The influence of **the** reciprocal hip joint link in the Advanced Reciprocating Gait Orthosis on standing performance in paraplegia

G. BAARDMAN*, M. J. UZERMAN*'**, H. J. HERMENS*, P. H. VELTINK**, H.B.K. BOOM** and G. ZILVOLD*'**

**Roessingh Research and Development b.v., Enschede, The Netherlands*
***Institute for BioMedical Technology, University of Twente, Enschede, The Netherlands*

Abstract

The effect of reciprocally linking the hip hinges of a hip-knee-ankle-foot orthosis on standing performance was studied in a comparative trial of the Advanced Reciprocating Gait Orthosis (ARGO) and an ARGO in which the Bowden cable was removed (A_GO). Six male subjects with spinal cord injury (SCI) at T4 to T12 level participated in the study, which was conducted using a single case experimental design. Standing balance, the ability to handle balance disturbances (standing stability), and the performance of a functional hand task during standing were assessed in both orthosis configurations in the order A_GO-ARGO-A_GO-ARGO.

No significant differences with respect to standing performance were found for the two orthosis configurations. However, the results indicate that the crutch force needed for maintaining balance during various tasks, especially for quiet standing with two crutches, may be much higher in the orthosis without Bowden cable. Therefore, it is very likely that the reciprocal hip joint link in the ARGO provides a substantial and clinically relevant reduction of upper body effort required for standing under functional conditions.

All correspondence to be addressed to Gert Baardman, Roessingh Research and Development b.v., P.O. Box 310, 7500 AH Enschede, The Netherlands. Tel: +31-53-4875777. Fax: +31-53-4340849. E-mail: g.baardman@rrd.nl

Introduction

Standing is a very important activity in the daily life of persons with paraplegia. Numerous therapeutic benefits of standing upright have been discussed in the literature: muscle contracture prevention, reduction of spasticity, reduction of bone mineral loss, improvement of lower extremity blood supply, prevention of pressure sores, and improvement of bladder and bowel function (Kunkel *et al,* 1993; Messenger *et al.,* 1989; Ogilvie *et al.,* 1993; Figoni, 1984). These preventive aspects above justify that standing is included in the rehabilitation programme for the spinal cord injured and that most paraplegics have some standing frame at home. Also, the psychological effect of being upright and able to communicate at eye level with healthy persons is very important (Nene *et al,* 1996).

As an alternative to a standing frame, an orthosis, in thoracic spinal cord injury usually a hip-knee-ankle-foot orthosis (HKAFO), can be used. An orthosis adds to the above-mentioned therapeutic benefits the possibility of functional use in daily life activities (Douglas *et al,* 1983; Motloch, 1992; Rose 1979; Winchester *et al.,* 1993). Besides, an orthosis offers the possibility of use outside the home environment.

It is clear that the functional characteristics of an orthosis are defined by its design. Traditionally, most attention in the design of orthoses has been directed to assistance in walking (Douglas *et al,* 1983; Motloch, 1992; Rose, 1979; Stallard *et al,* 1989; Stallard and

Major, 1993). As a result, the properties of an orthosis with respect to standing are a consequence rather than the objective of design choices. Since walking makes higher demands on the design of an orthosis than does standing, it is taken for granted that in well-designed walking orthoses the performance with respect to standing is of equal quality. However, functional standing, i.e. standing for the purpose of performing a (bi)manual task, imposes additional and possibly conflicting demands upon the design of an orthosis, especially with respect to stability and flexibility.

One of the important aspects related to orthosis design is energy consumption during gait (Stallard *et al,* 1989; Nene *et al,* 1996). Various principles and mechanisms have been described that contribute to a reduction of energy expenditure.

The alignment in the frontal plane was reported to have an impact on the lateral stability of an orthosis, and consequently on the effort required to balance the body (Rose, 1979). A similar effect was found in studies on lateral stiffness of orthoses (Stallard and Major, 1993; 1995).

An essential difference in the properties of currently prescribed HKAFOs arises from the application of a reciprocal coupling of the hip joints.

The LSU-Reciprocating Gait Orthosis (RGO) (Douglas *et al,* 1983) and the Advanced Reciprocating Gait Orthosis (ARGO) (Hugh Steeper Ltd., London, UK) (Jefferson and Whittle, 1990) incorporate hip hinges which are reciprocally coupled via one (ARGO) or two (RGO) Bowden cables. Recently, the Isocentric® RGO was designed which incorporates a reciprocal coupling via a lever with ball bearing (Motloch, 1992; Winchester *et al,* 1993). Though designed in the first place to provide energy transfer from the stance leg to the swing leg and vice versa during gait, a reciprocal coupling has great influence on standing as well. Since bilateral flexion or extension of the hips is made impossible, a stabilising effect results that forces the patient into an upright position, but may hinder the patient in reaching.

The Hip Guidance Orthosis or Parawalker (Rose, 1979) is mounted with free hip hinges with limited flexion and extension ranges. During gait, this hip mechanism allows the utilisation of gravity for the execution of the swing phase (Nene and Major, 1987). The other side of this favourable property with respect to energy consumption is the absence of mechanical stabilisation of the hips and trunk. Also during standing, the patient is free to flex and extend the upper body, which permits a free choice of posture, but may induce the need for substantial effort for maintaining balance. In practice two postures are applied in the Parawalker, namely the so called C-posture with extended hips, and a posture where the trunk is flexed and stopped by the hip flexion limits.

From the above it may be expected that reciprocal coupling of hip joints in an orthosis has both favourable and adverse effects on energy consumption during gait, in addition to an important effect on the functionality of standing, that is, on the applicability of standing in daily activities. In order to obtain directives for orthosis design, the authors studied these effects on both gait and standing performance. A direct comparative trial of, for example, the RGO, ARGO and Parawalker was not expected to be suitable for studying these effects, since differences between these orthoses other than the reciprocal hip joint linkage are likely to influence the results. In order to study the influence of the reciprocal hip joint link in isolation from other orthosis properties, the performance of the ARGO was compared with that of an ARGO of which the Bowden cable was removed. The results with respect to performance of gait have been reported separately (IJzerman *et al,* 1997). This paper focuses on the effect on standing performance.

Assessment of standing performance

In literature, standing performance has been associated with three different aspects. Standing balance, most often in relation to postural control, is commonly studied by means of ground reaction force measurements during varying support and visual or cognitive task conditions (Cybulski and Jaeger, 1986; Geurts *et al,* 1993; Goldie *et al,* 1989; Mayagoitia and Andrews, 1989; Slobounov and Newell, 1994). From such measurements, centre of pressure diagrams can be obtained which are mostly parameterised by the excursion or amplitude, the velocity and the frequency; elaboration both in terms of circular parameters or separated

anteroposterior and mediolateral parameters have been reported. It is assumed that amplitude parameters relate reciprocally to the effectiveness of balance. Velocity parameters are commonly associated with regulatory mechanisms (Mayagoitia and Andrews, 1989). Crutch support forces are incorporated in none of the reported analyses related to paraplegic standing.

The ability to maintain balance in the presence of disturbances, or standing stability, is an aspect particularly of interest in paraplegic standing, where many of the control mechanisms used in able-bodied standing are absent. The application of closed loop functional neuromuscular stimulation (FNS) control of the knee in paraplegia was reported to support voluntary response mechanisms of the upper body (Moynahan and Chizeck, 1993). The effort necessary to maintain an upright posture after unanticipated knee flexion disturbances was assessed by measurement of vertical arm force applied to a walking frame. The effect of the disturbance was measured by the time necessary to recover to a stable posture. The possibility of performing hand tasks during standing, or standing *functionality*, is a third and very important aspect of standing performance. The assessment of the ability to free the upper limbs from support and balancing tasks in order to manipulate objects was the subject of the development of the Functional Standing Test (Triolo *et al,* 1993). In this test 18 tasks requiring fine coordination, pushing, pulling, reaching horizontally, vertically and diagonally were included in order to allow evaluation of the effectiveness of different assistive devices for people with standing disabilities.

In the present study, an assembly of tests previously applied in comparable or related studies was made to allow comparison of functional standing performance in the ARGO with and without reciprocal hip joint link.

Methods

Subjects

Six complete thoracic spinal cord injured subjects participated in the study (Table 1). All had finished their rehabilitation programme and were well-trained and experienced ARGO users.

Informed consent was obtained from each subject prior to each measurement session. The study was approved by the local medical ethics committee.

Study design

The study was conducted using a single case experimental design. Subjects were assessed four times: two assessments of the ARGO were performed, and two of the ARGO with removed Bowden cable (hereafter referred to as A_GO) in the order A_GO - ARGO - A_GO - ARGO. A two weeks training period preceded each assessment in order to allow the subjects to get used to standing and walking in the orthosis configuration concerned.

Period effects, i.e. training and test effects, were avoided by applying a 4 weeks guided stance and gait training in the A_GO prior to the assessment phase of the study. All subjects had been previously involved in comparable studies and were well acquainted with testing equipment and procedures.

Training

At the start of the 4 weeks training programme, the Bowden cable was removed from the subject's ARGO. Flexion and extension stops were mounted to the hip hinges in the A_GO configuration and adjusted in

Table 1. Relevant subject information.

Subject	Sex	Age	Time post injury [years]	Lesion level	Weight [kg]
1	M	29	7	T4	79
2	M	40	21	T9	67
3	M	28	3	T4	73
4	M	34	5	T12	66
5	M	45	5	T9	90
6	M	57	5	T9	80

order to provide satisfactory hip angle ranges and step lengths. The training was directed at improving standing balance, obtaining unassisted, regular gait for at least 15 minutes without interruption, and improving physical aerobic capacity. If any objective had not been attained within the four weeks, the training period was prolonged.

Measurements

On each assessment day, a series of measurements was carried out in identical order and at the same time of day. Six measurements, each lasting approximately 1½ minutes including a subject installation procedure, were done for assessment of standing performance:

- 3 measurements comprising the Quiet Standing Test (dual crutch support) followed by the Balance Disturbance Test.
- 3 measurements comprising the Quiet Standing Test (single crutch support) followed by the Hand Function Test.

During the installation, subjects were positioned on a force plate (OR6-5 series, Advanced Mechanical Technology Inc., Newton, USA). The heels were aligned against a reference frame and the feet placed symmetrically with respect to the plate's centre line (Fig. 1). The lateral foot position was set using a centrally placed wedge and kept constant over all all assessments. The reference frame and the wedge were removed after the feet had been positioned.

Crutches instrumented with miniature load cells (LM-100KA, Kyowa Electronic Instruments Ltd., Tokyo, Japan) were used to measure axial crutch forces. Prior to the first assessment of each of the two orthosis configurations, the subject was asked to place the crutches in a comfortable position for prolonged standing. The crutch bottom end position was recorded and marked by a reference plate, and held identical during successive assessments of the same configuration. Different crutch positions for single and dual support were allowed. For single crutch supported standing, subjects were asked to use their non-dominant hand for support.

During all tests, the ground reaction force of the platform and axial crutch loads were sampled by a PC data acquisition setup at 50 Hz. The orientation of the crutches was identified by placing two retroreflective markers near the handle and the bottom end respectively, and measured using a 50 Hz, five camera 3-D movement analysis system (Vicon, Oxford Metrics Ltd., Oxford, UK). The crutch orientation recordings were used to calculate the normal components of the crutch ground reaction forces. All recordings were filtered off-line using a digital linear phase 2nd order Butterworth filter with cut-off frequency at 5 Hz. From the force data the centre of pressure (COP), i.e. the projection of the centre of gravity in the support plane, was calculated by weighted summation of the points of application of the normal ground reaction force components of feet and crutches.

Quiet Standing Test

This test was incorporated to assess stability during quiet standing in either orthosis with use of one or two cratches. Before the start of the 30 seconds test, the subjects were instructed to stand as still as possible and keep their eyes focused on one remote point.

The standing performance was expressed in terms of the range of the COP signal, both in mediolateral and anteroposterior directions. The support area, i.e. the area of the plane stretched by feet and cratches, was calculated from the position of the crutches and the estimated position of the feet (Fig. 1). Crutch axial forces were averaged over the test period to quantify the arm load necessary for maintaining balance.

Balance Disturbance Test

This test was performed during standing with double crutch support only. After the 30 seconds of the Quiet Standing Test had elapsed, three to four anteriorly or posteriorly directed force impulses were applied to the back tube of the subject's orthosis without warning in order to disturb standing balance. The impulses were applied in quasi-random order with 5-10 second intervals, and were generated by a nearly friction-free pneumatic cylinder with electronically operated valves in order to obtain highest possible reproduction. The levels and durations of the force impulses were set separately for anterior and posterior directions prior to the measurements, such that substantial but safe balance disturbances were obtained. Impulse settings were kept constant during all tests, for both ARGO and A_GO. The onset of

the force impluse was measured from the electronic valve actuator.

The effect of the balance disturbance was quantified by the anteroposterior and mediolateral COP ranges. The test performance was quantified by the time, T_{REC}, necessary for the subject to recover from the balance disturbance. T_{REC} for each disturbance was determined jointly by two observers during off-line visual inspection by setting markers in the combined anteroposterior position and velocity graphs of the COP. The criterion used for determining T_{REC} was that the position signal had stabilised at a value close to the value just prior to the onset of the disturbance, which could be accurately decided by simultaneous inspection of the velocity signal (Fig. 2). Visual inspection was preferred to automated calculation because the characteristics of the balance disturbance could not be determined in such a manner, that objective and subjective determinations of T_{REC} showed sufficient agreement.

Hand Function Test

This test was performed only during single crutch supported standing. Following the principles of the Jebsen Test of Hand Function, it consisted of reaching movements of the hand across the body median, while handling a heavy object (Jebsen *et al,* 1969; Triolo *et al,* 1993). The subject was standing in front of a table (width 80cm, depth 60cm), which was positioned at preferred workbench height and close to the body. Five cylindrical weights (1kg; height 15cm, diameter 5cm) were positioned approximately 15cm apart from left to right on the table's front end on 5 differently coloured foam circles. At the back end, identical foam circles were attached in reverse order. The subjects were instructed to move the weights-left to right-to the corresponding circle on the back end as quickly as possible, and back again from right to left. In this way, anteroposterior movements and mediolateral movements passing across the body median were combined in one test.

Prior to the series of three measurements of single crutch supported standing, the test was performed repeatedly in order to allow the subject to get used to the test. Also, for reference, the Hand Function Test was carried out three to four times by the subject while sitting in the wheelchair.

Test result was the time, necessary to complete the 10 displacements (T_{HFT}). Crutch axial force was analysed in order to obtain insight into the average and peak effort required to maintain balance during the test.

Data analysis

For each subject, all results obtained from two repeated measurements of either orthosis configuration were averaged in order to compensate for possible test effects. Variables were presented graphically in order to inspect whether their distributions deviated from a normal distribution. Differences in test results of ARGO and A_GO measurements were statistically tested by means of paired samples t-tests. For all tests, a p-value of 0.05 was considered significant. The results of the tests were expressed also in terms of 95% confidence intervals for the difference, in order to obtain better insight into the relevance of the results. All statistical analyses were performed using the Statistical Package for Social Sciences (SPSS).

Results

General

The results of the first and second ARGO assessments were compared for all subjects in order to check for test effects. Paired samples t-tests showed significant differences in one parameter of the Balance Disturbance Test (Centre of Pressure Anteroprosterior Range for anterior disturbance; $p < 0.03$) and in the reference time score for the Hand Function Test (T_{HFT} Sitting; $p = 0.02$). These differences imply that during the study, despite the measures taken in the design, some training effects were still present. The effects were reduced by averaging the results of the first and second assessment for both ARGO and A G O

None of the standing performance indicators showed substantial deviation from a normal distribution. Therefore, paired samples t-tests were performed on the data without prior transformations.

Further analysis of data was performed *post hoc* in order to study underlying mechanisms.

Quiet Standing Test

In the A_GO, 4 subjects preferred a standing posture in which the trunk was flexed and stabilized by the flexion stops in the hip joints.

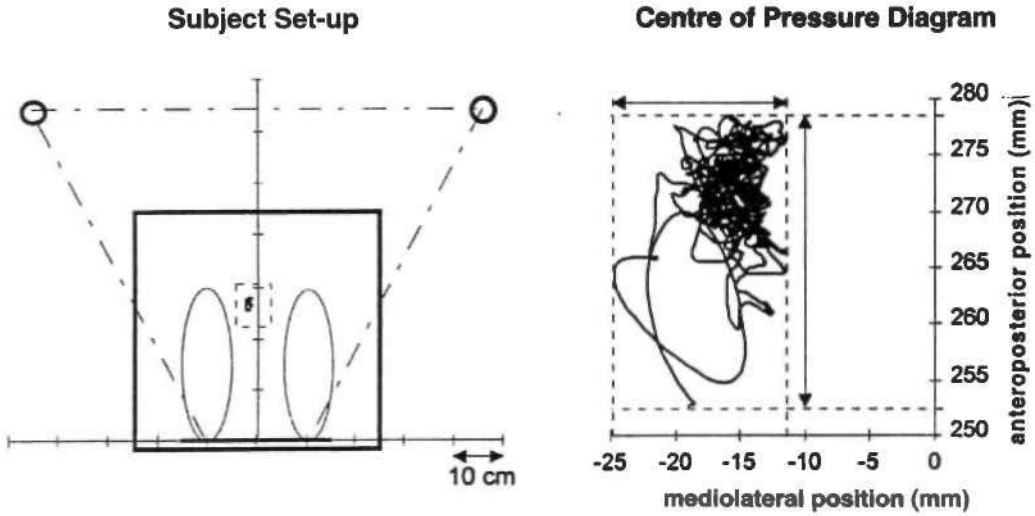

Fig. 1. Subject set-up and typical result of the Quiet Standing Test performed using two crutches. The left graph shows the position of the subject's feet (oval shapes) on the force platform (solid rectangle) and the position of the crutches (circles). The boundaries of the support area are indicated by the dash-dotted line. The small dashed rectangle near the centre of the force platform indicates the centre of pressure (COP) signal, which is presented in detail in the right graph. The right graph shows a typical 30 s recording of the mediolateral and anteroposterior position of the COP during a Quiet Standing Test with dual crutch support, taken from subject 4, Mediolateral and anteroposterior excursions are indicated by the horizontal and vertical arrows respectively.

The 2 other subjects, both having high lesion levels, preferred a C-posture with the hips extended towards the anatomical limit (Andrews *et al,* 1989).

A typical recording of the COP during the 30s test is shown in Figure 1. The COP was located well within the base of support bound by feet and crutches, and was typically located within the support area of the feet.

Standing balance in the ARGO and the A_GO were not significantly different, as can be seen from the mediolateral and anteroposterior ranges of the COP signal of both orthoses (Tables 2 and 3). The support areas selected by the subjects did not differ significantly between orthoses.

Tables 2 and 3 show that the crutch force required for quiet standing in the A_GO

Table 2. Summarised results of the Quiet Standing Test performed with dual crutch support. Presented data are mean values and standard deviations (between brackets). The 95% confidence interval data are presented in absolute values and in values relative to the mean values of the ARGO measurements.

Quiet Standing Test			Paired samples t-test		
A. Dual crutch support	A_GO	ARGO	A_GO-ARGO	p-value	95% C.I.
COP anteroposterior range [mm]	37.94	35.22	2.72	0.74	[-16.91, 22.36]
	(8.12)	(17.38)	(18.70)		*[-48%, +63%]*
COP mediolateral range [mm]	34.53	41.72	-7.19	0.41	[-27.60, 13.21]
	(17.68)	(31.35)	(19.43)		*[-66%, +32%]*
Support area [m²]	0.34	0.34	0.00	0.94	[-0.05, 0.06]
	(0.10)	(0.11)	(0.05)		*[-16%, +17%]*
Crutch axial reaction force [N]	53.94	39.59	14.35	0.06	[-0.48, 29.18]
(averaged left and right)	(11.99)	(17.33)	(14.13)		*[-1%, +74%]*

Table 3. Summarised results of the Quiet Standing Test performed with single crutch support. Presented data are mean values and standard deviations (between brackets). The 95% confidence interval data are presented in absolute values and in values relative to the mean values of the ARGO measurements.

Quiet Standing Test			Pared samples t-test		
B. Single crutch support	A_GO	ARGO	A_GO-ARGO	p-value	95% C.I.
COP anteroposterior range [mm]	30.92	34.83	-3.91	0.64	[-24.06, 16.23]
	(7.20)	(13.58)	(19.19)		*[-69%, +47%]*
COP mediolateral range [mm]	38.04	45.92	-7.88	0.55	[-39.06, 23.31]
	(13.98)	(39.46)	(29.71)		*[-85%, +51%]*
Support area [m²]	0.09	0.08	0.01	0.32	[-0.01, 0.02]
	(0.01)	(0.02)	(0.01)		*[-9%, +24%]*
Crutch axial reaction force [N]	59.30	43.26	16.04	0.32	[-21.27, 53.35]
	(27.15)	(23.30)	(35.54)		*[-49%, +123%]*

orthosis was substantially higher than for the ARGO. Especially for double crutch support, the 95% confidence interval for the difference A_GO-ARGO, relative to ARGO, indicates that this difference can be clinically relevant (relative 95% C.I. [-1%, +74%], $p < 0.06$).

When comparing the results for single and dual crutch support, a remarkable finding was that the (averaged) force per crutch required for maintaining balance was not significantly different for quiet standing with 2 crutches and standing with 1 crutch in either orthosis configuration. For the A_GO, the mean difference was 5.35N, i.e. the force applied to the single crutch was approximately 10% higher than the averaged force for dual crutch supported standing (the relative 95% C.I. was [-35%, +55%]; $p = 0.59$). For the ARGO, the mean difference was 3.67 N, or approximately 9% (relative 95% C.I.: [-38%, +57%]; $p = 0.64$).

For the A_GO, the COP anteroposterior range for standing with 2 crutches was not significantly higher than for standing with 1 crutch (relative 95% C.I. [-19%, +56%]; $p = 0.26$). The mediolateral range for dual crutch support was not significantly smaller: relative 95% confidence interval was [-65%, +44%] ($p = 0.65$). For the ARGO, the relative 95% confidence intervals for these differences were [-41%, +43%] ($p = 0.95$), and [-88%, +67%] ($p = 0.75$), respectively.

The support area was obviously much smaller (approximately 4 times) in the single crutch supported situation than in the dual crutch supported situation ($p = 0.001$ for both A_GO and ARGO).

Balance Disturbance Test

Figure 2 shows a typical recording of the anteroposterior aspect of the COP movement during an anterior balance disturbance (push). The first seconds of the recording clearly show the anterior shift of the COP position resulting from the force impulse applied. In the second part (time > T_{REC}) the position signal has returned closely to the pre-impulse value.

Typical force impulse for anterior disturbance was 325 N during 0.2 s (ranges: 300 to 400 N; 0.1 or 0.2 s), for posterior disturbance 300 N during 0.2 s (ranges: 250 to 375 N; 0.1 or 0.2 s). Tables 4 and 5 show that during the disturbance, the anteroposterior movement was typically 3 to 4 times as much as that during the Quiet Standing Test, while the mediolateral excursion of the COP was comparable to the quiet standing situation. These findings indicate that the disturbances were applied effectively in the anteroposterior direction.

There were no significant differences found in the recovery times of ARGO and A_GO for either anterior or posterior disturbances. Recovery times tended to be slightly lower for anterior disturbances than for posterior disturbances in both orthoses, but differences were not significant (95% C.I., relative to the anterior impulse recovery time, was [-90%, +42%], $p = 0.40$, for A_GO; relative 95% C.I. [-75%, +13%]. $p = 0.13$, for ARGO).

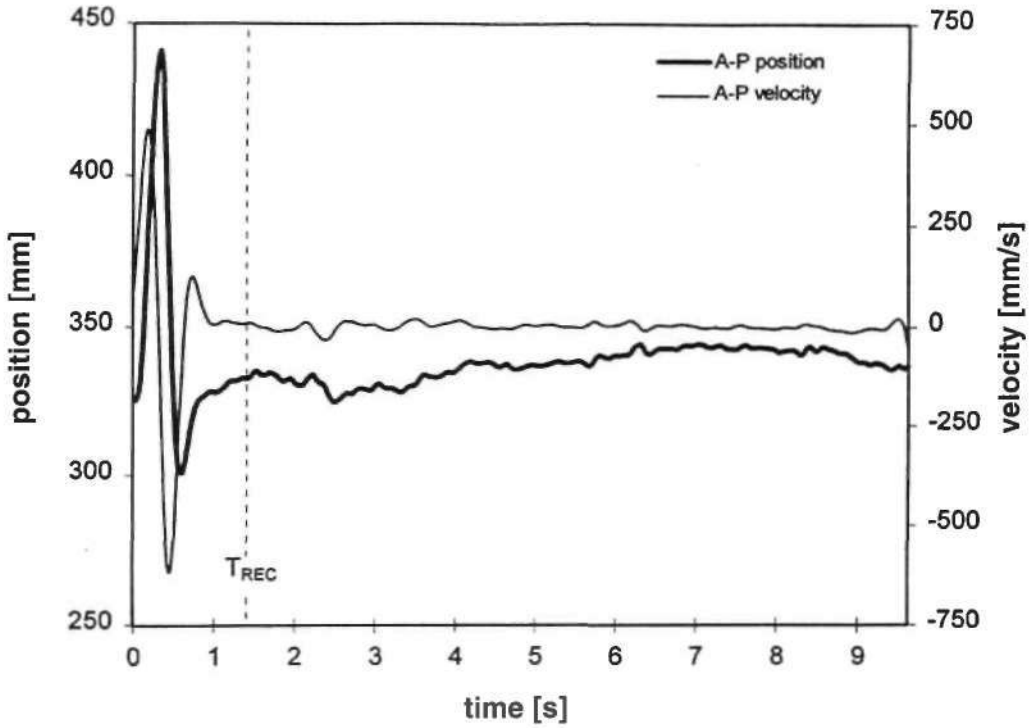

Fig. 2. Part of a centre of pressure (COP) recording measured during a Balance Disturbance Test from subject 4, The recording shown concerns an anterior disturbance with force level of 375 N and duration of 0.2 s, applied at time t=0 The bold solid line represents the anteroposterior position of the COP, the anteroposterior COP velocity is represented by the light solid line. The vertical dashed line indicates the recovery time T_{REC}, obtained from visual inspection of the COP anteroposterior amplitude and velocity signals (see text for details). For this particular case, T_{REC} was determined at 1.42 s.

Hand Function Test

Time scores were found to be higher for the standing situation than during sitting for both orthoses (Table 6). Paired samples t-tests showed that these differences were significant (A_GO: 95% confidence interval for the difference relative to T_{HFT} for sitting was [+1%, + 28%], p = 0.04; ARGO: p = 0.03, 95% confidence interval for the relative difference [+2%, +20%]).

The performance on the Hand Function Test, as indicated by T_{HFT} for stance, was not significantly different between A_GO and ARGO.

Table 4. Summarised results of the Balance Disturbance Test for anterior disturbances (push). Presented data are mean values and standard deviations (between brackets). The 95% confidence interval data are presented in absolute values and in values relative to the mean values of the ARGO measurements

Balance Disturbance Test			Paired samples t-test		
A. Anterior disturbance (push)	A_GO	ARGO	A_GO-ARGO	p-value	95% C.I.
T_{REC} [s]	2.51	2.75	-0.23	0.55	[-1.15, 0.69]
	(1.12)	(0.89)	(0.88)		*[-42%, +25%]*
COP anteroposterior range [mm]	127.31	114.58	12.73	0.50	[-32.03, 57.49]
	(58.63)	(37.01)	(42.64)		*[-28%, +50%]*
COP mediolateral range [mm]	39.44	40.99	-1.56	0.76	[-13.46, 10.35]
	(24.30)	(17.41)	(11.34)		*[-33%, +25%]*

Table 5. Summarised results of the Balance Disturbance Test for posterior disturbances (pull). Presented data are mean values and standard deviations (between brackets). The 95% confidence interval data are presented in absolute values and in values relative to the mean values of the ARGO measurements.

Balance Disturbance Test			Paired samples t-test		
B. Posterior disturbance (pull)	A_GO	ARGO	A_GO-ARGO	p-value	95% C.I.
T_{REC} [s]	3.11	3.61	-0.49	0.48	[-2.15, 1.16]
	(0.80)	(1.29)	(1.58)		*[-60%, +32%]*
COP anteroposterior range [mm]	90.83	100.72	-9.89	0.29	[-31.10, 11.32]
	(22.84)	(21.22)	(20.20)		*[-31%, +11%]*
COP mediolateral range [mm]	38.29	43.65	-5.36	0.60	[-29.95, 19.22]
	(21.87)	(17.54)	(23.42)		*[-69%, +44%]*

Crutch forces applied were typically twice as high as during quiet standing with single crutch support, with peak values of up to 4 times as high. As was found for the Quiet Standing Test, crutch forces necessary for balancing the A_GO during the Hand Function Test tended to be higher than for the ARGO, but differences were not significant.

Discussion

Various studies have been reported which compare the performance of different orthoses, or different orthosis configurations, for persons suffering from paraplegia (Jefferson and Whittle, 1990; Whittle and Cochrane, 1989; Whittle *et al*„ 1991; Winchester *et al*, 1993). These studies relate to walking and compare complete orthotic systems rather than specific design elements. The latter, e.g. lateral stiffness or hip transversal rotation, have been addressed in theoretical studies mainly (Stallard and Major, 1993; Ferrarin *et al.,* 1993; Ferrarin and Rabuffetti, 1996), one exception being a clinical evaluative study which shows that increased lateral stiffness in the Parawalker orthosis has a positive effect on the efficiency of paraplegic gait (Stallard and Major, 1995). It can only be speculated how the results of these studies relate to the pure influence of a component like the reciprocal hip joint coupling on the performance of standing. The present study was directed at adding a piece to this complex puzzle.

The results of this study show that in a variety of situations the presence of a reciprocal coupling of the hip joints in the ARGO has virtually no effect on the performance of standing. Standing balance, as assessed by the range of the COP during the Quiet Standing Test, was not significantly affected by removing

Table 6. Summarised results of the Hand Function Test. Presented data are mean values and standard deviations (between brackets). The 95% confidence interval data are presented in absolute values and in values relative to the mean values of the ARGO measurements.

Hand Function Test			Paired samples t-test		
	A_GO	ARGO	A_GO-ARGO	p-value	95% C.I.
T_{HFT} Sitting [s]	10.08	10.02	0.06	0.80	[-0.52, 0.65]
	(0.78)	(0.42)	(0.56)		*[-5%, +6%]*
T_{HFT} Standing [s]	11.54	11.12	0.42	0.25	[-0.40, 1.24]
	(1.73)	(1.16)	(0.78)		*[-4%, +11%]*
Crutch axial reaction force [N]	119.04	101.74	17.30	0.38	[-28.33, 62.93]
(Average)	(33.57)	(21.76)	(43.47)		*[-28%, +62%]*
Crutch axial reaction force [N]	198.00	179.75	18.25	0.40	[-32.80, 69.30]
(Peak)	(42.32)	(41.83)	(48.63)		*[-18%, +39%]*

the Bowden cable from the orthosis. The ability of maintaining balance in the presence of disturbances, i.e. standing stability, was not significantly different in the ARGO and the A_GO. The performance on the test of hand function was comparable for both orthoses.

There is, however, a strong indication that the crutch force required for maintaining balance in the ARGO was lower than in the A_GO, and that this difference may take on clinical relevance. The most likely explanation for this result is that a stable posture in the A_GO was achieved by most subjects by leaning against the flexion stops built into the orthosis' hip hinges. In this posture extra force is required for compensation of the horizontal component of gravity resulting from the forward inclination. The effect of this mechanical difference between standing in the ARGO and the A_GO may even be toned down in the results because two subjects preferred to stand in the A_GO in the so-called 'C-posture', i.e. 'leaning' against anatomical hip extension limits (Andrews *et ai,* 1989). Since the 'C-posture' is more upright than the flexed posture, the resulting horizontal component of gravity, and thus the extra arm force required for balancing, is smaller. It is clear that this inhomogeneity, as well as the low number of subjects, has an adverse influence on the statistical power of the study.

In the ARGO, the moment required for keeping the trunk erect is generated by the trunk corset, because necessary forces are transferred through the reciprocal link to the upper leg sections. Therefore, in this orthosis configuration only balancing forces have to be provided by upper body effort.

Crutch force, especially from a clinical point of view, is an important indicator in functional assessments, since shoulder and wrist problems form a major threat to the successful and prolonged application of orthotic devices in paraplegia (Gellman *et ai,* 1988). The finding that the difference in standing performance resulting from removing the reciprocal hip joint link from the ARGO lies exactly in the required upper body effort, is therefore greatly relevant.

The results of the Quiet Standing Test illustrate the mechanisms underlying the choice of posture in relation to stability. It was found that the force applied on each crutch was approximately the same for single and dual crutch supported standing. In other words, in the single crutch supported case the subjects took more weight on the feet than during double crutch support and, as a consequence, the mean position of the centre of pressure was shifted posteriorly.

This posterior shift is most likely a compensation for the changed geometry of the base of support. In single crutch supported standing, the anterior edge of the base of support extends diagonally from the front of the foot to the contralateral crutch contact point. Compared with the double crutch support situation, the distance from the anterior edge of the base of support to the COP is greatly smaller. As a consequence, the stability margin is reduced and a new optimum location of the COP must be found by posteriorly shifting weight (Karcnik *et ai,* 1995). Given this effect on stability, it is striking that the range of the COP was found not to be influenced by the number of crutches used for support. Apparently, for standing balance, the area of the base support is not relevant.

If the mechanical properties of standing in both orthoses is considered it would have been expected that differences would be found in the results of the Balance Disturbance Test for ARGO and A_GO. While quiet standing in the ARGO could be best compared with balancing an inverted pendulum, a more suitable description for the A_GO would be an inverted double pendulum. In the A_GO, four subjects chose to lean against the hip flexion stops in order to obtain a mechanically stable standing posture. It would then be expected that perturbations in anterior direction (i.e. pushes applied to the back tube of the orthosis) would cause a temporary deviation from this posture because a hip extension movement would occur. Consequently, the recovery from this change in posture would take less time than from a perturbation resulting from an indentical force impulse in the ARGO, since the inertia of the double pendulum would be lower. During the tests it was found however that, due to the high flexion moments around the hip in the A_GO, the described posture deviation did not occur at the force levels applied for balance perturbance. An analogue description holds for posterior balance disturbances applied to subjects that preferred the C-posture for standing in the A_GO. Though forces were indeed high enough to cause a temporary

flexion movement of the hips directly after the impulse, this effect did not result in significant differences in the posterior disturbance recovery time.

The results of the study lead to the conclusion that although the standing performance of ARGO and A_GO do not vary much, the reciprocal hip joint link in the Advanced Reciprocating Gait Orthosis provides a substantial and clinically relevant reduction of upper body effort necessary for maintaining a stable posture under functional conditions. Therefore, the incorporation of a reciprocal hip joint linkage, or any other mechanism providing the same stabilising properties, is highly recommendable in HKAFO design.

Acknowledgement

The authors wish to thank all volunteers participating in this study and J.Th.M.M. Cloostermans, PT, and J.H.C. Vorsteveld, MD, for their continuous support. The help of the Adaptation Technology department of Rehabilitation Centre Het Roessingh during the construction of the pneumatic balance disturbance device is gratefully acknowledged.

This study was supported by the Technology Foundation (STW)

REFERENCES

ANDREWS BJ, BARNETT RW, PHILLIPS GF, KIRKWOOD CA, DONALDSON N, RUSHTON DN, PERKINS TA (1989). Rule-based control of a hybrid FES orthosis for assisting paraplegic locomotion. *Automedica* 11, 175-199.

CYBULSKI GR, JAEGER RJ (1986). Standing performance of persons with paraplegia. *Arch Phys Med Rehabil* 67, 103-108.

DOUGLAS R, LARSON PF, D'AMBROSIA R, MCCALL RE (1983). The LSU reciprocating gait orthosis. *Orthopedics* 6, 834-839.

FERRARIN M, STALLARD J, PALMIERI R, PEDOTTI A (1993). Estimation of deformation in a walking orthosis for paraplegic patients. *Clin Biomech* 8, 255-261

FERRARIN M, RABUFFETTI M, (1996) On the improvement provided by hip transversal rotation on paraplegic gait walking with reciprocating orthosis. In: Neuroprosthetics - from basic research to clinical applications / edited by A Pedotti, M Ferarin, J Quintern, R Reiner.-Berlin: Springer-Verlag. p493-502.

FIGONI SF (1984). Cardiovascular and haemodynamic responses to tilting and to standing in tetraplegic patients: a review. *Paraplegia* 22, 99-109.

GELLMAN H, SIE I, WATERS RL (1988). Late complications of the weight-bearing upper extremity in the paraplegic patient *Clin Orthop* 233, 132-135.

GEURTS ACH, NIENHUIS B, MULDER THW (1993). Intrasubject variability of selected force-platform parameters in quantification of postural control. *Arch Phys Med Rehabill H,* 1144-1150.

GOLDIE PA, BACH TM, EVANS OM (1989). Force platform measures for evaluating postural control: reliability and validity. *Arch Phys Med Rehabil* 70, 510-517.

UZERMAN MJ, BAARDMAN G, HERMENS HJ, VELTINK PH, BOOM HBK, ZILVOLD G (1997). The influence of the reciprocal cable linkage in the advanced reciprocating gait orthosis on paraplegic gait performance. *Prosthet Orthot Int* 21, 52-61.

JEBSEN RH, TAYLOR N, TRIESCHNANN RB, TROTTER MH, HOWARD LA (1969). An objective and standardized test of hand function. *Arch Phys Med Rehabil* 50, 311-319.

JEFFERSON RJ, WHITTLE MW (1990). Performance of three walking orthoses for the paralysed: a case study gait analysis. *Prosthet Orthot Int* 14, 103-110.

KARCNIK T, KRALJ A, BAJD T (1995). Stability and energy criteria in healthy and paraplegic subjects gait. *In: Proceedings of the 5th Vienna International Workshop on Functional Electrostimulation.-Vienna: Austrian Society for Artificial Organs.* p87-90*

KUNKEL CF, SCREMIN AM, EISENBERG B, GARCIA JF, ROBERTS S, MARTINEZ S (1993). Effects of "standing" on spasticity, contracture and osteoporosis in paralyzed man. *Arch Phys Med Rehabil* 74, 73-78.

MAYAGOITIA RE, Andrews BJ (1989). Stability in FES and orthoses assisted standing. *In: Proceedings 3rd Vienna International Workshop on Functional Electrostimulation.-Vienna: Austrian Society for Artificial Organs* p85-88.

MESSENGER N, RITHALIA SV, BOWKER P, OGILVIE C (1989). Effects of ambulation on the blood flow in paralyzed limbs, *J Biomed Eng* 11, 249-252.

MOTLOCH W (1992). Principles of orthotic management for child and adult paraplegia and clinical experience with the Isocentric RGO. *In: Proceedings of 7th World Congress of the International Society in Prosthetics and Orthotics, Chicago, June 28 - July 3, 1992. -Alexandria VA: ISPO.* p28

MOYNAHAN M, CHIZECK HJ (1993), Characterization of paraplegic disturbance response during FNS standing. *IEEE Trans Rehabil Eng* 1, 43-48.

NENE AV, MAIOR RE (1987) Dynamics of reciprocal gait of adult paraplegics using the Para Walker (Hip Guidance Orthosis). *Prosthet Orthot Int* 11, 124-127.

NENE AV, HERMENS HJ, ZILVOLD G (1996). Paraplegic locomotion: a review. *Spinal Cord* 34, 507-524.

OGILVIE C, Bowker P, Rowley DI (1993). The physiological benefits of paraplegic orthotically aided walking. *Paraplegia* 31, 111-115.

ROSE GK (1979). The principles and practice of hip guidance articulations. *Prosthet Orthot Int* 3, 37-43.

SLOBOUNOV S, NEWELL KM (1994). Postural dynamics as a function of skill level and task constraints. *Gail Posture* 2, 85-93.

STALLARD J, MAJOR RE, PATRICK JH (1989). A review of the fundamental design problems of providing ambulation for paraplegic patients. *Paraplegia* **27,** 70-75

STALLARD J, MAJOR RE (1993). The case for lateral stiffness in walking orthoses for paraplegic patients. *J Eng Med* **207H,** 1-6.

STALLARD J, MAJOR RE (1995). The influence of orthosis stiffness on paraplegic ambulation and its implications for functional electrical stimulation (FES) walking systems. *Prosthet Orthot Int* 19, 108-114.

TRIOLO RJ, REILLEY BWB, FREEDMAN W, BETZ RR (1993). Development and standardization of a clinical evaluation of standing function: the functional standing test. *IEEE Trans Rehabil Eng* 1, 18-25.

WHITTLE MW AND COCHRANE GM (1989). A comparative evaluation of the Hip Guidance Orthosis (HGO) and the Reciprocating Gait Orthosis (RGO) *Health Equipment Information*, No. 192.-London: NHS Procurement Directotate

WHITTLE MW, COCHRANE GM, CHASE AP, COPPING AV, JEFFERSON RJ, STAPLES DJ, FENN PT, THOMAS DC (1991). A comparative trial of two walking systems for paralysed people. *Paraplegia* 29, 97-102.

WINCHESTER PK, CAROLLO JJ, PAREKH RN, LUTZ LM, ASTON JW, JR (1993). A comparison of paraplegic gait performance using two types of reciprocating gait orthoses. *Prosthet Orthot Int* 17, 101-106.