Original Article

Electromechanical gait training with functional electrical stimulation: case studies in spinal cord injury

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Study design: Single case studies.

Objectives: To describe the technique of intensive locomotor training on an electromechanical gait trainer (GT) combined with functional electrical stimulation (FES).

Setting: Neurological Rehabilitation Clinic, Berlin, Germany.

Methods: Four spinal cord-injured (SCI) patients, one tetraparetic, two paraparetic, and one patient with an incomplete cauda syndrome, more than 3 months postinjury, who were unable to walk at all, or with two therapists. They received 25 min of locomotor training on the GT plus FES daily for 5 weeks in addition to the regular therapy.

Results: The patients tolerated the programme well, and therapists rated the programme less strenuous compared to manually assisted treadmill training. Gait ability improved in all four patients; three patients could walk independently on the floor with the help of technical aids, and one required the help of one therapist after therapy; gait speed and endurance more than doubled, and the gastrocnemius activity increased in the patients with a central paresis.

Conclusion: This combined technique allows intensive locomotor therapy in SCI subjects with reduced effort from the therapists. The patients' improved walking ability confirmed the potential of locomotor therapy in SCI subjects.

Spinal Cord advance online publication, 2 March 2004; doi:10.1038/sj.sc.3101595

Keywords: rehabilitation; spinal cord injury; functional electrical stimulation; gait trainer

Introduction

Following the first description of a viable clinical system in 1987,4 goal-oriented treadmill training with partial body weight support (BWS) has evolved as a promising treatment option in spinal cord-injured (SCI) subjects. Several open clinical studies showed successful gait restoration in para- and tetraparetic subjects.2–11 Even paraplegic patients who were unable to walk on the floor exhibited a locomotor-like EMG activity of their leg muscles on the treadmill.4,5

Treadmill training with BWS enabled SCI patients to practise numerous gait cycles on the motor-driven belt. The enforced locomotion presumably activated spinal gait pattern generators, as in laboratory animals.12,13 Loading and unloading in the correct rhythm and hip extension during terminal stance phase were the major peripheral drives.14–16

One of the disadvantages of treadmill training was the effort for up to three therapists in lifting the paretic limbs, promoting hip extension, and assisting lateral weight transfer; this limits its use in routine clinical practice. To reduce the physical effort required of therapists, Hesse and Uhlenbrock designed an electromechanical gait trainer. Subject’s feet are placed on two plates, whose movements simulated stance and swing phases,17 and the vertical and horizontal movements of the pelvis were controlled in a phase-dependent manner by ropes attached to the harness. Nonambulatory hemiparetic patients have shown improved walking ability and more normal patterns of muscle activation on the gait trainer, and the treatment requires less effort from the therapists compared to gait training on the treadmill.18,19

Colombo et al.20,21 have chosen another solution to the problem of therapists’ overexertion: their subjects wore a powered gait orthosis on the treadmill, which flexed the subjects’ hip and knee joints during swing phase. Lower limb muscle activity of tetraparetic subjects on the treadmill was comparable during the manually assisted and the automated training with the driven gait orthosis.20,21

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Disclosure statement: REHA-STIM company holds the patent for the electromechanical gait trainer used in this study. The company is owned and operated by Dr Beate Brandt-Hesse, spouse of the co-author, Dr Stefan Hesse
The present case studies describe the potential of the gait trainer therapy in SCI subjects who were non-ambulatory or ambulatory with technical aids and personal assistance. FES of the thigh muscles was additionally applied to stabilise the knees during the stance phase.

Methods

Gait trainer therapy

On the gait trainer, the harness-secured patients were positioned on two foot plates, whose movements simulated stance and swing phases with a ratio of 60-40% between the two phases (Figure 1). Cadence, stride length, and thus velocity could be set individually from 0 to 2.5 km/h. A servo-controlled motor assisted the gait movement at constant speed. The vertical and horizontal movements of the centre of mass (CoM) were controlled in a phase-dependent manner by ropes attached to the harness and connected to the gear system. The lateral movement over the stance limb helped to unload the swinging limb. Despite this lateral movement, patients tended to take advantage of the ongoing plate support, and put approximately 10–15% of their body weight onto the ‘swing’ limb.

Partial BWS was provided via an overhead pulley, and the actual amount was indicated on the display. The support was not kept constant but could vary, according to the controlled movement of the CoM. Two programmable dual-channel stimulators (biphasic constant current pulses, 20 Hz, 0.2 ms, 80 mA max, adjusted for a strong but tolerable muscle contraction) stimulated the quadriceps and biceps femoris muscles of both sides via 5 × 5 cm² self-adhesive surface electrodes during the stance phase. The rotation of the gear system, detected with the help of a light cell, controlled the stimulator. In case of insufficient hip extension, an adjustable rear bar acting on the pelvis pushed the patients into extension. BWS compensated for the weakness of the affected lower limbs, and was reduced as soon as possible to enable full limb loading. This was achieved when the patients could extend their hips and carry the weight sufficiently, that is, not swing in the harness. The training velocity was set individually, between 1 and 2 km/h. We aimed for each patient to take 800–1250 steps per session, which required a net treatment duration of approximately 20–25 min. Breaks were optional after each 5 min of therapy. Physical help, example, for additional control of the paretic knees, was administered according to individual needs.

The patients practised without any orthoses each workday for 5 weeks, that is, 25 treatment sessions. In addition, they participated in a conventional comprehensive in-patient rehabilitation programme. Minimal clinical criteria for inclusion into the programme were successful wheelchair mobilisation, no fixed contractures of the lower limb joints, no skin ulcers, and a stable cardiovascular system.

When possible, comprehensive gait analysis (Infotronic) was performed on the floor. This system consisted of overshoe-slippers, with the option to record the dynamic electromyogram of up to eight muscles. A portable data logger stored the data, and on being transferred to a PC, commercial software calculated the limb cycle parameters including the relative single stance duration (% cycle), the derived symmetry ratio (left divided by right if less than 1, or vice versa), the vertical forces (% body weight) including the displacement of the centre of pressure, and the rectified and averaged electromyogram normalised with respect to the gait cycle. The 10 m-time and 6 m walking distance were recorded to assess gait velocity and endurance.

Results

Subject #1

Male, 50 years old, with a spastic tetraparesis, 11 months after C5/C6 injury, ASIA D. His MRC scores (0–5) were 2 for hip flexion (L&R), 4 for knee extension (L&R), and 2 (L) and 3 (R) for ankle dorsiflexion, totalling a motricity index (MI 0–100) of 75 (L) and 82 (R). The modified Ashworth spasticity scores (0–5) were 1 (ankle dorsiflexion), 2 and 3 (left/right knee extension), and 4 (hip extension). Within the clinic, he used a
wheelchair, and during the physiotherapy he could walk with the assistance of two therapists helping with balance and placing the paretic limbs a distance of 34 m at a velocity of 0.32 m/s. He needed a walker and a left ankle-foot orthosis (AFO). His gait was highly symmetrical (ratio 0.98), and he loaded each lower limb with approximately 80% body weight (BW) during a shortened single stance phase of 26%. The EMG of the shank muscles showed very little, rather tonic activity of both the tibialis anterior and gastrocnemius muscles (Figure 2).

Initially, he put on the harness in the wheelchair positioned in front of the gait trainer, and was transferred with the help of a crane and two therapists into the machine. While seated on a foldable chair, the feet were attached on the adjustable plates with Velcro, and pairs of self-adhesive surface electrodes were put on the quadriceps and the hamstring muscles of both sides. Then, the patient stood up with the help of the pulley, and the training started at a velocity of 1.2 km/h, a step length of 45 cm, and 20% body weight support. He tolerated the electrical stimulation of the thigh muscles well, but due to fatigue, the amplitude had to be increased every 7-8 min to elicit a visible muscle twitch. One therapist sat in front of the patient to further stabilise both knees. The adjustable rear bar assisted hip extension. The net walking time was 20 min with a break every 5 min. At the end, he was transferred back to his wheelchair with the help of the crane. After seven sessions, he could stand in front of the machine holding to a bar, put on the harness, and step into the machine with the help of one therapist. With FES, he could then practise independently for 25 min with one break after 10 min. The step length (velocity) was increased to 49 cm (1.8 km/h) and BWS was reduced to 10%. After 5 weeks, he could walk within the clinic independently while using a walker and a left AFO. He managed to climb one stair but preferred the elevator. Outside the clinic, he continued to use a wheelchair. The maximum walking distance (gait speed) on the floor was 296 m (0.81 m/s). The motor power and the spasticity scores did not change to a relevant extent. Gait analysis revealed an ongoing highly symmetric gait, a longer relative single stance duration (34%), and the ability to carry his entire BW on each lower limb. The EMG revealed a pronounced, phasic, and timely correct activation pattern of the tibialis and gastrocnemius muscles bilaterally (Figure 2).

Figure 2  Rectified, averaged, and normalised EMG of the tibialis anterior and gastrocnemius muscles of both sides of a tetraparetic subject (#1), level C5/C6, ASIA D, before and after therapy. Note an increased amplitude and the more phasic pattern of the shank muscles after locomotor therapy.
Subject #2

A 45-year-old woman with a spastic paraparesis 8 months after a D7/8 SCI, ASIA D. Her muscle power, tested for ankle dorsiflexion, knee extension, and hip flexion ranged from 2 to 4, totalling an MI of 76 (left) and 80 (right). The spasticity was moderate with modified Ashworth (0–5) scores of either 2 or 3. She could walk with the help of two therapists a maximum distance of 14 m at a velocity of 0.12 m/s. Her gait was asymmetrical (swing ratio of 0.65), with a short relative single stance phase (16%, left, 25%, right), and each lower limb carried approximately 50% body weight (BW). EMG showed a timely correct activation of both tibialis anterior muscles, whereas the activity of the gastrocnemius, vastus medialis, and rectus femoris muscles was rather small and not well modulated (Figure 3). Right from the beginning, the patient could step into the machine with the help of one therapist, and she could practise independently with the FES after 10 sessions. Training velocities ranged from 1.5 to 2.0 km/h, and the BWS from 15 to 10%. After the programme, she could walk independently on the floor using two AFOs and a walker for 99 m at 0.26 m/s. Within the clinic, she continued to use a wheelchair. Her lower limb motor power and muscle tone did not change considerably. She walked more symmetrically, and almost carried 75% BW on each limb. The EMG (Figure 3) revealed the following improvements: an increased and

![Graphs showing EMG activity](image)

**Figure 3** Rectified, averaged, and normalised EMG of the tibialis anterior, gastrocnemius, rectus, and vastus femoris muscles of both sides of a paraparetic subject (42), level TH7/TH8, ASIA D, before and after therapy. Note the increased amplitude of the gastrocnemius muscles and the more physiological pattern of the thigh muscles after locomotor therapy.
timely correct activation of the gastrocnemius and vastus medialis muscles, and a monophasic activation of the rectus femoris muscles with an onset during midstance (right) and preswing (left). On the other hand, the left tibialis anterior pattern tended to be more tonic.

Subject #3
A 62-year-old male with a paraparesis 18 months after a D1/2 SCI, ASIA C. His body weight was 120 kg. He could not walk at all, although a KAFO had been prescribed 4 months before admission. His MRC scores were 1 (ankle dorsiflexion), 1 and 2 (left and right knee extension), and 2 and 3 (left and right hip flexion) totalling an MI of 42 (left) and 61 (right). The muscle tone was mildly increased with a modified Ashworth score of either 0, 1, or 2. Throughout the treatment, he was transferred with the crane into the machine and he required the help of two therapists sitting in front to assist knee extension, as FES only elicited a weak muscle twitch. The initial BWS was 35%, the step length 43 cm, and the velocity 1 km/h, for a total of 14 min. After 16 sessions, he could practise a net treatment duration of 25 min at a velocity of 1.5 km/h and a step length of 47 cm. Two therapists still had to assist knee extension, but with less effort. After 5 weeks of daily therapy, he could walk with the help of one therapist, a walker and bilateral KAFO a maximum distance of 17 m at 0.15 m/s. The isometric muscle strength and the muscle tone had not changed.

Subject #4
A 44-year-old man, with a flaccid paraparesis 3 months after an incomplete conus/cauda syndrome, level L1/L2, ASIA D. His left lower limb was weaker with an MRC score of 1 (ankle dorsiflexion, hip flexion) and 2 when tested for knee extension (MI 42), the corresponding MRC scores for the right lower limb were 4 (MI 79). Wearing an AFO on both sides, he could walk with the help of two therapists and a walker a maximum distance of 13 m at 0.16 m/s. The gait was highly asymmetric (see Figure 4), due to an extremely short left relative single stance duration (12 versus 31%), and he only carried 24% BW on his left and 52% on his right limb. The electromyogram of the shank muscles revealed small amplitudes particularly on the left side while the pattern was preserved.

During therapy, he required the ongoing help of one therapist to assist left knee extension, as he tolerated FES only on the right side. The BWS ranged from 10 to 15%, the velocity ranged from 1.2 to 1.8 km/h, and the step length was 46 cm. After 5 weeks, he could walk independently with AFOs and the walker a maximum distance of 112 m at 0.65 m/s. Within the clinic he continued to use a wheelchair. The muscle strength of his left dorsiflexor and knee extensor muscles increased considerably (MRC 4), while the strength of the left hip flexor remained unchanged (MRC 1). The strength of the right limb improved less. Gait analysis revealed a more symmetrical gait (ratio of 0.75 versus 0.39) due to a prolongation, particularly of the left relative single stance duration (Figure 4). The only change in the shank muscles EMG was increased amplitude of the left dorsiflexor muscle.

Figure 4 Gait line (ie the trajectories of the centre of pressure beneath both feet) of subject #4, incomplete cauda syndrome, ASIA D, before and after therapy.
All patients enjoyed the repetitive walking on the machine, as it reminded them of their walking pattern before their accident, and they could fully recommend it for other patients. The therapists appreciated the reduced effort on the machine as compared to the manually guided treadmill training or gait practise on the floor. They rated it as a meaningful adjunctive tool in gait rehabilitation; for further improvement, they suggested the individual adjustment of the foot trajectories and the possibility to de-load fully the swinging limb.

Discussion

Automated gait training in combination with FES followed the principles of locomotor therapy in SCI subjects. The repetitive practice of 800–1200 steps per session over a period of 5 weeks resulted in a functionally useful improvement of gait ability of all four subjects who did not walk at all (one subject, ASIA C) or had required the assistance of two therapists in addition to orthotic devices (ASIA D) during floor walking before therapy. After therapy, one subject remained wheelchair-independent in the clinic all day long, two subjects could walk with aids for 99 and 114 m, and the most severely gait-impaired subject (ASIA C) managed to walk 7 m with the help of one person after therapy. Gait analysis revealed a more consistent, symmetric, and dynamic gait pattern, with the patients carrying more load on their lower limbs.

The machine was designed to reduce the therapeutic effort during the locomotor therapy. Instead of placing the limbs and controlling the trunk movements on the treadmill, the therapists sitting in front of the patients assisted the knee extension in combination with the FES. Less and less assistance was required during the course of the treatment and the therapists rated the work less strenuous as compared to the manually assisted treadmill training. The Lokomat, by Colombo et al., also reduced the therapeutic effort in SCI subjects; probably the applied exoskeleton provided more external guidance as compared to the gait trainer, which does not mechanically support the knees. Instead, the authors have chosen FES of the thigh muscles to stabilise additionally the knees in the stance phase. FES, however, excludes patients with a complete peripheral paresis of the thigh muscles and may provide less support than a mechanical joint, particularly in the beginning of the therapy when the muscles are atrophic. On the other hand, FES is actively therapeutic, promoting muscle strength, lower limb circulation, and bone mineralisation.

The observed improvement of gait ability of the SCI subjects corresponds with the literature on treadmill training with BWS. For instance, Wernig et al. studied 25 chronic patients, 5.5 months after injury. They were nonambulatory at all or required the help of two persons on the floor. After 12 weeks of daily locomotor therapy on the treadmill, 20 out of 25 could walk with a walker; eight subjects could even climb stairs. Voluntary muscle strength tested at rest only slightly improved.

Field-Fote reported on 19 subjects at least 12 months after an ASIA C injury. After 36 sessions of locomotor therapy in combination with FES, four of the six subjects, who had required manual assistance in addition to their orthotic devices before therapy, could walk without external help. For the whole group of patients, the mean overground speed improved from 0.12 m/s to 0.21 m/s. The mean treadmill distance per session increased from 93 to 243 m.

Our four cases improved to a similar degree after a much shorter period of training. Less therapeutic effort and a lower perceived exertion on the gait trainer probably resulted in a higher therapy intensity with the continuous practice of 800–1200 steps/session right from the beginning of therapy. On the other hand, a lesion interval of less than 12 months in three of the four subjects was lower than in most treadmill studies, so that spontaneous recovery, particularly in subject #4, may also explain this difference.

With respect to gait kinematics, Field-Fote and Tepavic also reported a more consistent gait pattern after locomotor therapy, and the present study additionally showed a more dynamic and symmetric gait with the patients carrying more weight on their lower limbs.

The results of the dynamic electromyography of the centrally paretic SCI subjects are confirming the results of Dietz et al., on treadmill training with BWS: the amplitude of the gastrocnemius muscle increased in the centrally paretic subjects, and one patient showed a more physiologic pattern of the tibialis anterior and thigh muscles after therapy.

These case studies do not prove the efficacy of the device. Further, clinical studies will include a controlled trial and examining the feasibility of using the gait trainer with wheelchair-dependent ASIA B patients. Future technical developments will include programmable footplates and a force control to adjust the foot trajectories individually and fully unload the swinging limb.

Acknowledgements

The work was supported by a BioFuture grant (BE011) of the Bundesministerium für Bildung und Forschung, BMBF. We acknowledge the help of Stephen Kirker, Cambridge, in writing this paper.

Suppliers

a. Reha-Stim, Dr. Beate Brandl-Hesse, Berlin, Germany.
b. Bentronic GmbH, Munich, Germany.
c. InfoTronic, Tubbergen, Netherlands.

References


Adaptive robotic rehabilitation of locomotion: a clinical study in spinally injured individuals

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Study design: Clinical study on six spinal cord-injured subjects. The performance of two automatic gait-pattern adaptation algorithms for automated treadmill training rehabilitation of locomotion (called DJATA1 and DJATA2) was tested and compared in this study.

Objectives: To test the performance of the two algorithms and to evaluate the corresponding patient satisfaction. We also wanted to evaluate the motivation of the patients to train with a fixed gait pattern versus training where they can influence and change the gait pattern (gait-pattern adaptation).

Setting: Spinal Cord Injury Center ParaCare, Balgrist, Zürich, Switzerland.

Methods: The experimental data were collected during six blinded and randomized training trials (comprising three different conditions per algorithm) split into two training sessions per patient. During the experiments, we have recorded the time courses of the six parameters describing the adaptation. Additionally, a special patient questionnaire was developed that allowed us to collect data regarding the quality, perception, speed, and required effort of the adaptation, as well as patients' opinion that addressed their motivation. The achieved adaptation was evaluated based on the time course of adaptation parameters and based on the patient questionnaire. A statistical analysis was made in order to quantify the data and to compare the two algorithms.

Results: Significant adaptation of the gait pattern took place. The patients were in most cases able to change the gait pattern to a desired one and have always perceived the adaptation. No statistically significant differences were found between the performances of the two algorithms based on the evaluated data. However, DJATA2 achieved better adaptation scores. All patients preferred treadmill training with gait-pattern adaptation.

Conclusion: In the future, the patients would like to train with gait-pattern adaptation. Besides the subjective opinion indicating the choice of this training modality, gait-pattern adaptation also might lead to additional improvement of the rehabilitation of locomotion as it increases and promotes active training.

Sponsorship: The work was supported by The Swiss Commission for Technology and Innovation (Project No. 4005.1).

Keywords: treadmill training; gait-pattern adaptation; adaptive control; robotic orthosis

Introduction

This manuscript reports results from a clinical study on rehabilitation of locomotion in spinal cord-injured (SCI) patients. The aim of the study was the evaluation of two algorithms for adaptive robotic rehabilitation with an automated treadmill training system.

First approaches to rehabilitation of locomotion in SCI individuals were developed in the 1980s based on the results of experiments performed with spinalized cats. These were shown to be able to walk on a treadmill in spite of spinal transection if their body weight was supported. This research led to the development of a so-called treadmill training rehabilitation exercise. In the manual treadmill training, two physiotherapists move the legs of the patient suspended over a treadmill in order to simulate walking. During the training, the body...
weight of the patient is counteracted by a special unloading system, since the patients are not able to maintain their equilibrium and to walk by themselves. In a later stage of rehabilitation, some patients regain the ability to walk by themselves, but might still require body-weight support to do so.

During the last two decades, treadmill training has become an established rehabilitation exercise for individuals with locomotor dysfunction such as SCI and stroke patients. This is due to faster and greater mobility improvement demonstrated with a combined treadmill training and conventional physiotherapy versus conventional physiotherapy only. The mechanism for the improvement of locomotor function in SCI individuals is likely the following one: a periodic excitation of the cutaneous, muscular, and joint receptors provides a periodic afferent input to the neural circuits located in the spinal cord (central pattern generator, CPG). These circuits are responsible for coordinated muscle activation that generates locomotion. Treadmill training reactivates and retrains the CPG, and therefore improves the generated muscle activation pattern. The result is a faster and better relearning of locomotion. Since assisting the patient’s leg movement in manual treadmill training is a very strenuous task for the physiotherapists (which limits the training duration) and also results in an irregular and not completely physiological gait pattern, several developments were undertaken to automate the training and to increase its duration. One such automated treadmill training system is the Lokomat system, which was developed and is in regular use at our rehabilitation center. Figure 1 shows the manual treadmill training (panel a) and the automated treadmill training with the Lokomat (panel b). Since 1999, more than 10 patients have gone through the regular Lokomat training that consisted of several weekly sessions (each having a duration of 1 h) over a period of 2–4 months. Current Lokomat training, however, only provides treadmill training with a fixed gait pattern without the possibility for the patient to influence the way he/she is walking on the treadmill. To encourage the patients to walk more actively and to increase their motivation, algorithms have been developed that enable automatic gait-pattern adaptation during the Lokomat training. These algorithms are used in combination with sensing of the remaining voluntary activity of the patient to determine the change in the gait pattern that the patient would like to achieve. The gait pattern is adapted in an automatic way. The desired adaptation is determined as a solution of a nonlinear optimization (minimization) problem that is solved online. A thorough treatment of the theoretical aspects of the algorithms and some initial results are described in Jezernik et al.

The goal of this work was a clinical evaluation of the two gait-pattern adaptation algorithms called DJATA1 and DJATA2 (DJATA stands for direct joint-angle trajectory adaptation). The evaluation took into account technical as well as subjective aspects. The latter were assessed based on a specially developed patient questionnaire.

Methods

Gait-pattern adaptation principle

The conventional Lokomat training is realized with a position controller that tracks reference hip and knee joint-angle trajectories. If the patient tries to change this reference movement and acts against the movement with his remaining voluntary activity, the controller is going to counteract the patient (from the ‘viewpoint’ of the controller the extra forces generated by the patient represent unwanted disturbances that cause deviations from the reference motion). The idea of the gait-pattern adaptation is, on the contrary, to allow the patient to change the controlled motion. The underlying principle of gait-pattern adaptation algorithms is that the gait pattern is adapted in a way that minimizes (reduces) the interaction between the Lokomat and the patient, so that the Lokomat ‘yields’ to the voluntary exerted patient forces. The Lokomat motion thus synchronizes with the desired patient movement. DJATA1 minimizes the interaction based on estimated desired angular acceleration changes. DJATA2 minimizes the interaction based on a so-called impedance control and on estimated desired angular changes. The interaction was estimated from the actuator force measurement.

Since the Lokomat training is based on nominal (reference) hip and knee angle trajectories, the adaptation of the gait pattern is achieved by changing these angle trajectories. To ensure that the adapted gait-pattern trajectories always result in a physiological gait pattern, the nominal hip and knee angle trajectories \( q_{i, \text{Nom}} \) were parameterized each with three parameters: \( a_i \) (amplitude scaling), \( c_i \) (time stretch), and \( d_i \) (amplitude offset). \( i = 1 \) for hip and \( i = 2 \) for knee joint-angle trajectory. The parameters were used to calculate the
Experimenat study design

The clinical study comprised five incomplete paraplegic and one incomplete tetraplegic patient. The study was approved by the local ethics committee and the patients have given a written consent to participate in the study. We certify that all applicable institutional and governmental regulations concerning the ethical use of human volunteers were followed during the course of this research.

Each experiment was carried out at a fixed treadmill speed that has ranged from 1.7 to 1.9 km/h across experiments (approximate stride periods were 2–2.5 s) and with an unloading that has ranged from 50 to 90%. The age, lesion heights, training parameters, and the ASIA motor and sensory scores of the participating patients are listed in Table 1.

Three different conditions were tested in the experiments. These conditions corresponded to trials 1–3 for DJATA1 and to trials 4–6 for DJATA2. The trials were conducted in two separate blinded experimental series per patient in a semirandomized order, since one full experiment would be too long and too tiring. Each experimental series (three trials) lasted about 1 h.

In condition 1 (trials 1 and 4), the patients were asked first to follow the motion of the Lokomat for 5 min (they had to try to synchronize their own gait pattern with the gait pattern of the Lokomat). During the next 5 min, they were asked to try to change the gait pattern in different ways. The changes belonged to a well-defined adaptation set (more/less hip flexion/extension, more/less knee flexion/extension, larger/shorter steps, faster/slower motion).

In condition 2 (trials 2 and 5), the patients had to follow the motion of the Lokomat for 2 min, then they had 8 min time to try to walk with their own, preferred gait pattern.

In condition 3 (trials 3 and 6), the patients had initially to follow an unphysiological gait pattern for 2 min, then they had 8 min time to try to walk with their own, more physiological gait pattern. The unphysiological gait pattern had an extensively scaled nominal hip motion (increased hip flexion and extension, scaled by 1.2) and a reduced knee flexion (scaled by 0.7 and added offset of +4°). The idea behind condition 3 was to demonstrate that an unphysiological gait pattern can

Table 1 Summarized data of the SCI patients who took part in the study

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<tr>
<th>Pat. No.</th>
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<th>Height of the lesion</th>
<th>ASIA–motor score</th>
<th>ASIA–I. touch sensory score</th>
<th>ASIA–p. prick sensory score</th>
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NA, not applicable
be changed to a physiological one (by the use of a gait pattern adaptation algorithm).

The adaptation was switched on after 20 strides, which corresponded to a time of approximately 40 s. In case that for some reason the patient could not perform a trial for the full duration of 10 min, the trial duration was shortened by 1–4 min.

The data needed for later evaluation included trajectory parameters $a_i, c_i, d_i$ that corresponded to the gait-pattern adaptation. The force measurements were not collected and analyzed as the focus of the clinical study was on the achieved gait-pattern adaptation and not on the technical correctness of the gait-pattern adaptation algorithms. A quantitative assessment of interaction forces that has shown correct adaptation was performed earlier in computer simulations and in some pilot experiments with SCI and healthy subjects (where the extent of reduction of interaction forces was evaluated based on direct force measurement and based on calculation of special factors related to the functional that was minimized with the steepest descent method). These results were reported in Jezernik et al.1,13

**Patient questionnaire** Besides the analysis of the obtained parameter adaptation, we have also evaluated patient’s subjective opinion about the achieved adaptation. For this purpose, a special questionnaire was developed. After each trial, the quality of adaptation was assessed (judged by the patient and ranked with 0 = bad, 1 = good, and 2 = very good with resolution of 0.5), and the patients were asked if they could perceive a difference between the initial and the adapted gait pattern. They were also asked how fast the adaptation was, if perceived, and what was their general well-being. Further, they had to quantify how exhausting each trial was for them. The Borg scale was used for the last question (0 = no effort, 5 = medium effort, 10 = maximally tiring, resolution = 1). The questions are summarized in Table 2.

Apart from the questions relating to the actual gait-pattern adaptation, several questions were also asked to determine the patient’s motivation for treadmill training with gait-pattern adaptation versus training with fixed gait pattern. These included the following questions: (1) ‘how do you judge the overall quality of the training with gait-pattern adaptation’, (2) ‘do you think that training with gait-pattern adaptation leads to better rehabilitation’, and (3) ‘in the future would you like to train with the fixed or with the adaptive gait pattern’. The patients also had to give reasons for their answer to the third question.

**Analysis of the experimental data**

The data analysis consisted of several calculations. To obtain a representative (average) adapted gait-pattern trajectory for conditions 2 and 3, an average of the parameters $a$–$d$ was calculated over the last 30 strides in case the parameter values settled after the adaptation (Figure 3a), or over the last 60 strides in case the parameter values did not settle (Figure 3b).

To assess the variation of the parameter values around the nominal values, the following para-meter-variation root-mean-square (RMS) values were calculated two times: (a) after the adaptation was switched on but before the patient tried to change the gait-pattern, and (b) after the adaptation was switched on and after the patient tried to change the gait pattern:

$$PV_{\text{RMS}}(a_i) = \sqrt{\frac{1}{N} \sum_{i=1}^{N} (a_i \cdot 1)^2}$$

$$PV_{\text{RMS}}(c_i) = \sqrt{\frac{1}{N} \sum_{i=1}^{N} (c_i \cdot 1)^2}$$

$$PV_{\text{RMS}}(d_i) = \sqrt{\frac{1}{N} \sum_{i=1}^{N} (d_i \cdot 1)^2}$$

The number of strides $N$ ranged from 76 to 253 across the calculations. The duration of the initial walking with switched-on adaptation, but no effort of the patient to change the gait pattern was about 2 min (equals to approximately 60 strides).

Furthermore, we also calculated the RMS differences between the adapted and the initial joint-angle trajectories (JAT-RMS)

$$JAT_{\text{RMS}} = \sqrt{\frac{1}{N} \sum_{i=1}^{N} (\theta_{\text{TR}1.i} - \theta_{\text{TR}2.i})}$$

The indices $\text{TR}1$ and $\text{TR}2$ stand for angle trajectory 1 and angle trajectory 2, respectively. The JAT-RMS was

<table>
<thead>
<tr>
<th>Question</th>
<th>Answer</th>
<th>Ranking</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Quality of adaptation</td>
<td>Bad</td>
<td>0</td>
</tr>
<tr>
<td>Good</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td>Very good</td>
<td>2</td>
<td></td>
</tr>
<tr>
<td>No</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>2. Difference between the initial and adapted gait-pattern perceived</td>
<td>Yes</td>
<td>1</td>
</tr>
<tr>
<td>Slow</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>Medium-fast</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td>Fast</td>
<td>2</td>
<td></td>
</tr>
<tr>
<td>Very fast</td>
<td>3</td>
<td></td>
</tr>
<tr>
<td>Very bad</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>Bad</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td>So-so</td>
<td>2</td>
<td></td>
</tr>
<tr>
<td>Good</td>
<td>3</td>
<td></td>
</tr>
<tr>
<td>Very good</td>
<td>4</td>
<td></td>
</tr>
<tr>
<td>5. Physical effort (Borg scale)</td>
<td>Borg Scale</td>
<td>1–10</td>
</tr>
</tbody>
</table>
calculated over three strides with $N = 450$ for three $TR_1/TR_2$ combinations:

1. $TR_1 = \text{nominal trajectory}/TR_2 = \text{adapted nominal trajectory};$
2. $TR_1 = \text{nominal trajectory}/TR_2 = \text{adapted unphysiological trajectory};$
3. $TR_1 = \text{nominal trajectory}/TR_2 = \text{unphysiological trajectory}.$

The last part of the experimental data analysis consisted of ‘expert’ evaluation (by the authors of this paper) of the recorded parameter adaptation observed in different conditions. In condition 1, the parameter adaptation could be judged quite easily, since it was known what kind of gait-pattern change the patient would like to achieve. In condition 3, we could see if the parameter adaptation led to a more physiological gait pattern. Condition 2 was more difficult to judge, since the preferred gait pattern was unknown a priori. Especially in this condition, but also in the other two, we have also judged the stabilization of the parameter values versus drift. All trials were analyzed according to the above criteria and afterwards ranked on the following scale: $0 = \text{bad adaptation,} 1 = \text{good adaptation,} 2 = \text{very good adaptation (resolution = 0.5).}$

**Statistical analysis**

*Patient questionnaire* Statistical analysis of the ranked replies to questions from the patient questionnaire included calculation of the mean, median, standard deviation, and the nonparametric Wilcoxon test in order to test for significant differences between the experiments performed with the two algorithms DJATA1 and DJATA2.

*Experimental data* For the PV-RMS and JAT-RMS values, we have also calculated the mean, median, and the standard deviation. F- and t-test were performed afterwards in order to compare the parameter/joint-angle trajectory variation before versus after the actual adaptation.

The ranks assigned to the parameter adaptation in each trial (expert opinion) were analyzed in the same way as the data stemming from the patient questionnaire (Wilcoxon’s test). Table 2 summarizes different questions that were analyzed by ranking.

**Results**

*Experimental results* The SCI subjects were able to influence the gait pattern with their remaining voluntary activity. The extent of gait-pattern adaptation depended, of course, on the force that they were able to produce. Usually, they could achieve large adaptation for periods of 5–8 min. In most cases, the patients were able to adapt the nominal gait pattern to be alike the desired one (condition 1), and they were satisfied with the outcome of the adaptation.

Most of the patients, however, had problems with following and adapting the unphysiological initial gait pattern (condition 3). Their feet (or one foot) hit the treadmill during the swing phase so that they were tripping over. We had to unload them almost completely to prevent this tripping, which was impossible to avoid entirely. Due to a large amount of unloading, the patients had difficulties in developing large moments in order to change the gait pattern. Therefore, in many cases the gait pattern remained unphysiological or sometimes even adapted to an even more unphysiological one. There were a few cases where the gait pattern became slightly more physiological.

Next, we will present a couple of typical experimental results based on figures showing the observed adaptation of the parameters. The six graphs in Figure 4 show

![Figure 4](image-url)
the time courses of the hip \((a_1, c_1, d_1)\) and knee \((a_2, c_2, d_2)\) adaptation parameters plotted against the stride number for condition 2, trial 5 (DJATA2). The time when the patient tried to adapt the nominal gait pattern to a preferred one is indicated by horizontal bars. A notable change in the parameters \(a_2, c_2, d_1,\) and \(d_2\) can be observed. To get a better picture of how the changes in the parameters influenced the gait pattern in terms of joint angles, the two graphs in Figure 5 show the average hip (left) and knee (right) joint angles for two cases: (1) mean trajectory between strides 50 and 150 (plotted with dashed line) and (2) mean trajectory between strides 150 and 250 (plotted with dotted line). The nominal trajectories are plotted with solid line for comparison. Two different mean trajectories are shown as the parameter \(d_1\) and partially also \(a_2\) show a different behavior during these two periods. Negative \(d_1\) during the first period shifted the nominal hip trajectory downwards, whereas positive \(d_1\) during the second period shifted the nominal hip trajectory upwards (compare dashed and dotted lines). We can conclude that the desired gait pattern in this case was not the same during these two periods. The adaptation algorithm has tracked the desired gait pattern in a dynamic way, that is, the gait pattern was continuously adapted according to the dynamic interaction between the patient and the Lokomat, and this interaction was not constant but has changed with time.

Figures 6 and 7 show two examples of parameter adaptation during condition 1. Figure 6 represents trial 1 (DJATA1), where the patient first tried to walk faster (indicated by solid bars), then tried to increase the hip flexion (dashed bars), and finally tried to walk with shorter steps (double bars). During the first period, the parameters did not change much. During the second period, hip and knee flexion increased due to increased \(a_1, d_1,\) and \(c_2, d_2.\) During the third period, the hip and knee flexion decreased due to decreased \(a_1, d_1,\) and \(c_2, d_2\) has increased, but the change in \(a_2\) had a greater effect on

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*Figure 6* Experimental data consisting of six adaptation parameters for experimental condition 1 and DJATA1 algorithm. See text for more details

the decrease of the knee flexion. Figure 7 represents trial 4 (DJATA2), where the first three periods (solid bars, dashed bars, and double bars) are the same as in Figure 6, and where during the fourth period (double dashed bars) the patient tried to increase hip extension. During the first period, \(d_1\) and \(d_1\) have increased. The changes in parameters during the second and third period were, as expected, opposite and had the desired effect on the gait pattern. During the fourth period, \(d_1\) has additionally decreased and the other parameters remained quite the same. Decreased \(d_1\) means increased hip extension (as desired).

The shown examples demonstrate that the adaptation took place, and that in most cases it corresponded to a change in the gait pattern that the patient tried to achieve.
Statistical results

The statistical results concerning the patient questionnaire are shown in Table 3 and Figure 8 (left columns are plotted for data from DJATA1 and right columns for data from DJATA2). No statistically significant differences were found between the two algorithms. The average values for the quality of adaptation were 1.08 and 1.15 for DJATA1 and DJATA2, respectively, which indicates good adaptation. Most patients were also able to perceive a difference between the nominal and the adapted gait pattern (mean = 0.8 for both algorithms). They stated that the adaptation occurred medium–fast to fast (mean = 1.08 and 1.58), which meant that the gait pattern adapted to a new one after four to six strides (8–12 s). In all cases, the well-being of the patients was between good (3) and very good (4). The mean effort of trials was 3.66 and 3.82 for DJATA1 and DJATA2 respectively (middle effort).

Table 4 shows the mean values of the parameter variation RMS (PV-RMS) calculated for all the six parameters and for both the algorithms, DJATA1 and DJATA2, before (left column) and after (right column) the patient tried to change the gait pattern. Statistically significant changes in PV-RMS are indicated by asterisks. The average values were almost always greater in the right than in the left column, and were in DJATA2 significantly greater in five out of six cases, which demonstrates that significant adaptation took place.

Table 5 shows the means and standard deviations of the differences between the nominal and adapted joint-angle trajectories (in degrees). The mean differences were higher for DJATA2 (2.09 and 2.71° for hip and knee angle trajectories, respectively) than for DJATA1 (0.69 and 1.20°). The differences between the nominal and unphysiological angle trajectories (2.67/1.68 and 6.74/8.59° for hip and knee angle trajectories,}

| Table 3 | Means ± SDs calculated from the ranks from the patient questionnaire (Table 2) across all DJATA1 and DJATA2 trials and across all SCI subjects |
|-----------------|-----------------|-----------------|-----------------|-----------------|
| Quality of adaptation | Difference init./adapt. gait pattern | Speed of adapt. | General well-being | Borg scale |
| DJATA1 | 1.08 ± 0.59 | 0.80 ± 0.41 | 1.08 ± 0.99 | 3.50 ± 0.51 | 3.66 ± 2.47 |
| DJATA2 | 1.15 ± 0.65 | 0.80 ± 0.41 | 1.38 ± 0.90 | 3.52 ± 0.51 | 3.82 ± 2.02 |

Figure 8 Bar graph showing the means (column bars) and standard deviations (lines) of ranked answers to the patient questionnaire (Table 2). The calculations were made across all experiments and separately for the two algorithms DJATA1 and DJATA2 to enable comparison.

| Table 4 | Mean values of the PV-RMS before the patient tried to change the gait pattern (first and third columns), and during gait pattern adaptation (second and fourth columns) |
|-----------------|-----------------|-----------------|-----------------|-----------------|
| DJATA 1: mean PV-RMS | DJATA 2: mean PV-RMS |
| a1 | 0.006 | 0.016* | 0.005 | 0.021** |
| c1 | 0.001 | 0.004* | 0.003 | 0.016** |
| d1 | 0.503 | 0.702 | 1.050 | 1.772* |
| a2 | 0.030 | 0.028 | 0.003 | 0.016** |
| c2 | 0.001 | 0.006** | 0.003 | 0.020** |
| d2 | 0.288 | 0.348 | 0.683 | 0.902 |

A paired t-test was made to analyze if the parameter variation significantly increased after the patients tried to adapt the gait pattern. The t-test was made for each parameter separately. Statistical significance is indicated by **(P < 0.05) or by * (P < 0.1)
Table 5  For conditions 2 and 3, also the RMS differences between the nominal and adapted joint-angle trajectories were calculated.

<table>
<thead>
<tr>
<th></th>
<th>Nom-adapt DJATA1</th>
<th>Nom-adapt DJATA2</th>
<th>Nom-adapt unphys. DJATA1</th>
<th>Nom-adapt unphys. DJATA2</th>
<th>Nom-unphys. DJATA1</th>
<th>Nom-unphys. DJATA2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td>0.69±0.35</td>
<td>2.09±1.51</td>
<td>1.68±0.53</td>
<td>3.00±0.82</td>
<td>2.67±3.63</td>
<td>1.68±0.58</td>
</tr>
<tr>
<td>Knee</td>
<td>1.20±1.02</td>
<td>2.71±2.81</td>
<td>6.57±2.40</td>
<td>8.41±1.42</td>
<td>6.74±2.76</td>
<td>8.59±1.93</td>
</tr>
</tbody>
</table>

Means±SDs for hip and knee trajectories and for condition 2 (first two columns), condition 3 (third and fourth column), as well as initial RMS differences between the unphysiological and nominal trajectories (last two columns).

Figure 9  Bar graph showing the means (column bars) and standard deviations (lines) of ranks given for the quality of parameter adaptation by experts (separate calculations for DJATA1 and DJATA2 algorithms)

respectively) did not decrease much or effectively decrease after the adaptation (1.68/3.00 and 6.57/8.41°) for DJATA1/DJATA2 algorithms.

The last statistical analysis dealt with the evaluation of the expert opinion of the quality of adaptation (based on parameter adaptation). The mean results are shown in Figure 9 together with standard deviations (mean±SD=0.78±0.52 and 1.00±0.50 for DJATA1 and DJATA2, respectively). The difference in means was not statistically significant. The means were close or equal to the ranking good (good = 1).

Patient motivation
All patients have noted (perceived) the gait-pattern adaptation and have, on average, judged it as good (see above). Furthermore, they all thought that the training modality with gait-pattern adaptation will lead to better rehabilitation outcome, and they would all prefer to train with the new training modality in the future. The given argumentation for the training with gait-pattern adaptation included the following reasons: (a) they can train with their own gait pattern; (b) the training with gait-pattern adaptation is more comfortable; (c) it is possible for them to train more actively (with more active effort); and (d) they are able to vary the gait pattern and therefore walk in a more differentiated way.

Discussion and conclusions
The aim of the clinical study was to evaluate the performance of the two algorithms for automatic gait-pattern adaptation for automated treadmill training. The emphasis was not only placed on the analysis of the performance, but maybe even more importantly, on the patient’s opinion and motivation. The study has shown the feasibility of gait-pattern adaptation, demonstrated good algorithm performance, and revealed that in the future the patients prefer to train with the gait-pattern adaptation (versus conventional training with fixed gait pattern).

It is important to remember that the subjects who want to adapt their gait pattern necessitate sufficient residual motor capacity to do so. If they do not possess this capacity, they will not be able to change the gait pattern. Therefore, the training with gait-pattern adaptation is meant only for incomplete SCI patients with sufficient remaining motor capacity. Since the algorithms have been successfully tested in extensive computer simulations and also with the healthy subjects, it is clear that if an SCI patient fails to generate the desired gait-pattern adaptation, then the cause for this is his insufficient motor capacity and not the algorithm failure. It is true that the disability limits the ability of the patient to generate enough force to change the gait pattern, but the patient still knows well in which way he/she wants to change the gait pattern.

The algorithm DJATA2 performed slightly better than DJATA1, but the differences derived from the performed analysis were not statistically significant. However, the adaptation of the gait pattern was more significant in the case of DJATA2 than DJATA1 (based on PV-RMS analysis), but this might have been a consequence of a higher chosen sensitivity in the case of DJATA2 algorithm. It was not possible to exactly equalize the two sensitivities as they depend on different factors and algorithm properties. In the opinion of the authors and according to additional testing and simulation results, DJATA2 should be preferred and is therefore recommended to be used in the future.

The problems that the patients experienced when trying to change the initial unphysiological gait pattern

Spinal Cord
Figure 10 Graphical user interface (GUI) for the adaptive gait-pattern rehabilitation as used by physiotherapists. The interfaces in the top right allow the physiotherapists to start and stop the training, to change the treadmill speed, and to change the extent of gait-pattern adaptation. The adaptation parameters are shown in four bottom-left windows.

(condition 3) can partially be explained by the fact that the unphysiological gait pattern prevented the patients to exert sufficient forces onto the Lokomat in order to adapt the gait pattern. However, treadmill training with unphysiological gait pattern anyway does not make much sense, and during an actual training the patients only have to adapt the initial nominal gait pattern.

Currently, the adaptation algorithms assume symmetry between the left and right legs. The present software/hardware estimates the interaction only on one body side/one leg (there exist, however, a software switch that allows us to use either the force measurements from the left or from the right leg). In the future, the adaptation algorithms will possibly be adapted to allow independent adaptation of the left and right leg trajectories (except for parameters c, which are related). However, unsymmetrical gaits are energetically nonoptimal and undesired because of other reasons (balance, comfort, smoothness of the movement, internal torques/forces). Regarding this issue, we would also like to mention that we have developed another adaptive training concept for treadmill training of stroke patients, which takes into account the existing asymmetry between the kinematics of the healthy and impaired legs. We believe that the asymmetry is more important in the case of stroke than in the case of SCI patients.

On the basis of the presented results, it can further be concluded that the treadmill training with gait-pattern adaptation increases the motivation of the patient and gives him/her the feeling that they are controlling the machine versus that the machine is controlling them. Gait-pattern adaptation also increases and promotes active training, which might lead to better rehabilitation outcome. The latter hypothesis will need to be tested in further clinical studies that will also include EMG measurements and follow-up of the rehabilitation outcome for different experimental groups (conventional versus adaptive treadmill training).

With a simple but powerful graphical user interface (Figure 10) developed for the physiotherapist (that also allows the gait-pattern adaptation to be switched off, so that a training with a fixed gait pattern can be done with the same software and that as well allows adjustment of the adaptation sensitivity), the gait-pattern adaptation will most likely substitute the current treadmill training modality in the future.

References


