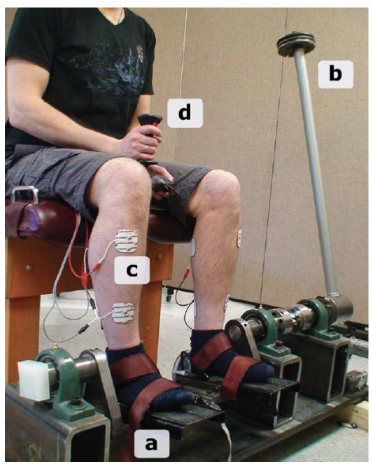
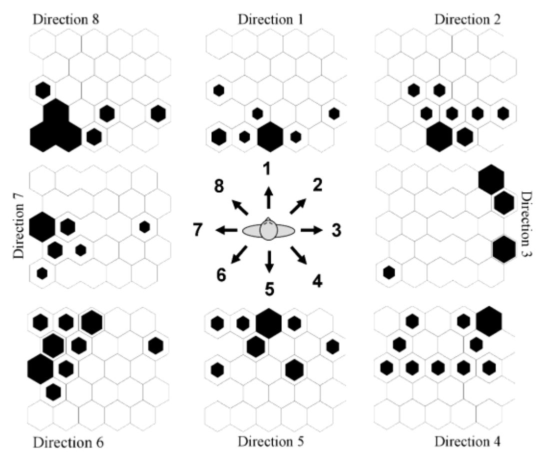
Rehabilitation Engineering Laboratory

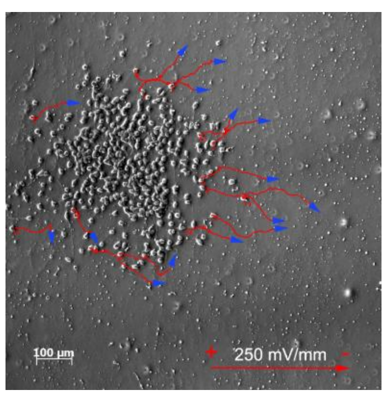
Institute of Biomaterials & Biomedical Engineering
UNIVERSITY OF TORONTO

Toronto Rehab
Everything Humanly Possible

REL
RehabilitationEngineeringLaboratory







Toronto Rehab, Lyndhurst Centre
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www.toronto-fes.ca

About the Rehabilitation Engineering Lab

History

The Rehabilitation Engineering Laboratory was established in 2001 at the Lyndhurst Centre of Toronto Rehab. In 2006 and in 2009, the laboratory underwent two major renovations, quadrupling the amount of space and equipment available for personnel and experiments.

What We Do

We develop advanced technologies for spinal cord injury (SCI) and stroke rehabilitation. These include brain/machine interfaces, assessment tools for determining an individual's level of function, and rehabilitation techniques for restoring walking, sitting, standing, reaching, and grasping ability. We also design neuroprosthetic systems to assist individuals with tasks such as walking, reaching, 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 to re-train weak muscles and the injured central nervous system.

Accomplishments – 2010-2012

- Completed a number of randomized control trials with stroke and SCI individuals involving FES therapies
- Developed advanced communication and stimulation technologies for neuroimplants
- Provided FES therapy to more than 50 individuals with SCI or stroke
- Published 41 peer-reviewed journal papers
- Obtained more than $1.1M in direct funding and collaborated on grants which total funding exceeded $0.6M
- Trained 8 postdoctoral fellows, and 22 graduate students in SCI and stroke research, who obtained an additional $1M in funding

Want to Get Involved?

We're always looking for participant for our studies, volunteers to help us with the experiments, and students and research collaborators. If you would like to join us, please feel free to contact us at 416-597-3422, Ext. 6206.

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Research at the Rehabilitation Engineering Laboratory

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, significantly improving their independence in activities of daily living. People who have SCI at C3-C7 level or severe stroke have used this system as a rehabilitation tool to assist in retraining voluntary 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 studies showed a significant improvement in walking speed and/or a reduction in the use of assistive devices for walking after using the neuroprosthesis. In this application the neuroprosthesis for walking is used as a short-term intervention for improving voluntary walking function.

Neuroprosthesis for Sitting

Trunk instability is a major problem for many people with SCI, affecting 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.

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.

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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-machine interfacing, which investigates the relationship between intended arm movement and electroencephalogram (EEG) or electrocorticographic (ECoG) signals from the motor cortex of the brain.

Novel Neuroprosthesis

People with spinal cord injury have impaired movements of their arms and/or legs, due to paralyzed muscles. Rehabilitation using electrical muscle stimulation is very advantageous for this patient population. However, one aspect that often limits the use of electrical stimulation is the rapid onset of muscle fatigue. One can potentially explain the muscle fatigue following electrical stimulation by the fact that electrical stimulation contracts muscle fibers simultaneously and that electrical stimulation is unable to contract all the fibers within the muscle. In the current project, we propose a new stimulation method that would activate most of the muscle fibers in a regulated cyclic pattern.

Equipment

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

- Compex II stimulators
- Body weight support treadmills
- Force plates
- Polhemus motion capture system
- Optotrack dual camera motion capture systems
- 6-camera Raptor Motion Analysis system
- ARMEO and ReJouce systems for upper limb rehabilitation
- ERIGO tilt table with motorized leg movement
- Electromagnetically shielded room for EMG and EEG measurements
- Vibration platforms
- REL-PAPPS perturbation system
- Biodex System 3
- Various EMG and EEG measurement systems

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Our People

Principal Investigators

- **Dr. Milos R. Popovic** (biomedical engineering), **Head of the Laboratory**, Toronto Rehab Chair in Spinal Cord Injury Research, Senior Scientist, and Professor
- **Dr. Cesar Marquez-Chin** (biomedical engineering), Research Scientist
- **Dr. Kei Masani** (exercise physiology), Research Scientist and Assistant Professor
- **Dr. Mary Nagai** (surgery), Research Scientist and Assistant Professor

Postdoctoral Fellows

- **Dr. Elias Daniel Guestrin** (biomedical engineering), Postdoctoral Fellow
- **Dr. Santa Concepción Huerta Olivares** (electronics), Postdoctoral Fellow
- **Dr. Reza Javaheri** (electronics), Postdoctoral Fellow
- **Dr. Masae Miyatani** (exercise physiology), Postdoctoral Fellow
- **Dr. Aravind Kumar Namasivayam** (speech/language), Postdoctoral Fellow
- **Dr. Hossein Rouhani** (biomedical engineering), Postdoctoral Fellow
- **Dr. Dimitry Sayenko** (space medicine), Postdoctoral Fellow
- **Dr. Jose Zariffa** (biomedical engineering), Postdoctoral Fellow

Graduate Students

- **Davide Agnello**, MASc student
- **Kathryn Atwell**, MASc student
- **Sara Ayatollahzadeh**, MEng student
- **Rob Babona-Pilipos**, PhD student
- **Martha Gabriela Garcia-Garcia**, MASc student
- **Diane Kostka**, MHSc student
- **Meredith Kuipers**, MSc student
- **Cheryl Lynch**, PhD student
- **Andresa R. Marinho**, PhD student
- **Steve McGie**, PhD student
- **Bahar Memarian**, MASc student
- **Matija Milosevic**, PhD student
- **Michael Same**, MASc student
- **Egor Sanin**, MASc student
- **Miyuki Tsukimoto**, MASc student
- **Albert Vette**, PhD student
- **Noel Wu**, MASc student
- **Takashi Yoshida**, PhD student

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Our People

International Research Placements

- **Stefanie Buley** - University of Applied Science Senftenberg, Germany
- **Ubaldo Garcia** - Universidad Iberoamericana, Mexico
- **Marisol Martinez** - Universidad Iberoamericana, Mexico
- **Karen Rodriguez** - Universidad Iberoamericana, Mexico
- **Catalina Villa Saenz** - School of Engineering of Antioquia, Columbia

Support Staff

- **Shaghayegh Bagher**, Research Engineer
- **Zina Bezruk**, Administrative Assistant
- **Betty Chan**, Grants & Accounts Coordinator
- **Naaz Desai**, REL Manager, Research Coordinator and Physiotherapist
- **Jennifer Holmes**, Occupational Therapist
- **Anne Hu**, Physiotherapist
- **Lorna Lo**, Occupational Therapist
- **Esther Oostdyk**, Secretary
- **Abdulazim Rashidi**, Research Engineer
- **Egor Sanin**, Research Engineer
- **Dr. Vera Zivanovic**, Research Coordinator

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Awards and Distinctions

Robert Babona-Pilipos

2010	Scientific Day 2010 Best "Lightning Round" Presentation	Institute of Biomaterials and Biomedical Engineering, University of Toronto
2011	St. George's Society Graduate Scholarship	University of Toronto
2011	CREATE – CARE Graduate Scholarship	Natural Sciences and Engineering Research Council of Canada
2011	Best Paper Award 2010-2011 in the category "Biomaterials, Tissue Engineering and Regenerative Medicine	Institute of Biomaterials and Biomedical Engineering, University of Toronto

Zachary Dulberg

2011	Summer Student Scholarship	Natural Sciences and Engineering Research Council of Canada

Martha Gabriela Garcia-Garcia

2011	Graduate Scholarship	Ontario Council on Graduate Studies of Canada and the National Council of Science and Technology of the United Mexican States
2012	Connaught Grant for Foreign Students	University of Toronto

Vasily Grigorovsky

2011	Finalist of the Centennial Thesis Award	Engineering Sciences, University of Toronto

Navid Javadi

2011	Summer Student Scholarship	Natural Sciences and Engineering Research Council of Canada

Meredith Kuipers

2010	Ontario Graduate Scholarships in Science and Technology	University of Toronto
2010	Student Travel Award, Young Investigators' Forum	Institute of Circulatory and Respiratory Health, Canadian Institute of Health Research

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Awards and Distinctions

Cheryl Lynch

2010	Best Student Paper Award – Third Place	15th International Functional Electrical Stimulation Society Conference
2011	Neural Engineering and Therapeutics Team Excellence Award	Toronto Rehabilitation Institute

Steve McGie

2010	Barbara and Frank Milligan Graduate Fellowship	University of Toronto
2011	Ontario Graduate Scholarship in Science and Technology (declined)	University of Toronto
2011	OSOTF: Graduate Student Studentship	Toronto Rehabilitation Institute
2012	OSOTF: Graduate Student Studentship	Toronto Rehabilitation Institute

Matija Milosevic

2010	Graduate Research Excellence Award (GREA)	Electrical and Computer Engineering, Ryerson University
2011	CREATE-Care Graduate Scholarship	Natural Sciences and Engineering Research Council of Canada
2012	School of Graduate Studies Conference Grant	University of Toronto

Dr. Masae Miyatani

2011	Postdoctoral Fellowship	The Craig H. Neilsen Foundation

Dr. Milos R. Popovic

2011	Elected Fellow	American Institute for Medical and Biological Engineering, Washington, USA
2012	TiEQuest Business Venture Competition: 1st Prize	TiEQuest, Toronto, Canada
2012	TiEQuest Business Venture Competition: Best Intellectual Property Award	TiEQuest, Toronto, Canada
2012	Innovation Award	Connaught Committee, University of Toronto

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Awards and Distinctions

Saswat Pradhan

2011	Summer Student Scholarship	Natural Sciences and Engineering Research Council of Canada

Dr. Hossein Rouhani

2012	Postdoctoral Fellowship	Swiss National Science Foundation

Dr. Dimitry Sayenko

2010	Best Poster Award	9th Annual Charles Tator-Barbara Turnbull Lectureship on Spinal Cord Injury, Toronto, Canada

Noel Wu

2010	Graduate Fellowship	Barbara and Frank Milligan Graduate Fellowship, University of Toronto

Dr. Albert Vette

2010	CIHR Strategic Research Training Post-Doctoral Fellowship in Health Care, Technology & Place	Canadian Institute of Health Research
2010	MITACS Industrial Fellowship Award	MITACS
2010	"Focus on Stroke 9" Postdoctoral Fellowship	Heart and Stroke Foundation of Canada, Canadian Institute of Health Research and the Canadian Stroke Network
2010	Best Paper Award 2009-2010 - category "Neural, Sensory System and Rehabilitation Engineering	Institute of Biomaterials and Biomedical Engineering, University of Toronto
2010	Best Student Paper Award – Third Place	4th National Spinal Cord Injury Conference, Niagara Falls.

Takashi Yoshida

2010	Institute for Technology Thesis Award	Institute of Biomaterials and Biomedical Engineering, University of Toronto
2010	CREATE – CARE Graduate Scholarship	Natural Sciences and Engineering Research Council of Canada
2012	Student Scholarship	Toronto Rehabilitation Institute, the Faculty of Medicine, University of Toronto

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Awards and Distinctions

Takashi Yoshida

2012	Award for Student Poster in Clinical Science	Aftab Patla Research Innovation Award, 20th International Society for Gait and Posture Conference, Trondheim, Norway
2012	Best Paper Award 2011-2012 – category “Neural, Sensory System and Rehabilitation Engineering	Institute of Biomaterials and Biomedical Engineering, University of Toronto

Dr. Jose Zariffa

2011	Clinical Research Fellowship	Canadian Paraplegic Association Ontario
2011	Winner of the Gordon Hiebert Prize for Best Postdoctoral Fellow Poster	ICORD Annual Research Meeting

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Recent Research Posters

Neuroprosthesis for Grasping

1. Brain-Controlled Functional Electrical Stimulation for Retraining of Grasp Function
2. FES Therapy for grasping in incomplete SCI: Randomized Control Trial
3. Toronto Rehabilitation Institute's Hand Function Test For Gross Motor Assessment

Neuroprosthesis for Sitting

4. Respiratory Function Increases with Trunk Muscle Stimulation Among Individuals with Spinal Cord Injury
5. Visualization of Trunk Muscle Synergies during Sitting Perburbations
6. Does Functional Electrical Stimulation of Trunk Muscles Alter Respiration in Spinal Cord Injury?
7. Linking FES-assisted sitting stability and breathing in spinal cord injury
8. A Complete, Universal, and Verifiable Set of Upper Body Segment Parameters for Three-Dimensional Dynamic Modeling
9. Self-Organizing Map for Data Mining of EMG Trunk Muscle Signals During Sitting
10. Quantifying Multi-directional Trunk Stiffness During Sitting with and without Functional Electrical Stimulation

Neuroprosthesis for Standing

11. Center of Pressure Velocity Captures Body Acceleration Rather Than Body Velocity During Quiet Standing
12. Decomposing the Ankle during Quiet Standing into Active and Passive Components using EMG
13. Controlling Balance during Quiet Standing via Co-Contraction in the Elderly May Not Be Advantageous

Cardiovascular

14. Decrease in Venous Return due to Postural Change is Mitigated by Functional Electrical Stimulation for People with Spinal Cord Injury
15. The Associations Between Aerobic Capacity and Arterial Stiffness in People with Chronic Spinal Cord Injury

Human-Machine Interfaces

16. Typing with Eye-Gaze and Tooth-Clicks
17. EMG-onset Detection using Change-point Analysis

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Brain Machine Interface

18. A Novel Assistive Device for Cursor Control
19. Estimation of Arm Kinematics From ElectroCorticoGraphic Recording
20. Using Electrocorticographic Recordings to Estimate Upper Limb Kinematics

Whole Body Vibration

21. Acute Effects of Passive Standing and Whole Body Vibration on Arterial Stiffness
22. Serum 25(OH)D Levels and Tibia Volumetric Bone Mineral Density (vBMD) in Chronic Spinal Cord Injury (SCI)
23. Usability of the Juvent™ and WAVE® Whole Body Vibration Plates for Men with and without Spinal Cord Injury
24. Whole Body Vibration and EMG Activation: Effects of Vibration Frequency, Amplitude and Posture
25. Development of a Sham Condition for Future Whole Body Vibration Intervention Trials

Other Projects

26. Bilateral Soleus H-Reflexes in Humans During Sitting and Passive Standing: Variability and Correlation.
27. Effects of Voluntary Exercise, Functional Electrical Stimulation and Vibration on Corticomuscular Coherence
28. Response of Individual Muscles to Neuromuscular Electrical Stimulation Training of Knee Extensors
29. Novel Methodology to Reduce Muscle Fatigue during Functional Electrical Stimulation using Spatially Distributed Sequential Stimulation
30. A Multi-Channel Current-Regulated Output Stage for an Electrical Stimulator
31. Including Non-Ideal Behaviour in FES Simulations
32. Including Non-Ideal Behaviour in Simulations of FES
33. Improved Fatigue Resistance in Leg Muscles during Spatially Distributed Sequential Stimulation

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Brain-Controlled Functional Electrical Stimulation for Retraining of Grasp Function

Steven McGie[1], Milos Popovic[1,2]

[1]University of Toronto, IBBME; [2]Toronto Rehab Institute

Introduction

• Functional electrical stimulation (FES) can facilitate rehabilitation following a spinal cord injury (SCI).
• Brain-machine interfaces (BMI) read neural signals, and use them to control external devices.
• BMI can be used to control an FES stimulator.
• This can restore innervation to otherwise paralyzed limbs.

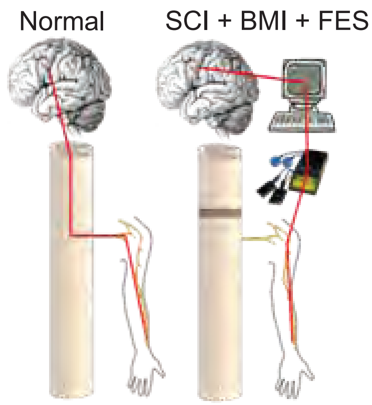

Fig. 1: Schematic of how BMI-controlled FES can circumvent a spinal lesion. The red line indicates alternate paths by which a motor command can reach the target muscles.

Hypothesis

• BMI-controlled FES may have a greater rehabilitative effect than regular FES, by following classic "Hebbian" rules of neuroplasticity.
• Synapses are strengthened when both sides are activated simultaneously.
•BMI-controlled FES may ensure consistent pairing of activity in both sides of the spinal synapse.
  • Descending commands control the BMI.
  • Antidromic firing caused in the periphery by FES goes back up to the synapse.

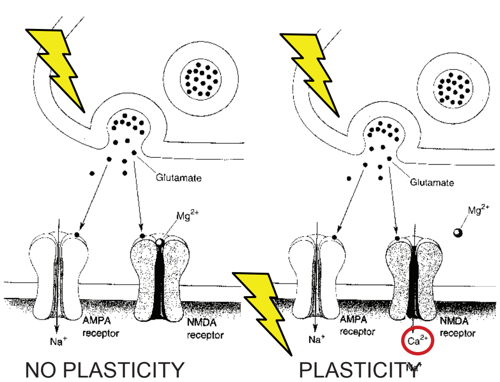

Fig. 2: Mechanisms of Hebbian plasticity. Ca^{2+} influx through NMDA channels causes synaptic strengthening. BMI-controlled FES may establish this contingency by ensuring consistent pairing of activity on both sides of the synapse.

Methods - System

• Neural recordings are non-invasively obtained from hand-related areas of primary motor cortex using electroencephalography (EEG).
• EEG recordings are analyzed in real-time to determine the bandpower in a specific frequency band.
• Sustained changes in bandpower across a threshold send a signal to switch FES from "Rest" to "Flexion" mode.
• Switch from "Flexion" to "Extension" is done via push-button, since artefacts from stimulation contaminate the EEG signal.
• Switch from "Extension" to "Flexion" occurs after 3.5 s.
• Extensive calibration routines are used to determine ideal parameter values.
  • Based on a short session in which the user is cued to either attempt a movement or to relax.
  •The system then finds parameters which allow for ideal discrimination between the two tasks.

Methods - Treatment

• Participants will be individuals with incomplete C4-C7 SCI
• The intervention will consist of 40 sessions (2 months) of using BMI-controlled FES to perform functional movements.
• Outcome measures will include self-report measures of functional independence, functional assessments, and physiological assessments.

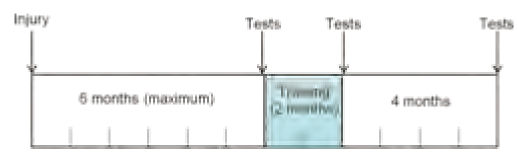

Fig. 3: Timeline of each subject's participation.

Discussion

• System completed and tested on able-bodied volunteers.
• Recruitment for individuals with high-level SCI opened; no participants at the time of writing.
• System needed modification to account for stimulation artefact.

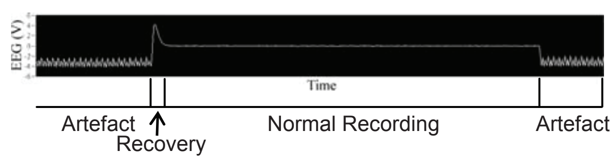

Fig. 4: Sample EEG recording showing periods with and without contamination from the stimulation artefact.

Conclusions

• The feasibility of BMI-controlled FES has been proven.
• The hypothesis (that BMI-controlled FES can provide greater rehabilitative effect than regular FES) remains to be tested; contingent on recruitment.

Made in Canada

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Partners:

FES Therapy for grasping in incomplete SCI: Randomized Control Trial

MR Popovic[1,2], N Kapadia[2], V Zivanovic[2], JC Furlan[2], BC Craven[2,3], K Nakazawa[4], and C McGillivray[2,3]

[1] Toronto Rehabilitation Institute, [2] IBBME - University of Toronto,
[3] Departments of Medicine and Health Policy Management and Evaluation - University of Toronto,
[4] Department of Life Sciences, Graduate School of Arts and Sciences, University of Tokyo, Japan

Introduction

- There are about 350,000 SCI survivors in US and Canada, and 50% of them are Quadriplegics.
- 50% of the Quadriplegics ranked return of arm hand function as their highest priority.

Methods

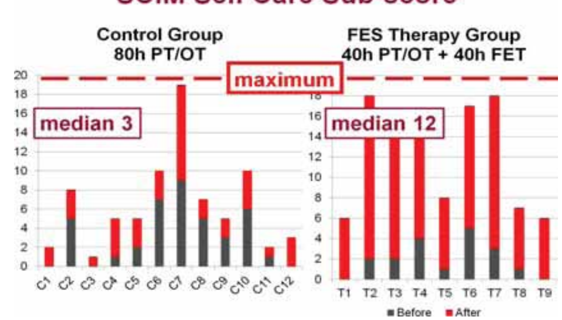
- Randomized control trial
- Control: n=12 and FES therapy: n=9
- Both groups received 40 sessions of therapy
- Control group received 2 hours of conventional occupational therapy (COT) and intervention group received 1 hour of COT plus 1 hour of FES therapy.
- Primary outcome measure was FIM Self Care Sub score and secondary outcome measures were SCIM Self Care Sub score and TRI-HFT.

Subject Demographics

Feature	Control Group	Intervention Group	p value
Age (years)			
Mean age ± SEM	44.75 ± 4.72	43.2 ± 5.45	0.896
Median age	51.5	50	
Age range	20 to 65	18 to 66	
Sex (n)			
Males	9	8	1
Females	3	2	
Cause of SCI (n)			
Motor vehicle accident	7	2	0.009
Fall	2	8	
Other causes	3	0	
Severity of SCI (n)			
AIS B	4	4	0.517
AIS C	8	5	
AIS D	0	1	
Level of SCI (n)			
C3	0	1	0.071
C4	7	3	
C5	4	1	
C6	1	5	
Time since SCI (days)			
Mean time ± SEM	58.33 ± 6.55	69.9 ± 14.11	0.974
Median time	63.5	50	
Age time	22 to 102	33 to 164	

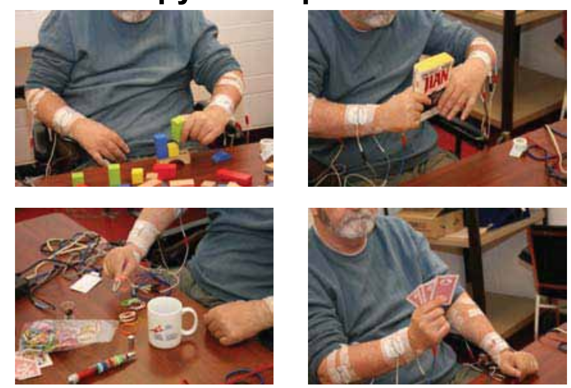
Results

- FIM self care sub-score: 20 points improvement (FES therapy) as compared to 10 (COT) (p=0.015)
- SCIM self care sub-score: 12 points improvement (FES therapy) as compared to 3 (COT) (p<0.0001)
- TRI-HFT: TRI-HFT total score for manipulation of objects 1 to 10 also approached statistical significance (p=0.0538)



TEST	Control Group		Intervention Group		p values (both hands)
	Before	*After*	*Before*	*After*	
FIM Self Care Sub-scores	7.8	17.8	8.1	28.2	0.015
SCIM Self Care Sub-score	3.3	6.4	1.9	12.1	<0.0001
TRI-HFT Components					
10 Objects	27.2	38.5	37.1	53.8	0.054
Rectangular Blocks	29.3	38.4	34.7	49.7	0.124
Instrumented Cylinder *Able to Hold*	1.90	1.33	1.0	1.7	0.033
Instrumented Cylinder *Torque Values (Nm)*	0.26	2.59	0.26	1.13	0.4247
Credit Card *Able to Hold*	1.33	1.41	1.0	1.7	0.035
Credit Card *Force Values (N)*	2.67	8.76	4.42	12.5	0.422
Wooden Bar *Able to Hold*	0.63	0.96	0.8	1.5	0.065
Wooden Bar Thumb Direction *Length Values (cm)*	2.88	10.5	1.67	10.94	0.622
Wooden Bar Little Finger Direction *Length Values (cm)*	3.17	11.85	5.56	12.78	0.767

FES Therapy Participant



Conclusions

- Restoration of voluntary hand function in incomplete SCI is possible using FES.
- Improvements in hand function and thereby increase in level of independence are significant with FES training.

Recommendations

- Flexible and programmable FES system.
- Repetitive daily treatments.
- FES in combination with OT.

References

- Popovic, Kapadia, Zivanovic, Furlan, Craven, and McGillivray, *Neurorehabilitation and Neural Repair*, vol. 25, No. 5, pp: 433-442, 2011.
- Popovic, Thrasher, Adams, Takes, Zivanovic, and Tonack, *Spinal Cord*, vol. 44, No. 3, pp. 143-151, 2006.
- Kapadia, Zivanovic, Furlan, Craven, McGillivray, and Popovic, *Artificial Organs*, vol. 35, No. 3, pp: 212-216, 2011.
- Kapadia, Zivanovic, Verrier, and **Popovic**, *Accepted for publishing in Topics in Spinal Cord Injury Rehabilitation in April 2011.*

Partners:

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The authors acknowledge the support of Toronto Rehabilitation Institute who receives funding under the Provincial Rehabilitation Research Program from the Ministry of Health and Long-Term Care in Ontario. The views expressed do not necessarily reflect those of the Ministry."

TORONTO REHABILITATION INSTITUTE'S HAND FUNCTION TEST FOR GROSS MOTOR ASSESSMENT

Kapadia N[1], Zivanovic V[1], Verrier M[1], Popovic MR[1,2]

[1] Toronto Rehabilitation Institute, [2] IBBME - University of Toronto

Introduction

The Toronto Rehabilitation Institute-Hand Function Test (TRI-HFT) is a new user-friendly assessment tool designed to evaluate unilateral gross motor function of the hand to perform power grasp and precision grip in tetraplegic individuals.

Methods

- Reliability, validity and sensitivity of TRI-HFT were evaluated within a randomized control trial.
- Control: n=12 FES: n=9.
- Both groups received 40 sessions of therapy. Control group received 2 hours of conventional occupational therapy (COT) and intervention group received 1 hour of COT plus 1 hour of FES therapy.
- Outcome measures were FIM Self-care Sub-score, SCIM Self-care Sub-score and TRI-HFT.

TRI-HFT

- The TRI-HFT consists of two parts:
  1) The first part of the test assesses the patients' ability to manipulate objects that they may encounter in their daily lives using a lateral pinch, pulp pinch or palmar grasp.
  2) The second part of the test measures the strength of their lateral pinch or pulp pinch, and palmar grasp.
- The entire evaluation for both hands can be completed in less than 30 minutes.

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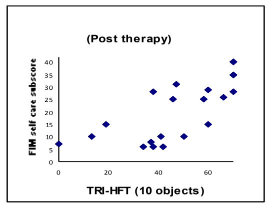
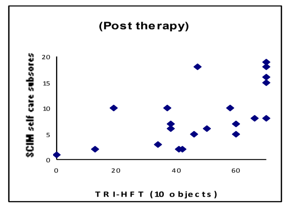
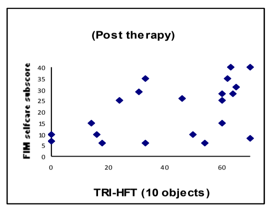
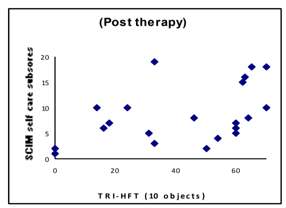
Results

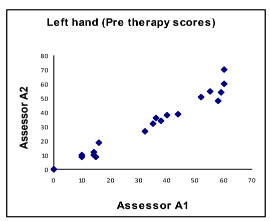
	Left Hand (r values)	Right Hand (r values)
Construct Validity		
TRI-HFT and FIM self-care sub-score	0.73	0.56
TRI-HFT and SCIM self-care sub-score	0.62	0.48
Inter-rater Reliability	0.99	0.99

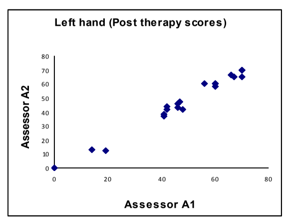


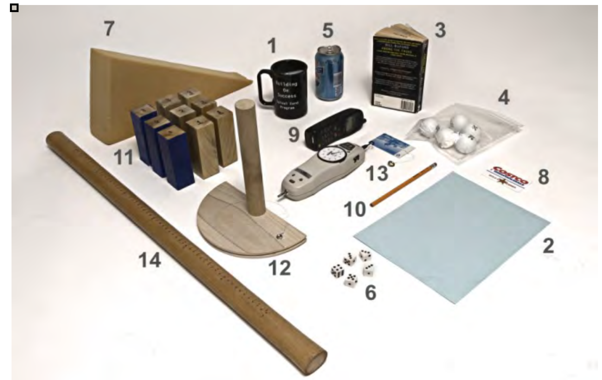












Conclusion

- TRI-HFT is a simple, reliable and valid measure.
- TRI-HFT takes less than 30 minutes to be administered and it is a publically available test.
- TRI-HFT was found to be highly responsive to change.

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Partners:

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RESPIRATORY FUNCTION INCREASES WITH TRUNK MUSCLE STIMULATION AMONG INDIVIDUALS WITH SPINAL CORD INJURY

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Introduction

Altered breathing following spinal cord injury (SCI) depends on the level and completeness of injury. Multi-system motor dysfunction, including lack of coordinated activation of respiratory and trunk muscles, and postural instability inhibit proper diaphragmatic function. In upright sitting, this results in reduced expiratory capacity[1] and tidal volume[2]. Functional electrical stimulation (FES) improves sitting posture, including pelvic orientation and spinal alignment, and lung capacity during a forced expiratory manoeuvre[3]. However, respiratory function remains to be investigated among individuals with SCI during the application of FES.

Objective

1) Describe the effects of sitting posture on respiratory function in able-bodied individuals (ABs).
2) Test the feasibility of using FES to change respiratory function in individuals with SCI.

Participants & Paradigm

Respiratory function, including tidal volume (VT), respiratory rate (RR), and minute ventilation (VE), was measured during 60s trials of unsupported sitting.

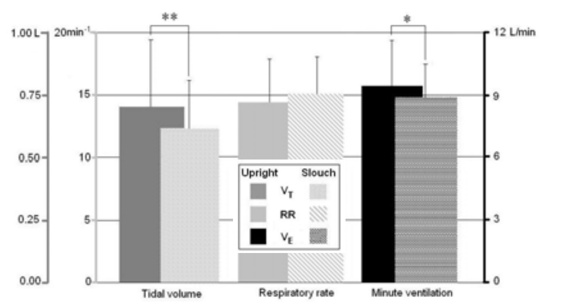
- **ABs** (N=10): upright sitting, SIT_{UP}; slouch sitting, SIT_{SL}
- **SCIs** (N=3): unsupported (SIT_{UN}); anterior-posterior trunk muscle FES, AP-FES; medial-lateral trunk muscle FES, ML-FES.

Results

Table 1. Participant characteristics

	Age (y)	Injury level	Dur (m)	Pack-years
ABs	31±5.9	--	--	6.3±3.7
SCI1	25	C5-I	31	8.0
SCI2	25	C5-MC	36	non-smoker
SCI3	33	C4-MC	192	2.3

Figure 1. ABs respiratory function during sitting



ABs: During SIT_{UP}, VT increased by 14% compared to SIT_{SL} (0.70±0.27l vs. 0.61±0.19l, p=0.009**); RR did not change significantly between SIT_{UP} and SIT_{SL} (14±3.47min-1 vs. 15±2.97min-1, p=0.18). Similarly, VE was 6% greater during SIT_{UP} compared to SIT_{SL} (9.40±2.20l/min vs. 8.83±1.60l/min, p=0.01*).

Table 2. SCIs respiratory function during sitting

		VT	RR	VE
SCI1	SITUN	0.51	18.8	9.7
	AP-FES	↓3% *	↑ 3% *	↑ 1% *
	ML-FES	↓2% *	↑ 6%	↑ 3% *
SCI2	SITUN	0.81	11.2	9.0
	AP-FES	↑ 10%	↑ 9%	↑ 12%
	ML-FES	↑ 19%	↓2% *	↑ 15%
SCI3	SITUN	0.75	13.8	9.7
	AP-FES	↑ 7%	↑ 5%	↑ 14%
	ML-FES	↓8%	↑ 32%	↑ 26%

** A difference of <5% was not considered meaningful.*

SCIs: For participants with complete SCI, FES appears to have a positive effect on VE. SCI1, who was the only participant with an incomplete injury, did not demonstrate any meaningful change in respiratory function with FES. There was an increase in measures of respiratory function with FES in participants SCI2 and SCI3: SCI2 and SCI3 demonstrated increases in VT and RR, depending on the site of stimulation. VE increased for all participants with the application of FES from 1%-26%. The prominent changes in VE for SCI2 and SCI3 may be due to the completeness of their injuries.

Discussion

The ultimate goal is to delineate customized stimulation paradigms that are patient-specific to optimize respiratory function during functional activities. The benefits of achieving optimal respiratory function are significant and may parallel achieving optimal posture (i.e. upright vs. slouch sitting), as demonstrated in able-bodied individuals. Individuals with high-level, motor and sensory complete SCI may benefit from trunk muscle FES given the potential to increase VE. It may be possible to use FES to improve VE and therefore physiological homeostasis. Further studies are required to determine if these preliminary findings can be reproduced, and must include participants with different levels and completeness of SCI to determine specificity of stimulation parameters and resultant changes in different components of respiratory function.

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Visualization of Trunk Muscle Synergies during Sitting Perturbations

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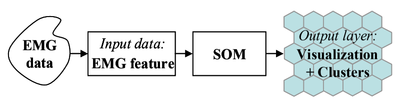
Introduction

Muscle Synergies
- *Synergy* = pattern of co-activations of muscles
- multiple muscles activate during sitting perturbations
- **Problem:** how to compare postural synergies?

Visualization using Self Organizing Maps (SOM)
- *SOM*: encodes of data on the map
- *Visualization:* data summary on a 2D map projection



Objectives:
- visualization of sitting trunk muscle synergies
- compare 8 perturbation direction synergies

Methods

Experiment [1]
- sample: N=13; healthy
- sitting perturbations: 8 directions
- EMG: 10 trunk muscles (bilateral)

Input Data
- EMG filtering: rectification, low-pass & window filter
- feature: phasic EMG response after perturbation

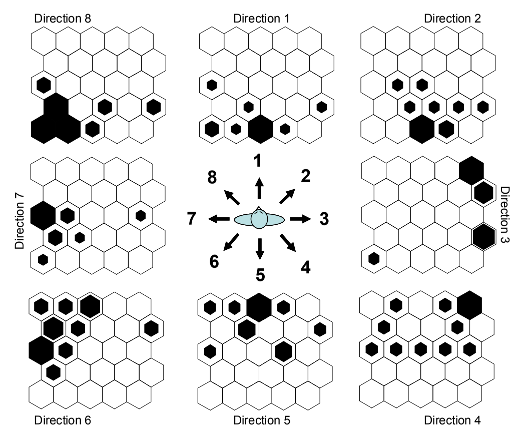
SOM
- SOMToolbox, Helsinki University of Technology
- normalized input data
- **Training** → EMG (all) + directions (8 direction)
  - input data ($\mathbf{x}_n$): 104 input vectors
  - output layer: 5x5 nodes have weight vectors $\mathbf{w}_j$
  - SOM: groups similar data patterns together

$$\mathbf{w}_j(n+1) = \mathbf{w}_j(n) + h_{cj}(n)\cdot[\mathbf{x}_n - \mathbf{w}_j(n)]$$

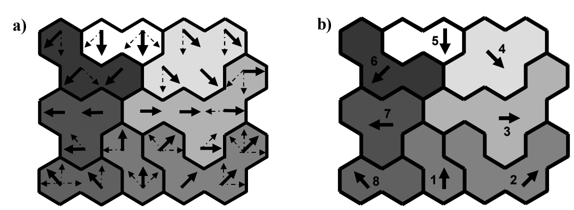
Output Layer
- 5x5 hexagonal map (2-D)
- **Visualization** → density / clusters analysis

Results

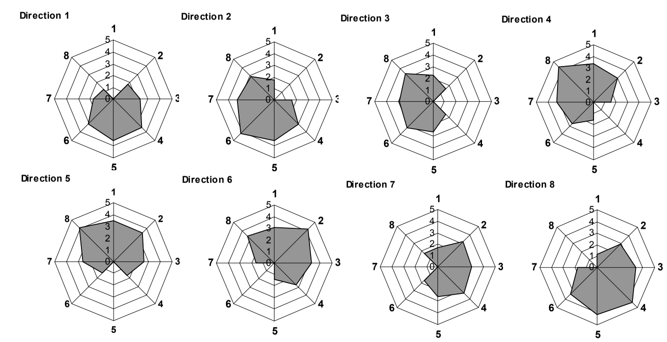
Visualization of Trunk Muscle Synergies
- EMG pattern for bilateral: rectus abdominis (RA), external oblique (EO), and internal oblique (IO), thoracic erector spinae (T9) and lumbar erector spinae (L3)



Perturbation Clusters
- clustering based on 'winning' directions on each node a) all directions (b) 'winning' cluster for 8 directions
- spatial proximity of clusters = similarity of synergies



Synergy Comparison
- Euclidian distance between the center of each cluster
- implies postural synergy similarities and differences
- small distance = similar; large distance = dissimilar



Discussion

Trunk Muscle Response
- direction-dependant differences during sitting [1,2]
- adjacent directions most similar responses
- progressively more dissimilar as perturbations change

Synergy Visualization
- SOM encodes complex muscle synergies
- describes synergistic muscle relationships at a glance
- clusters could uncover deviant neuromuscular responses
- compare muscle synergies of patients (eg. SCI, stroke)

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DOES FUNCTIONAL ELECTRICAL STIMULATION OF TRUNK MUSCLES ALTER RESPIRATION IN SPINAL CORD INJURY?

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Relationship between Postural Control and Respiratory Capacity

Introduction

- Individuals with SCI at or above L1 experience loss of voluntary control of trunk muscles from atrophy and paralysis
- During sitting, postural control is compromised; individuals assume a suboptimal posture to maintain sitting balance
- Lumbar kyphosis and posterior pelvic tilt lead to limitations in functional ability and reduced respiratory capacity
- Current interventions for corrective postural control are functionally restrictive
- Implanted stimulating electrodes delivering FES to rectus abdominus and erector spinae has improved vertebral alignment and respiratory capacity in a single SCI subject while sitting
- Transcutaneous functional electrical stimulation (FES) is a non-invasive technique for training motor recovery in SCI

Hypothesis

We hypothesize that transcutaneous FES of select trunk muscles will facilitate postural control and gains in respiratory capacity while sitting

Methods

- During each trial, data will be recorded: kinematic measures of head position, vertebral alignment, and pelvic tilt; respiratory measures of forced vital capacity, forced expiratory volume in one second, and the ratio of the two values (FVC, FEV_1, and FEV_1/FVC); dynamic sitting stability measured as center of pressure (COP).

Inclusion Criteria

- Subjects with:
  - Level C5-T12 injury
  - ASIA A-D impairment
  - No prior respiratory disease

Experimental Paradigm

Incentive spirometry

Multi-channel stimulation

Motion capture system (not shown)

Force plate

Tasks:

1a. Quiet sitting
1b Quiet sitting w/ FES
2a. Complex cognitive task
2b. CCT w/ FES

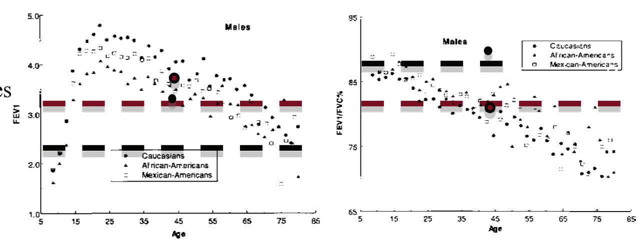

Figure 1. Plots comparing mean normative respiratory values FEV_1 and FEV_1/FVC by age cohort to: average values for subjects with paraplegia (red lines) and tetraplegia (black lines); and values for a single tetraplegic subject with (red dot) and without (black dot) implanted trunk muscle FES. Figure and data adapted from: Hankinson et al. 1999; Baydur et al. 2001; Triolo et al. 2009.

Discussion

- In SCI, respiratory impairments correlate with injury level and worsen with upright (sitting) posture: restrictive nature of impairments mean FEV_1 and FVC decrease proportionally so the ratio may be normal or increased
- In order to prove efficacy, transcutaneous FES must facilitate similar gains in respiratory capacity as current interventions: i.e. change on the magnitude of 10+% increase in FEV_1 and 20+% increase in FVC
- Other improvements of surface FES-assisted sitting for individuals with SCI may include an aesthetic and corrected-normal posture; pressure relief off the coccyx; and reduction in incidence of pressure ulcers.
- By training with FES in a multitask paradigm, there may be improvements in upper limb function from achieving better postural control in a sitting position.
- The benefits of achieving sitting balance are significant, contributing to independence, improved health status, and wellbeing in the SCI population

Acknowledgements

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Linking FES-assisted sitting stability and breathing in spinal cord injury

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Institute of Biomaterials and Biomedical Engineering[3]; Toronto Rehabilitation Institute[4].

Toronto Rehab is a teaching and research hospital affiliated with the University of Toronto.

Research Problem

Loss of voluntary control of the trunk muscles due to paralysis following spinal cord injury (SCI) is associated with abnormal posture, impaired sitting stability and reduced breathing capacity during sitting. Specifically, thoraco-lumbar kyphosis and posterior rotation of the pelvis are features of a typical maladaptive posture in SCI[1] that allow individuals without trunk control to maintain sitting balance yet limits some functional activity[2,3]. A similar posture in healthy individuals results in significant decreases in lung function[4],which is evidence of the adverse effects of such a position on breathing capacity. After an SCI, trunk muscle paralysis leads to loss of abdominal muscle tone such that in an upright position the effect of gravity and increased abdominal compliance causes a caudal shift of the abdominal contents[5]. This shift effectively removes the fulcrum of the diaphragm, affecting the ability of the diaphragm to do respiratory work in an upright position[6], further compromising the respiratory system. Stimulation of select trunk muscles using implanted electrodes has contributed to improvement of vertebral alignment and has facilitated gains in functional independence and breathing capacity[7].

Objective

To determine, using a proof-of principle paradigm, the feasibility of muscle-specific functional electrical stimulation (FES) to improve vertebral alignment, enhance diaphragmatic functiona, and optimize parameters of task-specific functions. Electrical stimulation is being delivered to select trunk muscles via surface electrodes to stiffen abdominal muscles and trunk extensors. Vertebral alignment, sitting stability and breathing capacity are being measured during static and dynamic conditions of unsupported and FES-assisted sitting.

Methodology

Individuals will be asked to participate based on the following criteria:

Inclusion	Exclusion
25-45 yrs	
Able-bodied *or* SCI	Acute or chronic respiratory illness
[C5-T12 injury level]	>10 pack year smoking history
[Post-acute phase]	BMI >25

Experimental Set-up

Device	Function	Outcome measures	Placement
Stimulating electrodes	Artificial muscle stimulation	N/A	Rectus abdominis, lumbar erector spinae
Force plate	Sitting stability assessment	Center of Pressure Displacement, range, velocity	Sitting surface
Motion analysis system	Vertebral alignment assessment	Head position Spinal curvature Pelvic tilt	C7, forehead L1, L3, L4, L5, S1 ASIS, PSIS
Spirometer	Respiratory assess-ment during pulmonary function test (PFT)	Expiratory volumes and flow rates	Flow sensor at mouth and nose clip

Pilot Studies of Tasks

Two individuals, one able-bodied and one chronic SCI, performed repeated trials of static and dynamic sitting tasks on the force plate. Representative COP diagrams for each type of task are shown below.

Does surface FES have an effect on sitting COP in able-bodied individuals?

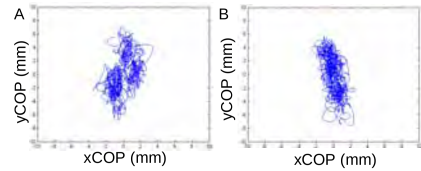

Figure 1. Center of pressure excursions for an able-bodied individual during quiet sitting (A) and with FES (B).

Does performing PFTs affect sitting COP in individuals with chronic SCI?

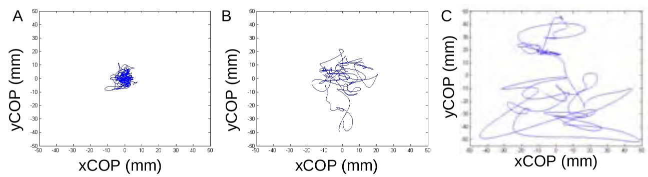

Figure 2. Center of pressure excursions for an individual with chronic SCI during quiet sitting (A), performing a PFT before (B) and after (C) a series of functional reaching tasks.

How do dynamic activities affect sitting COP in individuals with chronic SCI?

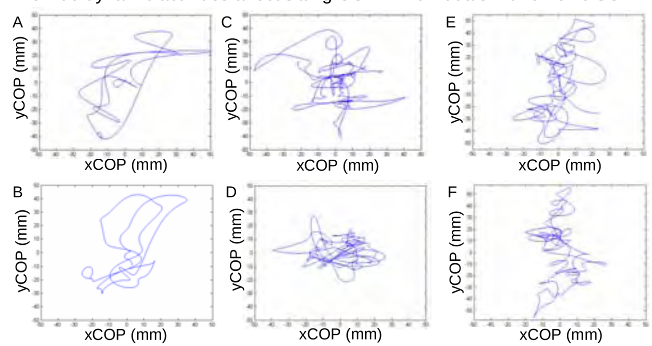

Figure 2. Center of pressure excursions for an individual with chronic SCI during multiple trials of forward reaching (A, B), lateral reaching (C, D), and posterior leaning (E, F).

Conclusions

In both quiet sitting (QS) and incentive spirometry (IS) conditions for able-bodied, surface stimulation of the abdominal and trunk extensor muscles reduced total excursion, velocity and frequency of the COP. The larger COP values measured during quiet sitting in chronic SCI suggest less sitting stability in comparison to able-bodied during unsupported sitting, as expected. In chronic SCI, the frequency of the COP was decreased and the excursion and velocity both increased during multiple trials of dynamic tasks. It is possible that individuals with chronic SCI use alternative motor strategies to complete dynamic tasks, each representing a characteristic COP.

Table 1. Summary of mean COP values for able-bodied individual under different conditions. All trials were 60sec in duration; each trial condition, quiet sitting (QS) and increased response effort (IRE), was repeated with transcutaneous FES (w/ FES).

Condition	No. Trials	Mean total excursion (mm)	Mean velocity(mm/s)	Mean frequency (Hz)
QS	3	11.1711	0.4965	1.0221
QS w/ FES	3	8.7536	0.3890	0.8851
IS	3	22.6846	1.0082	1.6045
IS w/ FES	3	17.1789	0.7635	1.2313

Table 2. Summary of mean COP values for individual with chronic SCI.

Condition	No. Trials	Mean total excursion (mm)	Mean velocity(mm/s)	Mean frequency (Hz)
QS	1	23.3581	1.0381	1.1481
pre PFT	3	37.4737	1.6655	0.3862
forward reach	4	23.0051	1.0224	0.1843
lateral reach	4	39.0721	1.7365	0.4089
lean back	3	40.2169	1.7874	0.2651
post PFT	3	25.2793	1.1235	0.2838
PFT arm raise	1	52.2974	2.3243	0.3131
QS final	1	34.9224	1.5521	1.6529

Implications

Ultimately, our goal is to develop comprehensive subject-specific protocols to test the relationship between sitting stability and breathing capacity in SCI. The next step is to test surface FES during a dynamic paradigm to determine the acute effects of FES and altered sitting stability on breathing.

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A Complete, Universal, and Verifiable Set of Upper Body Segment Parameters for Three-Dimensional Dynamic Modeling

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Center for Neuromotor and Biomechanics Research, University of Houston

Introduction

Background

Dynamic models of the human trunk have been extensively used to investigate the biomechanics of lower back pain and postural instability in different populations. Despite their diverse applications, these models rely on intrinsic upper body segment parameters (UBSP), e.g., each segment's mass-inertia characteristics. However, a comprehensive UBSP set allowing state-of-the-art, three-dimensional (3D) dynamic modeling does not exist to date.

The purpose of this study was to establish a UBSP database that is accurate, complete, and universal, i.e., independent of pre-defined (lumped) trunk portions such as 'upper trunk' and 'lower trunk' (**PART I**). To demonstrate the practicality of the UBSP, they were finally implemented in a 3D dynamic model of the upper body to predict lumbar joint torques from experimental kinematics during perturbed sitting (**PART II**).

Methods – Geometric Models and UBSP

Data Source:

- High-resolution, transverse color images from the **Male Visible Human** (MVH) [1]
- MVH anthropometrics: age 38; height 180 cm; weight 90 kg [1]
- Image resolution: 0.144 mm x 0.144 mm x 1 mm [1]

Body Segmentation:

- 24 vertebral trunk and head segments:
  - 5 lumbar segments (**L1 to L5**)
  - 12 thoracic segments (**T1 to T12**)
  - 6 cervical segments and head (**C2 to C7 and HD**)
- 2 x 2 upper limb segments:
  - left and right upper arm (**lUA and rUA**)
  - left and right forearm-hand complex (**lFA-H and rFA-H**)

3D Geometric Models of:

- MVH spinal vertebrae, spinal discs, and pelvis (PV)
- MVH body shell

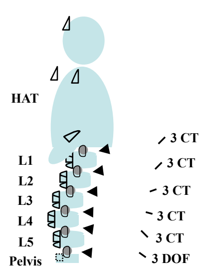
Identified Parameters:

- Spinal joint centers
- Segment masses
- Location of segment center-of-masses
- 3 x 3 moment of inertia tensor of segments

Methods – Inverse Dynamics

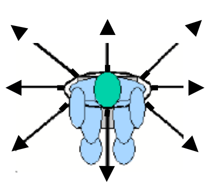
Model:

- Focuses on the action of the lumbar spine [2]
  - AP: flexion-extension
  - ML: lateral bending
  - RT: axial rotation
- Six rigid bodies [2]:
  - L1 to L5
  - Head-Arms-Thorax complex (HAT)
- 3 degrees-of-freedom (DOF) between at L5-PV joint [2]
- 3 x 5 constraints (CT) at the remaining 5 joints [2]



Experiments:

- 1 subject with MVH anthropometrics
- Horizontal perturbations during upright sitting
- 8 perturbation directions, 5 trials each
- Impulse force of ~200 N max, applied inferior to axillae

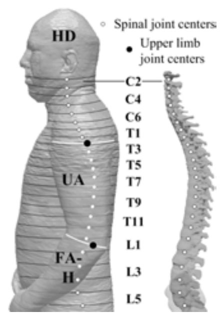


Joint Torque Estimation:

- Inverse dynamics from kinematics to joint torques
- 3 different dynamic implementations:
  - Newton-Euler formulation
  - Lagrange formulation
  - Simulink & SimMechanics (Mathworks Inc.)

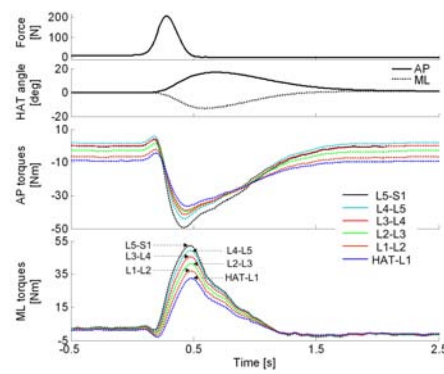
Results

Geometric Models and UBSP:



Surface models of the MVH body shell and spine. The identified joint centers were used to define the vertebral trunk segments and upper limb segments for which respective mass-inertia characteristics were calculated using the geometric method.

Inverse Dynamics:



Results of the inverse dynamics using the identified UBSP set. Shown are the AP and ML trunk angles and joint torques for an anterior-left diagonal perturbation during sitting. A visual inspection suggests that the torque outputs of the three different inverse dynamics implementations match very well (plotted on top of each other; R^2 = 99.9999 %). It can also be seen that the AP torque traces are affected by the AP curvature of the spine.

Discussion

Identified UBSP Set:

- **Complete** with respect to 3D dynamic modeling (all required parameters identified)
- Based on **highly accurate** 3D geometric models (1000 higher than existing sets)
- **Universal** with respect to potential model definitions (parameters can be lumped together)
- Uniquely **verifiable and expandable** (MVH dataset is freely accessible)

Conclusion

A comprehensive UBSP database has been obtained that can be implemented in 3D dynamic models to: (1) systemize thinking in postural control studies; (2) quantify the effect of impact forces on the head and trunk (e.g., during whiplash); (3) suggest population-specific experiments based on theoretical insights into trunk dynamics (e.g., regarding lower back pain); or (4) assess the feasibility of new surgical techniques (e.g., spinal fusion) and neuroprostheses (e.g., after spinal cord injury [3]).

References

[1] Spitzer V et al, *J Am Med Inform Assn* **3**, 118-130, 1996.
[2] Wilkenfeld AJ et al, *J Rehabil Res Dev* **43**, 139-152, 2006.
[3] Masani K et al, *Clin Biomech* **24**, 176-182, 2009.

We acknowledge the support of the Toronto Rehabilitation Institute, which receives funding under the Provincial Rehabilitation Research Program from the Ministry of Health and Long-Term Care in Ontario. The views expressed do not necessarily reflect those of the Ministry.

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Partners:

SELF-ORGANIZING MAP FOR DATA MINING OF EMG TRUNK MUSCLE SIGNALS DURING SITTING

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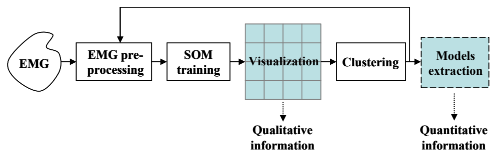
Introduction

Spinal Cord Injury (SCI)
- causes paralysis of lumbar musculature
- trunk instability is a major problem with SCI
- multiple muscles contribute to trunk stability

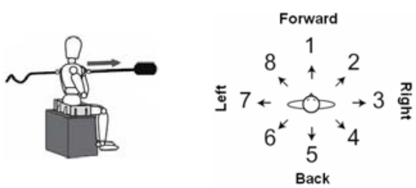
Data Visualization using Self Organizing Maps (SOM)
- *SOM* - encodes statistical properties of data on the map
- *Visualization* - data summary on a 2D map projection
- *Models* - SOM models for closed-loop control of sitting



Objectives:
- SOM for electromyography (EMG) data mining
- Visualization of high-dimension EMG data sets
- Model trunk muscle activation patterns during sitting

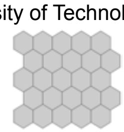
Methods

Experiment [1, 2]
- Sample: N=13, all healthy
- Sitting perturbations: 8 directions
- Record EMG: (5 trunk muscles x 2 sides)



Data processing
- EMG filtering: rectification, low-pass & windowing filter
- Model feature: max EMG response after perturbation

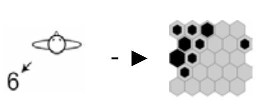
SOM (SOMToolbox, Helsinki University of Technology):
- 5x5 hexagonal map
- normalized input
- batch training mode
- Euclidian measure



- **Training** → EMG (10 muscles) + directions (8 direction)
  - nodes (index *j*) have a weight vectors $\mathbf{w}_j$
  - input vector ($\mathbf{x}_n$), iteration *n,* compared to nodes
  - algorithm selects (index *c*) nearest node to the input $\mathbf{x}_n$
  - this weight vector and region $h_{ci}(n)$ updated toward input $\mathbf{x}_n$

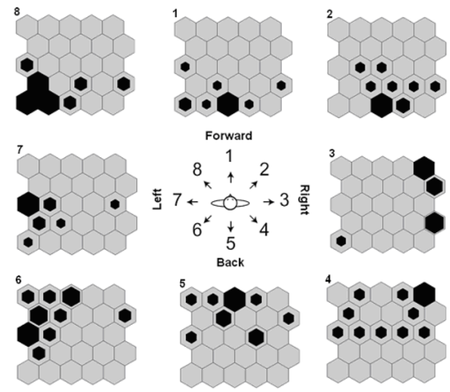
$$\mathbf{w}_j(n+1) = \mathbf{w}_j(n) + h_{cj}(n) \cdot [\mathbf{x}_n - \mathbf{w}_j(n)]$$

- **Visualization** → density / clusters analysis for all directions

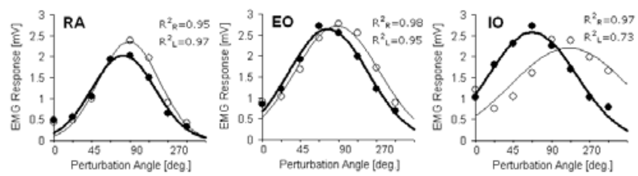
Input 6 

Results

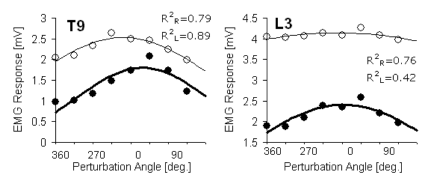
SOM Visualization



SOM Model
- Clustering – based on 'winning' directions on each node
- Gaussian regression models for all muscles from clusters
- Normal distribution; consistent with [1, 2]



Activation pattern for abdominal muscles left (gray line) and right (black line): rectus abdominis (RA), external oblique (EO), and internal oblique (IO)



Activation pattern for back muscles for left (gray line) and right (black line): thoracic erector spinae (T9), lumbar erector spinae (L3)

Discussions

SOM Visualization
- describes synergistic muscle relationships at a glance
- cluster could uncover deviant neuromuscular responses
- can assess and grade trunk function in individuals

SOM Models
- can model the activation patterns
- opposing muscles stabilizes perturbations [1, 2]

References

[1] Masani K, et al. Postural reactions of the trunk muscles to multi-directional perturbations in sitting. Clin Biomech, 2009
[2] Preuss R, Fung J Musculature and biomechanics of the trunk in the maintenance of upright posture. J Electrom Kinesiol, 2008

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Quantifying Multi-directional Trunk Stiffness During Sitting with and without Functional Electrical Stimulation

Able-Bodied Individuals

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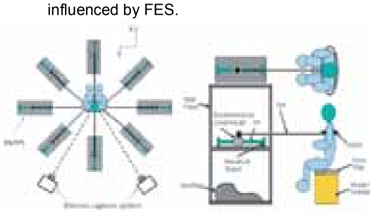
Objectives

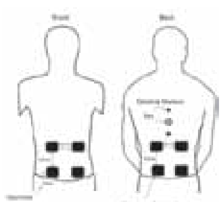
1. To develop a novel protocol to quantify and identify the multidirectional stiffness of the trunk during sitting
2. To test the effect of FES on the multidirectional stiffness of the trunk.

Background

- Inability to voluntarily control the trunk musculature for individuals with spinal cord injury (SCI) causes both functional problems and secondary health complications.
- One possible solution to improve sitting balance is to use a functional electrical stimulation (FES) to activate patient's own musculature to support the trunk [1].
- To build such FES system one needs to determine multidirectional stiffness of the trunk during sitting and to demonstrate that this stiffness can be influenced by FES.



Portable and Automated Postural Perturbation System (PAPPS)



Locations of FES and Motion Tracking

Experiment

Subjects:

- 15 able-body male subjects with averages: 27yrs, 176cm and 73kg

Task:

- Sitting upright with eyes closed
- Perturbed through a chest harness in 8 horizontal directions randomly using PAPPS, for a total of 160 trials [3]
- 20 trials for each perturbed direction: 10 without FES and 10 with FES
- FES applied to erector spinae & abdominal muscles in the lumbar region

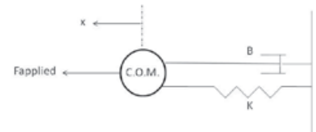
Measurements:

- Horizontal perturbation force
- Kinematics of T10 as COM of trunk, using motion capture system

Analysis:

- First 1s time period from the onset of the perturbation was analyzed
- Trunk was modelled as a mass-spring-damper system
- Translational and torsional trunk stiffness and damping were identified using an optimization algorithm (Matlab, Mathworks)

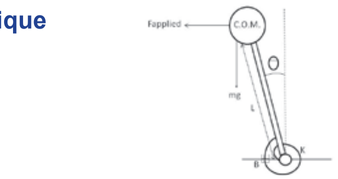
System Identification Technique



$$F_{!""\#\$\%\&} = m * \ddot{x} + b * \dot{x} + k * x$$

Model 1) Translational Mass-Spring-Damper

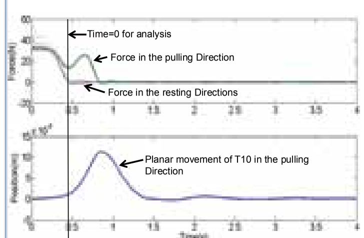
COM – center of mass of the system
b – damping coefficient
k – stiffness coefficient
$F_{applied}$ –force exerted on the COM
x – displacement from the equilibrium position



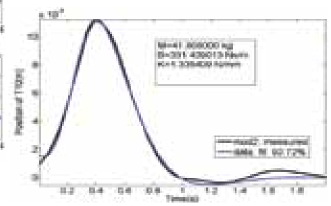
$$F_{!""\#\$\%\&} * L * \cos\theta + m * g * L * \sin\theta = I * \ddot{\theta} + B * \dot{\theta} + K * \theta$$

Model 2) Torsional Mass-Spring-Damper

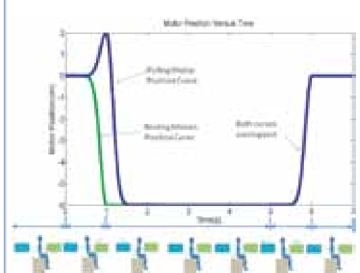
COM – center of mass of the system
I – moment of inertia
B – torsional damping coefficient
K – torsional stiffness coefficient
$F_{applied}$ – force exerted on the COM
L – length from L4/L5 joint to the COM
g – gravitational constant
θ – angular displacement from the equilibrium



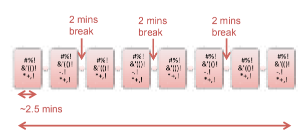
Example: Position and Force Profile of a Trial



Example of Curve Fitting to Obtain Mass, Stiffness, and Damping Ratio

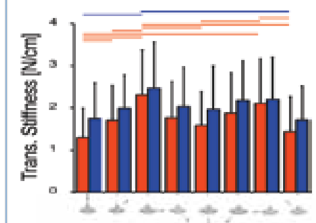


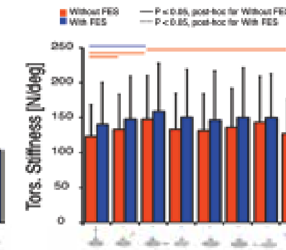
Position Curves of Pulling and Resting Motor



Protocol of Perturbation Timeline

Results





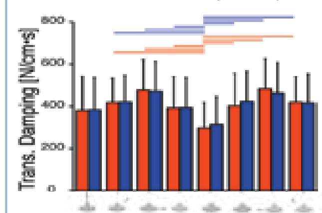
Translational Stiffness With and Without FES - 2 *Way ANOVA:*

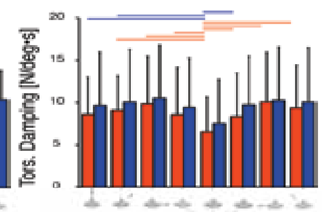
- Main Factor: NO FES vs. FES, p = 0.004
- Main Factor: Directions, p<0.001
- Interaction: p = 0.210

Torsional Stiffness With and Without FES - 2 *Way ANOVA:*

- Main Factor: NO FES vs. FES, p = 0.116
- Main Factor: Directions, p<0.001
- Interaction: p = 0.197

- *Both translational and torsional stiffness are significantly different between directions*
- *Translational stiffness is significantly different between with and without FES*
- *Torsional stiffness is not significantly different between with and without FES*





Translational Damping With and Without FES - 2 *Way ANOVA:*

- Main Factor: NO FES vs. FES, p = 0.842
- Main Factor: Directions, p<0.001
- Interaction: p = 0.559

Torsional Stiffness With and Without FES - 2 *Way ANOVA:*

- Main Factor: NO FES vs. FES, p = 0.357
- Main Factor: Directions, p<0.001
- Interaction: p = 0.369

- *Both translational and torsional damping are significantly different between directions*
- *Both translational and torsional damping are not significantly different between with and without FES*

Conclusion

- Multidirectional stiffness of the trunk is quantified and an increase in translational stiffness was shown with FES condition.
- These results can be used to develop an FES protocol for individuals with SCI to improve their sitting balance

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References:
[1] K. Masani, et al. Clinical Biomchanics; 24:176-182, 2009
[2] P Lee, et al. Human Factor; 49:100, 2007
[3] AH Vette, et al. ASME J. of Medical Devices; 2: 2008

Center of Pressure Velocity Captures Body Acceleration Rather Than Body Velocity During Quiet Standing

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3) Research Center for Advanced Science and Technology, University of Tokyo, Tokyo, JAPAN
3) Department of Motor Dysfunction, National Rehabilitation Center for Persons with Disabilities, Tokorozawa, JAPAN

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PURPOSE

To investigate the hypothesis that the fluctuation of COPv captures the fluctuation of ACC due to larger derivative gain in the neural control system during quiet standing

BACKGROUND

The center of pressure (COP) velocity (COPv) has been used as one of the most sensitive posturographic measures to detect changes in balance abilities due to aging and/or neurological diseases. Since the COP trajectory matches very well with the center of mass (COM) trajectory due to the body dynamics, the COP velocity has been believed to represent the COM velocity (COMv) in most of these studies. However, there is no study to date that has investigated the physical and physiological meaning of the COP velocity.

Based on the motion equation of an inverted pendulum,

$$COP \approx COM + ACC\frac{I}{mgh} \qquad \text{(eq 1)}$$

where *ACC* = body acceleration, *I, m, h* = body inertia, mass, and height, *g* = gravity coeff. According to this,

$$COPv \approx COMv + ACCv\frac{I}{mgh} \qquad \text{(eq 2)}$$

where *ACCv* = derivative of ACC (= jerk). Thus, COPv can be correlated with COMv.

On the other hand, as *COP* = *mg* * *TQ*, where *TQ* = ankle torque, COP represents the ankle torque, which controls COM during queit standing. Based on the suggested models [1, 2], TQ may be modeled as,

$$TQ \approx Kp(\theta - \tau) + Kd(\dot{\theta} - \tau) + Ki\int(\theta - \tau) + K\theta + B\dot{\theta} \qquad \text{(eq 3)}$$

where theta = body angle, and *Kp, Kd, Ki, K*, and *B* are controller gains (see below). Therefore,

$$dTQ \approx Kp(\dot{\theta} - \tau) + Kd(\ddot{\theta} - \tau) + Ki(\theta - \tau) + K\dot{\theta} + B\ddot{\theta} \qquad \text{(eq 4)}$$

Since it has been suggested that Kd is comparatively large in the control system [2, 3, 4], and considering ACC is similarly sensitive to COPv in detecting aging of postural control [5], Kd term may be dominant in dTQ, and as the result, dTQ and COPv are proportional to ACC.

Experiment

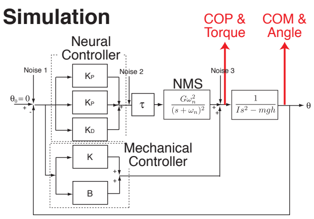
Subjects:
- Young healthy subjects: n = 27 (female 14), 27.2±4.5 yrs
- Elderly healthy subjects: n = 23 (female 12), 66.2±5.0 yrs

Task:
- Quiet standing with eyes open for 90 sec on a force plate, 5 trials

Measurements:
- Center of pressure (COP)
- Sway at L5 using a laser displacement sensor
- AP horizontal force

Analysis:
- The body center of mass (COM) was estimated using the sway at L5.
- The body acceleration was estimated using two ways:
  - AP horizontal force divided by the body mass (ACCf)
  - Second derivative of sway at L5 (ACCl)
- All derivatives (COPv, COMv, ACCfv, ACClv) were also calculated.
- To assess the amount of fluctuation, the root mean square (RMS) was calculated.

Simulation

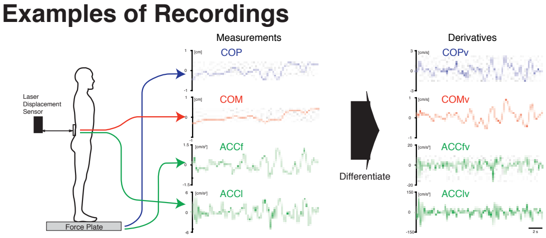


NMS (Neuro-musculo-skeletal system): 2nd order filter

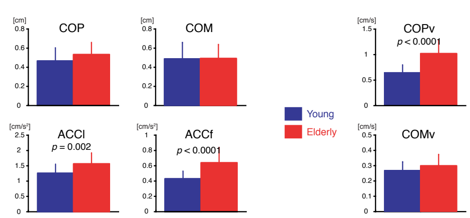
Model:
- Neural Controller: 0<Kp<500, 0<Kd<300, 0<Ki<300
- Motor delay = Sensory delay = 40 ms
- Mechanical Controler: 0<K<1000, B=5
- NMS: 0.1<1/omega<0.2 s
- Time constant of Noise 1, 2: 5, 100, 200 s
- Time constant of Noise 3: 0.025, 0.25, 0.5 s
- Pendulum 1: H=0.852 m, W=580 N, I=44.9 kgm^2
- Pendulum 2: H=0.986 m, W=957 N, I=85.7 kgm^2
- Pendulum 3: H=0.773 m, W=394 N, I=24.7 kgm^2

Stability:
- Nyquist stability with gain (2dB) and phase (5 deg) margins

1 Examples of Recordings

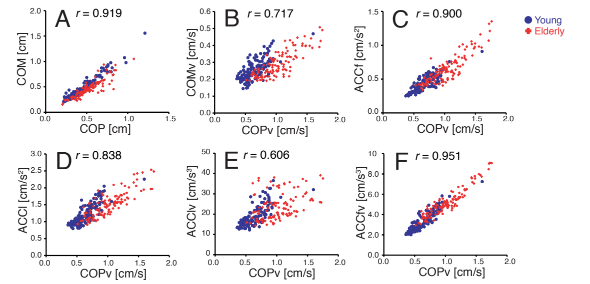


2 Amount of Fluctuation



- The COP, COM, COMv were not significantly different between the young and the elderly.
- The ACC (ACCf and ACCl) and COPv were significantly different between the groups.

3 Correlations between Variables

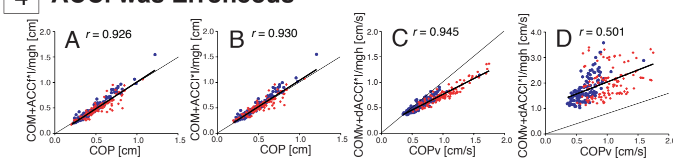


- The COM and COP were highly correlated (A), agreeing with the motion equation (eq 1).
- The COMv and COPv were correlated but not very strongly (B).
- The ACC (ACCf and ACCl) were more strongly correlated with COPv (C and D), agreeing with our hypothesis.
- While COMv and COPv were not highly correlated (B), ACCfv and COPv were very highly correlated (F) (Re: ACCfv (E), see below). This suggests that among the terms in eq 2, ACCv term is dominant in determining COPv.

[illegible]	[illegible]	[illegible]	[illegible]
[illegible]	[illegible]	[illegible]	[illegible]
[illegible]	[illegible]	[illegible]	[illegible]
[illegible]	[illegible]	[illegible]	[illegible]
[illegible]	[illegible]	[illegible]	[illegible]
[illegible]	[illegible]	[illegible]	[illegible]

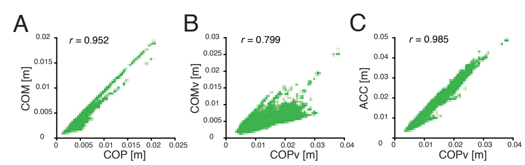
- COPv correlated with ACCf more than with COMv, not only in RMS but also in other parameters quantifying fluctuations.

4 ACCl was Erroneous

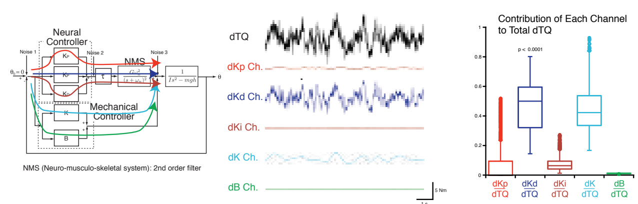


- The COP, COM and ACC satisfied eq 1 (A and B) suggesting that ACC is satisfactorily accurate.
- However, only ACCf satisfied eq 2 (C and D). ACCl may contain a higher level of high frequency measurement errors.

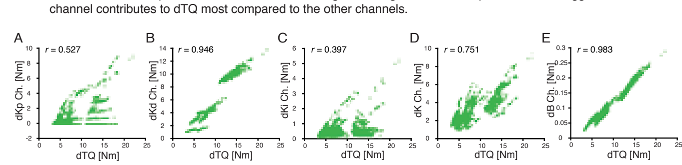
4 Simulation Study



- In the simulation, the same results as the experiment, i.e., 1) COP and COM were highly correlated (A), 2) COPv and COMv were not very highly correlated (B), 3) COPv and ACC were highly correlated (C), were obtained.



- The fluctuation of output from Kd channel was the largest among the terms in eq 4. This result suggests that Kd channel contributes to dTQ most compared to the other channels.



- Among the channels, Kd and B channels more correlated with dTQ compared to the other channels. Considering the contribution of B channel was very low, Kd channel must dominate the fluctuation of dTQ.

CONCLUSION

- The COP velocity fluctuation captures the COM acceleration fluctuation rather than the COM velocity fluctuation.
- The current results suggest that the neural motor command controlling quiet standing posture contains a significant portion that is proportional to the body velocity.

References
[1] Peterka. J Neurophysiol 88:1097-118, 2002.
[2] Masani K, et al. J Neurophysiol 100:1465-1475, 2008.
[3] Masani K, et al. J Neurophysiol 90: 3774-3782, 2003.
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[5] Masani K, et al. Neurosci Lett 422: 202-206, 2007.

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Decomposing the Ankle Torque during Quiet Standing into Active and Passive Components using EMG

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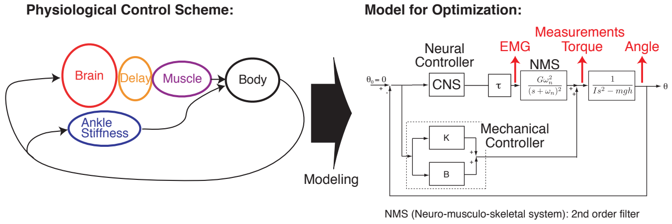
PURPOSE

The purpose of this study was to validate the methodology that we recently proposed, via a systematic simulation study.

BACKGROUND

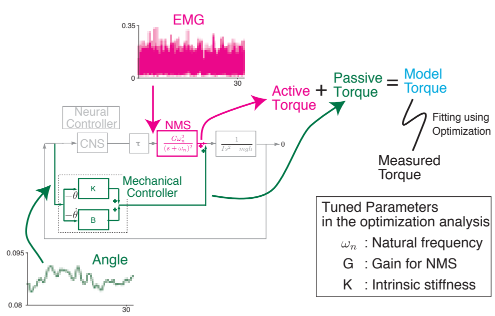
In the control system of quiet standing, the ankle torque generation is critical. The controlled ankle torque can be divided into 2 components: a passive torque component and an active torque component. The passive torque component is generated by the stiffness of the activated muscles, ligaments, and joint structures. The active torque component is generated by the muscle contractile elements, and is neurally controlled moment-to-moment. The passive component during quiet standing has been investigated by applying a quick perturbation at the ankle joint when subjects kept a standing posture. The identified stiffness in such perturbation experiments is about 70 to 90% of the required stiffness (Loram et al. 2002; Casadio et al. 2005). However, it is not certain that the stiffness actually contributes this much to the exerted ankle torque during quiet standing. In our recent study, we introduced a methodology decomposing the ankle torque during quiet standing into active and passive torque components using a simple model (Vette et al. 2010). However, internal noise properties in the control system may affect the outcomes of this approach. In this study, we investigated the practical feasibility of this methodology using a systematic simulation study.

Model (Vette et al. 2010)



NMS (Neuro-musculo-skeletal system): 2nd order filter

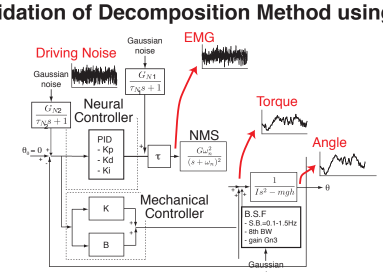
Passive and Active Torque Decomposition Process (Vette et al. 2010)



Tuned Parameters in the optimization analysis
- ω_n : Natural frequency
- G : Gain for NMS
- K : Intrinsic stiffness

- EMG was used as an input to the muscle model, and active torque was obtained as the output.
- Body angle was used as an input to the mechanical controller, and passive torque was obtained as the output.
- Active and passive torque were added, and the model torque was obtained.
- The model torque was fitted with the measured torque using the optimization technique.

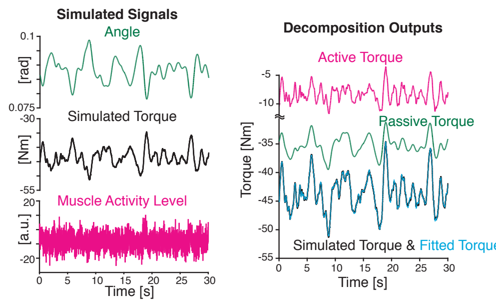
Validation of Decomposition Method using Simulation Study



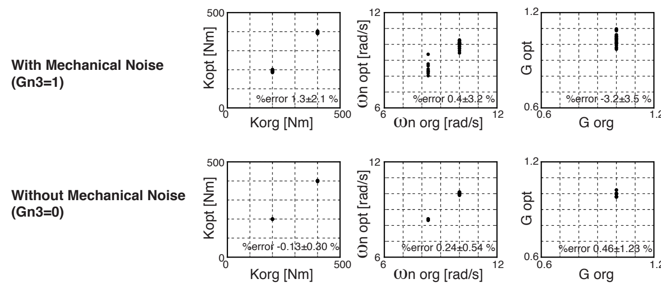
Simulation Parameters:
- Kp = 0 - 500 w step 200
- Kd = 0 - 300 w step 200
- Ki = 0 - 800 w step 200
- K = 0 - 500 w step 200
- B = 5
- ωn = 6.25 - 10
- G = 1
- (Gn1, τn1) = (5, 1) (100, 40) (200, 80)
- (Gn2, τn2) = (0.025, 1) (0.25, 40) (0.5, 80)
- Gn3 = 0, 1

- Postural sway during quiet standing was simulated, and the decomposition method was applied to the simulated time series.
- Mechanical noise simulated and slightly larger than the noise effect of hemodynamics described by Conforto et al. 2001, i.e., 0.4 N of the horizontal force.

Decomposition Results



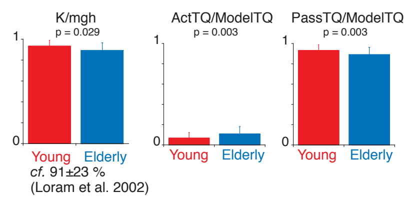
Simulation Parameters:
- Kp = 200
- Kd = 200
- Ki = 200
- K = 400
- B = 5
- ωn = 8.33
- G = 1
- Gn1 = 200, τn1 = 80
- Gn2 = 0.025, τn2 = 1
- Gn3 = 1, τn3 = 80



- The fitting exhibited good results. The errors of the identified parameters were quite small, especially when the mechanical noise was not present.
- The decomposition method detected K and omega used in the simulations with a sufficiently small error, suggesting that the decomposition method is valid and practically feasible.

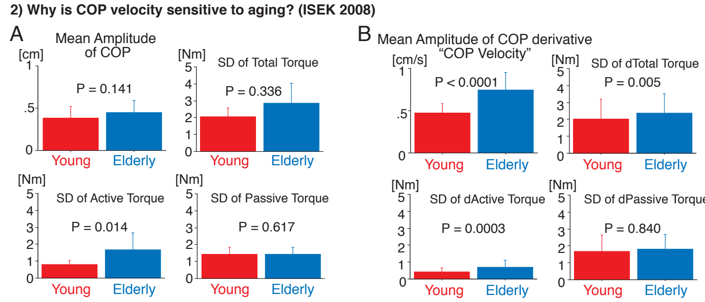
Examples of Application

1) Why do the elderly sway larger? (SfN 2008)



- Ankle stiffness agreed with the previous study using a perturbation. The ankle stiffness was significantly smaller in the elderly than in the young.
- The contribution of passive torque was dominant in the model torque, and lower in the elderly than in the young.
- The elderly tend to have lower stiffness, and as such rely on the active torque more, which makes the elderly less stable.

2) Why is COP velocity sensitive to aging? (ISEK 2008)



- The difference of SD of active torque between the groups was detected by the COP Velocity and the derivative of total torque (B), while COP amplitude nor SD of total torque failed to detect the difference (A).
- The results suggest that the COP Velocity is a sensitive measure in detecting changes of active torque fluctuation induced by aging, since the contribution of active torque is larger in the torque derivative than in the torque itself.

CONCLUSION

- The decomposition method detected K and omega used in the simulations with a sufficiently small error, suggesting that the decomposition method is valid and practically feasible.

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The views expressed in this poster do not necessarily reflect those of any of the granting agencies.

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Made in Canada

Controlling Balance during Quiet Standing via Co-Contraction in the Elderly May Not Be Advantageous

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812.16

PURPOSE

To compare postural sway during the periods with and without the co-contraction between plantarflexors and dorsiflexor, to investigate the role of the co-contraction frequently observed in the elderly during quiet standing

BACKGROUND

In the research field of human bipedal stance, many studies have focused on the control mechanism of the ankle joint torque during quiet standing, since the ankle joint has the primary role of maintaining center of mass (COM) equilibrium during standing. Due to the fact that the COM is in front of the ankle joint during the natural standing posture, a gravity-induced torque continuously accelerates the body forward from the upright position. Therefore, a corrective ankle extension torque is continuously required to resist the gravity effect and to ensure that the COM remains close to the equilibrium position.

The ankle extensors are responsible for generating the corrective ankle extension torque, since the ankle extensors show continuous activity whereas the ankle flexors are silent or only intermittently active in most of cases. However, it has been reported that the elderly use tibialis anterior (TA) for longer duration and with larger amount during quiet standing, compared to the young (Laughton et al., 2003; Panzer et al., 1995). As the plantarflexors are always active, the ankle joint is stabilized by co-contracting the plantarflexors and TA when TA is active. It remains unclear if this co-contraction precludes greater postural instability or if it is a compensatory response to other factors.

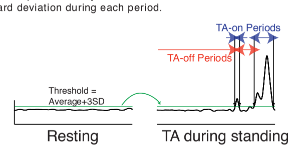
Experiment

Subjects:
- Young healthy subjects: n = 27 (14 females), 27.2±4.5 yrs
- Elderly healthy subjects: n = 23 (12 females), 66.2±5.0 yrs

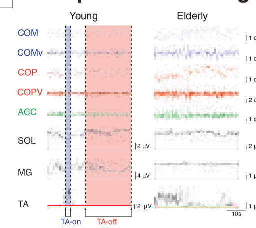
Task:
- Quiet standing with eyes open for 90 sec on a force plate, 5 trials
- Sitting rest for 30 sec

Measurements:
- Center of pressure (COP)
- AP horizontal force
- EMGs for soleus (SOL), medial head of gastrocnemius (MG) and tibialis anterior (TA)

Analysis:
- The body cener of mass (COM) and its velocity (COMv) were estimated using Gravity Line Projection Method (Zatsiorsky et al. 1998).
- The velocity of COP (COPv) was also calculated.
- The body acceleration (ACC) was estimated by dividing the AP horizontal force by the body mass.
- After low-pass filtering EMGs with a cutoff frequency of 1 Hz, periods when TA was active (TA-on) and inactive (TA-off) were identified.
- To idenfity these periods, we used several thresholds based on TA activity during sitting rest. The thresholds were the average of sitting rest + 3, 5, 7, 9, 11 SD. In this poster, we showed the results with sitting rest + 3 SD in most cases.
- The amount of postural sway and muscle activities were determined for TA-on/off using the standard deviation during each period.

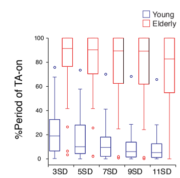


1 Examples of Recordings



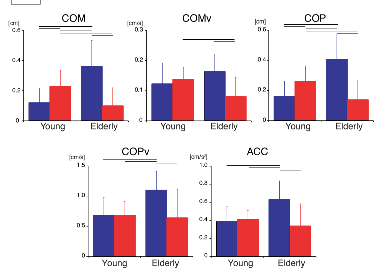
- The postural sway was larger in the elderly compared to the young.
- The periods for TA-on/off were identified and the fluctuations of the postural sway and muscle activities were quantified.

2 Durations of TA-on/-off



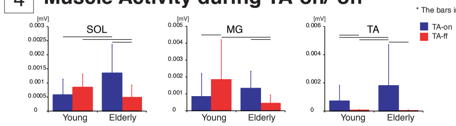
- The elderly showed longer activity in TA irrespective of the type of activity identification threshold used.
- TA-on period was approximately 80-90 % (Median) in the elderly, and approximately 10-20 % in the young.

3 Postural Sway during TA-on/off



- There were significant differences between the groups and between TA-on and -off as well as the interactions by 2-way RMANOVA ($p < 0.05$), in all measures except for the difference between the groups for COMv ($p = 0.494$).
- The elderly showed more sway on all postural sway measures during TA-on, while the young showed less sway.

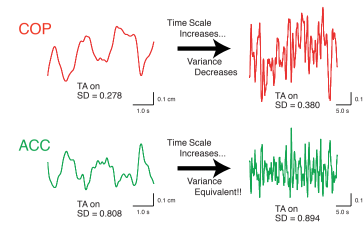
4 Muscle Activity during TA-on/-off



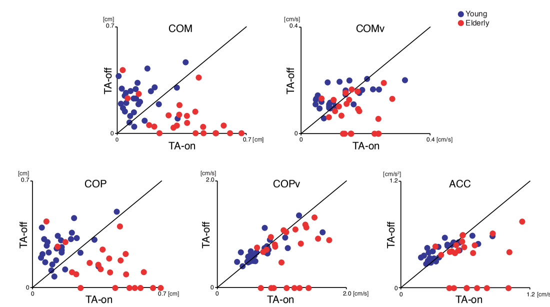
- For SOL and MG, there were significant interactions by 2-way RMANOVA ($p < 0.05$) , while there were no significant main effects. For TA, there was a significant difference between TA-on and -off.
- Larger fluctuations during TA-on for the elderly suggest that the additional activity in the TA may induce increased activity in the plantarflexors related to postural corrections.

5 Effect of Nonstationarity

- The COM and COP have an inherent characteristic, i.e., it shows larger variance with longer duration (not stationary for a short duration).
- Therefore, this result may depend on the different total durations between TA-on and TA-off.
- Unlike COMv, COPv and ACC are more stationary.



6 Comparison of Postural Sway between TA-on and -off



- For more stationary measures (COMv, COPv, and ACC), there is a tendency that they fluctuate more during TA-on in both groups, especially, for the subjects with a larger sway.
- This result implies that the co-contraction of the ankle joint muscles increases the postural sway more in comparison to the plantarflexors' activity alone, which could be a cause of larger postural sway in the elderly.

CONCLUSION

- The co-contraction frequently observed during quiet standing in the elderly does not successfully preclude the increased postural sway. Rather, the co-contraction induces further postural sway probably due to the increment of controlling parameters.
- The elderly show the co-contraction as compensating reactions rather than to reduce the postural sway.

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Decrease in Venous Return due to Postural Change is Mitigated by Functional Electrical Stimulation for People with Spinal Cord Injury

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INTRODUCTION

1) Orthostatic hypotension (OH) is caused by gravitational venous pooling and a subsequent decrease of the circulating blood volume.
2) OH can induce symptoms of cerebral hypoperfusion (e.g., headache, lightheadedness, and dizziness), which can disable people from withstanding sitting or standing.
3) People with spinal cord injury (SCI) are susceptible to OH because of their impaired sympathetic cardiovascular control.
4) Functional electrical stimulation (FES) generates dynamic muscle contractions that mimic the skeletal muscle pump.
5) Cyclic passive leg movements can increase stroke volume in people with SCI.

PURPOSE

To examine the effects of FES and passive stepping on the cardiovascular response of people with SCI under orthostatic stress.

HYPOTHESES

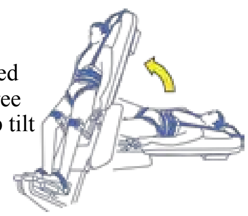
1) FES and passive stepping should independently induce venous return and mitigate a decrease in arterial pressure.
2) FES with passive stepping should be more effective than FES or passive stepping alone.

MATERIALS AND METHODS

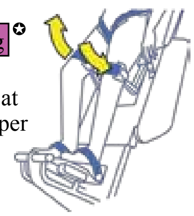
Experimental Conditions:

HUT = Tilt*
STEP = Tilt + Stepping✪
IFES = Tilt + FES✜
DFES = Tilt + Stepping + FES

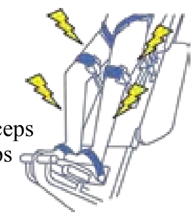
Tilt* Sustained 70-degree head-up tilt



Stepping✪ Passive stepping at 40 steps per minute



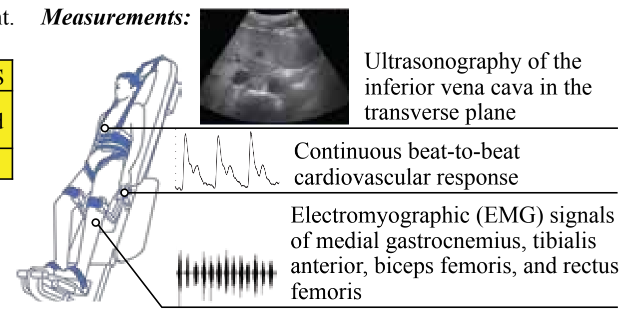
FES✜ Bilateral FES of tibialis anterior, hamstring, quadriceps femoris, and triceps surae muscle



Protocol: The order of these conditions was randomized for each participant.

Task	Rest	HUT	Rest	STEP	Rest	IFES	Rest	DFES
Position	Supine	Tilted	Supine	Tilted	Supine	Tilted	Supine	Tilted
Duration [Minutes]	10	10	10	10	10	10	10	10

Measurements:

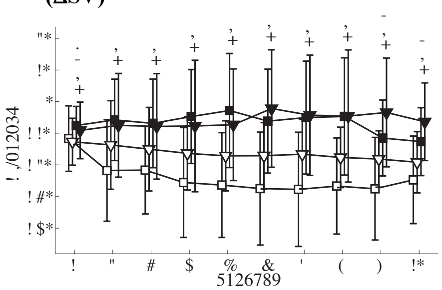


Participants:

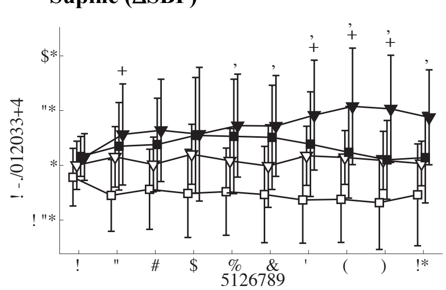
- 10 participants (6 males)
- 43.5±10.8 years old, 172±11 cm, and 87.0±20.6 kg
- SCI at or above T6 level (T6 to C4)
- 9.90±8.56 years post injury
- AIS: 5 A, 3 B, 1 C, and 1 D

RESULTS

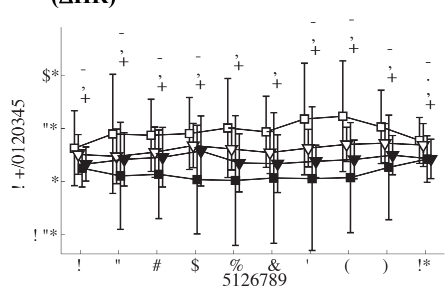
1 Change in Stroke Volume from Supine (ΔSV)



2 Change in Systolic Blood Pressure from Supine (ΔSBP)



3 Change in Heart Rate from Supine (ΔHR)



□ = HUT ▽ = STEP ■ = IFES ▼ = DFES

4 Main Effects and Interactions

Cardiovascular Parameter	FES		Passive Stepping		FES × Passive Stepping	
	$F_{(1,9)}$	p	$F_{(1,9)}$	p	$F_{(1,9)}$	p
ΔSV	15.6	0.003*	2.35	0.159	1.31	0.282
ΔSBP	14.3	0.004*	10.9	0.009*	0.542	0.480
ΔHR	7.07	0.026*	0.089	0.772	2.34	0.161

CONCLUSIONS

1) FES was more effective than passive stepping in maintaining stroke volume (SV).
2) FES and passive stepping maintained SV probably by inducing venous return, as indicated by ΔSBP.
3) FES and passive stepping did not interact synergistically.

The work presented in this poster is under review for publication as "Cardiovascular Response of Individuals with Spinal Cord Injury to Dynamic Functional Electrical Stimulation under Orthostatic Stress" in *IEEE Transactions on Neural Systems and Rehabilitation Engineering* (submitted Nov. 15, 2011).

ACKNOWLEDGMENT

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The Associations Between Aerobic Capacity and Arterial Stiffness in People with Chronic Spinal Cord Injury

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Abstract

People living with spinal cord injury (SCI) are a vulnerable population prone to coronary artery disease (CAD) resulting in high morbidity and mortality. Arterial stiffness assessed by pulse wave velocity measured between the common carotid and femoral arteries (cfPWV) is an established independent predictor of CAD morbidity and mortality in the able-bodied population. The European Society of Hypertension and the European Society of Cardiology have defined cfPWV values ≥1200 cm/s as a diagnostic threshold for subclinical vascular end-organ damage. Prior studies have demonstrated that aerobic capacity is inversely associated with arterial stiffness in able-bodied people. However, the relationship between arterial stiffness, particularly elevated cfPWV values (≥1200 cm/s), and aerobic capacity in people with SCI has not been described.

Objective: To explore the associations between aerobic capacity (Vo_2peak) and elevated cfPWV values (≥1200 cm/s) among adults with chronic SCI.

Methods: Forty-one men and women with chronic SCI (C2-T12; AIS A-D; 24 paraplegics and 17 tetraplegics; time post injury: 14.7±11.4 yrs; Age: 49.4 ±13.1 yrs; Height: 173.2 ± 9.8cm; and Weight: 81.7 ± 17.6 kg) participated in the study. cfPWV was measured using two Doppler flowmeters and Vo_2peak was measured by arm ergometery. Subjects were stratified into two groups according to cfPWV values (<1200 cm/sec: low cfPWV, ≥1200 cm/sec: high cfPWV). Logistic regression analysis was used to determine the contribution of aerobic capacity after adjustment for confounders (age, gender, metabolic syndrome, smoking status, Neurologiclevel of injury (NLI), and ASIA Impairment Scale (AIS) to High cfPWV. **Results:** Aerobic capacity was significantly and negatively associated with cfPWV after adjustments with other risk factors (r=-0.385, p<0.045). After correction for other confounding parameters, aerobic capacity was an independent predictor for high cfPWV (odds ratio = 2.20, 95% CI=1.06– 4.59, p<0.034). **Conclusion:** Aerobic capacity is an independent predictor of high cfPWV values (≥1200 cm/sec) among people with chronic SCI. Further research is needed to explore whether improvements in aerobic capacity will reduce arterial stiffness and adverse cardiac outcomes among people with chronic SCI.

Objective

To explore the associations between aerobic capacity (Vo_2peak) and the diagnostic threshold for subclinical vascular end-organ damage (defined by cfPWV values ≥1200 cm/s)[1] among adults with chronic SCI.

Methods

Data Collection:

Standard Measures

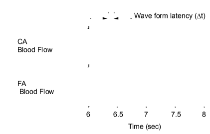
- cfPWV (See cfPWV measurements section)
- Aerobic capacity: VO_2peak (Arm ergometry, Work rate: increased at 30 kpm/min or 60 kpm/min at one minute intervals)
- Smoking status
- Metabolic syndrome:
  Presenting any three of the following five risk factors: 1) elevated waist circumference: ≥102cm in males, ≥88cm in females; 2) elevated triglycerides: ≥1.70 mmol/L or drug treatment for elevated triglycerides; 3) reduced HDL-C: <1.03 mmol/L in males, <1.32 mmol/L in females or drug treatment for reduced HDL-C; 4) elevated blood pressure: SBP ≥130mmHg or DBP ≥85mmHg or antihypertensive drug treatment; 5) elevated fasting glucose ≥5.6mmol/L or drug treatment for elevated glucose (defined in accordance with National Heart, Lung, and Blood Institute/American Heart Association)

SCI Specific Measures

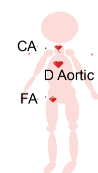
- Duration of injury (years)
- Age at injury (years)
- Neurologic Level of Injury (NLI: aboveT1: Tetraplegia, below T1: Paraplegia)
- ASIA Impairment Scale (AIS)

cfPWV Measurements:

1) Pulse wave latency (Δt) is measured simultaneously using two Doppler flow meters at common carotid (CA) and femoral artery (FA)



2) The distance (D) between the landmarks is measured over the surface of the body.



3) Calculation of cfPWV :

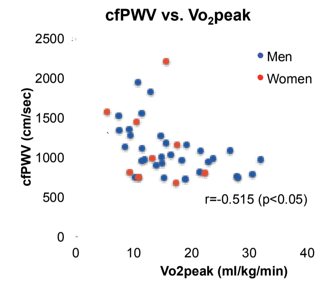
$$cfPWV\ (m/sec) = D_{Aortic} / \Delta t_{CA-FA}$$

Index of subclinical vascular end-organ damage (high cfPWV)
(The European Society of Hypertension and the European Society of Cardiology. 2007)
cfPWV (m/sec) ≥1200cm/sec

Participants:

	Low cfPWV group (n=30)	High cfPWV group (n=11)	p-value	Significance by t-test or chi² test (p<0.05)
Age (years)	59.5 ± 9.3	45.7 ± 12.4	0.002	*
Gender			0.795	NS
Men	23	8		
Women	7	3		
Height (cm)	175.7 ± 10.3	172.3 ± 9.7	0.329	NS
Weight (kg)	93.8 ± 18.1	77.3 ± 15.5	0.006	*
BMI (kg/m²)	30.3 ± 4.5	26.1 ± 5.1	0.023	*
aPWV (cm/s)	1570 ± 301	917 ± 153	< 0.001	*
Duration of Injury (years)	13.1 ± 14.1	15.3 ± 10.4	0.587	NS
Age at Injury (years)	46.3 17.4	30.4 14.2	0.005	*
Level of Injury			0.081	NS
Tetraplegia	10	7		
Paraplegia	20	4		
AIS			0.448	NS
A,B and C	25	8		
D	5	3		
Currently Smoke			0.167	NS
Yes (N)	9	1		
No (N)	21	10		
SBP (mmHg)	111.0 ± 19.9	116.0 ± 12.4	0.401	NS
DBP (mmHg)	69.9 ± 12.6	73.5 ± 10.6	0.358	NS
HR (bpm)	75.4 ± 13.3	68.6 ± 12.1	0.132	NS
Waist Circumference (cm)	113.9 ± 30.7	92.9 ± 13.9	0.004	*
VO_2 peak (ml/kg/min)	10.4 ± 3.1	18.1 ± 6.6	0.001	*
HDL Cholesterol (mmol/L)	1.1 ± 0.3	1.3 ± 0.3	0.160	NS
LDL Cholesterol (mmol/L)	2.3 ± 0.5	2.8 ± 0.8	0.084	NS
Total Cholesterol (mmol/L)	4.0 ± 0.8	4.6 ± 1.0	0.091	NS
Total Cholesterol/HDL	4.0 ± 1.3	3.9 ± 1.1	0.921	NS
Triglycerides (mmol/L)	1.9 ± 1.8	1.3 ± 0.6	0.174	NS
Fasting Blood Glucose (mmol/L)	5.7 ± 1.6	5.0 ± 0.6	0.060	NS
Metabolic syndrome			0.028	*
Yes (N)	4.0	5.0		
No (N)	26.0	6.0		

Results



cfPWV was significantly correlated with Vo_2peak. This remained significant after correcting for age, gender, NIL (tetraplegia or paraplegia), AIS (A,B,C or D) and metabolic syndrome (partial r=0.385, p< 0.05).

Adjusted Relative Risk (RR) for High cfPWV by Vo_2peak

	Vo_2peak (for each 1ml/kg/min decrease)		
Adjustments	RR	95%CI	P
Age and Gender	1.47	1.06 - 2.04	0.020
Above + metabolic syndrome and smoking	2.01	1.13 - 3.59	0.018
Above + NLI and AIS	2.20	1.06 - 4.59	0.034

According to the logistic regression analysis, after controlling age, gender, metabolic syndrome, smoking, NLI and AIS, Vo_2peak was a significant predictor of the presence of high cfPWV cfPWV (RR = 2.20, 95% CI=1.06– 4.59, p<0.034).

Conclusions

Aerobic capacity is an independent predictor of high cfPWV values (≥1200 cm/sec) among people with chronic SCI. Further research is needed to explore whether improvements in aerobic capacity will reduce arterial stiffness and adverse cardiac outcomes among people with chronic SCI.

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Acknowledgements

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EMG-onset Detection using Change-point Analysis

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Introduction

Detecting onset of muscle contraction from an electromyography (EMG) signal recording is an important task for applications in neurology, psychophysiology and EMG-controlled neuroprosthetic devices development.

Change-point detection problem is to detect abrupt changes in statistical properties of a signal.

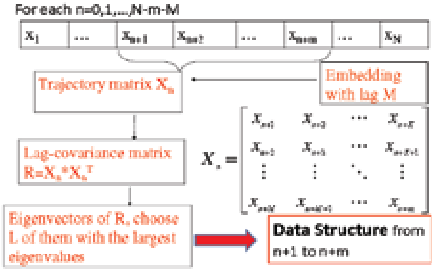
EMG onset is a change-point!

Muscle activation can represent such a change in EMG signal. We apply the change-point detection method [1] to EMG signals to investigate its usefulness for the detection of onset of muscle activity and compare its performance to other common onset detection approaches.

Change-Point Detection Process

I. Trajectory Matrix

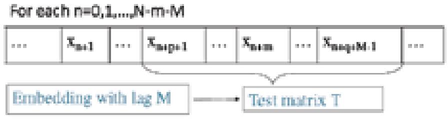
Choose window width m and the lag parameter M, such that M<=m/2. Set K=m-M+1.
For each n=0,1,...,N-m-M



$$X_n = \begin{bmatrix} x_{n+1} & x_{n+2} & \cdots & x_{n+K} \\ x_{n+2} & x_{n+3} & \cdots & x_{n+K+1} \\ \vdots & \vdots & \ddots & \vdots \\ x_{n+M} & x_{n+M+1} & \cdots & x_{n+m} \end{bmatrix}$$

II. Test Matrix

Pick values of parameters p and q, recommended values are p=K, and q=m+1
For each n=0,1,...,N-m-M



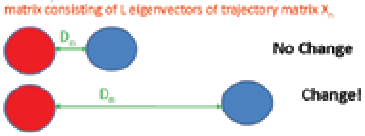
$$T_n = \begin{bmatrix} x_{n+p+1} & x_{n+p+2} & x_{n+p+3} & \cdots & x_{n+q} \\ x_{n+p+2} & x_{n+p+3} & x_{n+p+4} & \cdots & x_{n+q+1} \\ \vdots & \vdots & \vdots & \ddots & \vdots \\ x_{n+p+M} & x_{n+p+M+1} & x_{n+p+M+1} & \cdots & x_{n+q+M-1} \end{bmatrix}$$

III. D_n Statistic

Compute $D_n(T_n)$, the sum of squared distances between the space defined by columns of test matrix and the space defined by eigenvectors of trajectory matrix

$$D_n = \sum_{j=p+1}^{q} \left(\left(T_j^{(n)}\right)^T T_j^{(n)} - \left(T_j^{(n)}\right)^T U U^T T_j^{(n)} \right)$$

where $T_j^{(n)}$ are the columns of the test matrix T_n, and U is a matrix consisting of L eigenvectors of trajectory matrix X_n



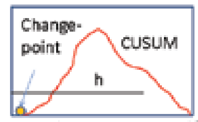
IV. CUSUM Statistic

To find precise locations of change-points compute:

v_n=estimator of D_n at time intervals where no change is occurring (e.g. beginning of signal).

$S_n = D_n/v_n$

CUSUM (cumulative sum) Statistic:



$$W_1 = S_1,$$

$$W_{n+1} = \max\left[0, W_n + S_{n+1} - S_n - 1/\left(3\sqrt{M(q-p)}\right)\right]$$

If W_n exceeds threshold $h = \frac{2t_\alpha}{M(q-p)}\sqrt{\frac{1}{3}(q-p)*(3M*(q-p)-(q-p)^2+1)}$ (where t_α is the $(1-\alpha)$-quantile of the standard normal distribution) then the change-point estimate is a first point with non-zero value of W_n before reaching this threshold.

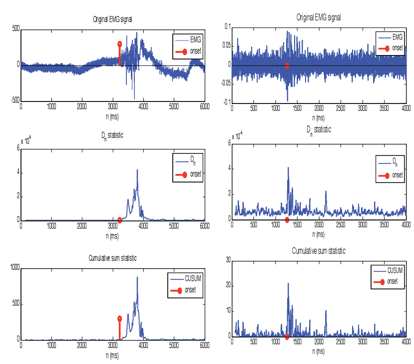
Methods Overview

Three methods of onset detection used for comparison with change-point method were: (1) visual detection by three specialists in EMG processing, (2) Hodges and Bui threshold-based algorithm [2], and (3) Donoho's wavelet-based denoising [3] followed by Hodges and Bui algorithm.

The EMG data was recorded from the extensor carpi radialis muscles, and from the muscles maintaining trunk stability when seating. Details in [4]

Results

Sample successful EMG onset detection by change-point analysis for the wrist and trunk muscles are shown in Figure 1.



A B

Figure 1: Sample results for onset detection in A) wrist muscle EMG, B) trunk muscle EMG. Circle marks detected onset

To evaluate the quality of the EMG onset estimates, the average absolute differences between the three visual estimates and each of the computer estimates were computed for all used EMG segments within datasets and compared using Kruskal–Wallis ANOVA with 5% significance level. Results in Figure 2.

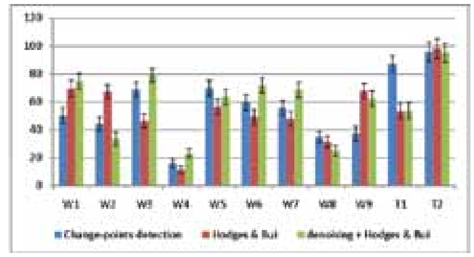


Figure 2: Median ranks of average absolute differences (smaller ranks -> better performance for 3 computer methods. Results shown for 9 wrist and 2 trunk EMG datasets .

Discussion

The change-point detection algorithm can be applied directly to raw signal, automatically "denoises" the signal, and does not require *a priori* knowledge of signal properties.

The main drawback of change-point detection is that it cannot identify what all the detected changes are due to. For example, it cannot distinguish between the changes due to tremor and true muscle activation, or between multiple changes in the signal, which are not due to muscle activation (common in trunk EMG data), based on the magnitude of detection statistic only. Use of change duration in addition to detection statistic's magnitude improves this ability.

Conclusions and Future Work

The change-point analysis method can be used to detect muscle activation in EMG signals. An application to detection of bursts of intermittently active muscle within the recorded continuous activity of other muscles is currently investigated. Additional detection statistics processing to improve the performance is also investigated.

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The views expressed in this poster do not necessarily reflect those of any of the granting agencies.

A Novel Assistive Device for Cursor Control

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A Combined Eye-Gaze Tracking and Brain-Computer Interface System

Background

Target Population
- Severely motor impaired and nonverbal individuals with intact cognitive abilities (e.g., victims of severe spinal cord injury or advanced amyotrophic lateral sclerosis)

Table 1: Common Access Modalities

Access modality	Resolution	Data transfer rate	Risk
Mechanical switch	Low	Low	Low
EMG-based	Medium	Medium	Low
Eye-gaze tracking	High	High	Low
Non-invasive BCIs	Low	Low	Low
Invasive BCIs	High	High	High

The EGT-BCI System

- Eye-gaze tracking is the most efficient modality for cursor position control, but the use of "dwell time" to generate click commands can result in unintentional actions (this is known as the "Midas' touch" problem)
- A brain-computer interface may provide a solution to the "Midas' touch" problem; it can also be used by severely disabled individuals

Eye-gaze tracker (EGT)
- Captures images of the user's eyes using a video camera and two near-infrared light sources
- Uses a mathematical model and eye features extracted from the images to calculate the user's point-of-gaze on a computer screen

Brain-computer interface (BCI)
- Uses a surface electrode to receive EEG signals from the user's left motor cortex
- The user generates an "activation" event using motor imagery of the right hand

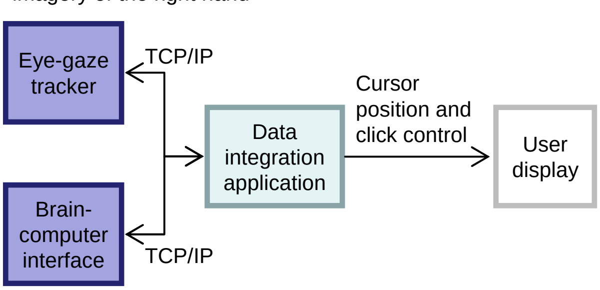

Fig. 1 – The input from the EGT and BCI are integrated to control a computer cursor.

Proof-of-Concept Study

Materials and Methods
- 1 participant (male, able-bodied, age 24)
- Measured typing speed and degree of control by asking the participant to type a sentence
- Used WiViK on-screen keyboard (http://www.wivik.com/) with two different cursor control approaches:
  (1) EGT-controlled cursor positioning and BCI-controlled mouse clicks (EGT-BCI system)
  (2) EGT-controlled cursor positioning and "dwell time"-controlled mouse clicks (EGT system only)

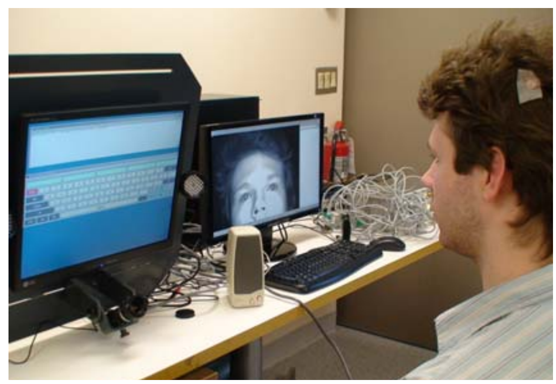

Fig. 2 – The participant controls the position of the cursor with his gaze and generates "click" commands by invoking motor imagery.

Results
- The EGT-BCI system resulted in a lower typing speed but allowed the participant a greater degree of control (assessed subjectively)

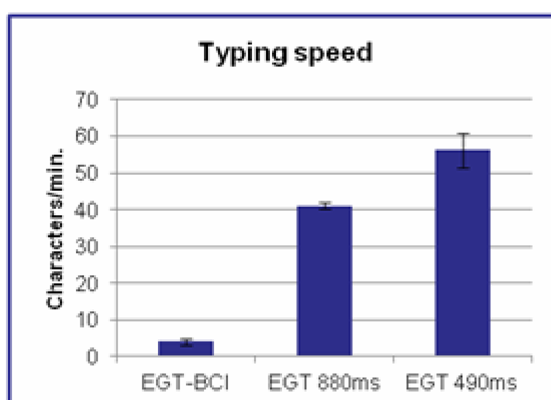

Fig. 3 – The participant typed a sentence using the EGT-BCI system, as well as the EGT system with 2 different dwell time settings.

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Partners:

Estimation of Arm Kinematics Using ECoG Recording

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Introduction and Method

Brain–Computer Interfaces (BCIs) interpret brain activities to user's intent in order to control an external actuator or computing device. Because these systems do not depend on peripheral nerves and muscle activity, they can be used by individuals with severe motor disabilities caused by amyotrophic lateral sclerosis, brainstem stroke or other neuromuscular diseases that make it impossible to move or to express themselves.

A BCI system identifies and decodes the brain activity related to the behavioural outcome. While BCI systems can be implemented using virtually any type of brain activity, most often they are implemented with electrical signals from the brain. Electrophysiological (ECoG) signals have already proven to be suitable for classification and decoding of a limited set of discrete arm movements. However, real-life applications often should consider continuous movements.

From a rehabilitation perspective, the ultimate goal is to be able to offer a technological solution whereby disabled individuals without the ability to move voluntarily can control external devices using brain activity. Such an interface must not only allow for high accuracy of control and a high communication rate between user and device, but be relatively non-invasive in its implementation.

In our simple prediction model, we hypothesize that Upper limb position and velocity are encoded into the brain activity and the functional relationship between ECoG recordings from primary motor cortex and upper limb velocity is linear. Thus, the dynamical behaviour of the arm can be predicted solely from ECoG recordings from primary motor cortex.

Two important physiological characteristics of the motor system are taken into consideration. The first is the delay between the cortical activity and the corresponding motor output. The lag represents a transmission delay between cortical motoneuron spike response and emergence of EMG/kinematic activity. We incorporate the delay into the linear model as a fitting parameter. The second physiological characteristic relates to event-related synchronization and desynchronization in the beta and gamma frequency bands. These events are used as indicators of movement initiation and termination.

To evaluate the accuracy of the model, we estimated the arm velocity of a single trial by using the remaining trials as a training set. Training and test data were then permuted. The average Pearson correlation between the estimated velocity and actual limb velocity was 84%. Examples of the predicted velocity profiles are illustrated in Figure (1).

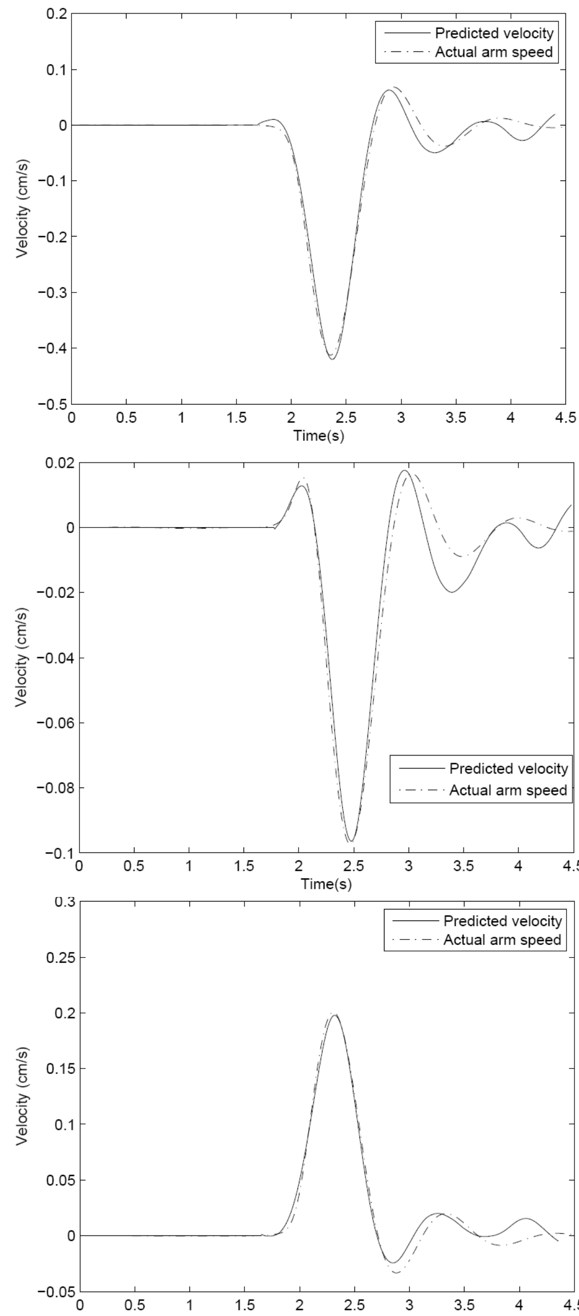

Fig1 – Arm velocity predicted by described multi-linear regression model from four ECoG electrodes is shown by solid line while actual arm velocity is shown by dashed-lines. The graphs show velocities on x-axis, y-axis, and z-axis, respectively.

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Partners:

Using Electrocorticographic Recordings to Estimate Upper Limb Kinematics

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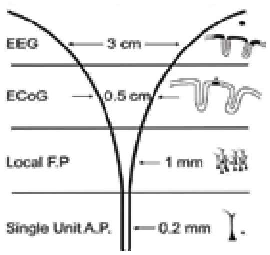
Background and Motivation:

Understanding and decoding human behaviour through electrophysiological means is one of the great challenges of modern science.
Brain–Computer Interfaces (BCIs) interpret brain activities to user's intent in order to control an external actuator or computing device. Because these systems do not depend on peripheral nerves and muscle activity, they can be used by individuals with severe motor disabilities caused by amyotrophic lateral sclerosis, brainstem stroke or other neuromuscular diseases that make it impossible to move or to express themselves.

A class of BCI system involves decoding motion kinematics from brain signals. The system identifies and decodes the brain activity related to the behavioural outcome.

Electrophysiological (ECoG) signals have already proven to be suitable for classification and decoding of a limited set of discrete arm movements. However, real-life applications often should consider continuous movements.
An important challenge to implement a continuous movement classifier is that voluntary arm movements are self-paced (i.e., they can occur at any moment) and

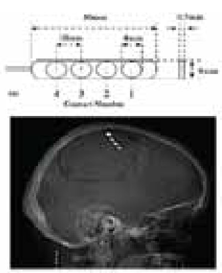


Adapted from Schwartz et al., 2006

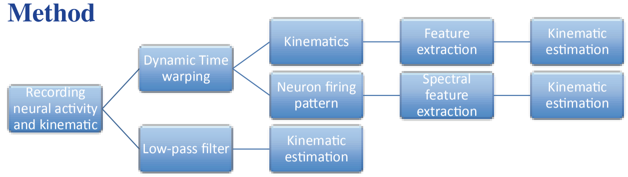
are prone to trail-by-trail variability in terms of initiation time, duration, and speed. We present four methods to adjust the movement variability and align the neural events in time or in time and frequency. Once, the neural events are time aligned, we can identify the neural processes attributed to the movement and uncover the relationship between them and upper limb kinematics.

Material

A 73 year old male with Parkinson's disease and a 65 year old female with essential tremor, participated in this study. The subjects were recruited from the Movement Disorders Clinic of the Toronto Western Hospital. They were medicated at the time of testing. Both participants received a system for direct brain stimulation for the treatment of tremor. This procedure began with the implantation of subdural electrodes. The upper limb movement was recorded by a six dimensional electromagnetic tracker. A sensor was placed over the dorsal aspect of the third metacarpal bone. The position of this point in three-dimensions in addition to its rotation was recorded while the participants were reaching to left, reaching to right, or flexing the wrist.

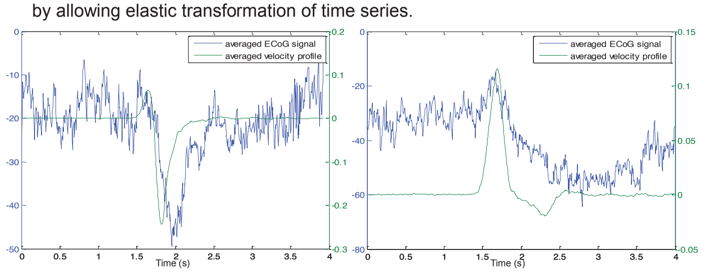


Method



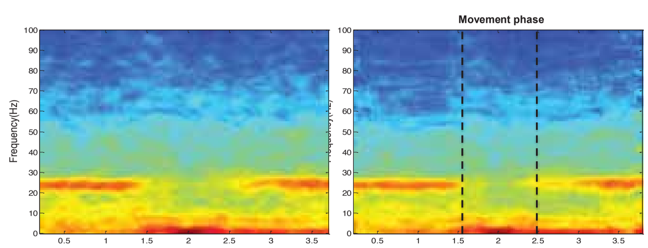
Neural activity temporal alignment

If the movement kinematic is locked to the neural activities, recorded brain activity can be adjusted for temporal variations introduced by how the task was performed. The time-alignment is achieved by elastic deformation of time. The Dynamic Time Warping (DTW) algorithm is extremely efficient as a time-series similarity measure which minimizes the effects of shifting and distortion in time by allowing elastic transformation of time series.



Neural activity spectral warping

The spectral content of neural activity has been shown to be indicative of changes in user's mental state and intention. The time transformation map obtain by warping the kinematics is applied to the spectral distribution of the ECoG signal. The advantage is that warping preserves the spectral content of the signal.

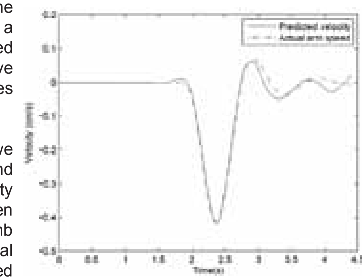


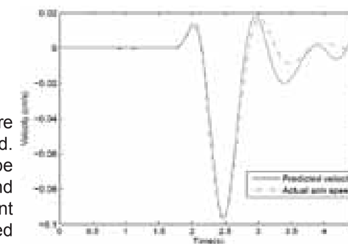
Averaged spectrogram (left) and averaged spectrogram after warping (right).

Estimation of movement kinematic parameters

From a rehabilitation perspective, the ultimate goal is to be able to offer a technological solution whereby disabled individuals without the ability to move voluntarily can control external devices using brain activity.

In our simple prediction model, we hypothesize that Upper limb position and velocity are encoded into the brain activity and the functional relationship between ECoG recordings from M1 and upper limb velocity is linear. Thus, the dynamical behaviour of the arm can be predicted solely from ECoG recordings from primary motor cortex (M1).

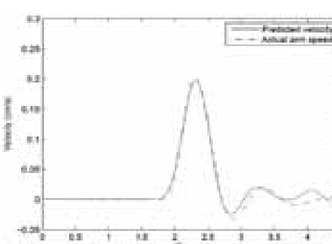




Conclusion

Voluntarily self-paced movements are inconsistent in time duration and speed. Thus, the neural event should be temporally aligned for comparison and statistical analysis purposes. Consistent patterns among the trials are identified to distinguish the movement and decode the arm kinematic parameters from the cortical activities.

We discovered certain functional relationship between the cortical recordings and motion parameters; e.g. position and velocity. Our result showed that the upper limb kinematic parameters can be estimated using only ECoG recordings from four electrodes implanted over the primary cortex. This is while other systems usually consider a grid of more than 64 electrodes to map the brain and estimate the movement kinematic parameters. Using four



recording sites dramatically simplifies the BCI system design and makes the system more feasible from engineering and clinical perspectives.
It is anticipated that the system will be integrated with non-invasive recording methods like EEG. If such paradigm can be deployed and standardized, it opens a path to a new generation of rehabilitation and movement restoration devices.

Acute Effects of Passive Standing and Whole Body Vibration on Arterial Stiffness

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ABSTRACT

The purpose of this study was to investigate the acute effects of Whole body vibration (WBV) with passive standing (PS) on arterial stiffness in able-bodied individuals. Thirteen able-bodied individuals (2 women, age: 24.6±7.2 yrs) performed WBV with PS (WBV-trial) and PS alone (PS-trial) on separate days. The WBV-trial consisted of 5-set of 4-min vibration with inter-set resting periods of 1min each. In the PS-trial, subjects passively stood for 25min. Arterial stiffness (Aortic Pulse Wave Velocity, aPWV) was measured at the baseline, 20min and 60min after both trials. aPWV significantly decreased at 20-min and recovered to the baseline at 60-min in both the WBV-trial, and PS-trial (P=0.01, 2way repeated measure ANCOVA with covariates of heart rate). There was no trial by time interaction. The results suggest that PS itself may be feasible as a therapy to improve arterial stiffness. More research with other WBV protocols are needed to examine the effect of WBV on arterial stiffness.

INTRODUCTION

- Increased arterial stiffness is an established coronary artery disease (CAD) risk among able-bodied people. Exercise reduces arterial stiffness and CAD related mortality
- WBV exercise decreases arterial stiffness (Otsuki et al. 2008) in able-bodied people.
- Among individuals with spinal cord injury (SCI) passive standing (PS) with an aid of standing frame has traditionally been recommended as part of therapeutic exercise programs to maintain the health.
- WBV with PS using a standing frame may be beneficial as an alternate exercise to decrease arterial stiffness in disabled people such as people with SCI.
- The **purpose** of this study was to investigate the effects of WBV with PS on arterial stiffness in healthy young men and women on aortic PWV.

METHODS

Subjects
- Eleven males and two females (Age: 24.6 ± 7.2 yrs; Height: 171.6 ± 7.9 cm; Weight: 64.5 ± 12.2 kg)

Protocol (Figure 1)
- Subjects participated in two sessions (one WBV and 1 PS) for 25 minutes following a 30 minute rest period
- Outcome measures were measured at baseline, 20 minutes post-vibration, and 60 minutes post vibration

Outcome Measures
- Heart rate (HR), systolic blood pressure (SBP), diastolic blood pressure (DBP) were measured. Mean blood pressure (MBP) was calculated as [DBP+ 1/3(SBP-DBP)].
- Aortic pulse wave velocity (aPWV) was measured as an index of arterial stiffness (Tanaka et al. 1998)

WBV
- Modified WAVE Pro platform (Whole-body Advanced Vibration Exercise, Windsor, Canada) with EasyStand 5000 (Figure 2). Subjects were exposed to 5 sets of 4min 25Hz, 1.2mm peak-to-peak displacement vibration with knees extended.

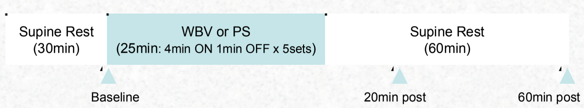

Figure 1. Outcome measures were taken at baseline, 20min post-vibration, and 60min post-vibration.

METHODS

Measurement of aPWV
- Time (Δt) was measured using two identical non-invasive transcutaneous Doppler flow meters (Smartdop50, Hadeco,Inc., Japan) (Figure 3a). The average of at least 20 waveforms were used to calculate Δt.
- Distance (D) was measured over the surface of the body using a metal tape measure (Figure 3b)
- aPWV (cm/sec) = D / Δt

Analysis
- Two-way repeated measures analyses of variance (ANOVA) with two within-subject variables (Condition: PS, WBV; time: 0, 20, 60) were conducted to analyze aPWV
- Analyses of covariance (ANCOVA) adjusting for effects of HR and MBP were also conducted

Figure 2. Subjects were exposed to WBV with a modified WAVE Pro platform fitted with an EasyStand 5000.

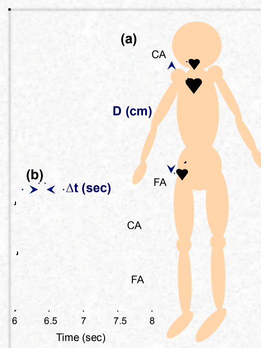

Figure 3. (a) measurement of wave form latency (Δt); (b) measurement of distance (D) between carotid artery (CA) and femoral artery (FA)

RESULTS (BP and HR)

*Table 1. Blood pressure and heart rate before and after trials (Mean±SD). * p<0.05 vs baseline.*

		Baseline	Post 20 min	Post 60 min
SBP (mmHg)	**PS**	119 ± 8	115 ± 8	116 ± 11
	WBV	119 ± 13	117 ± 10	114 ± 7
MBP (mmHg)	**PS**	88 ± 9	85 ± 8	88 ± 10
	WBV	87 ± 12	88 ± 10	83 ± 12
DBP (mmHg)	**PS**	73 ± 11	71 ± 9	75 ± 12
	WBV	72 ± 12	74 ± 11	68 ± 15
HR (beats/min)	**PS**	63 ± 10	60 ± 10	58 ± 7*
	WBV	61 ± 8	58 ± 8	58 ± 7

RESULTS (aPWV)

- A two-way repeated measures ANOVA did not reveal any significant main effects or interaction for aPWV (Figure 4). A main effect for time approached significance (p=0.065).
- When controlling for HR, ANVOCA revealed that there was a significant main effect for time (p=0.006). *Post hoc* analyses revealed that the source of the effect was due to subjects having a significant decrease in PWV 20 minutes following standing whether or not they received vibration (Figure 5). aPWV no longer differed from baseline 60minutes post standing.

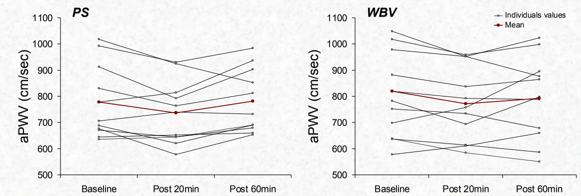

Figure 4. Raw aPWV scores for subjects during the PS (left) and WBV conditions (right).

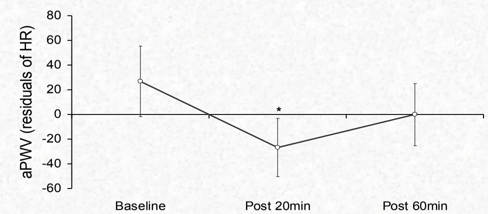

Figure 5. When controlling HR, aPWV was shown to significantly decrease 20min post standing regardless of PS or WBV condition. The effects was no longer present 60min post standing.

DISCUSSION

- WBV + PS and PS alone reduced arterial stiffness, although the mechanisms remains unclear. We speculate that the acute reductions of arterial stiffness after PS may be associated with the arterial functional changes (i.e. endothelial function) due to increased blood volume in lower body.
- More research with other WBV protocol (i.e. Duration, frequency and amplitude) is definitely needed to examine the effect of WBV on arterial stiffness.

CONCLUSION

- WBV + PS and PS alone improved arterial stiffness. The study results suggest that passive standing may be used as a therapy to reduce arterial stiffness.

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Serum 25(OH)D Levels and Tibia Volumetric Bone Mineral Density (vBMD) in Chronic Spinal Cord Injury (SCI)

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Toronto Rehab

Advancing Rehabilitation
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Background

Extensive sublesional bone loss occurs following SCI[1], resulting in a high fragility fracture susceptibility[2]. Current guidelines for determining fracture risk are unsuitable for the SCI population[3]. Identification of risk factors associated with bone loss may help distinguish between individuals with SCI who fracture and those who do not fracture.

Vitamin D intake is a modifiable factor that could change fracture risk after SCI[4]. Peripheral quantitative computed tomography (pQCT) allows for characterization of changes in vBMD and bone structure; however no prior work has explored the relationships between serum vitamin D status and lower extremity tibia vBMD or structure variables in the SCI population.

Purpose

To investigate the relationship between serum 25(OH)D levels and bone health outcomes (vBMD, cortical thickness, and trabecular structure parameters) from pQCT scans of the tibia in individuals with chronic SCI.

Methods

Study Design & Setting: Cross-sectional; Toronto Rehab, Lyndhurst Centre and McMaster University

Participants: 40 adults with chronic SCI (C2-T12 AIS A-D)

Assessment Techniques & Outcomes:

1. Serum 25(OH)D (nmol/L); DiaSorin LIAISON® CLIA
   Vitamin D status was classified as:
   - Sufficient (≥75 nmol/L)
   - Insufficient 2 (50-74 nmol/L)
   - Deficient (<50 nmol/L)
2. Bone quality outcomes; StraTec XCT 2000 pQCT
   - 4% tibia trabecular vBMD (mg/cm^3), connectivity (nodes), mean hole size (mm^2), maximum hole size (mm^2)
   - 66% tibia vBMD (mg/cm^3), cortical thickness (mm)

Analysis: Spearman correlations were used to assess the associations between 25(OH)D and bone quality outcomes. Correlations were described as: no (r=0.0-0.19), weak (r=0.2-0.39), moderate (r=0.4-0.69), strong (r=0.7-1.0) associations and $p<0.05$ was considered significant.

Table 1. Participant Characteristics

	Male	Female	Total
Total n	32	8	40
Age (years)	52 (±11.8)	50 (±14.1)	51 (±12.0)
Motor Complete Paraplegia (n)	16	4	20
Motor Incomplete Paraplegia (n)	3	2	5
Motor Complete Tetraplegia (n)	6	0	6
Motor Incomplete Tetraplegia (n)	7	2	9
Time post injury (years)	15 (±10.8)	15 (±9.3)	15 (±10.0)

Results

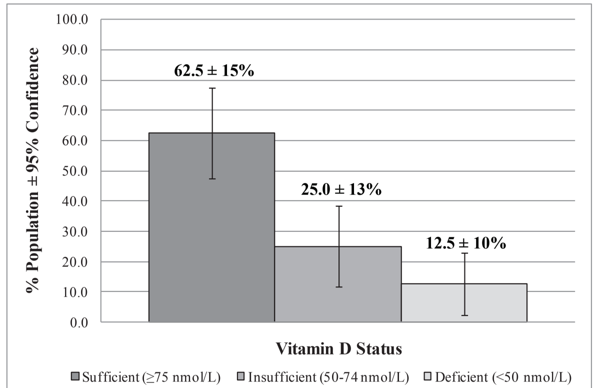

Figure 1. (above) Percent of cohort (N=40) ± 95% CI grouped by vitamin D status.

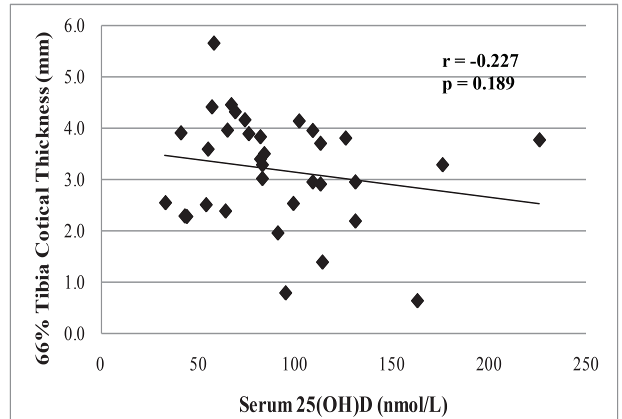

Figure 2. (above) Linear relationship between 66% tibia cortical thickness and serum 25(OH)D level (r=-0.227, p=0.189)

		Spearman Correlation with Serum 25(OH)D		
	Mean (± SD)	N	r	p value
4% Tibia Site				
Trabecular vBMD (mg/cm^3)	142.6 (±51.4)	36	-0.109	0.535
Connectivity Index (nodes/unit of connected network)	0.189 (±0.075)	15	0.036	0.900
Mean Hole Size Between Trabeculae (mm^2)	7.85 (±10.62)	15	0.032	0.910
Maximum Hole Size Between Trabeculae (mm^2)	628.27 (±327.41)	15	0.014	0.960
66% Tibia Site				
Shaft vBMD (mg/cm^3)	1080.3 (±56.8)	34	-0.176	0.311
Shaft Cortical Thickness (mm)	3.18 (±1.08)	34	-0.227	0.189

Table 2. (left) Summary of trabecular and cortical bone health outcomes and Spearman correlations with serum 25(OH)D level

Discussion

- There were no significant associations between serum 25(OH)D level with trabecular vBMD and trabecular structure outcomes in individuals with chronic SCI, contrary to findings in the non-SCI population[5].
- There was a weak negative correlation (r=-0.227, p=0.189) between serum 25(OH)D and 66% tibia cortical thickness, which was not significant.
- Limitations of this study include: small sample size, relative impairment homogeneity, duration of injury bias and single time point of data collection.
- Prospective study of the associations between bone quality and 25(OH)D status over time in a larger cohort of individuals may facilitate identifying those for whom:
  - Therapeutic range vitamin D enhances bone architecture thereby providing fracture protection
  - Insufficient levels enable fracture risk prediction.

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CIHR IRSC

Usability of the Juvent™ and WAVE® Whole Body Vibration Plates for Men with and without Spinal Cord Injury

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[1] University of Waterloo, [2] Toronto Rehabilitation Institute, [3] University of Toronto [4] Arcon Engineering

BACKGROUND

- Whole-body vibration (WBV) is a new therapy used for improving muscle strength, endurance, power and bone mineral density (BMD) in animals, post menopausal women, bariatric persons, elite athletes and children with neurological impairments.
- Although popular in many rehabilitation settings, the usability of WBV among patients with spinal cord injury (SCI) is unknown.

PURPOSE

- To assess SCI and non-SCI subjects' feedback on the usability of the Juvent™ and WAVE® WBV devices using specified knee angles, plate amplitudes and frequencies.

SUBJECTS

- Age: 20–50yrs;Weight:68–115kg; Height;168–188cm
- SCI: n=8 (C4-L2 AIS A-D); Non-SCI: n=10

WHOLE BODY VIBRATION

- Subjects were exposed to *apriori* selected combinations of vibration parameters on the modified WAVE® and Juvent™ WBV devices (See Figure 1).
- Vibration Parameter Combinations

  Knee angle: 140º, 160º

  Frequency: 25, 35 & 45 Hz

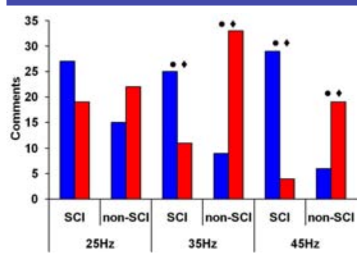
  Device/Amplitude: WAVE High (1.1mm)
  WAVE Low (0.7mm)
  Juvent (0.05mm)

  were presented systematically for two minutes with a one minute rest period

CODING FRAMEWORK

Type			Vibration Descriptors
Positive	Euphoria Pleasantness Easy to see Increased sensitivity	Massage Reduced spasticity Therapeutic Warmth	Optimal Enjoyment Fun Relaxing Smooth
Negative	Blurred vision Discomfort Induced spasms Nausea Worsening pain	Numbness Shaking head Uriticaria Vertigo	Annoyance Bothersome Irritating Pressure from angle Discomfort

CONDITION-SPECIFIC RESULTS



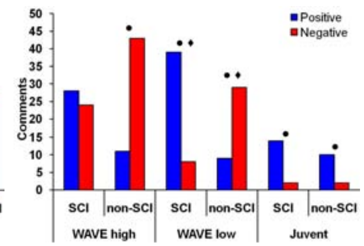
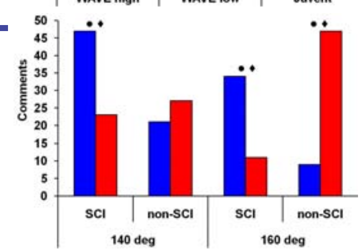


Figure 1. Modified WAVE® (left) and Juvent™ (right) WBV devices



♦ = significant differences between vibration parameters
● = significant differences between SCI and non-SCI

OUTCOME ASSESSMENT

- Subject's experiences during and after vibration exposure were solicited through semi-structured interviews
- Reports were coded as neutral, positive, or negative (Coding Framework; neutral not shown)

OVERALL RESULTS

- There were no overall difference in the number of positive (115) or negative (104) comments reported
- SCI subjects reported significantly more positive statements (81) than negative statements (34
- Non-SCI subjects reported significantly more negative statements (74) than positive statements (30)
- The parameter combinations generating the most negative comments were WAVE high, 140°, 25Hz, and WAVE high, 160°, 25Hz and 35Hz
- The parameter combinations generating the most positive reports were: WAVE low, 140°, 35Hz/45Hz, and WAVE low, 160°, 25Hz

CONCLUSION

- Whereas non-SCI subjects found higher frequency vibrations unpleasant, SCI subjects preferred these vibration parameter combinations. SCI related impairments alter WBV perception; potentially masking adverse effects.
- A intervention study using WBV parameters of WAVE low, 140º, 45Hz is underway based on this usability data suggesting likely adherence in SCI subjects.

The views expressed in this poster do not necessarily reflect those of any of the granting agencies.

Grant #: ONF-SCI-2006-WAVE-445

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Whole Body Vibration and EMG Activation: Effects of Vibration Frequency, Amplitude and Posture

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Toronto Rehab is a teaching and research hospital fully affiliated with the University of Toronto.

Background

- Individuals who sustain a spinal cord injury (SCI) experience muscle atrophy in their lower extremities as a result of denervation and lack of mechanical stimulus below the level of injury. There is a 18-46% reduction in lower extremity muscle cross sectional area six weeks after SCI compared to control.
- Current lower extremity rehabilitation methods include functional electrical simulation, body-weight support treadmill training, gait training, and passive standing.
- Whole-Body Vibration (WBV) is a novel therapy used for augmenting muscle activity and improving muscle strength in the able-bodied population.

Objectives

- To assess the effects of specific vibration parameters on lower extremity EMG activation in individuals with SCI during exposure to passive standing and WBV.
- To describe the effects of knee angle on EMG activation and determine the optimal angle to elicit EMG activation.
- To compare the Wave® and Juvent™ WBV platforms' effectiveness in eliciting lower extremity EMG activity in individuals with SCI.

Methods

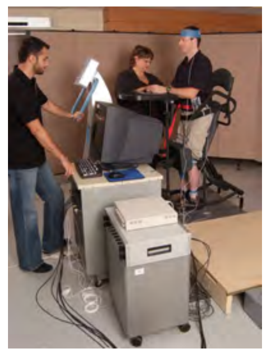
Subjects
- Able-bodied subjects (n=7)
- SCI subjects (n=5)

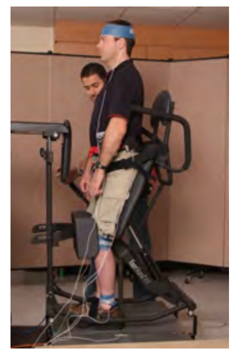
Controlled Parameters
- Vibration Frequency: 25, 35, 45Hz
- Vibration Amplitude: 0.6 & 1.2mm
- Knee Angles: 140°, 160°, 180°

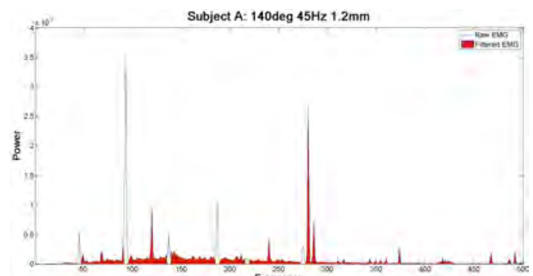
Quantitative analysis:
- Calculate EMG activity during WBV for different conditions.
- Compare magnitude of EMG during WBV and resting conditions.
- EMG activity recorded using bipolar silver-silver chloride surface electrodes on the following muscles (Gain = 2000, pre amplified, 60 Hz rejection ratio, Fs = 2000 Hz):
  - Tibialis Anterior
  - Soleus
  - Gastrocnemius Medialis
  - Rectus Femoris
  - Vastus Lateralis

Signal Filtering

- Custom filters were applied to the raw EMG signal to eliminate motion artifacts and electro-magnetic interferences from the vibration platforms.
- The EMG signal was notch filtered (power = 0) at the vibration frequency and its harmonics.
- The EMG signal was normalized by the # of FFT points after notch filtering.







Discussion

Effect of Frequency on EMG:
- Frequency variations had a significant effect with the WAVE® platform for both AB and SCI subjects for all muscles (except VL in SCI).
- 45 Hz vibration generated the greatest activation of EMG compared to the resting condition.

Effect of Amplitude on EMG:
- Amplitude variations had a significant effect with the WAVE® platform for all muscles in AB subjects and for GM and RF muscles in SCI subjects.
- 1.2 mm vibration elicited greater EMG activation than 0.6 mm with the WAVE® platform.

Effect of Posture on EMG:
- Posture variations did not have a significant effect on EMG activation.
- Greater EMG activation with 140° & 160° knee angles for both platforms.

Comparing Wave® and Juvent™:
- The WAVE® platform augmented EMG activity to a much greater degree than the Juvent™ platform (no activation with SCI subjects on the Juvent™).

Conclusions

- The passive standing and WBV platform configuration elicited eliciting EMG activity in the lower extremity muscles in individuals with SCI and able-bodied subjects.
- The suggested optimal parameters for greatest EMG activation in the lower extremities are 45 Hz, 0.6mm, 140° or 160°, during passive standing with the Wave® platform.
- Due to the higher amplitude of vibration provided by the Wave® platform, it created greater EMG activation compared to the Juvent™ platform, suggesting its greater potential for eliciting EMG activity.

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Development of a Sham Condition for Future Whole Body Vibration Intervention Trials

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Toronto Rehab is a teaching and research hospital fully affiliated with the University of Toronto.

Background

- Placebo effects are be defined as the positive physiological or psychological changes associated with the use of inert medications, sham procedures, or therapeutic symbols within a healthcare encounter. Placebos can also be active substances or real procedures that produce unexpected beneficial effects.
- In randomized control trials using device interventions, it is difficult to create a comparator or control condition similar enough to the intervention to be considered an appropriate or valid control. This is thought by many to be the only way to minimize the effects of bias on rehabilitation trial outcomes.
- Non-drug interventions in spinal cord injury (SCI) rehabilitation research such as acupuncture, magnetic field therapy, and specific exercise therapies lend themselves better to the usage of a placebo or sham condition.
- Whole body vibration (WBV) is a new and novel intervention which has received attention in SCI research for the treatment of sublesional osteoporosis, sarcopenia, obesity, metabolic syndrome and cardiovascular disease.
- Our ability to determine the effectiveness of rehabilitation interventions in phase III & IV trials is dependent upon creation of valid shams/ control conditions.

Objectives

- The purpose of this study was to develop a sham WBV condition.
- We sought to produce a device which is identical in appearance, while replicating two components of the real device including mechanical vibration via the application of vibration to the participant's upper extremities, and similar auditory stimuli during operation.
- Our *a priori* aim was to fool naïve participants 50% of the time.

Preamble

- A customized WAVE Pro vibration platform fitted to an EasyStand 5000 standing frame was used. Qualitative investigation of the device previously conducted at our centre identified the most salient stimuli resulting from exposure:
  - Vestibular
  - Auditory
  - Proprioceptive
- The study team attempted to replicate these stimuli with the sham device

This project was supported by the Toronto Rehabilitation Institute, which receives funding under the Provincial Rehabilitation Research Program from the Ministry of Health and Long-Term Care in Ontario. The views expressed do not necessarily reflect those of the Ministry.

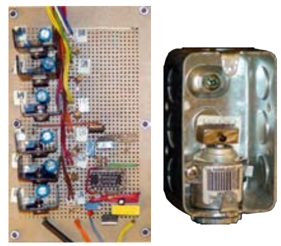

Figure 1. Circuit board controlling the five sham motors simultaneously (left). Motor with offset weight to generate the sham vibration (right).

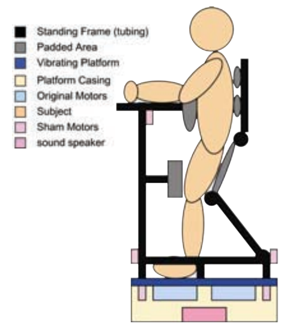

Figure 2. Schematic of the modified vibration device to generate the Sham condition.

Methods

1) Sham Development

Vibration Replication

- Three vibration setups were tested to replicate vibration:
  1) Three gaming joy-stick motors attached to the standing frame.
  2) Two small 3V/1A DC motors connected under the platform.
  3) Five small 3V/1A DC motors with 10g and 15g offset weights connected under platform and to the standing frame.

Sound Replication

- Comparisons between different sound recorders & small size speakers were done.
- We recorded the sound of the vibrating platform for replaying and "controlling" the sham motors.
- A sound speaker was placed under the vibrating platform to mimic the location of the sound source.

2) Evaluation

- The sham device will be evaluated by people with SCI and without SCI to determine whether it is a convincing sham.
- Participants will be exposed to two minutes vibration or sham vibration sequentially in a counterbalanced design.
- Questionnaires gauging certainty of real versus sham vibration will be administered. The rate of sham sufficiency will be reported.
- An additional questionnaire will inquire about loudness and perceived vibration generated across the two conditions.
- Staff administering questionnaires & participants will be blinded to group allocation (sham or regular device) during data collection.

Results

1) Device Development (Figure 2)

- The project team agreed that the optimal sham setup was the use of five small 3V/1A DC motors including three on the frame with 10g weights and two under the platform with 15g offset weights (Figure 1).
- A high definition recording conducted in a sound dampening chamber was used to generate a sound file which was played back using a Bose SoundDock Portable Digital Music System.
- A circuit was created to allow for the independent adjustment and initiation of five motors simultaneously. The circuit was powered a 5V and 12V DC outputs from an ATX computer power supply (Figure 1).

2) Evaluation

- Qualitative evaluation of the device is planned

Discussion

- A preliminary sham prototype has been developed. Plans are underway to prospectively evaluate the sham condition.
- Limitations of the current prototype include:
  - Effects of the surrounding environment on the sound recording
  - The audible difference between the "recorded sound" and the "actual sound" from the running vibrating platform
  - The sound that the small vibrating motor added to the device is audible.
  - Separate control interfaces for the real vibration (original WAVE interface) and sham vibration (custom circuit).
  - Further refinements to the sham device are pending qualitative feedback.

Acknowledgements

- We acknowledge funding from the Ontario Neurotrauma Foundation & the Ministry of Health and Long Term Care. We gratefully acknowledge the technical assistance of Egor Sanin and Elias Guestrin in creation of the Sham Device.

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BILATERAL SOLEUS H-REFLEXES IN HUMANS DURING SITTING AND PASSIVE STANDING: VARIABILITY AND CORRELATION

687.25

DG Sayenko[1], K Masani[1], MR Popovic[1,2]

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INTRODUCTION

The random fluctuation in H-reflex amplitude has been studied previously to understand its characteristics and origins [Nozaki et al., 1995]. It has been suggested that supraspinal centers contribute to the correlation in bilateral H-reflex through pre- and/or postsynaptic inhibition [Nozaki et al., 1996].

To investigate the contribution of presynaptic and postsynaptic mechanisms in the variability and correlation of bilateral H-reflexes, we examined the H-reflexes in both legs during different conditions: sitting and passive standing.

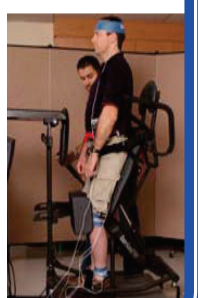
Because the H-reflex is inhibited in standing posture due to supraspinal input via presynaptic inhibition [Koseja et al., 1993], we hypothesized that the reflex magnitude will vary in sitting and standing, whereas its variability will not show significant difference in different postures. Further, if the supraspinal centers are involved in correlation of bilateral H-reflex responses through presynaptic mechanisms, then this correlation may depend on the posture.

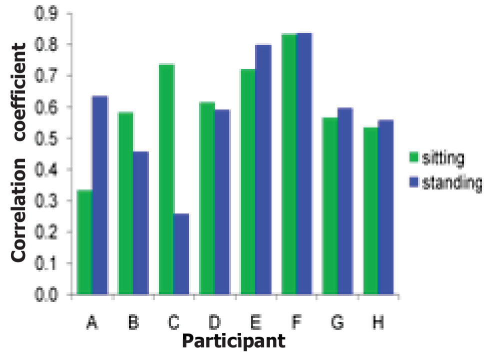
METHODS

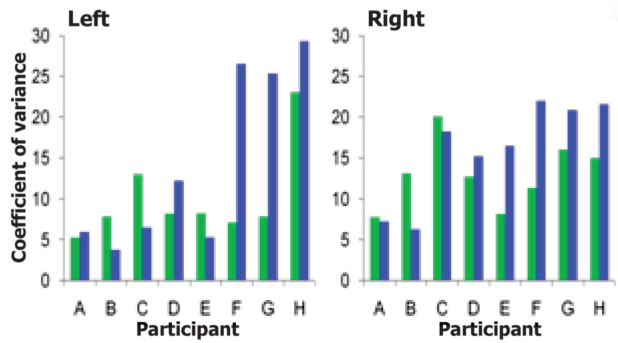
PARTICIPANTS: Eight able-bodied volunteers (2 females, 6 males), 20-40 years old

POSITIONS: *Sitting* and *Passive standing* in "EasyStand 5000" (Altimate Medical, Inc., USA)

H-REFLEX: 60 soleus H-reflexes were evoked in right and left legs simultaneously with interstimuli interval of 5 s.

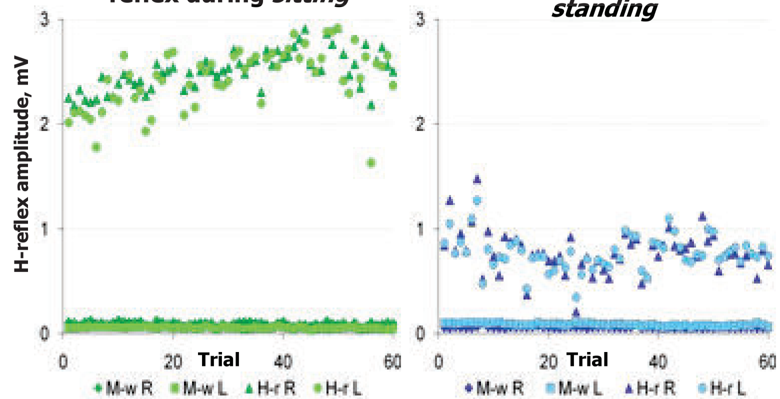






Participant F: Soleus H-reflex during *sitting*

Participant F: Soleus H-reflex during *passive standing*



RESULTS

1. The magnitude of the H-reflex was smaller during passive standing in comparison with sitting: 19.0 ± 9.8 % vs. 36.3 ± 4.6 % of the soleus maximal M-wave (Mmax), respectively ($p = 0.004$).
2. There was no significant difference in the coefficient of variation during sitting and passive standing: 11.7 ± 5.3 vs. 15.4 ± 9.2 %, respectively.
3. There was no significant difference in the correlation coefficient during sitting and passive standing: 0.62 ± 0.16 vs. 0.59 ± 0.20, respectively.

CONCLUSION

1. **These results concur with prior reports that significant inhibition of the soleus H-reflex occurs when changing posture from sitting to standing.**
2. **Our results support the findings that postsynaptic but not presynaptic mechanisms are predominant in the variability of the monosynaptic reflex.**
3. **The level of correlation of bilateral H-reflex did not depend on the posture suggesting that presynaptic mechanisms are not the leading factor in this phenomenon.**

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EFFECTS OF VOLUNTARY EXERCISE, FUNCTIONAL ELECTRICAL STIMULATION AND VIBRATION ON CORTICOMUSCULAR COHERENCE

74.19

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Toronto Rehab — Everything Humanly Possible
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INTRODUCTION

New findings in the field of adaptive neuroplasticity have demonstrated that goal-oriented exercises coupled with neuromuscular electrical stimulation or vibration may promote improvement of voluntary muscle control and functional abilities [1, 2, 3]. However, the physiological mechanisms underlying the observed benefits of the supplementary sensorimotor stimulation have not been investigated in detail.

In the current study, we aimed to investigate the *central* effects of the sensorimotor stimulation on the interaction between corticospinal drive and spinal motoneurones. It is believed that increased corticomuscular coherence can reflect a tighter cortical control of the muscle activity during the acquisition of the motor task [4].

PURPOSE

To investigate whether the somatosensory stimulation caused by neuromuscular electrical stimulation or vibration can potentiate corticomuscular coupling during voluntary movement.

METHODS

<u>Participants</u>: Five able-bodied volunteers (2 females, 3 males).

<u>Conditions</u>: intermediate 1 s dorsiflexion and rest periods during 10 min:
- voluntary movements at the level of 15% of MVC (VOL15);
- voluntary movements at the level of 7% of MVC (VOL7);
- FES driven movements at the level of 7% of MVC (FES);
- voluntary movements at the level of 7% of MVC supplemented with FES at the level of 7% of MVC (VOL+FES);
- voluntary movements at the level of 15% of MVC supplemented with constant vibration (VOL+Vibro).

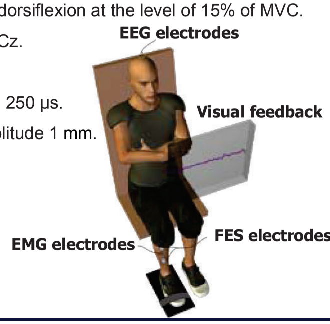
<u>Tests</u>: were performed before and after the experimental conditions, during 1 min isometric dorsiflexion at the level of 15% of MVC.

<u>EEG</u>: Cz and 5 cm lateral to Cz.

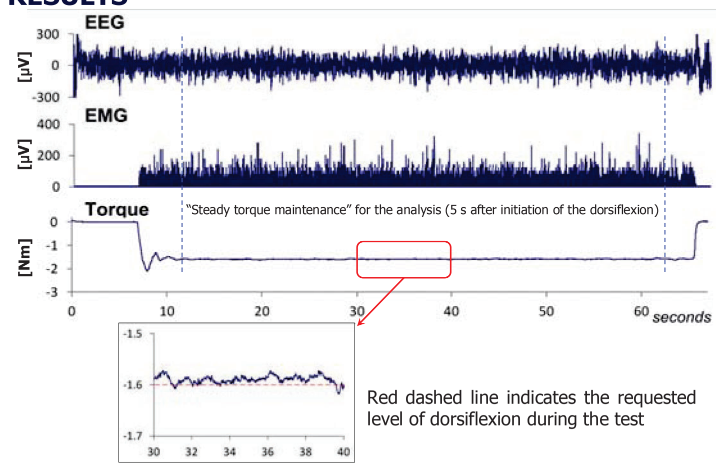
<u>EMG</u>: m. tibialis anterior.

<u>FES</u>: frequency 40 Hz, pulse 250 μs.

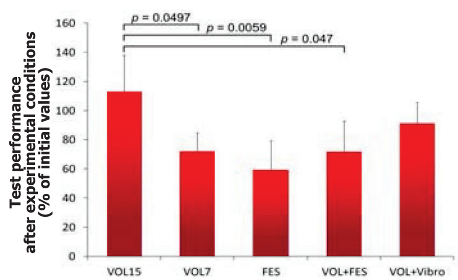
<u>Vibration</u>: frequency 60 Hz, amplitude 1 mm.



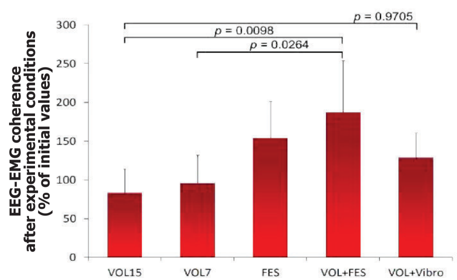
RESULTS



Red dashed line indicates the requested level of dorsiflexion during the test



1. Motor task performance increased after VOL7, FES, VOL+FES. The values are presented as percentages of the standard deviation values obtained prior to each experimental condition.

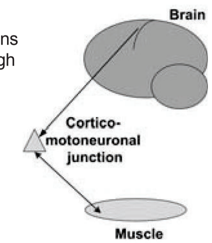


2. EEG-EMG coherence significantly increased after FES and VOL+FES. The values are presented as percentages of the coherence area obtained prior to each experimental condition.

DISCUSSION

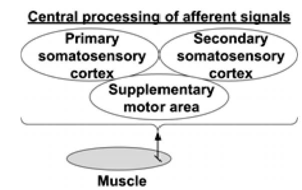
POSSIBLE MECHANISM I

Repetitive antidromic activation of the motorneurons enhanced recruitment of the motorneurons (through Hebb synpses) [2].



POSSIBLE MECHANISM II

Repetitive peripheral nerve stimulation enhanced the excitability of the cortical projections of corresponding muscles and improve motor performance [3].



CONCLUSION

1. Supplementary sensory stimulation during goal-oriented exercise can result in cortical reorganization and maximize transmission effectiveness of descending motor commands.
2. Contribution of the mechanisms underlining the observed effects will be determined in future research.

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Response of Individual Muscles to Neuromuscular Electrical Stimulation Trainig of Knee Extensors

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PURPOSE

1) To identify the unique force vector produced by individual knee extensors
2) To investigate the contribution of individual knee extensors to the force vectors produced by the global stimulation in neuromuscular stimulation training.

ABSTRACT

Neuromuscular electrical stimulation (NMS) is one of the most popular advanced methodologies for athletes to gain muscle strength. Usually, NMS uses a pair of large electrodes to stimulate one muscle group (Group Stimulation, GS) instead of stimulating individual muscles. To design effective training protocols, the contribution of individual muscle to GS must be examined. Purpose: 1) To identify the unique force vector produced by individual knee extensors, and 2) to investigate the contribution of individual knee extensors to the force vectors produced by GS in NMS training. Methods: Six subjects (30.0±8.9 yrs) sat on a chair and three knee extensors (vastus medialis, vastus lateralis, and rectus femoris) were individually stimulated. Additionally, knee extensors were stimulated simultaneously as a group using conventional NMS electrodes at three different locations (GS1-3). The unit force vectors in the isometric condition were calculated for individual muscles and for GS1-3. The relative contribution of individual muscles to GS1-3 was estimated based on an assumption that the force vector measured during each GS was a linear vector summation of force vectors of the three individual muscles. Results: The unit force vector was statistically different among the three knee extensors (MANOVA, $p < 0.001$). The contribution of the three knee extensors to the GS was statistically different among the three knee extensors (MANOVA, $p = 0.04$). Conclusion: Each knee extensor has a unique force vector. Distinct contributions from individual muscles were found, suggesting that NMS training may unevenly affect knee extensors.

Methods

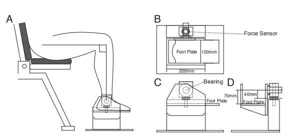
Subjects:
- Young healthy subjects: n = 6 (all males), 30,0±8.9 yrs

Stimulation Locations:
- Vastus medialis (VM), Vastus lateralis (VL), Rectus femoris (RF)
- Three global locations on knee extensors; GS1, GS2, and GS3

Measurements:
- 3D force exertion at the ankle joint using a force sensor (Kistler 9067) mounted on a custom made device (Nozaki et al. 2005)

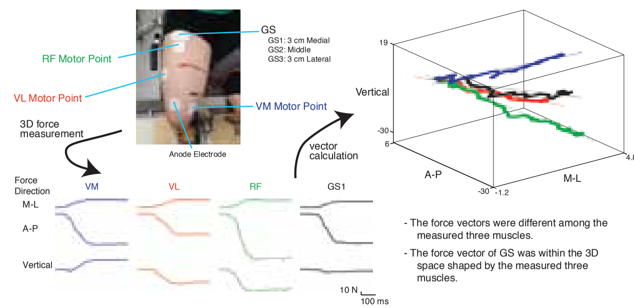
Protocol:
- The subject sat on the chair with knee and hip flexed at 90 degree.
- A motor point for each muscle was individually stimulated with the maximal, comfortably tolerable itensity, in the isometric condition.
- GS1-3 were also stimulated.

Analysis:
- The unit force vectors were calculated for individual muscles and for GS1-3, using principle component analysis.
- The relative contribution of individual muscles to GS1-3 was estimated based on an assumption that the force vector measured during each GS was a linear vector summation of force vectors of the three individual muscles.



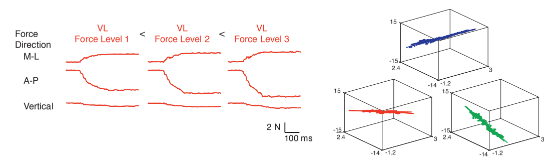
Force measurement system. A: subject seated on a chair with the right foot placed on a force measurement device. The lower leg was set to be vertical. The right foot was firmly fixed to the device by a strap. Top (B), side (C), and front (D) views of the force measurement device.The triaxial force sensor mounted just below the rotation axis can measure the force that the subject exerts at the ankle joint position.

1 Examples of recordings



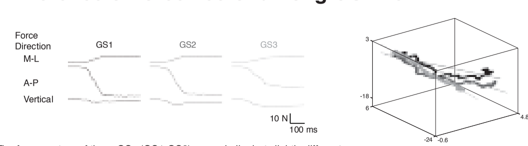
- The force vectors were different among the measured three muscles.
- The force vector of GS was within the 3D space shaped by the measured three muscles.

2 Independency of force vector to force level



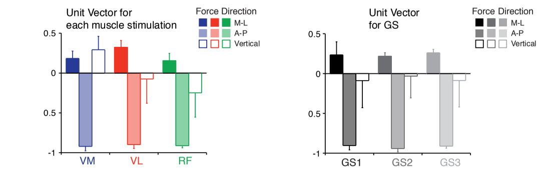
- Examples of three force levels for VL stimulation (left), and force directions for each measured muscle including three force levels respectively.
- The force vector for each muscle appears similar among different force levels.

3 Difference of force vector among GS1 - 3



- The force vectors of three GSs (GS1-GS3) were similar but slightly different.

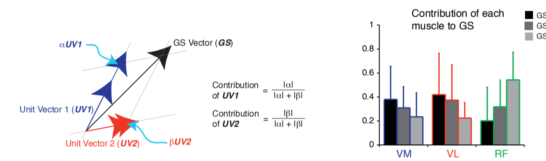
4 Unit vector for each muscle stimulation and GS



- The unit force vector was statistically different among the three knee extensors (MANOVA, $p < 0.001$).
- The measured three muscles exerted force in different directions.

- The unit force vector was not statistically different among the three GSs (MANOVA, $p = 0.840$).
- Considerably different patterns among the subjects might cause this insignificant result.

5 Contribution of each muscle on GS



- The contribution of the three knee extensors was statistically different among the three knee extensors (MANOVA, $p = 0.04$).
- When the global stimulation was applied, each muscle was stimulated with a different level.

CONCLUSION

- Each knee extensor has a unique force vector.
- Distinct contributions from individual muscles were found, suggesting that NMS training may unevenly affect knee extensors.

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Novel Methodology to Reduce Muscle Fatigue during Functional Electrical Stimulation using Spatially Distributed Sequential Stimulation

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PURPOSE

To investigate the feasibility of reducing muscle fatigue by distributing the electric field over a wider area using multiple surface electrodes placed over the muscle

BACKGROUND

Patients with neurological disorders such as spinal cord injury (SCI), stroke, and traumatic brain injury, frequently suffer from motor disability. Motor disability detracts from patients' independence and therefore quality of life. Additionally, motor disability causes disuse of muscles, and induces muscle atrophy. Many secondary impairments in those patients, such as pressure sores, low impact fractures, and deep vein thrombosis are thought to be at least partially related to muscle atrophy. Functional electrical stimulation (FES) has been used for neurorehabilitation of individuals with such neurological disorders. FES has been used as an assistive technology to help patients carry out tasks in daily life, as a therapeutic tool helping motor relearning known as FES therapy, and as a tool for muscle training known also as electrical muscular stimulation. However, one common aspect that limits its usage is the rapid onset of muscle fatigue, which severely restricts the duration FES can be used. It is believed that the reverse recruitment of motor units during electrical stimulation is the main cause of rapid fatigue. Another reason is that a localized electric field repeatedly excites only some nerve bundles that are innervating muscles of interest (i.e., muscles to be stimulated).

It is hypothesized that the second reason, localized electric field, can be addressed by stimulating a wider area of the nerve bundle, which activates more motor units. This can be realized by using multiple stimulation sites to deliver electrical stimulation. Indeed, studies have shown the effectiveness of sequential activation of multiple portions of nerves in reducing fatigue using multiple electrodes [1, 2, 3, 4]. However, all of these studies investigated stimulation that required invasive implanting of multiple electrodes, which is not suitable for clinical use. The purpose of this study is to investigate the feasibility of reducing muscle fatigue by distributing the electric field over a wider area using multiple surface electrodes placed over the muscle. If successful, such a technique would allow longer FES sessions due to reduced muscle fatigue, leading to better rehabilitation outcomes.

Experiment 1

Subjects:
- An individual with complete SCI, T3/T4, ASIA A, 4 years post injury

Task:
- Isometric, continuous plantar flexion in a sitting posture for 2 min
- Comparison between SES and SDSS (see the right column)
- 12 trials in separated days (6 SES and 6 SDSS)
- FES: Pulse frequency, 40 Hz; Pulse width, 250 us; Pulse intensity, 64±5 mA (SES) and 51±4 mA (SDSS)

Measurements:
- Planter flexion torque

Analysis:
- Fatigue Index (FI): Torque at the end of the 2 min stimulation normalized to the maximum torque
- Fatigue Time (FT): Difference in time from when the maximum torque was achieved to when the torque dropped by 3 dB
- Torque-Time Integral (TTI): Integral of torque over the entire 2 min fatiguing stimulation

Experiment 2

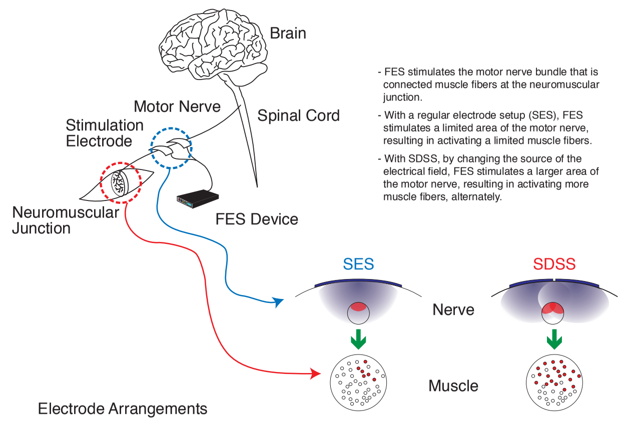
Subjects:
- Young healthy subjects: n = 13 (female 5), Age 25.7±7.0 yrs, Weight 64.8±6.9 kg, Height 169±7 cm

Task:
- Isometric, repetetive plantar flexion in a sitting posture for 2 min, 300 ms-on and 700 ms-off
- Comparison between SES and SDSS (see right column)
- 1 trial for each condition per subject, 2 trials in separated days
- FES: Pulse frequency, 40 Hz; Pulse width, 250 us; Pulse intensity, 42±6 mA (SES) and 39±6 mA (SDSS)

Measurements:
- Planter flexion torque

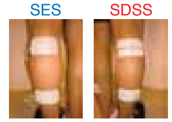
Analysis:
- FI and TTI

1 Spatially Distributed Sequential Stimulation (SDSS) & Single Electrode Stimulation (SES)



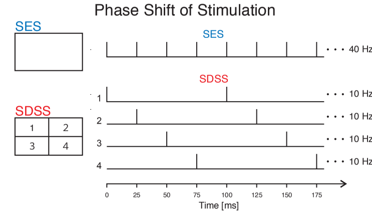
- FES stimulates the motor nerve bundle that is connected muscle fibers at the neuromuscular junction.
- With a regular electrode setup (SES), FES stimulates a limited area of the motor nerve, resulting in activating a limited muscle fibers.
- With SDSS, by changing the source of the electrical field, FES stimulates a larger area of the motor nerve, resulting in activating more muscle fibers, alternately.

Electrode Arrangements



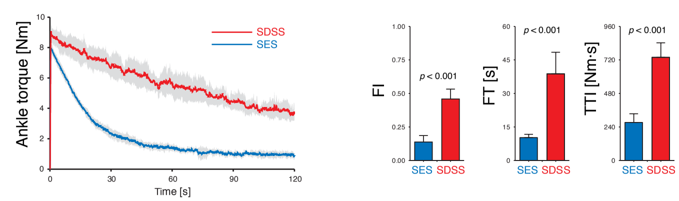
- In SES, a set of 9 cm - 5 cm electrodes was used for the active electrode and the reference electrode.
- In SDSS, an array electrode of four 4.5 cm - 2.5 cm electrodes was used for the active electrode, and a 9 cm - 5 cm electrodewas for the reference electrode.
- Each electrode in the array electrode was stimulated with 10 Hz, with 90 deg phase shift among the array electrodes.

Phase Shift of Stimulation



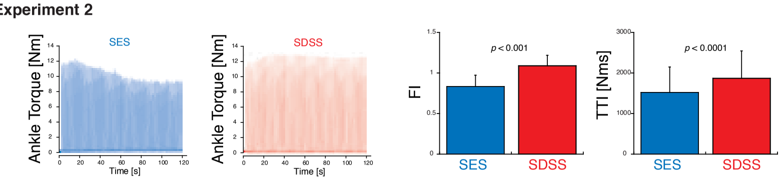
2 Comparison between SDSS and SES

Experiment 1



- For both modes, the maximum torques quickly developed after onset of stimulation and fatigue ensured immediately after that. However, SDSS resulted in clearly less fatigue than SES as evidenced by a lower rate of torque decline during the first quater of stimulation and a higher torque produced at the end of stimulation.
- All measures of FI, FT, and TTI quantitatively and statistically supports the abovementioned observation, i.e., SDSS less fatigue.

Experiment 2



- With the able-bodied individuals and repetitive contractions, the similar observation was obtained.
- For both measures of FI and TTI, SDSS showed less fatigue compared to SES.

CONCLUSION

- The effectiveness of using SDSS in planter flexors to reduce muscle fatigue was shown in one individual with SCI and able-bodied individuals.
- The results warrant the potential of SDSS to prolong training sessions of FES therapy.

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A MULTI-CHANNEL CURRENT-REGULATED OUTPUT STAGE FOR AN ELECTRICAL STIMULATOR

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Background and Motivation

Electrical stimulation (ES) therapy delivers charge pulses to excitable tissue to improve, modulate or restore lost or damaged neuronal and/or muscular function. Specific ES systems that have been developed for different applications are limited for two main reasons:

1. They often do not provide consistent charge delivery over the full range of tissue impedances.
2. They cannot be easily applied to different areas of the body without major adjustment or redesign.

This paper elucidates a novel electrical stimulator platform that solves the problems above by:

1. Achieving the same muscle contraction as state of the art solutions with significantly lower charge using sharper, better controlled pulses.
2. Employing a flexible digital platform to accommodate virtually all FES applications.

Stimulation Pulses and Specifications

In developing the new flexible output stage, several key stimulation characteristics had to be both tightly controlled and readily adjustable:

Pulse shape: Fig1 shows biphasic asymmetric (left) and biphasic symmetric pulses (right). For balanced pulses, the charge delivered and removed must be maintained over time, to avoid damaging charge retention in tissue.

Pulse duration (*PD*): should be long enough so the latent period, action potential and recovery take place. The larger the muscle groups, the greater PD needed(up to several ms).

Pulse frequency (f_{pulse}): should generate continuous muscle (tetanic) contractions (>40Hz) and discontinuous muscle (non-tetanic) contraction (between 16 Hz and 40Hz).

Pulse intensity (*PI*): should be high enough to produce an action potential. The deeper and/or larger muscles, the greater the PI needed (up to over 100mA).

Pulse rise time (t_{rise}): is the time it takes for a pulse to reach the desired intensity. Minimizing t_{rise} allows generation of action potentials with less electric charge (reduced PI).

Current vs. voltage control: current regulated pulses are preferentially used because they provide better control of the charge being applied when tissue impedance changes (during therapy or between sessions), providing more consistent stimulation.

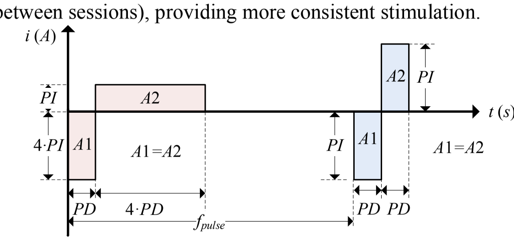

Fig1 – Key pulse characteristics.

Based on the abovementioned pulse requirements, past FES solutions, and the needs of FES practitioners and researchers, the specifications of the new system were established in Table 1.

Table 1 – New stimulator specifications

Feature	*Specification*
Variable pulse intensity	0.5mA – 125mA (resolution 0.5mA)
Variable pulse frequency	1Hz – 15kHz, (resolution of 1Hz)
Variable pulse duration	10µs – 10ms (resolution of 1µs)
Ultrafast pulse rise time	10ns
Variable pulse shape	Biphasic (Symmetric/Asymmetric) Monophasic (Positive/Negative)
Balanced pulses	No residual charge left in tissue
Voltage and current protection	Immediate shutdown if voltage or current exceeds allowable limit
Multiple stimulation channels	4 independently regulated stimulation channels

Methods

The four-channel stimulator consists of two stages of switched mode power supplies arranged in series. These supplies serve two important functions:

1. *First stage:* the battery (input) voltage is stepped up to several hundred volts with built in isolation from the patient.
2. *Second pulse shaping circuitry stage:* creates the desired stimulation pulses with tight and fast regulation of current amplitude level.

This novel setup has many advantages:

1. Built in charge balancing ensures no excess charge is present in the tissue of the patient following a single pulse delivery.
2. A digital controller for dynamic current regulation, providing exquisite pulse shapes and inherent current overload protection.
3. A programmable digital platform that generates many different types of pulses, varying in duration, amplitude and frequency.
4. The ability for closed loop functionality using inputs from various sensory modalities.

Results and Conclusion

Figs2 a and b show examples of pulses generated with the new output stage of the stimulator. Fig3 highlights the ultrafast rise time.

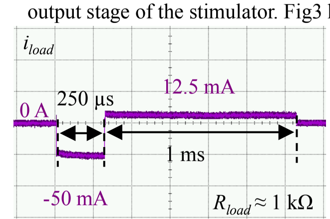

Fig2a – Assymetric Biphasic Pulse

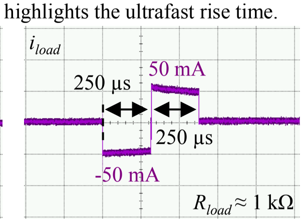

Fig2b – Symetric Biphasic Pulse

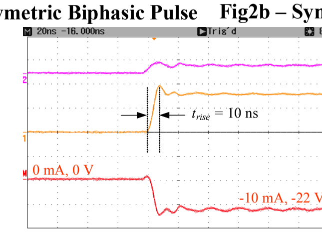

Fig3 – Ultrafast rise time

The new stimulator has several improvements over current system:

- Two orders of magnitude faster rise time, reducing amplitude
- Tighter and safer regulation of current delivered to the patient
- Broad performance envelope encompassing all ES applications.

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INCLUDING NON-IDEAL BEHAVIOUR IN FES SIMULATIONS

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Introduction

• Functional electrical stimulation (FES) can be used to restore or replace motor function in individuals with spinal cord injuries (SCI).
• New SCI applications must be thoroughly tested with SCI subjects – time-consuming and expensive, so simulations are used to refine design of FES applications prior to testing with subjects.
• FES simulations are often based on ideal stimulated muscle response, which may result in an overly optimistic assessment of FES system's likely real-world performance.

• We created a "non-idealities" block in Simulink with data from complete SCI subjects.
  • Modifies nominal stimulated muscle response to reflect undesirable behaviour seen in real world.
  • Represents the range of spasm, tremor, and fatigue behaviour exhibited by stimulated muscles.
  • Can be incorporated into existing simulations to analyze potential real-world performance of FES systems.

Methods

• Spasm, tremor, and fatigue waveforms were extracted from stimulated knee movements of complete SCI subjects.
  • Fatigue waveforms were scaled between 0 and 1 [1].
  • Each non-ideality waveform was classified as mild, moderate, or severe.

• Non-idealities block is implemented in Simulink, and allows user to choose severity of non-idealities included in particular instance of block.
  • Actual waveform that is used for each non-ideality is chosen randomly from group of waveforms having desired classification.
• Eqn 1 gives output of block, where \tau(t) is nominal stimulated muscle torque, \nu(t) is modified torque, and s(t), m(t), and fat(t) are instances of spasm, tremor, and fatigue waveforms.

• Fig 1 shows example of implementation of non-idealities block in FES simulation.
  • PID controller is used to track desired knee angle.
  • Based on Ferrarin & Pedotti's model of seated, stimulated knee extension against gravity [2].

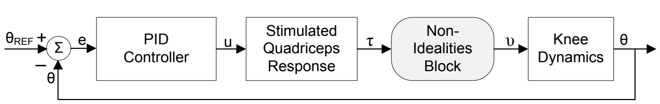

Fig 1 – Diagram of knee control simulation (θ is knee angle)

Results

• Fig 2 shows effect of non-idealities block. Dotted line is nominal knee torque at maximum stimulation. Solid line is modified torque for mild spasms, fatigue, and tremor.
• Fig 3 shows unit step tracking performance for PID controller. Dotted line is without non-idealities block, and solid line is with block (mild spasms, fatigue, and tremor).

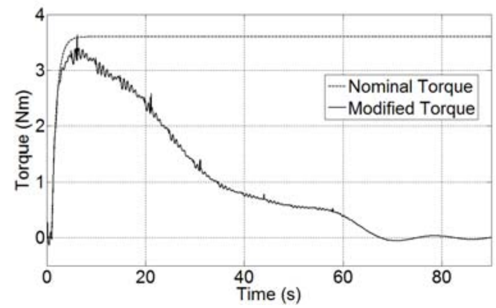

Fig 2 – Nominal torque at maximum stimulation, and modified torque from non-idealities block (mild spasms, fatigue, and tremor)

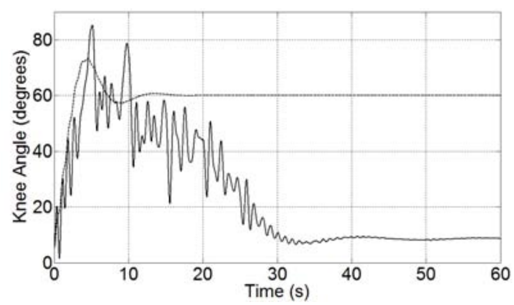

Fig 3 – Sample implementation – unit step tracking with PID control (dotted line is nominal case, solid line is with non-idealities block).

$$\nu(t) = (\tau(t) + s(t) + m(t)) \cdot fat(t)$$

Equation 1 – Output of non-idealities block

Discussion

• Modified torque produced by non-idealities block differs from nominal torque.
• Fatigue, tremors had a large effect on control performance.
• Real-world control performance may be modestly better than simulated results due to unmodeled muscle recovery.
• Isotonic contractions in trained muscles may have different fatigue profiles than those used in non-idealities block.
• Non-idealities block does not reflect all possible undesirable behaviour that can occur with real-world FES use.

Conclusions

• Non-idealities block allows researchers to assess likely real-world performance of FES systems prior to subject testing.
• MatLab code for block will be freely available on our website (targeted release date – early 2011).

References

[1] Graham GM, Thrasher TA, Popovic MR. *IEEE Trans Neural Syst Rehabil Eng*, 14:38-45, 2006.
[2] Ferrarin M, Pedotti A. *IEEE Trans Rehabil Eng*, 8:342-352, 2000.

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INCLUDING NON-IDEAL BEHAVIOUR IN SIMULATIONS OF FES

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Introduction

• Functional electrical stimulation (FES) can be used to restore or replace motor function in individuals with spinal cord injuries (SCI).
• New SCI applications must be thoroughly tested with SCI subjects – time-consuming and expensive, so simulations are used to refine design of FES applications prior to testing.
• FES simulations are often based on ideal stimulated muscle response, which may result in an overly optimistic assessment of FES system's likely real-world performance.

• We created a "non-idealities" block in Simulink with data from complete SCI subjects (injury levels C6 through T10).
  • Block modifies nominal stimulated muscle response to reflect undesirable behaviour seen in real world.
  • Represents the range of spasm, tremor, and fatigue behaviour exhibited by stimulated muscles.
  • Can be incorporated into existing simulations to analyze potential real-world performance of FES systems.

Methods

• Spasm, tremor, and fatigue waveforms were extracted from stimulated knee movements of complete SCI subjects.
  • Fatigue waveforms were scaled between 0 and 1 [1].
  • Each non-ideality waveform was classified as mild, moderate, or severe.

• Non-idealities block was implemented in Simulink, and allows user to choose severity of non-idealities included in particular instance of block.
  • Actual waveform that is used for each non-ideality is chosen randomly from group of waveforms having desired class.
• Eqn 1 gives output of block, where $\tau_{quad}(t)$ is nominal stimulated muscle torque, $\nu(t)$ is modified torque, and *s(t)*, *m(t)*, and *fat(t)* are instances of spasm, tremor, and fatigue waveforms

• Fig 1 shows implementation of non-idealities block in simulation.
  • Boundary-layer sliding mode controller (SMC) is used to track desired knee angle.
  • Based on Ferrarin & Pedotti's model of seated, stimulated knee extension against gravity [2].

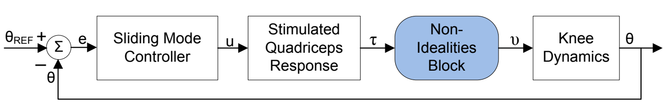

Fig 1 – Diagram of knee control simulation (θ is knee angle)

$$\nu(t) = \left(\tau_{quad}(t) + s(t) + m(t)\right) \cdot fat(t)$$

Equation 1 – Output of non-idealities block

Results

• Fig 2 shows the step response for the nominal case and the case with mild fatigue, mild spasms, and mild tremors.
• Fig 3 shows the response for a walking-like trajectory for the same cases shown in Fig 2.

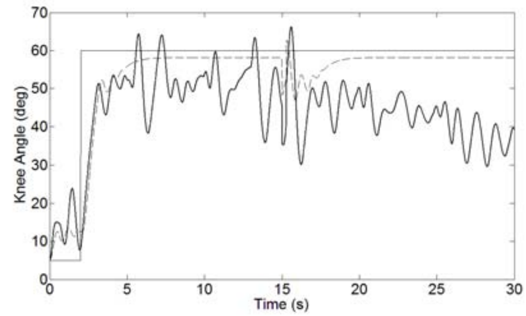

Fig 2 – Step response. Thin line is reference trajectory, dashed line is response for nominal knee model, and thick solid line is response for model with mild non-idealities included.

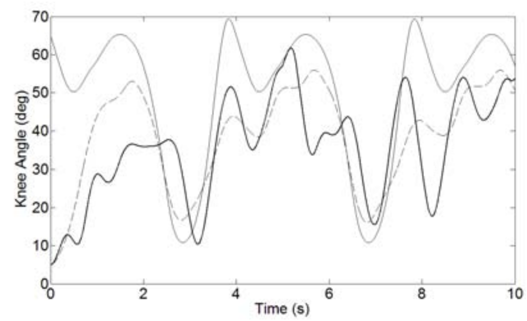

Fig 3 – Response for walking-like trajectory. Thin line is reference trajectory, dashed line is response for nominal knee model, and thick solid line is response for model with mild non-idealities.

Discussion

• Performance of SMC is good for nominal case and poor for cases that include non-idealities, as shown in Figs 2 and 3.
• Inclusion of non-idealities block shows that controller is not ready for testing with human subjects.
• Spasticity improved controller performance, possibly due to increased damping of the knee joint.
• Real-world control performance may be modestly better than simulated results due to unmodeled muscle recovery.
• Isotonic contractions in trained muscles may have different fatigue profiles than those used in non-idealities block.

Conclusions

• Non-idealities block illustrates that real-world performance of controller may be vastly different from nominal performance.
• Non-idealities block provides a convenient method of comprehensively testing FES systems in simulation.
• MatLab code for block will be freely available in early 2011 at www.toronto-fes.ca under Products.

References

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Improved Fatigue Resistance in Leg Muscles during Spatially Distributed Sequential Stimulation

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PURPOSE

To explore the fatigue-reducing ability of spatially distributed sequential stimulation (SDSS) for major lower limb muscle groups

BACKGROUND

Neuromuscular electrical stimulation (NMES) is used to promote physiological and functional improvement in paralysed limbs [1, 2] and counteract musculoskeletal atrophy [3, 4]. However, while it has succeeded in assisting individuals with neuromuscular disorders, a critical limitation with this rehabilitative approach is the rapid onset of muscle fatigue during repeated contractions [5], which results in the muscle force decay and slowing of muscle contractile properties [6].

The increased fatigability with NMES is thought by some researchers to reflect a reversal of the size principle of recruitment [7], when larger axons that innervate the more easily fatigable fibers are recruited at low stimulus magnitudes, and the smaller axons follow with increased stimulation levels [1, 5]. Another plausible explanation is that, while during voluntary contraction, the work is being shared between different motor units of the same muscle [5], conventional NMES does not permit alterations in recruitment of motor units because all parameters remain fixed during the bout [5].

One of the means to counter force loss during electrical stimulation is aimed at achieving an asynchronous behavior by delivering electrical stimulation through multiple electrode locations on a single site, producing a fused contraction with relatively low stimulation rates, and delaying the onset of fatigue. This type of spatially distributed and sequentially applied stimulation is referred to as 'sequential stimulation'. Fatigue was reduced with such stimulation, widely shown in animal experimental models [8-11]. Until now, only a few studies [12-14] investigated this stimulation method in human experimental setups, however, this method has not been successfully incorporated into clinical applications.

Experiment 1 [13]

Subjects:
- An individual with complete SCI, T3/T4, ASIA A, 4 years post injury

Task:
- Isometric, continuous plantar flexion in a sitting posture for 2 min
- Comparison between SES and SDSS (see the right column)
- 12 trials in separated days (6 SES and 6 SDSS)
- FES: Pulse frequency, 40 Hz; Pulse width, 250 us; Pulse intensity, 64±5 mA (SES) and 51±4 mA (SDSS)

Measurements:
- Planter flexion torque

Analysis:
- Fatigue Index (FI): Torque at the end of the 2 min stimulation normalized to the maximum torque
- Fatigue Time (FT): Difference in time from when the maximum torque was achieved to when the torque dropped by 3 dB
- Torque-Time Integral (TTI): Integral of torque over the entire 2 min fatiguing stimulation

Experiment 2

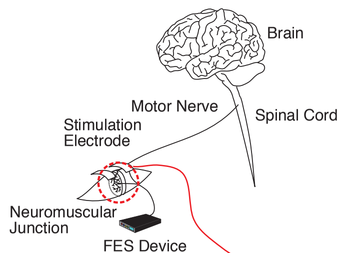
Subjects:
- Young healthy male subjects: n = 15

Task:
- Knee extension, knee flexion, plantar flexion, and dorsi flexion
- Isometric, repetetive contraction in a sitting posture for 2 min, 300 ms-on and 700 ms-off
- Comparison between SES and SDSS (see right column)
- 4 trials for each joint with each condition per subject in a day, 2 separated days for 2 conditions
- FES: Pulse frequency, 40 Hz; Pulse width, 250 us

Measurements:
- Joint torque

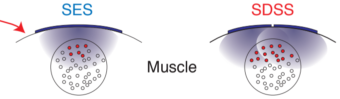
Analysis:
- FI

1 Spatially Distributed Sequential Stimulation (SDSS) & Single Electrode Stimulation (SES)



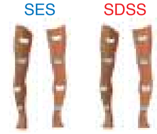
- FES stimulates the motor nerve bundle that is connected muscle fibers at the neuromuscular junction.
- With a regular electrode setup (SES), FES stimulates a limited area of the motor nerve, resulting in activating a limited muscle fibers.
- With SDSS, by changing the source of the electrical field, FES stimulates a larger area of the motor nerve, resulting in activating more muscle fibers, alternately.

Hypothesized Mechanism



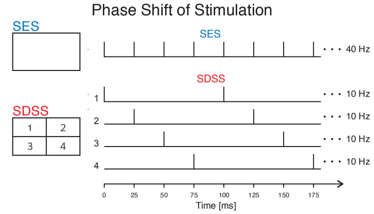
- In SES, a smaller number of muscle fibres are activated continuously with a high frequency.
- In SDSS, a larger number of muscle fibres are activated alternatively with a low frequency.

Electrode Arrangements



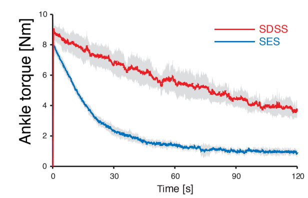
- In SES, a set of 9 cm - 5 cm electrodes was used for the active electrode and the reference electrode.
- In SDSS, an array electrode of four 4.5 cm - 2.5 cm electrodes was used for the active electrode, and a 9 cm - 5 cm electrodewas for the reference electrode.
- Each electrode in the array electrode was stimulated with 10 Hz, with 90 deg phase shift among the array electrodes.

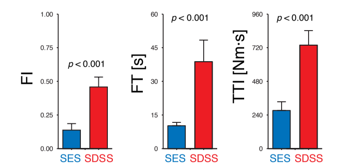
Phase Shift of Stimulation



2 Comparison between SDSS and SES

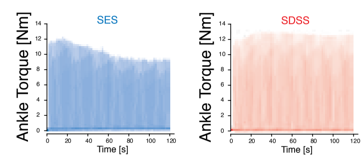
Experiment 1

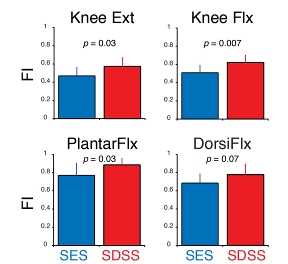




- For both modes, the maximum torques quickly developed after onset of stimulation and fatigue ensured immediately after that. However, SDSS resulted in clearly less fatigue than SES as evidenced by a lower rate of torque decline during the first quater of stimulation and a higher torque produced at the end of stimulation.
- All measures of FI, FT, and TTI quantitatevely and statistically supports the abovementioned observation, i.e., SDSS less fatigue.

Experiment 2





- With the able-bodied individuals and repetitive contractions, the similar observation was obtained.
- Except for dorsiflexors, SDSS showed significantly less fatigue compared to SES.

CONCLUSION

- The effectiveness of using SDSS to reduce muscle fatigue was shown in one individual with SCI for plantarflexors and able-bodied individuals for knee extensors/ flexors and plantarflexors .
- The results warrant the potential of SDSS to prolong training sessions of various FES therapy.

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