表 2　胸髄完全損傷者の装具歩行中の速度，平均心拍数，定常状態酸素摂取量，エネルギー消費量，エネルギーコスト

	Lesion level	Grade of ASIA	Gait Speed (m/min)	HR (beat/min)	VO 2 (ml/kg)	E consmp. (J/kg/sec)	E cost (J/kg/m)
A	T12	B	19.29	135.5	20.24	6.81	21.18
B	T12	A	20.06	99.2	16.01	5.39	16.12
C	T12	A	32.58	114.3	17.63	5.93	10.93
D	T12	A	27.22	140.2	14.91	5.02	11.06
E	T11	A	21.55	129.5	15.62	5.26	14.63
F	T10	A	19.99	132.5	24.20	8.14	24.44
G	T10	A	18.35	131.5	15.41	5.19	16.95
H	T8	A	11.64	110.1	16.75	5.64	29.05
I	T7	A	15.58	163.0	15.19	5.11	19.67
J	T6	A	17.09	143.7	24.83	8.35	29.33
K	T5	B	14.69	166.4	21.14	7.11	29.06
			19.8	133.2	18.4	6.2	20.2
			5.8	20.5	3.7	1.2	7.0

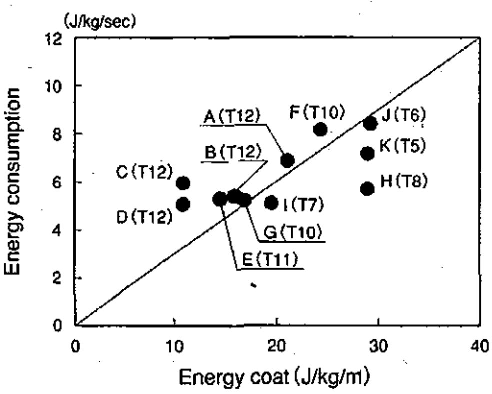

図 7　損傷高位別にみた装具歩行中のエネルギー消費量とエネルギーコストの関連
各プロットと原点を結ぶ直線の傾きが歩行速度を反映している．

度，PCI を比較すると，歩行速度は開始時の 140〜170%に増加し，PCI は 50〜60%に減少することがわかる．これらの変化は訓練によって装具歩行動作が獲得された結果，歩行中の運動効率，さらには呼吸循環機能が改善されたことを反映しているものと考えられる．とはいえ訓練初期は身体的負担度，とりわけ上肢に懸かる負担が大きく，数分の歩行訓練で疲労困憊に至るケースがほとんどである．この点は，受傷後のリハビリテーション過程における装具歩行実施を妨げる原因ともなるが，数カ月の訓練を経ることで，身体機能の維持・向上に適した運動強度での歩行が可能になることを考え合わせると，初期の身体的負担をあらかじめ認識したうえで，中長期的な装具歩行訓練を行うことも身体機能の維持・向上のためには大きな意義を持つものと考えられる．

表 2 には 10 週以上の装具歩行トレーニングを経た胸髄完全損傷者 11 名（T5-12）の歩行中の心拍数，酸素摂取量，エネルギー消費量およびエネルギーコストを示す．装具歩行中のエネルギーコストは，健常者の歩行中の値（2〜3 J/kg/m）と比較すると著しく高い値を示したが，酸素摂取量，心拍数から推察される運動強度は有酸素性作業能の向上に資する適正な範囲にあるものと考えられた．

2　損傷高位の違いによる装具歩行中の身体的負担度の変化

図 7 は表 2 のうち，横軸に歩行中の運動効率を反映するエネルギーコスト，縦軸にエネルギー消費量をとり，各被験者の値をプロットしたものである．原点からおのおののプロットへの直線は歩行速度を反映し，傾きが大きいほど歩行速度が速いことを示す．高位損傷者のプロットがグラフの右側に分布していることから明らかなように，損傷部位が高位に及ぶほど，装具歩行中のエネルギーコストが悪く，歩行速度が遅いことがわかる．リハビリテーション現場でもすでに経験的に理解されているように，装具歩行では損傷高位によっ

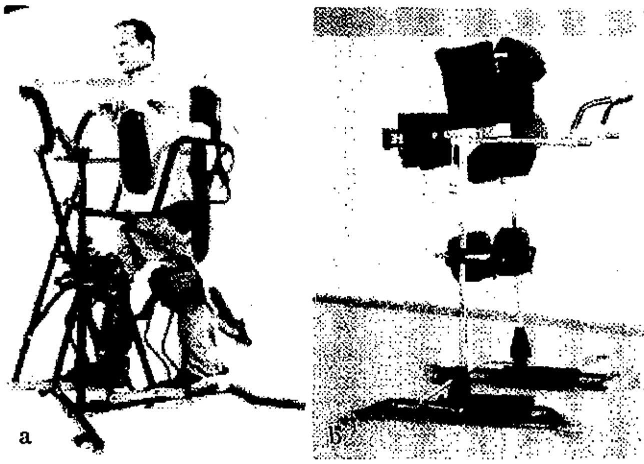

図 8　比較的簡便に立位歩行運動を実現できる装置
a：EasyStand 6000 Glider（Altimate Medical 社のホームページ http://www.altimatemedical.com/ より）.
b：Parapodium（Singlevision 社のホームページ http://www.singlevision.co.uk/concept.htm より）.

て歩行運動の実現可否が左右される．まして，頸髄損傷者では装具歩行の実現が困難であることから，この点は装具歩行の限界と言えよう．

ただし，立位歩行運動の必要性は，むしろ障害による身体機能への影響が深刻な高位損傷者ほど高い場合も考えられることから，頸損者を含む高位損傷者であっても立位歩行訓練を可能にするなんらかの方策が求められる．

3 装具歩行に代わる立位訓練

前項までに記したように装具歩行は習熟までに一定期間の訓練を要し，さらには頸損者や胸髄高位の損傷者では装具歩行の実現が困難であることから，立位歩行訓練の対象となるべき脊髄損傷者の総数に占める装具歩行の実施率はきわめて低くならざるをえない．

現在では比較的簡便に立位歩行運動を実現する装置がすでに開発されている．図 8 a の装置はアメリカの EasyStand という製品で，その名の通り脊髄損傷者であっても容易に立位姿勢を保持した状態で，上肢によるレバー操作によって脚の交互動作が可能である．当センター病院ではこの装置を用いた立位歩行運動の利点に注目し，すでに脊髄損傷後のリハビリテーションにおける運動療法で活用している．胸髄損傷者はもとより，運動麻痺が上肢に及ぶ頸髄損傷者であっても立位姿勢保持下での脚のダイナミックな運動が可能であることから，装具歩行が困難な高位損傷者であってもすでに述べたような効果が得られるものと期待される．実際にこのような運動が脊髄損傷者の身体機能維持・向上にどの程度貢献するのかについては，現在研究を進めているところである．

EasyStand は身体の移動を伴わないが，図 8 b の Parapodium という装置は左右への身体重心の移動と上肢のレバー操作によって歩行に近い運動を実現するための装置である．この装置による運動は装具歩行ほど長期の訓練と技術を必要せず，かつ遊具的な要素も兼ね備える利点も持ち合わせている．

おわりに

適切な強度での立位歩行運動が実現可能であれば，脊髄損傷後の身体機能に対して立位歩行運動の実施がきわめて高い効果を持つことは明らかである．しかし現状では，脊髄完全損傷者の立位歩行運動を実現するため方策が限られ，装具歩行や機能的電気刺激を用いた歩行訓練では運動効率の

悪さ，装具や装置の煩雑さなどの問題が十分な歩行運動の効果を得るうえでの制約となっている．したがって，より多くの脊髄損傷者の立位運動の実現と，それに伴う脊髄損傷者の積極的な健康維持・増進を実現するためには，科学的裏付けを得るための実証的研究，医療従事者と患者双方の立位運動の効果に対する認識の定着はもとより，有効な運動強度を実現するための装置の開発なども，きわめて重要となるものと考えられる．

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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 (±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 ±14 and ±9° motion in the hip and ankle joints, respectively (these values of range of joint motion depend on the subject's lower limb length). This range of motion of each joint is similar to the data for the normal walking provided by Winter (1990). Although only reciprocal leg movement can be induced at the default setting, synchronous or unilateral leg movement can be induced by removing the bolts that connect bars to both sides.

Experimental protocol

Before the experiment, we checked that the standing posture was stable and that no hypotension was observed. First, bilateral alternate leg movement was imposed for 3 min so that the subjects could experience the standing posture and the imposed leg movement. Then, the experiments were performed under the following three conditions: *1*) bilateral alternate (anti-phase) leg movement; *2*) unilateral leg movement; and *3*) bilateral synchronous (in-phase) leg movement. In the unilateral leg movement, the right leg was moved while the position of the left leg was fixed to be vertical. In the bilateral synchronous leg movement, both legs were passively swung simultaneously in the same direction. Throughout the exercise period, subjects were asked to grasp the bar in front of them and to keep their upper limbs relaxed (Fig. 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 ≥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

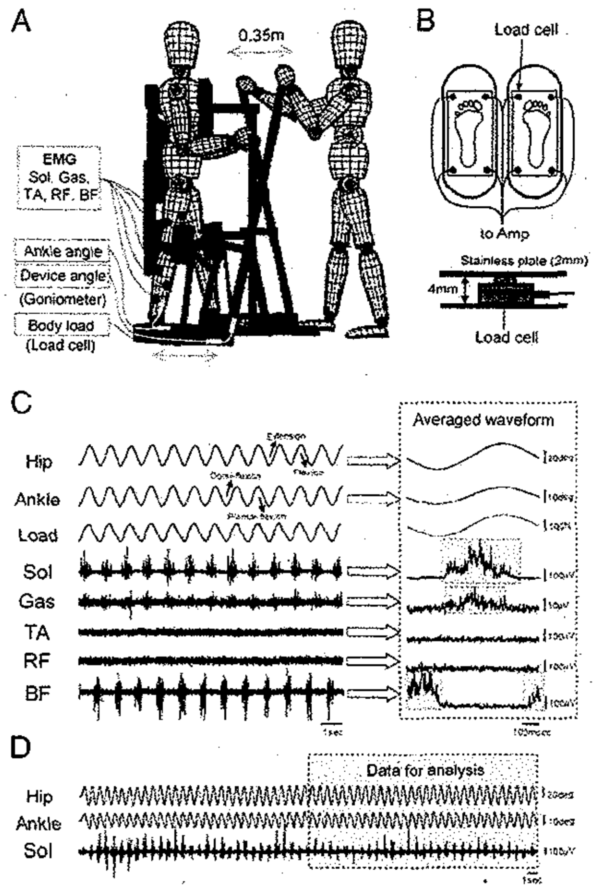

FIG. 1. *A*: overview of the experimental setup. During the experiment, the spinal cord injury (SCI) subject stood securely on a special device (Easy Stand Glider 6000, Altimate Medical). The subject's legs were moved passively by moving the lever back-and-forth in a sinusoidal manner in pace with the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

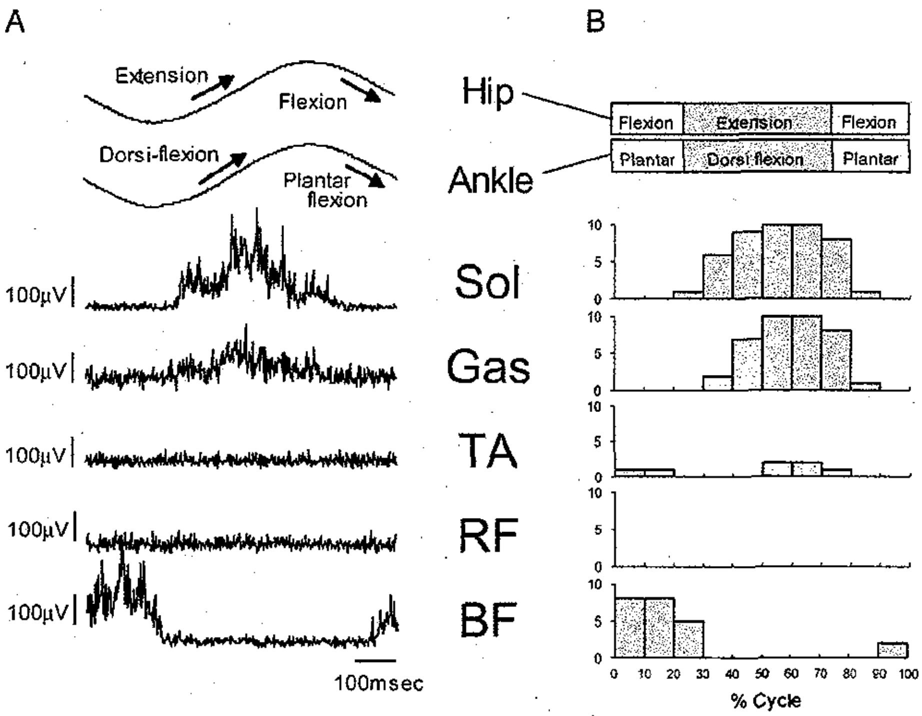

FIG. 2. *A*: ensemble averaged waveform of hip and ankle joint motion and induced EMG activity in each lower limb muscle obtained from an SCI patient. *B*: probability of the occurrence of EMG activity in the lower limb muscles during passive leg movement. Data show number of subjects who showed EMG 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 ± 3.8% in the unilateral leg movement condition and 64.3 ± 12.5% in the synchronous leg movement condition. On the other hand, the peak-to-peak load applied to the left leg was 22.5 ± 4.4% in the unilateral leg movement condition and 69.9 ± 11.9% in the synchronous leg movement condition compared with the alternate leg movement condition.

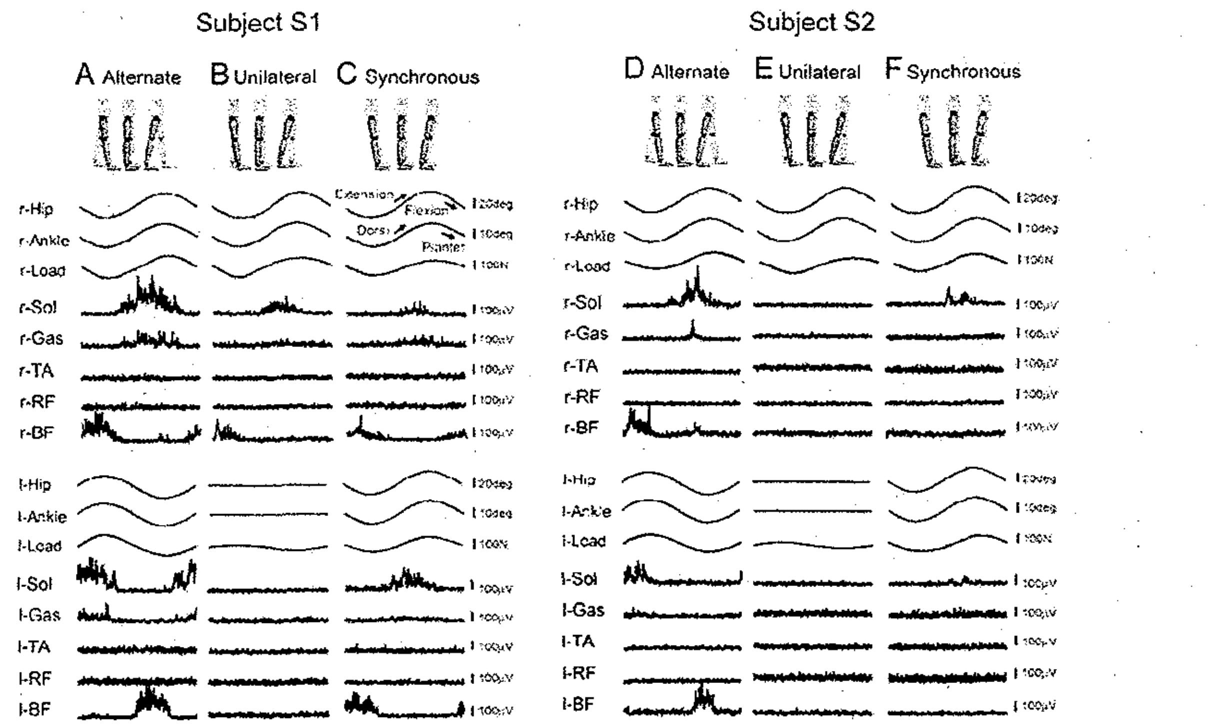

FIG. 3. Averaged waveforms of hip and ankle joint motion, load on the leg, and induced EMG activities of limb muscles recorded from subjects S1 and S2 (*top*, right leg; *bottom*, left leg). *A* and *D*: alternate leg movement condition. *B* and *E*: unilateral leg movement condition. *C* and *F*: synchronous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

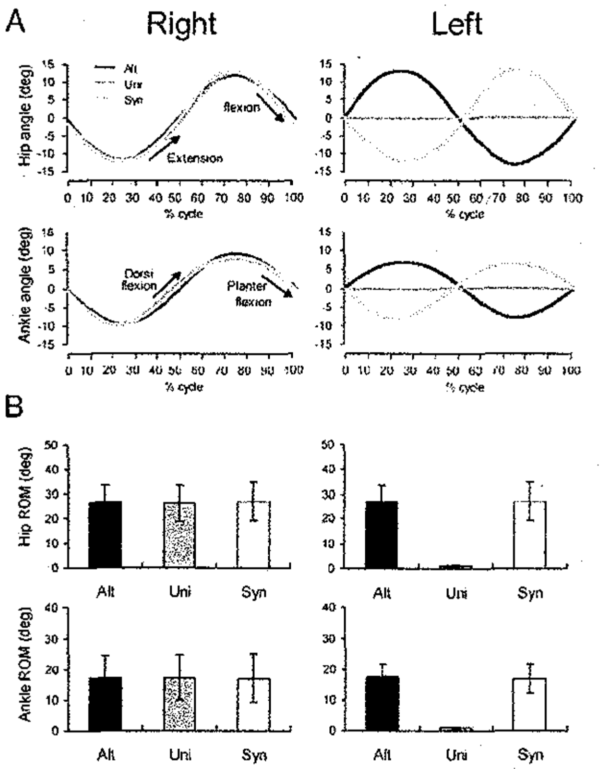

FIG. 4. Kinematical characteristics of imposed leg movement. *A*: changes of hip (*top*) and ankle (*bottom*) joint angular motions in 1 cycle of leg movement recorded from 1 subject using the VICON system. *B*: mean value of range of motion (ROM) of the hip (*top*) and ankle (*bottom*) joint ($n = 10$). 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

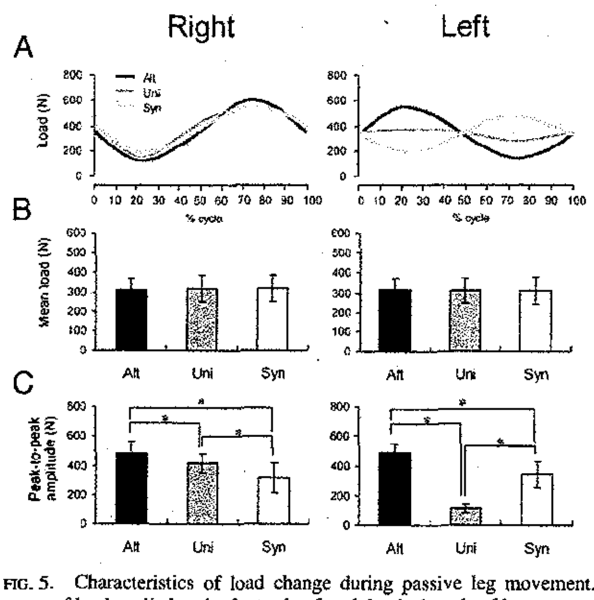

FIG. 5. Characteristics of load change during passive leg movement. *A*: changes of load applied to the foot sole of each leg in 1 cycle of leg movement. *B*: mean value of load averaged over 1 cycle ($n = 6$). *C*: mean value of peak-to-peak amplitude of load ($n = 6$). Alt, Uni, and Syn indicate the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

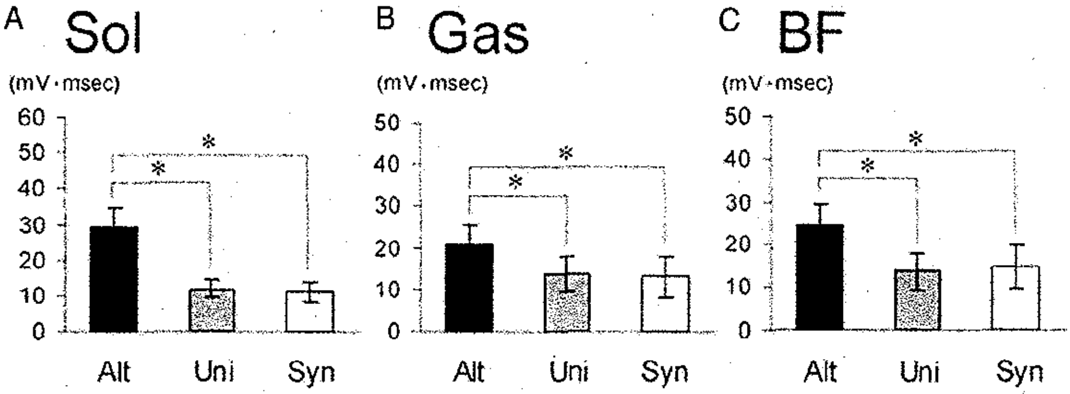

FIG. 6. Effect of leg movement pattern on muscle activity. Mean values of integrated rectified EMG induced by 3 types of leg movement. *A*: soleus (Sol). *B*: gastrocnemius (Gas). *C*: biceps femoris. Induced activity level of all 3 muscles was significantly larger ($P < 0.05$) during the alternate leg 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

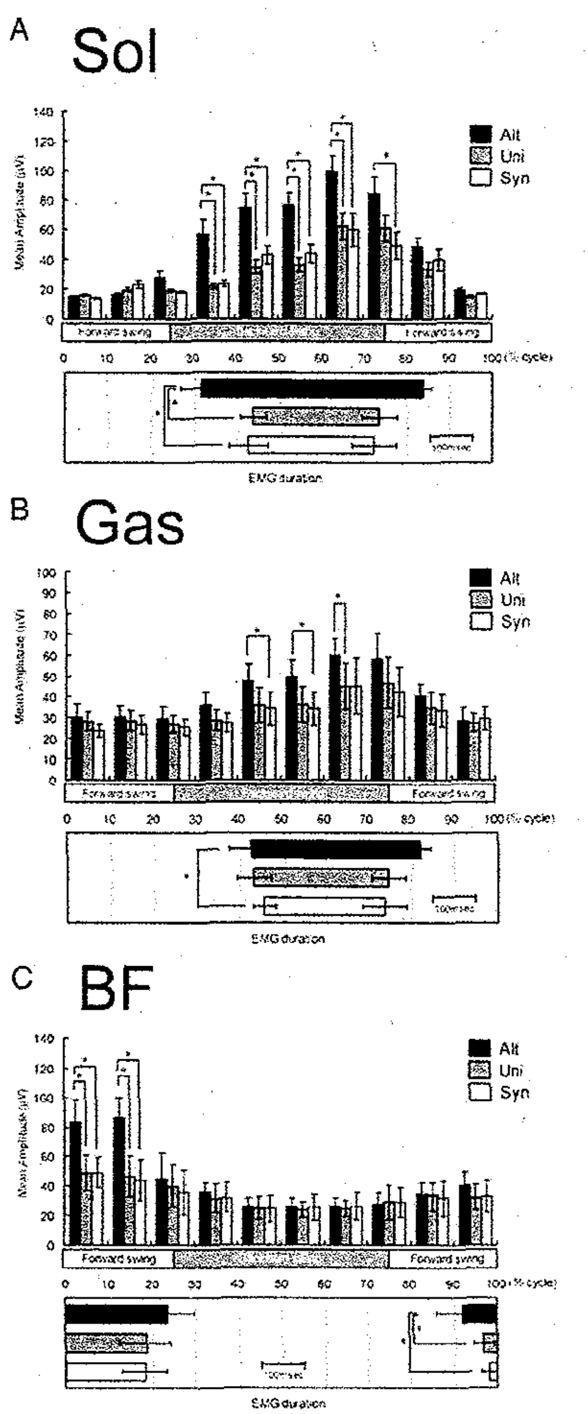

FIG. 7. Comparison of mean EMG amplitude in each 10% cycle bin (*top*) and duration of EMG activity (*bottom*) among the 3 experimental conditions. *A*: soleus (Sol). *B*: gastrocnemius (Gas). *C*: biceps femoris (BF). Alt, Uni, and Syn indicate the alternate, unilateral, and synchronous leg movement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

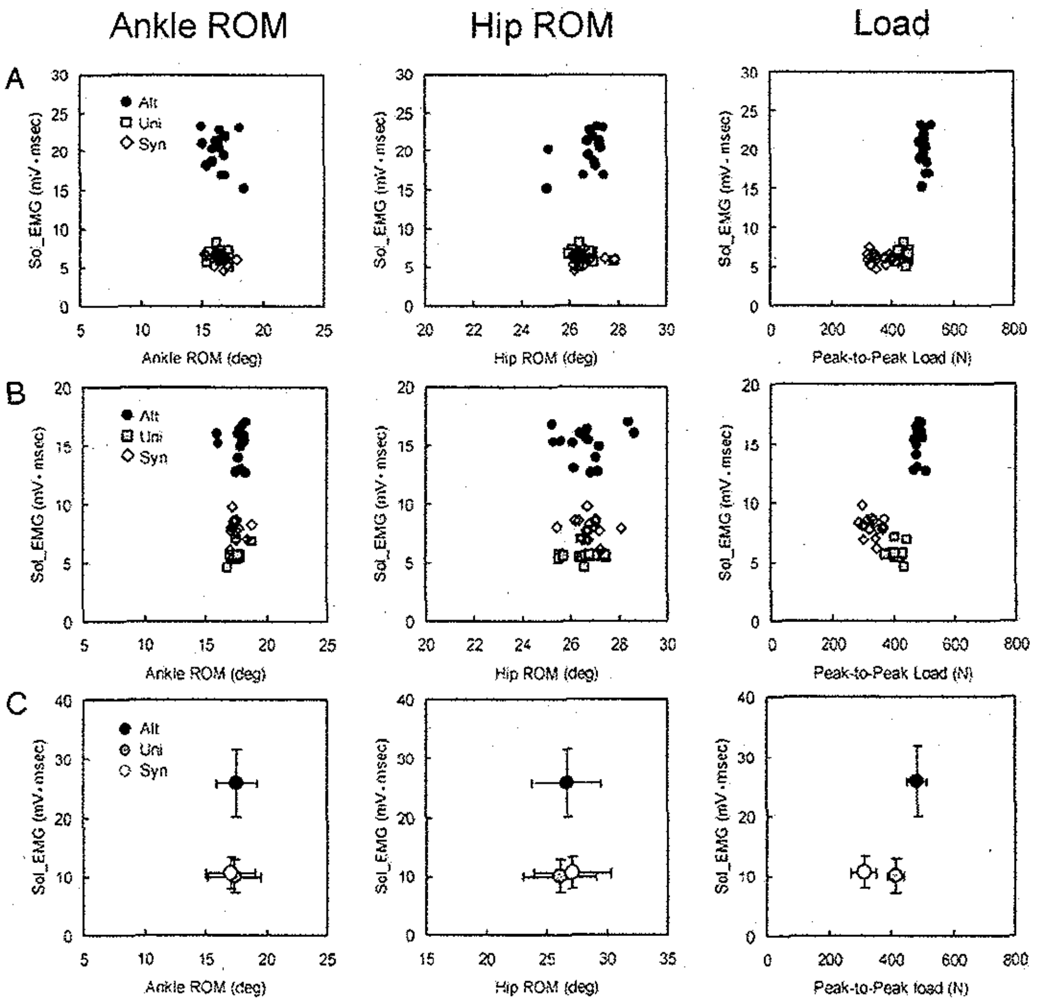

FIG. 8. Relationships between the right soleus EMG magnitude and ROM of the hip joint, ROM of the ankle joint, and peak-to-peak amplitude of load applied to the right leg. *A*: subject S1. *B*: subject S2. *C*: mean data ($n = 6$). In *A* and *B*, each point corresponds to the value obtained in each 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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13-84-3　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
－膝関節動作の有無による麻痺下肢筋活動の変化－

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股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

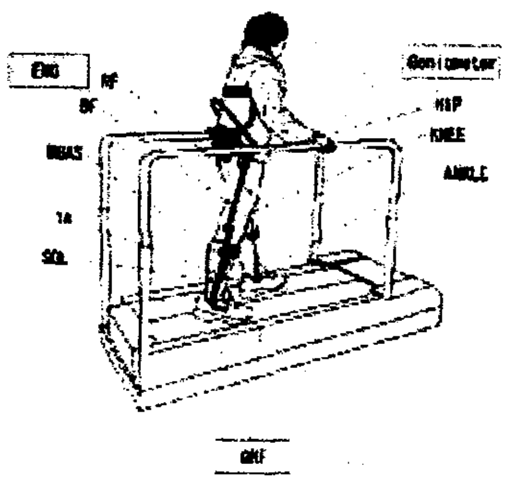

Fig.2 Experimental settings

ミルのベルト部に埋め込まれた3分力床反力計（ADAL-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．各筋から導出した筋電信号は整流・積分処理の後，遊脚期と立脚期それぞれの平均積分値を定量した．

5. 結果

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

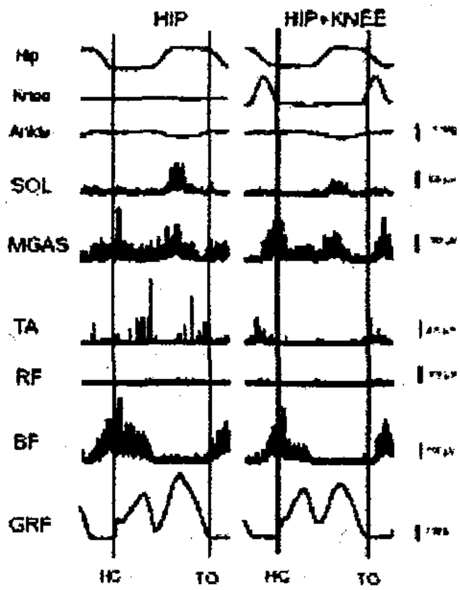

Fig.3 Typical data of the knematics (hip, knee, and ankle joint), electromyographic activity (SOL, MGAS, TA, RF, BF), and GRF during orthotic gait in a paraplegic patient. (HC: Heel contact, TO: Toe off)

Fig.4には被験者５名の歩行遊脚期，立脚期における各筋の積分値の平均値を示した．膝関節動作に伴う筋活動の変化は，必ずしも全ての被験者に共通の変化を認めなかったが，遊脚期における膝関節の屈曲伸展により直接筋長の変化がもたらされる大腿筋群，腓腹筋に変化を認める被験者が多かった．遊脚期における腓腹筋，大腿直筋の活動は，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ュエータを搭載することで，通常の装具歩行動作では実現されない膝の屈曲･伸展動作を可能にし，実現された膝関節動作に伴って歩行様筋活動が変化するのか否かを検討し

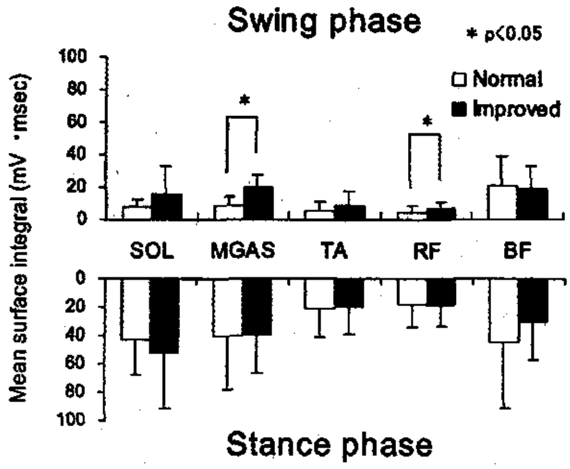

Fig.4 Comparison of the muscle activity between two types of orthosis in each lower limb muscle

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

7. 結論

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

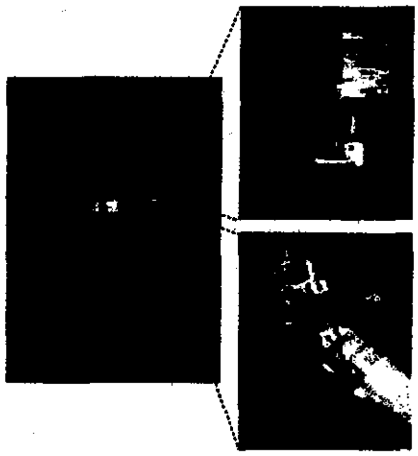

Fig.1 ARGO (left) and the devices mounted on the hip (right; above) and knee joint (below)

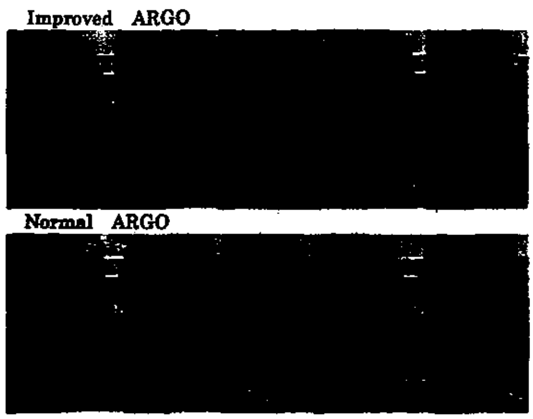

Fig.2 Gaits with the normal and improved ARGO

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（Th 8 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．

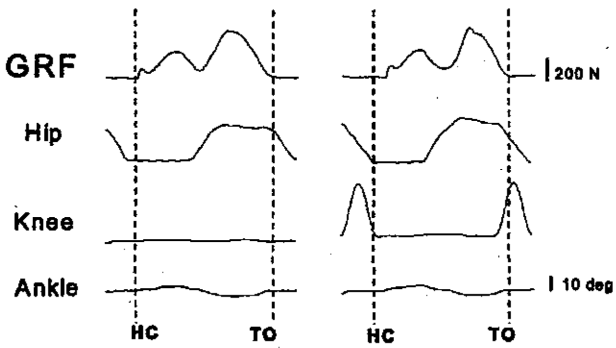

Fig.3 Changes in the ground reaction force (GRF) and the joint angles during a gait cycle. (HC: Heel contact, TO: Toe off)

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．

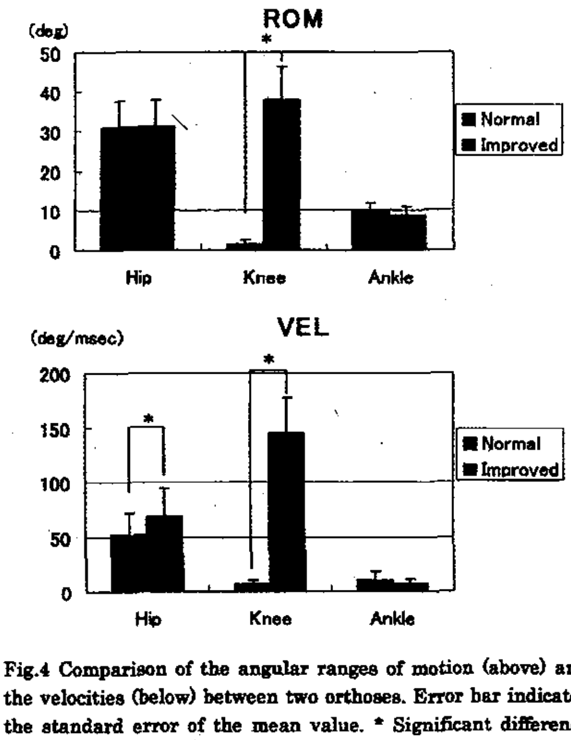

Fig.4 Comparison of the angular ranges of motion (above) and the velocities (below) between two orthoses. Error bar indicates the standard error of the mean value. * Significant difference (P<0.05).

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必がある．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で圧倒的に有利であり，車椅子との併用利用を前とした歩行器を実現する必要がある．もう一つ完全・不完全麻痺に対し，多くの研究データが部強制歩行が歩行 PG の再賦活を促進するこを示しており，完全対麻痺者における再賦活メニズム・可塑的変化を解析するとともに，その果を歩行再建に向けて生かすための具体的手を検討する必要がある．なお，この歩行機能支システムの拡張応用として，今後急増が予想さる自立歩行困難な高齢者・高齢障害者の立位歩トレーニングへの利用も可能と考えられ，寝たり高齢者・障害者数の減少，ひいては医療費・介護費用の削減に大きく貢献可能と考えられる．

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具による歩行訓練