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Saudi J Disabil Rehabil 2002;8(1):20-32

KNEE FLEXION DURING THE SWING PHASE OF ORTHOTIC GAIT: INFLUENCE ON KINEMATICS, KINETICS AND ENERGY CONSUMPTION IN TWO PARAPLEGIC CASES

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لقد تمت دراسة حيوية مشية ذوي الشلل السفلي والتي تشمل ثني الركبة خلال طور التأرجح لدى استعمال جهاز تعويضي للورك والركبة والكاحل والقدم ذي الميكانيزم الآلي من أجل التحكم في مفصل الركبة . عنصرا التجربة مصابان بعطب في الحبل الشوكي الأول ذكر مصاب عند مستوى الفقرة الصدرية ١٢ والثاني أنثى مصابة عند مستوى الفقرة الصدرية ٤ ، وكلاهما مكتمل الإصابة شاركا في تجربة مقارنة للمشي بما يشابه جهاز مشي تعويضي ترددي للاختبار بركبة مقفلة ونفس الجهاز التعويضي بمكانيزم تحكم في الركبة عن طريق التعشيق ، وقد سجلت النتائج الانطباعية فيما تتعلق بدرجة السلامة والاستعمال والناحية الحركية ، كما تم تسجيل بارائرات الحركة والتحريك واستهلاك الأوكسجين لتقييم الأداء للمشية وكفاءة الطاقة . استطاع عنصرا التجربة المشي بواسطة الجهاز التعويضي دون مساعدة لفترات طويلة على أسطح مغايرة وشعرا بأن ثني الركبة يساعد على التحرك فوق السطح غير السوي ويحسن من جمالة ومظهر المشية ، ونسبة لشعورهما بعدم الأمان والسلامة لم يود أي منهما استخدام جهاز الاختبار التعويضي بدلاً عن جهازيهما التعويضي الترددي في الأغراض اليومية ، وقد وجد بأن إدخال ثني الركبة منعدم أو قليل الأثر على متغيرات النتائج الكينماتيكية وقد مالت الآثار المتحصل عليها للاختلاف بين عنصري التجربة إذ وجدت فوارق كبيرة بالنسبة للعنصر الأول في بقوة العكاز ، حيث كان عالياً في الجهاز التعويضي ذي التحكم في ثني الركبة . كما وهناك تدني كبير في قوة العكاز العظمي خلال مرحلة التأرجح بالنسبة للعنصر الثاني . لقد وجدت بأن استهلاك الطاقة كان أعلى من حالة الجهاز التعويضي ذي التحكم في ثني الركبة للعنصر الأول وأدنى للعنصر الثاني . خلصت الدراسة إلى أن إدخال ثني الركبة في مرحلة التأرجح قد يقدم قيمة إضافية لمشية ذوي الشلل السفلي الذين يستخدمون أجهزة تعويضية وبقدر أقل بالنسبة لاستهلاك الطاقة ، ولكن في الغالب للمناحي الوظيفية والجمالية .

The viability of paraplegic gait incorporating knee flexion during the swing phase was studied in a Hip-Knee-Ankle-Foot-Orthosis with automatic mechanism for the control of the knee hinge lock.
Two complete thoracic spinal cord injured subjects (1 male T12, 1 female T4) participated in a comparative trial of gait in the ARGO-like test orthosis with locked knee and the same orthosis with engaged knee control mechanism. Subjective findings were recorded with respect to experienced safety, usability and cosmesis. Kinematics, kinetics and oxygen uptake were measured to assess gait performance and energy efficiency.
Both subjects were able to walk in the orthosis unassistedly for prolonged periods on varying surfaces. Knee flexion was felt to facilitate ambulation on uneven terrain, and to improve cosmesis of gait. Due to an experienced feeling of insecurity, neither of the subjects would prefer to use the test orthosis instead of their own ARGO for daily use.
The introduction of knee flexion in the gait pattern was found to have little or no effect on kinematic and kinetic outcome variables. The effects found tended to be different for subject 1 and subject 2. A significant difference was found for subject 1 in the crutch force time integral (or linear impulse), which was higher in the orthosis with controlled knee flexion. A significant reduction in the crutch peak force applied during the swing phase was found for subject 2. Energy consumption was found to be higher in the orthosis with controlled knee flexion for subject 1, and lower for subject 2.
It was concluded that the incorporation of knee flexion in the swing phase may provide a modest added value to paraplegic orthotic gait, not so much perhaps with respect to energy consumption, but more likely with respect to functional aspects and cosmesis.

Key Words: *Paraplegia, Orthosis, Knee Flexion, Gait Analysis, Oxygen Consumption*

INTRODUCTION

Walking has been an increasingly important item of rehabilitation programmes for spinal cord injured patients over the last two or three decades. To date, various mechanical devices for the restoration of walking are available for people with paraplegia, ranging from long leg braces or knee-ankle-foot-orthoses (KAFOs) for low level injured[1], via the Walkabout linked KAFOs[2], to hip-knee-ankle-foot-orthoses (HKAFOs) for thoracic paraplegia, such as

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the Reciprocating Gait Orthosis (RGO)[3], the Hip Guidance Orthosis or ParaWalker[4], and the Advanced Reciprocating Gait Orthosis (ARGO)[5].

It is well recognized that performance of gait in paraplegia, irrespective of the orthosis used, is limited[6]. Obviously, this is due in the first place to the impaired or absent control of the lower extremities: paraplegic walking is essentially an upper body effort. But also, an orthosis itself can only provide limited function. The reason for this is that an orthosis must provide sufficient stability to the user and allow movements which are related to the forward progression of the body[7]. To some extent, these requirements are contradictory. Since stability is the predominant factor, major compromises have been made in most orthoses at the expense of efficiency of progression.

This certainly holds for HKAFOs, in which the only movement not blocked is the rotation of the hip joints in the sagittal plane. As a result, only a rudimentary reciprocal type of gait can be applied in these devices. This type of gait has been indicated by the term compass gait, and has been associated with very high energy expenditure[8,9]. In gait of able bodied persons, various mechanisms provide a reduction of the vertical and horizontal movement of the centre of mass of the body, which has been assumed to connect to a reduction of metabolic energy requirements[9]. Pelvic transversal rotation, pelvic tilt (necessarily accompanied by knee flexion during the swing phase) and knee flexion in the stance phase flatten the arc described by the centre of mass during forward translation of the body over the stance foot. The roll-off mechanism provided by foot and knee smoothens the transitions between consecutive steps. Pelvic lateral displacement, finally, is reduced by relative adduction and abduction of the stance side and swing side hips respectively.

None of these so-called determinants of gait are present in currently available HKAFOs, apart from coincidental effects resulting from limitations in mechanical properties of these devices. It is not surprising therefore, that various studies have shown that energy efficiency of paraplegic gait is very low. Nene and Patrick[10] studied the energy requirements (as calculated from oxygen uptake data) of 10 complete thoracic paraplegic subjects who walked in the ParaWalker orthosis. They found that energy cost of walking, i.e. energy requirement per unit distance, in this group was 4.9 times the cost quoted from historical data for normal subjects, at a self-selected speed that was approximately 15% of normal walking speed. Identical results were found in a study on the energy requirements of 6 thoracic paraplegic persons walking in the RGO[11]. Massuci et al concluded the great energy cost during walking to be a major factor for abandoning orthotic devices in paraplegia[12].

These and other findings have brought engineers and clinicians to the conviction that reduction of the energy requirements of gait is a prerequisite for functional application of orthoses in paraplegia[7]. This conviction has indeed been the prime motive for research on principles that could improve the performance of orthoses.

Many studies have been directed to the improvement of characteristics of the orthosis design. The effect of increasing the lateral stiffness of ParaWalker hip hinges was tested in a study on three paraplegic subjects. It was found that in the ParaWalker mounted with new 70% stiffer hip hinges, the average Physiological Cost Index (PCI*) was 30% lower than in the original ParaWalker[13]. A reduction of 15% on average in crutch forces exerted during gait was found to result from altering the alignment in the frontal plane of the ARGO upper leg steels from 6° adduction (the standard alignment) to neutral or 3° abduction. This reduction in upper body effort shown in 5 subjects, however, was not reflected in oxygen uptake[14]. The Isocentric® RGO[15], an RGO in which the reciprocal cable coupling is replaced by a centrally pivoting bar, was compared to the conventional RGO in a study on 4 complete paraplegic subjects. No significant difference in the oxygen cost obtained from the Isocentric® RGO and the original RGO was found, but PCI* was found to be 28% lower in the Isocentric® RGO[17]. Harvey and co-workers concluded that the energy demands of the walkabout orthosis are higer than the Isocentric® RGO in T9-T12 paraplegia[18]. Yang and co-workers studied the effects of varying the hip flexion-extension ratio of the reciprocal hip joint link in 3 normal subjects walking in an RGO-like assessment orthosis[19]. They found that the average PCI in orthosis configurations with hip flexion-extension ratio of 2:1 or free hips was 20% lower than in comparable configurations with conventional 1:1 flexion-extension ratio, while walking speed in these

* $PCI = \frac{\text{heart rate (walking, steady state) - heart rate (rest)}}{\text{Walking speed}} [\text{beats.m}^{-1}]$[16]

configurations was 5% and 10% higher, respectively.

Fewer studies report on trying to incorporate the determinants of gait in the orthosis design. Ferrarin and Rabuffetti[20] have recently described a kinematic evaluation of a new hip hinge for the RGO which, in addition to the reciprocal coupling of flexion and contralateral extension, provides pelvic transversal rotation during gait. It was found that the vertical displacement of the body centre of mass was reduced from 39 to 31 mm. in one paraplegic subject (as opposed to 20 mm. in a normal subject), but the influence on energy expenditure was not assessed directly. In the study on three normal subjects mentioned before, Yang and co-workers also studied the effects of allowing the knee to flex, and allowing the ankle to plantarflex during early stance phase. Allowing the knee to flex reduced the average PCI and increased walking speed, especially when combined with the possibility to plantarflex the ankle[18]. The influence of incorporating knee flexion during the swing phase of paraplegic gait was addressed in a case study of a T6 complete paraplegic man, who walked in an RGO with bilateral lockable-unlockable knee joints (LUK-RGO). It was found that PCI when walking with a rollator was 12% lower in the LUK-RGO than in a conventional RGO; no data on walking speed were reported[21].

The present study was also directed to the incorporation of knee flexion in paraplegic gait, and was part of our research in the framework of the design of a user friendly, energy efficient hybrid orthosis. The aim of this study was to explore whether the incorporation of knee flexion during the swing phase would be advisable for improvement of orthotic paraplegic gait.

We conducted two comparative case studies of gait with locked knees and gait with knee flexion during the swing phase using an ARGO-like test orthosis. The opinion of the subjects was asked on experienced safety, usability and cosmesis. Kinetic and kinematic quantities, and oxygen uptake were measured to establish the influence of knee flexion on gait performance, and the role of walking speed herein.

METHODS

Subjects

The first subject was a 35 years old man with traumatic complete lesion at T12, who had 4 years of RGO and ARGO experience. He very regularly walked at home and had been involved in similar studies previously. The second subject was a 32 years old woman with traumatic complete T4 lesion, who had 4 years of ARGO experience. She had been regularly involved in clinical tests of (hybrid) orthoses. Other relevant subject information is included in Table 1.

Table 1. Relevant subject data

Subject	Gender	Age [years]	Weight [kg]	Lean Body Weight [kg]	Lesion Level	Time Post Injury [years]
1	M	35	66.0	50.1	T12 complete	5
2	F	32	59.0	44.5	T4 complete	6

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

Orthosis

A prototype Hip-Knee-Ankle-Foot-Orthosis was designed for this study, which incorporated a mechanism for the automatic control of the knee hinge lock (Figure 1). The orthosis had an ARGO-like open structure without upper leg mouldings. Two sets of adjustable straps were used for fixation of the pelvis and the chest. A reciprocal link of the hip joints, constructed with a single Bowden driving cable, was incorporated to provide stability to the trunk during standing and gait. For easy fitting to different subjects, the back tube width was adjustable and the knee hinge could be connected to their ARGO's lower leg sections. In this way, only the trunk corset and upper leg side steels had to be tailor-made.

Training

Both subjects were trained extensively in the orthosis during approximately 3 months prior to the measurements. The training was directed at obtaining unassisted, regular gait for a minimum of 15 minutes, with a special focus on the safe operation of the knee control mechanism. Some outdoor training was incorporated in order to establish the reliability and safety of the knee lock control mechanism under different circumstances, and to get an impression of its surplus value in daily life activities.

Measurements

Both subjects were assessed once in the orthosis with automatic control of the knee, and once in the same orthosis with the control mechanism disengaged.

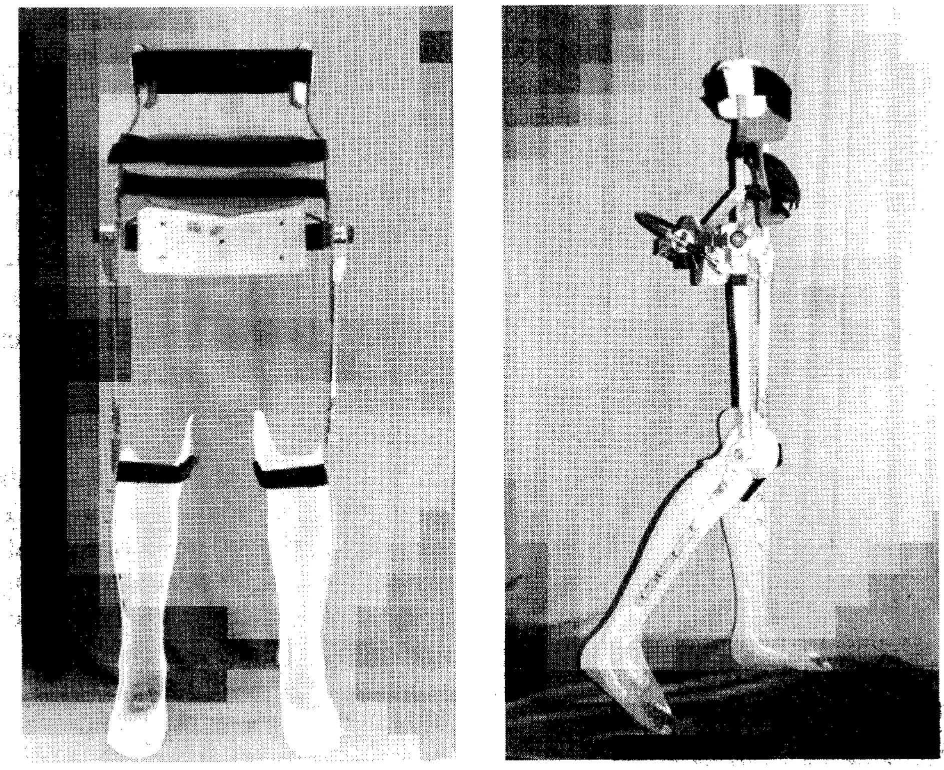

Figure 1. Frontal (left) and right sagittal (right) views of the orthosis prototype with built-in mechanism for the automatic control of the knee hinge lock. The right picture shows the knee lock control mechanism at work: the knee is flexed during the swing phase. A knee joint control mechanism was integrated in the upper leg sections of the orthosis, that senses the hip angle as well as the knee moment and actuates the knee hinge lock. At terminal stance phase, provided that the hip be extended and an extension moment be present at the knee, the knee hinge is unlocked, thus allowing flexion in the ensuing swing phase. At ternimal swing, the knee hinge is locked again as the knee is extended, in order to provide stability during the next stance phase. The control mechanism could be engaged or disengaged manually by means of the operating button which is placed on the hip joint axis.

Kinematics and kinetics of gait were measured from subject 1 at three different walking speeds, viz. slow, self-selected, and fast, in order to assess the influence of walking speed on the efficacy of knee flexion. Since subject 2 was unable to vary walking speed over a sufficiently large range, she was assessed at self-selected speed only.

Energy consumption of gait was measured separately. Since his self-selected walking speed had shown to vary between the orthosis configurations during the training, energy requirements were measured from subject 1 over a range of walking speeds on a treadmill. In this way, the speed versus energy uptake relations for both orthosis configurations could be established. For subject 2, a treadmill measurement was considered neither realistic nor safe. Since furthermore, self-selected walking speed in her case was the same in both orthosis configurations, a direct comparison of energy requirements at this speed was performed.

Kinematics and kinetics

A 50 Hz, 3-dimensional 5-camera video movement analysis system (Vicon 370, Oxford Metrics Ltd., Oxford, UK) was used to measure the kinematics and kinetics of the subjects' gait. Nine retroreflective markers were placed bilaterally on the shoulders, hip hinges, knee hinges and ankles, and centrally on the back tube of the orthosis. One reference measurement was performed while the subject was standing still in order to obtain offset values for the kinematic quantities to be calculated. Miniature load cells (LM-100KA, Kyowa Electronic Instruments Ltd., Tokyo, Japan), integrated in the lower tube of both crutches, were used to measure axial crutch reaction forces. Heel contacts were measured using capacitive switches which were attached to the shoe heels. These sensor signals were acquired also by the Vicon system at 200 Hz. For each combination of orthosis configuration and walking speed, the subjects walked up and down a calibrated video measurement space repeatedly, until approximately 10 strides had been successfully recorded.

Video and sensor data were filtered off-line using digital linear phase 2nd order Butterworth filters with cut-off frequency at 5 Hz. From the marker data transients of the knee angles, pelvic elevation and pelvic lateral sway were calculated using the offsets from the reference measurement. The heel contact signal was used to split the crutch force and kinematic data into gait cycle sections (left heel strike to left heel strike), which were then analysed. Peak knee flexion angle, pelvic elevation and lateral sway ranges, crutch peak forces at swing and stance leg sides, and crutch force time integral (or linear impulse) were calculated for each recorded cycle and averaged for either orthosis configuration and each walking speed. Averages for stride length, step frequency and walking speed were also calculated.

Energy consumption

Metabolic energy requirements during gait were assessed using an open spirometry system (Oxycon Alpha, Jaeger Benelux b.v., Breda, The Netherlands). Subjects were equipped with a face mask covering mouth and nose, which was connected to the spirometry system by means of a flexible air tube. Breath-by-breath analysis was performed on inspired and expired gases to yield values for oxygen consumption. Signals were sampled at 200 Hz.

Subjects were instructed to refrain from food and coffee for at least two hours prior to the measurements. Both were non-smokers.

For subject 1, both assessments started with a 5 min. measurement during quiet standing, in order to check for steady state rest metabolism. The treadmill speed was then increased in steps of approximately 0.1 ms^{-1} to maximum walking speed. Each speed setting was maintained during 5 minutes in order to obtain steady state oxygen uptake. Analysis of respiratory gases was performed on the last minute of each 5 minutes recording. Average oxygen consumption ($\dot{V}O_2$), i.e. oxygen uptake per unit of time, was calculated for each speed.

For subject 2, the measurements were performed in a long carpeted corridor. During the first assessment walking distance was recorded at regular time intervals. These data were used during the second assessment for instructing the subject to adjust her walking speed, if necessary. Both assessments comprised a 5 min. baseline metabolic measurement during quiet standing, followed by a 10 min. measurement of respiratory gases during gait. Average values for $\dot{V}O_2$ were calculated from the last 5 minutes of the recordings, in which metabolism had reached steady state.

Analysis

Data obtained at different walking speeds from subject 1 were grouped in order to establish the effect of walking speed on the efficacy of knee flexion, and the influence of knee flexion on kinematic and kinetic parameters and oxygen uptake. The relations between walking speed and outcome variables for the orthosis configuration with locked knees and the one with controlled knees were described by means of regression equations, and were compared using analysis of covariance.

For subject 2, averaged outcome variable data for both orthosis configurations were calculated. In view of the limited and varying numbers of observations obtained from the kine(ma)tic assessments of both orthosis configurations, comparisons were carried out using the non-parametric Mann-Whitney U test for independent samples. Differences between the results of oxygen consumption measurements were established using a t-test for paired samples. Statistical testing was performed with a significance level of 5 %.

RESULTS

General

After the training period, both subjects were able to walk unassistedly with two elbow crutches for prolonged periods. Walking was possible on flat surfaces as well as uneven terrain. Both subjects also seemed to be well capable of handling the orthosis when the knee control mechanism was engaged. Bringing the swing leg forward was felt to be easier than when the knees were locked.

On being asked, neither preferred the test orthosis with knee flexion control over their own ARGO for unassisted home use, since both felt less confident than at walking with locked knees. One reason for this was the uncertainty felt by the subjects due to a perception

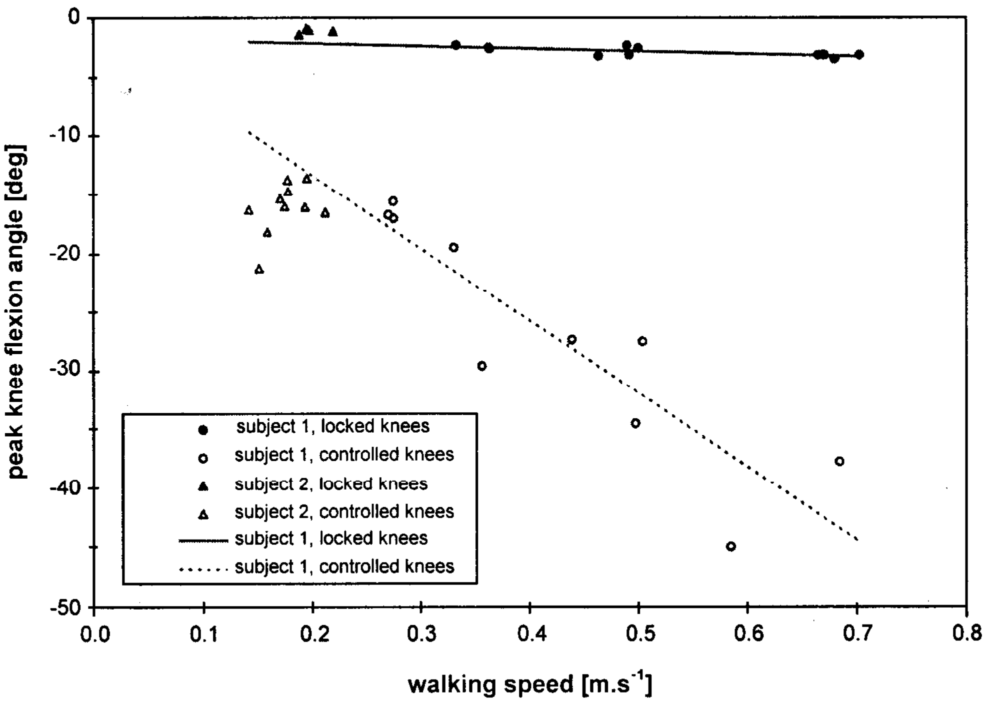

Figure 2. Speed vs. knee flexion relations of both orthosis configurations obtained from measurements at three different wlaking speeds for subject 1 (O), and at self-selected speed for subject 2 (Δ). Closed symbols represent the orthosis configuration with locked knees, open symbols represent the configuration with controlled knees. Presented data are averages for left and right peak knee flexion angles obtained from individual strides. The solid and dashed lines represent the linear regression equations of the data for subject 1 for the orthosis configuration with locked and controlled knees, respectively.

of reduced sensory feedback from the orthosis. A second reason was the feeling of unsafety or fear to fall, perceived mainly by subject 1 at higher walking speeds, caused by occasional uncontrolled occurrences of knee flexion.

Subject 1 indicated that the knee flexion control mechanism was especially useful for negotiating uneven surfaces and slanting pavements. On watching their video recordings, both subjects judged the cosmesis of gait with controlled knee flexion better than gait with locked knees, because it better resembled normal gait.

Kinematics and kinetics

Subject 1 was well capable of walking at three different speeds in both orthosis configurations. Slow walking speeds were approximately 30% lower, fast walking speeds approximately 30% higher than self-selected walking speeds in both configurations. Ranges of walking speeds were comparable for both orthosis configurations, and varied from approximately 0.3 $m.s^{-1}$ to 0.7 $m.s^{-1}$.

Self-selected speed and cadence for subject 2 were not significantly lower in the orthosis configuration with controlled knee flexion than in the configuration with locked knees, and was accompanied by significant smaller stride length (see Table 2).

The influence of walking speed on knee flexion efficacy

Figure 2 shows the relationship between walking speed and the peak knee flexion angle during the swing phase. Data presented are average peak knee flexion angles for left and right side. As can be seen, the peak knee flexion angle in the orthosis configuration with controlled knees varied linearly with walking speed for subject 1. Even at high walking speed, this angle was much smaller than the peak knee flexion angle observed during normal gait, which is approximately 60°.

At an average self-selected speed of 0.18 $m.s^{-1}$, average peak knee flexion angle for subject 2 was approximately 16°. As can be seen in figure 2, this result is not essentially different from the results obtained from subject 1 at that speed.

Table 2. Comparison of spatio-temporal, kinematic and kinetic parameters obtained for subject 2 from measurements at self-selected speed in two orthoses configurations. Presented data are means and standard deviations (in parentheses) as obtained from averaging several recorded strides. Peak knee flexion angle, and crutch force parameters presented are averages of left and right recordings. Significance of differences between the orthosis configurations was calculated using the Mann-Whitney U test for independent samples.

Subject 2: Kine(ma)tic assessment		Orthosis configuration		Difference*
		Locked knees	*Controlled knees*	
Walking speed	[m.s^{-1}]	0.20 (0.01)	0.18 (0.02)	n.s.
Stride length	[m]	0.84 (0.04)	0.79 (0.04)	p<0.05
Cadence	[strides. min^{-1}]	14.4 (1.5)	13.4 (1.2)	n.s.
Peak knee flexion angle	[deg]	-1.2 (0.5)	-16.1 (2.2)	p<0.05
PE_{RANGE}	[mm]	39.1 (0.8)	35.4 (4.0)	n.s.
PS_{RANGE}	[mm]	120.9 (29.0)	133.7 (25.0)	n.s.
CPF_{SWING}	[N]	322.7 (11.3)	295.4 (20.4)	p<0.05
CPF_{STANCE}	[N]	275.3 (20.0)	289.2 (17.8)	n.s.
CFTI	[N.s]	506.9 (36.4)	522.2 (37.7)	n.s.

*Non-parametric Mann-Whitney U test for independent samples
n.s. = not significant

The influence of knee flexion on kinematics and kinetics

The incorporation of knee flexion in the swing phase of gait was accompanied by small changes in the relation between speed and kinematic and kinetic quantities for subject 1. In both orthosis configurations, pelvic elevation range (PE_{RANGE}), pelvic lateral sway range (PS_{RANGE}), crutch peak forces on swing leg and stance leg sides (CPF_{SWING} and CPF_{STANCE}, respectively), and crutch force time integral (CFTI) all varied linearly with walking speed (v) (see Figures 3a through 3c). These relations were estimated by calculating linear regression lines.

Comparison of the outcome parameters obtained from both orthosis configurations was carried out by means of analysis of covariance (ANCOVA) while effect modification due to the incorporation of knee flexion was excluded (no significant difference in the slopes of regression lines obtained from both configurations)[22].

A significant difference in the slopes of the regression lines of the two orthosis configurations was found to occur in the relation between speed and peak knee flexion angle (ANCOVA: F=32.57; p<0.05). For all other outcome parameters, slopes of the regression

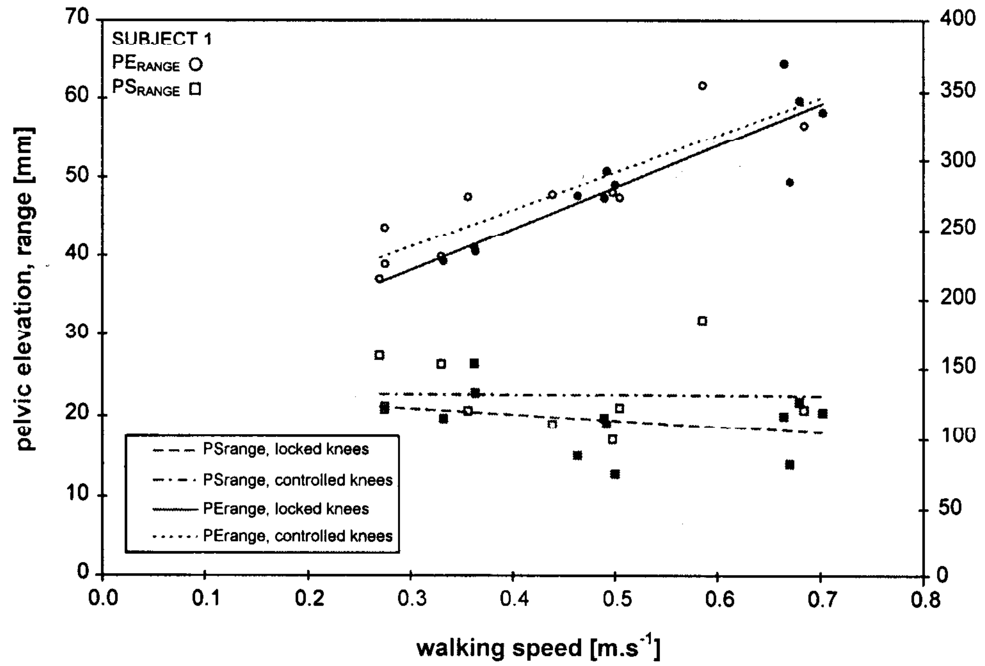

Figure 3a. Speed vs. pelvic elevation range (PE_{RANGE}, O) and pelvic lateral sway range (PS_{RANGE}, □) ; Closed symbols represent the orthosis configuration with locked knees, open symbols represent the configuration with controlled knees. Data points represent values calculated for individual strides. Crutch force data are averages for left and right side. Lines represent linear regression equations of the data. Analysis of covariance showed that differences in the regression lines of the two orthosis configurations were not significant except for CFTI.

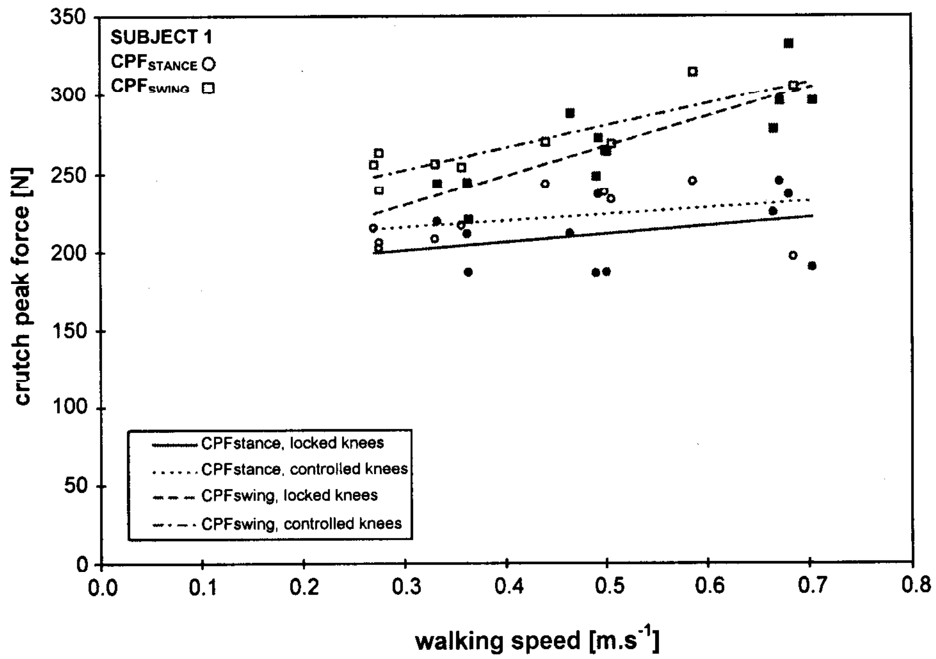

Figure 3b. Speed vs. crutch peak forces on stance side (CPF_{STANCE}, O) and on swing side (CPF_{SWING}, □)

lines calculated for the orthosis with locked knees and the orthosis with controlled knees were not significantly different. Further analysis revealed that the incorporation of knee flexion did not cause significant differences in the outcome parameters except for CFTI, which was higher in the configuration with controlled knee flexion (ANCOVA: $F=13.46$; $p<0.05$).

For subject 2, a significant favourable influence of incorporating knee flexion was found at self-selected speed for CPF_{SWING}, which was approximately 9% lower in the orthosis with controlled knees (Table 2). No significant differences were found for PE_{RANGE}, PS_{RANGE}, CPF_{STANCE} or CFTI.

Energy expenditure

Treadmill oxygen uptake measurements for subject 1 were carried out at speeds ranging from 0.19 to 0.59 m.s^{-1} for the configuration with locked knees, and at speeds ranging from 0.13 to 0.39 m.s^{-1} for the orthosis with automatic control of the knee lock. The maximum speeds found during the energy expenditure assessment were lower than the ones measured during the kinetics and kinematics assessments, which was probably due partly to the imposed continuous speed of the treadmill, and partly to the longer duration of the treadmill measurement, which induced a certain degree of fatigue.

Table 3. Comparison of energy requirements of gait obtained from subject 2 during the energy uptake measurements of both orthosis configurations. Presented data are averages and standard deviations (in parentheses) calculated from the last 5 minutes recording. $\dot{V}O_2$ was calculated with respect to lean body weight (LBW). Significance of the difference between the $\dot{V}O_2$ measurements was calculated using the t-test for paired samples and presented in the last column.

Subject 2: Energy requirements assessment		Orthosis configuration: *Locked knees*	Orthosis configuration: *Controlled knees*	Difference*
Walking speed	[m.s^{-1}]	0.15	0.15	-
$\dot{V}O_2$	[ml.min^{-1}.kg$_{LBW}$$^{-1}$]	17.37 (1.10)	15.88 (1.28)	1.49 (1.74) p<0.001

*T-test for paired samples
- = not tested

Figure 4 shows the results of the treadmill measurements of subject 1. Steady state $\dot{V}O_2$ varied directly with walking speed. Linear regression lines were fitted to the data in order to analyse the influence of knee flexion on the speed vs. energy consumption relation.

Analysis of covariance could be performed as the slopes of the two regression lines were not significantly different (ANCOVA: $F=4.83$) and revealed that $\dot{V}O_2$ was significantly higher in the configuration with controlled knees ($F=18.26$; $p<0.05$).

Subject 2 was well capable of controlling her walking speed over the 2 measurement trials, as can be seen from Table 3. Walking speed during the 10 minutes of the assessments was approximately 25% lower than during the kinematic assessments. Probably, self-selected speed during longer periods of gait is better described by the term comfortable walking speed, and is related to the cardio-vascular condition of the subject. This is not the case for the speed selected by the subject during the (short) kinematic assessments.

Steady state oxygen uptake was reached in both assessments within 3 to 4 minutes after walking had started.

A significant difference in energy consumption was found between the configurations. $\dot{V}O_2$ was approximately 9% lower in the orthosis with controlled knee flexion.

DISCUSSION

Knee flexion is probably the most important mechanism for reducing the vertical displacement of the centre of mass of the body and smoothening its pathway in normal gait, which both have been associated with reduction of energy expenditure[8,9] seems worthwhile therefore to try and incorporate knee flexion into the orthotically aided gait of people suffering from paraplegia. This presumption is supported by the results of recent work of Yang and colleagues, who studied the effects of various joint motion constraints on the gait of three normal subjects walking in a hip-knee-ankle-foot orthosis. Their results indicate that incorporating knee flexion into orthotic gait improves gait performance and contributes to the reduction of energy expenditure[19].

The orthosis used in the study of Yang allowed knee flexion both in the stance phase and in the swing phase of gait, and thus allowed near-normal gait in the subjects studied. The orthosis developed for the present study was intended for use by paraplegic subjects, which demands high safety and high stability during the stance phase. For this reason, the control mechanism for the knee lock in our orthosis was designed for operation during the swing phase only. As a consequence, the major effect of knee flexion on the vertical excursion of the centre of mass of the body, which is observed in normal gait during the stance phase, could essentially not be studied with our orthosis. One could not, therefore, expect the same effect on energy efficiency as has been described for knee flexion as a whole, namely an average reduction of oxygen cost of 23% in ten healthy subjects[23].

A major problem which accompanies the orthosis' knee hinge control mechanism is the absence of knee flexion at initial swing phase. In normal gait, the greater part of knee flexion (30 to 40 degrees) is obtained prior to the swing phase, which results in a biomechanically favourable starting point for clearing the swinging foot. In our orthosis, the knee hinge can only be unlocked on the condition that the ipsilateral hip is in extension and an extending moment occurs around the knee. Consequently, the swing phase is started typically with the knee in full extension. Substantial momentum must therefore be generated at the hip in order to propel the upper leg against its high inertia. It was found in our study that this can only be achieved at high walking speeds. The peak knee flexion angle during the swing phase was very small (16°) for subject 2, who walked at a speed of approximately 0.2 $m.s^{-1}$, which is a commonly observed walking speed for mid-thoracic paraplegics[14]. Subject 1 was able to achieve knee flexion angles of more than 40°, but only at speeds over 0.6 $m.s^{-1}$, which walking speed can only be attained by a small group of well-trained, low level paraplegics.

The problem of insufficient knee flexion, which is due to the fact that knee flexion is obtained by passive swing only, should typically be approached by incorporating electrical stimulation of the leg muscles. Preliminary studies in our laboratory on the enhancement of our test orthosis by means of a previously reported FES control strategy for the swing phase of gait, which utilizes the hamstrings muscles for improving knee flexion at initial swing[24], have yielded encouraging results.

It can easily be analysed that, depending on the anthropometry of the individual subject and thus the size of orthosis components, knee flexion less than approximately 40° with neutrally positioned ankles results in an increase of the distance between hip and toes (henceforth referred to as effective leg lengthening) and is therefore counterproductive with respect to clearing the swing leg off the ground. As a result, compensatory mechanisms must be addressed in order to be able to make a step, which could result in increased effort and energy requirements.

Since dorsiflexion of the ankle in our orthosis was impossible, two 'interdependent' options for compensation of effective leg lengthening at low walking speeds were available, namely increasing pelvic elevation and increasing pelvic lateral sway. From the results of both subjects, no significant difference could however be established in either of these parameters, indicating that on the average, no compensation was necessary. On the other hand, we did establish a difference in required effort for walking in subject 1, reflected in increased crutch force time integral in the orthosis with controlled knees. Since CFTI is composed of crutch force and time, this increase can be caused by an increase in either of the two, or both. Since no increase was found in the peak crutch forces, either extra crutch force was applied in the orthosis with controlled knee flexion for stabilisation, or cadence was lower at the same walking speed.

The results from subject 2 reflect a very small range of walking speeds, since measurements were performed at self-selected speed only. For this reason, interpretation of these data is more difficult. Since walking speed was not significantly different between orthosis configurations, however, the significant decrease found in CPF_{SWING} can be attributed to the incorporation of knee flexion.

The treadmill measurements of subject 1 provide insight into the energy requirements of the two orthosis configurations at a range of walking speeds. They allow comparison of energy uptake vs. speed relations rather than comparison of two - more or less arbitrary - observations, and give insight into possible speed dependence of results. The results of the analysis of the measurements taken from subject 1 show that energy consumption in the orthosis with controlled knee flexion was higher than in the orthosis with locked knees. This result indicates a real difference in the slopes of the $\dot{V}O_2$ versus walking speed relations of the two orthosis configurations, in other words effect modification by knee flexion, which could however not be demonstrated as such in the limited setting of the current case study.

The energy measurements taken from subject 2 merely provide single observations from the energy uptake vs. speed relations. This is obviously less informative than an overview of the speed versus energy requirements relation, but is certainly as relevant since measurements were performed at self-selected speed. A significant difference in the order of 10% in energy consumption between orthosis configurations was found, which must be attributed to the incorporation of knee flexion in the swing phase, since walking speed was kept constant over both

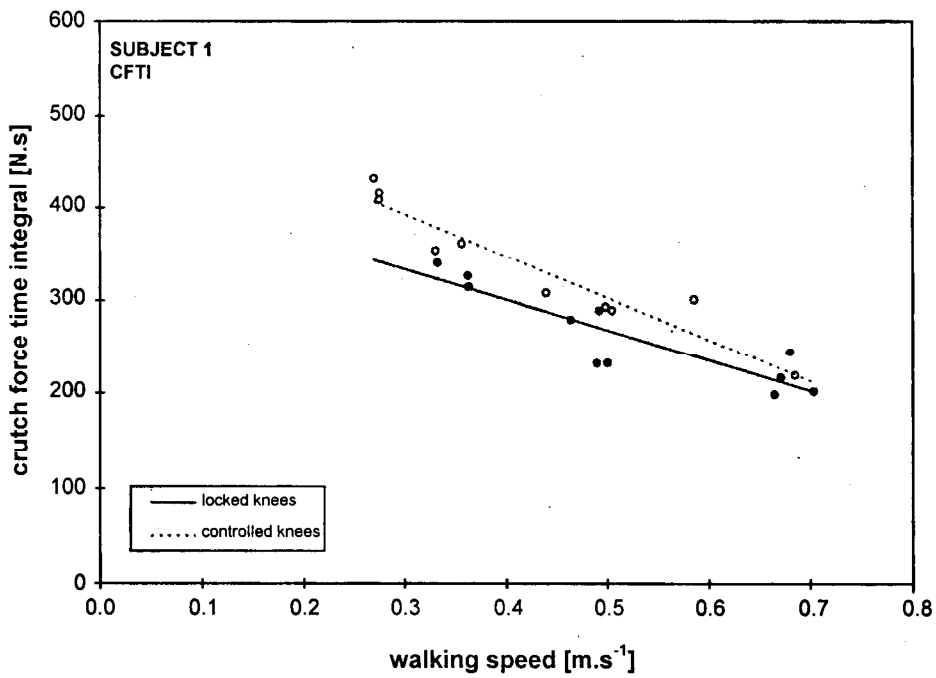

Figure 3c. Speed vs. crutch force time integral (CFTI), obtained from subject 1

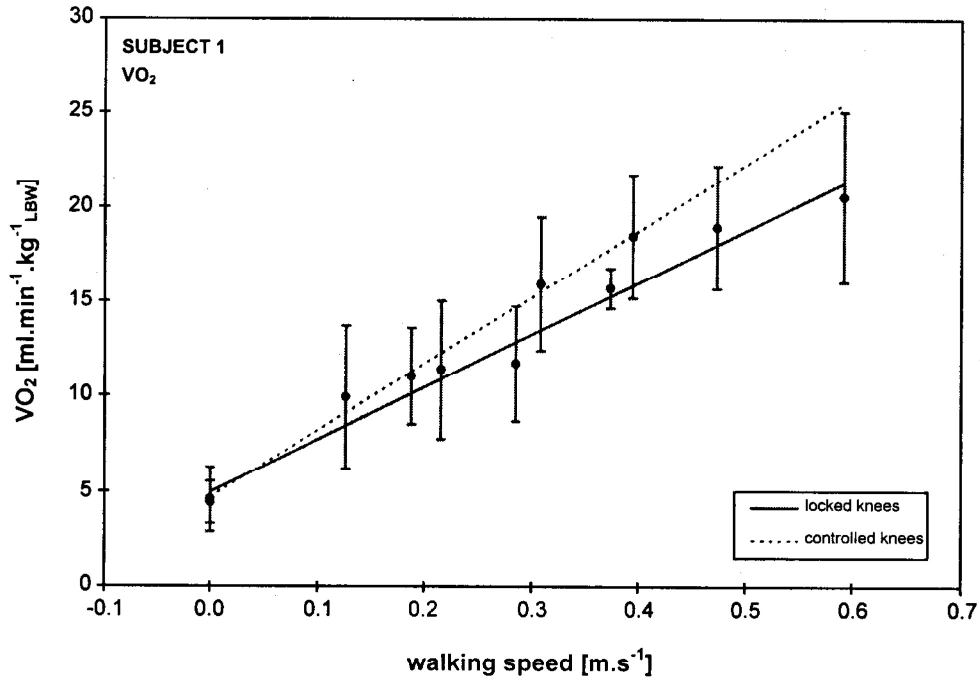

Figure 4. Speed vs. energy comsumption ($\dot{V}O_2$) obtained at a range of speeds in the treadmill measurements from subject 1. Closed symbols represent the orthosis confgiguration with locked knees, open symbols represent the configuration with controlled knees. Data points represent averages calculated for each speed over the last minute of each 5 minutes recording; error bars indicate standard deviations. Linear regression lines were fitted to the $\dot{V}O_2$ data. Analysis of covariance showed that the lines were significantly different.

measurements. This finding seems to be in agreement with the results of a study into the LUK-RGO, namely a reduction in PCI of 12% in comparison with the conventional RGO[21]. The latter study provides no data on walking speed, however, so it is not clear whether or not this result should perhaps be (partly) contributed to confounding.

Though energy consumption during gait is certainly a very important criterion in the design of orthoses, and has been widely advocated as such[7,11,25], the practical consequences of design choices for everyday use may play an equally important role. In this context, knee flexion can have important merits. On the basis of the experience gained during the training period, it was felt by subject 1 that knee flexion could be very useful for negotiating uneven terrain and slanting pavements. In contrast to the adverse effect effective leg lengthening can have during the swing phase, the possibility of flexing the knee for overcoming natural or architectural barriers was found very practical, which is an important finding with respect to possible outdoor use.

Another important aspect related to knee flexion is cosmesis of gait. The incorporation of knee flexion in the gait pattern visually reduces the gap between orthotic and normal walking, which may be valuable with respect to system acceptance.

A problem encountered with knee flexion during this study was the fact that neither of the subjects felt sufficiently safe for unassisted ambulation. This lack of safety forms an impediment to the functional application of the orthosis in daily life, and thus requires additional attention.

Changing the design of the knee lock control mechanism could take away part of the safety problem. In the LUK-RGO for instance, the knee hinge is unlocked only if the leg is unloaded, which is detected by a slight lengthening of the upper leg steel[21]. Although this principle avoids one problem found in the orthosis developed for our study, namely unintended knee flexion during the stance phase, it provides no solution either for the problems arising occasionally from dragging the foot at mid swing or insufficient knee extension at terminal swing.

Moreover, it introduces the danger of bilateral knee flexion during push-ups, e.g. at short turns. Combining the design principles presented in the LUK-RGO and the ones present in our test orthosis could yield a knee flexion control mechanism with higher overall safety than present in each of them separately.

One can conclude from this study that the incorporation of knee flexion in the swing phase may provide a modest added value to paraplegic orthotic gait. It is uncertain whether upper body effort and energy requirements of gait can be improved in a sufficiently large population. Better prospects can be attributed to the practical aspects of knee flexion during gait on uneven terrain, and convincing cosmetic improvement can be obtained. Furthermore, incorporation of a knee flexion control mechanism in the HKAFO prepares the orthosis for full benefit of augmentation of the swing phase of gait by means of electrical stimulation.

ACKNOWLEDGEMENT

The authors wish to thank all volunteers participating in this study, and J.W. van Dijk, M.Sc. and E.G.F. Stijnen, M.Sc. for their valuable contributions during preparatory work. The help of the Adaptation Technology department of Rehabilitation Centre Het Roessingh during the construction of the orthosis is gratefully acknowledged.

This study was supported by the Dutch Technology Foundation (STW).

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