DYNAMIC ASSESSMENT OF ABOVE KNEE PROSTHESES

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Summary

A dynamometer has been constructed for incorporation in lower limb prostheses. The dynamometer is tubular in form and carries electrical resistance strain gauges connected in bridge circuits to give signals corresponding to three orthogonal force components and three moments about corresponding axes. Signals from the dynamometer are amplified and recorded on a galvanometer recorder. The device can be included in below knee or above knee prostheses but the present results refer only to studies performed on an above knee amputee. A Goniometer is installed on the prosthesis for the present tests to measure knee angle and the output from this appears as a trace on the recorder paper. The values recorded are measured and recorded on punch tape for computer analysis. The quantity calculated in the present tests has been the hip moment exerted by the amputee. It is contended that the magnitude and duration of the hip moment action is a measure of the effort exerted by the amputee in using the prosthesis.

Tests have been performed using six knee mechanisms representative of the types which are claimed to provide some functional advantage to the amputee in walking. The results indicate that in all phases of the support period, the amputee exerts a greater hip moment than static analysis of the knee mechanism would indicate to be necessary.

Introduction

The modular system of assembly of prostheses is now a widely known principle. Modular assembly means assembly from premanufactured, standardised components and it seems likely that this type of system will, within the next few years, largely replace existing conventional means of manufacture. If such a system is to be widely applicable it would seem reasonable that there should be a range of knee mechanisms available to match the requirements of patients with varying functional capabilities. The knee mechanism of an above-knee prosthesis has two sets of functions which may be to a large extent separated. These functions relate firstly to the swing phase and secondly to the stance phase. In objective

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terms a great deal is known about the former but the stance phase has, it is believed, only been studied in a subjective manner. The design of mechanisms claimed to afford improved function to the amputee while the prosthesis is weight-bearing has generally been carried out in an intuitive fashion. This paper represents a report of the first stage, of a project to study prosthetic knee mechanisms in an objective fashion.

Stance-phase Mechanisms

During the time when the prosthesis of an above-knee amputee is in contact with the ground, the requirement of the prosthesis (and the incorporated mechanisms) is that it be stable at heel contact and as the amputee's weight is passing forward over the foot; and that the amputee should be able to flex the knee before toe-off, preparatory to swing through. It is seen that the voluntary control exerted by the amputee during this operation will be dependent on the use of the flexors and extensors of the hip — use of the extensor muscles will tend to stabilise the knee, use of the flexors will tend to flex the knee.

Basically there are three ways in which the knee mechanism may be designed to influence the performance of the prosthesis during stance phase:

1) The alignment of the knee axis in the anteroposterior direction relative to the line between hip and heel. A knee axis placed posterior to this line will produce a stable situation, in that weight applied to the prosthesis will tend to cause hyper-extension at the knee. Such an arrangement will however require increased hip moment to flex the knee before toe-off.

2) The development of a resisting moment at the knee. Many devices produce such a moment as a result of axial load on the prosthesis. As the moment developed resists motion, however, this also increases the hip moment required to flex the knee.

3) The use of polycentric knee mechanisms. These generally are designed so that, at the fully extended position and for the first few degrees of knee flexion, the centre of rotation of the shank relative to the thigh is above the anatomical knee joint. This has the effect of reducing the moment required at the hip to either flex or extend the knee.

All available prosthetic knee mechanisms utilise one or more of these principles. A decision having been made to measure objectively the effect on the amputee of different types of devices a representative selection of those commercially available was made.

Mechanisms tested and abbreviated references

1) Uni-axial (single axis) knee — entirely dependent on alignment stability. — S.A.

2) Blatchford stabilised knee — develops friction moment on application of axial load. — B.S.K.

3) Otto Bock safety knee — develops friction moment on application of axial load. — O.B.

4) Lammers knee — four-bar polycentric mechanism. — L. A.

5) University of California, Berkeley — four-bar polycentric mechanism. — U.C.B.

6) Greissinger knee — condylar type polycentric mechanism, develops friction moment with axial load. — G. R.

Apparatus

To measure the force actions transmitted at the junction between limb segments, it is necessary to measure the externally applied force actions and the configuration of the limb. This is frequently accomplished by the use of measuring platforms and photography. Since complete specification of the ground to foot force action is necessary a single step, six component force platform is required. A multiple step force platform would be advantageous but none is known which gives complete information on the force actions and which does not require abnormal gait to avoid right and left foot contact simultaneously on the same platform. It was decid-

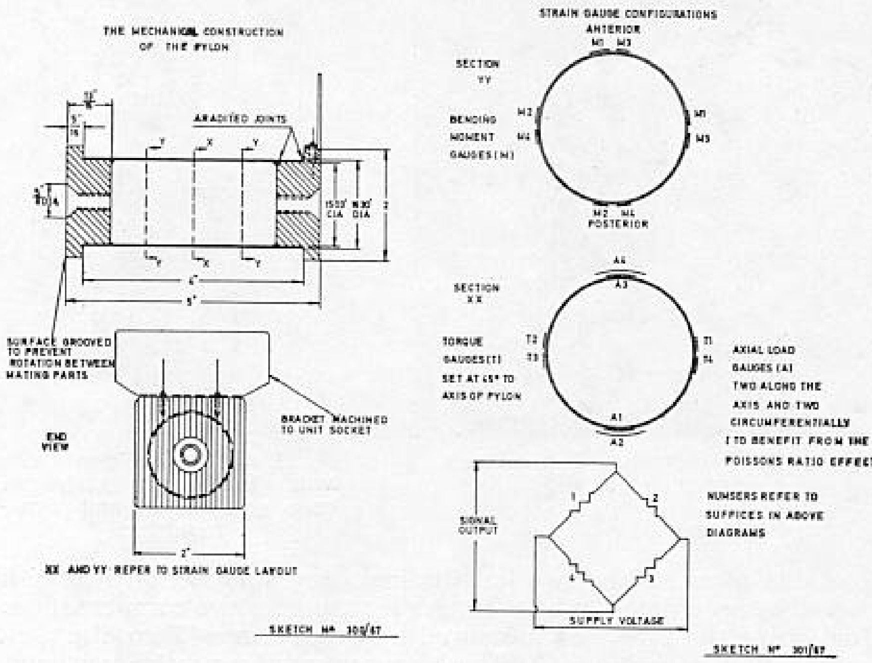

Fig. 1. Prosthesis dynamometer construction, strain gauge layout and circuit.

ed therefore to incorporate force and displacement measurement into the prosthesis.

A tubular dynamometer was constructed similar to that of Cunningham and Brown [1] but having a shorter axial length to allow its inclusion in below knee prostheses. This dynamometer and its associated strain gauge bridge circuits is shown in Figure 1. The tube is 1 $^5/_8$" in diameter, aluminium, 16 gauge. The grooved aluminium end plates are secured to it by epoxy resin. Anterior posterior and medio-lateral bending moments are measured at two sections marked YY by foil resistance strain gauges connected in bridge circuits giving automatic temperature compensation. Four gauges are mounted at each section for the measurement of each bending moment to allow the use of four active arm bridges. Axial

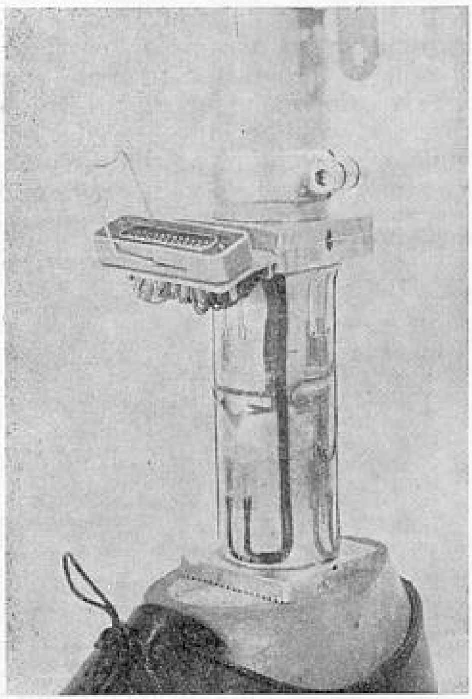

Fig. 2. General layout of prosthesis dynamometer.

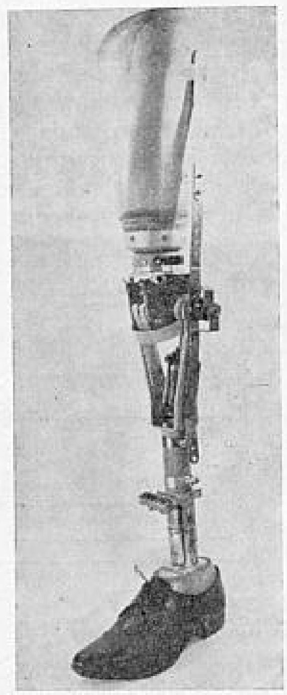

Fig. 3. Above knee prosthesis with Blatchford stabilised knee goniometer and dynamometer.

load was measured by two longitudinal gauges at section XX with two adjacent circumferential gauges for temperature compensation. Torsion of the tube was measured by proprietary 45° rosette gauges mounted at section XX. These gauges gave four active arm operations and self temperature compensation. The general layout of the dynamometer and gauges is shown in Figure 2.

Each bridge circuit was connected to an integral unit comprising DC bridge power supply and DC amplifier. The outputs from the amplifiers were connected to galvanometers in a 25 channel galvanometer recorder using an ultra violet light source. The signals from these six channels suffice to define the force actions transmitted at the proximal and distal joints of the shank of the prosthesis. To calculate the load actions at a reference axis through the hip joint it is necessary to define the position of the dynamometer relative to this. Continuous measurements were therefore made of the angle of knee flexion, using a precision variable resistor connected to the shank and so fixed to the socket of the prosthesis that rotation was transmitted to the rotor without any constraint on relative longitudinal movement. The assembly is shown in Figure 3. The potentiometer is energised from a stabilised DC power source and

TYPICAL U.V. RECORDER TRACE

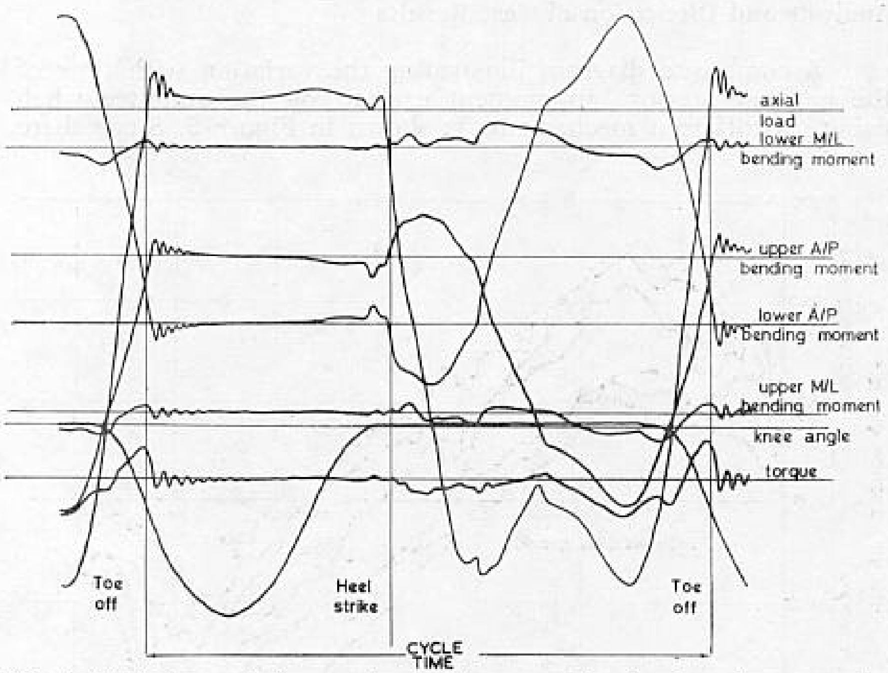

Fig. 4. Typical record from galvanometer recorder showing dynamometer and goniometer signals.

the output of its associated circuit is recorded directly in the recording galvanometer. A reproduction of the galvanometer records obtained from a typical walking cycle is shown in Figure 4.

The data on the galvanometer record is transcribed into punch tape using a proprietary trace analyser. The subsequent calculations are performed for every 1/100 sec. interval by digital computer, taking account of the sensitivity of individual channels by statically determined calibration factors. The graphical output facility of the computer was used to trace graphs to a base of time of the force actions transmitted at ankle, knee and hip.

The amputee used in the recent series of tests is a male of 162 lbs. weight, 5 ft. 9 1/2 ins. tall having a unilateral A-K amputation. A laminated epoxy resin quadrilateral socket was constructed and for each test was connected to the knee mechanism of interest through the associated alignment device or one of the "Staros-Gardner" pattern. A length of tube connected the knee mechanism to the dynamometer and a SACH foot was bolted to the lower face. The leg length and approximate alignment were set and the leg was fitted to the subject. Final alignment was then carried out using conventional methods of assessment.

Analysis and Discussion of Test Results

A composite diagram illustrating the variation with time of the antero-posterior hip moment exerted by the amputee when using six different mechanisms is shown in Figure 5. Since there

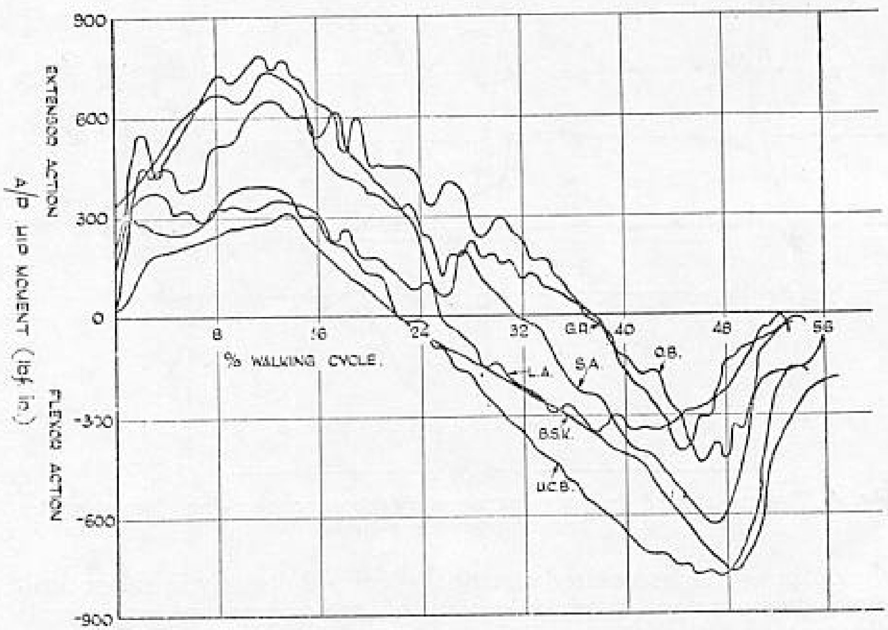

Fig. 5. Moments about medio-lateral hip axis, exerted by amputee on six prostheses with different knee mechanisms. (per legend see text)

were small differences in the cadence of the subjects the results are presented with time plotted as a percentage of the walking cycle measured from heel strike.

It will be seen that the maximum hip extension moment occurs at about 12% of the walking cycle and varies from about 300 lb. — in. to 750 lb. — in. over the range of devices tested. In locomotion of the normal subject there is at mid-stance a period of low hip moment lasting for approximately 15% of the walking cycle. In the devices tested this phenomenon did not occur and change-over from extensor to flexor action was continuous although the position of change-over varied with the device between 22% and 36% of the walking cycle. Maximum hip flexor action occurred at about 46% of the cycle and lay between 350 lb.-in. and 800 lb.-in.

If the geometry of the prosthesis is considered in conjunction with the ground-to-foot forces actually exerted by the amputee, a theoretical curve can be drawn to show the hip moment, M_T, necessary to maintain knee stability for each test. If M_T were plotted on Figure 5, a positive value would indicate a tendency for the knee to buckle which would require to be resisted by hip extensor muscle action. The greatest M_T value calculated for any test was 250 lb.-in. occurring at 8% of the walking cycle and the change-over from positive to negative M_T occurred in the range 12% to 18% of the walking cycle. A comparison of the individual curves of Figure 5 with the corresponding M_T curves showed that at every instant the amputee was exerting a greater stabilising hip moment than was theoretically required. It is recognised however that, since these results were obtained on one test subject and with one particular alignment setting, the diagram will not necessarily represent the complete range of performance of the mechanisms.

REFERENCES

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