Rehabilitation Engineering Laboratory

Toronto Rehab
Advancing Rehabilitation
Enhancing Quality of Life

VELUT ARBOR ÆVO

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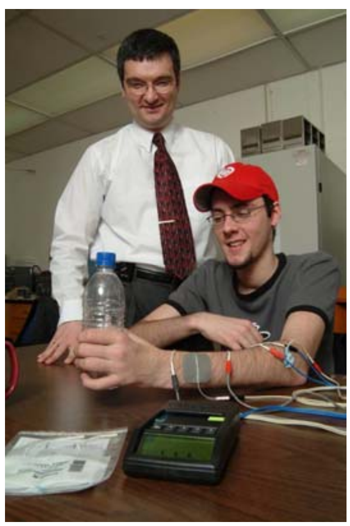

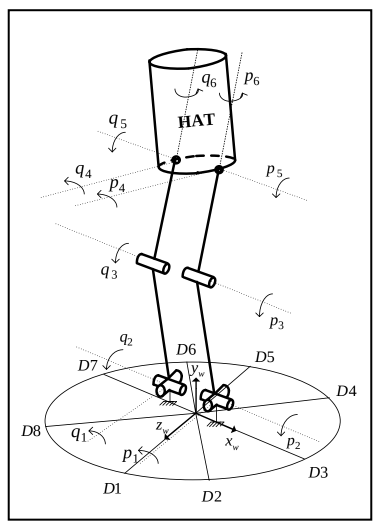

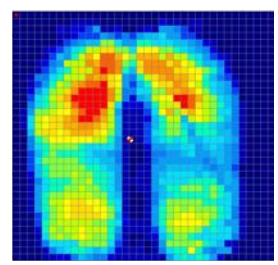

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Design: Cheryl Lynch, Rehabilitation Engineering Laboratory – 2006

About the Rehabilitation Engineering Lab

History

The Rehabilitation Engineering Laboratory was established in 2001 at the Lyndhurst Centre of Toronto Rehab. In 2006, the laboratory underwent a major renovation, doubling the amount of space available for personnel and experiments.

What We Do

We develop advanced technologies for spinal cord injury (SCI) and stroke rehabilitation. These include assessment tools for determining an individual's level of function and rehabilitation techniques for restoring walking and grasping ability. We also design neuroprosthesis systems to assist individuals with tasks such as walking, grasping, and balance during standing and sitting.

Most of our work is based on functional electrical stimulation (FES), which uses electricity to cause muscles to contract. FES can be used to provide movement to paralyzed muscles or to re-train weak muscles.

Accomplishments – 2001 to 2006

- Developed grasping and walking neuroprostheses for stroke and SCI subjects
- Developed FES therapies for neuro-rehabilitation in stroke and SCI
- Provided FES therapy to more than 50 individuals with SCI or stroke
- Published more than 30 peer-reviewed journal papers
- Obtained more than $2,000,000 in funding for SCI and stroke research
- Trained 8 researchers and 12 graduate students in SCI and stroke research, who obtained an additional $1,000,000 in personnel funding

Want to Get Involved?

We are always looking for participants for our studies, volunteers to help us with the experiments, students and research collaborators. If you would like to join us, please feel free contact us via e-mail at bezruk.zina@ torontorehab.on.ca.

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Research at the Rehabilitation Engineering Lab

Neuroprosthesis for Reaching and Grasping

Our reaching and grasping neuroprosthesis is designed for individuals who cannot reach and/or grasp voluntarily. These individuals are able to use the system to pick up and manipulate objects, and can significantly improve their independence in activities of daily living. People who have SCI at C5-C7 have used this system as a prosthetic device, and hemiplegic and SCI people have used it as a rehabilitation tool to assist in retraining reaching and grasping.

Neuroprosthesis for Walking

The purpose of the neuroprosthesis for walking program is to demonstrate the long-term benefits of FES therapy on walking function in patients with incomplete SCI and stroke. Our pilot study showed a significant improvement in walking speed and/or a reduction in the use of assistive devices for walking after using the neuroprosthesis. We are currently conducting a randomized treatment-vs-control study to verify that these benefits truly resulted from FES-assisted walking therapy.

Neuroprosthesis for Sitting

Trunk instability is a major problem for many people with SCI that affects their independence and ability to perform activities of daily living. The long-term objective of this project is to produce a new device that will improve sitting stability by stimulating paralyzed trunk muscles using FES. This sitting neuroprosthesis will improve the ability of people with SCI to perform such tasks as reaching and wheeling. We are currently studying the mechanisms of balance in the trunk and the consequences of muscle paralysis on these mechanisms. This analysis will form the basis for developing the FES system for balance during sitting.

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Neuroprosthesis for Standing

The neuroprosthesis for standing and balancing is a device that will allow some neurologic patients to stand up, perform stable “hands-free” standing, and sit down again. At least two applications of this technology are envisioned: 1) this device will be used as an independent system to allow complete SCI patients to stand; and 2) to retrain standing function and balance control in incomplete SCI, stroke and elderly patients through active, repetitive, balance training sessions. Besides the obvious functional benefits, this neuroprosthesis would also help maintain bone density and prevent pressure sores by allowing people to stand for extended periods of time.

Human-Machine Interfaces

Understanding the relationship between an assistive device and its user is a fundamental step towards designing better systems. The human-machine interface project focuses on developing new communication strategies and methodologies to allow users to have more natural control over an assistive device. One aspect of this work is our research into brain-computer interfacing, which investigates the relationship between intended arm movement and electroencephalogram (EEG) signals from the motor cortex of the brain.

Equipment

The Rehabilitation Engineering Laboratory has a variety of research equipment including:

- Compex II stimulators
- Body weight support treadmill
- Force plates
- Polhemus motion capture system
- Optotrack dual camera motion capture systems
- Erigo tilt table with motorized leg movement
- Electromagnetically shielded room for EMG and EEG measurements

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Our People

Researchers

- Dr. Milos Popovic, Head of the Laboratory
- Dr. Kei Masani (exercise physiology), Senior Scientist
- Dr. Judith Hunter (neuroscience), Postdoctoral Fellow
- Dr. Noritaka Kawashima (exercise physiology), Postdoctoral Fellow
- Dr. Masae Miyatani (exercise physiology), Postdoctoral Fellow
- Dr. Alan Morris (aerospace engineering), Postdoctoral Fellow
- Dr. Mary Nagai (pathology, toxicology, orthopaedic surgeon), Postdoctoral Fellow
- Marlene Adams, Occupational Therapist
- Veronica Takes, Occupational Therapist

Graduate Students

- Cesar Marquez Chin, PhD student
- Cheryl Lynch, PhD student
- Albert Vette, PhD student
- José Zariffa, PhD student
- Lorne Chi, MHSc student
- Robert Nguyen, MSc student
- Vivian Sin, MASc student
- John Tan, MASc student
- Lev Vaisman, MASc student

Support Staff

- Zina Bezruk, Administrative Assistant
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- Dr. Vera Zivanovic, Research Coordinator

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Recent Research Posters

Neuroprosthesis for Grasping

- “Functional electrical therapy: Retraining reaching and grasping functions in severe hemiplegic patients”
- “Restoration of grasping in patients with quadriplegia”
- “Upper extremity functional electrical stimulation”
- “REL hand function test”

Neuroprosthesis for Walking

- “Necessity of successive sensory feedback to update the internal model for walking”
- “Oxygen uptake during FES treadmill walking: Case study”
- “Neuroprostheses for walking”
- “FES for treadmill training”
- “Recovery of motor function following incomplete spinal cord injury”

Neuroprosthesis for Standing

- “Influence of the musculoskeletal system on the control mechanism of quiet standing”
- “Derivative gain in the neural controller accounts for the temporal relationship between body sway and muscle activity”
- “Implementation of a PD controller for improving human balance”
- “Balance control during standing: Stabilizing PD controller generates COM dynamics as observed in healthy individuals”
- “Dynamic modeling and analysis of arm-free paraplegic standing”
- “Optimal combination of active degrees of freedom to facilitate arm-free paraplegic standing”
- “Balance during quiet standing: The role of toes in the fine-tuning control of balance”
- “Voluntary knee response in healthy subjects”
- “Evaluating existing FES controllers”

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Neuroprosthesis for Sitting

- "Muscle synergies for pressure management during prolonged sitting"
- "Development of an automated perturbation system"

Human-Machine Interfaces

- "What is a brain-computer interface?"
- "Identifying movements from cortical signals"
- "Automatic classification of ECoG signals"
- "Source localization in peripheral nerves: Part I"
- "Source localization in peripheral nerves: Part II"
- "Predicting finger force from surface EMG"

Other Projects

- "Effect of random modulation of FES parameters on muscle fatigue"
- "Effects of isometric FES and dynamic FES on cardio-vascular parameters on an active tilt-table stepper"
- "Human responses to vibration therapy"
- "Arterial stiffness in persons with spinal cord injury: A pilot study"
- "Novel home-based care equipment for preventing the secondary disorders in paraplegic persons"
- "Simulation of elbow muscle force from EMG"

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FUNCTIONAL ELECTRICAL THERAPY

M.R. Popovic, A.T. Thrasher and V. Zivanovic

Toronto Rehabilitation Institute and University of Toronto

Retraining Reaching and Grasping Functions in Severe Hemiplegic Patients

Materials and Methods

PARTICIPANTS:

- **Stroke patients with severe unilateral upper extremity paralysis.**
- Chedoke McMaster Stages of Motor Recovery scores 1 or 2 measured at least three weeks after onset of stroke.
- Participants unable to voluntarily:
  - flex, extend, abduct, adduct, or rotate the shoulder;
  - flex or extend the elbow;
  - pronate or supinate the forearm;
  - flex, extend, abduct or adduct the wrist;
  - move any fingers;
- **Participants were acute (0 – 6 weeks post stroke) or long-term (12 months after stroke).**

  Group A - ***control group*** had standard physiotherapy and occupational therapy alone.

  Group B - ***treatment group*** had functional electrical stimulation (FES) therapy administered in addition to conventional physiotherapy and occupational therapy.

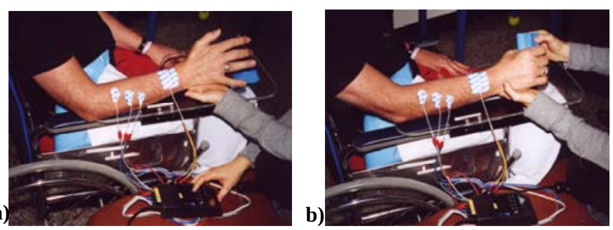

Fig. 1: a) Finger extension performed with the help of neuroprosthesis b) Voluntary finger flexion

FUNCTIONAL TESTS:

- Functional Independence Measure (FIM)
- Barthel Index (BI)
- Chedoke McMaster Stages of Motor Recovery (CMSMR)
- Fugl-Meyer Assessment (FMA)
- REL Hand Function Test (REL) [1]

FES THERAPY PROTOCOL:

- Participant was asked to execute a task with the impaired arm (e.g. reaching and grasping a pen) unassisted.
- **The components of the task that the participant was unable to carry out him/herself were assisted by the neuroprosthesis.**
- Compex Motion neuroprosthesis was used for FES therapy [1].
- In the early stages of the treatment, the arm/hand tasks were performed by the neuroprosthesis alone.
- **As the patient improved, the neuroprosthesis assistance was reduced to the necessary minimum and eventually was removed from the treatment protocol.**
- **The patient had three treatment sessions per week, 45 minutes per session, up to 16 weeks.**

CONTROL GROUP PROTOCOL:

- **Standard physiotherapy and occupational therapy**
- Control group sessions were equal in length and intensity to the FES therapy sessions.

STATISTICAL ANALYSIS:

- Wilcoxon rank-sum test

Results

SUMMARY:

- **The study was conducted with 24 stroke patients** (8 female, 16 male; average age 56):
  - **15 participated in FES therapy**
  - **9 were controls**
- The difference between mean scores obtained on discharge and admission for both Groups A and B are shown in Fig 2.
- **Significant differences were found between Groups A and B in terms of change of the FMA and the torque, force and eccentric load components of the REL Hand Function Test.**
- **The statistical analysis suggests that the FES therapy gives rise to greater improvement in arm and hand functions, compared to traditional physiotherapy and occupational therapy alone.**

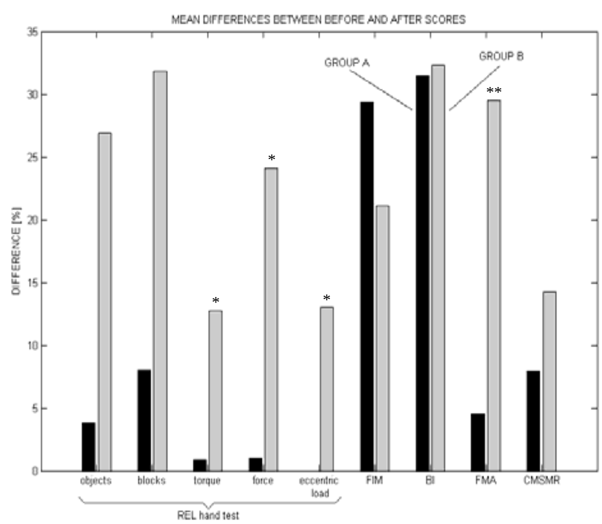

Fig. 2: Differences between the after and before mean scores for: 1-5) REL Test; 6) FIM; 7) BI; 8) FMA; and 9) CMSMR tests. Statistical significance of the difference is presented as follows: * $p < 0.05$, and ** $p < 0.01$.

References:

[1] Popovic M.R., Thrasher T.A., Zivanovic V., Takaki J., and Hajek V., *Neuroprosthesis for restoring reaching and grasping functions in severe hemiplegic patients.* Nauromodulation, 2005. 8(1): p. 60-74.

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Partners:

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Restoration of Grasping in Patients with Quadriplegia

M. Adams, V. Takes, A. Thrasher, A. Bulsen, V. Zivanovic, M. Tonack and M.R. Popovic

REL REHABILITATION ENGINEERING LABORATORY

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Introduction

Purpose of the study: To examine the carryover effect of using a neuroprosthesis for grasping.

Research question: Does the application of daily functional electrical stimulation (FES) training facilitate the restoration of neurological function in the wrist or fingers, and subsequently does it increase a person's level of independence in daily tasks?

Novelty: The study is a departure from the current approach whereby the neuroprosthesis is used as a permanent assistive device. The idea for this study came from previous experience and anecdotal reports in the literature describing improvement in active movements following a training period with the FES.

Methods

Subjects: 22 individuals with complete and incomplete acute SCI (less than one year after SCI) participated. All subjects had diagnoses of quadriplegia between the levels of C4 to C7.

- ***Incomplete SCI consumers:*** Patients with either no active movement in their wrists or fingers up to those with less than half the normal active range were chosen for the study. It was reasoned that patients with more muscle activity than this would be more likely to improve without FES.
- ***Complete SCI consumers:*** Patients who had some level of response to FES electrode stimulation. Patients that did not respond with movement of the wrist or fingers when the FES was applied were not included in either the treatment or control groups.

Participants were randomized

Therapy: ***Controls*** received regular OT and PT which may have included electrical stimulation for muscle strengthening without the functional component. ***Treatment group*** received individualized FES programs in which the neuroprosthesis was helping the patients complete only those movements they were unable to produce on their own. ***Treatment*** was five times per week, for up to three months.

Assessments: were performed before and after the treatment period. ***Hand function*** was evaluated using the REL Hand Function Test. ***ADL functions*** assessed using FIM and SCIM. All outcome measures were evaluated without FES.

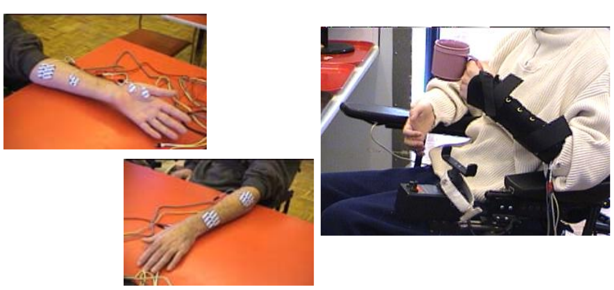

Figure 1: Placement of electrodes and neuroprosthesis in use

Results

- Treatment group demonstrated greater improvements in independence and hand function compared to controls. However the differences were not statistically significant.
- Participants who had motor complete injuries (ASIA A or B) received very little carryover from the neuroprosthesis. However, some patients with complete injuries were observed to use their hands in new activities, such as eating Smarties or playing ping pong, following sessions where they used the FES to complete these tasks.
- Participants' feedback has been positive and included comments on improved function, ability to perform ADL's, self-satisfaction and long-term commitment.

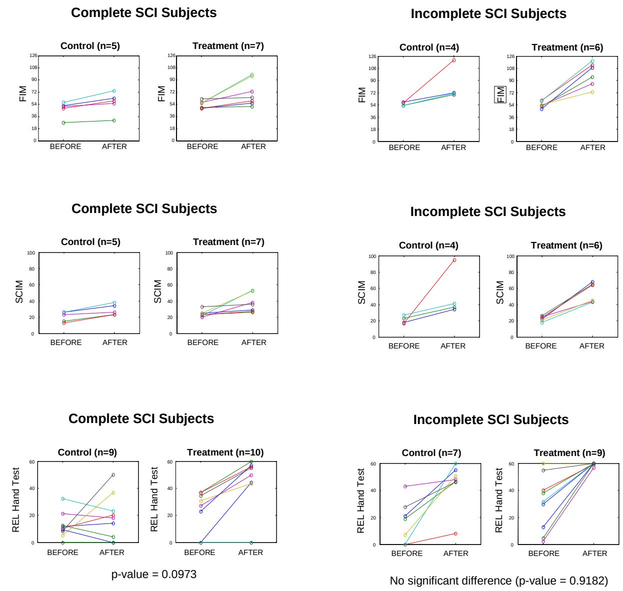

Figure 2: Raw pre and post treatment data

Conclusions

- Repetitive daily FES treatments facilitate the carryover in hand function in SCI individuals
- The most dramatic results were observed in participants with incomplete injuries, as expected
- Functional improvements were also observed in participants with complete injuries, which was not expected
- Due to small sample size the results of this study do not show statistically significant difference between treatment and control groups yet. Larger sample size is needed to test the hypothesis that FES therapy is beneficial in restoring hand function in individuals with SCI

References

- Popovic et al. *IEEE Trans Eng Med Biol Mag.* 20:82-93, 2001.
- Popovic et al. *Artificial Organs* 26(3):271-275, 2002.
- Catz et al. *Spinal Cord.* 2001 Feb;39(2):97-100.

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Upper Extremity Functional Electrical Stimulation: Case Study of Neuromuscular Restorative Therapy

Miller RC[1], Popovic MR[1,2,3], McIlroy W[1,3], Verrier MC[1,3]

1 Graduate Department of Rehabilitation Science, 2 Institute of Biomaterials and Biomedical Engineering, University of Toronto, Canada, 3 Toronto Rehabilitation Institute, Toronto, Canada

Introduction

Spinal cord injury (SCI) results in the impairment of motor and sensory function. Hand function is critical to independence[1]; however strategies to facilitate the recovery of upper extremity function are the focus of current research. The purpose of this study was to exploreof the use of functional electrical stimulation (FES) as a central element an intervention strategy targeted at upper extremity function for individuals with SCI.

1.1. Functional Electrical Stimulation

FES is used primarily as a neuroprosthetic. However there are reports of restoration of upper extremity function following FES[2,3,4] promoting the development of NRT.

1.2. Neuromuscular Restorative Therapy

NRT is an integrative intervention strategy using FES to facilitate muscle strengthening and functional training while attempting specific movement acquisition.

Principles of motor learning are combined with purposeful, meaningful, goal oriented movements. NRT, in addition FES motor training, has active (no FES) component (movement training and stretching). NRT is customized to the participants' functional and motor abilities as well as rate of recovery.

Methods

Stimulation: A Compex® Motion stimulator (Medicompex SA, Ecublens, Switzerland)[5]

FES parameters: blocked asymmetric biphasic pulse with a width of 300μs, frequency of 40Hz, and amplitude of 8-50mA

Assessements: ASIA[6] scores, SCIM[7] scores, and a kinematic assessment of the Reach, Grasp, Transport, and Release (RGTR) task.

2.1. RGTR Task

The RGTR consists of grasping a small wooden block in one of two targets. Five trials were performed each trial consisted of ten movements five to each target (in randomized order). Target A was perpendicular to the participant and Target B was at 45o to the Target A. Force Sensitive Resistors (FSR's) were used to measure the force production of the thumb, index and middle digits during the RGTR.

Analysis: A Labivew program was used determine successful object contact (SOC), by examining the FSR signal. SOC's were defined as the participants ability to trigger the FSR (cross a threshold of < 9Volts) and maintain sufficient contact with the FSR so that the threshold is only crossed at the begining and end of the grasp portion of the RGTR. Also both FSR had to be triggered within 150 msec of each other. This requires the ability of an individual to co-activate the finger and thumb flexors while the wrist is in either an extended or neutral position for the duration of the grasp and transport portion of the RGTR task.

ADD DIGITAL PIC of Hand Grasping in RGTR

Case Study 1

Participant 1	NRT Program Goals		FES Parameters
Right Dominant	Strengthen	Wrist Flexors Wrist Extensors Digit 3-5 Flexors	Motor Stimulation (mA)
24 Years			
Injury Onset 1994	Acquire	Digit 1-3 Flexors Digit 1-5 Extensors	ED (15.7 ± 2.3) MN: Thumb: (11.1 ± 1.2) FDS (16.5 ± 2.4) MN: Digit Flexors (12.2 ± 1.1)
C6 Sensory Incomplete			

Table 1: P1, Right arm recieved NRT

Case Study 2

Participant 2	NRT Program Goals		FES Parameters
Right Dominant	Strengthen	Wrist Extensors	Motor Stimulation (mA)
31 Years Old	Acquire	Wrist Flexors Digit 1-3 Flexors Digit 1-5 Extensors	ED (25.7 ± 1.6) MN: Thumb: (11.8 ± 0.6) FDS (15.5 ± 1.1) MN: Digit Flexors (20.2 ± 0.6)
Injury Onset 1996			
C4-5 Motor Incomplete			

Table 2: P2, Left arm recieved NRT

ED: Extensor digitorum, MN: Median Nerve, FDS: Flexor digitorum superficialis, FDP: Flexor digitorum profundus.

ASIA Scores and SCIM Scores Did Not Change

Participant 1

ASIA score remained "B", SCIM score 68/100

Participant 2

ASIA score remained "C", SCIM score 42/100

1. Improvement in Raw FSR traces

Participant 1

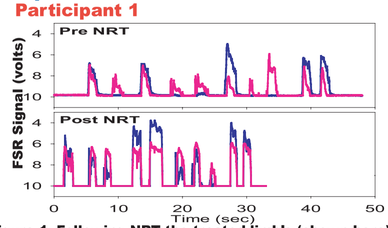

Figure 1: Following NRT the treated limb's (shown here) FSR trace was more syncronized and had more accurate FSR triggering

Participant 2

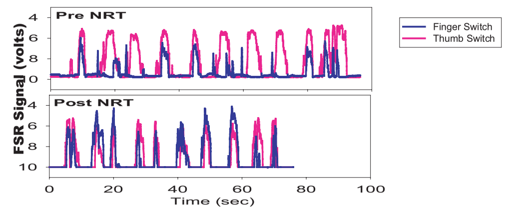

Figure 2: Following NRT the treated limb's (shown here) FSR trace was more syncronized and had more accurate FSR triggering

2. Improvement in Number of SOC's per Target

Participant 1

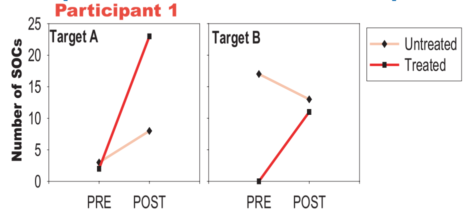

Figure 3: Untreated limb had small changes in both targets. However treated limb had substantial change in both targets.

Participant 2

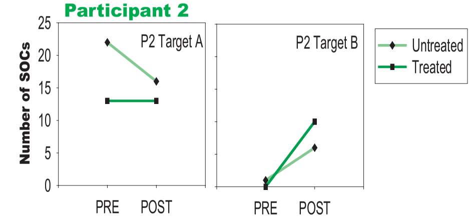

Figure 4: Untreated limb had small changes in both targets. However treated limb had substantial change in Target B, while Target A success remained unchanged.

3. Overall Mean SOC Score Signficantly Improves

Participant 1

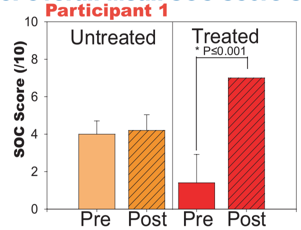

Figure 1: Mean change in SOC socre was 5.6/10

Participant 2

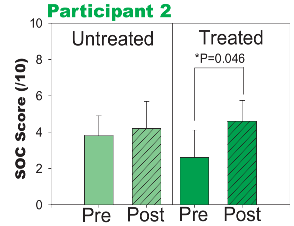

Figure 2: Mean change in SOC socre was 2/10

Conclusions

Following SCI, neural and functional recovery occurs for approximately 24 months following injury[1]. Interesting to note, although both participants' injuries occurred more than two years prior to their participation in NRT, the force porfiles for the RGTR task improved in the limb that received NRT.

It was not expected that ASIA of SCIM scores would change (these tests were not designed to assess specific hand abilities) assessments were used to characterize the participant's injury, as well as their global ability to function independently.

RGTR

Raw FSR singal for both participants in the untreated limb only. There is improvement in the timing of the triggering ot both FSR's and an increased ability to maintain contact on the FSR's during the RGTR.

Participant 1 has an increase in number of SOC regardles of the targetParticipant 2 has an increase in number of SOC to Target B only in the treated limb.Both participants had an overall significant improvement in the mean SOC socres.

Preliminary examination of the kinematic data from the RGTR task appears to demonstrate an improvement in hand motor ability. However, more exploration is needed to determine whether this increased ability is due to peripheral or central adaptations. The role of FES in these adaptations has yet to be determined.

References

[1] Waters et. al. Arch Phys Med Rehab, 75, 1994.
[2] Ditunno et. al. Arch Phys Med Rehab, 73, 1992.
[3] Popovic et. al. Spinal Cord, 39, 2001.
[4] Snoek et. al. Spinal Cord, 38, 2000.
[5] Popovic et. al. Med Eng Phys, 27, 2005.
[6] Maynard et. al. Spinal Cord, 35, 1997.
[7] Catz et. al. Disabil Rehabil 23, 2001.

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REL HAND FUNCTION TEST

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Toronto Rehabilitation Institute
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Assessing Grasping Improvement due to FES Application

Purpose of Assessment

- Evaluate gross motor function of hand post stroke/spinal cord injury
- Assess palmar and lateral grasp during functional hand activities

Technical Background

1. Neuroprosthesis for grasping using Functional Electrical Stimulation (FES) can enable restoration of two most frequently used grasps: palmar and lateral/pulp pinch grasps
2. Has implications for stroke and SCI rehabilitation
3. Occupational therapy focuses on functional outcomes of therapeutic intervention
4. Assessment items of the REL Hand Function Test are representative of activities of daily living

Materials and Methods

- REL Hand Function Test consists of two sets of tests
  - ***Object manipulation*** assesses palmar and lateral/pulp pinch grasps when hand is in pronation/neutral/supination
  - ***Strength test*** assesses grasp using weight, force and torque

OBJECT MANIPULATION TEST

- To test the **palmar grasp**, the subject is presented with the following five items: *mug, book, pop can, isosceles triangular sponge* and *mobile phone*. To test the **lateral or pulp pinch grasp**, the subject is presented with the following five items*: paper sheet, zip-lock-bag filed with five golf balls, die, credit card* and *pencil*.
- The objects are placed on a desk 20-30 cm in front of the subject in a pre-determined order. The subject is expected to pick up an object, lifting it in front of the chest, and move the object from a pronated to a neutral position, followed by a supinated position. In each position the subject is expected to hold the object for 20-30 s.
- ***Scoring:*** If the subject is unable to hold the object, he/she receives 0 points. If the subject holds the object for a short period of time (e.g., 2 to 10 s) and eventually drops the object, then the subject is awarded 1 point. Finally, if the subject is able to hold the object for 20-30 s, he/she receives 2 points.
- The following is the order in which the tasks are presented to subjects:
  1. **Mug** [*palmar grasp*]: Filled with wax to simulate load of beverage
  2. **Paper** [*pulp/lateral pinch grasp*]: A single sheet of paper A4 or Letter size
  3. **Book** [*palmar grasp*]: Standard paperback book with 150-200 pgs.
  4. **Ziplock bag** [*pulp/lateral pinch grasp*]: Filled with 5 golf balls
  5. **Can of pop** [*palmar grasp*]: Full can of pop
  6. **Die** [*pulp/lateral pinch grasp*]: A standard die
  7. **Sponge** [*palmar grasp*]: Triangle shape 40cm x 20cm x 10cm
  8. **Credit card** [*pulp/lateral pinch grasp*]: Standard card
  9. **Mobile phone** [*palmar grasp*]: Standard mobile/cellular phone
  10. **Pencil** [*pulp/lateral pinch grasp*]: Standard HB pencil

STRENGTH TEST

- To test strength of the **palmar and lateral or pulp pinch grasps** using the following four tools:
  11. **Rectangular wooden blocks** [*palmar grasp*]: Nine rectangular wooden blocks with dimensions of 40 mm x 40 mm x 120 mm. Three blocks weigh 100 g, the other three weigh 200 g, and the remaining three weigh 300 g. For each group of three blocks with the same weight: one should have a slippery surface made of overhead protecting covers; the second should have a polished wooden surface; and the third should have a non-slippery surface made of Dycem. ***Scoring:*** same as the object manipulation test.
  12. **Instrumented cylinder** [*palmar grasp*]: A cylinder with a diameter of 30 mm, the surface made of polished wood, and having a torque meter attached to it. The torque meter can record torques in the range of 0-5 Nm with a resolution of 0.05 Nm.
  13. **Credit card & dynamometer** [*pulp/lateral pinch grasp*]: A credit card attached to a dynamometer that has a range of measurable forces from 0-50 N with a resolution of 0.5 N.
  14. **Wooden bar** [*palmar grasp*]: A wooden bar that has an elliptic cross section (diameters of 35 mm and 40 mm), a length of 800 m and the weight being 600 g. ***Scoring:*** The subject should hold the bar in the middle of the handle using a palmar grasp. The person administering the test should slowly push the bar in one direction until the eccentricity of the load causes the grip to release the bar or the bar becomes inclined. Record the line on the bar with the highest number before the bar stops being horizontal and becomes inclined. The same test is done by pushing the bar in the opposite direction.

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Necessity of Successive Sensory Feedback to Update the Internal Model for Walking

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1. Background

Walking is a highly sophisticated movement in human behavior

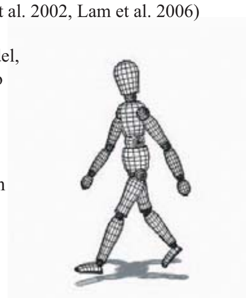
- Even when we walk on an unstable surface, we can unconsciously accomplish stable walking within a few steps.

▶ Some evidences for the involvement of *an internal model* in the control of gait were suggested (Hodgson et al. 1994, Timoszyk et al. 2002, Lam et al. 2006)

▶ Given the previous finding about the internal model, two major mechanisms have been hypothesized to underlie the above phenomenon of walking:

1. A predictive process according to previous experiences (i.e., using the internal model)
2. A modifying process that detects a mismatch between the planned and actual movement (i.e., updating the internal model).

(Kawato et al. 1987)



2. Gait Experiment

According to self-contemplation, H.O. strongly relies on visual information during walking. He told us the following interesting experiences;

▶ He cannot walk with his eyes closed and is scared to walk in darkness.
▶ Interestingly, he needs several tens of seconds after the beginning of walking untill he feels "walking is stable."

cf. Similar description was partly provided by Rothwell et al. (1982)

In order to evaluate such an interesting phenomena quantitatively, we have conducted the following *very easy and simple* gait experiment.

Methods

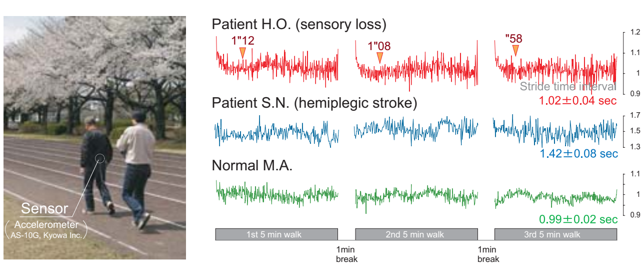
Subjects

① Patient H.O. (64yrs, Stroke patient who has complete sensory loss)
② Patient S.N. (61yrs, Stroke patient who has lacks both sensory and motor function)
③ Subject M.A. (34yrs, Able-bodied person)

Experiments consisted of three consecutive sets of walking at their comfortable speed. Stride time interval was evaluated from the data obtained from accelerometer (AS-10G, Kyowa Inc., Japan) placed on the lumber portion of the subjects (see below picture). The data was digitized (1 kHz) and stored in the memory of the data logging system (EDS-400, Kyowa Inc., Japan).

Results

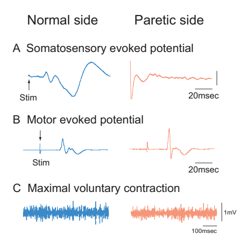
Stride interval had larger variability, but kept within a certain range. Averaged stride time interval (≒gait speed) of patient H.O. was longer (slower) than that of M.A., but shorter (faster) than that of S.N.



The subjects walked on track field with their confortable speed (left picture). The right panel shows time-series stride time interval calculated from acceleration data. Triangle markers indicate the timing when O.H. felt "walking was stable".

3. Patient's Information

H.O. is a 64 year old stroke patient. While he has complete sensory loss in the right side of the body, his motor function is very well preserved.



The results of the electrophysiological test showed that there are no somatosensory evoked potentials elicited by the electrical stimulation from paretic side of the ulnar nerve (A), but larger motor evoked potentials in the first dorsal interosseous (FDI) muscle elicited by transcranial magnetic stimulation on the motor cortex (B). Comparable EMG level during maximal voluntary contraction was observed in FDI of both side (C).

4. Discussion

Does H.O. have internal model for walking?

Because H.O. is six years post injury, his walking performance has already been well established. Therefore, H.O. could presumably plan and accomplish the task of walking by using the internal model.

Main Results and Discussion

H.O. can seemingly walk well, but it takes several tens of seconds after the beginning of walking until the subject's stride reaches a steady level. (Such adaptive changes were not observed in S.N.)

▶ Such adaptive changes might reflect the process which *compensates for the lack of sensory feedback using the visual* information.

Once H.O. stops walking and resumes walking after one minute of break, it takes much time again to reestablish stable walking.

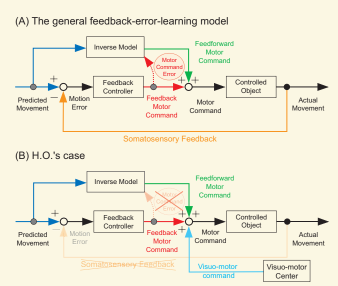
▶ Compensatory process accomplished by the visual information can be used only temporally. This process significantly differs from updating the internal model in able bodied individuals.

Proposed Model

- H.O. cannot detect a mismatch between planned and accomplished movement because of his loss of sensory feedback.

"*The difference between the predicted and actual sensory feedback can be used as an error signal to update a predictive model*"
(Wolpert and Ghahramani 2000)

- Compensatory process accomplished by the visual information cannot be preserved as an internal model.
- As a result, H.O. needs visually induced compensation for the sensory loss in each walk initiation.



The present results strongly suggest that successive sensory feedback is an essential factor for updating the internal model of walking.

References:

Hodgson et al. Can the mammalian lumber spinal cord kean a motor task? *Med Sci Sports Exerc* 26: 1491-1497, 1994
Kawato M et al. A hierarchical neural-network model for control and lerning of voluntary movement. *Biol Cybern* 57: 169-185, 1987.
Lam T et al. Contribution of feedback and feedforward strategies to locomotor adaptations. *J Neurophysiol* 95: 766-773, 2006
Rothwell JC et al. Manual motor performance in a deafferented man. *Brain* 105: 515-542, 1982
Timoszyk WK et al. The rat lumbosacral spinal cord adapts to robotic loading applied during stance. *J Neurophysiol* 88: 3108-3117, 2002
Wolpart D and Ghahremani Z Comptational principle of movement neuroscience. *Nature Neurosci* 3: 1212-1217, 2000

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Oxygen Uptake during FES Treadmill Walking: Case Study

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Introduction

•Tetraplegia patients need to exercise on regular basis to prevent secondary complications, such as cardiovascular diseases, osteoporosis and obesity.
•To achieve the required fitness level is challenging task for spinal cord injury (SCI) patients because they have difficulty to increase their exercise intensity by on their own .
•Functional electrical stimulation (FES) can restore movements by artificially contraction paralyzed or paretic muscles. Our previous study suggests that FES-assisted gait training has positive effect on SCI patients (Thrasher et al. 2005).
•It is likely that the FES-assisted gait training can provide required cardiopulmonary exercise in SCI patients by increasing their leg muscle activity and the energy consumption.
•However, to date, the physiological intensity of FES-assisted gait training has not been investigated.

Purpose

To examine the oxygen uptake and exercise intensity of the FES-assisted gait training in patients with the motor-incomplete tetraplegia.

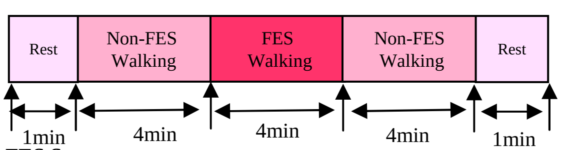
Methods

•**Case Study:** A 45-year-old men who had sustained motor-incomplete tetraplegia 2 years ago due to motor vehicle accident. The neurologic level for the motor disability was at C$ level and for the sensory disability was at C3 level (ASIA #). The subject completed a 4 month FES-assisted gait training prior to this experiment.
•**Control Subjects:** Two able-bodied men were used as controls to our case study.

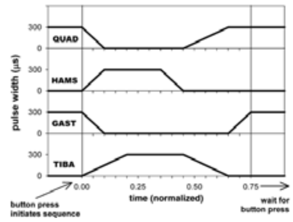
List of subjects

Subject	*Age (years)*	*Height (cm)*	*Weight (kg)*
Patient	45	175	77
Control 1	30	169.0	63.0
Control 2	39	174.0	75.7

•**Experimental Protocol**
- Walking Speed : 2.1km/h (the patient's comfortable speed)
- Protocol



•**FES System**
- Equipment: Compex Motion stimulators (Compex SA, Switzerland). Biphasic asymmetrical pulses (frequency 35Hz and pulse duration 0-300µs) were delivered to the body via eight self-adhesive gel electrodes.
- Target Muscle: quadriseps, hamstrins, gastroccnemium/soleus and tibialis anterior
- Pulse Amplitude: Patient: 75% of the maximum force induced by FES, Able-bodied subjects: 20-25mA
- FES Program: Feed-forward stimulation pattern triggered using two pushbuttons. For more details consult (Thrasher et al. 2005).

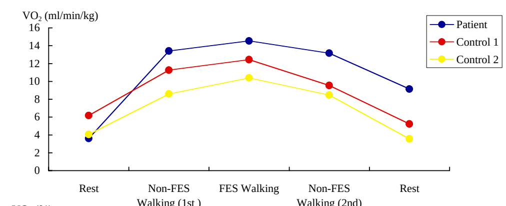


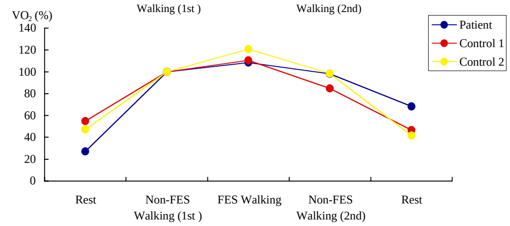
Stimulation program for a single leg. The stimulation begins from initial state representing late stance phase, and pushbutton triggers open-loop sequence beginning with swing phase (Thrasher et al. 2005).

•**Measurements**
- Device: Telemetric breath respiration mesurement system (K4b2, Cosmed, Italy)
- Measured Parameters: Oxygen Uptake (VO_2)
- Calculated Parameter: Metabolic Equivalent (METs)
(1MET = 3.5 ml/kg/m VO_2)

Results and Discussion

Oxygen Uptake during Rest, Non-FES Walking and FES Walking





VO_2 was increased during FES walking compared to the initial Non-FES walking, and then decreased again during the second Non-FES walking in both patient and control subjects (Upper Figure). The VO_2 increments from the initial Non-FES walking were 9.0, 10.6 and 20.8% in patient, control 1 and control 2, respectively (Lower Figure).

METs value of rest, Non-FES Walking and FES Walking

Subject	Rest	Non-FES Walking (1st)	FES Walking	Non-FES Walking (2nd)	Rest
Patient	1.0	3.7	4.0	3.7	2.5
Control 1	1.7	3.1	3.5	2.7	1.5
Control 2	1.1	2.4	2.9	2.4	1.0

1 METs corresponds to the resting condition.3-4 METs corresponds to the level of moderate walking.

Conclusion

Energy consumption was increased in FES Walking. The intensity of FES Walking in the patient was 4.0 METs. Thus, FES Walking has an advantage of increasing the exercise intensity for SCI patients who have difficulty to increase their exercise intensity by themselves.

Reference

TA Thrasher et al. Gait training regimen for in complete spinal cord injury using functional electrical stimulation. Spinal cord (2005)

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NEUROPROSTHESES FOR WALKING

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A Clinical Intervention for Incomplete Spinal Cord Injury

Introduction

There is growing evidence that walking assisted by Functional Electrical Stimulation (FES) can result in recovery of voluntary muscle control and significant improvements in function [1]. It has been hypothesized that direct stimulation of motor axons when coordinated with voluntary muscle effort can promote restorative modifications in the central nervous system [2]. In this pilot study, we put these ideas into practice.

Methods

Three adults with incomplete SCI, who were capable of walking unsupervised, were prescribed custom neuroprostheses for walking. The neuroprostheses were used 2-5 times per week for 13-16 weeks. The goal was to improve their ability to walk *without* FES.

Fig1 – Subject C using neuroprosthesis based on surface FES. A pushbutton is attached to the handle of the walker.

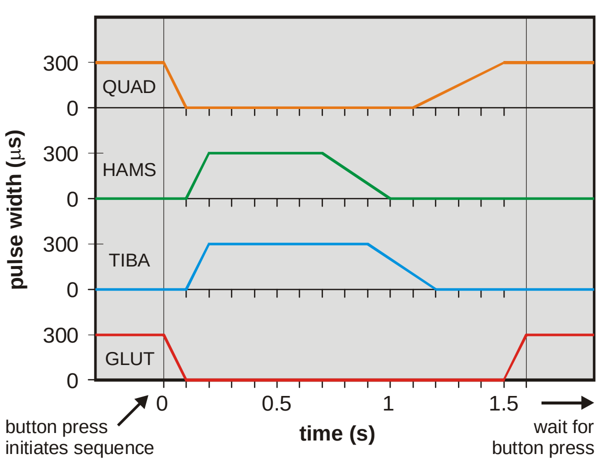

Fig2 – Stimulation sequence used by Subject C (7th revision). FES was applied to left leg only. Subject pressed button at the beginning of every stride to trigger the stimulation sequence.

Table 1 – Initial subject data

Subject	Age	Sex	SCI level	Injury duration	Cause	Assistive devices
A	36	M	T6-10	7	VHLS*	2 canes
B	24	F	T12	5	Fall	2 canes + LLB**
C	46	M	C5	7	Fall	2-wheel walker

* Von Hippel Lindau Syndrome
** Long leg brace

Before every session, subjects performed two walking tests (without FES) using their choice of assistive devices. For the first 2-4 weeks, subjects received stimulation while sitting on a bench. When the muscles began to respond well, walking exercises began.

Results

After 13 to 16 weeks of regular training with the neuroprosthesis, all three subjects demonstrated significant improvements to overall walking function *without* FES. As shown in Figure 2, subjects A & C walked significantly faster. At week 6, subject B no longer required a long-leg brace and began using an anti-hyperextension brace. Four weeks later, she was able to walk with no brace.

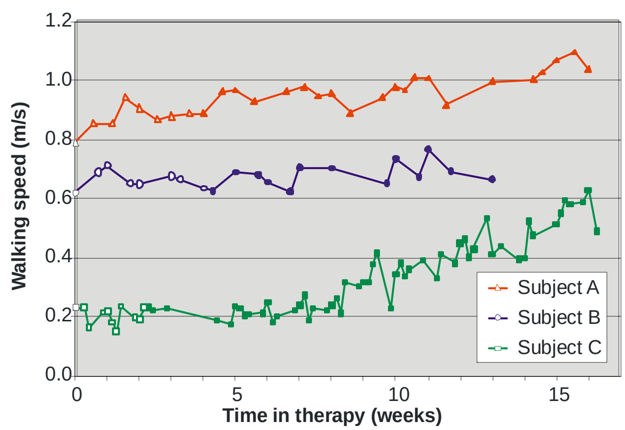

Fig3 – Walking speed over the course of treatment. Open points indicate that sitting exercises only were performed; closed points indicate that walking exercises were done.

Conclusions

The results suggest that FES-assisted walking can be an effective short-term intervention with long-term results, even for subjects that are many years post-injury. Further study is needed to compare the rehabilitative effects of FES to regular physiotherapy.

References

[1] Wieler et al. *Arch Phys Med Rehabil*, 80:495-500, 1999.
[2] Rushton. *Med Eng & Phys*, 25:75-78, 2003.

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FES FOR TREADMILL TRAINING

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Applying Passive Dynamics to Gait Rehabilitation

Introduction

- This study proposed a walking neuroprosthesis to be used in a rehabilitation setting for incomplete spinal cord injured subjects.
- This neuroprosthesis will replace the need for therapists to manually manipulate the patient's legs through the walking cycle during treadmill training.
- Since our intention is to use the neuroprosthesis as a rehabilitation tool, only a system that applies surface stimulation technology could be used, as shown in Fig. 1.

Fig. 1: Compex Motion Stimulator with surface electrodes (Compex SA, Switzerland)

- Hip flexor muscles are difficult to excite when using surface stimulation thus most past neuroprosthesis have used the peroneal nerve reflex, which is prone to habituation.
- To avoid use of the peroneal nerve stimulation it has been proposed that the participant walks downhill on the treadmill, so that gravity can induce hip flexion.

Methods

- To minimize fatigue and ensure that the hip flexors were not required during gait, a computer model for the swing phase of human gait was developed.
- To determine the appropriate stimulation current that would be applied by the neuroprosthesis during walking, the torque at the knee was measured as it moved through its range of motion (as shown in Fig. 2), for a variety of applied currents at a variety of speeds using a Biodex-2® (Biodex Medical System, USA).
- Isokinetic trials were conducted for both concentric and eccentric contractions at fixed angular velocities and fixed stimulation currents. Biphasic stimulation pulses were used at a constant pulse width of 250 µs and frequency of 25 Hz.

Results

- It was found through simulation and isokinetic FES trials that one could initiate the swing phase of gait without the use of hip flexors if the subject is walking on the 10% inclined plane.
- In Fig. 3 joint torque and FES protocol that generates the required torque to facilitate locomotion on an inclined surface are presented for the knee joint and hamstrings m.
- Initial trials have agreed with the theoretical analysis, as shown in Fig. 4, that by applying the inclined treadmill, one can facilitate FES supported locomotion, without stimulating the hip flexors. This is important because the surface FES that is ideal for rehabilitation application, cannot generate hip flexion.

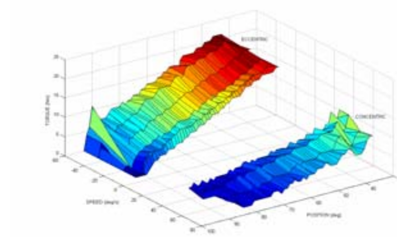

Fig. 2: Knee joint torque as a function of angular position and angular velocity, produced by hamstrings m. using FES for constant stimulation pulses that have 110 mA amplitude.

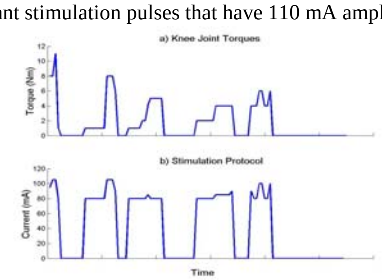

Fig. 3. a) Knee joint torque that facilitate "down hill" walking, b) Stimulation protocol applied to hamstrings m. that generates desired knee torque in a).

Discussion

- Further development of the proposed stimulation protocol is needed to include hill strike, load baring stance and hill off phases of the gait.

Fig. 4. Initial trials to validate that passive dynamics and surface FES can be applied to facilitate treadmill training in SCI subjects

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RECOVERY OF MOTOR FUNCTION FOLLOWING INCOMPLETE SPINAL CORD INJURY

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INTRODUCTION

- Distraction of protective musculoskeletal tissue and the spinal cord is a common mechanism of spinal cord injury (SCI) seen as a result of an accidental trauma or during spinal surgery.
- The computer-controlled intervertebral distraction model: (See Methods below)
  1. Applies acute tensile forces to the spinal cord
  2. Can generate three reproducible grades (mild, moderate, and severe) of injury

PURPOSE

To characterize the cellular events occurring in the spinal cord following a moderate grade spinal cord injury generated by a distractive force to the vertebral column of the rat

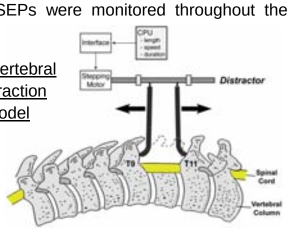
METHODS

Female Long-Evans rats, weighing 260-290g were used in all studies. All rats were acclimatized for a minimum of 1 week and maintained on a 12-hour light/dark cycle. Food and water were provided ad libitum. SEPs were monitored throughout the procedure.

Actual injury was confirmed after distraction by a decrease in SEP amplitude by ≥ 50%

-5 animals each
-Sacrificed at 1, 7, 14, and 28 days
-Immunohistochemical analysis performed looking at Synaptophysin, Glial Fibrillary Acidic Protein (GFAP), neurofilaments, and Nissl substance.

Intervertebral Distraction Model

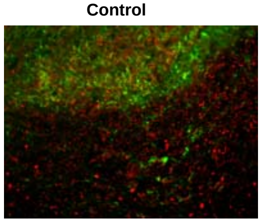


Highly controlled reproducible injury

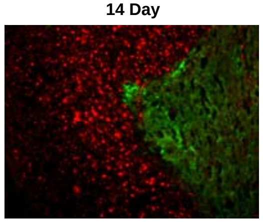
RESULTS

Axonal swelling in white matter after distraction injury

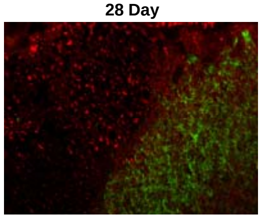
Control



14 Day



28 Day



Synaptophysin (green)
Neurofilament (red)

The red neurofilament stain shows swelling of axons in the white matter. This axonal swelling trailed the peak in gliosis being most apparent at 14 days following distraction injury (ventral horns with lateral white matter tracts are shown).

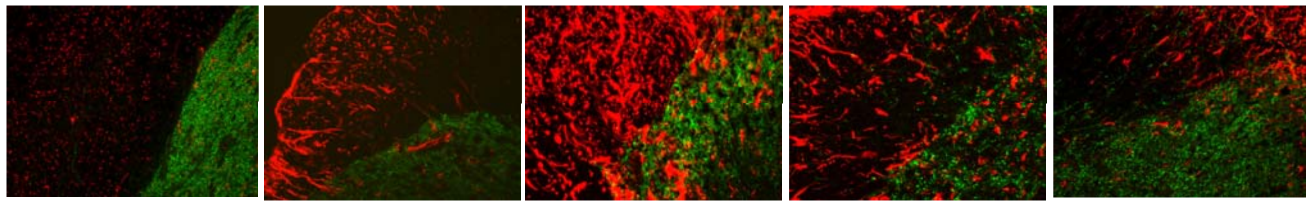
Reference: Dabney K.W et al. (2004) spine 29: 2357-2364.

Supported by a grant from the Nemours Foundation, USA.

RESULTS

White matter gliosis after distraction injury

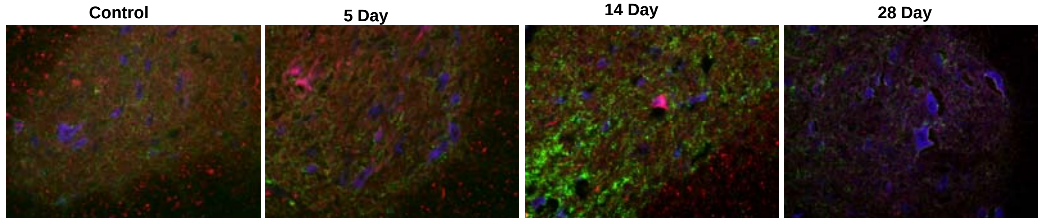
Control | 1 Day | 5 Day | 14 Day | 28 Day



Green synaptophysin staining highlights the grey matter. The red stain for glial fibriallary acidic protein (GFAP) shows robust gliosis predominantly limited to spinal cord white matter that peaks at 5 days post distraction injury (dorsal horn and lateral white matter tracts are shown).

Redistribution of synaptophysin in ventral horns after distraction injury

Control | 5 Day | 14 Day | 28 Day



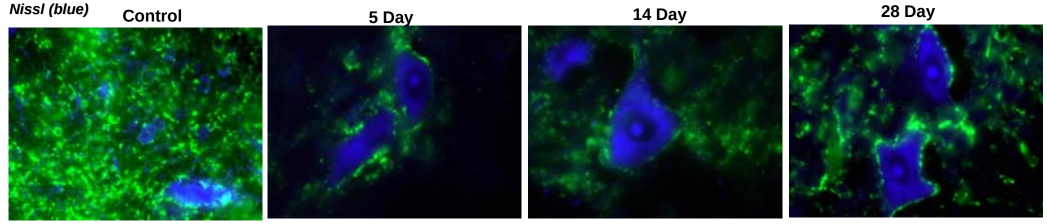
Synaptophysin (green) Neurofilament (red) Nissl (blue)

By Nissl staining to highlight the neuronal cell bodies, the grey matter appears spared of any significant injury. However, these merged images show a lasting redistribution of green synaptophysin immunoreactivity in the ventral horn grey matter. This was easily distinguished from control at 5 days and became more pronounced over the duration of the post-injury period (ventral horn is shown).

Redistribution of synaptophysin in ventral horns after distraction injury

Synaptophysin (green)
Nissl (blue)

Control | 5 Day | 14 Day | 28 Day



At high magnification, the green synaptophysin immunoreactivity is lost from the ventral horn neuropil with a corresponding increase in punctate staining along Nissl-reactive cell bodies of spinal motor neurons (asterisks) over 5 to 28 days. This suggests a reformation or accumulation of synaptic contacts along the neuronal cell body.

CONCLUSIONS

The histological pattern of moderate SCI observed following a distractive injury differs significantly from that observed in the classic weight-drop model of SCI

Significant necrosis involving both grey and white matter can be seen on gross histologic examination after weight drop injury

1) Moderate distraction SCI is primarily an: AXONAL INJURY (White matter injury)
2) Acute synaptic loss of motor neurons may cause temporary motor dysfunction of the hind limbs
3) Preserving the integrity of the neuronal perikaryon with formation/redistribution of synapses may play a role in recovery of motor function after incomplete SCI
4) By 28 days post SCI all rats recovered weight-supported walking

Influence of the Neuromuscular System on the Control Mechanism of Balance during Quiet Standing

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PURPOSE

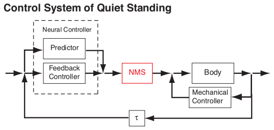
The purpose of this study is 1) to experimentally identify the neuromuscular system in standing posture, and 2) to investigate the influence of this component on the control mechanism of balance.

BACKGROUND

The ankle joint torque during quiet standing is controlled, at first, by the mechanical stiffness and viscosity, but this mechanical stabilizing torque was reported to be insufficient to stabilize the standing body. They provide 70-90% of the required torque but not all (Loram et al. 2002, Casadio et al. 2005). Therefore, the additional neural moment-to-moment intervention is inevitable. In the neural control mechanism, the feedback loop delays are problematic. The motor time delay is about 40ms from the brain to the calf muscles, and the sensory delay is also about 40ms. So the feedback loop includes at least 80ms in total. In our previous studies, we suggested that the controller with a large derivative gain can compensate the long feedback delay, even if the brain uses the simple linear controller strategy (Masani et al. 2003, 2006). Other studies proposed a feedforward strategy using a state predictor, which is popular in reaching arm control, to compensate the delay (Morasso et al. 1999, van der Kooij et al. 2001, Loram et al. 2005). However, no study considers the neuromuscular system (NMS). The NMS could generate an additional large phase delay in the control mechanism.

The neuromuscular system is concisely modeled as a 2nd order critically damped system in isometric contraction. In quiet standing, the muscle length change is very small, so we can use this concise approximation. The property of the transfer function is mostly determined by its natural frequency. And in the case of NMS, the natural frequency corresponds to the inverse form of the twitch contraction time of the muscle. Several studies reported the twitch contraction time of human calf muscles (Tani and Nagasaki 1996, Bellemare et al. 1983, Buchthal and Schmalbruch 1970). It is about 80-120ms. However, they used large electrical stimulation or voluntary contraction in isolated joint conditions. The value could be different from the standing condition.

Control System of Quiet Standing



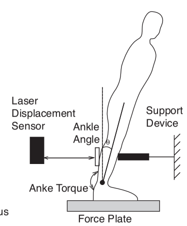
References

[1] Loram ID et al. J Physiol 545: 1041-, 2002.
[2] Masani K et al. J Neurophysiol 90: 3774-, 2003.
[3] Masani K et al. Gait and Posture 23: 164-, 2006.
[4] Morasso et al. Neural Netw 12: 1173-, 1999.
[5] van der Kooij et al. Biol Cybern 84: 103-, 2001.
[6] Loram ID et al. J Physiol 564: 281-, 2005.
[7] Tani and Nagasaki. J Electromyogr Kinesiol 6: 205-, 1996.
[8] Bellemare et al. J Neurophysiol 50: 1380-, 1983.
[9] Buchthal and Schmalbruch. Acta Physiol Scand 79: 435-, 1970.

Experimental Setup

Subjects:
10 (Male 9, Female 1, 21-39 yrs.)

Measurements:
- Ankle Torque
  <- Force Plate (AMTI, AccuSway)
- Ankle Angle
  <- Laser Displacement Sensor (Keyence, LK-2500)
- EMGs from right soleus (SOL), medial gastrocnemius (MG), lateral gastrocnemius (LG) (Delsys, Bangnoli 8)

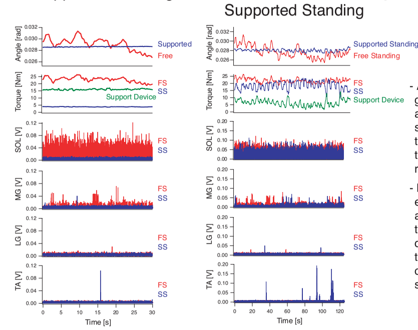


* Note: This figure is exaggerated.
The subjects stood naturally.

1 Supported Standing

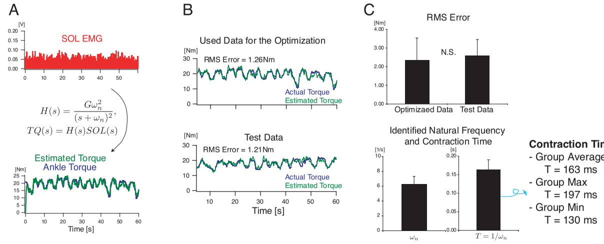
A: Supported Standing

B: Torque Exersion during Supported Standing



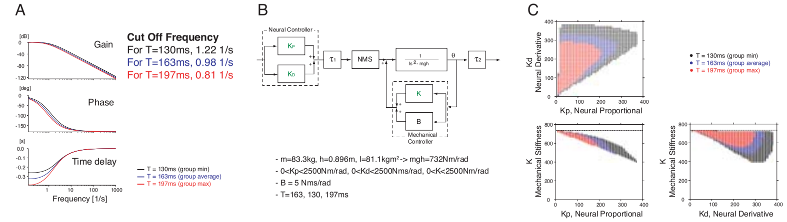
- A: The subject was supported by a ground-fixed device, keeping the ankle angle about same as the one in natural and free standing. Under this condition, the required torque was provided by the support device, so the exerted torque at the ankle joint becomes reduced to the resting level.
- B: Under this supported condition, the subject exerted the plantarflexion torque randomly but about the same level as free standing. Even in this task, the subject was supported by the device [right second, green line], and therefore the angle kept constant. Thus, we obtained the EMG-torque relationship under standing condition.

2 Identification of Nueromuscular System



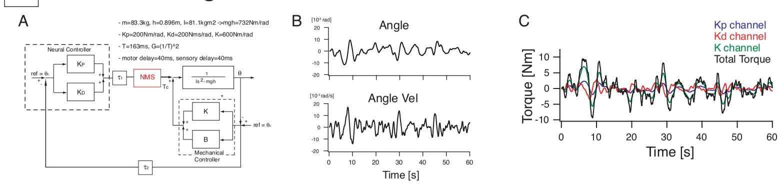
- A: The neuromuscular transfer function from SOL to the ankle torque was identified using a non-linear optimization.
- B: We used the last 60 sec for the optimization and the first 60 sec for the validatin test. In both cases, the torque estimated from EMG fitted the actual torque well.
- C: The group average RMS error were sufficiently small. Finally, the contraction time were identified.

3 Stability Analysis



- A: Bode plots of the obtained transfer functions, and the calculated time delay from the phase. The cut off frequencies are about 1 1/s. So the most of motor commands required for the balance control can pass through the transfer function, because the body sway fluctuation during quiet standing is mostly below 1Hz. However, there is a large phase delay in this frequency region, which could be a considerable destabilizing factor in the feedback loop.
- B: We executed an open loop stability test for the system that includes the NMS. We systematically changed the gains, and tested the stability of each system using Nyquist criterion.
- C: Gain sets that stabilized the whole system even with the NMS. Due to the induced phase delay, the system required the mechanical stiffness at least 400Nm/rad, which corresponds to 55% of the stiffness that stabilizes the body alone. We also can see that the neural derivative gain is larger than the neural proportional gain in most of cases. Especially in the cases that the mechanical stiffness is smaller, the derivative gain should be larger.

4 Simulation using Determined Gains



- A: We executed a simulation using a feedback model including the identified NMS, and a representative gain set determined by the stability analysis.
- B: Anglular displacement and velocity traces, which fluctuate in the physiological range.
- C: The mechanical stiffness provides most of the required torque. Kd-channel provides the same amount of the torque as Kp-channel.

CONCLUSION

- Using supported standing, we identified the NMS, i.e. the transfer function from EMG to ankle torque, during standing. The identified contraction time was slightly longer than the values determined using a single joint torque exertion in literatures.
- The motor command controlling the body successfully passes through the NMS. However, the NMS creates a large phase delay in the feedback loop, which is a significant destabilizing factor in the control mechanism.
- Due to this destabilizing effect, a large mechanical stiffness is necessary to stabilize the body. Additionally, a large neural derivative gain is required.
- The brain must use the body velocity information considerably in the calculation of the motor command. Further study is required to investigate the responsible neural mechanism.

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Partners:

Derivative Gain in the Neural Controller Accounts for the Temporal Relationship between Body Sway and Muscle Activity

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Jul 29 - Aug 4, 2006
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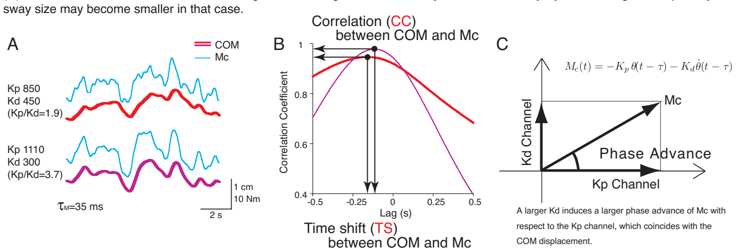
PURPOSE

The purpose of this study was to test the hypothesis that a more effective utilization of the body velocity information leads to a smaller body sway size during quiet standing.

Introduction

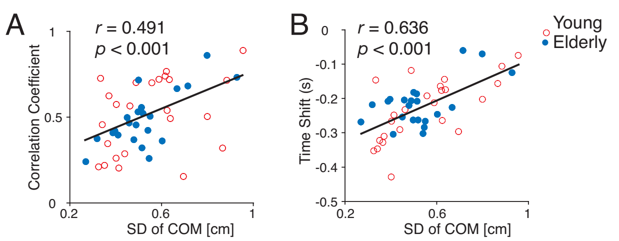
The nature of the control mechanism responsible for ensuring stability during quiet standing has attracted the attention of many researchers, since larger sway is supposed to relate to falling, which is a serious problem in the elderly. However, the mechanism that determines the sway size remains unclear.

A linear control system can account for many characteristics of human balance [1]. We have found that a linear PD controller with a large derivative gain can explain the observed temporal relationship between body sway and calf muscle activity [2, 3] (A, B below). Because the physiological feedback system includes a long loop delay, a larger derivative gain is necessary to overcome the delay (C). Therefore, if the PD controller with a larger derivative gain can effectively stabilize the body by overcoming the loop delay, the sway size may become smaller in that case.



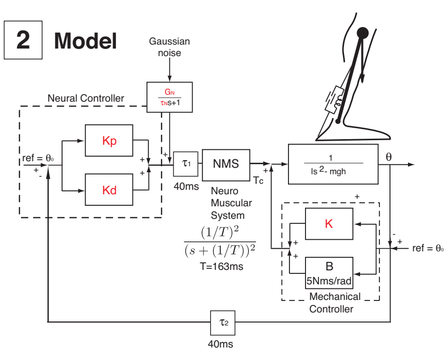
A larger Kd induces a larger phase advance of Mc with respect to the Kp channel, which coincides with the COM displacement.

1 Smaller CC and Longer TS Relate to Smaller Sway Size



- In the experiment, the correlation coefficients (CC, A) and the time shift (TS, B) highly correlated with the SD of COM, where CC is relatively scattered. This means that a person who sways less has a lower correlation and longer time shift between the COM displacement and EMG signals.
- We tested in the following analysis, which factors (i.e. neural gains, mechanical gains, noise amount, noise property) can explain these results.

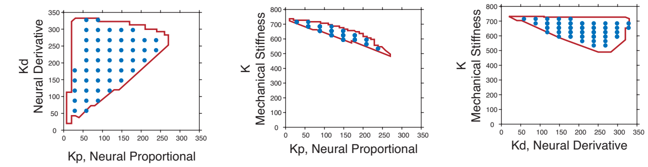
2 Model



- The body was modeled as a single inverted pendulum.
- The pendulum was controlled by two PD controllers: a mechanical controller and a neural controller.
- The neuromuscular system was considered [4].
- We systematically changed Kp, Kd, K, G_N, and τ_N and tested the stability of each system using Nyquist analysis [4].
- Seleted stable gain combinations were analyzed further.

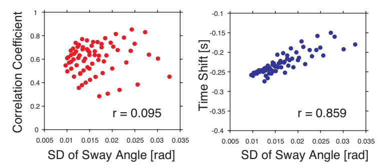
References
[1] Peterka. IEEE Eng Med Biol Mag 22(2):63-68, 2003.
[2] Masani K, Popovic MR, Nakazawa K, Kouzaki M, Nozaki D. J Neurophysiol, 90: 3774-3782, 2003.
[3] Masani K, Vette AH, Popovic MR. Gait Posture 23: 164-172, 2006.
[4] Masani K, Vette AH, Abe MO, Morris A, Nakazawa K, Popovic MR. 5th World Congress of Biomechanics, #7297, 2006.
[5] Loram ID, Kelly SM, Lakie M. J Physiol 532: 879-891, 2001.
[6] Peterka RJ. Biol Cybern 882: 335-343, 2000.
[7] Van der Kooij H, van Asseldonk E, van der Helm FCT. J Neurosci Methods 145: 175-203, 2005.

3 Stable Gain Combinations



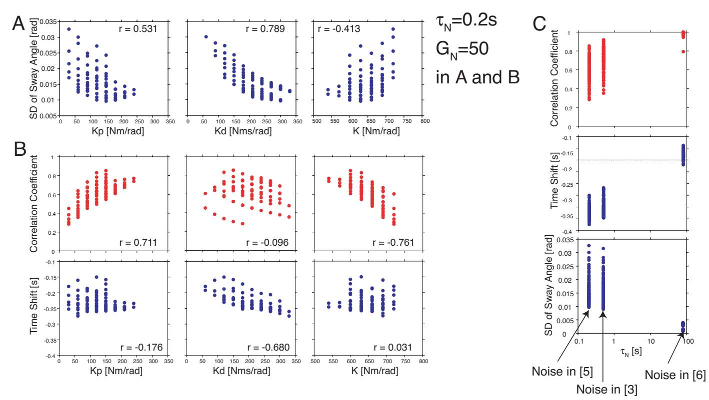
- The area covered by the red lines represents the gain combinations that stabilize the system [4]. This gain variation was assumed to correspond to the inter-subject variation.
- The blue dots represent the gain combinations used in the following analysis.

4 Longer TS Relates to Smaller Sway Size in Simulation



- Change in G_N did not affect the temporal relationship at all.
- The left figures show the result in the case of τ_N=0.2s and G_N=50.
- TS shows a strong correlation with the sway size. CC had no correlation with the sway size although it shows some tendency.

5 Neural Derivative Gain Accounts for Experimental TS Results



- Larger Kd was related to smaller sway size (A) and longer time shift (B). The variation in Kd accounts for the experimental results well, while Kp and K did not. These simulation results support our hypothesis.
- However, τ_N affected the temporal relationship (C). If the inter-subjects variation of τ_N is significant, Kd does not necessarily account for the experimental results.

CONCLUSION

The strong correlation between TS and sway size was demonstrated in the simulation as well as the experiment. Larger Kd shows a tendency to create a smaller sway and longer time shift. Therefore, the reulsts support our hypothesis. The results could be used for a "fear-free" balance test to investigate the control ability in the elderly.

However, the temporal relationship between the body sway and the muscle activity depends on the noise properties, as pointed in [7]. The inter-subject variation in the noise properties should be tested in future studies.

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Partners:

Implementation of a PD Controller for Improving Human Balance

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Introduction

Background of Study

Our simulation studies have demonstrated that a proportional and derivative (PD) controller can compensate for the sensory-motor time delay (Fig. 1A) and stabilize the body during upright stance. The required phase advance is generated by a large derivative gain K_D (Fig. 1B), resulting in a physiological motor command that precedes body sway by 100 to 200 ms [Masani *et al.*, 2006].

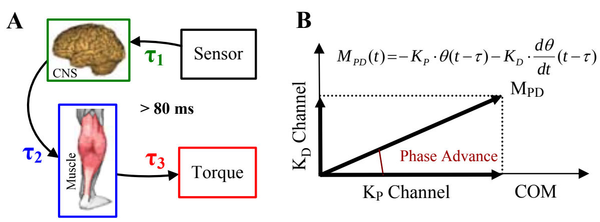

Fig. 1: Sensory-Motor Time Delay (A) and PD Phase Advance (B).

The purpose of the present study was to evaluate our theoretical results experimentally and verify that the PD controller can, in fact, improve balance during quiet standing.

Materials and Methods

Experiments

As shown in Fig. 2, the PD controller received its input from a laser sensor that estimated the anterior-posterior fluctuation of the subject's center of mass (COM).

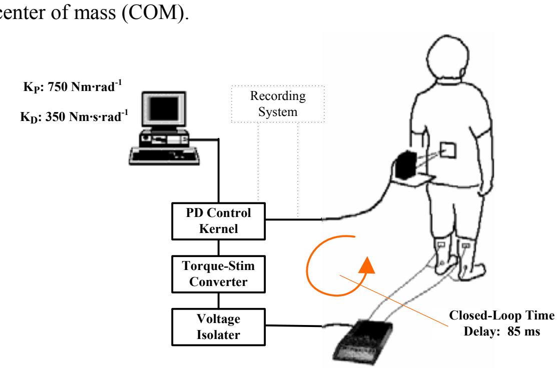

Fig. 2: Schematic Diagram of the Experimental Setup.

- Subject: – male, 36 years, 173 cm, 59 kg
  – experienced balance problems and gait impairments (*von Hippel-Lindau Syndrome*)

- Procedure: Comparison of the subject's standing performance for three different treatments (six trials each of 60 s):
  1. NTR: Natural performance without stimulation
  2. CST: Performance with constant stimulation
  3. CTR: Performance with PD controlled stimulation

Stability Analysis

Application of measures of postural steadiness to center of pressure (COP) and COM fluctuation [Prieto *et al.*, 1996]: I) Distance measures (MDIST: mean distance, RDIST: rms distance, RANGE); and II) velocity measures (MVELO: mean velocity, RVELO: rms velocity).

Results

Time Advance of Motor Command (CTR)

The controller effort generated a motor command that preceded COM by 203 ms (Fig. 3).

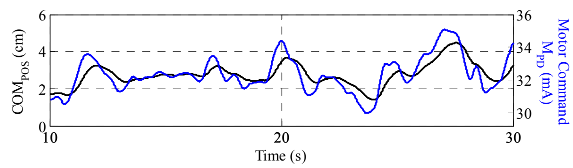

Fig. 3: COM and Stimulation Fluctuation for CTR.

Stability Results

Table 1: Average Stability Results for Each Treatment (Mean±SD).

	COM	NTR	CST	CTR
I	MDIST* [cm]	0.832 ± 0.166	1.052 ± 0.243	**0.690 ± 0.154**
	RDIST* [cm]	1.083 ± 0.231	1.253 ± 0.269	**0.871 ± 0.217**
	RANGE [cm]	6.003 ± 1.382	5.701 ± 1.194	**4.498 ± 1.030**
II	MVELO* [cm/s]	0.939 ± 0.115	0.882 ± 0.102	**0.776 ± 0.086**
	RVELO* [cm/s]	1.237 ± 0.152	1.131 ± 0.136	**1.019 ± 0.091**
	COP	**NTR**	**CST**	**CTR**
I	MDIST* [cm]	1.009 ± 0.136	1.158 ± 0.229	**0.857 ± 0.126**
	RDIST* [cm]	1.308 ± 0.181	1.423 ± 0.270	**1.095 ± 0.169**
	RANGE* [cm]	8.399 ± 0.898	7.767 ± 1.390	**6.723 ± 0.862**
II	MVELO [cm/s]	**2.607 ± 0.444**	2.691 ± 0.478	2.759 ± 0.740
	RVELO [cm/s]	**3.542 ± 0.550**	3.582 ± 0.608	3.761 ± 1.099

- Smallest average value among the three treatments in bold font.
- Significant differences ($\alpha = 0.05$) for seven out of ten measures (*).

Discussion and Conclusion

- Since all COM measures were smallest for CTR, body sway has evidently been reduced by the proposed FES system.

- The lower COP distance measures in CTR compared to NTR can be related to a higher level of stability. The higher velocity measures can be explained by a higher regulatory activity of the control system [Prieto *et al.*, 1996], proving the effectiveness of the FES system.

The results suggest that the PD controller significantly reduced body sway in CTR and that one of the system's key characteristics – the preceding time of the motor command – is almost identical for both the simulations and CTR. Consequently, human balance can be improved by means of a PD controlled FES system that mimics the active control task of an intact CNS. Furthermore, the system's effectiveness shows that the CNS adopts a control strategy that relies highly on the velocity information.

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Balance Control during Standing: Stabilizing PD Controller Generates COM Dynamics as Observed in Healthy Individuals

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Introduction

Background of Study

Our simulation studies have demonstrated that a proportional and derivative (PD) controller can compensate for the sensory-motor time delay (Fig. 1A) and stabilize the body during upright stance. The required phase advance is generated by a large derivative gain K_D (Fig. 1B), resulting in a physiological motor command that precedes body sway by 100 to 200 ms [Masani *et al.*, 2006].

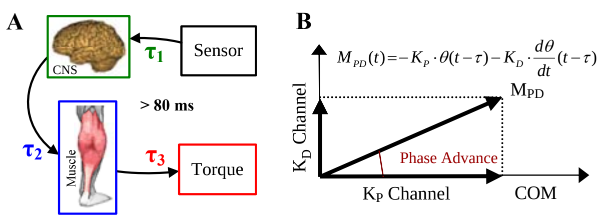

Fig. 1: Sensory-Motor Time Delay (A) and PD Phase Advance (B).

After verifying experimentally that the PD controller can regulate balance during quiet stance [Vette *et al.*, 2005], the purpose of the present study was to investigate whether the cross-correlation (CCF) between the center of mass (COM) and PD motor command (M_{PD}) indicates system dynamics as identified in able-bodied experiments.

Materials and Methods

Experiments

The PD controller received its input from a laser sensor that estimated the anterior-posterior fluctuation of the COM (Fig. 2).

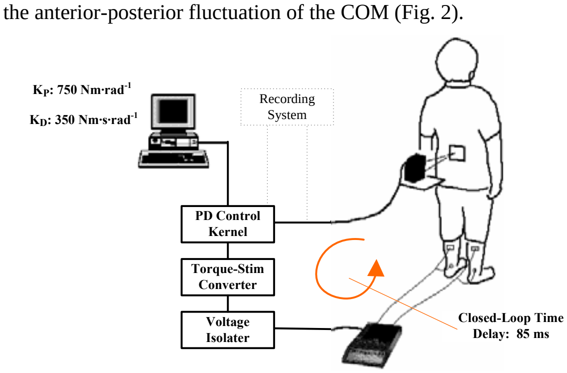

Fig. 2: Schematic Diagram of the Experimental Setup.

- Subject:
  - male, 36 years, 173 cm, 59 kg
  - experienced balance problems and gait impairments (*von Hippel-Lindau Syndrome*)
- Closed-Loop Time Delay:
  - 51 ms (*FES*) + 33 ms (*filter*) + 1 ms (*laser*) = 85 ms
  - in the range of sensory-motor time delay (>80 ms)
- Procedure:
  - quiet standing with eyes open (6 × 60 s)
  - COM and M_{PD} measurement (1 kHz)
  - low-pass filtering (order: 4, cut-off: 5 Hz)

Cross-Correlation Analysis

- Two CCFs: 1) CCF between the COM position (COM_{POS}) and M_{PD}; and 2) CCF between the COM velocity (COM_{VEL}) and M_{PD}.
- The average time shifts from COM_{POS} to M_{PD} and from COM_{VEL} to M_{PD} were related to respective time shift ranges of healthy individuals [Masani *et al.*, 2003]:
  - COM_{POS}-M_{EMG}: –155 ± 46 ms
  - COM_{VEL}-M_{EMG}: +121 ± 134 ms and –620 ± 134 ms,

where M_{EMG} is the EMG of the right medial gastrocnemius muscle.

Results

- The controller effort generated a motor command M_{PD} that preceded the COM_{POS} fluctuation in spite of the closed-loop time delay (Fig. 3):

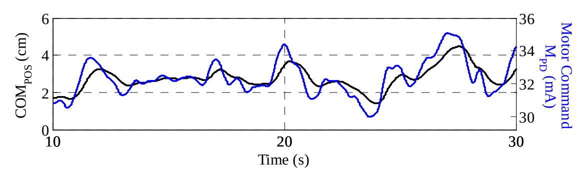

Fig. 3: COM_{POS} and M_{PD} Fluctuation.

- The peaks of the average CCFs – one for COM_{POS}-M_{PD} and two for COM_{VEL}-M_{PD} – lay within the specified target ranges (Fig. 4):

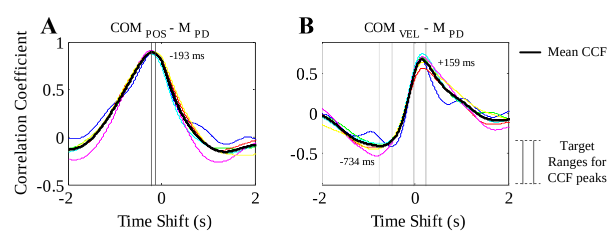

Fig. 4: CCFs between: (A) COM_{POS} and M_{PD}; (B) COM_{VEL} and M_{PD}.

Discussion and Conclusion

- Since the preceding time of M_{PD} with respect to COM_{POS} lies not only in the target range, but is fairly large in comparison with healthy subjects, the PD controller evidently provides a sufficiently preceding motor command and compensates for the closed-loop time delay.
- Due to the fact that the peaks of the CCF between COM_{VEL} and M_{PD} lie within the specified target ranges, M_{PD} generated dynamics as seen in healthy subjects.

The fact that the performed CCF analysis is in agreement with the findings in healthy individuals indicates that a velocity accentuated PD controller can mimic the physiologic control task, compensate for long neurological delays and regulate balance when certain neuromuscular disorders are present.

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Dynamic Modeling and Analysis of Arm-Free Paraplegic Standing

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Introduction

The present study provides a dynamic analysis of human double-support stance that is needed to investigate the feasibility of closed-loop control of FES-assisted standing for paraplegics. The stability analysis for two nonlinear controllers suggested that the developed 3D dynamic model could achieve asymptotic stability if only six out of twelve degrees of freedom (DOF) in the lower limbs are controlled. This result implies that the dynamic redundancy of the physiologic system of bipedal stance allows the selection of an ideal subset of DOF in a particular patient to design a neuroprosthesis for standing.

Methods

3D Dynamic Modeling

The 3D dynamic model shown in Fig. 1 represents quiet standing. The HAT (head-arms-trunk) was modeled as a rigid body, whereas each leg consisted of six DOF. ${}^{w}\boldsymbol{F}_o$ denotes the external force and moment applied to the HAT COM (center of mass). Denavit-Hartenberg notation was used for kinematic modeling of the legs. Newton-Euler and Lagrange formulations were used for dynamic modeling of the HAT and the legs, respectively. Details of the model can be found in [1].

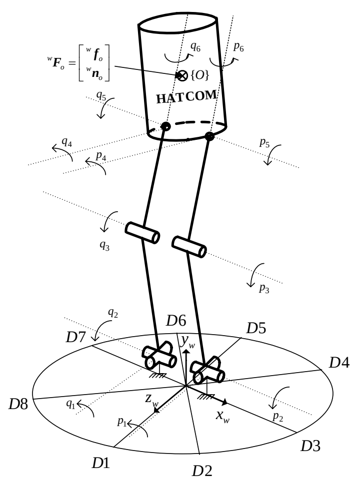


Fig. 1: Dynamic model of quiet standing with twelve DOF (q_i and p_j) and eight perturbation directions (D1 to D8).

Inverse Dynamics Solution

By utilizing Nakamura's method [2], the inverse dynamics solution was found such that a unique torque input at the minimum number of six DOF can generate the desired kinematic output of the dynamic model. Table 1 describes the six combinations of the minimum number of DOF such that the inverse dynamics solution exists. Note that Table 1 is based on the assumption that q_1 and p_1 as well as q_6 and p_6 were always passive DOF.

FES Control Strategy

Based on the above results, two controllers, a PD (proportional and derivative) plus gravity compensator and a computed torque controller, were proposed for FES-assisted arm-free standing of paraplegic individuals. The theoretical stability analyses for both controllers suggested that controlling the minimum number of DOF described in Table 1 can obtain asymptotic stability for the nonlinear dynamic model.

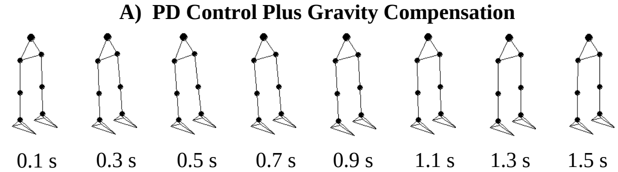
Table 1: Six cases of six active DOF combinations such that the inverse dynamics solution exists (D/P: Dorsiflexion/Plantarflexion, F/E: Flexion/Extension, and A/A: Abduction/Adduction).

DOF	Ankle D/P		Knee F/E		Hip A/A		Hip F/E	
	q_2	p_2	q_3	p_3	q_4	p_4	q_5	p_5
I		●	●	●		●	●	●
II	●		●	●		●	●	●
III	●	●		●		●	●	●
IV	●	●	●			●	●	●
V	●	●	●	●	●		●	
VI	●	●	●	●		●	●	

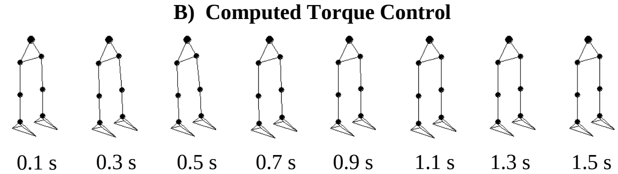
Simulation Results

- **Subject**: A complete paraplegic individual with 66.7 kg of weight and 1.72 m of height.
- **Initial Condition**: Subject being in the upright posture.
- **Controlled DOF**: Case VI (see Table 1).
- **External Disturbance**: 100 N at the HAT COM in the direction of D6 (see Fig. 1) for 0.5 s.

A) PD Control Plus Gravity Compensation



B) Computed Torque Control



Conclusion

By means of inverse dynamics analysis and control synthesis of closed-chain robotic systems, we demonstrated that asymptotic stability can be obtained by regulating only six out of the model's twelve DOF. This result implies that dynamic redundancy is, in fact, an advantage with respect to a practical implementation of an FES system for paraplegic standing. That is, it is not necessary to control all DOF in the lower limbs to obtain stable standing of paraplegic individuals, but rather only six DOF need to be actuated. As such, one can choose a feasible set of DOF to be controlled depending on the individual.

References

[1] J. Kim, "3D Dynamic Modeling, Control Synthesis, and Controllability Analysis for FES-Assisted Paraplegic Standing," M.A.Sc. Thesis, University of Toronto, 2005.

[2] Y. Nakamura and M. Ghodoussi, "Dynamics computation of closed-link robot mechanisms with nonredundant and redundant actuators," *IEEE Trans. Robotics and Automation*, vol. 5, pp. 294-302, 1989.

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Partners:

Optimal Combination of Active Degrees of Freedom to Facilitate Arm-Free Paraplegic Standing

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Introduction

Background of Study

Arm-free paraplegic standing via functional electrical stimulation (FES) has drawn much attention in the field of rehabilitation as it might allow an individual with paraplegia to stand and at the same time use both arms to perform daily activities. In this context, we have recently shown that only six of the existing twelve degrees of freedom (DOF) in the lower limbs need to be actuated to facilitate stable standing when external perturbations are present [Kim *et al.*, 2006].

After identifying the various combinations of six DOF that allow stable standing during perturbed arm-free standing, the purpose of the present study was to determine the *optimal* DOF combination, i.e., the combination that requires the minimum total torque to regulate balance.

Methods

3D Dynamic Model

The 3D dynamic model, which represents the human body during double-support stance, is shown in Fig. 1 [Kim *et al.*, 2006].

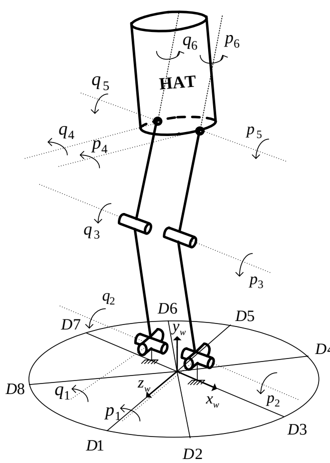

Fig. 1: Dynamic model of quiet standing with twelve DOF (q_i and p_j) and eight perturbation directions (D1 to D8).

- HAT:
  - Head-Arms-Trunk
  - rigid body with constant mass and moment of inertia
- Legs:
  - six DOF each
  - sufficient number of DOF for a reasonable approximation of the double-support characteristics of human standing.

Inverse Dynamics Solution

Application of a computationally efficient inverse dynamics method [Nakamura *et al.*, 1989]:

- driven by kinematic data from four healthy subjects.
- torque estimation for eight perturbation directions (D1 to D8 in Fig. 1)
- torque estimation for six combinations of six active DOF (Cases I to VI in Fig. 2) as identified in [Kim *et al.*, 2006].

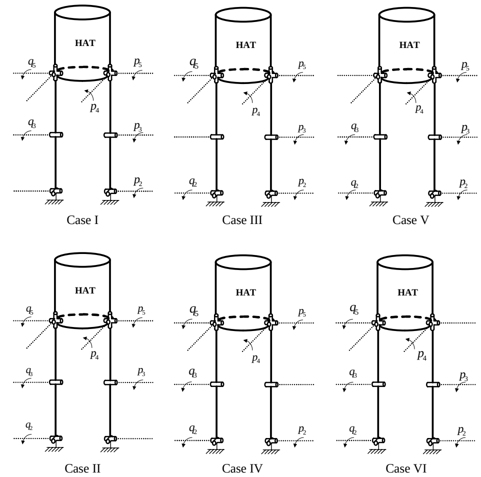

Fig. 2: Six cases of six DOF that need to be actuated in order to facilitate stable standing when external perturbations are present.

Results and Discussion

The torque sums for each case and each perturbation direction are shown in Fig. 3.

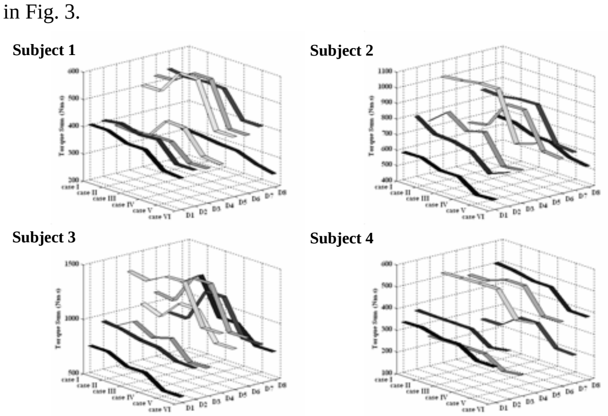

Fig. 3: Torque sums of the joint torques at all active DOF, shown for each of the six cases and all directions of perturbation (D1 to D8).

- Cases V and VI (Fig. 2) exhibited the smallest torque sums (Fig. 3). for all subjects and perturbations.
- Due to the symmetry of the two legs, this results in four active DOF combinations that are optimal with respect to energy consumption.

Conclusion

The determined combinations can achieve paraplegic arm-free standing while minimizing the forces produced by the FES-controlled muscles when one of these specific sets of DOF in the lower limbs is actuated. As such, FES-assisted, arm-free standing is, in fact, feasible, even when only six DOF in the lower limbs of a paraplegic are being controlled to account for the difficulty of accessing or actuating various DOF.

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CRSNG NSERC Natural Sciences and Engineering Research Council of Canada / Conseil de recherches en sciences naturelles et en génie du Canada

BALANCE DURING QUIET STANDING

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The Role of Toes in the Fine-Tuning Control of Balance

Introduction

The human body can be approximated as an inverted pendulum that rotates around the ankle joint. The ankle torque should satisfy the following equation concerning the foot segment:

$$\tau + f_v u \approx 0$$

where f_v is the vertical component of the ground reaction force, and u is the centre of pressure (COP) position. In most studies, it is usually supposed that ankle torque is controlled by regulating the muscle force exerted by the ankle extensors, and the COP position is proportional to the amount of ankle torque. However, if other muscles, such as the toe muscles, can move the COP position independently from the ankle muscles, the ankle torque can then be modulated. Therefore, we hypothesized that toe muscles can modulate COP.

Methods

- Subjects: 10 adults, 5 males and 5 females (age: 26.4±3.1 yrs.).
- Stimulation target muscles: Flexor hallucis brevis and flexor digitorum brevis of both legs.
- Electrical stimulation:
  - Device: Compex Motion, Switzerland.
  - Stimulation strength: 0.8, 1.3, 1.5 and 2 times the motor threshold (MT).
- Measurements:
  - COP: measured with two force plates (Kistler, Type 9366AB05)
  - Centre of Mass (COM): estimated with a laser displacement sensor (Keyence, LK-2500).
- Subjects stood quietly with eyes open for 2 min. Muscles were stimulated 20 times for 1 s in random intervals.

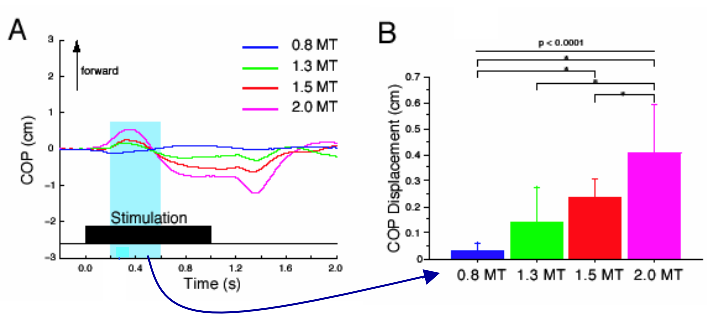

Fig. 1 – Ensemble averaged COP traces from one subject

Results

- As shown in Fig. 1, after the forward movement a backward COP movement was induced. Then, the COP kept a constant position. After the cessation of the stimulation, a backward COP movement occurred, and then, the COP went back to the initial position.
- As shown in Fig. 2.A, COP started moving at 0.2 s while COM did not move. Backward ACC started at 0.2 s and was at peak around 0.4 s, when the deviation between COP and COM (COP-COM) appeared the largest.
- At 0.4 s, COM started moving backward, and then COP and COM coincided at 0.7 s when ACC returned to zero.
- The amount of COP-COM between 0.2 s and 0.6 s depended on the stimulation level (Fig. 2.B), as well as the amount of ACC (Fig. 2.C).

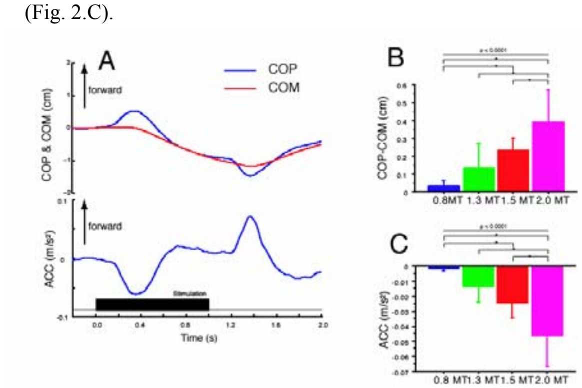

Fig. 2 – Stimulation induced COP-COM as well as ACC

- According to the inverted pendulum model, COP-COM and ACC must satisfy the following equation:

$$ACC = -(mgh/I)(COP - COM)$$

Therefore the slope of the regression line for the relationship between COP-COM and ACC must be equal to *–mgh/I*. For each subject, the theoretical value of the slope was compared with the slope of the regression line, and there was no significant difference between them.

The results indicate that the deviation between COP and COM generated ACC according to the dynamics of the inverted pendulum model.

Conclusion

- The level of stimulation can change the amount of body movement.
- The results suggest that toe muscles have a role in balancing the body during quiet standing.
- The results also show the feasibility of using electrical stimulation on toe muscles to control the COP.

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VOLUNTARY KNEE RESPONSE IN HEALTHY SUBJECTS

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Developing Control Design Criteria for FES-Based Knee Movement

Introduction

• The most commonly used technique for restoring motor function in individuals with spinal cord injury (SCI) is functional electrical stimulation (FES), which involves electrically stimulating the neuromuscular system to generate skeletal muscle contractions.

• SCI individuals have successfully used FES for standing [1], cardiovascular exercise [2] and gait neuro-rehabilitation [3].

• Currently, all clinical FES systems require regular input from the user.

• There are many potential applications of FES that require the systems to work autonomously, i.e. without user intervention. These applications include FES systems for balancing during standing, torso control during sitting, and FES-assisted walking.

•Autonomous FES systems require closed-loop control systems, which are not currently available for FES systems.

• We are currently developing a closed-loop FES control system for knee movement. The voluntary knee response in healthy subjects provides an excellent benchmark that can be use to design the controller for this system. This poster discusses how this benchmark data was obtained.

Hypotheses

• For a step trajectory, we expected that the voluntary knee response would overshoot the desired trajectory and then quickly settle to the desired knee angle.

• For a sinusoidal trajectory, we expected that the voluntary knee response would slightly lag the desired trajectory and that there would be some error in the amplitude of the response.

Methods

We measured how closely the voluntary knee movements of seated healthy volunteers followed a prescribed knee trajectories:

- Step trajectory - desired knee position goes from 90 degrees flexion to full extension as quickly as possible
- Sinusoidal trajectory - knee repeatedly flexes and extends

These two trajectories were selected because they are considered standard design benchmarks in control system design [4].

Seven subjects were seated on a padded bench, and their thigh was held stationary with a strap. Visual cues were provided to indicate the desired trajectory using LEDs mounted in a board (see Fig. 1). The actual knee movements of the subjects were recorded as they attempted to follow the trajectory. Each subject performed three trials for the step trajectory, and three trials for slow, medium, and fast sinusoidal trajectories. The frequencies of the sinusoidal trajectories were chosen to cover the extremes of knee movement speed.

Results

The step response metrics (delay, 10-90% rise time, 2% settling time, percent overshoot, steady-state error) and sinusoidal response metrics (lag on knee extension and flexion, root-mean-squared error) were calculated for each trial [4]. The averages across all trials are shown in Tab. 1 and 2.

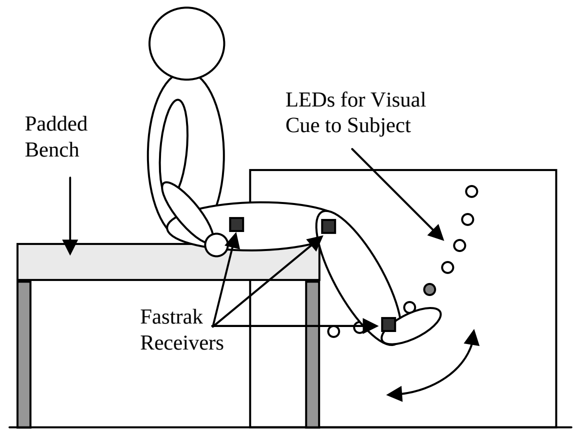

Figure 1 – Experimental Apparatus

Table 1 – Step Response Metrics (Knee Extension)

	Delay	Rise Time	Settling Time	Over-shoot	S. S. Error
Average	0.24 s	0.20 s	0.77 s	6 %	14 %

Table 2 – Sinusoidal Response Metrics

Trajectory Speed	Extension Lag	Flexion Lag	RMS Error
Slow (0.14 Hz)	0.30 s	0.14 s	7.69 rad
Medium (0.27 Hz)	0.16 s	0.16 s	4.88 rad
Fast (1.36 Hz)	0.60 s	0.55 s	10.30 rad

Discussion

As expected, the average subject overshot by 6% on the step response (knee extension) task, and settled to the steady-state value in less than 1 second. Subjects had trouble following the slowest and fastest sinusoidal trajectories, but did relatively well when following the 0.27 Hz trajectory, as shown by the small lag and RMS error. The performance on the extension and flexion parts of the sinusoidal trajectory were similar.

We will use this data to develop controller performance criteria for the closed-loop FES system, and to compare different designs to the behaviour of the healthy knee.

References

[1] Abbas JJ & Riener R. Neuromodulation. 4:187-95. 2001.

[2] Davoodi R & Andrews BJ. Trans Biomed Eng. 51:541-3. 2004.

[3] Thrasher TA & Popovic MR. 25th EMBS Conf. 1:1566-8. 2003.

[4] Phillips CL & Harbor RD. Feedback Control Systems, 3rd edition. Prentice Hall, Upper Saddle River, NJ, USA. 1996.

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EVALUATING EXISTING FES CONTROLLERS

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[2]Toronto Rehabilitation Institute

Will existing control methods work for a clinical FES system?

Introduction

• The most commonly used technique for restoring motor function in individuals with spinal cord injury (SCI) is functional electrical stimulation (FES), which involves electrically stimulating the neuromuscular system to generate skeletal muscle contractions.

• SCI individuals have successfully used FES for standing, cardiovascular exercise, and gait neuro-rehabilitation.

• We are developing a new FES system that can operate automatically, with minimal user intervention; this is called a "closed-loop" system. A critical step is finding a controller for regulating the muscle stimulation so that the desired joint angle or torque can be achieved.

• Several published closed-loop controllers looked promising for our new FES system. We evaluated these controllers in simulation to determine if their reported performance could be achieved in a realistic test, namely controlling knee angle by stimulating the quadriceps muscle.

• We tested three closed-loop FES controllers:
1. a gain scheduling controller composed of a number of locally defined linear controllers [1]
2. a sliding mode controller, which ensures system stability and drives the error between the desired and actual knee angles towards zero [2]
3. a standard proportional-integral-derivative (PID) controller.

Hypotheses

We expected that the gain scheduling controller would perform well, provided that the local controllers adequately captured all of the behaviour of the stimulated knee. We also expected that the sliding mode controller would perform well. We anticipated that the PID controller would perform the best, since this type of controller can be tuned to provide excellent performance as long as the system does not change with time.

Methods

We tested the controllers using [3], which is a physiologically-based model of knee response to quadriceps stimulation that was characterized using data from SCI subjects. We implemented this model in Simulink, which allowed us to test the controller's performance when tracking a step or a sinusoidal trajectory. Also, a disturbance was applied at 6 s to evaluate how quickly the controllers were able to recover from this perturbation.

Results

Fig. 1 is the unit step response for the three controllers. In this case, the desired "step" movement was a knee extension starting from π/2 radians to π/4 radians. Fig. 2 shows the sinusoidal response for the three controllers.

Discussion

The PID controller performed best in both tracking tasks. The sliding mode controller performed well, but exhibited an error in both tracking tasks that was likely due to modeling errors. The gain scheduling controller was under-damped for the step trajectory, but behaved well when tracking sinusoids within a limited range of knee angles. Its failure outside these angles was probably due to an insufficient set of local linear controllers.

We plan to conduct more experiments with these controllers to assess their performance in the presence of muscle fatigue and spasticity, since we expect the controllers to perform less well under these conditions. We will use the results of this testing to design the controller for the new closed-loop FES system.

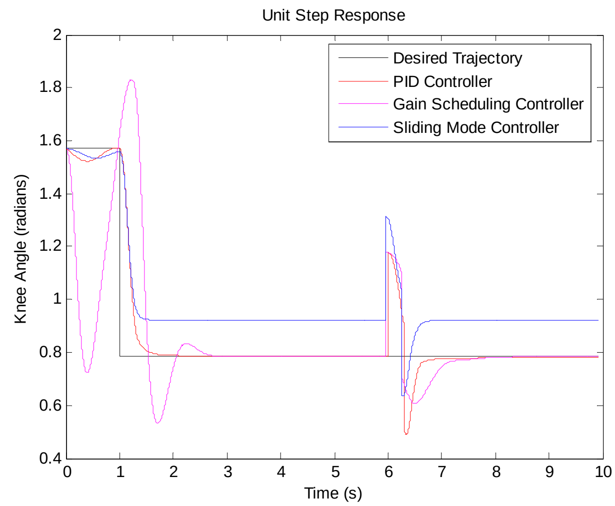

Figure 1 – Unit Step Response

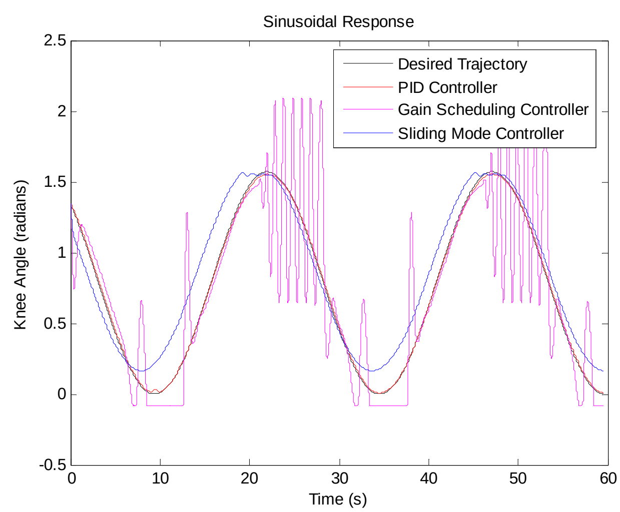

Figure 2 – Sinusoidal Response

References

[1] Previdi F & Carpanzano E. Trans Contr Syst Technol. 11(3):310-24. 2004.

[2] Jezernik S, Wassink RGV, & Keller T. Trans Neural Syst Rehab Eng. 12(1):1-2. 2004.

[3] Ferrarin M & Pedotti A. Trans Rehab Eng. 8(3):342-52. 2000.

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Muscle synergies for pressure management during prolonged sitting

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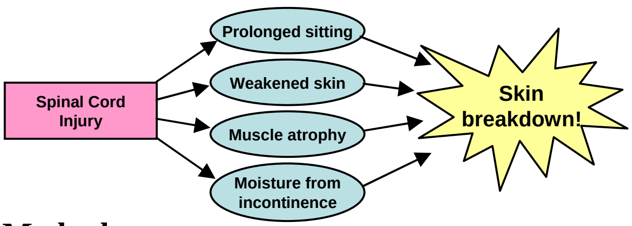
[3]Universidad Iberoamericana, Mexcio City, Mexico

Introduction

- Pressure ulcers are one of the most significant secondary complications following spinal cord injury (SCI). They occur in up to 56% of the SCI population [1].
- Stimulation of the gluteal muscles and hamstrings has been shown to reduce interface pressure in the gluteal region [2]. Little is known about how these muscles interact naturally to relieve pressure during prolonged sitting.
- The purpose of this study was to identify patterns of muscle contraction that could be used by a neuroprosthesis to reduce the risk of pressure sores in SCI.



Methods

- 8 able-bodied subjects and 8 subjects with complete SCI (>1 year post-injury) sat for 1 hour in a customized wheelchair while watching a documentary alone in a quiet room.
- Sitting pressure profile was measured using an XSensor pressure-mapping mat, with 36x36 measuring cells sized 1 cm^2 each.
- In the able-bodied subjects, surface EMG was recorded from six muscles bilaterally: 1) Gluteus Maximus, 2) Gluteus Medius, 3) Adductor Longus, 4) Biceps Femoris, 5) Vastus Lateralis, and 6) Lumbar Erector Spinae.
- Equilibrium states and transitional states were determined by observing changes in the Center of Pressure (COP). A principal component analysis was conducted on the EMG signals to determine patterns of muscle coordination during the sitting task.

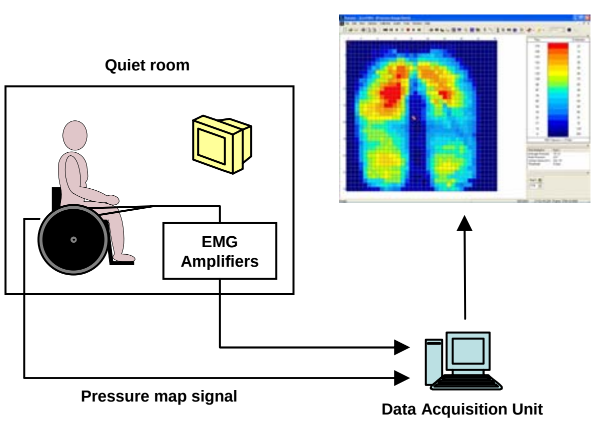


Fig1 –Subjects remained seated for 60 minutes with their prescribed cushion. Pressure profile was recorded.

Results

Table 1 – Frequency of weight shifts in able-bodied subjects

Subject	# of weight shifts	Duration of equilibrium Mean ± S.D. (sec)
A1	25	131 ± 83
A2	64	38 ± 29
A3	12	258 ± 210
A4	21	153 ± 109
A5	50	44 ± 28
A6	18	185 ± 105
A7	9	293 ± 180
A8	12	237 ± 248

- Able-bodied subjects shifted weight on average **34 times per hour** (every 1.8 minutes).
- SCI subjects shifted weight on average **8 times per hour** (every 7.5 minutes).

- In SCI, weight shifts were always conscious, whereas able-bodied subjects were usually unaware of when they were shifting.
- All six muscle groups that were recorded participated in weight shifting and all maintained steady levels of tonus when the COP was stationary. However, EMG levels varied between equilibrium states. General patterns of muscle co-activation were found; they were represented by **five** principal components.

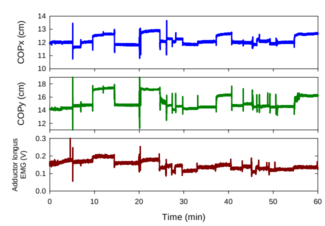


Fig2 – Sample pressure profile with matching EMG signal from one able-bodied subject (A4)

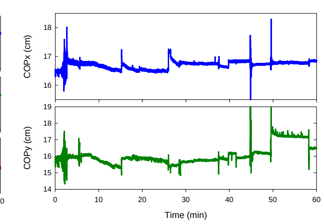


Fig3 – Sample pressure profile from one SCI subject (S1)

Conclusions

- Five principal components were identified which describe muscle synergies during normal sitting. All 6 muscle groups recorded (bilaterally) played a role in sitting pressure management.
- The results suggest that surface FES could be used to control COP position in the SCI population. However, to restore normal muscle activation patterns requires a complex multi-channel approach.
- We conclude that SCI patients should be trained to increase the frequency of weight-shifting movements as a first step to prevent pressure ulcers.

References

1. Ash D. An exploration of the occurrence of pressure ulcers in a British spinal injuries unit. *J Clin Nurs*. 2002 Jul;11(4):470-8.
2. Ferguson AC, et al. Reduction of seating pressure using FES in patients with spinal cord injury. A preliminary report, *Paraplegia*. 1992 Jul;30(7):474-8.

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Development of an Automated Perturbation System

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Rehabilitation Engineering Laboratory - Toronto Rehab Institute
***University of Toronto – Institute of Biomaterials and Biomedical Engineering**

Introduction

- Trunk instability is a major problem for people with spinal cord injury (SCI), since their lumbar muscles cannot produce sufficient forces to stabilize the lumbar spine. People with SCI often cope with trunk instability by assuming a slumped posture and leaning against the backrests of their chair. When reaching, they use one arm on their lap or thrown over the back of the chair to provide the external forces necessary to keep the trunk from bending forward uncontrollably. Although a significant body of work has been published in the field of sitting balance and posture control, there are still many fundamental questions that remain unanswered.

- The Rehabilitation Engineering Laboratory has played a significant part in developing and evaluating new stimulation neuroprostheses to regain function such as standing, grasping, and walking. We believe that a neuroprostheses to maintain stable sitting posture for individuals in wheelchairs would have clear benefits.

- We are developing a series of postural response investigations to identify the muscles and neuromuscular mechanisms necessary to stabilize the trunk while sitting, reaching, and standing.

Experimental Device

- To develop accurate models of postural stability in all directions about the torso, it is essential to devise a series of experiments with the following ***requirements***:
  - place the individual in a seated position
  - record the forces, moments, and centres-of-pressure (COP) under each side of the buttocks
  - provide perturbations in all directions (360 degrees) about the body
  - to finely control the perturbation profile (force, velocity, or position)
  - high level of safety (mechanical, software, and electrical systems)

- Based on manual tests of perturbing able-bodied individuals, the following ***specifications*** were made:
  - 8 actuators positioned in a circle (45 degrees apart) about the subject
  - linear actuators capable of :
    - peak force (tension): 600 N
    - peak velocity: 0.5 m/s
    - stroke length: 60 cm
  - each of the control profiles (position, velocity, and force) will be precisely modulated through feedback to follow:
    - step input
    - gaussian (bell)-shaped profiles
    - constant velocity (ramp-and-hold)
  - randomized perturbations

Data Recorded

- To record both the biomechanical and physiological aspects of postural control, a large number of measurements were performed:
  - actuator force profile
  - actuator displacement
  - body movement (Optotrak motion analysis system)
  - electromyography (surface electrodes)
  - forceplate recordings

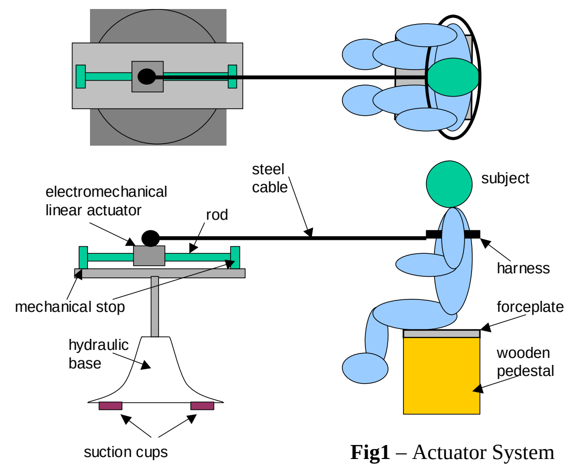

Fig1 – Actuator System

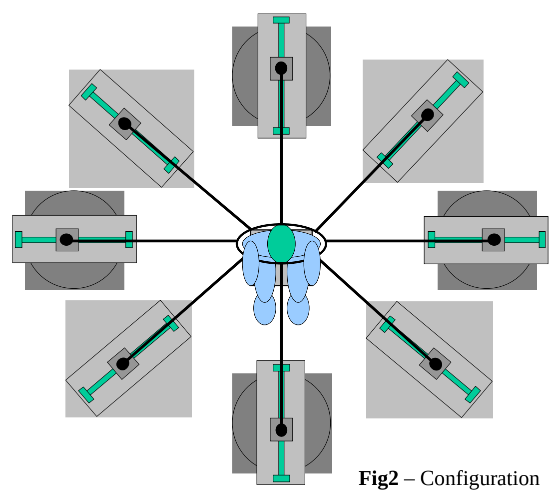

Fig2 – Configuration of Actuators

Conclusions

- Our development of a novel system of actuators will enable us to monitor biomechanical and physiological measures of the postural response to highly repeatable perturbations of the torso.
- These experiments will allow for us to identify the muscles and neuromuscular mechanisms necessary to stabilize the trunk.

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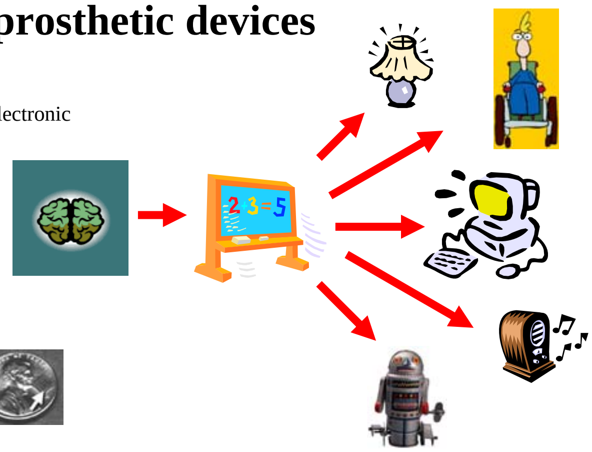
What is a Brain-Computer Interface?

César Márquez Chin[1,2] and Milos R. Popovic[1,2]

[1]Institute of Biomaterials and Biomedical Engineering, [2]Toronto Rehab Institute

A different face of neuroprosthetic devices



Definition

A brain-computer interface (BCI) is a device that uses signals from the brain to control electronic devices.

Relevance for Rehabilitation

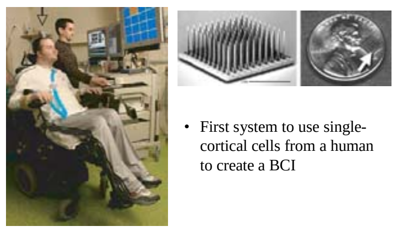
BCI have an enourmous potential to assist individuals with severe mobility impairments such as advanced stages of amyotrophic lateral sclerosis (ALS), brain stem stroke, spinal cord injury, and severe cerebral palsy.

The Graz BCI

- Distinguishes EEG generated with motor imagery

•*IEEE Trans Rehab Eng* (2000) **8** (2), pp. 222-226.

BrainGate



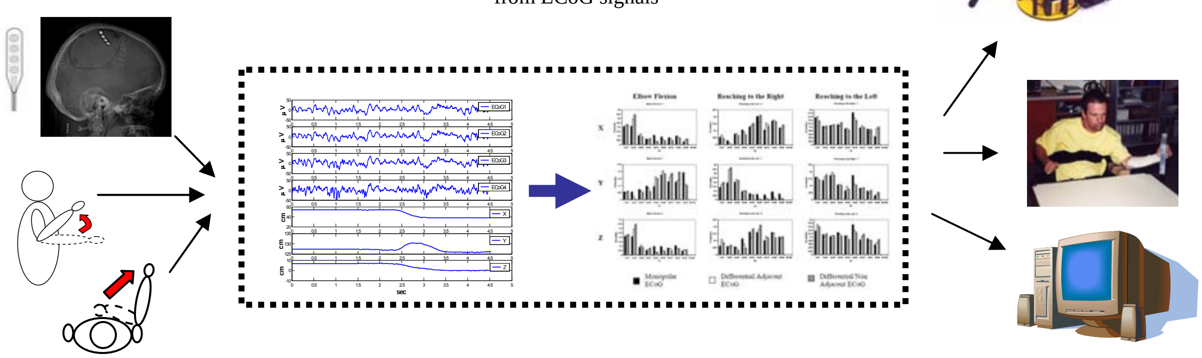
- First system to use single-cortical cells from a human to create a BCI

•*Nature* (2006) **442** (7099), pp. 164-171.

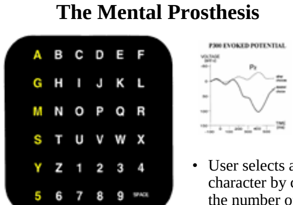
BCI at Toronto Rehab Lyndhurst

•The BCI project started in 2001 at the Rehabilitation Engineering Laboratory
•Dedicated to the understanding and development of BCI technology

•Long-term goals:
-To assess the capabilities of intracranial electrodes for the development of BCIs
-To explore the reconstruction of movement from ECoG signals



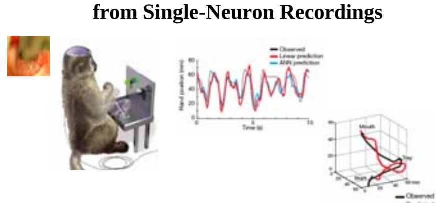
The Mental Prosthesis



- User selects a character by counting the number of times the desired character flashes.

•*IEEE Trans Rehab Eng* (2000) **8** (2), pp. 174-179.
•*Clin Neurophysiol* (2002) **113** (6), pp. 767-791.

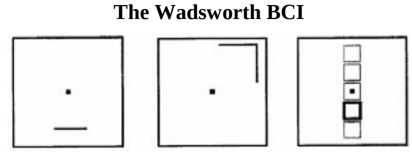
Prediction of Hand Trajectory from Single-Neuron Recordings



- Neuronal recordings of the motor cortex of monkeys to predict trajectory of movements

•*Nature* (200) **408** (6810), pp. 361-365.
•http://www.nicolelislab.net/NLNet/Load/index.htm

The Wadsworth BCI



- Self-regulation of μ or β rhythm amplitude controls a cursor.

•*IEEE Trans Rehab Eng* (200) **8**(2), pp. 222-226.

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Partners:

Identifying Movements From Cortical Signals

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Electrocorticographic Signals Reflect Specific Motor Tasks

Long –Term Goal

- To create a system that uses the electrical activity of the brain to control electronic devices: a brain-computer interface.

Purpose of the Study

- Explore the changes elicited by specific movements in the electrical activity of the primary motor cortex.
- Identify features in the electrocorticographic (ECoG) signals that show consistent behaviour in relationship to movement.

Background

- Different frequency bands in the electrical activity of the brain experience changes in amplitude during voluntary movement and/or preparation to perform such movement.
- These changes have been used to identify the occurrence of specific movements.
- The electrical activity of the brain can be recorded with intracranial electrodes (subdural electrodes) placed on the surface of the brain.
- The signals recorded with these electrodes are referred to as ECoG signals.

Materials and Methods

Participants
- Two individuals were implanted with subdural electrodes (four contacts) for the treatment of movement disorders: Parkinson's disease (PD) and Essential Tremor (ET). (See Fig. 1 and Table I)
- The site of implantation corresponded to the arm area of the primary motor cortex, confirmed with electrical stimulation.

Experimental Procedure
- Participants performed specific movements (see Table I).
- ECoG signals and position (X, Y, Z coordinates) of the hand were recorded simultaneously.

Table 1. Participants of this study.

Subject	Age / Gender	Diagnosis	Performed Movements
1	Male / 73	PD	•Elbow Flexion (EF) •Reaching to the Right (RTR) •Reaching to the Left (RTL)
2	Female / 65	ET	•Closing Hand (CH) •Reaching to the Right (RTR) •Reaching to the Left (RTL)

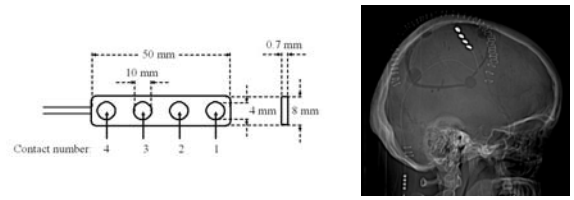

Figure 1. Subdural electrodes used for this study.

Analysis
- ECoG signals were analyzed in different configurations including monopolar as well as the difference between adjacent and non-adjacent electrodes.
- The time-frequency distribution was estimated for each ECoG signal with a resolution of 1.5 Hz
- The spectral components with the highest correlation values with each one of the kinematic dimensions were identified for each movement
- These frequency components were then grouped in a histogram
- The magnitude of each column represents the probability that a specific frequency band is correlated with a kinematic dimension of each movement.

Results

- The consistency of the distributions of correlated spectral components suggest that these features might be used to classify ECoG signals automatically.

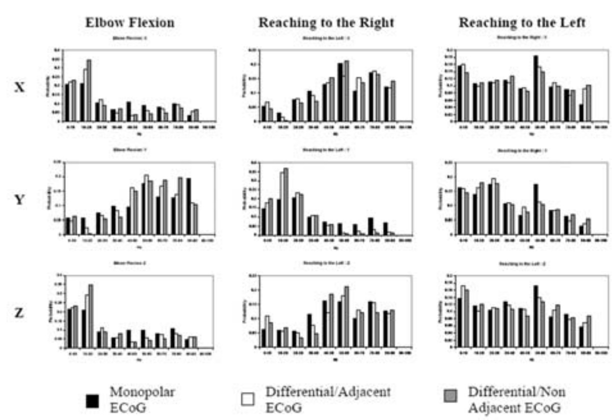

Figure 2. Average distribution of ECoG spectral components correlated with movement.

Discussion

- The consistency of the distributions of correlated spectral components suggest that it might be possible to use it as features to automatically classify ECoG signals.

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Partners:

Automatic Classification of ECoG Signals

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Toronto Rehabilitation Institute

Nearest-neighbour based brain-computer interfacing

Purpose

The purpose of this study was to create a system to identify automatically specific movements performed by an individual by analyzing the electrical activity of the motor cortex recorded with subdural electrodes.

Introduction

- Brain-compute interfaces (BCI) use brain signals to control electronic devices.
- A fundamental problem in BCI is classifying brain signals.
- Each classified brain signal represents a command that can be issued to an external device

Materials and Methods

Experiments

- Two individuals had subdural electrodes implanted over the primary motor cortex for clinical reasons independent from the development of this project.
- ECoG signals and limb kinematics were recorded while each subject performed movements with the upper limb contralateral to the site of implantation.
- Movements included elbow flexion (EF), closing of the hand (CH), reaching to targets placed 30 cm to the right (RTR) and left (RTL) from the subject's midline
- Each movement was repeated 30 times.

The following combination of signals were subjected to our analysis:

- Four monopolar (MP) signals: ECoG1, ECoG2, ECoG3, ECoG4
- Three differential adjacent (DA) signals resulting from subtracting potentials of the adjacent electrodes: ECoG1- ECoG2, ECoG2- ECoG3, ECoG3- ECoG4
- Three differential non-adjacent signals (DN) representing the difference between potentials of the non-adjacent electrodes: ECoG1- ECoG3, ECoG1- ECoG4, ECoG2- ECoG4

Feature Extraction

- The time-frequency distribution for each ECoG signal was estimated (resolution 1.5 Hz)
- A correlation coefficient was calculated between each time series representing the evolution in time of power in a spectral component over time and each of the six kinematic signals (X, Y, and Z, roll, yaw and pitch).
- The frequencies that were found to be highly correlated with movement were grouped in a histogram.
- The columns of the histograms represented different frequency ranges (e.g. 0-10 Hz, 10-20 Hz, …, 90-100 Hz)
- A different histogram was created for each one of the six kinematic coordinates of the executed movement.
- The magnitude of each bin in the histogram indicates the probability that the frequency it represents is correlated with the movement performed by the subject at the time of the recordings.

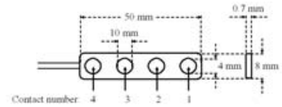


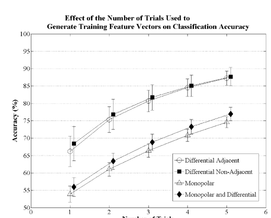
Figure 1. Subdural electrodes used for this study.

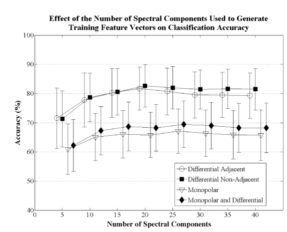
Classification Tests

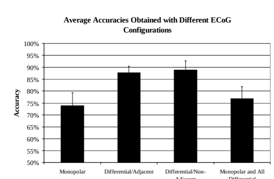
- We created a nearest neighbour classifier (NNC) to identify off-line which movement was performed by an individual during the recordings
- The magnitude of each column for each kinematic signal (X, Y, Z, roll, yaw and pitch) were concatenated to form a feature vector.
- The following three effects were statistically analyzed: 1) effect of the number of trials used to train the classifier; 2) effect of the number of frequency components used to train the classifier; 3) effect of the type of ECoG signals.

Results

- The best classification accuracy of our classifier was 97.3 %
- The worst accuracy was 38.7 %
- The classification accuracy was highly dependent on the number of trials used for the classifier, with the best accuracies obtained using at least four trials for training
- The number of spectral components used to create training vectors to train the classifier had a positive effect on the classification accuracy up to 20 spectral components, were no significant improvement was found
- The classification accuracy was also found to be dependent on the type of ECoG signals used: the highest classification accuracy was achieved differential (DN) signals while the worst results were obtained using monopolar signals (MP).







Conclusion

The best classification accuracy using the feature extraction method presented here can be obtained using differential signals, with five trials to train a NNC and 20 spectral components.

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SOURCE LOCALIZATION IN PERIPHERAL NERVES

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Toronto Rehabilitation Institute, Canada

Part I – Forward Problem

Introduction

- Benefits of precise source localization in peripheral nerves include:
  - –Improved mapping of neural pathways
  - –Insight into neural control mechanisms
  - –Improvements in neural prosthetics control
- We propose to achieve this using a source localization procedure similar to that used with EEG/MEG recordings.
- Using recordings from a multi-contact cuff electrode, try to localize current sources anywhere within a given cross-section of the nerve.
- Process divided into two parts: the forward and inverse problems.

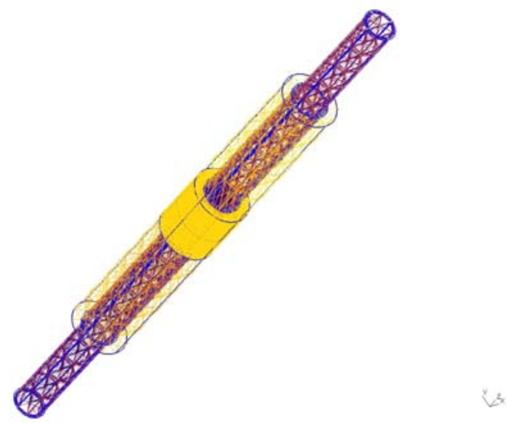

Fig1 – FE model of rat sciatic nerve

Role of the forward problem

- Solving the inverse problem requires a description of the relationship between the measurements and the current sources within the nerve.
- Because electric fields add linearly, this relationship can be expressed as shown in Equation 1.
- The purpose of the forward problem is to compute the matrix L in Equation 1. For each current location, compute the measurements produced by a source at that location.
- Use finite-element (FE) modeling

FE model of a rat sciatic nerve

- Model includes the following regions and objects:
  - –Endoneurium layer of the nerve
  - –Perineurium layer of the nerve
  - –Epineurium layer of the nerve
  - –Connective tissue layer between nerve and cuff
  - –Saline layer between nerve and cuff
  - –Cuff electrode
  - –Saline bath
- Mesh is made finer in the middle portion of the nerve. Coarser mesh outside the region of interest reduces the number of variables and makes the inverse problem better defined.
- FE model created with Gmsh using a geometric description of the nerve and the electrode (Fig1).
- Model allows us to compute the measurements produced by a current dipole at any mesh element (Fig2).

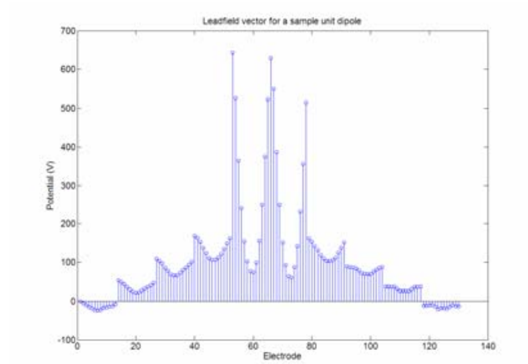

Fig2 – Influence of a unit current dipole on the electric potential at each electrode contact. The dipole corresponding to this vector is oriented in the x-direction and situated at the periphery of the endoneurium, halfway up the cuff electrode.

Equations

$$v = Lj + \varepsilon \qquad (1)$$

v: measurements
j: current dipole magnitudes
L: leadfield matrix
ε: noise

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SOURCE LOCALIZATION IN PERIPHERAL NERVES

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Part II – Inverse Problem

Introduction

- Goal is to localize activity anywhere within a segment of peripheral nerve, using recordings from a multi-contact cuff electrode.
- Using a small number of measurements (~100), we seek to determine the magnitude of a large number of current dipoles (~10,000): problem belongs to the class of ill-posed inverse problems.

Methods for solving the inverse problem

- The inverse problem (Equation 1) has an infinite number of solutions. In order to arrive at a unique solution, need to impose additional constraints based on our knowledge of the problem at hand:
  - –Anatomical constraints
  - –Biophysical constraints
  - –Temporal constraints
- The general solution of the problem is shown in Equation 2. Different properties of the solution can be emphasized depending on the choice of the matrix H (e.g. depth compensation, smoothness) [1].
- Can apply process iteratively, basing H on results of previous step, in order to obtain a more focal source estimate [2].

Example Using Rat Sciatic Nerve Model

- Two current dipoles are placed in one cross-section of the nerve (Fig1), and the forward problem is solved to obtain the measurements at the electrode contacts (no noise is applied for this example)
- Localization problem is solved using Equation 2, with H defined such that deeper sources are given more weight to compensate for their weaker influence on the measurements (Fig2). The estimate has correct peaks but is more distributed than the actual distribution.
- Solution in Fig2 is iteratively refined to obtain a focal source estimate (Fig 3). The locations of the current dipoles are now recovered perfectly.

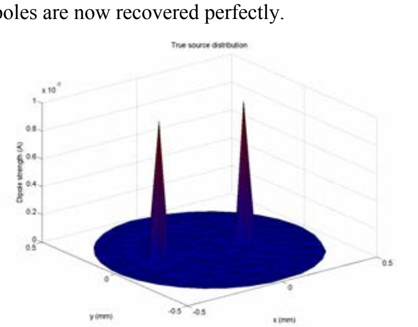

Fig1 – True source distribution (cross-section at z = 10mm)

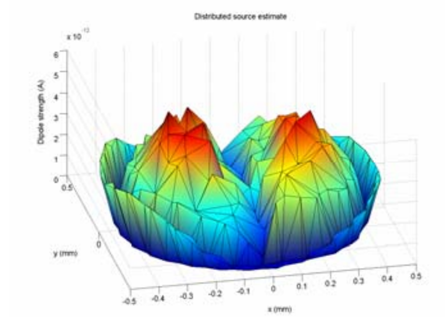

Fig2 – Distribution estimate using Eq. 2 (cross-section at z = 10mm)

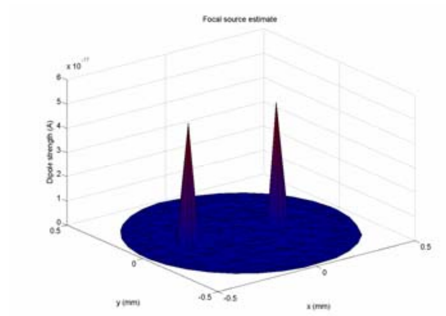

Fig3 – Focal estimate obtained by iteratively refining the results of Eq. 2 (cross-section at z = 10mm)

Equations

$$\hat{j} = \arg\min_{j}\{\left\|C_{\varepsilon}^{-0.5}(Lj - v)\right\|^{2} + \lambda\left\|Hj\right\|^{2}\} \quad (1)$$

$$\hat{j} = (H^{t}H)^{-1}L^{t}[L(H^{t}H)^{-1}L^{t} + \lambda C_{\varepsilon}]^{-1}v \quad (2)$$

j: true source distribution
ĵ: estimated source distribution
v: measurements
L: leadfield matrix
H: weight matrix
λ: regularization parameter
C_{ε}: noise covariance matrix

References

[1] R.D. Pascual-Marqui. *IJBEM*, 1(1):75-86, 1999
[2] I.F. Gorodnitsky et al. *Electroencephalogr. Clin. Neurophysiol.*, 95(4):231-251, 1995

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Partners:

PREDICTING FINGER FORCE FROM SURFACE EMG

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[1]Institute of Biomaterials and Biomedical Engineering, University of Toronto
[2]Toronto Rehabilitation Institute

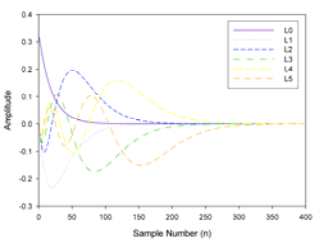
Applying the Laguerre Expansion Technique

Introduction

The purpose of this research is to model the relation between surface electromyography (EMG) and force produced by the index finger. Modelling the relationship has importance for developing interfaces that require direct input from the index finger.

Methods

Twelve able-bodied subjects produced isometric forces with the dominant index finger. The subjects tracked touch stimulations (0.25-9 Hz) provided on the opposite hand by producing a proportional finger force. Three trials were performed on each subject. The forces were measured using a load cell placed under the finger tip. Surface EMG was measured from the extensor indices muscle, extensor digitorum muscle, flexor digitorum superficialis muscle, and first dorsal interosseous muscle. The signals were sampled at 2000 Hz and down-sampled to 500 Hz.



Discrete Time Laguerre Function

$$L_j(n)=\alpha^{(n-j)/2}(1-\alpha)^{1/2}\sum_{k=0}^{j}(-1)^k\binom{n}{k}\binom{j}{k}\alpha^{j-k}(1-\alpha)^k$$

Parameters
- j: function order
- n: discrete time
- α: system memory ($0 < \alpha < 1$)

Fig. 3 – Six Laguerre Functions ($\alpha = 0.9$)

The EMG-force relation was modelled with six Laguerre functions using the Laguerre expansion technique (Fig. 1). The Laguerre functions were used to derive the 1st-order kernels (one per muscle) used in the model (Fig. 2). The models were developed using data from trial 1 and tested using data from trials 2 and 3. EMG-force models were also developed to predict forces before the finger forces occurred. This was done by time-advancing the force recordings with respect to the EMG signals.

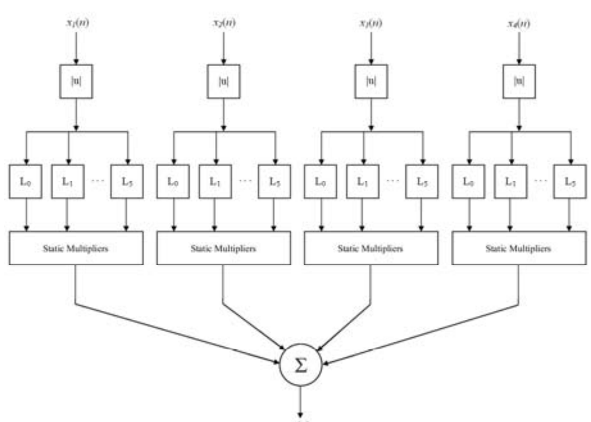


Fig. 2 – EMG-force model with inputs $x_i(n)$ and output $y(n)$

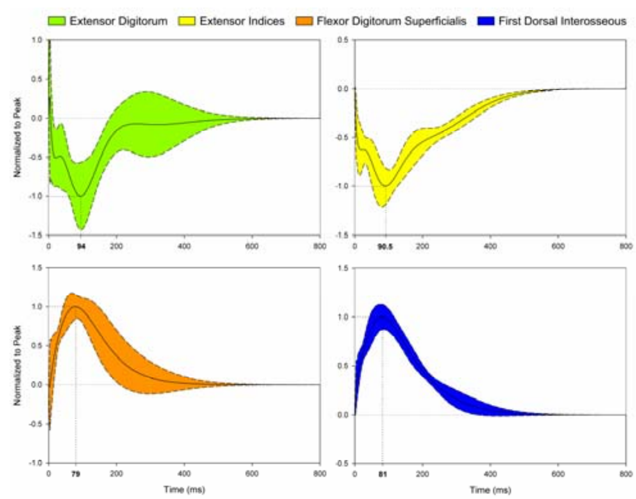
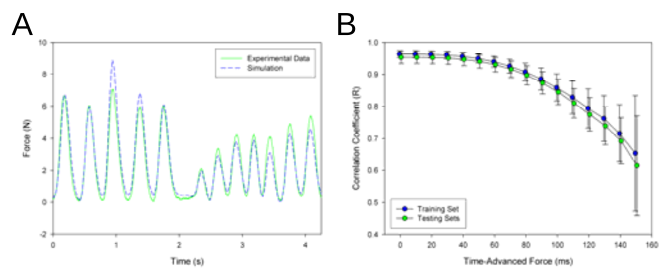


Fig. 3 – 1st-order kernels of the four surface EMG signals

Results

- The shape of the 1st-order kernels was consistent for all subjects and resembled the twitch response of skeletal muscle (Fig. 3).
- Predictions from the EMG-force models had a correlation coefficient $R = 0.95 \pm 0.02$ (Fig. 4B).
- The predictive models were able to predict finger force 50 ms before the forces were generated by the finger. These models provided a suitable strategy for predicting motion while keeping modelling error to a minimum ($R = 0.94 \pm 0.02$) (Fig. 4B).



Figs. 4 – Performance of the EMG-force models

Conclusion

- Finger force was predicted from surface EMG using the Laguerre expansion technique with a high level of correlation.
- The models account for the physiological relationship involved in contraction and can be easily implemented in real-time.
- The predictive EMG models (predicting 50 ms ahead) can help circumvent the electromechanical time delay involved in contraction and may be of importance for improving the response time involved in sending motor commands to an interface.

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Effect of Random Modulation of FES Parameters on Muscle Fatigue

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Lyndhurst Centre, Toronto Rehabilitation Institute
Institute of Biomaterials and Biomedical Engineering, University of Toronto

Introduction

- ***Objective:*** to reduce the rate of muscle fatigue in FES induced contractions of spinal cord injury (SCI) subjects by randomly modulating the three main FES parameters: current amplitude, frequency, and pulse width.
- ***Motivation:*** Many FES applications, such as standing and walking, are greatly limited due to the rapid rate of muscle fatigue in induced contractions.
- ***Previous research:*** This problem was addressed in a preliminary study [1] where improved endurance of isotonic contractions of the quadriceps was achieved in one complete SCI subject by randomly varying the interpulse interval (frequency).
- We extended the notion of stochastic modulation to amplitude and pulse width and extended testing to include multiple subjects.

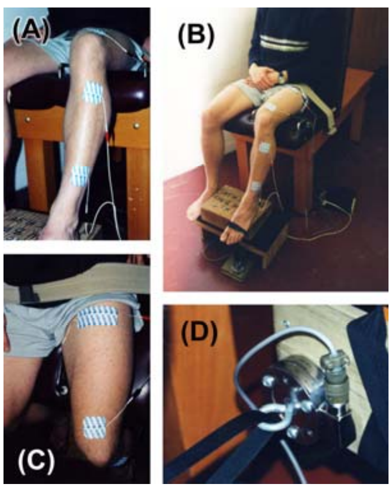

Figure 1 –Electrode placement on the (A) tibialis anterior and (C) quadriceps. (B) shows the upright sitting position of each subject and (D) shows the uniaxial, strain gauge-based load cell used to measure force (Honeywell Sensotec, Columbus, Ohio, USA).

Methods

- ***Subjects:*** 7 individuals with complete SCI participated.
- ***Experimental setup:*** While subjects were seated, FES was applied to the quadriceps and tibialis anterior muscles bilaterally using surface electrodes (see Figure 1) and isometric knee extension and dorsiflexion forces were measured by a load cell.
- ***Trials:*** To each muscle, four modes of FES were applied in random order with a 10-minute rest period between applications: (1) constant stimulation, (2) randomized current amplitude, (3) randomized frequency, and (4) randomized pulse width. Stimulation parameters were stochastically modulated in a range of +/- 15% using a uniform probability distribution.
- ***Outcomes:*** Force time series were collected for an equivalent of 22 muscles. The time for muscle fatigue to occur was chosen to be the time from the first onset of stimulation (time 0) until the isometric force dropped to –3dB of the full value of the maximum force measured. The force time integral (FTI) from time 0 until the –3dB time, was also calculated for each trial. Differences between the four modes of stimulation were tested using repeated means ANOVA with a significance level of $p < 0.05$.

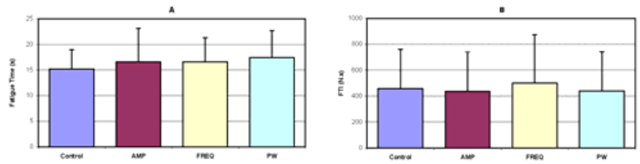

Figure 2 – Average values over all muscles of fatigue time (A) and FTI (B) for the four treatment groups.

Results

- ***Fatigue time:*** Although random modulation of the amplitude, frequency and pulse width produced slightly higher fatigue time measurements than the control trials, the differences were not significant ($p = 0.1911$). See Figure 2A.
- ***Fatigue-time integral:*** There was also no significant effect of random modulation on FTI ($p = 0.7649$). See Figure 2B.
- ***Trial order:*** The FTI measurements were highly dependent on the trial order ($p < 0.0001$), while the fatigue time measurements appeared to be independent of trial order ($p = 0.1354$). Maximum force was seen to consistently decrease from one trial to the next ($p < 0.0001$).
- ***Correlation of maximum force with outcome measures:*** As shown in Figure 3A, there was no apparent correlation with the maximum force recorded in each trail and the corresponding fatigue time measurement. However, as shown in Figure 3B, the maximum force correlated very closely with FTI.

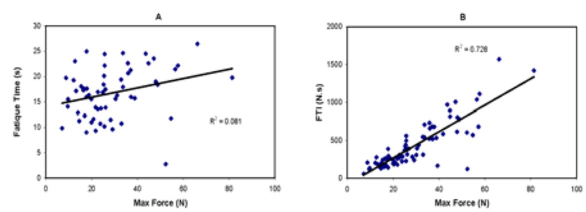

Figure 3 – Average values over all muscles of fatigue time (A) and FTI (B) for the four treatment groups.

Conclusions

- ***Observed effect of random modulation:*** Random modulation of frequency, amplitude and pulse width during stimulation did not appear to have any effect on the fatigue rate of isometric contractions of the quadriceps and tibialis anterior muscle of subjects with complete SCI. Therefore, we cannot conclude that these are viable techniques of fatigue reduction in practice.
- ***Rest period between trials:*** The order of the trials significantly affected the maximum force and the FTI measurements but not the fatigue time measurements. We conclude that a rest period of ten minutes between stimulation trials is insufficient to allow full recovery of muscle strength. Fatigue time may be a sufficiently robust measure of fatigue that does not require a full rest.

Reference

[1] Graupe et al. *Neurol. Res.* 22:703-704, 2000.

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Effects of isometric FES and dynamic FES on cardio-vascular parameters on an active tilt-table stepper

Thrasher TA[1,2], Keller T[3,4], Lawrence M[4], Popovic MR[1,2]

[1]University of Toronto, Canada [2]Toronto Rehabilitation Institute, Canada

[3]University Hospital Balgrist, Zurich, Switzerland [4]ETH-Zurich, Switzerland

Introduction

- Orthostatic hypotension prevents approximately 64% of acute SCI patients from participating in activities involving mobilization [1].
- Functional Electrical Stimulation (FES) of the lower extremities can increase blood pressure and improve circulation, however it is not known if these effects can be maintained for extended periods of time. Previous studies have only analyzed isometric stimulation.
- We evaluated the benefit of applying FES in synchronous with movement to produce concentric contractions. FES was combined with active tilt-table therapy and tested on healthy normals.

Methods

- Subjects: 16 young, healthy adults (Mean ± S.D.; Age: 26 ± 4.1, Height 1.72 ± 0.07 m, Weight: 65.4 ± 9.1 kg)
- Subjects were put on an active tilt-table stepper, which provided robotic-assisted stepping movements to the legs. This device was developed by University Hospital Balgrist, Zurich in collaboration with University Hospital Heidelberg, Germany [2].
- Surface FES was synchronized with the actuators and delivered to the quadriceps and gastrocnemius/soleus bilaterally via a 4-channel stimulator (Compex Motion, Switzerland).
- Randomized cross-over design with three test conditions:
  - **A. No FES, subject immobilized**
  - **B. Isometric FES, subject immobilized**
  - **C. Dynamic FES, subject's leg moved by actuators**
- In all trials, subjects were supine for 10 minutes, and then tilted to 80º from horizontal for 30 minutes or until syncope occurred.
- BP and HR were measured continuously using a tonometric wrist monitor (Colin Medical Instruments Corp., Japan).

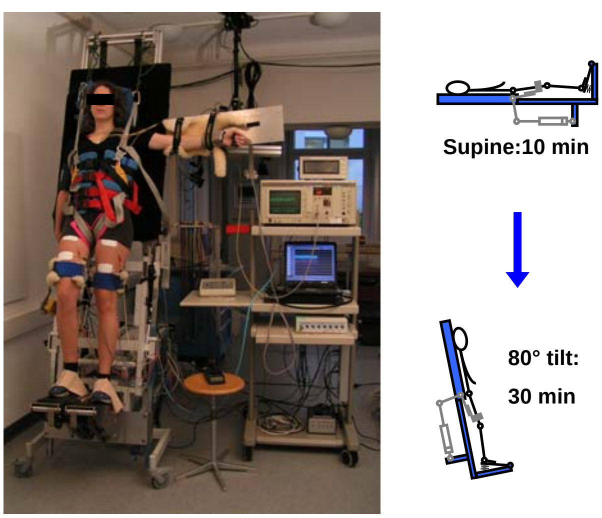

Fig1 – Experimental setup. Subjects remained supine for 10 minutes, then were tilted 80º from horizontal for up to 30 minutes. BP, HR and EKG were recorded throughout all trials.

Results

- Six of the 16 subjects experienced syncope or pre-syncope during at least one trial. The duration of the trials for these six subjects is shown in Figure 2. No one experienced syncope during Trial C.

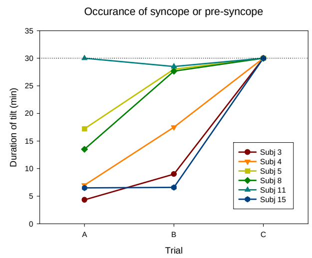

Fig2 – Subjects who had syncope or pre-syncope during static tilt

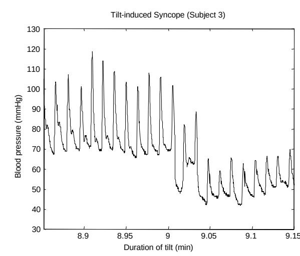

Fig3 – Sudden BP drop occurs in a only a few heartbeats

- Maximum HR and BP were significantly lower in Trial C (p=0.01).
- Isometric FES did not prevent tilt-induced syncope, but it delayed it an average of 6.5 minutes.
- Variable extensor tone was observed when FES not used (Fig 5).

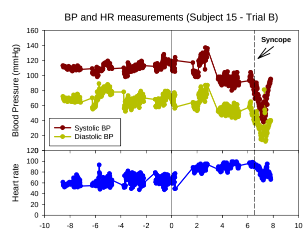

Fig4 – HR increase at tilt and gradual drop in BP seen in all subjects who had syncope.

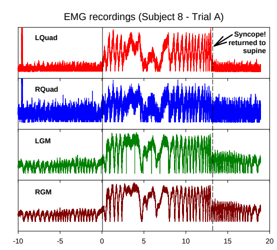

Fig5 – EMG shows variable extensor tone during tilt.

Discussion

- Our results show that although isometric FES increased systolic and diastolic BP at onset, it was not able to prevent sudden decreases after several minutes of head-up tilt.
- When FES was combined with movement, BP was maintained for the full 30 minutes. This suggests that the amount of lower limb muscle activity induced by isometric FES of the gastrocnemius-soleus and quadriceps was insufficient to prevent the sudden loss of venous tone that occurs in some able-bodied individuals.

References

1. Illman A et al. The prevalence of orthostatic hypotension during physiotherapy treatment in patients with an acute spinal cord injury. *Spinal Cord*. 2000 Dec;38(12):741-7.
2. Rupp R, Eberhard S, Schreier R, Colombo G. Reha-Stepper locomotion therapy in early rehabilitation of paraplegic patients. *Biomed Tech* (Berl). 2002;47 Suppl 1 Pt 2:708-11.

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HUMAN RESPONSES TO VIBRATION THERAPY

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Towards a Procedure for Maintaining Bone Density via Mechanical Stimulation

Introduction

Vibration therapy has been used at the Toronto Rehabilitation Institute in an attempt to maintain bone density in spinal cord injured patients. Several studies have shown that mechanical strain can be used to regulate bone density [1]. Clinical results have also demonstrated that only the application of certain types of strain fields will stimulate increases in bone density [2]. In this study, experiments were performed to ascertain the accelerations present in the lower body during vibration therapy with subjects of different body types. This clinical data is being used to develop a biomechanical model of the body during vibration therapy. The goal of this research is to predict the strain fields present in bones during vibration therapy, and thus assist in the development of an effective procedure for maintaining bone density using vibration therapy.

Modeling

- Equations were developed treating the body as a system of four rigid bodies
- The bodies were connected with linear springs and dampers
- Both feet, both shanks, and both thighs were each represented by a rigid body
- The trunk, arms, and head were also represented by a single body
- Equations of motion were obtained using Lagrange's Equation
- Simulations were done in Matlab's Simulink environment

Experiments

- Two spinal cord injured and two healthy subjects of similar body types participated
- Tri-Axial accelerometers were attached to the base plate, shank, thigh, and hip of each subject
- Data was recorded using two different vibration therapy devices, and with and without a constraining device.
- Each vibrating device was designed to apply perturbations in a single direction. A vertically vibrating device used in previous studies and a horizontally vibrating device that is part of a new study were used to apply vibrations to the subjects.

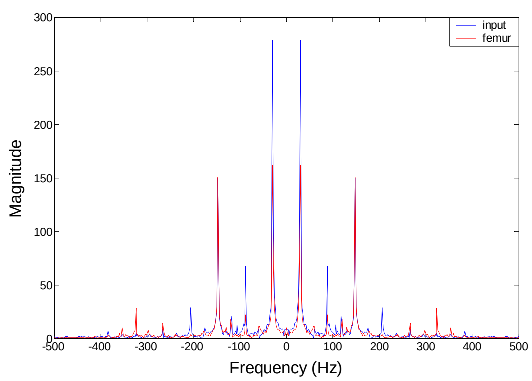

Fig. 1 – Frequency Content of Input and Scaled Femur Signals (x-direction)

Results

Both experimental and simulation results were generally as expected. Analysis of the data showed the following:

- There were negligible differences in the mechanical properties of SCI and healthy patients
- Accelerations on the same order of magnitude were present in all three directions during therapy with both devices
- The constraining device caused either an increase or decrease in base plate to body segment acceleration ratios, depending where the subject was constrained. Taller subjects had higher acceleration ratios during therapy with the constraining device because it supported their upper body more, and shorter subjects had lower acceleration ratios during therapy with the constraining device because it did not support their upper body.
- FFT analysis showed linearity of human subjects for small amplitude perturbations in this configuration (see Fig.1).
- The model matched experimental data reasonably well (see Fig. 2).

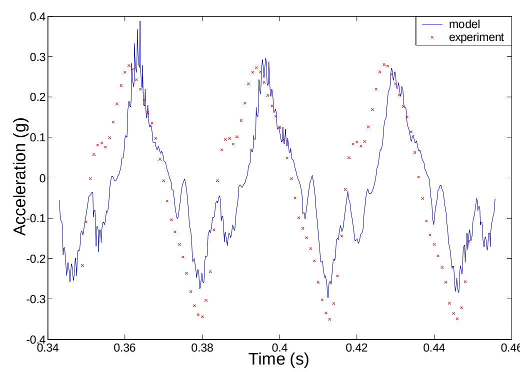

Fig. 2 – Acceleration of Femur, Model and Experimental Results

Acknowledgements

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References

[1] J. C. Fritton, C. T. Rubin, Y.-X. Qin, and K. J. McLeod, "Whole-Body Vibration in the Skeleton: Development of a Resonance-Based Testing Device," *Annals of Biomedical Engineering*, **25**: 831-839, 1997.

[2] C. T. Rubin and L. E. Layton, "Osteoregulatory Nature of Mechanical Stimuli: Function a Determinant for Adaptive Remodeling in Bone," *Journal of Orthopaedic Research*, **5**: 300-310, 1987.

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Arterial Stiffness in Persons With Spinal Cord Injury: A Pilot Study

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Background and Research questions of the project

Why do we focus on Arterial Stiffness?

• The stiffening in central or cardiothoracic arteries is an emerging risk factor for coronary artery disease (CAD). Decreases in the elastic properties of arteries reduce the buffering capacity of the arteries, leading to increased pulse pressure, aortic impedance, and left ventricular wall tension, all of which augment the workload of the heart, thereby increasing the CAD risk.

What is Pulse Wave Velocity (PWV)?

• PWV is the speed of the blood pressure wave of the arterial system. PWV is affected by the elasticity and other properties of the artery (eg., wall thickness and blood density). Thus, PWV correlates with arterial distensibility and stiffness. Higher velocity corresponds to higher arterial stiffness and lower distensibility.

• PWV is a noninvasive, simple and reproducible measurement.

What are the determinants of PWV?

• Several risk factors such as aging, obesity, poor cardiorespiratotory fitness and low physical activity, have been identified as determinants of PWV in able body population.

Why do we need to know about PWV in SCI patients?

• These risk factors are also well-known characteristics of individuals with spinal cord injury (SCI). Thus, it is hypothesized that individuals with SCI have high risk of adverse vascular health characterized by increasing PWV.

What are aims of the project?

• To test the hypothesis that SCI individuals have greater arterial stiffness (PWV) than age-matched non-SCI peers,

• To compare PWV in SCI individuals with that of the persons with CAD

• To investigate the relationship between cardiorespiratory fitness, body composition and PWV in these populations.

Methods

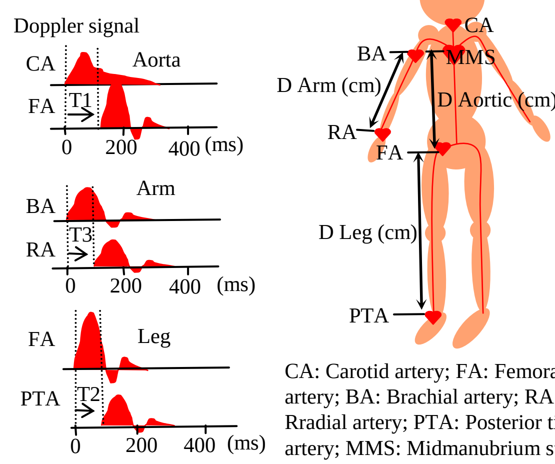
♦**Subjects** (Age: 35-85 years):

• 6 with traumatic SCI with CAD
• 6 with traumatic SCI without CAD
• 6 with CAD not participating in exercise cardiac rehab
• 6 with CAD participating in exercise cardiac rehabilitation
• 6 healthy age and gender matched controls

♦**Measurement:**

• Body composition using Dual energy X-ray Absorputiometry
  - Measurements: Adipose tissue mass, lean soft tissue mass, waist circumference
  - Region: Arms, legs, and trunk
• Cardiorespiratory fitness during arm ergometry
  - Measurements: Maximum oxygen consumption
• PWV using Noninvasive two identical transcutaneous Doppler flow meters (Smartdop50, Hadeco,Inc., Kanagawa, Japan)
  - Measurements: PWV
    * Probes were placed at two vascular landmarks and their blood flows were simultaneously measured by two trained technicians.
    * The pulse wave (1) between the carotid and the femoral artery (aortic PWV); (2) between the femoral and posterior tibial artery (leg PWV); and (3) between the brachial and radial artery (arm PWV) were assessed.
    * The distance traveled by the pulse wave were measured over the surface of the body using a metal tape measure.

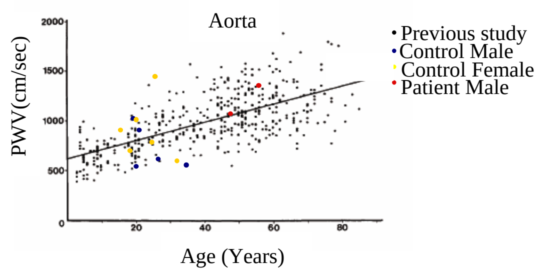
- Calculation of PVW



CA: Carotid artery; FA: Femoral artery; BA: Brachial artery; RA: Rradial artery; PTA: Posterior tibial artery; MMS: Midmanubrium sterni

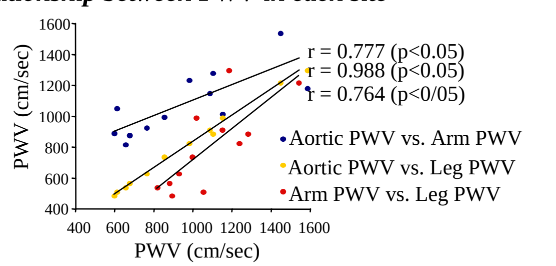
• Aortic PWV (cm/sec)= D Aortic / T1
  * *We assum that the PWV between CA and MMS and between MMS and FA are same.*
• Arm PWV (cm/sec) = D Arm / T3
• Leg PWV (cm/sec) = D Leg / T2

PWV in each subject



• So far, PWV data of 6 male control, 5 female control and 2 male patients (complete paraplegias) were collected as preliminary data.
• Our data agrees with the previous study (Avolio et al. 1985).

Relationship between PWV in each site



• There were significant relations between PWV in each site. The result shows the measured PWV represents the subjects' arterial property.

Conclusion

• PWV is the reproducible parameter and represents the patients' arterial property, which is usable in assessing the CAD risk.
• PWVs in two measured patients in this study were in the range of the healthy persons. Further measurements are required.

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2nd National Spinal Cord Injury Conference, October 26-28, 2006 in Toronto, Canada

Novel home-based care equipment for preventing the secondary disorders in paraplegic persons

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1. Background

Motor paralysis

Isolation from gait and standing

Immobilization
- reduced joint motion
- reduced mechanical (gravitational) stress
- Hypo-circulation in the paralyzed area

Miss-use syndrome
- Atrophy
- Loss of BMD
- Joint contracture

Secondary disorders
- Pressure sores
- Deep venouse thrombosis

Preventive measures

Muscle stretching and passive joint motion are usually prescribed.

Daily regimen is most effective method for inactivity induced deteriorations.

Objectives:

1. To develop a rehabilitation device for preventing deterioration of the neuromuscular system affected by the spinal cord injury (SCI)

Concepts for developing the device.
1) Easy-to-use, and lightweight for home use
2) High affinity with the wheelchair
3) Physiological evidenced based setting for motion

2. To test the feasibility of the device through physiological evaluation.

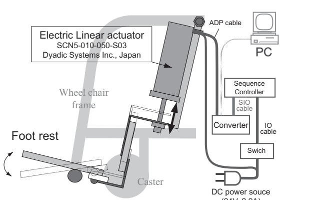
cf. Reflexive muscluar activity can be induced during passive motion (Kawashima et al. J Neurophysiol 2005)
Muscle oxygenation level is altered during motion (Kawashima et al. Med Sci Sports Exers 2005)

2. Foot Rest Device

We developed a motor assistive devise by improving the foot cast on the wheelchair. By the use of a portion of the wheelchair, the patients need not to transfer any other devise.

We intended to make daily application of the passive ankle movements by the use of motorized foot cast.

motorization



A Linear actuator

Stroke length	50 mm
Maximal velocity	400 mm/sec
Minimum velocity	10 mm/sec
Size (demension)	45-202-68 mm
Weight	1.1 kg
Accuracy of position	±0.01 mm

B Foot rest

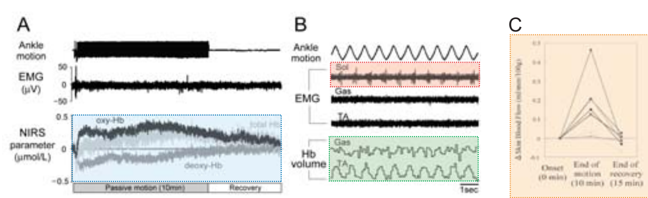
Range of Motion	36 deg (88 deg - 124 deg)
Maximal velocity	288 deg/sec
Minimum velocity	0.72 deg/sec

The proposed device used a pair of linear electric actuators attached to the wheelchair at one end and the backside of the foot rest at the other end. The vertical motion of the linear actuator translated into rotational motion about the ankle joint.

3. Physiological Assessments

To test the feasibility of the developed device, electromyographic activity, muscle oxygenation, muscular mechanical changes and skin blood flow in the paralyzed muscles were evaluated in twenty individuals with motor complete SCI.

Methods
- Subject: Twenty SCI patients (Complete T5-12)
- Protocol: alternate ankle passive motion with 24 degrees range of motion and 0.7 Hz motion frequency for 10 minutes
- Measurement: EMG (Sol, MGas, TA)
  NIRS (MGas, TA)
  Skin Blood Flow (lower limb)



Subject	Sex	Age (yr)	Duration of Paraplegia (yr)	ASIA Grade	Modified Ashworth Scale	EMG	Spasticity	Δ Skin Blood Flow (ml/min/100g)
S1	M	35	5	A	1+	-	-	-
S2	M	20	3	A	1	+	+	-
S3	M	30	3	A	2	-	-	-
S4	M	23	5	A	2	++	++	-
S5	F	35	20	A	2	+	++	0.21
S6	F	42	24	A	1+	+	-	0.46
S7	M	21	3	B	1+	-	-	0.13
S8	M	46	6	A	2	++	++	0.01
S9	M	22	3	A	1	-	-	0.15
S10	M	38	20	A	0	-	-	0.20
S11	M	33	6	A	1	+	-	0.03
S12	M	35	6	A	0	-	-	0.12

During passive ankle motion, (1) rhythmical muscular activity in the anlke extensor muscle, (2) an alteration in the muscle oxygenation level, and (3) periodic muscle mechanical (stretch-shortening) changes were induced. (4) An enhancement of the blood flow in the paralyzed area was also observed.

4. Summary and Conclusion

- A foot rest on the wheelchair was motorized for generating passive motion in the paralyzed ankle joint.
- The device developed in this study has a very simple structure and is very effective in generating rhythmic movement in the paralyzed ankle.
- The results obtained from the physiological assessments suggest that the passive ankle motion could enhance not only the muscle neuronal activity but also the muscle circulatory conditions.
- The developed device seems to be a useful and effective rehabilitation alternative for individuals with SCI.

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Simulation of Elbow Muscle Force from EMG

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***University of Toronto – Institute for Biomaterials and Biomedical Engineering**

Introduction

- Modelling and simulation of muscle recruitment has potential application to simulating functional outcomes of: functional electrical stimulation, orthopaedic surgery, and prosthetic limbs.
- The estimation of muscle force from surface electromyographic (sEMG) signals has been a significant biomechanical challenge of properly characterizing the dynamic behaviour of muscle and incorporating it into a mathematical model driven with sEMG data, and the second is to experimental validation of this behaviour.

Model

- Muscle's dynamic (force) characteristic is non-linearly based upon the muscle and tendon state (length, velocity, present force) and geometric properties.
- A new computational, lumped-parameter Hill-type muscle model was developed in Simulink (Mathworks Inc.) (see Fig1) incorporating:
  - series-elastic element (tendon modelling)
  - parallel-elastic element (passive muscle stiffness)
  - active element (active muscle contraction)
  - force-length relationship
  - force-velocity relationship
  - pennation angle
  - optimal fibre length
  - optimal fibre length/activation relationship
  - tendon slack length
  - electromechanical delay
  - maximum isometric muscle force
  - EMG-activation transformations

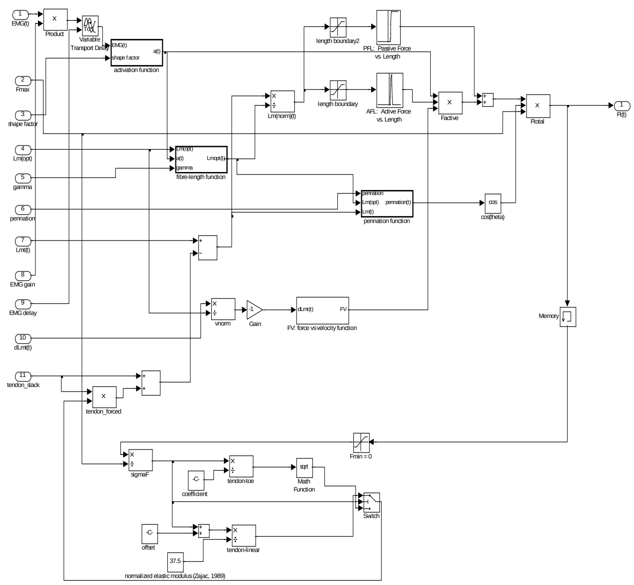

Fig1 – muscle model in simulink

Model Validation

- Elbow function was chosen as a task to both record and model. A rehabilitation dynamometer (Biodex II, Biodex Corp. Shirley, NY) allowed for restriction of limb movement to achieve 14 testing conditions:
  - isokinetic @ 30 / 90 / 180 / 300 deg/sec
  - isotonic @ 13.56 / 27.12 / 54.23 N.m
  - eccentric (54.23 N.m) @ 30 / 60 / 120 deg/sec
  - isometric @ 0 / -45 / -90 deg
- Six able-bodied male subjects voluntarily maximally flexed and extended there elbow through the complete range-of-motion for 10 seconds while the dynamometer modulated the resistance and mode of contraction (see Fig2).
- EMG signals of six muscles were recorded: biceps brachii, brachialis, brachioradialis, and triceps brachii (medial, intermediate, and lateral).
- Dynamometer data was recorded: joint angle, angular velocity, and joint moment.
- A maximal trial @ 30 deg/sec was used to normalize the EMG signals.

Parameter Optimization

- The only means of evaluating the muscle model was through the creation of a 6-muscle models within an elbow model incorporating:
  - muscle length and moment-arm as a function of joint angle
  - muscle-specific parameters
  - global muscle parameters
  - driven with rectified/normalized surface-EMG signals
- While a single set of muscle parameters for all muscles and all subjects would be desirable, it may not mimic reality. Global optimization was to identify sets of parameters for each individual that would provide the minimum difference between simulated and recorded joint moments (see Fig3).

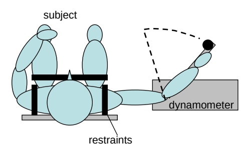

Fig2 – experimental setup (top view)

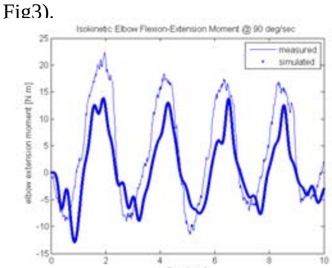

Fig3 – simulation results example

Results

- Model estimates of net joint moments compared to experimentally measured moments across 14 experimental conditions indicated a strong mean correlation across all trials (r = 0.83) and acceptable joint moment error (RMSE = 20.06%). The elbow model's predictive performance varied across contraction type. The new muscle model and its validation demonstrate the application of electromyographic and kinematic data to accurately predict muscle forces/moments for dynamic activity

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