

The Influence of Frontal Alignment in the Advanced Reciprocating Gait Orthosis on Energy Cost and Crutch Force Requirements During Paraplegic Gait

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Abstract

Reduction of energy cost and upper body load during paraplegic walking is considered to be an important criterion in future developments of walking systems. A high energy cost limits the maximum walking distance in the current devices, whereas wrist and shoulder pathology can deteriorate because of the high upper body load.

A change in alignment of the mechanical brace in the frontal plane, i.e. abduction, can contribute to a more efficient gait pattern with sufficient foot clearance with less pelvic lateral sway. A decrease in pelvic lateral sway after aligning in abduction results in a shift of the centre of mass to the swing leg crutch which may result in a decrease in required crutch force on stance side to maintain foot clearance.

Five paraplegic subjects were provided with a standard Advanced Reciprocating Gait Orthosis (ARGO) and an ARGO aligned in 4 different degrees of abduction (0°, 3°, 6° and 9°). After determining an optimal abduction angle for each of the subjects, a cross over design was used to compare the ARGO with the individually optimised abducted orthosis. An abduction angle between 0° and 3° was chosen as optimal abduction angle. Subjects were not able to walk satisfactory with abduction angles 6° and 9°.

A significant reduction in crutch peak force on stance side was found (approx. 12% , $p < 0.01$) in the abducted orthosis. Reduction in crutch force time integral (15%) as well as crutch peak force on swing side (5%) was not significant. No differences in oxygen uptake as well as oxygen cost was found. We concluded that an abduction angle between 0° and 3° is beneficial with respect to upper body load, whereas energy requirements did not change.

Key words: paraplegia, crutch forces, energy cost, orthotics, frontal alignment.

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Although intended to be functional, orthosis supported walking in paraplegics has primarily been advocated because of its psychologic and therapeutic value [11, 23]. Various devices have been developed during the last decades in order to provide ambulation to paraplegic patients. Mechanical braces are most commonly used and include those with a reciprocating cable linkage, i.e. Advanced Reciprocating Gait Orthosis (ARGO) and Reciprocating Gait Orthosis (LSU-RGO), and those without this linkage, i.e. Hip Guidance Orthosis (HGO). Functional Neuromuscular Stimulation can be added to the mechanical brace to generate propulsion [10, 15, 16].

One of the problems of the available walking systems preventing a functional use in daily live, is the high exertion of the upper body during walking [1, 5]. Moreover,

the high repetitive upper body load may increase the prevalence of carpal tunnel syndrome as well as shoulder impingement syndrome in paraplegic individuals [2, 8]. Precaution must be taken when prescribing a walking system to a population at risk with respect to upper body pathology. Beside the addition of FNS to a mechanical brace, several aspects relating to properties and alignment of the mechanical device may contribute to a reduction in upper body load and energy requirements.

The reciprocating cable linkage in ARGO as well as RGO may prevent a free ballistic swing of the swing leg [22]. It has been suggested that a change of either the reciprocal coupling mechanism [25] or the transmission ratio may be beneficial [26]. Three aspects have been described which contribute to a facilitation of foot clear-

ance during early and mid swing phase., i.e. knee flexion, alignment in frontal plane and lateral stiffness of the orthosis. By allowing knee flexion during swing phase the patient can use the bending of the knee rather than pelvic lateral sway to obtain foot clearance [4]. Both, lateral stiffness as well as frontal alignment have been described in HGO because of an enhanced foot clearance with less pelvic lateral displacement [7, 14, 18, 20, 21]. Nene found that in some adult patients deformation at the hip hinge region on the stance side reached such proportions that the swing leg failed to clear [14, 15]. Pushing harder on the swing side crutch increases lateral forces and causes even more deformation at the stance side. Collapsing of the orthosis can be compensated by applying force on the stance as well as on the swing side crutch. In addition, deformation can be prevented by increasing lateral stiffness in the orthotic structure or by stimulating stance leg gluteal muscles [20].

An orthosis which is aligned in slight abduction can also contribute to a more efficient ratio between foot clearance and lateral sway of the body [18, 27]. In addition, due to a decrease in pelvic lateral sway, the centre of gravity during mid swing is shifted towards the swing leg crutch. This change in body posture during mid swing may lead to an increase in force on swing side crutch and a decrease in force on stance side crutch. It is expected that alignment in abduction yields a better utilisation of swing crutch force for propulsion rather than for pelvic sway and a reduction in stabilizing forces on stance leg crutch. Rose used a mathematical description of mechanical stress at the hip hinge in order to determine optimal abduction angle [18]. He found that 5° of abduction was associated with least mechanical stress in an orthosis with an average leg length of 1 metre and a standard pelvic width. However, an optimal abduction angle must be determined for each subject individually because of the differences in anthropometry and their preference with respect to walking with larger step widths.

This study was conducted in order to assess the differences between ARGO and ARGO aligned in slight abduction with respect to upper body load and energy requirements. Crutch forces, oxygen uptake and subjects' opinion were used as outcome measures. The study comprised an optimisation procedure in which an optimal

abduction angle was determined for each subject and a comparative trial in which the ARGO was compared with this individually optimised abducted orthosis.

Methods

Subjects

Subjects were selected from the rehabilitation centre if they had complete thoracic lesions and could walk independently in the ARGO with a regular walking pattern for 10 minutes or more. Subjects were included after they had given their written informed consent. Five paraplegic subjects were included according to this procedure (table 1). The study was approved by the local medical ethical committee.

System description

The standard ARGO, used as reference system, is aligned in 6° adduction at the level of the hip joints, whereas the ankle foot braces are mounted in slight abduction. The ARGO was aligned in 6° abduction by mirroring the hip joints. Applying different bent rods resulted in abduction angles of 0°, 3° and 9° (figure 1).

Study Design

Optimisation

The optimal abduction angle was determined for each subject. Five different systems were assessed at random on one day: ARGO and ARGO in 0°, 3°, 6° and 9° of abduction (Abducted Orthosis: AO0°, AO3°, AO6° and AO9° respectively). Before the assessment, patients were asked to practice in the orthosis configuration to be tested. All patients walked using normal crutches and a four point gait pattern. No differences in walking technique between different orthoses were allowed.

Comparative trial

In order to prevent training effects, a separate training period was included between optimisation and comparative trial. We continued with the comparative trial after the subjects became familiar with the orthosis configuration. A cross over trial (sequences: ARGO-AO_{opt} and AO_{opt}-ARGO) was used to compare the ARGO with the individually chosen abduction angle (AO_{opt}). Comparability of groups was obtained using level of lesion as matchings factor [17]. Two subjects were selected for an ARGO-AO

Table 1. Subjects included in the study. Weight is measured in kilogram (kg) and kilogram lean body weight (kgLBW). Measurement sequence 'A' corresponds with measurement of ARGO followed by the Abducted Orthosis (AO), sequence 'B' for AO followed by ARGO.

	1	2	3	4	5
age/sex	35/m	32/f	45/m	41/m	31/m
level of lesion	T12	T4, T5	T9	T8	T11
weight (kg and kgLBW)	66/52.2	60/41.8	90/71	66.5/53.9	100/75.2
measurement sequence	A	B	B	A	B

Frontal alignment in ARGO

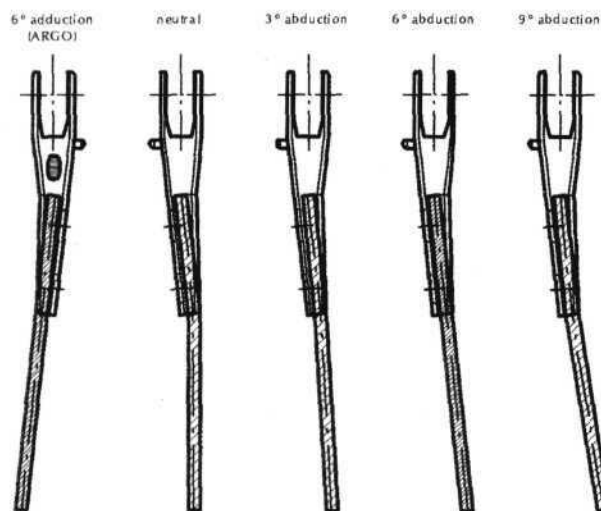


Figure 1. Obtaining different hip abduction angles in Advanced Reciprocating Gait Orthosis (ARGO). The left hip hinge is drawn in frontal view. ARGO is aligned in 6° adduction at the hip hinge. Mirroring the lower hip joint section yields a 6° abduction angle. Two bent rods were used to create 0°, 3° and 9° abduction.

sequence in the cross-over trial and three subjects for an AO-ARGO sequence (table 1). Because of the small sample size only one factor could be used for stratification. Level of lesion was chosen, since this was considered as the most biasing factor.

Measurements

Biomechanical assessments were performed during the optimisation phase in order to determine the optimal abduction angle. Biomechanical as well as physiological assessments were performed during the comparative trial. In addition, during the optimisation phase as well as the comparative trial, subjects were asked for their grading of either orthosis.

Biomechanical assessments

Kinetic and kinematic assessments were performed in the gait lab using a 5 camera 3D movement analysis system (VICON 370, Oxford Metrics, Oxford, UK). Each measurement consisted of 10 walks along a 5 metre gait lane in order to obtain 20-30 strides for averaging. Marker positions of ankles were sampled at a frequency of 50 Hz. Crutch forces and heel contacts were recorded simultaneously at 200 Hz using strain gauges and foot switches respectively. All data were filtered (linear phase 2nd order Butterworth, F_{-3dB} = 5 Hz) and splitted into gait cycle intervals using the heel strike data. Stride length (m) and cadence (strides.min⁻¹) were calculated using the ankle markers. Crutch Force Time Integral (CFTI) and crutch peak force on stance as well as on swing side (CPF_{stance} and CPF_{swing}) were calculated and normalised for body weight.

Physiological assessments

Subjects were asked to refrain from coffee, food and cigarettes for at least 2 hours prior to arrival to the unit. Breath-by-breath measurement of inspired and expired gases was conducted by using a metabolic cart (OXYCON-alpha, Jaeger, the Netherlands). Subjects were provided with a heart rate belt (Sport tester, PE3000, Polar Electro, Finland) and a facemask containing a flexible gas-tube. Measurement of rest metabolism was performed during 5 minutes, while the patient sat quietly. Subsequently, subjects were asked to stand up. When heart rate approached a stable level, subjects were instructed to walk at a comfortable, self-selected speed during 10 minutes along a 125 metre circular pathway. After walking subjects were asked to sit quietly for another 10 minutes in order to determine their recovery. Heart rate (beats.min⁻¹), V_{O2} (ml.min⁻¹.kg⁻¹), V_{cO2} (ml.min⁻¹.kg⁻¹), Respiratory Exchange Ratio and expiratory volume (l.min⁻¹) were measured. Mean heart rate (HR), Vo₂, Vc_{o2}, respiratory exchange ratio (RER_{steady state}) and expiratory volume (Ve) were calculated during the last 5 minute interval of walking assuming a delayed steady state. Vo₂ and Vc_{o2} were corrected for percentage fat and expressed in lean body weight (kg LBW). Oxygen cost during steady state (EO₂) was calculated by:

$$EO_2 \text{ (ml.m}^{-1}\text{.kg}^{-1}\text{)} = Vo_2 / v_{\text{steady state}}$$

Where V_{steady state} is walking speed during steady state.

Peak value of Respiratory Exchange Ratio (RER_{peak}) was recorded during the recovery period. Oxygen debt (O₂-debt) was determined by calculating the difference between Vo₂ prior to walking and Vo₂ during recovery. The O₂-debt is expressed as the amount of oxygen above rest Vo₂ during the first 10 minutes of recovery.

Subjects' preference

Subjects were asked to give their grading for either orthosis. We did not ask subjects to compare the devices, since these comparisons may be subject to information bias [19, 24]. Subjects' grading was scored on a 10 point scale, ranging from dislike to excellent orthosis.

Data analysis

Optimisation

Distribution of each of the variables (CFTI, CPF_{swing}, CPF_{stance}, walking speed and grading) was examined and natural log transformation was applied to transform skewed variables to "normality".

All data for each system and each subject were pooled and statistically tested by means of two way analysis of variance (ANOVA) and two way analysis of covariance (ANCOVA). Factors were system (ARGO and either abducted orthosis) and subject. Walking speed was included in the analysis of crutch force data as covariate (ANCOVA) in order to either correct for confounding or obtain a more precise estimation [12]. Post-hoc testing of significant differences was performed with paired t-tests. A Bonferroni correction was applied to adjust the level of significance during post-hoc testing. A Bonferroni correc-

tion can be obtained by dividing the significance level by the number of tests to be performed.

Determination of the optimal abduction angle

An optimal abduction angle was selected for each subject using subjects' grading and crutch force outcome as criterion.

Comparative trial

Only parameters which could support clinical decision making (CFTI, CPF_{stance}, CPF_{swing}, walking speed, Vo₂ and E_{O2}) were subject to statistical testing. Distribution of each of the variables was examined and natural log transformations were applied to transform skewed variables to "normality". Period effects were not statistically tested because of the small sample size. Moreover, it was assumed that period effects were reduced since all data were pooled. Paired t-tests were used to calculate 95% confidence intervals for the differences between ARGO and AO_{opt}. All 95% confidence intervals are presented as relative differences with respect to the ARGO assessments.

Two way ANCOVA was used to adjust for walking speed if walking speed was different between ARGO and AO_{opt}. All analyses were done using SPSS. A p-level of 0.05 was considered significant.

Results

Optimization

An abduction angle of 9° was excluded from the analyses because of the large step widths in this system. Subjects were not able to walk with the system. Figure 2 shows a typical crutch force pattern of the left crutch of one of the subjects. The left crutch is stance leg crutch during the first and third peak, after left heel strike. Swing leg crutch peak force (second peak) is higher than stance leg crutch peak force [14]. Stance leg crutch peak force is reduced in the

AO_{0°} and AO_{3°} orthosis, whereas the swing leg crutch peak force did not change in these systems with respect to the ARGO. A reduction in peak force of swing leg as well as stance leg crutch was found for the AO_{6°}. The decrease in swing leg peak crutch force may be caused by differences in walking speed.

Reduction in walking speed in AO_{6°} was not significant (p < 0.16, ANOVA, table 2). A significant difference was found with respect to subjects' grading for either orthosis

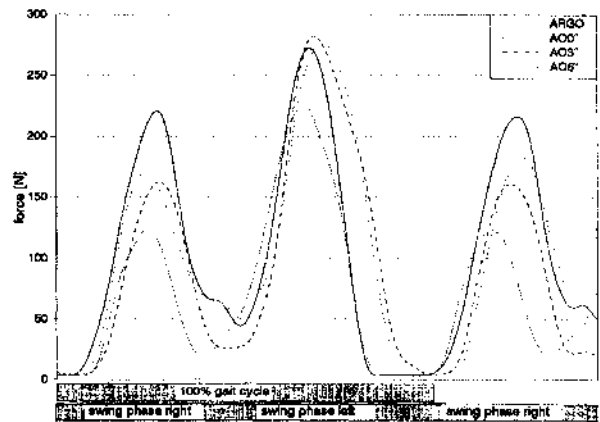


Figure 2. Typical example of left crutch force patterns of one subject walking in the ARGO and AO with different abduction angles. The left crutch is stance leg crutch during the first and third peak, after left heel strike. Swing leg crutch peak force is higher than stance leg crutch peak force. Stance leg crutch force is reduced in the AO_{0°} and AO_{3°} orthosis, whereas swing leg crutch force did not change in these systems with respect to the ARGO. The decrease in swing leg crutch force (second peak) may be caused by differences in walking speed.

Table 2. Results of ARGO and either abducted orthoses obtained during the optimisation phase. Mean (and standard deviation) of walking speed, crutch forces (Crutch Peak Force (CPF) and Crutch Force Time Intergral (CFTI)) and subjects' preferences. F- and p-values of two way ANOVA and ANCOVA are presented. Factors in the ANOVA were subject and system. Walking speed was added in the ANCOVA as covariate. The p-value of the covariate indicates whether the difference between adjusted and crude data is significant.

	CFTI	CPF _{stance}	CPF _{swing}	walking speed	grading
ARGO	0.59 (0.12)	0.39 (0.05)	0.43 (0.02)	0.29 (0.09)	7.60 (0.55)
AO _{0°}	0.57 (0.12)	0.36 (0.04)	0.40 (0.02)	0.28 (0.09)	7.60 (0.55)
AO _{3°}	0.56 (0.13)	0.36 (0.04)	0.41 (0.03)	0.28 (0.10)	8.40 (1.30)
AO _{6°}	0.59 (0.21)	0.33 (0.07)	0.40 (0.07)	0.26 (0.11)	6.00 (0.70)
ANOVA					
system	0.53 / p < 0.67	9.42 / p < 0.01	0.82 / p < 0.51	2.03 / p < 0.16	6.70 / p < 0.01
ANCOVA					
system	1.34 / p < 0.31	6.17 / p < 0.01	2.01 / p < 0.17		
covariate	p < 0.05	p < 0.82	p < 0.10		

($p < 0.01$, ANOVA, table 2). Although walking speed was not significantly reduced, differences in walking speed between orthoses configurations may hamper the interpretation of crutch force outcome. Analysis of covariance (ANCOVA) was used to adjust for walking speed. ANCOVA can only be used reliably if there is no effect modification or interaction between covariate and outcome measure [12]. Effect modification in epidemiology is the condition where the relationship of interest is different at different values of the extraneous variable. Visual inspection of scatter plots was used rather than a formal statistical test, since we had too few data to perform a test on interaction (figure 3). Differences in CFTI as well as CPF_{swing} were not significant ($p < 0.31$ and $p < 0.17$ respectively, ANCOVA). CPF_{stance} was significantly reduced in the abduction orthoses ($p < 0.01$, ANCOVA). The crude and adjusted differences between ARGO and either abduction orthosis were calculated (figure 4). Differences in CFTI as well as CPF_{swing} were larger after adjusting for walking speed. The difference between crude and adjusted data was significant for CFTI ($p < 0.05$). Average reduction in CFTI and CPF_{swing} in the AO_{3° was 9.5% and 7% respectively. CPF_{stance} is not affected by walking speed (covariate: $p < 0.82$). The average reduction in CPF_{stance} was approximately 8% for AO_{0° as well as AO_{3° .

Since neither subjects' grading nor CPF_{stance} were biased by walking speed, post-hoc testing was performed using paired t-tests. Subjects found no difference between ARGO and AO_{0° , preferred the AO_{3° and did not like the AO_{6° (table 3). CPF_{stance} was significantly reduced in the AO_{6° . CPF_{stance} was not significantly reduced in AO_{0° and AO_{3° .

Determination of the optimal abduction angle

A decrease in CPF_{stance} as well as CPF_{swing} of approximately 10% and 16% respectively was found for the AO_{6° . However, subjects preferred the AO_{3° because of some

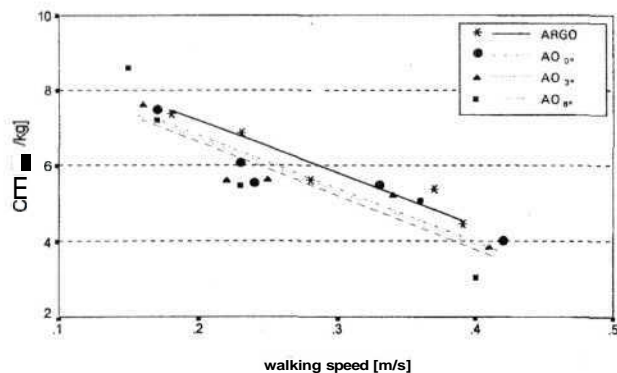


Figure 3. Scatter plot of CFTI against walking speed or different abducted orthoses. The slopes are not significantly different, indicating no effect modification. The differences in intercepts between the Abducted Orthoses and ARGO represent the walking speed adjusted mean difference.

interfering user aspects in the AO_{6° . Sitting in a wheelchair was not possible with such abduction angles. Consequently, independent donning and doffing was not possible. Therefore, subjects' grading was used as main criterion for determining the optimal abduction angle. Four subjects chose a 3° , and one subject chose a 0° abduction aligned orthosis.

Comparative trial

An estimation of the differences in physiological and crutch force measurements was made using 95% confidence intervals (figure 5). No difference in subjects' grading was found with respect to either ARGO or AO_{opt} . The increase in walking speed in AO_{opt} was not significant (95% CI: [-25%, 2%]). No significant difference in oxygen cost (E_{O_2}) and oxygen uptake (Vo_2) was found in the AO_{opt} . Table 4 summarizes the other physiological variables. Heart rate and Ve increased in the AO_{opt} . A trend of increased RER_{peak} , $RER_{steadystate}$ as well as O_2 -debt in the AO_{opt} can be observed.

The reduction in CPF_{stance} in AO_{opt} was significant (95% CI: [1% , 19%]). The reduction in CFTI as well as CPF_{swing} was not significant (95% CI: [-1% , 32%] and

Table 3. Post hoc testing of subjects' grading and CPF_{stance} by means of paired t-tests. Mean, (standard deviation) and p-value of differences are presented. The actual significance level is 0.017 due to Bonferoni correction.

	subjects' grading	CPF_{stance}
ARGO- AO_{0°	0.00 (1.00), $p = 1.0$	0.03 (0.03), $p < 0.10$
ARGO- AO_{3°	-0.80 (1.10), $p < 0.18$	0.03 (0.02), $p < 0.07$
ARGO- AO_{6°	1.60 (0.55), $p < 0.05$	0.06 (0.02), $p < 0.01$

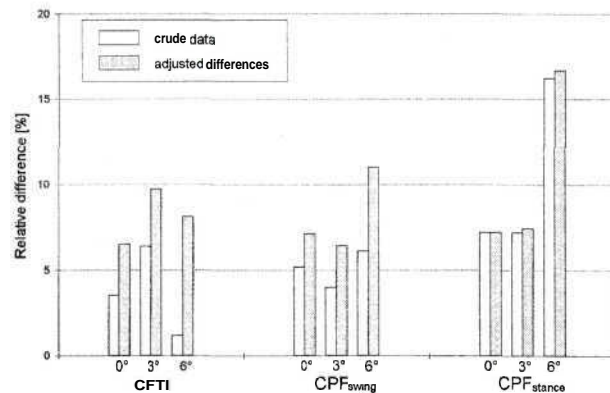


Figure 4. Crude and walking speed adjusted differences for crutch force variables using analysis of co-variance. Differences are presented as relative change with respect to reference system ARGO. Positive differences indicate higher values in ARGO.

Frontal alignment in ARGO

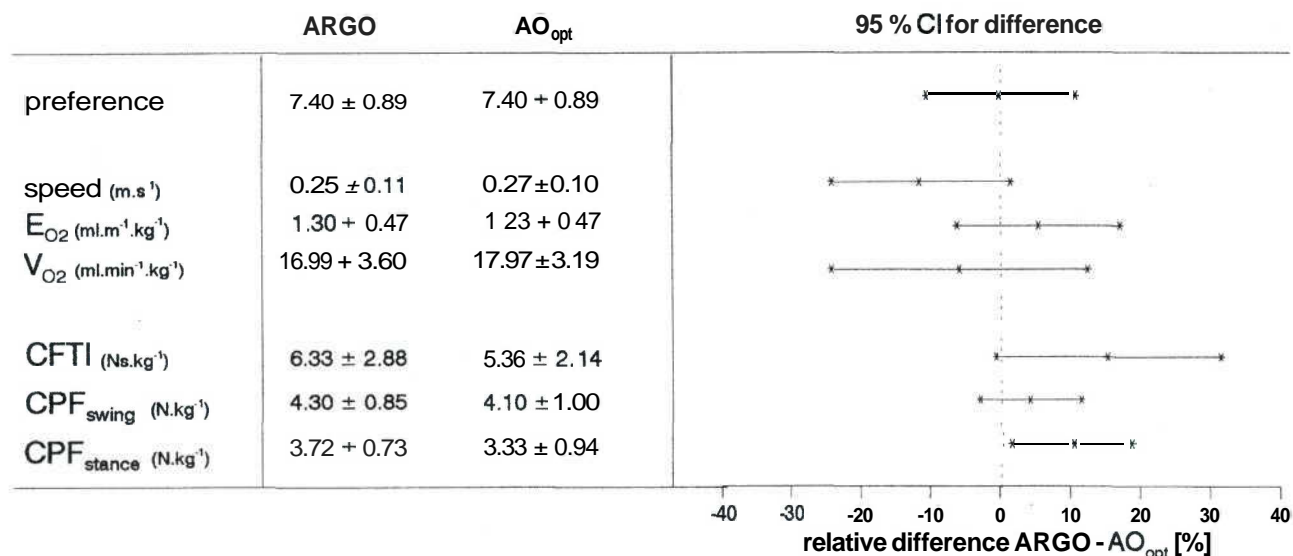


Figure 5. Estimation of differences between ARGO and AO_{opt} with respect to energy requirements, walking speed, crutch force requirements and subjects' grading. Estimations were made using 95% confidence intervals. The intervals are represented as relative change with respect to the reference ARGO. Positive differences indicate higher values in ARGO.

Table 4. Results of physiological assessments during the comparative trial. Heart rate (HR), expiratory volume (Ve), oxygen debt (O2-debt), respiratory exchange ratio during steady state walking (RER_{steadystate}) and peak value of RER during recovery (RER_{peak}). Presented data are mean and (standard deviation).

	HR b.min ⁻¹	Ve l.min ⁻¹	O2-debt ml.kgLBW ⁻¹	RER _{steadystate}	RER _{peak}
ARGO	136.4 (19.1)	36.6 (12.7)	17.7 (5.6)	0.99(0.10)	1.13 (0.10)
AO _{opt}	143.0 (16.4)	39.2 (12.5)	21.3 (6.7)	1.01 (0.05)	1.16 (0.07)

[-3%, 12%] respectively). Two way ANCOVA was used to adjust for walking speed in order to obtain more precise estimations of differences in crutch force and energy requirements. A summary of crude and adjusted differences

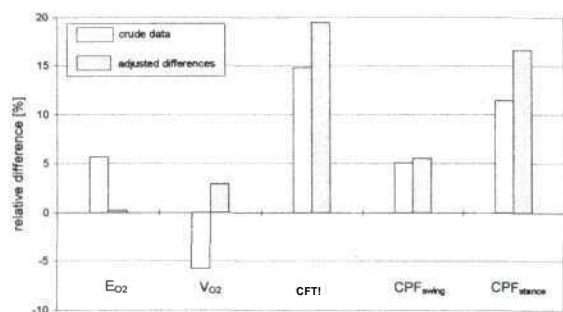


Figure 6. Crude and walking speed adjusted differences between ARGO and Abducted Orthosis. Relative differences are presented. Positive differences indicate higher values in ARGO.

is given in figure 6. The walking speed adjusted differences in CPF_{stance} as well as CFTI were slightly larger than the crude differences. However, adjusting for walking speed did not significantly change the outcome for these variables.

Discussion

Several authors have outlined that reduction of upper body load and energy expenditure during walking are important design criteria for walking systems [1, 13]. Functional use of a walking system requires an energy efficient walking pattern and minimal upper body load. This may improve gait performance and prevent wrist and shoulder pathology as well. Most evaluations which have appeared in the literature are comparisons between clinical systems [23, 25] or studies on the effect of functional neuromuscular stimulation [10, 13, 16]. Few articles have been published on the influence of orthosis components on energy cost and upper body load. The benefits of orthosis alignment in abduction have been described theoretically by Rose [18], but he gave credit to Herzog and Sharrard

[9]. Although a slight abduction alignment has been used in the Hip Guidance Orthosis, the benefits of this modification on energy cost and crutch force requirements have not been investigated.

The optimisation procedure in our study was included in order to gain insight into the size of the optimal abduction angle. This is why we have chosen only four different abduction angles. A more precise estimation of the abduction angle was expected to be senseless, since subsequent differences of 1° would have been too small to detect. We concluded from the optimisation that an abduction angle between 0° and 3° was sufficient in order to gain the benefits. All five subjects had the same experience of more stability and ease of foot clearance.

Variations in walking speed between orthoses hamper a reliable comparison, since most physiological and biomechanical measures are speed dependent. Moreover, the influence of walking speed as a confounding factor in the relation between orthosis and oxygen cost was noticed by Marsolais et al. who presented scatterplots of walking speed against $\dot{E}O_2$ in his comparison of RGO and RGO-FES [13].

The walking speed dependent behaviour is clear with respect to CFTI. CFTI decreases at higher walking speeds, since the integration interval is decreasing. Adjusting for walking speed by means of regression analysis or analysis of covariance is a common procedure in order to correct for confounding bias. However, in case of CFTI one should be aware of the behaviour of CFTI itself. By integrating the force one introduces a time dependent behaviour. Adjusting for walking speed implies a correction for time and results in an estimation of the force component in the CFTI. CFTI therefore, may not give additional information with respect to crutch peak forces or crutch average forces per stride. In addition, the time dependent behaviour only introduces a very difficult interpretation in cases of different walking speeds.

In the optimisation phase walking speed was mainly reduced in the AO 6° , which can be explained by the inability of the subjects to walk in this system satisfactorily.

As a consequence, a proper comparison of ARGO and AO 6° on the basis of the crude crutch force outcome is difficult. In addition, it can also be expected that more gait training in the abducted orthoses can diminish the differences in walking speed. Adjusting for walking speed may then be doubtful. Walking speed in the AO 0° and AO 3° was only slightly changed with respect to the ARGO, so a comparison of these systems is more appropriate. We found that CPF_{stance} as well as CPF_{swing} were reduced in AO 0° as well as AO 3° .

Walking speed in the AO_{opt} increased in the comparative trial with 11% on average. In healthy subjects, a comfortable walking speed is closely related to the metabolically most efficient walking speed, i.e. the walking speed with the lowest oxygen consumption per unit distance [6]. In paraplegia, however, it is not clear whether comfortable walking speed can be determined by means of the relation with oxygen consumption. Another possible physiological

explanation of comfortable walking speed is the onset of blood lactate accumulation (OBLA) [5]. An increase in comfortable walking speed therefore indicates that the OBLA will be reached at higher walking speeds. This can be interpreted as a functional improvement, i.e. higher walking speed during steady state. In the present study, $\dot{V}O_2$ did not change in the AO_{opt}, indicating that the aerobic part of energy delivery was equal to the ARGO. The anaerobic part of energy delivery can be estimated roughly by means of RER and O₂-debt [3, 13]. RER_{peak}, RER_{steady state} as well as O₂-debt were higher in AO_{opt}. If the OBLA was reached at higher walking speeds we had expected that there was no difference between the estimators of anaerobic energy delivery. Comfortable walking speed in the AO_{opt} is possibly more related to personal preference rather than to a physiological mechanism.

A clinically probably more relevant outcome variable with respect to the perceived exertion of walking is crutch peak force. Beside the relation between upper body load and prevalence of wrist and shoulder pathology, a reduction in upper body load may also contribute to the experience of a less fatiguing walking pattern. A clinically relevant and significant reduction in CPF_{stance} (crude: 12%) is therefore the most important finding in the study. The reduction in CPF_{swing} (crude: 5%) is considered clinically relevant as well, since absolute values of CPF_{swing} are higher than those of CPF_{stance}. Alignment of the ARGO in abduction may therefore be indicated for those individuals whose walking distance is limited by wrist and shoulder pain rather than metabolic energy delivery.

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List of abbreviations

AO = Abducted Orthosis, ARGO = Advanced Reciprocating Gait Orthosis, $\dot{V}O_2$ = Oxygen uptake [$\text{ml}\cdot\text{min}^{-1}\cdot\text{kg}^{-1}$], $\dot{E}O_2$ = Oxygen cost [$\text{ml}\cdot\text{m}^{-1}\cdot\text{kg}^{-1}$], CFTI = Crutch Force Time Integral, CPF = Crutch Peak Force, ANOVA = Analysis of Variance, ANCOVA = Analysis of CoVariance.

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