Rehabilitation Engineering Laboratory

Institute of Biomaterials & Biomedical Engineering
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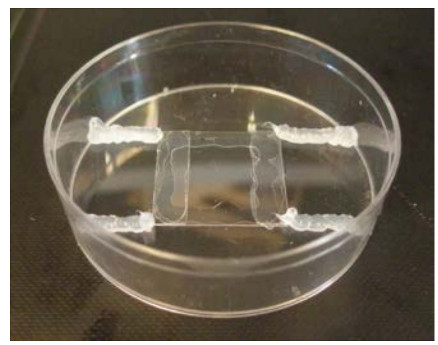

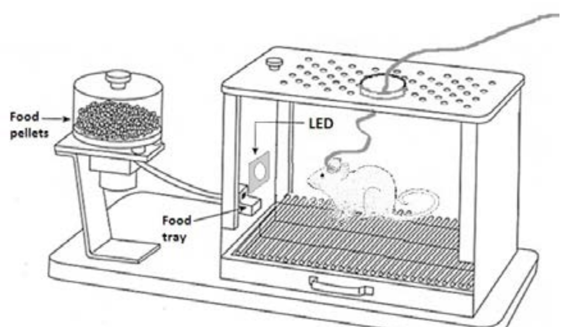

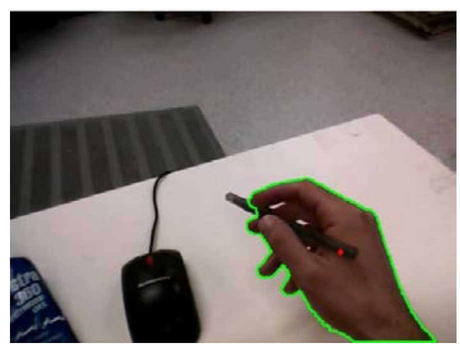

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About the Rehabilitation Engineering Lab

History

The Rehabilitation Engineering Laboratory was established in 2001 at the Lyndhurst Centre of Toronto Rehab, University Health Network. 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, reaching, and grasping ability. We also design neuroprosthesis 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 or to re-train weak muscles or the central nervous system.

Accomplishments –2012-2014

- MyndTec Inc. commercialized our FES Therapy for reaching and grasping. The product is called MyndMove™
- Developed EEG-based brain machine interface that can distinguish 5 different grasping postures
- Completed a randomized control trial with FES Therapy for walking in chronic incomplete SCI patients
- Developed electrical stimulation tool to direct and accelerate movement of the neural precursor cells
- Published more than 30 peer-reviewed journal papers and filed 6 patents
- Trained 7 postdoctoral fellows, 7 PhD students and 11 MSc students

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, or www.toronto-fes.ca

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 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.

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) 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 treadmill
- Force plates
- Polhemus motion capture system
- Optotrack dual camera motion capture systems
- ERIGO tilt table with motorized leg movement
- Electromagnetically shielded room for EMG and EEG measurements
- Vibration platforms
- REL-PAPPS perturbation system
- Biodex System 3
- ARMEO and ReJouce systems for upper limb rehabilitation
- Ultrasound system
- Transcranial Magnetic Stimulation System Medtronic Mag Pro R30
- 6-camera Raptor Motion Analysis system
- Various EMG and EEG measurement systems

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. Kei Masani (exercise physiology), Research Scientist
Dr. Jose Zariffa (biomedical engineering), Research Scientist
Dr. Cesar Marquez-Chin (biomedical engineering), Research Scientist

Postdoctoral Fellows

Dr. Hossein Rouhani (biomedical engineering), Postdoctoral Fellow
Dr. Aravind Kumar Namasivayam (speech/language), Postdoctoral Fellow
Dr. Masae Miyatani (exercise physiology), Postdoctoral Fellow
Dr. Robart Babona-Pilipos (biomedical engineering), Postdoctoral Fellow
Dr. Austin J. Bergquist (exercise physiology), Postdoctoral Fellow

Graduate Students

Rob Babona Pilipos, PhD student
Takashi Yoshida, PhD student
Andresa R. Marinho, PhD student
Steve McGie, PhD student
Matija Milosevic, PhD student
Kathryn Atwell, MASc student
Bahar Memarian, MASc student
Michael Same, MASc student
Martha Gabriela Garcia-Garcia, MASc student / PhD student
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Betty Chan, Grants & Accounts Coordinator
Naaz Desai, REL Manager, Research Coordinator and Physiotherapist
Abdulazim Rashidi, Research Engineer
Dr. Vera Zivanovic, Research Coordinator
Esther Oostdyk, Secretary

Awards and Distinctions

Kai-Lon Fok

2013	Summer Student Poster Presentation Award – 1st Place	Electrical, Bioelectrical and Computer Science Group, Engineering Science, University of Toronto

Chaim Katz

2014	Undergraduate Student Research Award	Natural Sciences and Engineering Research Council of Canada

Sara Mahallati

2014	Graduate Scholarship	Natural Sciences and Engineering Research Council of Canada, CREATE – CARE

Kramay Patel

2014	The Summer Student Oral Presentation Award – 1st Place	In Biomechanical Group, Division of Engineering Science, University of Toronto
2014	Undergraduate Student Research Award	Natural Sciences and Engineering Research Council of Canada
2014	The Kenneth Carless Smith Engineering Science Research Fellowship	Division of Engineering Science, University of Toronto
2013	Distinction - Rank 1st in the Division of Engineering Science	Division of Engineering Science, University of Toronto.
2013	Student Presentation Award - 1st Place	Institute of Biomaterials and Biomedical Engineering Undergraduate Summer Research Program, University of Toronto,

2013	Undergraduate Student Research Award	Natural Sciences and Engineering Research Council of Canada

Milos R. Popovic

2013	Business Award of Excellence – Technology and Innovation Award Winner	Mississauga Board of Trade
2013	University of Toronto Inventor of the Year Award	University of Toronto
2013	Morris (Mickey) Milner Award for Outstanding Contributions in the Area of Assistive Technologies	The Health Technology Exchange

Michael Same

2013	Student Travel Award	School of Graduate Studies, University of Toronto

Hossein Rouhani

2014	Postdoctoral Fellowship	Swiss National Science Foundation, Bern, Switzerland
2013	Postdoctoral Fellowship	Spinal Cord Injury Ontario Postdoctoral Fellowship

Takashi Yoshida

2013	Graduate Scholarship	Toronto Rehabilitation Institute
2012	Team Excellence Award	Neural Engineering and Therapeutics Team, Toronto Rehabilitation Institute

Recent Posters

Neuroprosthesis for Grasping

1. Functional electrical stimulation for upper extremity therapy in chronic stroke: A Case Study
2. Functional Electrical Stimulation for Severe Upper Extremity Hemiparesis: A Randomized Controlled Study
3. FES Therapy for Grasping in Chronic Incomplete SCI: Pilot Randomized Control Trial
4. Short-Term Neuroplastic Effects of Brain-Controlled Functional Electrical Stimulation

Neuroprosthesis for Sitting

5. Assessment of Sensitivity to Anatomical Landmark Misplacement Error in Human Trunk Motion Analysis
6. Functional Electrical Stimulation of Trunk Muscles Increases Trunk Stiffness
7. Effects of Foot Support on Sitting Postural Stability after Spinal Cord Injury
8. Multi-Segment Kinematic Assessment of Human Trunk: Sensitivity to Skin Artifacts

Neuroprosthesis for Standing

9. Comparison of single- and double-link inverted pendulum models for human standing posture
10. Spatially distributed sequential stimulation improves fatigue resistance in plantar flexors

Walking

11. Anticipatory postural adjustments and step execution during gait initiation in water: A pilot study
12. The effects of the aquatic environment on the control of upright stance: A pilot study comparing COP displacements in water and on dry land

Human-Machine Interfaces

13. Novel Paradigm to Elicit Volitional Modulation of Cortical Neuronal Activity in a Rat Model

Other Projects

14. Modulation of emotion using FES of the “Duchenne marker”
15. Cardiovascular Response of Individuals with SCI to Functional Electrical Stimulation and Passive Stepping

Functional Electrical Stimulation for Upper Extremity Therapy in Chronic Stroke: A Case Study

Noritaka Kawashima[1,2,3], Milos R. Popovic[1,2], and Vera Zivanovic[1]

1. Toronto Rehab, University Health Network; 2. Institute of Biomaterials and Biomedical Engineering, University of Toronto; and 3. National Rehabilitation Centre for Persons with Disability, Tokotozawa, Japan

Introduction

50,000 Canadians and 795,000 Americans will experience a new or recurrent stroke each year. Despite receiving weeks of rehabilitative therapy, the majority of stroke survivors are unable to incorporate the affected upper extremity into daily activities at 6 months post-stroke.

Functional electrical stimulation therapy integrates electrical stimulation to peripheral sensor and motor nerves with repetitive functional movement. Randomized controlled trials have shown that FES therapy can restore voluntary upper limb movement in sub-acute stroke and spinal cord injury patients [Ref 1,2]. The neural mechanisms underlying the improvements are not fully understood and it remains uncertain whether FES therapy is effective in chronic stroke with severe hemiparesis.

Objective

To investigate the impact of intensive Functional Electrical Stimulation (FES) therapy on neuromuscular changes in the upper limb (UL) of stroke patients with severe hemiparesis. [Ref 3]

Methods

Design: Open label case study. To eliminate contributions from spontaneous recovery, a chronic patient (>2 yrs post-stroke) was recruited.

Participant: 22-year-old female with severe upper limb paresis 2 years after a hemorrhagic stroke in the right frontal parietal area, secondary to an anteriovenous malformation bleed.

Intervention: FES Therapy administered for 1 hour, twice daily for 12 weeks for a total of 108 treatment sessions.

Clinical Assessments: Chedoke McMaster Stages of Motor Recovery (CMSMR), Motricity Index, Maximum Voluntary Contraction (MVC), and Modified Ashworth Scale (MAS)

Electrophysiological Assessment: H-reflex and Maximum motor response (M_{max})

Upper Arm Joint Kinematic Assessment: Dynamic Range of Motion (ROM) test and Drawing Test (at 6 and 12 weeks). (**NOTE:** These assessments were added to the study as a result of remarkable improvements observed after 6 weeks of therapy, no baseline captured)

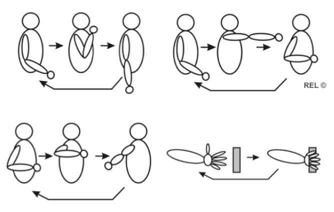
FES Therapy Program

FES Therapy consisted of two components:

1) Pre-programmed, coordinated surface electrical stimulation of multiple muscle groups to coincide with the phase and type of arm or hand motion a patient is striving to achieve.
2) Manual assisted (externally generated) passive motion in order to establish physiologically correct movement.

The FES system offers a full range of reaching and grasping movements to facilitate shoulder, elbow, wrist and hand function.

As the patient recovers voluntary function, neuroprosthesis assistance is reduced and eventually removed.



Results

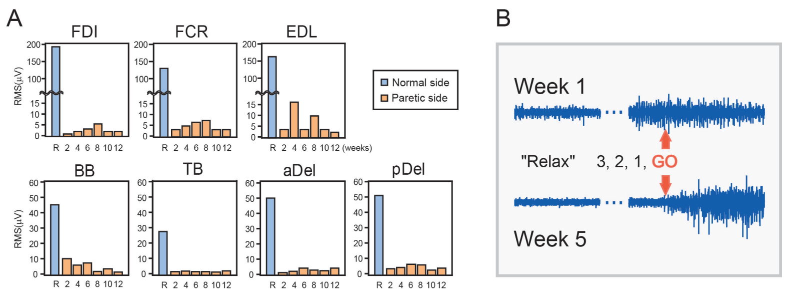
Clinical Assessments

The patient completed all training sessions and assessments. The CMSMR scores remained unchanged over the course of the study. Likewise, the Motricity Index did not change during the 12 week treatment period.

Modified Ashworth Scale scores decreased from 3 to 2 for the hand and from 4 to 3 for the arm during the 12 week study period.

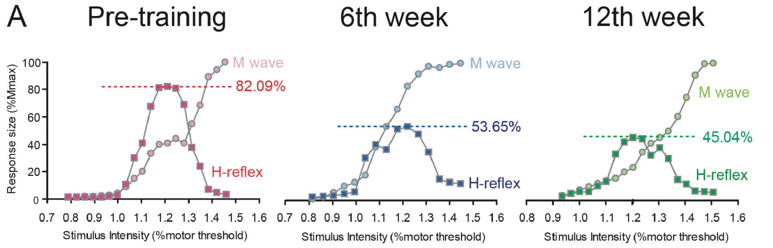
Maximal Voluntary Contraction (MVC)

MVC of the affected arm was remarkably smaller than that of the less affected arm and showed no significant changes over the course of therapy (Figure A below). Some muscle groups (i.e., TB and FDI) showed improved EMG activity as a result of the FES Therapy (Figure B)



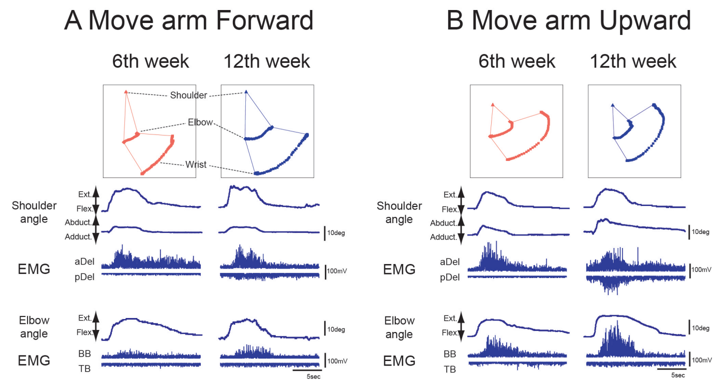
H-reflex and M_{max}

H-reflex, which reflects spinal motoneuron excitability, decreased considerably over the treatment period.



Range of Motion (ROM)

As a result of dramatic improvements observed in the first 6 weeks, voluntary arm ROM was captured using a three-dimensional tracking device (FASTRAK). For the shoulder and elbow joints, ROM tended to be larger at week 12 relative to week 6.



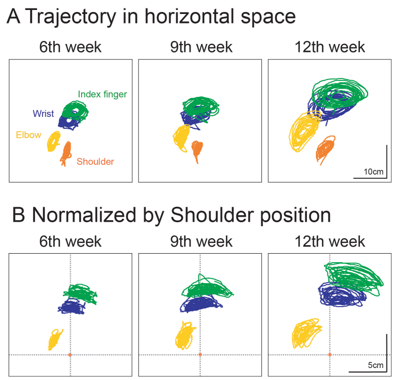
On enrollment the patient rarely used her paretic arm for functional activities.

On discharge from the study, the patient could relax her arm and hand voluntarily, allowing the arm to hang by her side when she is not using it.

Following 12 weeks of the FES Therapy, she was able to pick up a thin object and to touch her nose, movements she had been unable to perform before the therapy.

Drawing Test

At baseline, the patient was unable to draw a circle. Over the course of FES Therapy, the patient was able to proficiently draw increasingly larger circles. The following figure shows the trajectory of the shoulder, elbow, wrist and index finger as the patient performed the circle-drawing test: (a) absolute positions of individual joints; and (b) positions normalized with respect to the shoulder joint.



Conclusion

While motor function assessments such as CMSMR and MVC did not show remarkable changes, the chronic stroke patient showed significant improvement in upper extremity functional motion following FES Therapy.

Improvements in upper-limb function observed following intensive FES Therapy can be attributed to a) regained ability to voluntarily contract muscles of the affected arm; b) reduced spasticity and improved muscle tone in the same muscles; and c) increased range of motion of all joints.

References and Declaration of Interest

[1] Thrasher et al. Neurorehabilitation and Neural Repair. 2008; 22(6): 706-714.
[2] Popovic et al. Neurorehabilitation and Neural Repair, 2011; 25(5): 433-442.
[3] Kawashima et al. Physiotherapy Canada 2013; 65(1): 20-28

*Declaration of Interest - Dr. Popovic is a founder, a shareholder and the Chief Technology Officer of MyndTec Inc, a healthcare company created to commercialize technologies described in this presentation.

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

Functional Electrical Stimulation for Severe Upper Extremity Hemiparesis: A Randomized Controlled Study

Milos R. Popovic[1,2], Vera Zivanovic[2], and Adam T. Thrasher[3]

1. Toronto Rehab, University Health Network; 2. Institute of Biomaterials and Biomedical Engineering, University of Toronto; and 3. Department of Health and Human Performance, University of Houston

Introduction

50,000 Canadians and 795,000 Americans will experience a new or recurrent stroke each year.

Stroke remains the leading cause of long-term disability in North America and long-term disability is often associated with the persistent impairment of the upper extremity.

Despite receiving weeks of rehabilitative therapy, the majority of stroke survivors are unable to incorporate the affected upper extremity into daily activities at 6 months post-stroke.

Effective new treatment options are required to enhance a patient's independence and quality of life and to relieve the financial pressures incurred by the individual, their family, and the healthcare system.

Objective

To investigate whether treatment with a novel, non-invasive, functional electrical stimulation (FES) therapy improves recovery of voluntary arm function in severely disabled subacute stroke patients.

Methods

Design: Randomized controlled, two-arm, parallel group, single blind (assessor), single centre study

Participants: Twenty-one (21) stroke patients with severe upper extremity paralysis, i.e., individuals with Chedoke McMaster Stages of Motor Recovery scores of 1 or 2, who were at least two weeks (less than 6 months) after onset of stroke, took part in the study.

Interventions: The patients were randomized to receive either 1 hr/day of FES Therapy (Treatment group) or an equivalent dose (length and intensity) of conventional upper extremity therapy (Control group).

Assessments: Upper Extremity Fugl-Meyer (UE-FMA), Chedoke McMaster Stages of Motor Recovery (CMSMR), Barthel Index (BI), Functional Independence Measure (FIM™), and Self-Care FIM™ subscore (SC-FIM™).

FES Therapy Program

FES Therapy provides pre-programmed, coordinated muscle stimulation that coincides with the phase and type of arm motion a patient is striving to achieve.

Figure 1. The FES system offers a full range of reaching and grasping movements to facilitate shoulder, elbow, wrist and hand function.

As the patient recovers voluntary function, neuroprosthesis assistance is reduced and eventually removed.

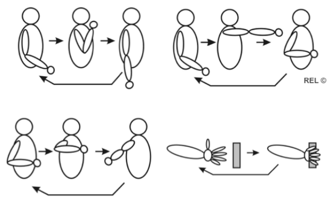
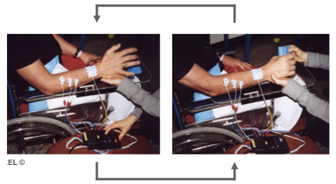


Figure 2. Shows a therapy session in which finger extension was preformed with neuroprosthetic assistance, and finger flexion was performed voluntarily. Hand function therapy sessions occur in the latter stages of the treatment program.



Results

Table 1: Summary of Baseline Patient Characteristics

Patient Characteristics	CONTROL (n= 11)	FES Therapy (n=10)
Age (years)		
mean (± SD)	64.8 (± 20.3)	51.0 (± 14.7)
range	(29 – 82)	(32 – 74)
Sex (number (%))		
male	6 (55%)	7 (70%)
female	5 (45%)	3 (30%)
Index Stroke Type (number (%))		
hemorrhagic	4 (36%)	3 (30%)
ischemic	7 (74%)	7 (70%)
Days from stroke to 1st treatment		
mean (± SD)	31.5 (± 11.6)	27.5 (± 12.0)
range	(19 – 47)	(16 - 57)

FES Therapy treatment group received an average of 40.4 (± 6.3) FES sessions. Control group received an average of 42.9 (± 8.4) sessions of conventional therapy. (1 session = 1 hour/day)

Functional Outcomes

The FES Therapy group realized statistically significant improvements in UE-FMA, CMSMR (arm & hand), BI, and self-care FIM™ over the Control group (Table 2). The FES group reported overall higher FIM™ compared to the Control group, but did not reach statistical significance.

Table 2: Functional Outcome Measures

Assessment	Control (n=11)		FES Therapy (n=10)		p-value
	Before	After	Before	After	
CMSMR (arm & hand)	3.5 (± 0.8)	4.3 (± 0.8)	3.1 (± 0.9)	5.4 (± 1.6)	< 0.02
UE-FMA	4.4 (± 4.6)	9.6 (± 13.7)	3.4 (± 4.8)	30.6 (± 15.5)	< 0.001
Barthel Index	42.7 (± 9.3)	74.5 (± 17.5)	42.5 (± 7.5)	89.5 (± 9.8)	< 0.05
FIM™	60.2 (± 11.6)	94.3 (± 19.2)	62.7 (± 9.1)	106.4 (± 6.6)	0.139
Self-Care FIM™	8.9 (± 3.5)	17.9 (± 8.8)	8.1 (± 3.3)	30.9 (± 6.6)	0.005

Five (5) of 10 patients in the FES Therapy group reported SC-FIM™ scores of 36 and 38, representing 86% and 90% of maximum SC-FIM™ = 42 (complete independence). No patient in the control group exceeded 30 points. The majority of the Control group remained ≤20 points, with 3 individuals in the Control group remaining highly dependent (≤10). (Table 3)

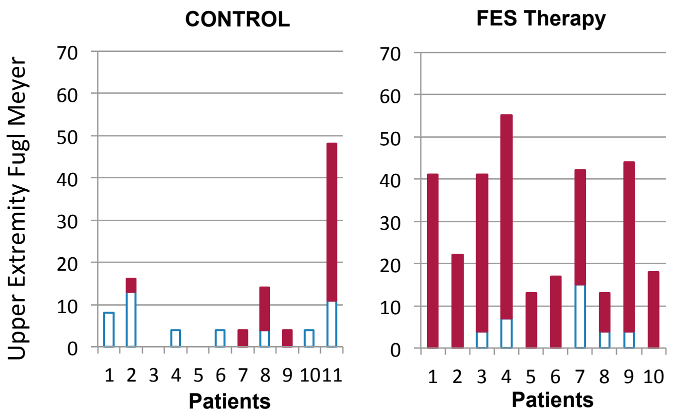
Table 3: Individuals in different SC-FIM™ ranges (min=6 indicates complete dependence; max=42 independence) before and after treatment.

Self Care-FIM™ Range	CONTROL (n=11)		FES Therapy (n=10)	
	Before	After	Before	After
≥31				
21 - 30				
11 - 20				
6 - 10				

Upper Extremity Fugl-Meyer (UE-FMA)

Every patient in the FES Therapy group realized a clinically significant gain in UE-FMA (median gain 24.5 points, range 9 – 48 points) while only 2 of 11 patients (18%) in the Control group realized gains of greater than 6 points. The median gain for the Control group was zero (0) (Figure 3).

Figure 3: Upper Extremity FMA for individual patients before ( ) treatment and Gain( ) realized after treatment (Maximum UE-FMA = 66 points)



Conclusion

As compared to an equivalent dose of conventional rehabilitation therapy programs, functional electrical stimulation (FES) therapy significantly improved voluntary motor function and self-care functional independence in stroke survivors with severe upper extremity impairment.

References

[1] Thrasher et al. Neurorehabilitation and Neural Repair. 2008. 22(6): 706-714.

[2] Popovic et al. Neuromodulation. 2005. 8(1): 60-74.

*Declaration of Interest - Dr. Popovic is a founder, a shareholder and the Chief Technology Officer of MyndTec Inc, a healthcare company created to commercialize technologies described in this presentation

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Made in Canada

FES Therapy for Grasping in Chronic Incomplete SCI: Pilot Randomized Control Trial

N Kapadia[1], V Zivanovic[1] and MR Popovic[1,2],

[1] Lyndhurst Centre, Toronto Rehabilitation Institute

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

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

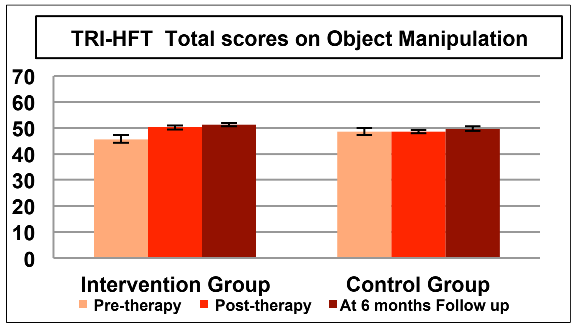
- Pilot randomized control trial in chronic (>24 months) incomplete C4-C7 SCI
- Control: n=3 and FES therapy: n=5
- Both groups received 39 sessions of therapy
- Control group received 1h of conventional occupational therapy (COT) and intervention group received 1h of FES therapy.
- Primary outcome measure was TRI-HFT and secondary outcome measures were FIM Self Care Sub score and SCIM Self Care Sub score.

Subject Demographics

Feature	Control Group	Intervention Group	P value
Age (years)			
Mean age ± SEM	49 ± 15.35	47.2 ± 2.23	0.922
Median age	62	54	
Age range	15-70	25 to 73	
Sex (n)			
Males	3	5	1.0
Females	0	0	
Cause of SCI (n)			
MVA	1	2	0.250
Fall	0	1	
Other causes	2	2	
Level of SCI (n)			
C3	0	0	
C4	2	0	0.393
C5	1	5	
C6	0	0	
Time since SCI (years)			
Mean time ± SEM	4.83±1.04	10.6±4.43	
Median time	6	5	0.436
Time range	2.5 to 6	2 to 26	

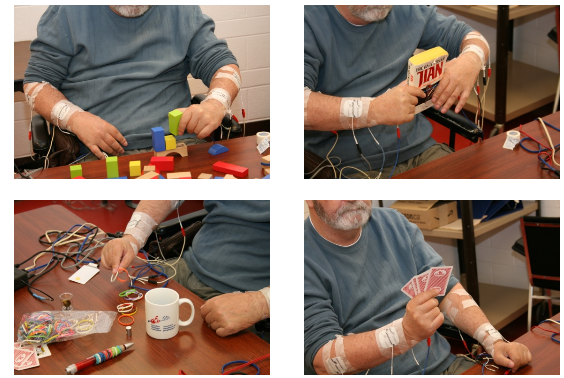
Results

- **FIM self care sub-score** improvements at discharge and at 6 months: ***FES therapy***: 4.6* and 8.3* points, & ***COT:*** 0 and 1.4 points, respectively.
- **SCIM self care sub-score** improvements at discharge and at 6 months: ***FES therapy***: 2.2* and 3.6* points & ***COT:*** 0.7 and 1.7 points, respectively.
- **TRI-HFT 10 objects** improvements at discharge and at 6 months: ***FES therapy***: 4.3 and 5.5 points, & ***COT:*** 0 and 1.1 points, respectively.



Test	Control Group (Mean scores)			Intervention Group (Mean scores)		
	Before	*After*	*6 Month*	*Before*	*After*	*6 month*
FIM Self Care Sub-scores	16.6	16.6	18	19	23.6*	27.3*
SCIM Self Care Sub-score	5.6	6.3	7.3	7	9.2*	10.5*
TRI-HFT Components						
10 Objects	48.5	48.6	49.7	45.7	50.1	51.3
Rectangular Blocks	30	31.2	26	40.8	41.1	42.9
Instrumented Cylinder *Torque Values (Nm)*	3.1	3.1	6.16	1.7	5.0*	7.6
Credit Card *Force Values (N)*	13.3	16.5	15	8.6	15.4*	14.3
Wooden Bar Thumb Direction *Length Values (cm)*	10	10	10	0.9	3.6	2.8
Wooden Bar Little Finger Direction *Length Values (cm)*	10	10	10	6	18*	11.3

FES Therapy Participant



Conclusions

- Restoration of voluntary hand function in chronic (>24 months) incomplete SCI is possible using FES therapy.
- Improvements in hand function and thereby increase in level of independence are significant with FES therapy.

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.
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Short-Term Neuroplastic Effects of Brain-Controlled Functional Electrical Stimulation

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Introduction

- Functional electrical stimulation (FES) can facilitate motor recovery after an incomplete spinal cord injury (SCI)
  - Thought to work by pairing signals in the upper motor neurons and α motor neurons, which meet at the spinal synapse.
  - This induces neuroplasticity through "Hebbian" mechanisms..
- FES is typically controlled by push-button.
- More empirical control may provide more consistent pairing of signals, and therefore greater neuroplasticity.
  - Based on detection of movement intention
    - Brain-machine interface (BMI)
    - Electromyography (EMG)

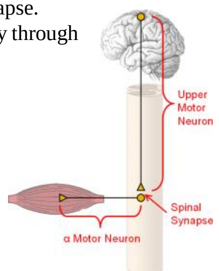

Fig. 1: Diagram of the corticospinal tract

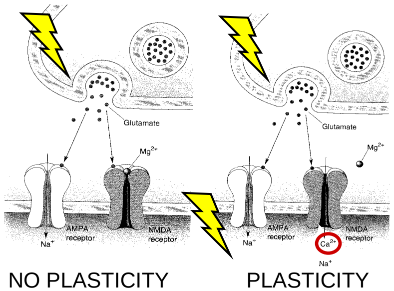

Fig. 2: Mechanisms of Hebbian plasticity. NMDA channels are only open when both pre-and post-synaptic cells are simultaneously active. This in turn causes Ca2+ influx through NMDA channels, which in turn causes synaptic strengthening.

Hypothesis

- Brain-controlled FES (BMI-FES) and muscle-controlled FES (EMG-FES) should produce greater neuroplasticity than conventional FES.

Methods

- 10 able-bodied participants
- Each received all 5 interventions, each involving 1h of grasping practice:

Grasp \ Control	Voluntary	EEG	EMG
Voluntary	VOL	BMI	
FES	FES	BMI-FES	EMG-FES

Fig. 3: Intervention outline

- FES facilitates grasping movements.
- BMI is non-invasive (electroencephalography (EEG)-based).
- EMG is non-invasive.

- Assessments pre- and post-intervention
  - Grip force
  - Maximum voluntary contraction (MVC)
  - Motor evoked potential (MEP):

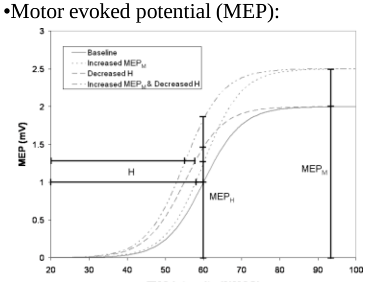

Fig. 4: Outline of MEP-based outcome measures

- Paired-samples t-tests (within-subjects design)

Results

- No voluntary BMI control obtained:
  - Chance accuracy: 50%
  - Range: 43.5-58.1%
  - Mean: 49.4%
- MEP upregulated, but not grip force or MVC:

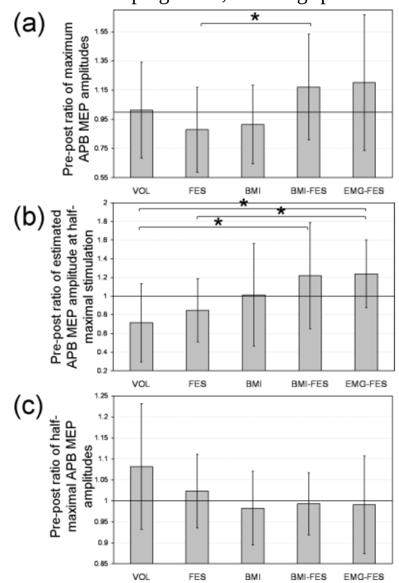

Fig. 5: Pre-post changes in MEP-based measures: (a) MEP_M (b) MEP_H (c) H

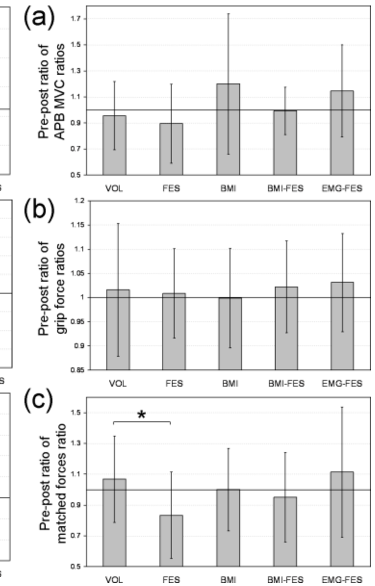

Fig. 6: Pre-post changes in non-MEP-based measures: (a) MVC (b) grip force (c) matched forces ratio

Conclusions

- BMI-FES and EMG-FES provide greater neuroplasticity than FES or voluntary grasping.
- This suggests their potential as novel therapeutics for motor recovery following SCI.

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Assessment of Sensitivity to Anatomical Landmark Misplacement Error in Human Trunk Motion Analysis

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Introduction

The reliability of kinematics measurement of the trunk and vertebral column have been of concern in a variety of clinical evaluations for patients with low back pain, spinal cord injury, etc. Among different types of experimental errors, anatomical landmark misplacement errors made by examiner while palpating, have been considered as a major source of error in assessment of 3D joint angles. Such induced errors have been widely discussed for lower limb joints, but were never assessed for multi-segment trunk joints; which this study aimed to investigate.

Methodology

a) *Measurement protocol*

- Trunk modeled as a multi-segment inverted pendulum: upper thoracic (UT), mid-upper thoracic (MUT), mid-lower thoracic (MLT), lower thoracic (LT), upper lumbar (UL), lower lumbar (LL), and sacral (SC).
- Eleven healthy subjects. Three trials for each subject. Bending in 5 directions (Fig1). Six cameras (VICON, UK) recorded markers trajectories.

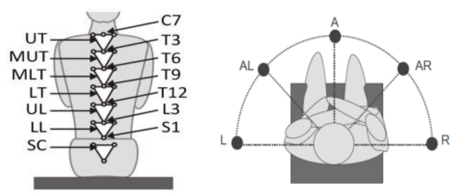

Fig1: Markers 10 cm apart. The five directions of bending: Left, Anterior Left, Anterior, Anterior right, Right.

b) *Data Analyses*

- Group of three markers over a segment formed a local coordinate system. 3D joint angles between consecutive segments were calculated based on joint coordinate system (JCS) convention. Then the range of motion (ROM) in the time frame was calculated.

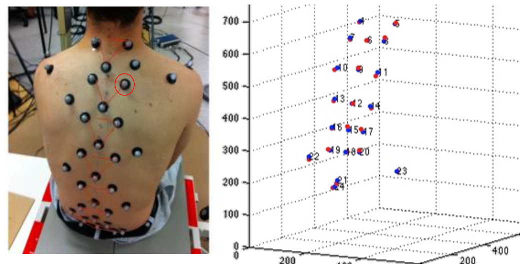

Fig.2: a) landmark palpation error region shown as a red circle around a marker. b) Simulation of original trunk markers(blue) and erroneous trunk markers (red).

b) *Data Analyses (cont.)*

- Anatomical landmark misplacement error was simulated as a Gaussian error with dispersion of 7 mm at each of the two directions on the skin surface in vicinity of the bony landmark on which the marker was placed.
- This random error was added to the marker trajectory and the erroneous 3D joint angle was calculated (as an angular curve during trunk bending).
- The difference between the range of erroneous and original angles divided by the original range was calculated as the relative error (%).

Results

As shown in the ROM of original angles in Fig.3., in all the three planes range of motion of the joints doesn't go beyond 10 degrees. In anterior bending the sagittal plane has the major contribution in bending motion and in the more lateral bending (Left or right bending) the angular motion in the frontal plane have close range of motion to sagittal angles.

Based on the relative errors in measurements of ROMs (Fig.4.) the error in sagittal plane is less than %8 for all directions, however it is significantly larger in frontal and transverse angles. Specifically in MLT-LT joint and UL-LL joint that exceeds %30.

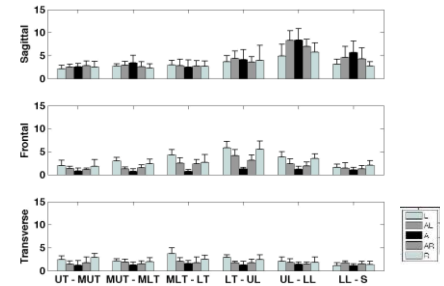

Fig.3: The range of angular motion (ROM) of the joints between trunk segments (Segment1-Segment2). ROMs are presented in degree as median ±SD among subjects.

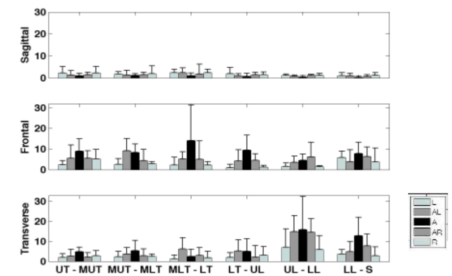

Fig.4: The errors propagated in ROMs of the joints between trunk segments due to anatomical landmark misplacement. The relative ROM errors are presented in percentage of the original ROM as median ±SD among subjects.

Conclusion

Based on the results, the marker misplacement error caused ignorable errors in the joints' ROM in the sagittal plane for all directions. Therefore the measurements of the sagittal motion is reliable under marker misplacement error.

However, in anterior bending, the induced relative errors in frontal and transverse motions were much larger, which is due to the small amplitude of the joints' ROM in those planes.

Therefore, especially if the data is used for clinical assessments, the frontal and transvers angle measurements in joints with small range of motion (i.e. MLT-LT; UL-LL) should be considered with caution.

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Functional Electrical Stimulation of Trunk Muscles Increases Trunk Stiffness

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Introduction

Sitting Balance
- maintained by activating trunk muscles
- neurological impairment (e.g., spinal cord injury) could cause trunk muscle paralysis and poor sitting balance

Functional Electrical Stimulation (FES)
- activates muscle with short electric impulses
- generates muscle co-contractions which can increase stiffness of the trunk
- used as an assistive-device (i.e., neuroprosthesis)

Objectives

Study 1: Characterize sitting stability while FES is applied on the trunk muscles - an experimental study

Study 2: Investigate the impact of stiffness on sitting stability - a simulation study

Study 1: FES-assisted Sitting

Methods

- **Participants**: n = 15; able-body; 26.7±4.6 yrs
- **Measurements**: center of pressure (COP) on the seat for anterior-posterior (AP) and medial-lateral (ML) fluctuations: 60s trial during quiet sitting
- **Assessments**: mean distance (MD), mean velocity (MV), mean frequency (MFREQ), centroidal frequency (CFREQ), frequency dispersion (FREQD) and 95% power (P95) [1]

- **Experimental Conditions**

  UN = unsupported sitting

  FES = FES-assisted sitting: Compex Motion (Switzerland) surface stimulation (300µsec pulse duration; 40Hz frequency) to co-contract the rectus abdominis (RA): 20.3±3.8 mA and erector spinae (L3): 24.6±7.4 mA

Fig. 1 - Experimental setup

Results

- AP sway velocity was increased during FES condition

Measures		UN	FES	Sig.
MD (cm)	AP	0.61±0.26	0.54±0.24	
	ML	0.52±0.26	0.46±0.28	
MV (cm/s)	AP	2.80±0.41	3.04±0.61	**
	ML	2.02±0.54	2.30±0.77	
MFREQ (Hz)	AP	0.97±0.30	1.13±0.35	**
	ML	0.82±0.22	1.01±0.27	**
CFREQ (Hz)	AP	1.72±0.23	1.82±0.25	
	ML	1.71±0.19	1.67±0.27	
FREQD (-)	AP	0.59±0.05	0.56±0.03	*
	ML	0.56±0.05	0.54±0.06	
P95 (Hz)	AP	0.72±0.18	0.85±0.18	*
	ML	0.81±0.28	0.87±0.31	

** $p < .05$; * $p < .01$

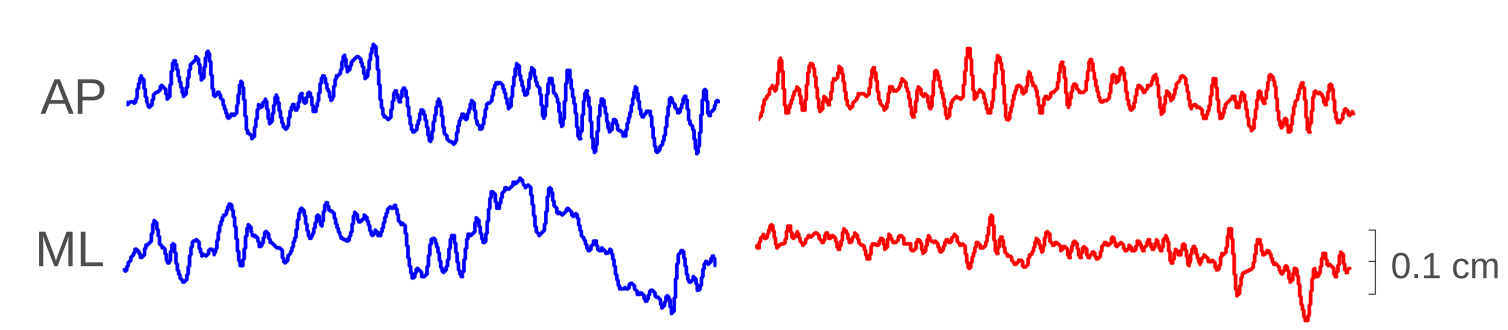

Fig. 2 - COP fluctuations during 15 s comparing UN and FES

Study 2: Effects of Increased Stiffness

Methods

- **Inverted pendulum**: m = 49.2kg; h = 0.38m; I = 7.14 kg·m²
- **Stiffness and damping**: 0 < K < 300 Nm/rad; B = 6 Nms/rad [2]
- **Neural controller**: PD control; 0 < K_P < 500 Nm/rad; 0 < K_D < 300 Nms/rad [2]; transmission delays: $\tau = \tau_1 + \tau_2$ = 40 ms; neuromusculoskeletal (NMS) torque generating process: $T = 1/\omega_n$ = 120, 170 and 220 ms [3]
- **Simulation**: stable gain combinations were used to simulate COP fluctuations with various stiffness

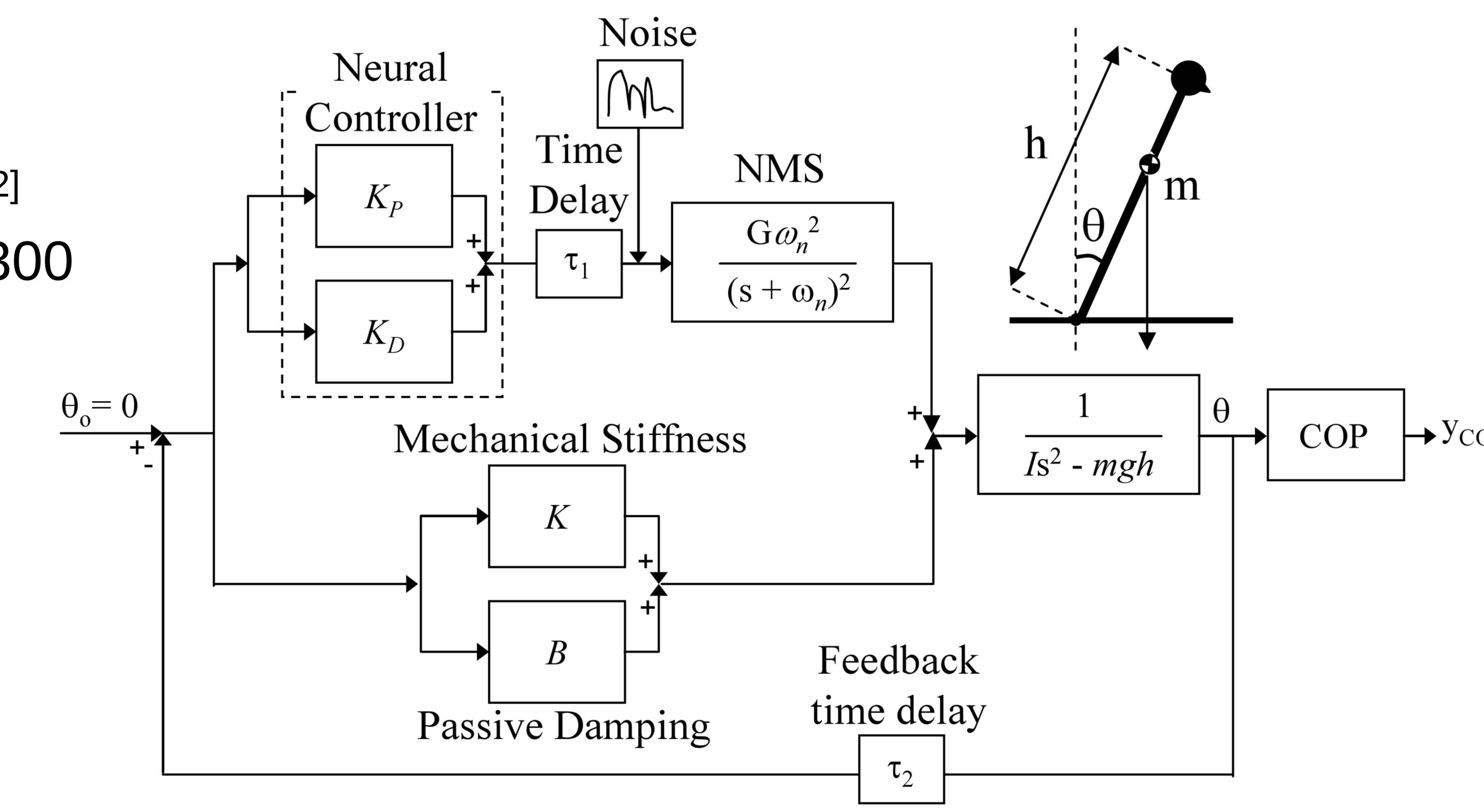

Fig. 3 – Simulation model

Results

- partial correlations between stiffness and COP parameters
- increased mechanical stiffness had same effect on postural stability as FES-assisted sitting

Measures	K	
MD	-0.035	
MV	0.283	*
MFREQ	0.927	*
CFREQ	-0.034	
FREQD	-0.543	*
P95	0.422	*

* $p < .01$

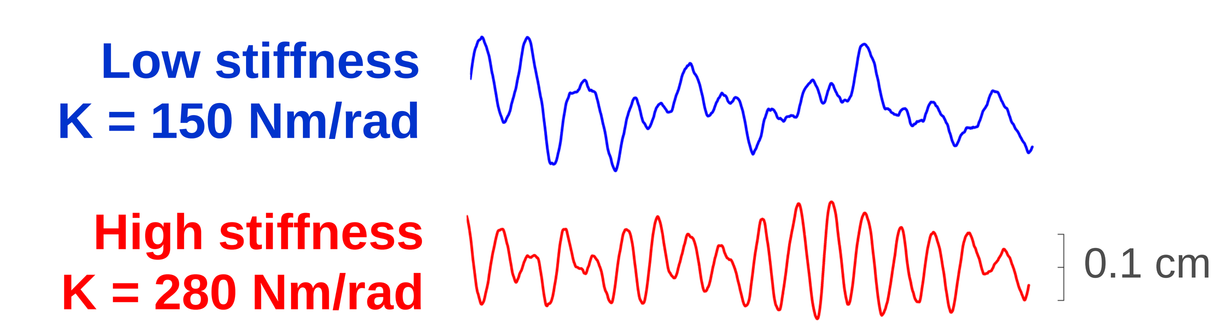

Fig. 4 - Simulated COP during 15 s comparing low and high stiffness (K_P = 50 Nm/rad; K_D = 50 Nms/rad; B = 6 Nms/rad; T = 120 ms)

Conclusion

- FES increases trunk stiffness that increased fluctuation velocity during quiet sitting
- AP fluctuations velocity was increased because RA and L3 muscles control AP posture during sitting [3]
- increased stiffness with FES can improve dynamic balance and help individuals with neurological impairments [4]

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[1] Prieto TE et al., *IEEE Trans Biomed Eng*, 43, 956-966, 1996

[2] Goodworth AD et al., *J Neurophysiol*, 102(1), 496-512, 2009

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Effects of Foot Support on Sitting Postural Stability after Spinal Cord Injury

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Introduction

Sitting Balance
- maintained with activation of trunk muscles
- foot support likely affects sitting balance but mechanisms are not fully understood

Spinal Cord Injury (SCI)
- cervical SCI results in impairment of trunk and lower limbs
- people with cervical SCI usually have difficulty sitting

Objectives
1) compare sitting of able-body (AB) and cervical SCI group
2) analyze the effects of foot support on sitting

Methods

Participants
- AB: n = 10; 31.0±5.9 yrs; 174.4±9.5 cm; 68.5±13.2 kg
- SCI: n = 6; cervical 4 - 6 level; 41.3±18.1 yrs; 175.3±5.1 cm; 75.1±9.8 kg

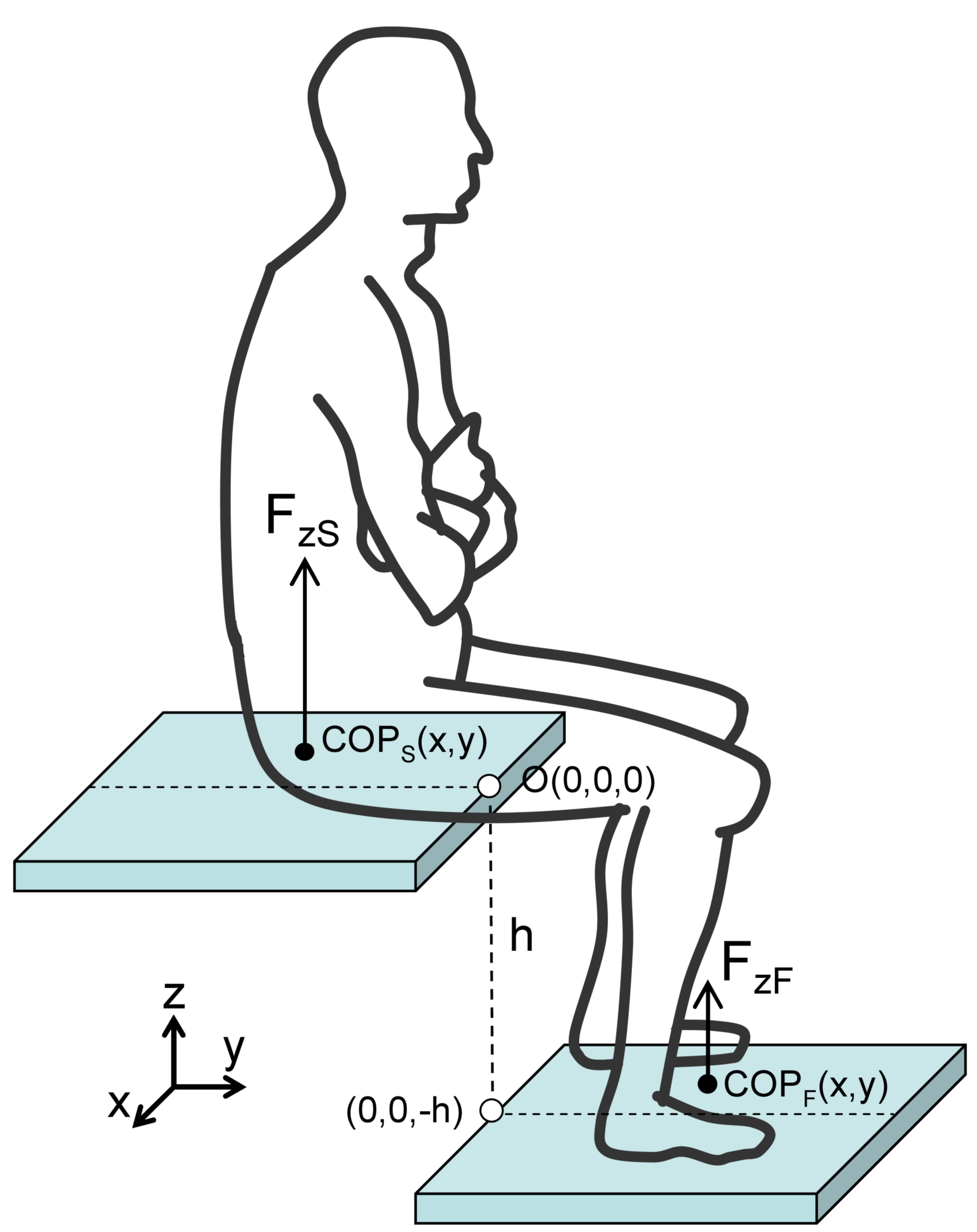


Fig. 1 - Experimental setup

Center of Pressure (COP)

- **Measures**: trunk fluctuations (COP_S) and foot support (COP_F) fluctuations using two AMTI AccuSway^{Plus} force plates: 2 x 60s during quiet sitting (Fig 1)
- **Calculated**: global (COP_G) fluctuations of the whole body
- **Assessments**: medial-lateral (ML), anterior-posterior (AP) and resultant distance (RD) direction fluctuations mean distance (MD), mean velocity (MV), 95% confidence ellipse (AREA-CE), mean frequency (MFREQ), centroidal frequency (CFREQ) and frequency dispersion (FREQD) [1]

Results

- SCI group has larger sway compared to AB group for global and seat fluctuations
- fluctuations with foot support (COP_G) were faster compared to no foot support (COP_S), mainly affecting AP sway
- vertical (F_Z) and shear (F_X and F_Y) forces were not different

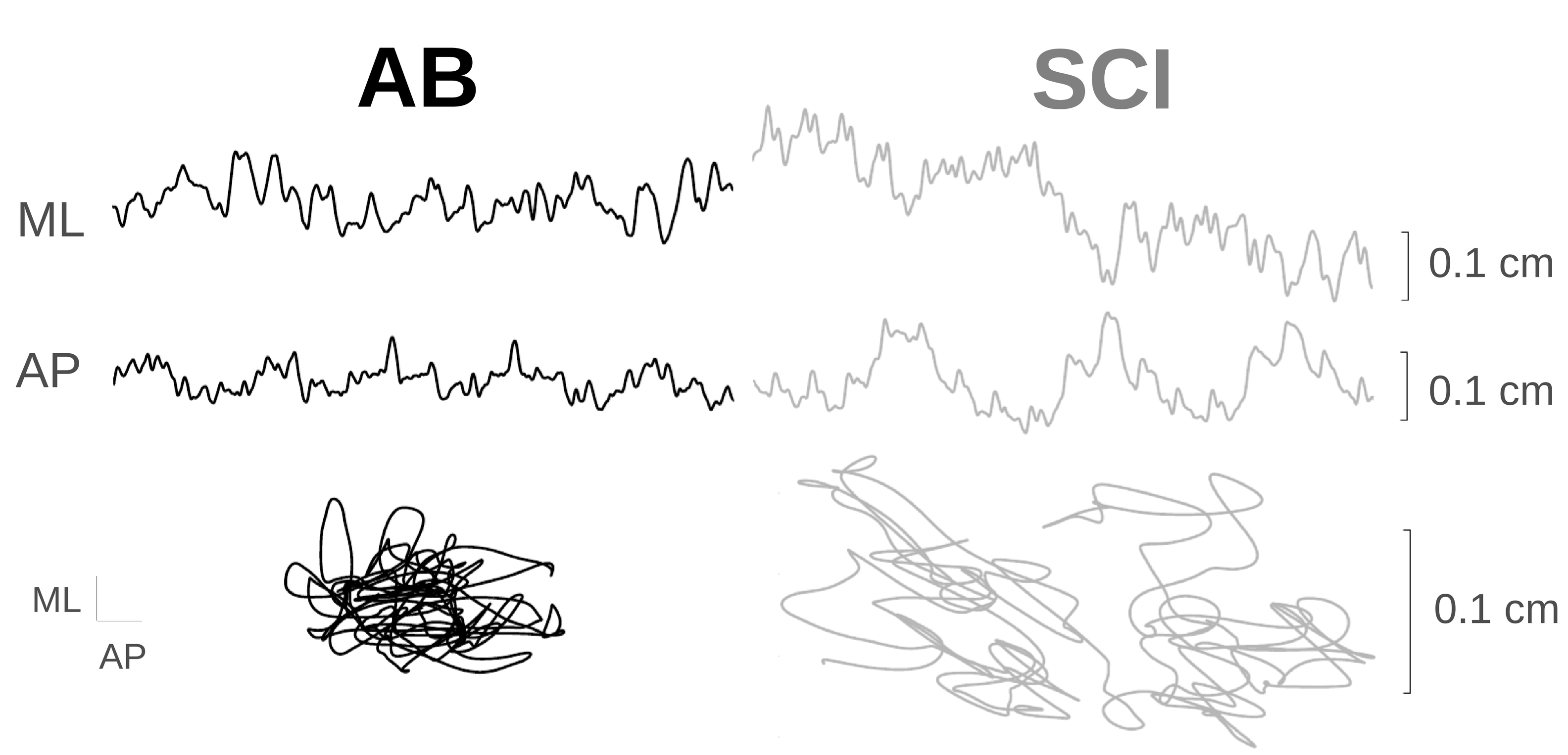


Fig. 2 - COP_G fluctuations during 15 s for one individual in each group

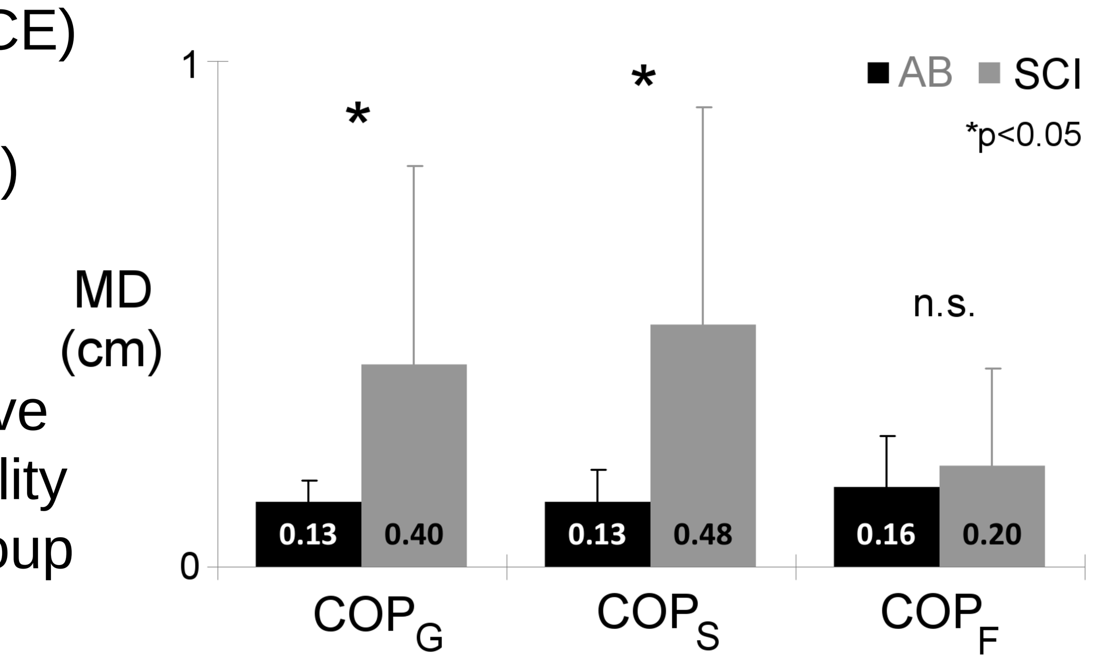
$$COP_G(x,y) = COP_S(x,y)\cdot\frac{F_{zS}}{F_{zS}+F_{zF}} + COP_F(x,y)\cdot\frac{F_{zF}}{F_{zS}+F_{zF}} + COP_{shear}(x,y)$$

$$COP_{shear}(x,y) = \left(h\cdot\frac{F_{xF}}{F_{zS}+F_{zF}},\ h\cdot\frac{F_{yF}}{F_{zS}+F_{zF}}\right)$$

Sitting after Cervical Injury: AB vs. SCI

- SCI group sway amount (MD) larger for COP_G and COP_S
- sway area (AREA-CE) larger in SCI group
- frequency (MFREQ) larger in SCI group

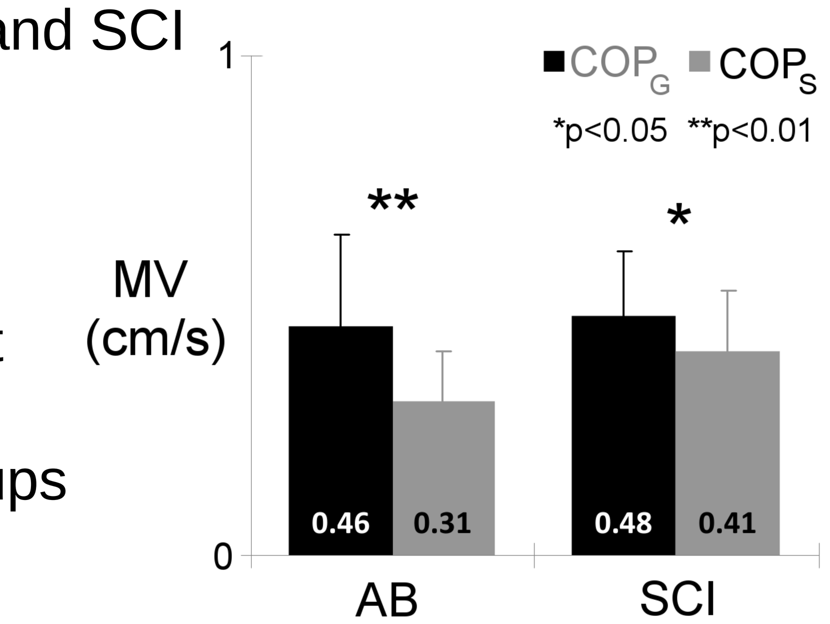
Interpretation:
- people with SCI have lower postural stability compared to AB group



Effect of Foot Support: COP_G vs. COP_S

- sway velocity (MV) of COP_G (with foot support) larger than COP_S (no foot support) in AB and SCI
- frequency (CFREQ) also larger with foot support

Interpretation:
- foot support increased amount of postural regulatory activity
- foot support same in both groups



Conclusion

- foot support is approximately 20% and head, arm and trunk (HAT) is 80% of total body weight during sitting
- deficient trunk control is the dominant mechanism of sitting instability in individuals with SCI
- foot support has a passive function in postural stability
- foot support has to be measured to assess sitting posture

References

[1] Prieto TE et al., *IEEE Trans Biomed Eng*, 43: 956-966, 1996
[2] Grangeon M et al., *Gait Posture*, 36: 572-579, 2012

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Multi-Segment Kinematic Assessment of Human Trunk: Sensitivity to Skin Artifacts

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Introduction

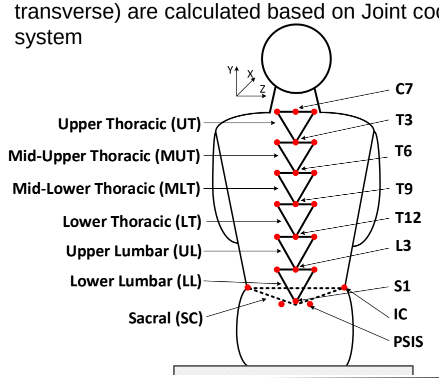
- Kinematic assessment of the trunk based on a multi-segment trunk model is valuable in for evaluations of a wide range of back pathologies and conditions (e.g. low back pain, scoliosis and spinal cord injuries)
- The standard practice is using motion capture systems and skin mounted reflective markers.
- The challenge is that the skin and soft tissue cause the markers to move relative to the underlying bone.
- A study that investigated the role of skin artifacts in trunk motion analysis is not yet available.

Objective

To investigate the propagation of skin artifacts to the 3D inter-segmental joint angle assessment in the seven-segment trunk model.

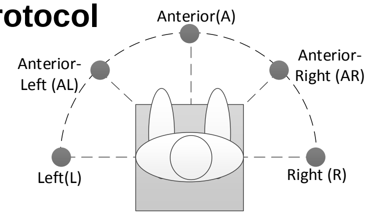
Multi-segment Model of the Trunk

- Every 2,3 vertebrae are considered as a segment
- 3+ reflective markers placed on and besides anatomical landmarks (spinous processes) for motion assessment of each segment (23 reflective markers, 7 trunk segments)
- 3D inter-segmental angles (sagittal, coronal, transverse) are calculated based on Joint coordinate system



Experimental Protocol

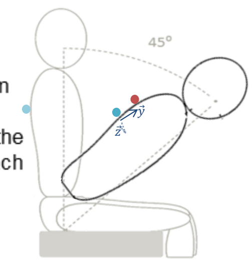
- 11 healthy subjects
- Age: 28.5±3.3
- Trunk Height: 0.75±0.04 m
- 45° trunk bending in 5 directions
- 3 trials per direction



Skin movement artifact Simulation

1. Measured maximum displacement in the $\overline{yz}$ plane of the anatomical frame at 45° bending for each marker location:
   - Mean(μ) and standard deviation (σ) of 5 subjects
2. Assigned a Gaussian number in the maximum bending position for each marker:

$$MaxD_{ij}^{d} \sim N(\mu, \sigma^2)$$



$$\begin{cases} j = y,z \ \textit{coordinates in anatomical frame} \\ d = \textit{direction of the bending}\,(A, AL, AR, L, R) \end{cases}$$

3. Determined the skin movement artifact throughout the course of the bending task

$$D_{ij}^{d}(t) = C(t) \times MaxD_{ij}^{d} \quad ; \quad C(t) = \frac{Proj_T\,\overrightarrow{CS}(t)}{Proj_T\,\overrightarrow{CS}(t_{45^\circ})}$$

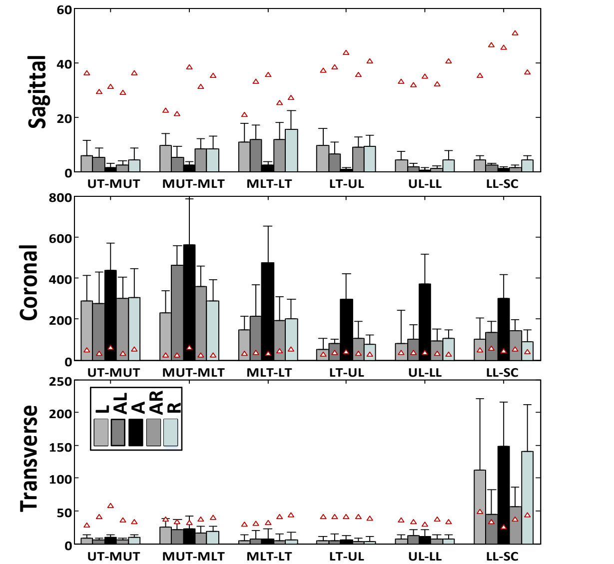
$\overrightarrow{CS}$ = the distance vector between C_7 and S_1 in the global frame

4. 1000x random error generated

Calculating error propagated to angles

- Skin movement artifact added to marker trajectory and 3D joint angles re-calculated
- Relative error in range of motion(ROM), 95% of 1000 runs

Results



Error in inter-segmental ROMs in **percentage** of the original ROM
Inter-Subject variability (SD/mean%)

Conclusions

- Implementing similar sensitivity analysis is a necessary step before kinematic analysis of the trunk.
- Error propagation introduces distortions:
  - To the coronal plane angles to the extend that information relative to the angular movements of the underlying bone is concealed.
  - To LL-SC and MUT-MLT angles on the transverse plane.
- Angles on the sagittal plane are relatively reliable.
- These findings are guidelines for interpreting clinical evaluations and developing skin artifact compensation techniques.

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Comparison of single- and double-link inverted pendulum models for human standing posture

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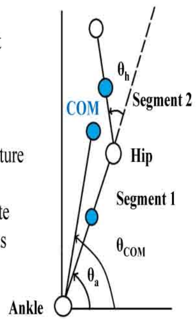
Purpose

To determine the feasibility of the single-link inverted pendulum (SIP) model for the design of a functional electrical stimulation (FES) controller for human standing

Background

- SIP model assumes the human body to be one segment that rotates about the ankle joint, which has been used in many studies
- Recent studies claim that a double-link inverted pendulum model (DIP), should be used even when quiet standing posture is analyzed
- There may be a misunderstanding that SIP is not appropriate any more for modelling quiet standing posture. However, as long as ankle joint control is considered, SIP must still be appropriate.



Methods

Subjects

- 10 healthy young adult males; age 19.9 ± 2.3 yrs (16 – 23 yrs); height 174 ± 5.8 cm (168 – 187 cm); weight 69 ± 8.7 kg (55 – 85 kg)

Task

- Stand with arms crossed against chest for 120 seconds with eyes open or closed
- Stand with a natural voluntary sway in eight directions (10 times each)

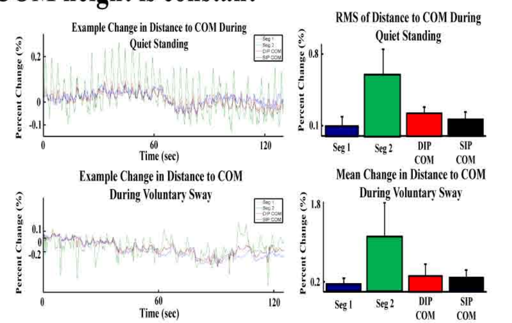
Measurements

- 3D body kinematics by a motion capture system (Cortex V3.1.101)
- Force components and centre of pressure by 2 force plates (AMTI Accusway)

Data Analysis

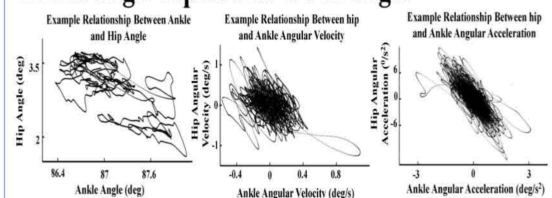
- Body segment centre of masses (COM) were calculated using a 14 segment model with 29 reflective markers
- Ankle, hip and COM angles were calculated using kinematics
- Ankle and hip torques were calculated using kinematic information
- Ankle torque was also calculated using kinetic information from force plates

COM height is constant

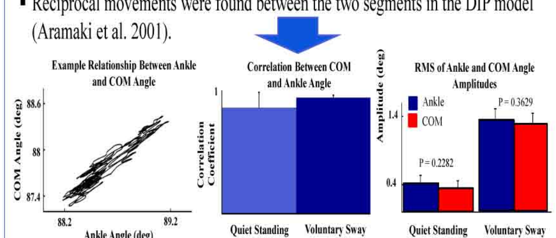


- The height of COM (distance between ankle and COM) changed less than 1%, suggesting that quiet standing posture can be modelled as a rigid segment
- As the change of length of each segment in DIP also showed very small changes, DIP model is also appropriate

Ankle angle represents COM angle

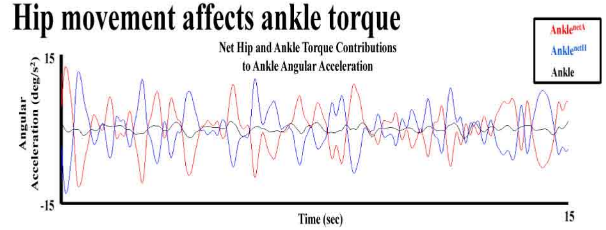


- Reciprocal movements were found between the two segments in the DIP model (Aramaki et al. 2001).

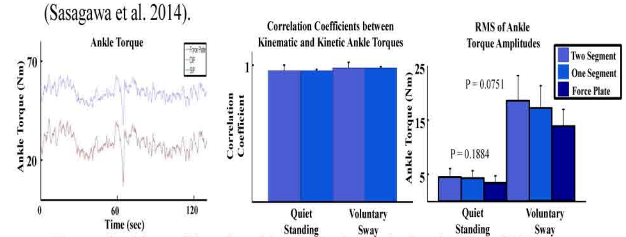


- Even with the reciprocal behaviour of leg and trunk segments, the ankle angle represents COM angle

Hip movement affects ankle torque



- Trunk behaviour affects ankle angular acceleration, i.e. induced acceleration (Sasagawa et al. 2014).



- Even under this condition, the ankle torque calculated using the SIP and DIP models from kinematic information was equivalent to the ankle torque calculated using kinetics (gold standard) – DC offset is due to the force plate and marker coordinate systems being misaligned
- Thus suggesting that the ankle torque is controlling the COM regardless of segmental movements above the ankle joint

Conclusion

- We confirmed that the segmental movement during quiet standing and voluntary sway was represented by the DIP more than SIP model.
- However, we verified that the ankle torque is controlling the COM whose height is approximately constant
- The results suggest that it is feasible to use a SIP model for an FES controller for the ankle joint in human standing

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Spatially distributed sequential stimulation improves fatigue resistance in plantar flexors

378.20

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INTRODUCTION

Neuromuscular electrical stimulation (NMES) is used to promote physiological and functional improvement in paralysed limbs [1, 2] and counteract musculoskeletal atrophy [3, 4]. A critical limitation with NMES approach is the rapid onset of muscle fatigue during repeated contractions, which results in the muscle force decay and slowing of muscle contractile properties [5, 6]. Recently, we demonstrated that spatially distributed sequential stimulation (SDSS) shows a greater fatigue-reducing ability compared to a single active electrode stimulation (SES) [7]. The goal of the present study is to extend the reported findings on the fatigue-reducing ability of interleaved sequential stimulation in the able-bodied population.

PURPOSES

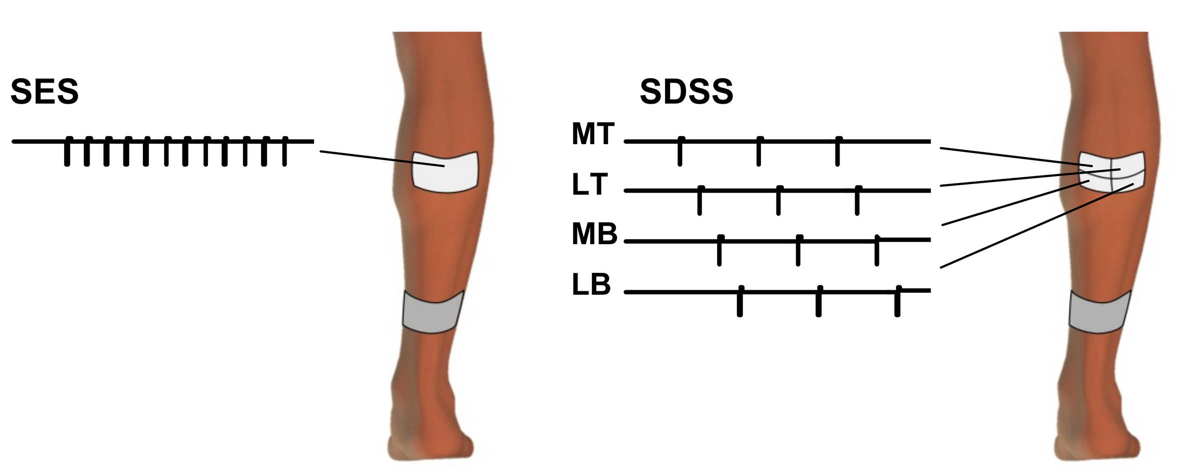
1. To explore the fatigue-reducing ability of SDSS in more details especially focusing on the muscle contractile property.
2. To investigate the mechanism of this effect using array-arranged electromyogram (EMG).

HYPOTHESES

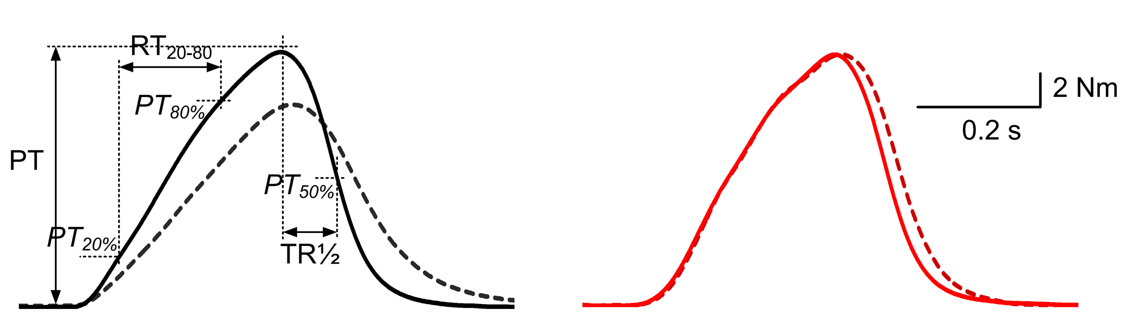
1. Reduced muscle fatigue during SDSS may be achieved in non-paralyzed muscles as evidenced by the muscle force decay and slowing of muscle contractile properties.
2. During SDSS, the magnitude of the muscle response in different portions of the muscle will depend on the location of the stimulation.

METHODS 1

SDSS was delivered through four active electrodes applied to the muscle of interest, sending a stimulation pulse to each electrode one after another with 90° phase shift between successive electrodes. For comparison, a single active electrode stimulation (SES) was tested too. For both modes of stimulation, the resultant frequency to the muscle as a whole was 40 Hz.

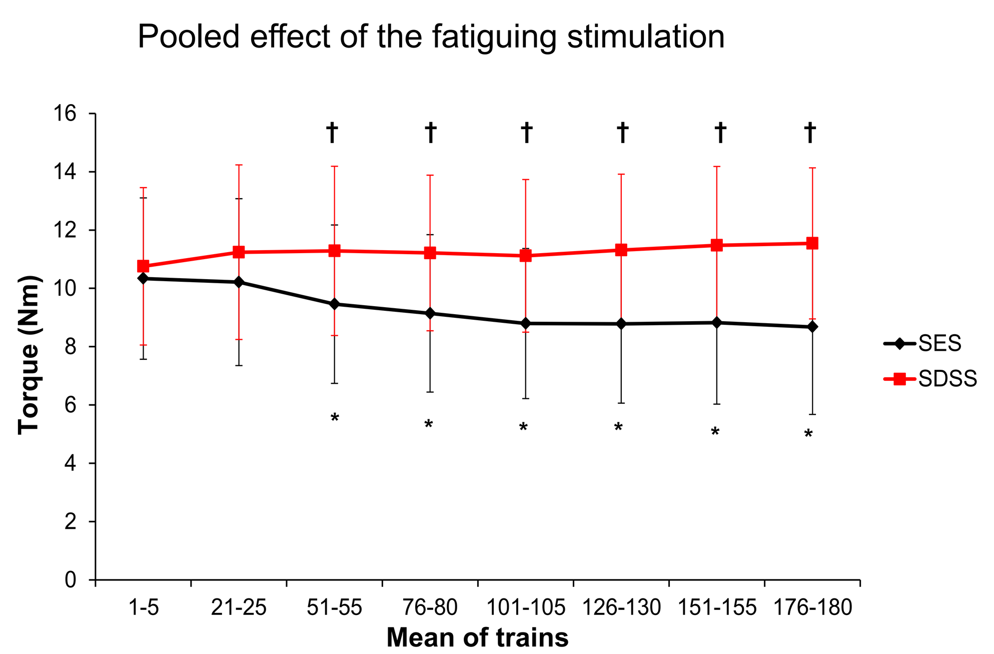


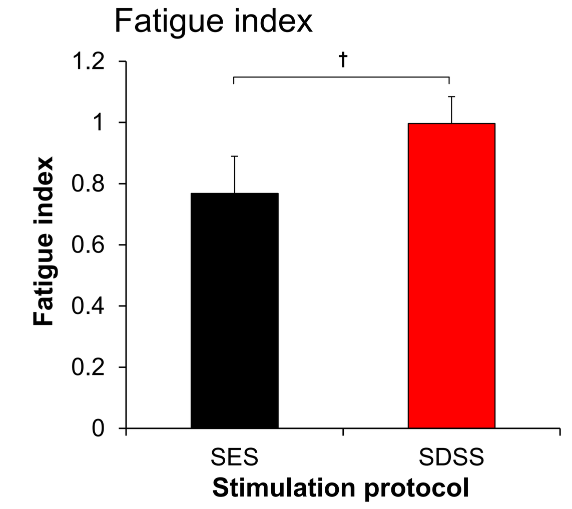
The exerted ankle torque during a 2 minute fatiguing plantarflexion was measured and compared between SDSS and SES.

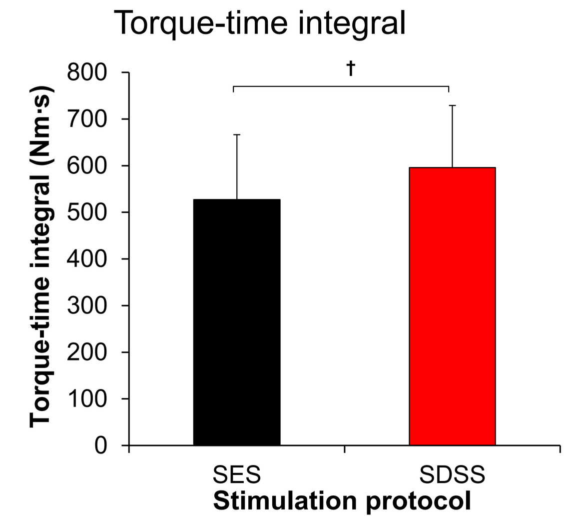


RESULTS 1

1

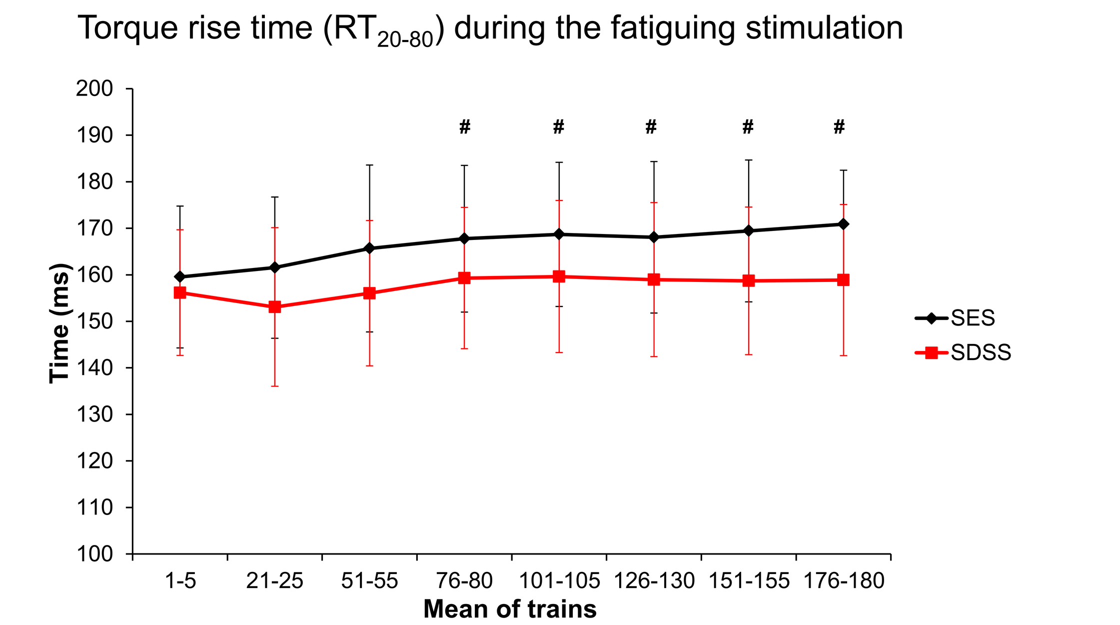






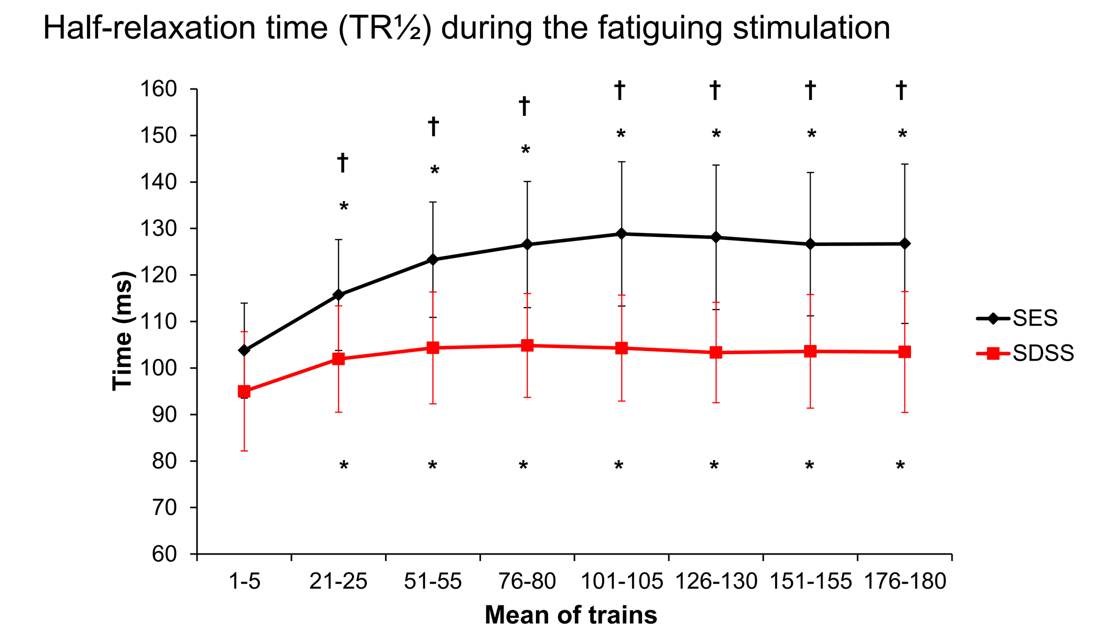
1) During SES and SDSS the muscle force decay developed differently across time.
2) Fatigue index was higher during SDSS.
3) Torque-time integral values were higher during SDSS.

2



4) RT_{20-80} increased during SES, and did not change during SDSS.

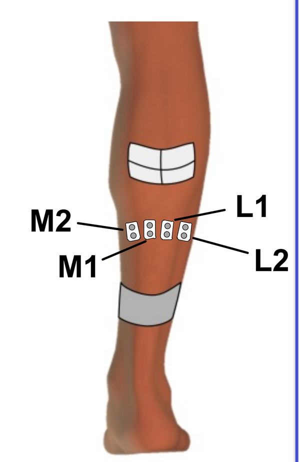
3



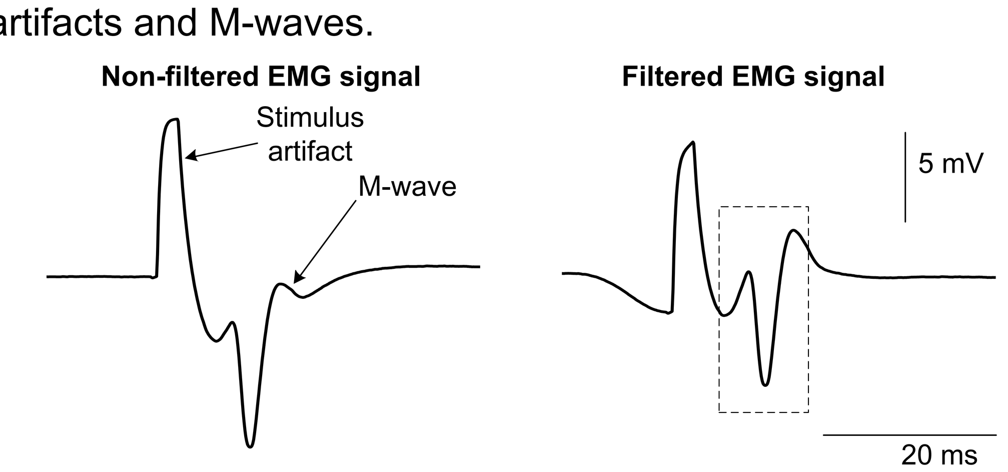
5) Significant prolongation in muscle relaxation during the fatiguing stimulation occurred during both SES and SDSS protocols.
6) The reduction of muscle relaxation was larger during SES than during SDSS.

METHODS 2

Surface EMG signals were recorded via four bipolar surface electrodes (Ag/AgCl, diameter of 9 mm, inter-electrode distance of 22 mm, Bortec Biomedical Ltd., Canada) placed at equal distance from each other on the right soleus muscle (M2–L2).

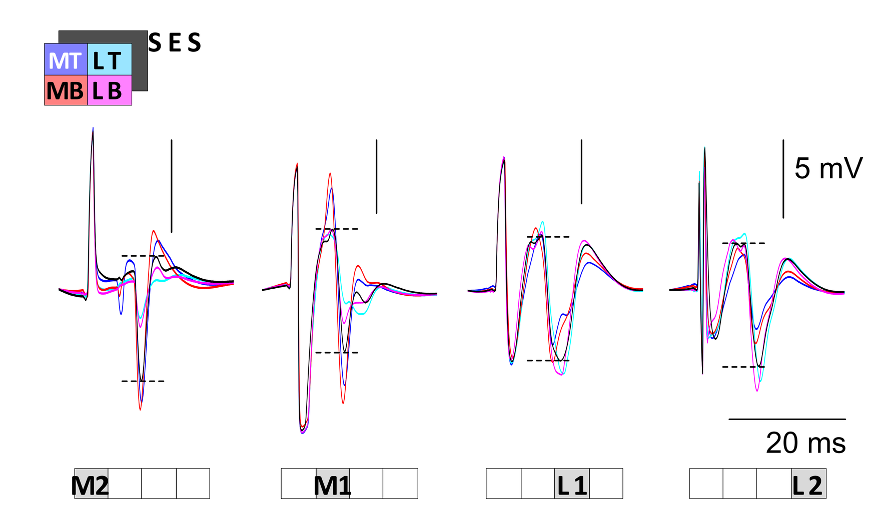


The EMG data were filtered with a high-pass filter with cut-off frequency of 50 Hz to minimize the interference from the overlapping of stimulation artifacts and M-waves.

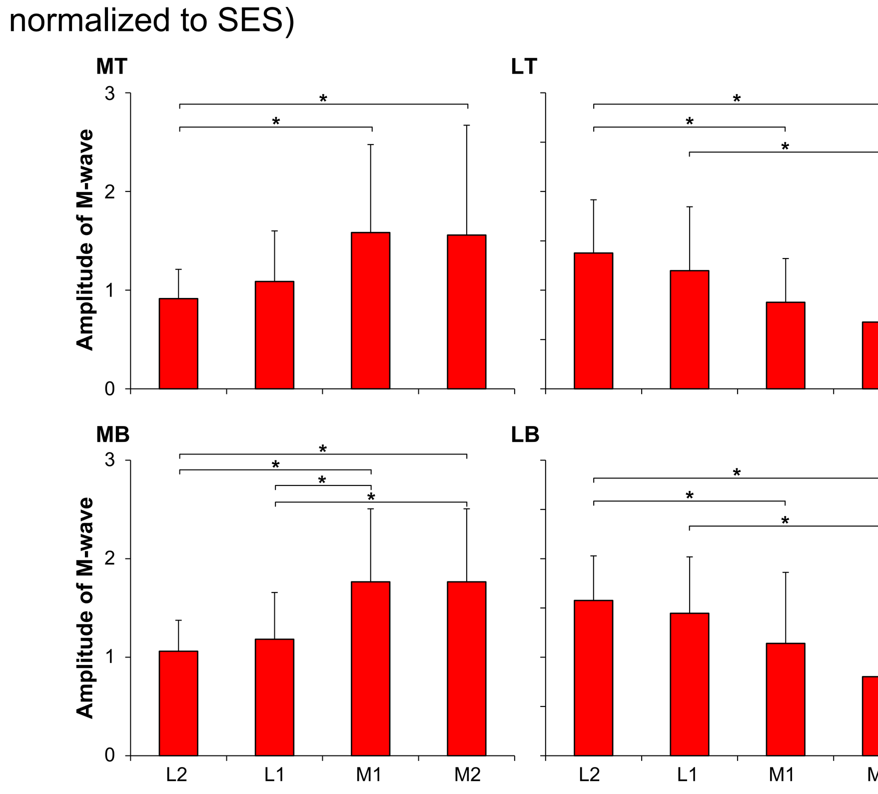


RESULTS 2

M-waves recorded in each active electrode location during SDSS and SES in one participant



Normalized amplitude of the M-waves from each of the four muscle portions dependent on the location of the stimulation during SDSS (data normalized to SES)



7) During SDSS, the magnitude of the muscle response in different portions of the muscle depends on the location of the stimulation.

CONCLUSIONS

1. The present study demonstrated the fatigue-reducing ability of spatially distributed sequential stimulation in able-bodied population through higher fatigue indices, higher torque-time integral, as well as less change in the muscle contractile properties as compared with the single active electrode setup.
2. The observed effects were most likely caused by different sets of muscle fibers being activated by different electrodes during SDSS, which is closer to physiological (voluntary) activation.

ACKNOWLEDGEMENTS

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ANTICIPATORY POSTURAL ADJUSTMENTS AND STEP EXECUTION DURING GAIT INITIATION IN WATER: A PILOT STUDY

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Background and Aim

Aquatic therapies are often used in attempt to facilitate gains in postural stability (*Berger et al., 2006*) and mobility (*Tripp and Krakow, 2014*) in people with disabilities. It has been reported that aquatic environment has the potential to improve mobility due to its properties such as buoyancy and resistance (*Becker, 2009*). Although biomechanical parameters of steady-state gait have been examined in water (*Barela et al. 2006*), to date, none has been reported on the anticipatory postural adjustments (APAs) and on the execution of the first step in water. We investigated how immersion in water changes APAs, the center of pressure (COP) parameters and kinematics of the initial step when able-bodied subjects walk in water in comparison to their performance on dry land.

Methods

Subjects

Seven female able-bodied volunteers from Cruzeiro do Sul University (*Table 1*).

Task

Participants initiated gait following an auditory cue and walked 4 steps forward. Initial foot position was self-selected and maintained in all trials. Arms' positions were maintained at approximately 90 degrees of elbow flexion during trials in water and on dry land. Five trials were used for analysis.

Apparatus and data collection

Participants initiated gait on a waterproof force plate (AMTI, ORP-WP-1000, USA). A digital camera (Sony DCR-HC28) was positioned approximately 3 m distance from the sagittal plane of the movement. Passive circular markers were placed on the 5th metatarsal, lateral malleolus, lateral femoral epicondyle, and greater trochanter. Sampling frequency for forces and images were 1024 Hz and 60 Hz, respectively. Low pass filter of 20 Hz and 8 Hz were applied to the forces and kinematic data, respectively.

Dependent variables

COP trajectory was divided into three sections: **S1** (APA)**, S2** (COP shifts laterally to stance limb)**, and S3** (COP moves forward during initial step) *(Hass et al. 2008) and* **duration, displacement and velocity** were analyzed (*Figures 1 and 2, Table 2*). **AP and ML peak of APA,** and **time to unload** were also examined. Kinematic parameters were **step length, duration and velocity, toe-clearance, ankle and knee ROM** during swing phase of the first step.

Results

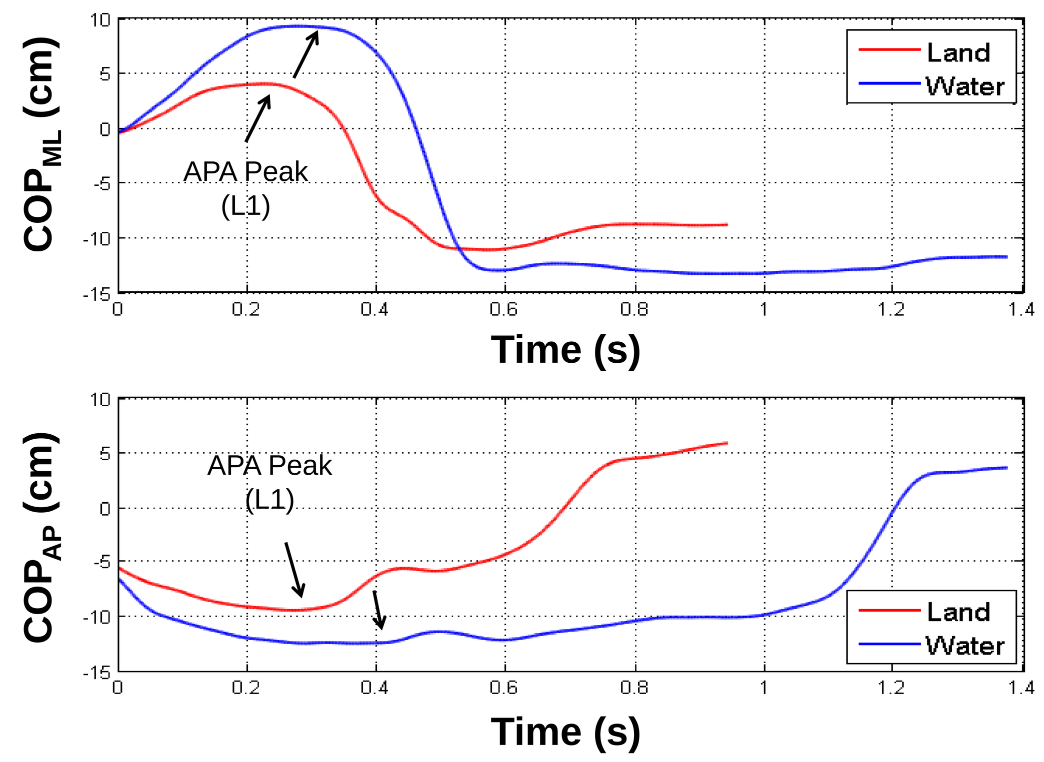
Table 1. Subjects' characteristics

Subjects	Gender	Age (years)	Height (cm)	Body Mass on Land (kg)	Body Mass Index (BMI)	Apparent Body Mass in Water (Kg)	% BM Off in Water
1	F	25	151	61.8	27.1	19.4	68.7
2	F	54	158	55.2	22.1	21.6	60.9
3	F	22	164	69.7	25.9	28	59.9
4	F	27	158	58.2	23.3	24	58.8
5	F	29	167	57.4	20.6	24.3	57.7
6	F	23	167	61.3	22.0	27.5	55.1
7	F	27	160	59.7	23.3	20.9	65.0
Mean		29.6	160.7	60.5	23.5	23.7	60.9
SD		11.0	5.8	4.7	2.3	3.3	4.6

$$\% \text{ BM offloading} = [(BM_{land} - BM_{water})/BM_{land}]\times 100$$

- % body mass offloading (**% BM Off**) significantly and negatively correlated with **height** (**p=0.019, r=-0.835**)
- % BM offloading was not significantly correlated with BMI (p=0.089, r=0.685)

Figure 1. COP trajectories in AP and ML during APA and step initiation



Landmark 1 (L1) = the point where COP moves to the most posterior and lateral toward the initial swing limb (**heel-off of swing limb = release**)

Landmark 2 (L2) = when COP shifts from lateral to anterior during execution of first step (**toe-off of swing limb = unload**) (*Figure 2*)

Table 2. COP trajectories in anteroposterior (AP) and mediolateral (ML) directions during anticipatory and execution phases of the first step

COP Trajectories	S1 (APA)			S2			S3		
Dependent Variables	Land	Water	p value	Land	Water	p value	Land	Water	p value
Duration (s)	0.29 ± 0.06	0.36 ± 0.07	0.009	0.32 ± 0.07	0.41 ± 0.05	0.006	0.56 ± 0.07	0.83 ± 0.18	0.003
COP_{AP} trajectory (cm)	2.31 ± 0.84	4.19 ± 1.00	0.004	4.12 ± 0.87	5.63 ± 1.22	0.003	11.20 ± 1.24	14.03 ± 2.49	0.065
COP_{ML} trajectory (cm)	3.90 ± 1.33	6.09 ± 1.45	0.003	13.68 ± 1.76	17.32 ± 2.88	0.002	2.63 ± 0.86	4.93 ± 2.43	0.042
COP_{AP} MVELO (cm/s)	8.75 ± 4.32	12.65 ± 4.06	0.074	13.59 ± 5.04	13.82 ± 3.06	0.832	20.64 ± 4.88	17.47 ± 3.32	0.057
COP_{ML} MVELO (cm/s)	14.92 ± 6.93	18.46 ± 6.45	0.075	45.18 ± 11.86	44.86 ± 11.47	0.922	4.74 ± 1.52	5.72 ± 2.30	0.315
APA Peak in AP (cm)	6.87 ± 2.46	7.58 ± 3.74	0.330	---------------	---------------	---------	--------------	-------------	-------------
APA Peak in ML (cm)	3.76 ±1.30	6.44 ± 1.82	0.001	---------------	---------------	---------	--------------	-------------	-------------
Time to unload (s)	--------------	---------------	-------------	0.61 ± 0.11	0.78 ± 0.11	0.001	--------------	--------------	--------------

Figure 2. COP trajectories in AP x ML during step initiation

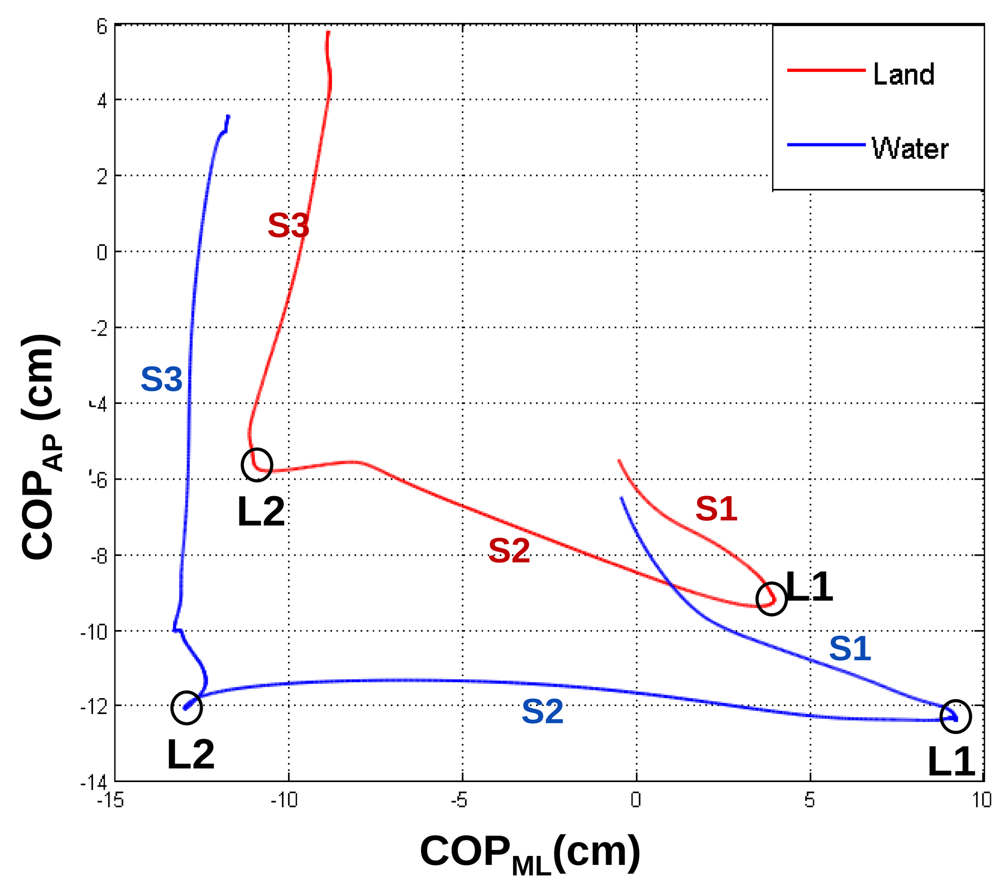


Table 3. Kinematic parameters during step initiation

Kinematic parameters	Land	Water	p value
Step length (m)	0.53 ± 0.08	0.50 ± 0.08	0.105
Step duration (s)	0.56 ± 0.06	0.90 ± 0.12	0.001
Step velocity (m/s)	0.95 ± 0.18	0.55 ± 0.09	0.001
Toe-clearance (cm)	6.83 ± 1.91	8.09 ± 3.81	0.194
Ankle ROM (deg)	9.87 ± 3.97	17.92 ± 7.14	0.032
Knee ROM (deg)	42.18 ± 8.16	40.11 ±13.32	0.434

Conclusion

The aquatic environment leads to a longer COP displacement, especially in ML direction, during APA and step execution. Water resistance appears to increase duration and decrease velocity of the first step. A longer time to unload the leading limb and to execute the first step along with facilitation of mediolateral transfer could potentially favor gait initiation training in the early stages of locomotor recovery. Further investigation is required.

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The effects of the aquatic environment on the control of upright stance: A pilot study comparing COP displacements in water and on dry land

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Introduction

Aquatic therapies are shown to benefit postural control and mobility functions in persons with neurological disorders[1,2]. Studies have shown that exercise in water significantly improved weight-shift and walking performance in comparison to therapies on dry land[1,3]. However, none of these studies has investigated how the aquatic environment influences postural control during individuals performance in water, as measured by biomechanical parameters. The present study is pioneering the investigation of center of pressure (COP) excursions during quiet standing in thermal water in comparison to dry land performance.

Objectives

To investigate the control of quiet standing performance by comparing COP excursions in water and on dry land in both eyes open and eyes closed conditions.

Methods

Subjects

Seven female able-bodied subjects from the Cruzeiro do Sul University volunteered for the study (*Table 1*).

Task

Participants were requested to stand on an AMTI waterproof force plate for 10 30-seconds trials, in 1-meter water depth (*Figure 1*) and on dry land. Participants were instructed to remain "stable" without moving their feet on the force plate, to focus on a fixed target about 2 meters distance and to maintain their upper limbs crossed over the chest. Vision condition was randomly assigned to all 10 trials, 5 with eyes closed (EC) and 5 with eyes opened (EO).

Outcomes and statistics

- Body sway area (AREA-SW)
- Mean sway amplitude (MSA_{AP} and MSA_{ML})
- Mean sway velocities ($MVELO_{AP}$ and $MVELO_{ML}$)
- Frequencies: predominant (fpred), 95% (f95) and 50% (f50)

* 25 seconds of quiet standing were analyzed. Multivariate analysis were used for comparisons.

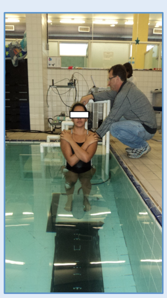

Figure 1. Experimental set-up (Cruzeiro do Sul University – Campus Sao Miguel, Sao Paulo, Brazil)

Results

Table 1. Subjects characteristics

Subjects	Gender	Age (years)	Height (m)	BM on Land (Kg)	BM Index (BMI)	BM in Water (Kg)	% BM Off Water
AB 1	F	25	151	61.8	27.1	19.4	68.7
AB 2	F	54	158	55.2	22.1	21.6	60.9
AB 3	F	22	164	69.7	25.9	28.0	59.9
AB 4	F	27	158	58.2	23.3	24.0	58.8
AB 5	F	29	167	57.4	20.6	24.3	57.7
AB 6	F	23	167	61.3	22.0	27.5	55.1
AB 7	F	27	160	59.7	23.3	20.9	65.0
Mean		*29.6*	*160.7*	*60.5*	*23.5*	*23.7*	*60.9 **
SD		*11.0*	*5.8*	*4.7*	*2.3*	*3.3*	*4.6*

* % BM (body mass) offloading significantly and negatively correlated with height (p=0.019, r=-0.835). % BM offloading was not significantly correlated with BMI (p=0.089, r=0.685).

COP amplitudes in anterior-posterior (AP) and mediolateral (ML) directions, in water and on dry land

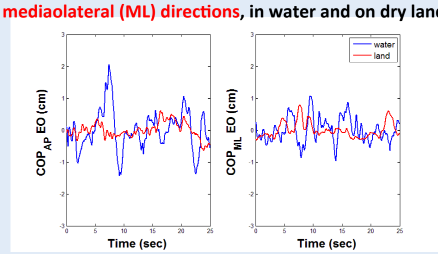

Figure 2. Representative sample of time series for 1 subject during 25 sec of quiet standing in water and on dry land with EO.

Mean Sway Amplitude (MSA_{AP} and MSA_{ML})

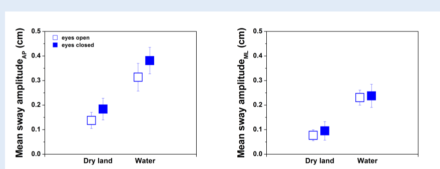

Figure 3. Mean (±SD) of MSA in the AP and ML directions (n=7).

MSA_{AP}
- p<0.0001 for differences between dry land and water
- p=0.001 for differences between EO and EC conditions

MSA_{ML}
- p<0.0001 for differences between dry land and water
- p=0.122 for differences between EO and EC conditions

Mean Sway Velocity ($MVELO_{AP}$ and $MVELO_{ML}$)

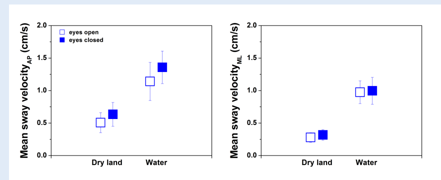

Figure 4. Mean (± SD) of MVELO in the AP and ML directions (n=7).

$MVELO_{AP}$
- p<0.0001 for differences between dry land and water
- p=0.006 for differences between EO and EC conditions

$MVELO_{ML}$
- p<0.0001 for differences between dry land and water
- p=0.441 for differences between EO and EC conditions

Body Sway Area (AREA-SW)

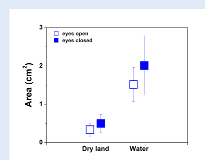

Figure 5. Mean (± SD) of AREA-SW (n=7).

AREA-SW
- p=0.001 for differences between dry land and water
- p=0.018 for differences between EO and EC conditions

Frequencies: fpred and f95 were significantly higher in water than on dry land (p<0.05). Frequencies were not affected by visual condition.

Discussion

•The percentage of BM offloading in water (55% to 69%) and the height (1.51 to 1.70 m) of participants were inversely correlated. The variance of % BM offloading may account for the higher variance of the measures in water compared to dry land.

•The increased body sway in water compared to on land is possibly due to the properties of the aquatic medium which challenges lower limb control used to maintain stable posture. Further investigation is needed.

Conclusion

• The aquatic environment provides increased challenges for quiet standing and could be valuable to stimulate upright stance control during functional recovery in neurorehabilitation.

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Novel Paradigm to Elicit Volitional Modulation of Cortical Neuronal Activity in a Rat Model

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Introduction

•Conventional brain-machine interface (BMI) paradigms translate neural activity into output commands through complex mathematical algorithms.
•Open-loop control is used to predict states of the peripheral motor system to move the actuator in real-time.
•Requires intensive training to achieve moderate accuracy.

Objective

•Test the feasibility of single neuron control in a BMI.
•Implement new paradigm that elicits volitional modulation of single neuron activity within single training sessions using operant conditioning and closed-loop control.

Methods

•Implanted a chronic microelectrode array in the motor cortex of a Long-Evans rat.
•Implemented spike sorting to obtain the firing rate from a single neuron in the array.
•Trained the rat in operant conditioning chamber with food/water dispensers to reward the rat upon target achievement.
•Provided visual feedback and auditory cuing (Fig. 1).

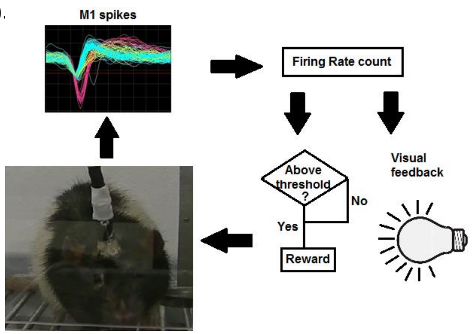

Fig. 1. Experimental setup

•**Behavioral task**: up-regulation of the firing rate for 1 second.
•We incremented the target threshold as the rat mastered the task.

Results

•The rat showed improvement in short periods.
  ○5 to 15 minutes of practice
•Fig. 2 shows three consecutive experiments where we incremented the target from 8 to 14 spikes/second.
•Performance initially dropped, but the rat was able to up-regulate the firing rate of the neuron (p<0.001) after 5 minutes of practice.
•Acquired as many targets as with 8 spikes/sec.

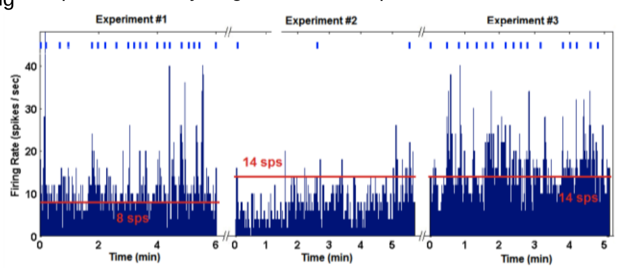

Fig. 2. Firing rate of the selected neuron during consecutive experiments. Tick marks denote target achievement.

•Initial vs. peak performance comparison (Fig. 3):
  ○All conditioned neurons
  ○All threshold increments
•Rat acquired more than twice as many targets per minute (p<0.05).
•Obtained rewards in half the time (p<0.001).

•Monitored the rat's gross activity with accelerometer and video recordings.
  ○Initially, the rat might have relied on gross motor behavior to up-regulate activity of the neuron.
  ○Eventually learnt to do it in absence of movement.

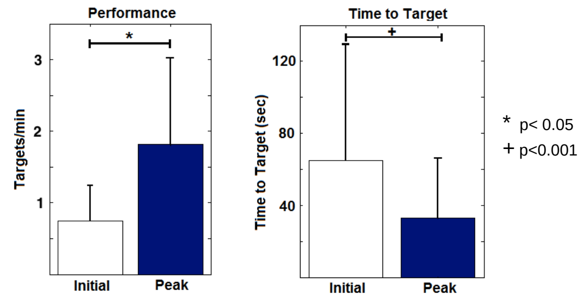

Fig. 3. Pair-wise comparisons of initial vs. peak performance and time to reach target.

Conclusions

•Rats can learn to readily control a one DOF BMI within minutes of practice even with increasing levels of difficulty.
•Rat was in full control of the food dispensing process and even anticipated obtaining the reward, as shown in Fig. 4.
•Our paradigm exploits the natural motor learning circuitry of the brain to control the BMI.
•Can be used to study neuroplasticity *in-vivo*.

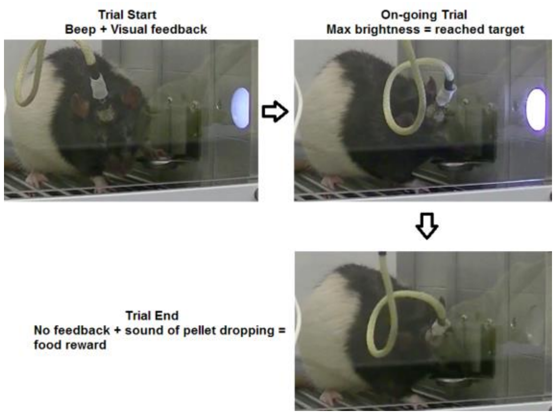

Fig. 4. Video frames of a representative trial.

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Modulation of emotion using FES of the "Duchenne marker"

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Introduction

- Pharmaceutical treatments for major depressive disorder (MDD) can have side-effects and are not always effective. New approaches for treating MDD are needed.
- Facial expressions are known to modulate mood, and there is evidence that manipulating expressions to activate certain facial muscles can enhance this effect.
- Spontaneous smiles due to positive emotions involve a specific pattern around the eyes characterized by a raising of the cheeks and the appearance of wrinkles next to the eyes (the "Duchenne marker"). Smiles without emotional involvement lack this marker. These two types of smiles involve different neural pathways. **There is a close neural link between the activity of facial muscles in the Duchenne marker and the brain regions involved in regulating emotion**.

We hypothesized that functional electrical stimulation (FES) of these facial muscles might modulate the activity of brain regions related to positive emotions.

Methods

•Two groups of 12 able-bodied subjects, each for a single 1-hour session.

Intervention (FES) group:

•25 2-minute blocks consisting of 4 30-second tasks:

1) Continuous voluntary smile including the Duchenne marker while receiving FES (no cognitive task)
2) Neutral expression while performing a cognitive task
3) Continuous voluntary smile while performing a cognitive task and receiving FES
4) Neutral expression while performing a cognitive task.

•FES consisted of 150 μs biphasic pulses delivered at 60 Hz, with amplitudes in the 3-9 mA range, on the orbicularis oculi and zygomatic major muscles bilaterally (1 and 2 respectively in Fig. 1).

Control (no FES) group:

•Same as above, but without receiving FES.

Assessments:

•The Positive and Negative Affect Schedule – Expanded Form (PANAS – X), a standardized assessment of mood, was administered before and after the experiment.

•Primary outcomes were the base item "happy" and the aggregate scores "positive affect" and "joviality", and all other PANAS-X items were considered secondary outcomes.

Deception:

•To avoid assessment bias, the subjects were given a false rationale for the study, such that they were not aware until the end that the true purpose was to investigate mood modulation.

Fig. 1 – Stimulation locations (bilateral)

Results

- No significant between-group differences were found in the change scores for our primary outcomes, the PANAS-X item "happy" and aggregate scores "Joviality" and "Positive Affect" (Fig. 2). Significant differences were, however, detected for secondary outcomes "determined" ($p = 0.03$), "daring" ($p = 0.04$), "scared" ($p = 0.03$) and "concentrating" ($p = 0.04$) (Fig. 3), all of which are relevant to MDD.

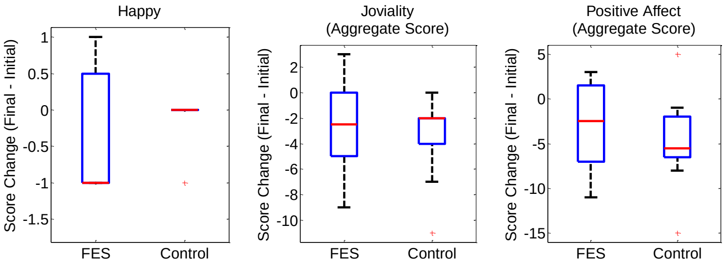

Fig. 2 – Differences between FES and control groups for primary outcomes

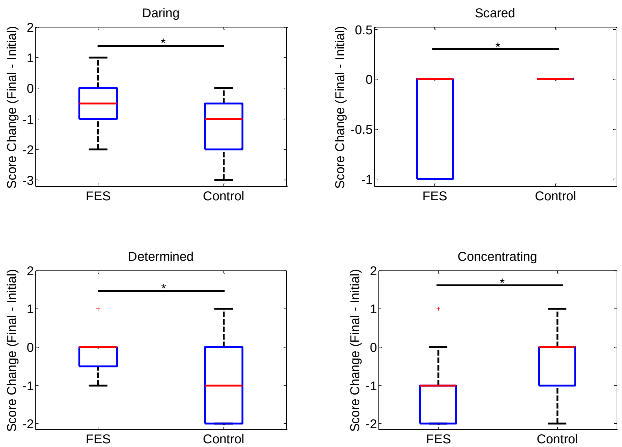

Fig. 3 – Differences between FES and control groups for outcomes showing significance

Conclusion

- Our results suggest that modulating emotion using FES may be possible, but is difficult to target accurately. Further work is warranted to explore FES applications to MDD.

References

- J. Zariffa J, S.L. Hitzig, and M.R. Popovic, "Neuromodulation of emotion using functional electrical stimulation applied to facial muscles", *Neuromodulation*, 2013, Epub ahead of print.

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Cardiovascular Response of Individuals with SCI to Functional Electrical Stimulation and Passive Stepping

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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 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

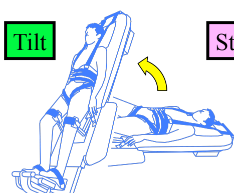
1) FES and passive stepping should independently induce venous return and mitigate a decrease in arterial pressure.
2) FES with passive stepping should be more effective than FES or passive stepping alone.

MATERIALS AND METHODS

Experimental Conditions:

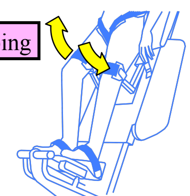
HUT = Tilt *
STEP = Tilt + Stepping
IFES = Tilt + FES
DFES = Tilt + Stepping + FES

Tilt



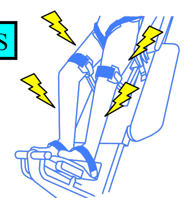
Sustained 70° head-up tilt

Stepping



Passive stepping 40 steps/min

FES



FES of tibialis, hamstring, quadriceps femoris, and triceps surae

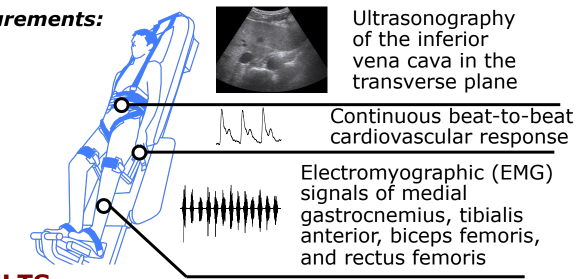
Protocol: The order of these conditions was randomized for each participant.

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

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

Measurements:



RESULTS

Cardiovascular parameter	HUT		STEP		IFES		DFES	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
HR [BPM]	18.9	3.57	12.2	1.73	2.88	2.59	8.40	1.54
SBP [mmHg]	-10.4	2.59	1.78	1.55	6.69	3.33	14.6	5.43
DBP [mmHg]	0.63	0.56	4.23	1.01	2.99	1.16	7.60	2.10
MBP [mmHg]	-4.78	0.92	1.29	1.06	2.51	1.63	8.64	3.15
SV [mL]	-24.1	4.91	-16.4	2.01	-6.77	3.09	-6.02	2.27
SVR [dyn*s/cm]	-5.36	57.4	21.0	22.8	29.0	55.2	61.3	46.2

CONCLUSIONS

- DFES generated the most favorable outcomes during severe orthostatic stress condition caused by a 70° head-up tilt:
  - SBP ↑, DBP ↑, MBP ↑, SV ↑, HR →
- Future studies to evaluate utility of the DFES in retraining sympathetic nervous system to cope with orthostatic stress

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