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RESEARCH ARTICLE

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Modulation of locomotor activity in complete spinal cord injury

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Abstract The aim of this study was to evaluate the modulation of muscle activity during locomotor-like movements by different walking speeds in subjects with a motor complete spinal cord injury (SCI) compared to actively- and passively-walking control subjects without neurological deficit. Stepping movements on a treadmill were induced and assisted by a driven gait orthosis. Electromyographic (EMG) muscle activity of one leg (rectus and biceps femoris, tibialis anterior and gastrocnemius) was recorded and analyzed at three stepping velocities with similar body weight support in both subject groups. In SCI subjects, the EMG amplitude of biceps femoris, tibialis anterior and gastrocnemius was in general similar or weaker than in passively- and actively-stepping control subjects, but that of rectus femoris was larger. The degree of co-activation between tibialis anterior and gastrocnemius was higher in SCI than in control subjects. A significant velocity-dependent EMG modulation was present in all four-leg muscles in both subject groups. In SCI subjects, this EMG modulation was similar to that in actively stepping control subjects. It is concluded that in complete spastic SCI subjects, spinal neuronal circuits underlying locomotion can to a large extent adequately respond to a change in external drive to adapt the neuronal pattern to a new locomotion speed. The application of various speeds might enhance the effect of locomotor training in incomplete SCI subjects.

Keywords Electromyography · Locomotion speed · Leg muscle EMG · Spinal cord · Paraplegia

Abbreviations ANOVA: Analysis of variance · BF: m. biceps femoris · BWS: Body weight support · DGO: Driven gait orthosis · EMG: Electromyography · GM: m. gastrocnemius medialis · MF: Modulation factor · RF: m. rectus femoris · RMS: Root mean square · SCI: Spinal cord injury · TA: m. tibialis anterior

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Introduction

In the last years, increasing evidence has arisen that a locomotor pattern can be induced and trained in human subjects with a complete spinal cord injury (SCI) (Jilge et al. 2004; for reviews see Dietz and Harkema 2004; Barbeau and Rossignol 1994), corresponding to what was previously shown for the cat (Barbeau et al. 1987; De Leon et al. 1999; for review see Edgerton et al. 2004). The main difference in the locomotor pattern recorded in subjects with an SCI, compared to control subjects without neurological deficit, is a greatly reduced amplitude of leg muscle EMG activity (Dietz et al. 1995). Consequently, these patients require an unloading of up to 80% of their body weight to allow for the induction of assisted stepping movements.

The essential stimulus to evoke locomotor activity and to obtain training effects in patients with spinal cord injury was shown to arise from load and hip joint-related afferent input (Dietz et al. 2002). Stretch reflexes were shown to play only a minor role in the leg muscle activation during locomotor movements in SCI subjects (Harkema et al. 1997; Dietz et al. 1998a, b; Beres-Jones and Harkema 2004).

In two studies (Dobkin et al. 1995; Beres-Jones and Harkema 2004) the effect of walking velocity on modulation of leg muscle activity was assessed in SCI subjects. The observation of an increase in EMG amplitudes and a shortening of EMG burst durations with increasing walking velocity might be compromised by the manual assistance of the leg movements resulting in varying trajectories. With the availability of a driven gait orthosis (DGO) to assist the locomotor movements (Colombo et al. 2000, 2001), well-controlled movement

trajectories for SCI and control subjects can be provided by a closed-loop position control strategy over a range of movement speeds. Thus, the aim of this study was to evaluate the degree to which the pattern of leg muscle activation induced by the DGO in these two subject groups becomes modulated by different speeds. Furthermore, the extent to which the leg muscle activation pattern in SCI subjects is modulated by a given externally imposed locomotion speed was studied. To assess the potential difference to the unaffected nervous system, EMG modulation was compared to that of actively- and passively- walking control subjects within the DGO. The answers to these questions could have consequences for locomotor training therapy, as the issue of optimal locomotion speed for achieving the best training effects in patients with SCI has remained open (cf. Dietz et al. 1998b; Dietz and Harkema 2004).

Methods

With the permission of the local Ethics Committee and the informed consent of the participants, leg muscle electromyographic (EMG) activity at different velocities of locomotion was recorded and analyzed in nine control subjects without neurological deficit (age 29 ± 3 years, ranging from 25 to 33 years) and 13 subjects with SCI (age 37 ± 14 years, ranging from 21 to 67 years). According to the classification of the American Spinal Injury Association (Maynard et al. 1997), the participating SCI subjects had sensory incomplete but motor complete (ASIA B) or sensory and motor complete (ASIA A) lesions with a neurological level of lesion between C4 and L1 without signs of peripheral nerve damage (see Table 1 for details). All SCI subjects showed spastic signs in the legs, such as exaggerated reflexes, and most of them also displayed muscle hypertonia. Eight SCI subjects were on antispastic medication (Baclofen) when they participated in the experiments.

Table 1 Characteristics of subjects with SCI included in the study

Patient	Sex	Age (years)	ASIA	Level of lesion	Duration of lesion (years)	Muscle tone (Ashworth)
P1	M	45	B	C6	2	3
P2	M	30	A	T10	0.5	3
P3	F	22	A	T5	2.5	4
P4	M	54	A	T4	33	3
P5	M	21	A	C4	1	4
P6	M	32	A	T6	0.2	2
P7	M	41	B	C4	7.5	2
P8	F	24	B	L1	2	2
P9	M	29	A	T6	1	2
P10	M	27	B	T5	4	2
P11	M	67	A	T5	0.5	4
P12	M	55	A	T4	1.5	3
P13	F	30	A	T9	12	1

Sex: *F* female, *M* male. Classification according to the American Spinal Injury Association (ASIA). *A* sensory and motor complete, *B* motor complete, sensory incomplete. Neurological level of lesion: *C* cervical, *T* thoracic, *L* lumbal. Muscle tone of hip and knee flexors and extensors assessed by the Modified Ashworth Scale (Bohannon and Smith 1987)

Stepping movements on a treadmill were induced and assisted by a driven gait orthosis. A detailed description of the driven gait orthosis (DGO) "Lokomat®" (Hocoma AG, Volketswil, Switzerland) can be found elsewhere (Colombo et al. 2000, 2001). Briefly, the subjects are fixed to the DGO by straps around the waist, the thighs and the shanks. The orthosis can be adjusted in size at the different segments and, therefore, can be adapted to the different subjects. The DGO provides drives for flexion and extension movements of the hip and knee joints of both legs, whereas the feet can move freely. The trajectories for hip and knee represent a physiological gait pattern and are identical for all SCI and control subjects. For the SCI subjects, passive foot lifters (elastic straps/springs) are required to prevent drop-foot posture and to provide safe foot clearance. Foot lifters were also applied in control subjects to ensure similar afferent input and ankle trajectories. Stepping cadence differed at the same speed for different subjects because of the different leg lengths. Unloading was achieved by a parachute harness connected to counterweights. For comparison, both SCI and control subjects walked with the same body weight support (BWS) of 70% of the body weight. For SCI subjects, such an unloading is necessary to allow stepping movements to be induced (Dietz et al. 1995) and to prevent stumbling.

Prior to the experiments, the subjects became familiarized with the DGO. Before the recording session reported here, all SCI subjects experienced some DGO walking within other research projects.

All SCI and control subjects performed treadmill walking within the DGO at three velocities: 1.5, 2.0 and 2.5 km/h (0.42, 0.56, 0.69 m/s), each for 20 gait cycles. Walking at 2 km/h was the first block of 20 steps and was repeated for the last block of 20 steps in order to see any systematic alteration of leg muscle activation, e.g. a decrease in EMG amplitude during a walking session as recently reported for chronic SCI subjects (Dietz and Müller 2004). The relatively low walking speeds are representative for training sessions according to current clinical practice. Pilot studies in patients with complete SCI showed that higher speeds might, for these patients, lead to un-physiological stress. The total duration of one experiment was about 60 min, the measurement time to less than 5 min for SCI subjects, and 15 min for control subjects. Subjects did not report fatigue within this time. Control subjects were instructed for trial (i) to walk passively (i.e. to have stepping movements induced by the DGO), for trial (ii) to actively walk as normal as possible within the DGO, and for trial (iii) to walk on the treadmill with the same body weight support (i.e. 70% BWS) but without the DGO. In the passive DGO walking condition, the control subjects were instructed to relax as much as possible and not to intervene with the leg movements imposed by the DGO.

During walking, the EMG activity in the rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA) and medial gastrocnemius (GM) muscles of one leg was recorded. Surface electrodes were fixed over the muscle belly with an inter-electrode distance of 25 mm. For data recording and signal analysis, Soleasy Software (ALEA Solutions GmbH, Zurich, Switzerland) was used. EMG signals were amplified, filtered (band pass 30–300 Hz) and transferred to a PC via a 12-bit A/D-converter. In addition, the two trigger signals identifying the beginning of the stance phases of the right and left leg were recorded. All signals were sampled at 1 kHz.

The EMG activity of each leg muscle was quantified for every gait cycle by calculating the root mean square (RMS) of the EMG signal over the whole gait cycle, which represents the mean or effective amplitude (see De Luca 1997). The values for each muscle were averaged over the 20 gait cycles at each speed tested.

A modulation factor (MF) was calculated by the following equation

$$\mathrm{MF} = \frac{\mathrm{EMG}(2.5\,\mathrm{km/h}) - \mathrm{EMG}(1.5\,\mathrm{km/h})}{\mathrm{EMG}(2.0\,\mathrm{km/h})} \times 100\%,$$

where "EMG( )" represents the RMS value of the respective muscle at the corresponding velocity. By this calculation the change in leg muscle EMG activity from 1.5 to 2.5 km/h walking relative to the activity level during walking at 2.0 km/h was analysed and quantified.

Furthermore, the co-activation level in the lower leg muscles during locomotion was assessed. The EMG signals of TA and GM were normalized to 1,000 samples per step, rectified, and filtered (5th order Butterworth low-pass filter with 10 Hz). Thereafter, the signals were classified into active or non-active using the method of Yakovenko et al. (2005): The (voltage) threshold for the classification of activity or inactivity was based on the EMG signal of this muscle recorded during each task (for details see Yakovenko et al. 2005). To prevent misinterpretation of sporadic supra-threshold EMG spikes, a continuous EMG activity of 20 consecutive samples was the prerequisite for the acceptance of begin and end of an activity period. The co-activation signal of TA and GM was obtained by multiplication of the classified signals (logical conjunction of the individual signals) and averaging over the gait cycle. This analysis was done separately for each subject, each walking condition and each speed, because the level of activity varied between conditions. Setting of a more general threshold might have underestimated the co-activation level in conditions with lower activity.

For statistical analysis of the absolute EMG values and of the degree of co-activation, an analysis of variance (ANOVA) for repeated measures was used for each of the four muscles (RF, BF, TA, GM) to compare the values at the three different velocities (1.5, 2.0, 2.5 km/h) for the three walking conditions (and their interaction) applied in control subjects (passive DGO, active DGO, treadmill without DGO) and to compare the effect of speed during the single walking condition in SCI subjects. To assess potential differences between the SCI and the control group, the single condition (with DGO) in SCI subjects was compared to each of the three walking conditions in control subjects using a repeated-measures ANOVA. The statistical analysis of the modulation factor was performed using an ANOVA for repeated measures comparing the four muscles and the four walking conditions. Pair-wise comparisons were performed by the t-test with additional Bonferroni's correction. The significance level was set to $P < 0.05$. The statistical analyses were performed with SPSS (SPSS Inc., Chicago IL, USA). For comparisons within the subject groups the general linear model function (GLM), and for comparison between the subject groups the mixed model function (MIXED) function were used.

Results

All subjects were able to walk at the three velocities within the DGO without any problems and without reporting fatigue. No systematic change was observed in the locomotion EMG pattern between the first and the second block of steps at 2 km/h.

Strength of leg muscle activity

Figure 1 displays the EMG activity patterns of four leg muscles from all SCI and control subjects at two locomotion speeds. A moderate increase in EMG amplitude, especially in the lower leg muscles, at the higher compared to the lower speed was visible in control subjects. When stepping movements were passively induced in control subjects, which corresponds to the condition in SCI subjects, relevant leg muscle EMG was only recorded at higher speed (Fig. 1a (i)). In control subjects, no relevant difference was visible between active walking within and without DGO (see Fig. 1a (ii), (iii)) in the leg muscle activation except for TA, which was higher in the condition without foot lifters (Fig. 1a (iii)).

In the SCI subjects, BF, TA and GM EMG amplitudes were smaller and RF amplitudes were larger than in the actively walking control subjects (Fig. 1b vs. Fig. 1a (ii), (iii)).

The absolute RMS values of the EMG activity for the entire study population are shown in Fig. 2. Statistical analysis of the upper leg muscles in control subjects (Fig. 2a (i–iii)) showed no significant influence by task, speed or their interaction (P and F values see Table 2). The lower leg muscles showed a significant speed- and task-dependent absolute activity. Specifically, the absolute RMS activity in TA and GM increased with increasing speed, significantly from 1.5 to 2.0 and 1.5 to 2.5 km/h (t-test with Bonferroni correction, $P < 0.05$). For TA, comparisons of the walking conditions showed pair-wise significant differences; passive DGO walking

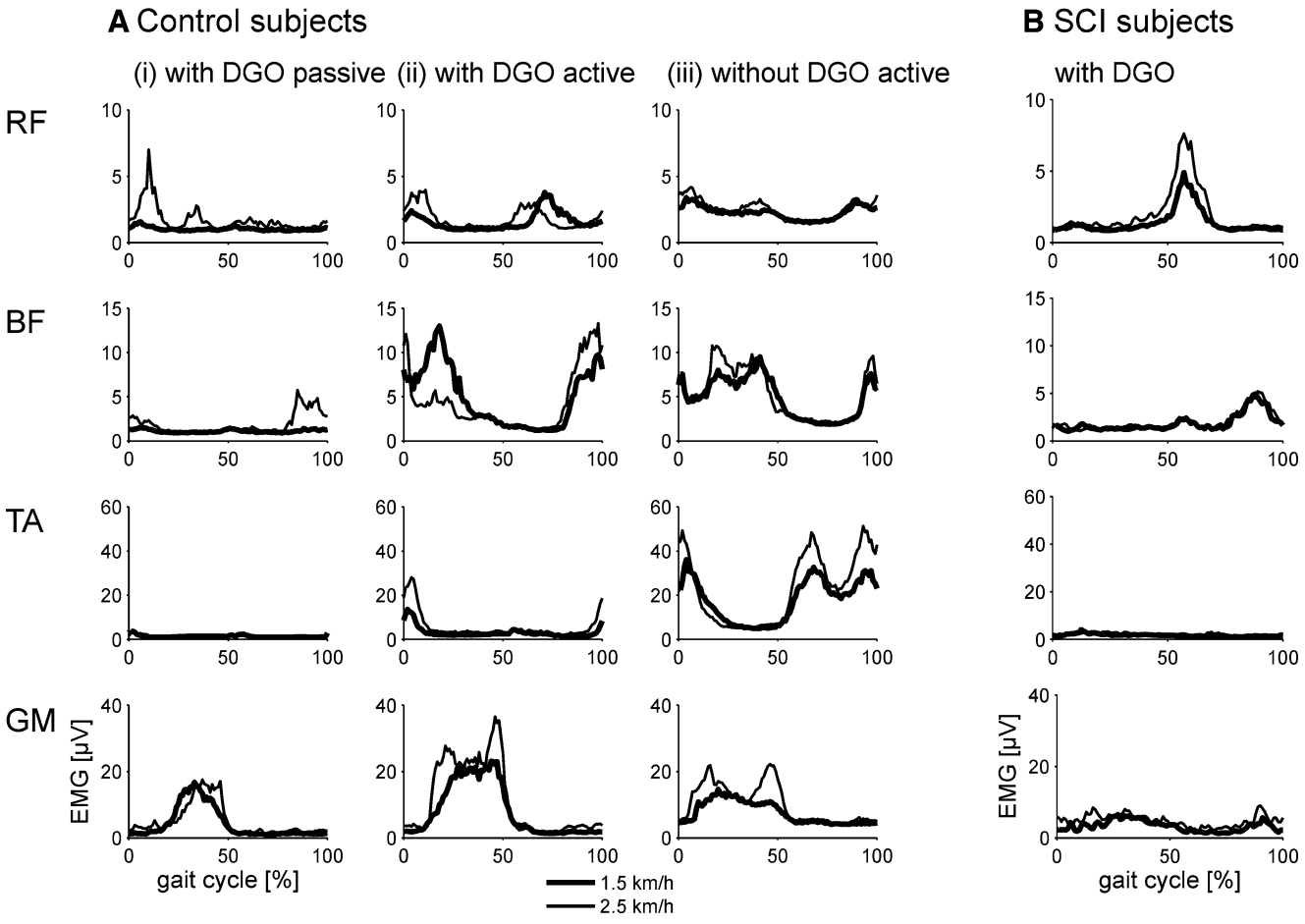

Fig. 1 **a**, **b** Grand means of EMG activity of four leg muscles (*RF* rectus femoris; *BF* biceps femoris; *TA* tibialis anterior; *GM* medial gastrocnemius) for the group of (**a**) control subjects walking with the same body weight support (*i*) passively, (*ii*) actively within the DGO, and (*iii*) on the treadmill without DGO, and (**b**) SCI subjects walking within the DGO. Activity at low stepping velocity (1.5 km/h) is indicated by *black*, and at higher velocity (2.5 km/h) by *grey* lines

resulted in the lowest, while active treadmill walking resulted in the highest activity (t-test with Bonferroni correction, $P < 0.05$). In the control subjects, the GM showed higher absolute EMG amplitudes during active than passive DGO walking (t-test with Bonferroni correction, $P = 0.01$). The relation of absolute RMS-EMG activity and velocity was different for the different walking conditions for TA (interaction: $P = 0.02$), but not for GM (interaction: $P = 0.51$). For TA, the absolute EMG activity was higher for both 2.0 and 2.5 km/h compared to 1.5 km/h during passive DGO walking; it was higher for 2.5 km/h than for 1.5 km/h during active DGO walking and was higher for 2.5 km/h compared to both 1.5 and 2.0 km/h during active treadmill walking (t-test with Bonferroni correction, $P < 0.05$).

The absolute EMG activity in SCI subjects (Fig. 2B) increased with walking velocity. However, this influence was significant only for RF (factor speed: $P = 0.03$) but not for BF ($P = 0.15$), TA ($P = 0.29$) and GM ($P = 0.55$). The absolute RF activity was significantly lower for 1.5 km/h than for the other two walking velocities (t-test with Bonferroni correction, $P = 0.03$).

In the SCI subjects, absolute RMS–EMG activity was significantly lower compared to that of control subjects during active DGO and active treadmill walking for GM, and during active DGO walking for TA (factor task/group: $P = 0.01$ for all; for details see Table 2).

Velocity-dependent modulation of EMG activity

The change of EMG activity associated by the different walking velocities was assessed by the modulation factor (MF, i.e. the change in RMS–EMG activity from 1.5 to 2.5 km/h relative to the activity at 2.0 km/h). In control subjects, significant modulation (MF > 0 with $P < 0.05$) was present in the upper leg muscles (RF, BF) during passive DGO walking and in the lower leg muscles (TA, GM) during active DGO and treadmill walking (Fig. 3). In SCI subjects, the lower leg muscles (TA and GM) and the RF (but not BF), showed a significant modulation (MF > 0 with $P < 0.05$).

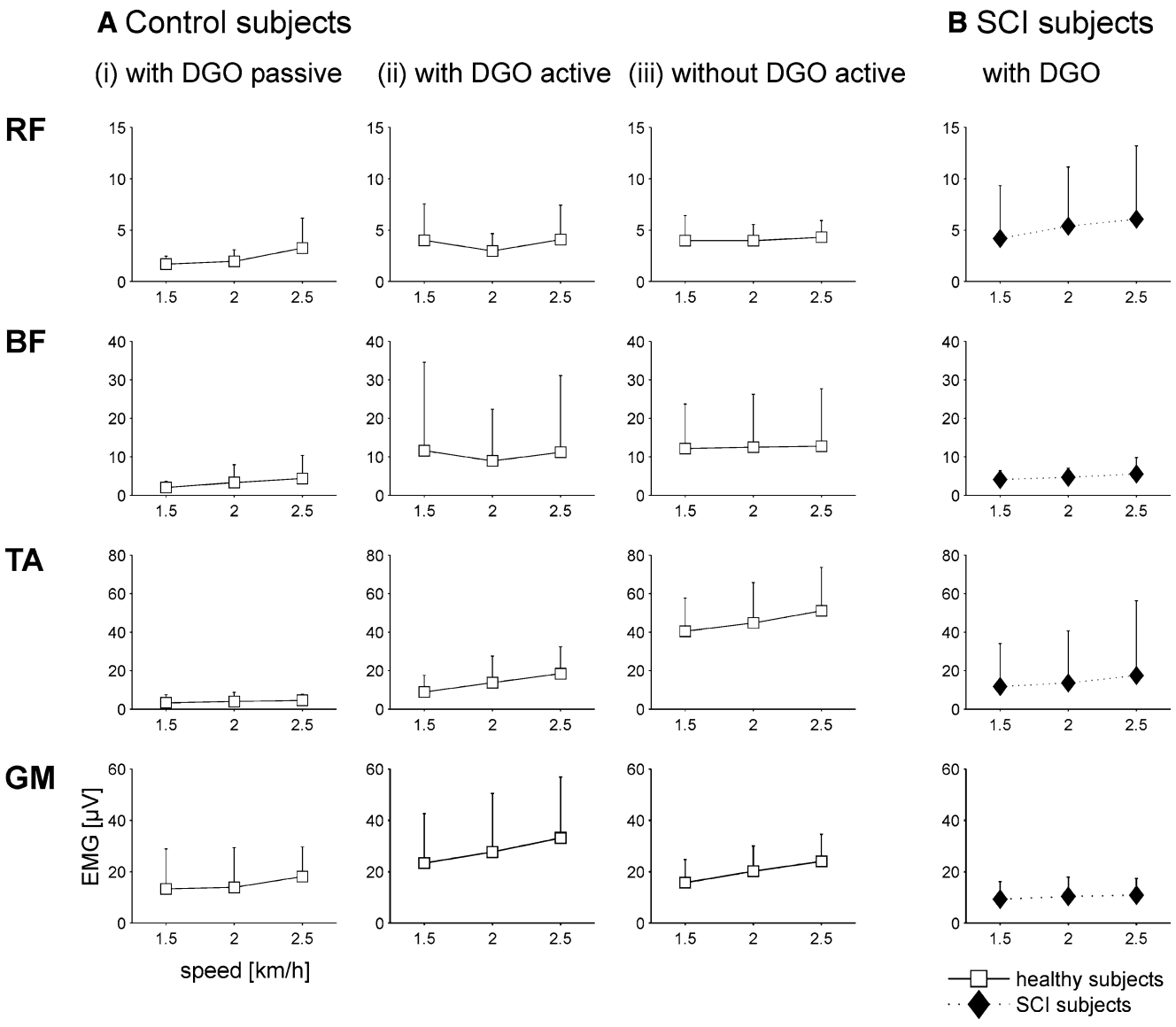

Fig. 2 a, **b** Changes in absolute EMG activity of the four leg muscles (*RF* rectus femoris; *BF* biceps femoris; *TA* tibialis anterior; *GM* medial gastrocnemius) at three gait velocities for all control (**a**) and SCI (**b**) subjects (mean and standard deviation). Each of the three walking conditions of control subjects ((*i*) passive, (*ii*) active walking within the DGO, and (*iii*) treadmill walking without DGO) and the single DGO walking condition in SCI subjects are plotted

During passive DGO walking MF showed no difference for any leg muscle between control and SCI subjects (t-test, all $P > 0.05$). During active DGO walking, MF for TA was higher in control than in SCI subjects ($t(20) = 2.597$, $P = 0.02$). There was no difference for the other muscles (RF, BF, GM, $P > 0.05$) between control and SCI subjects. During active treadmill walking, the MF for all leg muscles showed no difference between control and SCI subjects ($P > 0.05$).

Co-activation of leg muscles

The degree of co-activation was calculated as the average proportion of the gait cycle during which the TA and GM muscles were simultaneously active (for details see Methods). The degree of co-activation (Fig. 4) was highly variable in the SCI group, ranging from less than 1 to 53% of the gait cycle. For the analysis of the co-activation level, the two blocks of steps at 2.0 km/h were not pooled because the co-activation differed significantly between these blocks in 7 of the 13 SCI subjects, with a smaller level of co-activation in the second block for 5 and an higher level in 2 subjects (per-subject Wilcoxon signed ranks test on 2 sets of 20 steps in each phase respectively, $P < 0.05$), i.e. the proportion of the gait cycle during which TA and GM muscles were co-activated was usually smaller at the end of the recording session compared to the start. However, this difference was not significant for the whole SCI group (Wilcoxon signed ranks test on the mean degree of co-activation over all 20 steps in each phase, $z = -1.334$, $P = 0.18$).

Table 2 Summary of statistical analysis of the different factors influencing the absolute RMS EMG activity of leg muscles and differences between control and SCI subjects

Factor	Intercept	Task/group	Speed	Task × speed
Within control subjects				
RF	<0.001*	0.123	0.075	0.131
	$F(1,8) = 44.759$	$F(2,16) = 2.394$	$F(2,16) = 3.067$	$F(4,32) = 1.923$
BF	0.058	0.053	0.392	0.595
	$F(1,8) = 4.884$	$F(2,16) = 3.536$	$F(2,16) = 0.994$	$F(4,32) = 0.704$
TA	<0.001*	<0.001*	<0.001*	0.022*
	$F(1,8) = 50.244$	$F(2,16) = 27.056$	$F(2,16) = 18.769$	$F(4,32) = 3.318$
GM	0.002*	0.036*	<0.001*	0.507
	$F(1,8) = 21.497$	$F(2,16) = 4.117$	$F(2,16) = 16.553$	$F(4,32) = 0.845$
Within SCI subjects				
RF	0.008*	NA	0.030*	NA
	$F(1,12) = 10.113$		$F(2,24) = 4.085$	
BF	<0.001*	NA	0.146	NA
	$F(1,12) = 38.835$		$F(2,24) = 2.088$	
TA	0.105	NA	0.289	NA
	$F(1,12) = 3.079$		$F(2,24) = 1.307$	
GM	<0.001*	NA	0.552	NA
	$F(1,12) = 33.965$		$F(2,24) = 0.610$	
SCI subjects versus control subjects during passive DGO walking				
RF	0.001*	0.167	0.003*	0.609
	$F(1,20) = 13.731$	$F(1,20) = 2.055$	$F(2,40) = 6.615$	$F(2,40) = 0.502$
BF	<0.001*	0.306	0.015*	0.760
	$F(1,20) = 30.173$	$F(1,20) = 1.105$	$F(2,40) = 4.665$	$F(2,40) = 0.276$
TA	0.081	0.302	0.290	0.595
	$F(1,20) = 3.370$	$F(1,20) = 1.095$	$F(2,40) = 1.276$	$F(2,40) = 0.527$
GM	<0.001*	0.267	0.054	0.321
	$F(1,20) = 34.139$	$F(1,20) = 1.303$	$F(2,40) = 3.146$	$F(2,40) = 1.169$
SCI subjects versus control subjects during active DGO walking				
RF	<0.001*	0.480	0.097	0.064
	$F(1,20) = 17.545$	$F(1,20) = 0.518$	$F(2,40) = 2.473$	$F(2,40) = 2.944$
BF	0.008*	0.278	0.381	0.358
	$F(1,20) = 8.632$	$F(1,20) = 1.243$	$F(2,40) = 0.990$	$F(2,40) = 1.055$
TA	0.014*	0.956	0.011*	0.709
	$F(1,20) = 7.285$	$F(1,20) = 0.003$	$F(2,40) = 5.037$	$F(2,40) = 0.346$
GM	<0.001*	0.010*	0.001*	0.016*
	$F(1,20) = 36.900$	$F(1,20) = 8.110$	$F(2,40) = 8.672$	$F(2,40) = 4.566$
SCI subjects versus control subjects during active treadmill walking				
RF	<0.001*	0.588	0.040*	0.174
	$F(1,20) = 20.584$	$F(1,20) = 0.303$	$F(2,40) = 3.489$	$F(2,40) = 1.830$
BF	<0.001*	0.054	0.314	0.803
	$F(1,20) = 20.847$	$F(1,20) = 4.179$	$F(2,40) = 1.192$	$F(2,40) = 0.221$
TA	<0.001*	0.012*	0.005*	0.591
	$F(1,20) = 28.082$	$F(1,20) = 7.667$	$F(2,40) = 6.041$	$F(2,40) = 0.533$
GM	<0.001*	0.008*	<0.001*	0.021*
	$F(1,20) = 81.917$	$F(1,20) = 8.683$	$F(2,40) = 9.372$	$F(2,40) = 4.284$

NA Not applicable, *RF* rectus femoris, *BF* biceps femoris, *TA* tibialis anterior, *GM* medial gastrocnemius
P values are derived from RM-ANOVA (GLM) for within group comparison and MIXED model for comparison between SCI and control subjects. Asterisks indicate significant factors ($P < 0.05$)

Comparing all four blocks of steps in SCI subjects (1.5, 2.5 and 2.0 km/h twice), no overall effect of walking velocity on the co-activation was observed (RM-ANOVA, $F(3,36) = 1.308$, $P = 0.29$). In the control subjects, the co-activation did not depend on the walking conditions (passive DGO, active DGO, active treadmill), walking velocity or the interaction of both (RM-ANOVA: factor task: $F(2,16) = 0.178$, $P = 0.85$; factor speed: $F(3,24) = 0.718$, $P = 0.55$; interaction speed × task: $F(6,48) = 1.147$, $P = 0.35$).

There was a significantly higher co-activation level in SCI subjects during DGO walking condition than in any of the three walking conditions of control subjects (passive DGO: $F(1,20) = 5.791$, $P = 0.03$; active DGO: $F(1,20) = 4.663$, $P = 0.04$; active treadmill: $F(1,20) = 6.381$, $P = 0.02$). The interactions were not significant (passive DGO: $F(3,60) = 2.023$, $P = 0.12$; active DGO: $F(3,60) = 0.938$, $P = 0.43$; active treadmill: $F(3,60) = 2.014$, $P = 0.11$).

Discussion

The aim of this study was to evaluate the modulation of leg muscle EMG activity during locomotor-like movements by changes of afferent input—here different

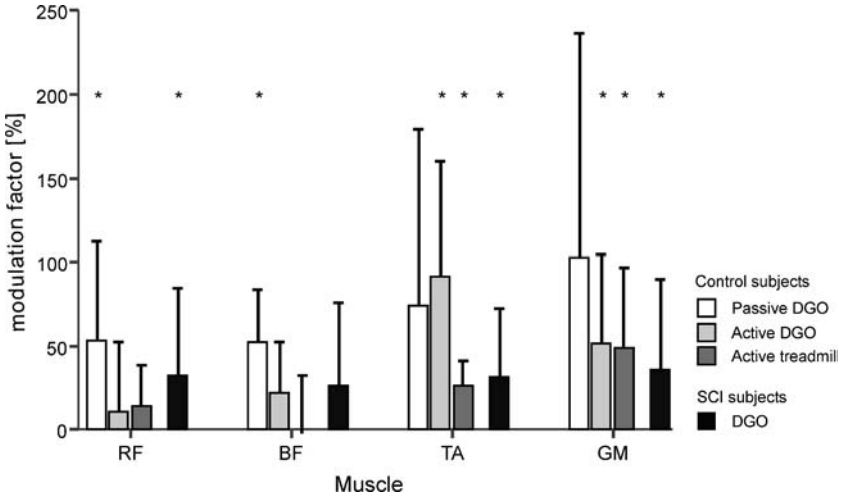

Fig. 3 Modulation factor (MF) of EMG activity of four leg muscles (*RF* rectus femoris; *BF* biceps femoris; *TA* tibialis anterior; *GM* medial gastrocnemius) during three walking conditions of control subjects (*light grey bars*) and the DGO walking condition of SCI subjects (*black bars*). The MF reflects the change in RMS-EMG activity from 1.5 to 2.5 km/h relative to the activity at 2.0 km/h. Significant modulation (compared to zero) is indicated by *asterisks* ($P < 0.05$)

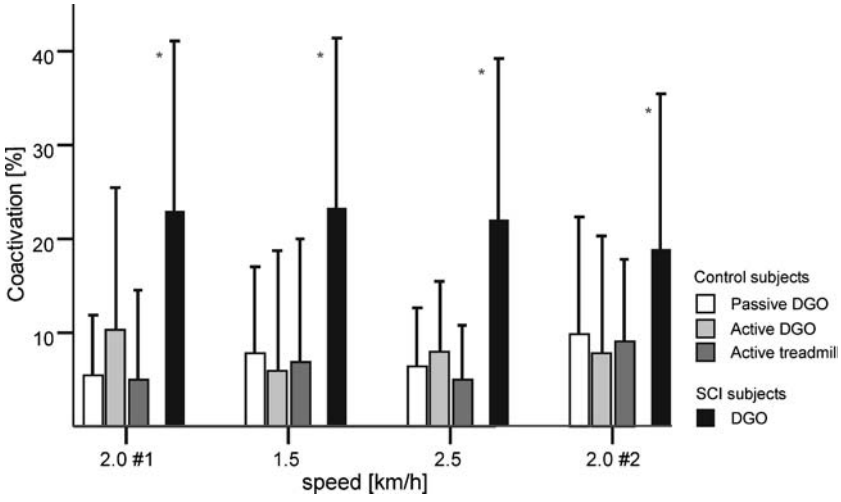

Fig. 4 Levels of EMG co-activation of TA and GM muscles at three walking velocities during three walking conditions in control and the single DGO walking condition in SCI subjects. The co-activation level was calculated as the proportions during which TA and GM were simultaneously active (for details see Methods). *Asterisks* indicate significant differences ($P < 0.05$) of the co-activation level in SCI subjects compared to all walking conditions in the control subjects. *#1* and *#2* indicate the first and second block of steps at velocity 2.0 km/h, respectively

walking velocities—in subjects with motor complete SCI and to compare it with control subjects. More specifically, how similarly does the spinal cord completely isolated from volitional supraspinal control respond to a change in external drive when compared to the intact condition in control subjects?

Structure of locomotor pattern

As described earlier, a locomotor-like activation pattern of leg muscles can be induced in patients with complete SCI if the locomotor movements are assisted and the essential afferent input is provided (Harkema et al. 1997; Dietz et al. 2002; Dietz and Harkema 2004). By the application of a robotic device, which allows for the induction of consistent leg movements in SCI subjects, the influence of the walking velocity, on leg muscle activation could be evaluated. The stepping condition of complete SCI subjects would best correspond to that of control subjects passively moving in the DGO. During passive stepping of control subjects (i.e. without voluntary supraspinal drive) at least some appropriate leg muscle activity became automatically generated, especially at higher velocities.

In the present study, the locomotor pattern in control subjects was similar during active walking within and without the DGO. However, differences in muscle activation between active treadmill and DGO walking were reported recently (Hidler and Wall, 2005). The difference in interpretation might be due to the fact that specific modulation occurred in some gait phases while the fundamental activation pattern was unchanged. A more influential factor was whether the control subjects walked passively or actively in the DGO.

It cannot be decided from this study whether the leg muscle activation induced in the SCI subjects is produced by a spinal locomotor pattern generator (CPG) as demonstrated in lower vertebrates (Grillner 1985). The contribution of stretch reflexes to the muscle activity in SCI subjects can hardly be deduced from the present experiments because walking velocity and rate of muscle stretch were linked via limb trajectory and cadence. It was previously supposed that stretch reflexes play a

minor role during locomotion of SCI subjects (Dietz et al. 1998b, 2002; Beres-Jones and Harkema 2004). In the light that simple rhythmic muscle stretch or loading alone does not lead to a meaningful leg muscle activation (Dietz et al. 2002), a combination of afferent inputs has to be assumed as the underlying mechanism of the EMG pattern described here.

Influence of external factors on the locomotor pattern

The main observation made in this study was that the modulation of the EMG pattern during locomotor-like movements seen in SCI subjects was in the range of that observed in passively- and actively-stepping control subjects. The generation of such a pattern in complete SCI subjects requires injury-induced changes in the spinal networks for locomotion. These changes appear to be associated with the presence of spasticity, as only in spastic SCI subjects could a "locomotor-like" EMG activity be induced during assisted locomotor movements (Dietz et al. 1995). In contrast, only weak, or absent, leg muscle activity was observed with a pharmacologically reduced muscle tone (e.g. as induced by clonidine; Dietz et al. 1995).

The present observations might indicate that during locomotion after a loss of supraspinal drive, the spinal neuronal circuits can be driven by peripheral afferent input to produce an EMG pattern similar to that obtained in control subjects, still of somewhat smaller amplitude. According to recent observations, such a drive is thought to mainly arise from a combination of load and hip joint-related afferent input (Dietz et al. 2002). Findings in the cat (Bouyer and Rossignol 2003) indicate that the afferent input from the skin around the feet might also contribute. Furthermore, the afferent input has been demonstrated to play an important role in modulating neurotrophin levels in the spinal cord, and hence ultimately affect spinal cord neural plasticity (Gomez-Pinilla et al. 2004).

The degree of co-activation (calculated as temporal proportion of simultaneous supra-threshold EMG activity in TA and GM) was higher in SCI than in control subjects during all walking conditions. Real co-activation, i.e. simultaneous neuronal excitation of both muscles, and/or a slower switch off of EMG activity after an appropriate muscle activation are possible underlying mechanisms. The lower degree of co-activation in the second compared to the first block of walking at 2.0 km/h in SCI subjects might be due to a reduction of spasticity during the walking session (Lünenburger et al. 2005).

Conclusions

The velocity-dependent modulation of locomotor activity observed in motor-complete SCI subjects might be of interest for defining an optimal speed for the locomotor rehabilitation of subjects with incomplete SCI or other neurological gait disorders. Of course, the range of speeds tested in the present study was relatively small and generally low. However, even within the limited range of speeds applied, a velocity-dependent modulation of leg muscle activity was found in SCI subjects which is in line with a recent report (Beres-Jones and Harkema 2004). Higher speeds have been claimed to be favourable for achieving optimal training effects in stroke subjects (Pohl et al. 2002). Our observations in complete SCI concur with the proposal of higher training speeds. However, this factor has to be balanced with the suggestion of a minimal unloading (Dietz et al. 2002; Dietz and Harkema 2004) and with the actual functional ability of a patient.

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