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THE EFFECT OF A DISTALLY PLACED ROCKER BAR ON KINEMATIC AND KINETIC PARAMETERS IN HEALTHY SUBJECTS

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الهدف : تحديد أثر القضيب المهزاز «المتأرجح» الموضوع في نهاية الحذاء على خصائص كل من الحركة والحركة الكيناميتية وتكييف نمط المشي . التصميم : تم اختيار ثلاثة حالات هي (١) زوج حذاء طبيعي (٢) أحذية وضعت على نهاياتها قضبان مهزازة وهي ذات نعول صلبة (٣) مثل مافي (٢) مع تحديد لحركة الانثناء الظهري لدى الكاحل . أجرى تحليل مشية ثلاثي الأبعاد للحصول على الخصائص المكانية الزمانية والكيناميتية لدى مفصلي الركبة والكاحل قد طرحت قوى ردود الأفعال الأرضية للحصول على العزوم لدى مفصلي الركبة والكاحل . هذا وقد مشى عناصر التجربة ١٠ مرات على مسار للمشي بسرعة مشي مريحة . موضع التجربة : معهد بحثي تابع لمستشفى تأهيلي كبير . عناصر التجربة : تم اختيار عشرة أشخاص أصحاء لهذه التجربة . النتائج : هناك زيادة بكل من عزم تمديد أو بسط الركبة التام وفي الذروة خلال مرحة الترجح لدى المشي مع القضيب المهزاز في نهاية الحذاء . لم تحدث الزيادة المتوقعة كتكييف على نمط المشي في الانثناء الظهري لدى مفصل الكاحل في الحالة (٢) . منعت الإضافة المحدودة في الانثناء الظهري لدى الكاحل في الحالة (٣) الانثناء الظهري لدى مفصل الكاحل ولكنها لم تزد من عزم بسط الركبة مقارنة مع الحالة (٢) . الخلاصة : إن القضيب المهزاز الموضوع في نهاية الحذاء يزيد من ثبات واستقرار الركبة لدى الأشخاص الأصحاء ، وفيما يبدو فإن الرفع المبكر للقدم عند نهاية مرحلة الترجح يستخدم كتكييف لدى المشي بحذاء وضع القضيب المهزاز في نهايته .

Objective: To determine the effect of a distally placed rocker bar on kinetic and kinematic parameters as well as on the adaptation of the walking pattern. ***Design:*** Three conditions were tested on one day. Condition A: normal pair of shoes; condition B: shoes with a distally placed rocker bar and stiffened sole; condition C: same as condition B, but including a limitation of the dorsal flexion movement at the ankle. A three-dimensional gait analysis was performed to obtain spatio-temporal and kinematic parameters at the knee and ankle joints. Ground reaction forces were deduced to acquire moments at the knee and ankle joints. Subjects walked 10 times along a walkway at a comfortable walking speed. ***Setting:*** Research institute affiliated with a large rehabilitation hospital. ***Subjects:*** Ten healthy subjects were selected for this study. ***Results:*** There is an increase in both peak and the integral of the knee extension moment during stance phase when walking with a distally placed rocker bar. An increase in the dorsal flexion at the ankle joint in condition B, which was expected to be an adaptation of the walking pattern, did not occur. The addition of a limited dorsal flexion at the ankle in condition C prevented the dorsal flexion at the ankle joint but did not increase the knee extension moment in comparison to condition B. ***Conclusion:*** A distally placed rocker bar increases the knee stability in healthy subjects. Early lifting of the foot at the end of the stance phase seems to be used as adaptation when walking with a distally rocker bar.

Key Words: *Rocker Bar, Kinetic and Kinematic Parameter, Walking Pattern*

INTRODUCTION

Provision of shoes is often used to improve walking in subjects with various deficiencies of the musculoskeletal system. Probably the most commonly prescribed shoe provision to influence the roll off, is the rocker bar[1]. Normally the roll off point is situated at the metatarsophalangeal (MTP) joints. By stiffening the sole and using a rocker bar the roll off point may be changed, either proximally or distally with respect to the MTP joints. A rocker bar proximally to the MTP joints (prox-rocker bar) causes an earlier heel rise whereas a rocker bar distally to the MTP joints (dist-rocker bar) causes a delayed heel rise.

The actual choice on shape and type of a rocker bar

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depends on the desired effect of the rocker bar and the patient's specific (foot) problems[2]. Broadly, the following functions of a rocker bar are distinguished:

1. unrolling the foot from heel-strike to toe off without bending the MTP joints (all rocker bars with stiffened sole)
2. providing an easier roll-off movement (prox-rocker bar)
3. diminish the pressure at the level of the MTP joints (prox-rocker bar)
4. more knee stability during walking (dist-rocker bar)

The existence of the effects for the first and second function is generally accepted. The third function has been subject of several studies[3-9]. All studies show a decreasing pressure at the level of the MTP-joints when using a prox-rocker bar.

In contrast with this, no studies were found that demonstrate the functional effects of a dist-rocker bar. There is only few and not congruent information about the effect of a rocker bar on walking speed, knee- and ankle joint motion, ground reaction force (GRF) and knee- and ankle moments. For multiple sclerosis patients it was shown that a prox-rocker bar causes an increased walking speed and single stance time compared to normal shoes[10]. The study of Peterson et al.[11] among healthy women showed no significant differences in walking speed, single limb support time and terminal peak vertical GRF between walking with athletic shoes and shoes with a rigid wooden sole applied with a prox-rocker bar. The double limb support time, stride length and ankle plantar flexion at the end of stance phase decreased while walking with a prox-rocker bar.

No studies were found, describing the effects of a dist-rocker bar on motion at the ankle and knee joint, GRF and knee and ankle moments. Conceptually, a dist-rocker bar is used to compensate for knee instability during gait, by increasing the extension moment on the knee joint during the second part of the stance phase[2]. A knee extension moment occurs when the GRF vector is in front of the centre of the knee joint (Figure 1). When the mass of the lower leg is neglected, the actual knee moment equals the product of the lever arm (=distance between knee joint centre and line of action of the GRF vector) and the magnitude of the GRF vector[12]. During the period of gait when the complete foot is in a flat position, a dist-rocker bar is expected to result in a more distal point of application of the GRF. This could result in a larger lever arm and therefore a larger extension moment at the knee joint.

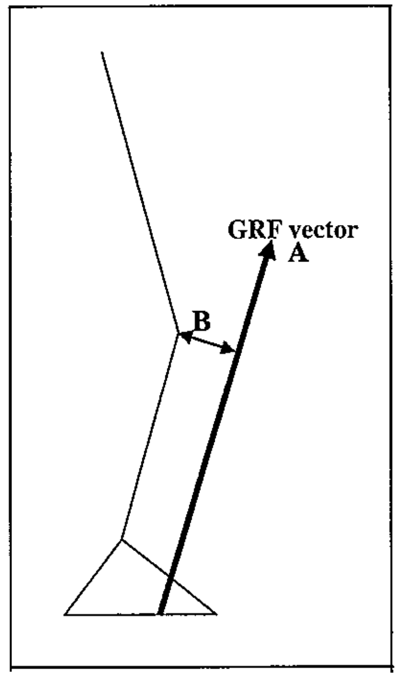

Figure 1. Illustration of knee extension moment. A: magnitude of the ground reaction force (GRF) vector; B: distance between joint centre and line of action of the GRF vector

Subjects might adapt their walking pattern when their shoes are provided with a dist-rocker bar. One possibility is a decrease of walking speed[13,14]. The knee extension moment decreases when walking speed decreases. Stance phase duration, double stance phase duration and GRF in early stance phase decrease with higher walking speed[15]. A second possibility is an increase of dorsal flexion at the ankle joint during midstance. This results in a decrease of the lever arm and therefore a decrease of the knee extension moment.

Nowadays, the exact prescription of a dist-rocker bar is largely based on clinical experience and good co-operation between physician and orthopaedic shoe technician[1]. Evidence is lacking concerning the assumed working mechanism of a dist-rocker bar. In addition, no protocols exist when prescribing a dist-rocker bar.

The purpose of this study is to determine the effects of a dist-rocker bar on kinetic and kinematic parameters in healthy subjects, like walking speed, ankle motion, knee and ankle moments and GRF. Also adaptations of the walking pattern are studied.

METHODS

Subjects

Ten healthy subjects were selected, two female and eight males between 24 and 39 years old. All subjects had no current or previous history of any neurological or orthopaedic disorder affecting their lower extremities or walking capacities. No subject had muscle weakness or limitation of range of motion. Only subjects with the same foot size were selected, so it was possible to use the same pair of shoes for all subjects.

Table 1. Subjects data

Subject	Sex	Age	Body length (cm)	Body weight including shoes (kg)	Sequence of measure-ments
1	female	33	184	73.5	ABC
2	male	23	185	68.0	CAB
3	male	29	193	79.0	BAC
4	male	24	188	85.0	BCA
5	male	34	179	84.0	CBA
6	male	29	187	85.0	BCA
7	male	34	190	83.5	BCA
8	female	28	183	70.0	BCA
9	male	32	179	82.0	ACB
10	male	39	179	81.0	ACB

A: normal shoes; B: shoes with a distally placed rocker bar and stiffened sole; C: shoes with a distally placed rocker bar and a limitation of dorsal flexion at the ankle

Before participating in this study, each subject signed an informed consent, according to the standards of the local medical ethical committee.

Table 1 shows the sex, age, body length, body weight and the sequence of measurements of the participating subjects.

Interventions

Three conditions were tested. In all three conditions the same type of shoes were used.

Condition A: confectional shoes with supple rubber sole

Condition B: a dist-rocker bar and stiffened sole are applied to confectional shoes

Condition C: same pair of shoes as in condition B, with a limitation of the dorsal flexion movement.

The shoes in condition B were made as follows: the original sole was removed and a carbon layer was attached to the innersole to stiffen the sole. A dist-rocker bar and a heel of cork were attached to the carbon layer. A layer of 4 mm of rubber was attached to the cork.

The dist-rocker bar begins at about 65% (lateral side) from the heel side of the shoe, anterior to the metatarsal heads. The anterior aspect of the shoe was 23 mm off the ground. The arc formed by the resulting roll-off had a 10-cm radius. The dist-rocker bar made an angle of 3 degrees from the medial to the lateral edge of the shoe in line with the metatarsal heads.

To limit the dorsal flexion at the ankle, a fitted rigid carbon layer was placed on the ventral side of the shank and the dorsal side of the foot. The plantar flexion angle was not limited by the carbon layer. As the carbon layer had to be used in all patients it was worn on the outside of the shoes using adhesive tape.

Gait analysis and instrumentation

All measurements were performed at the Roessingh Research and Development gait laboratory. To assess gait characteristics a VICON 3D-motion analysis system (Oxford Metrics Ltd, Botley, Oxford, UK) and two force plates (AMTI) were used. The VICON 3D system is a three-dimensional video/computer gait analysis system and was used for movement data acquisition. It consists of 5 standard ccd cameras, fitted with infrared filters, which record the position of the retroflective markers that were placed on standardised landmarks on the subject. The cameras were linked to a datastation (VICON 370). Datastation software (version 2.8) for three-dimensional data collection was used for capturing kinematic data. Forceplates, which were embedded in the centre of the walkway, were used to register the GRF. The video signals were acquired at 50 Hz, force-plate data at 200 Hz. The datastation was connected to a workstation. Workstation software (version 1.21) was used to determine the three-dimensional marker trajectories.

A biomechanical model developed by Koopman et al.[16] was then used to calculate the joint angles and joint moments during the gait cycle. Kinematic data like walking speed and stride characteristics were calculated.

Experimental protocol

For every subject three conditions, A, B and C, were tested in one session. All individual measurements were executed on the same day. The sequence of the conditions was determined randomly (Table 1). The whole session lasted about 2-3 hours. Subjects were asked to walk at their comfortable walking speed.

In order to get used to the different shoes each subject walked approximately 15 minutes with the shoes before starting the measurements. It was important that the subject struck the force plate with the complete foot. Therefore, a fixed starting point was determined. Subjects were instructed to look forward while walking.

Before the gait measurements started anthropometric data as body length, body weight, knee and ankle diameter were measured. The retroflective markers were fixed to the skin using double-sided adhesive tape.

Every session started with a static measurement, with the subject standing in erect position, in order to calibrate the markers. Ten walking trials were performed. Mean values of ten trials for walking speed, cadence, stride length, stride time, double stance phase time, stance phase time, dorsalflexion at ankle, flexion at knee, horizontal GRF (GRF-H) and vertical GRF (GRF-V) (peak and integral values), extension moment at the knee (peak and integral values) and dorsal flexion moment at ankle (peak and integral) were calculated. The term 'moments' is used to indicate external moments of force.

Data analysis

Peak GRF-H and GRF-V were normalised with body weight (BW), by expressing the GRF as a percentage of BW. For all parameters only the results of the right leg will be presented, because the results on both sides of the body are assumed to be dependent.

The statistical analysis was carried out with SPSS. All data were statistically tested by means of either two-way analysis of variance (ANOVA) or two-way analysis of covariance (ANACOVA). Walking speed was included in the ANACOVA as covariate in order to correct for confounding[17]. A covariate is a quantitative variable that is related to the dependent variable. Due to the limited dataset, correction for confounding was only possible in absence of effect modification. Effect modification is the condition where the relationship of interest is different at different values of the covariate. Visual inspection of scatter plots was used to determine whether confounding or effect modification occurred. In this study it has been assumed that when the slope of the regression lines for the three conditions differs little, that only confounding occurs. In these cases a correction for walking speed was executed.

Post-hoc testing of significant differences was performed with paired t-tests. This supplies 95% confidence intervals (CI) for the mean (paired) differences between the conditions. A p-level of 0.05 was considered significant. A Bonferoni correction was applied to adjust the level of significance during post-hoc testing. This correction was obtained by dividing the significance level by the number of tests performed.

RESULTS

The mean difference in walking speed between conditions A, B and C was not significant (Table 2). However, the mean maximal intra-individual difference (condition

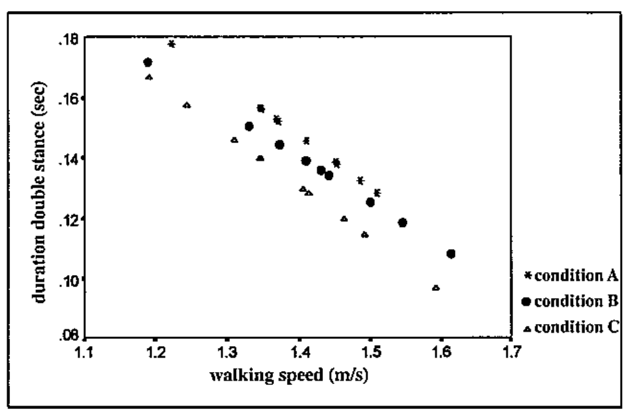

Figure 2. Relation between walking speed and double stance phase time. A: normal shoes, B: shoes with a distally placed rocker bar and stiffened sole, C: shoes with a distally placed rocker bar and limitation of dorsal flexion at the ankle. Slopes of lines are almost equal, meaning that only confounding occurs.

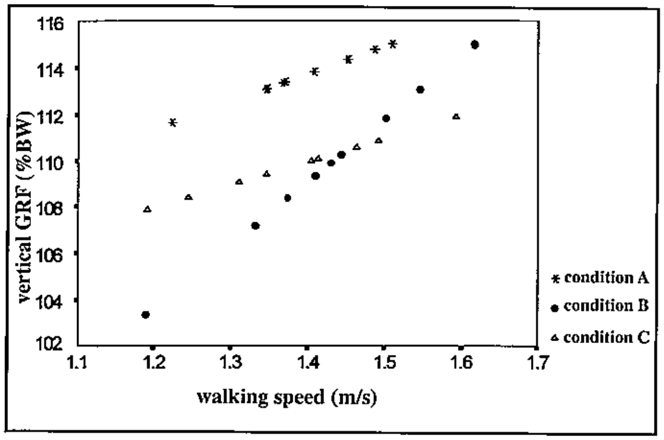

Figure 3. Relation between walking speed and maximal vertical ground reaction force. A: normal shoes, B: shoes with a distally placed rocker bar and stiffened sole, C: shoes with a distally placed rocker bar and limitation of dorsal flexion at the ankle. Slopes of lines have different directions, meaning that confounding and effect modification occurs.

A =100%) between the three conditions was 5.3% (sd= 3.7%), which may hamper a good inter-pretation of some parameters. Therefore, a correction for walking speed, by means of analysis of covariance (ANACOVA), has been executed for the following parameters: double stance phase time and stance phase time. For these parameters walking speed was a confounder (Figure 2). For the parameters: cadence, stride length, stride time, peak GRF-V and peak GRF-H, walking speed was not only a confounder but also an effect modifier (Figure 3). For these parameters no correction was executed.

Kinematic parameters

Table 2 shows the results of walking speed, stride length, stride time, duration of double stance phase and duration of complete stance phase. Duration of the double stance phase in condition B and C is shorter, compared to condition A. The 95% (CI) of the mean difference (in % of the value in condition A) between condition A and B is 2.7% - 12.8% and for condition A and C this is 4.1% - 15.5%. The stance phase duration is significantly shorter in condition B compared to condition A. The 95% CI of the mean difference (in % of the value in condition A) between condition A and B is 1.4% - 5.6%.

To verify whether the AFO prevents dorsal flexion,

Table 2. Kinematic parameters for the different conditions. Mean values and sd ar presented. The p-value according to the ANOVA is also presented.

	A Mean (sd)	B mean (sd)	C mean (sd)	p-value ANOVA	significant difference between conditions
walking speed (m/s)	1.39 (.08)	1.42 (.12)	1.38 (.12)	.07	no
cadence (strides/min)	51.71 (2.80)	52.86 (3.38)	51.83 (3.55)	.01	yes (A<B, B>C)
stride length (m)	1.62 (.10)	1.62 (.10)	1.60 (.10)	.19	no
stride time (sec)	1.16 (.07)	1.14 (.08)	1.16 (.08)	.01	yes (A>B, B<C)
double stance phase time* (sec)	.148 (.014)	.136 (.019)	.134 (.020)	.00	yes (A>B, A>C)
stance phase time* (sec)	.733 (.029)	.708 (.040)	.717 (.041)	.00	yes (A>B)

*: parameter is corrected for walking speed.
A: normal shoes, B: shoes with a distally placed rocker bar and stiffened sole, C: shoes with a distally placed rocker bar and limitation of dorsal flexion at the ankle

ankle motion was recorded (Figure 4). As shown in Figure 4, there is a clear difference in the amount of plantar flexion at heel contact (gait cycle at 0%) for the different conditions. Opila et al.[18] found that the position of the foot in several types of footwear when standing still can be different. A different amount of plantar flexion at heel contact makes it impossible to compare the absolute values of the dorsal flexion at the ankle joint in the three conditions.

The range of motion at the ankle joint (difference between maximal plantar flexion and maximal dorsal flexion during stance phase) can be used as a measure for the difference in dorsalflexion at the ankle joint. This can be justified by considering that no differences in stride length in the three conditions are apparent, indicating that the maximal plantar flexion at the ankle joint during the initial stance phase is the same for the three conditions.

The same applies for the knee flexion. The range of motion at the knee joint during stance phase (difference between maximal extension and maximal flexion) is taken as a measure for the amount of knee flexion.

Table 3. Range of motion at the ankle and knee during the stance phase and the plantar flexion at the end of stance phase (relative to the plantar flexion at heel-strike). The mean values and the sd for the different conditions are presented. The p-value of the executed ANOVA is presented.

	A mean (sd)	B mean (sd)	C mean (sd)	p-value ANOVA	significant difference between conditions
range of motion at ankle (deg.)	16.38 (4.39)	17.13 (6.49)	10.48 (3.00)	.00	yes
range of motion at knee (deg.)	17.10 (2.87)	16.97 (3.79)	16.21 (2.65)	.18	no
plantar flexion at the end of stance phase (deg.)	14.25 (6.13)	8.51 (8.06)	8.17 (4.74)	.013	yes (A>B,A>C)

A: normal shoes, B: shoes with a distally placed rocker bar and stiffened sole, C: shoes with a distally placed rocker bar and limitation of dorsal flexion at the ankle

Table 3 shows the range of motion at the ankle joint during stance phase, the range of motion at the knee joint during stance phase and the extent of plantar flexion at the ankle joint at the end of stance phase (relative to the plantar flexion at heel strike). In condition A and B the range of motion at the ankle joint is significantly greater compared to condition C. The 95% CI of the mean difference (in % of the value in condition A) between condition B and C is 11.3% - 69.9% and for condition A and C is 19.6% - 52.4%.

Kinetic parameters

The peak GRF-H and peak GRF-V at the end of the stance phase, as well as the peak and the integral of the knee extension moments and peak and the integral of the dorsal flexion moments at the ankle joint are shown in Table 4. For all parameters the values in condition A are significantly different from condition B. Comparing condition B and C, only the peak dorsal flexion moment shows a significant difference.

Figure 5 shows a typical example of the moment at the knee for one subject. The 95% CI of the mean difference (in % of the value in condition A) in peak extension moment at the knee joint between condition A and B is 12.3% - 66.1% and between condition A and C is 16.5% - 69.5%. For the integral of the knee extension moment the 95% CI of the mean difference (in % of the value in condition A) between condition A and condition B is 15.9% - 122.4% and between condition A and C is 14.9% -133.9%. The 95% CI for the mean difference (in % of the value in condition A) in the peak dorsal flexion moment between condition A and B is 5.9% - 24.2% and for the integral of the dorsal flexion moment is 7.8% - 22.0%.

Table 4. Maximal vertical and horizontal ground reaction force (GRF), peak and integral extension moment at the knee and peak and integral dorsal flexion moment at the ankle during the stance phase. All values are measured during the later stance-phase. The mean values and the sd for the different conditions are presented.

	A mean (sd)	B mean (sd)	C mean (sd)	p-value ANOVA	significant difference between conditions
peak horizontal GRF (% BW)	22.46 (3.31)	17.65 (4.59)	16.83 (3.19)	.00	yes (A>B, A>C)
peak vertical GRF (% BW)	113.69 (2.81)	109.76 (5.11)	109.72 (5.47)	.00	yes (A>B, A>C)
peak extension moment at knee (Nm)	14.3 (13.6)	19.9 (14.8)	20.5 (12.5)	.01	yes (A<B, A<C)
integral of extension moment at knee (Nms)	2.11 (1.93)	3.57 (2.41)	3.68 (2.09)	.01	yes (A<B, A<C)
peak dorsal flexion moment at ankle (Nm)	96.2 (14.6)	110.7 (16.1)	100.2 (12.2)	.01	yes (A<B, B>C)
integral of flexion moment at ankle (Nms)	21.1 (3.7)	25.9 (4.9)	23.9 (3.0)	.05	yes (A<B)

A: normal shoes, B: shoes with a distally placed rocker bar and stiffened sole, C: shoes with a distally placed rocker bar and limitation of dorsal flexion at the ankle.

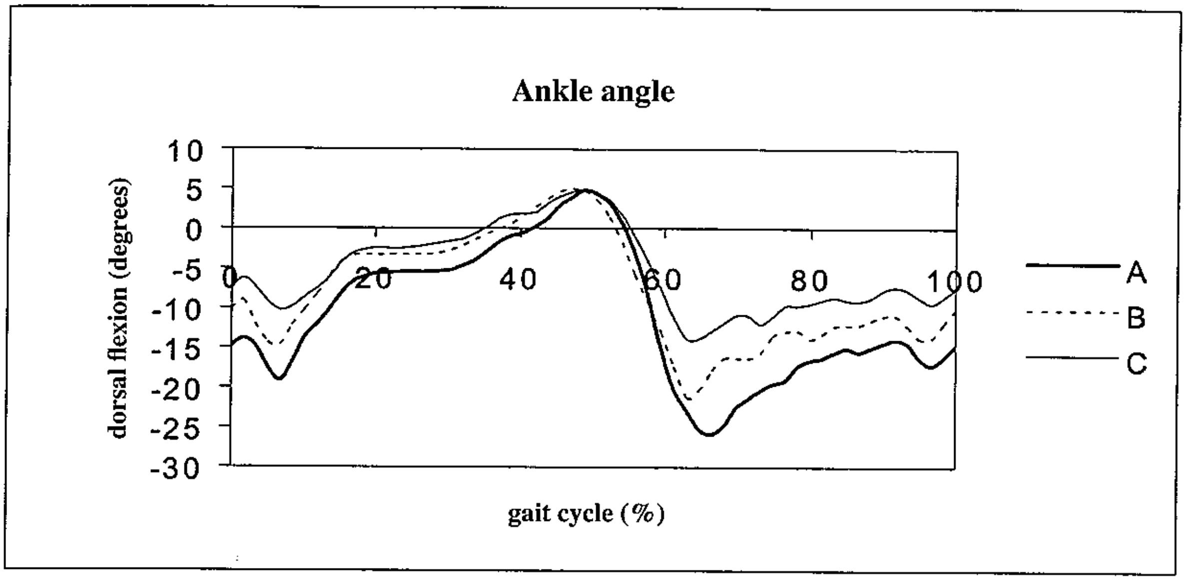

Figure 4. Typical example of ankle angle (degrees) in one participant during the gait cycle for the different conditions. A: normal shoes, B: shoes with a distally placed rocker bar and stiffened sole, C: shoes with a distally placed rocker bar and limitation of dorsal flexion at the ankle.

DISCUSSION

The main expected effect of a dist-rocker bar on gait is an increase in the knee extension moment, at the end of stance phase. The results of this study confirm the expectations regarding the peak knee extension moment and the integral of the knee extension moment. The integral is clinically most interesting because this is a measure of the total extension moment at the knee joint during stance phase, and thus a measure for the total stability at

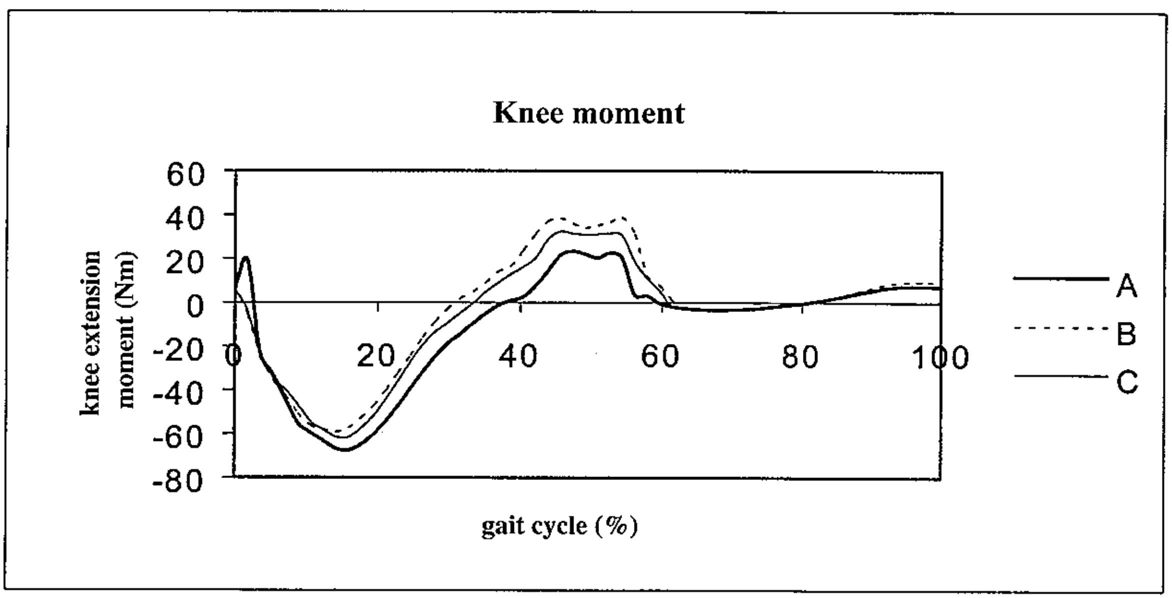

Figure 5. Typical example of knee moment in one participant during the giat cycle for the different conditions. A: normal shoes, B: shoes with a distally placed rocker bar and stiffened sole, C: shoes with a distally placed rocker bar and limitation of dorsal flexion at the ankle

the knee joint during stance phase.

Another expected effect of the dist-rocker bar is an increased dorsal flexion moment at the ankle joint during the second part of stance phase caused by a greater lever arm. The subjects in this study show an increased peak dorsal flexion moment and an increased integral of the dorsal flexion moment at the ankle joint when walking with a dist-rocker bar. When the peak dorsal flexion moment at the ankle joint increases, more activity of the calf muscles at the end of stance phase is needed for push off. This should be taken into account when prescribing shoes with a dist-rocker bar. In condition C the peak dorsal flexion moment decreases compared to condition B. This means that a decrease in peak dorsal flexion moment is caused by the ventral AFO. This might be the reason to use a combination of a dist-rocker bar and an AFO in patients with a decreased function of the calf muscles.

As, in case of a dist-rocker bar the bearing surface of the shoe is increased, it was expected that heel-rise starts later in gait cycle in condition B, compared to condition A. However, it was found that the double stance phase time and stance phase time is significantly shorter for condition B in comparison to condition A. Stance phase consists of 3 periods: 1.'heel strike' till 'foot flat'; 2. 'foot flat' till 'heel rise'; 3. 'heel rise' till 'toe off'.

A dist-rocker bar does not effect the duration of period 1 and is expected to increase the duration of period 2. So, a shorter duration of stance phase in condition B can only mean a shorter duration of period 3. This may mean that the subjects lift their feet before they have completely rolled off to prevent from an uncomfortable roll off. Peterson et al.[11] found that only a few subjects made a complete roll off when using a rocker bar. Most subjects unloaded their feet abruptly and did not complete the normal roll off. The same phenomenon seems to have happened in this study.

When subjects want to prevent an uncomfortable roll-off they will lessen the amount of the force used to push off. The extent of the GRF, especially the GRF-V, at the end of the stance phase is a measure for the push off at that moment[12]. In this study a significantly lower GRF-V and GRF-H, during the late stance phase, for condition B and C in comparison to condition A is seen. This might be an indication for early lifting of the feet.

Another indication of early lifting is that the extent of plantar-flexion at the end of stance phase, relative to the plantar-flexion at heel strike, is significantly greater in condition A compared to condition B and C. A diminished plantar-flexion at the end of stance phase means that the roll off at the end of stance phase is not complete.

So, the conclusion seems to be justified that walking with a dist-rocker bar shortens the duration of the period 'heel rise' till 'toe off' because the foot is lifted from the floor earlier and turning around the rocker bar is not completed.

In this study, it was assumed that during walking with a dist-rocker bar the amount of dorsal flexion at the ankle joint during midstance might be enlarged. However, no differences between condition A and B are demonstrated, indicating that adaptation of the walking pattern in the sense of enlargement of dorsal flexion at the ankle joint does not occur. So, a ventral AFO is not needed to decrease the dorsal flexion movement at the ankle joint during midstance.

The results show that the range of motion at the ankle joint when walking in condition C is smaller compared to condition A and B, as to be expected. However, the knee extension moment in condition C is not increased compared to condition B. Again, a reason of this might be the early lifting of the foot. When using a ventral AFO the calf muscles might be loaded less, and heel-rise will take place earlier.

In practice, shoes with a dist-rocker bar are prescribed to increase stability at the knee joint in cases of mild paresis of the knee extensors. This study confirms the supposed effect of a dist-rocker bar in a group of 10 healthy persons.

The expected increase in dorsal flexion at the ankle joint when walking in condition B, as an adaptation of the walking pattern, did not occur in this study. Early lifting of the foot at the end of the stance phase seems to be a more important adaptation of the walking pattern. However, early lifting does not eliminate the desired effects of a dist-rocker bar.

So, the dist-rocker bar is a useful supply to increase knee stability, at least in healthy subjects. This study also makes it clear that a limitation of the dorsal flexion movement at the ankel is not needed to create the effect of a dist-rocker bar. However, whether these conclusions are also valid for a group of patients should be subject of further research. For instance patients with CP or polio, in contrary to healthy subjects, have no possibility of selective musle control and therefore may need an AFO to limit the dorsal flexion movement.

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THE RELATIONSHIP BETWEEN THE SPEED OF WHEELCHAIR USE AND THE EFFICIENCY OF MUSCLE CONTRACTION

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يختلف الكرسي المتحرك عن الدراجات في افتقاره إلى تروس لتغيير السرعة التي يتم بها الدفع ، ورغم ذلك فمن المعروف في الدراجات أن كل من البدال المنخفض السرعة أو المرتفع السرعة تصاحبه كفاءة عمل متدنية . بذل جهد في السنوات الأخيرة لتصميم كراسي متحركة بها صناديق تروس في العجلات لتوفير خيارات لسرعة الدفع كما في الدراجات بينما لم يُفعل سوى القليل لتحديد السرعة المناسبة لدفع الكراسي المتحركة ، الشيء الذي كان له تبعاته من حيث التصميم وتدريب المعالج الطبيعي لمهارات الكراسي المتحركة . من هذا المنطلق يختبر البحث الحالي كل من استهلاك الأكسجين والتهوية »تغيير الهواء« وحاصل التنفس وإنتاج حمض اللبنيك والاستخدام العضلي خلال سرعات دفع تراوحت بين ٣٠ - ١٠٠ دورة في الدقيقة . أظهرت النتائج أن السرعة المناسبة هي ٥٠ دورة في الدقيقة وأن السرعات الأعلى يصاحبها عدم كفاءة في العمل وتحول لإنتاج حمض اللبنيك في العضلات أما العمل عند سرعة ٣٠ دورة في الدقيقة يصاحبه مزيد من الإجهاد العضلي بينما لا يظهر استخداماً أعلى للأكسجين حسب قياسات جهاز تخطيط العضلات.

Unlike bicycles, wheelchairs do not have gears to change the speed with which pushing is accomplished. And yet, for bicycles, it is well known that high or low cycling rates are associated with very low efficiency of work. In recent years there has been an effort to design wheelchairs with a gearbox in the wheel to allow this same selection of push speed as cyclists have. But very little has been done to determine the optimum speed to push a wheelchair. This has consequences forboth design and physical therapist training of wheelchair skills as well. Therefore, the present investigation examined oxygen consumption, ventilation, respiratory quotient, lactic acid production and muscle use during push speeds between 30 and 100 rpm. The results show that the optimum speed is at 50 rpm. Higher speeds were associated with inefficient work and a switch to lactic acid production in muscle. Work at 30 rpm, while not showing larger oxygen use than work at 50 rpm, was associated with more muscle fatigue than work at 50 rpm as assessed by the surface EMG.

Key Words: Exercise, Cycle Ergometry, Wheelchair, Efficiency, Paralysis

INTRODUCTION

Muscle is not an ideal energy transducer. In fact, much of the energy from the breakdown of glucose or fats is wasted as heat, and must be dealt with by the thermoregulatory system. While the remainder of the energy goes to external work[1,2]. The ratio of the external work accomplished divided by the energy consumed by the body in doing the work is called the mechanical efficiency of the work[1,3]. Various types of exercise are more efficient than other types of exercise. For example, swimming is often considered a very inefficient form of exercise and the efficiency has been estimated to be as low as 2%[2,4,5]. In contrast, exercise on a cycle ergometer or on a bicycle is considered to be one of the most efficient forms of exercise with efficiencies as high as 30%[3,6]. The maximum work that can be sustained in bicycle ergometry is typically as high as 600 watts[1].

Other studies report the efficiency of running to be as high as 45%[7]. But even for a given form of exercise, efficiency can vary with the speed that work is accomplished. Walking increases in efficiency up to about 3 mph at which time efficiency drops exponentially[2]. Above this transition point, running is more efficient than walking. Exercise on a cycle ergometer is best at 50-60 rpm. At speeds above and below this, efficiency drops[3,6]. For example, exercise at 50 rpm is 25% effi-

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