THE UNIVERSITY OF CAPE TOWN LOWER-LIMB PROSTHETIC PRINCIPLE *

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COMPLEXITY OF NORMAL GAIT

Many attempts have been made to describe normal gait, but few descriptions have been sufficiently adequate to assist us in our work. It is important to appreciate that variations, major and minor, occur in normal gait, dependent upon many factors. Gait may be rapid or slow. It may be carried out uphill or downhill. It may be vigorous or slovenly, youthful or aged. It varies with the roughness or smoothness of the terrain, with slipperiness, dryness, stability and lighting. The gait of a baby is different from that of an adult. Differences are brought about by footwear and items of clothing, e.g. the mincing high-heeled gait hampered by tight skirts, compared with the free unencumbered strides of the bare-footed scantily-clad African tribesman. Walking is studied by stage artists and can, in fact, be varied to bring about profound differences in stage expression.

No single description of the biomechanics of gait can be adequate as a standard upon which to base all normal fundamentals, unless allowance is made for the above factors and many others not mentioned. Walking is essentially a conscious or subconscious mechanism, a curious trait of the human race.

The mere consideration of anatomic and physiological processes involved in the act of walking would be incomplete without making due allowance for the part played by cerebration and the interplay of vast fields of different conditioned reflexes. Without apparently giving conscious thought to the matter, the walking person between footsteps normally clears the ground over which he walks by a minimal distance consistent with his not stumbling. Thus, gait over an even floor which is well lighted and well seen by the walker may have such a small limit of clearance that an unobserved minor obstruction of as little as an eighth to a quarter of an inch in height can trip the person. Walking over the same terrain at night, the same person will tend to clear the ground with slightly higher steppage for safety's sake, unless he is absolutely familiar with the floor over which he is walking. Should the ambler's eye or his under-foot sensation reveal an unevenness in the surface, he will alter his gait even though he may not always be consciously aware of appreciating the situation.

Non-stumbling and non-accident-proneness may, to a certain extent, be due to the ability to respond to varying circumstances by adopting an appropriate gait learnt from experience. This involves processes of complex conditioned reflexes. People on board a ship can become adjusted to the heaving deck and rapidly attain what appears to be a normal gait. This applies even to the passenger who is not a habitual sea traveller. The sailor who has had a greater training in this process would be more capable of adjusting his walk when placed on such a deck, and would be safer.

Expenditure of Energy

When considering the expenditure of energy in walking, some scientists have confined their attention to the power required by the muscles working across joints to bring about the required range of movement at each joint. This would not be an adequate consideration of normal two-legged gait if the many factors mentioned above are taken into account.

Energy expenditure in gait must be approached in terms of maximal and minimal expenditure. A man carrying a maximum load and climbing up a ladder would represent a maximum gait-energy expenditure, while an unclad, unhampered walker down a gentle slope may well represent a minimal gait-energy expenditure. Walking down a steep hill would once again require an expenditure of greater energy, unless the gait in such a case were 'parkinsonianly' uncontrolled. In walking up steps a person has to lift his total body weight the height of each step with every stride. In walking down he has to lower the same mass under control.

Basing our concepts on previous experience in the field of electromyography with synchronized photography,[1] it would seem reasonable to accept that the human makes as much use of the force of gravity as he can in all his orthograde activities. In standing he will tend to expend minimum energy, and being in a state of near-balance will use only as much muscle power as is required to obtain and maintain balance. In the main, in flexing his trunk from this orthograde stance he will, in preference, allow his trunk to fall, using his posterior muscles to control this fall. His anatomic flexors would only come into action should he wish to bend forward against resistance or carry the action out rapidly.

Man thus uses the force of gravity as a prime mover and merely expends muscle energy to control its effects. His legs act as props balancing the trunk on the ball-and-socket joints of the hip, the flexion-permitting knee joints and the hinge joints of the ankle, with the complex joints of the foot permitting a distribution of body weight on to heel and forefoot under conditions of only essential muscle function.

Thus, also, in walking man tends to harness gravitational and other available physical forces to his own ends. The walking man can be considered as a mass allowed to fall. The fall is controlled by the use of props—the lower limbs. The props are so used as to allow the forward movement of the body mass with minimal lowering; muscle power is then applied in the appropriate way, using the components of the props to lift the body mass again in readiness to repeat the process.

Standing, man can be seen as a system of balancing bodies or components; walking, as a controlled system of falling components, lifted after falling in readiness

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to repeat. Conservation of energy is a hall-mark of efficiency in human function.

Application of these Views to the UCT Research Project

Without much preliminary experimentation, the approach to motorization of lower-limb splintage was based on the abovementioned knowledge, and it was assumed for practical purposes that the act of walking was largely a matter of controlled falling. Four major forms of energy expenditure were considered in this control:

1. Energy for thrust-off.
2. Energy for maintenance of balance.
3. Energy for increasing walking speed and for walking against resistance.
4. Energy for slowing gait impetus.

Accordingly, in the consideration of the earliest design of the University of Cape Town (UCT) Limb as described in this *Journal* in 1960,[2] it was proposed to minimize the lifting of the body and pelvis over the artificial limb during the weight-bearing phases of walking, and to obtain flexion at the knee, which was at all times under control. It was felt that in using standard types of prosthesis amputees were expending a great deal of energy by virtue of their having to lift the body weight excessively with each pace, by their having to brace the knee back for safety, and by their having to maintain control over an almost rigid and somewhat equinus foot. The 1960 UCT Limb can now be regarded as only a first-stage advance towards a definite improvement. Further advances required a review of our own concepts of the basic pattern of gait.

BASIC PATTERN OF GAIT

Orthograde Stance Preparatory to Walking

The body weight carried successively through the bodies of the vertebrae of the spinal column is shared out evenly in normal orthograde stance on the two lower limbs, a pelvic position being selected by the person concerned compatible with his trained, acquired or desired posture. Each half-weight load can be considered to be carried through the heads of the femurs, the knee joints and the ankle joints and then divided again; ultimately to be shared by the underparts of the os calcis and of the heads of the fifth and first metatarsals. Minimal muscle action is used to maintain this position and variations are numerous. Such variations occur according to bodily habitus, shape and design of the bones, joints and muscles, and the election of stance controlled, either consciously or subconsciously, by the individual. The terrain again plays a part in producing variations; ground slope, its surface, the type of footwear being worn, a bag carried on the back, a bag held in the hand, even the holding forward or backward of an arm or a hand; all these will necessitate a change to establish or control safe equilibrium.

The human in normal stance appears to be acutely aware (though subconsciously) of the balance of the component parts of the body. This is seen in extreme examples in the highly-trained balancing acts of tightrope walkers and trick cyclists. We appear to be able to learn how to find a balance of our bodies under the most amazing conditions. Not only is the balancing feat of orthograde stance an amazing phenomenon, but it also becomes more amazing to see African women carrying, with unerring accuracy and safety, large loads on their heads—often walking for many miles in this way.

It would seem, then, that our ability to sense a balancing requirement extends beyond our bodies to tangential things. We can rapidly become proficient at balancing on skates and stilts. Thus, the somewhat easier act of balancing in normal standing stance is regarded as a natural commonplace essential of human life and its non-acquisition requires an adequate pathological explanation. The tired body stands differently from the fresh, vigorous one, and this may account for the occasional Trendelenburg stance in certain statues, the models concerned having adopted a posture designed essentially for a protracted period of holding a particular pose. For our purposes, however, we prefer to consider the standing position as the one described, with the knees not rotated and backlocked, and the feet positioned comfortably so that the body weight tends to run out over the second toe.

The First Pace from Stance

From the stance position we now consider the requirements for walking, once the person concerned has decided to start to walk. The total body weight is allowed to fall slightly forwards, one leg gradually taking more of this weight than the other. This lessening of weight on the limb which is about to be swung forward is assisted by a lift of the pelvis on its side and the gradual transference of body weight over the femoral head of the weight-bearing side. At the same time the pelvis on the free-swinging side is swung forwards. The opposite arm simultaneously tends to swing forward and the arm on the free-swinging side goes backward and slightly outward into abduction, ostensibly to offer additional balancing control. As the one side of the pelvis rises, which it tends to do for $1\frac{5}{8}$ inches, as a rough standard mean, the last evidence of body-weight transference through the limb which is about to become free-swinging is felt under the first and fifth metatarsal heads.

The knee having flexed to approximately 11° from its previous position, and the ankle having dorsiflexed at first approximately 15°, and subsequently plantar-flexed to somewhat below a right angle between the sole of the foot and the mean axis of the leg, the continuance of the forward thrust of the pelvis, aided by the forward thrust of total body weight and, possibly, aided in the first pace by hip flexors, thrusts the free-swinging thigh forward. (Note: in subsequent paces gravity plays an important role in swinging the thigh.)

The knee now bends, giving clearance to the still somewhat plantar-flexed forefoot over the ground. The free-swinging side of the pelvis is then slowly lowered, and while this is happening the knee is straightened—at first slowly, then very rapidly and finally slowly again—in such a manner as to bring the heel of the still somewhat equinus foot to the ground. During this process the opposite arm is progressing forwards and with heel impact it commences its return. The ipsilateral arm almost simultaneously reaches its excursion backwards and commences to swing forward again. The shoulder and body drop slowly once more, and the effect of a general body sway is brought to a near-neutral position as the arms approach the sides of the body and the weight becomes shared once again on both legs.

The Significance of Heel and Foot Fall

For practical purposes heel fall is an instant of great importance. The heel, and shortly thereafter the sole of the foot, senses not only the nature of the terrain, but also the reaction of increasing body weight. At this instant circumstances of action and reaction can be studied at all the component parts involved—the heel itself, its sensory mechanisms, the ankle joint, the joints of the foot, the muscles crossing these joints, the knee, the hip, the pelvis, the spine, the neck mechanisms, the balancing mechanisms and, ultimately, the central nervous system (CNS) mechanisms. We may divide our analysis of heel-fall reaction into two major categories: (1) the effects of the reaction of foot fall upon the person, and (2) the results and effects of the person's predetermination to take the step. Essential to this predetermination are the host of variables involved in the personal selection of the method of gait. An example is the ballet dancer who, for reasons of training and stage appearance, selects to walk with externally-rotated limbs.

It is from this position, i.e. the moment of heel fall, that we choose conventionally to start the consideration of the act of walking.

I have deliberately described the stage ahead of this conventional starting point because we have discovered in the course of our artificial-limb work that the first pace has many and great differences from subsequent paces. In the main, these differences are brought about by the fact that in commencing to walk from the standing position we are obliged to overcome inertia and, additionally, have not as yet determined our selected gait speed. In the construction of a mechanical limb it was easier to incorporate the mechanism for producing free swing for the first pace than for subsequent paces.

The proprioceptive, sensory, peripheral, CNS and autonomic nervous impulses set in train by the sensation resulting from the reaction of the heel and forefoot ground contact predisposes and triggers control of the total mechanism of gait. Again the response may be very varied, but the mechanism we have elected as our guide is as follows:

The whole body tends to sway over towards the new weight-bearing side, the foot gradually receiving this body weight by its transference down the spine through the pelvis, the head of the femur, the knee, the ankle and the foot bones; the foot being placed conveniently in a position of about 5 - 10° of external rotation on the floor. The weight-bearing anterior superior iliac spine commences to pass through the first 90° of a sine curve as described by Steindler,[3] Inman[4] and others. The knee flexes and, for our purposes, we have accepted 11° of knee flexion as being its practicable working extent, in contradistinction to other authorities who claim knee flexion to be as great as 22° and over. At about 11° a line down the mean weight-bearing axis of the femur would be found to fall on, or reasonably near to, the line joining the weight-bearing points of the first and fifth metatarsal heads. While the knee is flexing, the opposite anterior superior spine commences to be thrust forward at greater velocity than the weight-bearing ipsilateral pelvic side.

The whole body sways to maintain the shifting mobile equilibrium, the arms and shoulder girdle swinging contrawise to the pelvic girdle and legs, and the body weight being thrust ever forwards. The whole process would appear to be a means of allowing the body weight to be somewhat lifted on to the weight-bearing limb, and thrust forward at the same time with the minimum wastage of energy, as determined by the conditions under which this process is being carried out.

The impression is gained from our study of these events that a minimal amount of joint movement takes place during weight-bearing conditions. At the top of the first rise of Steindler's weight-bearing sine curve, hip-joint and knee-joint movements are only slight and slow. It would seem that certain individuals, while walking on flat terrain, tend at this stage to extend the knee slightly, but for our purposes we have elected to accept that the knee is still flexed to approximately 11°. The hip is still slightly flexed and the foot has become neutral or slightly dorsiflexed. Body-weight transference proceeds smoothly over the foot, the heel begins to rise, the knee straightens out and the hip extends. The body sways over again towards the mid-line of gait direction, preparatory to the emplacement of the still free-swinging opposite leg on the ground and the sharing once again of body weight between the two limbs. We have accepted that from a mechanical point of view the most efficient time to impart a forward thrust to a pace would be during the period when the body weight has passed ahead of the weight-bearing foot base and is, in fact, a falling body whose direction of fall is merely controlled by the weight-bearing limb, which is itself elongating by extending at the knee and by plantar flexing at the ankle and foot.

This process may be likened to the wrestler's manoeuvre of imparting a most effective and relatively minor force in the direction of established or relative imbalance. After the walking person has reached his desired gait speed, the rhythm of these actions becomes established to carry his body mass smoothly and evenly forward. In the main, our own findings are substantially in agreement with those of others, succinctly described by Charles W. Radcliffe in the June 1962 edition of *Artificial Limbs*,[5] and are therefore not now repeated.

I feel it important, however, to stress the following mechanisms, whose subtlety is not commonly appreciated, and which will be emphasized in describing the principles of action aimed at in our UCT design concept.

APPLICATION OF ABOVEMENTIONED CONCEPTS OF GAIT TO ARTIFICIAL PROSTHESES

Limp

Limping may be defined in different ways according to the manner in which it is being studied. Since it was our object to design an artificial limb as part of a larger research scheme, which was to design and construct mechanisms for motorizing paralysed limbs, it became necessary to decide upon a definition which was helpful in this work. As our work progressed in the field of betterment of artificial lower-limb design, it gradually became impressed on me that one of the most important features about normal gait is the maintenance of an even rhythm of action. The component parts of the body are required to play their part, in accordance with this even transference of body weight ever forwards at any evenly varying velocity, and from side to side over each weight-bearing limb in turn. A disturbance of this regular rhythm constitutes a limp.

In the design of our prototype prosthesis,[2] only a very simple advance was made in the direction of obtaining a more natural appearance in gait. The pantostat design of that model enabled a through-the-hip amputee to walk, but the pelvis tended to fall away at the end of the weight-bearing phase. This required an excessive expenditure of energy on the part of the good limb in counteracting this gravitational interference, and the patient had to lift her body weight preparatory to taking the next step. On attempting to apply this design to an above-knee amputee selected by the British Commonwealth Ex-Servicemen's League (BCESL), difficulty was at once encountered, partly because this amputee had acquired abnormal walking habits through the use of standard prostheses, and partly because the rate at which the pantostat knee extended during the swing-through was predetermined by its design and therefore remained constant. This meant that the wearer was able to walk at one speed only. Any desire on his part to increase gait velocity upset his walking rhythm.

A study of the significance of this finding led me to appreciate that probably the most meritorious feature in the standard artificial limb was the happy empirical accident of the leg being free to swing on the axle at the knee. The wearer of an artificial limb of such design could, within limits, vary the speed of his gait since this free-swinging leg could be made to carry out its excursion through space according to the desire and ability of the wearer. This had two profound effects on our research: (1) it determined our present definition of 'limping', and (2) it necessitated a protracted period of fresh lower-limb prosthetic design and the building and testing of numerous new prototypes.

Definition of Limp

Limping can be said to occur when any of the mechanisms of gait do not take place at the correct moment of time.

From this definition it will be appreciated that we have come to realize that the timing of all the mechanisms of an artificial leg must be made to approximate as nearly as possible to those of the person's normal limb. It is not enough to design a device which imitates natural function in appearance if it does not do so with the correct timing.

Limp Prevention in Prosthetic Design

Having arrived at a decision that a free-swinging leg was an essential for establishing gait-speed selection, it became necessary to consider ways and means of detaching the pantostat apparatus to permit of free swing, and re-engaging this mechanism for the weight-bearing phase of action. This could be done in the bias-control of the four-bar-kinematic-chain or by an equivalent design. Numerous ingenious methods were tried and a number of model prototypes made. One incorporated columns of ball-bearings which, by shooting through a gate into a magazine on weight being relieved, enabled a shaft to shorten suddenly and to allow up to 90° of flexion at the knee. Two sliding sleeves were later designed to obtain the same effect. Consideration was given to the possibility of having a simple axle through the knee, permitting the free swinging, an engagement pinion locking this to the pantostat knee mechanism for the weight-bearing phase.

Ultimately, however, it was decided to carry out the free-swinging action within the hydraulic bias apparatus, by allowing an increased excursion of the hydraulic piston and opening valves to permit the oil to flow through the piston, thus bringing about this free-swinging action. Again, several different designs were tried, with varying degrees of complexity of the piston and the valves attached to it.

Research Aims

While this work was in progress detailed aims were formulated for the mechanical requirements of the new artificial leg. These could be tabulated as follows:

1. The placing of the heel on the ground was required to pre-set the whole mechanism for safe weight bearing.

2. Thereafter the weight of the person on the limb, applied through the ischial tuberosity, and perhaps augmented by some forward thrust of the amputee's stump where long enough, should bring about 11° of flexion at the knee. This flexion at the knee was required to be resilient and, preferably, hydraulically-controlled.

3. The ankle joint should permit of about 15° of weight-bearing dorsiflexion, after which both knee and ankle should tend to extend so that in the passage of the body weight over the limb in the weight-bearing phase the body was ultimately supported on the forefoot in as near as possible a normal anatomic manner.

4. The passage of the body weight over the limb should load springs which, before take-off, could discharge their energy and aid forward propulsion while also tending to lift the pelvis, thus emulating to some extent the thrust-off muscle action of the normal walking leg and foot.

5. Immediately the weight had passed from the weight-bearing artificial leg to the newly-emplanted normal leg, the release of weight from the forefoot should entirely disengage the resilient knee-flexion control and permit the free-swinging action similar to that of standard designs of prosthesis.

6. All the improvements, additional to the above, in our description of the 1960 UCT Limb[2] should, wherever possible, be incorporated. For example, the design was to be light and skeletal-like, covered with light foam plastic and, ultimately, rubber skin for aesthetic, cosmetic and clothing-protection purposes.

Subsequent study of these aims made it clear that it was essential for the hindfoot to be virtually independent in its action, compared with the forefoot. Additionally, the forefoot required to be sprung in such a manner as to allow its forward weight-bearing points to adapt themselves adequately to any reasonable angle of footfall on the ground and to any imperfections and undulations in the terrain. The advantages of such a flexible foot are many, and include such things as the production of a safer footbase, a profound reduction in the tortional and other mechanical stresses and strains placed on the apparatus as a whole, and the reduction in back thrust and strain brought to bear on the patient's stump and body.

It is tempting to describe some of the interim designs experimented upon before arriving at the present design, but, for the purposes of this publication, I have decided to confine myself to a description of an over-simplified schematic representation which will cover the essential action-principles of the UCT Limb.

OVERSIMPLIFIED SCHEMA TO ILLUSTRATE PRINCIPLES OF FUNCTION (FIG. 1A)

Thigh Housing

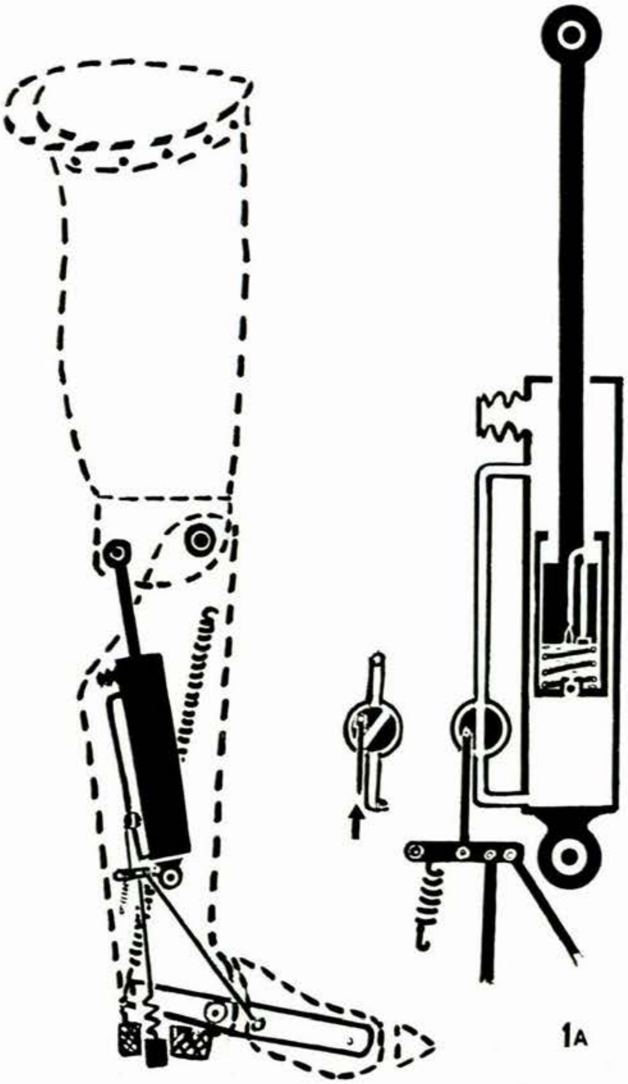
(Dotted line in schematic drawing.)

This is essentially unaltered from the 1960 description, being attached to the pelvic bucket in through-the-hip and hemi-pelvectomy amputees by our own design of hip hinge as described in 1960[2] (Fig. 1B). Above-knee amputees are fitted with standard-type buckets.

Leg Housing

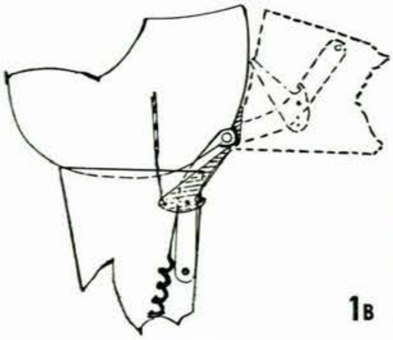
(Dotted line in schematic drawing—Fig. 1A.)

By virtue of the greater simplification in the knee-acting parts, it is now possible to attach the leg housing to the thigh housing with a coronally-placed axle, positioned well interiorly. This axle is similar in design to the standard prosthetic axle in most artificial legs. The leg housing itself is easily adaptable in length, as is the thigh housing, for fitment according to patient variables. At the lower end the leg housing is fitted with a casting or moulding adequately fashioned to permit of the passage of a control rod from the heel and another control from the forefoot.



1A

Fig. 1A. Oversimplified schematic drawing showing the principles of UCT lower-limb prosthetic function. Hydraulic apparatus shown enlarged.



1B

Fig. 1B. UCT modification of pelvic bucket for through-hip and hemi-pelvectomy amputees. Described in this *Journal* in 1960.[2]

This casting acts as a mechanical stopper to dorsiflexion of the spring - steel levers forming the essential part of the forefoot.

As in the 1960 design, weight is kept to a minimum by the making of aluminium housings of small dimensions, and clothing these with foam plastic and rubber skin. As far as possible, working parts of heavier metals are kept as near to the knee as they can be, to minimize leg weight.

Foot

(Blade drawn in continuous line— wood cover in dotted lines—Fig. 1A.)

The foot has three components, viz. (*a*) a heel component entirely independent in its intrinsic action from the forefoot, which consists of (*b*) and (*c*) two spring-steel levers placed on edge to give upward and downward strength for leverage purposes and yet permit of considerable tortional action within their length. Each of these two levers pivots independently of its mate on an anteriorly-placed ankle axle, so that if an unevenness in terrain is encountered, one lever will rise higher than the other, establishing a reasonably even forefoot base. In definitive designs these blades are angled on their pivots to toe the forefoot outwards. Posteriorly, the thrust-off springs attach to these blades, being fastened above at their other ends to the upper part of the leg housing. Forefoot components in the form of wood, with inter-placed rubber or elastic material, give the normal shape to the foot. These are attached to the two spring-steel levers in such a manner that the above actions can take place, while permitting an even forefoot base to be obtained should the leg be put to the ground in abduction or adduction. Additionally, this design permits considerable relief of tortional stress transmitted through the leg to the patient.

Both parts of the forefoot and the hindfoot are covered on the undersurface by bonded rubber, preferably in rubber-slipper form, the heel portion of this rubber base being suitably thickened under the heel without interfering with the action of the heel press-button. When clad in its shoe, allowing for the extra thickness of the shoe heel, the footbase must give a good working footbase for all phases of gait, being neither too equinus nor too calcaneus. The toe tip is cut to give a representative metatarsal weight-bearing transverse or slightly oblique under-edge, which is then rounded off for smooth take-off.

The Schematic Hydraulic Apparatus

(Shown black in the composite, and in schematic detail alongside—Fig. 1A.)

In principle this consists of a shaft attached to a piston working in an inner travelling cylinder, which itself is a piston working inside the main cylinder. A bypass permits communication between the lower and upper parts of the main cylinder.

Bypass valve. In this conceptual drawing a rotatory bypass valve is shown in the bypass circuit. The flow is shown shut off in the small detached sketch and opened in the hydraulic schematic drawing.

Other hydraulic components. At the very top of the main cylinder a compensator bellows is fitted to accommodate rapid-volume oil-displacement variations. The piston is fitted with two valves; a bleeder valve acting during its downstroke, a flap valve opening rapidly on the upstroke. The piston descends on to a compression spring in its downstroke. This spring is contained in the lower part of the travelling cylinder. In the bottom of this travelling cylinder a ball or flap valve is fitted which closes on the downstroke.

The attachment of the hydraulic apparatus. The upper end of the shaft is attached to the lower posterior part of the knee portion of the thigh housing by means of a pivot and bearing. A pivot and bearing attaches the lowest end of the outer cylinder to the shin housing in its lower third. The bypass valve is in effect harnessed to two separate controls, one attached to the heel press-button and the other differentially attached to the two forefoot spring-steels anterior to their shinpiece pivots. These controls are designed to open the bypass when weight is removed from both the heel press-button and the forefoot, and to close the bypass whenever pressure or weight is felt anywhere under the foot.

DESCRIPTION OF THIS SCHEMATIC APPARATUS IN ACTION DURING GAIT

(See Figs. 1A and 2. In Fig. 2 the gait phase positions are referred to by the letters A - J.)

First-pace Cycle

If the heel is placed on the ground as at A (Fig. 2), the heel press-button is pressed upwards. This operates its attached control and closes the bypass valve in the hydraulic system (see small side sketch in Fig. 1A). The patient's weight is transmitted down the thigh housing posterior to the shin-thigh axle, and therefore the knee will tend to flex at B. The hydraulic system being filled with oil and the bypass valve now being closed, the shaft and attached piston are driven downwards. The ball valve in the bottom of the travelling cylinder shuts off, as also does the flap valve in the piston. Thus, with only the bleeder valve in operation in the piston, the piston can only travel downwards on the oil and the spring in the travelling cylinder a limited distance, the travelling cylinder being unable to move. This limited distance of piston movement is arranged to permit only 11° of flexion at the knee, as at C.

The body weight is carried forward over the forefoot and the hip moves towards extension, as at D. The pelvis is elevated on a Steindler's curve, which, for practical purposes, is normal. As the body weight is transferred on to the forefoot and beyond, the pelvis following the normal Steindler's curve fall, the heel tends to lift, as at shortly after D. The thrust-off springs become stretched by the forefoot becoming dorsiflexed through about 15°, and the bypass valve remains closed by virtue of the action of the controls attached to the forefoot spring-steels.

Thus the controlled 11° of knee flexion is maintained and, with the body weight moving ever forward, as at E, the leg rises on the front of the forefoot. The stage is reached when the opposite leg is about to be placed on the ground and the body weight has moved sufficiently ahead of the artificial forefoot so that its effect is rapidly diminished, as between E and F.

The amputee is momentarily off balance and this is the moment to apply his thrust-off, which is done by the recoil of the thrust-off springs. In practice this thrust-off begins surprisingly early in the last part of the weight-bearing phase and, coupled with this action, the knee tends to straighten out and the ankle to plantar-flex, while the pelvis is pushed slightly forward and upward. The mechanism of this is as follows: The piston flap valve opens and the piston rises on the spring in the travelling cylinder. The thrust-off springs acting through the foot levers enhance this action.

The instant the body weight is taken off the prosthetic limb, as at slightly before G, the bypass valve opens and allows the flow of oil through the whole system. The free swing at the knee is thus permitted. The lift and forward thrust of the pelvis on the prosthetic side imparts a forward momentum to the thigh housing and to the upper part of the shin housing. The lower shin housing tends to lag behind, the resultant effect being flexion at the knee, which varies according to the speed of the gait and the velocity of pelvic forward thrust elected by the amputee, as in G, H and I. In this manner the toe clears the ground, and the whole limb continues to move forward. With the relative slowing of the thigh as it reaches the walking hip-flexion position, and with the retardation of pelvic velocity, the knee becomes straight again preparatory to subsequent heel fall and the commencement of the second pace.

If a friction or damping factor is introduced between the shin and thigh at the commencement of the free-swing phase, the whole timing of the free swing is altered, the upward excursion of the heel is cut down, the toe and foot ground clearance is diminished and the extended leg reaches the heel-bearing stage sooner. It must be appreciated that this back-damping requirement differs in the UCT Limb principle from that needed in the standard lower-limb prosthesis. Many factors appear to be responsible for this difference, of which the take-off from a forward-thrusting flexed-knee position is certainly a major one.

This is in contradistinction to the standard types of artificial limb, in the use of which the wearer needs to keep the knee braced back in extension during all weight bearing; the toti-resultant of body weight and forward thrusting forces at take-off is a vector falling behind the forefoot weight-bearing line (Fig. 3A). With the thigh swinging forward the knee describes a forward moving arc of a circle, the radius of which is the length of the femur, and the centre the hip joint. These things are happening in the extension range of the hip joint's action. The knee's arc of movement must be downwards towards the ground at the very time that the leg is required to rise on to its forefoot preparatory to breaking into the free swing. A considerable reactive force is thus set up, tending to cause the foot to lift upwards and slightly backwards, not only upsetting the timing of the free swing, but also necessitating

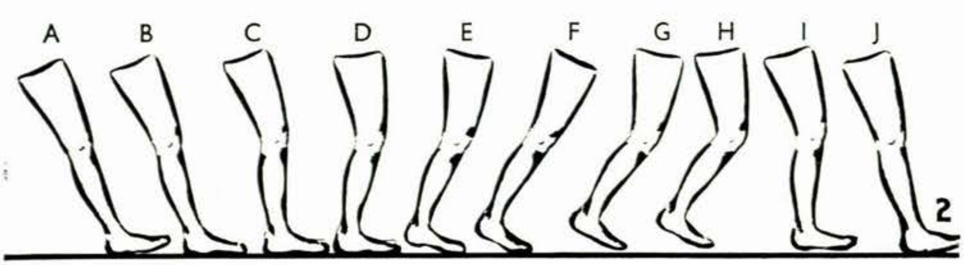


Fig. 2. UCT gait cycle. This should be studied together with the schematic diagram, Fig. 1A.

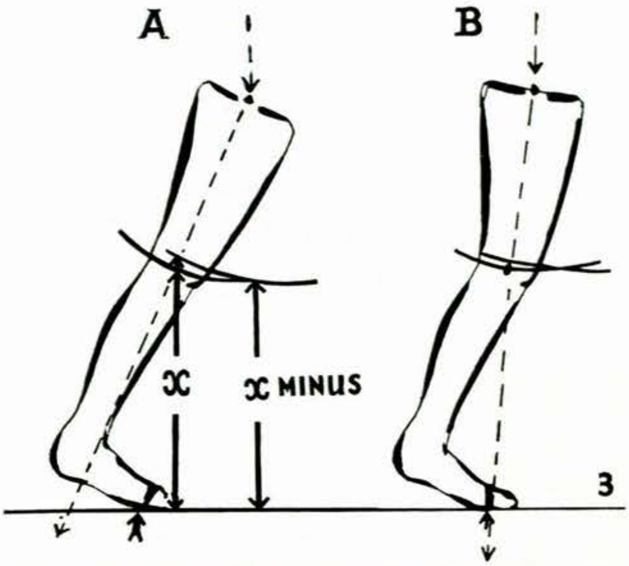


Fig. 3. Illustrating the difference in take-off reactionary forces, using (A) back-braced knee, and (B) flexed knee. The 'pip-squeeze action'.

other anatomically-divergent habits on the part of the wearer, these varying in different amputees.

In the UCT Limb principle, resilient knee-flexion control and ankle movement ensure that minimal backward and upward lifting reactive force exists, and the free swing can be more naturally obtained. Should they be found desirable later, when sufficient experience over a wide range of limb wearers has shown the need for damping at the back of the free swing, there are numerous simple ways of obtaining these damping effects in a hydraulic system of this design.

A damping effect, however, which we have considered it wise to include at this stage, is the hydraulic slowing-down of extension at the forward end of the free swing. The two main reasons for its inclusion are to lessen extension jarring and to obtain a more anatomic timing of the gait cycle. Incidentally, the need for this type of damping is also lessened in the UCT Limb principle by the fact that the amputee does not have to thrust stump and pelvis backwards to ensure extension of the knee for safety; in fact, the desirable thing is the opposite—the normal act of thrusting forward to produce a flexed knee.

Sequence of Events once Gait-speed is Obtained, i.e. Subsequent Paces (Figs. 1A and 2)

Heel-fall (A) and foot-fall (B) (Fig. 2) remain as before, only becoming more rapid. Knee and ankle flexion, as at C, require that the amputee's ischial-bearing weight is ensured, and that he thrusts forward with thigh stump and pelvis.

Owing to the 11° of knee flexion reducing the 'pip-squeeze action', with the foot assuming a fixed relationship to the shin piece between D and E, the limb rises up on its forefoot. The thrust-off springs continue to load and there is a growing tendency, by virtue of the intrinsic design of the knee and ankle with bias systems working between them, for the knee to extend slightly and the ankle shortly thereafter to plantar-flex. Between E and F the body weight falls increasingly ahead of the position of the forefoot on the ground. It is during this period that the thrust-off spring comes into action by unloading. This recoil has the effect of lengthening the limb overall and, at the same time, of tending to increase the length of the stride, thus imitating the thrust-off of normal limb gait. Footfall of the limb on the opposite side occurs shortly after F and the swing-through phase of the artificial limb commences shortly before G.

The swing-through is similar thereafter to that seen in the use of a standard artificial type of leg, but because it commences from a position of some degree of knee flexion and is assisted by a pelvis and stump moving forwards, aided by the thrust-off mechanism, the swing-through occurs in more natural timing and with less upward fling of the heel. Immediately before G the pelvis on the prosthetic side is lifted. The removal of weight from the foot disengages the hydraulic weight-bearing conditions effectively and permits the leg to swing freely on the knee pivot. At G the thigh is being propelled rapidly forwards, the knee travelling the fastest. This carries the upper shin piece forwards at the same rate, depending on the mass of the leg and its inertia at the knee. The foot tends to lag behind. It lifts in its swing and thus clears the ground. As the thigh begins to slow down in its forward passage at H and I, the lower part of the shin continues to travel forward, being slowed slightly towards the end of its excursion by a simple narrowing of the bypass oil port in the hydraulic cylinder. It thus assumes the position at J, which is the same as the position at A, in readiness to recommence the cycle without any jarring.

DISCUSSION OF CERTAIN PRINCIPLES LEARNED

The Free-swinging Knee

Probably one of the most important lessons we have had to learn in the course of our research is the importance of the free-swinging knee in the swing-through phase of gait. This is a subtlety which has been taken for granted by most prostheticists. It was not until we appreciated that the placing of a simple pivot at the knee was perhaps the happiest single empirical accident in prosthetic history, that we were able to break away from the 1960 pantostat design, in which we used a bias system to control the behaviour of the limb during the swing-through. This pantostat design had worked admirably on a through-the-hip amputee, but when applied to men from the BCESL who had above-knee amputations, we discovered that such amputees could only walk at one speed, viz. the speed governed by, and which took its timing from, the toe-off to heel-fall phase. Certainly one of the most difficult tasks in the development of the UCT Limb was the incorporation of the benefits of a resilient knee-flexion control, i.e. the weight-bearing advantages of the pantostat limb with the hitherto ill-appreciated free-swinging action of the standard prosthetic device.

Significance of the Differential UCT Design of the Foot

It would not have been adequate for our leg to be safe after heel-fall only, since the transference of body weight on to the forefoot in the presence of a flexed knee would have resulted in a fall. It therefore became necessary to draw yet another lesson from nature. We decided to make the heel a sensitizing device which pre-set the hydraulic apparatus for weight-bearing function. Since heel sensation often has the opposite function to that of the forefoot, we slowly evolved the concept that the forefoot should be entirely independent of the hindfoot, but that, nevertheless, the transference of weight on to the forefoot should maintain the same weight-bearing properties in the limb as pressure on the hindfoot. Thus, by coupling the forefoot to the same hydraulic control, we were able to obtain complete safety under all weight-bearing conditions.

The leg could walk up a steep hill, making its landing on each occasion on the forefoot alone, and still always permit a further safe 11° of knee flexion and 15° of ankle dorsiflexion. Furthermore, by dividing the foot into independently-sprung medial and lateral halves, we were able to obtain a foot which could adjust itself to unevennesses on the walking surface and also to its emplacement in extreme positions of abduction or adduction footfall.

Sitting

In sitting, using a UCT Limb, all that is required is that the weight be taken off the prosthetic limb. The knee will then bend to the desired degree. In fact, in our research model we have elected to allow approximately 130° of knee flexion, which enables a person to kneel in comfort. Over the major range of this knee flexion it is still possible, if the foot is placed on the ground immediately, to arrest the flexion and thereafter to permit only a further controlled 11°. Thus, if the amputee should stumble, his knee will arrest his fall by this mechanism. Additionally, by inclusion of a valve in the piston mechanism, the person may at any time extend the knee without lifting the foot from the ground. These features have been incorporated to enhance the mechanical imitation of the normal limb.

While we have not carried out any particular tests on stairs, it is very possible that these additional features may be found useful, and we expect them to be developed further by subsequent research workers in due course.

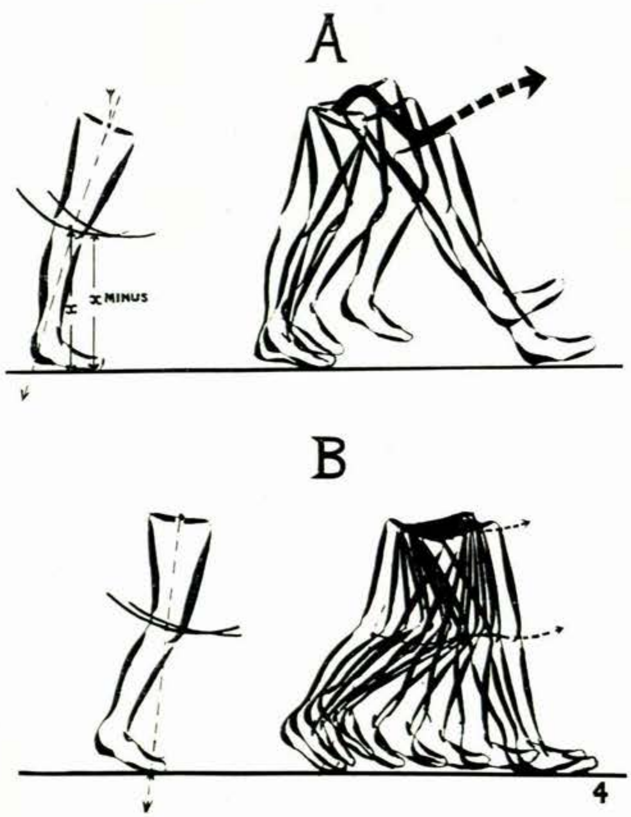


Fig. 4. Free-swinging phase. *A.* The high lift and dumping action of standard prosthetic gait. *B.* The more natural swing-through of the UCT principle.

Energy Expenditure Factors

Any feature in an artificial limb that will cause the body weight to be lifted higher or dropped lower than it should be with regard to the phase timing of gait, will result in a tendency to limp, expenditure of energy, and stress and strain on apparatus and amputee. In Fig. 4A a somewhat exaggerated example of a commonplace free-swing action, using a standard artificial limb, illustrates both excessive body lifting and downward plunge. This is due to the need for constant back-bracing of the knee, coupled with the mechanics resultant upon a fairly rigid, somewhat equinus, foot. In Fig. 4B a free-swing tracing of the UCT Limb in action has been made, and this appears to be fairly faithful to Steindler's normal gait curve. Pelvic lift for take-off is not great and, once mastered, does not appear to result in any serious loss of energy or in distress to the amputee.

Any feature in an artificial limb preventing smooth ankle and knee movement through an anatomic excursion will result in a faulty timing of action during the weight-bearing phase. This is illustrated in Fig. 5A, which is made from commonplace examples of the use of standard artificial legs and illustrates how, with the foot snubbered in relation to the shin piece in order to produce back-setting of the knee, and with the amputee back-bracing his stump and pelvis to produce this same safety effect,

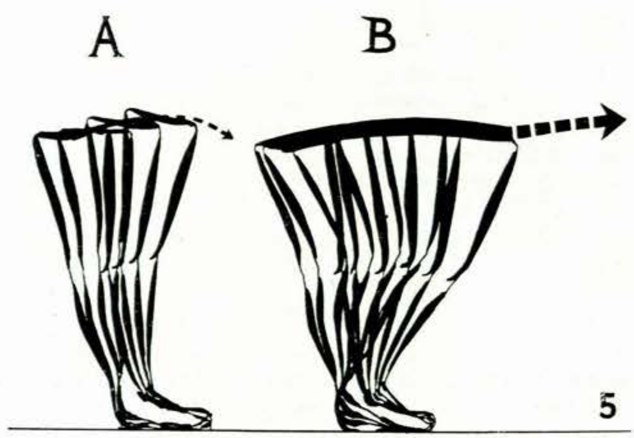


Fig. 5. Weight-bearing phase. *A.* The limited and energy-consuming action range of the standard prosthesis. *B.* The full carry-over action of the UCT principle.

Note the flatter bucket-top tracing of the UCT principle during comparable gait timing.

it is not possible for such a leg to use more than the first 90° of Steindler's weight-bearing sine curve. Furthermore, such a limb must once again raise body weight unnaturally high. In Fig. 5B, for contrast, the tracing of the curve described by the UCT Limb bucket-top is found to pass through the full length of the weight-bearing trochanteric sine curve and, owing to the safety position of knee flexion and the controlled ankle dorsiflexion before the limb rises on the forefoot, this curve is now as near to that of an anatomically normal limb as we could make it. Additionally, all components appear to track through space in correct or nearly correct timing.

In Fig. 3A, what appears to be a very important principle is illustrated. The lower end of the femur or thigh piece describes a circle whose centre is the hip joint. At the end of the weight-bearing phase, if the thigh is in extension, such a circle will approach the ground for a part of its curve. If the hip joint is swung forwards slightly during this movement, a fresh arc of circle can be described which, likewise, will approach the ground. In the case of a standard prosthesis in which the mean thigh axis passes through the hindfoot on protraction, the weight-bearing forefoot-to-knee will become squeezed, producing excessive pressure on the forefoot. This, in turn, will result in an upward thrust on the pelvis and a reactionary force tending to put the knee joint into flexion with a sudden jar and causing the heel to swing backwards and upwards out of correct timing and too high.

In our workshop we have at times simplified complex reasoning by referring to analogous effects. Since this action seemed similar to the shooting of an orange pip from between an index finger and thumb, we have come to refer to this as the 'pip-squeeze action'. In contradistinction to this effect, Fig. 3B shows the advantage of knee flexion before take-off. The protracted mean thigh line moves forward to the weight-bearing point of the forefoot. Under such circumstances no backward thrusting force is felt and the limb can move forward into a more natural free swing.

Our limited experience using the UCT Limb has led us thus far to feel that a small amount of 'pip-squeeze action' is possibly a desirable thing, and this is one of the features which has made us, in preference, keep to the 11° of knee flexion rather than a greater angle. Should later research prove that a greater angle of resilient knee flexion control is desirable, it would be obtained easily by increasing the excursion of our inner piston in its travelling cylinder. In all ways the diminished 'pip-squeeze action' at the foot is regarded as a real advance in our prosthetic research, enhancing not only the smoothness in correct timing of the rhythm of gait, in removing shocks and strains from the prosthesis and the person, but also in altering the need for damping at the back of the swing phase.

Prosthetic Walking Habits

A major difficulty encountered was the attachment of a weight-bearing knee-flexion prosthesis to people who had been trained to walk on standard prostheses. Weight-bearing safety on a standard prosthesis, requiring that the wearer keep the knee braced backwards at all times, results in habits of gait which are the reverse of normal. A number of the prototypes designed and made by the UCT research team could obviously be used immediately by amputees who had not acquired these habits. At one stage we were almost resigned to the acceptance of the separate design of an alternate principle for use by amputees who had worn standard prostheses for any length of time.

However, with the development of the present principle we have ultimately been able to overcome this problem. It is no longer essential in the present limb design for the amputee to bend his knee. In fact, he may elect to use almost any mannerism of gait and the limb will function reasonably to accommodate such gait election. Nevertheless, it is obvious that it is advisable for any such person, desiring to use the UCT type of limb permanently, to overcome his acquired habits, and to master those more natural actions which will bring out the best characteristics in the new limb. This limb should be made almost to leg length, and certainly should not be as grossly shortened as standard prostheses.

Thalidomide Babies

In the case of M.G., a baby from South West Africa born without arms and legs, whose mother may or may not have taken drugs during her pregnancy, I have personally made two consecutive pairs of legs of wood with mechanical parts of 'tufnol'. Experience with this case indicates that a number of smaller UCT-type limbs should be made for these children, even though the expense may be great. They are surprisingly agile with the parts nature has given them, and quickly learn to use such devices. Where arms are missing in addition to lower limbs, the danger of not being able to break a fall, with consequent head and shoulder injury, must not be forgotten. In the baby mentioned above, a walking frame is regarded as essential.

This problem is a new and important one, which should be attacked vigorously. Our own experience has shown that the principles of function of the UCT lower-limb prosthesis go a long way towards a solution, and if we could have afforded it we would have used a hydraulic means of making M.G.'s legs, rather than the mechanical method.

Testing

It is not expected that the principles of lower-limb design incorporated in this work will immediately or even rapidly result in full-scale manufacture. We appreciate that there is a great deal of work in pre-manufacture development. From our point of view it would be utterly wrong to take one of our limbs and judge it or condemn it by any form of existing laboratory or bench test. The only test of definitive value is the field test or man test, and our work is expected to stand ultimately only in the face of such testing over a wide range of amputees. In giving this work to the world, therefore, we are proud of the university in which it has been developed thus far. We are also proud to have carried out this work in South Africa. We ask in return that these facts be commemorated by the perpetuation of our principle of limb function under the title of 'The University of Cape Town Lower-limb Prosthetic Principle'. No royalties or financial reimbursement are sought—it is the desire of those concerned to help the people who need our work; in return we request a reasonably sympathetic acceptance of these principles and the genuine furtherance of their objectives by those in other lands who make use of them.

BRIEF HISTORY OF THE DEVELOPMENT OF THE UCT LOWER-LIMB PROSTHETIC PRINCIPLE

This work has been carried out by a very small research team working at times with minimal resources under difficult conditions. Our first guinea-pig found ample material to write a book about her experiences.[6] The members of our team could similarly each write a separate book concerning our day-to-day experiences, the perpetual frustrations, the seemingly insuperable obstacles—all of which have been mastered the hard way and, for some unaccountable reason, in every instance the most complex way to begin with. After beating a problem by constructing a most complex 'Heath Robinson' device, we have set to work again to turn such a device into simplicity. Only those who have been intimately associated with this struggle will ever fully appreciate the meaning of these words. A sense of continuous urgency has been with us through the years, and this has been heightened by the knowledge that the world needed our results. The reality of this was brought home to us continuously by the daily spate of letters we received from limbless persons.

This has not merely been an attack on a single component such as a knee or ankle or foot but, in fact, has been at all times the incorporation of principles of limb function in the total design of artificial legs.

When the limb research project of this Department was begun, we did not have a clear conception of what was desirable in an artificial limb and did not know to what extent it would be possible to attain our objectives by means of mechanical contrivances. The team was small, consisting of myself with the orthopaedic and electromyographic background, and Mr. L. V. Holmgren with the technological training in similar research fields, and with a particular aptitude for design and the interpretation of biomechanical requirements into a mechanical end-product. We were subsequently joined by Mr. R. C. Hampton, whose precision turning, fitting and machining left little to be desired. Our policy was to make a device based on what seemed a correct principle, put this on to a patient and test it, learn from our mistakes, and develop the succeeding stage therefrom.

The first major milestone was the building of our 'pantostat limb', the account of which was published in this *Journal* in 1960.[2]

Resulting directly from our early tests on the BCESL subjects, the second major stage of the work consisted of developing what we have called the 'resilient knee-flexion control' (a term suggested by Dr. T. Ritchie of Roehampton). These prototypes have been redesigned and altered many times, each alteration being dictated either by the addition of a new principle of requirement or by an imperfection or failure brought to light by a human test.

The history was punctuated at regular intervals by our kicking ourselves for 'not having thought of that one before'. In fact, it now becomes apparent that the most important end-product is not the design of the limb itself, but the major principles of action to which lower-limb design will have to conform. The same effects could be obtained in many different ways. It is obvious to us that the possible refinements, embellishments, improvements and alterations are legion. The time has come for us, in the name of the University of Cape Town, and on behalf of all those who have in any way helped this work along, to place on record the essentials of these principles so arduously learned. This we have done in the present paper. Our small team also feel that the time has come for the world to do what we have done so many times ourselves and say, 'Why didn't we think of these simple things before?'

ACKNOWLEDGEMENTS

The Research Team

Mr. L. V. Holmgren, A.I.S.T., has worked in close liaison with me. His has been the task of designing and making the many devices and prototypes tested. It was originally intended that his name should appear with mine as co-author of this publication, but the final decision not to include the detailed specification of our latest prototype affords me this opportunity of recording my deepest appreciation of the dogged and determined efforts of this outstanding research technologist.

The third team member is Mr. R. C. Hampton. It has been his task to machine the parts designed. His contribution has been indispensable to the success of our work, and he has endeared himself to us all.

Mr. R. B. Westgate has often volunteered his valuable assistance, sharing some of the long nocturnal periods of work.

Helpful People

Sir Solly Zuckerman, *K.T., C.B.*, M.D., D.Sc., F.R.S., indirectly determined my progress which led to this work, by his recognition in 1947 of some value in my early electromyographic work.

Drs. T. Ritchie, D. S. McKenzie and others of Queen Mary's Hospital, Roehampton, gave useful criticism, while Brig. H. Hopthrow, *C.B.E.*, of Imperial Chemical Industries arranged this valuable collaboration.

The late Mr. H. Morris of Optical Supplies (Pty.) Limited made limb housings and parts for some of the earlier experiments.

Mr. A. R. McDonald of the Trigonometrical Survey Unit in Cape Town helped with occasional advice.

Messrs. G. McManus, W. Hall and C. Goosen gave help in photography, and in plastic and moulding work.

'Guinea-pigs'

Mrs. J. Tribelhorn, the late Mr. H. Wanliss and Mr. Alan Pargiter bore the brunt of the hazardous tests.

Dr. A. W. S. Verster of the Pensions Department arranged the selection of amputees.

Financial and other Assistance

Thanks are due to the University of Cape Town, the South African Council for Scientific and Industrial Research, the National Council for the Care of Cripples in South Africa, the Cape Cripple Care Association, the South African Legion of the British Commonwealth Ex-Servicemen's League and the Hospitals Department of the Cape Provincial Administration.

Valuable cooperation has been enjoyed with the Orthopaedic Workshops of the Cape Provincial Administration, under the directorship of Mr. D. Louw.

Interest in our work was shown by the other Departments in the University of Cape Town, especially those of Surgery, Medicine, Anatomy and Physiology.

Private Financial Donors

The late Mr. Francis George Connock. Mr. and Mrs. Harold Jones.

Pre-manufacture Development

On behalf of Messrs. J. E. Hangar & Co. Limited, Mr. J. B. Waggott has kindly acceded to our immediate request to consider the possibility of manufacture and pre-manufacture development and testing in Britain of the latest University of Cape Town Limb prototype. We have his assurance that no undue exploitation will occur.

SUMMARY

1. Lower-limb normal and prosthetic gait is discussed.

2. The end-product of the past five years of lower-limb prosthetic research by the University of Cape Town's Orthopaedic Department Research Unit is hereby made public as follows:

(*a*) The concept of what appears to us to be a correct approach to the basic pattern for lower-limb prosthetic gait is described under the title of 'The University of Cape Town Lower-limb Prosthetic Principle'.

(*b*) A schematic drawing of a mechanical limb is described to illustrate these principles of action.

3. This work has now been carried sufficiently far to make it clear that the above-mentioned principle may be obtained in different ways; thus the emphasis is placed on the concepts and principles of function rather than on the specific limb design.

REFERENCES

1. Allen, C. E. Lewer (1948): Brit. J. Phys. Med., new series, **11**, 66.
2. *Idem* (1960): S. Afr. Med. J., **34**, 125.
3. Steindler, A. (1953): J. Bone Jt Surg., **35A**, 540 and 728.
4. Saunders, J. B. de C., Inman, V. T. and Eberhardt, H. D. (1953): *Ibid.*, **35A**, 543.
5. Radcliffe, C. W. (1962): Artif. Limbs, **6**, 16.
6. Tribelhorn, J. (1962): *The Challenge and the Triumph.* Cape Town: Tafelberg Uitgewers.