A two-degree-of-freedom motor-powered gait orthosis for spinal cord injury patients

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Abstract: A number of orthoses have been developed to restore stance and walking in paraplegic subjects. Compliance, however, has been limited, mainly owing to walking effort. Use of the forces produced by actuators is an effective way to solve the problem of the considerable effort required for orthotic gait, namely high muscular effort and high energy expenditure. The purpose of the present study was to investigate the effects of assistance by external actuators on the orthotic gait of spinal cord injury (SCI) patients. Two kinds of linear actuator were developed by using direct current (d.c.) motors for assisting the knee and hip joint of a gait orthosis. They were mounted on the knee and hip joint of a commercial advanced reciprocating gait orthosis (ARGO), and a new two-degree-of-freedom externally powered gait orthosis was thus developed. The orthosis was assessed through inter-subject experiments on five male adult complete SCI patients. Owing to the short training period available for the assisted gait, simultaneous operation of both joint actuators was not conducted: either the knee actuation or the hip actuation was executed only. Thus, the knee actuator and the hip actuator were assessed with a T12 subject and with subjects for T5, T8, T11, and T12 respectively. The motions of the gaits, assisted by the linear actuators, were measured by a Vicon 370 system, and the general gait parameters and compensatory motions were evaluated. Results demonstrated that (a) all subjects could walk without falling, assisted either by the knee or the hip actuator; (b) both the knee and hip joint actuator increased the gait speed and the step length; (c) the knee flexion produced by the orthosis improved the dynamic cosmesis of walking; and (d) lateral compensatory motions as well as vertical ones tended to decrease when the hip joint was assisted, which could contribute to a reduction in walking effort.

Keywords: spinal cord injury patients, powered gait orthosis, linear actuator, inter-subject experiment, compensatory motion

1 INTRODUCTION

Once spinal cords have been completely or partially damaged, most people have no choice but to be confined for their remaining life to a wheelchair, even with current state-of-the-art assistive technologies. Although at present the wheelchair is the most efficient transportation means for spinal cord injury (SCI) patients, being in a sitting posture for long periods weakens them gradually both mentally as well as physiologically. Since the various benefits that stance posture or bipedal locomotion brings have been recognized, a number of passive mechanical gait orthoses, such as the Parawalker (also known as the hip guidance orthosis (HGO)) [**1**], the reciprocating gait orthosis (RGO) [**2**, **3**], the advanced reciprocating gait orthosis (ARGO) [**4–6**], and the Walkabout [**7**], have been developed and marketed. In practice, however, these orthoses are not necessarily used, and in order to investigate the reason for their abandonment, many clinical studies have been conducted. Bernardi *et al.* [**8**] estimated the efficiency of RGO locomotion as the ratio between mechanical work required to move the patient–orthosis system and energy expenditure for ten SCI patients, and pointed out that the large metabolic energy cost of the orthotic gait was the main factor limiting the walking capability of paraplegic subjects. Massucci

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et al. [**9**] also reported that the large metabolic energy cost was considered to be the main reason for the frequent abandonment or low utilization of the ARGO orthosis. Through these studies, the energy cost of the gait using these orthoses has come to be recognized as one of the major reasons for abandonment.

In parallel with the clinical studies, biomechanical investigations have also shown the importance of the energy cost in exerting the large effort required for the orthotic gait. Jefferson and Whittle [**10**] assessed the walking performance of one patient who was proficient in the use of three devices: a Parawalker, RGO, and an ARGO. (The latter was a development of the RGO system from Hugh Steeper Ltd. The hip mechanism was a modified version of that found in the RGO, incorporating only a single reciprocal cable.) From the viewpoint of limb abduction and ground clearance, these researchers recorded the motion of the lower limbs, pelvis, and upper torso by the Vicon system, and reported that the energy cost of walking in the Parawalker could be expected to be less than in the other two devices, owing to the smaller compensatory motion of the pelvis. Greene and Granat [**11**, **12**] also pointed out that upper limb loading was a limiting factor in mechanical gait orthoses for paraplegics. As mentioned above, the orthotic gait requires a large effort for SCI patients, and the major reason for abandonment is thought to be high metabolic energy cost and muscular strength of the upper limbs necessary to provide ground clearance and leg swing. The method for solving this problem would be a purely mechanical improvement of passive orthoses, but the use of additional assisting forces is also effective in reducing the effort for walking. Functional electrical stimulation (FES) and motorization by external actuators are typical sources of power.

Stallard and Major [**13**] pointed out the importance of maintaining some degree of abduction for effective reciprocal walking, and managed to reduce the energy cost of ambulation further by increasing the lateral rigidity of a Parawalker mechanical orthosis and by stimulating the gluteal muscles with a hybrid FES system. Thoumie *et al.* [**14**] also developed an RGO hybrid FES orthosis that activated the quadriceps and the contralateral hamstrings to produce the swing phase and contralateral push-off, and reported that the hybrid orthosis could increase the maximal walking distance of the SCI patients. Nene and Patrick [**15**] stated that the Parawalker with FES reduced the force applied through the crutches during the gait cycle and found that the device could also reduce the energy cost. In addition, many other researchers reported the effect of FES on the reduction of the effort in walking [**16–19**].

There is also a long history of the development of powered gait orthoses using the assistance of external actuators. In reference [**20**], Hughes introduced the research study conducted by Tomovic, Vukobratovic, and their colleagues at the Mihailo Pupin Institute in Belgrade. Their work is described in several contributions featured in the volumes entitled *Advances in external control of human extremities* (1963–1990) [**21**]. They also constructed an exoskeleton incorporating external power for paraplegic patients [**22**]. A series of subsequent developmental studies are detailed in references [**23**] and [**24**]. In 1997, Ruthenberg *et al.* [**23**] developed a single-degree-of-freedom powered exoskeleton gait orthosis with direct current (d.c.) motors to provide bipedal locomotion to individuals with physical impairment. Yano *et al.* [**24**] also developed a weight-bearing control (WBC) orthosis with a rigid exoskeletal frame, a reciprocal link device on the rear pelvic girdle, and a gas-powered foot device that varied its sole thickness for providing ground clearance. The exoskeletal frame consists of a thoracic girdle and two long leg braces with hip and knee joints and with reciprocal connection. The reciprocal device consists of a link connecting both hip joints through a steel bar and a link connecting the medial uprights of the long leg brace at the medial proximal aspect. They reported that the WBC orthosis enabled thoracic-level paraplegic patients to walk at relatively higher speed than with conventional orthoses, under similar energy expenditure [**25**]. Recently, in 2004, Kang *et al.* [**26**] proposed a powered hip joint for T11 or lower patients by using an air muscle actuator. In the present study, in order to assist SCI patients in walking with a gait orthosis requiring a large effort on a commercially available ARGO, a prototype gait orthosis, fitted with d.c. motors which assisted both knee and hip joint rotation, was developed, and the effects of the actuators on gait assistance were assessed through inter-subject experiments.

2 DESCRIPTION OF THE PROPOSED TWO-DEGREE-OF-FREEDOM MOTOR-POWERED GAIT ORTHOSIS

2.1 Mechanism of the motor-powered knee joint

A linear motion actuator was developed by combining a low-inertia d.c. motor (3042, 20.6 W, 156 g, diameter 30 mm, length 63 mm, by Faulhaber, Switzerland) and a ball screw (GY0802GS, shaft diameter 8 mm, lead of 2 mm, length 200 mm, by

Kuroda Precision Industries Ltd, Japan). The actuator was mounted at the back of the knee joint of a commercially available ARGO (Hugh Steeper Ltd, UK), which was chosen because of the high stability of bipedal locomotion provided by the reciprocating cable. One actuator was mounted on each knee joint. The stroke length of the linear actuator and the maximum knee flexion angle were set at 150 mm and 70° respectively. Figure 1 provides schematic diagrams of the link mechanism and photographs of the actuator. As can be seen, the mounted mechanism is compact; its mass is 0.8 kg. The modified knee joint is locked by the original mechanism of ARGO at full-extension position. The actuator unlocks it at the beginning of the swing phase, during which it rotates. At heel strike, the actuator locks the joint again at the full-extension position. The purpose of this locking mechanism is to allow SCI patients to walk safely while flexing the knees during the swing phase. The power source was provided by 12 Ni–MH rechargeable batteries featuring light weight and large capacity (HR-3US, 1600 mAh, Sanyo Co., Japan) for each actuator (a total of 24 batteries for both knees). The batteries last about 1 h under regular gait conditions.

2.2 Mechanism of the motor-powered hip joint

One of the features of the ARGO is the achievement of steady bipedal locomotion enabled by the hip-driving cable, which was improved by another linear actuator. It consists of a d.c. motor (3557, 14.5 W, 275 g, diameter 35 mm, length 80.6 mm, by Faulhaber, Switzerland) with a planetary gearhead (30/1, 66 : 1, 171 g, diameter 30 mm, length 63.7 mm, by Faulhaber, Switzerland), and a rack-and-pinion gear (EP150, 94 × 48 × 62 mm (width, depth, height), 0.7 kg, stroke 100 mm, maximum thrust 300 N, rack gear rod 250 g, by Asahi Seiko Co. Ltd, Japan). Its mass is 1.4 kg. Figure 2 provides schematic diagrams of the original hip-driving mechanism of the ARGO and the hip joint actuation, as well as photographs of the mounted linear actuator.

One actuator operating the connecting cable (or the inner reciprocal cable) was mounted on the left side of the orthosis. The 12 Ni–MH batteries mentioned

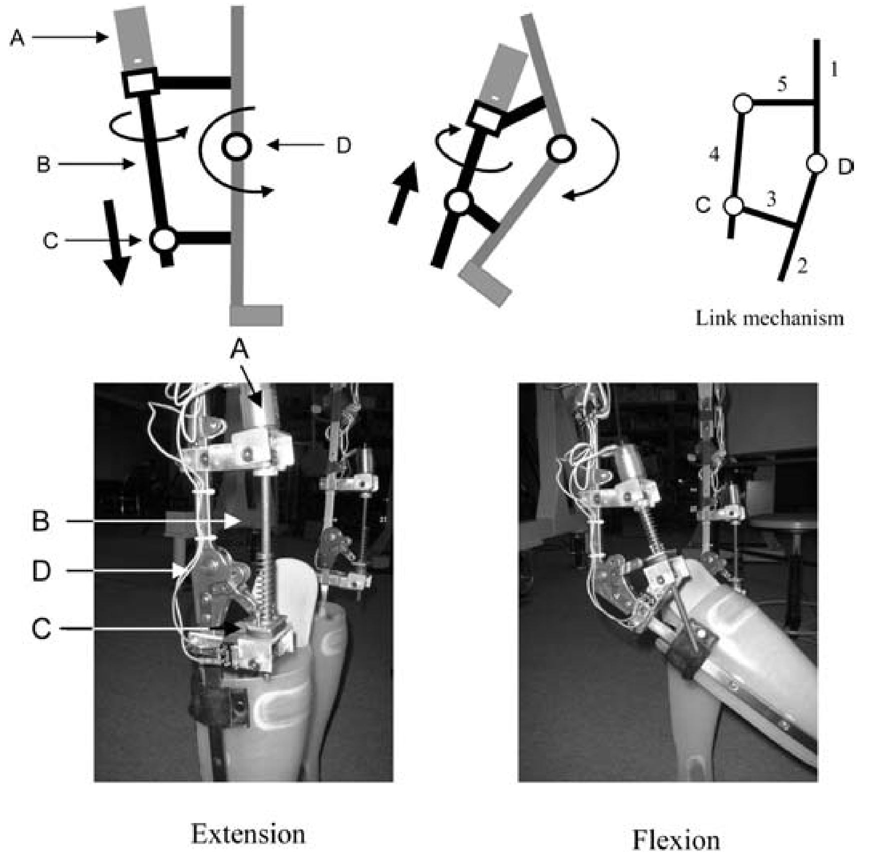

Fig. 1 Schematic diagrams and photographs of the motor-powered knee joint: A, d.c. motor; B, screw; C, nut (joint); D, knee joint. Bars 5 and 3 are fixed on bars 1 and 2 respectively. When the rotation of the screw by the d.c. motor shortens the length of bar 4, the knee joint flexes

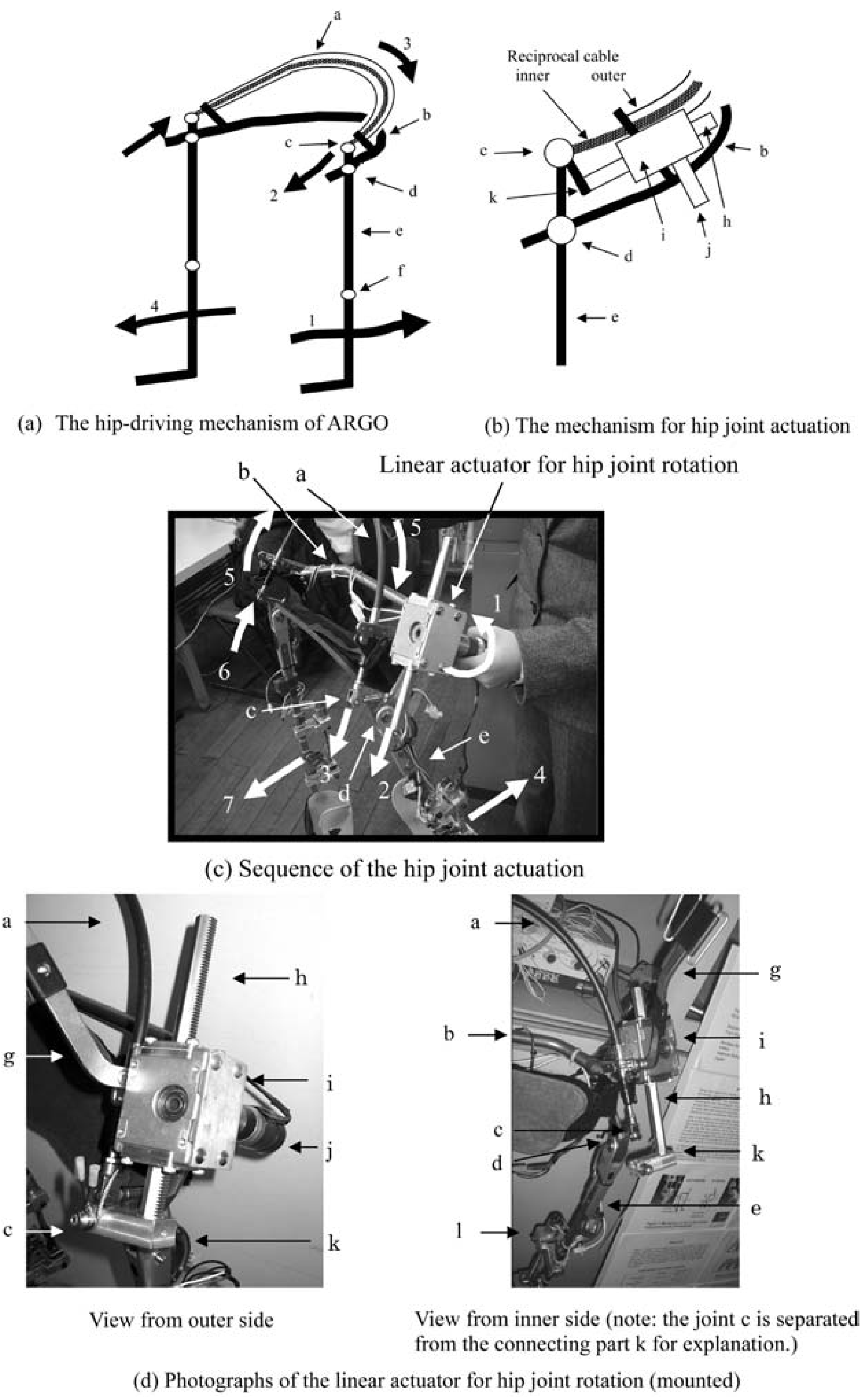

Fig. 2 Schematic diagrams and photographs of the motor-powered hip joint (a, reciprocal cable (outer); b, back tube; c, joint between the inner reciprocal cable and the leg frame; d, hip joint; e, leg frame; f, knee joint; g, thoracic bar of the ARGO; h, rod of the rack gear; i, rack-and-pinion gear box; j, d.c. motor with a planetary gearhead; k, connecting part; l, knee actuator)

previously were used for the actuator. Figure 3 shows a photograph and drawing of the powered orthotic gait device with the three actuators mounted (one for the hip and two for the knee). A compact exoskeleton-type two-degree-of-freedom gait orthosis could be achieved. In order to prevent users from being caught by the cable and falling down, an infrared switch was used to control the actuators. The switch was placed on the handle of an arm crutch and was pushed by users at every step. The control of the joint rotation was simply designed as follows. The joint rotates while the switch is being pushed. When it is released, in the case of the knee joint, it returns to the extension position. In the case of the hip joint, it stops moving.

3 KINEMATIC MEASUREMENT OF THE ASSISTED GAIT OF SCI PATIENTS

The powered gait orthosis developed was assessed through experiments on the SCI subjects from the viewpoint of gait kinematics. One male (complete SCI of the T12 level) and four other male subjects (complete SCI of the level of T5, T8, T11, and T12) participated in the experiments for the evaluation of the knee and hip assistive mechanism respectively. At the time of the experiment each subject had learned to use a normal ARGO for 3 months or more, and the measurements were conducted after the subjects had become accustomed to the gait with the powered ARGO. Through experiments, all subjects could walk without falling, assisted by the device. A subject wearing the powered orthosis walked with crutches at his own pace, and the gait motions for a distance of about 6 m were measured 5 times by the Vicon 370 system. Eighteen reflective markers were put on the subject and the crutches for the measurement of general gait parameters: top of head, vertebra prominence, acromion, head of radius, radial styloid process, greater trochanter, lateral condyle of femur, lateral malleolus of fibula, second metatarsals, and tip of the crutch. The sampling rate of the Vicon system was set at 60 Hz. The general gait parameters and compensatory motions in the lateral direction as well as the vertical direction were analysed from the measured motion data. Then, to investigate the effect of the actuators on the gait functions, statistical inter-subject comparisons were made by the paired *t*-test. All experiments were approved by the ethics

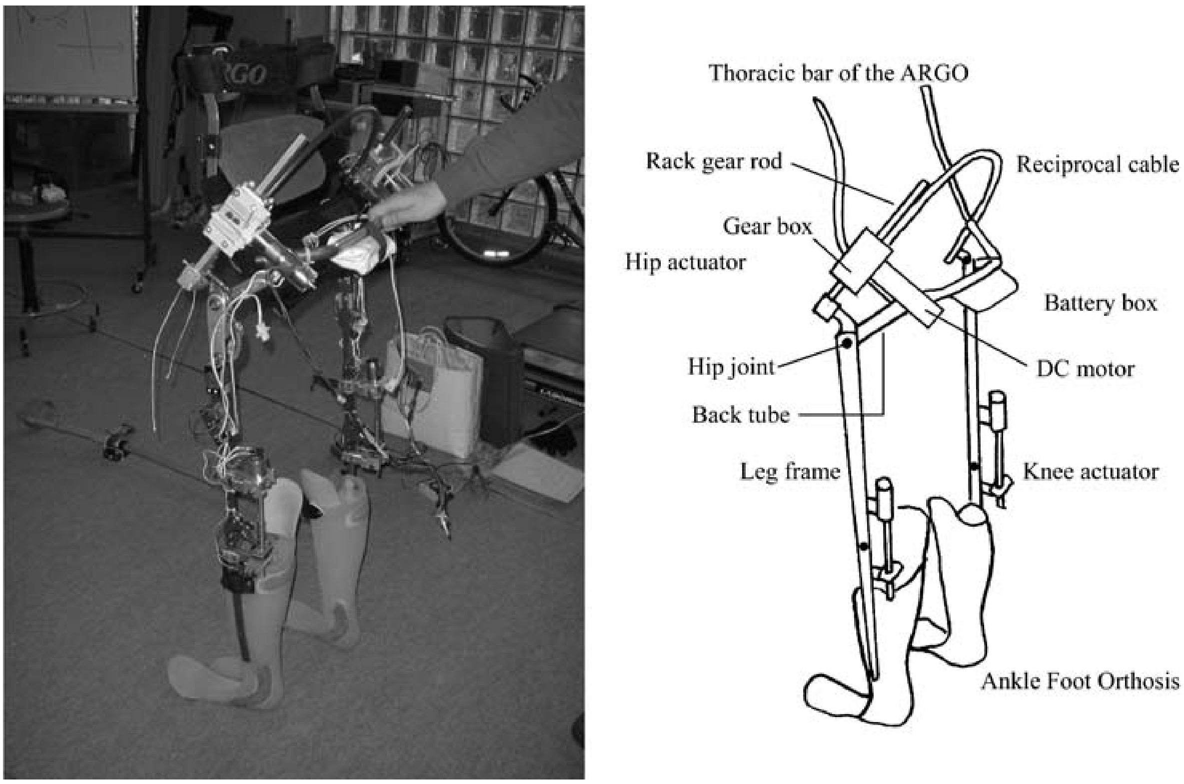

Fig. 3 Photograph and line drawing of the two-degree-of-freedom motor-powered gait orthosis (Note: the actuator mounted on the right hip joint shown in the photograph was found to be unnecessary through the trial gain experiments with SCI patients and was removed. One actuator is enough to swing both legs.)

committee of the National Rehabilitation Centre for Persons with Disabilities (Tokorozawa, Japan). The experimental procedures were also explained and informed consent was obtained from each subject. (Note that in the following experiments simultaneous operation of both joint actuators was not conducted: only the knee actuation or the hip actuation was executed. Many gait practices for several months were required for the subjects to learn to use the normal ARGO, and more were required to learn to use the hip-powered or knee-powered ARGO. Owing to discharge from the hospital, they did not have enough time to learn the ARGO with both hip and knee actuation. More time was needed for them to acquire the skills, such as the operation of the two switches and the more complicated upper trunk movements.)

4 RESULTS

4.1 Experiments with gait assistance by the knee actuator

Figure 4 shows an example of the gait motion (T12, right leg) on a sagittal plane measured by the Vicon system when the knee joint was rotated by the actuator. As shown by the arrows, the knee joints flexed during leg swing. The range of the rotational angle of the knee, angular speed during flexion, and the one at extension were $31.8 \pm 1.9^\circ$, 152.4 ± 1.5 deg/s, and 204.4 ± 6.2 deg/s respectively. The extension speed was faster than the flexion speed. Regarding the gait parameters, as shown in Table 1, the knee actuator increased the gait speed and the step length by 23 and 7 per cent respectively. By flexing the knee joint, a gait closer to a normal one could be achieved with respect to cosmesis. However, in comparison with a normal gait, the subject had to bend the upper torso forward using crutches, which could result in a large muscular effort to the upper limbs.

4.2 Experiments with gait assistance by the hip actuator

Table 2 shows the gait speed and the step length acquired from the gait of four subjects with and without the assistance of the hip actuator. As shown, the actuator tended to increase the gait speed as well as the step length in all cases. All subjects gave the impression that the gait effort had decreased. In the

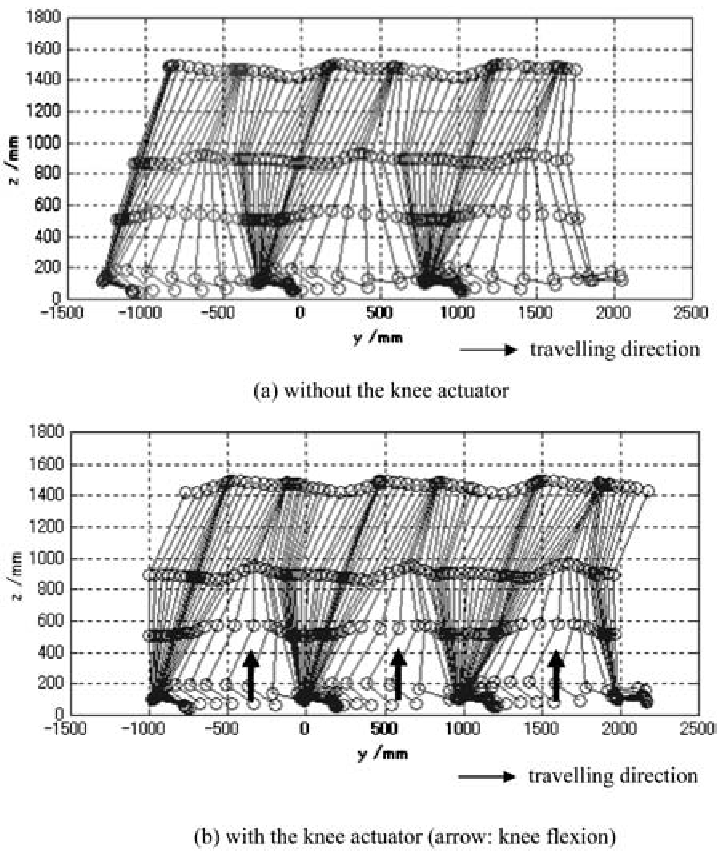

Fig. 4 Examples of the motions of the right leg on a sagittal plane with and without the knee actuator (T12 subject)

Table 1 Gait speed and step length with and without the knee actuator (T12 subject, ±standard deviation)

	Without actuator, normal	With actuator, knee flexion	Increase (%)
Gait speed (m/s)	0.43 ± 0.01	0.53 ± 0.03	23*
Step length (cm)	101.4 ± 1.5	108.3 ± 2.3	7

*$p < 0.01$.

Table 2 Gait speed and step length with and without the hip actuator (±standard deviation)

	Gait speed* (m/s)			Step length* (cm)		
Subject	PA	Normal	Increase (%)	PA	Normal	Increase (%)
T5	0.28 ± 0.03	0.27 ± 0.01	4	89.6 ± 3.4	85.3 ± 7.1	5
T8	0.25 ± 0.01	0.23 ± 0.01	9†	80.9 ± 5.8	78.7 ± 7.1	3
T11	0.35 ± 0.02	0.34 ± 0.01	3	96.0 ± 5.4	96.2 ± 8.4	–
T12	0.57 ± 0.02	0.54 ± 0.02	6	122.0 ± 3.0	119.4 ± 2.6	2

*PA = with power assistance; normal = without power assistance.
†$p < 0.05$.

case of the regular orthotic gait where the knee joint is fixed at an extension position, ground clearance during leg swing has to be provided by tilting the upper trunk laterally. This compensatory motion was evaluated from the trajectory of the marker placed on the subject's head, since this motion might put a higher load on the upper limbs. Figure 5(a) shows examples of the lateral excursion of the marker on the transverse plane with and without the hip actuator (T5 subject). The horizontal axis and the vertical axis represent the travelling distance and the lateral displacement respectively. As shown, the subjects walked moving the trunk from side to side at every step. In addition to the lateral tilting, the upper trunk has to be hiked to swing the paralysed leg. This compensatory motion could increase the load on the arms [**12**]. Figure 5(b) shows examples of the vertical excursion of the same marker on the sagittal plane.

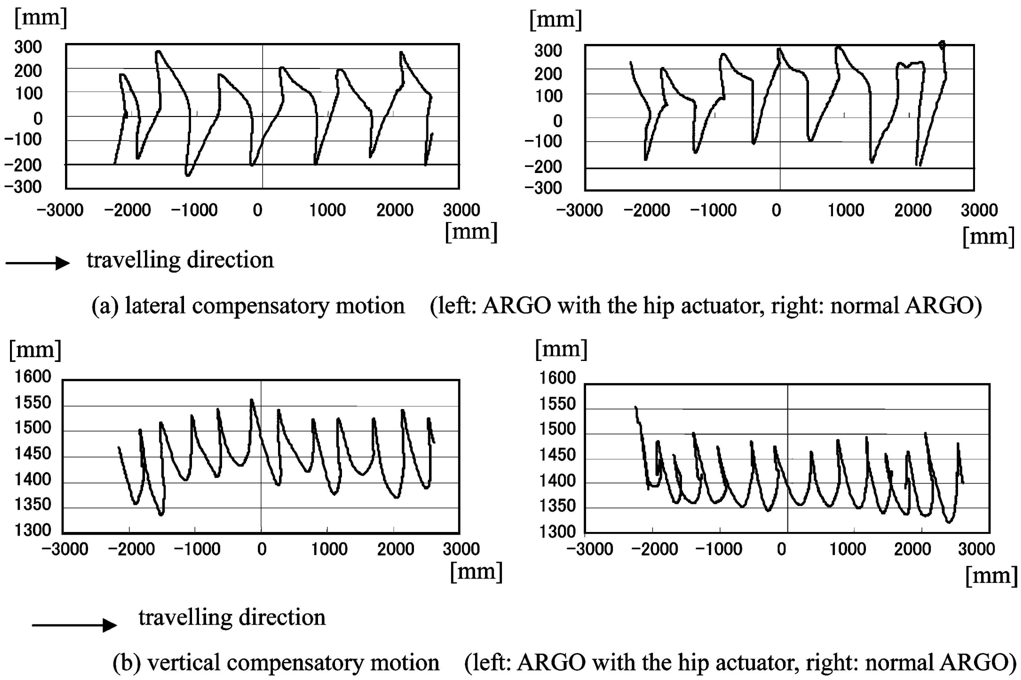

Fig. 5 Examples of compensatory motions with and without the hip actuator (T5 subject). The compensatory motion was evaluated as the trajectory of the marker placed on the subject's head

Here too the horizontal and vertical axes represent the travelling distance and the vertical displacement respectively. In the same way as the lateral excursion, it was found that at every step the subject moved the upper trunk upwards and slightly backwards. Based on the trajectory of the head marker, the lateral and vertical compensatory motions were evaluated. Table 3 shows the average amplitude of the compensatory motion from the T5–12 subjects. The amplitude was measured as the distance from peak to bottom within a wave. (Since a few steps just after the onset of the 6 m gait motion tended to be unstable, the data in the middle or later part of the marker trajectory were used to calculate the average amplitude. Several series of motion measurements were conducted under each experimental condition until sufficient data for ten steps ($n = 10$) were obtained.) These data show that the patient with the higher lesion level has the larger extent of marker excursion in both the lateral and vertical directions, and that, with power assistance, the lateral compensatory motion and the vertical one tended to decrease by 1–18 and 11–25 per cent respectively. The actuator swung the paralysed legs instead of the trunk muscles, which might have resulted in the decrease in the lateral and vertical movement of the upper torso.

5 DISCUSSION

In the present study, in order to assist an SCI patient in walking by gait orthosis, which requires a large effort, and to improve the degree-of-freedom of the gait device, two assistive actuators were developed using small d.c. motors, and were assessed through kinematic analysis of the gaits of SCI patients. As stated in the methodology section, simultaneous operation of both joint actuators could not be conducted: only the knee actuation or the hip actuation was executed, owing to the limited available training period. Thus, the knee actuator and the hip actuator were assessed with a T12 subject and with subjects for T5, T8, T11, and T12 respectively. The results showed that both actuators tended to improve the gait function. It was also shown that the assistance of the hip joint had a tendency to decrease the compensatory motion of the upper part of the body. Since these experiments were conducted on five subjects, statistical experiments with a larger number of subjects would be necessary to verify the effects on the compensatory motion for ground clearance and to measure the muscular effort of the upper limbs and the energy cost for walking. It might also be interesting to mount these actuators on the Parawalker to have the additional feature of high lateral rigidity and to make biomechanical comparisons of the compensatory motions between these two designs of orthosis. However, it is important also to consider what each user needs in daily living and what he/she expects from the gait orthosis, because successful use of gait orthosis is never determined simply by quantitative parameters such as the walking function, energy required, etc. [**27**, **28**].

With reference to the pioneering research conducted by Tomovic and his colleagues and the various subsequent developments, a motor-powered two-degree-of-freedom gait orthosis was developed in the present study. The actuators developed are small and easy-to-install for commercial orthoses. During the stance phase, the knee joint was securely locked at the extension position by the knee actuator and the subject (T12) wearing it could walk without falling. All subjects assisted by the powered hip joint gave the impression of a reduction in walking effort, probably owing to a reduction in the compensatory motion. The motor under operation was quiet. One of the issues for future work on the developed device would be the simultaneous operation of both joint actuators. In this case, the subject would have to walk with the device, controlling the two types of actuator at the appropriate gait phase. An automatic switch using the gait phase detection system [**29**] might be effective considering the decrease in the effort to the

Table 3 Amplitudes of the compensatory motions with and without the hip actuator ($\pm$standard deviation)

	Lateral amplitude* (cm)			Vertical amplitude* (cm)		
Subject	PA	Normal	Decrease (%)	PA	Normal	Decrease (%)
T5	37.6 ± 2.0	42.7 ± 5.8	12†	11.6 ± 2.8	13.1 ± 1.1	11
T8	21.5 ± 2.4	21.8 ± 1.8	1	10.5 ± 2.4	12.5 ± 1.8	16
T11	17.5 ± 3.8	19.1 ± 1.8	8	6.1 ± 1.8	8.1 ± 2.2	25†
T12	3.7 ± 1.5	4.5 ± 1.7	18	5.9 ± 0.5	7.4 ± 0.8	20‡

*PA = with power assistance; normal = without power assitance.
†$p < 0.05$.
‡$p < 0.01$.

upper limbs as well as the fall risk. Since power assistance using external actuators still presents many problems, as do the pure FES and hybrid systems, selection from various assistive techniques would depend on the user's preference, as long as the performances offered are almost the same.

Two points are considered in the present study. One is the effect of the knee flexion on the orthotic gait. Edwards and Bataweel [**30**] and Yang *et al.* [**31**] developed knee flexion devices and reported that the knee flexion mechanisms contributed a reduction in energy consumption, whereas Baardman *et al.* [**32**] showed opposite results. Greene and Granat [**11**] also reported that knee flexion provided no decrease in the compensatory mechanisms because of the increase in the effective leg length. (The distance from hip to toe lengthens because of the lack of a moveable ankle joint.) The knee flexion affects energy consumption and the compensatory motions are still controversial; in order to clarify them, a set of statistical experiments on many subjects has to be conducted. The other point is the neuro-rehabilitation effect on the paralysed muscles. Kojima *et al.* [**33**] and Colombo *et al.* [**34**] examined the lower-limb electromyographic (EMG) activities from SCI subjects with gait devices and observed cyclic EMG activation patterns from paralysed muscles. These could be explained as follows: the orthotic gait would stimulate the proprioreceptors in the paralysed lower limbs, and thereby afferent inputs from them could activate the locomotor centre in the isolated spinal cord [**35**] (the stretch reflex is a typical example). These experimental findings should be considered in further developments of gait devices for patients with neural disorders.

For effective reactivation of the paralysed locomotor centre by stimulating the proprioreceptors, training at a certain gait speed and progressively faster has to be conducted regularly from the beginning of the rehabilitation. Actually, it is difficult to execute this, because SCI patients tire easily. Therefore, it would be appropriate to start with treadmill training using the powered gait device. Then, with the gait function gradually recovered, a trainee would be released from the treadmill and be able to walk around freely with the gait device, which might also be expected to improve motivation for training. At this time, the device developed in the present study cannot be a substitution for the wheelchair as the daily means for ambulation. Moreover, a patient fitted with it cannot sit on a wheelchair owing to the mechanical interference of the hip actuator. Therefore, compatibility of wheelchairs with the device has to be much improved so that SCI patients might acquire therapeutic benefits through frequent use of the device, even away from treadmill training at hospital.

6 CONCLUSIONS

In an attempt to reduce the large effort and high metabolic energy consumption required by SCI patients in walking with a gait orthosis, a two-degree-of-freedom externally powered gait orthosis system was designed by modifying the commercially available ARGO. The linear actuators used are of small size, lightweight, and can be easily installed. Experiments on five male SCI patients (complete T5–T12), using the ARGO device, assisted by either the knee or the hip actuators, demonstrated that all subjects could walk without falling and both actuators tended to increase the gait speed and the step length. Knee flexion was found to improve the dynamic cosmesis of walking. Kinematic analysis of the gait assisted by the hip actuator also revealed a tendency towards a decrease in lateral and vertical compensatory motions in all cases, which could contribute to a reduction in the walking effort. The system developed in the present study could be expected to lead to the expansion of walking opportunities for SCI patients.

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