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A new modular six-bar linkage trans-femoral prosthesis for walking and squatting

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Abstract
Four-bar linkage mechanisms produced by many designers of knee joints for trans-femoral prostheses can provide knee rotation to permit walking only. In Afro-Asian countries people are accustomed to a squatting posture in their daily activities. A six-bar linkage knee-ankle mechanism trans-femoral prosthesis is described which was developed and fitted to an amputee. The motion patterns of the ankle, knee and thigh during walking and squatting (obtained using a flickering light emitting diode system) for the above prosthesis is compared with motion patterns obtained for normal subjects. The closeness between both the patterns establishes the suitability of the new modular trans-femoral prosthesis for producing near normal patterns of motion during walking and squatting. The additional facility of cross-legged sitting provided in the prosthesis makes it functionally suitable for Afro-Asian amputees.

Introduction
People in Afro-Asian countries are in the habit of sitting in a squatting posture for many activities starting from the use of the toilet to farming operation and in cross-legged sitting posture for relaxation and during their daily prayers. Amputees wearing usual trans-femoral prostheses are not in a position to perform the above functions. Most of the models for trans-femoral prostheses are of single knee axis with solid ankle except a few which have the provision for ankle dorsiflexion (Radcliffe and Lamoreux, 1972; Seliktar and Kenedi, 1976) and polycentric action at the knee joint (Radcliffe and Lamoreux, 1972; Cortesi, 1975; Cappozzo *et al.*, 1980). The required amount of ankle dorsiflexion with knee flexion during squatting is not possible with any of the existing models except for one (Chaudhry *et al.*, 1972) designed so far. But this model is an exoskeletal single axis prosthesis which makes it difficult to provide a proper cosmetic cover. This being a fixed, single axis knee, requires more effort at the start of flexion during stance phase than the polycentric knee prosthesis. This is because of the smaller effective lever arm of the single axis prosthesis as compared to the polycentric knee axis prosthesis.

In normal walking two major muscle groups of the lower limb control the swing phase of walking. The prosthetic leg without any swing phase control arrangement at the knee joint behaves like a pendulum and without any control produces an unnatural gait. In earlier designs this problem was solved by mechanical means or by providing frictional resistance at the knee joint (Murphy, 1964). Later different hydraulic control systems (Lewis, 1965) were introduced. The disadvantage of a mechanical frictional resistance system lies in the fact that this system cannot produce natural gait. Moreover, these devices get damaged due to wear. The hydraulic control unit can provide a better control of swing but generally due to the

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weight of the unit the prosthesis becomes heavy and also due to leakage of oil the prosthesis becomes dirty. On the other hand, when a pneumatic control is provided at the knee joint both the above disadvantages can be removed and a better walking pattern can be achieved. The pneumatic swing phase control unit was first developed in the Biomechanics Laboratory at the University of California (Radcliffe and Lamoreux, 1968; Zarrugh and Radcliffe, 1976). But this prosthesis is not able to provide the knee ankle coordinated motion necessary for squatting. Therefore, it was felt necessary to develop a new prosthesis which (i) can provide the basic requirement of a stance phase stability with proper polycentric action (such that the hip ankle reference line during stance phase passes in front of the variable knee axis), (ii) should be of modular design and will provide ease of walking and sitting in and rising from the squatting posture, (iii) should be able to provide swing phase control.

In this paper, details of a new modular six-bar linkage trans-femoral prosthesis, with facilities for (i) swing phase control, (ii) coordinated motion between knee and ankle (provided by a six-bar linkage), (iii) squatting and (iv) cross-legged sitting, are described. The prosthesis is fitted to an amputee and his motion patterns (obtained experimentally) are compared with normal patterns.

Methodology

The prosthesis designed and developed for different functional improvement is shown schematically in Figure 1. Different functional capabilities of the prosthesis are described below.

Polycentric action at knee

The trans-femoral prosthesis has a six-bar linkage arrangement at the knee (Fig. 2) by which the motion from the thigh can be transmitted to the foot during squatting action and during the swing phase of walking. The four bars 'a', 'b', 'c' and 'd' form a four-bar linkage mechanism with a short posterior link 'c', designed after several trials, so as to create an instantaneous centre in full extension located well above a corresponding single axis knee centre and posterior to the hip ankle line. This results in stability of the prosthesis during stance phase. With little effort of the hip

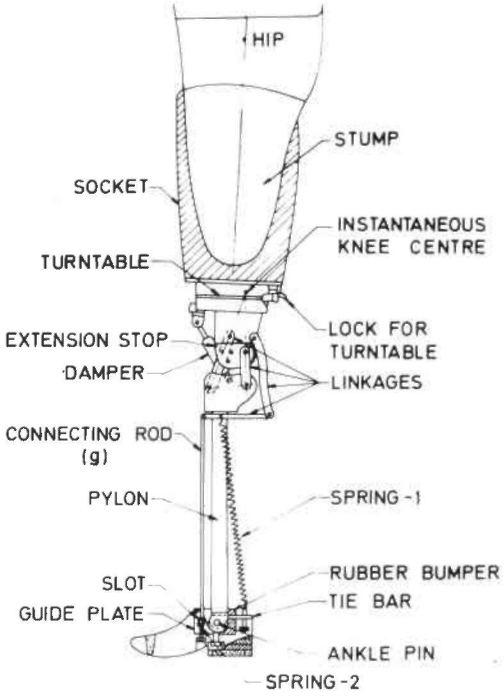

Fig. 1. Schematic diagram of the new modular six-bar linkage trans-femoral prosthesis.

muscles, flexion can be initiated and the instantaneous centre moves rapidly down to the natural position of the anatomical knee joint. Up to 10° of knee flexion, the centre of rotation is well above the location of a single axis knee joint, thus the amputee will be able to control both extension and flexion voluntarily over this critical range of motion. With this linkage arrangement a flexion angle of 150° can be achieved for squatting action.

Coordination of knee-ankle motion during squatting

To achieve the coordinated knee-ankle motion during sitting in the squatting posture two links 'e' and 'f' (Fig. 2) are arranged in such a fashion that the links 'e' and 'f' with the four bar linkages 'a', 'b', 'c' and 'd' form a six-bar linkage mechanism consisting of two loops CABDC and GFEBDG. An analysis of relative rotations between the linkages was carried out by solving two vector loop equations obtained from the dimensions and orientations of the linkages. With thigh motion, the relative rotation obtained between the link 'b' and link 'f' is transferred to the ankle by connecting a

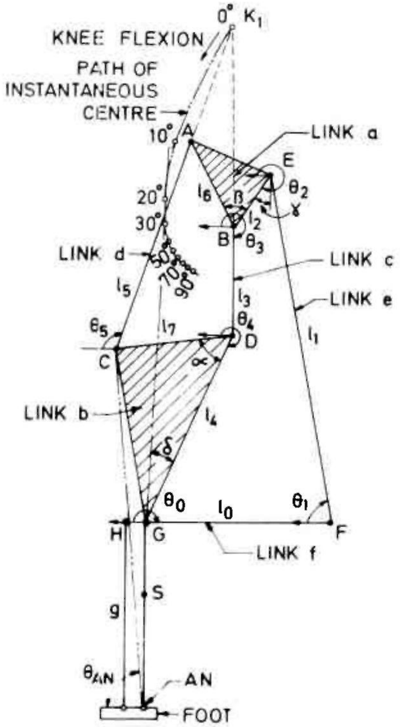

Fig. 2 Six-bar linkage knee-ankle mechanism used in the new prosthesis

rod 'g' in parallel with the pylon connecting the anterior end of the link 'f' to the foot unit. The position of the pivotal point E and dimensions and orientations of links 'e' and 'f' were selected after several trial solutions of the above mentioned loop equations so as to have a variation of knee and ankle angles with the thigh angles similar to those of normal persons.

The position of links in the mechanism and the locus of instantaneous knee centre for rotation of the thigh unit relative to the shank is shown in Figure 3. During squatting the upward movement of the lower end of the connecting rod 'g' is restrained by the upper end of a slot provided at the foot and the shank rotates forward with knee flexion to provide the co-ordinated flexion of the ankle joint.

Ankle dorsiflexion and plantarflexion during walking

During normal walking after heel strike the ankle undergoes plantarflexion of about 8° until the foot flat position, after which the shank rotates forward with foot flat on the ground to provide a dorsiflexion of about 13°, before heel off. With knee flexion after heel off, the ankle angle starts increasing by rotation of the foot pivoting at the toe. Dorsiflexion at the ankle joint after midstance facilitates a smooth pattern of walking. In the case of a SACH or conventional foot flexion at the ankle joint is very limited and the amputee has to follow an unnatural gait by raising the hip with much physical effort. In the present design to provide facilities of dorsiflexion after midstance and plantarflexion after heel off, the lower end of the connecting rod 'g' is allowed to move in a slot (Figs. 1 and 6) provided in the foot unit. Normally when the prosthesis is straight i.e. when knee flexion is zero the pin connected at the lower end of the rod 'g' touches the upper surface of the slot thus restricting the relative movement between the shank and the foot and thus behaves almost as a single foot-shank unit during the early phase of stance. The initial plantarflexion at heel strike is obtained by the compression of a rubber bumper provided behind the ankle joint. A spring (marked 2 in Figure 1 and marked 12 in Figure 6 inside the slot) and the rubber bumper facilitate a

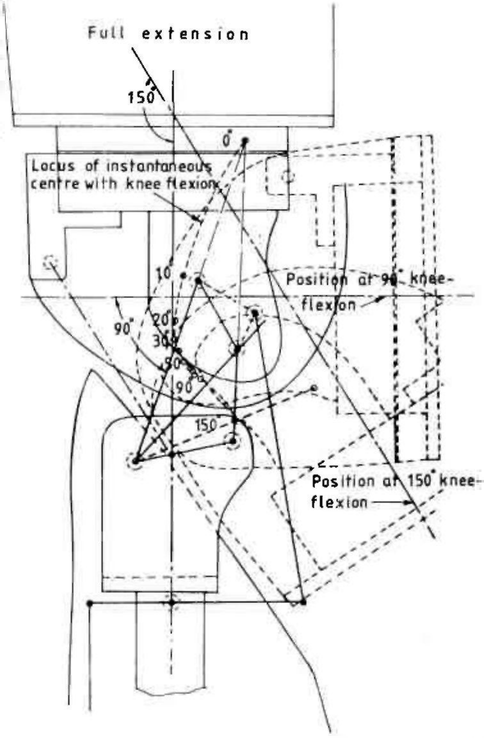

Fig. 3. Locus of instantaneous knee centre for rotation of thigh relative to shank up to 150° of flexion.

restraining action at ankle during plantarflexion. A relatively strong tension spring (marked 1 in Figure 1) connected to the pylon at a point posterior to the ankle joint facilitates rising from the squatting position and is also useful in minimising the inherent knee instability during flexion by resisting ankle dorsiflexion.

Although a total of eight links are used for transferring motion from thigh to foot, the prediction of motion of the prosthesis and estimation of forces in the pneumatic damper have been obtained from the kinematic analysis of the two loop equations for a six-bar linkage knee mechanism consisting of the linkages 'a', 'b', 'c', 'd', 'e' and 'f'.

Swing phase control

During the normal swing phase of walking the motion of the foot and shank is controlled by quadriceps muscles during flexion and by hamstring muscles during extension of knee. Quadriceps action restricts excessive heel rise after push off and provides acceleration in the initial part of swing phase followed by deceleration by hamstrings so as to have smooth entry (heel strike) into the next stance phase. In the trans-femoral prosthesis to achieve similar control during swing phase, a pneumatic damper may be provided between thigh and shank at the knee joint. This damper is basically a double acting cylinder with a piston moving inside. With knee flexion and extension, the air inside the cylinder is compressed and provides resistance to pendulum motion of the shank. On the top and bottom end of the cylinder, provisions are made for air leakage to achieve resistance characteristics similar to resisting moments developed in natural knee joint during swing phase.

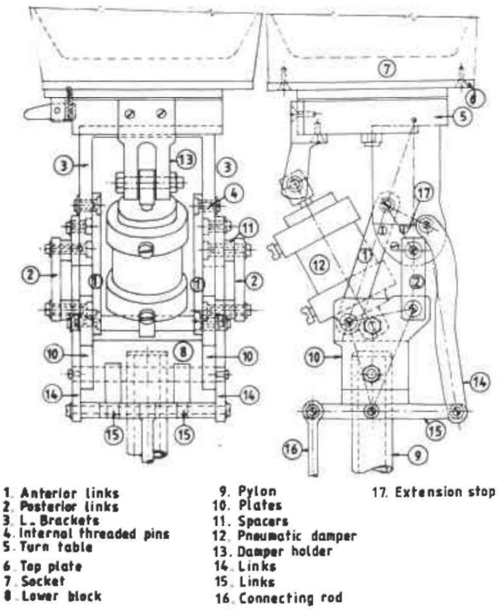

Fig. 4. Details of knee unit.

Cross-legged sitting

Sitting on the ground in cross-legged posture is a regular habit in Afro-Asian countries. A provision is made in the present design for cross-legged sitting with the help of a turntable located above the knee joint (in the thigh portion) of the prosthesis. A lock is provided to prevent motion of the turntable during walking. Before sitting in cross-legged posture the lock is operated manually to allow the rotation of the shank about the thigh axis.

Modular and endoskeletal design

The prosthesis is made of high strength aluminium and the different parts fabricated separately, are assembled together to provide an endoskeletal structure. The modular design by its fabrication facilitates mass production and replacement of parts of the prosthesis. The dimensions of different linkages are selected such that the endoskeletal structure can be provided with a soft cosmetic cover.

Details of the mechanical design of the prosthesis

The mechanical design description of different parts of the prosthesis are given below.

Knee unit

The knee unit (Fig. 4) consists of two sets of aluminium linkages (part numbers 1 and 2) connecting the upper and lower portions of the joint. The anterior links (1) are longer than the posterior links (2). The upper ends of the linkages are connected to two L-brackets (3) by specially designed internally threaded pins and screws (4), so that the joints can provide mobility during motion. The L-brackets are fitted to the back of the turntable (5). A thin plate is fitted to the top of the turntable to connect the whole unit with the socket (7). The socket is made of moulded resin cast from the positive mould of the amputee's stump.

The lower portion of the knee unit is fitted to the shank and it consists of an aluminium block (8) to which the pylon (9) is fitted. Two specially shaped aluminium plates (10) are fitted on both sides of the lower block. The lower ends of the linkages for knee rotation are connected to these plates. The planes of rotation of the shorter links and longer links are separated by spacers (11), so that the upper part of the knee joint can move freely relative to the lower part of the knee mechanism up to 150° of knee flexion without any interference from link motion. Space is provided in between the two L-brackets for fitting the swing phase control unit (12) which is connected to the upper portion of the knee unit at the desired position by a bracket (13). The lower portion of the swing phase control unit is connected to the block (8) of the lower portion of the knee unit.

To transmit the knee motion to the ankle, another set of links (14) and (15) are connected between the upper portion and the lower portion of the knee unit. The links (14) are connected at suitable selected points on the L-brackets and links (15) are pivoted at intermediate points of the pylon vertical axis. The two sets of links are connected together at their ends. The projected ends of the link (15) are connected to the foot and ankle unit by a connecting rod (16). An extension stop (17) is provided to stop hyperextension of the prosthesis during stance phase.

Turntable

The turntable consists of two aluminium circular discs; the upper one fitted inside the groove of the lower one to provide sufficient bearing surface (Fig. 5). Steel balls provided between the two discs act as ball-bearings and minimize friction and facilitate smooth relative rotation between them. A tension spring is fitted inside along a circular groove, connecting the top and bottom discs. When the bottom disc rotates counter clockwise relative to the top disc during cross-legged sitting, the spring is extended and helps, when it is released, the lower portion of the prosthesis to come back to its initial position. The turntable can rotate 110° of axial rotation for ease of sitting in the cross-legged position. In order to avoid any accidental rotation of the prosthesis about the long axis during walking a locking arrangement is provided. In the normal position of the prosthesis this lock will hold the two discs together. At the time of cross-legged sitting the lock has to be manually operated for unlocking. The upper disc and the lower disc are connected by a central bolt.

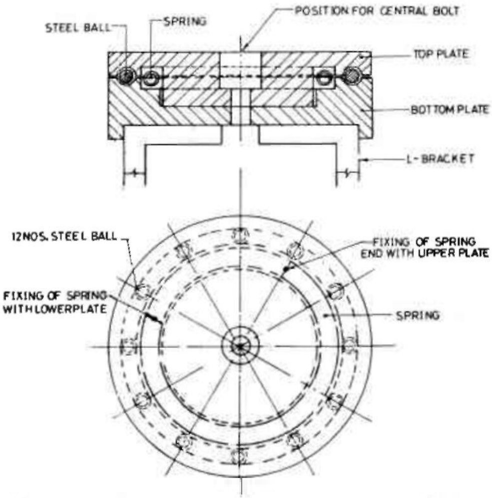

Fig. 5. Details of turntable used for cross-legged sitting.

Pneumatic damper

The pneumatic damper consists of an aluminium cylinder inside which a brass piston moves. The cylinder is closed by two aluminium caps on each end. The caps are each provided with a one way check valve to allow the air to enter into the chambers when the piston is moving away from the ends. When the piston moves towards the cap after compressing the air for a specified distance, a spring-loaded pin is pressed to leak the air through a throttling valve. The lower connecting point of the damper is adjusted so that the piston can be fully extended up to 60° of knee flexion after which the extension of piston will cease for further flexion of the knee during squatting. The force developed inside the cylinder can be adjusted by the flexion throttling screw or the extension throttling screw (as the case may be), thus providing the necessary resistance for control of swing. The fully extended length of the damper is 140 mm.

Foot and ankle unit

The conventional foot used in the trans-femoral prosthesis has generally a wooden solid ankle with the provision of heel cushioning and toe flexion by rubber blocks. In the present design

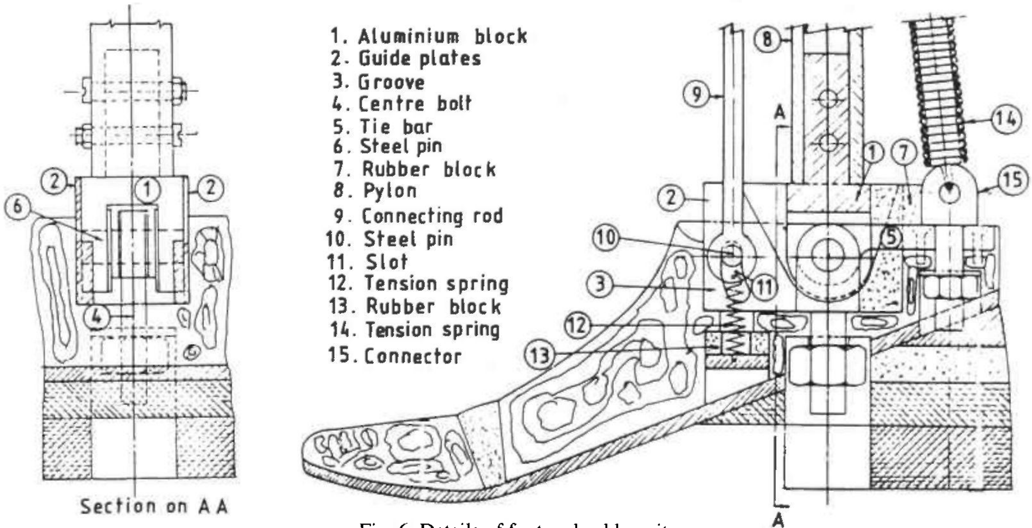

Fig. 6. Details of foot and ankle unit.

the foot has been modified to provide ankle flexion during walking and squatting (Fig. 6). An ankle unit fabricated separately, is fitted in a groove made in the wooden ankle block of the foot unit.

An U-shaped aluminium upper block (1) rests on two guide plates (2) which are embedded in a rectangular groove (3) on the foot unit at the ankle position. This upper block is connected with the foot unit by a central bolt (4) and a tie-bar (5). A steel pin (6) connecting the upper block, central bolt and a tie-bar is supported on both sides by the walls of the guide plates. The central bolt is fitted with the foot from the bottom side by a nut. The tie-bar is screwed to the wooden portion of the foot on the posterior side. This arrangement helps in resisting the horizontal force transmitted to the ankle joint during motion. A hard rubber block (7) is placed at the posterior side of the upper unit to resist the ankle plantarflexion. The pylon (8) is connected to the upper U-block. The lower end of the connecting rod (9) from the knee unit is connected to the ankle unit by a steel pin (10) which is allowed to move along a slot (11) made on both guide plates to allow ankle dorsiflexion and plantarflexion during walking. To provide additional resistance to ankle plantarflexion at heel strike, the free end of the rod (9) is loosely connected to the foot by a tension spring (12) and a plate (which is cushioned by a rubber block (13)). A tension spring (14) is connected from the pylon rod to the back side of the ankle unit by a nut and bolt arrangement (15) to provide the necessary resisting force to ankle dorsiflexion during squatting and to help the foot to rise after heel off during walking. The total mass of the prosthesis is 4 kg which is much less than the mass of the lost limb of the amputee.

Results

Figure 7 shows computed variations of knee and ankle rotations with thigh rotation as independent variable during squatting posture for the selected dimensions of the links of the prosthesis. The analytical graphs for the prosthesis mechanism for knee and ankle rotations are obtained by giving the thigh rotations obtained experimentally from normal walking as input to the loop equations. Corresponding normal patterns are superimposed on Figure 7 and it is observed that the pattern, given by the prosthetic mechanism closely follows the normal pattern of squatting.

The walking pattern of the amputee, wearing the trans-femoral prosthesis was recorded using a modified method of flickering light emitting diode system originally proposed by Soderberg and Gabel (1978) and as detailed below. Flickering light emitted diodes (LEDs) are fitted at hip, above and below the knee joint, ankle joint, toe and heel. In addition four LEDs (of different colour) are fitted at the pivoting points

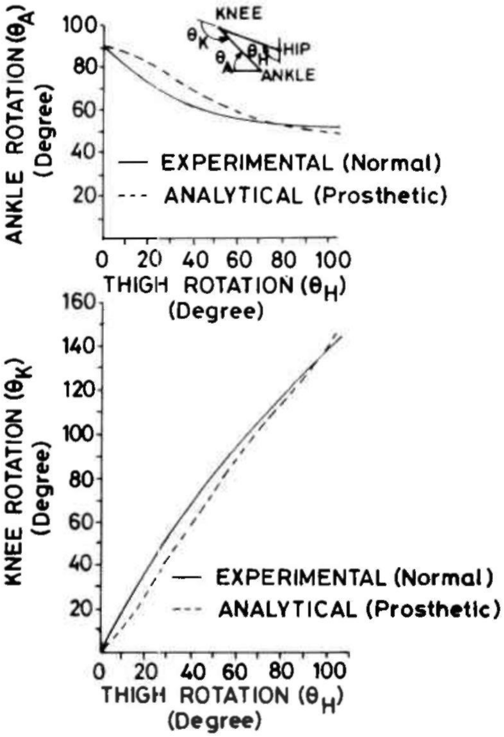

Fig. 7. Knee and ankle angle variations during squatting.

of the four-bar linkages (a b c d) to record the motion of the relative centre of rotation of the shank and thigh, required for the dynamic analysis of the prosthetic motion. The LEDs connected with the toe and heel switches are attached to thigh and shank portions of the prosthesis to record the temporal factors of walking. When the amputee is walking, motions of flickering LEDs are recorded on a single frame of a colour film using a still camera whose shutter is kept open in a dark room during one complete cycle of walking. The displacement pattern of hip, knee, ankle joints and toe of the prosthesis, recorded on the film, are projected on a screen and a stick diagram is drawn.

The direction of the line passing through the hip and centre of contact of the foot with the ground (obtained from the force platform, recorded simultaneously with the walking pattern) and the position of instantaneous knee centres for the rotations of the thigh relative to the shank of the prosthesis, are drawn for different phases of amputee walking (Fig. 8). The positions of instantaneous knee centres with reference to lines joining the centre of contact of the foot with the ground and hip joint for different phases, indicate that the design is suitable for providing stability during stance phase with little voluntary control of hip musculature and is also suitable for easy flexion during push off phase.

From the analysis of the six-bar linkage knee mechanism during prosthetic walking the motion of the different points of interest on the prosthesis can be predicted when the co-ordinates of the hip displacement and angular rotation of the thigh (obtained from LEDs

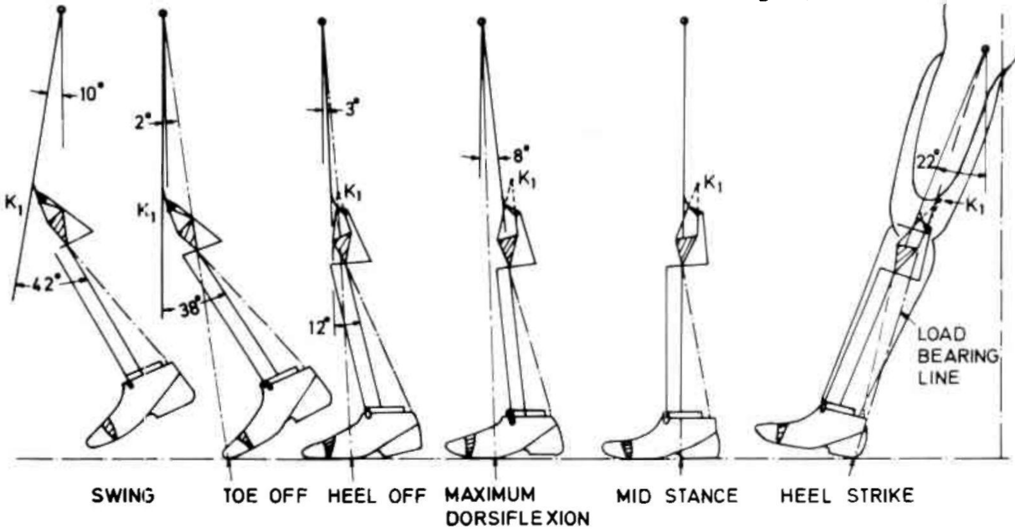

Fig. 8. Positions of instantaneous knee centre of the prosthesis with reference to the line passing through hip joint and centre of contact of foot with ground at different phases of walking

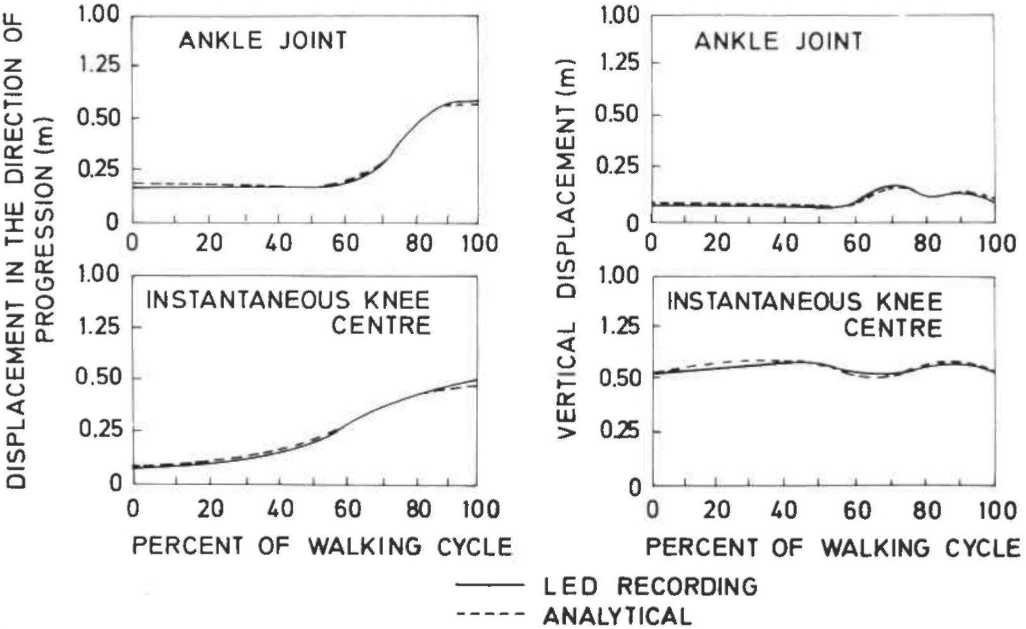

Fig. 9. Patterns of variations of horizontal and vertical components of displacements of the instantaneous knee centre and the ankle joint

study) are given as input. Figure 9 shows typical variations of horizontal and vertical components of displacements of the instantaneous knee centre and the ankle joint. The corresponding patterns obtained by LED recording of amputee walking pattern are superimposed for comparison. The results show close agreement between the analytical and experimental values.

A comparative study, of normal walking, prosthetic walking using the newly designed prosthesis with pneumatic damper and the old conventional single axis prosthesis, was carried out (Fig. 10). It is observed that the motion patterns are closer to that of the normal than the corresponding patterns for the amputee wearing the conventional prosthesis.

The pressure developed inside the two chambers of the pneumatic damper during knee flexion was recorded by properly mounted pressure sensitive transducers mounted on the flexion and extension chambers of the damper. The transducer is a piezo-resistive type and its resistance changes in accordance with the pressure acting on it above the atmospheric pressure. The output pressures are recorded simultaneously with the ground reaction forces obtained from the force platform during amputee walking. The pressures in the flexion chamber and extension chamber are in the order of 580 kPa and 210 kPa, respectively. A hardly audible noise (due to exhaustion of the air from the pneumatic chambers into the atmosphere) is heard during amputee walking and this does not cause any discomfort either to the amputee or the surroundings.

To examine the effect of the pneumatic damper on the walking pattern, the amputee walking patterns wearing the new prosthesis are recorded both for prosthesis fitted with and without pneumatic damper. Figure 11 shows the variations of (i) relative rotations, (ii) relative angular velocities and (iii) relative angular accelerations at knee during amputee walking fitted with the new prosthesis with and without pneumatic damper. It is seen that the new prosthesis with pneumatic damper has a relative knee angle pattern similar to the normal pattern (Fig. 10). But the prosthesis without pneumatic damper gives rise to higher relative angular accelerations of the knee (at the end of stance and beginning of swing phase) as compared to the corresponding values for the new prosthesis provided with pneumatic damper. Comparing the angular variations at knee for the new prosthesis with and without pneumatic damper, it is found that the prosthesis fitted with pneumatic damper has a pattern more closely

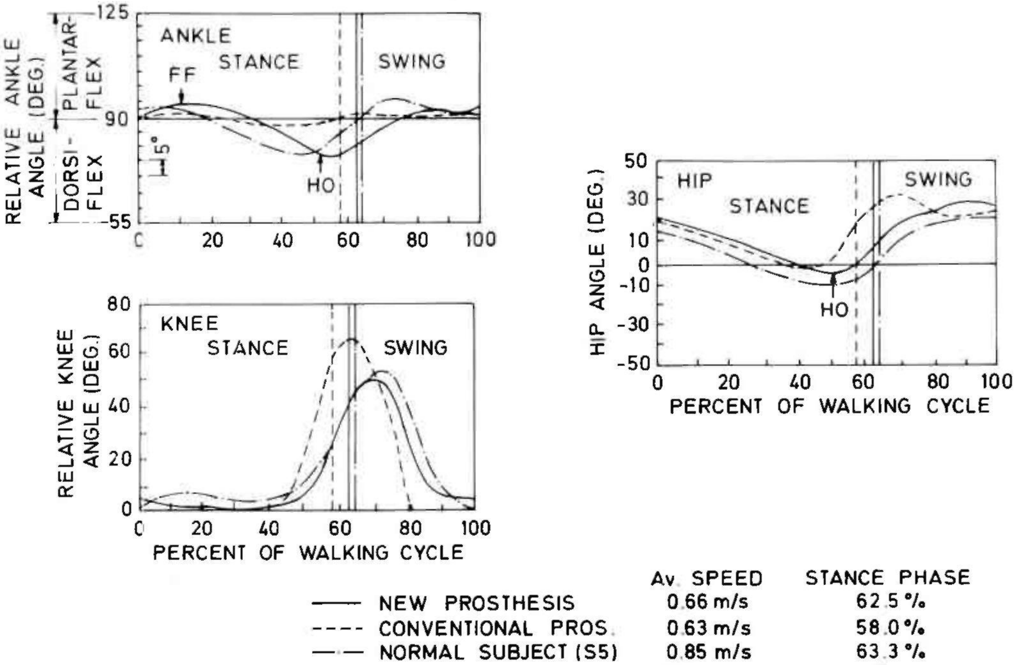

Fig. 10. Relative angular rotations of ankle, knee and hip during amputee walking with (i) new prosthesis and (ii) single knee axis conventional prosthesis and their comparison with the corresponding patterns for normal subjects.

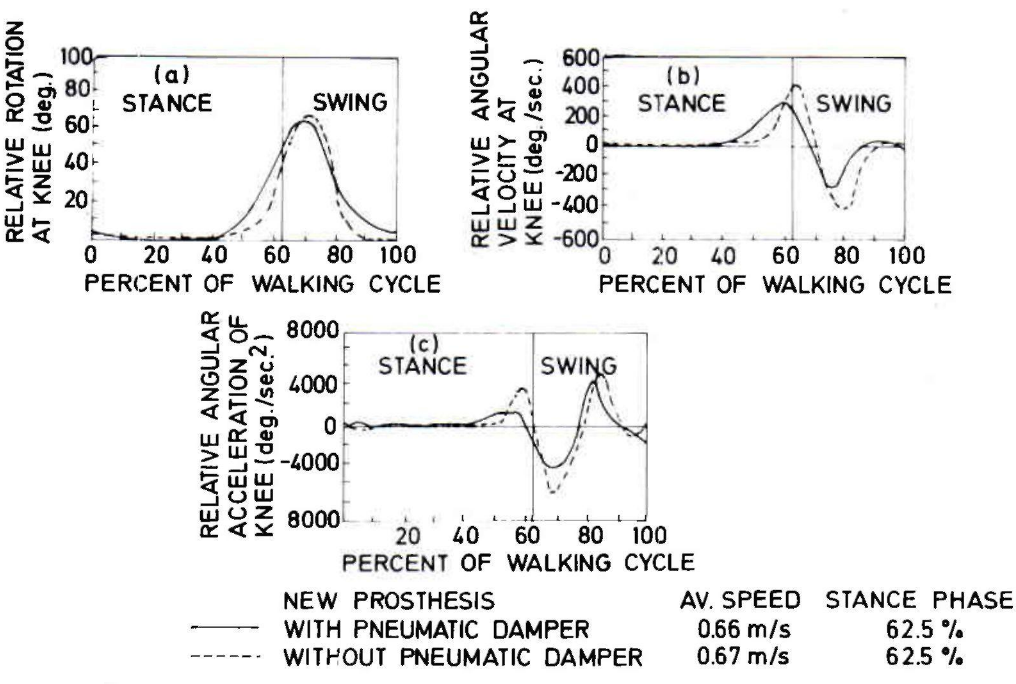

Fig. 11. Variations of (a) relative rotations, (b) relative angular velocities and (c) relative angular accelerations at knee during amputee walking with new prosthesis with and without pneumatic damper.

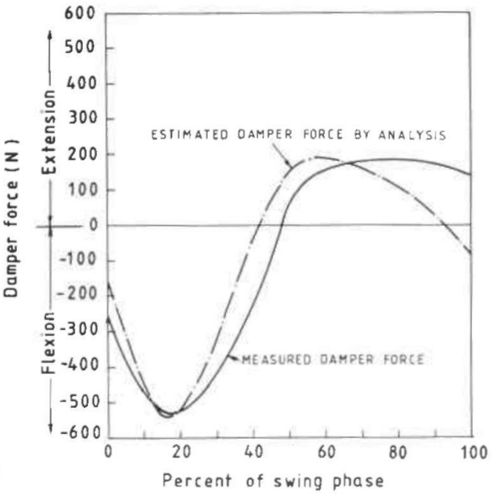

Fig. 12. Variations of forces in pneumatic damper during swing phase of walking obtained from dynamic analysis of prosthesis and experimental measurements.

following the pattern for the normal subject.

The damper force (obtained by multiplying the recorded pressure by the cross-sectional area of the respective chamber), is plotted as a percentage of swing phase (Fig. 12). Damper forces obtained from the dynamic analysis of swing phase of amputee walking is superimposed on the experimental measured pattern. The result shows a close agreement between estimated damper forces and the measured values.

Figure 13 shows the amputee wearing the new prosthesis in walking, squatting and cross-legged sitting postures. It is observed that the new prosthesis is able to provide considerable improvement in reproducing normal motion patterns of ankle, knee and hip angles as compared to a conventional trans-femoral prosthesis.

Conclusions

Positions of the instantaneous knee centre of the prosthesis with reference to the hip ankle reference line drawn at different phases of walking showed stability of the prosthesis during stance phase and ease of flexion for the initiation of swing. The motion patterns, of ankle, knee and hip during walking and squatting, obtained experimentally for the six-bar linkage mechanism trans-femoral prosthesis, were compared with the corresponding patterns obtained for normal subjects. The closeness in the two patterns establishes the suitability of the new prosthesis for producing near normal patterns of motion during walking and squatting.

The cross-legged sitting feature makes it functionally suitable for Afro-Asian amputees. The modular nature of the prosthesis enables variation of the size of the pylon very easily to suit different heights of amputees and makes the mass production of parts possible, thereby potentially reducing the cost of production. The endoskeletal nature of the prosthesis facilitates

(a) (b) (c)

Fig. 13. The amputee (a) walking, (b) squatting and (c) sitting with the new prosthesis.

the provision of a suitable cosmetic cover in the future development of the prosthesis.

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