

Dynamics of reciprocal gait of adult paraplegics using the ParaWalker (Hip Guidance Orthosis)

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Abstract

Force plate studies with nine adult traumatic complete paraplegics with lesion at thoracic level have shown that, unlike children walking in the Oswestry Hip Guidance Orthosis (ParaWalker) system, they have to apply extra stabilizing forces through the crutches for safe walking. These forces are described and are attributed to an increased crutch force applied to prevent lateral deformation of the orthosis in stance phase.

Introduction

Clinicians and engineers involved in the rehabilitation of paraplegics with thoracic level lesion have long wished to achieve an effective form of ambulation for this heavily handicapped group. Swing through gait using crutches with legs braced in long leg calipers (knee-ankle-foot or hip-knee-ankle-foot orthoses) has been used for many years. However, this method incurs the penalty of high energy consumption. Other devices such as the Swivel Walker (Edbrooke, 1970; Rose & Henshaw, 1972; Stallard et al, 1978) and the Parapodium (Motlock & Elliott, 1966) permit low energy ambulation but are limited to flat, smooth surfaces.

Rose (1979) outlined the principles of the hip guidance orthosis, which was initially designed for children suffering from congenital paraplegia. Over the years design modifications have been made based mainly on clinical experience with these children (Stallard et al, 1986). Lately, with the help of improved understanding of the concept, adult paraplegics have been fitted with the orthosis. Twenty adult

traumatic paraplegics have been supplied with the ParaWalker (the name given to the device in routine clinical supply) since 1981. Seventeen of these are still actively using the orthosis. Observation of this group showed that none of them performed as well as junior patients in that they were less confident and ambulation endurance was often limited by pain in the hands. A biomechanical study of their performance was undertaken to see if any reason for this difference could be identified.

Subjects, method and equipment

Subjects were nine adult traumatic paraplegics with ages ranging from 22 to 33 years and lesions at thoracic level between T4 and T9. All were experienced ParaWalker users and all used elbow crutches.

Each subject was asked to ambulate along the 20' (6.1m) walkway in the ORLAU Gait Laboratory. A dynamic analysis of their gait was obtained using video recordings from two orthogonally placed cameras and monitoring the outputs of a six channel force platform (Kistler 9261B) mounted in the centre of the walkway. The force platform data was sampled and analysed using a Commodore 4032 Computer interfaced via a 12 bit, 8 channel A/D convertor sampling at 66.67 Hz and presented in a graphical form on a Hewlett Packard plotter.

To enable correlation of the recordings of the events in the gait cycle with the graphical force output, a Real Time Video Vector System was used. It produces a force vector superimposed instantaneously on a television picture of the subject by electronically processing the video signal (Tait & Rose, 1979). The resulting composite signal can be recorded and played back and also contains a display of the video field number.

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An inherent limitation of the force platform is that data relating to a specific component or limb segment is invalid if more than one support is in contact with the measuring surface at any one time, or if the support area overlaps the force platform boundary. Hence, using a single force plate, the complete picture of ground reaction forces from two limbs and two crutches requires a minimum of four walks. A walk was considered representative when the subject achieved this without visually detectable hesitation, and with only the limb or crutch in question striking the platform fully. To obtain a complete set of data to represent one gait cycle, subjects were asked to perform a number of walks along the walkway and force platform readings along with video recordings were taken. From these video recordings average time intervals between the events in the gait cycle were determined for each patient. This allowed the average foot and crutch contact times to be established and then the force platform outputs most closely matching these times were selected and temporally adjusted to exactly fit the time base. This enabled a composite graph to be produced of all ground reaction forces of an average gait cycle for each patient. The graphical results from all nine patients were similar in form and an idealised representation of this is shown in Figure 1.

To ensure acceptable accuracy the total impulse, represented by the summed area under the graphs of vertical force against time for each leg and each crutch during the gait cycle, was measured with a video position

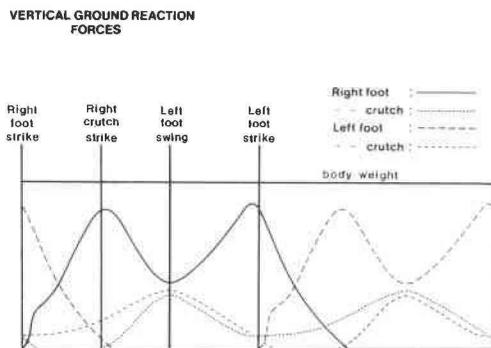


Fig. 1. General form of the graphical results of vertical ground reaction forces against time for adult ParaWalker users. This is an idealized diagram constructed from the general form of all 9 users (see text).

analyser and compared with the product of body weight and duration of the gait cycle. In all patients the difference between the two measurements was less than 5%. The range of forces obtained is presented in the following description.

Description of forces in the gait cycle

Convention

In the following description the forces are those applied by the orthosis or crutch to the ground, not ground reaction forces.

Vertical forces

Phase 1: Right heel strike

When the (R) heel strikes the floor the (L) foot force is beginning to decline after reaching a peak and is approximately 0.8–0.95 BW, and the (L) crutch loading is at a minimum of about 0.075–0.15 BW. The (R) crutch force is very low and is rapidly decreasing as (R) crutch swing approaches.

Phase 2: Right crutch strike

Commencement of body movement to right

At the time of (R) crutch strike the loading on the (R) leg is peaking at approximately 0.83–0.9 BW, and the (L) crutch loading is beginning to increase in order to get the (L) foot off the ground, the load on the (L) crutch reaches approximately 0.28–0.4 BW, and at the same time the load on (R) crutch is also seen to be increasing. It may reach up to 0.28–0.35 BW.

Phase 3: Left early to mid swing and lateral tilt to right

At the beginning of the swing phase vertical loading on both crutches is peaking at approximately 0.32–0.4 BW, and the (R) foot force is at its mid-stance lowest between 0.29–0.47 BW. As the swing phase progresses the crutches begin to unload and at the same time the (R) foot force begins to increase.

Phase 4: Left heel strike

(R) foot force is beginning to decline having reached its peak between 0.8–0.98 BW, and the (R) crutch force is at a minimum of about 0.11–0.18 BW. The (L) crutch force is rapidly decreasing as it approaches (L) crutch swing.

The process continues in the similar fashion with the left leg on the floor and the body moving towards the left until the right heel strikes again.

Horizontal forces

Depending upon the walking style, at the time of foot strike there may be a force directed either forwards (anteriorly) or backward (posteriorly). The reason for this is that some subjects reach their limit of forward leg movement and then allow the foot to swing back and down producing a peak in the posterior direction, whilst the remaining subjects produce an anterior peak because they do not reach their limit of forward movement but bring the foot down whilst it is in forward motion. During the majority of the stance period the horizontal force is predominantly posterior.

The crutch forces in the horizontal direction follow a symmetrical pattern. During the first period the force is anteriorly directed changing to a posterior direction for the second peak producing the forward propulsion.

Transverse forces

Transverse crutch forces do not show medial directional forces at any time during the gait cycle. They are directed laterally throughout the crutch contact.

There is some medially directed force from each foot at the time of footstrike.

Discussion

Dynamic analysis of the force patterns of adult ParaWalker patients has enabled the understanding of the principle of ParaWalker walking to be further advanced. The previous dynamic study of this orthosis (Major et al. 1981) used a subject aged eight years and demonstrated that the crutch vertical force pattern showed only one peak during the later period of crutch contact, whereas the present study with adult paraplegics shows two.

The authors believe that with any large orthosis it becomes difficult to maintain the same relative rigidity as is achieved in smaller devices since bending deformation is proportional to the cube of the length of individual components. With adult patients this deformation may reach such proportions that the swing leg fails to clear the ground. Pushing harder on the crutch has the effect of increasing lateral forces and causing still more deformation in the orthosis particularly in the region of the hip joint (Figure 2b). This can eventually lead to a loss of extrinsic stability

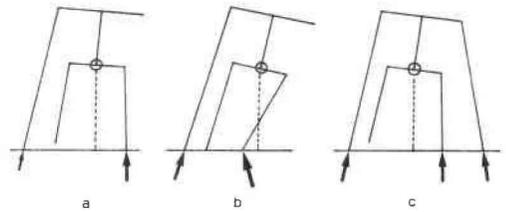


Fig. 2. Diagram showing various modes of ParaWalker crutch use (after Rose 1979). Deformation of the structure is exaggerated for clarity.

- a) Ideal situation with a stiff structure enabling a single crutch force to produce foot clearance.
- b) Low stiffness in the orthosis allows the stance hip to adduct producing instability as the centre of mass passes outside the support area.
- c) Introduction of a force through the remaining crutch restores stability and also reduces loading on the stance foot hence relieving the adducting moment at the hip and restoring more normal geometry.

(Rose 1979). To overcome this problem the patient is obliged to apply large forces through both crutches. Hence two peaks in the vertical crutch force graph occur (Figure 1), the second being the essential force required for normal ParaWalker walking to tilt the body over and clear the swing leg, the first being the force introduced to compensate for instability due to the lack of lateral stiffness in the orthosis (Figure 2c). The forces described here where two peak forces are seen from each crutch and the previous description of forces in which only one peak is seen from each crutch represent two ends of the spectrum of performance of the ParaWalker users. Performance will vary between these two extremes as the relative lateral stiffness of the orthosis changes. Although in theory the stiffness could be increased by the use of larger sections or stiffer materials the scope for improvement in this area is limited since space for the lower hip joint member is restricted as patients need to be able to sit in their wheelchairs and it is difficult to find a material stiffer than steel with suitable failure characteristics.

Consequently alternative methods of compensating for the lack of structural rigidity by means of a hybrid system incorporating both a mechanical orthosis and electrical stimulation of selected paralysed muscles now needs to be investigated. Initial exploratory trials with such a system (Patrick & McClelland, 1985) show promising potential which is worthy of further investigation. The reported study of the

dynamics of the mechanical system on the adult patients will provide a useful basis of comparison for trials in Centres undertaking such work.

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