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The functional use of the reciprocal hip mechanism during gait for paraplegic patients walking in the Louisiana State University reciprocating gait orthosis

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Abstract

Reciprocally linked orthoses used for paraplegic walking have some form of linkage between the two hip joints. It has been assumed that flexion of the swinging leg is driven by extension of the stance leg. The aims of this study were to investigate the moments generated around the hip joint by the two cables in a Louisiana State University Reciprocating Gait Orthosis (LSU-RGO). Six (6) subjects were recruited from the Regional Spinal Injuries Centre at Southport, who were experienced RGO users. The cables were fitted with strain gauged transducers to measure cable tension. Foot switches were used to divide the gait into swing and stance phases. A minimum of 20 steps were analysed for each subject. Moments about the hip joint for each phase of gait were calculated.

There were no moments generated by the front cable in 4 of the subjects. In only 2 subjects did the cable generate a moment that could assist hip flexion during the swing phase. These moments were very low and at best could only have made a small contribution to limb flexion. The back cable generated moments that clearly prevented bilateral flexion. It was concluded that the front cable, as used by these experienced RGO users, did not aid flexion of the swinging limb.

Introduction

Walking in a reciprocal manner with the aid of an orthosis for people with a thoracic level spinal cord injury requires bracing from hip to ankle, and often includes trunk support. A range of devices is available, most of which consist of bilateral knee ankle foot orthoses (KAFO) and a trunk section. Knee and ankle joints are fixed and hip joints provide a limited range of motion in flexion and extension. Where options differ most markedly is in the method of limiting flexion and extension at the hip joint. These orthoses generally fall into two categories, reciprocally linked orthoses which have some form of reciprocal linkage between the hip joints, and free swing orthoses which allow free hip movement in flexion and extension between stops. Examples of reciprocally linked orthoses include the Advanced Reciprocating Gait Orthosis (ARGO) (Jefferson and Whittle, 1990) which has a single cable link, the Isocentric Reciprocating Gait Orthosis (IRGO) (Davidson, 1994) which has a rocker bar link, and the Louisiana State University Reciprocating Gait Orthosis (LSU-RGO) (Douglas *et al.*, 1983) which has a dual cable reciprocal link. The LSU-RGO was developed from previous designs by Motloch (Durr-Fillauer, 1983). The ParaWalker (Stallard *et al.*, 1986) which has trunk support, and the Walkabout (Middleton *et al.*, 1997) which has a pair of medially linked KAFOs , are examples of free swing orthoses.

Walking in all of these orthoses is achieved with the help of walking aids, either a rollator or crutches. The user inclines the trunk to one side by pushing with their hands on the contralateral side of their walking aid to produce vertical clearance for the swinging leg. The trunk is then moved forward over the stance foot using the walking aid. While the trunk is progressing over the stance foot, the swing leg moves from hip extension to flexion. In reciprocally linked

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orthoses it is reported that swing hip flexion is driven by stance hip extension through the reciprocal link. Additionally the reciprocal link provides support for both hip joints during double support periods (Beckman, 1987). In free swing orthoses, hip flexion on the swing side is achieved by gravity and inertia from the previous step. The hip joints guide the path of the hip joint in adduction and abduction, but allow free swing in flexion and extension between stops (Moore, 1988).

The benefit of the reciprocal link in an orthosis has been assessed indirectly by comparing the energy expenditure of patients walking in reciprocally linked orthoses and in free swing orthoses. Hirokawa *et al.* (1990) measured the energy expenditure per metre walked in the LSU-RGO (reciprocally linked orthosis) over a range of walking speeds (0.1 to 0.4m/s) and compared this to values from the literature on the ParaWalker (free swing orthosis). At low speeds the energy expenditure per metre for walking in the LSU-RGO was lower than the energy expenditure per metre for walking in the ParaWalker. The energy expenditure per metre for walking in the ParaWalker became lower than that when walking in the LSU-RGO at higher speeds. In the double support phases of the gait cycle, the reciprocal linkage of the LSU-RGO supports the hip joints so that less energy is expended resisting bilateral hip flexion. During the swing phase the ParaWalker has a freely swinging leg, the hip joint has lower friction and the swing leg is not constrained by the action of the stance leg, thus the leg requires less energy to be expended in moving it forward. As the speed of walking increases the double support phases of gait decrease proportionally compared to the swing phase. Therefore at slower speeds (where double support is a greater proportion of the gait cycle) walking in the LSU-RGO uses less energy, and at faster speeds (where the swing phase becomes more important) walking in the ParaWalker uses less energy.

The ParaWalker differs from the LSU-RGO in other areas besides the reciprocal linkage. It has a higher lateral stiffness than the LSU-RGO (Jefferson and Whittle, 1990), and the angle of ankle fixation is often different (Isakov *et al.*, 1992; Stallard *et al.*, 1986). The effect of those differences means that it is uncertain whether the relative changes in orthotic function shown by Hirokawa *et al.* (1990) are entirely due to the reciprocal linkage. Ijzerman *et al.* (1997) addressed these problems by measuring oxygen cost and speed of patients walking in the ARGO (reciprocally linked orthosis) and walking in the same orthosis with the reciprocal link replaced by flexion stops (free swing orthosis). The patients in the study with T4 lesions generally walked slower than those with lower level lesions (T9 - T12). In line with the findings of Hirokawa *et al.* (1990) the oxygen cost of patients with high thoracic lesions (slower gait) walking in the ARGO was lower than in the free swing orthosis, while the oxygen cost of patients with lower level thoracic lesions (faster gait) walking in the ARGO was higher than walking in the free swing orthosis.

The most effective way of assessing the action of the reciprocal link is to measure the use to which the cable is put during gait. Petrofsky and Smith (1991) attached load cells to both cables of an LSU-RGO and measured the force in them while spinal cord injured patients were walking. The force measured in the cables was less than 230N during level walking. The distribution of the cable force with respect to the phase of the gait cycle was shown graphically, but the beginning and end of the phases are difficult to determine precisely. Unfortunately no attempt was made to distinguish between force patterns in the two cables.

In order to assess the effect of the reciprocal link on the gait of a person with spinal cord injury it is necessary to quantify the variation in tension in each cable with time during gait. The aim of this project was to measure the forces in the cables of an LSU-RGO and the resultant moments developed at the hip joints with respect to the phases of the gait cycle during walking of spinal cord injured subjects.

Equipment

All the subjects in the study walked using a dual cable Louisiana State University Reciprocating Gait Orthosis (LSU-RGO). The LSU-RGO is shown in Figure 1.

At the hip joint Bowden cables were used which would only transmit forces and motion when in tension. Each cable consisted of an inner cable attached to the lower section of the hip joint, and an outer conduit, which is attached to the trunk section of the orthosis. The lower member of the hip joint was in the form of a T-

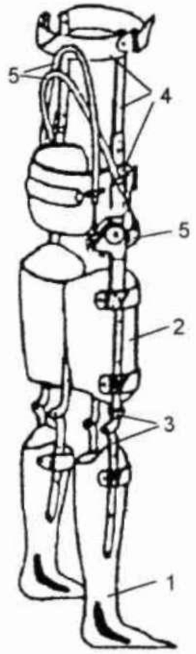

Fig. 1: The LSU-RGO (drawing adapted from Davidson 1994). 1 – polypropylene AFO section; 2 – polypropylene thigh sections; 3 – steel side-members with posterior offset knee joints and bale locks for sitting; 4 – a trunk section which consists of two steel uprights, polypropylene sacral band, stomach pad, and posterior and anterior Velcro straps at chest height; 5 – two uniaxial hip joints; the flexion and extension of which is constrained to reciprocate by a dual cable system. (Douglas *et al.*, 1983)

piece and a cable was attached to the front and back of each joint (Fig. 2). The cable attaching to the front of the left hip joint was also attached to the front of the right hip joint and is hereafter referred to as the front cable. The cable that was attached to the back of the right hip joint was also attached to the back of the left hip joint and is hereafter referred to as the back cable. When both feet are on the ground the back cable prevents bilateral hip flexion and the front cable prevents bilateral hip extension.

Therefore when a subject is in double support the walking frame is only needed for balance. During swing, when only one foot is in contact with the ground, the cables act in a reciprocal manner. Thus when one hip is flexed, the other hip extends by an equal angle. If one hip is driven into flexion by extension of the contralateral hip, the front cable will be in tension. Conversely if one hip is driven into extension by flexion of the contralateral hip, the

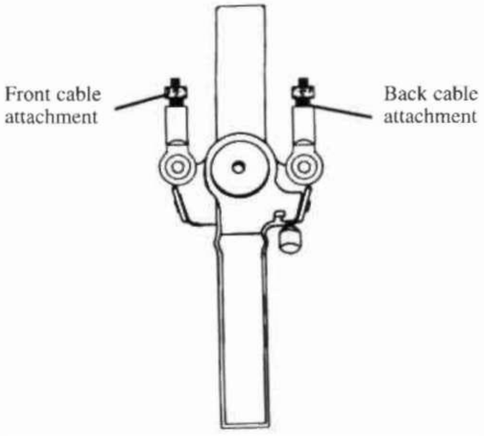

Fig. 2: Schematic representation of the hip joints and reciprocal cable link of an LSU-RGO.

back cable will be in tension. In theory during paraplegic gait in the LSU-RGO the back cable is in tension during standing and double support phases to prevent the patient collapsing into flexion at the hips. On the other hand the front cable is in tension during swing phase as the stance leg hip extension is used to drive contralateral hip flexion.

To measure cable tension two strain gauge transducers were constructed. Four strain gauges connected in a bridge arrangement were attached to an aluminium alloy cylinder (40mm by 12mm diameter) for each transducer, so that axial force was measured. The transducers were fitted between the cable and its attachment to the lower member of the hip joint (Fig. 3). Since the effective length of the cable was increased by insertion of the transducer a bracket was constructed to hold the outer cable. A slot in this bracket meant that the relative length of the front and back cables could be adjusted to fit each subject. One transducer was fitted to each of the front and back cables of the orthosis used by each subject.

To distinguish between stance and swing phases of the gait cycle, foot switches were attached to the heel and toe of each foot. Square force sensitive resistors (FSR) (Interlink Electronics, Luxembourg) 35mm square were placed on the sole of the AFO. The gain of the amplifier to which the FSRs were connected was set so that the output was at base line when the switch was not in contact with the floor and saturated high when the switch was in contact

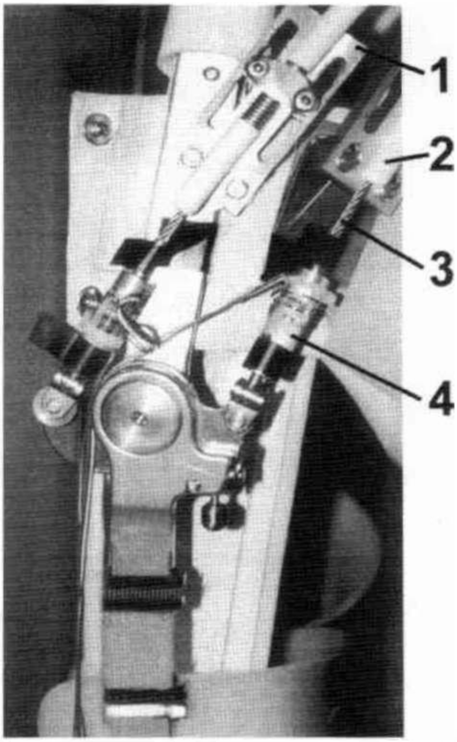

Fig. 3: Cable force transducers on the front and back cable of an LSU-RGO. 1 – extension bracket; 2 – outer conduit; 3 – inner cable; 4 – cylinder with strain gauge transducer setting.

with the floor and minimal weight on the foot.

During walking trials data were collected from the force transducers and foot switches by a computer, sampled through an A/D converter at 50Hz, via an umbilical cable.

Method

Six (6) adult spinal cord injured subjects participated in the study (Table 1). All were using the LSU-RGO as part of an exercise programme at the Regional Spinal Injuries Centre, Southport. Five (5) subjects walked with a rollator and one subject (F) walked with crutches.

Before the trial commenced each orthosis was modified to fit a force transducer and associated bracket to each cable. On arrival the subject donned the orthosis and foot switches were attached. The subject walked at a self-selected speed along a straight 10 metre track. The subject then turned and rested for a few minutes before repeating the trial. The trial was repeated at least four times so that between 20 and 40 steps were available for analysis for each subject. A sagittal plane video of the trials was taken to provide a record of the walking action.

Analysis

The data for each cable were divided into phases of the gait cycle. These were double support right leg back, double support left leg back, right swing and left swing. Division into four sections was chosen, as the patients' gait was not assumed to be symmetrical with respect to left and right steps. The swing phase was defined to start from toe off and end with heel strike. The transition between low and high saturation took approximately 0.2 seconds (approximately 10 data points), so it was necessary to set a threshold to pinpoint where toe off and heel strike occurred. If extraneous switch signals were present, for example if the subject scuffed their foot during swing phase, these were edited manually so that they did not affect gait phase determination.

The time scale of the data for each gait phase was transformed to a percentage scale. Graphs of the mean force and standard deviation for each subject and cable were plotted over the entire gait cycle. Maximum force in the cable and the percentage of gait phase that the cable was in

Table 1. Details of subjects included in the study. Walking speed is the average speed measured during the trials.

Subject	Sex	Injury level	Complete/ incomplete	Age	Body mass (kg)	Time since injury (years)	RGO use	Walking Aid	Walking speed (m/s)
A	M	T5	complete	37	54	15	17 months	rollator	0.18
B	M	T12	complete	44	67	10	7 months	rollator	0.17
C	M	C5/6	complete	54	63	6	17 months	rollator	0.16
D	M	T4/5	complete	28	63	2	5 months	rollator	0.18
E	F	T11/12	complete	28	54	1	6 months	rollator	0.17
F	F	T7	incomplete	40	84	15	11 years	crutches	0.42

tension were calculated. The moment arm of the front and back cables at the hip joint of the orthosis with the joint in neutral was measured. The mean distance was determined and used to calculated the moment produced about the hip joint of the orthosis by the tension in each cable during the gait cycle. The maximum moment during a gait phase for each cable and patient was calculated. To eliminate artefact caused by noise a threshold of 10N (0.3Nm) was chosen and all outcome measures were only calculated using data points that were above this threshold.

The average speed at which each patient walked during a trial was calculated from the video of those trials. The time taken for the patient to walk between lines of known spacing marked on the gymnasium floor was timed using a stopwatch.

Results

Force cable data, for front cable and back cable, transformed so that each gait cycle is the same length are shown in Figure 4 for 2 subjects. From this information the mean and standard deviation of cable force throughout the gait cycle was determined. A plot of the mean value is shown for all subjects in Figure 5. For most subjects the front cable showed no tension above 10N (0.3Nm) at any point in the gait cycle. Subject B had a peak in front cable tension at the end of the double support phase (left leg back) (Fig. 4). Two (2) subjects had peaks in the front cable in swing phase, subject A for both swing phases and subject E for the right swing phase only. Typically the back cable force was high during stance phase, and tailed off during the first half of swing phase to rise again towards the end of swing phase. The back cable force often built up during double support to a maximum in a series of peaks during double support phases (Fig. 4), but this detail was lost when the traces were averaged.

The means and standard deviations of percentage time above threshold and maximum moment developed at the front cable, for all subjects grouped by gait phase, are shown in Figures 6 and 7.

During stance phase the percentage time that the front cable was in use was less than 20% for all subjects, and was close to 0% in both double support phases for 4 subjects and in one double support phase for 1 subject (B). The front cable was in tension for 60% to 80% during swing phase for 1 subject (A) and was in tension for 25% during the right swing phase for 1 subject (E). For the other subjects and the left swing of subject E, the front cable was not in tension during the swing phase. For subjects C, D and F the maximum force in the front cable during stance and swing phases was 0N and therefore the effective moment at the hip was 0Nm (underneath the 0.3Nm threshold). The maximum hip moment produced by the front cable by subject A was around 3Nm in both double support phases and right swing, and 4Nm in the left swing phase. For subject B the maximum moment produced by the front cable during double support left leg back was 3Nm and 0Nm for all other gait phases. There was a maximum moment produced in the front cable of 5Nm during the right swing phase of subject E, and 0Nm for the other swing phase and both double support phases.

For the back cable the means and standard deviations of the percentage time of gait phase that the cable force was above threshold and maximum moment for all subjects by gait phase are shown in Figures 8 and 9, grouped by gait phase. The back cable was in tension between 97% to 100% of stance phase for all subjects. During swing phase the back cable was in tension for less than 100% of the phase in 4 patients. The other 2 patients had the back cable in tension 100% of swing phase for one leg but not for the other. Excluding these the back cable was in tension from 40% to 90% of the swing phase. The maximum moment in the back cable during stance was greater than during the swing phase for 5 out of 6 subjects. The range of maximum moments was greater in the double support left leg back phase than the double support right leg back phase. The maximum moment ranges from 35Nm to 12Nm during double support. The maximum moment during swing phase was more consistent between subjects than for the double support phases with little difference between right and left leg swing phases. The maximum moment during swing phase ranged from 8Nm to 14Nm, apart from the left leg swing for subject F which produced a maximum moment of 18Nm.

Discussion

The pattern of front cable use showed distinct variations between subjects, although the pattern was consistent within subjects. The pattern of

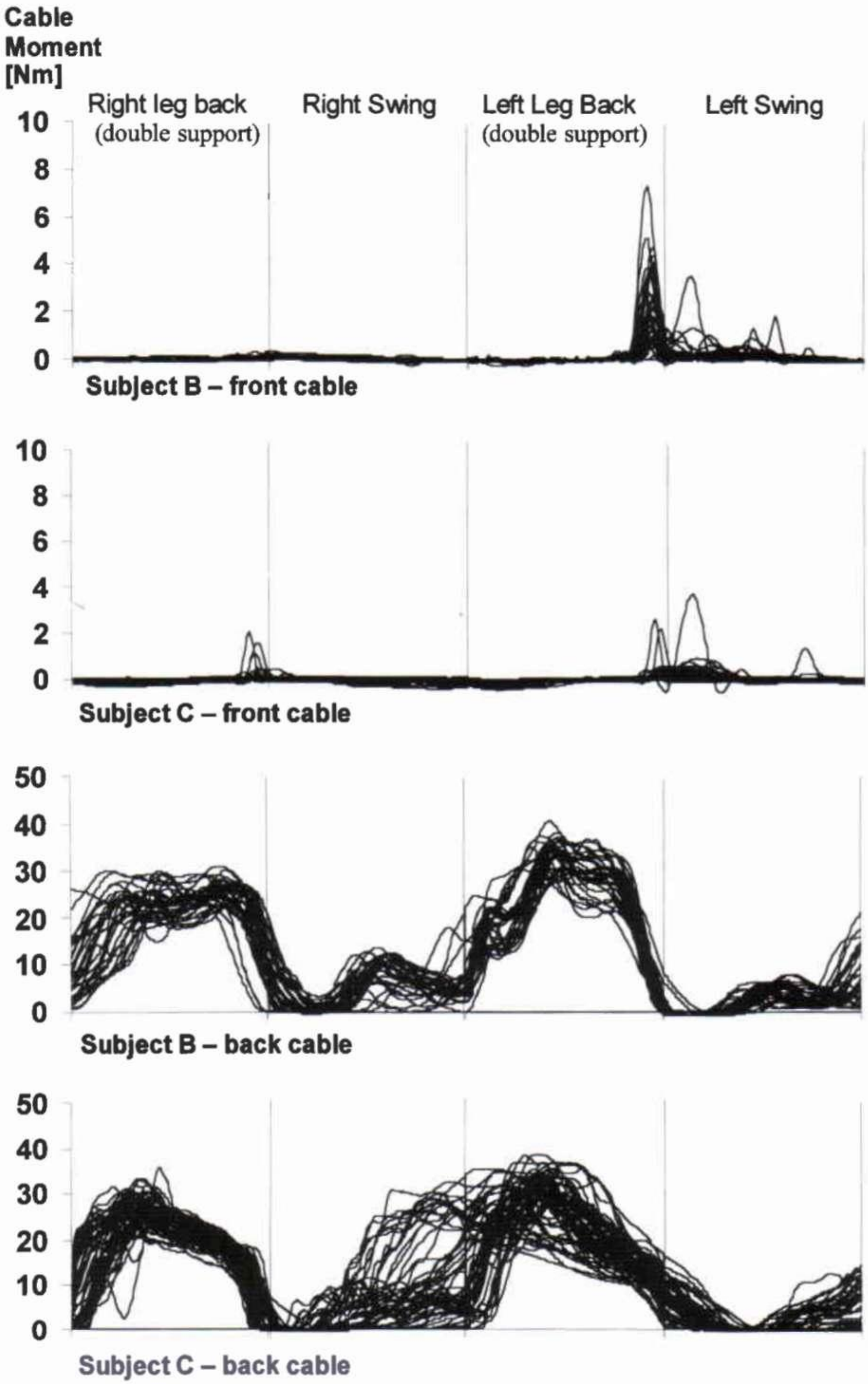

Fig. 4: The hip moment (Nm) produced by cable force in the front and back cables during the gait cycle, for each step taken by two subjects. The time scale of the graph has been modified so that each phase of the gait cycle is the same length for each step taken.

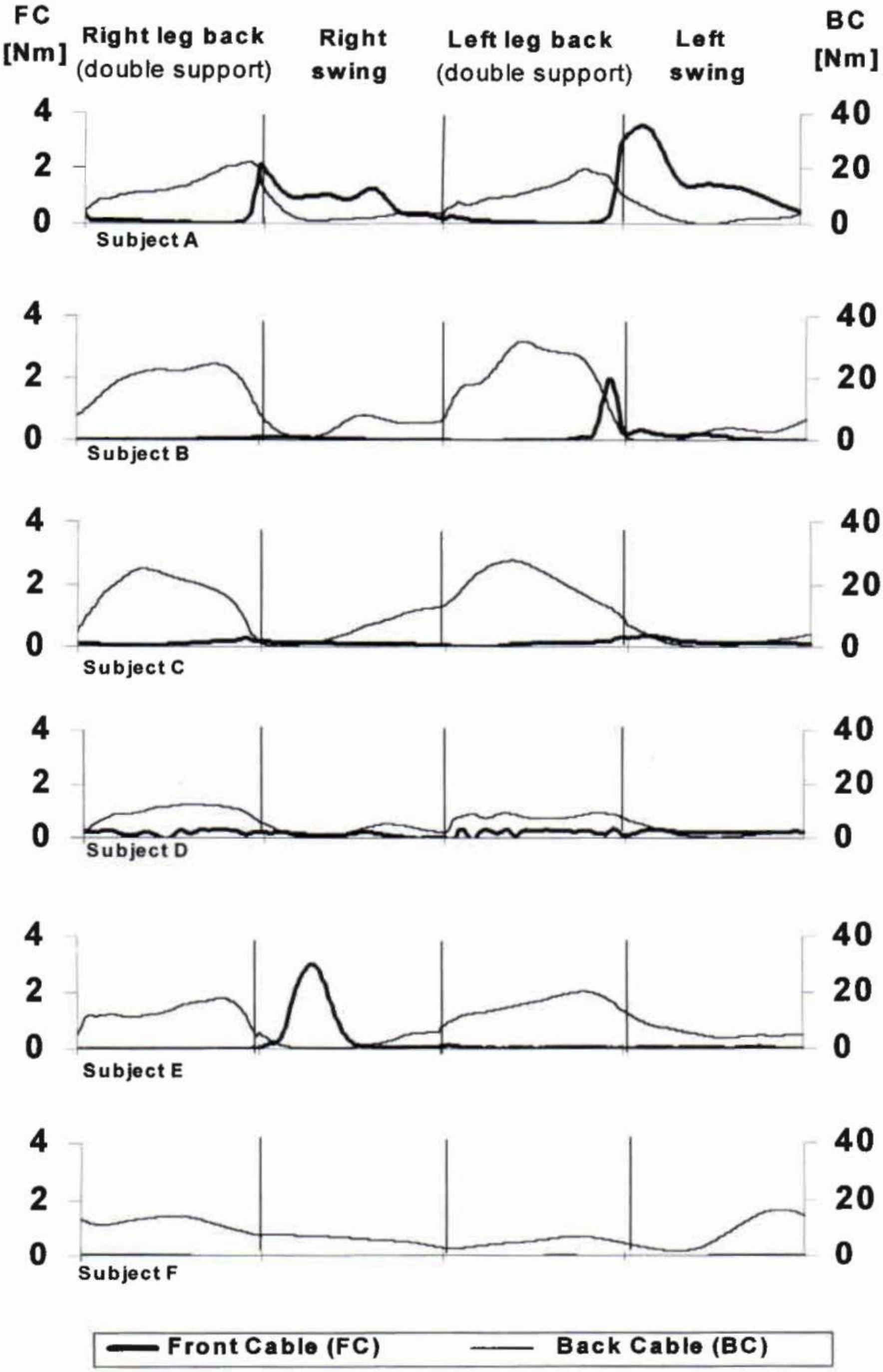

Fig. 5: Mean hip moment (Nm) produced by cable force in the front and back cables during the gait cycle for each subject. The time scale of the graph has been modified so that each phase of the gait cycle is the same length for each step taken.

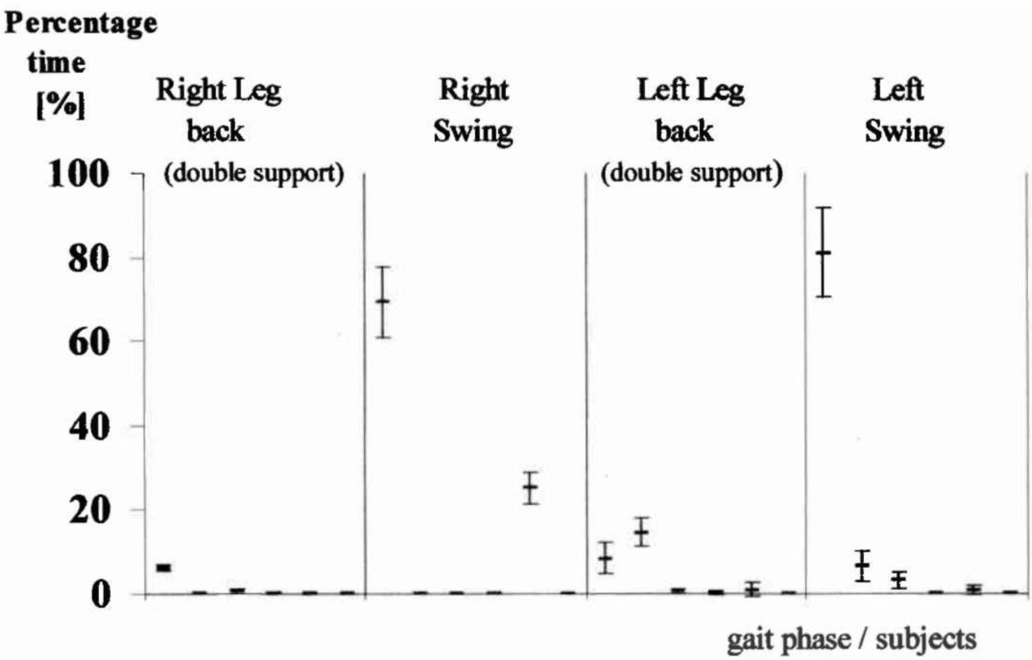

Fig. 6: Mean and standard deviation of the percentage that the front cable force was above the threshold, grouped by gait phase, for all subjects. Within each phase subjects are shown in order (A-F).

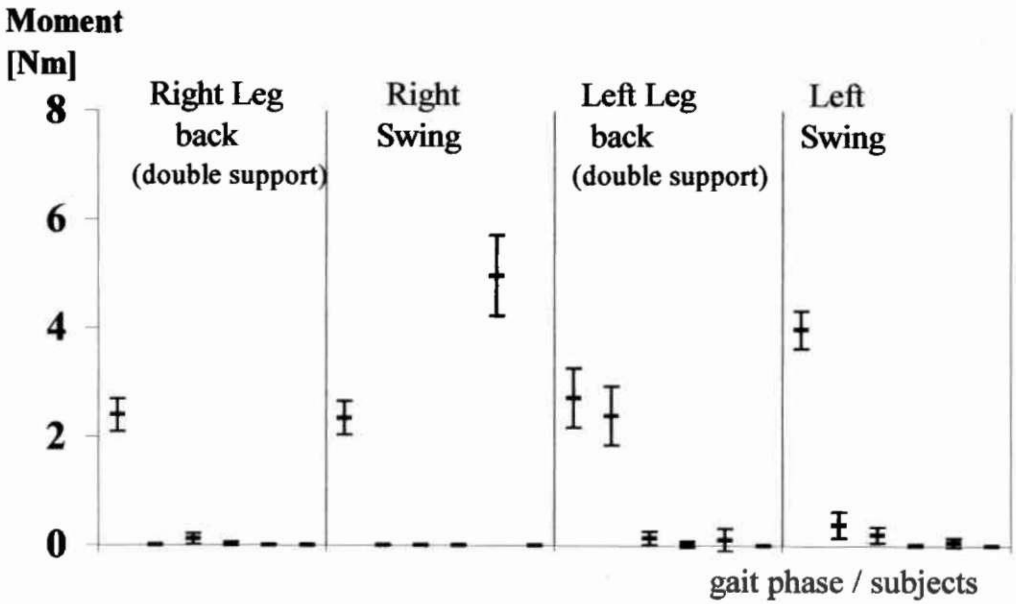

Fig. 7: Mean and standard deviation of the maximum hip moment produced by cable force in the front cable, grouped by gait phase, for all subjects. Within each phase subjects are shown in order (A-F).

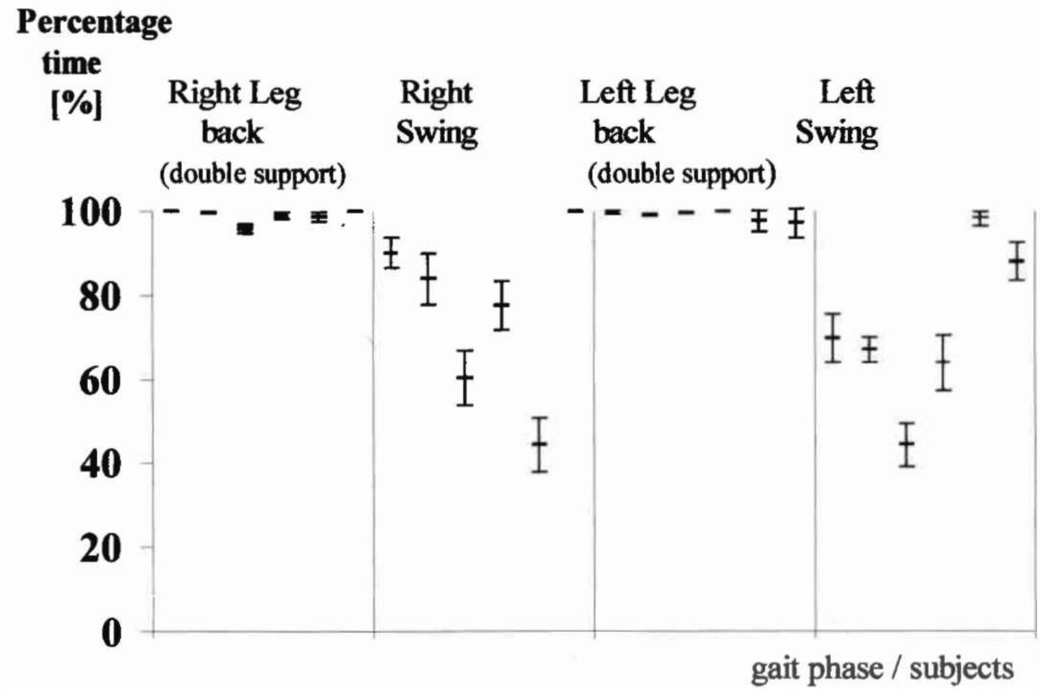

Fig. 8: Mean and standard deviation of the percentage time that back cable force was above threshold, grouped by gait phase, for all subjects. Within each phase subjects are shown in order (A-F).

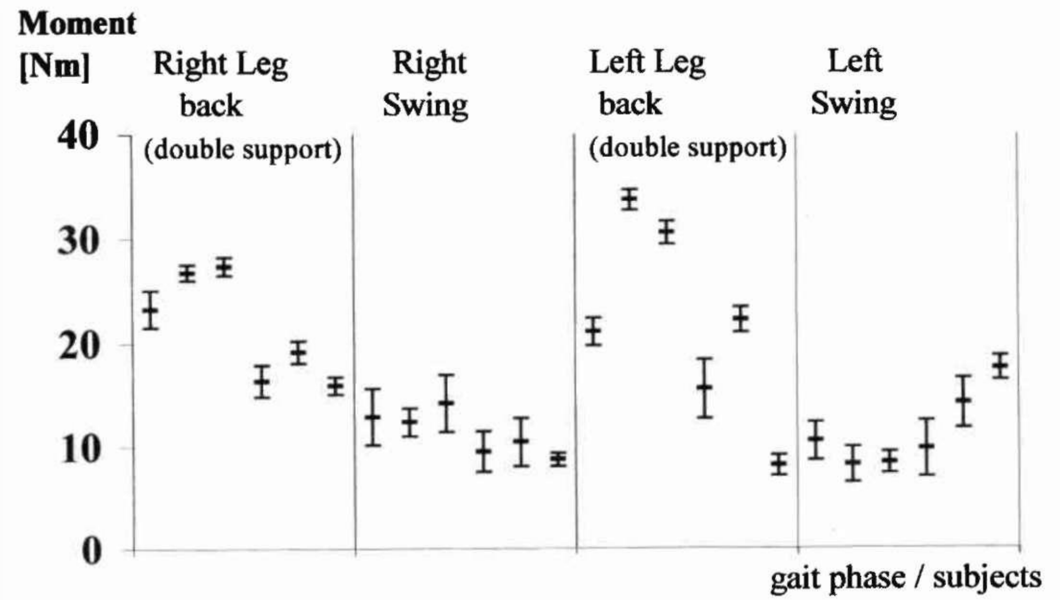

Fig. 9: Mean and standard deviation of the maximum hip moment produced by cable force in the back cable, grouped by gait phase, for all subjects. Within each phase subjects are shown in order (A-F).

back cable use was more consistent between subjects.

It has been assumed that the reciprocal link is used during swing phase, so that swing hip flexion is driven by stance hip extension. Thus tension in the front cable is thought to be generated and dissipated during the swing phase. This was clearly not the case for the majority of patients in this study. For 3 subjects the front cable was not in tension for the whole gait cycle and had a maximum moment in the cable of 0Nm, demonstrating that the cable was not being used at all during gait. There was tension in the front cable of subject E, during the right swing phase of the gait cycle. The force built up and declined from 10% to 50% of the swing phase with a peak value of 5Nm aiding swing hip flexion. The build up of tension in the front cable during swing phase indicated that stance hip extension was the cause of tension in the cable. Subjects A and C built up tension in the front cable during the last 5% of the double support phase to a hip flexion moment of 3 Nm, which, since both feet were on the ground, must have been caused by bilateral hip extension. The tension in the front cable of the orthosis of subject C dissipated before toe off and so did not contribute in any way to swing hip flexion. The tension built up in the front cable of subject A during double support was reduced during swing phase. Thus during the initial part of swing phase, swing hip flexion was being assisted by tension in the cable to a value of 3Nm.

Winter (1990) measured the moments acting on the hip joint during normal walking. The maximum hip flexion moment was 0.7Nm/Kg body mass, which for subjects in this study would be approximately 40Nm. The maximum moment created to aid hip flexion during the swing phase of gait in the front cable in this study was 5Nm. Thus the moment produced at the hip in those 2 users who were using the front cable to aid hip flexion was one eighth of that used during normal walking, and could not have been the sole driving force of hip flexion in the leg.

The pattern of use of the back cable was consistent between subjects and full use of the back cable was made during the double support phase resisting a hip flexion moment of 12Nm to 35Nm. At the start of swing phase all subjects showed a reduction in the tension in the back cable. The percentage of the gait phase over which this happened varied considerably, between 10% and 50% of gait phase, but could generally be said to have been a continuation of the decline in force started at the end of double support. Force in the back cable is built up during the stance phase as the subject leans forward towards the walking aid, some subjects relax this flexed position by standing straighter just prior to swing, whereas others try to stand straighter during swing phase itself. Those who stand straighter before the start of swing relax the force in the cable earlier, so the fall is shorter during the swing phase.

During the second half of the swing phase the moment produced by the back cable at the hip built up to between 3Nm and 14Nm. One foot was not in contact with the ground so the cable, at this point, was acting in a reciprocal fashion. Tension in the back cable should therefore drive hip extension. However, as the weight of the swinging hip was less than the weight of the trunk, this was unlikely to be the case. The swing leg continued to flex and it is likely that the back cable was acting as a retarding force on the swinging hip, possibly restricting the length of swing and slowing down gait speed.

The build up of force in the back cable during double support did not generally occur as one smooth motion, but rose to the maximum force in a series of peaks and troughs. Subjects with higher level lesions built up tension in the back cable in a smoother fashion than subjects with lower level lesions. It is assumed that subjects with lower thoracic level lesion placed less reliance on the use of the back cable during stance phase as this support could be periodically reduced, whereas subjects with higher level lesions relied on the back cable to support their hips far more. Hence back cable support was more continuous.

Maximum moment and percentage that cable force was above threshold were not related to the subject's age, lesion level or time since injury. Subject F, the only incomplete user, had been using the LSU-RGO for 11 years, 9 years longer than any other subject in the study, and was the only user to walk with crutches. She walked substantially faster than the other subjects and had a lower maximum moment in the back cable during stance than the other patients. It was impossible to conclude whether this difference in use of the back cable and speed of progression were due to the incomplete nature of subject F's injury, to the length of time she had been using the orthosis, to the use of crutches as the walking aid, or to her own natural ability. Excluding subject F, maximum moment and percentage that cable force is above threshold were not related to length of time using the LSU-RGO, or speed of progression.

Conclusions

1. The front cable was not used to any effect in 4 out of 6 subjects.
2. There were two subjects that used front cable tension to assist hip flexion of the swing leg, but each used a different action to produce this assistance. Subject A used bilateral hip extension in double support to generate tension in the front cable. Subject E used hip extension prior to stance phase to assist swing hip flexion.
3. The maximum hip flexion moment induced by front cable tension was 5Nm; at best one eighth of normal hip moments.
4. The back cable was mainly used during double support to resist bilateral hip flexion to a maximum of 12Nm to 35Nm.
5. Tension was built up in the back cable during the latter part of swing phase, to between 3Nm and 14Nm, and may have restricted stride length.

It is suggested that the reciprocal linkage with respect to the front cable is not being used as expected, and that an orthosis providing the function of the back cable in double support and with no provision for front cable function will be as effective as the current LSU-RGO.

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