Research Article

Impact of an implanted neuroprosthesis on community ambulation in incomplete SCI

Lisa M. Lombardo¹, Rudolf Kobetic¹, Gilles Pinault¹, Kevin M. Foglyano ¹, Stephanie N. Bailey¹, Stephen Selkirk^{1,2}, Ronald J. Triolo^{1,3}

¹Louis Stokes Cleveland Department of Veterans Affairs Medical Center, Cleveland, Ohio, USA, ²Department of Neurology, Case Western Reserve University, Cleveland, Ohio, USA, ³Department of Orthopaedics and Biomedical Engineering, Case Western Reserve University, Cleveland, Ohio, USA

Objective: Test the effect of a multi-joint control with implanted electrical stimulation on walking after spinal cord injury (SCI).

Design: Single subject research design with repeated measures.

Setting: Hospital-based biomechanics laboratory and user assessment of community use.

Participants: Female with C6 AIS C SCI 30 years post injury.

Interventions: Lower extremity muscle activation with an implanted pulse generator and gait training.

Outcome Measures: Walking speed, maximum distance, oxygen consumption, upper extremity (UE) forces, kinematics and self-assessment of technology.

Results: Short distance walking speed at one-year follow up with or without stimulation was not significantly different from baseline. However, average walking speed was significantly faster (0.22 m/s) with stimulation over longer distances than volitional walking (0.12 m/s). In addition, there was a 413% increase in walking distance from 95 m volitionally to 488 m with stimulation while oxygen consumption and maximum upper extremity forces decreased by 22 and 16%, respectively. Stimulation also produced significant ($P \leq 0.001$) improvements in peak hip and knee flexion, ankle angle at foot off and at mid-swing.

Conclusion: An implanted neuroprosthesis enabled a subject with incomplete SCI to walk longer distances with improved hip and knee flexion and ankle dorsiflexion resulting in decreased oxygen consumption and UE support. Further research is required to determine the robustness, generalizability and functional implications of implanted neuroprostheses for community ambulation after incomplete SCI.

Key Words: Electrical stimulation, Gait training, Neural prosthesis, Rehabilitation, Spinal cord injury, Walking

Introduction

Appropriately timed application of electrical stimulation (ES) to the motor nerves of the paralyzed or paretic muscles of individuals with incomplete spinal cord injuries (iSCI) has the potential to facilitate stepping and significantly enhance walking performance over voluntary function alone.¹ Improvements of one or more levels in functional ambulation category² have been reported with neuroprosthesis-assisted walking after iSCI, and incorporating ES into over ground gait training can improve volitional walking speed and distance over conventional rehabilitation techniques.³

When spasticity interferes with stepping, stimulated exercise can reduce extensor tone while increasing

voluntary muscle strength.^{4,5} In certain individuals with iSCI, an exaggerated extensor tone enables weight bearing but prevents step initiation. In these cases, one or two channels of peroneal nerve stimulation can inhibit lower extremity extensor tone and help initiate stepping⁶ with the flexion withdrawal reflex.⁷ In others, one leg can be completely paralyzed while the other is sufficiently strong to allow independent standing. This group requires unilateral stimulation of the knee extensors in addition to the hip and knee flexion and ankle dorsiflexion initiated with withdrawal reflex for effective stepping.⁸ Still others require bilateral knee extension along with peroneal nerve stimulation for stepping. Thus, the motor deficits and stimulated responses in iSCI indicate the necessity to customize the stimulation system to the particular needs of the individual.

Although prior studies strongly suggest that stimulation has the potential to significantly enhance

Correspondence to: Lisa M Lombardo, MPT- Louis Stokes Cleveland Department of Veterans Affairs Medical Center, 10701 East Boulevard, Cleveland, OH, 44106, USA. Email: llombardo@fescenter.org

Color versions of one or more of the figures in the article can be found online at www.tandfonline.com/yscm.

ambulation in individuals with iSCI,^{9–11} generating selective and strong muscular contractions, coordinating stimulus timing with residual function during the gait cycle, supporting the body weight and producing smooth and efficient movements in the presence of increased muscle tone and fatigue remain challenges to routine application of ES for ambulation after iSCI. It is difficult or impossible to selectively stimulate muscles deep to the skin surface (such as the hip flexors), responses are not always repeatable from day to day, and surface electrodes can irritate the skin.^{12, 13} As a result, application of surface stimulation have been done mostly for therapeutic muscle re-strengthening and motor re-education. Percutaneous intramuscular (IM) electrodes have addressed many shortcomings of surface stimulation by providing safe,¹⁴ selective, and repeatable muscle stimulation.^{15–22} However, their major drawbacks are the need for daily care of the percutaneous sites, risk of infection and the need to replace electrodes that break or move, making them applicable for short-term clinical use and assessment.^{23–25} For long-term clinical applications, implanted pulse generators (IPGs) based systems offer many advantages over surface and percutaneous stimulation including convenience, cosmesis, reliability, maintenance, and repeatability.^{26–32}

The purpose of this single-subject exploratory study was to quantify and assess the effect of multichannel implanted neuroprosthesis on walking in an individual with iSCI with limited ability to ambulate in the community. We hypothesize that activating key muscles that are inaccessible to surface stimulation with an implanted system will increase walking speeds and distances and require less effort than volitional walking alone, thus improving potential access to the community without the use of a wheelchair. Gait speed, distance, oxygen consumption and UE forces during walking as well as gait kinematics were measured to quantify the effects of neuroprosthesis-assisted walking over voluntary function, and usage data were collected to document system utilization and value.

Methods

The subject was a 52-year-old female who sustained a C6 incomplete SCI ASIA Impairment Scale (AIS) C with significant lower extremity weakness and moderate weakness of the trunk and upper extremities. Subject was 30 years post injury and able to ambulate short community distances (80 m) with bilateral forearm crutches. Her ambulatory distance was primarily limited by upper extremity fatigue secondary to compensating to clear her feet when walking. Her gait presented with significant

Figure 1 Components of the neuroprosthesis: (a) 8-channel implantable pulse generator; (b) intramuscular stimulating electrode; (d) external components: External Control Unit with attached communicating coil.

genu recurvatum (32° on left, 28° on the right) and difficulty in advancing the foot during swing marked by plantar flexion, adduction and internal rotation of bilateral ankles. Subject reported no knee or ankle pain during ambulation. After approval by the local Institutional Review Board, informed consent was obtained prior to implantation of a neuroprosthesis consisting of an 8-channel IPG and eight intramuscular stimulating electrodes.^{2,3,33–36} The IPG was powered and controlled by a wearable external control unit (ECU) (Fig. 1) that transmitted information to the implant via an inductive communication link maintained by a coil affixed to the skin.

The following electrodes were implanted bilaterally to target muscles to address her primary gait deficits: sartorius for hip and knee flexion, tensor fasciae latae for hip flexion and abduction, short head of biceps for knee flexion, and one electrode positioned near common peroneal nerve to activate tibialis anterior and peroneus longus for balanced ankle dorsiflexion. All electrodes were inserted near the nerve that innervate the target muscle. These muscles were chosen specifically for this subject based on her gait kinematics and manual muscle testing to determine which muscle groups would have the greatest impact on events in the gait cycle.

Implantation was performed in a 6-hour surgery under general anesthesia. After surgery the subject remained in the hospital for 4 days. The subject was advised to refrain from vigorous activities for six weeks to promote healing before stimulated exercise

was initiated. Stimulation patterns that modulate the pulse width (0–250 μ s) and frequency (20 or 25 Hz) with constant amplitude (20 mA) were developed for exercise and walking. The subject was prescribed exercises to be performed at home to improve strength and endurance of the stimulated muscles. Her exercise patterns were designed to initiate and improve swing phase of gait and consisted of 1) stimulation to the hamstring muscle for knee flexion, 2) stimulation to hip flexors and ankle dorsiflexors, and 3) stimulation to the tibialis anterior and peroneous longus for dorsiflexion. The ECU logged the dates, times and durations each pattern of stimulation was activated to track the subject's adherence to the home exercise program.

Activation patterns for stepping with stimulation were developed to address her individual gait deficits, and adjusted heuristically to improve gait performance. Initially, her gait training was with a free-running stimulation pattern, that was initiated and stopped by a hand switch and automatically cycled between left and right steps at a fixed cadence. Once comfortable walking with this pattern, she began training with a stimulation pattern where steps were initiated by movement of forearm crutches used for balance and support. The stimulation for each step was triggered by forward placement of the contralateral crutch (Fig. 2). Crutch strike was detected by a wireless accelerometer placed in baby shoes she attached to the crutch tips to provide her with extra stability (Fig. 3). An algorithm was developed to ensure that the crutch was actually moved forward in preparation to take a step and not just repositioned to maintain balance. Detection accuracy was 100% when tested on level ground in hospital hallways.³⁷

After 18 sessions of exercise and gait training with the system, data were collected both with and without stimulation to quantify the effects of the neuroprosthesis on gait speed, maximum walking distance, upper extremity forces on the crutches, gait kinematics and oxygen consumption. At baseline testing, 10 m walk tests and maximum walking distances were collected. At long term follow up (after 21 months of home use with the system), 10 m walk tests, maximum walking distances, oxygen consumption, gait kinematics and forces on the instrumented crutches were collected while walking with and without stimulation.

Data on 10 m walk tests were collected in the laboratory with ten minutes of rest between each trial to minimize the effect of fatigue. Two trials with and without stimulation were randomized using a coin toss. Each trial was timed with a stopwatch and the speed was calculated in m/s. The two trials were averaged.

One maximum walking distance test was performed for each condition with a minimum of one hour between each test condition to minimize the effect of fatigue. The subject was instructed to walk as far as possible and was permitted to slow down or stop to rest while standing if necessary. The subject used bilateral forearm crutches instrumented with accelerometers to trigger steps for all data collections.

Walking distances were measured with a calibrated measuring wheel, and timed with a stop watch. In addition, steps were counted with a hand tally counter and recorded every 2 minutes. Average step length and cadence were calculated using the first 12 minutes (volitional walking time) of walking for both conditions.

Gait kinematics were captured using a 16 camera Vicon MX motion capture system (Oxford, UK) while walking along a 10 m walkway. Data were collected both with and without stimulation on the same day to eliminate variability in marker placement. Fifteen markers were placed on the pelvis and lower extremities according to Vicon's Lower Extremity Plug-In Gait model³⁸ which was used for kinematic calculations. Four trials of each condition, volitional and stimulated gait, were collected with a five-minute rest break in between to reduce the effect of fatigue. Trials were randomized for conditions. Three steady state strides were selected from each trial to yield twelve strides per condition. Peak hip flexion, peak knee flexion, ankle angle at foot off, ankle dorsiflexion/plantar flexion angle at mid-swing, ankle adduction/abduction angle at mid-swing, ankle internal/external rotation angle at mid-swing were recorded/analyzed.

Upper extremity effort during gait was captured using six-axis load cells (AMTI, Inc., Watertown, MA, USA) that were integrated into the uprights of standard forearm crutches to measure the vertical component of the forces applied on the crutches. Data were collected during one 10 m walk in the lab. Average and maximum total UE support forces during volitional and neuroprosthesis-assisted walking were collected.

Metabolic energy consumption was measured by a K4b² (COSMED, Rome, Italy) portable gas exchange measurement system. The system provided breath by breath analysis of respiratory effort, pulmonary gas exchange, and also received heart rate information from a chest worn heart rate monitor (Polar Electro, Kempele, Finland). Baseline data were recorded in sitting for five minutes prior to walking and five minutes after walking. Ten points (in 15 second increments) during the last 2:15 minutes (when steady state was reached) of a 6-minute walk test with and without stimulation were selected for analysis and were tested

Figure 2 Diagram of the stimulation pattern used for walking with accelerometers. The following muscles are being stimulated: Tensor Fasciae Latae (TFL), Sartorius (Sart), Short Head of Biceps Femoris (Ham), and Tibialis Anterior (TA).

for normality in SPSS (IBM Corp., Armonk, NY, USA) with the Shapiro-Wilk test.

For parametric data including step length, cadence, gait kinematics, UE forces and oxygen consumption with and without stimulation, visual inspection of the Q-Q plots showed that the data were normally

distributed. Paired *t*-tests were performed in SPSS to determine significant differences between walking with and without stimulation.

At long-term follow up the subject completed the Quebec User Evaluation of Satisfaction with Assistive Technology (QUEST)³⁹ which uses a scale from 1 (not

Figure 3 Components of forearm crutches with accelerometers in the crutch tip used to trigger steps during walking.

satisfied at all) to 5 (very satisfied) to rate 8 questions about the device and 4 about the services provided. In addition, the subject was asked to list the 3 satisfaction items that were most important. Subjective report was obtained to determine how often the system was used for exercise and walking. This was also compared to a data log from the ECU to document usage patterns while at home and in the community.

Results

The speed during the 10 m walk tests remained consistent from baseline to follow-up with no difference with and without stimulation. At baseline, speed was 0.22 m/s without stimulation. At 21-month follow up, her walking speed with and without stimulation was 0.20 m/s and 0.19 m/s, respectively.

At baseline, maximum walking distance was 80 m and speed was 0.22 m/s without stimulation. At long term follow up, her maximum distance was 95 m at a speed of 0.12 m/s without stimulation and 488 m at a speed of 0.22 m/s when using stimulation. This demonstrates a 413% increase in distance and 77% increase in speed with stimulation after 21 months of community use. This increase in speed was the result of significant ($P \leq 0.003$) increase in cadence and step length when walking with stimulation. Cadence increased from 24 ± 3 to 30 ± 1 steps per minute with stimulation. Step length increased from 0.33 ± 0.02 to 0.47 ± 0.04 m.

Significant ($P \leq 0.001$) changes in gait kinematics including increased peak hip flexion, increased peak knee flexion during swing, decreased ankle angle at toe off and during mid-swing were detected when walking with stimulation vs. volitional walking (Table 1). The ankle was evaluated at foot-off and at mid-swing. At foot off with stimulation, the ankle went from an average of 33° to 13° of plantarflexion on the left and from 36° to 11° on the right. There was a significant ($P < 0.001$) reduction with stimulation in ankle adduction and internal rotation during mid-swing, moving the foot into a more neutral position. The average percent difference for these parameters ranged from 20% to almost 120% change between volitional and with stimulation (Fig. 4).

The oxygen consumption for walking with stimulation decreased by 1.87 ml of O_2 /min/kg ($P = 0.001$) or 22% when compared to walking without stimulation. The time for recovery was similar for both walking conditions.

There was no significant ($P \geq 0.124$) reduction in average upper extremity forces on crutches when walking with stimulation in comparison to walking

volitionally. However, a significant (left: $P = 0.022$, right: $P = 0.005$; total: $P = 0.002$) decrease of 16% was seen for maximum upper extremity forces during neuroprosthesis-assisted walking.

The results of the QUEST indicate a rating of “very satisfied” for both the assistive device and the services provided. “Durability”, “ease of use” and “comfort” were rated the most important satisfaction items. Self-reports indicate that the device was used more than four times per week for both exercise and walking. The ECU logged usage of the system on 93 out of the 108 days monitored (86%), averaging 5 walks per day of approximately 7 minutes each. On average she used her system for walking for 33.4 minutes per day. Data logs indicate she used the system almost twice (1.7 times) per day for exercise, for a total of approximately 12 minutes each day.

Discussion

When comparing volitional and neuroprosthesis-assisted maximum walking distance, there was a 413% increase in distance with stimulation from 95 m to 488 m at long-term follow-up. Similarly, walking speed increased by 77% from 0.12 m/s to 0.22 m/s with stimulation during the maximum walk. This increase in gait speed was the result of significant ($P \leq 0.003$) increases in cadence and step length. When walking with stimulation, the subject was able to maintain her speed over long distances, as no decrease in speed was seen for the maximum walk when compared to the 10 m walk (0.22 m/s and 0.20 m/s, respectively). In contrast, there was a noticeable decrease in gait speed from 0.19 m/s (10 m) to 0.12 m/s during longer distances when walking volitionally. She was unable to maintain her speed when walking longer distances volitionally due to a decrease in cadence and step length over time. Despite the relatively slow speed, this increase in maximal walking distance is significant enough in this individual to facilitate community access without relying on a wheelchair for mobility as self-reported.

Several kinematic changes were measured with stimulation that allowed the subject to ambulate with a more normal gait pattern. Her gait was most limited during the swing phase due to difficulty clearing the floor with her feet. Peak hip flexion with stimulation increased from 49° to 60° on the left and from 47° to 62° on the right to assist with foot-floor clearance. In addition, significant ($P \leq 0.001$) changes were seen at the knee and ankle, allowing improved clearance during swing (Fig. 5). At the knee there was an increase in peak knee flexion. The ankle was evaluated at foot-off and at mid-swing. At foot-off with stimulation, the ankle was in

Table 1 Means and standard deviations for gait kinematics with and without stimulation.

	VOL - L		STIM - L		VOL - R		STIM - R	
	Mean	StDev	Mean	StDev	Mean	StDev	Mean	StDev
Peak hip flexion	49.04	2.13	60.25	6.35	47.07	1.51	62.3	3
Peak knee flexion	46.6	1.89	69.17	6.95	50.22	2.59	61.23	5.29
Ankle angle at foot-off	-32.7	11.95	-13.44	8.51	-36.05	9.75	-10.96	7.02
DF/PF angle at mid-swing	-13.16	6.81	0.3	1.11	-17.41	6.9	3.09	1.3
Add/abd angle at mid-swing	-11.65	1.87	1.52	2.51	-12.92	1.65	1.29	1.21
Int/ext angle at mid-swing	39.38	4.69	-6.58	10.07	41.93	3.67	-6.31	4.94

13° of plantarflexion on the left and 11° on the right, which is comparable to normal ankle position (9° of plantarflexion) at foot-off for an individual walking at a slow cadence.⁴⁰ With stimulation her ankle was in a more favorable position at foot-off than during volitional walking when both of her ankles were significantly internally rotated and abducted. Additionally, during swing both feet were in a more neutral position with only slight external rotation and adduction with stimulation. These changes in gait kinematics with stimulation resulted in improved foot-to-floor clearance without foot dragging and contributed to an increase in cadence and step length which resulted in being able to walk longer and farther with stimulation while maintaining speed.

In this study the upper extremity forces were only measured during 10 m walks, which might not show as significant of a difference that could occur during longer walks. Her primary reason for stopping the volitional maximum walk test was due to upper extremity fatigue with notable dragging of her feet during

stepping. Due to the changes in gait kinematics with stimulation, the improved clearance may contribute to decreased forces on the upper extremities when walking longer distances. The subject reported a decrease in shoulder pain during maximum walk while walking with stimulation which allowed her to continue to walk longer. For future studies upper extremity forces should be measured during maximum walk tests.

Another important finding of this study is the decrease in oxygen consumption (22%) with neuroprosthesis-assisted walking compared to volitional function alone. This may have contributed to her ability to walk longer distances with stimulation and not feel fatigued. In addition to being able to walk farther, she was also able to walk for almost three times longer with stimulation (37:10 minutes) than volitionally (12:45 minutes). This may be the result of not only the reduction in oxygen consumption but also the decrease in maximum upper extremity forces and improved gait kinematics with stimulation.

Figure 4 Average difference between the mean joint angles for peak hip flexion, peak knee flexion, ankle at foot off (FO), and ankle angle (dorsiflexion (DF), plantar flexion (PF), adduction (Add), abduction (Abd), internal rotation (Int), and external rotation (Ext) at mid-swing (MS) with and without stimulation.

Figure 5 a) Vicon collection of gait kinematics without stimulation. Note internal rotation and dragging of the leg throughout the gait cycle during volitional walking. b) Vicon collection of gait kinematics with stimulation.

Gait speed is an important component of being able to ambulate in the community safely. Additional major muscles providing the power for forward progression such as hip extensors and ankle plantar flexors⁴¹ should be considered to increase walking speed with future implanted systems. In this study, gait speed did not increase significantly with stimulation, which is expected because only hip flexors were activated due to the limited number of available stimulus channels. With additional channels of stimulation, it may also be possible to address issues such as genu recurvatum and other gait deviations that can compromise walking efficiency and gait speed exhibited by this subject. In this study, the stimulated knee flexion generated by the short head of the biceps femoris was insufficient to prevent the significant genu recurvatum exhibited by this subject.

Because of the heterogeneous iSCI population, the results from this single subject are not generalizable to the general population. Instead, this study points to the need to customize stimulation systems to the needs of each individual with iSCI and that stimulation provided with a multichannel IPG has the potential to significantly improve walking in the community. Documented usage patterns indicate that the subject utilized the system almost daily for walking and therefore valued the resulting improvements it made in her gait.

Conclusion

Walking with an implanted neuroprosthesis allowed for significant increases in maximum walking distance in an individual with iSCI. Decreased oxygen consumption and decrease in maximum upper extremity forces may have contributed to this increase. Speed did not

change between volitional walking and walking with stimulation for shorter distances. However, volitional walking speed over long distances steadily decreased over time, while neuroprosthesis-assisted walking speed remained constant throughout the trial. Gait kinematics also improved when walking with stimulation and was characterized by increased peak hip and knee flexion, improved ankle angle at foot-off, and a more neutral position of the ankle during swing. The multichannel IPG was valued by the recipient as evidenced by almost daily usage for walking. Targeting the primary gait deficits of this individual with iSCI with an implanted neuroprosthesis shows great potential for facilitating community ambulation, improving physical access to wheelchair inaccessible places and deriving the physiological benefits associated with walking exercise.

Disclaimer statements

Contributors The authors would like to thank the subjects for their participation.

Funding This material was based on work supported by the Department of Veterans Affairs Rehabilitation Research and Development Service (Merit Review award B7692R).

Conflict of interest None.

Ethics approval Subject signed informed consents from Louis Stokes Cleveland Department of Veterans Affairs Medical Center's Institutional Review Board (12004-H02).

ORCID

Kevin M Foglyano <http://orcid.org/0000-0002-4781-1404>

References

- Bajd T, Kralj A. Functional Electrical Stimulation: Standing and Walking After Spinal Cord Injury. (1989) CRC Press, Inc, Boca Raton, Florida, USA.
- Hardin E, Kobetic R, Murray L, Corado-Ahmed M, Pinault G, Sakai J. Ambulation after incomplete spinal cord injury using an implanted FES system: A case report. *J Rehabil Res Dev* 2007; 44(3):333–46.
- Bailey SN, Hardin EC, Kobetic R, Boggs LM, Pinault G, Triolo RJ. Neurotherapeutic and neuroprosthetic effects of implanted electrical stimulation for ambulation after incomplete spinal cord injury. *J Rehabil Res Dev* 2010;47(1):7–16.
- Granat MH, Ferguson ACB, Andrews BJ, Delargy M. The role of functional electrical stimulation in the rehabilitation of patients with incomplete spinal cord injury- observed benefits during gait studies. *Paraplegia* 1993;31(4):207–15.
- Marsolais EB, Kobetic R, Jacob J. Comparison of FES treatment in the stroke and spinal cord injured patient. In Popovic D (ed): *Advances in External Control of Human Extremities X*. Belgrade, Yugoslav Committee for ETAN, 1990;219–24.
- Bajd T, Kralj A, Turk R, Benko H, Sega J. Use of functional electrical stimulation in the rehabilitation of patients with incomplete spinal cord injuries. *J Biomed Eng* 1989;11(2):96–102.
- Kralj A, Bajd T, Kvesic Z, Turk R. Electrical stimulation of incomplete paraplegic patients. *RESNA 4th Annual Conference*, Washington D.C. 1981; 226–8.
- Bajd T, Kralj A, Turk R, Benko H, Sega J. Assessment of functional characteristics relevant to FES application in incomplete SCI patients. In Popovic D (ed): *Advances in External Control of Human Extremities IX*. Belgrade, Yugoslav Committee for ETAN, 1987. Pages 129–40.
- Bajd T, Andrews BJ, Kralj A, Katakis J. Restoration of walking in patients with incomplete spinal cord injuries by use of surface electrical stimulation- preliminary results. *Prosthet Orthot Int* 1985;9(2):109–11.
- Ferguson ACB, Granat MH. Evaluation of functional electrical stimulation for an incomplete spinal cord injured patient. *Physiotherapy* 1992;78(4):253–6.
- Bajd T, Kralj A, Turk R, Benko H. FES rehabilitative approach in incomplete SCI patients. *RESNA 9th Annual Conference*, Minneapolis, MN, 1986;316–8.
- Stanic U, Acimovic-Janezic R, Gros N, Trnkoczy A, Bajd T, Kljajic M. Multichannel electrical stimulation for correction of hemiplegic gait. *Scand J Rehab Med* 1977;10(25):175–92.
- Petrofsky JS, Phillips CA. Closed-loop control of movement of skeletal muscle. *CRC Critical Reviews in Biomedical Engineering* 1985;13(1):35–75.
- Mortimer JT, Kaufman D, Roessmann U. Intramuscular electrical stimulation: tissue damage. *Ann of Biomed Eng* 1980; 8(3):235–44.
- Marsolais EB, Kobetic R. Functional walking in paralyzed patients by means of electrical stimulation. *Clin Orthop Rel Res* 1983;(175):30–6.
- Marsolais EB, Kobetic R. Implantation techniques and experience with percutaneous intramuscular electrodes in the lower extremities. *J Rehabil Res Dev* 1986;23(3):1–8.
- Marsolais EB, Kobetic R. Development of a practical electrical stimulation system for restoring gait in the paralyzed patient. *Clin Orthop Rel Res* 1988;(233):64–74.
- Nene AV, Jennings SJ. Hybrid paraplegic locomotion with the ParaWalker using intramuscular stimulation: A Single Subject Study. *Paraplegia* 1989;27(2):125–32.
- Stein RB, Prochazka A, Popovic D, Edamura M, Llewellyn MGA, Davis LA. Technology transfer and development for walking using functional electrical stimulation. *Advances in External Control of Human Extremities X*. Nauka, Belgrade, YU, 1990. Pages 161–75.
- Saito C, Ichie M, Honda T, Takahashi H, Kameyama J, Tanaka Y, et al. FES-controlled locomotion in the paraplegic patient. *Advances in External Control of Human Extremities X*. Nauka, Belgrade, YU 1990. Pages 91–7.
- Kagaya H, Shimida Y, Sato K, Sato M. Changes in muscle force following therapeutic electrical stimulation in patients with complete paraplegia. *Paraplegia* 1996;34(1):24–9.
- Stein RB, Belanger M, Wheeler G, Wieler M, P DB, Prochazka A, et al. Electrical systems for improving locomotion after incomplete spinal cord injury: an assessment. *Arch Phys Med Rehab* 1993;74(9):954–9.
- Kagaya H, Sharma M, Polando G, Marsolais EB. Reliability of closed double helix electrode for totally implantable FES system. *Clin Orthop Rel Res* 1998;(346):215–22.
- Scheiner A, Polando G, Marsolais EB. Design and clinical application of a double helix electrode for functional electrical stimulation. *IEEE Trans Biomed Eng* 1994;41(5):425–31.
- Daly JJ, Kollar K, Debogorski AA, Strasshofer B, Marsolasi EB, Scheiner A, et al. Performance of an intramuscular electrode during functional electrical stimulation for gait training post stroke. *J Rehabil Res Dev* 2001;38(5):513–26.
- Wilemon WK, Mooney V, McNeal DR, Reswick JB. Surgically implanted peripheral neuroelectric stimulation, Internal Report of Rancho Los Amigos Hospital, Downey, CA, 1970.
- Brindley GS, Polkey CE, Rushton DN. Electrical splinting of the knee in paraplegia. *Paraplegia* 1979;16(4):428–37.
- Holle J, Frey M, Gruber H, Kern H, Stohr H, Thoma H. Functional electrostimulation of paraplegics. Experimental investigations and first clinical experience with an implantable stimulation device. *Orthopaedics* 1984;7(7):1145–55.

- 29 Kilgore KL, Peckham PH, Keith MW, Thrope GB, Wuolle KS, Bryden AM, Hart RL. An implanted upper extremity neuroprosthesis: A five patient follow-up", *J Bone Joint Surg Am* **1997**;79(4):533–41.
- 30 Davis R, Eckhouse R, Patrick JF, Delehanty A. Computer-controlled 22-channel stimulator for limb movement. *Acta Neurochirurgica*, **1987**;39:117–20.
- 31 Davis R, Houdayer T, Andrews B, Emmons A, Patrick J. Paraplegia: prolonged closed-loop standing with implanted nucleus FES-22 stimulator and Andrews' foot-ankle orthosis. *Stereotact Funct Neurosurg* **1997**;69(1–4 Pt 2):281–7.
- 32 Rushton DN, Perkins TA, Donaldson N, Wood DE, Harper VJ, Tromans AM, *et al.* LARSI: How to obtain favorable muscle contractions. Proceedings of the Second Annual IFESS Conference (IFESS '97) and Neural Prosthesis: Motor Systems 5 (NP '97), Burnaby, British Columbia, Canada, 16–21 Aug., **1997**, 163–4.
- 33 Davis JA, Triolo RJ, Uhler JP, Bhadra N, Lissy DA, Nandurkar S, *et al.* Surgical technique for installing an eight channel neuroprosthesis for standing. *Clin Orthop* **2001**;385:237–52.
- 34 Memberg WD, Peckham PH, Keith MW. A surgically-implanted intramuscular electrode for an implantable neuromuscular stimulation system. *IEEE Trans Biomed Eng* **1994**;2(2):80–91.
- 35 Kobetic R, Marsolais EB. Synthesis of paraplegic gait with multi-channel functional neuromuscular stimulation. *IEEE Trans Rehabil Eng* **1994**;2(2):66–79.
- 36 Kobetic R, Triolo RJ, Marsolais EB. Muscle selection and walking performance of multichannel FES systems for ambulation in paraplegia. *IEEE Trans Rehabil Eng* **1997**;5(1):23–9.
- 37 KM Foglyano, JR Schnellenberger, R Kobetic, LM Lombardo, G Pinault, S Selkirk, NS Makowski, RJ. Triolo, Accelerometer-based step initiation control for gait assist neuroprosthesis, *J Rehabil Res Dev*, **2016**;56(6).
- 38 Vicon plugin gait product guide- Foundation notes. Revision 1.0. Los Angeles (CA): Vicon; **2008**. P. 11–2.
- 39 Demers L, Weiss-Lambrou R, Ska B. Item analysis of the Quebec User Evaluation of Satisfaction with Assistive Technology (QUEST). *Assist Technol*. **2000**;12(2):96–105.
- 40 Winters, D (1991). *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological*. Ontario, Canada: Waterloo Biomechanics.
- 41 Kepple TM, Siegel LK, Stanhope SJ. Relative contributions of the lower extremity joint moments to forward progression and support during gait. *Gait Post* **1997**;6(1):1–8.