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Effects of Intramuscular Trunk Stimulation on Manual Wheelchair Propulsion Mechanics in Six Subjects with Spinal Cord Injury

Ronald J. Triolo, PhD^{1,2}, Stephanie Nogan Bailey, BSE¹, Lisa M. Lombardo, MPT¹, Michael E. Miller, MS¹, Kevin Foglyano, BSE¹, and Musa L. Audu, PhD²

¹Louis Stokes Cleveland Department of Veterans Affairs Medical Center (Cleveland, OH)

²Case Western Reserve University (Cleveland, OH)

Abstract

Objective—To quantify the effects of stabilizing the paralyzed trunk and pelvis with electrical stimulation on manual wheelchair propulsion.

Design—Single-subject design case series with subjects acting as their own concurrent controls.

Setting—Hospital-based clinical biomechanics laboratory.

Participants—Six (4M, 2F age 46±10.8yrs) long-time users (6.1±3.9yrs) of implanted neuroprostheses for lower extremity function with chronic (8.6±2.8yrs) mid-cervical or thoracic level injuries (C6-T10).

Interventions—Continuous low level stimulation to the hip (gluteus maximus, posterior adductor or hamstrings) and trunk extensor (lumbar erector spinae and/or quadratus lumborum) muscles with implanted intramuscular electrodes.

Main Outcome Measure(s)—Pushrim kinetics (peak resultant force, fraction effective force), kinematics (cadence, stroke length and maximum forward lean), and peak shoulder moment at preferred speed over 10m level surface; speed, pushrim kinetics and subjective ratings of effort for level 100m sprints and up a 30.5m ramp of approximately 5% grade.

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Corresponding author: Ronald J. Triolo, Ph.D., Louis Stokes Cleveland Department of Veterans Affairs Medical Center, 10701 East Boulevard 151W/APT, Cleveland, OH 44106, Phone: 216-791-3800 x4138, Fax: 216-707-6420, Ronald.triolo@case.edu.

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Dr. Triolo Clarifies the statements on the disclosure forms:

My mistake. I have no financial interest. I have disclosed inventions to the University and VA regarding this work. Nothing has been patented or licensed. There was no commercial support for the study.

4. Dr. Audu has no financial interests to disclose, either.

Please let me know if you need anything else.

RJT

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Results—Three out of five subjects demonstrated reduced peak resultant pushrim forces ($p = 0.014$) and improved efficiency, ($p = 0.048$) with stimulation during self-paced level propulsion. Peak sagittal shoulder moment remained unchanged in three subjects and increased in two others ($p < 0.001$). Maximal forward trunk lean also increased by 19-26% ($p < 0.001$) with stimulation in these three subjects. Stroke lengths were unchanged by stimulation in all subjects, and two showed extremely small (5%) but statistically significant increases in cadence ($p = 0.021$). Performance measures for sprints and inclines were generally unchanged with stimulation, however subjects consistently rated propulsion with stimulation to be easier for both surfaces.

Conclusions—Stabilizing the pelvis and trunk with low levels of continuous electrical stimulation to the lumbar trunk and hip extensors can positively impact the mechanics of manual wheelchair propulsion and reduce both perceived and physical measures of effort.

Keywords

Neural Prostheses; Spinal Cord Injuries; Posture; Electrical Stimulation; Torso

Many of the nearly 300,000 people in the US with spinal cord injuries (SCI) rely on manual wheelchairs [1,2]. Paralysis of the trunk and hip extensors after SCI can lead to spinal instability, increased risk of rotator cuff injuries and other chronic health problems with long term manual wheelchair use [3]. Belts, straps or custom seating adaptations that restrict functional movements are often the only alternatives for stabilizing the torso, preventing falls and maximizing potential for upper extremity use. For these reasons, individuals with SCI consistently identify trunk stability as one of their highest priorities for motor recovery [4].

Up to 68% of individuals with paraplegia exhibit degenerative changes of the shoulder, and individuals with tetraplegia are at greater risk due to the increased instability of their trunks [5]. More than 70% of all individuals with SCI experience shoulder pain [6] which appears to be related to duration of manual wheelchair use. Prevalence of pain and rotator cuff tears (63%) is higher after paraplegia than able-bodied controls (15%), again highlighting the importance of trunk stability [7].

Propulsion velocities and stroke distances decrease as SCI increases from low-thoracic to cervical levels, implicating increasing paralysis of the trunk musculature among other factors such as impaired autonomic control and variations in upper extremity strength and hand function [8,9]. Maximal speeds attained by individuals with C6 tetraplegia are slower than typical community demands, illustrating the importance of trunk stability to an independent lifestyle [9]. Exaggerated trunk flexion due to fatigue may be an effective compensatory strategy to create additional forces helpful to propulsion [10]. Similarly, forward trunk lean is beneficial during ramp ascent [11]. Moreover, trunk muscle activity increases with propulsion speed [12], and both trunk lean and shoulder muscle activity increase with slope during ramp ascent [13], highlighting the importance of providing a stable base of support for the arms.

Previous studies have explored the feasibility of stabilizing the trunk and hips with electrical stimulation to stabilize sitting posture [14,15]. Continuous stimulation of trunk and hip

extensors can positively alter pelvic tilt, expand bimanual workspace [16], facilitate forward reaching and active pulling, and restore upright sitting from forward flexed positions [17]. Activation of paralyzed trunk musculature can also improve skeletal alignment and deep breathing, and sufficiently enhance sitting stability to reduce reliance on chest straps [18]. Yet few studies have systematically investigated effects of stabilizing the torso with electrical stimulation on manual wheelchair propulsion mechanics. Yang *et al.* showed that continuous surface stimulation of the lumbar trunk extensors significantly increased gross mechanical efficiency (GME) of propulsion in experienced wheelchair users [19]. While GME improved 8.7-10.8% with stimulation, shoulder muscle activity, trunk ROM and peak flexion angles remained unchanged. Thus, stabilizing the trunk with stimulation may allow the shoulder muscles to be used more effectively.

The purpose of this investigation was to examine the benefits and limitations of low levels of continuous stimulation to the gluteal and lumbar paraspinal muscles via implanted intramuscular electrodes on manual wheelchair propulsion. This feasibility study was designed to isolate the relative effects of stiffening the hips and trunk with intramuscular stimulation compared to volitional effort alone on 1) pushrim kinematics/kinetics, shoulder moments, forward trunk lean and propulsion efficiency at preferred speeds over level surfaces, 2) time to completion (average speed), pushrim kinematics/kinetics and propulsion efficiency during level sprinting and ascending ramps, and 3) subjective perceptions of relative difficulty during level sprints and ramp ascents.

Methods

Subjects

Six manual wheelchair users with implanted neuroprostheses for exercise, standing or transfers participated. All subjects had received their implanted systems and original rehabilitation during participation in other studies. Subjects exhibited chronic (8.6 ± 2.8 yrs) motor complete injuries at mid-cervical to thoracic levels (C6-T10). Three subjects had preserved but impaired sensation (ASIA B). All were long-term users of implanted technology with mean follow-up of 6.1 ± 3.9 yrs post-implant at enrollment. Subjects were predominantly (67%) male and averaged 46 ± 10.8 years of age, 72.1 ± 13.5 kg in weight and 173.6 ± 7.6 cm in height (Table 1).

Subjects either received 8-channel implanted receiver-stimulators (IRS-8) [20] for standing and transfers [21-23] or 16-channel implanted stimulator-telemeters (IST-16) [24] for ongoing studies of standing balance [25-27] or new nerve cuff electrodes [28-30]. The implanted neuroprostheses were approved for experimental use by Investigational Device Exemptions from USFDA and all procedures for the investigation were approved separately from the original implantation studies by the Institutional Review Board of the LSCDVAMC.

All participants had intramuscular stimulating electrodes [31] inserted at the L1-L2 spinal nerves bilaterally to recruit the lumbar erector mass of the segmentally innervated paraspinal muscles [32,33]. All subjects had also received intramuscular or epimysial stimulating electrodes [34] to activate gluteus maximus bilaterally for hip extension [35]. Hamstrings

(semimembranosus) were utilized to control the pelvis in one volunteer (S5) with a non-functional gluteus electrode, and one subject (S3) had intramuscular electrodes at T12-L1 spinal nerves to activate quadratus lumborum for medial-lateral stability. Posterior portion of adductor magnus was stimulated in one subject (S6) to counteract thigh external rotation and assist hip extension.

Both 8- and 16-channel implants delivered constant current, charge balanced biphasic stimulation at nominal values of 20mA and 20Hz except when stimulus amplitudes were reduced to avoid activation of abdominal muscles. Pulsewidth was modulated independently (0-250 μ sec) on each channel. Threshold and saturation pulsewidths (defined as the minimum eliciting a just noticeable contraction and the maximum above which no increase in contractile force was observed, respectively) were determined and comfortable patterns of stimulation were constructed to extend the hips and trunk symmetrically. Pulsewidths on individual channels were customized to maintain sagittal and coronal alignment without compromising respiration or pressing the trunk firmly against the back rest.

Self-Selected Speed

Subjects sat in their own wheelchairs just outside the 10m measurement volume of a Vicon (Vicon Inc., Oxford UK) MX digital motion capture system. The wheelrim on the dominant side was replaced by a 24in SmartWheel instrumented pushrim with a solid tire (Outfront, Mesa AZ). The contralateral wheel was replaced by an uninstrumented wheel provided by the SmartWheel vendor of identical dimensions, tire and pushrim configuration. Retro-reflective markers (Figure 1) were taped to C7 and T6 spinous processes, and third metacarpal joint, radial/ulnar styloid midpoint, lateral epicondyle, and acromion process of the dominant arm, and a triad of orthogonal markers was strapped to the sternum [36]. Subjects started at least 2 strokes outside of the motion capture volume and propelled at comfortable speed across the level walkway. After exiting the 10m data capture volume, subjects continued to propel their wheelchairs for another 2-3 cycles before decelerating. After sufficient practice to ensure consistent speeds, 10 trials with and 10 without stimulation were collected in alternating order separated by one minute rest. Typically, 3-6 steady-state cycles, defined as strokes free from starting or stopping transients, were collected each trial. One representative cycle as close to the midpoint of the 10m data capture volume as possible per trial was selected for analysis.

Pushrim forces and moments were combined with Vicon kinematics as described by Cooper *et al.* to derive sagittal shoulder moments [36]. Mean stroke length, peak resultant pushrim force (after initial impact), peak shoulder moment and fraction of effective force (FEF) – an indicator of propulsion efficiency defined as the ratio of tangential to total pushrim force [37, 38] – were computed from the SmartWheel data. Cadence and peak forward trunk lean, defined as relative angle of the C7-T6 segment with vertical, were derived from the Vicon kinematic measurements. Average power and preferred propulsion speed [10,39,40] were computed and checked for consistency between conditions to isolate the effects of stimulation.

Sprints

A straight, 100m course through hospital corridors was defined and cleared of obstacles. Subjects sat in their own wheelchairs retrofitted with the SmartWheel as described above and completed the course as fast as possible. Trials with and without stimulation were performed in random order separated by 10min rest. A total of three trials of each condition were collected. The seven point ordinal Usability Rating Scale (URS), ranging from “very difficult” (-3) to “very easy” (+3), was applied after each trial to assess perceived effort [41,42]. Every other stroke centered about the midpoint of the sprint distance were identified offline from each trial, and assumed to represent steady state propulsion free of starting and stopping transients. At least 3 non-contiguous cycles from each trial for a total of 10 independent strokes per condition were included in subsequent statistical analyses. Differences in average speed, average power, peak resultant pushrim force, stroke length, cadence, FEF, and URS with and without stimulation were analyzed for significance.

Inclines

From a stationary start, subjects propelled their wheelchairs as fast as possible up a 30.5m ramp of 4.7% grade (2.7°). Trials with and without stimulation were performed in random order separated by 10min rest. Three trials under each condition were collected. The URS was applied after each trial to determine perceived effort. Spatiotemporal and kinetic variables from the SmartWheel were analyzed for at least three non-contiguous strokes centered about the ramp midpoint from each trial, which were considered to be free of starting and stopping transients and significant fatigue. A total of 10 independent propulsion cycles under each condition were included in the statistical analyses.

Statistical Analyses

Data were analyzed as a series of individual case studies with subjects acting as their own controls. Repeated measures of each variable were treated as the population for the outcome of interest for each subject [43-45]. Statistical significance of changes due to stimulation on 1) peak resultant pushrim force, 2) peak shoulder moment, 3) propulsion efficiency (FEF), 4) forward trunk lean, 5) stroke length, and 6) cadence were determined for self-paced level propulsion. In addition, the effects of stimulation on average speed (time to completion), average power and perceived effort/difficulty (URS) were determined during 100m sprint and 30.5m ramp ascent maneuvers. Statistical significance of changes due to stimulation to the kinematic and kinetic variables were determined by two-tailed Student t-test ($p < 0.05$) for repeated measures from each individual for each maneuver (self-paced 10m, 100m sprint and 30.5m ramp). Overall effects of stimulation on all kinematic and kinetic variables were assessed by MANOVA. Significant differences in ordinal URS values with and without stimulation for the group were determined by the Wilcoxon Signed Rank Sum test. Positive changes indicated that maneuvers were perceived to be easier with stimulation. The same number of trials with and without stimulation was included in the analyses for each outcome variable.

Results

Self-Selected Speed

The effects of stimulation for level propulsion at self-selected speeds for all subjects are summarized in Table 2. Kinetic data for S6 were unavailable during the 10m trials due to a malfunction of the SmartWheel. No clinically relevant changes in mean average velocity were observed between stimulated and non-stimulated trials for any subject. Four subjects exhibited no statistically significant changes ($p = 0.113$) with stimulation, while two varied by $<10\%$ between stimulation conditions. Average velocities were consistent between conditions for each subject and ranged from 0.6 ± 0.06 to 1.4 ± 0.05 m/s without stimulation and from 0.7 ± 0.09 to 1.3 ± 0.07 m/s with stimulation. Similarly, stimulation produced no changes to average power for any subject. Average power ranged from 7.5 ± 1.6 to 29.6 ± 3.9 W without stimulation and from 8.3 ± 1.8 to 29.9 ± 3.2 W with stimulation across all subjects. Because average power and velocity were consistent between conditions for each individual, changes in the other outcome variables were attributed to the application of stimulation.

Peak resultant force (Figure 2a) exerted during contact phase of the stroke cycle decreased significantly ($p = 0.014$) with stimulation in three of five subjects (S1, S2, S3). Relative reductions ranged from 13% to 26% in these individuals. Changes with stimulation were not different than zero for the other two subjects. Peak sagittal shoulder moment (Figure 2b) also remained unchanged for three of five subjects, but increased for S2 (71%) and S3 (17%) with stimulation ($p < 0.001$).

Stimulation caused no change in average stroke length in any subject (Figure 3a). Cadence (Figure 3b) increased (5%) in only 2 subjects (S3, S5) with stimulation, and was unchanged in all other subjects. Although statistically significant ($p = 0.021$), this change was not considered clinically relevant. FEF increased significantly ($p = 0.048$) in the same three subjects (S1, S2, S3) who showed significant declines in peak resultant pushrim force with stimulation (Figure 4a). Increases in propulsion efficiency with stimulation ranged from 10-22% over unstimulated cases in these individuals. Stimulation did not significantly alter FEF in the remaining subjects. Forward trunk lean increased significantly ($p < 0.001$) with stimulation (Figure 4b) by 19-27% for these same subjects. No effects of stimulation were observed in the other two subjects.

Sprints and Inclines

Trunk angle and peak shoulder moment were not computed during the sprint and ramp trials because they required kinematic data only available within the measurement volume of the Vicon motion capture system. Stimulation had no significant effects on mean cadence, stroke length, average velocity (time to completion) and peak resultant force in any of the six subjects during the 100m sprint. In one instance (S4) stimulation caused a decrease in FEF of 17% ($p = 0.034$) indicating reduced propulsion efficiency. No other significant changes were observed with stimulation for any subject. Most subjects (S1, S2, S3, S5) rated sprinting easier with stimulation (Figure 5a). Median changes in URS for these subjects ranged from +1 to +3. Stimulation produced no change in perceived effort in the remaining

subjects. Overall median change in URS was positive (+1) and statistically significant ($p<0.001$), indicating consistent improvement in perceived effort with stimulation.

Subject S5 was unable to ascend the 30.5m ramp either with or without stimulation. Continuous activation of hip and trunk muscles produced no changes in cadence, stroke length, average power, peak resultant force or FEF in any of the five subjects who completed the task. Only S1 took 7sec (17%) longer to ascend the ramp with stimulation ($p=0.026$). Stimulation had no effect on completion time (average speed) for the remaining subjects, with differences between conditions of approximately 1sec or 5% ($p=0.397$) in all cases. Four of the five subjects completing the task (S1,S2,S3,S6) perceived propelling up the 30.5m ramp to be easier with stimulation (Figure 5b). S4 demonstrated a small decrease of 1 point in median URS score, indicating slightly higher perceived effort with stimulation. Overall, stimulation resulted in modest positive median change in URS (+1), which was not statistically significant ($p=0.181$).

Overall Effects of Stimulation

The MANOVA applied to the outcomes (peak resultant pushrim force, peak shoulder moment, cadence, stroke length, FEF and trunk lean) of the level 10m self-paced trials across the five subjects with complete data sets indicates that stimulation had a significant effect on overall propulsion kinetics and kinematics ($p<0.001$, $F=7.679$). This result is consistent with the individual analyses performed with subjects acting as their own control as described above. There were no significant differences due to stimulation for either the 100m level sprint ($p=0.621$, $F=0.742$) and 30.5m ramp ($p=0.900$, $F=0.355$) maneuvers across all six subjects.

Discussion

Stimulation significantly decreased peak resultant wheelrim forces, and improved efficiency and the ability to lean forward in a majority (S1,S2,S3) of subjects during level self-paced propulsion. These parameters were unchanged by stimulation in the other subjects. The magnitude and consistency of these results suggest that hip and trunk stimulation may be beneficial for manual wheelchair users during the normal course of their daily lives. The average speed and power of level self-paced propulsion were consistent between conditions for all subjects, implying that these changes can be attributed to the effects of stimulation.

Peak sagittal shoulder moment remained unchanged with stimulation in three cases (S1,S4,S5). However, subjects exhibiting increased peak shoulder moment with stimulation (S2,S3) also demonstrated decreases in peak resultant pushrim forces, as well as significant increases in forward lean and propulsion efficiency. Improved efficiency implies a more horizontally directed wheelrim force, and increased forward lean could affect the moment arms for calculating shoulder reactions. Both factors may have contributed to the increased peak shoulder moments, particularly if improved forward lean increased the horizontal moment arm through which a more horizontally directed wheelrim force was acting.

Stimulation neither increased stroke length nor decreased cadence, which might have been interpreted as beneficial effects. This implies that the major action of continuous low-level

stimulation during self-paced propulsion may be on the kinetic rather than kinematic characteristics of the stroke cycle. Peak resultant pushrim force, stroke length, cadence and peak shoulder moment have all been linked to upper extremity pain or pathology [46-48]. Stimulation appears to decrease peak resultant pushrim force without changing stroke length or cadence, which might suggest a protective effect at the shoulder. The mixed shoulder moment results, however, prevent drawing any conclusions regarding the potential of stimulation to impact chronic rotator cuff injury.

Increases in forward trunk lean may be due to improved confidence that stimulation would prevent forward falls by resisting hip and trunk flexion. Stimulation did not appear to adversely restrict trunk ROM in any individual. The low levels of tonic stimulation applied in this study appear to be sufficient to stiffen the torso while preserving the ability to flex the trunk, which is important during ramp ascent or when fatigued [10].

FEF increased significantly in a majority of subjects during level propulsion at self-selected speeds. This suggests that stiffening the torso to stabilize the base of support for the upper extremities can improve ability to transfer propulsive forces to the wheelrim. The few subjects who did not increase FEF with stimulation showed no decrease in efficiency during level self-paced propulsion. Since FEF either increased or remained unchanged with stimulation, stiffening the hips and trunk was never observed to hamper propulsion efficiency.

The kinematic and kinetic outcome measures exhibited by subjects in this study during level, self-paced propulsion without stimulation were consistent with values of the same parameters in the literature. The average speed (1.11m/s), cadence (1.09strokes/s) and stroke length (97.2°) without stimulation are similar to those in Cowan *et al.* [49], which were reported to average 1.13m/s, 1.0strokes/s and 100.6°. Kotajarvi *et al.* [50], reported a faster average speed (1.48m/s) but similar cadence (0.926strokes/s). The mean average power across all subjects without stimulation in this study (18.7W) also fell within the range reported by other groups [30,40], and increased when ascending ramps, as would be expected from the literature [30]. The mean peak resultant wheelrim force exhibited by subjects in this study (65.5N) was close to the 70.9N reported by Boninger *et al.* [51], although somewhat less than that of Koontz *et al.* [52], which were both obtained at lower speeds (0.9m/s). Peak sagittal shoulder moments without stimulation averaged 29.3Nm and ranged from 20.4 to 36.9Nm, which are comparable to those reported by Veeger *et al.* [53] at similar speeds (21.6-49.2Nm, mean 35.6Nm), although somewhat higher than Robertson *et al.* [54]. Finally, the average FEF exhibited without stimulation by study participants (55.6) coincides with that observed in other reports under similar conditions [50].

The results of this study are consistent with previous reports utilizing surface stimulation [21]. Although surface systems provide a convenient method to investigate the effects of stabilizing the paralyzed trunk with stimulation in the laboratory, implanted systems provide repeatable stimulated responses and access to deep neuromuscular structures without the complicating issues related to external componentry and adverse sensation from cutaneous sensory receptors. The intramuscular electrodes inserted at the lumbar spinal nerves and deep motor points of the gluteal muscles utilized in this study yield stronger and more

consistent stimulated output than surface electrodes. Implanted systems are also personal, intimate and obviate complications related to donning and doffing surface electrodes, which is particularly challenging for posterior locations on the back and gluteal region, and minimize daily variability in stimulated responses due to changes in electrode location. Intramuscular stimulation was chosen for this study because of these advantages.

Core trunk stability is one of many factors influence propulsion mechanics and efficiency. For example, the relative position of the seat and axle can significantly impact pushrim forces [47,50,55]. Moving the axle position from an anterior to a more posterior position relative to the shoulder has been reported to change peak resultant pushrim force by an average of 2.17% [55]. In this study, the mean change in peak resultant pushrim force with stimulation was 9.57% over all subjects tested, with a maximum reduction of 26.2% (23.3N) in one subject (S2). Since subjects were tested in their own wheelchairs, it was assumed that their seating configurations had already been optimized and that the changes observed with stimulation were beyond those already achieved by adjusting seat and wheel positions.

The positive effects of stiffening the trunk and hips with stimulation evident at comfortable speeds on level surfaces disappeared during the more challenging tasks of ramp ascent and sprints. This was evident from the group MANOVA results, as well as those from each individual subject. This may be because the forward lean required during phases of ramp ascent and sprinting exceeds that allowed by continuous stimulation. In addition, propulsion speed (time to completion) was largely unaffected. Continuous stimulation at low, comfortable levels may be insufficient to meet the demands of these activities, which may require coordinating phasic excitation of the trunk and hip flexors and extensors with the stroke cycle. Skill level and wheelchair configuration may also have masked any potential effects of stimulation since they play increasingly important roles in activities such as sprints and inclines that require maximal effort. However, sprints and ramps were generally perceived as easier with stimulation, indicating that subjects found value and sensed some benefit from the core stability provided.

Limitations

The small sample size compromises rigorous exploration of inter-subject differences and interactions with injury level, preserved sensation or motor function and other variables such as time-post-injury. Sprints and ramps were difficult and fatiguing for some subjects, which limited the number of repetitions possible. Larger scale studies with more subjects and repetitions are needed to thoroughly understand the impact of trunk stimulation on wheelchair propulsion and the segment of the SCI population it would most benefit. There is potential for selection bias since inclusion was not randomized and only individuals who satisfied the rigorous selection criteria for other investigations and were experienced users of the implanted systems could participate. Additionally, the implanted systems were all optimized for functions other than wheelchair propulsion and installed by different surgical teams. Variations in stimulated responses from different electrode insertion techniques and the amount of practice with stimulation patterns designed specifically for wheelchair propulsion could influenced the results.

Finally, wheelchair configuration was assumed to be optimal for each individual. Future studies should review and adjust the seating system of each subject for maximal performance prior to testing with stimulation.

Conclusions

Low levels of continuous stimulation to hip and trunk extensors can positively impact manual wheelchair propulsion mechanics over level terrain at self-selected speeds. Stabilizing the spine and pelvis with stimulation can enhance ability to transmit propulsive forces to the wheelrim without increasing demands on the shoulder, thus improving propulsion efficiency. Low, continuous stimulation can also allow manual wheelchair users to lean farther forward during the stroke cycle by resisting trunk flexion. Stimulation is generally perceived to make propulsion easier, particularly during sprints. However, benefits of continuous low level stimulation disappear during level sprints and ramp ascents that either require more core stiffness, or modulation of stimulus timing and intensity. Further study is needed on larger samples of manual wheelchair users to determine effects of such factors as injury level, time-post-injury, and degree of retained sensation and voluntary strength. Research into new methods of modulating stimulation timing and intensity with the demands of the stroke cycle may further improve propulsion mechanics during challenging tasks of ramp ascent or changing speeds.

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Abbreviations

ASIA	American Spinal Injury Association
C5	5 th cervical vertebrae
C7	7 th cervical vertebrae
ECU	external control unit
Hz	hertz
IRS-8	eight-channel implanted receiver-stimulator
FEF	Fraction of Effective Force
GME	Gross Mechanical Efficiency
IST	implanted stimulator-telemeter
GME	Gross Mechanical Efficiency
IST	implanted stimulator-telemeter
kg	kilogram
L1	1st lumbar vertebrae
L2	2nd lumbar vertebrae

LSCDVAMC	Louis Stokes Cleveland Department of Veterans Affairs Medical Center
μsec	microsecond
mA	milliamp
N	newton
Nm	newton-meter
ROM	range of motion
SCI	spinal cord injury
T6	6 th thoracic vertebrae
T10	10 th thoracic vertebrae
URS	Usability Rating Scale
USFDA	United States Food and Drug Administration
W	watts

Figure 1.
Experimental set-up for assessing effects of stimulating the trunk and hip extensors on level propulsion at self-selected speeds with instrumented pushrim and reflective markers for digital motion capture.

Figure 2.

Average ($\pm$ SD) peak resultant wheelrim force (a) and peak sagittal shoulder moment (b) with and without stimulation to the hip and trunk extensors for each subject during level self-paced propulsion. (* = statistically significant at $p < 0.05$)

(a)

(b)

Figure 3.

Average ($\pm$ SD) stroke length (a) and cadence (b) with and without stimulation to the hip and trunk extensors for each subject during level self-paced propulsion (*=statistically significant at $p<0.05$).

Figure 4.

Average ($\pm$ SD) Fraction of Effective Force (a), and average maximal forward trunk lean (b) during level self-paced propulsion with and without stimulation for each subject (* = statistically significant at $p < 0.05$)

Figure 5.

Change in subjective perceptions of effort with activation of the hip and trunk muscles relative to propulsion without stimulation during (a) level sprints of 100m and (b) ascent of a 30.5m ramp. S5 could not complete the ramp maneuver either with or without stimulation. (* = statistically significant at $p < 0.05$)

Table 1

Summary of clinical characteristics of study participants.

Subject	Age ¹ (years)	Gender	Height (cm)	Weight (kg)	Injury Level	ASIA Class ²	Time Post Injury ¹ (years)	Time Post Implant ¹ (years)	Muscles Stimulated ³
S1	56	F	167.6	67.7	T5-6	B	6.6	2.3	ES, GM
S2	58	M	174.0	80.9	T6	A	11.8	10.6	ES, GM
S3	40	F	167.6	57.3	C7	B	13.0	11.4	ES, GM, QL
S4	50	M	185.4	97.7	T6	B	7.8	6.4	ES, GM
S5	26	M	165.1	63.2	C6	A	6.3	4.7	ES, GM, SM
S6	47	M	181.6	65.5	T10	A	5.8	0.9	ES, GM, AM

¹ At time of initial enrollment and testing

² A = motor and sensory complete; B = motor complete with sensory sparing; C = Motor and sensory incomplete

³ ES = Erector Spinae; GM = Gluteus Maximus; QL = Quadratus Lumborum; SM = Semimembranosus; AM = Posterior portion of Adductor Magnus

Individual statistical results for 10m self-paced level propulsion. Statistically significant differences ($p < 0.05$) between stimulation-assisted and baseline voluntary outcomes are shaded with arrows indicating direction of the changes caused by stimulation.

Table 2

Subject	Peak Force	Peak Shoulder Moment	Mean FEF	Average Forward Lean	Cadence	Stroke Length
S1	↓		↑	↑		
S2	↓	↑	↑	↑		
S3	↓	↑	↑	↑	↑	
S4						
S5					↑	
S6	N/A	N/A	N/A			N/A