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

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. 1*B*). 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. 1*C*). 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. 1*D*). 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. 1*C*). 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 2*A* 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 2*B* 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

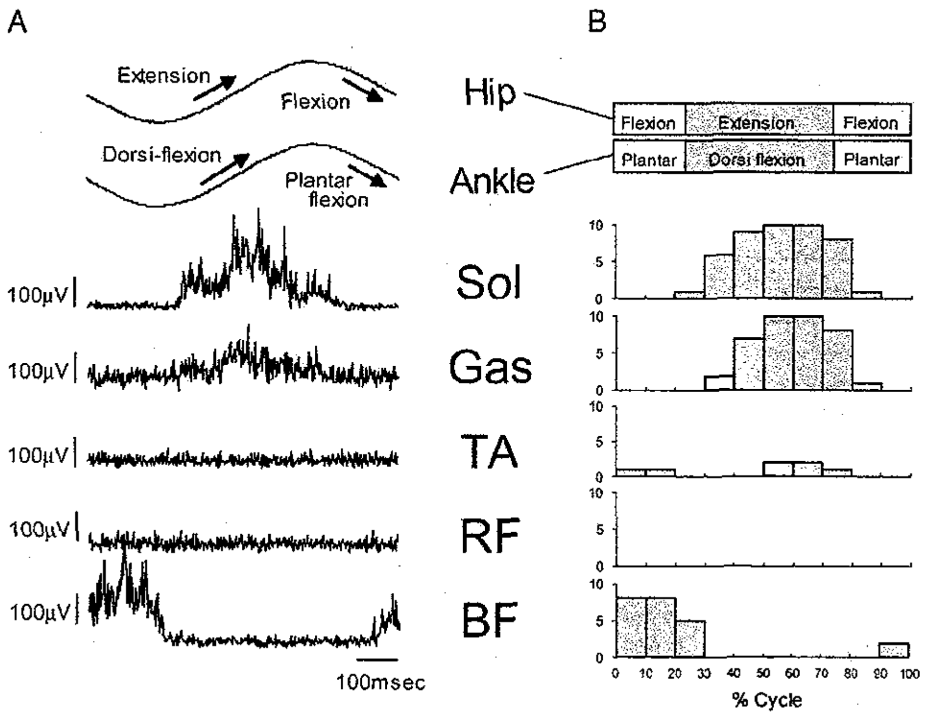

FIG. 2. *A*: ensemble averaged waveform of hip and ankle joint motion and induced EMG activity in each lower limb muscle obtained from an SCI patient. *B*: probability of the occurrence of EMG activity in the lower limb muscles during passive leg movement. Data show number of subjects who showed EMG activity in each leg movement cycle (total number of subjects is 10). Muscle activity was determined to be active when its averaged EMG signal consistently exceeded the level of resting EMG activity (mean value + 3 × SD).

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 4*A* 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. 4*B*). In the unilateral leg movement condition, the left leg movement was kept at almost zero (Fig. 4*B*). It should be noted that the data in Fig. 4*B* 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 5*A* 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. 5*B*), there was a statistically significant ($P < 0.05$) difference in the peak-to-peak load among the three experimental conditions (Fig. 5*C*). In comparison with the alternate leg movement condition, the load applied to the right leg was $85.5 \pm 3.8\%$ in the unilateral leg movement condition and $64.3 \pm 12.5\%$ in the synchronous leg movement condition. On the other hand, the peak-to-peak load applied to the left leg was $22.5 \pm 4.4\%$ in the unilateral leg movement condition and $69.9 \pm 11.9\%$ in the synchronous leg movement condition compared with the alternate leg movement condition.

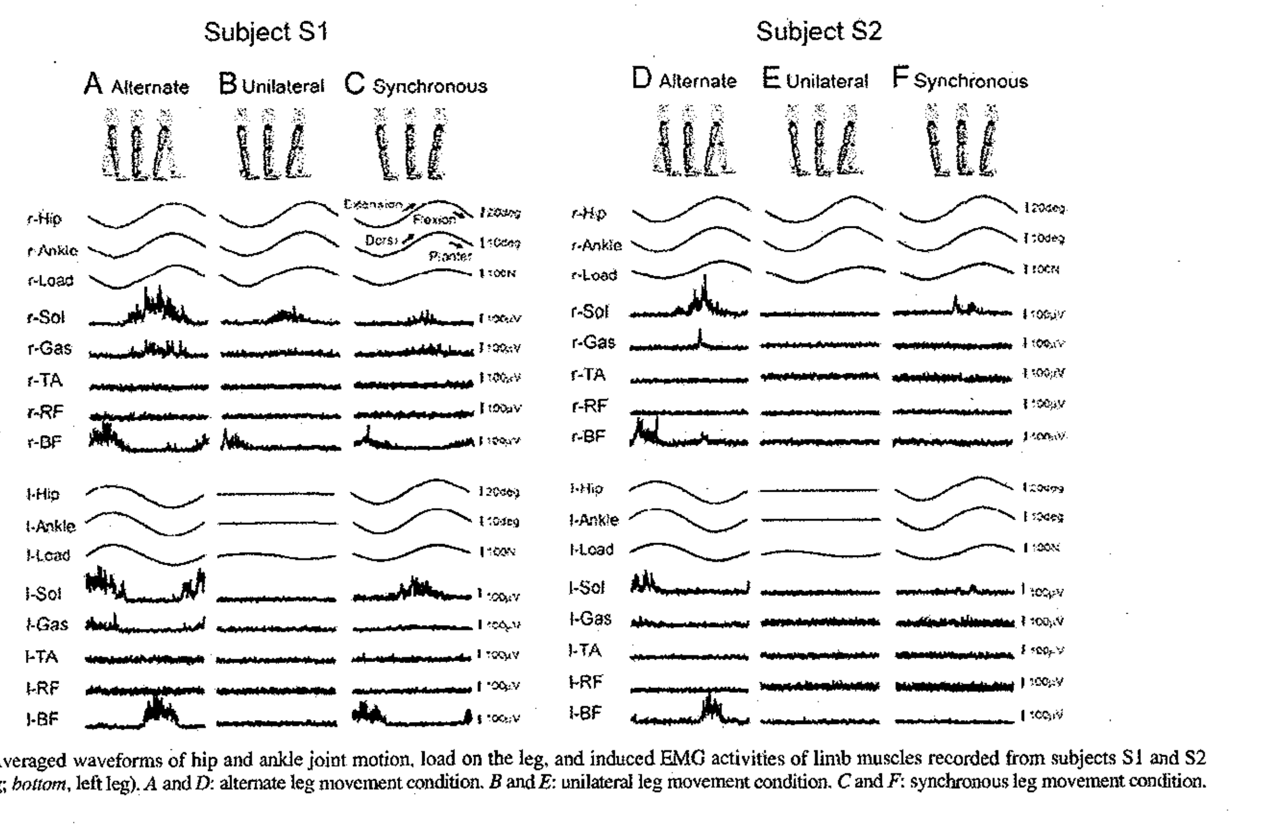

FIG. 3. Averaged waveforms of hip and ankle joint motion, load on the leg, and induced EMG activities of limb muscles recorded from subjects S1 and S2 (*top*, right leg; *bottom*, left leg). *A* and *D*: alternate leg movement condition. *B* and *E*: unilateral leg movement condition. *C* and *F*: synchronous leg movement condition.

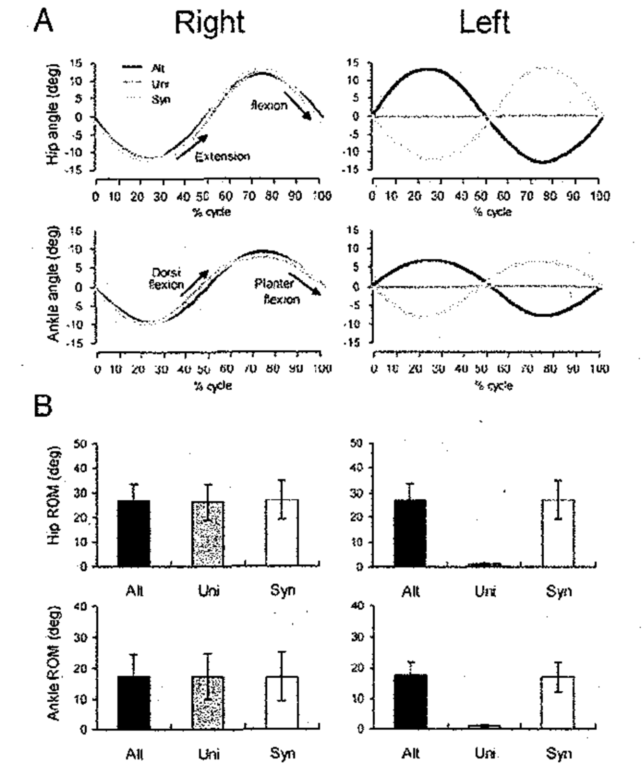

FIG. 4. Kinematical characteristics of imposed leg movement. *A*: changes of hip (*top*) and ankle (*bottom*) joint angular motions in 1 cycle of leg movement recorded from 1 subject using the VICON system. *B*: mean value of range of motion (ROM) of the hip (*top*) and ankle (*bottom*) joint ($n = 10$). Alt, Uni, and Syn indicate the alternate, unilateral, and synchronous leg movement conditions, respectively. Error bars indicate SE.

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 ± 70, 163 ± 16, and 278 ± 71% 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. 7*A*). 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. 7*A*). 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. 1*A*) 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

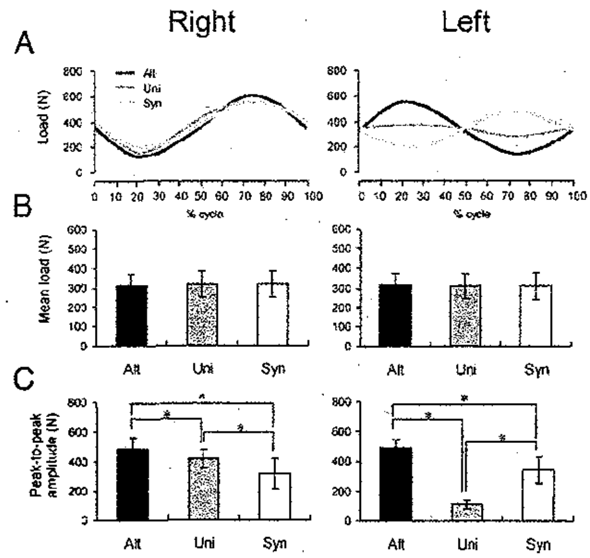

FIG. 5. Characteristics of load change during passive leg movement. *A*: changes of load applied to the foot sole of each leg in 1 cycle of leg movement. *B*: mean value of load averaged over 1 cycle ($n = 6$). *C*: mean value of peak-to-peak amplitude of load ($n = 6$). Alt, Uni, and Syn indicate the alternate, unilateral, and synchronous leg movement conditions, respectively. Error bars indicate SE. *Significant difference ($P < 0.05$).

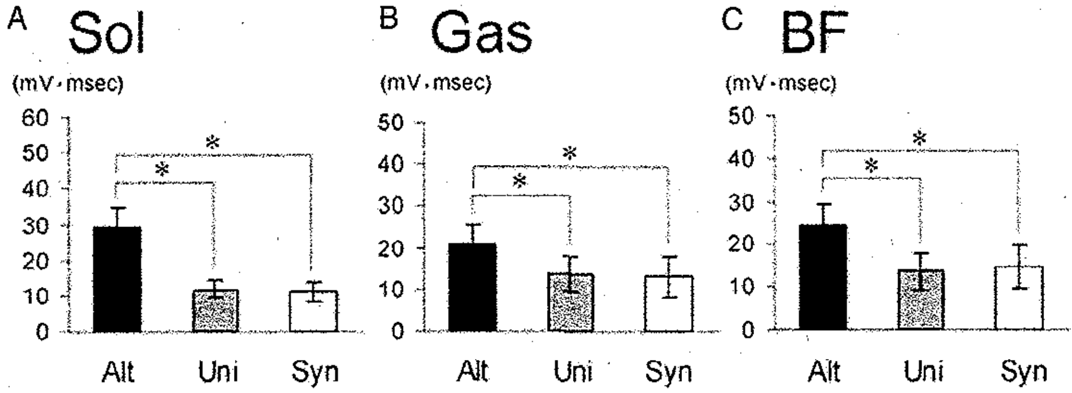

FIG. 6. Effect of leg movement pattern on muscle activity. Mean values of integrated rectified EMG induced by 3 types of leg movement. *A*: soleus (Sol). *B*: gastrocnemius (Gas). *C*: biceps femoris. Induced activity level of all 3 muscles was significantly larger ($P < 0.05$) during the alternate leg movement (Alt) than the unilateral (Uni) and synchronous (Syn) leg movement conditions. Error bars indicate SE. *Significant difference ($P < 0.05$).

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

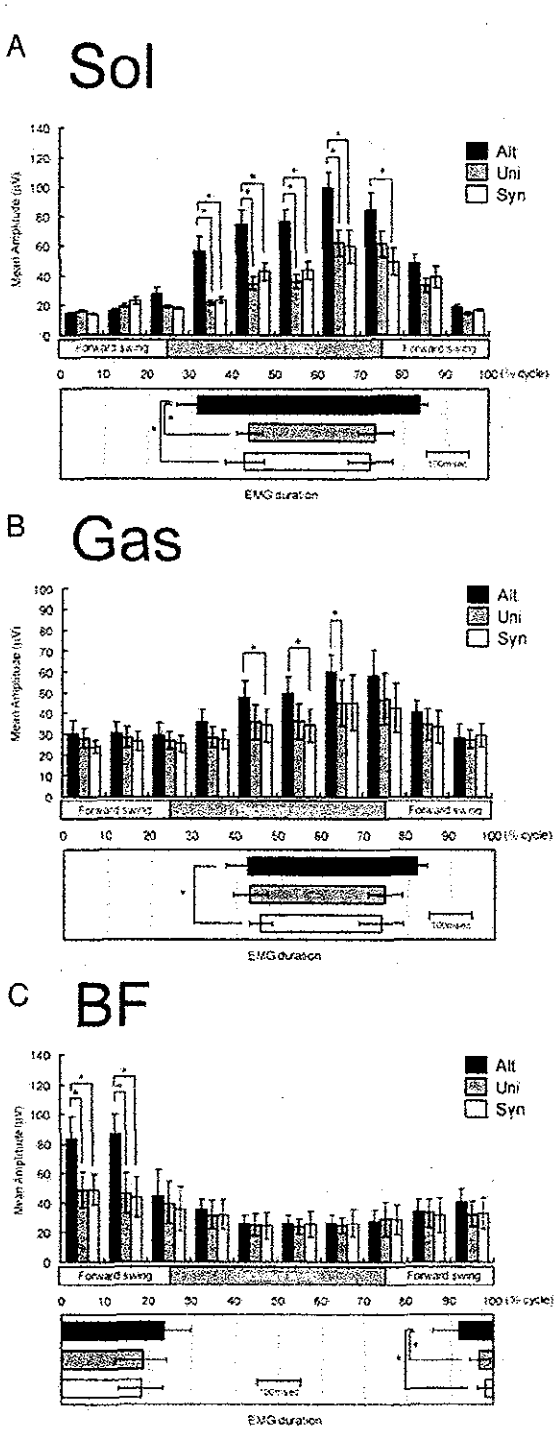

FIG. 7. Comparison of mean EMG amplitude in each 10% cycle bin (*top*) and duration of EMG activity (*bottom*) among the 3 experimental conditions. *A*: soleus (Sol). *B*: gastrocnemius (Gas). *C*: biceps femoris (BF). Alt, Uni, and Syn indicate the alternate, unilateral, and synchronous leg movement conditions, respectively. Error bars indicate SE. *Significant difference ($P < 0.05$).

(Harkema et al. 1997; Kojima et al. 1999). 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. 8*B*, *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).

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.

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.

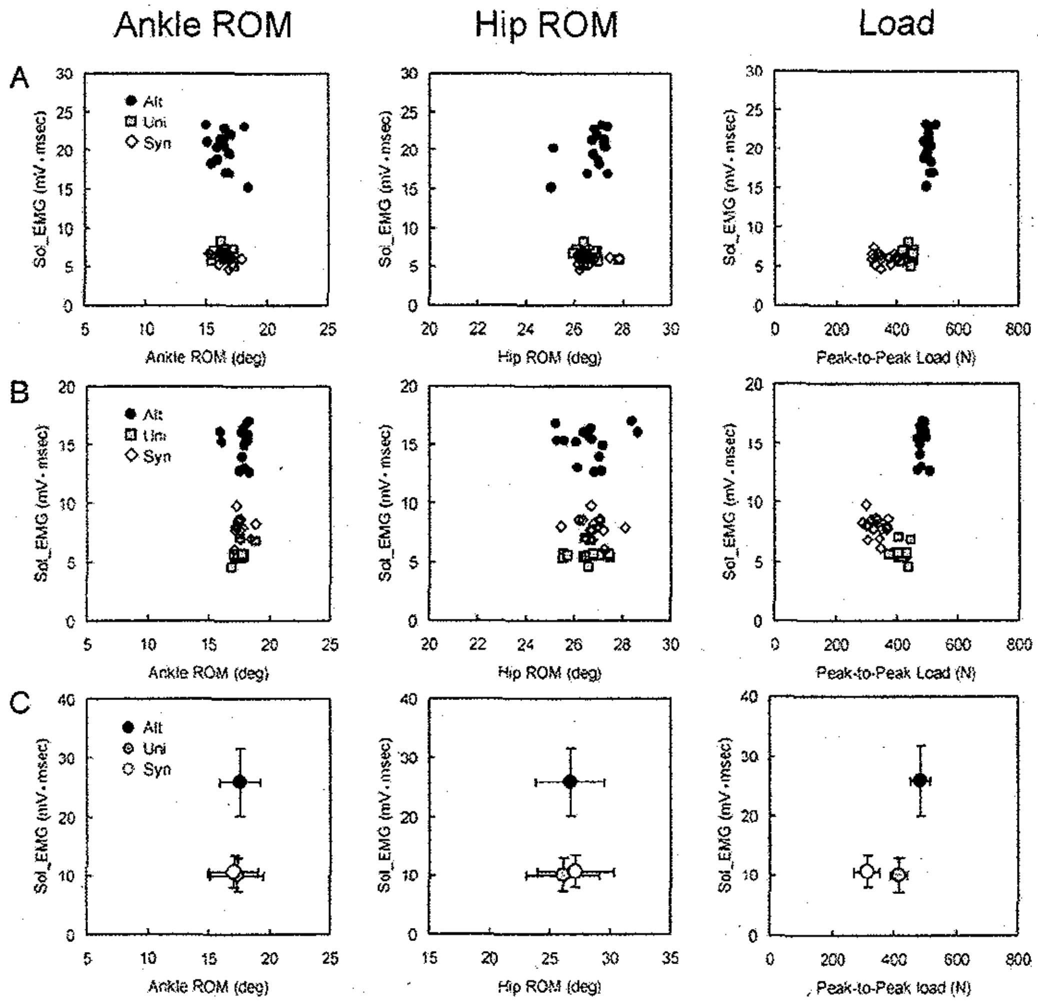

FIG. 8. Relationships between the right soleus EMG magnitude and ROM of the hip joint, ROM of the ankle joint, and peak-to-peak amplitude of load applied to the right leg. *A*: subject S1. *B*: subject S2. *C*: mean data ($n = 6$). In *A* and *B*, each point corresponds to the value obtained in each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

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Muscle Oxygenation of the Paralyzed Lower Limb in Spinal Cord–Injured Persons

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ABSTRACT

KAWASHIMA, N., K. NAKAZAWA, and M. AKAI. Muscle Oxygenation of the Paralyzed Lower Limb in Spinal Cord–Injured Persons. *Med. Sci. Sports Exerc.*, Vol. 37, No. 6, pp. 000–000, 2005. **Purpose:** Even in the paralyzed lower limb muscle, EMG activity can be induced by imposing passive leg movement in standing posture in persons with spinal cord injury (SCI). The purpose of the present study was to ascertain whether the oxygenation level of the paralyzed lower limb muscle covaried with the muscle EMG activity during imposed passive leg movement. **Methods:** Six motor-complete SCI subjects and four neurologically normal controls were placed on a gait-training apparatus that enabled the SCI subjects to stand and move their legs passively. After a 1-min resting stage, consecutive passive alternate leg movements were performed at different frequencies (0.8, 1, 1.2, and 1 Hz, for 3 min at each stage). To obtain postexercise data, subjects were kept in a standing posture for 5 min after passive movement ceased. The EMG activity and concentration changes in the oxygenated (oxy-) and deoxygenated hemoglobin (Hb) (deoxy-Hb) were continuously measured using near-infrared spectroscopy (NIRS) from the gastrocnemius muscle. **Results:** In all SCI subjects, muscle EMG activity was observed during passive leg movement. The oxy-Hb level gradually increased, whereas the deoxy-Hb decreased, and these changes were independent of the total Hb changes. In the recovery stage, the total Hb level was found to exceed the preexercise level. In contrast to the SCI patients, the normal subjects showed neither EMG activity nor changes in oxy- or deoxy-Hb. **Conclusion:** The present results demonstrate that passive leg movement can induce not only muscular activity but also alteration of muscle oxygenation level in the paralyzed lower leg. Particularly, induced muscular activity seems to correlate with increased perfusion of the muscle. **Key Words:** SPINAL CORD INJURY, PARALYZED MUSCLE, OXYGENATION LEVEL, NEAR INFRARED SPECTROSCOPY, REFLEXIVE MUSCLE CONTRACTION, PASSIVE MOVEMENT

Previous studies have indicated that spinal cord injury (SCI) leads to extreme muscle atrophy (7,19), fiber type transformation toward fast-fatigable fibers (13,20), and lower bone mineral density (BMD) (11,33). This musculoskeletal degeneration can be attributed largely to the dramatic reduction of muscular activity and mechanical stress in the paralyzed limbs, which is due primarily to the motor paralysis following SCI. Furthermore, long-term immobilization of the paralyzed limb may bring about vascular effects such as reduction in vessel diameter (4,26,27), and changes in muscle blood flow (24) and vascular compliance (17,26,27). Because chronic inactivity and hypocirculation of the paralyzed area are especially crucial factors in cardiovascular-related complications such as pressure sores and deep venous thrombosis (5), enhancement of the metabolism and circulation in the paralyzed area is particularly important in preventing these problems.

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It is now well recognized that, even in the paralyzed muscles of SCI patients, locomotion-like muscle activity can be induced by imposing stepping movement on a treadmill (8–10). Induced muscle activity is believed to have the potential to prevent degeneration of the musculoskeletal system in SCI patients. From the perspective of muscle metabolism, an important issue is whether the muscular activity induced by imposed passive leg movement is accompanied by alterations in the oxygenation level and/or circulation in the paralyzed area. The present study was designed to address this question by simultaneously recording the EMG activity and the muscle oxygenation using near-infrared spectroscopy (NIRS). NIRS, a noninvasive and reliable technique for measuring oxygenation and hemodynamics in tissue, is based on the principle that the near-infrared light absorption properties of hemoglobin (Hb) and myoglobin (Mb) depend on their O_2 saturations. Recently, NIRS has been applied in clinical fields to mea-

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sure metabolic and circulatory patterns in a variety of diseases, and is recognized to be a useful method for identifying impairment of muscle metabolism (for a review, see Boushel et al. (6)).

The purpose of the present study was to ascertain whether the oxygenation level of the paralyzed lower limb muscle changed with muscle EMG activity during imposed passive leg movements. If muscle oxygenation and circulation can be facilitated by imposing passive movement, it may have significant ramifications for rehabilitation in cases of SCI, and especially in the prevention of secondary impairment following SCI. In the present study, we hypothesized that the muscle oxygenation level should change with the appearance of EMG activity in the paralyzed lower limb muscle.

METHODS

Participants

Six men with SCI (26.4 ± 4.4 yr) and four neurologically normal subjects (25.3 ± 2.4 yr) participated in the present study. All SCI patients had traumatic SCI at the thoracic level (between T4 and T12) and had complete paralysis of their lower limb muscles (American Spinal Injury Association (ASIA) Class A or B) (22) with moderate spasticity. Their postinjury time was longer than 6 months. The physical characteristics of the subjects are summarized in Table
T1 1. The subjects gave their written informed consent for 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 the Disabled, Tokorozawa, Japan.

Experimental Procedure

Passive leg movement. To impose locomotion-like
F1 movement on the legs, we used an apparatus (Fig. 1A) developed for the physical exercise of persons with disabilities (Easy Stand Glider 6000, Altimate Medical, Inc., Morton, MN). This apparatus enables SCI subjects to stand securely by immobilizing their trunk and pelvis using front and back pads, and by preventing hyperextension of the knee joint using a kneepad. It also enables them to swing their legs by moving a handle connected to a foot plate. In the present study, the experimenter manually moved the handle back and forth in a sinusoidal manner.

TABLE 1. Characteristics of the SCI subjects.

Group	Subject	Age (yr)	Weight (kg)	Lesion Level	ASIA Grade	Duration of Paraplegia (months)
SCI	S1	24	75	T12	A	26
	S2	21	60	T12	B	25
	S3	30	74	T8	A	14
	S4	19	53	T5	B	26
	S5	39	67	T12	A	15
	S6	32	68	T12	A	34
	Mean	27.5	66.2			23.8
	SD	7.56	8.42			7.58
Normal	Mean	26.4	64.2			
	SD	4.54	5.43			

Protocol. Subjects were asked to abstain from alcohol and caffeine for at least 12 h before the experiment. The subjects were placed in the device and held in standing posture. We verified that the standing posture was stable and that there was no hypotension. We had initially planned to measure postexercise data for 10 min, but some subjects showed orthostatic hypotension for 8 or 9 min after the cessation of the exercise in the preliminary experiment. We therefore set the duration of postexercise measurement at 5 min. After a 1-min resting stage, consecutive passive movements were performed for 3 min at each of the following frequencies: 0.8, 1, 1.2, and 1 Hz. This protocol was used to examine whether EMG activity and muscle oxygenation are dependent on the frequency of passive movement. The 1-Hz movement was repeated to examine time-dependent changes of muscle activity and oxygenation. During the movement, the hip joint range of motion was set at 40. The experimenter manipulated the lever, keeping pace with the rhythm of a metronome. The experimenter had conducted a sufficient number of practices before the resting session so that they could adjust the leg motion to the predetermined pattern (i.e., the range of motion and swing frequency) by monitoring the angle data from an electrogoniometer displayed on an oscilloscope. Since our aim was to estimate the muscle oxygenation due to EMG activity in the paralyzed muscle, the subjects were asked to relax their upper limbs.

Near-infrared spectroscopy. During passive leg movement, the oxygenation levels of the medial head of the gastrocnemius (MG) muscle were continuously measured by a NIRO-300 (Hamamatsu Photonics, Inc., Hamamatsu, Shizuoka, Japan) with dual-channel near-infrared laser diodes. The NIRS signal has been assumed to reflect the combined absorption of the oxygenation level of Hb and Mb. Though it is impossible to distinguish between Hb and Mb because of identical spectral characteristics, contribution from myoglobin to the overall signal is quite small. Changes in oxygenated- (oxy-) and deoxygenated Hb (deoxy-Hb) were calculated by measuring light attenuation at 775-, 813-, 850-, and 913-nm wavelengths, and were then analyzed with an algorithm incorporating the modified Beer–Lambert law. The NIRS probe was placed on the upper portion of the bellies of the MG muscle, and a calibration procedure was carried out to ascertain whether the range of measurement was within the optimal range. Before the beginning of the passive leg movement, subjects were kept in standing posture on the apparatus until the total Hb value reached a constant level, that is, until the pooling of venous blood was completed. At that time, the concentrations of each Hb value were set at zero. Changes in the Hb values were calculated relative to the resting level, and are represented in micrometers.

Electromyography. The surface EMG signal was recorded from the MG muscle using bipolar electrodes. Because it is impossible to place both the NIRS sensor and the EMG electrode at the same place, they were placed proxi-

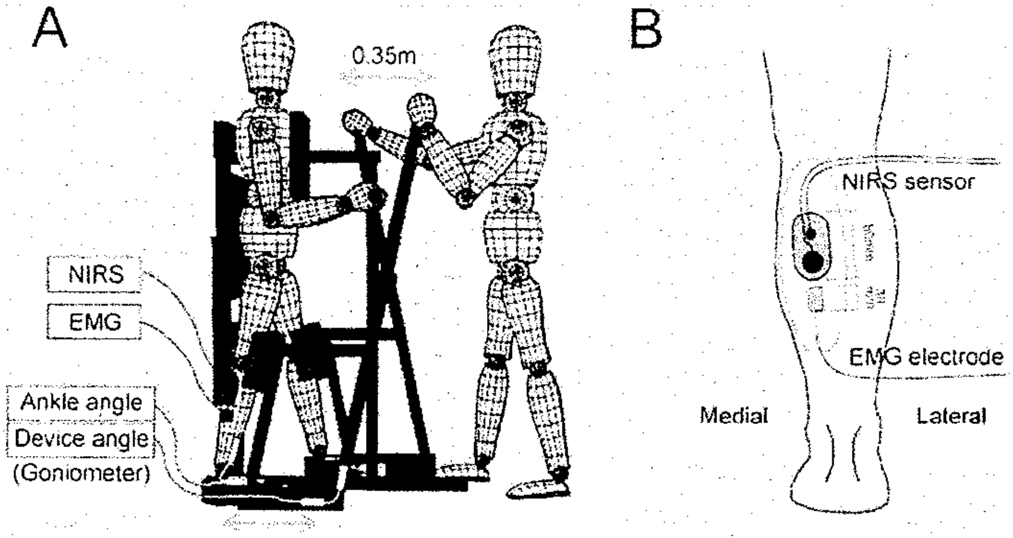

FIGURE 1—A. Experimental setup. This apparatus enables SCI patients to stand securely by immobilizing their trunk and pelvis using front and back pads, and by preventing hyperextension of the knee joint using a kneepad. It also enables them to swing their legs by moving a handle connected to a foot plate. In this study, the experimenter manually moved the handle back and forth in a sinusoidal manner by matching the movement frequency with the sound of a metronome. B. Location of the EMG electrode and the near-infrared spectroscopy (NIRS) sensor. Because it was impossible to place both the NIRS sensor and the EMG electrode at the same place, they were placed proximally and distally on the medial side of the muscle.

mally and distally on the medial side of the muscle (Fig. 1B). The electrode (DE-2.3, DelSys, Inc., Boston, MA) was placed at least 2 cm proximal to the end point of the MG muscle. This electrode has parallel bars (1 cm long and 1 mm wide) spaced 1 cm apart, and is designed with a built-in filter from 20 to 450 Hz. The common mode rejection ratio at 60 Hz is greater than 80 dB. SCI patients tend to have larger impedance in their paralyzed legs, and special care was thus taken to eliminate any artifacts of the EMG recording. The electrodes were attached using double-sided adhesive tape after careful preparation of the skin. The EMG signal was amplified (Bagnoli-8 EMG System, DelSys, Inc.).

Electrogoniometer. In order to ascertain the similarity of the leg motion throughout the exercise session, the angle of the device was recorded by an electrogoniometer (Goniometer System, Biometrics Ltd., Ladysmith, VA) with sensors placed on the lateral aspect of the apparatus.

Heart rate. To confirm whether central circulation is enhanced by imposing the passive leg movement, HR was continuously measured by using an integrated telemetric monitor (HR meter, Polar, Vantage, Finland) in two SCI patients and two normal subjects.

Data Analysis

During the experiment, all data were continuously monitored by PowerLab software (Chart ver. 4, AD Instruments Inc., Milford, MA) and were digitized at 1 kHz for later analysis. For NIRS data, the average value in the last 30 s at each stage and those at 1, 3, and 5 min postexercise were evaluated for each parameter. The EMG signals were full-wave rectified after subtraction of the DC component. The magnitude of the EMG activity was quantified by the mean amplitude and integrated area of the EMG activity during the last 1 min of each stage.

Statistical Analysis

Values are given as means ± SD. Statistical differences in the size of EMG value and each Hb value were tested by ANOVA with repeated measures. Tukey's *post hoc* test was applied to identify differences between the conditions. The statistical software SPSS 11.0 was used to carry out all analyses. Significance was accepted at $P < 0.05$.

RESULTS

Figure 2 shows a typical example of the EMG activity, F2 NIRS values, and leg motion during an experiment in a SCI patient (Fig. 2A) and a normal subject (Fig. 2B). As clearly shown in this figure, there are remarkable differences in both EMG activity and NIRS parameters between the two groups. It was confirmed that the leg motion was maintained within similar range throughout the exercise in both SCI and normal subjects.

EMG activity. During passive movement, all SCI patients showed EMG activity in the gastrocnemius muscle. The active phase of the EMG activity corresponded to the backward phase of leg movement. Despite the fact that the total (integrated) response area increased with the frequency of the movement, there was no remarkable change in amplitude (Fig. 3). F3 Although the movement frequency in both the second and fourth stages was set at 1 Hz, the EMG amplitude in the fourth stage was significantly lower than that in the second stage (second vs fourth: 42.08 ± 3.73 vs 29.66 ± 6.19 μV, $P < 0.05$). In contrast to the SCI patients,

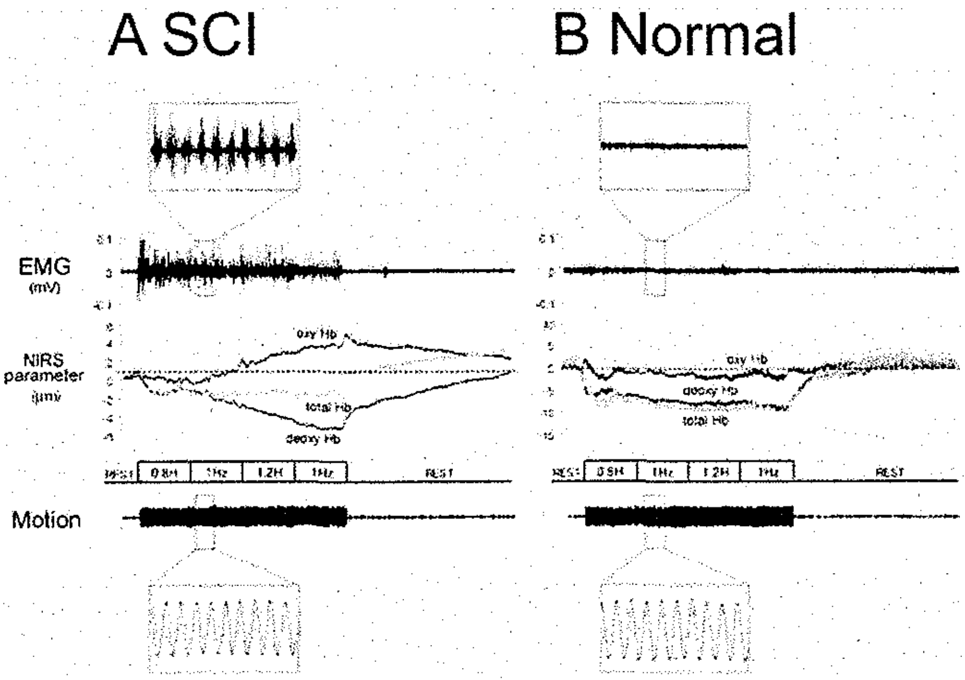

FIGURE 2—Typical example of the EMG activity and concentration changes in each hemoglobin parameter in a patient with motor-complete spinal cord injury (SCI) (A) and a neurologically normal subject (B). Note that there are remarkable differences in both EMG activity and near-infrared spectroscopy (NIRS) parameters between the two subjects. The motion of the apparatus was maintained within a similar range throughout the exercise for both subject groups.

normal subjects showed no visible EMG activity in the gastrocnemius muscle at any time during the exercise.

NIRS parameters. In both the SCI and normal groups, the concentrations of total Hb and deoxy-Hb showed rapid decrements following the onset of the exercise and remained at lower levels compared to the resting value while the legs were passively moved. The degree of the decrease of total Hb in the first stage was much smaller in the SCI group than in the normal group (SCI vs normal: 2.79 ± 0.99 vs 7.04 ± 2.18 μm). During the exercise period, an increase in oxy-Hb and a decrease in deoxy-Hb, which were independent of the changes in total Hb, were observed in the SCI group but not in the normal group (Figs. 2 and 4). In the recovery stage, F4 the total Hb level exceeded the resting value in the SCI group, whereas it merely recovered to the pretest level in the normal group.

Heart rate. Figure 5 shows the change of the HR at rest F5 and during passive leg movement and the recovery period obtained from two SCI patients and two normal subjects. As shown in this figure, HR increased just after the onset of passive leg movement in all subjects.

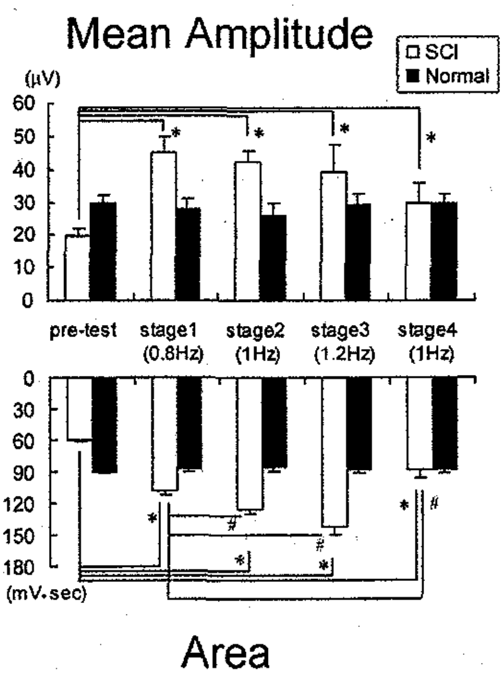

FIGURE 3—Mean amplitude and area of muscle EMG activity in the gastrocnemius muscle. The error bars indicate the SEM value. * Significant difference ($P < 0.05$) compared with the resting value. # Significant difference to the first set value. SCI, spinal cord injury.

DISCUSSION

The present study was designed to examine whether the oxygenation level of the paralyzed muscle is altered with the EMG activity induced by imposed passive leg movement. Our primary observations are the following: (i) during passive movement, all SCI patients showed EMG activity in the gastrocnemius muscle, whereas none of the normal subjects showed such activity; (ii) during the exercise period, an increase in oxy-Hb and a decrease in deoxy-Hb, both of which were independent of changes in total Hb, were observed in the SCI group; and (iii) in the recovery stage, total Hb exceeded the preexercise value in the SCI group. A possible mechanism for these changes in oxygenation level in the SCI patients and its implications for rehabilitation are discussed below.

Muscle activity during passive movement. Despite the motor paralysis in their lower legs, all six SCI subjects showed EMG activity in the paralyzed gastrocnemius muscle during passive movement. On the other hand, no normal subjects showed any EMG activity in the gas-

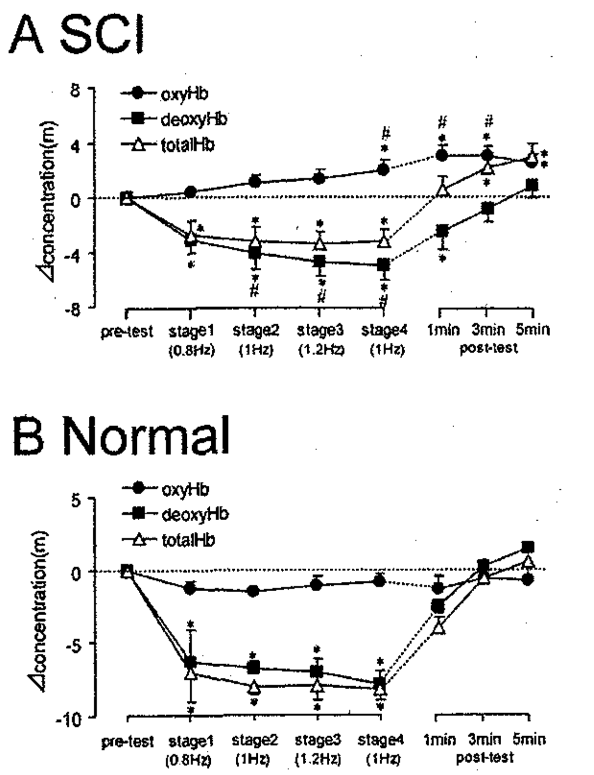

FIGURE 4—Concentration changes in total, oxygenated hemoglobin (oxyHb), and deoxygenated hemoglobin (deoxyHb) throughout the experiment for patients with spinal cord injury (SCI) (A) and normal subjects (B). The error bars indicate the SEM value. * Significant difference ($P < 0.05$) compared with the resting value. # Significant difference to the first set value.

trocnemius muscle, even though the applied leg movements were identical to those applied to the SCI patients. In our previous data, it was found that the passive leg movement can also induce EMG activity in other lower leg muscles, for instance, the soleus and biceps femoris muscles (18). It is possible that the observed muscle activity consisted of complex spinal reflexes rather than simple stretch reflex responses induced by rhythmical stretching of the muscle tendon (14,18), and that the lack of EMG activity in normal subjects can be partly explained by the inhibitory neural input from a higher center to the spinal motor neurons (8). We do not discuss any further details of the neural mechanism of this EMG activity here because this article is concerned primarily with the relationship between the magnitude of muscular activity and the degree of the Hb value. The neural mechanism underlying this EMG activity has been described in detail in the previous research (for a review, see Harkema (14)).

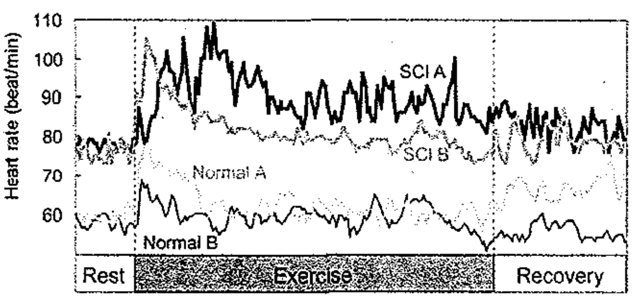

FIGURE 5—HR changes at rest, during passive leg movement, and in the recovery period obtained by two patients with spinal cord injury (SCI) and two normal subjects.

The degree of changes in the NIRS signals should strongly depend on the muscle contraction level. It is therefore important to know how much the muscle activity occurs during passive leg movement. However, it is difficult to evaluate the muscle contraction level using the percentage of the maximal voluntary contraction (%MVC), which is commonly used to normalize and evaluate the muscle contraction level because SCI patients cannot accomplish voluntary contraction. When the EMG activity of the paralyzed muscle is expressed with size relative to the MVC obtained by normal subjects (average: 417.7 ± 43.24 μV), it corresponds with approximately 10% MVC. Given the muscle atrophy of the paralyzed muscle (7,19), it can be assumed that the SCI patients have an MVC lower than that in the normal subjects. Therefore, we estimate the contraction level observed in the SCI patients as no less than 10% MVC.

Changes in Hb concentration during exercise. In the present study, both the SCI and normal groups showed a rapid decrease in the total Hb concentration following the onset of exercise and maintained the lower value while the legs were passively moved. We considered venous blood in the calf to be complete, that is, to have reached plateau level, at the beginning of the passive leg movement because subjects were kept in standing posture until the total Hb value stabilized. By imposing passive leg movement, the pooling venous blood might be expelled from the calf because intramuscular pressure is increased due to the imposed length changes in the muscle (28,30), irrespective of the appearance of EMG activity. Therefore, it seems reasonable to assume that the decreased total Hb observed in this study during movement is explained by this expulsion of the pooling venous blood in the calf.

In addition, the degree of the concentration changes in the total Hb was much larger in the normal group than in the SCI group. This result is consistent with the report of van Beekvelt et al. (36) that muscle pump activity induced by imposing electrical stimulation is reduced in SCI subjects compared with healthy subjects. It has been suggested that this reduced muscle pump activity might be explained by the muscle atrophy and low venous capacity found in SCI subjects (17). A possible explanation for this result is that SCI patients have more fat because larger amounts of fat result in lower NIRS signals (37). However, as described later, other our results which the oxy- and deoxy-Hb showed AQ: 2
in the opposite concentration changes would not be expected from fat for the same reason.

In the present study, a gradual increase in oxy-Hb and decrease in deoxy-Hb that were independent of changes in total Hb were observed in the SCI subjects during the exercise period. On the contrary, there is no obvious concentration change of the oxy-Hb in the normal subjects who showed no EMG activity during passive leg movement. If no muscle oxygen consumption and/or supply was induced by the imposed movement, both the oxy-Hb and the deoxy-Hb should vary in a manner related to the concentration changes in the total Hb. Taken together with the occurrence of EMG activity in the SCI group, this change in the muscle oxygenation level can be attributed to the muscle activity produced by imposing passive leg movements. These results are in good agreement with a recent report by Bhambhani et al. (3), who suggest that changes in the oxygenation level in the paralyzed rectus femoris muscle during cycling movement are generated by functional electrical stimulation.

With respect to muscle oxygenation during exercise, previous studies have reported that continuous muscle contraction at moderate intensity follows the increments of deoxy-Hb because of the oxygen consumption in the acting muscle (6,34). Although we hypothesized that muscle was "active" during passive leg movement, the present results did *not* show increments of the deoxy-Hb. According to the general principle, concentration changes in oxy- and deoxy-Hb are dependent on the dynamics of the equilibrium between tissue oxygen demand and supply (2,16). Therefore, a possible reason for our result is that oxygen delivery far exceeds the oxygen extraction in the acting muscle. The enhancement of HR during passive leg motion (Fig. 5) provides evidence to support this notion.

Changes in Hb concentration after exercise. After the cessation of the passive leg movement, the concentration of total Hb in the SCI group exceeded the preexercise level, whereas that in the normal group simply recovered to the preexercise level. Because the total Hb reflects the degree of muscle blood flow (6), these changes may suggest that enhancement of the muscle blood flow occurred in the SCI group, possibly resulting from the muscle contraction and oxygenation during the exercise period. It is likely that the excess total Hb following exercise resulted from the pooling of blood in the calf. Nevertheless, in this study, the subjects were kept in a standing posture on the apparatus before the initiation of the exercise period until the total Hb value reached a constant level; therefore, the above total Hb changes during the recovery stage cannot be explained solely by blood pooling in the calf. These total Hb changes may be due to postexercise hyperemia (35).

HR changes by imposing passive leg movement. As shown in the Figure 5, the HR increased after the onset of the passive leg movement in both SCI and normal subjects. These results provide evidence of the enhancement of central circulation by imposing passive leg motion even in the SCI patients. The simplest explanation is that increments of the venous return due to the muscle pump activity result in the central circulation (29). However, taken together with the results of differences in the EMG activity, there would be different mechanisms underlying the enhanced HR between two groups. In the case of the normal subjects, it is plausible that the enhancement of the HR is induced by the neuronal factor, which is an afferent neural signal from the mechanoreceptor by inducing muscle stretching (12). On the other hand, this neuronal factor is not a suitable explanation for SCI results because of the sensory paralysis. Rather, our results, the appearance of muscular activity and an alteration of the NIRS signals, imply that a metabolic change accompanied by muscle contraction seems to play a primary role in the enhancement of the central circulation. Because we do not still have any direct evidence, further investigations are needed to clarify this point.

Implications for rehabilitation. As mentioned at the beginning, chronic inactivity and hypocirculation of the paralyzed area are crucial factors in secondary impairment in SCI subjects (25). The present results provide indirect evidence that passive leg movement performed in a standing posture could alter the oxygenation level of the paralyzed muscle and has the potential to facilitate circulation of the paralyzed area. Given that the muscle contraction level during normal walking is about 15% MVC (21), it is considered that the muscle contraction level observed in this study is adequate to facilitate neural activity and circulation of the paralyzed area.

On a practical level, the subjects in the present study did not move their upper limbs and trunk voluntarily, because our aim was to examine whether the oxygenation level of the paralyzed muscle was altered by imposing passive leg movement. In a nonexperimental situation, however, patients would commonly operate the device themselves by manipulating the lever with their upper limbs. It is possible that the additional voluntary upper limb movement could enhance circulation not only in the voluntarily acting area, but also in the paralyzed area.

Although muscular activity in the paralyzed area can also be induced by applying electrical stimulation, as is the case in functional electrical stimulation (FES) (1,23,31), there are essential differences between our method and the FES technique. Previous investigations pointed out that one of major disadvantages of the FES technique is that it is difficult to generate FES-induced continuous muscle contractions without fatigue (for a review, see Stein et al. (32)). The muscle fatigue can be attributed to the fact that the fatigable motor unit is preferentially recruited by imposing electrical stimulation, in that large motor nerves are more easily activated than smaller ones. In contrast, in the case of the passive leg movement produced in the present study, the motor units are presumably recruited according to the size principle (15), because the afferent input was offered from proprioceptors by imposing muscle stretch and body load. Furthermore, passive leg movement is simpler and more practical than FES, and has a lower risk of misuse. Therefore, this type of passive leg movement might be a useful and efficient method for rehabilitation following SCI.

CONCLUSION

The present results demonstrate that passive leg movement can induce not only muscular activity, but also alteration of the muscle oxygenation level in the paralyzed lower limb. There may be increased oxygen consumption, but this could not be ascertained from the measurements in this study. Further study will be needed to clarify this issue.

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脊髄損傷者の装具歩行における股関節動作の動力補助
—動力補助による装具歩行動作，エネルギーコストの変化—

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キーワード　脊髄損傷者，装具歩行，動力補助

抄録

交互歩行装具（Advanced Reciprocating Gait Orthosis：ARGO）を用いた脊髄完全損傷者の装具歩行について，遊脚期の股関節屈曲動作を動力補助する装置を考案した．動力補助の有無が装具歩行パフォーマンスにいかなる影響を及ぼすかを検討するために，胸髄完全損傷者8名（Th5～12）を対象として，装具歩行中の動作力学的計測，代謝計測を実施した．動力補助の直接の対象である股関節屈曲動作には動力補助に伴う顕著な変化は認められなかったが，体幹の筋群に運動麻痺が及ぶ胸髄高位損傷者では，歩行中の定常状態酸素摂取量と歩行速度から算出したエネルギーコストが動力補助によって改善する傾向を示した．本研究の結果は，脊髄損傷者の装具歩行動作に動力補助を施すことによって過度の身体的負担を軽減できる可能性を示すと共に，動力補助による影響が損傷高位によって異なる可能性を示すものであった．

1．はじめに

外傷等によって脊髄を損傷すると，多くの場合下肢の運動機能障害によって立位・歩行が困難となる．立位姿勢からの隔離は，骨・筋萎縮に代表される麻痺領域の退行の直接的な原因となるばかりか，日常生活における身体活動量の低下を引き起こし，心臓循環系疾患や生活習慣病の発現リスクを増加させる[1]．したがって，脊髄損傷後のリハビリテーションにおける立位・歩行訓練は他の運動では代替不可能な重要な役割を担うものと考えられる．しかし一方で，下肢に完全麻痺を有する脊髄損傷者の装具歩行は，杖を介した上肢・体幹等の残余運動機能によって代償的に脚を前方に振り出す運動様式を採ることから，歩行運動に要する労力が極めて大きくなることが指摘されている[2]．先行研究では，装具歩行中のエネルギー消費が健常者の歩行時の約3倍におよび，運動効率を反映するエネルギーコストに至っては約15倍

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Novel device for the hip motion assistance during orthotic gait for paraplegic patients

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に相当することが報告されている[6,8]．このような過度の身体的負担は短時間での疲労困憊を招き，身体機能の維持・向上を図るための適切な運動強度の実現を妨げる要因となる．

本研究では，交互歩行装具（Advanced Reciprocating Gait Orthosis：ARGO，**図1A**）を用いた脊髄完全損傷者の装具歩行時に，股関節動作を補助することで装具歩行動作の改善が図られるのではないか，との着想を持ち，歩行遊脚期における股関節屈曲動作を動力によって補助する装置を試作・評価した．歩行運動中の過度の身体的負担が動力補助によって軽減できれば，脊髄損傷後の身体機能の維持・向上を図る上で，より適切な身体的負荷を実現することが可能になるものと考えられる．

2. 研究方法

2-1 被験者

本研究の主旨を理解し，研究参加への同意を得た脊髄完全損傷者8名（19～34歳，Th5～12損傷）を対象とした（**表1**）．被験者は一般的なリハビリテーション訓練を受けた後，少なくとも10週間の装具歩行トレーニングを実施した．被験者には事前に本研究の実施に際しての危険性，身体的苦痛について十分に説明し，書面をもって研究参加に対する同意を得た．本研究で実施するプロトコルは国立身体障害者リハビリテーションセンター生物倫理委員会の是認を受けて実施した．

2-2 動力補助の概要

ARGOおよび股関節部に装備した動力機構の外観を**図1**に示す．ARGOは左右の股関節部を連結するヒップドライビングケーブルの作用（**図1B**参照）により，左右の脚を交互に振り出すことを可能にしている[9]．本研究ではヒップドライビングケーブルの末端部分にモータアクチュエータを取り付けることにより，歩行遊脚期の股関節屈

表1　被験者の身体特性

Subject	Age (year)	Weight (kg)	Injured level	Grade of ASIA
A	19	53	Th5	B
B	21	68	Th7	A
C	30	74	Th8	A
D	26	48	Th10	A
E	34	73	Th10	A
F	19	53	Th12	A
G	29	67	Th12	A
H	26	75	Th12	A

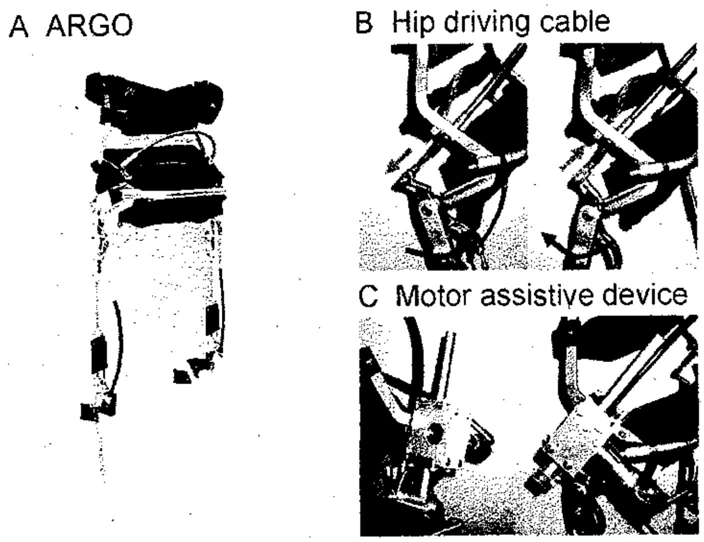

図1　交互歩行装具ARGO

A：外観．B：ARGO股関節部のヒップドライビングケーブル（左：股関節伸展時，右：股関節屈曲時）．
C：本研究で試作した動力装置．アクチュエータによってヒップドライビングケーブルの末端部を動作させる．

曲動作を補助する動力装置を試作した（**図1C**）．モータアクチュエータはDC-マイクロモータ，遊星ギア(ともにMINIMOTOR社)，ラックアンドピニオン(旭精工(株))で構成され，ニッケル水素電池を電源として駆動する．動力機構による股関節動作は，ロフストランド杖のグリップ部に設置した赤外線リモコンによって使用者本人が操作した．以降，動力装置を装備した装具，通常の装具をそれぞれ，M-ARGO（Motor assist-ARGO），N-ARGO（Normal ARGO）と表記する．

2-3　歩行動作解析

歩行動作の解析には三次元動作解析装置VICON370 system（Oxford Metrix製）を使用し，身体各部位に貼付したマーカの3次元座標データを取得した．同時に床面に埋め込まれた3分力床反力計(Kistler社製)により，身体および杖の床反力を測定した．M-ARGO, N-ARGO各々について，自己快適速度での歩行を5試行実施した．

両装具による歩行動作は，①歩行速度，②股関節動作，③床反力の各計測項目によって比較した．股関節動作に関してはマーカの座標データより関節可動域（ROM：Range of Motion），股関節角速度のピーク値（Hip VEL）を定量した．また，床反力に関しては身体荷重，杖各々の積分値により定量した．

2-4　代謝計測

携帯型代謝測定装置（K4-RQ, Cosmed社製）を用いて両装具による20分間の自己快適速度による装具歩行中の呼気ガス計測を実施した．N-ARGO, M-ARGO各々の計測は異なる日の同時間帯に実施した．呼気ガスはリアルタイムで分析され，酸素摂取量，二酸化炭素排泄量等の換気パラメータを15秒間隔で得た．同時にテレメトリ心拍計にて心拍数を記録した．実験後，歩行中の定常状態酸素摂取量と既定の係数[10]（$K=20.19$ J/ml, 1 ml $O_2=4.825$ cal, 1 cal $=4.184$ J）の積によりエネルギー消費量を，さらに歩行速度で除すことによりエネルギーコストを算出した（エネルギー消費量は単位時間あたり（秒）に要する熱量，エネルギーコストは1 mの移動に要する熱量をそれぞれ示す）．なお，各変数は体重あたりに換算した値を用いた．

$$\text{Energy consumption}(\text{J/kg/s}) = \frac{\text{Ambulatory min } \dot{V}_{O_2}(\text{ml/min})}{\text{Weight}(\text{kg}) \times 60} \times K$$

$$\text{Energy cost}(\text{J/kg/m}) = \frac{\text{Ambulatory min } \dot{V}_{O_2}(\text{ml/min})}{\text{Speed}(\text{m/min}) \times 60} \times K$$

2-5　統計解析

本研究で定量した変数は，各被験者の値は平均値±標準偏差，すべての被験者の平均値については平均値±標準誤差で表記した．改良型，通常型ARGO双方の装具歩行時の差異は対応のあるt検定を用いて検討した．また，損傷高位と各変数の関連をピアソンの積率相関係数を用いて検討した．本研究における統計的有意性の判断は5%水準とした．

3. 結果

図2にARGOによる脊髄完全損傷者（Th12損傷）の装具歩行の様子，歩行動作解析，代謝計測によって得られるデータの典型例を示す．**図3**にはM-ARGO, N-ARGO両装具による歩行中の平均速度を示した．被験者全員の平均値で比較すると，両装具間には有意な差異は認められなかったが（N-ARGO vs. M-ARGO：18.87±6.18 vs. 19.56±6.02 m/min, **図3上段**），損傷高位との関連を見ると，損傷部位が高い被験者において，動力補助による歩行速度の増加傾向が認められた（**図3中段**）．N-ARGO歩行時の速度に対する変化率（**図3下段**）で示すと，特に高位損傷者において動力補助による歩行速度の増加が実現されていることが伺える．

3-1　歩行動作解析

図4には両装具による歩行中の動作力学的計測から得られた各変数の，群間の平均値(**図4左**)，損傷高位との関連(**図4右**)を示した．動力補助の直接の対象である股関節動作については，被験者全員の平均値では可動域（N-ARGO vs. M-ARGO：44.04±3.82 vs. 42.94±4.02 deg），関節角速度（148.80±15.34 vs. 142.18±12.78 deg/sec）共に両装具間に有意な差異を認めなかったが，杖の床反力は高位損傷者で減少する傾向が認められた．

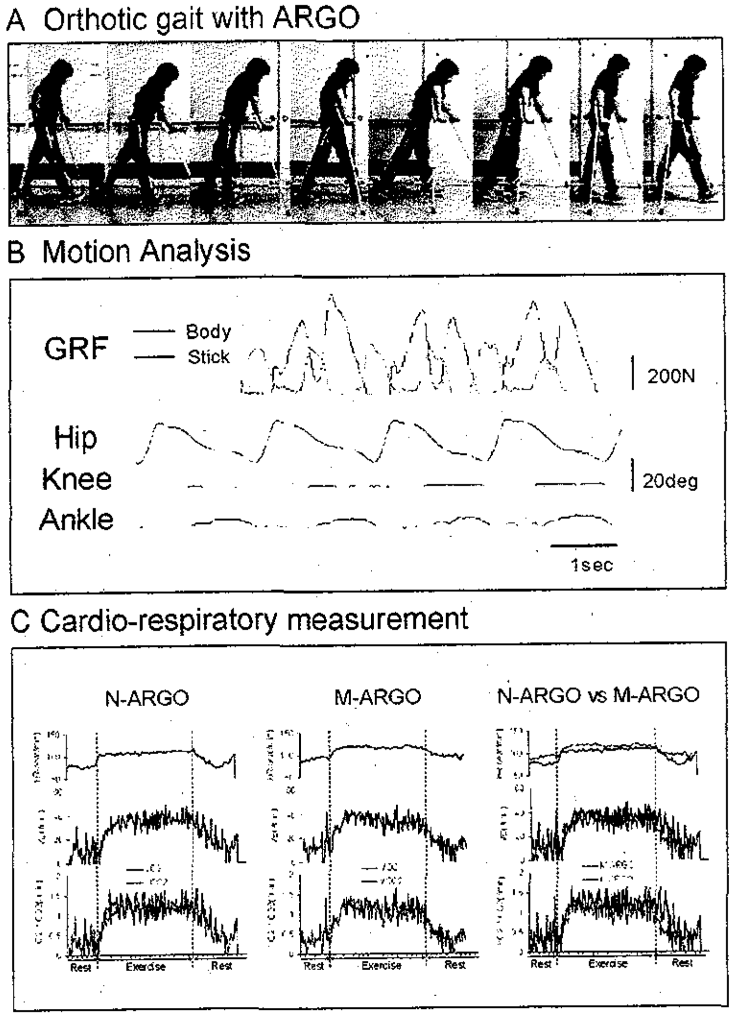

図 2

A：ARGO による歩行の様子．B：歩行中の動作解析から得られる床反力および関節角度データの典型例．C：歩行中の代謝計測から得られる換気パラメータの典型例（左：N-ARGO 歩行時，中央：M-ARGO 歩行時，右：両装具の比較）．

3-2　エネルギー消費量，エネルギーコスト

表 2 には呼気ガス計測から得られた両装具による歩行中の心拍数，酸素摂取量を示す．酸素摂取量は安静時と比較して装具歩行時では約 3 倍を示し，この値は先行研究における報告[3]とほぼ一致した．**図 5** には上記酸素摂取量と歩行速度から算出した，歩行中のエネルギー消費量とエネルギーコストの結果を示した．エネルギー消費量は，動力補助の有無による顕著な差異を認めなかった一方で，エネルギーコストは高位損傷者ほど減少する結果を示した．

4.　考　察

本研究では，交互歩行装具 ARGO の股関節部に動力機構を装備し，動力補助の効果を歩行動作，運動中のエネルギー消費の観点から検討した．機械的な動力補助対象である股関節動作には，動力補助の有無で大きな変化は認められな

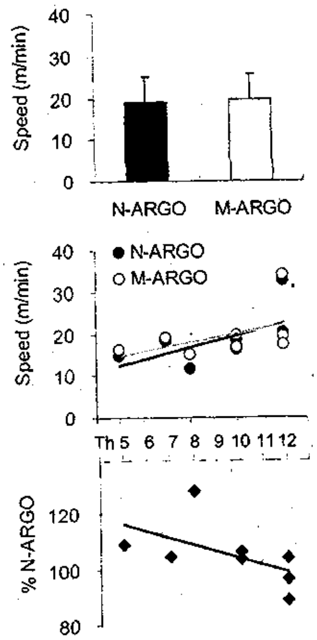

図 3　両装具による歩行中の平均速度

上段：全被験者の平均値，中段：損傷高位と歩行速度の関連，下段：N-ARGO 歩行時に対する M-ARGO 歩行時の変化率．

かったものの（図4），歩行中の運動効率を反映するエネルギーコストは，高位損傷者で減少する傾向が認められた（図5）．以下の考察では，今回試作した動力補助の作用について，ARGO を用いた装具歩行の特性，損傷高位による効果の違いの観点から議論する．

4-1　動力補助の作用

本研究で対象とした脊髄損傷者はいずれも下肢運動機能に完全麻痺を有する者であり，随意的な股関節屈曲動作が困難な者であった．ARGO を用いた脊髄損傷者の装具歩行では，杖を介した上肢，体幹等の残余機能によって上半身を前後傾させ，左右の股関節部を連結するヒップドライビングケーブル[9] を作用させることによって，脚を前方に振り出す運動形態を採る．本研究では ARGO のヒップドライビングケーブルの作用に着目し，ケーブルの終止部にアクチュエータを装備することによって，歩行遊脚期における股関節屈曲動作の動力補助を試みた（図1C）．評価実験の結果，直接的に動力補助を施した股関節屈曲動作には動力装置の有無による顕著な差異は認められなかったが，酸素摂取量計測から得た装具歩行中のエネルギーコストは，高位損傷者で動力補助によって減少する傾向を示した．また，高位損傷者では動力補助によって杖の床反力が減少する結果を示した（図4）．

本研究で対象とした脊髄損傷者の受傷部位は，体幹の筋群を支配する胸髄中～下位の髄節にわたるため，上記の結果には体幹の残存運動機能の有無が影響している可能性が高い．杖の床反力が減少した結果を合わせて考えると，本研究で認められた高位損傷者におけるエネルギーコストの変化は，動力補助が股関節動作を生み出すための上肢の代償動作に必要な労力を緩和したことによって生じた可能性が高い．

4-2　装具歩行の動力補助とリハビリテーション効果の関連

本研究では，脊髄損傷者における装具歩行中の過度の負担[8] が指摘されている現状を踏まえ，身体機能の維持・向上のためのより適切な運動強度を実現するための支援策として，股関節動作の動力補助を考案した．歩行運動によるリハビリテーション効果を最大限に得るためには，装具歩行に要する高い身体的負担を軽減することは極めて重要であり，上記の企図が実現されれば脊髄損傷者の装具歩行訓練に新たな展開をもたらす可能性がある．

脊髄損傷では損傷部位が脳に近いほど麻痺に陥る部位が広範に及ぶことから，損傷部位が高位になるほど，装具歩行の実現可能性もより低くなることは容易に想像される．事実，装具歩行パフォーマンスと損傷高位との関連について検討した我々の先行研究では，本研究と同様に損傷高位とエネルギーコストの間に負の相関関係を認めている[1]．しかし，歩行リハビリテーションの効果は損傷高位を問うものではなく，むしろ身体諸機能の障害の程度が深刻な高位損傷者ほど立位・歩行の必要性が高い場合もあり得る．こうした視座に立てば，運動麻痺によって実現が困難な動作に動力補助を施すことにより，より適切な運動強度の実現を図ることは極めて重要な意味を持つものと考えられる．

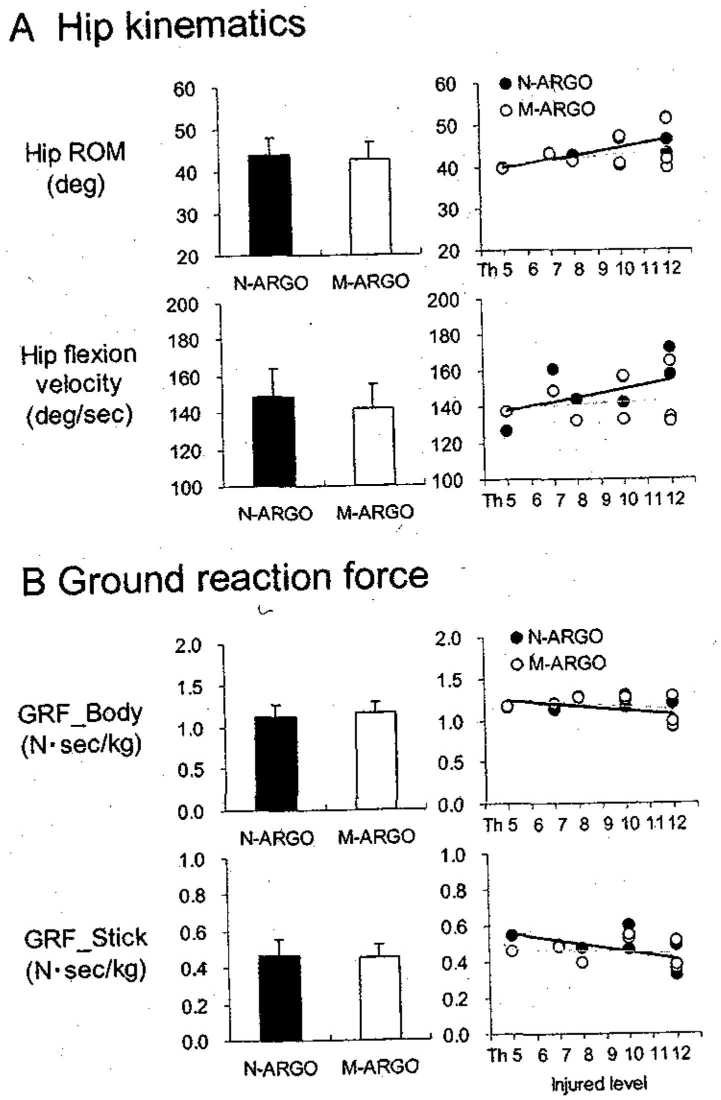

図 4　両装具による歩行中の股関節動作（A），床反力（B）の比較
左：全被験者の平均値，右：損傷高位と歩行速度の関連．

4-3　今後の課題―より適切な動力補助を目指して―

前項に記した目的を達するには，装置のさらなる改良が必要である．今回試作した装置は，アクチュエータの出力が一定であったが，被験者個々の歩行速度，股関節の動作特性に応じてアクチュエータの出力を調節することができれば，より適切な動作補助が可能になるものと考えられる．また，動力装置の操作は杖のグリップ部に設置したスイッチを被験者当人が操作することによって行わせたが，この操作方法では，動力補助の駆動位相や時間が被験者のボタン操作に依存する．これに対して，他の身体部位の運動位相，あるいは床反力を参照して股関節屈曲の位相を同定し，動力装置の駆動位相を制御するなどの方策が取れれば，より効果的な動力補助が実現できるものと考