Methods

Subjects
In all, 10 SCI patients who met the following criteria participated in this study: (i) injured at thoracic level, (ii) complete motor paralysis in the lower limb muscle (ASIA classification; grade A or B), (iii) no history of cardiorespiratory disease. All patients were at least 6 months since time of injury, with time since injury ranging from 8 to 32 months (Table 1). Each subject gave his or her informed consent to the experimental procedure, which was approved by the local biological ethics committee of the National Rehabilitation Center for the Persons with Disabilities (NRCD).

Orthotic gait training
All patients had undergone a standard rehabilitation program, consisting of muscle stretching, balance training, and transfer activity, and participated in orthotic gait training with a weight-bearing control orthosis (WBC) or advanced reciprocating gait orthosis (ARGO). Eight of 10 patients have kept the orthotic gait training for 15 weeks, and the other two patients have kept for 10 (patient E) and 4 weeks (patient G), respectively. Although there is individual variation, in many cases, lower thoracic level of paraplegic patients could walk after 10 weeks of gait training independently, while it needs more practice for higher thoracic level of patients. After the training period, each subject could perform the orthotic gait (patients F and G still required light support to avoid falling) independently, and were able to walk continuously for at least 20 min.

Apparatus
Appearance and sequential picture of walking with WBC and ARGO were shown in Figure 1. The mechanical features of the WBC have been fully described elsewhere.[11,12] This orthosis consists of a rigid frame that supports the user's body weight, a special hip joint device that reciprocally propels each leg forward, a gas-powered foot device that varies the sole thickness of the device for foot/floor clearance, and a control system of the orthosis. As a whole these mechanical features enable a user to ambulate at a faster speed and with less energy expended.[6] The ARGO also has a special hip joint device named 'hip driving cable' which connects both sides of the leg frame. With this device a torque exerted by the right (left) hip joint is mechanically transmitted to the left (right) hip joint, resulting in the torque to the opposite direction exerted by the left (right) hip joint.

Physical intensity during orthotic gait
On a separate day, cardiorespiratory responses at rest and during orthotic gait were measured. Subjects were asked to abstain from alcohol and caffeine for at least 12 h before the experiment. The temperature and humidity on the experiment were 23.5±4.2°C and 68.3±3.3%, respectively. The experimental procedure was as follows: 5 min at rest in the sitting position, 20 min of continuous walking at the most comfortable speed. The cardiorespiratory parameters at rest and during walking were measured continuously with a telemetric device (K4 Cosmed, Italy) and were analyzed in real time. The telemetric device consists of a transmitting unit, a face mask to sample the expired gas, a heart rate chest strip, a battery, and a receiving unit. The following cardiorespiratory parameters were

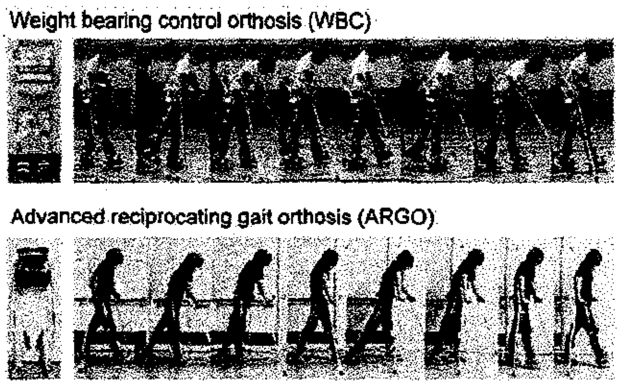

Figure 1 Appearance and sequential picture of walking with weight-bearing control orthosis (WBC; above) and advanced reciprocating gait orthosis (ARGO; below)

Table 1 Characteristics of the patients

Patient	*Sex*	*Age (years)*	*Height (cm)*	*Weight (kg)*	*Lesion level*	*Grade of ASIA*	*Duration of paraplegia (months)*	*Orthosis*
A	M	28	173	63	Th8	A	12	WBC
B	M	27	175	60	Th10	A	10	WBC
C	M	22	175	68	Th12	A	8	WBC
D	M	21	167	46	Th12	B	32	WBC
E	M	36	178	73	Th11	A	20	ARGO
F	M	19	175	53	Th5	B	24	ARGO
G	F	26	156	45	Th10	A	13	ARGO
H	M	30	178	67	Th12	A	13	ARGO
I	M	34	165	54	Th6	A	28	ARGO
J	M	23	168	65	Th8	A	26	ARGO

obtained: oxygen uptake (V_{O_2}) and heart rate (HR). Walking speed in the steady state during walking and rating of perceived exertion (RPE) score were also recorded. After the experiments, the energy consumption and walking energy cost were calculated. The terms adopted were those of Nene and Patrick[9] and calculations performed according to their protocol:

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

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

where $K = 20.19$ J/ml, since 1 ml $O_2 = 4.825$ cal and 1 cal = 4.184 J.

NK cell activities

Blood samples were drawn from an antecubital vein with the patient in the seated position before and just after orthotic gait exercise. The exercise consisted of 20 min of continuous walking at the most comfortable speed in the inside of the hospital ward.

NK cell activity was determined by (superscript: 51) Cr-release cytotoxicity assay using the K562 cell-line which derived from cells with chronic myelogenous leukemia as targets, and calculated using the following formula: %NK cell activity = {(experiment mean − spontaneous mean)/(total mean − spontaneous mean)} * 100. In all NK cell activity testing, percent of spontaneous release was less than 5% of total release. Effector:target (*E:T*) ratios used were 20:1. Controls included cultures of untreated cells (spontaneous release) and cells treated with 3% sodium dodecyl sulfate (SDS) (total release).

Statistical analysis

Values are given as the mean ± SD. Statistical difference in NK cell activity between pre- and postexercise was tested by paired *t*-test. Significance was accepted at $P < 0.05$.

Results

Physical intensity during orthotic gait

The average walking speed during orthotic gait was 18. 01 ± 2.22 m/min. Eight of 10 patients were able to walk continuously, without stumbling, for 20 min. Table 2 shows the cardiorespiratory responses, energy consumption, energy cost, walking speed, and RPE during orthotic gait. During walking, cardiorespiratory parameters clearly showed a significant increase compared with resting rate. The steady-state value of the V_{O_2} ranged from 14.20 to 24.83 ml/kg (average value = 18.13 ± 3.92 ml/kg), and HR was 99.2–166.4 b/min (average value = 142.53 ± 19.84 b/min). The energy consumption and energy cost during walking were 5.94 ± 1.16 J/kg/s and 19.63 ± 5.04 J/kg/m, respectively. The RPE score just after exercise ranged from 13 to 19 (median value: 15).

NK cell activity

The average value of NK cell activity in the SCI patients at rest (12.7 ± 5. 28; ranging from 7.6 to 23.4) was remarkably lower than the standard value in the healthy nondisabled persons (32.9 ± 15.8). Nine of 10 patients showed enhancement of NK cell activity in response to the 20 min of orthotic gait exercise. The total average value of the postexercise NK cell activity was significantly higher than that of the pre-exercise (pre versus post; 12.7 ± 5.28 verses 17.76 ± 6.71, $P < 0.05$, Table 3, Figure 2).

Table 2 Physical intensity during orthotic walking

Patient	V_{O_2} (ml/kg) Rest	V_{O_2} (ml/kg) Exercise	*HR (beat/min)* Rest	*HR (beat/min)* Exercise	*Energy consmption (J/kg/s)*	*Energy cost (J/kg/m)*	*Walking speed (m/min)*	*RPE score (unit)*
A	8.84	17.39	84.6	154.0	5.85	17.55	20	15
C	6.76	14.67	104.4	132.9	4.94	16.45	18	15
D	4.57	18.05	96.0	145.1	6.07	16.56	22	13
E	6.71	15.62	78.3	131.5	5.26	17.19	18	13
F	8.29	21.14	104.1	166.4	7.11	29.06	15	19
G	6.78	24.20	62.11	132.5	8.14	26.22	18	17
H	6.70	16.01	40.1	99.2	5.39	16.12	20	13
I	9.75	24.83	81.0	143.6	8.35	29.33	17	17
J	5.80	15.19	107.8	163.0	5.11	19.67	16	17
Mean	7.03	18.13	85.18	142.53	6.10	20.61	18.01	15 (median)
SD	1.53	3.92	21.23	19.84	1.32	5.40	2.22	-

Table 3 NK cell activity in pre- and postexercise

Patient	*Pre*	*Post*	*Δ(%Pre)*
A	10.1	14.4	142.36
B	11.3	14.6	129.19
C	23.4	27.1	115.87
D	18.0	19.4	107.78
E	9.8	17.3	176.53
F	18.3	16.6	90.71
G	9.3	30.8	331.18
H	7.6	11.6	152.63
I	11	17.3	157.27
J	8.3	8.6	103.61
Mean	12.70	17.76	150.71
SD	5.28	6.71	68.87

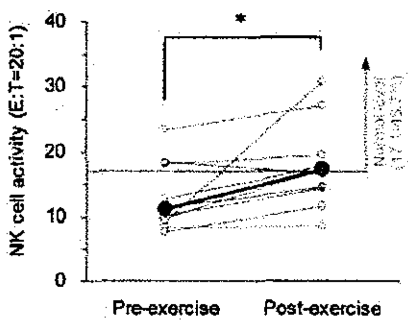

Figure 2 Natural killer (NK) cell activity in response to 20 min of orthotic gait exercise. Thick and thin lines indicate the total averaged ($n=10$) and each subject's value, respectively. NK cell activity was represented by an E/T ratio (20:1)

Discussion

In the present study, we aimed to examine the effect of orthotic gait exercise on the NK cell activity in SCI patients. The main observations made here were as follows: (i) the NK cell activity in SCI patients was remarkably lower than the standard value in healthy persons; (ii) the NK cell activity was significantly increased through 20 min of orthotic gait exercise, and (iii) the one patient who showed decrement of NK cell activity in response to exercise had an injury of the highest level (Th5) and showed the higher energy cost during orthotic gait exercise.

Previous investigations have reported that SCI was accompanied by depression of immune system including decrement of NK cell activities.[3] These findings, taken together with the fact that paraplegic patients have significantly reduced $\dot{V}_{O_2}$ peak values as a consequence of the reduction in the daily activity levels,[10] suggest that the decrements of NK cell activity may be the result of limitation of the patient's physical activity.

Because immune resistance is generally regarded as an essential factor for health care, it is conceivable that enhancement of immune function is important for SCI patients to maintain decent physical condition. In this regard, many investigations have reported the possibility of enhancement of immune function through moderate exercise not only in normal persons[1,7,8] but also in SCI patients.[11] Kliesch *et al*[3] demonstrated restoration of immune function through rehabilitation therapy in treated subjects by comparison with those not receiving treatment. The present result of exercise-induced enhancement of NK cell activity is in good agreement with these reports. In the present study, the steady-state value of the $\dot{V}_{O_2}$ during orthotic gait was 18.13 ± 3.92 ml/kg, and HR was 142.53 ± 19.84 b/min (Table 2). The level of physical intensity implied by these values was considered to be suitable for promoting the general health of SCI patients. Further, all of our subjects, with the exception of patient F, could walk for a considerable time and distance without exhaustion. It is therefore considered that the enhancement of NK cell activities was the result of the suitable aerobic condition during orthotic gait.

Finally, we considered why only patient F showed a decrement in the NK cell activity in response to the orthotic gait exercise. As mentioned above, this patient showed the higher energy cost during orthotic gait and had the highest level of injury in all eight patients. The orthotic gait for SCI patients requires compensatory motion of the residual trunk and upper limbs to swing the paralyzed leg.[12,13] Patient F, who was injured at Th5, could not contract his trunk muscles due to motor paralysis. Consequently, the excess energy expenditure and burden on his upper limbs made it impossible for him to achieve suitable exercise intensity for enhancement of immune function during orthotic gait. His higher RPE score (19: very very hard, Table 2) reflects greater energy consumption than that of the other patients during orthotic gait.

To date, many researchers have reported extremely high-energy requirements of orthotic gait.[4,14,15] Although many devices have been developed to improve this problem to date, it is still unknown whether the orthotic use contribute to facilitate the health care for SCI persons. The present result provides evidence of the effectiveness of the orthotic gait exercise for promotion of the general health of these SCI patients. However, the question remains whether regular exercise training leads to chronically elevated NK cell activity. Further study will be needed to clarify this issue.

Acknowledgements

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RESEARCH ARTICLES

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Induction of locomotor-like EMG activity in paraplegic persons by orthotic gait training

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Abstract This is, to our knowledge, the first report demonstrating the effects of orthotic gait training on the activity of the spinal locomotor neural networks. Three subjects with complete spinal cord injury (SCI) performed 1-h training with reciprocating gait orthosis 5 days/week for 12 weeks. The results showed that after 3 (n=1) or 6 weeks (n=2) of training, EMG activities synchronized with locomotor rhythm appeared in the soleus muscle (SOL) in all subjects, although very little EMG activity accompanied the orthotic gait at the early training stage. Our results suggest that the induced modulation in the SOL EMG waveforms might be attributable to changes in the orthotic gait movement pattern, and/or changes in the interneuronal activities of the spinal locomotor neural networks, as a result of orthotic gait training.

Keywords Spinal cord injury · Locomotion · Gait orthosis · Plasticity

Introduction

It has been well established that the human spinal cord has the potential to generate the basic locomotor pattern by interaction of the locomotor neural networks and peripheral sensory information concomitant with limb movements (Harkema et al. 2000). For example, several researchers have demonstrated that in severe spinal cord-injured (SCI) subjects, locomotor-like coordinated electromyographic (EMG) activity can be induced in paralyzed lower limb muscles by passive stepping on a moving treadmill with partial body-unloading (Dietz et al. 1994, 1995; Dobkin et al. 1994; Wernig et al. 1995). However, the nature of the neural networks involved in generation of locomotor EMG activities in SCI persons is not yet fully understood.

During a specific type of upright walking with gait orthosis called weight bearing control orthosis (WBCO) (Yano et al. 1997), modulation of lower leg-muscle activities that synchronize with that particular locomotor cycle can be induced (Kojima et al. 1998). Because the WBCO gait, like other reciprocating gait orthoses, is a "stiff-leg" gait, i.e., a gait with the knee locked in full extension and the ankle in a neutral position, the afferent information thought to primarily contribute to inducing the locomotor-like EMG activity would be associated with hip-joint movement and load on the leg (Dietz et al. 2002). This in turn might mean that use of the orthotic gait would allow us to investigate the contribution of the involvement of hip extension/flexion movement or load on the leg to generation of locomotor-like EMG modulation, specifically in the "lower leg" muscles, which are remote from the hip joint. However, in our experience, little EMG activity appears during the WBCO gait when the user is not well trained, whereas it has been demonstrated in a well-trained SCI subject that locomotor-like EMG is observed (Kojima et al. 1998). These empirical observations might be explained as follows: (1) afferent inputs concomitant with limb movements would not be sufficient to evoke locomotor EMGs during the untrained orthotic gait; and (2) the orthotic gait training induces an alteration in interneuronal activities in spinal neural networks, which would generate the locomotor EMG even when the pattern and amount of afferent inputs are the same. To test these possibilities, we first had to longitudinally evaluate changes in EMG activities in paralyzed muscles, and to relate these EMG changes with gait-motion changes during the time course of training. The purpose of this study, therefore, was to clarify: (1) whether alteration in the EMG activities in the lower leg muscles occurs during the time course of orthotic gait training, and (2) the relation between the EMG activities and kinetic/kinematic alteration of the gait motion due to the training.

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Materials and methods

Subjects and orthotic gait training

Three clinically complete paraplegic men (22–28 years; 60–68 kg; 173–177 cm) with traumatic spinal cord injuries (Th8–Th12) voluntarily participated in this study. The physical characteristics of the subjects are shown in Table 1. The American Spinal Injury Association (ASIA) impairment scale was used for the clinical assessment of subjects, and in each case the sensorimotor functions were classified as A, which means no motor or sensory function below the level of the lesion. The subjects gave their 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 the Disabled, Tokorozawa, Japan. The long brace reciprocating gait orthosis, the WBCO, was used for the training. The subjects performed orthotic gait training with WBCO for 12 weeks, at 1 hour/d for 5 days/week. The mechanical features of the WBCO have been fully described elsewhere (Yano et al. 1997; Kojima et al. 1998). In short it has two specific features that the other existing gait orthoses do not have. The first one is a special gas-powered system to control the foot sole thickness. It can switch the sole thickness depending on the gait phase; when a leg is swinging forward, the sole of this particular leg is held at the thinner position; and just before the heel strikes, the sole gets changed to the thicker position. With this system a user can swing their legs easier without leaning the body sideward to make a clearance between a foot sole and floor. The second one is a special hip joint device. With this device a torque exerted by the right (left) hip joint is mechanically transmitted to the left (right) hip joint, resulting in the torque to the opposite direction exerted by the left (right) hip joint. This system assists each leg to reciprocally propel forward. As a whole these mechanical features enable a user ambulate at faster speed and with less energy expended (Kawashima et al. 2003).

Experiments

To evaluate the kinetic and kinematic changes in the orthotic gait motion during the course of training, the gait motion was measured with a three-dimensional motion-analysis system (VICON370, Oxford Metrics, UK). The motion-analysis system consists of a conventional video-analysis system with seven cameras and Kistler force plates. The force plates, sized 160×450 cm, consisted of two 80×200 cm plates and four 40×250 cm plates. These separate force plates enable us to measure ground reaction forces (GRF) under the feet and canes on both sides, separately. The orthotic gait motion was recorded along with electromyographic (EMG) activities in the right soleus (SOL) and tibialis anterior (TA) muscles. EMGs were recorded by two surface electrodes (Ag/AgCl, 0.8 cm diameter) attached along the muscle fibers over the belly of each muscle and set at an interelectrode distance of 0.5 cm. The EMG signal was detected by a bipolar differential amplifier with upper and lower cutoff frequencies of 50–3000 Hz. Very thin elastic nylon bandages were used to firmly hold both electrodes and lead lines to the body, preventing any small displacement of electrodes and lines that might cause artifacts. These measurements were carried out three times (after 1, 6, and 12 weeks) in subject A; six times (after 1, 2, 3, 4, 10, and 12 weeks of training) in subject B; and twice (after 1 and 6 weeks) in subject C during the training period. For the measurement, subjects ambulated along a 10-m walkway several times at comfortable cadences. They repeated the trials with short-time intermissions, usually a couple of minutes, until the minimum required number of data was obtained. We sampled at least six step cycles for the analysis. Many step cycles, for example more than ten cycles, could not be recorded in the measurements of this study, since high quality VICON data could be obtained only for one or two steps of around five steps in a trial. At the beginning of training, especially, the experimenter had to walk beside the subjects to prevent a fall. This disturbs the motion capture with the VICON system, and makes the space in which the motion capture is possible small. Due to this limitation it would have taken a relatively longer time for the subjects to record many step cycles. To reduce time for the experiment we decided to finish the measurement when at least six step cycles were obtained in good quality.

Changes in the following kinetic and kinematic variables were evaluated from the measured VICON data throughout the training period: kinetic variables, including the impulse and mean vertical GRF (mGRF) under the foot, and kinematic variables, including the stance time and swing time, velocity, cadence and step length, joint range of motion (ROM), and peak velocity during the stance phase of the hip and ankle joints. The digitized EMG signals were full wave-rectified after rejection of the DC component. Then, from the rectified EMG signals, mean values (mEMG) for the stance phase were calculated and normalized by those values at rest.

Stretch reflex test

The reflex EMG responses elicited by mechanical stretches at various velocities were tested to verify whether the stretch reflex mediated the induced EMG activity in the SOL during the orthotic gait. Stretch reflex responses were evoked by imposing a quick dorsiflexion with an amplitude of 10 deg to the SOL muscle, while the subjects were seated comfortably in a chair with the right leg fixed to a foot plate connected to a servo-controlled torque motor (Senoh Inc., Japan). The hip, knee and ankle joints were set at 80 deg, 60 deg flexed and 10 deg plantar-flexed positions (anatomical position is 0 deg), respectively. All of 25 perturbations, each consisting of various angular velocities (50–350 deg/s), were applied to the ankle joint in random order. In the present study, the short latency reflex component, M1, was evaluated, since only M1 component was induced in the three subjects. The onset of the first EMG response was defined as the moment when the rectified EMG activities reached levels higher than the average resting potential plus three times its standard deviation (BGA+3SD), and the response duration was defined as 30 ms from the response onset. The average rectified EMG value above the resting potential level over the response duration was considered as the M1 level, and the relation between the imposed stretch velocity and the M1 level was analyzed for each subject.

Additional experiments

Additional experiments were done for the subject A in order to ascertain whether: (1) the EMG activity was induced with another conventional gait orthosis (advanced reciprocating gait orthosis, ARGO), and (2) in order to compare how different gait velocities affected the induced EMG activity before and after the training. Because this subject continued the training for over half a year,

Table 1 *Clinical characteristics of subjects studied*

Subject	Sex	Age (years)	Height (cm)	Weight (kg)	Injury level (segment)	ASIA	Time postinjury (months)	Etiology
A	M	27	177	60	T10	A	10	Trauma
B	M	22	174	68	T12	A	8	Trauma
C	M	28	173	63	T8	A	12	Trauma

measurements could be taken at the 1st week (1-W), 4th week (4-W), and 20th week (20-W) of the training. In the measurements, the subject ambulated at various velocities, speeding up his pace on the basis of his comfort.

Statistics

The measured kinetic and kinematic variables during the training period were compared with the first measurement values using the Student's *t*-test. Data are presented as mean ± SD. Significance was accepted as $p<0.05$.

Results

All subjects could ambulate independently with the aid of canes in the 1st week of training. In subjects A and B, gait velocities were respectively increased from 7.7 to 13.2 m/min, and from 11.8 to 21.2 m/min after the 12-week training; these increases are concomitant with increases in step length, ROM, angular velocity of hip and ankle joints, and the mGRF during the stance phase. In subject C, however, both the hip and ankle-joint angular velocities were decreased; the hip joint's ROM was decreased; and the ankle joint's ROM was increased. These findings were likely due to the markedly faster gait velocity (22.4 m/min) of this subject, as compared to the other two subjects, at the beginning of the training. In none of the subjects was clear modulation in the EMG activities of either muscle observed at the beginning of the training. After three (subj. A) or six (subjs. B and C) weeks of training, however, synchronized EMG bursts with the stance phase commonly appeared in the SOL in all three subjects, whereas no clear modulation was observed in any of the TA EMG waveforms (Fig. 1A). Figure 2 shows changes in the walking velocities, hip and ankle joint angular velocities, mGRFs, and SOL EMGs during the time course of training in the three subjects. As mentioned above, the SOL EMG activities increased during the training period in all subjects, and this increasing time course was qualitatively most similar to the gait velocity and mGRF. The increasing patterns in the SOL EMGs were not necessarily in parallel with those in the hip and ankle joint velocities. In subject C, specifically, the hip joint and ankle joint velocities demonstrated a tendency to decrease, though the SOL EMG increased.

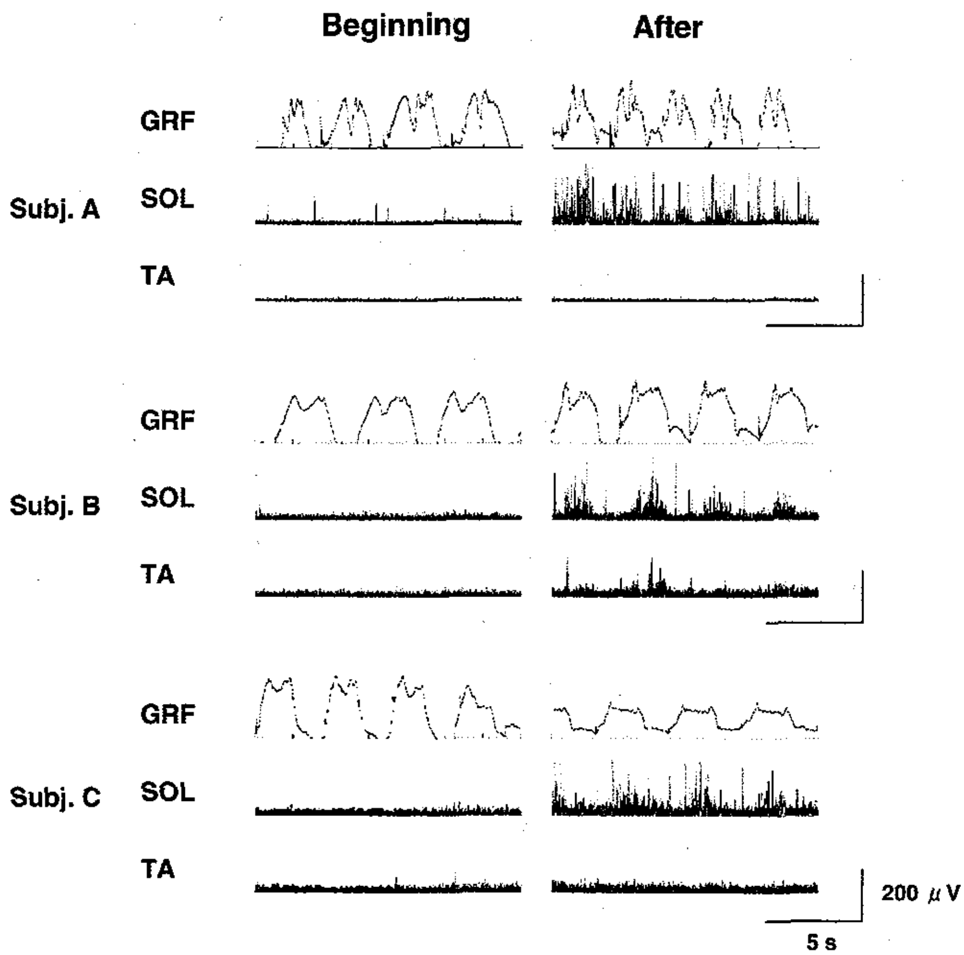

Fig. 1 Changes in EMG activities in the lower leg muscles during orthotic gait before and after the training. Results from the three subjects are shown (*GRF* the vertical ground reaction force, *SOL*, *TA* rectified EMGs from the soleus and tibialis anterior muscles, respectively)

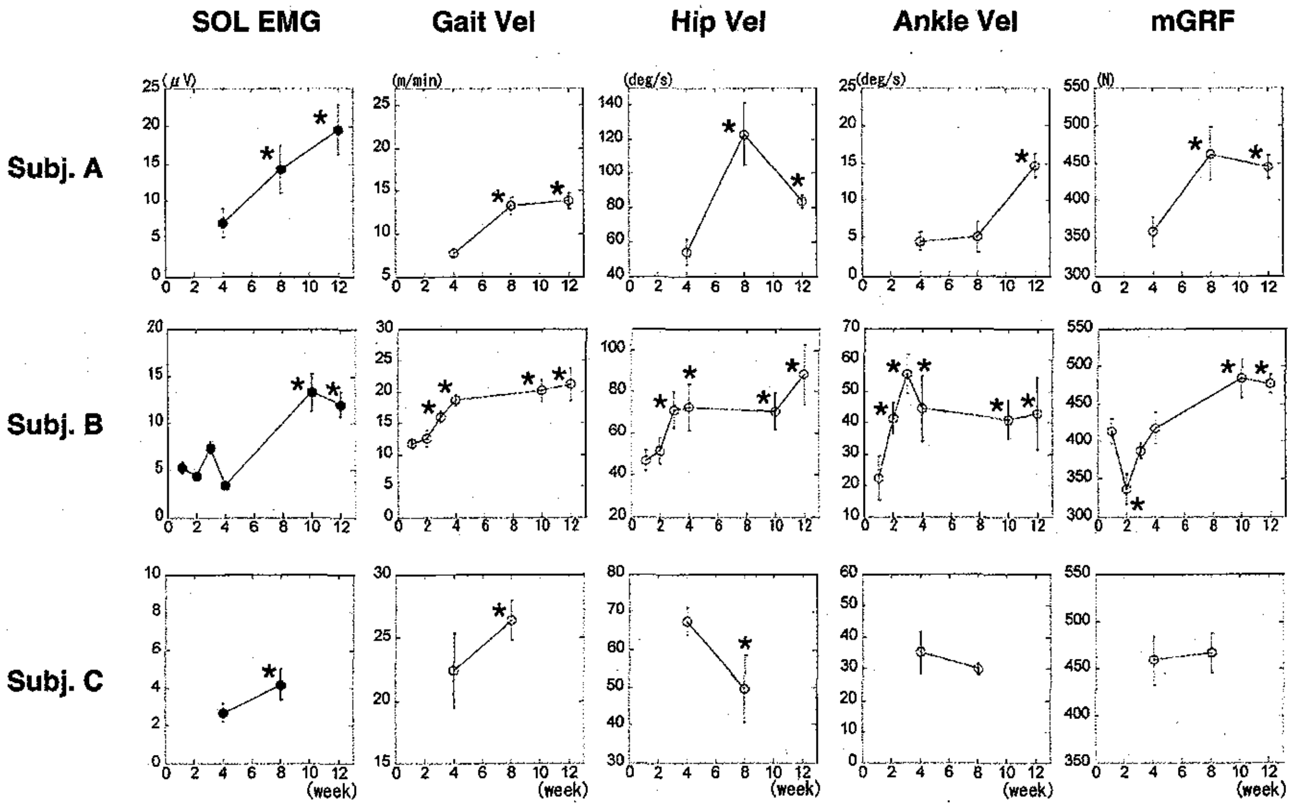

Fig. 2 Changes in the SOL EMGs, the gait velocities, the hip and ankle joint angular velocities, and the ground reaction forces over the training period for the three subjects

Additional experiment

Figure 3 shows the SOL EMG activities during the orthotic gait at three different gait velocities at the 1-W and 4-W measurements. It was demonstrated that the synchronized EMG burst with the stance phase increased drastically with the ambulation velocity after 4 weeks of training; no such clear modulation was observed at the 1st week measurement.

Relationships of the SOL EMGs during the stance phase with the gait velocity, hip velocity, ankle velocity, and ground reaction force, respectively, in each measurement are demonstrated in Fig. 4. In the 1-W measurement, the SOL EMG did not clearly increase with the gait velocity or the other kinetic and kinematic variables; the measurement reflected no qualitatively clear modulation. In the 4-W and 20-W measurements, however, the SOL EMGs covaried with the gait velocity and the other variables. It should be noted that the levels of SOL EMGs in the 4-W and 20-W measurements were greater than those in the 1-W measurement, even though the kinetic and kinematic variables were in similar ranges. This result suggests that the observed increase in the SOL EMG during the time course of training was not merely dependent on the kinetic and kinematic factors; neurological factors were to some degree involved.

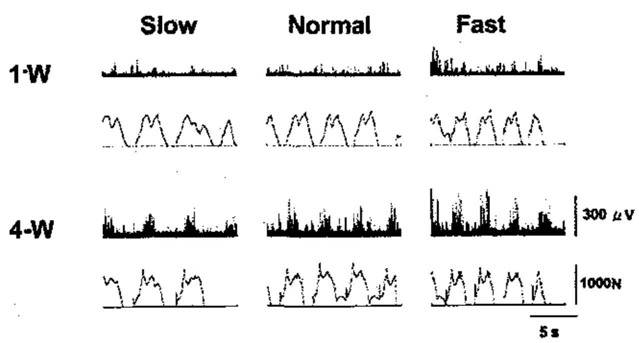

Fig. 3 A comparison of the velocity dependent changes in the SOL EMGs during orthotic gait before and after the 4-week training with the other gait orthosis

Stretch reflex test

Figure 5 demonstrates the relationships between stretch velocity and the reflex EMG responses in the SOL for the three subjects. It was indicated that in all three subjects, the stretch reflex EMG response was induced when the applied stretch velocity was faster than 100 deg/s, meaning the threshold velocity of the reflex was around 100 deg/s. These threshold velocities were well above the peak ankle dorsiflexion velocities observed during the orthotic gait in the three subjects, suggesting that the SOL EMGs

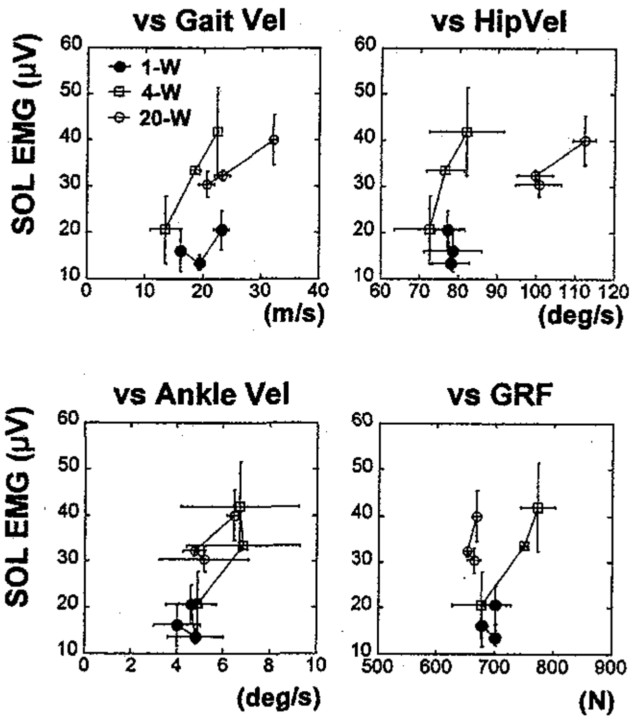

Fig. 4 Comparisons of the relationships of the SOL EMGs to the kinetic and kinematic variables before and after 1 month and 5 months of training. Abbreviations: *vs Gait Vel*, *vs Hip Vel*, *vs Ankle Vel*, and *vs GRF* refer to the relationships between the SOL EMG and the gait velocity, hip joint angular velocity, ankle joint angular velocity, and ground reaction force, respectively

observed were not merely mediated by the stretch reflex pathway.

Discussion

The results in the present study demonstrated that intense orthotic gait training induced modulation of EMG activities in the ankle extensor SOL muscle in individuals with clinically complete SCI. These results constitute neurologically significant indirect evidence that knee (and to some degree ankle) movements are far less important than hip movement and loading for the induction of locomotor EMG activity, at least in the SOL. Further, these results have great clinical significance especially in terms of gait rehabilitation of patients with incomplete spinal cord injury. It is worth noting that the results must be carefully interpreted, given that many factors may contribute the observed phenomena. Considering that possibility, we divided the various factors into two different types: (1) kinetic and/or kinematic, related to changes in the orthotic gait movement itself, and (2) neurological factors, namely, supposed changes in the interneuronal activities of the spinal locomotor networks as a result of training. We believe that neither the kinetic/kinematic nor the neurological factors alone can fully explain the observed results

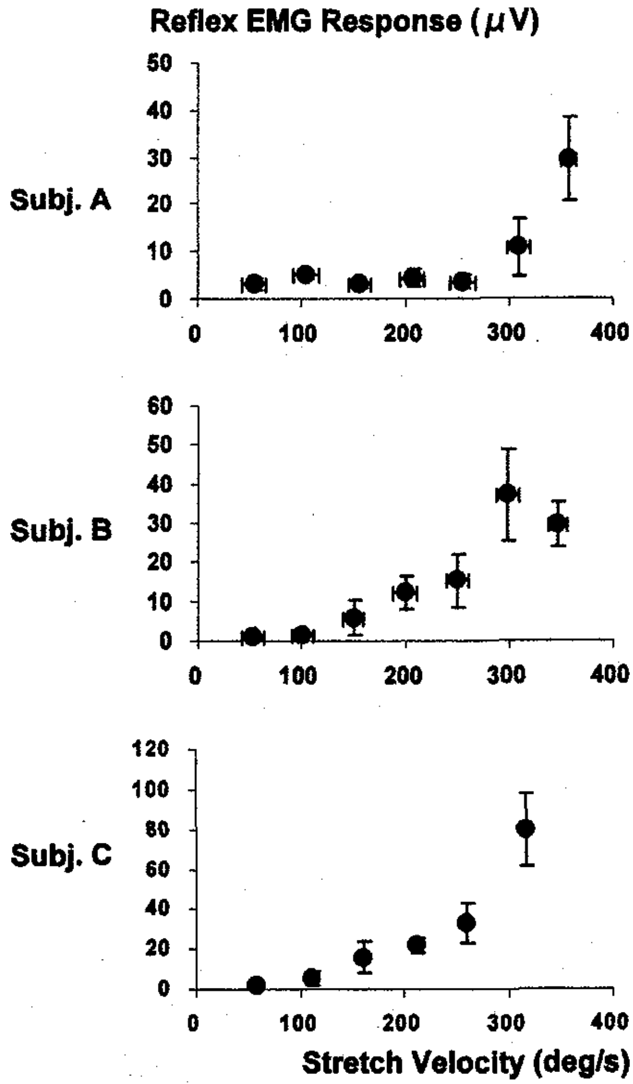

Fig. 5 Relationships between the stretch reflex EMG activities in the SOL and the applied stretch velocity for the three subjects. Note that the estimated threshold velocity to elicit the reflex is over 100 deg/s, which is far faster than the peak ankle joint velocity during the orthotic gait in any of the three subjects

Effects of the orthotic gait motion on modulation of EMG activity

The results of the stretch reflex test in the SOL indicated that the threshold velocities of stretch reflex in the SOL were far higher than the observed ankle dorsiflexion velocity during the orthotic gait. This result strongly suggests that the observed SOL EMGs during the orthotic gait were not induced merely by the spinal stretch reflex. Rather, in the present study, supposed spinal locomotor networks appeared to play a role in the induction of SOL EMGs during orthotic gait.

The observed modulation in the SOL EMG activities synchronized with gait phase confirmed our previous observation that even the knee-locked "stiff-legged" gait with a gait orthosis is effective to induce locomotor-like

EMG activity in lower limb muscles (Kojima et al. 1999). Because the stiff-legged gait is a gait with the knee locked in full extension and the ankle in a neutral position, the afferent information thought to primarily contribute to inducing the observed modulation in the EMG activity would be associated with the hip-joint movement and load on the leg. This result is consistent with recent observations from infant stepping experiments (Pang and Yang 2000) and experiments using a driven gait orthosis for paraplegic subjects (Dietz et al. 2002). Dietz et al. (2002), on the basis of their elegant series of studies on locomotor capacity of human spinal cord and relevant animal and human studies, concluded that "afferent input from hip joints, in combination with that from load receptors, plays a crucial role in the generation of locomotor activity in the isolated human spinal cord". The result in the current study would constitute indirect evidence to support this notion. With respect to afferent input from the hip joint, further, Schimit and Benz (2002) recently demonstrated that imposed hip joint extension/flexion movements in spinal cord-injured subjects induce a unique, stereotypical reflex response in hip, knee, and ankle joints, suggesting that hip movement would activate spinal interneuronal pathways associated with coordinated motor behaviors such as posture and locomotion. Given this notion, the results in the current study suggest that afferent information associated with hip extension during the orthotic gait might activate the spinal neural network responsible, at least in part, for the synchronized EMG activity in the SOL, which may be the common network generating the extensor reflex. With regard to the effect of limb loading on the locomotor activity in the paraplegic human, a growing body of indirect evidence from studies on humans has indicated that load-related afferent inputs play an essential role in the generation of locomotor-like efferent patterns by the human spinal cord (Harkema et al. 1997; Pang and Yang 2000, 2001; Dietz et al. 2002). The observed close relationship between the SOL EMG levels and mGRF during the training period (Fig. 2) and within a single experiment (Fig. 4) in the current study is consistent with that found in previous reports and our previous study (Kojima et al. 1999), in which we demonstrated that the levels of lower limb muscle EMGs during orthotic gait were well correlated to the level of limb loading.

Alteration of the intrinsic property of spinal neural networks due to training

The observed EMG alteration in the SOL might not solely depend on gait-motion changes due to training, but also on alterations in the intrinsic properties of neural networks. This hypothesis is supported by the result of an additional experiment, in which the EMG modulation occurred even under kinematic and kinetic profiles of orthotic gait similar to those of the pre-training gait after 4 weeks of training (Fig. 4). In addition, the effect of changing gait velocity was obviously different before and after several weeks of training, suggesting that the input and output properties in the spinal neural networks during orthotic gait were altered due to the training. Another observation that supports this hypothesis might be the observed EMG changes in subject C, whose orthotic gait motion was at a higher level (i.e., faster velocity and larger ROM, etc.) at the first stage of training and did not largely change during the training period. Nevertheless, modulation of the SOL EMG was induced in this subject, despite rather reduced angular velocities of hip and ankle joints after 8 weeks of training.

Use-dependent plasticity is now a well-known property of spinal neural networks (Hodgson et al. 1994; Muri and Steeves 1997; Raineteau and Schwab 2001). Repeated afferent input accompanying gait training might result in improvement in the transmission efficacy within the neural network responsible for the SOL EMG activity during orthotic gait. This possibility is extremely important with regard to rehabilitation strategy for SCI patients (Field-Fote 2001; Protas et al. 2001). If the act ambulation with an orthosis itself has the potential to improve neuronal activity in the spinal locomotor neuronal networks, a specific gait orthosis one could be designed and developed for locomotor training. Future studies should explore the optimal design for a gait orthosis that can effectively activate the spinal locomotor neural network, using the findings in the current study as the first step in such a series of investigations.

Finally, almost no EMG modulation appeared in the TA in the current study. This might be explained by the following: (1) because the correlation between the load on a limb and EMG activity is less in the TA than in the SOL (Harkema et al. 1997; Kojima et al. 1999), load-related afferent information during the orthotic gait in our study might not have been sufficient to evoke TA activity; and (2) although an imposed hip flexion can induce the ankle flexor response (Schmit and Benz 2002), the flexion amplitude or velocity during the orthotic gait was not sufficient to induce the TA activity. Further, the ankle joint was mechanically immobilized in the orthosis in our study, and the absence of ankle motion might have prevented the elicitation of TA activity. However, further training might induce modulation in TA EMG activity, based on the fact that we observed a reciprocal EMG activity pattern between the ankle extensor and flexor muscles during the orthotic gait in a well-trained SCI subject (Kojima et al. 1998). Further studies are needed to clarify this issue.

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13-84-3　膝関節屈曲ー伸展動作を実現する脊髄損傷者用歩行補助装具の開発
ー膝関節動作の有無による麻痺下肢筋活動の変化ー

A device for the knee motion assistance during orthotic gait for spinal cord injured individuals

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1. はじめに

脊髄損傷者の多くは下肢の運動麻痺のため，日常生活の殆どを車椅子上の座位で過ごす．長期の立位・歩行からの隔離は筋萎縮，骨萎縮をはじめとする身体機能の負の適応を招くことから，脊髄損傷後のリハビリテーションでは，装具を用いた歩行訓練が良く行われる．立位歩行運動により，全身持久性の維持，筋・骨萎縮の防止，免疫の活性化，消化機能の改善などが期待されることから[1]，脊髄損傷後の体力・健康の維持に極めて重要であると言える．

下肢装具を用いた脊髄損傷者の歩行動作では健常者の歩行とは異なり，運動周期全般にわたって膝関節が伸展位で固定されるため，歩行中の膝関節の動的運動が行われない．我々は，より効果的な立位歩行訓練を実現するための一方策として，対麻痺者用の交互歩行装具（ARGO: Advanced Reciprocatig Gait Orthosis）に改良を加え，歩行遊脚期に膝関節屈曲-伸展動作を実現できる動力装置を試作した．

Fig.1 The knee joint actuator mounted on the

2. 目的

本研究では，試作した動力装置による膝関節動作の実現効果を，歩行中の麻痺領域の神経活動の変化の観点から検討することを目的とした．先行研究では既に，脊髄完全損傷者の装具歩行中に麻痺下肢筋に歩行周期に同調した筋活動（以下，歩行様筋活動）が生じることが報告されていることから[2]，膝関節の屈曲-伸展動作の有無によって，筋活動がいかなる変化を示すのかを観察した．

3. 装置の概要

本研究で試作した動力装置は，ARGO の膝関節部にリニアアクチュエータを装備することにより，歩行遊脚期に膝関節屈曲-伸展動作を実現するものであった（Fig.1）．また，ARGO のレシプロ機構（ヒップドライビングケーブル）の末端部にもアクチュエータを装備し，股関節・膝関節モータの回転運動を連携させることにより，両関節の動作位相・時間を制御可能な機構を考案した．以降，膝屈曲有の装具を改良型 ARGO し，通常の装具を 通常型 ARGO と表記する．

4. 実験方法

下肢運動機能に完全麻痺を持つ胸髄完全損傷者５名（第５～１２胸髄損傷）を対象とした．被験者はARGOを用いた３ヶ月以上にわたる歩行トレーニングを経ており，実験時には歩行時の動作はほぼ安定し，歩行周期に同調した歩行様筋活動も高い再現性を示した．

被験者は，通常型ARGO，改良型ARGOによるトレッドミル上での歩行を行った(Fig.2)．本研究では，予め被験者の快適歩行速度を基準に股関節のアクチュエータの動作間隔を決定し，同一の歩行速度条件下での両装具による歩行動作を比較することにした．

両装具による歩行中の麻痺筋EMG活動を以下の筋より計測した．ヒラメ筋（SOL），内側腓腹筋（GAS），前頸骨筋（TA），大腿直筋（RF），大腿二頭筋（BF）．筋活動データは生体アンプにて1000倍に増幅後，A/D 変換器を通して周波数６００Ｈｚにて記録した．同時に、トレッド

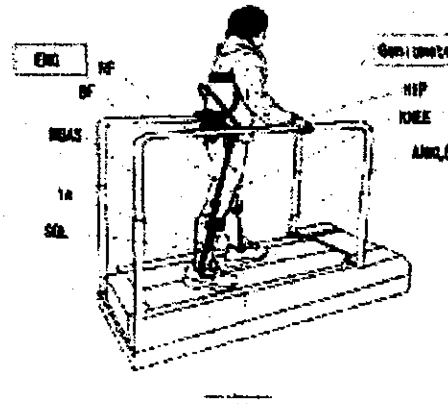


Fig.2 Experimental settings

ミルのベルト部に埋め込まれた3分力床反力計（ADAL-3I
Techmacine製）から歩行中の床反力を，さらに股関節，膝関
足関節に電気角度計（Biometrics製）を貼付することにより，
節角度変位を計測した．各筋から導出した筋電信号は
流・積分処理の後，遊脚期と立脚期それぞれの平均積分
を定量した．

5. 結果

関節角度，筋活動電位，床反力の１０試行分の加算平
波形を算出した（Fig.3）．図中に示すように，改良型AI
では歩行遊脚期に膝関節屈曲伸展運動が実現され，この
作に伴って下肢の筋活動に変化を認めた．膝関節の動作
除いては，両装具による歩行中の股関節・足関節の動
床反力には装具間に大きな差異は認められなかった．

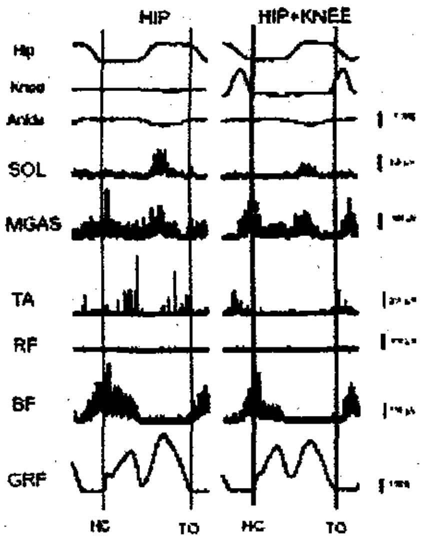
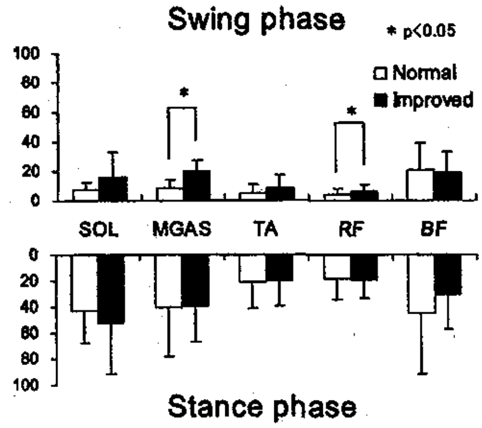


Fig.3 Typical data of the knematics (hip, knee, and ankle joint), electromyographic activity (SOL, MGAS, TA, RF, BF), and GRF during orthotic gait in a paraplegic patient. (HC: Heel contact, TO: Toe off)

Fig.4には被験者５名の歩行遊脚期，立脚期における
筋の積分値の平均値を示した．膝関節動作に伴う筋活動
変化は，必ずしも全ての被験者に共通の変化を認めなか
たが，遊脚期における膝関節の屈曲伸展により直接筋長
変化がもたらされる大腿筋群，腓腹筋に変化を認める被
者が多かった．遊脚期における腓腹筋，大腿直筋の活動
両装具間に有意な差を認めた．

6. 考察

本研究では，交互歩行装具ARGOの膝関節部にアクチ
エータを搭載することで，通常の装具歩行動作では実現
れない膝の屈曲･伸展動作を可能にし，実現された膝関
動作に伴って歩行様筋活動が変化するのか否かを検討



4 Comparison of the muscle activity between two types
rthosis in each lower limb muscle

膝関節動作伴う変化を認めた大腿直筋・腓腹筋はいず
膝関節の動作に関与する筋であり，膝関節の動作に伴
筋長変化が生じるものと考えられる．したがって，本
で認められた両装具間の筋活動の差異は装具歩行中
関節動作に伴って発現する末梢性神経入力の変化に
ものと考えられる．本研究による膝関節動作付与の試
動作の面では床面とのクリアランス確保や運動効率
上の効果を持つものと考えられるが，本研究の結果は，
節動作の実現が麻痺領域の神経活動を変化させるこ
示すものであった．

論

装具における麻痺下肢の歩行様筋活動の差異は，運動
の変化に伴う末梢性感覚入力の変化に応じて，麻痺領
支配する脊髄神経回路が可変的に応答することを示
果であった．したがって，動力装置による膝関節動作
現は，脊髄損傷者の麻痺領域の神経活動を賦活する方
なり得るものと考えられる．

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13-84-4　膝関節屈曲 - 伸展動作を実現する脊髄損傷者用歩行補助装具の開発

- 装置の概要とトレッドミル上での歩行による動作評価 -

A walking assistance equipment providing knee flexion-extension movement in the orthotic gait for spinal cord injured patients

-Outline of the device and evaluation with walk on treadmill-

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1．はじめに

脊髄損傷者の多くは移動手段として車椅子を使用する．しかし一方で，立位歩行からの隔離は身体機能に様々な負の適応をもたらすため，装具等を用いた歩行訓練の重要性は極めて高いと考えられる．

下肢装具を用いた脊髄損傷者の歩行動作は健常者のそれとは異なり，運動周期全般にわたって膝関節が伸展位で固定される．この点は歩行遊脚期における下肢振り出しの際に床面とのクリアランス確保を困難にする上，装具歩行の運動効率を悪化させる可能性がある．そこで本研究では，対麻痺者用の交互歩行装具 (Advanced Reciprocating Gait Orthosis) に改良を加え，歩行遊脚期に膝関節屈曲-伸展動作を実現できる動力装置を考案・試作することとした．

2．目的

本研究では，交互歩行装具 ARGO の膝関節・股関節部にアクチュエータを装備することにより歩行遊脚期に膝関節の屈曲-伸展運動を実現できる動力装置を開発し，トレッドミル上での歩行動作の評価を行うことを目的とした．

3．装置の概要

膝関節屈曲-伸展動作は ARGO の膝ジョイント部（両脚）にリニアアクチュエータ（DC モータとボールネジの組み合せ）を装備することによって実現し，立脚期には装具のロック機構を利用して伸展位を保持する機構を考案した．また，ARGO のヒップドライビングケーブルの末端（片側のみ）に別途リニアアクチュエータ（Minimotor3042 並びに減速機 30/1 66:1，Faulhaber 社製，および電動プッシャーEP40，旭精工株，全 1.3kg）を取り付けることにより，歩行遊脚期における股関節屈曲動作を補助する機構を考案した．

股関節動作と膝関節動作を連動させるために，シーケンサを用いて３つのアクチュエータの動作を制御し，対象者の歩行速度に合わせて動作位相・時間の設定を行った（以降，膝屈曲有の装具を改良型 ARGO とし，通常の装具を 通常型 ARGO と表記する）．Fig.1 に膝関節屈曲-伸展機構の概観を，Fig.2 に両装具での歩行の様子を示す．

4．評価実験

ARGO による歩行訓練を３ヶ月以上経験し，装具歩行動作に習熟した脊髄損傷者（胸髄完全損傷）６名を対象とした．２種の装具によるトレッドミル上での歩行を実施し，トレッドミルのベルト部に埋め込まれた3分力床反力計（ADAL-3DC, Techmacine 製）から歩行中の床反力を，さらに股関節，膝関節，足関節に電気角度計（Biometrics 製）を貼付することにより，関節角度変位を計測した．予め被験者の快適歩行速度を基準に股関節のアクチュエータの動作間隔を決定し，同一の歩行速度条件下での両装具による歩行動作を比較した．

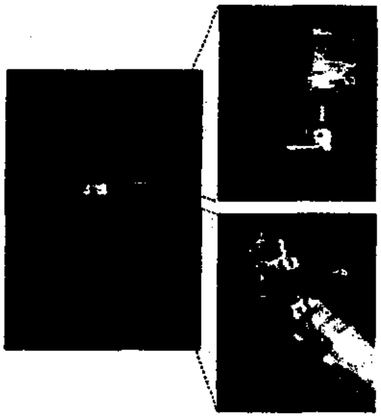

Fig.1 ARGO (left) and the devices mounted on the hip (right: above) and knee joint (below)

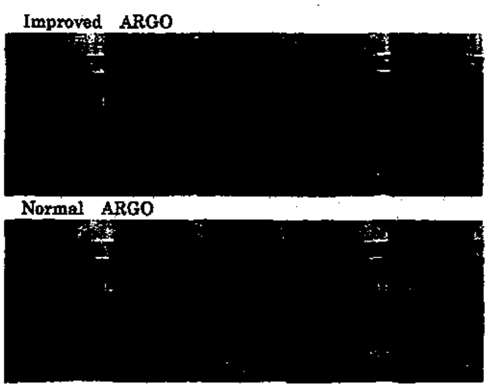

Fig.2 Gaits with the normal and improved ARGO

５．結果及び考察

Fig.3 に通常型（左）および改良型 ARGO（右）を装着して歩行した際の床反力，関節角度変位の典型例（Th８完全損傷者の歩行１０歩分の加算平均波形）を示す．改良型 ARGO では，膝関節が伸展位で固定された通常型 ARGO による歩行時と同様の床反力特性，股関節動作を保持しつつ，歩行遊脚期における膝関節の屈曲-伸展動作が実現されていることが分かる．

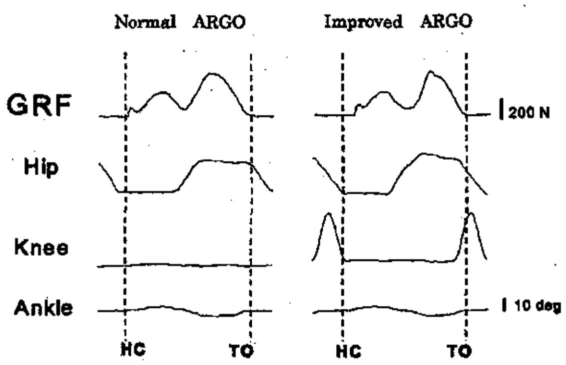

Fig.3 Changes in the ground reaction force (GRF) and the joint angles during a gait cycle. (HC: Heel contact, TO: Toe off)

また Fig.4 に歩行１周期における股関節，膝関節，足関節の最大可動域（上段），平均屈曲角速度（下段）の平均値を示す．角速度については，股関節は遊脚期における屈曲角速度，膝関節は同じく遊脚期の伸展角速度，足関節は立脚期における背屈角速度を定量した．

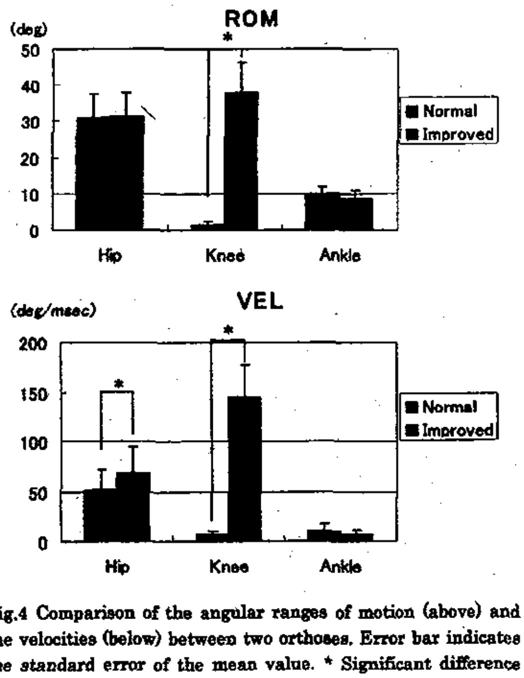

Fig.4 Comparison of the angular ranges of motion (above) and the velocities (below) between two orthoses. Error bar indicates the standard error of the mean value. * Significant difference (P<0.05).

その結果，関節可動域については，股関節，足関節では両装具間に有意差を認めず，動力化により他の動作を減ずることなく膝関節動作が実現されたことが明らかとなった．一方，股関節の屈曲角速度については有意差が認められ，動力化により増加する結果を示した．股関節が屈曲する局面と膝関節が伸展する局面は歩行遊脚期の同位相であることから，膝関節の伸展動作が加わったことで慣性が大きくなった結果，股関節の屈曲角速度が増加したと考えられる．

６．まとめ

股関節・膝関節動作に動力補助を加えた改良型 ARGO による歩行では通常の装具歩行時と同等の股関節可動域，床反力特性を再現しながら，歩行遊脚期初期に膝関節の屈曲-伸展動作を実現することが可能であった．本研究では直接的に検討できなかったが，運動周期に応じて膝関節の動的運動が実現されることで，床面とのクリアランス確保，運動効率の改善等の効果が期待できる．また，運動周期に応じて膝関節の動的運動が実現されることから，関節運動に伴って膝関節周辺の麻痺筋にも張力変化が生じるものと考えられ，麻痺領域の神経活動励起などの生理学的側面への影響も考えられる．

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健康・福祉工学におけるME技術の応用と今後の展開
脊髄損傷者のための歩行支援技術

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Assistive technologies for locomotion in spinal cord injury patients

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脊髄損傷と歩行：脊髄損傷者は移動に際し車椅子利用が前提となるが，車椅子のみに頼り歩行放棄すれば，長期的に呼吸循環機能，筋量，骨密度，免疫，消化機能等は低下し，また心理的側面も問題となる．この問題に対し補助装具を用いた自立歩行が心身両面に良好な効果を及ぼすことが確かめられており，日常で簡便に利用可能な支援機器が求められる．しかし現在までに開発されてきた歩行補助装具の多くは，着脱の煩雑さ，少ない関節自由度，無動力源などから，歩行労力も大きく日常的に利用されているものはない．

動力化歩行器の開発：これに対する試みとして，著者らは写真１に示す動力化歩行器の開発を進めてきている．これは市販の歩行装具 Advanced Reciprocating Gait Orthosis の股・膝両関節に小型 DC モータを搭載し，両関節の駆動アシストを実現したもので，充電電源の利用などによりコンパクトな歩行支援器となっている．これを用い脊髄損傷者４名を対象に装具歩行動作を計測し，動力化の有効性（歩行能力，労力減少，免疫能力など）を確認した．更に重要な点として，歩行中の麻痺筋の活動量計測を行った結果，装具歩行中の kinematics の変化により求心性入力が変化し，それに応じて歩行様筋活動が変化することを見出した．これは，従来，受傷後変わらないと考えられていた脊髄神経パターンが歩行訓練により活性化・可塑変化することを意味しており，旧来のリハビリテーションの考え方に大きく変更を迫る画期的な発見と考えられる．この結果により，もはや自立歩行は脊髄損傷者にとって単に心身面での効果があるのみでなく，神経筋機能の回復，すなわち脊髄内に存在する歩行のための Pattern Generator(PG)の再活性化にも効果をもたらす可能性が見えてきたといえるため，写真２に示すように，より長時間の安定した歩行訓練を実現すべく，動力化歩行器をトレッドミル上でベルト速度に合わせて制御するよう改良した．現在 T5～T12 の計７名の被験者を対象に歩行 PG への効果の検討を進めている．

まとめ：開発した動力化装具の２つの展開可能性につき述べた．一つは，完全対麻痺者に対する心身機能維持・二次障害防止を目的とした利用であり，今後，現実レベルで簡便に自立歩行の機会が提供可能な歩行装具デバイスを開発する
がある．ただし脊損者の移動を現実的に考え
移動速度，社会的受入態勢から車椅子が現状
圧倒的に有利であり，車椅子との併用利用を
とした歩行器を実現する必要がある．もう一
完全・不完全麻痺に対し，多くの研究データ
部強制歩行が歩行 PG の再賦活を促進する
を示しており，完全対麻痺者における再賦
ニズム・可塑的変化を解析するとともに，そ
果を歩行再建に向けて生かすための具体
を検討する必要がある．なお，この歩行機能
システムの拡張応用として，今後急増が予想
る自立歩行困難な高齢者・高齢障害者の立位
トレーニングへの利用も可能と考えられ，
り高齢者・障害者数の減少，ひいては医療
護費用の削減に大きく貢献可能と考えられ

写真１　動力化歩行器

写真２　トレッドミル上での動力化装
よる歩行訓練

N15

トルクユニットによる松葉杖のパワーアシスト

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Power assist control of crutches by using torque unit

Toyo University Asami Muraki, Tomomi Naka, Akihiro Matsumoto, Ochanomizu University Yuji Ohta

A pair of crutches for the walking assistance that are equipped with motor has been developed in this paper. Power assist is obtained by the conservation law of angular momentum using torque unit. The benefit of this approach is the ease of maintenance and the ease of changing dynamics. In this report, the prototype has been designed and evaluated.

1.はじめに

歩行における安定性や持久力は加齢や疾病による身体・精神機能の低下に伴って低下する。下肢の筋力が弱く体重を支えられない、下肢の骨折で患部に体重をかけられないなどの場合に松葉杖が使われている。握りを持ち、横木を胸の横で固定するので支持性に優れているが、十分な上肢の筋力が必要である[1][2]。そのため、筋力の弱い高齢者の場合松葉杖を使うのは困難であると思われる。

本研究では、Torque Unit Manipulator(以下TUM)に着目し、松葉杖をパワーアシストしようと考えている。そこで、トルクユニットを製作し、どれくらいのアシストトルクが発生するか実験的に求める。

2.トルクユニット

2.1 トルクユニットとは

トルクユニットとは、宇宙構造物へのアクチュエータとして提案されたもので、慣性ロータとアクチュエータから構成されておりロータに角加速度を与え、それによって生じる反動トルクを駆動力とするアクチュエータである[3]。トルクユニットは各リンクの任意の位置に装着されており、トルクユニットで発生するトルクは、関節にトルクを与えるのではなく、装着されているリンクに直接トルクを与える装置である。

TUMの実現において次のような利点が生まれるものと予想される。

(1) 保守の容易性：アクチュエータは、リンクとは独立に存在しており、マニピュレータから脱着が自由である。よって、アクチュエータが故障した場合には、トルクユニットを交換するだけでよい。

(2) 動特性の変更可能性：トルクユニットはマニピュレータの任意の位置に装着することができるので、装着位置によりマニピュレータの動特性を容易に変えることができる。

2.2 トルクユニットの原理

Fig.1において、mを松葉杖の質量、Iを松葉杖の重心周りのモーメント、rを接地点から松葉杖の重心までの距離、Mをトルクユニットを含めた松葉杖の質量、m_dを円盤の質量、I_dを円盤の慣性モーメント、l_dを松葉杖の回転軸から円盤の回転中心までの距離、θを松葉杖の回転角度、φを円盤の松葉杖に対する相対回転角度、τをトルクユニットのトルクとする。角度はすべて反時計回りを正とし、運動方程式を導くと、

$$J\ddot{\theta} + I_d\ddot{\varphi} = Mgr\sin\theta \qquad (1)$$

$$(J = I + I_d + mr^2 + m_d l_d^2)$$

となる。松葉杖の発生するトルクτは以下のように表せる。

$$\tau = I_d\ddot{\theta} + I_d\ddot{\varphi} + Mgr\sin\theta \qquad (2)$$

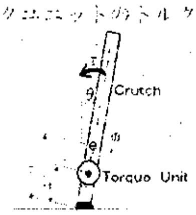

Fig.1 TUM

3.松葉杖のモータ周り・円盤の設計

モータの支持具の材料は、厚さ2.5㎜のアルミの角材を使用した。駆動力の伝達にはプーリと歯付きベルトを用いた。モータは、Maxon社のDC 90Wで減速比4.8のギアヘッド付きのものを使用する。ロータとして厚さ10mm、直径120mmのもの(円盤1)と、同じく厚さ10mmで、直径160mmのI型断面(Fig.4)のもの(円盤2)を設計した。円盤2は円盤1とほぼ同じ質量で慣性モーメントを増やした断面形状とした。Table 1にそれぞれの円盤の慣性モーメントを示す。

Table 1 円盤の比較

円盤	直径[mm]	質量[kg]	慣性モーメント[kgm²]
1	120	1.92	0.0035
2	160	1.93	0.0071

Fig.2 松葉杖全体

Fig.3 モータ周り

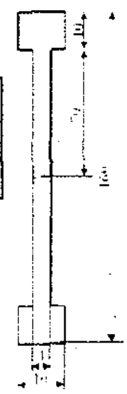

Fig.4 円盤の断面

4.スイッチ

4.1スイッチのタイミング

Fig.5 左より、松葉杖の傾きによってスイッチのタイミングを決める。人間が杖を支えにして、体を前方に持っていく時、約80°時に(Fig.5 右)ONになるようにスイッチを設定し、ON OFFさせる。

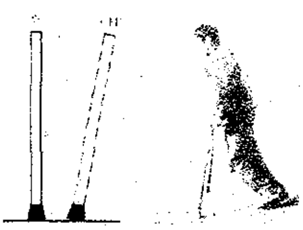

Fig.5 タイミング

4.2 回路図・フローチャート

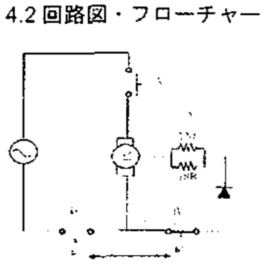

Fig.6 回路図

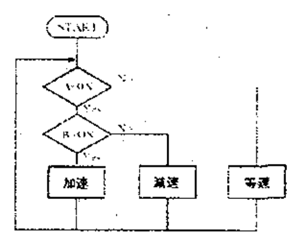

Fig.7 スイッチの流れ

スイッチは手先用(A)のスイッチと杖先のスイッチ(B)を使用する。Fig.6より、Bのスイッチだけだと地面に接地している間(b=On、b'=OFF)モータが回り続けてしまうため、Aのス

イッチで電源の供給 ON/OFF する．また，慣性モーメントで電源を OFF にしても止まるまで時間がかかるため，ブレーキをかける必要がある．スイッチの流れを Fig.7 に示す．ここで加速とは，円盤が最大角速度になるまで加速し，減速とは，速度ゼロになるまで減速させ，等速とは，電源を OFF にした状態(フリーモーション)を示す．

5.特性の測定

5.1 モータの角加速度の測定

松葉杖を鉛直上向きの状態で，20V 8A でモータを稼動した．モータに付属するタコジェネレータから，角速度を測定し，そこから角加速度を測定する．データの取り込みには KEYENCE 社製の NR-2000 を用いた．これを円盤 1，円盤 2 両方の場合において測定する．その結果を Fig.8 に示す．

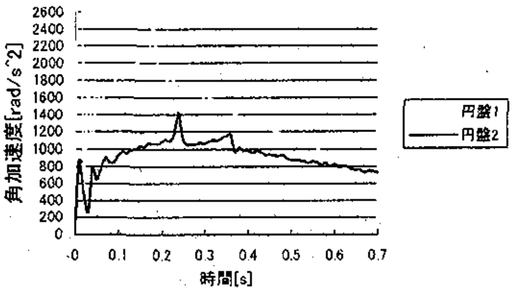

Fig.8 モータの角加速度変化

部分的にノイズが生じているのは，ベルトのたるみとギアのバックラッシュによる非線形項であり，モータの最大角加速度は，円盤 1 で 2.0×10^3 rad/s^2，円盤 2 で 1.1×10^3 rad/s^2 となり，これに減速比 4.8 がかかり，円盤 1 では 416.7 rad/s^2，円盤 2 では 229.2 rad/s^2 となった．

5.2 杖の角加速度測定

松葉杖にマークを 2 ヶ所取り付け，モータを稼動させた際の杖の動きを撮影した．撮影には Photron 社製 FASTCAM Rabbit-mini (250fps) の高速度カメラを用いた．この撮影された画像を画像処理ソフト HALCON によって画像解析し，各マークの座標値を求めた．2 ヶ所のマークの座標値より，角加速度を計算した．松葉杖の角加速度変化は Fig.9 のようになった．

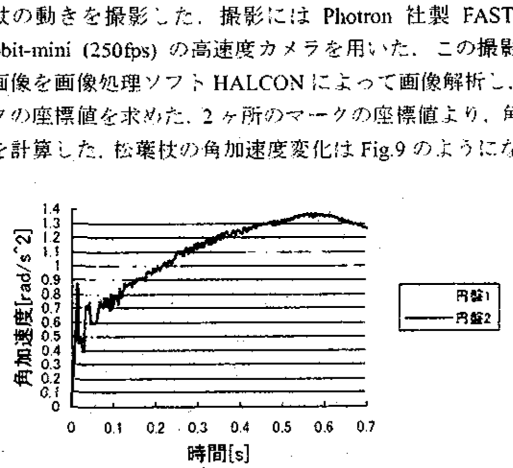

Fig.9 松葉杖の角加速度変化

5.3 松葉杖のトルクアシスト量の測定

Fig.10 のように杖を鉛直にたて，モータに 20V 8A を与え，地面との接触部 1m の地点での松葉杖の発生力の最大値をした．測定はバネ秤によって測定し，バネ秤は杖と垂直に配置した．荷重の変化量からトルクを計算した．これを 10 回行い平均を求めた．この結果，発生トルクは，

Fig.10 実験風景

円盤 1 では 1.63 N·m，円盤 2 では 2.18 N·m となった．

5.4 動特性の測定

5.3 節の測定と同様に松葉杖を鉛直にたて，モータに 20V 8A を与え，円盤 2 のトルクユニット装着位置が接地点から 0.17 m と 0.37m における松葉杖の発生力の最大値を測定した．これを 10 回行い，平均を求めた．この結果発生トルクは，接地点から 0.17m の地点では 2.18 N·m，接地点から 0.37m の地点では 1.72 N·m となった．

6.発生するトルクの計算

円盤 1 に関しては，モータの角加速度 416.7 rad/s^2，この時の杖の角加速度は 1.03 rad/s^2，杖の傾きは 2.0° となった．TUM の慣性モーメント 0.0035 kg·m^2，杖全体の慣性モーメント 0.4559 kg·m^2，より発生されるトルクの量を式(2)から求める．杖の長さは 1.22m とし，TUM 装着位置は地面から 0.17m，杖の重心は 0.275m，重さは 3.6kg であった．ここから発生トルクを計算すると 1.80 N・m となった．

円盤 2 に関しては，モータの角加速度 229.2 rad/s^2，この時の杖の角加速度は 1.06 rad/s^2，杖の傾きは 3.5° となった．TUM の慣性モーメント 0.0071 kg·m^2，杖全体の慣性モーメント 0.4595 kg·m^2，より発生されるトルクの量を式(2)から求める．杖の長さは 1.22m とし，TUM 装着位置は地面から 0.17m，杖の重心は 0.275m，重さは 3.6kg であった．ここから発生トルクを計算すると 2.21 N・m となった．

円盤 1 と円盤 2 を比較するために，上記の計算値と 5.3 節に示した実測値を Table 2 に示した．

Table 2 トルクの比較

トルク[N·m]	円盤 1	円盤 2
計算	1.80	2.21
実測	1.63	2.18

予想通り，円盤 2 の方が実測・計算値 共に，円盤 1 よりも高い値を示した．また，計算式で算出した値と ほぼ同様の値を示した．実測値が計算値より小さい値を示したのは，摩擦などで消費したためと考えられる．また 5.4 節の実験結果から，装着位置が接地点に近いほど，トルクが大きいことを利用して，ユーザが動特性を変化できる．

7.考察

TUM を使って，松葉杖のパワーアシストが可能であることが分かった．ただし，パワーアシストするために必要な電流値が大きいので，バッテリーに対しての要求が大きくなる．また，全質量が重くなってしまった．そこで，改良点としてはトルクユニットの慣性モーメントを保ちつつ，軽量化すること必要がある．

8. おわりに

今回トルクユニットを製作し，どれくらいのトルクが発生するかを実験し，理論値との比較をした．今後は 現在は松葉杖 一本で計測を行っているので，二本での実験を行い，実際に人間が松葉杖を使用した際の TUM の効果を検証し，歩行に合わせた制御タイミングの検討を行う．

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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 EMG activity in the bilateral alternate leg movement condition. These results suggest that the sensory information generated by alternate leg movements plays a substantial role in amplifying the induced locomotor-like muscle activity in the lower limbs.

INTRODUCTION

It is now well recognized that the paralyzed lower limb muscles of a person with spinal cord injury (SCI) can be activated by body weight–supported stepping movement on a treadmill (Dietz et al. 1995, 2002; Dobkin et al. 1995; Ferris et al. 2004; Harkema et al. 1997) or by locomotion with the use of a gait orthosis (Kojima et al. 1999; Nakazawa et al. 2004). Since the stepping movement accompanies joint rotation, it is possible that such muscle activity might merely reflect the reflex responses induced by rhythmic muscle-tendon stretches (Dobkin et al. 1995; for review, see Harkema 2001). Nevertheless, this muscle activity is considered to come from the interaction of the central pattern generator (CPG) in the spinal cord (Dimitrijevic et al. 1998) with the sensory input rather than a mere reflex response (Dietz et al. 2002).

This interpretation is justified by several recent findings. Harkema et al. (1997) have shown that the magnitude of EMG activity induced in SCI persons is more closely related with the peak load applied to the leg than with muscle-tendon stretch. Similarly, Dietz et al. (2002) have shown, using a driven gait orthosis, that the ankle muscle activity observed in ordinary locomotion movement was not induced either when the ankle joint alone was moved or when the body weight was completely unloaded. The crucial role of the sensory information of load and hip position in generating and/or shaping the rhythmic output pattern from the CPG has been considerably emphasized by previous studies using reduced animal preparations (Duysens 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

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spinal cord somewhere between T_5 and T_{12}, 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. 1*A*) 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 ($\pm$17.5 cm from default position) in a sinusoidal manner (Fig. 1*C*) by matching the movement frequency with the sound of a metronome (1 Hz). This handle movement could induce approximately $\pm$14 and $\pm$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. 1*A*). 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 $\geq$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*

Subject	Age, yr	Weight, kg	Lesion Level	Grade of ASIA	Duration of Paraplegia, mo
S1	19	53	T_5	B	24
S2	30	74	T_8	A	12
S3	24	75	T_{12}	A	24
S4	32	68	T_{12}	A	32
S5	22	70	T_{10}	A	11
S6	39	67	T_{12}	A	13
S7	35	55	T_6	A	46
S8	27	75	T_{12}	A	22
S9	30	72	T_{11}	B	23
S10	30	60	T_8	A	17

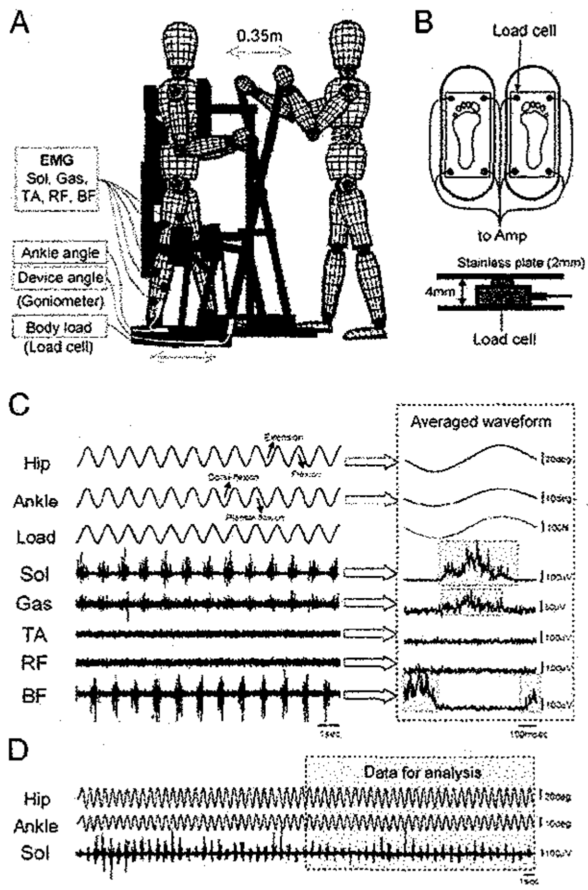

FIG. 1. *A*: overview of the experimental setup. During the experiment, the spinal cord injury (SCI) subject stood securely on a special device (Easy Stand Glider 6000, Altimate Medical). The subject's legs were moved passively by moving the lever back-and-forth in a sinusoidal manner in pace with the 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, T_{12}) 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