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A KINEMATIC EVALUATION OF AN EMG-CONTROLLED ABOVE-KNEE AMPUTEE PROSTHESIS

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ABSTRACT

Rehabilitation of an above-knee (A/K) amputee relies upon a well-designed prosthesis to give him functional support, and aesthetic posture during ambulation. To evaluate such a device requires thorough understanding of the bio-mechanics involved and extraction of kinematic information.

The success of electromyographic (EMG) control in powered upper-limb prosthesis has prompted the use of EMG control in a lower-limb prosthesis which varies the amount of viscous damping across the knee joint. In a former project, an EMG-controlled prototype was built, and positive results were noted.

In this study, the hydraulic and electronic designs of the original prototype were improved and modified to make it suitable for long term testing. This was followed by extensive evaluation which includes data acquisition and analysis on several experimental runs conducted during a training period. The performance of gait is compared to available data.

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CHAPTER 1

INTRODUCTION

Physiological deficiencies as a result of trauma, disease, congenital anomalies, and aging can lead to loss of lower or upper extremities. Replacement of the extremities requires prosthetic devices which can eventually duplicate the normal function of a healthy knee joint.

The case of an unilateral above-knee (A/K) amputee deserves extra attention because it involves factors common to all lowerlimb amputees. It provides perhaps the best example in the locomotion study of prostheses.

The majority of A/K prostheses in use today employ mechanical friction in the knee joint design, which does not permit the most efficient or normal appearing gait. Improvements have been achieved with mechanically 'programmed' hydraulic and pneumatic units; that is, the amount of damping resistance in the units is predetermined. However, functions like varying cadence, and walking up or down stairs, on ramps, and over rough terrain still cannot be performed in a normal manner. The use of myoelectric signals to control knee actions has been demonstrated at least to some extent by Horn (1972), who designed a knee unit which can be locked to flexion voluntarily by the wearer and is unlocked automatically at full extension, allowing for free swing. This design allows for weight bearing on a flexed knee but does not allow for varied cadence.

An EMG-controlled A/K prosthesis has been developed at the Univ. of Manitoba based upon data from normals. A hydraulic damper was built inside the knee unit, which can vary the knee action from 'free swing' to 'lock' and from 'lock' to 'free swing'. This variation is achieved by controlling the resistance of fluid flow around a damping cylinder with bypass valves through electromyographic (EMG) signals from the hamstrings of the stump (Dyck, 1974). A schematic of the prototype is shown in

Fig. 1.1. It consists of a hydraulic damper, whose resistance is varied by the control of two solenoid valves and one check valve. The check valve opens only during the extension phase, as locking is desirable only in flexion.

The mechanical and electronic portions of the prototype built in Dyck's work was bulky, and the electronics did not work reliably. Therefore, a new improved version was designed. This new version was then benched tested and tried by the amputee.

The main purpose of this study is to perform extensive evaluation on the new unit. To do this, the amputee experimented with it once every other week for approximately six months. During this period, locomotion runs were monitored via an existing video-computer system (Winter, et al, 1972). Based upon the kinematic information extracted, velocity and acceleration profiles of the limb segments could be obtained. When these profiles are coupled with anthropometric data, parameters such as force, torque, and energy levels could be calculated. Necessary computer programs were rewritten to accommodate the analysis of asymmetrical gait. The performance of the amputee-prosthesis system is then compared and analyzed for different control conditions, and tasks such as walking on ramps and climbing stairs.

It was very difficult to establish concise criteria for evaluation because of the large amount of data and the number of variables present. In spite of these difficulties, results of this study strongly indicate that a controlled EMG pattern can be conditioned through training in such a way that subconscious control is possible. Performance of the amputee on this prosthesis was found to be satisfactory.

CHAPTER 2

BACKGROUND

(I) Walking cycle

Essentially, a walking cycle consists of a stance phase, which constitutes approximately 60% of the cycle; and a swing phase, which constitutes approximately 40% of the cycle. The stance phase begins when the heel makes contact with the ground (HC) which is positioned as the 0% mark. The foot flattens (FF) at about the 20% mark; then by the plantar flexion of the foot and the initiation of knee flexion by the forceful action of the calf muscle, the body is propelled forward as the heel leaves the ground (HO) at about the 40% mark. The stance phase ends when the entire foot rises from the ground. The swing phase begins with toe-off (TO) at about the 60% mark, and ends when the heel strikes ground at the 100% mark again.

At the end of the swing, there is a period of deceleration when the forward motion of the leg is restrained by the stretching of the hamstring muscles to control the position and velocity of the foot immediately before HC. The period when both legs are on the ground is called the double support phase (DS). These parameters are shown on Fig. 2.1. The duration of the DS period is related to cadence, and the shortening or the absence of DS indicates that a person is running rather than walking.

(II) The general walking pattern of A/K amputees

It appears that in walking, the body is constantly converting energy from potential to kinetic and then back to potential energy again (Inman, 1965). The path of centre of gravity (CG) has been described as the intersection of the horizontal and vertical displacements shown in Fig. 2.2 during level walking of normals (Saunders, 1953). That is to say, the CG is continuously displaced sinusoidally in both vertical and horizontal directions. Therefore normal gait is harmonious, and energy among body parts can be balanced and transferred.

The gait pattern of A/K amputees varies considerably from normal. An A/K amputee has to elevate his body in order to keep the harmonious path of CG just before HC of the artifical leg when the trunk is descending. If the landing leg were the normal leg, such action could be accomplished by planter flexion of the ankle and flexion of the knee. Due to the deficiencies present in both ankle and the knee of the artificial leg, a different pattern results in angulation of the natural knee and ankle, or the tilting of pelvis (see Fig. 2.3; Eberhart, 1968).

Until the prosthetic heel rises, the leg rotates forward about the ankle joint of the foot. The knee is consequently in an extended position at the time the hip joint reaches its maximum elevation. This increases the vaulting sensation.

At this time, the hip joint is descending while the prosthetic leg is still on the ground. As it is unable to control stability, the normal leg is called upon for aid by landing earlier on the ground.

During the swing phase, the forward swing of the normal leg is initiated by powerful contraction of the plantar flexor of the ankle. Lack of such a mechanism in the prosthetic leg forces the amputee to rely heavily on his hip muscules for initiating the forward prosthetic swing. Fortunately the hip muscles are under voluntary control and consequently can adapt themselves within limits.

The lack of muscular control at the knee joint severely reduces its ability to create the normal knee moments. The A/K amputees compensate by shortening the steps in order to reduce the moment developed to flex the knee (Peizer & Wright, 1974). The inability of the knee to flex during early stance phase increases the amount of energy required to walk. This mainly results from the excessive rise of centre of gravity during the prosthesis stance.

(III) Basic requirements of A/K prosthesis

Essentially, an A/K design requires that the prosthesis does not collapse during the stance phase. This does not mean that locking the knee throughout the entire stance phase is optimum. In normal gait, the knee is fully extended at heel contact. It then flexes slightly (accompanied by planter flexion) and re-extends during weight bearing. This action allows the hip to follow a more harmonious path and therefore requires less energy. Conventionally, prostheses are aligned in such a way that the centre of rotation of the knee is dorsal to the load line (see Fig. 2.4), so as to prevent the knee from collapsing during weight bearing. This alignment applies a hyperextension to the hinge, locking it against the stops.

During the swing phase, an A/K prosthesis can be considered as a pendulum subjected to driving forces by the hip and thigh. At the beginning swing, a forward acceleration acts on the upper part of the prosthetic leg (socket portion and thigh muscles). If no resisting moment is supplied by the knee mechanisum, an excessive heel rise will occur. Also, at the end of swing, the shank will have a rotational velocity and must stop abruptly. Both problems become more serious as walking speed increases (see Fig. 2.5). Therefore, the minimum requirement is that some damping must be provided during the swing phase.

One thing to note is that, during the double support phase, the energy of forward translation of human body comes from a force couple, that is, from the push-off and the deceleration of the swinging leg. The ratio of the components from push-off and deceleration of the swinging leg has been found to be approximately five to eight (Inman, 1965). This phenomenon explains that if a prosthesis is too light, the amputee will have difficulty in developing sufficient kinetic energy at the end of the swing phase to be fed back into the system to maintain the forward velocity of the body.

Other factors that require consideration are: safety and reliability; flexibility for sitting; whether it is cosmetically acceptable; and if the energy cost is at a minimum.

(IV) Designs for A/K prosthesis

A) Constant friction

This provides friction in the knee bolt, which rotates with the shank, and a mating surface fixed to the upper leg. Variation of the resistive torque is adjusted by the contact force in the brake. "Bumper" springs have been provided to help control deceleration at the end of swing. This system provides a relatively constant resistive force independent of walking speed and angle of flexion.

B) Intermittent friction

This provides a resistive torque that is a function of the knee flexion angle. Mechanical friction is provided by the pressure between the three disks mounted concentrically with the long axis of the knee bolt. The resistance offered by each individual disk is varied by a wheel during different intervals of the swing phase. Essentially it provides a better torque profile than the constant friction type. It functions best only at one cadence and is purely dissipative.

C) Fluid control designs

Recently, mechanisms having orifice-flow damping have received much attention. The Henschke-Mauch (Mauch, 1968) and Dupaco "Hermes" units are well-developed examples. They contain flow channels and orifices in the piston cylinder, so that the moving piston can sucessively block off escape of the fluid and thus vary the resistance throughout the swing phase in order to approximate the 'ideal' moment curve. Check valves switch fluid flow to separate channels for extension and flexion, thereby allowing independent adjustment of flexion and extension resistance.

D) Pneumatic control design

An air dashpot can be used as a resistive element. The main advantage is that fluid leakage is not a problem and therefore the cost of manufacturing is lowered. The air in the cylinder serves both as an energy dissipator and a spring.

E) Polycentric knee

The most promising mechanism at present uses four-bar linkages together with fluid damping (Radcliffe, 1957). Since the action of a normal knee joint is also polycentric, the four-bar linkage provides different instantaneous centres (usually are very small distances apart) with relative changes of the positions of the limb segments.

Besides satisfying the stability requirement as described by Radcliffe (Radcliffe, 1957), the polycentric action also contributes somewhat to the sense of knee position as in normal individuals.

All of the mechanisms described above provide predetermined amount of damping resistance, and the control is passive.

F) Myoelectrically controlled knee locking mechanism

The design by Horn (Horn, 1972) uses myoelectric signals to lock the knee at any desired angle. Once locked, the knee can only extend, and unlocks itself at full extension. The advantage of this system is that a wearer can bear weight on a slightly bent knee, allowing for a more normal appearing gait. Horn also claims that it can be used for climbing stairs. This design does not, however, allow the amputee to vary the resistance as a function of the knee angle, nor to control this resistance in order to accommodate different gaits.

CHAPTER 3

EXPERIMENTAL DESIGN

(I) Hydraulic system

The hydraulic system of this study is a modification of the work done by Dyck (Dyck, 1974). A hydraulic damper with a discrete damper control is considered adequate to be implemented with a proportional myoelectric controlled scheme.

In Dyck's work, data from normals was gathered to determine the required characteristics for this hydraulic system, such as the pressure rating of each component, the rates of flow, the resistance to flow versus percent walking cycle, and the number of bypass valves and their resistance. Also in Dyck's work, with the knowledge that a valve's resistance to flow is characterized by its C_v factor (which is defined as the flow of 60^{o}F water in gal./min. through the valve with one p.s.i. pressure drop across it), the desired C_v needed across the damping cylinder was calculated and adjusted for the desired hydraulic fluid.

By curve fitting, he then determined the number and type of valves needed . Fig. 3.1 shows his resultant C_v curve versus walking cycle plot which calls for two solenoid bypass valves and one check valve, all in parallel with the damping cylinder. Dyck's design used two solenoid valves, V1 and V2, with C_v factors of 0.02 and 0.1, respectively, while the check valve has a C_v factor of 0.08.

The hydraulic prototype used in Dyck's project was made up of several individual components - a hydraulic cylinder, two electrical actuated valves, and a check valve. The hydraulic lines were connected together externally using elbows, unions, and plastic tubings as they were intended for temporary usage. In this study, it was felt that a more compact and permanant assembly was required for a prolonged period of training. Therefore, a new version of the hydraulic prototype was designed to reduce the bulkiness of the unit. It has all the components mounted into a single unit: the check valve was built into the piston , while two new bypass solenoid valves were mounted on the side (in parallel with the hydraulic cylinder) to be controlled by EMG signals.

The schematic and appearance of the new hydraulic unit are shown in Figs. 3.2 and 3.3 respectively. These figures also show two external valve screws in which adjustment can be made on the size of the orifices of the two bypass valves, either to increase or decrease the resistance to flow.

With this arrangement, the amputee is given a wider choice of resistive damping settings by selecting different combinations of the closure of the two bypass valves and the check valve according to his needs. With the maximum settings on the two bypass valves, the Cv factors for the designated High Resistive Valve (HRV), and Low Resistive Valve (LRV) are approximated to be 0.04 and 0.14 respectively, with the Check Valve (CHV) at 0.16 (see Appendix A). A coil spring was added to the top of the cylinder to provide energy storage which supplies an extra lift during push-off.

In the rest state, both solenoid valves are normally open; and while standing, the amputee will achieve his stability by prosthesis alignment. In the flexion mode, since the CHV is inoperative and since the two solenoids have different C_V factors, there exist three additional controllable levels of resistance - one of which is 'lock'. During extension, all corresponding levels of resistance will be decreased due to the release of the bypass CHV. These combinations (Fig. 3.4) constitute the same 'four-mode' operation as described by Dyck (1974).

(II) Myoelectronic system

The basic concept of myoelectronic control is that when a muscle contracts, the contraction is accompanied by an electrical signal, called a myo-electric or electromyographic signal (EMG). This signal can be detected by placing electrodes on the surface of the skin. One of the most useful characteristics of the EMG signal is that its root-mean-square (RMS) value is proportional to the isometric contractal force of the muscle involved. An EMG signal processor had been designed by Dyck (Dyck, 1974) to control the operation of solenoid valves utilizing such a characteristic. However, repairs on this circuit were needed constantly because of a poor wiring technique was employed. Therefore it was decided that a new printed circuit (PC) was to be built with a reduced size in mind. A block diagram of this processor is shown in Fig. 3.5, which is basically derived from Dyck's design. Also, the new circuit is shown in Fig. 3.6.

The typical EMG signal has peak amplitudes from 100 mV to 4mV and lies in the frequency band of 20-500 Hz. Therefore, it requires high gain amplification and 60 Hz.

noise rejection at the front end. The first stage of the processor consists of two high input impedance instrumentation amplifiers (AD 521, a data sheet is provided in the Appendix B). The first one acts as a low gain differential amplifier (with a gain of 20), which has a built-in feature of high common mode rejection ratio (CMRR) rated at 90 db to 60 Hz noise. As the skin-electrode interface generates DC potentials which may be up to 100mV in magnitude, it is filtered out by means of a high-pass filter with a corner frequency (f_c) of around 15 Hz, in order to prevent saturation before entering into the second high gain amplifier. The second amplifier has an adjustable gain of 100 to 1,000; therefore the total amplification can be set from 2,000 to 20,000.

The second stage is a full wave rectifier, the output of which is a signal whose DC component is proportional to the EMG signal's RMS value. To obtain this DC component, it is fed through a second order low-pass filter (LPF). The output from the LPF is a DC voltage that corresponds to the level of muscular tension.

In order to monitor this DC voltage produced at the integrator, three predetermined reference voltage levels are set up in a quad-comparator. Whenever the incoming signal is higher than the reference levels, the comparator(s) will be activated. With three different reference levels, there are four possible parallel selections, namely: (i) no comparator is ON, being the first or rest mode; (ii) comparator I is ON, being the second mode; (iii) comparators II and I are ON, being the third mode; (iv) comparators III, II and I are ON, being the fourth mode.

By utilizing an Exclusive-Or logic, it is possible to represent each mode by a single gate. No input represents the rest mode. The outputs of comparators I and II are fed into the Exclusive-Or Gate (Ex-Or) I, so Gate I is high only when comparator I is On. Outputs from comparators II and III are fed into Gate II, Gate III, respectively. These two signals are gated with V^+ separately. By these arrangements, the 'four-mode' operation is fully represented.

Finally, Gate I is 'Or'-ed with Gate III. Thus, the High Resistance Valve (HRV) is energized when either the second or fourth mode being activated. This leaves Gate II the sole control of the Low Resistance Valve (LRV), and the four-modes of operation can be achieved (see Fig. 3.7).

In order to eliminate the pick-up noise, the circuit was mounted on a printed circuit (PC) board. A power pack of 10 rechargeable "C" size batteries, together with the PC board, were placed inside a metal chassis. This metal chassis, which is about half the size of a tape-cassette recorder (see Fig. 3.8), was then mounted on a belt to be worn by the amputee.

Considerable problem was encountered with the batteries because of the packaging technique. The commercially available battery holders do not give good contact with batteries, causing noticeable contact resistance. At the time when the solenoids are energized, they draw 1 ampere of current through the contacts of the battery holders. Thus, the contact resistance drops the supply voltage by such an amount that the whole circuit became inoperable. One obvious solution is to add another battery which would then increase the weight and size of the package. Such an idea was immediately discarded and the problem was finally resolved by a custom-built battery holder which gives good contact with the batteries. Voltage drop across the batteries under load due to the internal and contact resistance is then kept minimal.

(III) Interfacing the hydraulic and the electronic systems

In order to achieve the C_V factor curve in Fig. 3.1, certain criteria had to be considered. At the end of the swing, a moderate resistive torque is required to decelerate the rotating shank. During the weight bearing period in the stance phase, an additional amount of damping is required to minimize the tendency to collapse while the knee is flexing in order to transfer weight to the other leg. The required schedule and the resulting C_V curve are shown in Fig. 3.9 and Fig. 3.10. According to this schedule, the level of EMG activities is required only to change four times. In fact, the amputee is required only to exert tension on his stump muscle only once per cycle for a prolonged period. He only has to remember to begin this process at the end of the swing in order to energize the first solenoid valve (HRV), with increasing effort at mid-stance to energize the second solenoid valve (LRV) (which can be accomplished with minimum effort as the whole weight of the body is pivoted on the prosthetic leg at mid-stance). Then when he releases the muscular tension gradually, the solenoids will be de-energized one after the other until the end of stance. By that time, he is bringing forth his stump for another free swing. This minimizes the requirement as prescribed by Dyck (1974) where muscular control has to be activated and de-activated twice within one walking cycle.During training, the amputee finds the present method very natural to proceed as it is rhythmic and changes in muscular activities are gradual.

(IV) Locomotion data acquisition

Experimental runs are monitored by an existing video-computer system developed by Winter, et al (1972). The major advantage of this system is that the data collected is immediately available for processing by the use of digital computers equipped with appropriate interface devices. Early analysés were done by manual photography and film processing, and data extraction of this kind was tedious and painstaking. In this study, foot-switch data were also taken to obtain temporal relationship between the left and right legs.

A) Foot-switch data

A pair of foot-switches were mounted under each shoe corresponding to the positions of toes and heel. When either part makes contact with the ground, a switching operation was carried out inside a pulse circuit, and the output is recorded through telemetry on a chart recorder. Therefore timings of events such as heel strike and toe-off can be related to the other foot with simultaneous recordings. By averaging several strides, the phasic relationship between the legs can be expressed as the percent walking cycle of the other.

To make certain that the correct sequence of solenoids being energized was achieved, pulses corresponding to energizing and de-energizing solenoid were recorded simultaneously with the foot-switch data. A typical amputee record is shown in Fig. 3.11. The recordings on the operation of the solenoid valves are made possible by transmitting signals from two differentiating circuits attached to the valves (see Appendix C). The resulting pulse with a large amplitude corresponds to the ON (energizing) operation, and the small pulse corresponds to the OFF (de-energizing) operation.

The results show that the swing phase of the intact leg is much shorter than that of the prosthetic leg due to the fact that the amputee feels less secure when his better leg is in the air, and he always tries to return it to the ground as soon as possible. The valve operation shows that the HRV was energized at the end of the swing and last after mid-stance, while the LRV was energized only once during mid-stance during one normal level walking cycle. These two valves were de-energized prior to toe-off to give the amputee a free swing.

These recordings, however, do not show whether the knee was flexing or extending, which would affect the fluid flow significantly due to the action of the check valve. Nevertheless, the general trend of movement of the prosthetic leg duplicates that of the intact leg in the swing phase, and angular changes during the stance phase are minimal. So the model developed from the data of normals is still valid under these circumstances. Angular changes in the knee and ankle will be explored further in later sections.

B) Video data

Body markers, consisting of halves of ping-pong balls covered with reflective tape, were attached at anatomical landmarks with double-sided adhesive tape (see Fig. 3.12). Background markers consisting of larger discs of reflective tape provide a spatial reference for the TV recording system. The amputee was tracked using a TV camera and monitor mounted on a cart at about three meters to the side (see Fig. 3.13). The lighting was adjusted to provide a very high contrast image which enables a one-bit conversion of this data into the computer (1 for white, inside a marker; 0 for black, outside a marker). Fig. 3.14 shows a computer print-out of one converted TV field, and the resulting trajectory plot of the spatial movement of body markers is shown in Fig. 3.15.

The TV recording is replayed on a video tape recorder and converted into digital format via a TV-computer interface. Computer programmes reduce the converted data, cluster the points from each marker and calculate the absolute x,y co-ordinates or the geometric centres of the markers. Because of the presence of high frequency noise, a second order digital Butterworth filter having a cut-off at 6 Hz was used to reduce the noise level (Winter, et al, 1974b).

Since the present video system consists of a single camera, in order to obtain a complete data set for an asymmetrical gait, video data on each leg of an amputee have to be recorded independently. During the locomotion runs, video data were recorded. They were then analyzed after conversion to determine the heel contact points according to the method described by Winter,et al (1974). Positional data of a single stride was normalized to 100% of a walking cycle by computer programs (see Appendix D). From the knowledge of the temporal information obtained as described earlier,digital data sets from both legs could then be aligned to form a complete data set to be stored on a direct access file.

By sucessive differentiation of the co-ordinates, velocities and accelerations of each marker can be obtained.

The location of the centre of mass of each limb segment was determined from anthropometric data provided by Contini (Contini, 1972) (see Fig. 3.16) and by relating this data to the location of the markers. For example, the x,y co-ordinates of the centre of mass of the shank are:

$$X_{sh} = X_4 + a(X_3 - X_4)$$

$$Y_{sh} = Y_4 + a(Y_3 - Y_4)$$

where a is the fraction of the distance that the centre of mass is located between marker 3 and 4. Absolute limb angle of the shank can be obtained by the simple relationship:

$$\theta_{sh} = \tan^{-1} (Y_3 - Y_4 \ / \ X_3 - X_4)$$

Angular velocity $\dot{\theta}$ and angular acceleration $\ddot{\theta}$ are obtained similarly by successive differentiation.

From the knowledge of the above kinematics and using the anthropometric information of the subject, computer programs have been developed by Winter's group to determine the joint and floor reaction forces, segment energies, joint torques, instantaneous energy and the power of each limb. In this study, they have been either modified or rewritten to accomodate asymmetric gaits. The results are being compared for symmetry and efficiency.

(V) Kinematic data runs

The purpose of this study is to evaluate an experimental prototype based upon known kinematic data of normals and other A/K prosthesis, and the effect of training on performance. The total period of testing and training lasted for about six months. However, the actual usage time the amputee spent on the prosthesis amounted to approximately 10 hours. Locomotion runs, which include video and foot-switch data, were recorded at the beginning and the end of this period. They are, for analytical purposes, classified under six different catagories which are shown in Table 3.1. These runs were taken from a single amputee during level walking at his most comfortable speed. The amputee had participated in an earlier project. He is in his fifties, his right leg was amputated in 1965 approximately 6" above the knee. Presently he wears Henschke-mauch model A/K prosthesis.

(1) Experimental prosthesis - during early training

This is the run taken at the early stage of training after he has managed to control the electronic device and has got used to the experimental prototype. An alignment unit (Staros-Gardner coupling) was installed for easy alignment purpose.

(2) Commercial prosthesis

This is a control run taken on the amputee when he was wearing his regular Henschke-Mauch model prosthesis. This prosthesis is considered to be a good but expensive model available for A/K amputees. Performance is expected to be above average, as the amputee has been wearing it for years.

(3) Conventional prosthesis with no damping:

This data set represents the experimental leg with the damping cylinder taken out. The amputee has to walk cautiously and extremely slowly in order to prevent falling. Therefore only small driving forces are expected at the prosthetic knee joints and the resulting shank motion will probably be similar to that of a frictionless compound pendulum.

(4) Experimental prosthesis with control electronics disabled:

This data set is similar in all respects to the Run-1, except that control electronics was turned off during walking. With no electronic control, the HRV and LRV are always open while the check valve opens only during extension. This operation provides a form of passive control as contrasted to the EMG activated control.

(5) Experimental prosthesis - after training:

Seeing that the amputee was always tired out after ten minutes of practising, he was asked to try out the prosthesis without paying attention to whether he has been able to turn on the correct sequence of valves operation. It was found that he could walk on the prosthesis in a much relaxed manner. Because of the change in the method of controlling the device, the amputee showed inconsistency in managing the walking process. It was then decided that further training was required.

During the last stage of training, not much improvement was observed, and the alignment unit slipped quite often to cause misalignment. To improve the performance, the socket was shifted forward in hope of getting a better mechanical advantage. But at the same time, the load line was shifted just behind the knee centre so that the geometric lock was eliminated, which meant that between heel contact and early stance, the only way that the amputee could stabilize the knee was by activating the EMG-controlled device. Much improvement was observed visually. The socket was finally molded in permanently after the prosthesis was properly aligned. Data were taken afterwards.

(6) Experimental prosthesis - fast cadence

This data set was taken at the same time with Run-5 except that the amputee was asked to walk at a faster pace.

CHAPTER 4

OBSERVATION AND EVALUATION OF DATA

A complete evaluation of a prosthesis is a task that involves defining a set of performance indices,followed by a subsequent rating of the prosthesis against these criteria. Since there are so many parameters which are pertinent, the evaluation process is very difficult. Therefore in order to determine the effectiveness of this new prosthesis, analyses were focused on the following areas to extract various aspects of gait from the recorded data: (I) Gait pattern;

(II) Vertical displacement profile of body markers;

(III) Angular changes in the limb:

(A) Profiles of limb angles,

(B) Velocities of the knee and shank angles,

(C) Accelerations of the shank angles;

(IV) Knee moment during the swing phase;

(V) Energy in the limbs;

(VI) Task performance.

The results are compared to available published data in order to determine whether this prosthesis is suitable for daily usage. Generally, results from one aspect of gait are either the direct result of, or reflected in the other aspects. Since Run-5 represents the present final stage of this new prosthesis, most of the comparisons are made only with respect to, or between both legs of Run-5 in order to simplify the analyses. The comparisons considered in this study are listed in Table 4.1. The results from the analyses performed will then indicate the presence or absence of features which are crucial to a better performance.

I) Gait Pattern

Experiments conducted by Bard & Ralston (Bard, 1959) and Radcliffe & Ralston (Radcliffe, 1963) to measure the metabolic energy required for normals and A/K amputee indicate that individuals will adopt an optimal speed of walking that is very nearly equal to that at which minimal energy expenditure occurs. For A/K amputees, this minimum occurs at what is called their 'natural' or 'comfortable' speed of walking, a speed at which they can maintain indefinitely without physiological embarrassment.

A paper by Murray, et al (Murray, 1966), contains extensive data on normals during normal and fast walking. Their results conclude that normals accomplish the forward speed of faster walking by increasing both the stride length and the cadence. They also have shown that the amplitude of vertical displacement of the trunk increases as a result of fast walking.

Data on gait pattern on a large group of A/K amputees are also available through the work of James &

Öberg (James, 1973). Their results show that A/K amputees walk at a slower speed than normal individuals (38% less) as referred to by Murray, 11% less even if they walk at a rapid speed. A/K amputees also walk with increased stride length (31% greater than that of Murray's subjects), which remains essentially unchanged with the change of cadence. The prosthetic gait is further characterized by striking asymmetry with regard to the stride length, the duration of stance, and the swing phase. This asymmetry does not seem to change with speed. Typical stance and swing phase diagrams are shown on Fig. 4.1; and for comparison, data from both papers are summarized in Table 4.2.

The essence of this data is that an A/K amputee walks much more slowly than normals in order to minimize the metabolic cost. As he relies heavily on his intact leg for support, the prosthetic swing period is extended, and so is its stride length. These factors contribute in part to the asymmetrical gait and are parameters to be measured to determine if improvement has been made.

In this study several steps from both the video and foot-switch data were averaged, and the stride lengths and time durations of one walking cycle were determined. By taking the average between the left and right legs, the average speed of progression was obtained. These are shown in Figs. 4.2, 4.3, and 4.4.

From Figs. 4.2, 4.3 and 4.4, several points are observed:

(i) Run-5 shows more than a 15% increase in the speed of progression from Run-1, which indicates that the amputee had shown more confidence in the new prosthesis. In contrast, Run-3 shows an extremely slow gait because of the lack of damping in the knee, and he had to walk cautiously and slowly to avoid falling.

(ii) In Run-3 the asymmetry is worse than in the other runs.

(iii) The average speed in Run-5 is the highest among those in all the runs (with the exception of Run-6). This higher speed is derived from a longer stride length and shorter step duration on both legs, which hopefully means a good and secure gait.

(iv) The average stride length on the prosthetic side is always longer than that on the intact side, and the velocity on the prosthetic side is higher than that on the intact side. Time duration is always shorter on the prosthetic side except for Run-2, possibly due to the fact that he is more familiar with his own prosthesis.

Temporal information from the footswitch data is recorded in Fig. 4.5. Phase lags between heel contacts of legs were obtained in terms of the percent walking cycle. Based upon this information, data on the prosthetic leg were shifted to coincide with data on the intact leg on each run as discussed in the previous chapter.

For a normal person, such phase lag is expected to be one half of the walking cycle or 50% cycle; that is to say, the timing of the heel contacts of one leg is always at the half step intervals of the other leg. Variation from the 50% cycle will lead to asymmetrical gait. From Fig. 4.5, it is obvious that Run-2 is closer to the 50% cycle for its phase lag between heel contacts. This is no surprise since the amputee has been wearing it for years. Run-5 shows quite an improvement from Run-1, and is only 3% from normals.

As mentioned before, a normal person will probably have a stance-to-swing ratio of 60%:40%, Once again, Run-2 represents a closer-to-normal gait with the smallest difference between legs; while Run-3 has the worst score.

The double support (DS) durations are recorded in Fig. 4.6. The general trend is that DS periods are longer for the intact legs, the obvious reason being that the intact leg has a tendency to stay on the ground longer after the prosthetic HC in order to secure a stabilized swing. DS periods are far from identical in both legs of

Runs-2 and 3, and their total DS periods in one walking cycle are well over 20% cycle time. These could be attributed to the slower progression speed and the different kinds of mechanisms employed in the knee joints, and it could mean instability. These results from Fig. 4.6, confirm that the total DS period for one walking cycle shortens as the walking speed increases.

II) Profiles of vertical displacement of body markers

A trajectory plot of the spatial movement of body markers has been shown in Fig. 3.15. It is obvious that most changes occur in the ankle, heel, and toe. To recall, one of the functional requirements of the A/K prosthesis is the provision of damping at the beginning of the swing. Measurements of the vertical amplitude of heel and toe markers during this period will then indicate the performance of the knee joints. Typical vertical displacement profiles from both legs of Run-5 are shown in Figs. 4.7 and 4.8, while the maximal values of heel marker heights from all data runs are shown in Fig. 4.9.

Because of the longer period of the prosthetic leg swing, the maximum heel rise occurs earlier than in the intact leg. Generally, amplitudes of heel-rise are smaller in the prosthetic leg as compared to that of the intact leg, except for Run-3 which has the highest prosthetic leg heel-rise due to the lack of damping mechanism. Since the velocity of the shank-foot is the major factor that affects the amplitude of the heel-rise, amputees usually try to reduce their walking speed in order to limit shank velocity, and subsequently the heel-rise. Run-5 shows more success in maintaining an acceptable heel-rise level in spite of its higher speed as compared to the first four runs.

Simultaneously raising the elevation of hip joints on both sides will substantially increase the energy cost due to the rise of the centre of gravity (CG) of the whole body. However, if the pelvis only acts as lever, one leg will compensate the motion of the other by knee flexion or extension in order to maintain a constant path of CG. In normal walking, the hip joints work between these two extremes to produce a sinusoidal path of CG. Therefore, examining the range of variation in the hip markers independently (which is shown in Fig. 4.10) will provide information on its symmetry, but only a rough indication as to the cost of energy.

Generally, one would expect the prosthetic hip joint to have the maximum hip-rise during early stance due to its limited ability to flex the knee at that time. However, from Fig. 4.8, the results show that maximum hip-rise occurred during the prosthetic swing, that is, the time that hip muscles are most active in lifting up the prosthetic leg. In addition,this hip action was much higher on the intact side. In Fig. 4.10 , Run-3 showed an extra amount of hip motion on the intact leg in order to compensate for the instability caused by the supporting prosthesis.Instantaneous CG of the body was determined from the intersection of the equidistances of the hip markers. As mentioned earlier, excessive rise in CG will greatly increase energy cost. The vertical displacements of CG are plotted in Fig. 4.11 to determine if the changes in CG are excessive and asymmetric. The minimum value of the CG during walking was subtracted from the two peak values of CG on each run (which correspond to the intact swing and prosthetic swing) to obtain the range of variation, or the relative changes in amplitude of the CG during both swings. These values are shown in Fig. 4.12. As expected, the prosthetic leg caused a larger change in the vertical displacement of CG during prosthetic swing (or within the intact leg stance period). In contrast, most peaks caused by the prosthetic hip (Fig. 4.10) had been smoothed out on the path of CG, resulting in gaits of lesser asymmetry and energy requirement.

III) Angular changes in the lower limbs

(A) Profiles of limb angles

From the body markers, limb angles were derived as described in Chapter 3. The typical profiles of four significant angles, namely those of the thigh, the shank the knee, and the ankle from Run-5 are shown in Figs. 4.13-4.16; their ranges of variation are also shown in Figs. 4.17-4.20, respectively.

From Fig. 4.17, it is apparent that there is no significant thigh angle (θ_{th}) variation between the legs on each run. Even though Run-3 shows a maximum difference of 4^{o} between the left and right thighs, the general trend is that the right thigh (amputated side) followed the left thigh closely. This result is a surprise since the amputee has only 0.2m. of stump which is approximately half of the total length of his normal thigh. Higher values are shown in both Run-5 and Run-6 due to a higher cadence.

From Fig. 4.18, the ranges of shank angles (θ_{sh}) have larger variations than those of the thigh angles.

A lower range on the prosthetic side of all runs (except for Run-3) is probably due to the high damping components in the units. This is more apparent when the damping cylinder was disconnected in Run-3, where the range on the prosthetic shank was increased and had a higher value than on the intact shank. A much higher difference between legs in Run-6 indicates that the inertial damping of the prosthesis limits the rise of shank during the rapid free swing and this enchanced the asymmetry. So the setting on this prosthesis may only be suitable for walking at a moderate cadence.

From Fig. 4.19 and 4.20, the ranges of both knee angles (θ_k) and the ankle angles (θ_a) have a much larger variation between legs in all runs. The intact leg had a larger range during all recordings. Once again, the limited range of the knee angles on the prosthetic side shows that the damping resistances were too high on the experimental prosthesis.

(B) Velocities of limb angles ($\dot{\theta}_{th}$, $\dot{\theta}_{sh}$, $\dot{\theta}_{k}$, $\dot{\theta}_{a}$)

The convention used in this study (refer to Fig. 3.16) for the direction of the various limb angle velocities is the same as for the limb angles and is shown in Table 4.3.

Table 4.3

angular vel.	+ve value stands for	-ve value stands for
$\dot{\theta}_{th}$	forward movement of thigh	backward movement of thigh
$\dot{\theta}_{sh}$	forward movement of shank	backward movement of shank
$\dot{\theta}_{k}$	knee flexion	knee extension
$\dot{\theta}_{a}$	plantar flexion	dorsiflexion

Angular velocities of the four angles given in Table 4.3 are shown in Figs. 4.21 - 4.24. Generally, the data from the prosthetic leg follows the pattern of the intact leg with a small lead. It can be easily demonstrated that the forward movement of the shank is associated with the extension mode of the knee angle after the initiation of swing by aligning the $\dot{\theta}_k$ and $\dot{\theta}_{sh}$ records.

The motion of the shank is best illustrated by its angular velocity and acceleration, and the shank can be thought of as an output device of the thigh. Furthermore, every $\dot{\theta}_{sh}$ record shows that the shank moves forward only once in a walking cycle at mid-swing. This velocity appears to bear a relationship with the velocity of progression and the amount of damping present. The resulting peak angular velocities of the shank in the forward direction of all runs are shown in Fig. 4.25. These peak velocities on the prosthetic side are always smaller than those of the intact leg (except for Run-3), and their forward swinging periods are longer. In spite of the fact that there was a 15% increase in speed from Run-1 to Run-5 (referring back to Fig. 4.4), the amputee was able to maintain the peak angular velocity of the rotating prosthetic in Run-5 at about the same value as in Run-1.

From Fig. 4.23, thigh angular velocities ($\dot{\theta}_{th}$) generally were the same for both legs, except that during the 52-66% cycle period, the prosthetic thigh is slowed down, which could have some significant bearing on the continuity of the motion of the shank.

There is a much bigger contrast between the two ankles in Fig. 4.24. In normal walking, the plantar flexion of the ankle during heel contact can help to absorb energy and reduce impact. Further, this motion stretches the length of the leg and increases the ankle moment to push the body forward. Since both the new prosthesis and the amputee's own prosthesis employ the SACH-foot, which is solely a passive device, the angular variation of ankle depends upon the magnitude and direction of the impact force; while the intact ankle will dorsiflex and plantar flex actively according to the need due to an increase in cadence. The results show that angular change in the prosthetic ankle was insufficient to follow that of the intact leg.

(C) Accelerations of shank angles ($\ddot{\theta}_{sh}$)

According to Radcliffe (1963), there is a phasic relationship between the changes in knee moment and the shank angular acceleration ($\ddot{\theta}_{sh}$). After further examination, his paper suggested that the $\ddot{\theta}_{sh}$ is a sensitive indicator of the action either of the muscle groups acting about the knee or of the equivalent fluid-controlled action of a prosthetic mechanism. Therefore, the $\ddot{\theta}_{sh}$ records are useful in the evaluation of lower-extremity prosthesis, when an indication of the phasic effect of the knee mechanism on the swing of the shank-foot is desired. A composite idealized record from Radcliffe is shown in Fig. 4.26, which contains characteristics typical of most recordings taken with lowerlimb prostheses. Based upon this information, the result from the prosthesis of this study (Run-5) is being evaluated here.

The $\ddot{\theta}_{sh}$ from both legs of Run-5 are shown in Fig. 4.27. There are six features to be noted. They associate with the prosthetic swing phase of $\ddot{\theta}_{sh}$ and have been redrawn and labelled in Fig. 4.28.

Prior to the swing, there is a deceleration (a) which corresponds to the reduction of heel-rise. Then an acceleration of the shank (b) brings the shank-foot assembly forward. It is followed by deceleration so as to clear the tip of the SACH-foot off ground (c). From Fig. 4.13, region (c) corresponds exactly to a thigh angle of 90°. Therefore it is suspected that circumduction of the thigh occurred at or before that moment. Further swinging action (d) then brings the shank-foot ahead of the knee-joint. Gravitational deceleration appears at the end of the swing (e), which is further aided by the energization of the HRV (from Fig. 3.11), and further by the closing of the Check Valve (f) when the knee changes from extension to flexion.

From Fig. 4.24, the angular velocity of the prosthetic ankle shows little or no dorsiflexion at the period that corresponds to region (c) of Fig. 4.28. A much smoother acceleration record of the shank can be obtained if some form of dorsiflexion is available to the passive SACH-foot at that precise moment to clear the floor easier. A dotted path is plotted on Fig. 4.28 to represent such an ideal knee mechanism.

In order to better identify the available features of the knee units, the shank angular acceleration records on the prosthetic side for all runs are plotted in Fig. 4.29. The knee flexion and extension phases during the toe-off and swing period are determined from knee angular velocity ($\dot{\theta}_k$) records. Generally, during mid-swing, the magnitude of forward shank acceleration is smaller than that of the intact leg. The sharp acceleration in Run-3 during this period indicates a sudden undamped swing which is undesirable. The action of the Check Valve which caused some additional deceleration can also be identified in Run-1, Run-4, Run-5 and Run-6 right after the end of the extension phase of swing.

IV) Knee moment during swing phase

By the use of free body diagrams, the moments of the shank-foot about the knee joint during the swing phase can be calculated from the video data (see Appendix E). Based upon the results from normals, and after accounting for the difference in limb masses, Radcliffe presented an ideal moment curve for A/K prosthesis (Radcliffe, 1957) which is shown in Fig. 4.30. The paper assumes that the amputee will be able to control the motion of the thigh socket in a comparatively normal manner. It actually represents values of moment required about a knee joint of an artifical leg of average weight to provide swing-phase characteristics similar to those of the normal leg during level walking (Wilson, 1968). With the sign convention adopted in this analysis, if the direction of the resultant force (from all inertial forces) is passing behind the knee centre, the moment created will be negative and would tend to flex the knee; while a positive moment will indicate that same force is passing in front of the knee joint, tending to extend it. The knee joint moment calculated from Run-5 on the prosthetic side is included in Fig. 4.30 for comparison. The phasic similarities between these two curves indicate that the new prosthesis is an acceptable design.

As for other types of prosthesis, Wilson has recorded the knee-moment patterns of various units at the Veterans Administration Prosthetics Center (see Fig. 4.31). The knee units were adjusted for intermediate resistance, and were put on a test-jig subjected to 43 cycles per minute (equivalent to 86 steps per minute). From these results, Wilson concluded that both the fluid and pneumatic controls show better characteristics than the ones with constant friction and intermittant friction. The result of this study (Fig. 4.30) shows a similar trend in its flexibility to vary the resistance, even though it is calculated from gait recordings.

V) Energy in the limbs

An analysis of the energy components of the normal gait in the sagittal plane has been published by Winter, et al (1976). Their main findings were: (i) the thigh conserves about 1/3 of its energy changes; (ii) the shank has the largest total energy changes during the swing phase, but virtually no exchange between kinetic and potential energy components; (iii) the stride to stride changes in total body energy were less than 4% of the average body energy; and (iv) the rotational kinetic component of the shank during the swing phase is significant. While their calculations were based on the peak to peak changes in the energy levels of limbs, no assessment was made on the transfer of energy across segments so as to depict the efficiency of the knee joint.

The total energy of each limb segment can be calculated from: $E_i = m_i g h_i + 1/2\, m_i v_i^2 + 1/2\, I_i \dot{\theta}_i^2$; where m_i is the mass of the segment, h_i the height of the centre of mass above a reference datum, v_i the linear velocity of the centre of mass, I_i the rotational moment of inertia about the centre of mass, and $\dot{\theta}_i$ the angular velocity of the i^{th} segment.

The resulting shank and thigh energy curves of both legs of the amputee from Run-5 are shown in Figs. 4.32 to 4.35. The results from the intact leg (Fig. 4.32, 4.33) generally agree with the results from Winter's group. While the results from the prosthetic leg are also resonable, however, there are no data available to compare to.

From Winter's energy data on five normal individuals, there seems to be a correlation between the changes in the shank to those of the thigh. As the changes in the limb energy are the amount of work done on or by the limb (normally the muscle groups), the changes in the thigh during swing have to be used for initiating the shank motion which causes the changes in shank energy. Therefore, the ratio between their changes could become a means of determining the efficiency of the knee joint. In addition, the recorded maximum energy changes occur simultaneously during the early swing phase in both shank and thigh, therefore data taken at this moment on both limb segments are time correlated. This ratio is defined here as the change in shank energy (ΔE_{sh}) with respect to the change in thigh energy (ΔE_{th}), or simply $\Delta E_{sh} / \Delta E_{th}$. The results from Winter's normals have been calculated in Table 4.4, while the results from the locomotion runs of this study are shown in Table 4.5. Actual values are also plotted in Fig. 4.36.

From Table 4.5, most changes in the prosthetic shank (ΔE_{sh}) are about half of those of the intact leg; while changes in both thighs are nearly identical (with the exception of Run-6). From Fig. 4.36, it is apparant that the change in thigh energy plays an important role in determining the effectiveness of the prosthesis. A small change in shank energy in respect to that of the thigh is not acceptable, as excessive muscular work would then be required if a larger variation in shank energy is desired. Also, if one leg is not secured well during stance, it is difficult to couple the transfer of the floor reaction forces across the pelvis to develop a large ΔE_{sh} to ΔE_{th} ratio, or an effective swing through the other leg. While the intact leg produced a ΔE_{sh} to ΔE_{th} ratio (except for Run-3) nearly similar to that of normals, it is important to note that there is no muscular input in the shank-foot portion of the prosthesis, and the changes in the shank energy (mainly the energy dissipated in the damping cylinder or the knee joint) have to be initated by the thigh. So a higher ΔE_{sh} to ΔE_{th} ratio in both legs in Run-5 provides evidence that improvement in transfering energy changes has been made.

VI) Task performance

In order to determine the adequacy of this experimental prosthesis, the amputee was asked to do tests after the training period on fast walking, upramp, downramp, stairs-descending and climbing. These results are briefly summarized as follows:

(A) Fast walking

During this test, the amputee was asked to walk at a faster pace with the control electronics set at the 4-mode-operation. Both video and footswitch data were recorded. The video data, which were converted into kinematic data, have been analyzed among other locomotion runs in the previous sections of this chapter.

From the footswitch data, it can be seen that the HRV was energized for a longer period than level walking required at the amputee's 'free speed' (see both Figs. 4.37 and 3.11) to reduce the velocity of the moving shank. It is common to expect that A/K amputees would exert more hip muscle effort to initate a faster swing on the prosthesis during fast walking. So when walking with any particular velocity at which a particular amputee could minimize his hip effort while maintaining a good aesthetic posture. When walking faster than this velocity, the response of the prosthetic leg would not be able to keep up with the normal leg, requiring compensation from the normal leg which would further increase the asymmetry.

From kinematic data it is seen that there is an increase in asymmetry in Run-6. From data in Fig. 4.37, it is noted that at least one valve was being energized approximately 75% of a cycle time. It is felt that too much damping existed because insufficient angular movements were observed on the prosthetic limb (Fig. 4.18, 4.19) as compared to that of the intact limb. Unnecessary damping can be reduced if the 3-mode-operation is to be substituted which will definitely improve the symmetry and the performance during fast walking.

(B) Upramp and downramp

Because the available video equipment was set up for level walking, only footswitch data could be recorded. Electronics on the prosthesis was set to the 3-mode-operation for easier handling on the ramps. The amputee was asked to practise for approximately 15 minutes before the test began.

During upramp, footswitch data taken show that the amputee was able to lock his knee (at state-3) to flexion during mid-stance in order to propel his body upward and forward (see Fig. 4.38). Free swing was also possible. The amputee walked extremely slowly (about 70 steps per minute), and the prosthetic side visually showed a lesser flexion in knee angle as compared to level walking.

During downramp, footswitch data show no definite pattern of the valves' operation. Some steps, however, do show that some levels of resistive torque were present most of the time during stance and swing on the prosthesis (see Fig. 4.39). The presence of resistive torque indirectly increased the time taken for every step; otherwise, he would have felt less secure.

The stance to swing ratios recorded on the ramps are about the same as those during level walking.

(C) Stairs

In ascending and descending stairs, the knee joint is extended and flexed while supporting the body weight on one leg. The extent of the movement of the knee is from full extension to a 90° flexion. The functional stabilizers in the normal knee joint are the patella and the quadriceps acting through it. Therefore, to climb stairs is considered to be one of the most difficult tasks for an A/K amputee since he does not have the necessary quadricep muscles across the knee joint to provide extension torque during the weight support period. In this study, even though the amputee can lock his prosthesis to flexion, he cannot "jack" himself up by pivoting on his prosthetic knee joint. Another fact is, his stump is only about 8 in. long, which provides little leverage or the mechanical advantage for such an operation. An extension moment has to be provided by an external power source to give a lift, so climbing stairs is not feasible in this prosthesis, if not impossible because it is a passive device.

As for descending stairs, it is analogous to walking down a ramp, except that the amputee has no idea of when locking is necessary as each step has to be very precise. He has to avoid crushing the edge of stairs when weight bearing is being transferred to the prosthetic leg. At the same time, his whole body has to descend with a yielding lock at flexion. Too much damping at the knee joint would take a longer time for the intact leg to land, and the close-to-lock situation could cause him to slip on the prosthetic leg. Less damping present could cause the prosthesis to collapse. These difficulties probably cause the amputee to lack confidence, particularly because of a lack of practice and, in turn, resulted in no meaningful and consistant correlation between the valves' operations and the walking cycle periods as each step takes considerable time to be executed.

CHAPTER 5

RECAPITULATION OF RESULTS

(1) The main progress from early training (Run-1) to after training (Run-5) is that the amputee increased his comfortable walking speed subconsciously. The increased speed on the prosthetic leg is a result of a longer stride length, and a shorter step period. The increase of confidence builds up as the use of this prosthesis increases, which is demonstrated by a shorter intact leg stance phase, and shorter double support periods.

(2) The kinematic results of this prosthesis at this stage of development (Run-5) are comparable with the commercially available prosthesis, namely the Henschke-Mauch leg in Run-2. In some ways (e.g., the stance-swing ratio, and the prosthetic vertical hip displacement), the performance of Run-2 is better than that of Run-5 as the amputee has been acquainted with his own prosthesis for years. A smaller variation of knee angle on the

prosthetic side of Run-5 indicates that damping might be too high in the knee joint. However, it does not mean that Run-2 has a better symmetrical gait. Run-5 outperforms Run-2 in the vertical displacement of the centre of gravity, and the discrepancy in double support periods of Run-2 illustrates that compensation from the intact leg still plays a major role in his daily walking.

(3) Run-3 clearly illustrates the shortcomings of the absence of damping in the knee joint during normal walking. A slower cadence surprisingly improves the symmetry in the shank and knee angular changes, and in the vertical displacement of centre of gravity of the body. However, discrepancies in stride length, walking cycle duration, double support periods, stance-swing ratio, and maximum heel-rise are more apparent in spite of its slow progression speed and smaller shank velocity.

(4) Physically, Run-4 and Run-5 are of the same prosthesis. Run-5 uses an active control scheme, while Run-4 represents the passive control aspect of the prosthesis.

There are no significant kinematic differences between Run-5 and Run-4 except that a higher progression speed and $\Delta E_{sh}/\Delta E_{th}$ are achieved in Run-5. Moreover, locking is only possible with the electronic control on in Run-5 to prevent stumbling.

(5) Generally, the results of fast walking (Run-6) are similar to level walking (Run-5) with the proportional increase of all the amplitudes in angular changes, except that changes in ankle angle (θ_a) remain quite constant. It is felt that reduction of resistive damping in the knee joint can be achieved if the 3-mode-operation is to be substituted which could have improved discrepanices in the motion of both legs.

(6) The knee moment calculated for this prosthesis (Run-5) shows some phasic similarities to the idealized moment curve from Radcliffe (1957). Moreover, this result is comparable to the other types of available prostheses.

CHAPTER 6

DISCUSSION & CONCLUSION

The primary concerns of evaluation lie in three areas: the amount of training required, the level of control achieved, and the performance of gait. All of them rely directly on how good a design is. Conversely, a good design should provide stability (as in control) and symmetry (as in gait pattern) to an amputee.

The stability of a prosthesis depends upon a sound and consistant controlling scheme, which in this study is the EMG pattern of the stump muscles. During the training period, it was found that the amputee began to be aware of his own controlling power in providing the required knee joint torque for walking. This awareness enchanced his confidence in both prosthetic stance and swing. This form of control, even though voluntary, does not necessarily have to be conscious. Training shows that when subconscious control was called for, he was able to trigger the required sequence of valve operation; and he did not get tired so frequently. Therefore, it is felt that by conscious training, the EMG pattern adapts during daily walking routine. This adapted EMG pattern enables the amputee to exert control over the electronics which varies the damping resistance accordingly with his minimal attention. He can then direct most of his attention to the form of posture and cadence variation.

In view of the fact that during level walking, no significant kinematic differences were observed after the control electronics were turned off, one might ponder if such a controlling scheme is of any benefit. It is felt that during non-level walking, the demand for damping would increase a great deal because an increase in muscular activity is expected. In the event of accidental falls, it is natural for the amputee to tense all his muscles which will automatically provide an 'emergency brake' through this controlling scheme. Therefore as far as safety is concerned, this type of active control is a necessity.

It is difficult, if not impossible for an unilateral amputee to perform as well as normals. At least, in some ways, he will attempt to let his prosthetic limb follow the pattern developed by his intact limb. Unfortunately, the pattern developed by his intact limb has been somewhat 'disturbed'due to the compensation required to support the function of the prosthetic limb. Therefore the only valid criteria for gait pattern evaluation is through the degree of symmetry accomplished by both limbs.

The overall symmetry is best illustrated by the stance-swing ratio on both legs and the path of the centre of gravity of the amputee from Run-5. Even though the best stance-swing ratios are accomplished by the amputee with his own prosthesis (which should be expected), the path of the vertical displacement of the centre of gravity clearly demonstrates a closer to normal gait from this new prosthesis.

Any design has to have a way to specify its effectiveness in order to provide room for future improvement. To determine the ΔE_{sh} to ΔE_{th} ratio as discussed in Chapter-4 appears to be one of the solutions since the knee joint unit is in the path of energy flow across the thigh and shank-foot of the prosthesis assembly. The total energy changes in the shank determines the speed and driving forces during swing, which in turn are the result of the necessary muscular activities of the thigh muscles. Therefore, it is felt that a large ratio of $\Delta E_{sh}/\Delta E_{th}$ is desirable as it would then be an indication of reduced effort from the thigh muscles. This study has shown that a rise in this ratio occurred after training, and a drop appeared after the elimination of the damping element. Furthermore, these ratios are larger on the intact legs.

There are a few areas where this prototype can still be improved. This design was built with the assumption that flexion is possible during early stance. But generally, the amputees are not able to provide flexion during prosthetic stance, and this fact is evident in all the prosthetic knee angle (θ_k) records of this study. The resulting gait looks 'stiff' during this period which could be an extra cost for the energy expenditure. A design similar to a four-bar linkage system may be able to alleviate this problem.

The inability of the SACH-foot to dorsiflex before mid-swing causes a problem. As a result of this inability, the thigh and shank motions were slowed down in this study. A hydraulic device might be useful in order to shorten the total length of the prosthesis during swing so as to minimize the loss of momentum developed.

Nevertheless, one has to appreciate that this prosthesis is only a passive (non-powered) device fitted on an amputee with only half of his stump muscles. Besides, improvements on all aspects mentioned are possible which make the EMG-control a very viable and promising scheme for lowerlimb prostheses.

REFERENCES

1. Bresler, B. and Berry, F. R. 'Energy and power in the leg during normal level walking'. Prosthetic Devices Research Project, U. of California, Berkeley, Series 11, Issue 15, May 1951.

2. Bresler, B., et al, 'Energy and power in the legs of A/K Amputees during normal level walking', U. of California, Berkeley, Series 11, Issue 31, May 1957.

3. Eberhart, H.K., et al., 'The locomotor mechanism of the amputee', in "Human Limbs and Their Substitutes", edited by Klopsteg, P.E., and Wilson, P.D., Hafner, 1968.

4. Dyck, W. R., 'A voluntarily controlled variable resistance A/K prosthesis', M.Sc. Thesis, U. of Manitoba, 1974.

5. Dyck, W.R., et al., 'A voluntarily controlled variable resistance A/K prosthesis', Bulletin of Prosthetics Research, pp. 169-186, Spring 1975.

6. Gage, H., 'Accelerographic analysis of human gait', ASME Paper No. 64-WA/HUF-8.

7. Horn, G.W., 'Electro-control: and EMG controlled A/K prosthesis', Medical and Biological Engineering, Pergamon Press, 1972, Vol. 10, pp. 61-73.

8. Inman, V.T., 'Conservation of energy in ambulation', Bulletin of Prosthetics Research, pp. 26-34, Spring 1968.

9. James, U. and Öberg, K., 'Prosthetic gait pattern in Unilateral above-knee amputees', Scand J Rehab Med 5:35-50, 1973.

10. Mauch, H.A., 'Stance control for above-knee artifical legs - Design consideration in the S-N-S knee', Bulletin of Prosthetic Research, pp. 61-73, Fall 1968.

11. May, D.R.W., 'The development of a comprehensive external prosthetic knee control', in "Human Locomotor Engineering", The Institution of Mechanical Engineers, London, pp. 111-119, 1974.

12. Murray, M.P., et al, 'Comparison of free and fast speed walking patterns of normal men', Am J Phys Med 45:8-24, 1966.

13. Peizer, E. and Wright, D.W., 'Human locomotion', in "Human Locomotor Engineering", The Institution of Mechanical Engineers, London, pp. 196-214, 1974.

14. Quanbury, A.O., et al, 'Instantaneous power and power flow in body segments during walking', J. of Human Movement Studies, 1975, Vol. 1, pp. 59-67.

15. Radcliffe, C.W., 'Biomechanics Laboratory, U. of California, Berkeley, Series 11, Issue 33, Oct. 1957.

16. Radcliffe, C.W., 'Locomotion and lower-limb prosthetics', Bulletin of prosthetics Research, pp. 167-187, Fall 1974.

17. Ralston, H.J. and Lukin, L., 'Energy levels of human body segments during level walking', Ergonomics, Vol. 12, No.1, pp. 39-46, 1969.

18. Saunders, J.B., et al, 'The major determinants in normal and pathological gait', J. of Bone and Joint Surgery, Vol. 35-A, No.3, 543-558, July 1953.

19. Thornton-Trump, A.B. and Daher, R. 'The prediction of reaction forces from gait data', J. Biomechanics, Vol. 8, pp. 173-178, 1975.

20. Wilson, A.B., 'Recent advances in A/K prosthetics', Artifical Limbs, Vol. 12, No. 2, pp. 1-27, 1968.

21. Winter, D.A., et al, 'Television-computer analysis of kinematics of human gait;, Comput, Biomed. Res., Vol. 5, pp. 498-504, 1972.

22. Winter, D.A., et al, 'Measurement and reduction of noise in kinematics of locomotion', J. of Biomechanics, 7, p. 157-159, 1974b.

23. Winter, D.A., et al, 'Analysis of instaneous energy of normal gait', J. of Biomechanics, Vol. 9, pp. 253-257, 1976.

24. Bard, G. and Ralston, H.J., 'Measurement of energy expenditure during ambulation, with special reference to evaluation of assistive devices', Arch. Phys. Med., 40, 415-420, 1959.

25. Hogan, N., 'A review of the methods of processing EMG for use as a proportional control signal', Biomedical Engineering, pp. 81-86, March 1976.

26. Radcliffe, C.W., and Ralston, H.J., 'Performance characteristics of fluid-controlled prosthetic knee mechanisms', Biomechanics Laboratory, U. of Califronia, San Francisco - Berkely, Number 49, February 1963.

27. Andriacchi, T.P., et al, 'Walking speed as a basis for normal and abnormal gait measurements', J. of Biochanics, Vol. 10, pp. 261-268, 1977.

28. Contini, R., 'Body segment parameters, Pt. II.', Artifical Limbs, pp. 1-19, 1972.

APPENDIX A

Data relating to the hydraulic unit

Type of solenoid hydraulic valves used:

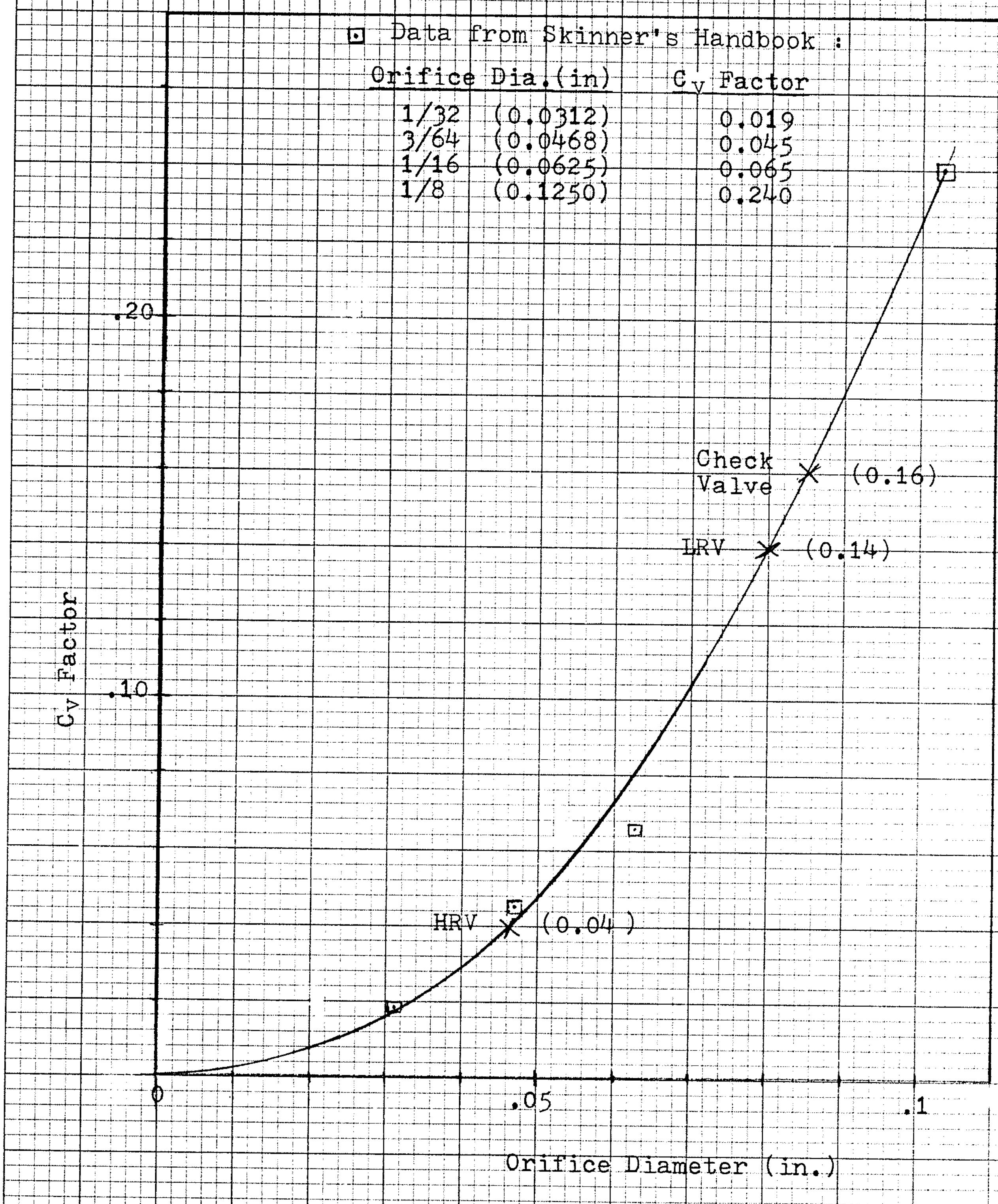
Skinner B1 DA1 400

(two way normally open subminature valves, type B1)

Because of the type of commercial valves available, adjustment had to be made on the size of orifices. The C_V factors are given on certain types from Skinner's handbook, while C_V factors for the modified valves of the study were extrapolated from these data (shown on the following page). The sizes of orifaces and corresponding C_V factors are listed in the accompanying table.

Two needle valves were added to the solenoid valves to provide variable resistances. The corresponding C_V factors can only be taken as approximations.

Dyck's design		New modification	
Valves	C_V factor	Orifice diameter (in.)	C_V factor
Check valve	0.08	~0.088	~0.16
High R. V.	0.02	~0.031 to 0.045	~0.02 to 0.04
Low R. V.	0.10	~0.065 to 0.084	~0.06 to 0.14



Extrapolation of the C_v factors.

APPENDIX B

Specifications for Instrumental Amplifier

AD 521

SPECIFICATIONS

(typical @ $V_S = \pm15V$, $R_L = 2k\Omega$ and $T_A = 25°C$ unless otherwise specified)

MODEL	AD521J	AD521K	AD521S
GAIN			
Range (For Specified Operation. Note 1.)	1 to 1000	•	•
Equation	$G = R_S/R_G$ V/V	•	•
Error from Equation	(±0.25 −0.004G)%	•	•
Nonlinearity (Note 2)			
1≤G≤1000	0.1% max	•	•
Gain Temperature Coefficient	±(3 ±0.05G)ppm/°C	•	±(15 ±0.4G)ppm/°C
OUTPUT CHARACTERISTICS			
Rated Output	±10V, ±10mA min	•	•
Output at Maximum Operating Temperature	±10V @ 5mA min	•	•
Impedance	0.1Ω	•	•
DYNAMIC RESPONSE			
Small Signal Bandwidth (±3dB)			
G = 1	>2MHz	•	•
G = 10	300kHz	•	•
G = 100	200kHz	•	•
G = 1000	40kHz	•	•
Small Signal, ±1.0% Flatness			
G = 1	75kHz	•	•
G = 10	26kHz	•	•
G = 100	24kHz	•	•
G = 1000	6kHz	•	•
Full Peak Response (Note 3)	100kHz	•	•
Slew Rate, 1≤G≤1000	10V/μsec	•	•
Settling Time (any 10V step to within 10mV of Final Value)			
G = 1	7μsec	•	•
G = 10	5μsec	•	•
G = 100	10μsec	•	•
G = 1000	35μsec	•	•
Differential Overload Recovery (±30V Input to within 10mV of Final Value) (Note 4)			
G = 1000	50μsec	•	•
Common Mode Step Recovery (30V Input to within 10mV of Final Value) (Note 5)			
G = 1000	10μsec	•	•

VOLTAGE OFFSET (may be nulled)			
Input Offset Voltage (V_{os_I})	3mV max (2mV typ)	1.5mV max (0.5mV typ)	**
vs. Temperature	15μV/°C max (7μV/°C typ)	5μV/°C max (1.5μV/°C typ)	**
vs. Supply	3μV/%	*	*
Output Offset Voltage (V_{os_O})	400mV max (200mV typ)	200mV max (30mV typ)	**
vs. Temperature	400μV/°C max (150μV/°C typ)	150μV/°C max (50μV/°C typ)	**
vs. Supply (Note 6)	0.005V_{os_O}/%	*	*
INPUT CURRENTS			
Input Bias Current (either input)	80nA max	40nA max	**
vs. Temperature	1nA/°C max	500pA/°C max	**
vs. Supply	2%/V	*	*
Input Offset Current	20nA max	10nA max	**
vs. Temperature	250pA/°C max	125pA/°C max	**
INPUT			
Differential Input Impedance (Note 7)	3 x $10^9\Omega\|\|1.8$pF	*	*
Common Mode Input Impedance (Note 8)	6 x $10^{10}\Omega\|\|3.0$pF	*	*
Input Voltage Range for Specified Performance	±10V	*	*
Maximum Voltage without Damage to Unit, Power ON			
or OFF Differential Mode (Note 9)	30V	*	*
Voltage at either input (Note 10)	V_S ±15V	*	*
Common Mode Rejection Ratio, DC to 60Hz with 1kΩ source unbalance			
G = 1	70dB min (74dB typ)	74dB min (80dB typ)	**
G = 10	90dB min (94dB typ)	94dB min (100dB typ)	**
G = 1000	100dB min (104dB typ)	104dB min (114dB typ)	**
G = 1000	100dB min (110dB typ)	110dB min (120dB typ)	**
NOISE			
Voltage RTO (p-p) @ 0.1Hz to 10Hz (Note 10)	$\sqrt{(0.5G)^2 + (150)^2}\mu V$	*	*
RMS RTO, 10Hz to 10kHz	$\sqrt{(1.2G)^2 + (30)^2}\mu V$	*	*
Input Current, rms, 10Hz to 10kHz	15pA(rms)	*	*
REFERENCE TERMINAL			
Bias Current	3μA	*	*
Input Resistance	10MΩ	*	*
Voltage Range	±10V	*	*
Gain to Output	1	*	*
POWER SUPPLY			
Operating Voltage Range	±5 to ±18	*	*
Quiescent Supply Current	5mA max	*	*
TEMPERATURE RANGE			
Specified Performance	0 to +70°C	*	-55 to +125°C
Operating	-25 to +85°C	*	-55 to +125°C
Storage	-65 to +150°C	*	*
PRICE			
(1-24)	\$12.75	\$18.00	\$30.00
(25-99)	\$10.20	\$14.40	\$24.00
(100-999)	\$8.50	\$12.00	\$20.00

*Specification same as AD521J.
**Specification same as AD521K.

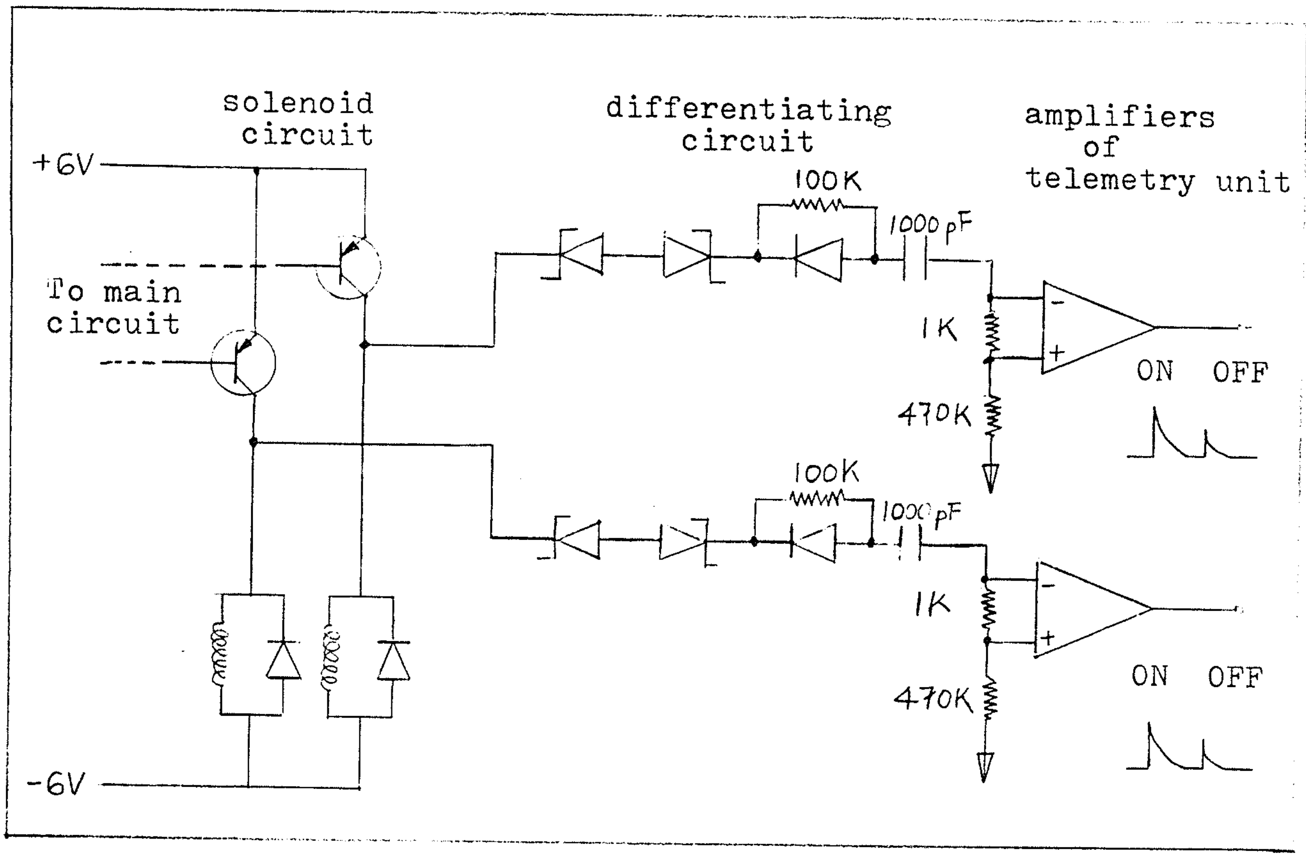
Specifications and prices subject to change without notice.

APPENDIX C

Solenoid switching detector

In order to obtain the temporal operation of the solenoid valves with respect to the walking cycle, simultaneous recordings of the ON-OFF operation of valves and the foot-switches data are desirable. Unused channels on the telemetry unit in the Locomotion Laboratory come with differential amplifiers as inputs. They were being utilised in the following manner: a couple of differentiating circuits were attached to the collector of the power transistors that drive the solenoids. Outputs from these differentiating circuits were fed into the inputs of the telemetry unit. By varying the resistance during the negative cycle, two different pulses can be obtained whenever the solenoids are in operation.

ON and OFF operations are distinguished by its sizes shown below:



Schematic of the solenoid-switching detector

APPENDIX D

Special computer programs

Two Fortran programs were written for data manipulation. The first one is for normalizing a set of points, it requires another subroutine called "PLOT" for output on a line printer; while the second one rotates the points, it requires another subroutine called "FILTER" to smooth the data at the merging junctions. This digital filtering procedure has been described by Winter, et al (1974b).

(1) Program 'Normalization'

```
C***
C***  THIS PROGRAM NORMALIZE A SET OF POINTS IN THE NEIGHBOURHOOD OF
C***  ONE HUNDRED INTO ONE HUNDRED POINTS, VALUES ARE LINEARLY EXTRE-
C***  PULATED. VAULES ARE PLOTTED USING A SUBROUTINE CALLED 'PLOT'.
C***
      SUBROUTINE NORMAL(NCOUNT,X,Y,NPTPLT)
C***
C***  NCOUNT -  NUMBER OF ORIGINAL POINTS (INCLUDING END POINTS)
C***  X      -  INPUT ARRAY (HAS TO BE LESS THAN 150)
C***  Y      -  OUTPUT ARRAY (INCLUDES THE END POINTS WHICH EQUALS 101
C***            POINTS)
C***  NPTPLT -  0 : PRINT RESULTS ONLY
C***            1 : PRINT AND PLOT RESULTS
C***
      DIMENSION Y(101),X(150),XDATA(1)
      IPT=6
      Y(1)=X(1)
      F=101./NCOUNT
      IF(F-1.) 1000,2000,3000
 1000 CONTINUE
      NN=NCOUNT-1
      DO 100 I=1,NN
      S1=I*F
      S2=(I+1)*F
      I1=S1
      I2=S2
      NDIFX=I2-I1
      IF(NDIFX.NE.1) GO TO 100
      Y(I2+1)=X(I+1)+(X(I+2)-X(I+1))*(I2-S1)/F
  100 CONTINUE
      Y(101)=X(NCOUNT)
      GO TO 4004
 2000 DO 200 I=2,100
```

```
      Y(I)=X(I)
  200 CONTINUE
      GO TO 370
 3000 CONTINUE
      N=0
      NTEMP=0
      DO 360 M=2,100
      I=M-N
      S1=(I-1)*F
      S2=(I-2)*F
      I1=S1
      I2=S2
      NDIFX=I1-I2-NTEMP
      IF (NDIFX.NE.2) GO TO 300
      N=N+1
      S2=S2+1
  300 Y(M)=X(I-1)+(X(I)-X(I-1))*(I1-S2)/F
      IF(NDIFX.EQ.2)NTEMP=1
      IF(NDIFX.NE.2)NTEMP=0
  360 CONTINUE
  370 CONTINUE
      Y(101)=X(NCOUNT)
      DO 4000 I=NCOUNT,149
      X(I+1)=0.0
 4000 CONTINUE
 4004 CONTINUE
      IF(NPTPLT.NE.1)RETURN
      XDATA(1)=0.
      CALL PLOT(Y,101,1,0.,1.,100.,0,XDATA,1,0,1)
      RETURN
      END
```

(2) Program 'Shift'

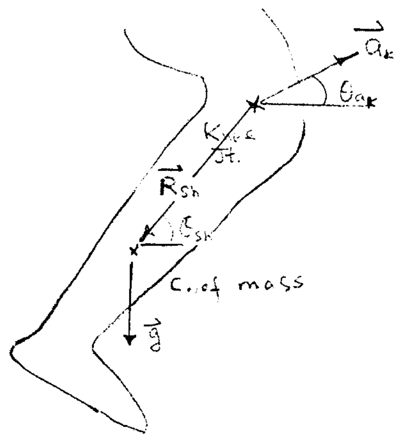
```
C***
C***  THIS PROGRAM ROTATES AN ARRAY OF DATA BY A CERTAIN NUMBER
C***  OF FRAMES TO THE LEFT OR TO THE RIGHT. IT REQUIRES ANOTHER
C***  SUBROUTINE CALLED 'FILTER' TO SMOOTH THE DATA AT THE
C***  MERGING JUNCTIONS.
C***
      SUBROUTINE SHIFT(XYS,ISHIFT,LORR,NPP,NTORTH)
C***
C***  XYS    -  ARRAY OF DATA TO BE MODIFIED
C***  ISHIFT -  NUMBER OF FRAMES TO BE SHIFTED
C***  LORR   -  0:ROTATE TO THE LEFT
C***            1:ROTATE TO THE RIGHT
C***  NPP    -  0:DO NOT PRINT THE RESULT
C***            1:PRINT RESULT
C***  NTORTH -  NUMBER 2 OR 3 FOR THE NUMBER OF PASSES REQUIRED
C***
      DIMENSION XYS(101,7,3),TEMSTO(105)
      IS=ISHIFT+2
      IDP=11
      IPT=6
      IRD=5
      IF (LORR.EQ.1) GO TO 55
      IS=101-IS
C MOVE X-DATA FORWARD BY HALF STEP
      XSHIFT=(XYS(101,7,1)-XYS(1,7,1))/2.
      DO 300 J=1,7
      DO 300 I=1,101
      XYS(I,J,1)=XYS(I,J,1)+XSHIFT
  300 CONTINUE
   55 CONTINUE
      NS=IS+1
```

```
      DO 50 K=1,NTORTH
      DO 50 J=1,7
      TEMDIF=XYS(101,J,K)-XYS(1,J,K)
      DO 60 M=1,IS
      TEMSTO(NS-M)=XYS(101-M,J,K)-TEMDIF
   60 CONTINUE
C SHIFT XYS
      NQ=1
      DO 80 M=NS,105
      TEMSTO(M)=XYS(NQ,J,K)
      NQ=NQ+1
   80 CONTINUE
      CALL FILTER(TEMSTO,1,105)
      CALL FILTER(TEMSTO,-1,103)
      DO 85 I=1,101
      XYS(I,J,K)=TEMSTO(I)
   85 CONTINUE
   50 CONTINUE
C FIND MINIMUM OF Y5,Y6,Y7 MARKERS AND MAKE ADJUSTMENTS
      YMIN=XYS(1,5,2)
      DO 95 J=5,7
      DO 95 I=1,101
      IF (XYS(I,J,2).LT.YMIN) YMIN=XYS(I,J,2)
   95 CONTINUE
      DO 120 J=1,7
      DO 130 I=1,101
      XYS(I,J,2)=XYS(I,J,2)-YMIN
  130 CONTINUE
  120 CONTINUE
      IF(NPP.EQ.0)RETURN
C THE FOLLOWING IS A PRINTOUT
      WRITE (IPT,70)IS
   70 FORMAT(' NO. OF FRAMES SHIFTED= ',I2)
      DO 100 K=1,NTORTH
      DO 100 J=1,7
      WRITE(IPT,90)(XYS(I,J,K),I=1,101)
   90 FORMAT(8X,10F9.2)
      WRITE(IPT,110)
  110 FORMAT(//)
  100 CONTINUE
      RETURN
      END
```

APPENDIX E

Determination of knee joint moment during swing phase

From the free body diagram below, we can write the moment



equation for equilibrium:

External moment - Inertia torque = 0

or $\quad M_{ext} \quad - \quad \tau \quad = 0$

Let M_k be the net moment about the knee joint, then if there is no motion,

Knee joint moment - moment due to gravity = 0

or $\quad$ Knee joint moment = moment due to gravity

that is, M_k (no motion) $= \vec{R}_{sh} \times M_{sh}\vec{g}$

where $\vec{R}_{sh}$ - displacement vector of centre of mass of Shank-foot from knee centre,

M_{sh} - mass of shank-foot

and $\vec{g}$ -acceleration due to gravity.

If there is motion,

M_k = Moment due to gravity + moment due to dynamic torque

or $\quad M_k = (\vec{R}_{sh} \; M_{sh} \; \vec{g}) + (I^*_{sh} \; \ddot{\theta}_{sh} + \vec{R}_{sh} \times \vec{A}^*_{sh} \; M_{sh})$

where I^*_{sh} - the moment of inertia of shank-foot about the centre of mass

$\ddot{\theta}_{sh}$ - absolute angular acceleration of shank in space

$\vec{A}^*_{sh}$ - acceleration of shank about centre of mass, which equals to $\vec{A}_k + (\vec{\dot{\theta}}_{sh} \; (\vec{\dot{\theta}}_{sh} \; \vec{R}_{sh}) + \vec{\dot{\theta}}_{sh} \; \vec{R}_{sh})$

($\vec{A}_k$ is the acceleration of shank about knee centre)

If we take moment about the knee joint, then

$\vec{A}^*_{sh} = \vec{A}_k$, and $\vec{I}^*_{sh} = \vec{I}_{sh}$, where $\vec{I}_{sh}$ is the moment of inertia about the knee joint

therefore, $\quad M_k = (\vec{R}_{sh} \times M_{sh} \; \vec{g}) + (\vec{I}_{sh} \; \ddot{\theta}_{sh} + \vec{R}_{sn} \; \vec{A}_k \; M_{sh})$

Or, in scalar quantity,

$$M_k = I_{sh}\ddot{\theta}_{sh} - R_{sh}M_{sh}g \cos(\theta_{sh}) - R_{sh} \; A_k M_{sh} \sin(\theta_{a_k} - \theta_{sh})$$

where θ_{a_k} is the direction of A_k in space.

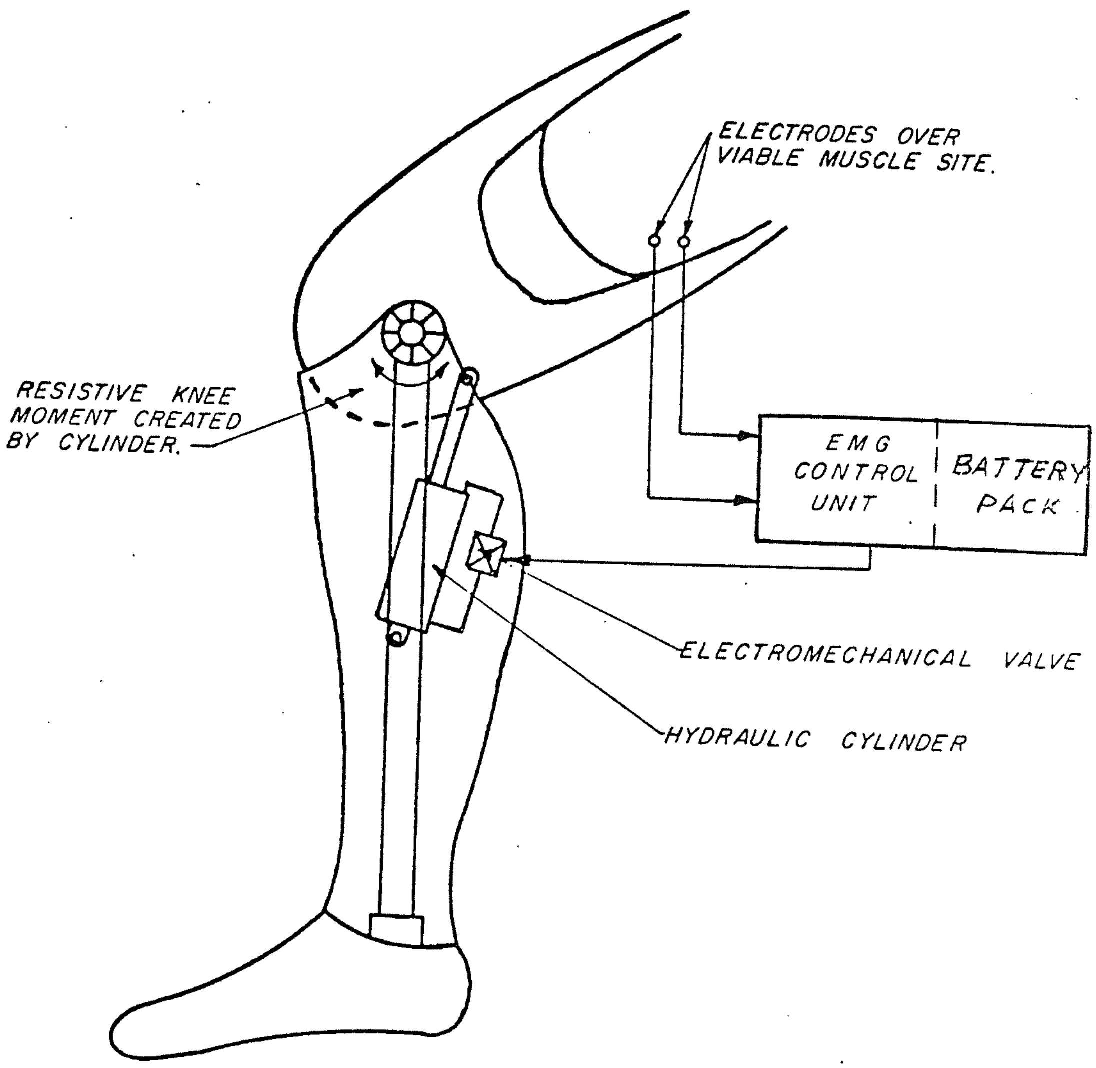


Fig. 1.1 Schematic of electro-hydraulic knee unit.

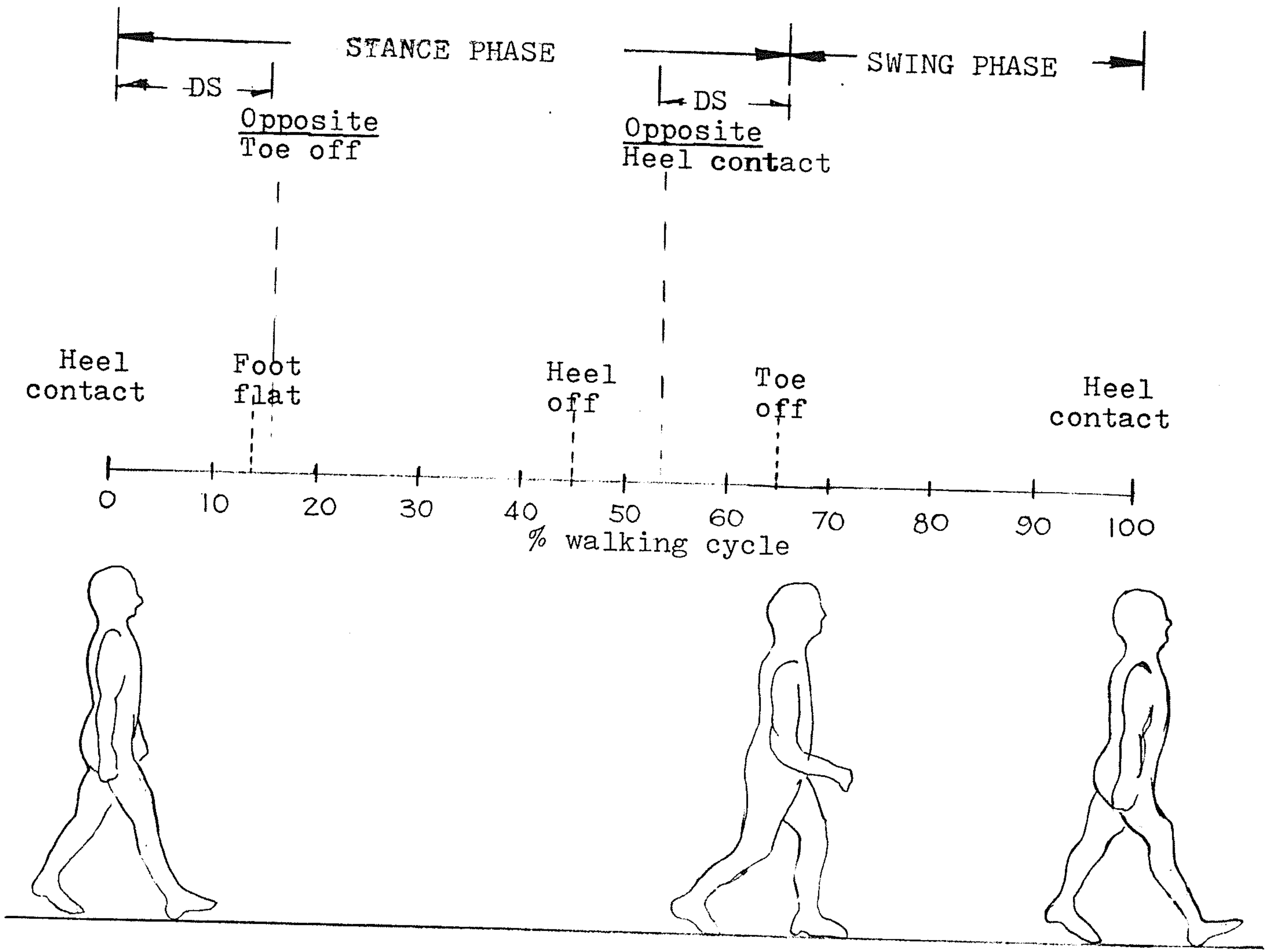


Fig. 2.1. Parameters of a typical walking cycle.

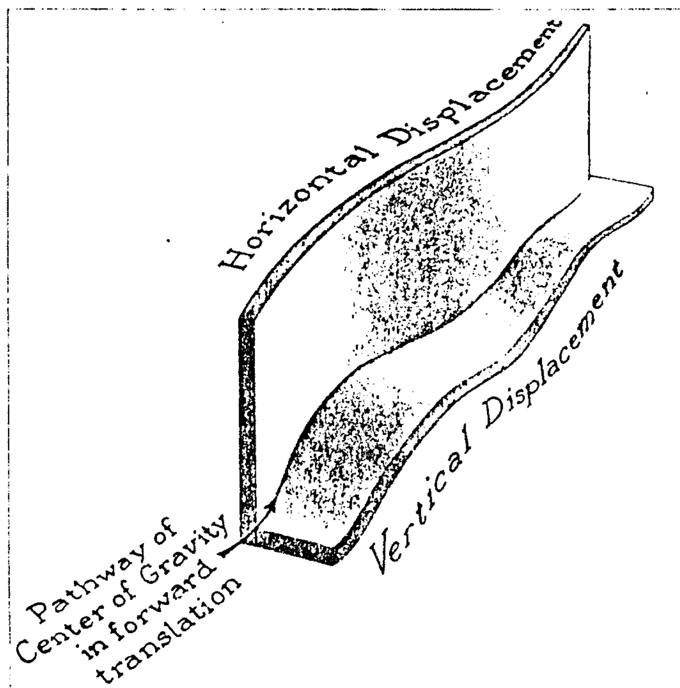


Fig. 2.2 The intersection of the horizontal and the vertical displacements produces the pathway of the center of gravity in locomotion. (Saunders, 1953)

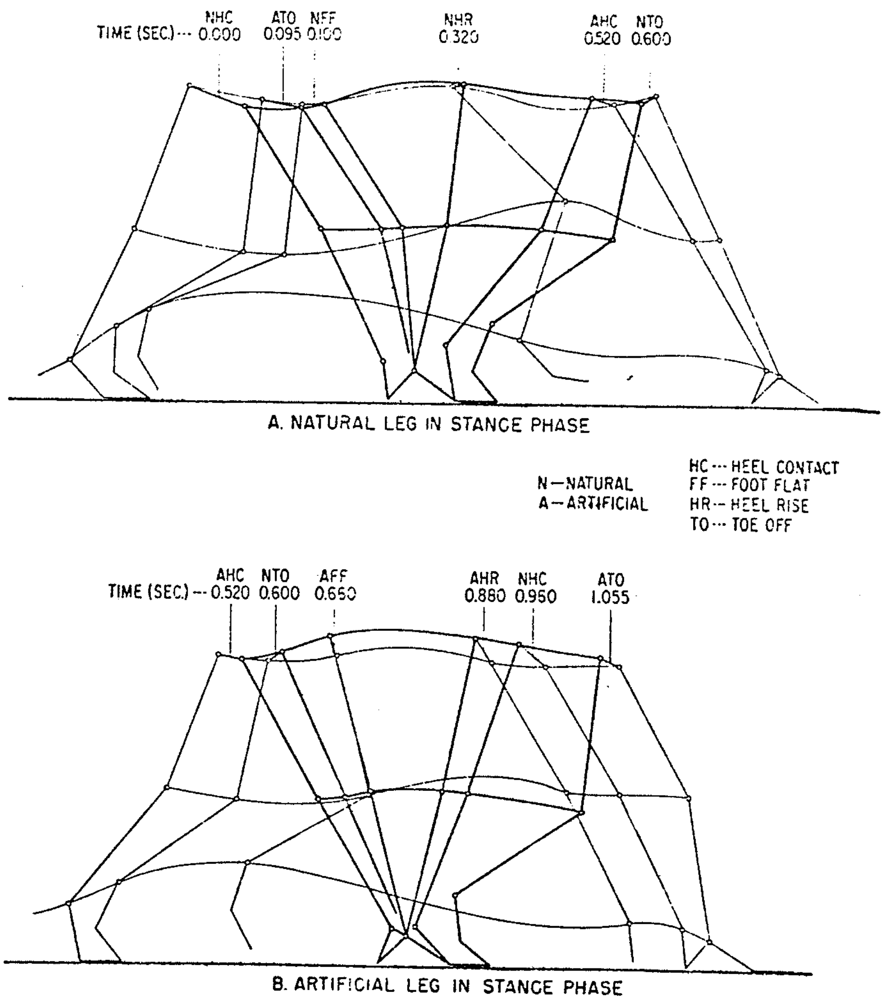


Fig. 2.3 Walking pattern of an above-knee amputee. (Conventional knee, suction socket, well-fitted, with excellent function and appear-ance. Eberhart, 1968.)

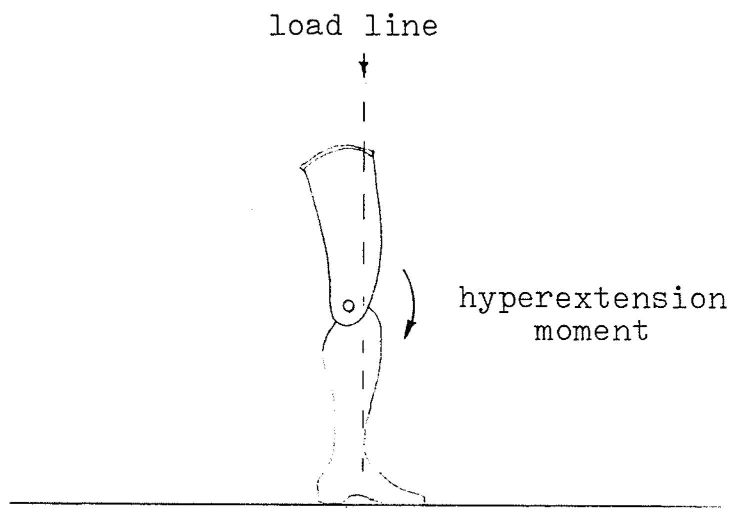


Fig. 2.4. Stance stability resulting from geometric alignment.

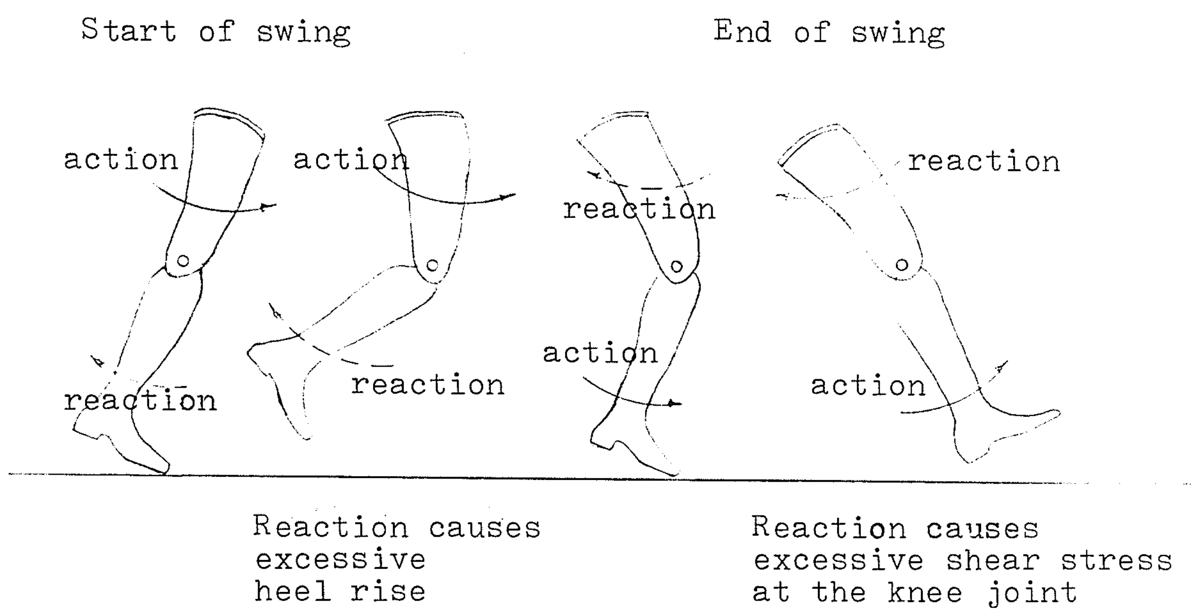


Fig. 2.5. Problems with an undamped knee hinge during swing phase.

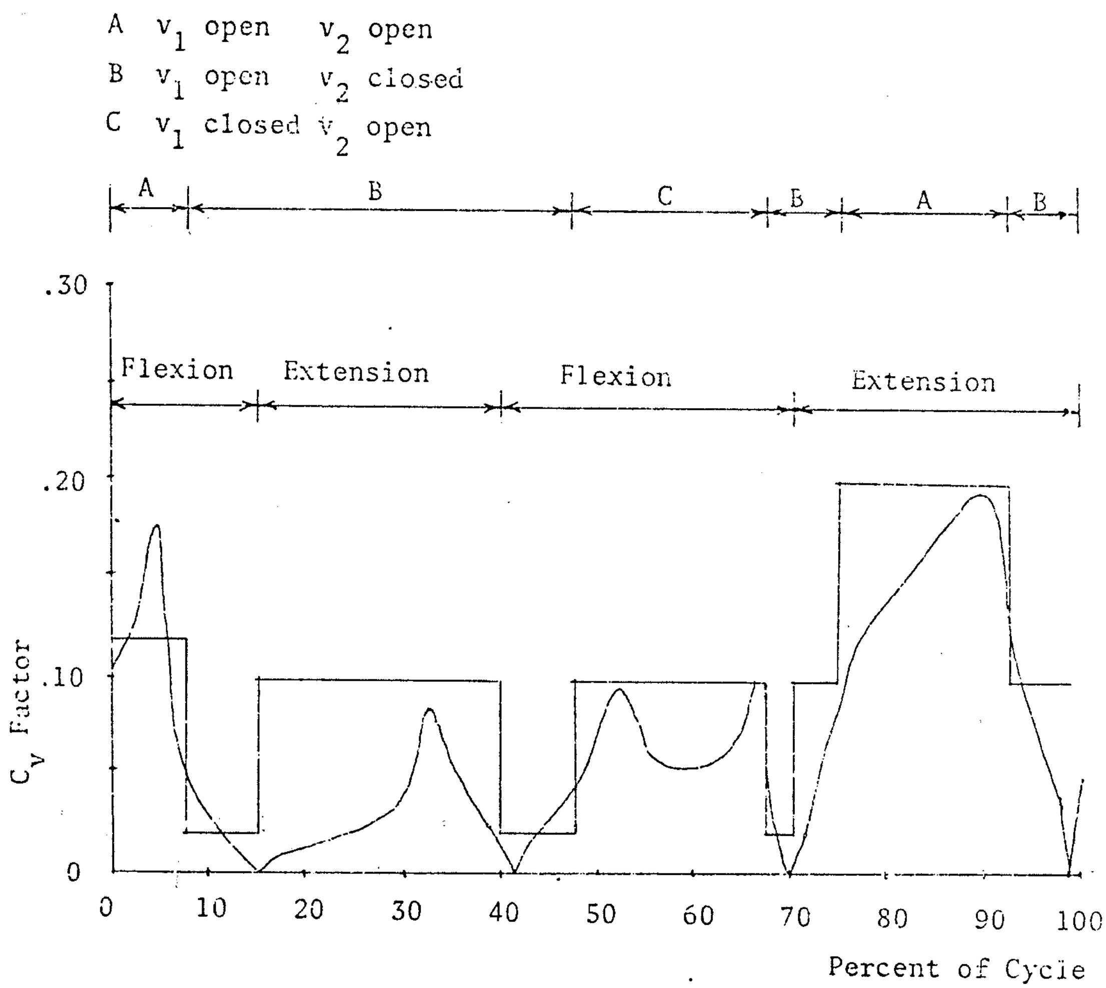


Fig. 3.1 C_v factor vs. Percent Walking Cycle and the C_v curve fit. (Dyck, 1974)

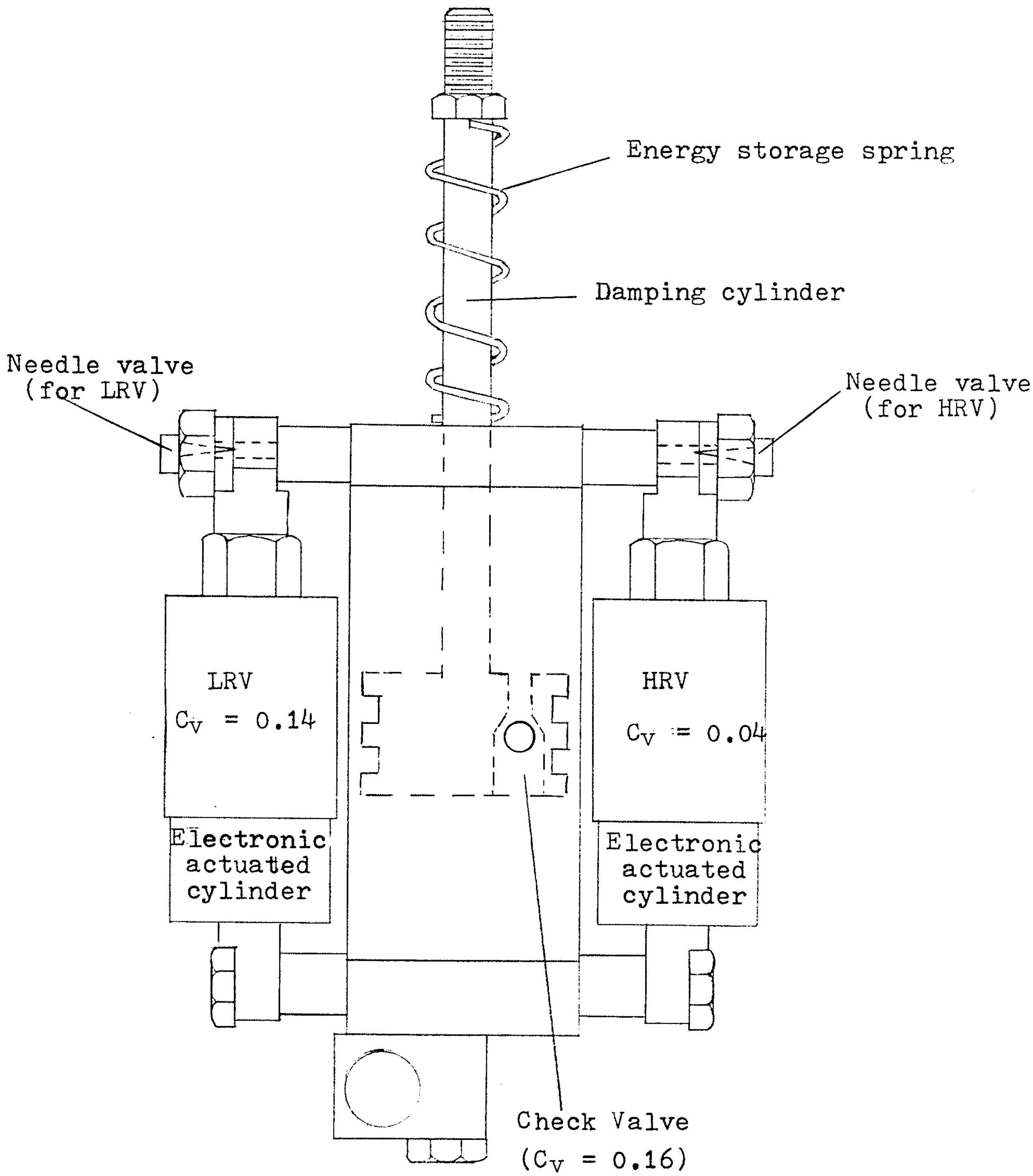


Fig. 3.2 Schematic of the hydraulic unit

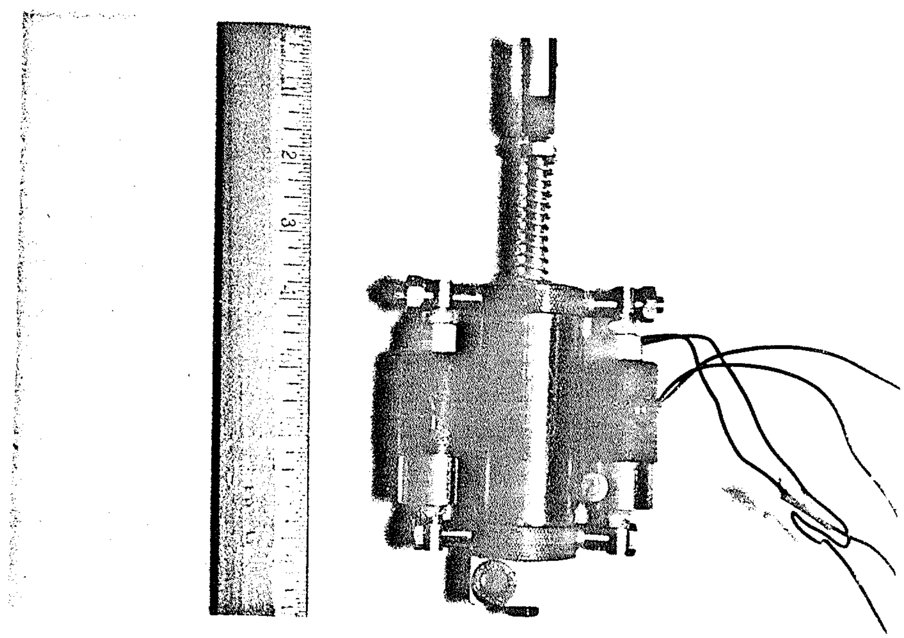


Fig. 3.3. Appearance of the hydraulic unit.

Mode	HRV C_v:0.04	LRV C_v:0.14	Check V C_v:0.16	Total C_v	Possible states
1	0	0	0	0.34	A
2	1	0	0 Extension Phase	0.30	B
3	0	1	0	0.20	C
4	1	1	0	0.16	D
1	0	0	1	0.18	E
2	1	0	1 Flexion Phase	0.14	F
3	0	1	1	0.04	G
4	1	1	1	0	F

1 - close
0 - open

$\underbrace{\qquad\qquad\qquad}$
To be controlled
by EMG-electronics

Fig. 3.4. Possible states achieved by knee unit.

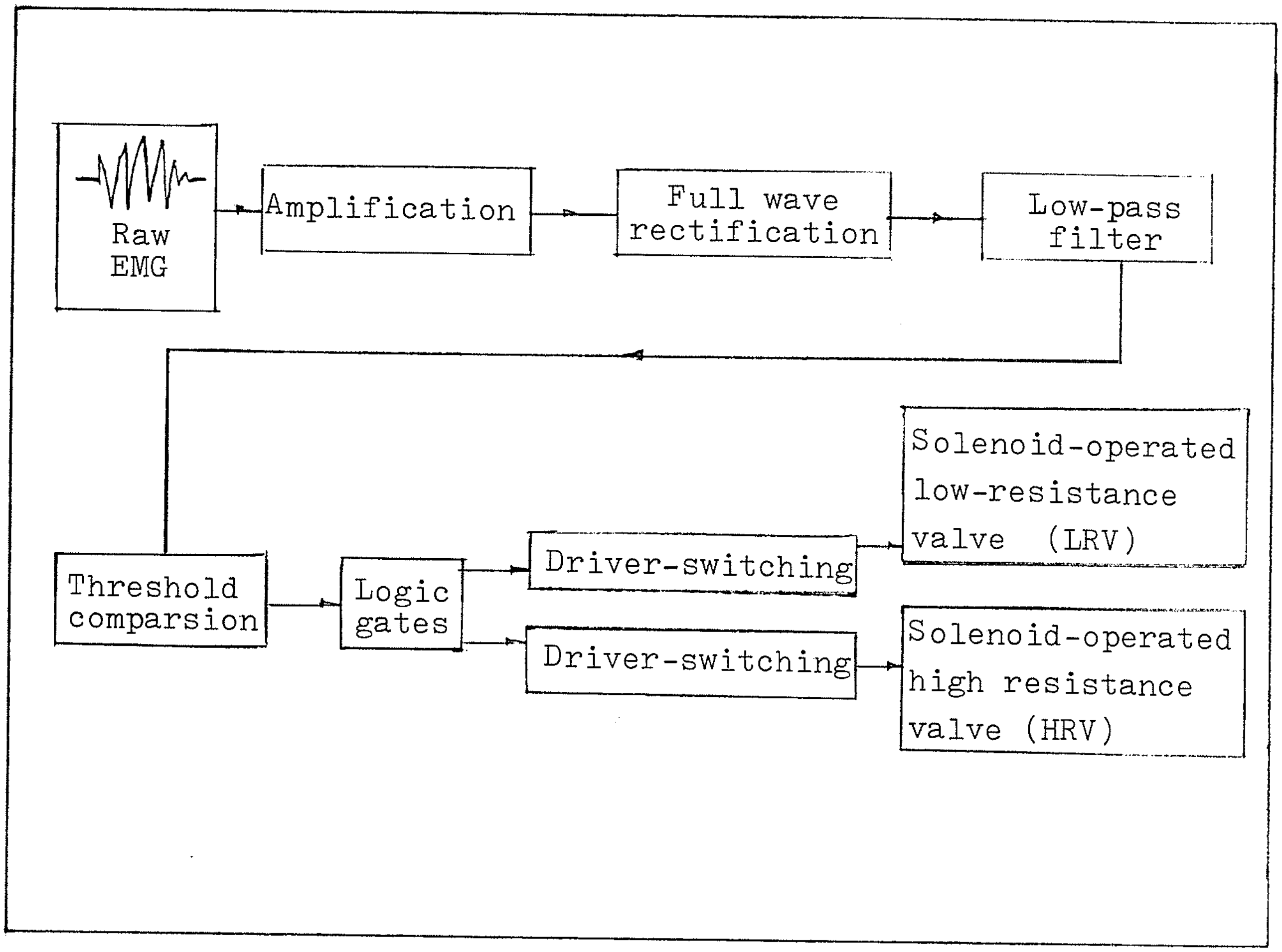


Fig. 3.5. Block diagram of the EMG control electronics.

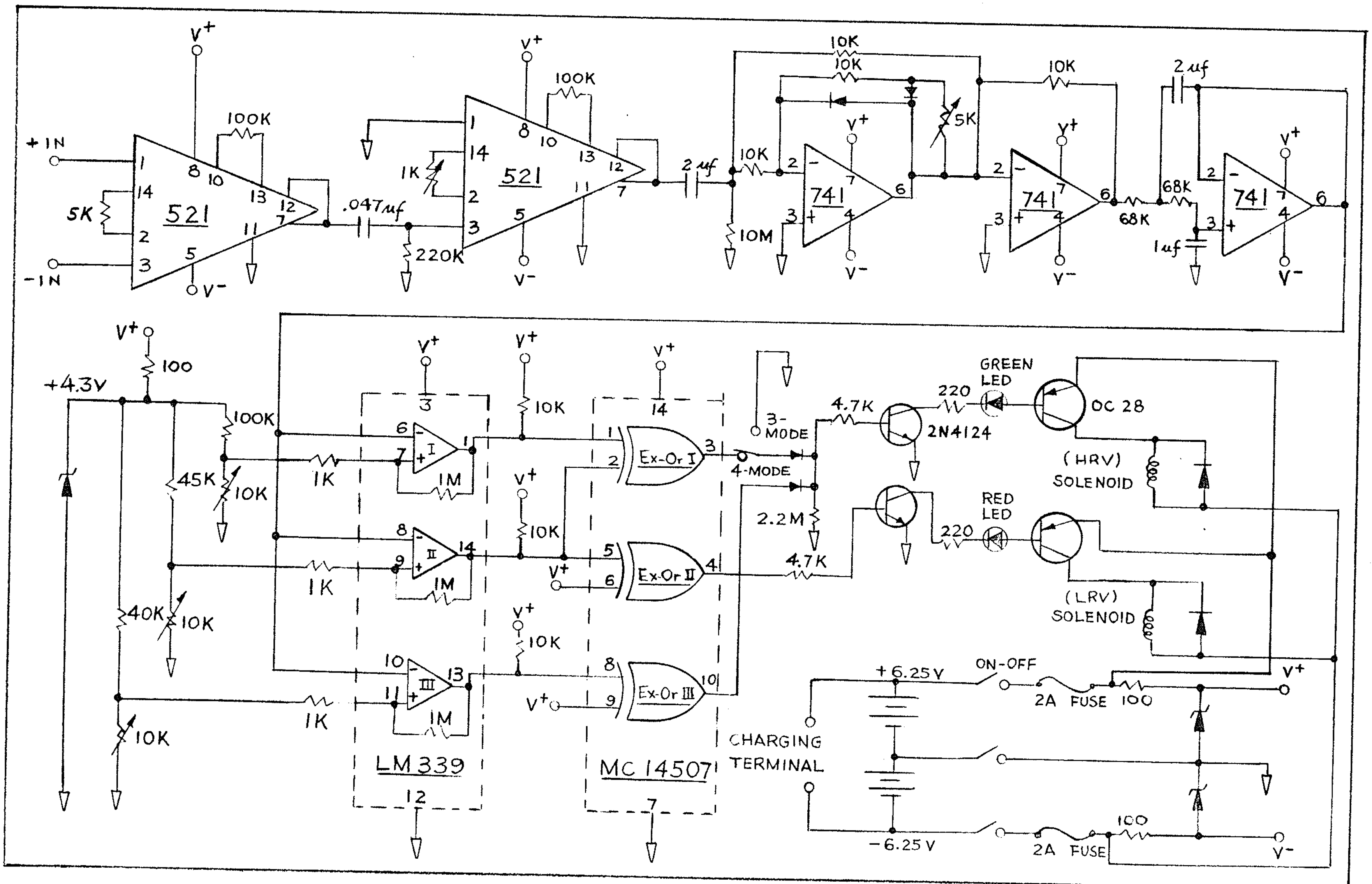


Fig. 3.6. Circuit diagram of EMG control electronics.

Mode	HRV (Green LED)	LRV (Red LED)
1 (rest mode)	0	0
2	1	0
3	0	1
4 (lock mode)	1	1
	0 - open 1 - close	

Fig. 3.7 Four-mode operation

Fig. 3.8. Electronics package

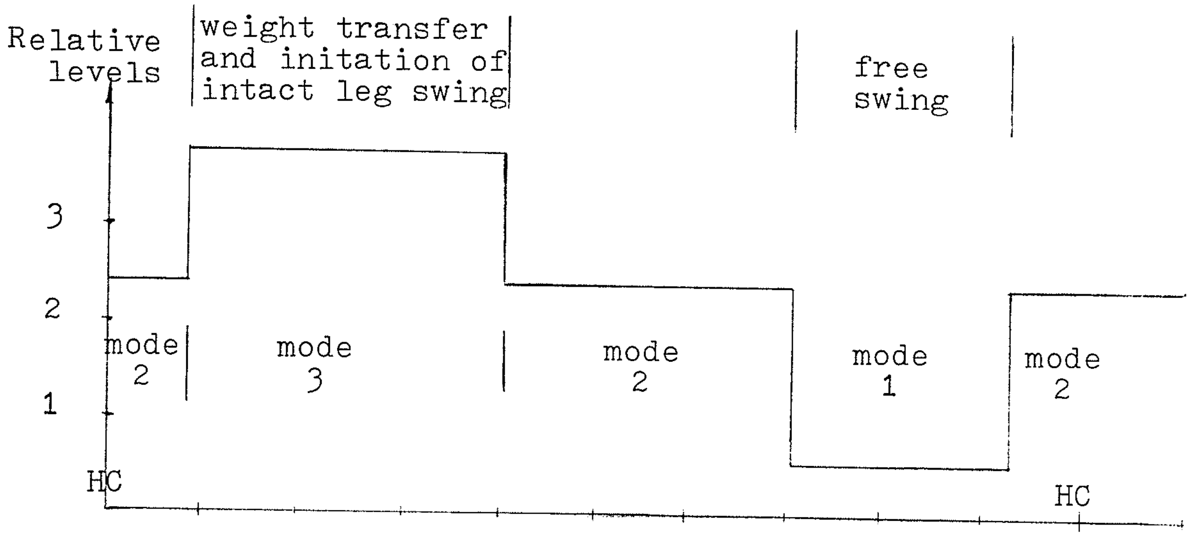


Fig. 3.9. Relative levels of muscular contraction required to produce the C_v curve in Fig. 3.10.

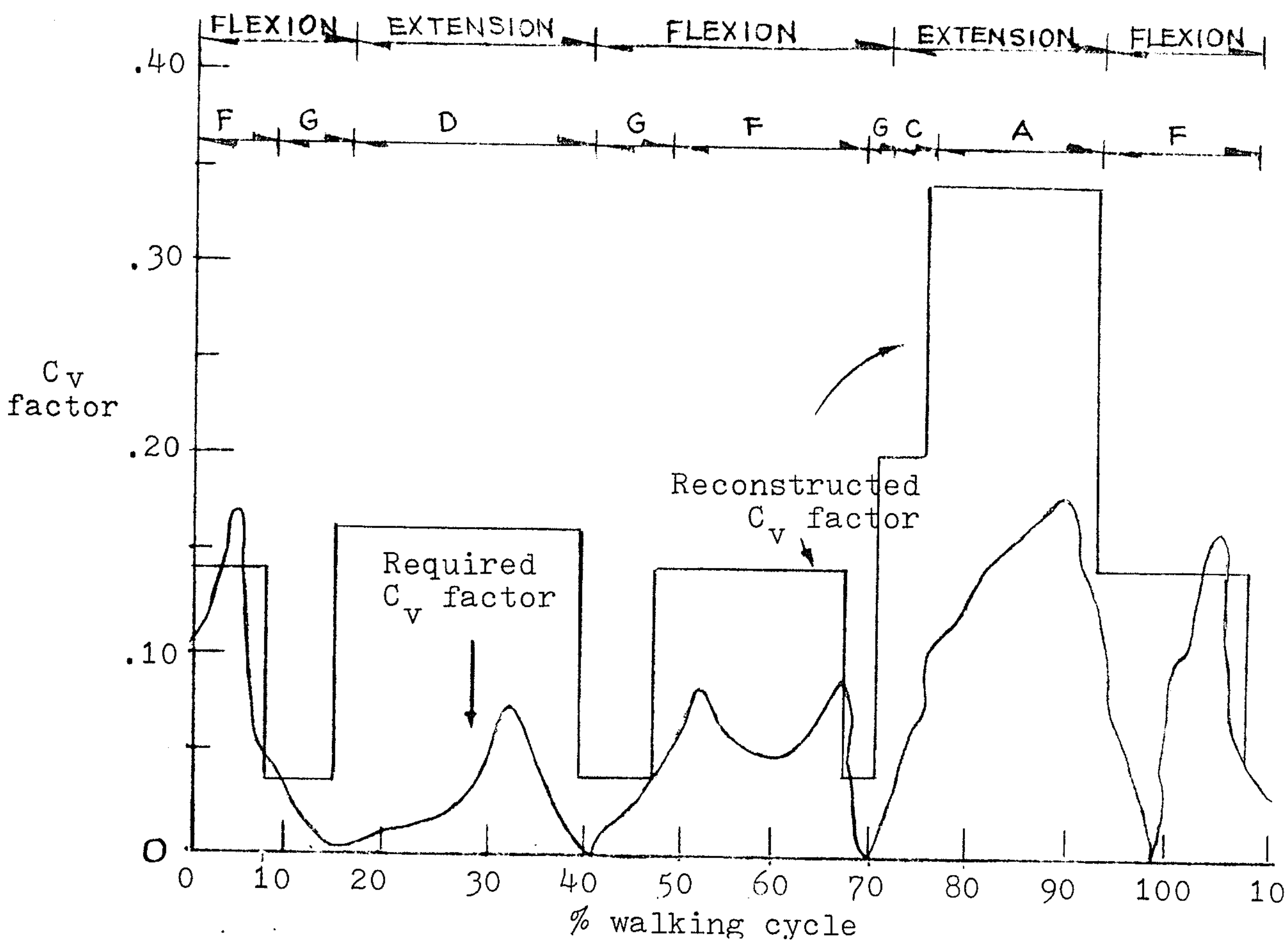


Fig. 3.10. C_v curve reconstructed.

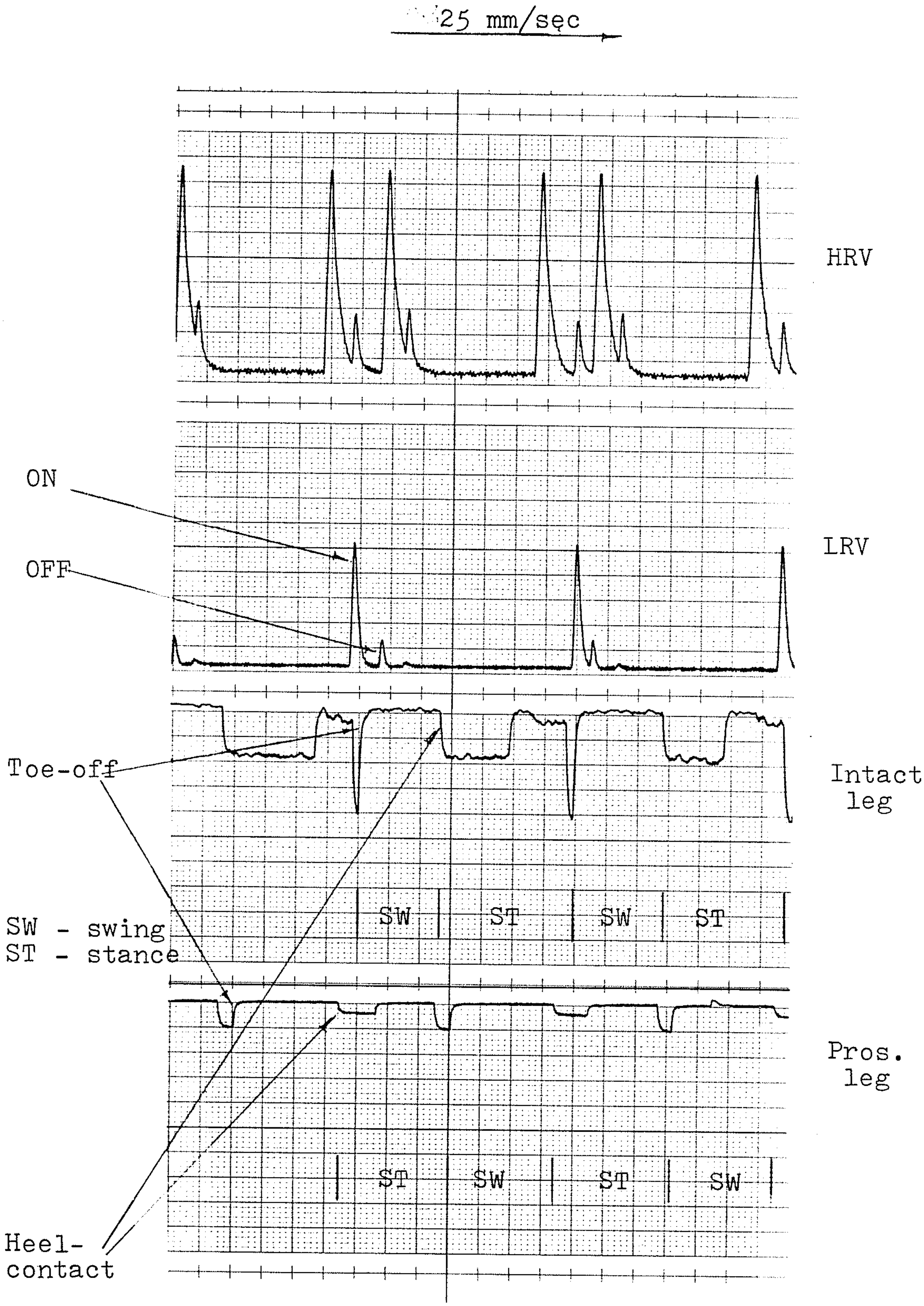


Fig. 3.11. Foot-switch recordings on level walking (at free speed).

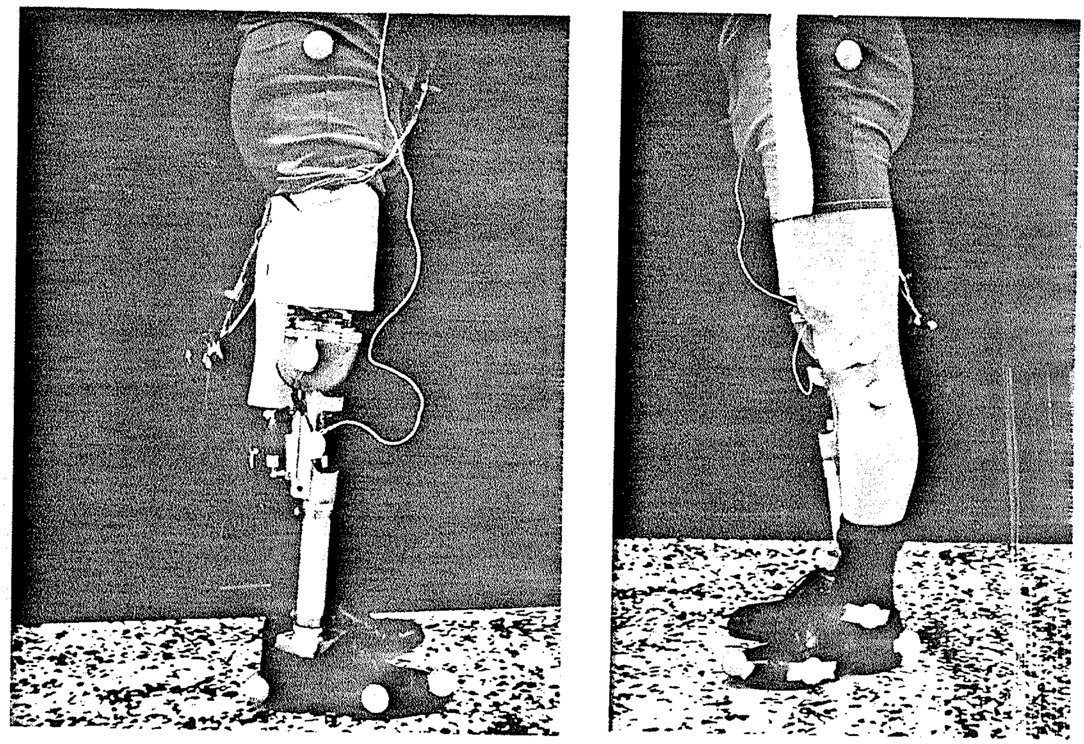


Fig. 3.12. Anatomical landmarks (body markers) used in video data processing.

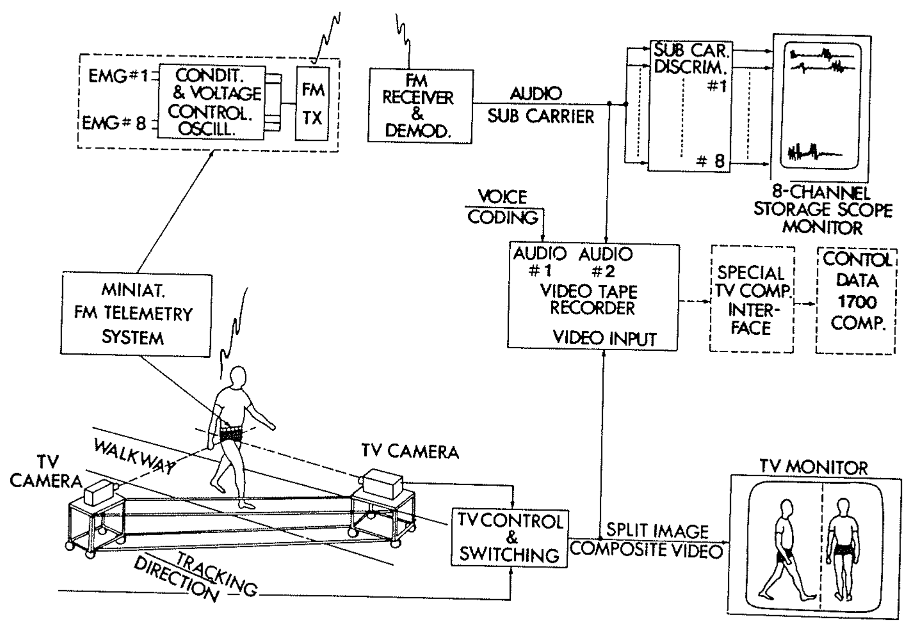


Fig. 3.13. Gait studies data acquisition system.

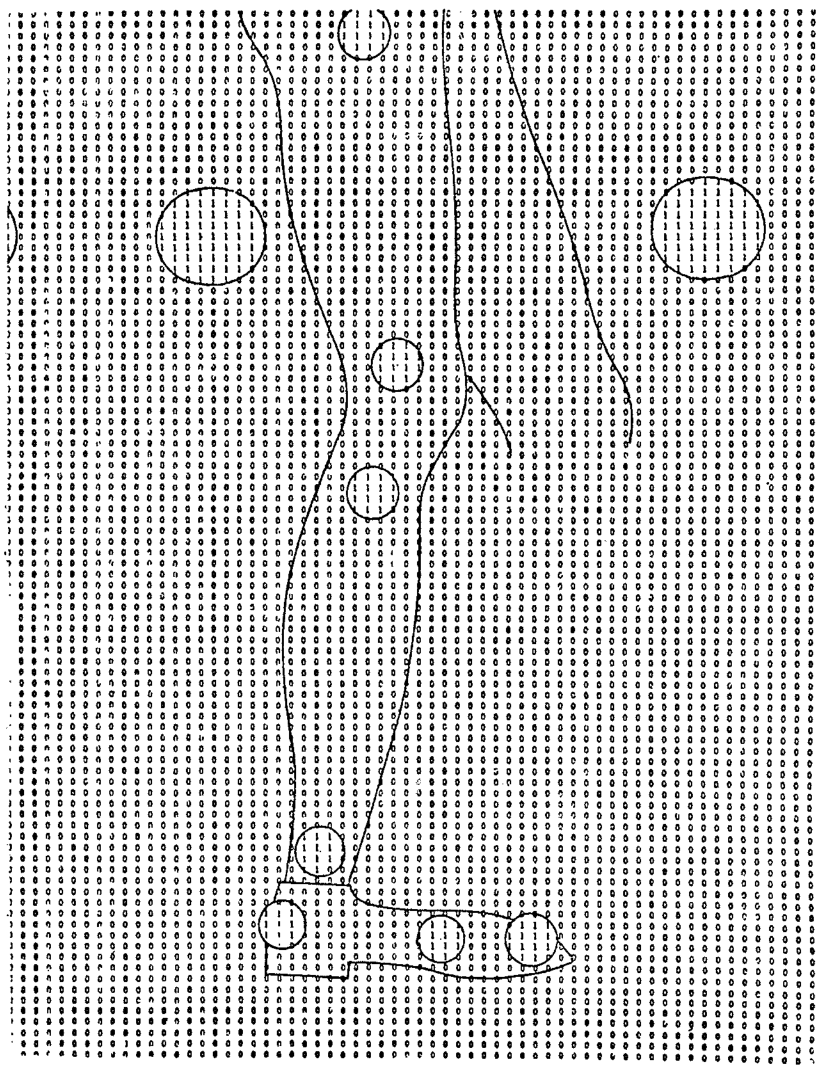


Fig. 3.14. Computer print-out of one converted TV field (from Winter, et al, 1972).

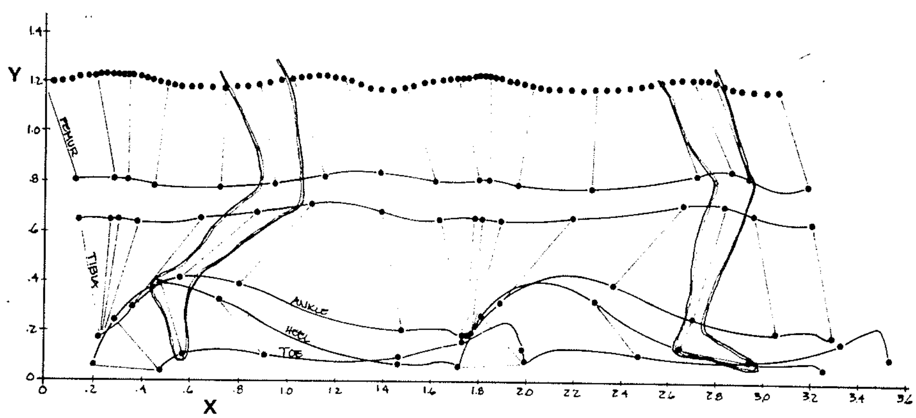


Fig. 3.15. Trajectory plot of spatial movement of body marker.

X - distance along walkway (meters);
Y - height (meters).

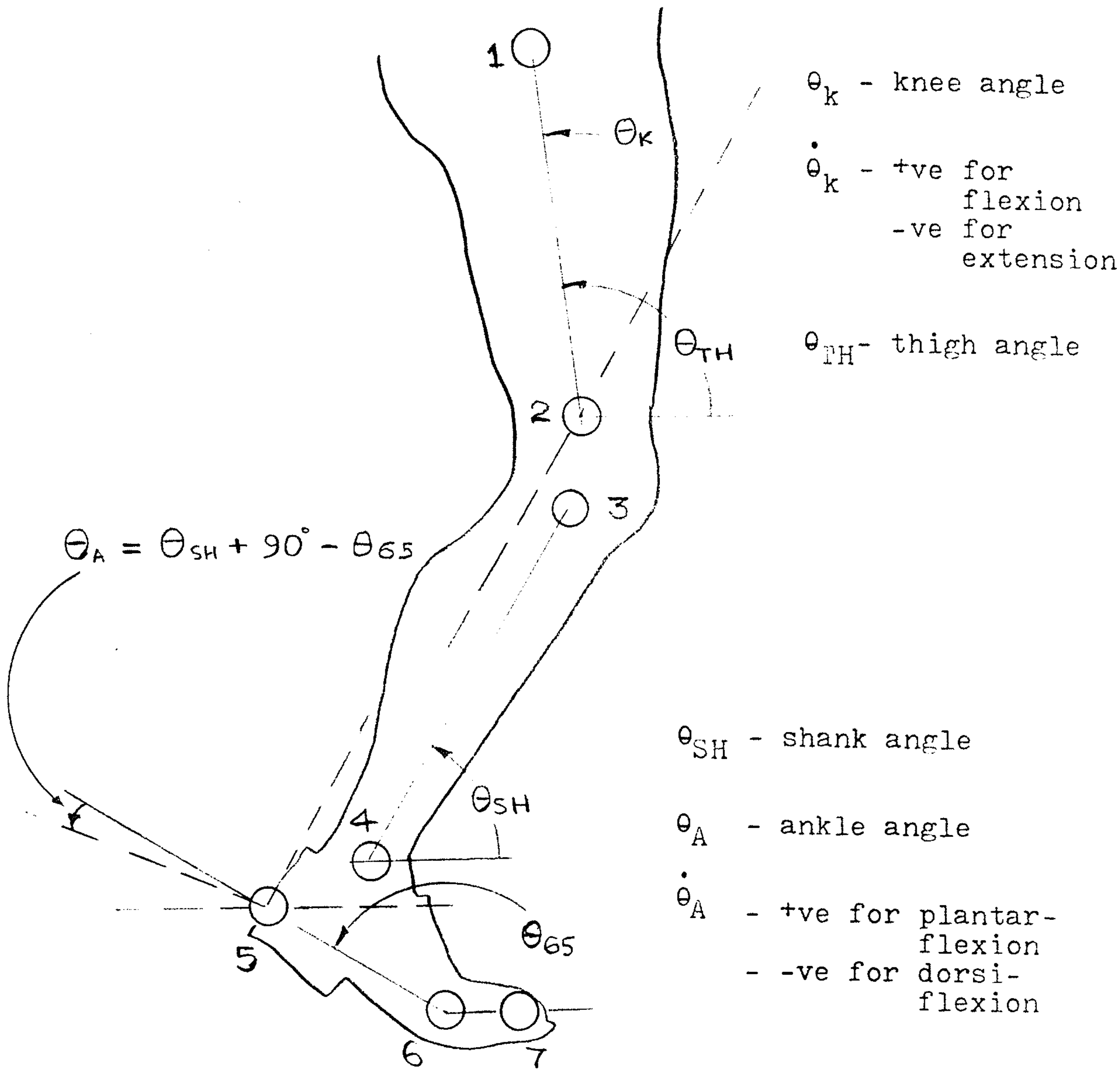
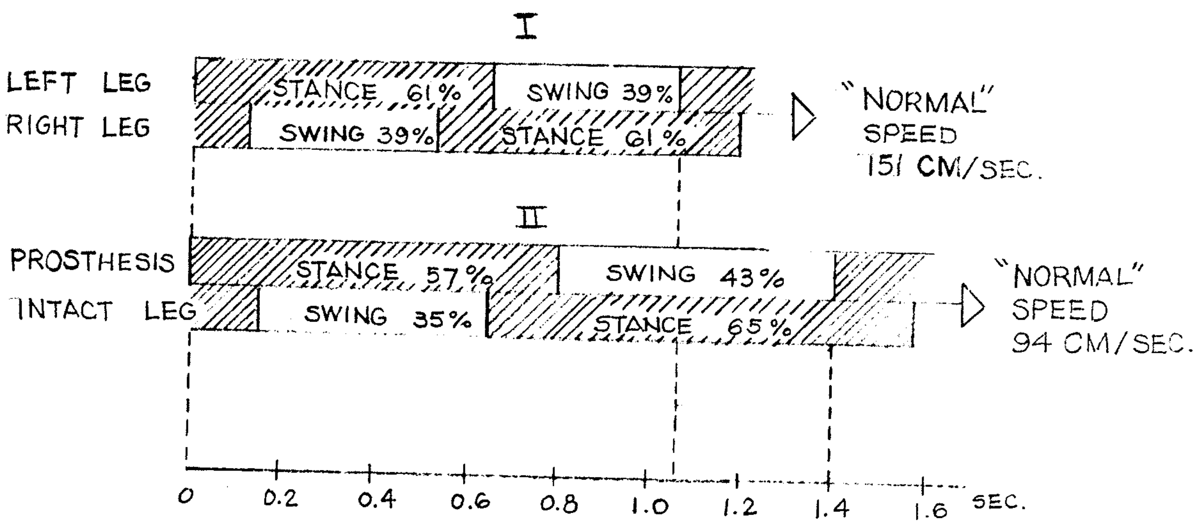


Fig. 3.16 Body markers relating limb angles.



I "Normal" reference material (Murray et al 1966)

II Unilateral above-knee amputees (James, 1973)

Fig. 4.1. Stance and swing ratios on normals and A/K amputees (from James, 1973).

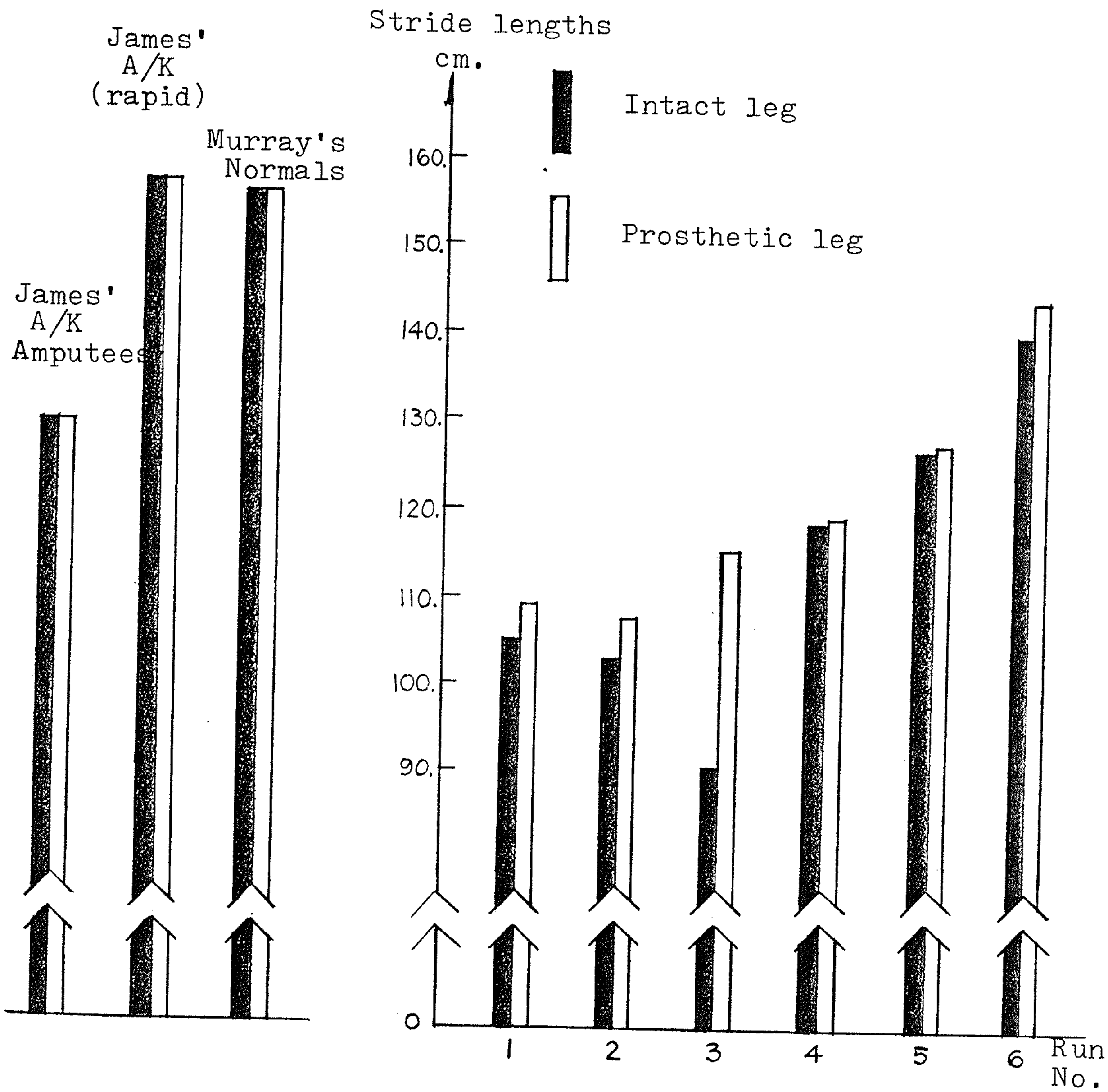


Fig. 4.2. Stride lengths per walking cycle.

Run-1: Expt. pros. - during early training,
Run-2: Henschke-Mauch pros.,
Run-3: Conventional pros. - no damping,
Run-4: Expt. pros. - electronics disabled,
Run-5: Expt. pros. - after training,
Run-6: Expt. pros. - fast cadence.

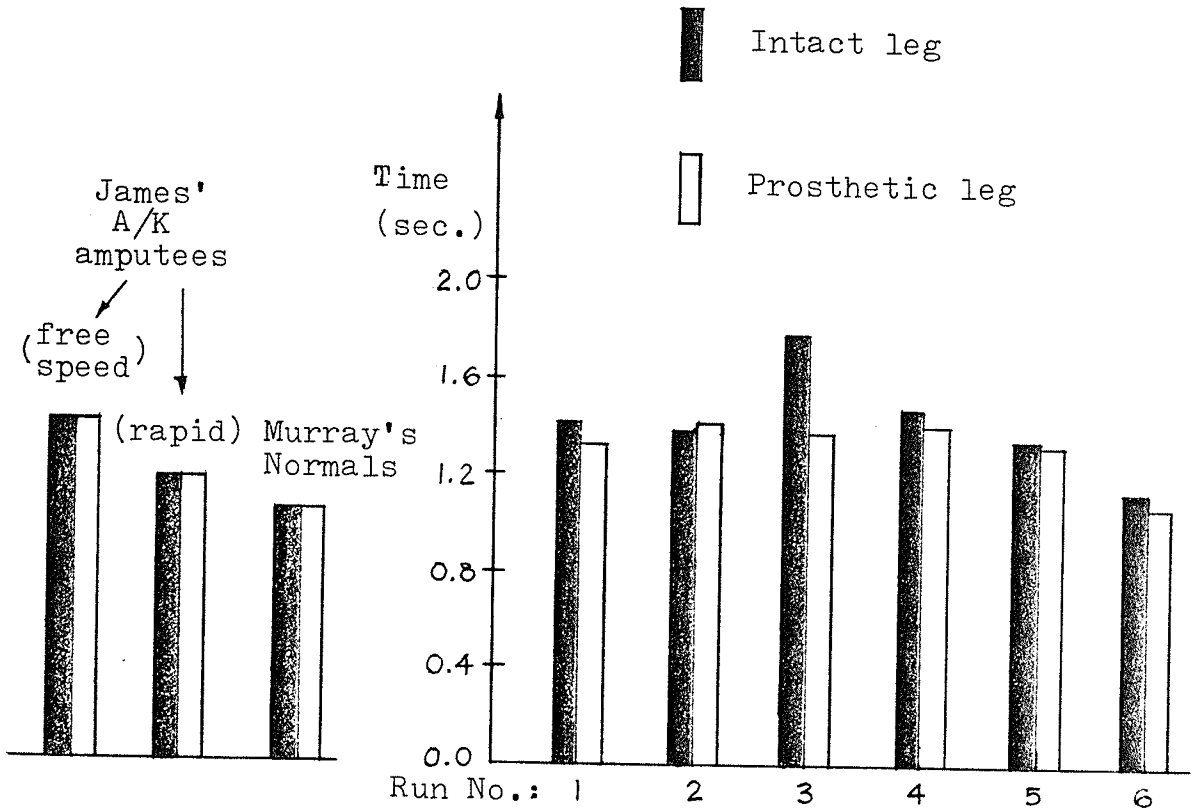


Fig. 4.3. Time durations per walking cycle.

Run-1: Expt. pros. - during training,
Run-2: Henschke-Mauch pros.,
Run-3: Conventional pros. - no damping,
Run-4: Expt. pros. - electronics disabled,
Run-5: Expt. pros. - after training,
Run-6: Expt. pros. - fast cadence.

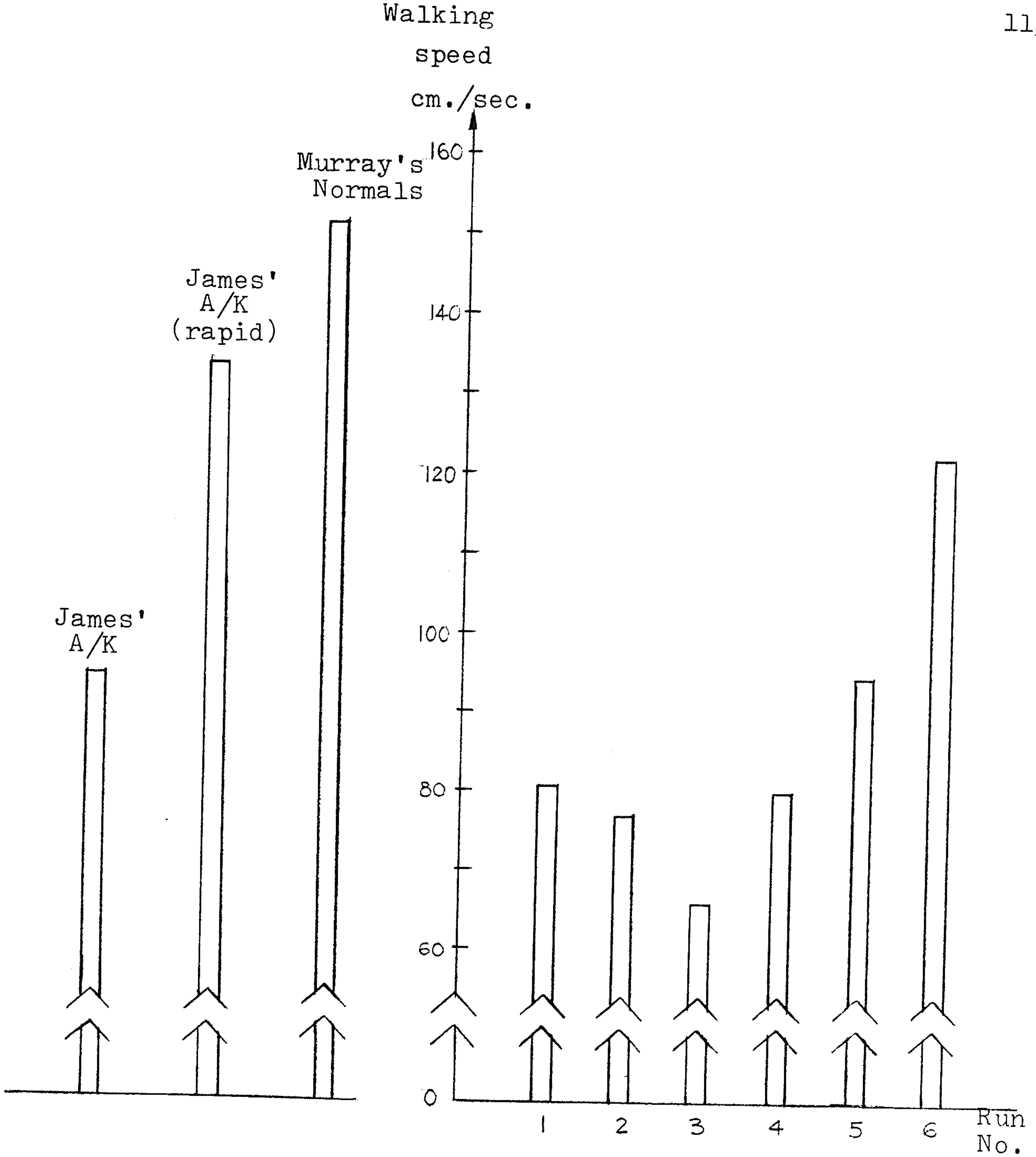


Fig. 4.4. Speeds of progression of body.

Run-1: Expt. pros. - during early training,
Run-2: Henschke-Mauch pros.,
Run-3: Conventional pros.- no damping,
Run-4: Expt. pros.- electronics disabled,
Run-5: Expt. pros.- after training,
Run-6: Expt. pros.- fast cadence.

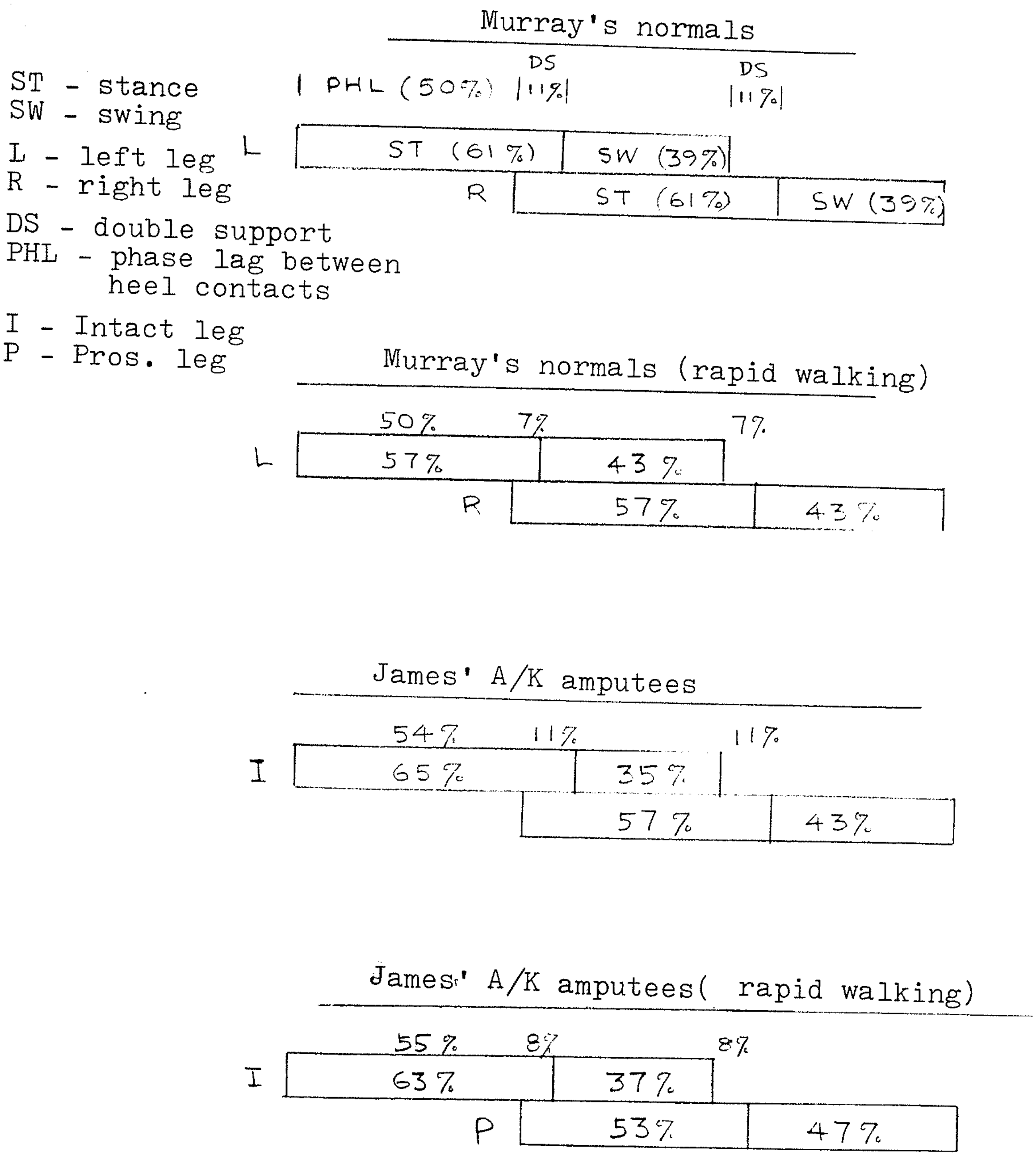


Fig. 4.5(a). The phasic relationship between legs of known data.

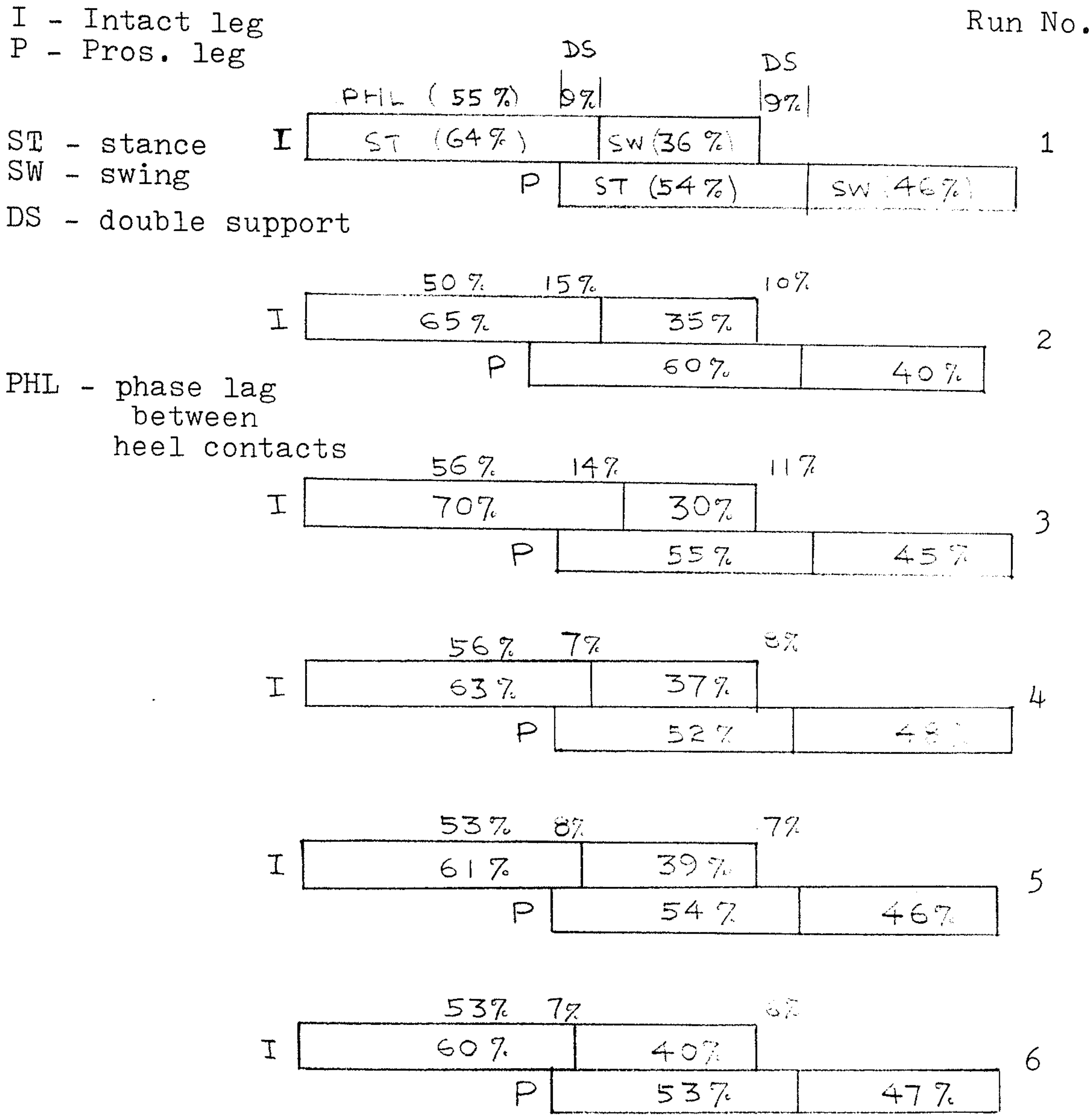


Fig. 4.5(b). The phasic relationship between legs (from the runs of this study).

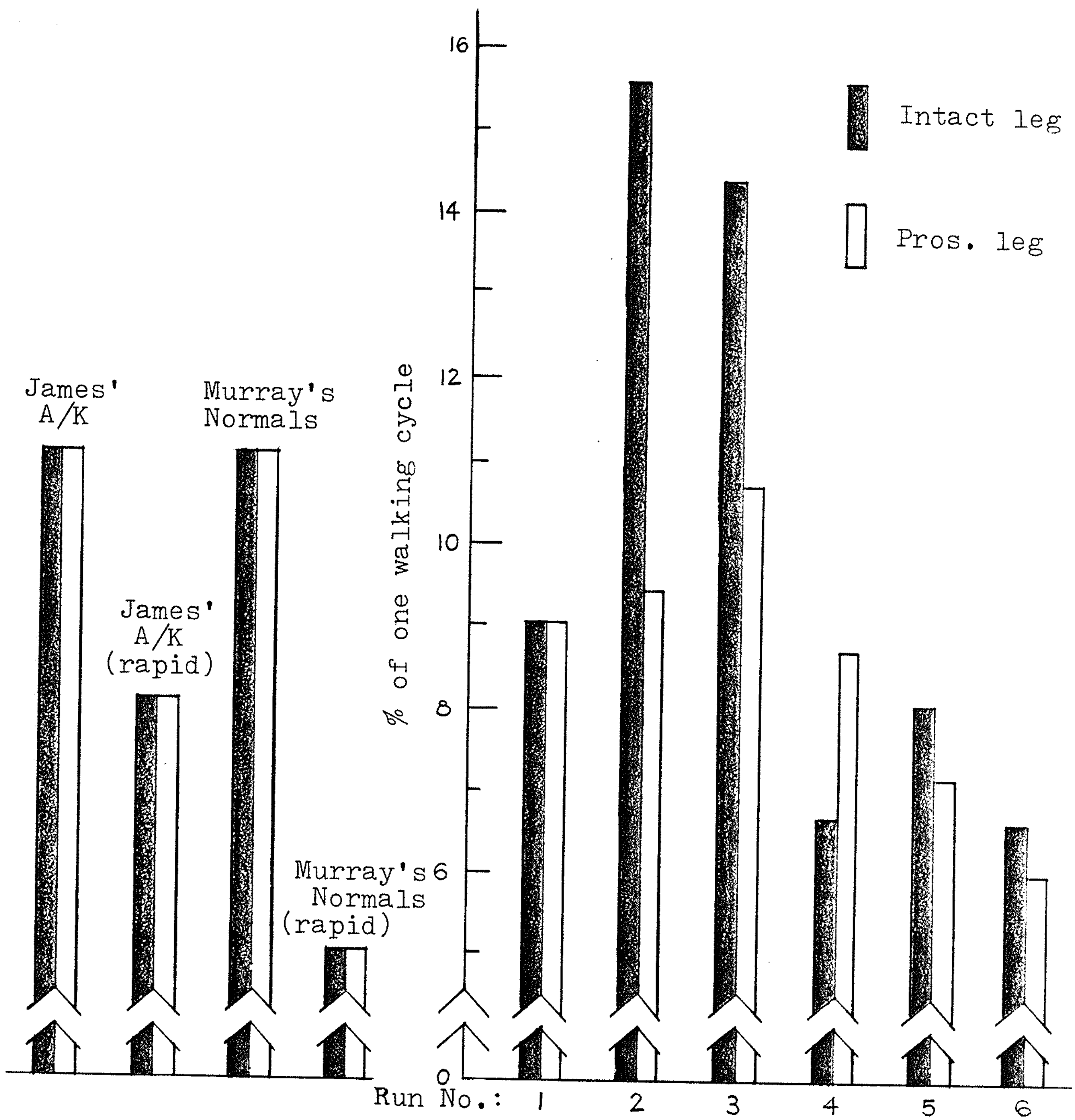


Fig. 4.6. Double support periods per one walking cycle.

Run-1: Expt. pros. - during early training,
Run-2: Henschke-Mauch pros.,
Run-3: Conventional pros. - no damping,
Run-4: Expt. pros. - electronics disabled,
Run-5: Expt. pros. - after training,
Run-6: Expt. pros. - fast cadence.

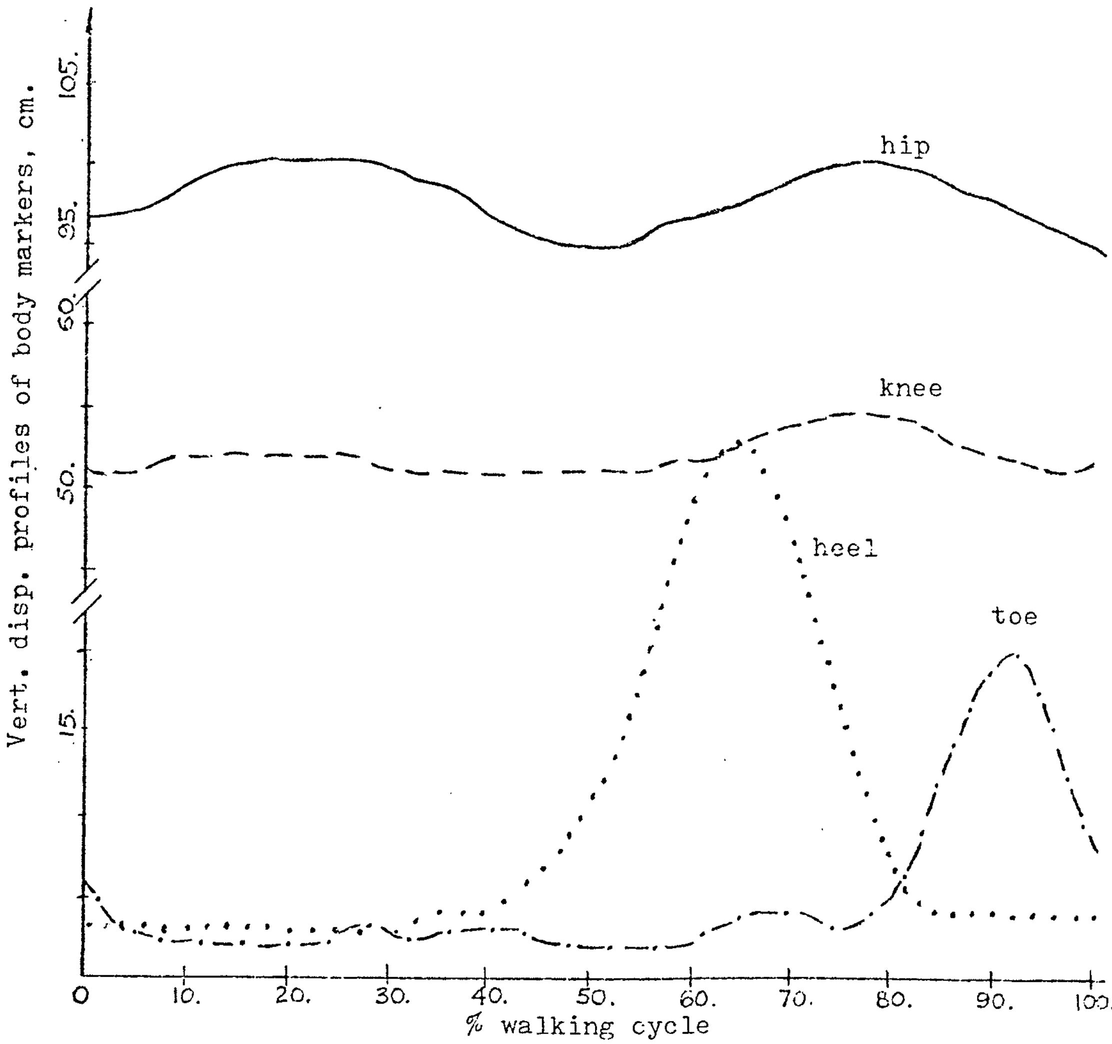


Fig. 4.7. Vertical displacement profiles of body markers. (Intact side of Run-5).

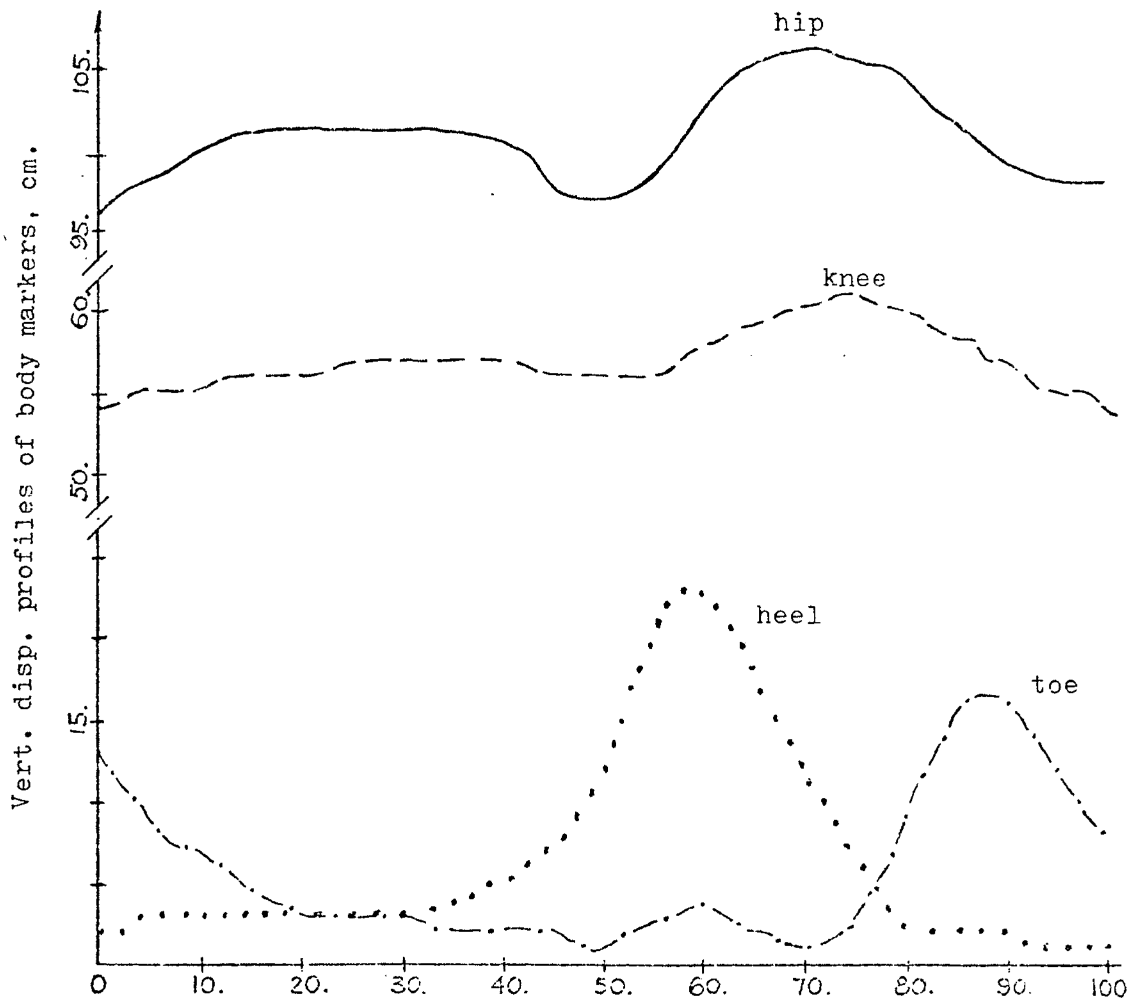


Fig. 4.8. Vertical displacement profiles of body markers (prosthetic side of Run-5).

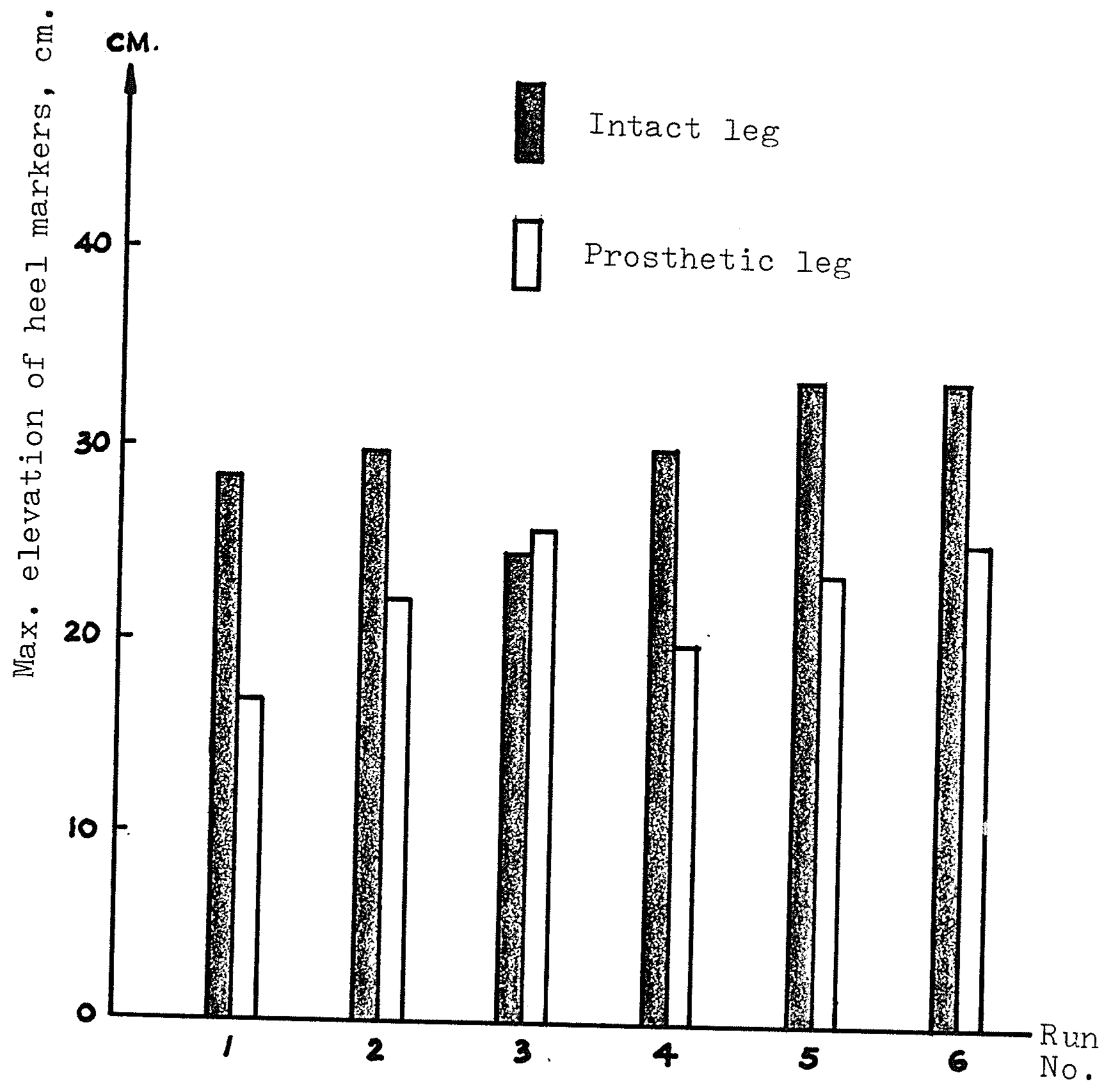


Fig. 4.9. The maximal elevation of the heel marker.

Run-1: Expt. pros. - during early training,
Run-2: Henschke-Mauch pros.,
Run-3: Conventional pros. - no damping,
Run-4: Expt. Pros. - electronics disabled,
Run-5: Expt. pros. - after training,
Run-6: Expt. pros. - fast cadence.

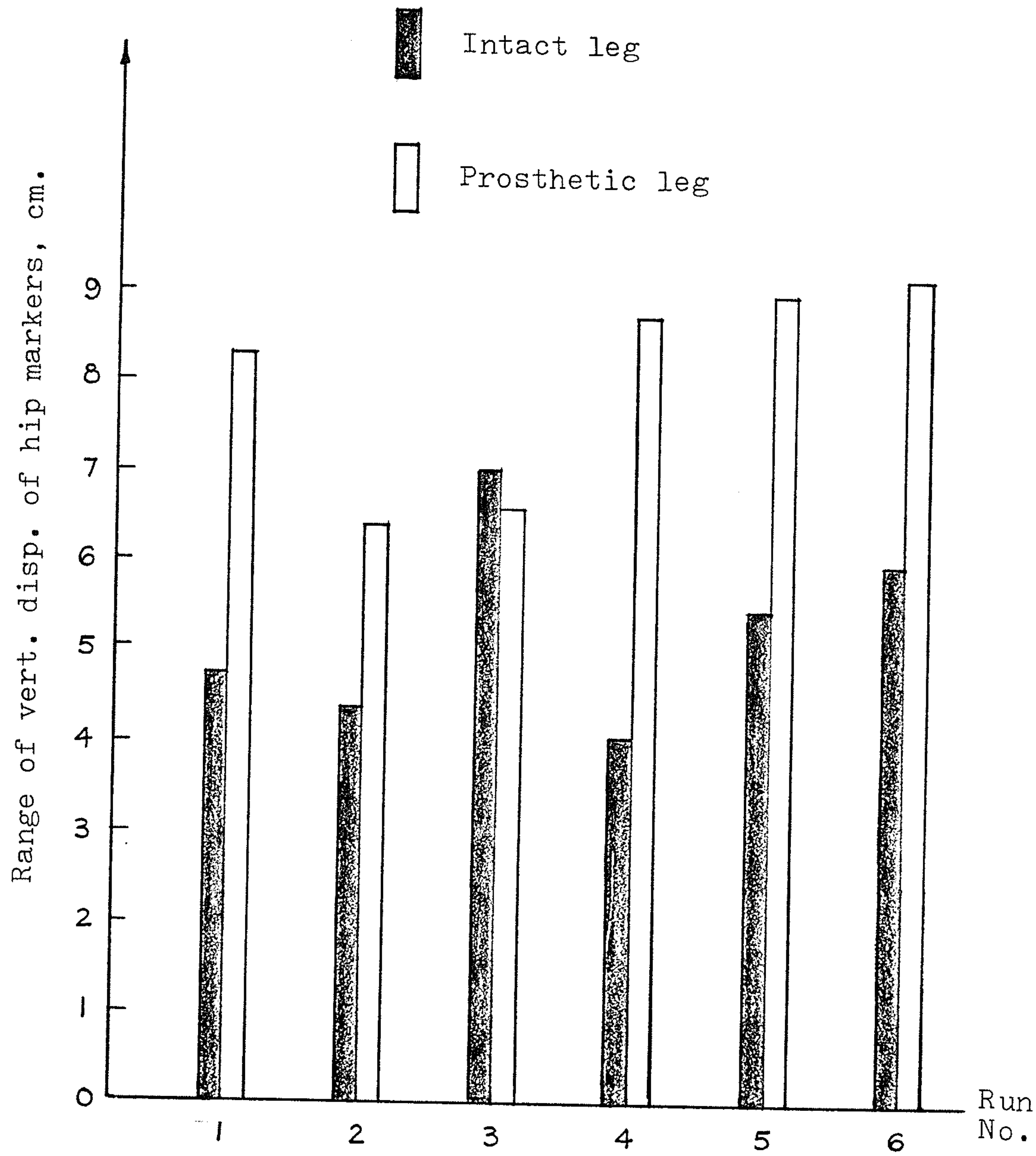


Fig. 4.10. The range of vertical displacement of the hip markers.

Run-1: Expt. Pros. - during early training,
Run-2: Henschke-Mauch Pros.,
Run-3: Conventional pros. - no damping,
Run-4: Expt. pros. - electronics disabled,
Run-5: Expt. pros. - after training,
Run-6: Expt. pros. - fast cadence.

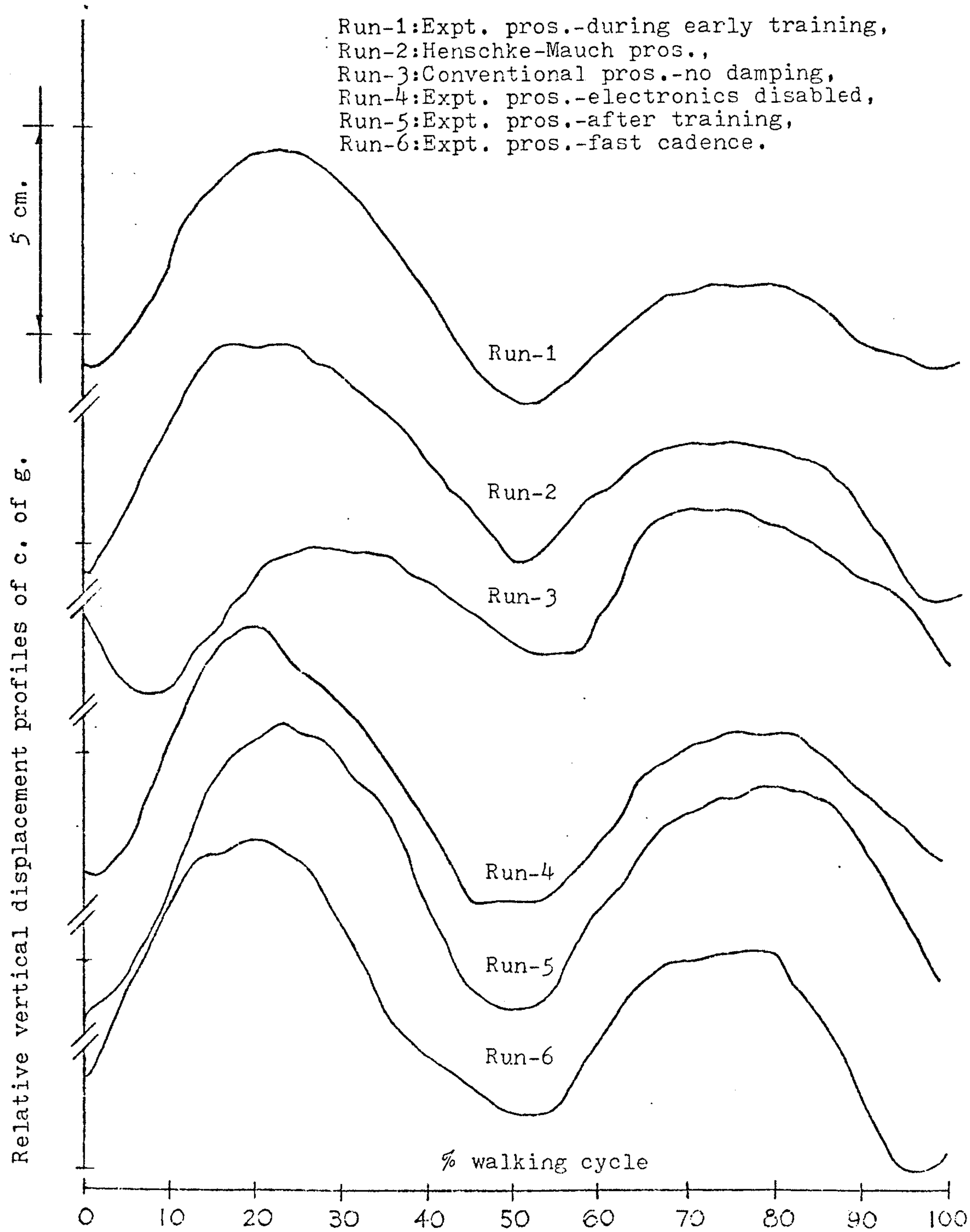


Fig. 4.11 Vertical displacement profiles of centre of gravity of the body.

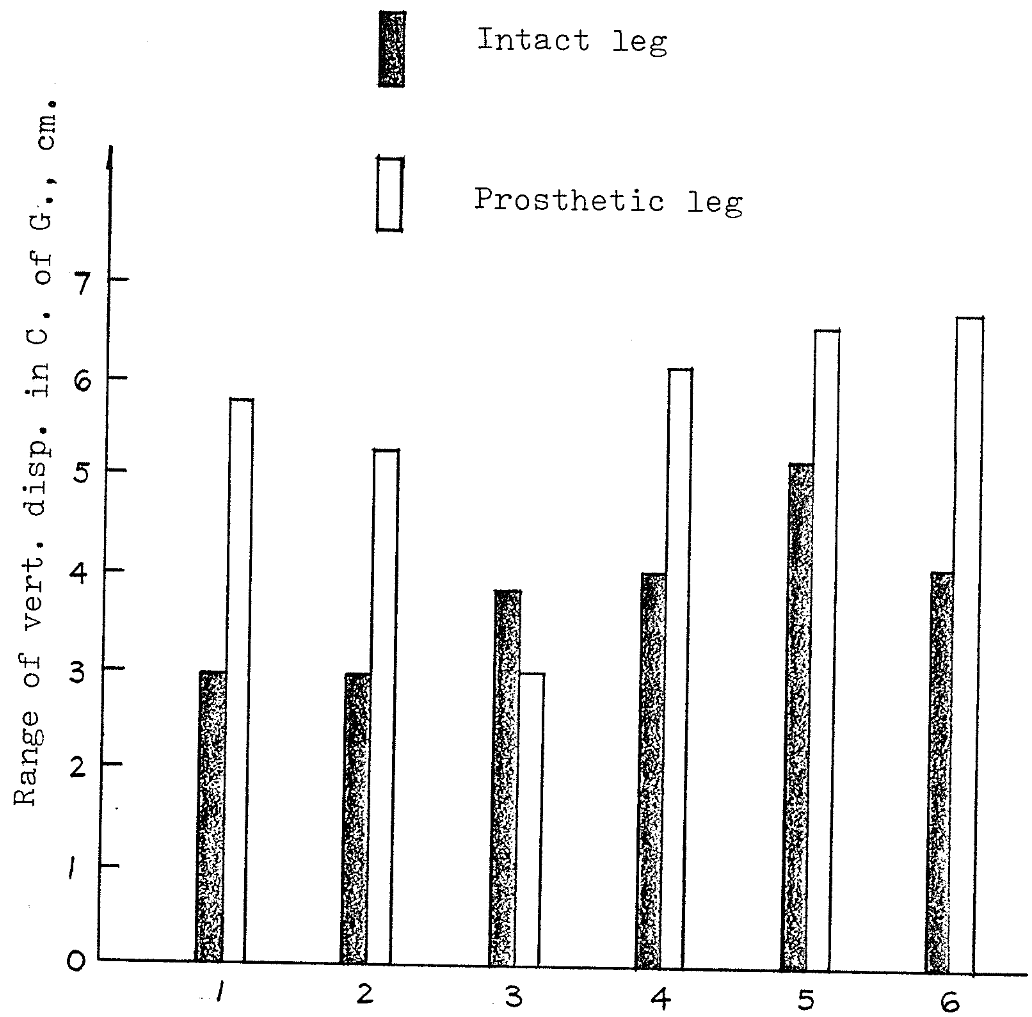


Fig. 4.12 The range of vertical displacement in the centre of gravity of the body.

Run-1: Expt. Pros. - during early training,
Run-2: Henschke-Mauch pros.,
Run-3: Conventional pros. - no damping,
Run-4: Expt. pros. - electronics disabled,
Run-5: Expt. pros. - after training,
Run-6: Expt. pros. - fast cadence.

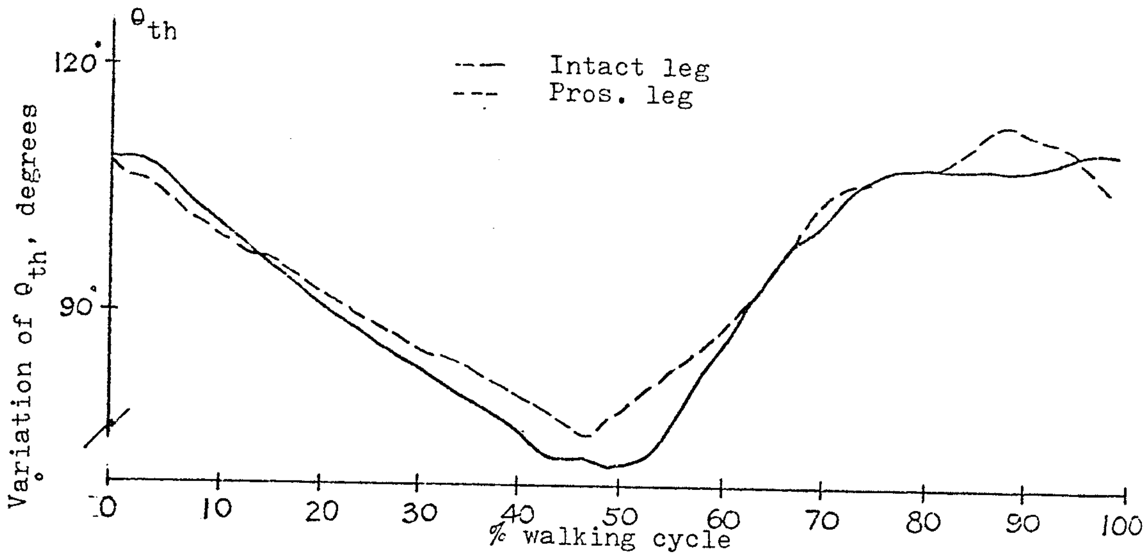


Fig. 4.13. The variation of the thigh angles(Run-5).

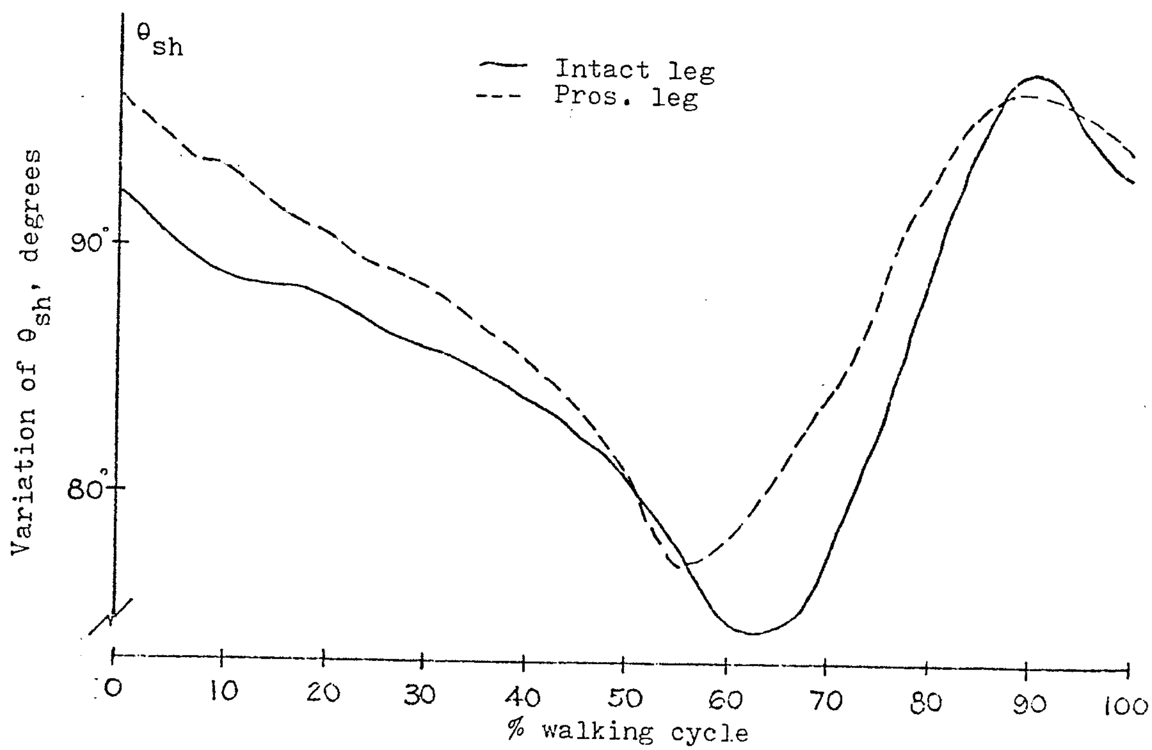


Fig. 4.14. The variation of the shank angles(Run-5).

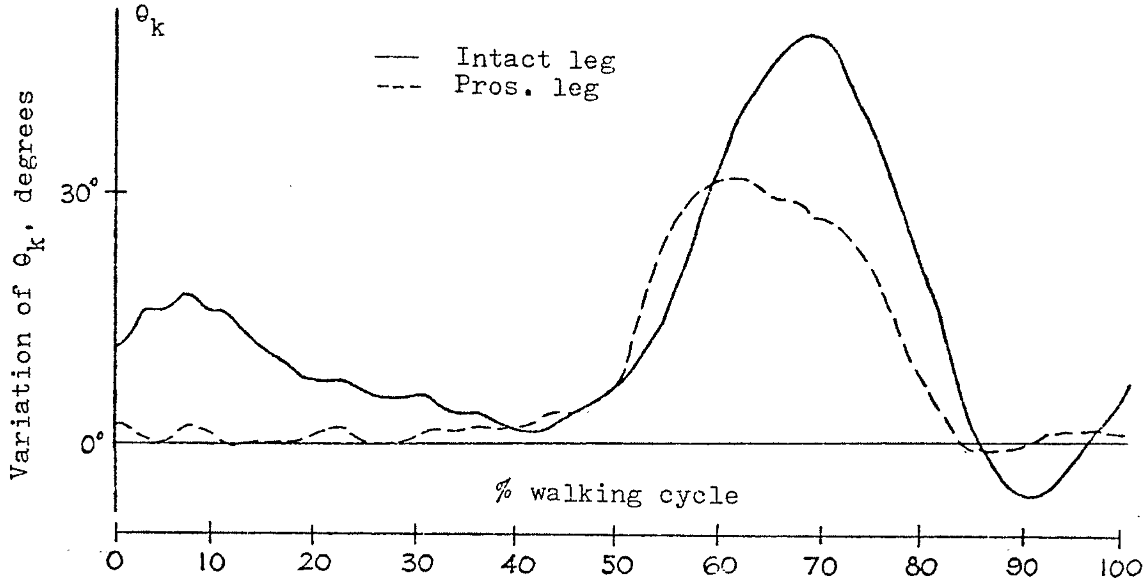


Fig. 4.15. The variation of the knee angles(Run-5).

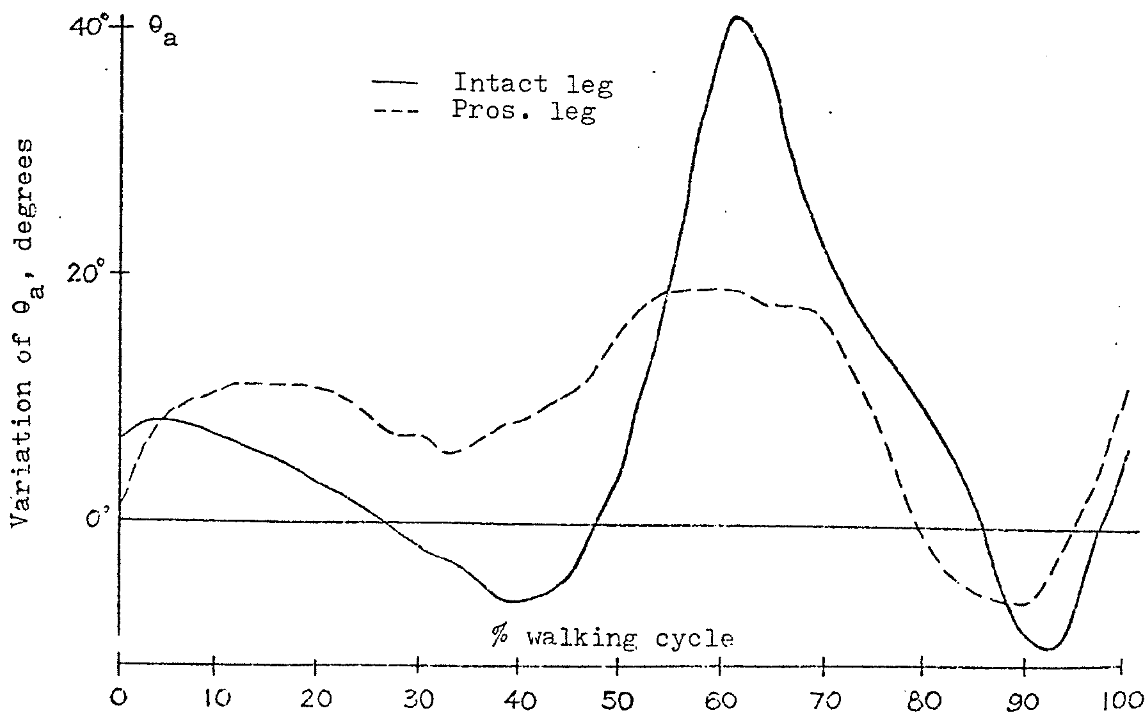


Fig. 4.16 The variation of the ankle angles(Run-5).

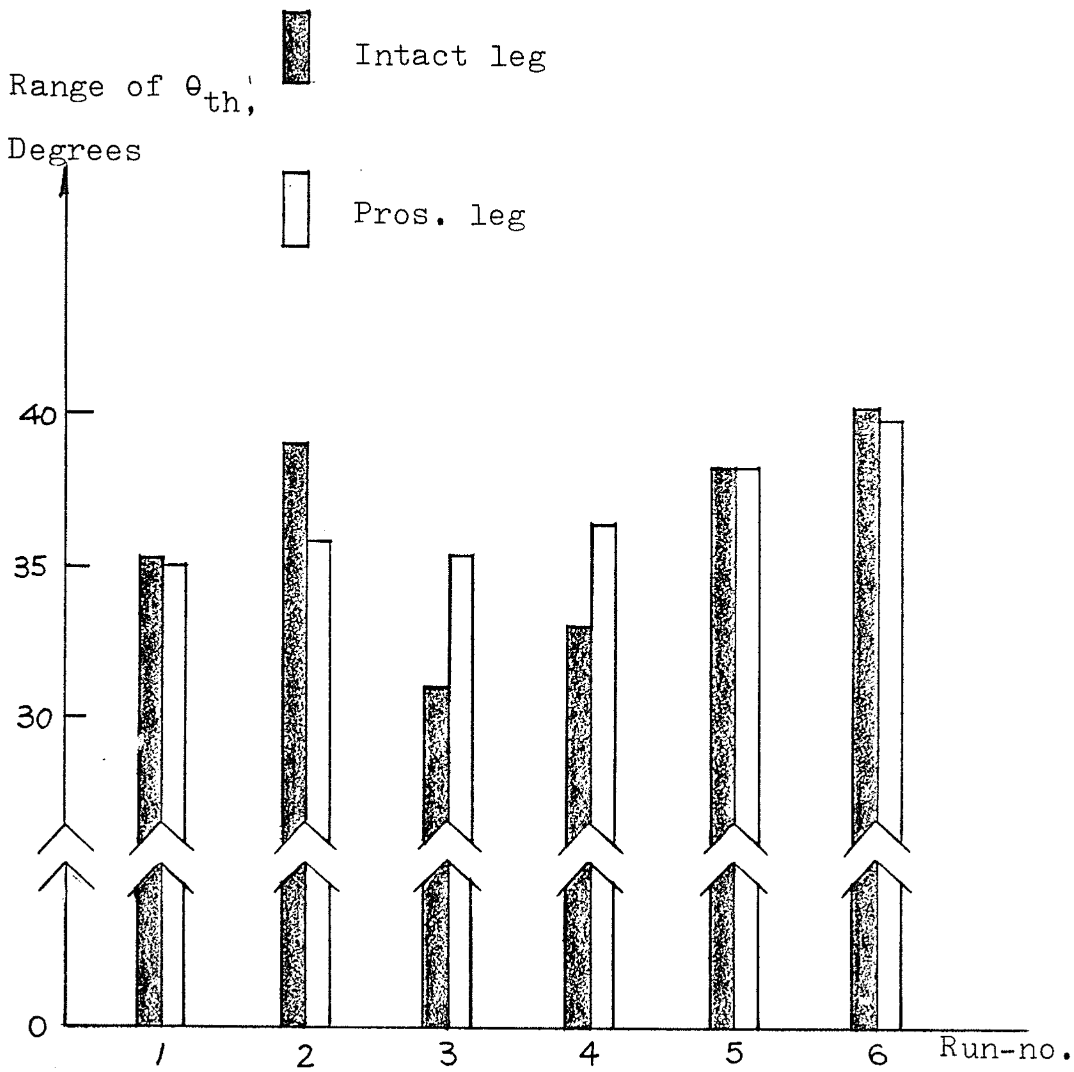


Fig. 4.17. The range of angular changes of the thighs.

Run-1: Expt. pros.-during early training,
Run-2: Henschke-Mauch pros.,
Run-3: Conventional pros.-no damping,
Run-4: Expt. pros.-electronics disabled,
Run-5: Expt. pros.-after training,
Run-6: Expt. pros.-fast cadence.