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Paraplegia

Reciprocal Aided Gait in Paraplegia

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Summary

A group of 9 male paraplegics, experienced in the use of walking aids for ambulation, were tested using an alternate four-point gait. Data were collected pertaining to the axial load transmitted through the crutches, the upper limb joint displacements and the moments about the elbow and shoulder joints during the period of contact of the walking aid with the ground. This gait was found to be slow with long periods of load transmission through the walking aids. Comparatively high values were calculated for the moments about the joints of the upper limb.
Key words: *Crutches; Gait; Paraplegia; Biomechanics.*

The need for early mobilisation of the paraplegic patient, in an attempt to minimise the post-traumatic complications of his condition, has been recognised throughout the literature. Hancock *et al.* (1980) found that there was a retardation of the loss of bone mass with such mobilisation, and several authors, including Guttmann (1973) and Bedbrook (1981, 1985) have stressed the importance of ambulatory activities in the prevention of the more common complications of the condition.

Walking with crutches is not an easy activity for the paraplegic. The energy costs of such locomotion are considerable (Gordon and Vanderwalse, 1956), there are numerous practical difficulties associated with the application of the necessary orthoses to support the subject during walking, and the individual may feel that, with improved wheelchair design and access provisions, there are few advantages in continuing to use crutches for ambulation. The eventual abandonment of crutches by paraplegics has been reported by Childs (1964) and Hussey and Stauffer (1973), who remarked that only 50% of paraplegics who had the capacity and training to walk could be expected to do so at a 'social' level.

The gait patterns which have been investigated and reported in much of the literature have been symmetrical gaits, characterised by a cyclic progression in

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which the crutches are moved forwards simultaneously, followed by an upward lift of the body, using the upper limbs for propulsion, with a forwards swing of the hips and lower limbs (Childs, 1964). Following foot contact with the ground the trunk is moved such that the body weight acts through the support base of the feet. The crutches are then moved once more.

Alternate four-point gait is a phasic pattern, featuring a reciprocal motion of one lower limb and the crutch held in the opposite hand. It appears a 'cosmetically acceptable' gait, in that the user seems to maintain a more or less constant speed throughout the entire cycle, as compared to the swing-through gait pattern in which greater fluctuations of the body velocity may be encountered. Some subjects describe a feeling of 'walking more normally' when using the alternate gait, and the fact that the ground reaction force is shared between the crutch and the contralateral foot will mean that lower absolute axial forces will be expected to be transmitted through the crutches in this gait than would be the case with a swing-through gait.

Alternate four-point gait has been described as the slowest and most difficult of the paraplegic gaits, only being achieved by accomplished walkers. It has the advantage of facilitating turning and manoeuvring in confined spaces (Bromley, 1985).

The purpose of this study was to investigate the kinetic and kinematic characteristics of the alternate four-point gait pattern in a small population of paraplegic subjects.

Materials and methods

Walking aids were instrumented with strain-gauge transducers, sensitive to loading related to the principal axes (Opila, 1985). Signals were amplified and transferred to a microcomputer (BBC B+) through an analogue-to-digital convertor. The computer incorporated an accurate clock device.

Tests were carried out on a flat walkway with a total of 9 subjects ambulating at their preferred speed using standard design elbow crutches fitted at their customary height.

Table I Clinical status of subjects

Subject	Age	Level of lesion	Duration	Orthosis
1	27	Incomplete T12	7	KAFO
2	50	Incomplete T12	1	AFO
3	35	Incomplete L1	20	AFO
4	33	Complete T11	2	KAFO
5	28	Incomplete C5	4	AFO
6	26	Incomplete L1	4	AFO
7	21	Complete L1/2	3	KAFO
8	23	Incomplete C6	5	AFO
9	22	Complete L2	4	KAFO

AFO: Ankle-foot orthosis
KAFO: Knee-ankle-foot orthosis

Subjects used orthoses appropriate to their clinical status (Table I). In all cases these devices were low profile moulded plastic orthoses. Ten-second periods of data collection were performed over a total of 5 tests for each subject. Analysis was carried out on data derived during the crutch/ground contact phase of each aided gait cycle.

Subjects were fitted with body markers over anatomical landmarks and were filmed simultaneously from both sides and from the front using crystal-locked motor driven cine cameras which were remotely operated in synchrony with the acquisition of kinetic data. The sampling frequency for film and load transducers was 25 Hz.

Following data acquisition, the kinematic files were processes using an image analysis system (Kontron Videoplan). The processed data realised three-dimensional spatial co-ordinates for the body and walking aid markers. These data were transferred to a mainframe computer and merged with the relevant kinetic data file and analysed using customised software and standard statistical packages (Minitab).

Data derived from the tests allowed calculation of temporal characteristics of the gait, the ground reaction force components acting on the walking aid during

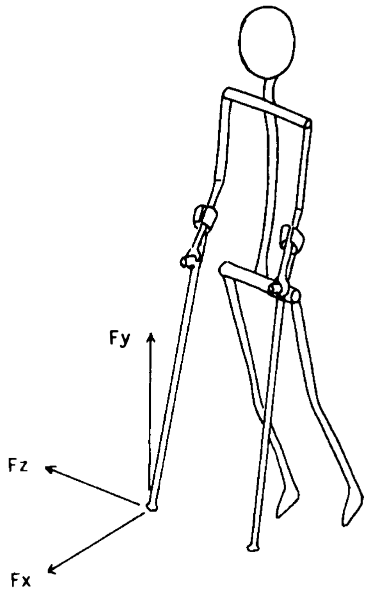

Figure 1 Ground reaction force components applied to walking aid tip.

the contact phase of the cycle (Fig. 1), the angular displacements of the upper limb joints, and the turning effects produced by the force components applied to the crutch tip with respect to the three-dimensional axes located at the elbow and shoulder joints.

The pattern of axial loading of the walking aids throughout the period of ground contact was studied. The load/time curves produced were analysed in conjunction with data related to contact time and displacement of the foot and the motion of the upper limb segments. These data were combined to permit analysis of the principal features of the reciprocal gait pattern.

The rotational effects of the applied loading at the upper limb joints were calculated. Specifically, joint moments were computed about the axes of flexion and extension at the elbow and shoulder joints. Also calculated were the moments of abduction and adduction about the shoulder joint.

The levels and characteristics of spinal cord lesion displayed by the subjects were variable (Table I) and thus produced some inconsistency regarding inter-subject analysis. Nevertheless, individual subjects displayed repeatable, consistent gait patterns and a number of general trends could be identified with respect to kinetic and kinematic parameters.

As a means of normalising the data, all forces applied to the walking aid, and turning effects produced about the joints of the upper limb were divided by the subject's body weight (in Newtons). Thus applied axial loading was expressed in dimensionless terms as $N.bw^{-1}$, while joint moments were expressed in $Nm.bw^{-1}$. Subjects were sufficiently similar in height (mean = 1·74m; SD = 0·06) for it to be deemed unnecessary to normalise data with respect to this parameter.

Results

The average velocity for the group of subjects was 0·47 $m.s.^{-1}$ (SD = 0·27), with a frequency for each complete cycle of 0·41 Hz (SD = 0·12). The sequence of crutch and foot motion, expressed as a ratio of the total cycle, is illustrated in Figure 2.

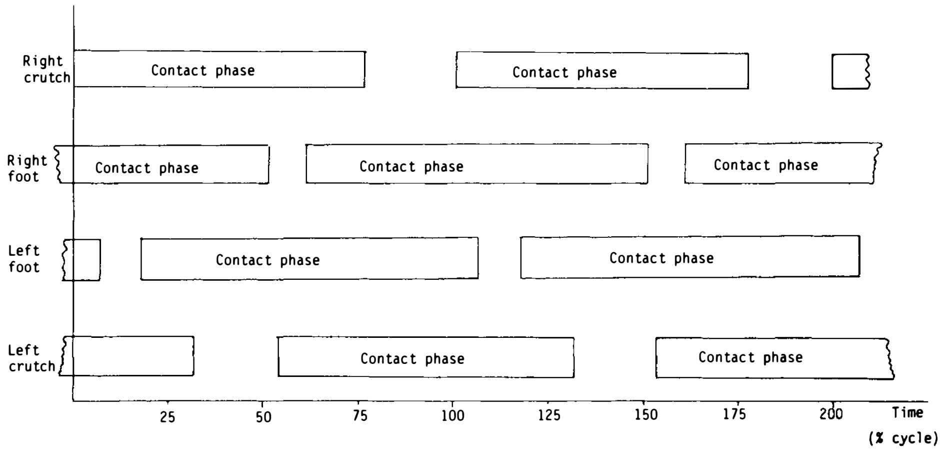

Figure 2 Temporal parameters of alternate four-point gait with respect to the total gait cycle.

For the purposes of this study, the cycle was considered to begin and end with contact of the right side crutch with the ground. The cycle was composed of a support phase and a swing phase, the relative timing of which was repeatable for each individual subject. The integration of foot and crutch motion is illustrated in Figure 2 and values are presented for the support and swing phase temporal relationships (Table II). The temporal events are generally discrete, with no significant difference between the duration of crutch/ground contact on one side compared to the other ($p > 0·2$). Duration of contact between the crutch and the

Table II Temporal relationships of foot/aid ground contact

	Initiation of support cycle time (s)	End of support cycle time (s)
Right crutch	0	0·77 ± 0·06
Right foot	0·6 ± 0·065	0·5 ± 0·08
Left foot	0·17 ± 0·01	0·06 ± 0·06
Left crutch	0·5 ± 0·06	0·32 ± 0·05

ground averaged 2·06 seconds (SD = 0·64) per cycle. It was observed that some subjects initiated foot swing phase as the contralateral walking aid contacted the ground, or even before this. Other subjects were more cautious in their gait and did not terminate stance until the opposite side crutch had been positioned in a secure manner.

The pattern of axial loading was derived for each subject and was found to be highly repeatable for each individual. Some inter-subject variability was noted, however, particularly with regard to the magnitude of loading detected (Table

Table III Peak axial load values recorded during aid/ground contact phase

Subject	Right side peak load ($N.bw^{-1}$)	Left side peak load ($N.bw^{-1}$)
1	0·505	0·497
2	0·207	0·215
3	0·358	0·379
4	0·419	0·400
5	0·259	0·210
6	0·237	0·206
7	0·218	0·343
8	0·482	0·466
9	0·399	0·381

III). Superimposition of load curves suggested that there were similarities between individuals in the kinetic characteristics of the gait pattern (Fig. 3).

Normalising the curve shape as a function of the maximum amplitude indicated an idealised load pattern (Fig. 4). The typical curve shape, with an initial peak value corresponding in time to motion of the contralateral foot, and a second peak at about the time of the ipsilateral foot 'toe off', was observed in most individuals. The load sharing by the two feet when one crutch was moving minimised the loading on the opposite side crutch, which retained contact with the ground.

The pattern of motion of the upper limbs and trunk throughout the gait cycle varied to some extent with individual morphology, habit and physical limitations imposed by the clinical condition. The general features of the elbow and shoulder joint motion throughout the contact phase are illustrated (Fig. 5). This suggests that during the contralateral limb swing phase, with the occurrence of the first peak axial load, the subject positioned the elbow in a minimum of flexion with the

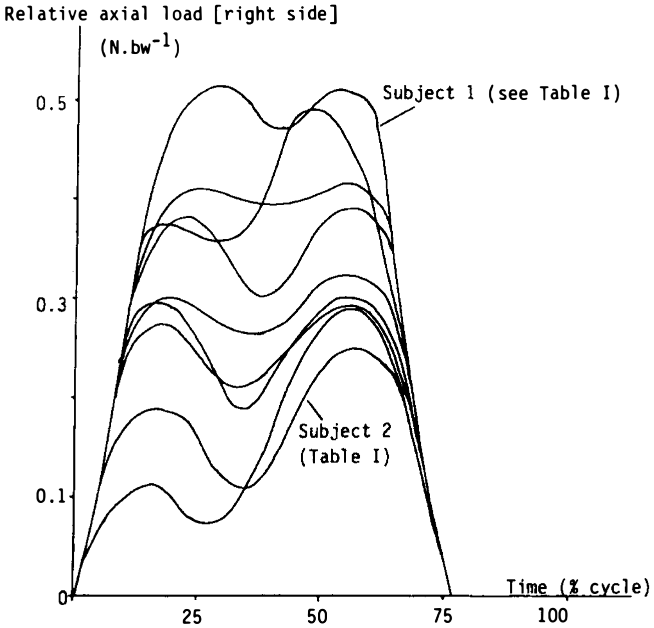

Figure 3 Superimposed axial loading on right side walking aid during support phase.

shoulder flexed forwards to the maximum. The upper limb position would have been dictated by such parameters as step length, walking aid height and upper limb length.

As the cycle progressed towards double foot contact and movement of the opposite aid, the elbow flexed and the shoulder extended, maintaining a position of slight abduction (Fig. 5). At the time of motion of the ipsilateral foot the axial load increased, the elbow joint reached a maximum angle of flexion and the shoulder attained maximum extension and abduction. This position, representing the extreme of motion in these directions, was then reversed as the axial load decreased and the subject prepared to move the crutch forwards in advance of the next contact phase.

The pattern of moments derived for these joints reflected the changing axial load value and the positional changes of the upper limb through the contact phase (Fig. 6). It was noted that the period of motion of the contralateral foot created significantly lower values for the moment imposed on the joints compared to that detected during motion of the ipsilateral foot ($p < 0{\cdot}001$).

The magnitudes of the imposed joint moments were related to the magnitudes of the axial loads and thus varied from subject to subject. Peak flexion moments of between 0·01 and 0·03 $Nm.Bw^{-1}$ were calculated for the elbow joint, with abduction moments at the shoulder of between 0·02 and 0·06 $Nm.bw^{-1}$ and extension moments at the shoulder of between 0·01 and 0·06 $Nm.bw^{-1}$.

The displacement of the load vector from the assumed centres of rotation of the elbow and shoulder joints was calculated. This was only performed where the moments were unidirectional throughout the contact phase. It was found that the mean displacement throughout the loading phase was 52 mm (SD = 24) anterior

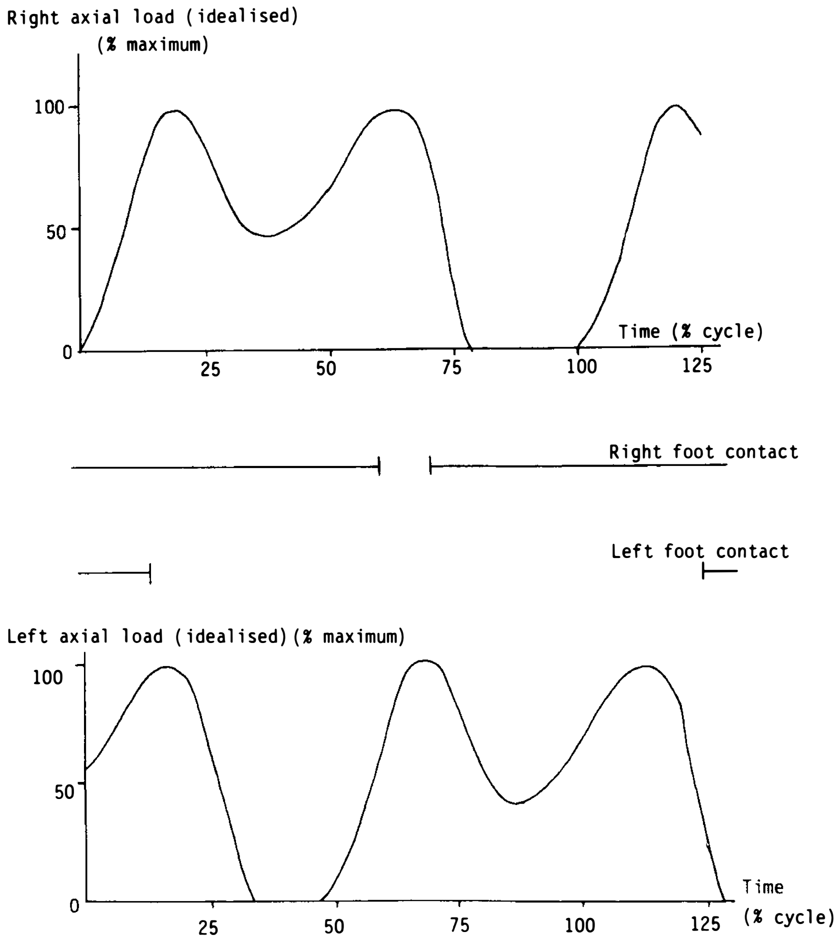

Figure 4 Axial loading patterns on walking aids.

to the elbow and 112 mm (SD = 67) lateral to the shoulder joint.

Discussion

Subjects using an alternate four-point gait pattern are normally those with low spinal level or partial (i.e. incomplete) cord lesions. The patchy distribution of motor activity which can be characteristic of the latter group of subjects renders them particularly susceptible to asymmetries and inconsistencies within the aided gait pattern. Nevertheless, there would appear to be certain features of the reciprocal gait pattern which were found in all subjects tested and which were defined by the general characteristics of an alternate four-point gait.

The gait was comparatively slow, with relatively long durations of crutch-ground contact. In subjects using ankle-foot orthoses only, abnormalities of motor control led to an exaggerated foot lift. Heel strike was generally absent, and the foot made contact with the ground either in a plantigrade position, or with the forefoot

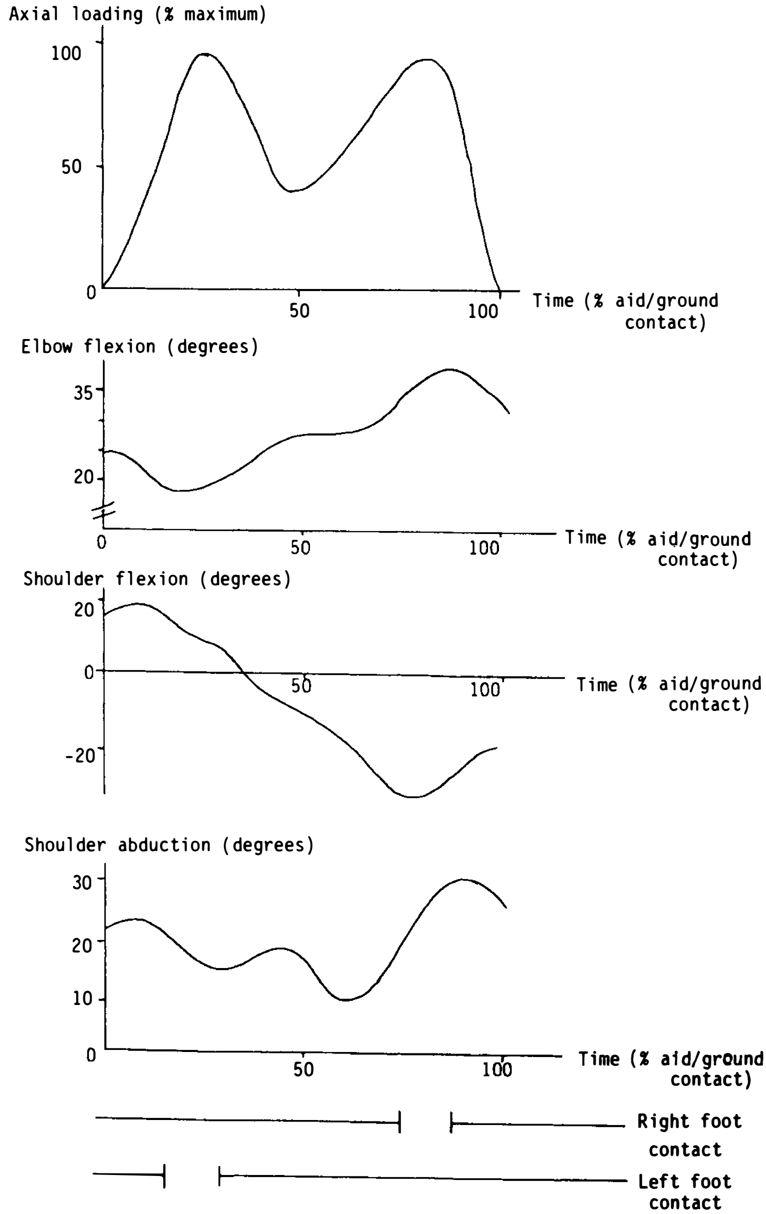

Figure 5 Right side upper limb joint displacements during loading of walking aids.

making first contact. Subjects using knee-ankle-foot orthoses had no free knee motion and advanced the lower limb using pelvic rotation about the vertical and sagittal axes.

Although cyclic in its nature, there was a decomposition of the normal unaided gait in that the upper and lower limb motions were not synchronously reciprocal, but were out of phase, with the upper limbs leading the lower limbs. This strategy preserved lateral stability in walking, since the base of support was of a

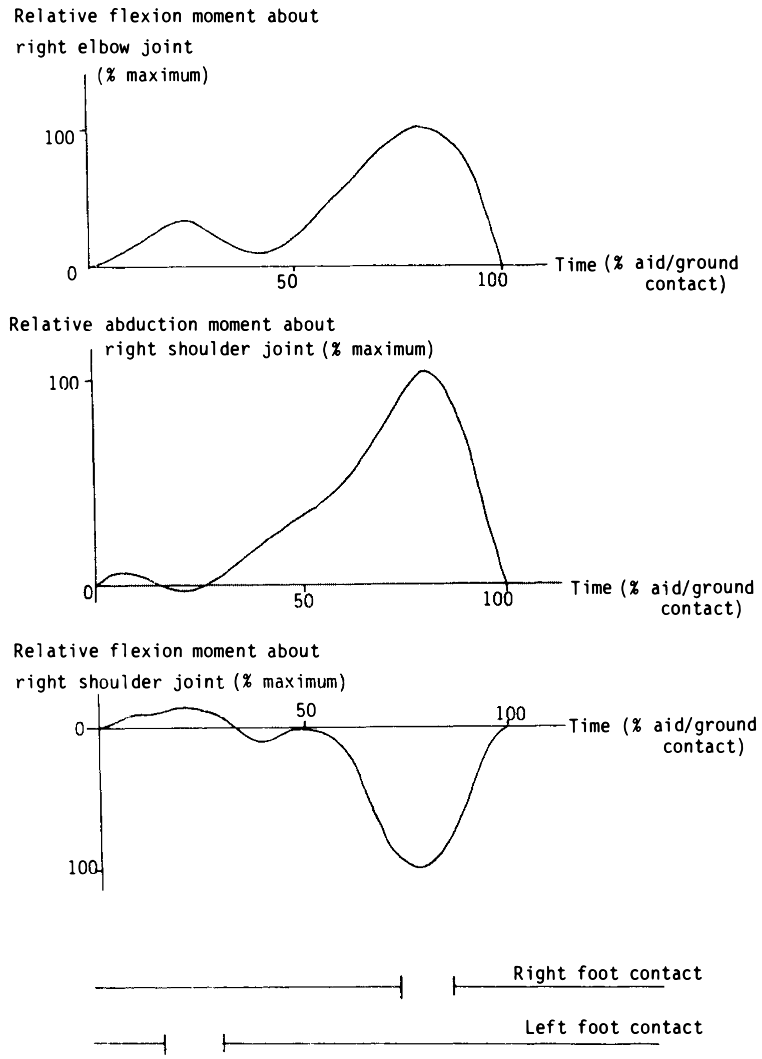

Figure 6 Right side upper limb joint moments during loading phase.

comparatively large area due to the preservation of three points of ground contact. In some individuals it was noted that the foot swing phase as initiated in advance of contact of the contralateral aid with the ground. These individuals (i.e. subjects 3 and 6) had suffered low level incomplete lesions of the spinal cord.

It might have been expected that the slow forward speed would have produced more marked lateral displacement of the body mass, such as occurs in unaided gait at reduced speeds (DuCroquet *et al.*, 1968). The reduced motor activity in the abdominal and lower trunk muscles created the potential for a marked degree of pelvic instability in the frontal plane during lateral transference of the subject's body mass from one foot to the other. These factors might have been expected to exaggerate the subject's transverse instability of the pelvis and produce a markedly 'waddling' gait.

Inspection of the data suggested that during single leg stance, both crutches were loaded to a similar extent. This would indicate a more or less central orientation of the body's mass centre. The subjects would seem to have compensated for the lateral instability with a bilateral increase in axial loading of the walking aids during the period of one-legged stance.

There was a considerable spread in the magnitude of axial loading imposed on the crutches (Table III), with peak values of axial load ranging from 15% of body weight to 50% of body weight (Fig. 3). This was consistent with the individual's requirements for mechanical support and impulse during the gait cycle. Subjects with relatively good pelvic girdle and lower limb muscle function loaded the crutches to a lesser extent than did those with more extensive motor loss. The extent of the neurological deficit is clearly a major factor in the determination of the aided gait characteristics. Detailed neurofunctional profiles were not available for each subject, but all participants in the study were at least capable of independent ambulation and were able to rise to stand from sitting without the assistance of any other person. This would therefore confine the study to subjects whose neurological deficit was relatively limited.

Paraplegics, as a group, tend to favour crutches adjusted to a length greater than that conventionally prescribed. This facilitates foot clearance of the ground by allowing a greater upwards displacement of the body's mass centre during the foot-swing phase of the aided gait cycle using a swing-through pattern. When used in the alternate gait, these longer than usual crutches may encumber free ambulation.

In the group of subjects tested, all used the walking aids in swing-through and alternate four-point gaits. The choice of gait was determined by the particular circumstances under which the subject ambulated. No adjustment was made to the preferred length of the walking aid. Consequently, crutches were often held away from the body and close to a vertical position in the frontal plane, with the shoulder joint abducted and internally rotated. Such a posture would tend to increase the displacement of the ground reaction vector from the sagittal axis of the shoulder joint, with a consequent increase in the abduction moment. Such a finding was, in fact, a feature of this gait.

The typical pattern of walking aid and upper limb motion detected during analysis of this gait involved placement of the tip of the walking aid well anterior to the feet, progressive elbow flexion throughout the period of crutch contact and extension and abduction of the shoulder joint. The varying forwards and lateral displacements of the body during the gait cycle meant that the line of the axial load fell outwith the mechanical centres of rotation of the joints of the upper limb, in some cases to a considerable extent. The elbow joint was normally displaced into flexion under the load effect, while the shoulder was abducted. Peak moment values about the shoulder of up to 0·06 $Nm.bw^{-1}$ were calculated during the gait cycle, with up to 0·03 $Nm.bw^{-1}$ calculated about the elbow joint. When this was compared to standard data for lower limb joints during normal gait it became apparent that the joints of the upper limb would have been subjected to considerable intersegmental loading. According to data provided by Winter (1987) the peak moment at a free knee joint during unaided gait can be expected to reach slightly over 0·06 $Nm.bw^{-1}$.

It was likely that such applied moments would demand isokinetic and isometric responses from the relevant periarticular muscles. These would probably be the Triceps brachii and the Latissimus dorsi. The former is the only antagonist to elbow flexion and would work concentrically and eccentrically as appropriate, depending upon the nature of motion occurring at the elbow joint while the applied moment was active.

The shoulder joint moved into an extended position during the period of crutch contact with the ground, while at the same time being abducted by the effects of the ground reaction forces. The Latissimus dorsi muscle is known to be active in extension and adduction of the arm, particularly in an orientation close to the anatomical position. Such a position was typically found during the crutch contact phase of this gait pattern.

Reduction of the loading demands on the upper limb segments during this gait would be a desirable objective in the rehabilitation of paraplegic gait. It would appear from this study that some of these demands might be reduced by altering the length of the walking aid, thus permitting a closer approximation of the load vector and the joint centre. This would be particularly advantageous during motion of the ipsilateral foot. A consequence of such an intervention, however, might be to decrease step length and so further to reduce gait velocity. There may be some possible design alterations which would permit a range of walking aids to be available to the paraplegic subject. The design of walking aid selected for use could more accurately reflect the ambulatory requirements of the user.

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