Modulation of leg muscle activity and gait kinematics by walking speed and bodyweight unloading

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Abstract

During rehabilitation, many patient groups are being trained using bodyweight-supported treadmill training. However, little is known about modulation of time and distance parameters, joint movements and leg muscle EMG patterns by very low walking speeds or partial bodyweight unloading. We collected data from 20 healthy young volunteers who walked on a treadmill at walking speeds varying between 0.5 and 5.0 km h⁻¹ (0.14–1.39 m s⁻¹) and with 0%, 25%, 50% and 75% bodyweight unloading. We found that cadence and stride length were largely influenced by walking speed, while bodyweight unloading influenced these measures only at 75%. However, the relative duration of the gait phases changed largely only at walking speeds less than 2.5 km h⁻¹, but was influenced by all different bodyweight unloading conditions. Joint trajectories of knee and ankle joint, as well as leg muscle EMG activity patterns changed largely at walking speeds slower than 2.5 km h⁻¹ and bodyweight unloading conditions less than 50%. Therefore, standards for EMG activity and joint angle trajectories should only be compared when the training is done with velocities higher than 2.5 km h⁻¹ and less than 50% body weight unloading.

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1. Introduction

In several patient groups, functional gait is successfully being improved using (automated) bodyweight-supported treadmill training (BWSTT). More recent studies show that incomplete spinal cord injured (SCI) subjects [1], subjects with brain injuries [2,3] and Parkinson’s disease [4,5] can improve walking ability with (automated) BWSTT. Initially, BWSTT developed from earlier animal experiments in which, for example, cats with a complete spinal cord injury were successfully trained to walk using this training method (for review see [6,7]). This was due to a set of neurons within the spinal cord, which could create a locomotor pattern named the central pattern generator (CPG). Although the existence of the CPG in humans is based on indirect evidence [7], BWSTT could contribute to the recovery of locomotor function in patients for additional reasons. BWSTT can be considered as a functional task, with a low risk of falling and with several options to adjust the intensity of the training to the ability of the patient, such as walking speed, external assistance or bodyweight-support. Especially at onset of BWSTT, the walking velocity is slow due to the impairments of the patient. However, even during and after training, many patients do not reach normal walking speed, which could be defined at about 4.7 km h⁻¹ or 1.31 m s⁻¹ [8–12]. In our spinal cord injury center, most SCI patients train at and achieve walking speeds between 1.0 and 3.0 km h⁻¹. To evaluate whether the training is successful, qualitative and quantitative clinical tests are applied, as well as changes in gait parameters, joint kinematics and muscle EMG activity. When such parameters become more comparable to normal values, an improvement in locomotor function can be assumed to have taken place. However, there is no normative data for gait parameters, joint kinematics and muscle EMG activity that has been measured in regular intervals from normal to very low walking speeds.
In addition, the influence of different amounts of bodyweight unloading on these parameters, especially at lower walking speeds, is unclear. Previous studies showed that mainly extensor muscles become influenced by body unloading [13,14]. However, in these studies only a limited number of muscles were investigated and changes in neither joint kinematics nor gait parameters were described. In another study, kinematics were studied, but only at a walking speed of 1.25 m s\(^{-1}\) [15].

Therefore, in the present study, we investigated changes in gait parameters, joint kinematics and EMG patterns of several leg muscles during (1) very slow to normal walking speeds and (2) various amounts of bodyweight unloading. The goals were to provide insight into the changes in these parameters due to differences in walking speed and bodyweight unloading and to provide a normative data set of the parameters, which might serve as a reference for future research and rehabilitation related studies. Finally, the clinical significance of the results will be discussed.

2. Methods

2.1. Subjects

Twenty subjects were selected from a convenience sample (mainly students). The 10 males and 10 females had an average age of 23.8 ± 3.4 (S.D.) years (range: 19–32), were 1.74 ± 0.07 m tall (range: 1.60–1.80) and their weight accounted to 67.9 ± 9.5 kg (range: 50–86). The subjects had no known cardio-vascular, neurological or orthopedic diagnoses and were naïve to the purpose of the study. The study was approved by the local ethic committee and the subjects were informed and gave written consent.

2.2. Measurement protocol

One goal was to determine differences in EMG activity and joint kinematics due to different walking speeds. Therefore, we measured the subjects walking at speeds starting at 0.5 km h\(^{-1}\) (=0.14 m s\(^{-1}\)). After the subjects became familiarized with walking at the selected speed, we recorded 25 gait cycles. We increased the speed with steps of 0.5 km h\(^{-1}\) and repeated the recording protocol until the final measurement was performed at a speed of 5.0 km h\(^{-1}\) (=1.39 m s\(^{-1}\)).

Another goal was to investigate differences in EMG activity and joint and gait kinematics due to body unloading. The subjects walked on the treadmill at three different speeds (1.5, 2.0 and 2.5 km h\(^{-1}\) or 0.42, 0.56 and 0.69 m s\(^{-1}\)) wearing a harness that did not interfere with the normal gait pattern, when no bodyweight unloading was applied (Fig. 1). The harness was connected to counter weights, which were used to partially unload the bodyweight. We investigated the influence of bodyweight unloading at 0%, 25%, 50% and 75% of the total bodyweight, starting with the 0% bodyweight unloading condition. After familiarization to the bodyweight unloading condition, 25 gait cycles were measured at 1.5 km h\(^{-1}\), after which we repeated the same procedure at the two higher walking speeds. This whole procedure was repeated for 25%, 50% and 75% body unloading conditions. There was no large break in-between the several measurements.

2.3. Recording of the measurements

The vertical forces, which were measured by one-dimensional force sensors that were located under the split-belt treadmill (Fig. 1), were used to determine the different phases of the gait cycle. EMG and joint movement data were all collected from the right leg. We recorded the EMG signals of eight leg muscles using surface electrodes. The skin was shaved and peeled to reduce skin resistance. Two electrodes (diameter 8 mm) were placed 2.5 cm from each other in parallel to the muscle fibers. The recorded muscles were: gluteus maximus (GL), rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), lateral hamstrings (HL), medial hamstrings (HM), tibialis anterior (TA) and the gastrocnemius medialis (GM). The EMG signals were amplified, band-pass filtered (30–300 Hz) and transferred together with the biomechanical signals to a PC using SOLEASY software (ALEA Solutions GmbH, Zurich, Switzerland) via a 12 bit AD converter. Furthermore, we measured the joint movement trajectories using electrogoniometers (Biometrics Ltd., Gwent, UK), which were fixated at the lateral side of the hip, knee and ankle joint. All data were sampled at 1000 Hz.
2.4. Data analysis

The EMG signals were rectified and filtered using a median-filter (25 ms). We used this filter for its good ability to remove signal artifacts and to prevent temporal shifting of the EMG signal. Using the forces on the treadmill, initial foot contact of the left and right foot was determined and the data series were cut into separate gait cycles. The start was defined at initial foot contact of the right foot. We normalized each of the 25 gait cycles to 1000 points and averaged these curves for each muscle and joint. In addition, the EMG signals were normalized for each subject and each condition by setting the difference between the lowest and highest EMG amplitude at 100% and normalizing the curve according to this value. Due to this, we could analyze the course of the EMG patterns well, but not the EMG amplitude. Previous studies showed that EMG amplitude increases by increased loading [13] or walking speed [16]. Although, usually, EMG amplitude normalization is done for one condition (for example peak EMG amplitude at maximum walking speed), we normalized for each condition to prevent that fine changes in EMG patterns became masked by the large influence of for example walking speed on EMG amplitude (see [16]). Finally, we calculated the averaged curve for each muscle and joint using the data of all 20 subjects. Furthermore, we calculated the cadence; stride length and the changes in the percentages of swing phase duration (SW), first and second double stance duration (DSD1 and DSD2, respectively) and the single stance duration (SS). We normalized the stride length by stature [17].

We calculated Pearson’s correlation coefficient (r) to quantify the linear relationship between the cadences and normalized stride lengths for the increase in walking speed. In addition, r was calculated for the joint trajectories and EMG patterns between the 5.0 km h⁻¹ condition versus the other speeds and between the 0% bodyweight unloading conditions versus the other unloading conditions at 2.0 km h⁻¹. For the EMG patterns, the correlation coefficient depended not on the normalization method, since r is scale-independent.

Finally, we analyzed the parameters for statistical differences using analysis of variance (ANOVA) for repeated measures with subsequent Bonferroni’s correction for pair-wise comparisons. To analyze differences caused by walking speed, a one-way ANOVA for repeated measures was used (10 levels). To analyze differences due to body unloading, we used a two-way ANOVA for repeated measures (bodyweight unloading with four levels, walking speed with three levels) and their interaction; α was set at 0.05.

3. Results

3.1. The influence of walking speed

These tests were performed at speeds varying between 0.5 and 5.0 km h⁻¹, in successive steps of 0.5 km h⁻¹.

3.1.1. Time and distance parameters

The influence of walking speed on the cadence, stride length and the different phases of the walking cycle are presented in Fig. 2. Both the cadence (Fig. 2A) and normalized stride length (Fig. 2B) increased when the walking speed increased and the variability, reflected in the standard deviation, decreased at higher walking speed. The cadence was significantly affected by walking speed [F(9, 171) = 613.6; P < 0.001]. Pair-wise comparisons showed that the difference was just not significant between 4.5 and 5.0 km h⁻¹ (P = 0.08) and significant between all other speeds (between 4.0 and 4.5 km h⁻¹: P = 0.048; between 3.5 and 4.0 km h⁻¹: P = 0.002; for all other comparisons: P < 0.001). The normalized stride length was also significantly affected by walking speed [F(9, 171) = 767.5; P < 0.001]. Pair-wise comparisons showed that the stride length differed between each successive speed (for all: P < 0.001). The Pearson’s correlation coefficient, calculated between the averaged cadence and stride length for the different walking speeds was 0.995.

With increasing speed, the relative swing phase duration and single stance duration increased, while the relative duration of the first and second double stance decreased (Fig. 2C). All four measures became affected by walking speed (for all: P < 0.001). For walking speeds less than
2.0 km h\(^{-1}\), pair-wise comparisons showed that DSD1 and DSD2 decreased and SS and SW increased significantly between successive walking speeds with \(P < 0.001\). Between 2.0 and 2.5 km h\(^{-1}\), an increase in SW (\(P = 0.003\)) and SS (\(P < 0.001\)) occurred, while DSD1 and DSD2 decreased (for both: \(P < 0.001\)). Finally, a slight but significant increase in DSD2 occurred between 2.5 and 3.0 km h\(^{-1}\) (\(P = 0.045\)). Remarkably, the relative swing phase duration at a walking speed of 0.5 km h\(^{-1}\) was less than 20% of the gait cycle and increased towards almost 40% at higher walking speeds.

3.1.2. Joint kinematics

Changes in joint kinematics due to different walking speeds are shown in Fig. 3. Table 1 shows the correlation coefficients calculated between the joint trajectories recorded during the several walking speeds and the trajectory at a walking speed of 5.0 km h\(^{-1}\). Although the amplitude of the hip joint movement clearly increased as walking speed increased (Fig. 3A), there were no large differences in movement pattern, except at very slow walking speeds. A similar finding was found for the knee joint (Fig. 3B, Table 1). Clear changes in amplitude and movement pattern occurred in the ankle joint (Fig. 3C, Table 1), already at speeds slower than 3.0 km h\(^{-1}\).

3.1.3. Leg muscle EMG activity

Fig. 4 shows the changes in muscle activity due to changes in walking speed. Table 1 shows the correlation coefficients calculated between the leg muscle EMG activity patterns recorded during the several walking speeds and the EMG pattern at a walking speed of 5.0 km h\(^{-1}\). In general, the EMG patterns of all muscles became less similar with the pattern recorded at a walking speed of 5.0 km h\(^{-1}\), as the walking speed decreased. The largest differences occurred between 2.0 and 2.5 km h\(^{-1}\) for most muscles (Table 1). Fig. 5 shows the mean trajectories ± one standard deviation of the TA muscle. This should provide some insight into the variability of the muscle EMG pattern, which decreased as walking speed increased.

3.2. The influence of bodyweight unloading

Four conditions of bodyweight unloading (0%, 25%, 50% and 75%) were tested during three walking speeds (1.5, 2.0 and 2.5 km h\(^{-1}\)). It was difficult for the subjects to walk with 75% bodyweight unloading. Some of the subjects had to walk on their forefeet, to keep in contact with the treadmill.

3.2.1. Time and distance parameters

There appear to be no large differences in cadence or stride length due to different amounts of bodyweight unloading (Fig. 6). However, ANOVA for repeated measures found that the cadence was significantly affected by bodyweight unloading [\(F(3, 57) = 7.6; P < 0.001\)], walking speed [\(F(2, 38) = 190.7; P < 0.001\)] and their interaction [\(F(6, 114) = 4.7; P < 0.001\)]. Pair-wise comparisons showed that the cadence differed in the 75% bodyweight unloading condition from the 0% (\(P = 0.015\)), 25% (\(P = 0.002\)) and 50% (\(P < 0.001\)) conditions. The normalized stride length was also affected by bodyweight unloading [\(F(3, 57) = 11.4; P < 0.001\)], walking speed [\(F(2, 38) = 454.0; P < 0.001\)] and their interaction [\(F(6, 114) = 4.3; P < 0.001\)]. Again, the stride length in the 75% bodyweight unloading condition differed from the other conditions (for all: \(P < 0.001\)).

Both the relative swing phase and single stance duration increased, with an increase in bodyweight unloading, while the duration of the double support decreased. All four measures became significantly affected by bodyweight unloading, speed and their interaction (for all: \(P < 0.001\)). Pair-wise comparisons showed that these measures differed...
Table 1

Linear relation of kinematics and EMG patterns between different walking speeds

<table>
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All correlation coefficients are calculated between the reference condition (5.0 km h\(^{-1}\)) and the presented conditions. Abbreviations: GL, gluteus maximus; RF, rectus femoris; VL, vastus lateralis; VM, vastus medialis; HL, lateral hamstrings; HM, medial hamstrings; TA, tibialis anterior and GM, gastrocnemius medialis.

between each bodyweight unloading condition (for DSD1 in 25–50%: \(P = 0.004\); SW in 25–50%: \(P = 0.006\); for all other comparisons: \(P < 0.001\)).

3.2.2. Joint kinematics

Fig. 7 shows the influence of bodyweight unloading on the joint trajectories, recorded at a walking speed of 2.0 km h\(^{-1}\). Table 2 shows the correlation coefficients calculated between the joint trajectories recorded during the several bodyweight unloading conditions and 0% bodyweight unloading. Although the \(r\) of the hip joint remained high during the three bodyweight unloading conditions (Table 2), the maximal extension decreased during the 50% and 75% bodyweight unloading condition (Fig. 7A). The \(r\) of the knee joint was considerably lower during the 75% bodyweight unloading condition (Fig. 7B). The ankle joint trajectory became also largely changed at 75% bodyweight unloading, which is reflected in the low \(r\) (Table 2) and can be visually inspected (Fig. 7C).

3.2.3. Leg muscle EMG activity

Fig. 8 shows the differences in muscle activity patterns recorded at a walking speed of 2.0 km h\(^{-1}\) of several bodyweight unloading conditions. Table 2 shows the correlation coefficients calculated between the leg muscle EMG patterns recorded during the several bodyweight unloading conditions and 0% bodyweight unloading. The GL muscle (Fig. 8A) showed some changes during all bodyweight unloading conditions compared to 0% bodyweight unloading (Fig. 8A and Table 2). The EMG pattern of the RF muscle (Fig. 8B) became clearly influenced by 75% bodyweight unloading (Table 2), but not so much in the other bodyweight unloading conditions. This could also be observed for the VL (Fig. 8C and Table 2), while the EMG pattern of the VM muscle (Fig. 8D) was more influenced by 75% and 50% bodyweight unloading. The largest differences were observed for the HL EMG patterns (Fig. 8E, Table 2), where the 50% and 75% body weight unloading conditions differed largely from the 0% condition. Similar findings were shown for the HM muscles (Fig. 8F), although less strong than for the HL muscle. The TA muscle (Fig. 8G) appeared to be only slightly influenced by bodyweight unloading (Table 2), while in contrast the EMG pattern of the GM muscle (Fig. 8H) was more strongly influenced by the amount of bodyweight unloading, mainly at 75% unloading (Table 2).

4. Discussion

The main results were as follows: (1) Cadence and stride length are largely influenced by walking speed, while bodyweight unloading influences these measures only at 75% unloading. (2) The relative duration of the gait phases changes largely at walking speeds less than 2.5 km h\(^{-1}\) and is influenced by all different bodyweight unloading conditions. (3) Joint trajectories of knee and ankle joint, as well as leg muscle EMG activity patterns change largely at walking speeds slower than 2.5 km h\(^{-1}\) and 75% bodyweight unloading. (4) Low walking speed seems to increase the variability of leg muscle EMG activity between subjects.

With increasing velocity, the product of cadence and stride length increases [18,19]. This occurs approximately linearly, as the high correlation coefficient shows. The reduced significance of the difference in cadence at high walking speeds can be explained by the decreased relative change in walking speed. The variability of the cadence and stride length between the subjects decreases as the speed increases, which indicates a more consistent walking pattern. Bodyweight unloading appears to have less impact on the cadence and stride length. Although in the 75% bodyweight unloading condition statistically significant differences are found for both parameters, the differences are so small that they might not be of any clinical significance.
Fig. 4. Influence of walking speed on EMG activity patterns. EMG activity patterns during different walking speeds of the (A) gluteus maximus (GL), (B) rectus femoris (RF), (C) vastus lateralis (VL), (D) vastus medialis (VM), (E) lateral hamstrings (HL), (F) medial hamstrings (HM), (G) tibialis anterior (TA) and (H) gastrocnemius medialis (GM) muscles.
The changes in the relative timing of the different gait phases during the gait cycle are strongly influenced by walking speed and bodyweight unloading. Walking speeds below 2.5 km h\(^{-1}\) influence the relative timing of the gait phases considerably. Since joint kinematics and muscle EMG patterns are closely related to the gait phase, both parameters will change considerably at lower walking speeds, as the present results show.

Many changes in muscle EMG patterns occur at walking speeds slower than 2.5 km h\(^{-1}\). While one study did not investigate such slow walking speeds [20], another study investigated walking speeds slower than 1.0 km h\(^{-1}\), but excluded the gap between 1.0 and 3.0 km h\(^{-1}\) [16], which is exactly the range of speeds achieved by our SCI patients. Indeed, many changes in leg muscle activity patterns can be explained by the change in relative duration of the gait phases. At 0.5 km h\(^{-1}\), the stance phase accounts for over 80% of the gait cycle. This influences the joint trajectories and leg muscle activity patterns, which are normalized for time, considerably. Since extensor muscles are mainly active...
during stance, the activity of GL, RF, VL, VM and GM lasted longer during the gait cycle at slow speeds. In addition, activation of flexor muscles (HL, TA and especially HM) lasted longer during the stance phase at low walking speeds. Although not quantified, this has to result in an increased co-contraction of flexor and extensor muscles.

Previous studies [13,14] showed that muscle EMG activity differed strongly between different bodyweight unloading conditions. Similar to these studies, the EMG activity pattern of TA, a flexor muscle, was much less influenced than GM, an extensor muscle. However, we could not generalize this result to more proximal muscles. The EMG activity patterns of HL and HM were strongly influenced by bodyweight unloading, in contrast to VL and VM, which were less influenced. This difference could be explained by the different parameters investigated. Previous studies [13,14] showed that muscle EMG activity differed strongly between different bodyweight unloading conditions. Similar to these studies, the EMG activity pattern of TA, a flexor muscle, was much less influenced than GM, an extensor muscle. However, we could not generalize this result to more proximal muscles. The EMG activity patterns of HL and HM were strongly influenced by bodyweight unloading, in contrast to VL and VM, which were less influenced. This difference could be explained by the different parameters investigated. Previous

Table 2
Linear relation of kinematics and EMG patterns between different bodyweight unloading conditions

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All correlation coefficients are calculated between the reference condition (0% bodyweight unloading) and the presented conditions. Abbreviations: GL, gluteus maximus; RF, rectus femoris; VL, vastus lateralis; VM, vastus medialis; HL, lateral hamstrings; HM, medial hamstrings; TA, tibialis anterior and GM, gastrocnemius medialis.
Fig. 8. Influence of bodyweight unloading on EMG activity patterns. EMG activity patterns during different bodyweight unloading conditions (0%, 25%, 50% and 75%) of the (A) gluteus maximus (GL), (B) rectus femoris (RF), (C) vastus lateralis (VL), (D) vastus medialis (VM), (E) lateral hamstrings (HL), (F) medial hamstrings (HM), (G) tibialis anterior (TA) and (H) gastrocnemius medialis (GM) muscles during 2.0 km h⁻¹. Note the difference in calibrations of the y-axis.
studies evaluated mainly the EMG amplitude to quantify load dependency in leg muscle activation, while we analyzed the EMG pattern of the curve. Due to the normalization method, we are unable to make any conclusions about changes in EMG amplitude.

4.1. Methodological considerations

An important question is whether the present results could also be applied to walking over ground, since some patients can walk over ground, but at a slow walking speed. In addition, new devices allow patients to walk over ground with partial bodyweight unloading. There are some differences between treadmill and over ground walking. During treadmill walking, increases in hip range of motion, maximum hip flexion joint angle and cadence occur, while the stance time decreases [21]. Knee kinematics appear to be comparable [22], while vertical ground reaction forces show similar patterns, but differ slightly in force magnitude during mid- and late stance [23]. Leg muscle EMG activity is somewhat smaller in over ground versus treadmill walking [24,25]. In these studies, the authors observed no large differences in timing of EMG bursts, except for the hamstrings. Activity of the hamstrings occurred earlier at onset of the stance phase during treadmill walking [24] and especially at slow walking speeds [25]. We expect that some parameters (knee kinematics, most leg muscle activity patterns) that we investigated in our study might be generalized to slow or partially bodyweight unloaded over ground walking. However, future research will need to address this question more accurately.

Furthermore, we measured only young healthy subjects in these experiments. How well this data can be applied to elderly subjects remains questionable. Some differences in gait parameters, joint kinematics and EMG patterns have been described between young and elderly subjects [26], as well as differences in load dependency [13]. Therefore, investigating gait parameters, joint kinematics and muscle EMG patterns in a sample of healthy elderly subjects would be interesting for a future study, since many gait impairing disorders occur in the elderly.

4.2. Clinical relevance

We present a clinical example of a male subject (age: 26 years, stature: 1.75 m, weight: 59 kg), who acquired a traumatic incomplete SCI at cervical 5. Initially, he was unable to stand or walk. After 3 weeks, he started with BWSTT. Six weeks after SCI, he was able to walk on the treadmill without bodyweight support, but only at very slow speed (maximum 1.5 km h\(^{-1}\)). Fig. 9 shows the 'normal' TA activity patterns recorded at walking speeds of 1.5 km h\(^{-1}\) (TA1.5) and 5.0 km h\(^{-1}\) (TA5.0) from our database and the TA EMG activity patterns of the incomplete spinal cord injured patient, all recorded at 1.5 km h\(^{-1}\), at 6 weeks (SCI06), 15 weeks (SCI15) and 23 (SCI23) weeks after the injury.

Six weeks after injury, his TA EMG pattern correlated only moderately \((r = 0.57)\) with the TA5.0, but well with the TA1.5 \((r = 0.82); \text{Fig. 9}\). As the SCI patient recovered over time, his TA EMG pattern became more similar with TA1.5 (after 15 weeks: \(r = 0.90\); after 23 weeks: \(r = 0.92\)). These small improvements in TA EMG pattern occurred in parallel to improvements in over ground walking capacity measured with timed walking tests [27]. Six weeks after injury, he walked 10 m in 42 s and 145 m within 6 min at preferred walking speed. Fifteen weeks after SCI, he walked 10 m in 8 s and 445 m within 6 min, which implies normal walking speed. While his TA EMG pattern, recorded at 1.5 km h\(^{-1}\), resembled TA1.5 well, it still correlated only moderately with TA5.0 \((r = 0.68)\). The correlation with TA5.0 did not further improve after 23 weeks \((r = 0.61)\).

This example shows that although he was initially unable to walk at normal walking speed, his TA activity pattern resembled the normal TA activity pattern recorded at 1.5 km h\(^{-1}\) well. However, his TA activity pattern differed largely from normal when compared to the TA pattern at 5.0 km h\(^{-1}\), which substantiates the importance of a speed-adjusted database. Furthermore, improvement in over ground walking capacity appeared to occur in parallel to an improvement in similarity between the TA EMG patterns recorded at similar walking speed, which might validate the assumption that an improved EMG pattern could have a functional significance.

We suggest that BWSTT should preferably be performed at a minimal walking speed of 2.5 km h\(^{-1}\) and with bodyweight unloading less than 75%, since this resembles closely the normal walking pattern. Practically however, this might be too intensive for many patients. For example, SCI patients who exhibit poor voluntary control and additional spasticity at the onset of their rehabilitation program will be unable to train at such conditions, even with adequate support of physical therapists or robotic devices. For these patients, the slow walking speed or high bodyweight unloading will result in somewhat different gait parameters, joint kinematics and muscle activity patterns, which might...
slow down the rate of functional recovery. However, even at slow walking speeds, BWSTT can positively affect other components of locomotor function. It resembles the functional task of walking well and remains a relative safe and easily controllable training tool for physical therapists.

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References