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Biomechanics of the hip disarticulation prosthesis[1]

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Introduction

More than twenty years ago, McLaurin (1954) published his first report on the Canadian hip disarticulation prosthesis. His work was aimed at improving the conventional design of the saucer-type and tilting table prostheses.

A measure of his success can be seen in that the Canadian-type has made the other two almost obsolete. Since its introduction, the prosthesis has changed very little. Modifications have been made to the socket and hip joint, but the principles have remained the same.

Two papers have been published to date which deal with the biomechanics of the prosthesis, Radcliffe (1957) and McLaurin (1969). Although both authors gave an indication of force directions, it was not possible for either to indicate magnitudes since no experimental results were available at that time.

The test reported here provides quantitative information on the variation with time of hip, and knee moments in both antero/posterior and medio/lateral planes, axial torque and axial force in the shank during the stance phase.

There are two main features in McLaurin's prosthesis compared to the other types:

(a) the design and position of the hip joint,
(b) the socket design.

The hip joint is a broad hinge placed well forward and below the anatomical hip joint. The joint allows free rotation in the antero/posterior plane, i.e. flexion/extension. It is rigidly fixed in the medio/lateral plane thus giving lateral stability.

The socket is a bucket-type seat which fits snugly around the pelvis and thus minimizes stump/socket movement. The suspension is provided by moulding over the iliac crests.

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Other important aspects of the prosthesis are:

(a) the knee joint is set posterior to the hip joint in such a manner that if a line is drawn through the hip and knee centres it will strike the ground about 20 mm behind the heel, this is to ensure knee stability especially at heel strike,

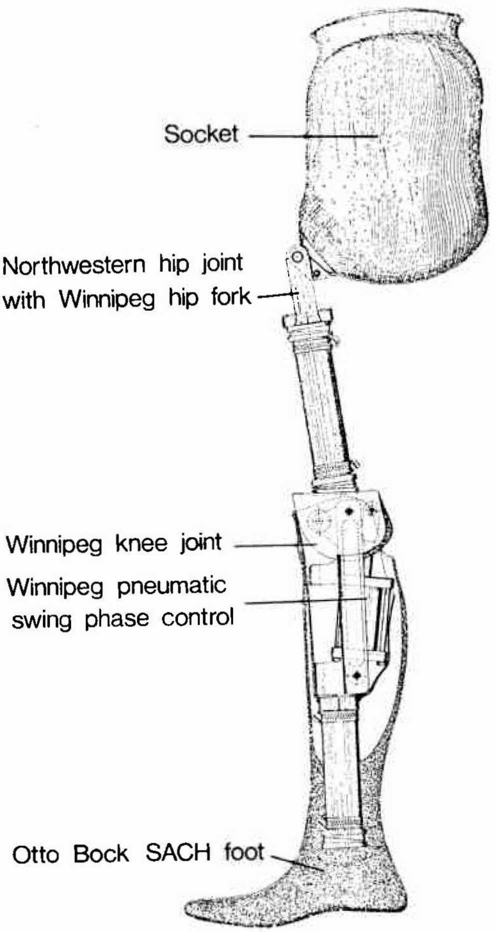

Fig. 1. The Winnipeg Modular Hip Disarticulation Prosthesis.

[1]Previously published in Orthopadie Technik, 4/76, 58–60.

(b) excessive hip flexion is prevented by a limiting device,

(c) hip joint hyper-extension is prevented by means of a "hip bumper" or extension stop secured to the bottom of the socket.

Subject and Prosthesis

The subject used in the test was a 21 years old active female amputee of mass 51 kg approximately and with a height of 1.65 m. Amputation had been performed on the right side for pathological reasons and she had been wearing a hip disarticulation prosthesis for two years. The prosthesis used was the Winnipeg Modular Hip Disarticulation Prosthesis (Fig. 1) incorporating a Northwestern hip joint and uniaxial knee joint fitted with pneumatic swing phase control. The flexion limiter consisted of a leather strap and buckle attached between the hip fork and the seat of the socket. An Otto Bock SACH foot was used and no cosmesis was fitted. Socket manufacture and alignment were as recommended by McLaurin and Hampton (1962).

Equipment and Procedure

A strain gauged pylon dynamometer and a knee goniometer, as described by Lowe (1969), were incorporated in the shank of the prosthesis. Following dynamic alignment the patient was asked to walk about for about half an hour in order to accustom herself to the limb. The types of activity investigated were level walking at various speeds and walking up and down a ramp (1:7 gradient). Analysis of the results was partly manual and partly with the aid of a PDP 12 computer. The transducer raw data were sampled at a frequency of 10 Hz.

Results and Discussion

Using the information obtained from the dynamometer, force analysis allowed the calculation of:

(a) antero/posterior (A/P) moments at hip, knee and ankle,

(b) medio/lateral (M/L) moments at hip, knee and ankle,

(c) torque about the long axis of the leg,

(d) axial load along the axis of the pylon dynamometer,

(e) A/P and M/L shears at right angles to the pylon axis,

(f) resultant force.

The results for A/P knee moment (level, up-ramp and down-ramp walking), A/P and M/L hip moment, axial force and torque (level walking only) during the stance phase are shown graphically. Six cycles for level walking and three cycles for up-ramp and down-ramp walking were analysed.

The results display considerable scatter which cannot be wholly attributed to the dynamometer or associated instrumentation and must therefore be due to uncontrollable parameters related to the patient, such as effects of change of speed. The first and last 5 per cent of the stance phase cannot be considered as accurate because the sampling frequency used was too low for these areas.

Axial force. This displays a double peak pattern (Fig. 2). There is, however, a much slower increase in value of the axial force when compared with the results obtained with BK and AK amputees (U.C.B. 1947). The peaks occur later in the stance phase, are closer together and less pronounced. The maximum value is approximately 650 N.

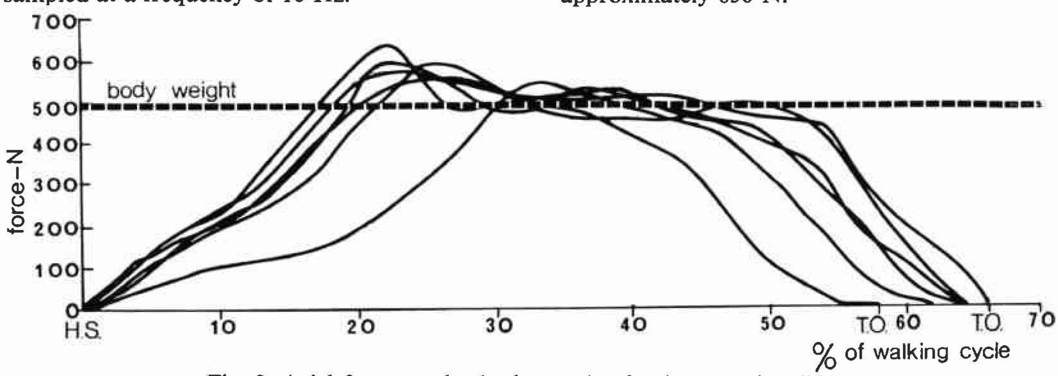

Fig. 2. Axial force on shank along axis of pylon. Level walking.

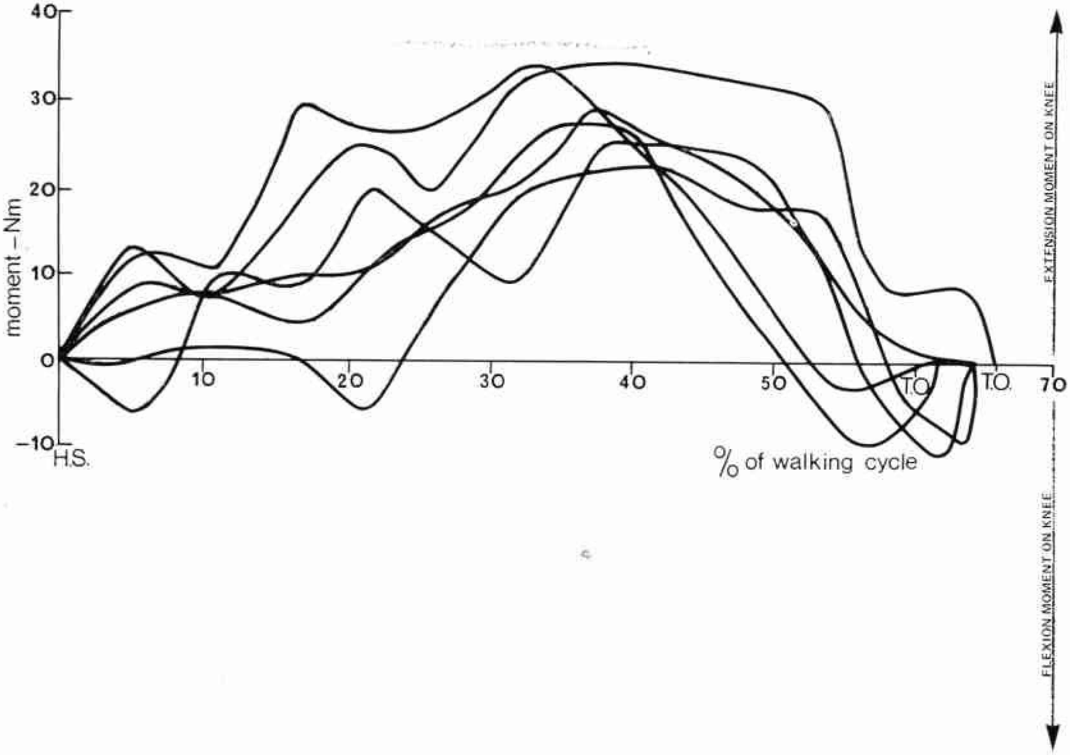

Fig. 3. A/P knee moment. Level walking.

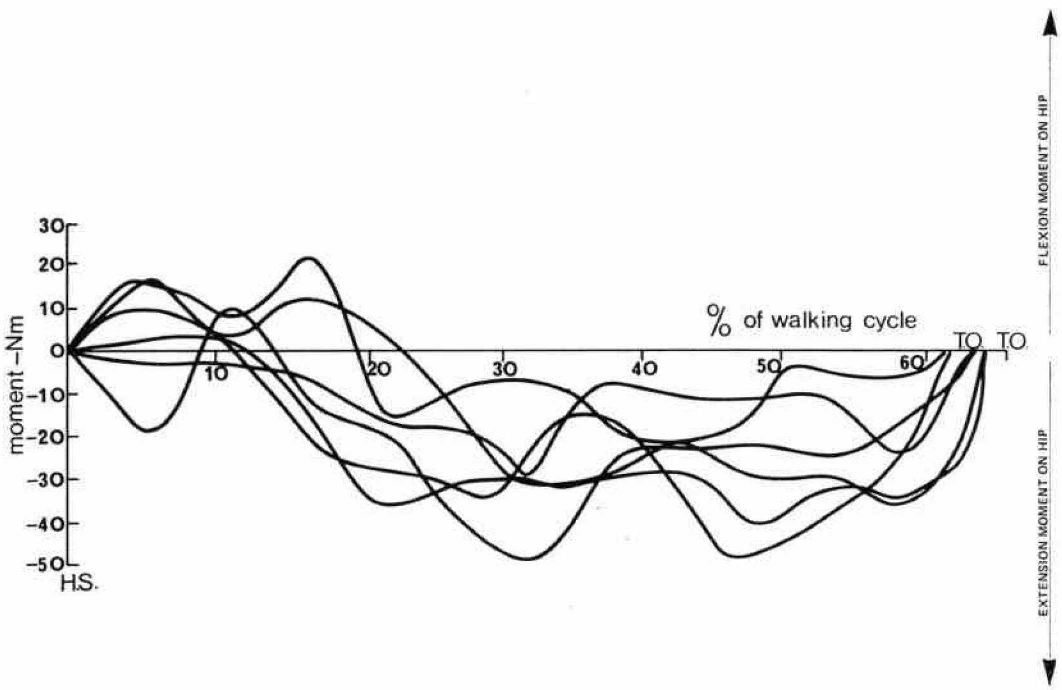

Fig. 4. A/P hip moment. Level walking.

A/P knee moment (level walking). Knee extension moment is low up to about 30 per cent of the cycle (Fig. 3). This is due partly to the slow increase in axial load and partly to the backward A/P shear which exists in the first half of the stance phase. The maximum knee extension moment is about 35 Nm. The flexion moments recorded at the beginning and end of the stance phase can be absorbed by the restraining action of the swing phase control unit.

A/P hip moment (level walking). The graph (Fig. 4) indicates that following heel strike there is a flexion moment which is due to the resistance of the hip flexion limiter. Between 10 and 20 per cent of the cycle the direction of the moment reverses, indicating that the hip extension rubber stop comes into action in this region of the cycle and remains in contact to the end of the stance phase. The peak extension moment is about 50 Nm. Prior to toe off the patient initiates knee flexion preparatory to the swing phase as the line of action of the resultant force between stump and socket passes progressively posteriorly.

M/L hip moment (level walking). The shape of this curve is as expected (Fig. 5) and is similar to that of a normal subject. The moment is predominantly adducting the hip joint with a maximum value of approximately 55 Nm.

A/P knee moment (walking up-ramp and down-ramp). As expected when walking up-ramp the knee is stable whereas down-ramp it is unstable (Figs. 6 and 7). This is due to the geometry of the limb configuration in relation to the force actions. The maximum values are 80 Nm for up-ramp and 10 Nm for down-ramp walking. It was also clear during the tests that the amputee had little difficulty in walking up-ramp but great difficulty walking down-ramp.

Torque (level walking). The torque curves (Fig. 8) show a pattern remarkably similar to that obtained for normal locomotion with maximum value of the order of 8 Nm.

Conclusions

The wide scatter of the results, however, makes it difficult to form precise judgements concerning assessment of amputee performance. Before this can be studied in greater depth, the reasons for the scatter must be determined, and attempts made to eliminate the scatter as far as possible. Thereafter, by acquiring a sufficient volume of data, a realistic statistical assessment could be made. However, the information presented in this paper is of value to the designer of prosthetic devices as it provides information on the nature and approximate value of the load actions.

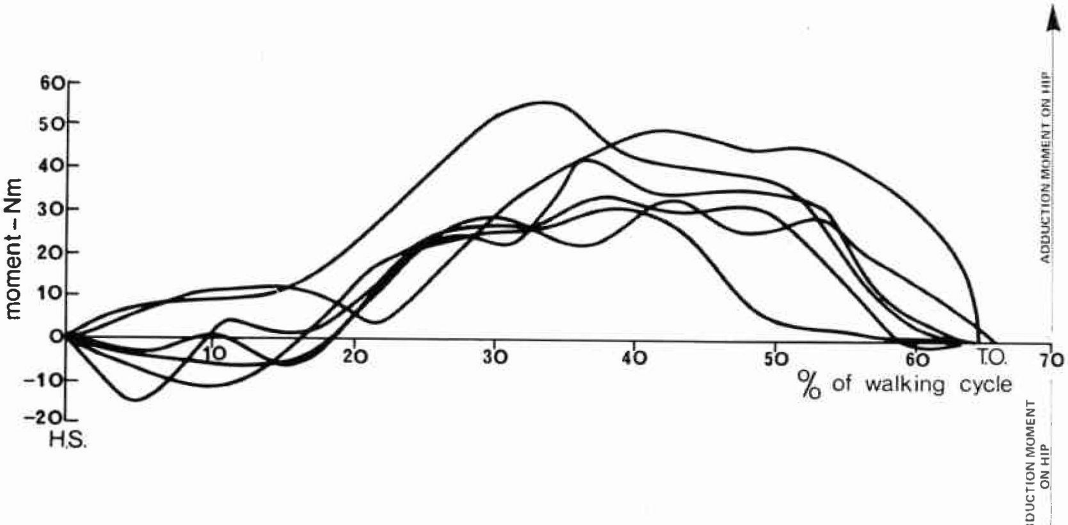

Fig. 5. M/L hip moment. Level walking.

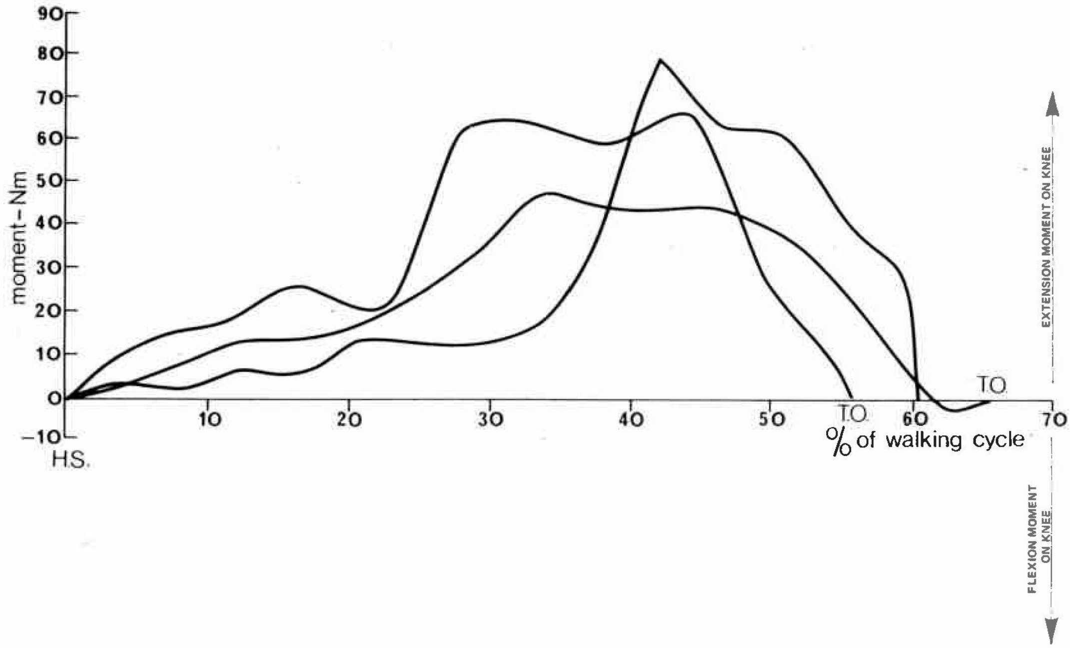

Fig. 6. A/P knee moment. Walking up ramp.

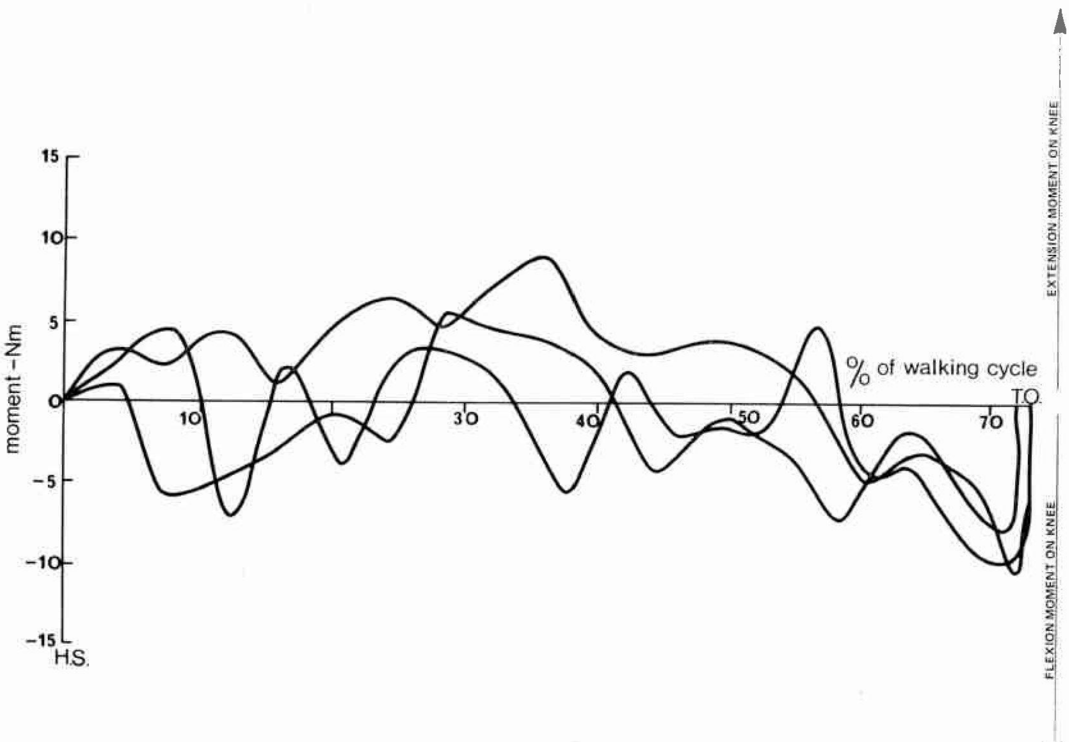

Fig. 7. A/P knee moment. Walking down ramp.

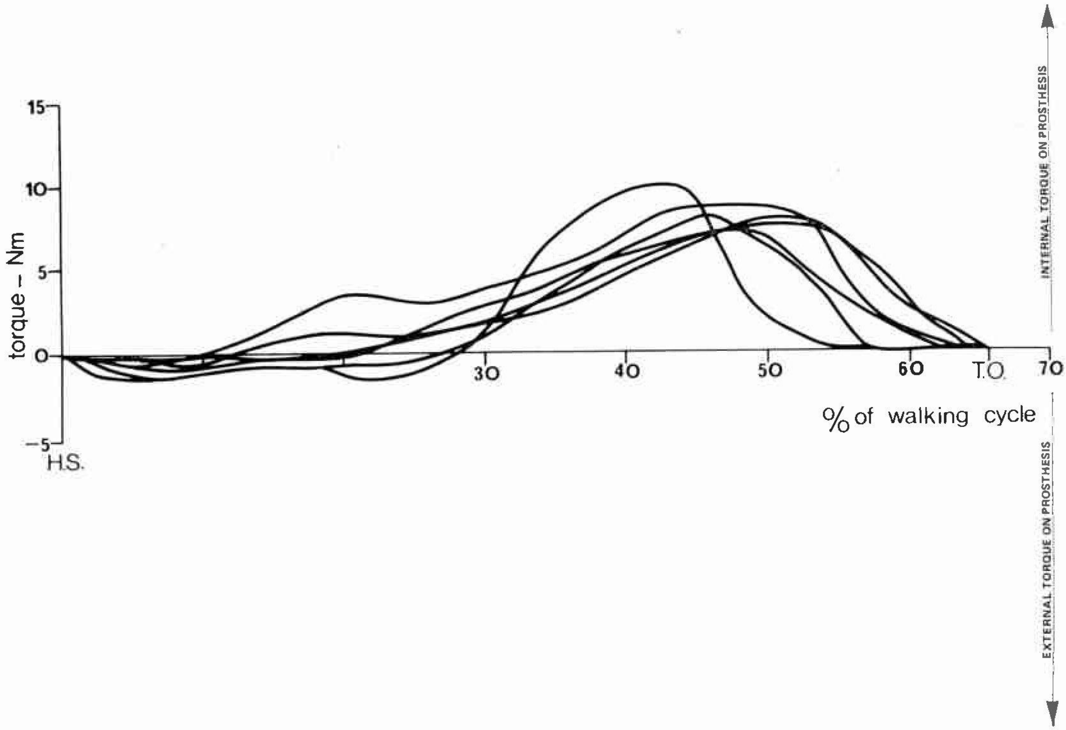

Fig. 8. Torque. Level walking.

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