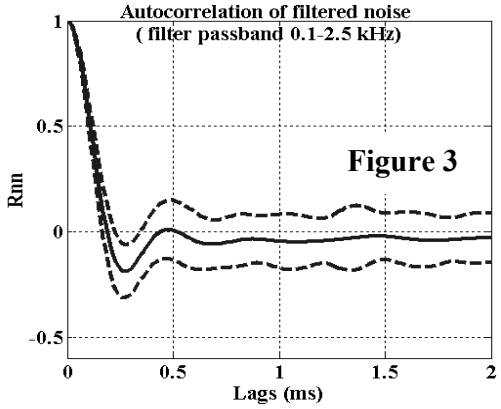


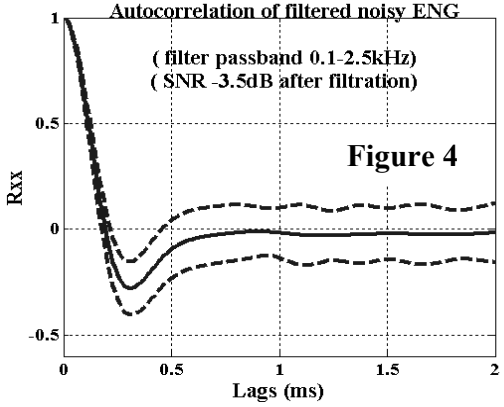
ACF of noise-free bladder ENG For pure bladder ENG, figure 2 shows the average ACF and curves for 3 standard deviations. Its shape is defined by the first zero crossing, the first negative peak and the time of the first positive peak. The time of the first negative peak showed least variance, so it was used as the descriptor of ACF shape. A range of 0.35 to 0.4ms was used to classify bladder signals.

Classification The ENG and noise were mixed at different SNR, filtered (Butterworth, 0.1-2, 0.1-5 or 0.1-10 kHz, 80dB/decade cutoffs) and autocorrelated. The time of the first negative peak (lag)

was compared to the detection range. If it fells inside the range, it was classified as bladder ENG. For each SNR level and filter passband combination, there were 50 runs in order to calculate error rate (Figure 4).







RESULTS

Effect of filters on SNR & classification rate.

Classification rate >50% shown in grey.

	Without filter	With filter		
		0.1 - 10kHz	0.1 - 5kHz	0.1 - 2.5 kHz
SNR level (dB)	0	7	10	12.5
Detection rate (%)	**88**	**96**	**98**	**100**
SNR level (dB)	-5	1.8	4.5	8
Detection rate (%)	**82**	**94**	**94**	**100**
SNR level (dB)	-10	-3.5	0	2.5
Detection rate (%)	**64**	**60**	**94**	**86**
SNR level (dB)	-13	-6		0
Detection rate (%)	**none**	**50**	**84**	**42**
SNR level (dB)	-16	-10	-6.5	-3
Detection rate (%)	**none**	**42**	**60**	**0**
SNR level (dB)	-18	-12	-9	-6
Detection rate (%)	**none**	**20**	**40**	**0**

DISCUSSION

The table gives the correct classification rate at different SNR levels. The SNRs in 3rd, 4th and 5th columns are the signal to noise ratios after the noisy ENGs are filtered. The SNRs showed in 2nd column are that of the corresponding unfiltered signal.

For a bladder ENG whose energy falls between 0.3-2 kHz, the SNR should be higher than –6.5 dB in order to achieve 60% correct rate or better. Although narrower band filtering (0.1-2.5 kHz)

significantly improves SNR, it classifies the nerve ACF less well. The middle pass band (0.1-5 kHz) is optimal.

Therefore, the use of this classification method can be considered under three circumstances. First, with good quality signals (SNR > -10 dB before filtering), it is possibly an effective way to identify signal origins, as long as one knows the signals are carried by nerve fibres of distinct diameter. Second, when SNR are less good (-16 to -10 dB), it still works for those applications requiring only 50% or lower classification rate. However, the optimal band is significantly wider than the signal spectrum. If the nerve characteristics are uncertain, it would be safer to choose a wider band (also faster sampling rate). Finally, it will not work if SNR is poor (< -16dB).

At present, the amplitude of the recorded bladder ENGs from animal experiments is small /5/, the SNR levels are lower than the limit set above. There are also disputes about fibre types on which bladder sensory information is carried. In cat, de Groat /8/ asserts that they are Aδ (2-10μm) and C (<1μm) fibre afferents /9/. For human, Schalow /6/ claims that they are larger fibres ("ST" fibre, >10μm). Until we have more human data on propagation velocity for bladder, rectal and cutaneous afferents, and realistic SNRs, the feasibility of autocorrelation for classification in neuroprostheses is uncertain.

REFERENCES

/1/ Sinkjaer T., Rijkhoff N. et al., Electrographic(ENG) recordings from intradural dorsal sacral nerve roots in a patient with a suprasacral spinal cord injury. 5th IFESS Ann. Meeting, Proc. 2000.
/2/ Kirkham APS., Craggs MD. et al., Acute and chronic use of a sacral posterior and anterior nerve root stimulator to increase bladder capacity in spinal cord injury. 6th IFESS Ann. Meeting, Proc.2001, 172-174.
/3/ Sheriff MK, Shah PJ, Fowler C, Mundy AR, Craggs MD., Neuromodulation of detrusor hyper-reflexia by functional magnetic stimulation of the sacral roots. Br J Urol. 1996 Jul;78(1):39-46.
/4/ Jezernik S, Grill WM, Sinkjaer T., Detection and inhibition of hyperreflexia-like bladder contractions in the cat by sacral nerve root recording and electrical stimulation. Neurourol Urodyn. 2001; 20(2):215-30.
/5/ Jezernik S., Towards a novel implantable closed-loop neuroprosthetic device to control the overactive bladder. PhD Thesis University of Aalborg, 1999, Denmark.
/6/ Schalow G., Conduction velocities and nerve fibre diameters of touch, pain, urinary bladder and anal canal afferents and α and γ-motoneurons in human dorsal sacral roots. Electromyogr. Clin. Neurophysiol., 1991, 31, 265-296.
/7/ Struijk J., The extracellular potential of a myelinated nerve fiber in an unbounded medium and in nerve cuff models. Biophysical Journal, Vol. 72, June 1997, 2457-2469
/8/ de Groat W., Central neural control of the lower urinary tract. Neurobiology of Incontinence. 1990, Wiley, Chichester (Ciba Foundation Symposium 151), p27-56.
/9/ Mathers L., The peripheral nervous system. 1985, Addison-Wesley Publishing Company. P70.

ACKNOWLEDGMENTS

This study is supported by The Engineering and Physical Sciences Research Council (EPSRC), United Kingdom.

AUTHOR'S ADDRESS:

Lixia zhou, Implanted Devices Group, University College London, 11-20 Capper Street, London WC1E 6JA. U.K. e-mail: lzhou@medphys.ucl.ac.uk

INDEX OF AUTHORS

INDEX OF AUTHORS

INDEX OF AUTHORS

INDEX OF AUTHORS

INDEX OF AUTHORS