Alternate Leg Movement Amplifies Locomotor-Like Muscle Activity in Spinal Cord Injured Persons

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Kawashima, Noritaka, Daichi Nozaki, Masaki O. Abe, Masami Akai, and Kimitaka Nakazawa. Alternate leg movement amplifies locomotor-like muscle activity in spinal cord injured persons. J Neurophysiol 93: 777–785, 2005. First published September 22, 2004; doi:10.1152/jn.00817.2004. It is now well recognized that muscle activity can be induced even in the paralyzed lower limb muscles of persons with spinal cord injury (SCI) by imposing locomotion-like movements on both of their legs. Although the significant role of the afferent input related to hip joint movement and body load has been emphasized considerably in previous studies, the contribution of the “alternate” leg movement pattern has not been fully investigated. This study was designed to investigate to what extent the alternate leg movement influenced this “locomotor-like” muscle activity. The knee-locked leg swing movement was imposed on 10 complete SCI subjects using a gait training apparatus. The following three different experimental conditions were adopted: 1) bilateral alternate leg movement, 2) unilateral leg movement, and 3) bilateral synchronous (in-phase) leg movement. In all experimental conditions, the passive leg movement induced EMG activity in the soleus and medial head of the gastrocnemius muscles in all SCI subjects and in the biceps femoris muscle in 8 of 10 SCI subjects. On the other hand, the EMG activity was not observed in the tibialis anterior and rectus femoris muscles. The EMG level of these activated muscles, as quantified by integrating the rectified EMG activity recorded from the right leg, was significantly larger for bilateral alternate leg movement than for unilateral and bilateral synchronous movements, although the right hip and ankle joint movements were identical in all experimental conditions. In addition, the difference in the pattern of the load applied to the leg among conditions was unable to explain the enhancement of the EMG activity induced in SCI persons using a gait-training apparatus. The knee-locked leg swing movement pattern should contribute to the generation and/or shaping of the rhythmic output pattern from the CPG has been considerably emphasized by previous studies using reduced animal preparations (Duyens and Pearson 1980; Grillner 1985; Pearson 1995), and these findings also support the view that the muscle activity induced in SCI persons reflects the output from the locomotory CPG.

However, it remains unclear to what extent the induced muscle activity (locomotor-like muscle activity) is actually “locomotory,” partly because almost all of the previous studies have not paid attention to a substantial feature of human bipedal locomotion, i.e., alternating leg movements. This is the main point that we focused on in this study. To investigate the significance of alternating leg movements to locomotor-like muscle activity, we compared the magnitude of the EMG activity induced in the complete SCI subjects using a gait-training apparatus in the following different conditions: 1) ordinary bilateral alternating leg movement, 2) unilateral leg movement, and 3) bilateral synchronous (in-phase) leg movement. If the spinal CPG is actually involved in the neural mechanism of the locomotor-like muscle activity, the alternating leg movement pattern should contribute to the generation and/or coordination of the muscle activity.

Part of this study has been presented in abstract form (Kawashima et al. 2003).

METHODS

Participants

Ten male SCI persons (28.8 ± 6.0 yr) participated in this study. All of the subjects had an injury at the thoracic (T) level of the spine.

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spinal cord somewhere between T5 and T12, complete motor paralysis in their lower limb muscles (American Spinal Injury Association Class: ASIA A or B; Maynard et al. 1997), and moderate degrees of spasticity. At least one-half a year had passed since they were injured. The physical characteristics of the subjects are summarized in Table 1. Each subject gave written informed consent to the experimental procedures, which were conducted in accord with the Helsinki Declaration of 1975 and approved by the ethics committee of the National Rehabilitation Center for Persons with Disabilities, Tokorozawa, Japan.

Passive leg movement apparatus

To impose locomotion-like movements on their legs, we used an apparatus (Fig. 1A) originally developed for physical exercise for persons with disabilities (Easy Stand Glider 6000, Altimate Medical). This apparatus enables the SCI persons to stand securely by immobilizing their trunk and pelvis using front and back pads and by preventing hyperextension of the knee joint using the knee pad. It also enables the subjects to swing their legs by moving the handle connected to the foot plate. In this study, the experimenter manually moved the handle back and forth (± 17.5 cm from default position) in a sinusoidal manner (Fig. 1C) by matching the movement frequency with the sound of a metronome (1 Hz). This handle movement could induce approximately ±14 and ±9° motion in the hip and ankle joints, respectively (these values of range of joint motion depend on the subject’s lower limb length). This range of motion of each joint is similar to the data for the normal walking provided by Winter (1990). Although only reciprocal leg movement can be induced at the default setting, synchronous or unilateral leg movement can be induced by removing the bolts that connect bars to both sides.

Experimental protocol

Before the experiment, we checked that the standing posture was stable and that no hypotension was observed. First, bilateral alternate leg movement was imposed for 3 min so that the subjects could experience the standing posture and the imposed leg movement. Then, the experiments were performed under the following three conditions: 1) bilateral alternate (anti-phase) leg movement; 2) unilateral leg movement; and 3) bilateral synchronous (in-phase) leg movement. In the unilateral leg movement, the right leg was moved while the position of the left leg was fixed to be vertical. In the bilateral synchronous leg movement, both legs were passively swung simultaneously in the same direction. Throughout the exercise period, subjects were asked to grasp the bar in front of them and to keep their upper limbs relaxed (Fig. 1A). The experimenters had conducted a sufficient number of practices before the testing session so that they could adjust the leg motion to the predetermined pattern (i.e., the range of motion and swing frequency) under all experimental conditions by monitoring the angle data from an electrogoniometer displayed on an oscilloscope. The duration of each session was 1 min, and an interval of ≥1 min was taken for rest between sessions. The order of conditions 1), 2), and 3) was randomized.

Table 1. Characteristics of the SCI subjects

<table>
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Fig. 1. A: overview of the experimental setup. During the experiment, the spinal cord injury (SCI) subject stood securely on a special device (Easy Stand Glider 6000, Altimate Medical). The subject’s legs were moved passively by moving the lever back-and-forth in a sinusoidal manner in pace with the tempo of a metronome (1 Hz). B: load applied to the foot soles of each leg was measured by 4 load cells placed under the stainless foot plate. C: typical example of hip and ankle joint motion and induced EMG activity of an SCI subject (level of lesion, T12) during passive leg movement (Sol; soleus, Gas; medial head of gastrocnemius, TA; tibialis anterior, RF; rectus femoris, BF; biceps femoris). To quantify level of muscle activity, the averaged EMG signal was calculated (right). D: adaptation-like phenomenon in EMG activity in the Sol. It took ~30 s for the EMG activity to reach steady state. Therefore data from the 1st 30 s were disregarded in the calculation of the averaged EMG signals (C).

Data recording

The surface EMG signal was recorded from the soleus (Sol), the medial head of the gastrocnemius (Gas), the tibialis anterior (TA), the rectus femoris (RF), and the long head of the biceps femoris (BF) muscles of both legs with the use of a bipolar electrode. Care was taken to exclude any artifact in the EMG signal (e.g., the skin was washed with a scrub gel and rubbed with sandpaper to reduce
the resistance of the skin). The EMG signal was amplified (Bag- 

noli-8 EMG System, DELSYS) with band-pass filtering between 20 and 450 Hz. Ankle joint motion was recorded with an electro- 
goniometer (Goniometer System, Biometrics), whose two sensor heads were placed on the lateral part of the shank and foot of the subject (Fig. 1A). Hip joint motion was estimated from the data recorded by using another goniometer attached to the lateral aspect of the apparatus (Fig. 1A).

In six subjects, the VICON 370 system (Oxford Metrics) was used to analyze the lower limb motion more accurately. Eight markers were attached to the right and left sides of the subject on the skin overlying the following landmarks: the acrominon (SHO), greater trochanter (GTR), lateral malleolus (AKL), and the top of the great toe (TOE). We defined the hip and ankle joint angles as the angles formed by the SHO, GTR, and AKL and by the GTR, AKL, and TOE, respectively. Furthermore, in these subjects, the actual load applied to each foot sole was measured using four load cells (LMA-A-1KN, Kyowa, Tokyo, Japan) placed under the four corners of the stainless foot plate (Fig. 1B). During the experiment, all data were continuously monitored by Power Lab software (Chart version 4, AD instruments) and were digitized at 1 kHz for later analysis.

**Data analysis**

The digitized EMG signal was full-wave rectified after the DC component was subtracted. It was then averaged over the last 30 locomotion cycles (Fig. 1C). The data of the first 30 cycles were discarded, because the EMG activity often showed gradual decay, and it took ~30 s (i.e., 30 cycles) to become stationary (Fig. 1D). The locomotor-like EMG activity was quantified using the integrated value of the averaged EMG signal and the duration over which the muscle was active (Fig. 1C). We regarded the muscle to be active when its averaged EMG signal consistently exceeded the level of resting EMG activity (mean value + 3 × SD). Furthermore, to examine the phase-dependent changes in the EMG activity, the averaged EMG signal was divided into 10 bins, and the mean amplitude in each bin was calculated. The ranges of hip and ankle movements were calculated from the data obtained by electrogoniometers, and those were compared with the VICON data. The load applied on each foot sole was quantified by calculating the summation of the data from four load cells.

**Statistics**

Values are given as means ± SE. Two-way ANOVA was used to test the difference in the EMG magnitude, duration, and hip and ankle joint range of motion among the three conditions. Tukey’s post hoc test was applied to identify differences among the conditions. Significance was accepted at $P < 0.05$.

**RESULTS**

**Pattern of the locomotor-like EMG activity**

Figure 2A shows the averaged waveform of the joint angle (estimated by electrogoniometers) and the EMG activity obtained from an SCI subject during alternate leg movement. In this subject, EMG bursts modulated with the locomotion cycle were observed in Sol, Gas, and BF. A similar muscle activation pattern was observed in other subjects. Figure 2B indicates the number of subjects whose muscle activity was judged to be significant in each of 10 leg movement phases. For all subjects, the EMG activity was observed in Sol and Gas during the backward leg swing phase corresponding to the stance phase in normal locomotion. Similarly, the EMG activity was observed...
in BF for 8 of 10 subjects during the hip-flexion phase corresponding to the swing phase in normal locomotion. Namely, the active phase of these muscles mainly corresponded with the phase during which they were mechanically stretched. The EMG activity of the TA was observed for two subjects, and no EMG activity was induced in the RF. In the results and discussion sections, we will focus only on these activated muscles (Sol, Gas, and BF).

Typical averaged waveforms of the EMG activity for three experimental conditions obtained from two subjects are shown in Fig. 3 (A and D, bilateral alternate; B and E, unilateral; C and F, bilateral synchronous leg movements). As clearly shown in these waveforms, the amount of EMG activity varied from condition to condition. In the unilateral leg movement (Fig. 3, B and E), no EMG activity was observed in the nonmoving left leg. The magnitude of the EMG activity was smaller for the unilateral leg movement condition (Fig. 3, B and E) than for the ordinary bilateral alternate leg movement condition (Fig. 3, A and D). In the bilateral synchronous leg movement condition, the EMG activity was present for both legs (Fig. 3, C and F); however, its magnitude was smaller than that for the bilateral alternate leg movement condition (Fig. 3, A and C).

**Leg motions and load to foot sole**

Figure 4A shows a typical example of the hip and ankle joint angle movements obtained using the VICON system. In the right (experimental) leg, both the hip and ankle joint angles moved in a similar manner among three conditions. On the other hand, the left leg movement was completely out of phase between the alternate and synchronous leg movement conditions, and no obvious hip and ankle motion was observed during the unilateral leg movement condition. There was no significant difference in the range of motion of each joint among three conditions for the right leg and between the alternate and synchronous leg movement conditions for the left leg (Fig. 4B). In the unilateral leg movement condition, the left leg movement was kept at almost zero (Fig. 4B). It should be noted that the data in Fig. 4B contain the data measured with electrogoniometers, because the joint angle movement estimated using electrogoniometers was not different from that measured directly using the VICON system.

Figure 5A shows a typical example of the load applied to the foot sole in the three conditions. The load was modulated almost sinusoidally with the leg movement cycle. The load was maximal and minimal, respectively, when the hip joint was maximally extended and flexed. Although the load averaged over time was not different from condition to condition (Fig. 5B), there was a statistically significant (P < 0.05) difference in the peak-to-peak load among the three experimental conditions (Fig. 5C). In comparison with the alternate leg movement condition, the load applied to the right leg was 85.5 ± 3.8% in the unilateral leg movement condition and 64.3 ± 12.5% in the synchronous leg movement condition. On the other hand, the peak-to-peak load applied to the left leg was 22.5 ± 4.4% in the unilateral leg movement condition and 69.9 ± 11.9% in the synchronous leg movement condition compared with the alternate leg movement condition.
Difference in the induced EMG activity among experimental conditions

Figure 6 summarizes the integrated EMG activity of the Sol, Gas, and BF in three experimental conditions. The integrated EMG activity induced by bilateral alternate leg movement was significantly larger ($P < 0.05$) than that induced during the other conditions. The values of the percentage increase in EMG magnitude induced by alternate leg movement compared with that induced by unilateral leg movement were 291/0.0670, 163/0.16, and 278/0.0671% for Sol, Gas, and BF, respectively.

Figure 7 shows the mean EMG amplitude in each 10% bin of the locomotion cycle (top) and in the period during which the muscle was evaluated to be active (bottom). The amplitude of the Sol EMG activity in the bilateral alternate movement was significantly larger ($P < 0.05$) than that in the unilateral movement from the 30 to 60% cycles, and significantly larger than that in the synchronous movement from the 30 to 70% cycles (Fig. 7A). The duration of the EMG activity of the Sol muscle during alternate leg movement was significantly longer ($P < 0.05$) than that during the other conditions (Fig. 7A). Such an amplifying effect of alternate leg movement on the EMG activity was also observed for the Gas and BF muscles (Fig. 7, B and C).

**DISCUSSION**

These results show that the locomotor-like EMG activity was significantly larger for alternate leg movement than for unilateral and bilateral synchronous movements. In the DISCUSSION section, the neuronal mechanism underlying these results, mainly in the context of what is known about the spinal locomotor system that was revealed in previous animal and human studies, will be addressed.

**Muscle activity induced by passive leg movement**

We used the gait-training apparatus (Fig. 1A) to impose the locomotory movement. However, the leg movement achieved by this apparatus is different from the ordinary stepping movement in the following two ways. First, the knee joint is locked in an extended position throughout the entire locomotion cycle. Second, the sole always touches the foot plate even during the forward leg swing phase. That is, the sensory information from the foot sole exists even in the swing phase, and there is no clear instant that corresponds to “heel contact.” Despite these differences in the movement pattern, the EMG activity was observed in the paralyzed lower limb muscles during the passive leg movement, as was shown during the body weight-supported stepping movement on a treadmill in previous reports (Dietz et al. 1995, 2002; Dobkin et al. 1995; Ferris et al. 2004; Harkema et al. 1997; Ivanenko et al. 2003). This is because several factors that are important to this phenomenon, i.e., hip joint motion (Andersson and Grillner 1983; Grillner and Rossignol 1978) and load information (Dietz and Duysens 2000; Duysens and Pearson 1980), were well preserved, even in our experimental setting. In fact, as for the first difference regarding the knee joint motion, Dietz et al. (2002) have shown...
that the knee-locked stepping movement (hip walking) does not affect the induced muscle activity. The only difference between normal and hip walking was that RF activity was almost absent in hip walking (see Fig. 4 in their study), a finding that agrees with our result (Figs. 3 and 5). The second difference regarding foot contact might influence the load information associated with the ordinary locomotion cycle; however, as shown in Fig. 5, we ensured that the load applied to the leg was periodically changed with the leg motion cycle in our experimental setting. The load was maximal when the hip joint was nearly maximally extended (Fig. 5), and this loading pattern resembled that observed when a stepping movement was imposed on a treadmill (Ferris et al., 2004). It is therefore likely that a considerable portion of the afferent neural inputs during normal walking could be preserved in our experimental setting.

In all subjects, coordinated EMG bursts can be induced by imposing passive leg movement in the lower limb muscle. As shown in Figs. 2 and 7, the phase in which the muscle activity was observed coincided with the phase in which it was mechanically stretched. That is, Sol and Gas were active while the leg swung toward the backward, and BF was active while the leg swung toward the forward. It is therefore possible that the muscle activity was associated with the stretch reflex response. However, these results show that the muscle activity was observed even in the muscle’s shortening phase (Figs. 2 and 7). Concerning this point, Dietz et al. (1998) have also observed that the leg muscle activity is equally distributed during shortening and it seems therefore likely that the locomotor-like muscle activity results from the complex interaction of the afferent inputs and the spinal neural circuits rather than simple stretch reflex.

**Contribution of alternate leg movement**

One of the most substantial features of human bipedal locomotion is alternating leg movement. Therefore investigating how such an alternate leg movement pattern affects the amount of locomotor-like EMG activity would give us important information, especially regarding the problem of whether the activity is actually “locomotory” or not. A relevant approach has been partly taken by Ferris et al. (2004). They found that muscle activity could be induced for complete SCI patients even in the nonmoving leg when the stepping movement was imposed only on the other leg. Their results have provided evidence that the human spinal cord has a mechanism to efficiently realize alternating leg movement. However, we did not observe any muscle activity in the nonmoving left leg (Fig. 3). This result was similar to the results of the study by Dietz et al. (2002), who ascribed the contradiction with the work of Ferris et al. (2004) to the difference in the speed of stepping and the amount of the load (Dietz and Harkema 2004). Likewise, one of the possible reasons for the contradiction between the results of Ferris et al. (2004) and our results is the difference in the load pattern on the nonmoving leg. In this study, the load was tonically applied and the amount of modulation was small (Fig. 5), while in their study, a load pattern resembling normal stepping was applied.

On the basis of the absence of muscle activity in the nonstepping leg, Dietz et al. (2002) referred the possibility that the interlimb coordination observed in normal subjects requires the supraspinal systems. Concerning this point, a recent study revealed that the interlimb coordination includes the activity of the supplemental motor cortex area (Debaere et al. 2001). However, our data have provided strong evidence that the spinal cord has an ability to coordinate the movement of both legs. Figure 8 shows the relationships between Sol EMG activity and ankle ROM (left), hip ROM (center), and the peak-to-peak load (right) on the right foot sole. The EMG level was significantly larger for locomotion-like alternate leg movement than for unilateral and bilateral synchronous movements, although the hip and ankle joint movements were kept identical in all experimental conditions. This result also indicates that the stretch reflex alone is insufficient to explain the modulation of the EMG activity. If the EMG activity were merely a response to the rhythmic muscle-tendon stretches, the level of muscle activity should have been independent of the contralateral leg movement.

One remaining concern is the difference among the three conditions in the load applied to the right leg (Fig. 8, right), because the load-related afferent inputs, such as proprioceptive inputs from the extensor muscle and the sole of the foot, are known to influence the magnitude of the EMG activity.
Therefore the larger EMG activity in the alternate leg movement condition could simply result from the load on the right leg having larger peak-to-peak amplitude. However, this is unlikely because the distribution of the Sol EMG activity with respect to the peak-to-peak amplitude of the load is distinctly different from other two conditions (Fig. 8, right). Therefore it is difficult to explain such a drastic enhancement of Sol EMG activity based only on the difference in load. In addition, although the peak-to-peak load was larger in the unilateral condition than in the synchronous condition, the Sol activity was almost similar between these two conditions (Fig. 8, A and C, right) and even smaller for the unilateral condition in subject S2 (Fig. 8B, right), suggesting that the Sol activity does not depend only on the load modulation.

Therefore our results strongly suggest that the afferent input from the contralateral leg plays a substantial role in amplifying the induced locomotor-like muscle activity in the lower limb. In particular, the contralateral leg movement has to be out of phase so that the muscle activity of the ipsilateral leg is well amplified. That is, the alternate leg movement should be added to the recipes for generating locomotor-like muscle activity that have been previously suggested, such as hip joint motion and the load applied to the lower limbs (Pearson 1995).

In summary, this study was designed to investigate to what extent the alternate leg movement influences the locomotor-like EMG activity in the lower limbs of SCI subjects. These results indicated that the alternate leg movements play a substantial role in amplifying the induced muscle activity, and not only suggest the existence of neuronal circuits enabling interlimb coordination within the spinal cord, but might reinforce the interpretation that the muscle activity induced by passive stepping movement is actually locomotory.

Interlimb coordination generated within the spinal cord

Previous animal studies, using a variety of preparations, indicate that basic neuronal circuits that generate the locomotive motor output exist in the lumbar level of the spinal cord (Forssberg et al. 1980; Pearson and Rossignol 1991; for a review, see Duysens and Van de Crommert 1998). Such neuronal circuits can operate in the absence of any afferent input (Grillner 1985), whereas the significance of the interaction of such a spinal neuronal circuit with the afferent input has also been pointed out (Duysens and Pearson 1980; Pearson 1995). Recent human studies have shown that the afferent signal from one limb affects the muscle activity of the contralateral limb in locomotory movement in a functional way (Pang and Yang 2001; Ting et al. 2000). However, since these studies were conducted in infants (Pang and Yang 2001) or in healthy subjects (Ting et al. 2000), the supraspinal system’s contribution remains unclear. Although the supraspinal system such as the supplementary motor area might contribute to the interlimb coordination (Debaere et al. 2001), these results indicate that some mechanism coordinating the alternate leg movement might exist within the human spinal cord itself. The precise mechanism(s) are unknown at this stage, but it is possible that the neuronal circuits associated with our results have a common origin in the crossed flexor/extensor reflex (Duysens and Loeb 1980; Duysens et al. 1991). Further research is needed to clarify this point.
REFERENCES


FIG. 8. Relationships between the right soleus EMG magnitude and ROM of the hip joint, ROM of the ankle joint, and peak-to-peak amplitude of load applied to the right leg. A: subject S1. B: subject S2. C: mean data (n = 6). In A and B, each point corresponds to the value obtained in each cycle. Alt, Uni, and Syn indicate the alternate, unilateral, and synchronous leg movement conditions, respectively. In C, error bars indicate SE.


