



Analysis of assisted-gait characteristics in persons with incomplete spinal cord injury

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Ambulatory assistive device use can improve functional independence following spinal cord injury and, potentially, quality of life. However, the interaction between aids and user in this population is poorly understood.

Objectives: To determine the influence of walkers, crutches and canes on assisted-gait following incomplete spinal cord injury.

Study design/methods: Outcome parameters evaluated in ten individuals included orthogonal forces exerted on instrumented assistive devices, walking speed, cadence, step length, trunk and thigh angles, as well as knee and ankle joint angles. Kinetic data included axial compressive force, and medio/lateral and antero/posterior bending forces.

Setting: Canada.

Results: Results indicated that walkers ($n=5$) provided the greatest vertical support (up to 100% body weight), but resulted in slow gait with a forward flexed posture. Elbow crutch users ($n=3$) walked faster (greater step length and cadence) and had a more upright posture than the walker users. Crutches supported up to 50% of the subject's body weight, granted lateral stability, and provided restraint in the antero/posterior direction. Canes ($n=2$) offered restraining and propulsive assistance, some lateral stability, and the least amount of vertical support.

Conclusion: Ambulatory devices affected posture and walking speed while fulfilling various assistive functions during locomotion. The conclusion drawn is that rehabilitation specialists are advised to match device characteristics to user needs when prescribing walking aids.

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Keywords: spinal cord injury; gait; ambulatory assistive devices; walkers; crutches; canes

Introduction

Little is known about the influence of ambulatory assistive devices (AAD) on gait. Assistive devices, such as walkers, crutches and canes, are prescribed to persons for a variety of reasons: to decrease excessive weight bearing on the lower extremity, to correct imbalance, to reduce fatigue, or to relieve pain secondary to loading of damaged structures.^{1–3} AAD also assist in force production by using the upper extremities to compensate for a patient's weak lower extremity. Selecting an appropriate AAD should depend on objective assessments of a person's functional requirements and physical capabilities. However, clinical guidelines for AAD prescription are not standardized and often depend on the clinician's experience. Prescription often does not take into

account the influence that AAD may have on the user's resultant gait pattern.^{4,5}

Individuals requiring AAD have a decreased ability to provide the supporting, stabilizing, propulsive or restraining forces necessary for forward progression. In principle, weight is partially borne by the upper extremities in order to relieve loading on the lower extremities.⁶ Compensatory forces are provided by the upper extremities and are transmitted through the wrist and hand to the AAD and subsequently to the floor. To understand the biomechanics governing assisted gait, forces exerted on the AAD have to be examined since the upper extremities are now involved in providing complementary forces for locomotion.^{7–10} The main goal for AAD use is the provision of safe and independent ambulation. To improve gait of AAD users, the influence of AAD on gait needs to be quantified.

Long-term AAD users include persons with incomplete spinal cord injury (SCI). Walking is not only a

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personal wish for individuals with SCI, but standing and walking also have known physiological benefits. These benefits include the prevention of contractures, minimization of osteoporosis, amelioration of circulation, and improvement of renal function.^{11,12}

Because more spinal cord injuries are now incomplete, survivors are functioning at a higher level and many are able to walk with an AAD.¹³ When devising gait-training strategies, it is important to understand the influence of AAD on an individual's gait characteristics. The objective of this study was to quantify the biomechanical characteristics of assisted gait in terms of: (a) walking speed, cadence and step length; (b) trunk, hip, knee and ankle joint angles; and (c) forces exerted on the AAD.

Methods

Subjects

Ten subjects (three women and seven men) with varying levels of incomplete SCI participated in the study (Table 1). All subjects had sustained an incomplete SCI more than 1 year prior to the study, had completed their formal rehabilitation, and could walk at least 20 m with their AAD. Subjects fitting these criteria were contacted via their treating physician. Subjects were recruited from The Rehabilitation Centre in Ottawa. Ethical approval was obtained and participants gave informed consent prior to participation in the study. Participant ages ranged between 24 and 72 years (mean 41 ± 24 years). The AAD used by the subjects included a straight cane with a 'T-handle', elbow crutches (also referred to as forearm or 'Canadian' crutches) and a walker with four rubber tips.

Experimental procedures

All subjects were asked to walk with a height-adjustable instrumented AAD, similar to the assistive device they would normally use.¹⁴ Each AAD was instrumented with 16 strain gauges at the tip of each leg. These strain gauges measured forces exerted orthogonally on the AAD: the axial compressive force along of the AAD

shaft, the antero/posterior bending force, and the medio/lateral bending force (Figure 1A and B). The total axial compressive force for the walker is calculated by adding the axial compressive force found on each of the four legs. The antero/posterior bending force exerted on the AAD can be positive or negative. A positive value refers to a 'restraining' force and a negative value refers to a 'propulsive' force.¹⁵ Data was collected at 300 Hz. Three reflective markers were placed in a triangular formation on the AAD (near the top, middle and distal end of the device) to determine device orientation in space and the particular orientation of the local reference system relative to the global reference system (eg, X, Y and Z axes, the Y-axis being vertical and the X-axis being co-axial to the line of progression). The three orthogonal forces were then expressed in terms of the global reference system through trigonometric functions.

In this report, the antero/posterior and medio/lateral bending forces were only reported for subjects walking with crutches and canes. The walker frame was considered a non-deformable structure with four legs with low deviation from the vertical (ie, bending forces were two orders of magnitude lower than the axial force).

Seven reflective markers were attached to bony landmarks on the subject's weaker side. These landmarks included the acromion, greater trochanter, lateral tibial plateau, lateral malleolus, posterior calcaneus, head of the fifth metatarsal and the tip of the fifth toe. Each subject was then asked to walk over a 10 m flat pathway at their regular speed while being filmed by a video camera (60 Hz). Subjects were instructed to walk according to their usual walking style. Five walking trials were collected for each subject and rest periods were provided as needed.

Data processing

The video data were digitised using the Ariel Performance Analysis System (Ariel Inc., USA). Data was digitally filtered at 10 Hz. Trunk angle with respect to the vertical (referred to as trunk alignment in this article), thigh angle with respect to the vertical (referred to as hip angle in this article), and the relative knee and ankle joint angles were calculated from the digitized data points and normalized to 100% of the gait cycle. Flexion and dorsiflexion angles were considered as positive joint excursion for convention with the baseline reference as 0° (neutral alignment). Walking speed, cadence, and step length were calculated from the digitized data.

During each walking trial, three consecutive AAD contacts were analysed and averaged for each of the five trials (15 in total). The axial compressive force was low-passed filtered at 25 Hz, expressed in terms of percentage of the user's total body weight (%BW) as well as to 100% contact time. The forces in the medio/lateral and antero/posterior directions were also filtered at 25 Hz and were normalized to user's total

Table 1 Subject characteristics

| Subject | AAD | Age (years) | Body weight (N) | Injury level | Weaker side | Year of injury |
|---------|--------|-------------|-----------------|--------------|-------------|----------------|
| WA1 | walker | 39 | 732 | C5 | right | 1989 |
| WA2 | walker | 70 | 817 | T12 | right | 1992 |
| WA3 | walker | 67 | 790 | T6 | left | 1991 |
| WA4 | walker | 32 | 701 | T11/T12 | right | 1990 |
| WA5 | walker | 36 | 709 | C5/C6 | right | 1988 |
| CR1 | crutch | 24 | 654 | C7 | right | 1988 |
| CR2 | crutch | 24 | 653 | L1 | left | 1994 |
| CR3 | crutch | 29 | 706 | T10 | right | 1994 |
| CA1 | cane | 72 | 557 | T10 | left | 1994 |
| CA2 | cane | 58 | 785 | T6/T7 | right | 1989 |

body weight (%BW), lever arm (AAD's length) and 100% of contact time.

Results

Walking pattern

A typical pattern for the walker users in this study involved advancing their walker, taking one or two

steps and then advancing the walker again. All four walker legs were placed on the ground at the same time. The walking pattern was interrupted because the subjects had to stop forward progression to advance the walker. The crutch users walked with a '4-point' gait pattern, where the ADD and legs were advanced reciprocally.⁴ For example, the right crutch would be advanced, then the left leg, then the left crutch, and subsequently the right leg. Cane users held their cane

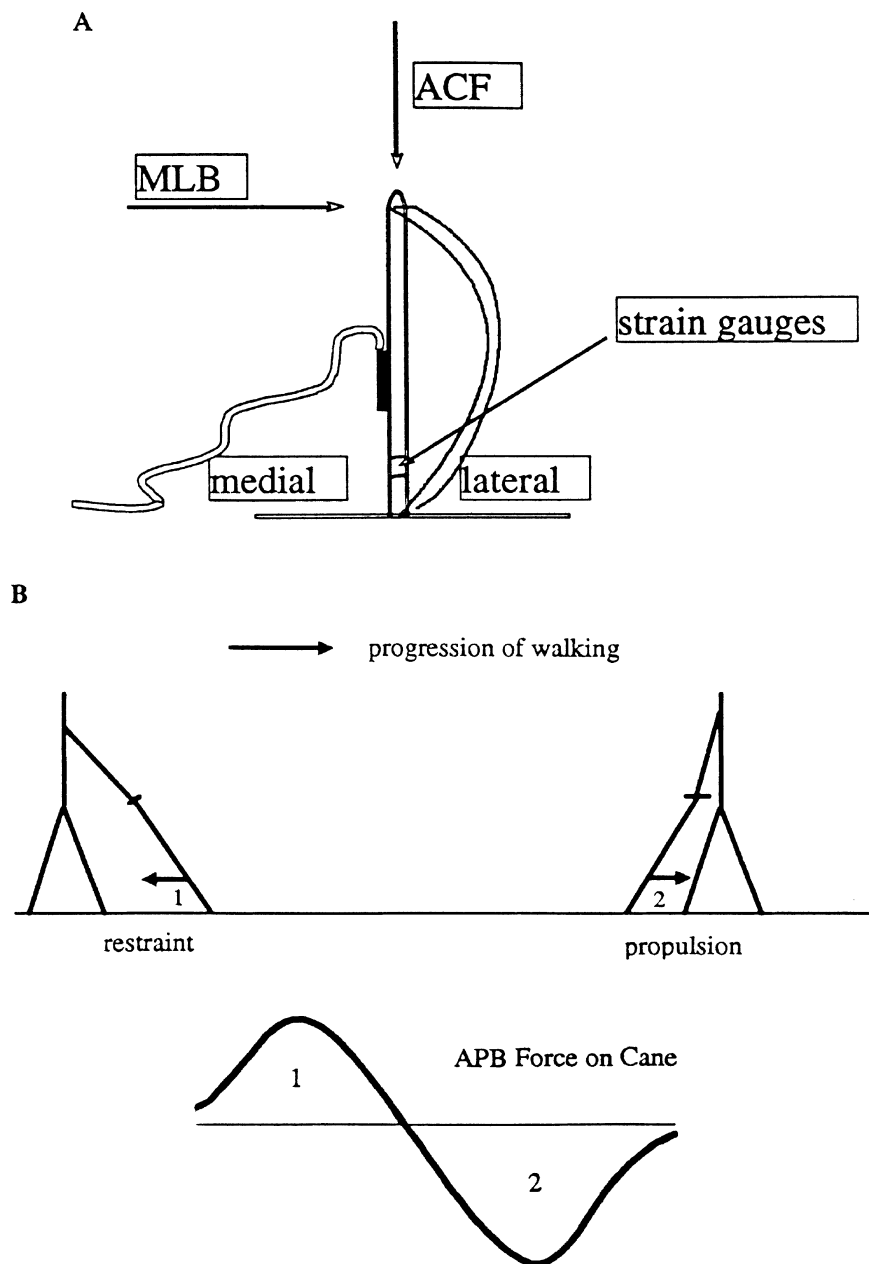


Figure 1 Posterior view of the instrumented cane (A). As force is exerted on the AAD, the axial compressive force (ACF) as well as the medio/lateral bending (MLB) and antero/posterior bending (APB) forces are measured. The APB force is in the direction of progression, orthogonal to the MLB force and the ACF. As the subject walks with the cane from left to right (B), the cane changes spatial orientation to provide a restraining (positive value) and propulsive (negative value) function in the A/P plane

on the side of their stronger leg (ie, contralateral cane use). During walking, the cane was advanced at the same time as their weaker leg.

Walker user temporospatial parameters and kinematics

Walking speed, cadence, step length, and stance/swing ratio for the five walker users are depicted in Table 2. Speeds varied from 0.05 to 0.29 m s⁻¹, with an average walking speed of 0.17 (± 0.04) m s⁻¹. The data indicated that the temporospatial parameters were repeatable for each subject. The inter-subject average values were 30 (± 13) steps min⁻¹ for cadence and 0.30 (± 0.11) m for step length. The fastest walker user (WA3) had the largest step length but not the highest cadence. The speed of the slowest walker user (WA4) was influenced by an extremely slow cadence of 13 steps min⁻¹ and a step length that did not differ from the other walker users. Subject WA4 spent a long time in stance (97%) and a relative short time in swing (3%) during the gait cycle. The stance/swing ratios for the other walker users varied between 73:27 and 95:5.

Kinematic evaluations revealed that all five walker users walked with the trunk flexed forward during the entire gait cycle (Figure 2A and B). Trunk flexion angles varied between 10° and 40° (Figure 3). Subject WA1 was able to achieve hip extension and a total hip excursion of 40°. However, this subject walked with marked knee flexion during stance and only achieved a step length of 0.22 m. Subject WA2 was not able to extend the hip during walking and remained in approximately 7° of hip flexion during stance (Figure 2). Total hip excursion for this subject was only 15°. The small hip excursion could contribute to subject WA2's short step length (0.22 m). This subject's knee remained in flexion and the ankle remained in plantarflexion during stance. In order for this subject to achieve the second fastest walking speed (of the walker users), he walked with a high cadence of 85.5 steps min⁻¹.

Subject WA3 was also able to extend his hip, thereby increasing step length, and was the only

subject who could extend the knee joint to zero during the stance phase of gait. Subject WA4 had approximately 10° of hip extension during stance; however, the total hip excursion was only 20°. The small hip excursion was related to a short step length (0.21 m), and was not compensated for by a high walking cadence. This resulted in an extremely slow walking speed (0.05 m s⁻¹). These factors were possibly related to the difficulty subject WA4 experienced in the unloading of the leg. Subject WA4 also remained in marked knee flexion and plantar flexion during the stance phase of gait. Subject WA5 was able to achieve hip extension and, although full knee extension was not achieved, a knee excursion of approximately 55°.

Forces on the walker

Maximal axial compressive forces exerted on the walker averaged 74 \pm 39%BW and varied from 20%BW for subject WA1 to 100%BW for subject WA4 (Figure 4A). The shape of the force-time curves were consistent for all subjects except WA4. The WA4 force-time curve had one peak at approximately 35% of the loading cycle while the other subjects had a two-peak pattern that coincided with the extra upper extremity force needed during loading and unloading of the legs.

Crutch user temporospatial parameters and kinematics

The crutch user's walking speed ranged from 0.30 m s⁻¹ (subject CR1) to 0.80 m s⁻¹ for subject CR2 (Table 2). Subject CR1's slower walking speed was more influenced by a lower cadence than a smaller step length. The stance/swing ratio was similar for all three crutch users. Average walking speed, cadence and step length were higher than that of the walker users.

The crutch users walked with the trunk in extension (Figures 2 and 3). Subject CR1 in particular, remained in trunk extension throughout the gait cycle. All crutch users were able to extend the hip past 0° during the stance phase. On average, 14° (± 4) of extension and 31° (± 8) of total hip excursion were obtained. Subject CR1 was not able to reach complete knee extension during stance but had ankle movements that were similar to other crutch users. Subjects CR2 and CR3 differed with respect to knee extension and ankle excursion; CR3 obtained full knee extension and had the largest ankle excursion (20°).

Forces on crutches

Maximal vertical forces exerted on crutches ranged from 15%BW to 50%BW (Figure 4B). Subjects CR2 and CR3 applied forces of similar magnitude on the left and right crutch; however, a marked asymmetry was noted for subject CR1 (CR1 was the slowest walker in the crutch user group). More force was

Table 2 Individual temporospatial parameters

| Subject | Walking speed m.s ⁻¹ (\pm SD) | Step length m (\pm SD) | Cadence steps.min ⁻¹ (\pm SD) | Stance/swing ratio |
|---------|------------------------------------------------|------------------------------|------------------------------------------------|--------------------|
| WA1 | 0.17 (0.01) | 0.22 (0.01) | 46.7 (2.7) | 91/9 |
| WA2 | 0.22 (0.02) | 0.22 (0.02) | 58.5 (2.8) | 73/27 |
| WA3 | 0.29 (0.03) | 0.59 (0.10) | 31.1 (3.6) | 75/25 |
| WA4 | 0.05 (0.02) | 0.21 (0.03) | 13.1 (4.6) | 97/3 |
| WA5 | 0.16 (0.01) | 0.33 (0.01) | 28.5 (0.3) | 95/5 |
| CR1 | 0.30 (0.04) | 0.43 (0.02) | 42.0 (2.2) | 69/31 |
| CR2 | 0.80 (0.07) | 0.60 (0.04) | 89.3 (4.9) | 74/26 |
| CR3 | 0.76 (0.06) | 0.67 (0.05) | 67.8 (3.4) | 70/30 |
| CA1 | 0.66 (0.08) | 0.51 (0.05) | 76.6 (5.7) | 72/28 |
| CA2 | 0.88 (0.06) | 0.60 (0.15) | 87.7 (6.2) | 70/30 |

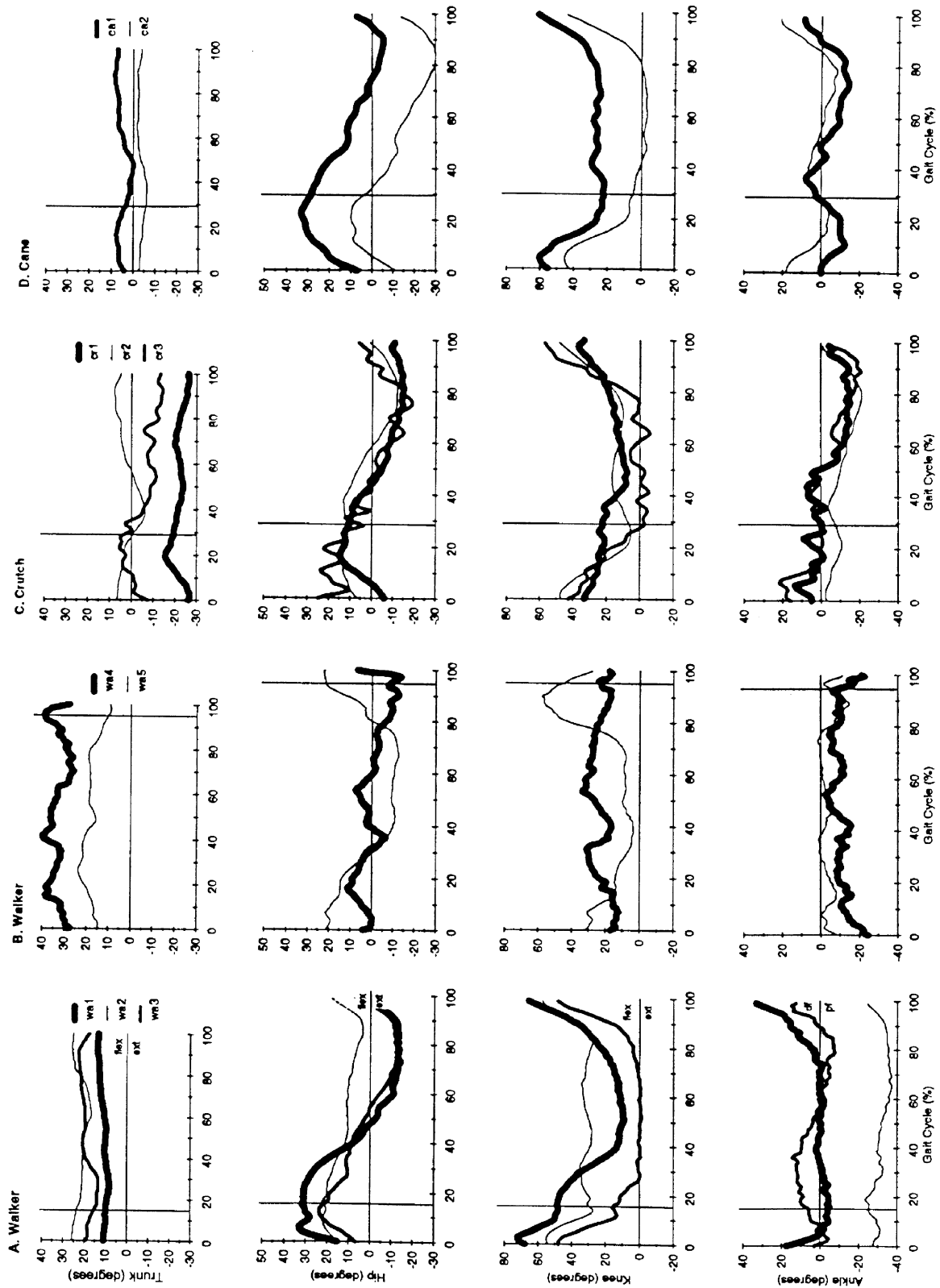


Figure 2 Trunk, hip, knee and ankle joint angles are depicted for subjects WA1, WA2 and WA3 (A), WA4 and WA5 (B), those walking with crutches (C) and those with canes (D). The X-axis on graphs (A) and (C) start with the toe-off of the filmed leg and end at the subsequent toe-off of the same leg. The X-axis on graph (B) starts with heel strike and terminates with the subsequent heel strike of the same foot. The vertical lines indicate the point of swing/stance transfer (A) and (C) or that of stance/swing transfer (B only)

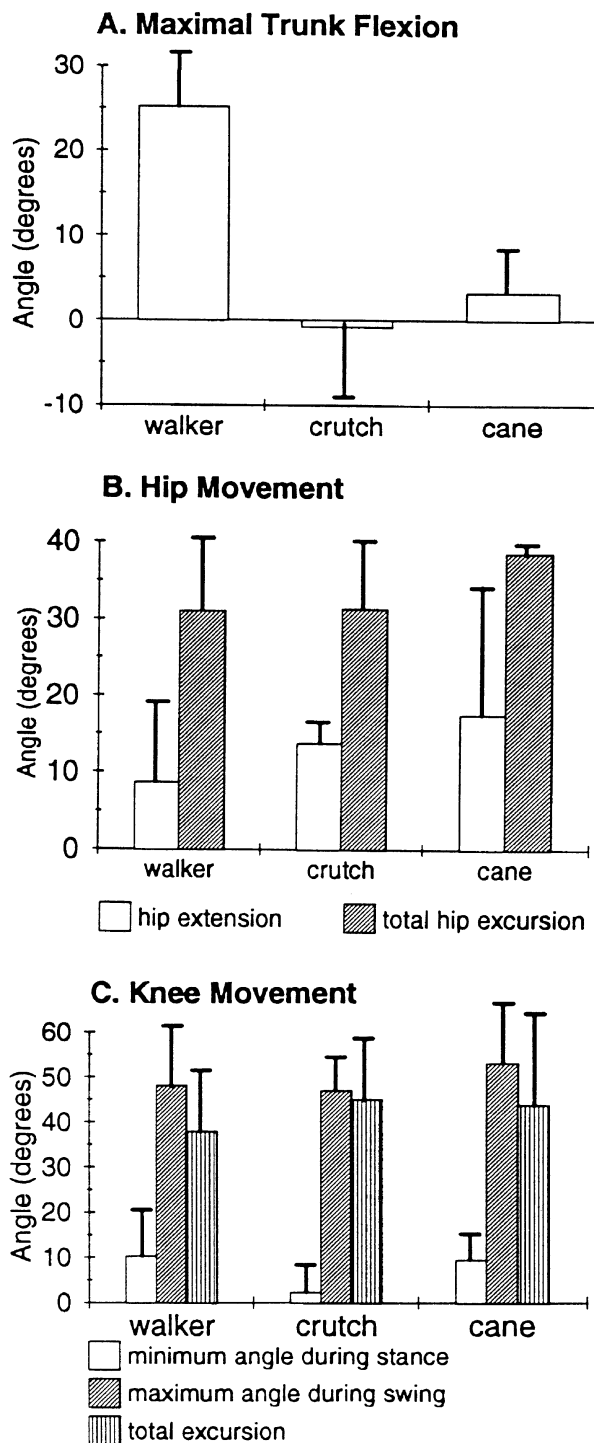


Figure 3 Maximal trunk flexion is depicted in (A), where negative values represent trunk extension. Hip extension and total hip excursion during stance are depicted in (B). The minimum flexion angle during stance, maximal flexion angle during swing and total excursion of the knee joint are depicted in (C). The error bars in all graphs represent standard deviations

exerted on CR1's left crutch to compensate for a weaker right lower extremity. The two peaks noted in Figure 4B1 coincided with the advancing of the lower extremities.

When analysing the medio/lateral bending forces exerted on crutches (Figure 5), it is clear that subject CR1 required the most external assistance in both the medio/lateral and the antero/posterior plane. As with the axial compressive force on the left crutch, the medio/lateral and antero/posterior bending forces were compensating for a weaker right lower extremity. These forces reached a peak value when the lower extremity was advanced, relating to difficulties in lower extremity loading and unloading. Crutch users CR2 and CR3 were quite similar in medio/lateral force exertion. These forces were less than the results for subject CR1.

Crutch users CR2 and CR3 utilized their AAD mostly to restrain their forward motion, as reflected by a positive antero/posterior bending force throughout loading (Figure 5). Visual inspection of crutch orientation indicated that, for both subjects, the crutches were tilted backwards throughout the gait cycle. However, subject CR1 used the right crutch to restrain forward motion during most of the loading cycle but used the crutch to push off during the latter part of crutch contact.

Cane user temporospatial parameters and kinematics

Walking speed, cadence, step length and stance/swing ratio of the two cane users were similar to that of the fastest crutch users, but were less variable between subjects (Table 2). Subject CA2's faster walking speed was more influenced by an increase in cadence than an increase in step length. Trunk alignment was similar and near vertical for both cane users (Figures 2 and 3). However, the two cane users differed in the amount of hip and knee excursion observed during swing and stance. Subject CA1 flexed the hip to 35° and extended the hip to only 5° for a short time period during stance (just prior to toe-off). Subject CA2 obtained only 10° of hip flexion during swing but extended the hip to 30° during stance. Subject CA1 was not able to obtain knee extension during stance and remained in 20° of flexion, whereas subject CA2 did not extend the knee completely and even went into slight knee hyper-extension during stance. Even though similar joint excursions were obtained at the hip and the knee, the joint positions during swing and stance were different for the two subjects. The ankle movements were similar for both cane users.

Force on canes

The two subjects exerted less than 15%BW (average $13.0 \pm 2.4\%$ BW) axial load on the cane (Figure 4C). The axial compressive force increased during the first 20% of floor contact, plateaued as weight was shifted

onto the weaker lower extremity, and then decreased in the last 20% of contact time. The medio/lateral bending forces for these subjects were of similar magnitude as those for subjects CR2 and CR3.

The antero/posterior bending forces were exerted in a pattern similar to that described for the right crutch of subject CR1. The cane restrained forward motion in the first half of cane contact and provided a propulsive function in the second half.

Discussion

The temporospatial, kinematic and kinetic parameters present different characteristics depending on the subject and the type of AAD used. Walker users' cadence and walking speed were slow when compared to that of the other AAD users. Previous research indicated that when healthy subjects were asked to walk with a walker, their preferred walking speed

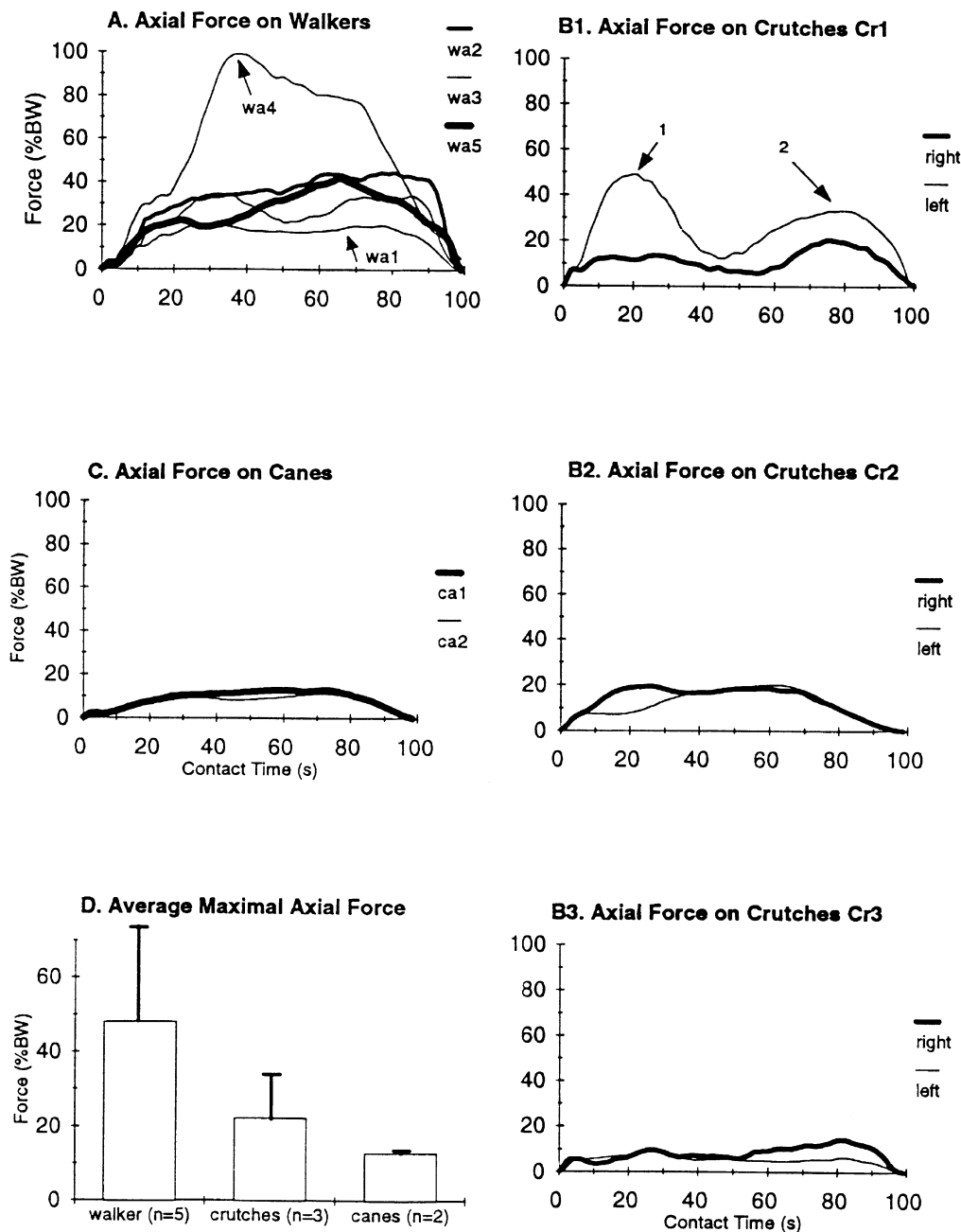
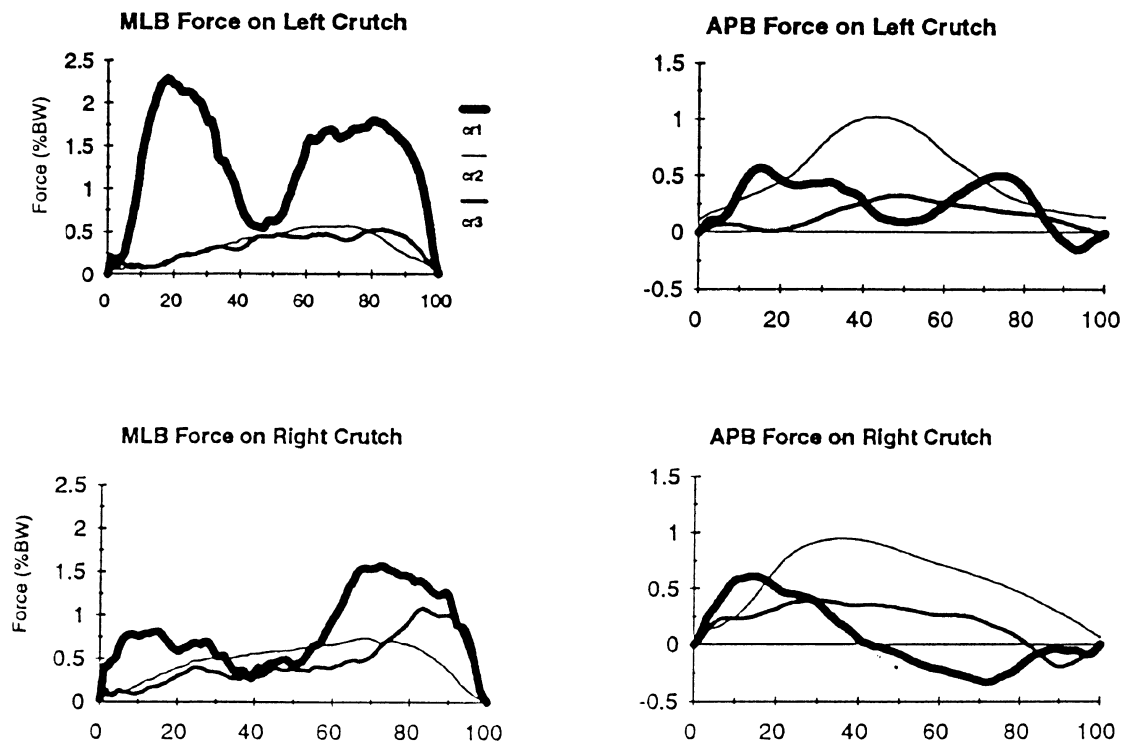


Figure 4 The axial compressive force (ACF) exerted by SCI subjects on walkers (A), crutches (B1, B2 and B3) and canes (C). Arrows 1 and 2 correspond to the peak ACF exerted on the left crutch when the lower extremities were advanced (B). All forces are averaged over five trials. Average ACF and standard deviation are depicted in (D)

decreased from 1.25 to 0.30 m s^{-1} since the gait cycle was interrupted when the walker is picked up and advanced.¹⁶ It should be kept in mind however, that persons with a disability may not be able to walk at all without the AAD and direct comparisons with healthy subjects should be made with caution. Weakness of leg muscles, spasticity, and decreased range of motion experienced by persons with incomplete spinal cord

injury, as in this study, may make it more difficult for subjects to advance their legs. The prolonged time spent in the stance phase may be indicative of the difficulty walker users experience in quickly unloading their leg and thus moving the leg forward. A decreased hip excursion and reduced step length further contribute to the slow gait pattern observed with these subjects.

A. Crutches



B. Canes

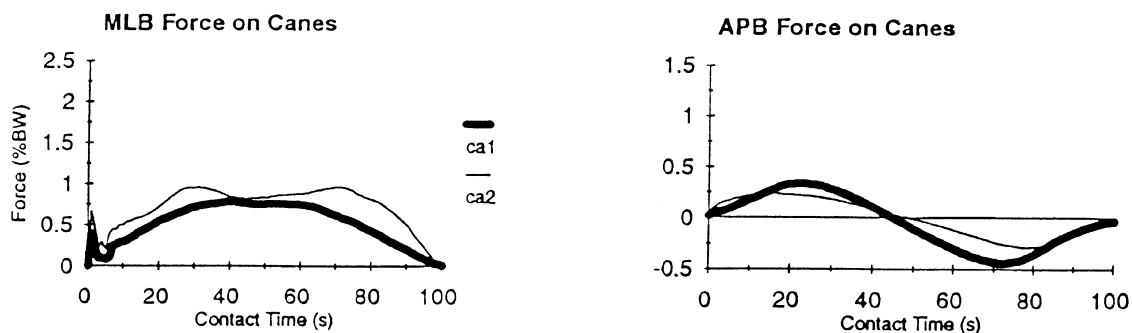


Figure 5 Medio/lateral bending (MLB) and antero/posterior bending (APB) forces exerted on crutches (A) and on canes (B)

The fact that the walker is advanced forward while the lower extremities remain in place, may result in a marked flexed trunk posture. Other authors have expressed concern that this flexion posture may effect adaptive tissue changes around the hip.¹⁷ Although hip extension was not limited for all walker users in this study, marked hip flexion was noted for at least one walker user and decreased hip excursion was noted for other subjects. When prescribing a walker, hip excursion should be assessed and, when necessary, gait-training strategies should be geared to increase hip range of motion during walking. This should be done during functional tasks, so that the hip is not merely exercised in isolation. Other assessments and therapies involving the knee and ankle should not be neglected.

The walker, being a stable device, allowed for a large force to be supported by the upper extremities (up to 100%BW). Although this may permit a person to be independent for locomotion (at extremely slow walking speeds), it also requires adequate upper extremity strength. The long-term effect of applying such a force through the arms has not been studied for walker users; however, studies on crutch users have shown that moments around the upper extremity joints are high.⁸ Since the upper extremities are not designed to bear weight, over-use injuries may occur and users should be assessed for early signs of injury. Prescription of AAD should take into account the magnitude of loading on the assistive devices and the projected duration of use.

It seems that although a high force can be supported by the walker, most walker users apply less than 100%BW of force on their device. The amount of force applied to the walker will depend on a number of factors, including the subject's balance, upper and lower extremity strength, degree and distribution of spasticity, and the ability to load and unload the legs.

Since the walker is firmly planted on the floor, AAD movement is greatly restricted in the M/L and A/P directions. Coupled with a slow walking speed, walkers can be considered very stable devices for those who do not have the balance and lower extremity strength to walk.

Walking with greater walking speed, cadence and step length is possible when using crutches. Crutches do not need to support a great amount of vertical force. In this study, the vertical forces exerted on crutches ranged from less than 10% to 50%BW. Axial compressive force values ranging from 20% to 70%BW have been reported.^{8,18,19} Since these forces are exerted by one upper extremity (as opposed to two when using a walker), a large force is transmitted through a shoulder joint that is mechanically not designed to bear weight.⁸ Especially with higher level SCI injuries where upper extremity strength is likely to be impaired, the potential for shoulder joint injury should be considered. In these subjects, when using a '4-point' gait pattern, the crutches are held in front and lateral to the user. This position allows the

crutches to act mostly as a restraining device that prevent subjects from falling forwards or sideways. Crutches can also be used to produce a propulsive force. The restraining and propulsive crutch functions produce a faster and smoother walking pattern.²⁰ Crutches allow the user to be more mobile since it allows him/her to ascend and descend stairs. They provide a restraining force to forward motion which has to be overcome in order to move forward. Overcoming the restraining force may place increased energy requirements on the body, especially when the large upper extremity contribution is considered.⁷ The cardiovascular fitness needs of the client should be addressed with long and short-term AAD users.

As compared with other AAD, cane users maintained a more upright trunk posture and had faster walking speeds, faster cadences and longer step lengths. The cane provided the least amount of vertical support; up to 15%BW unilaterally. Glendenning and collaborators²¹ reported that the cane becomes unstable when more than 20% to 25%BW is exerted on the device. While all loads in this study remained below 25%BW, but forces exerted on the AAD should be assessed when prescribing a cane. A simple method of approximating the vertical force on the cane is to have the client walk over a scale (weight scale) such that only the cane touches the scale. Even though this measure is not precise, the stability of the device should be evaluated if values over 25%BW are recorded.

Important functions of a cane are the provision of restraining and propulsive forces in the A/P direction as the user walks forward. This is similar to the fore/aft component of the ground reaction force vector acting on the foot as a subject walks over a force plate.²² The cane provides external forces in the same direction as forces generated by a normal, stronger lower extremity. Bennett and collaborators,¹ in their study on canes, concluded that some individuals used canes as restraining devices, while others used them as propulsive devices, depending on the needs and gait deficits of the users. Results reported in this study show both restraint and propulsion within the same stride.

Criteria for AAD progression have not been described in the literature and progression to date has been based on subjective assessments. In addition to adequate balance, postural control, and upper extremity strength, the axial compressive force may be an indicator for AAD progression. Although the current study has provided initial objective information to understand the influence of AAD on assisted-gait characteristics of individuals with incomplete SCI, biomechanical parameters for progression criteria need further investigation.

Conclusion

When prescribing AAD, one should be aware of the function the AAD fulfils. This should complement

the user's functional capabilities and the ability of the subject to adjust dynamically to changes in external support. A walker provides a large amount of vertical support but will affect trunk alignment. Perhaps other types of walkers will affect trunk alignment differently; such as walkers with wheels or a posterior walker.²³ Studies that compare different walker types are necessary to determine if other walkers would be better for persons with incomplete spinal cord injury.

Hip flexibility should be assessed and tissue adaptations should be anticipated and prevented for walker users. Upper extremity joint integrity, specifically the shoulder, should be evaluated for adequate strength when prescribing AAD, because substantial amounts of force are transmitted through the arm. Fitness assessments should be performed for any AAD user.

A systematic evaluation of the forces exerted on the AAD should be assessed in a larger group of AAD users undergoing gait training in order to determine specific progression criteria. Other factors to be evaluated include upper and lower extremity strength, balance, personal desire to progress and patient confidence.

Keeping the advantages and limitations of each of the devices in mind, the user's needs should be matched with the device that complements the user's physical capabilities. Proper matching, in combination with gait training strategies and appropriate AAD progression, should help to optimize client function and independence.

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