

THE UNDERSTANDING OF THE BIOMECHANICS OF
STANDING-UP AND SITTING-DOWN

by

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ABSTRACT

Functional Neuromuscular Stimulation (FNS) has been shown to be effective in restoring certain upper and lower limb movements of spinal cord injured patients. The successful application of FNS to achieve standing-up and sitting-down, in persons with lower limb paralysis, depends largely on two factors: firstly, the torques generated at the hip, knee and ankle joints during these movements, and secondly, the extent and distribution of the accompanying muscular activity. With this knowledge it is possible to determine the electrical stimulation parameters and sequence required to restore the above movement in spinal cord injured patients. This thesis has a two-fold aim to study these two factors.

A mathematical model to predict the torques generated at the hip, knee and ankle joints of individual participants was developed to achieve the first aim. This model is presented in the form of a computer program which requires anthropometrical and experimental data from the subject. The model was developed both from kinematic and kinetic aspects. The kinematic approach requires the varying angles at the hip, knee and ankle joints as inputs, whereas the kinetic approach requires the position of the ground reaction force in relation to the lateral malleolus, together with the ankle angles, as inputs. The various experiments performed to obtain the above mentioned variables are described. The second aim was achieved using electromyography to determine the sequential muscle activity, of the lower limb, during standing-up and sitting-down procedures. A description of the method used to obtain these electromyograms, together with a diagrammatic representation of the results, is presented.

As a result of the above mentioned theoretical modelling as well as simulation and validation experiments, the following conclusions can be made: (1) The mathematical model does predict reasonable torque values compared to that obtained in the literature. (2) The diagrammatic representation of the muscular activity is an acceptable reflection of the actual muscle activity. (3) In the light of the various problems encountered, a number of recommendations relating to both the mathematical model and the sequence of muscular activity are made.

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Nomenclature

W_{foot} = W_f : weight of the foot in Newtons

W_{calf} = W_c : weight of the calf (leg) in Newtons

W_{thigh} = W_t : weight of the thigh in Newtons

W_{HAT} = W_H : combined weight of the head, arms and trunk

in Newtons

SCI = spinal cord injured

COG = centre of gravity

GRF = ground reaction force

pGRF = distance between the line of action of GRF and the
lateral malleolus

calf_l = length of the calf in m

thigh_l = length of the thigh in m

x_foot = distance between COG of foot and lateral malleolus

x_calf = distance between COG of calf and lateral malleolus

x_thigh = distance between COG of thigh and lateral malleolus

x_hat = distance between COG of HAT and lateral malleolus

T_A = torque about the ankle joint

T_K = torque about the knee joint

T_H = torque about the hip joint

Glossary

FNS	Functional Neuromuscular Stimulation may be defined as the external control of innervated paretic or paralytic muscles, by electrical stimulation of the corresponding intact peripheral nerves or muscles, to achieve functional movement for rehabilitation purposes (Kralj et al, 1977(a); Benton et al, 1981).
MSS	The musculo-skeletal system, MSS, is a linkage system consisting of bones, muscles and ligaments with saddle, hinge, ball and socket etc joints.
EMG	Electromyogram is the graphic recording of the intrinsic electrical activity of skeletal muscle when contracting.
GRF	The ground reaction force, GRF, is the upward force acting underneath each foot when in contact with a surface. Also known as the centre of pressure.
COG	The centre of gravity is the point through which the GRF acts, provided that there is no horizontal acceleration.
HAT	In modelling the human body one can view it as a symmetrical linked-segment model of which one of the segments can be the head, arms and trunk (HAT).

CHAPTER 1

INTRODUCTION

The human body is a complex combination of various systems. One such system, the musculoskeletal system (MSS), consisting of bones joined together by ligaments and muscles, provides the framework for the human body. The activity of the MSS is controlled by an intricate network of nerves, thus enabling a myriad of coordinated functions and movements to be performed.

The act of rising from a seated to a standing position is such a common occurrence that it is taken for granted. Despite its outward simplicity, this process, and that of maintaining a standing posture, requires a complex interaction between the musculoskeletal and neuronal systems. This becomes evident when, due to severe damage to either of these systems controlling the lower limbs, an individual experiences difficulty in standing. Damage to the spinal cord could result in the neural pathway between the brain and the muscles being disrupted, resulting in either partial or total paralysis of these muscles. The actual muscles affected by the paralysis will depend on the level of the spinal lesion. The nerves for the upper limb muscles originate from the cervical region, whereas those for the lower limb muscles, originate from the lower thoracic/lumbar region. Therefore, the more cephalic the lesion, the greater the extent of the paralysis. Tetraplegics (quadriplegics) are persons who have a cervical lesion resulting in minimal or no control of their upper limbs and generally none of their lower limbs. Paraplegia is the result of a lower lesion level (thoracic or lumbar), consequently, paraplegics have the use of their upper limbs and trunk but not of their lower limbs.

Following trauma to the spinal cord an individual is usually confined to a wheelchair, generally their principal means of

perambulation for the rest of their lives. After a relatively short period in the wheelchair the muscles of the paralyzed limb start to atrophy due to disuse (Vodovnik, 1981). Other associated complications are pressure sores, reduced blood flow to the paralyzed limbs, osteoporosis in the paralyzed limbs, and malfunctioning of the urinary tract as well as other internal organs. There are also associated psychological problems.

Research work, in the area of spinal cord injury, has focused on neural regeneration, spinal cord transplants, chemotherapy and electrical stimulation of the paralyzed muscles, (Kao et al, 1983). Neural regeneration, which would obviously provide the ideal solution, is very much in its infancy.

Spastic tetraplegics and paraplegics, who suffer from upper motor neuron lesions generally have a normal spinal arc reflex. The lower neurons and the muscles they innervate are therefore still functional, but remain uncontrolled as they receive no input from the brain. The paralyzed muscles of these persons can be activated by electrical stimulation. Functional Neuromuscular Stimulation (FNS) may be defined as the external control of innervated paretic or paralytic muscles, by electrical stimulation of the corresponding intact peripheral nerves or muscles, to achieve functional movement for rehabilitation purposes (Kralj et al, 1977(a); Benton et al , 1981). FNS utilizes the still intact lower motor neuron system to provide control of the muscles. FNS can either complement or replace bracing which is the only reliable, albeit cumbersome, way of rehabilitation at the present. FNS, at the correct frequency and intensity, can be used to restrengthen these muscles to such an extent that supported standing and walking can be achieved by them under strict .

laboratory conditions.

The pioneering work of Liberson and his co-workers in 1961, using electricity to stimulate the dorsiflexors of the ankle joint in hemiplegics synchronously with gait, has led to numerous research projects in the field of FNS. In paraplegics and some partial tetraplegics, standing and walking with the aid of mechanical orthotic supports (callipers) has been the conventional method of rehabilitation. The success achieved by these methods has been very limited, as most users discard their appliances soon after discharge from the rehabilitation centre, invariably finding them too difficult to attach to their spastic limbs, (Isakov et al, 1986; Rosman and Spira, 1974). Much of the research work therefore, has been concentrated on achieving walking in paraplegics, with varying degrees of success.

A prerequisite for the use of FNS for walking is its use in standing. In practice, standing requires, at least, bilateral stimulation of the quadricep muscles, (Marsolais and Kobetic, 1982). Only once standing has been achieved by FNS can the problem of walking be addressed. Standing by the person with spinal cord injury is likely to reduce the chances of the development of pressure sores and improves the functioning of the internal organs as well as the patient's psychological state. Weight bearing on the legs also has a positive effect on the long bones in that it prevents osteoporosis. Very little quantitative biomechanical analysis of standing-up has, however, been done to date.

Several mathematical models relating to the biomechanical of movement have been developed. There are a large number of

muscles and ligaments crossing each joint in the lower limb. Mathematical modelling of the human body in motion has therefore resulted in an undetermined set of muscle-force equations with an infinite number of solutions. Various researchers have approached the indeterminacy of the muscle-force equations in a number of ways. While Crowninshield et al, (1978) used the reduction method, Chao and Rim, (1973) developed an interactive method to calculate the joint torques. A popular method used was to estimate the muscular forces so as to optimise some kinetic criteria (Seireg and Arvikar, 1973,1975; Crowninshield, 1978; Hardt, 1978; An et al, 1984 (a)). Other groups utilized force plates and stroboscopic photography to calculate the joint torques required for standing up (Bajd et al, 1982; Hart et al, 1984). However, there remains a certain amount of uncertainty as to the maximum knee-joint torque required, with the reported value ranging from 30 to 200 Nm (Bajd et al, 1982; Hardt et al, 1984; Kelley et al, 1976).

In view of all the complexities involved in the use of optimisation, a model which best suited the actual problem at hand seemed warranted, particularly one which would predict the various joint torques for the standing-up process and which would also be applicable to different anthropometrical data.

The aims of this thesis are therefore:

1. To develop a mathematical model of the lower limb for the standing-up and sitting-down process.
2. To establish an acceptable prescribed standing-up and sitting-down processes for normal persons. The dynamics of these movements will then be analysed.

The thesis is divided into four sections. The first section deals with the necessary preparatory material, including a literature survey, a brief discussion of the anatomy of the lower limb, a survey of the relevant nerve and muscle physiology, the physiology of electromyograms and a brief introduction to FNS. Section II presents the initial development of the mathematical model. From the development of the model it was possible to establish which variables were required for its completion. Some of these variables were obtained either from the literature, direct measurement, or experimental work. Electromyographic experiments were also conducted to establish the activity of the vastus medialis, rectus femoris, biceps femoris, peroneus tertius, gastrocnemius and gluteus maximus muscles of the lower limbs during standing-up. Section III utilises the experimentally or otherwise obtained values for the variables to finalize the mathematical model. The development of the computer program is also included. The final section, Section IV, contains the results obtained using the model, a summary, discussion and the conclusions reached during this research work. Finally, some recommendations for future work in this field are given.

SECTION I
PREPARATORY MATERIAL

CHAPTER 2

LITERATURE SURVEY

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There are typically two aspects of human movement that can be modelled mathematically: 1) the biomechanical system, and 2) the neuro-control of the biomechanical system (Winter et al, 1981). The musculoskeletal system (MSS), as a linkage system consisting of bones, muscles and ligaments with saddle, hinge, ball and socket etc joints, lends itself well to modelling. The biomechanical modelling of the MSS is the most challenging task for researchers in this field (as is evident by the numerous models available), (Seireg and Arvikar, 1973 and 1975; Chao and An et al, 1978; Jaeger, 1986; Crowninshield, 1978; Crowninshield and Brand, 1981; Hardt, 1978).

Human motion requires the co-ordinated contraction and relaxation of many muscles, which may cross one or more joints. The joints themselves may move about one, two or three axes. The purpose of mathematical models of the MSS is to predict the forces generated by the muscles and in the joints, under various loading conditions and movements. Knowledge of these forces is essential for the design of implants, orthoses, surgical procedures and medical equipment, and is valuable in understanding the various pathologies of the joints.

2.1 Mathematical Models of Musculo-skeletal System (MSS)

Quantitative *in vivo* experiments are often not feasible. Not only are they invasive, but the system studied is so complex that deductions of loading from purely experimental findings would be extremely difficult, if not impossible. Rydell, (1966) attempted to obtain the hip contact force during gait by using an instrumented prosthesis. However, Pauwels, (1980) found that the forces he recorded did not equal the forces in a normal joint.

Ghista et al, (1976) used electromyography (EMG) to determine muscular forces, using the assumption that the force exerted by a muscle depends exponentially on the time-derivative of the rectified integrated EMG signal. Although EMG can give some indication of muscular activity (Hogan, 1976), it does not provide the quantitative accuracy required, nor does it allow access to all muscles of interest (Crowninshield, 1978; Hardt, 1978). It is as the result of all these difficulties that mathematical modelling of the MSS has not been accurate enough and thus has not found wide application in clinical problems.

The calculation of the joint forces and moments is thus an extremely difficult task. The large number of muscles and ligaments crossing a joint, being the unknown variables, far exceed the number of equilibrium equations, thus usually rendering the problem indeterminate. This complexity compels the modeller to make "gross" anatomical and functional simplifications to render the equations solvable.

The majority of lower limb modelling work has been biased towards gait research. As this has significant relevance to standing up, it is prudent to survey these models.

2.2 The Problem of Indeterminacy in Mathematical Models

The early work in gait analysis utilised photographic measurement (Murray et al, 1964). Bresler and Frankel, (1950) calculated the forces and moments at the leg joints for normal level walking by combining the simultaneous leg position and foot-floor reaction with the mass moments of inertia. Their work can be considered as classical. Consequently their techniques have laid down the

foundations for further research and have been repeatedly used by researchers worldwide.

The indeterminacy of the muscle-force equations, resulting from mathematically modelling the lower limb, has been approached in a number of ways. One of these is called the "reduction" method (Crowninshield et al, 1978). The excess unknown variables are reduced by either grouping functionally similar muscles and their origin and insertion points, or by eliminating them with the aid of EMG studies. The knee joint model by Morrison, (1968) attempted to compute the muscle and ligament forces during gait. The redundancy problem was solved by eliminating forces in ligaments which became slack, during various phases of gait. The bone contact forces as joint forces thus obtained varied considerably from subject to subject but were comparable for the same subject who underwent multiple tests.

Models of the hip joint were initially formulated in basically the same way as those of the knee (King, 1984). Paul, (1967) assumed that the hip joint transmitted a contact force and that no more than two muscles were active at any instant of gait. Kinematic and force plate data were required by this model. An alternative method was to introduce additional equations based on the hypothesis that force distribution between muscles depends on their cross-sectional area during maximum force contractions (Amis et al, 1980).

Chao and Rim, (1973) developed an iterative process to calculate joint-torques. Because of the inaccurate method of obtaining limb accelerating by differentiating data obtained for limb position they used the "inverse dynamic approach". By

taking an initial moment profile during gait and then working backwards through the kinetics and kinematics, they were able to obtain a limb position profile. By optimising this profile, with the known experimental profile, they were able to compile an acceptable moment profile.

The more common method of solving the indeterminacy problem is to estimate the muscular forces so as to optimise (minimize or maximize) some kinetic criteria (called the objective function), such as the summation of muscular forces, (Crowninshield, 1978; Seireg and Arvikar, 1973) or the weighted, linear combinations of muscle and joint forces (Seireg and Arvikar, 1975). Crowninshield and Brand, (1981) also included muscle endurance as an objective function by examining stress and relating it to endurance.

Thus a bound optimisation procedure is used in which the solution is obtained from the chosen objective functions. Since both the equations describing the system and the objective function are linear functions of muscular forces and joint moments, use is made of linear optimisation procedures to predict muscle and joint forces. This approach was originally proposed by Seireg and Arvikar, (1975). The evaluated muscle forces and joint reactions for different static postures were calculated, and correlated well with EMG signals. Using the results from this study, Seireg and Arvikar, (1973) considered quasi-static walking. As before, muscle insertions and origins were estimated, and a simplified foot-ground reaction force was used in calculating the muscle and joint forces.

Crowninshield, (1978) used limits on muscle strength, equal to an

estimate of the maximum force capacity of the muscle, based on its physiologic cross-sectional area, as an objective function in his study on isometric and isokinetic elbow flexion. The resulting muscle force prediction was shown to correlate well with EMG data. As for previous methods, however, the method still has many limitations in that they all result in significant anatomical and functional simplifications. Joint centre location, muscle moment arm and physiologic cross-sectional area still have to be estimated. The effect of the antagonistic muscles is also, incorrectly, excluded.

Hardt, (1978), in determining the muscular forces in the leg during normal walking, first used an objective formula which minimized the muscular force. Although he obtained good results for some of the muscles when compared with EMG data an unrealistic muscular force was predicted for the tensor fascia lata muscle whilst no muscular force was predicted for others, despite electromyographic evidence of their activity. This was because the model ignored both the physiological and dynamic properties of the muscles. In an attempt to incorporate these properties, Hardt, (1978) defined six important properties of muscles to be incorporated into his model. To do this he defined an objective function which minimises the instantaneous energy requirement of the muscles. This involved representing muscle thermodynamics. As this was not available, he was forced to approximate muscle energetics while conforming to the linear programming format. This he did by making the following assumptions:

1. When actively contracting, the energy flux into a muscle is proportional to the mechanical power developed.
2. The energy requirement of the actively lengthening muscle

is non-zero and is a small constant value which is independent of force or velocity.

3. The internal work during isometric contraction provides a basis for the entire positive power mode.

The results were more favourable in that more muscles were included, but were still far from a true representation. The major flaw, however, is that a basically static optimisation is being performed on a dynamic system.

From the above survey of the use of linear optimisation techniques to predict muscular forces, it appears that linear optimisation was used more for reasons of mathematical simplicity rather than for physiological accuracy (Hardt, 1978; Crowninshield and Brand, 1981). In the light of this, a nonlinear optimisation method using combinations of unknown variables as the objective function, was used to solve for the muscular forces during gait. (Crowninshield and Brand, 1981). They used the hypothesis that muscular force prediction, such as to minimize the sum of muscular stresses to the third power, will result in predicted muscle action that maximizes activity endurance. Their results predicted reasonable muscular forces for normal gait which also correlated well with EMG patterns. The nonlinear optimisation procedure is however, extremely complicated mathematically, and attainment of a solution is not ensured.

An et al, (1984(a)) developed a new optimisation procedure based on the linear optimisation method. Additional inequality conditions were introduced, based on the stress of the muscle. A unique solution for the muscle forces was obtained by minimizing

the muscle stress. The objective function was the upper limit for muscle stress and a prediction of the muscle forces for the elbow joint was obtained by minimizing the muscle stress. The results correlated well with both EMG data and the work of Crowninshield, (1981).

In standing, one uses both agonistic and antagonistic muscles (Bajd et al, 1982). Lombard, (1903) put forward the hypothesis that those muscles which cross both hip and knee joints have better leverage as extensors than as flexors, and in the sit-to-stand movement, the simultaneous extension of both hip and knee joints is caused mainly by the hamstrings and rectus femoris muscles. Kelley et al, (1976), in using EMG to study the superficial muscle activity of the lower limbs during the standing-up procedure, observed this co-contraction of the rectus femoris and bicep femoris muscles. Lowery and Johnson, (1985) used a three-linked segment model, (tibia, femur, and head, arm and trunk (HAT)), to study the standing-up procedure in the sagittal plane.

Trnkoczy et al, (1976) developed a dynamic model of the ankle joint in which agonistic and antagonistic muscle groups, which perform the dorsiflexion and plantarflexion of the ankle, were under functional neuromuscular stimulation. The model described the behaviour of the joint in isometric and in free movement conditions.

Jaeger, (1986), developed two mathematical models to investigate quiet standing (standing without the use of a balancing aid), in spinal cord injured people. The first model, a three-linked segment model, was used to predict the torques generated at the

hip, knee and ankle joints to maintain static, upright standing. The second model was a single link-model with antero-posterior sway, which Jaeger identified as the greatest area of instability. The hip and knee joints were assumed to be held rigidly allowing motion to occur only about the ankle joint, where stability was maintained by the ankle dorsiflexors and plantarflexors.

2.3 Standing-up With Functional Neuromuscular Stimulation

The restoration of locomotion to people with spinal lesions is one of the main objectives of rehabilitation. For this purpose various aids have been designed. Unfortunately, all orthotic aids are basically passive appliances and generally difficult to utilize efficiently. As a result, most people discard them, preferring their wheelchair as their only means of locomotion.

The use of FNS as an electrical orthosis has opened up a new area of research. The main advantages of FNS are its easy implementation and the fact that it utilizes the person's own muscles. For the past 25 years, it has been used in the rehabilitation of persons suffering from partial or total paralysis of the upper and/or lower limbs. FNS was first applied by Liberson et al, (1961) to correct foot drop in hemiplegics through the stimulation of the peroneal nerve. Since then it has been used extensively for a wide variety of applications such as reversing muscle atrophy and increasing the range of motion of joints (Peckham et al, 1976). Because of the ability of a paralysed skeletal muscle to contract after either transcutaneous or intramuscular stimulation, the possibility exists of attaining standing-up, standing, walking and sitting-down in people with

spinal injuries. A number of laboratories have utilized this concept and have developed technology involving the use of electrical stimulation for experimental ambulation in spinal cord injured people.

Brindley et al, (1978) used FNS to achieve standing-up and walking in two cases. They implanted nerve stimulators into the femoral, inferior and superior gluteal nerves to stimulate the vasti muscles to achieve standing-up, excluding rectus femoris because of its hip flexor action. The knee joint was stabilised by stimulating the quadricep muscles continuously. This resulted in rapid fatiguing of the quadricep muscles. In an attempt to overcome this, they alternated the stimulation between the three superficial quadricep muscles. This proved to be unsuccessful as they found that fatigue was almost fully cumulative between successive cycles of stimulation and rest. They then put their patients on a training program in an attempt to increase the muscle fatigue resistance and found that, after a two month period, the fatigue resistance increased substantially.

The problem of rapid fatiguing of the muscles was studied by Kralj and Jaeger, (1982) and Kralj et al, (1986). They achieved prolonged standing in spinal cord injured people using FNS by means of posture switching. Once the correct erect posture was achieved, stimulation of the gastrocnemius and soleus muscles was increased. Simultaneously, stimulation of the quadriceps muscles was gradually decreased till it was discontinued. They found that this prolonged standing times two to five fold.

Kralj et al, (1980) used an orthotic device in conjunction with FNS to enable standing-up. They also noted that FNS offers

further advantages such as the prevention of muscle atrophy, the improvement of blood flow in the legs, prevention of muscle contractures and the reduction in spasticity.

Standing-up in a spinal cord injured (SCI) person by a two-channel electrical stimulator was analyzed by Bajd et al, (1982) and compared with standing-up in an able-bodied person. On stimulation of the quadriceps together with the assistance of the arms, standing-up from a wheel-chair was achieved. Once upright standing was achieved, the knee joints were locked through the electrical stimulation of both knee extensors. Using a force plate and stroboscopic photography, they calculated the maximum hip, knee and ankle joint torques produced during standing-up, without arm support, to be 85Nm, 80Nm and 10Nm respectively. With arm support, however, they mentioned that the knee joint torques required during standing-up is decreased by half of the calculated value, namely 40Nm.

Reciprocal walking of SCI people was generated by a four-channel stimulator (Bajd et al, 1983). In the double stance phase of gait, the knee extensor muscles of both knees were stimulated. In the single phase, the knee was kept extended only on the supporting side, while swing of the opposite leg was achieved by stimulating the afferent nerves cutaneously and thus triggering the flexor reflex mechanisms.

Chizeck et al, (1984) and Marsolais and Kobetic, (1987) in using an open-loop timing system together with implanted intramuscular electrodes under computer control and a standard walking frame, achieved standing-up and limited reciprocal walking. The electrodes were implanted, in the flexors, extensors, abductors

and adductors of the hip; the extensors of the knee; and the plantar and dorsi-flexors of the ankle by means of hyperdermic needles. A basic pattern of stimulation was adapted for each individual in which each step was initiated by depressing a hand-operated switch. The torques at the joints, as produced by electrical stimulation, were measured with a Cybex dynamometer. The walking pattern was evaluated using an instrumented walkway and qualitatively analyzed by means of video tapes. Dual force plates were used to measure the vertical forces on the feet and angular measurements were made with goniometers.

Petrofsky et al, (1985 and 1986) combined the two technologies - orthoses and FNS - to achieve standing-up and walking. By using braces, they were able to restrict the hip and knee joints to two-dimensional movement which increased stability and reduced the number of muscles required to achieve standing and walking. This achieved smooth standing and walking movements with low energy expenditure. The use of an orthosis however, removes the main advantage of FNS, namely ease of use.

The above experiments involved stimulation via surface electrodes (Kralj et al, 1980, 1982 and 1983; Bajd et al, 1982 and 1983; Petrofsky et al, 1985 and 1986), and implantable electrodes by either direct nerve stimulation (Brindley et al, 1978), or intramuscular stimulation (Braun et al, 1985; Marsolais and Kobetic, 1987). Each approach has its own advantages and disadvantages.

Surface electrodes have the advantage of easy application but the disadvantages of loosening, thus causing skin irritation and a bad cosmetic appearance. Implantable electrodes offer the

potential advantage of better control, no external wiring and ease of use when left inside the body but can be plagued with high infection and failure rates (Marsolais and Kobetic, 1987). Finally, if skin irritation or burns did occur with surface electrodes, no functional damage would be done to the muscle or nerve. With implantable electrodes however, permanent damage could occur to the nerve or muscle itself.

CHAPTER 3

RELEVANT ANATOMY AND PHYSIOLOGY OF MUSCLES AND NERVES

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To maintain equilibrium of the body, the torque produced by the weight of the head, arms and trunk and lower limbs about the hip, knee and ankle joint centres must be equalled by an opposing torque produced by tension in the muscles and ligaments surrounding the hip, knee and ankle joints. (Williams and Svensson, 1968).

The determination of which muscles are active in resisting the body torque produced whilst rising from a seated position, can be done either experimentally using electromyographic data or subjectively from an engineering aspect with regard to the necessary conditions for equilibrium of the joint. Both of these methods, however, require knowledge of the anatomy of the lower limb.

The description of the anatomy of the lower limb here is limited for the following reasons:

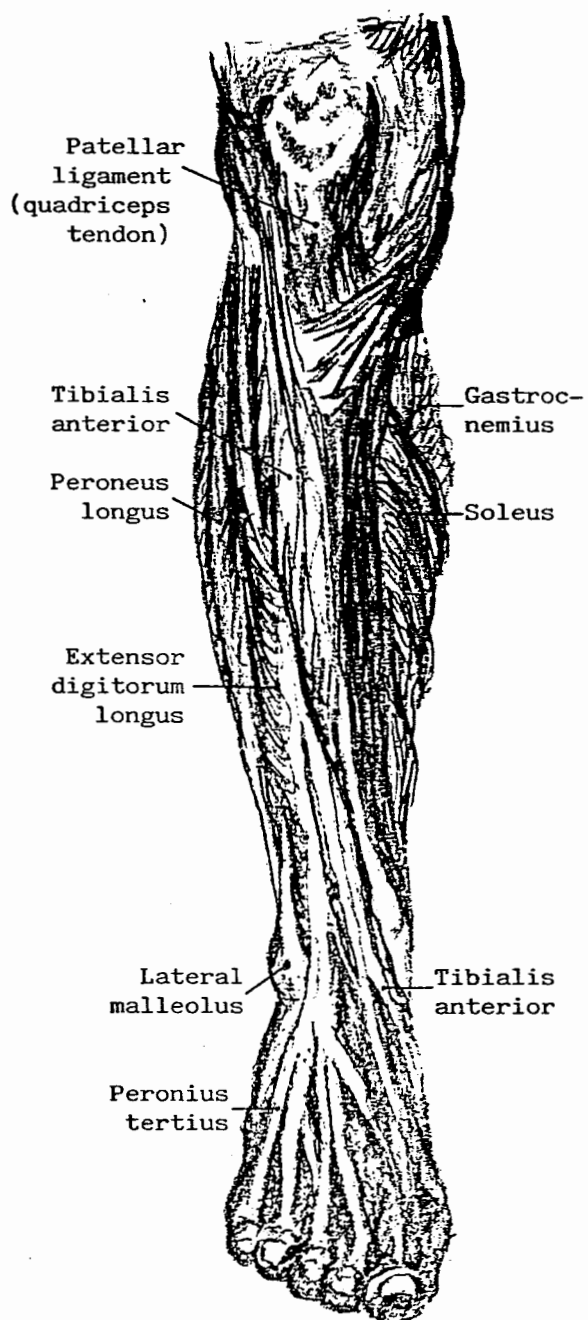
a) Since the present technology available at the University of Cape Town allows for surface electrical stimulation, only the superficial muscles will be accounted for.

b) Since standing up occurs mainly in the sagittal plane, only the muscles providing active resistance to the body in the sagittal plane will be accounted for.

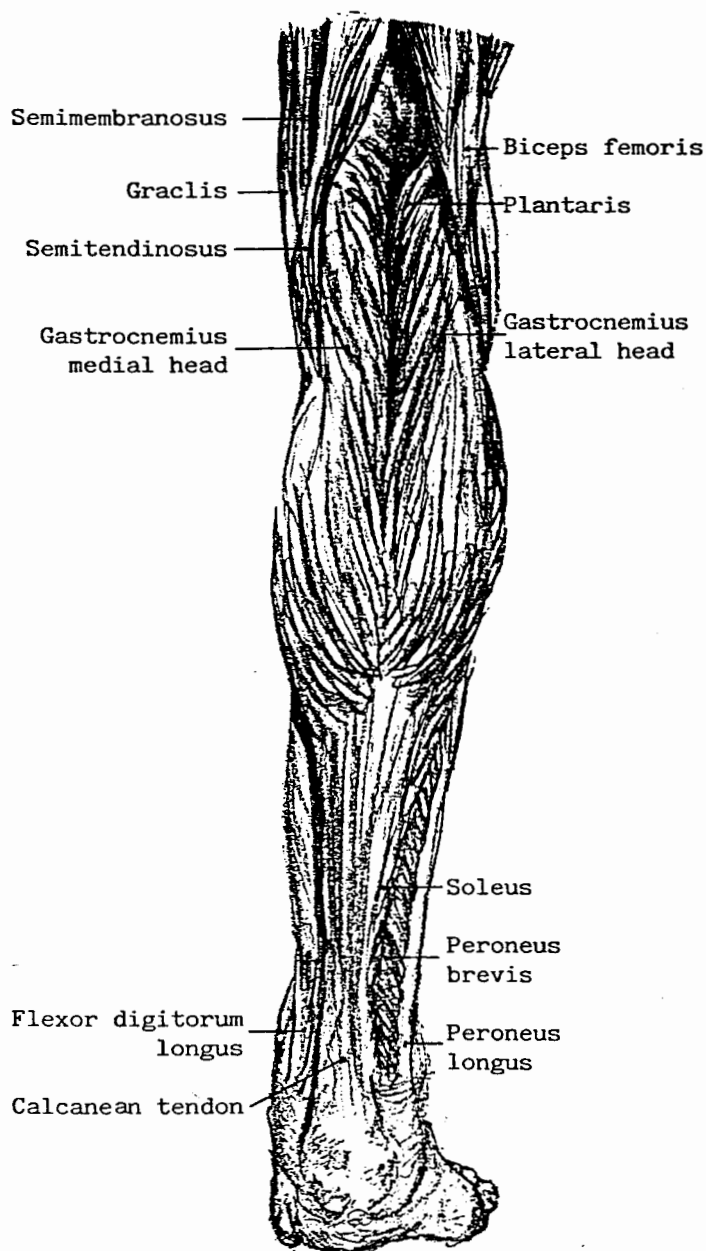
The muscles reviewed, together with their abbreviations and groupings are given in Table 3.1 (page 23), and shown in Figures 3.1 and 3.2 (pages 24 and 25).

ACTION	MUSCLES
Hip extensors	Gluteus maximus (GM) (with knee extended) Semimembranosus (SB) (with knee flexed) Semitendinosus (ST) Biceps femoris (long head) (BF)
Hip flexors	Rectus femoris (RF) Tensor fascia latae (TFL) Iliopsoas (IP)
Knee extensors	Rectus femoris (RF) Vastus medialis (VM) Vastus lateralis (VL) Vastus intermedius (VI) Iliotibial tract (IT)
Knee flexors	Semimembranosus (SM) Semitendnosus (ST) Biceps femoris (long head) (BF) Gastrocnemius (medial & lateral heads) (GN)
Ankle dorsi-flexors	Tibialis anterior (TA) Extensor hallucis longus (EHL)
Ankle plantar-flexors	Gastrocnemius (medial & lateral heads) (GN) Soleus (S)
Ankle evertors	Peroneus longus (PL) Peroneus brevis (PB) Peroneus tertius (PT)
Ankle Invertors	Tibialis anterior (TA)

Table 3.1
Lower limb muscles reviewed for model.



(a)



(b)

Figure 3.1
The superficial muscles of the right anterior thigh (a) and right gluteal posterior thigh (b). (Davies and Davies, 1962.)

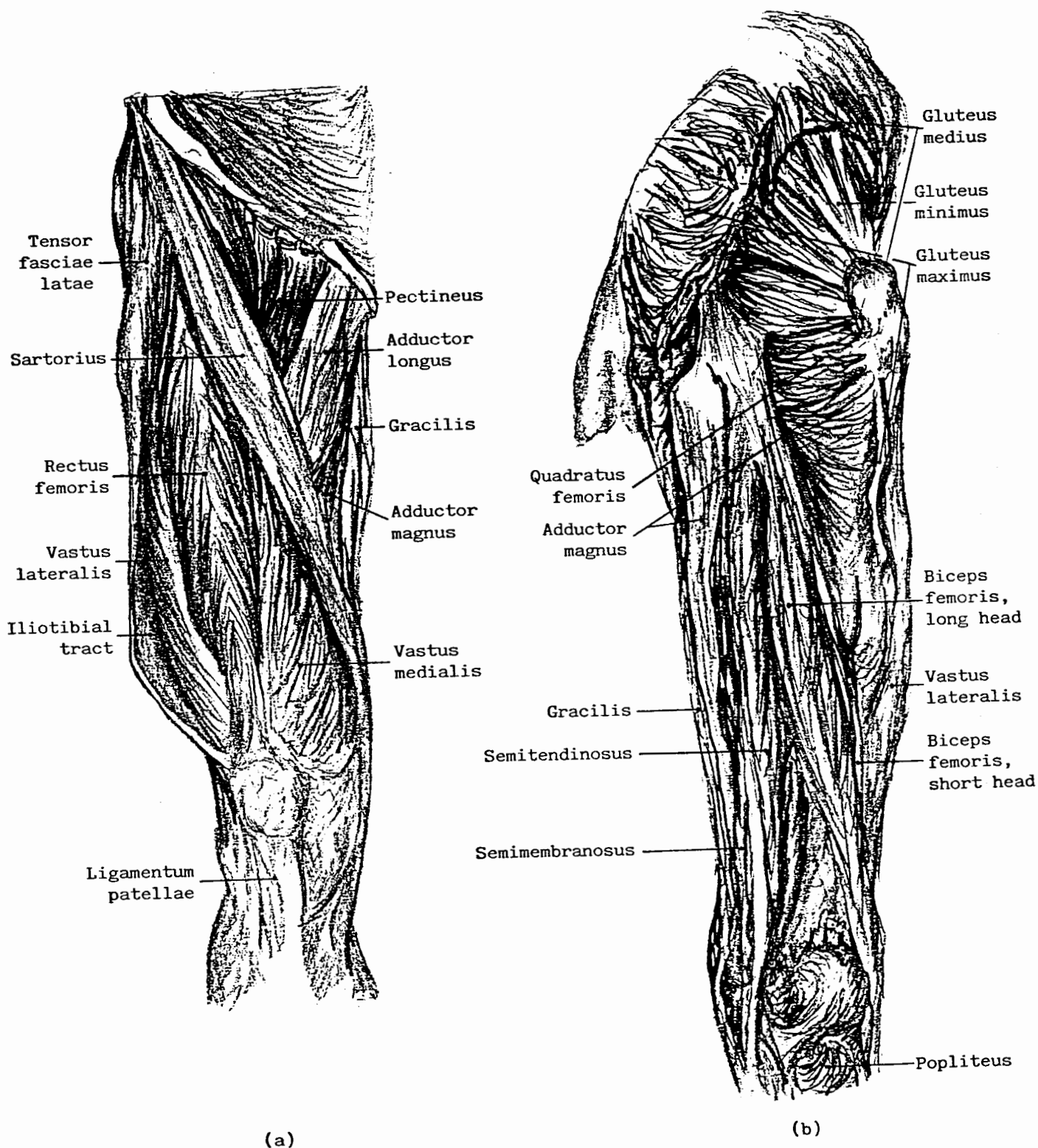


Figure 3.2
The superficial muscles of the right anterior leg (a)
and calf (b). (Davies and Davies, 1962.)

3.1 Muscle Description

Gluteus Maximus

Nerve Supply: Inferior Gluteal Nerve

- Action:
- i. With femur free, extends hip joint by pulling the femur backward.
 - ii. With the foot fixed, the body is raised from sitting to standing position; extends hip joint by moving the ilium backwards.
 - iii. By tightening the iliotibial tract, GM supports and steadies the knee joint in the extended position. In this action it is assisted by tensor fasciae latae, which corrects the tendency of GM to draw the iliotibial tract backwards.

Hamstrings (SB, ST, BF)

Nerve supply: Sciatic nerve

- Actions:
- i. Flexes the knee joint if leg is free to move.
 - ii. When knee is extended, as in standing, the hamstrings extend hip joint, as in raising the body from a stooping position.
 - iii. With knee semi-flexed: medial rotation of leg by the medial hamstrings.
lateral rotation of leg by biceps femoris.

Quadriceps Femoris (RF, VM, VL, VI)

Nerve supply: Femoral nerve

Actions: i. Extends knee joint when the tibia is mobile.

ii. Rectus Femoris flexes the hip joint.

iii. With the foot fixed, the vasti will bring the femur forwards and straighten the knee, as in rising from a sitting position.

Note that only the vasti brings the femur forward, and not the rectus femoris.

Gastrocnemius

Nerve supply: Tibial nerve

Actions: i. Plantar flexes the ankle joint.

ii. With toes on the ground, raises calcaneus off the ground: standing on the toes.

iii. Because of its origins on the femur, gastrocnemius is also able to flex the knee.

Soleus

Nerve supply: Tibial nerve - 2 branches

Actions: Plantar flexes the ankle joint.

Tibialis Anterior

Nerve supply: Deep Peroneal Nerve

Actions: i. When foot is free, dorsiflexes the ankle joint.

ii. Inverts the foot.

- iii. When foot is fixed, as in walking when the advancing foot has reached the ground, tibialis anterior brings the leg forward.

Peroneus Longus and Peroneus Brevis

Nerve Supply: Superficial peroneal branch of the lateral popliteal nerve.

- Action:
- i. Everts the foot.
 - ii. Plantar flexes the ankle joint (weak).

Peroneus Tertius

Nerve Supply : Deep Peroneal Nerve.

- Action :
- i. Everts the foot.
 - ii. Dorsiflexes the ankle joint.

3.2 The Nerve Cell and its Processes

The nervous system, together with the endocrine system, provides most of the control functions for the body. In general, the nervous system controls the rapid activities of the body such as muscular contractions, whereas the endocrine system regulates the metabolic functions of the body. The main functional unit of the nervous system is the nerve cell, consisting of a cell body and the fibrous processes termed the axon and dendrites, (Figure 3.3). Nerve fibre diameters range from 0.2 to 20 microns. Generally the greater the fibre diameter, the more rapid the conduction velocity. The range of conducting velocities is between 0.5 and 120 meters per second. (Guyton, 1981). There are

two main types of nerve fibres, viz. types A and C. Type A fibres are further divided into α , β , δ and γ fibres. They are motor and sensory in function. Type C fibres constitute more than half the sensory fibres in most peripheral nerves and also all of the post-ganglionic autonomic fibres (Guyton, 1981). The various characteristics of these fibre types are listed in Table 3.2.

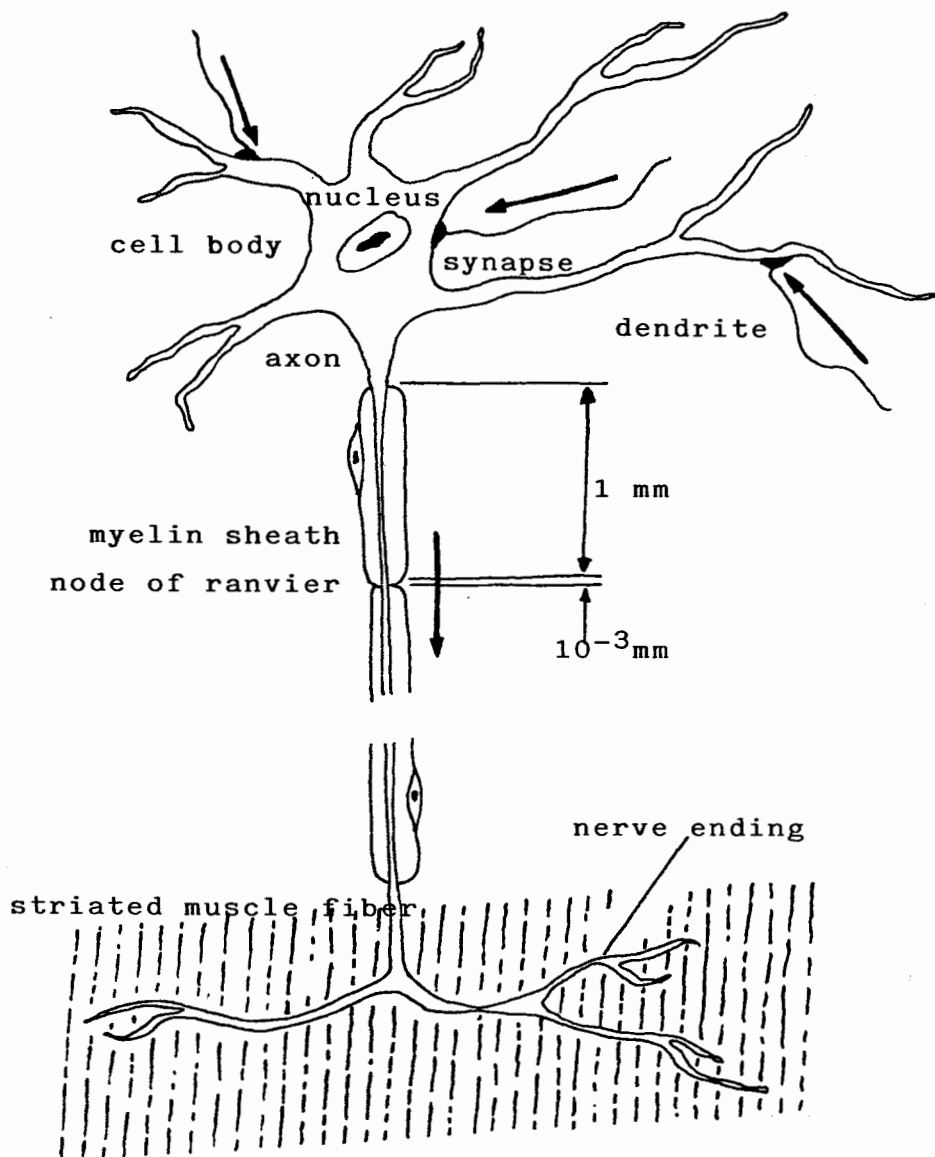


Figure 3.3

A motor neuron. (Guyton, 1981.)

Table 3.2 Mammalian Nerve Fiber Types and Characteristics

Fiber Type	Fiber Diameter (um)	Conduction Velocity (m/sec)	Spike Duration (msec)	Absolute Refractory Period (msec)	Peripheral Organ	Receptor Organ	Function
A fibers (motor)							
alpha	12-20	70-120	0.4-0.5	0.4-1	Muscle		Somatic motor
gamma	3-6	15-30					Motor to muscle spindle
A fibers (sensory)							
Group Ia	12-20	70-120	0.4-0.5	0.4-1	Muscle	Annulospiral Spindle endings	Proprioception
Group Ib						Tendon organs of Golgi	
Group II	5-12	30-70	0.4-0.5	0.4-1	1. Extensor muscles	Flower spray of spindle	Touch, pressure, vibratory receptors
					2. Flexor muscles	Flower spray of spindle	
					3. Skin	Touch-pressure receptors	
Group III	2-5	12-30	0.4-0.5	0.4-1	1. Muscle	Unknown - pain receptors(?)	Pain-(fast?), temperature (cold, heat)
					2. Skin	Pain	
C fibers							
Group IV	0.5-1	0.5-2	2	2	Muscle and skin	Pain	Pain-(slow), temperature, mechanoreceptors
B fibers	1-3	3-15	1.2	1.2			Preganglionic sympathetics

3.3 Electrophysiology of Nerves and Muscles

An electrical potential exists across the membranes of virtually all cells of the body. Nerve and muscle cells are excitable in that they are capable of generating electrochemical impulses at their membranes and can utilise these impulses to transmit signals along the membrane. This membrane potential is the result of an unequal active ionic distribution across the membrane. This imbalance of charges between the two sides of the membrane is created by: (1) active transport of sodium and potassium ions, by a sodium-potassium pump and (2) diffusion of ions through the membrane as a result of ion concentration differences between the two sides of the membrane. The cell membrane is thus polarized with the inside surface electronegative and the outside surface electropositive. A balance between the inward and outward flow of sodium and potassium ions is established when the membrane potential inside falls to about -90 millivolts. This -90 millivolts is the resting potential of the nerve membrane. At this resting membrane potential, there is a very high concentration of sodium ions on the outside of the cell surface and a very high concentration of potassium ions on the inside. Also inside the nerve fibre are large numbers of anions, which cannot diffuse through the nerve membrane.

Membrane potentials are used by nerve fibres to transmit nerve impulses, which is the means by which informational signals are transmitted from one part of the nervous system to another. This is achieved by means of action potentials, which are abrupt pulse-like changes in the membrane potential, lasting a few ten thousandths to a few thousands of a second (Guyton, 1981). The

action potential moves along the nerve, thus giving rise to a nerve signal.

The action potential results from rapid changes in membrane permeability to sodium and potassium ions, which occur in two separate stages called membrane depolarisation and membrane repolarisation. At the onset of the action potential, there is a sudden large increase to sodium permeability, resulting in a rush of sodium ions to the inside and causing the normal negative resting potential to change to a positive potential of about +40 millivolts. This is known as the reversal potential. This sudden loss of normal negative potential inside the fibre is called depolarisation. This is followed, almost instantaneously, by a return of the permeability of sodium ions to its original value and a considerable increase in permeability to potassium. The inflow of sodium ions thus stops, but there is a nett flow of potassium ions out of the fibre. Since potassium ions are positively charged, the positive charges inside the fibre are transferred back out of the fibre and the normal negative resting membrane potential returns. This is called membrane repolarisation. The imbalance of sodium and potassium ions is reversed by means of a sodium - potassium pump. The cell membrane is thus ready for the next stimulus. An action potential may be generated by many different kinds of stimuli (chemical, electrical, thermal, mechanical), but the sequence of excitation is always the same, viz, membrane depolarisation followed by a subsequent repolarisation. The stimulus in each case must be of adequate strength (intensity) and sufficient duration to initiate depolarisation. This threshold differs in nerves and muscles. A stimulus which does not result in an action potential is known as a subthreshold stimulus. This results in

local potentials which are important in that they can summate to result in an action potential (see Figure 3.4).

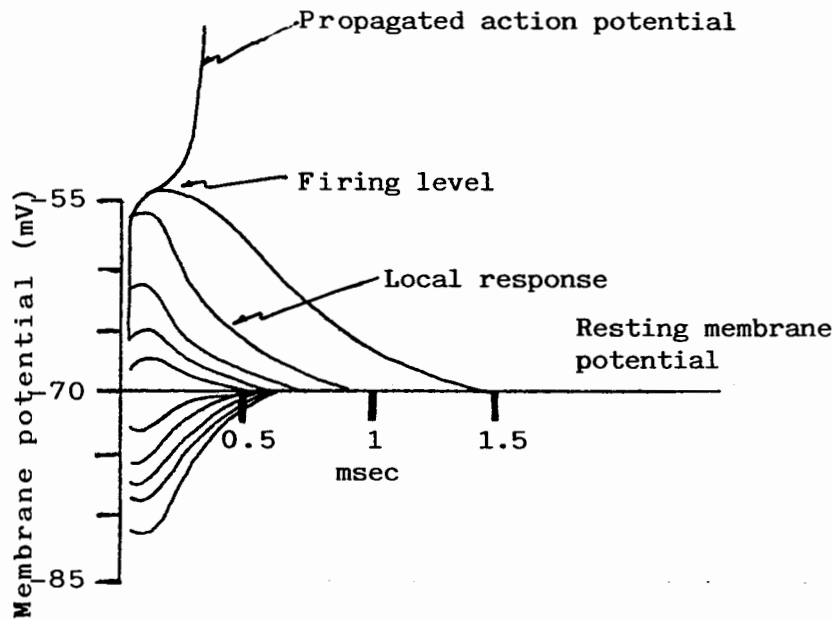


Figure 3.4
Electronic potentials and local response.
(Based on a diagram from Hodgkin: The subthreshold potentials in a crustacean nerve fiber. Proc Roy Soc London s.B 126: 87, 1938.)

The action potential also excites adjacent portions of the cell membrane, resulting in the depolarisation of these areas of the membrane as well. This depolarisation process travels in both directions along the entire extent of the fibre, thus propagating the action potential.

Once an action potential has been elicited, the depolarisation process will travel over the entire membrane. This is known as the "all-or-nothing" principle. Following the action potential is an absolute refractory period, during which the nerve fibre cannot be excited. Consequently, nerve fibres cannot remain continuously excited, but rather they respond to the stimulation with discrete, equal-amplitude impulses. (The stimulation is graded in terms of its "firing-rate").

3.4 Excitation and contraction of skeletal muscle

Muscle fibres, like nerve fibres, can be excited to produce an action potential. This, in turn, activates a contractile mechanism, provided that the threshold of excitation for the muscle tissue is reached. Skeletal muscle is made up of muscle fibres which in turn contain many bundles of myofibrils. Each myofibril consists of many myofilaments. Myofilaments are composed of thick myosin and thin actin filaments which are large polymerized protein molecules that are responsible for muscle contraction.

3.5 The Motor Unit

The unit of contraction in the muscle is not an individual muscle fibre but a motor unit. A motor unit normally comprises a single motorneuron and all the muscle fibres it supplies. The motor neuron has its cell body in the anterior horn of the spinal cord and an axon leading to the muscle. The axon may have many branches. Each branch of the axon joins a muscle fibre at approximately its mid-point, at a structure called a neuromuscular junction. The muscle fibres of a motor unit are not grouped together but are spread out over an area in the muscle and thus the muscle fibres of several motor units intermingle. The size of a single motor unit is related to the function of the innervated muscle. In general, small muscles that react rapidly and are responsible for fine movements have very few muscle fibres in each motor unit. Large muscles which do not require such fine degree of control have several hundred muscle fibres in a single motor unit.

Under normal conditions the muscle fibres of a motor unit are activated by a nerve impulse travelling down a motoneuron which branches to supply all the muscle fibres. When it arrives at the neuromuscular junctions the nerve impulse causes the release of a chemical transmitter substance, acetylcholine, into the synaptic cleft.

This increases the muscle membrane permeability to sodium ions resulting in a rush of sodium ions into the muscle fibre which, in turn, changes the membrane potential thus creating a local potential called the end-plate potential. This end-plate potential initiates the action potential on the muscle fibre membrane. The previous discussion regarding initiation and conduction of action potentials in nerve fibres applies equally well to skeletal muscle fibres, except for quantitative differences (Guyton, 1981). The action potentials elicit electrical currents that spread through the fibre causing the release of large quantities of calcium ions into the muscle fibre sarcoplasm surrounding the myofibrils. These calcium ions activate the attractive forces between the actin and myosin filaments resulting in muscle contraction. Since the neuromuscular junction is located near the middle of the muscle fibre, the action potential will spread from the middle towards the ends, ensuring coincident contraction. The summation of muscle contraction, which is the adding together of individual muscle twitches to make a strong muscle contraction, occurs in two different ways: (1) by increasing the number of motor units contracting simultaneously and (2) by increasing the rate of contraction (firing rate) of individual motor units.

3.5.1 Spatial summation (recruitment)

Within a single muscle there are different sized motor units. The order of recruitment is according to size, with the smaller motor units recruited first, and increasingly larger motor units recruited as the force output increases. This effect causes the gradations of muscle strength during weak muscle contraction to occur in very small steps, which become progressively greater as the intensity of contraction increases because the larger motor units then begin to contract.

3.5.2 Temporal summation (firing rate)

In a normal muscle, the motor unit does not discharge at constant intervals. Therefore, discharges of a motor unit are measured in terms of the average firing rate. As the frequency of muscle twitches increases, the muscle twitches will start to overlap, thus progressively increasing the degree of muscle shortening. At more rapid rates of contraction, the degree of summation of successive contractions becomes greater and greater because the overlapping is increasing. As the muscle is stimulated at progressively greater frequencies, a frequency is reached at which the successive contractions fuse together and cannot be distinguished from one another. This state is called tetanisation and the lowest frequency at which it occurs is called the critical frequency. Figure 3.5 illustrates the principles of temporal summation and tetanisation. As can be seen from Figure 3.5, once the critical frequency for tetanisation has been reached, further increase in the firing rate (rate of stimulation) results in a minimal increase in the force of contraction.

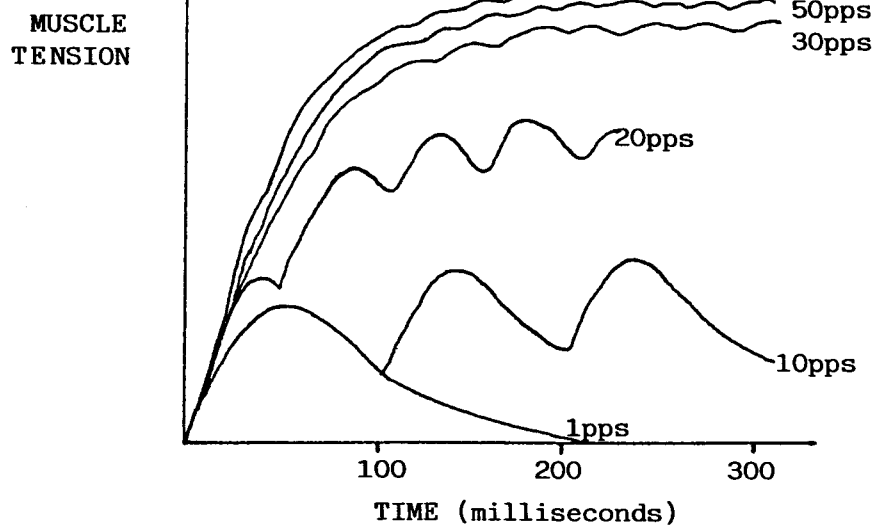


Figure 3.5

Wave summation and tetanisation. (Benton et al, 1981)

Milner-Brown et al, (1975) found that most recruitment of motor units occurred when low-force levels were required. As the required force level was increased, fewer (but larger) motor units were recruited. In most normal muscles, motor unit recruitment occurs up to approximately 75 percent of maximum voluntary contraction.

Although there appears to be generally orderly progression of motor unit recruitment as the force output increases, the neuromuscular control scheme is not absolutely rigid (De Luca, 1979). The order of recruitment for a muscle is consistent while the muscle is carrying out its normal function, but is not consistent when the function changes (Person, 1974).

3.5.3 Muscle fatigue and synchronization

Prolonged and strong contraction of a muscle leads to the well-

known state of muscle fatigue. This results from inability of the contractile and metabolic processes of the muscle fibres to continue supplying the same work output. Synchronization, the tendency for a motor unit to fire regularly at or near the time that other motor units discharge, describes their behaviour with respect to each other. The relationship between synchronization and contraction force and time is not well understood. There is some evidence however that motor units generally synchronise with a corresponding drop in frequency during fatigue, physiological tremors, after exercise and in some disease states (De Luca, 1979).

3.5.4 Summary

Although spatial and temporal summation have been discussed separately, they very rarely occur separately in normal muscle function. Another important feature is that the different motor units fire asynchronously (one is contracting whilst the other is relaxing), which ensures that the tension exerted by the whole muscle is still continuous and non-jerky, even when tetanisation of individual motor units is not occurring.

At the beginning of an isometric force varying contraction, the recruitment of motor units is the dominant factor with the smallest motor units being recruited first with a minimum firing rate of 6 pulses/second. This recruitment remains the dominant factor up to approximately 30 percent of maximum voluntary contraction with progressively larger motor units being recruited with resultant increase in force. As a secondary factor, the firing rate also increases. The firing rate replaces recruitment as the dominant factor for force levels ranging from 30 to 75

percent maximum voluntary contraction with the recruitment progressively playing the secondary role. At force levels above 75 percent maximum voluntary contraction, recruitment tends to cease and the further increase in the firing rate is the dominant factor. In addition to this, the rate of increase of a muscle contraction force appeared to have a significant effect on the correlation between recruitment and firing rate.

Synchronization of motor units is observed when a muscle fatigues, during both constant force and force varying isometric contractions.

3.6 Recording the Motor Unit Action Potential

The recording of nerve impulses can be done by placing two recording electrodes on the outside of a nerve fibre. The record that is obtained is biphasic due to the nature of the generation of an action potential and direction of the impulse. Figure 3.6 illustrates such a reading showing a potential change first in one direction and then in the opposite direction. The depolarisation of the muscle membrane results in local electrical currents flowing between the depolarized and unexcited regions of the muscle fibre. These local currents generate a potential difference in the extracellular medium in the vicinity of the depolarised membranes. Two recording electrodes located in this field will detect the potential difference and result in a signal. This monitored action potential which is known as the electromyographic signal, is generally, although incorrectly, referred to as the "recorded action potential".

The recorded action potential of a muscle fibre, like that of a nerve fibre, is biphasic. The recording electrodes are bipolar and the signal is amplified differentially. The sign of the phases depends on the direction from which the muscle membrane depolarisation approaches the electrode. In Figure 3.6 the recording electrode contacts are aligned parallel to the muscle fibres. In the diagram, a depolarisation approaching from the left side causes electrode A to become positive with respect to electrode B, which is reflected as a negative phase in the action potential, and vice versa. The shapes and, therefore, the frequency spectrum of the recorded action potentials depend on the orientation of the electrodes with respect to the active fibres, and on the tissue between the muscle fibre and the electrodes.

The tissue creates a low-pass filtering effect, where the bandwidth decreases as the distance increases. This is more pronounced for surface electrodes than indwelling electrodes because of the ability to locate the latter closer to the muscle fibres (De Luca, 1979).

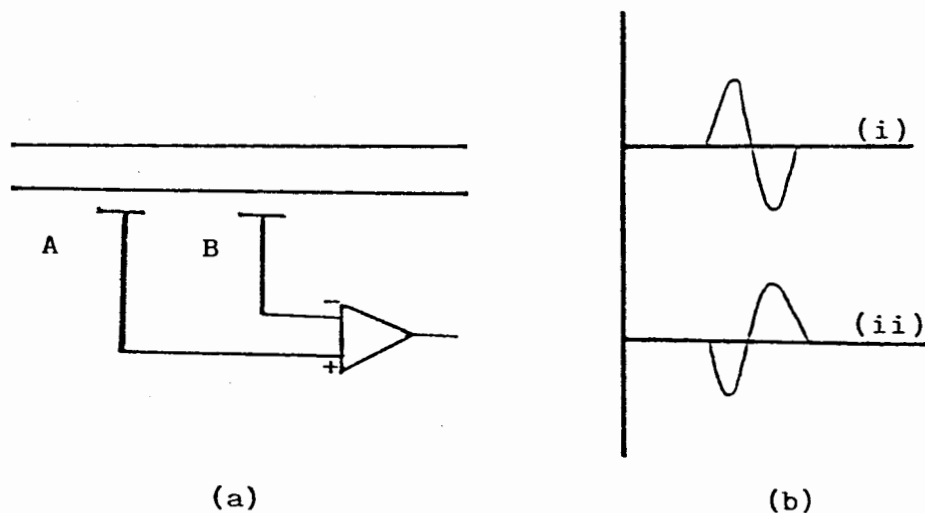


Figure 3.6

Recording of a biphasic action potential. (De Lucia, 1979.)

The amplitude of the action potential depends on the diameter of the muscle fibre, the distance between the active muscle fibre and the recording site and the filtering properties of the electrodes. These, in turn, are a function of the size of and the distance between the recording contacts, and also the chemical properties of the metal electrolyte interface (De Luca, 1979). Finally, the duration of the waveform depends on the conduction velocity of the action potential along the length of the fibre.

Since the depolarisations of the muscle fibre of one motor unit overlap in time, the resultant signal constitutes a spatial-temporal superposition of the contributions of the individual action potentials. The contributions of the action potentials of each muscle fibre are a function of the square of the distance between the muscle fibre and the recording electrode. The further away the muscle fibre is from the recording electrode, the less the contribution to the composite signal. The resultant signal is called the motor unit action potential and is designated as $h(t)$ in Figure 3.7. The shape and amplitude of the motor unit action potential are dependent on the geometric relationship between the electrode and the active muscle fibres, as well as the above mentioned factors affecting the action potential (De Luca, 1979). Figure 3.8 shows a typical motor unit action potential, the time duration of which may vary from 5 to 20 milliseconds. As can be seen the shape is no longer biphasic. This is because all the muscle fibres of a motor unit do not depolarize simultaneously. Thus the amplitude and shape of a motor unit action potential are a function of the properties of the motor unit, muscle tissue and recording electrodes.

A motor unit action potential will result in a twitch of the muscle fibres. In order to sustain a muscle contraction, the motor units must be repeatedly activated. The resulting sequence of motor unit action potentials is called a motor unit action potential train. The shape of the motor unit action potentials within a motor unit action potential train will remain constant if the geometric relationship between the electrode and the active muscle fibres remains constant and if the properties of the electrode do not change in the muscle tissue (De Luca, 1979).

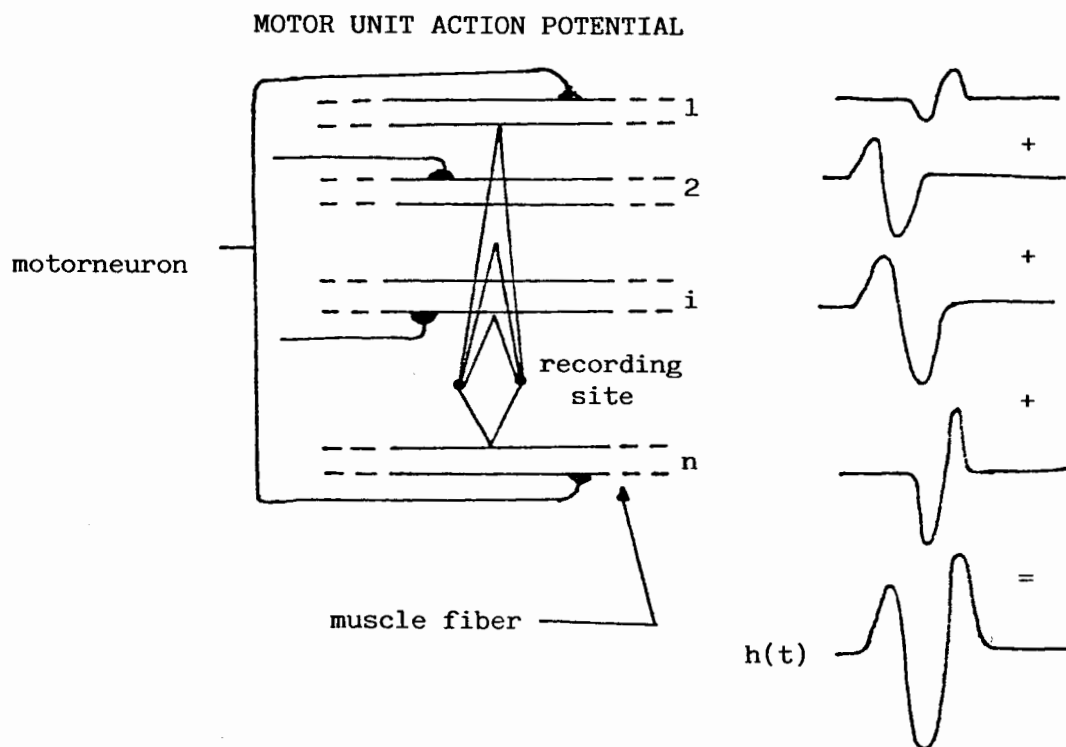


Figure 3.7
Schematic representation of the generation of the motor unit action potential, $h(t)$. (De Lucia, 1979.)

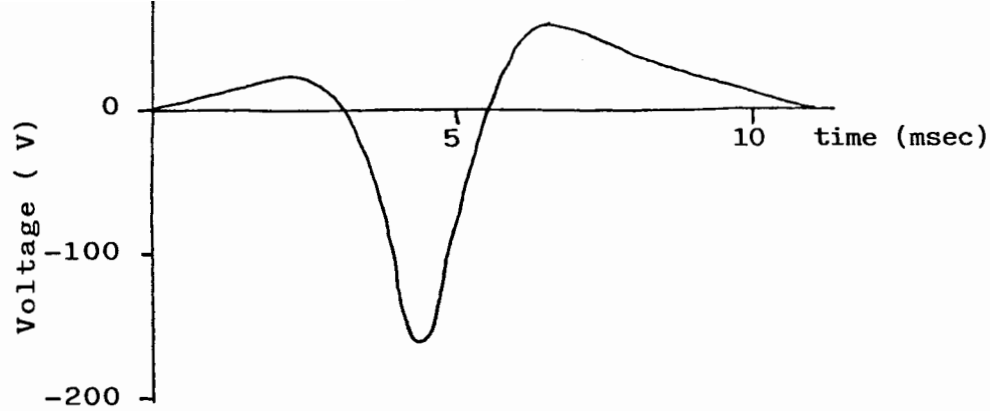


Figure 3.8

A typical motor-unit action potential. (De Lucia, 1979.)

To observe a single motor unit action potential is only possible during a very weak muscle contraction when the fibres of only one motor unit are active. As the contraction muscle force increases, more motor units are activated resulting in several motor unit action potential trains, thus making it difficult to identify all the motor unit action potentials of any particular motor action potential unit train (De Luca, 1979).

3.7 The Electromyogram

The propagation of an action potential along a muscle fibre results in local electrical currents which spread away from the muscle to as far as the skin. If many muscle fibres contract simultaneously, the summated electrical potentials at the skin can be very great. By placing two recording electrodes on the skin surface or inserting them into the muscle, an electrical recording, called the electromyogram, can be made when the muscle is stimulated.

The electromyogram (EMG) is a graphical indication of muscle activity which generally increases in magnitude with muscle tension. The many variables which can influence the signal include: fatigue, the type of recording electrode, reflex activity and velocity of muscle contraction, and abnormalities of muscle excitation (Soderberg and Cook, 1984; Crowninshield and Brand, 1981). Care must therefore be exercised when drawing conclusions about the strength of muscle contractions.

3.7.1 Electrodes

There are two groups of electrodes: surface or indwelling. Surface electrodes consist of discs of metal, usually silver/silver chloride, of about 1cm diameter. These electrodes detect the average activity of superficial muscles and give more reproducible results than do indwelling types. Bipolar electrodes are preferred to monopolar electrodes as they are ideal for single unit recording in large muscles, because they pick up the potentials from muscle fibres close to the electrode, and attenuate the potentials from distant fibres (Komi and Buskirk, 1973).

There are two types of indwelling electrodes: needle and fine wire. Indwelling electrodes are used for the assessment of fine movements recorded from deep muscles.

3.7.2 The surface electromyogram

The electromyographic signal measured differentially by a pair of electrodes is the result of the summation of the spatially and temporally dispersed trains of motor unit action potentials. A

schematic representation of this is shown in Figure 3.9. The superposition of the motor unit action potential trains at the recording site forms the physiological myoelectric signal, m_p , which is not observable. When the signal is detected, an electrical noise, n , is introduced. The detected signal will also be affected by the filtering properties of the electrode and possibly also by the recording equipment. The resulting signal is the observable electromyographic signal m . The positioning of the electrodes will determine the shapes of the motor unit action potentials, $h(t)$, as previously described (De Luca, 1979).

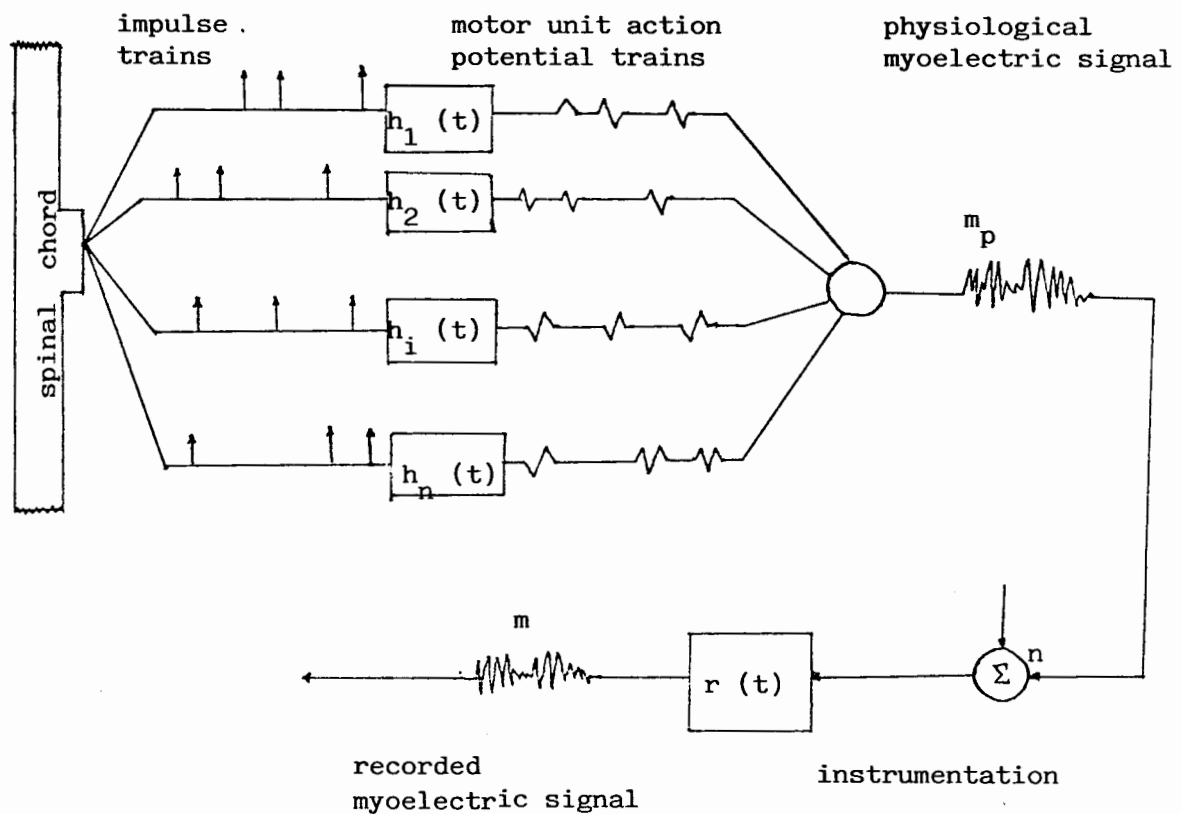


Figure 3.9
Schematic representation of the generation of the myoelectric signal as recorded by surface electrodes. (De Lucia, 1979.)

3.7.3 Frequency response

It is important to know the bandwidth of the EMG signal so that the frequency response of the amplifying, filtering and processing

systems can be optimised. The spectrum of the electromyographic signals has a frequency range from 5Hz to 2kHz. For surface electrodes however, significant information will be contained in frequencies below 1000Hz. Figure 3.10 shows a typical EMG spectrum from which it can be seen that most of the signal is concentrated in the 20 to 300 Hz range. The peak frequency, for surface electrodes, is between 50 and 60Hz (Kwatny et al, 1970).

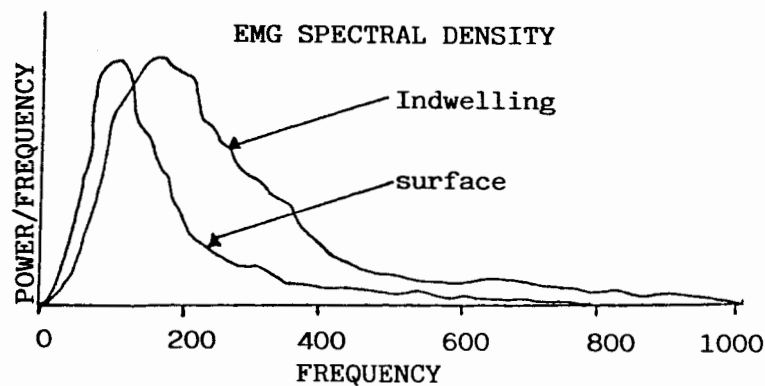


Figure 3.10
Frequency spectrum of EMG as recorded via surface and indwelling electrodes. (Kwatny et al, 1970.)

For frequencies below 20Hz one gets noise from sources such as movement artefacts (induced potential because of movement between the skin and electrode), as well as polarization potentials. A major source of interference comes from power line hum which is 50Hz. The hum lies near the middle of the electromyogram spectrum, unfortunately, so nothing can be done to filter it out. The shape of the spectrum of the EMG signal is dependent on the state of the muscle. In work done to see how well electromyographs can predict tension during muscle shortening and lengthening, Komi, (1973) found that the EMG amplitude remains fairly constant in spite of decreased tension during shortening

and increased tension during eccentric contractions.

However, a considerable change in the spectrum occurs when the muscle is fatigued. Muscle fatigue occurs when the muscle tissue cannot supply the metabolism at the contractile element (Guyton 1981). This change is noted by a shift of the power in the spectrum to the lower frequencies (Kwatny et al, 1970). This effect can be due to three possible causes:

- (i) lower conduction velocity of some or all action potentials (Mortimer et al, 1970).
- (ii) the slower motor units with their larger motor unit action potentials remain active while some of the faster motor units with shorter motor unit action potentials drop out.
- (iii) the tendency for the motor units to fire more synchronously (De Luca, 1979).

These effects all lead to an increase in the amplitude of the EMG during fatigue.

If fatigue does not occur, and the muscle force is increased, the number of muscle fibres recruited increases and so, to a certain degree, does the firing rate. One would expect an upward shift in the frequency spectrum of the EMG signal. This, however, is not the case. Hogan and Mann, (1980) showed that the predominant effect of a change in force is to change the amplitude of the spectrum and not its shape, and that there is a greater difference in the spectra obtained from different locations at the same contraction force than from the same location at different contraction forces.

SECTION II

INITIAL MODEL DEVELOPMENT AND EXPERIMENTAL PROCEDURES

CHAPTER 4

PRELIMINARY STUDIES

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4.1 Introduction to the Mathematical Model

The human body remains in equilibrium in its gravity field if the vertical line drawn through its centre of gravity falls within the boundaries of its base of support, namely the feet. The centre of pressure or ground reaction force (GRF) is the upward force acting underneath each foot. Provided that no horizontal acceleration occurs, GRF acts vertically upwards, through the centre of gravity (COG). If the COG passes outside the base of support, the body loses its state of equilibrium as a toppling moment has been introduced. The resulting loss of balance is regained either by a rapid righting reaction, or by moving the base of support by stepping forwards or backwards.

The standing-up procedure has been suggested to be a quasi-static one in which the acceleration of the various body segments is negligible (Bajd et al, 1982). This has been verified experimentally by measuring the GRFs using a force plate for the standing-up and sitting-down procedures (Section 4.8.3, page 71).

Kinetics is the general term given to the forces, both internal and external, that cause movement. The internal forces arise from muscle and ligament activities, whereas the external forces come from the ground (ground reaction forces) or from external bodies such as when a rugby player is being tackled by another player. Kinematics refers to the measurement of the variables involved in execution of a movement, and is independent of the forces that cause the movement. Such variables would include linear and angular displacements, velocities and accelerations. The identification of the magnitude and direction of the separate muscle forces is, however, an extremely difficult

task as explained in the foregoing literature survey. The development of a generalized model to determine these muscle forces would be subject to inaccuracies because of individual differences in anatomical structure and of the indeterminacy associated with the resulting model.

To enable one to rise from the sitting to the standing position or visa versa, a torque has to be generated at the hip, knee and ankle joints. This torque is brought about by the action of various postural muscles acting across the joints as shown in Figures 4.1 and 4.2. This torque will be dependent on the weight of the various parts of the body as well as the distance of their COGs from the hip, knee and ankle joints.

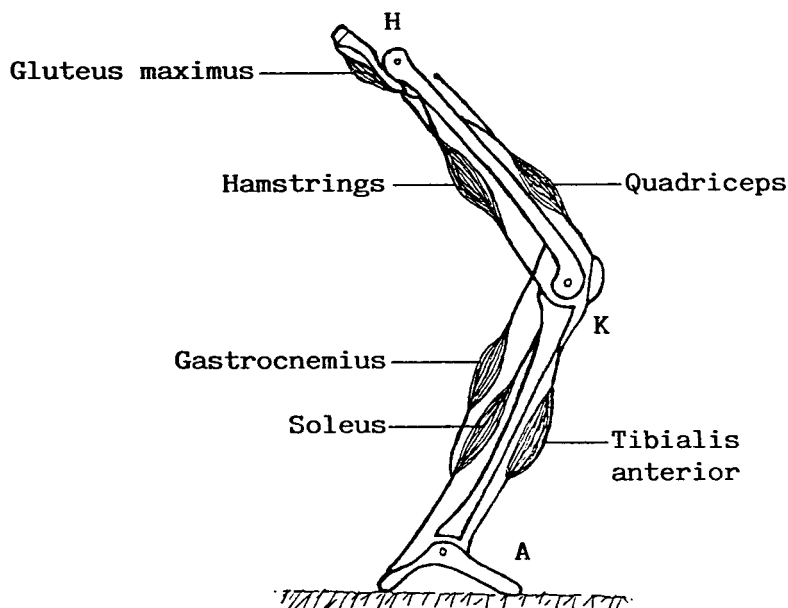


Figure 4.1
Sketch of the muscles of the lower limb.

At present, standing-up is generally achieved in paraplegics by electrical stimulation of the quadriceps muscles only (Bajd et al, 1982). There is a certain amount of uncertainty as to the amount of knee-joint torque required to achieve standing. From

the literature the maximum value ranges from 30 to 200 Nm (Bajd et al,1982); Hart et al,1984). There exists therefore the need for a mathematical model which will predict the torques generated at the joints during the standing-up procedure which is "tailor-made" for the person concerned. With this data, coupled with the anatomical knowledge of the lower limb, it is possible to determine the electrical stimulation parameters and sequence required to achieve standing-up in paraplegics.

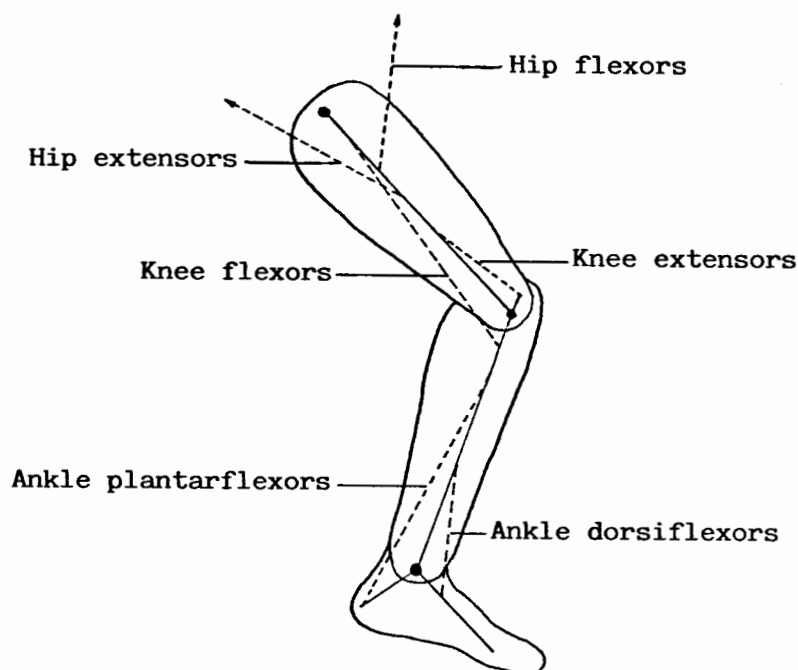


Figure 4.2
Idealised resultant muscle forces of the lower limb.

4.2 Development of the Mathematical Model

The human body was viewed as a symmetrical linked-segment model, with four moving segments; head-arms-trunk (HAT), left and right thigh and calf and foot. The anthropometrical parameters used to calculate the weights of the various body segments were taken from the work of Chandler et al, (1975). The weight and centre of gravity for each segment were computed using the Chandler-Vaughan regression equations (Vaughan, 1980). The four-linked-segment

model is given in Figure 4.3 where the hip, knee and ankle angles are represented as ϕ , β and α respectively. The moving segments were modelled in the sagittal plane and bilateral symmetry was assumed. This implied that the upper torso weight was equally borne by the lower limbs and thus the torques for each limb were calculated using the weights of half HAT, one thigh, calf and foot.

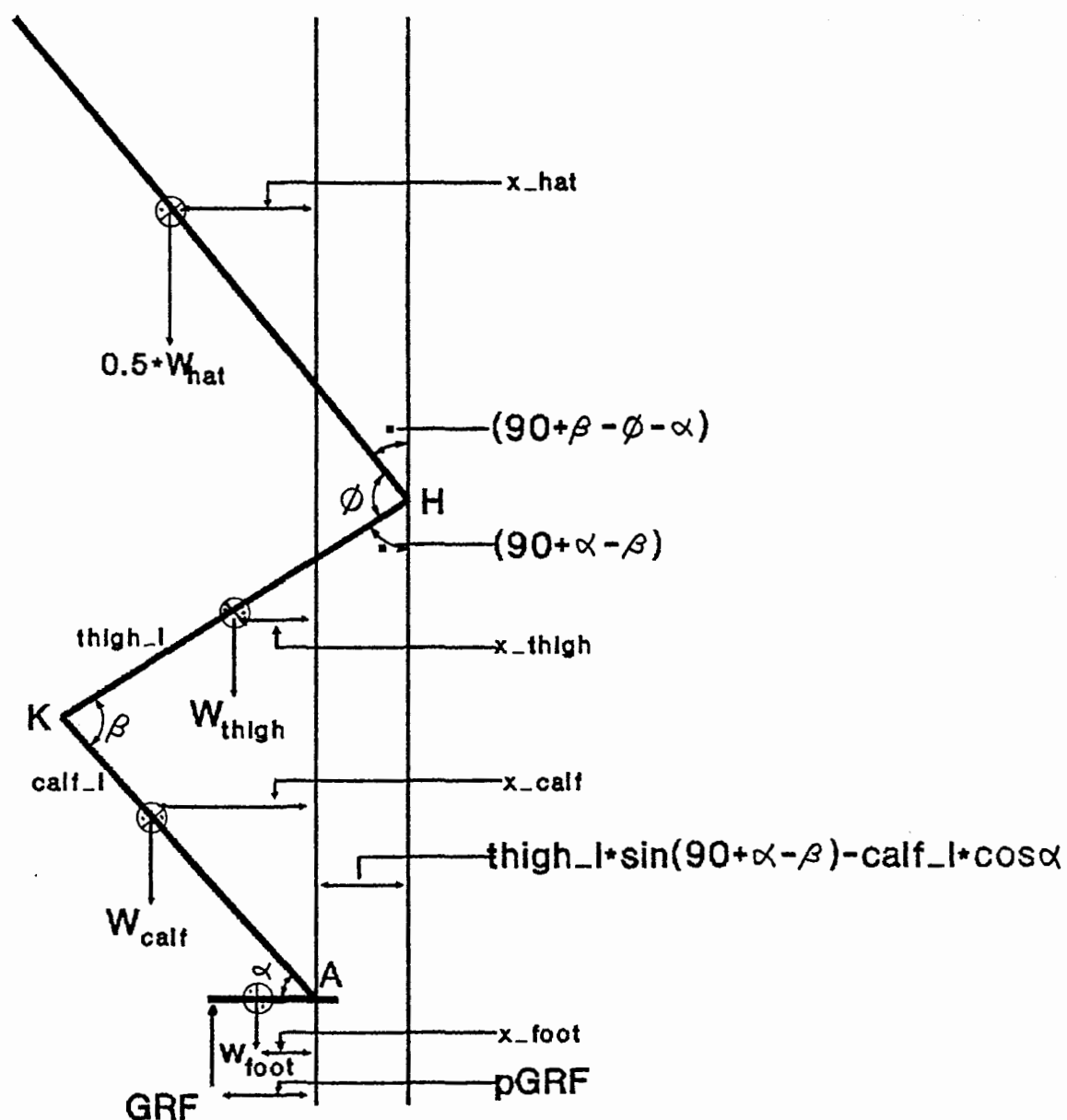


Figure 4.3
Free body diagram of the body.

The model was developed from two aspects:

(i) Kinematically, in which angles α , β and ϕ for the standing-up procedure were obtained experimentally using a digital camera system, the micron eye. These angles, together with the subject's anthropometrical data, were used to calculate the net torques generated at the hip, knee and ankle joints.

(ii) Kinetically, in which the positions of the ground reaction force, (pGRF), with respect to the lateral malleolus, and the corresponding ankle angles (α), were required. This data, together with the subject's anthropometrical data and a mathematical relationship between β and ϕ , was used to calculate the net torques generated at the hip, knee and ankle joints.

4.2.1 Assumptions

The main considerations for developing the model are summarised as follows:

1. The model does not take into account the neurophysiological or metabolic muscle conditions.
2. The muscles are assumed to produce tensile forces only.
3. The dependence of muscle force upon muscle length is neglected.
4. The model is derived with no upper limb involvement.
5. The standing-up procedure is assumed to be a quasi-static one for which the inertia forces and moments are assumed to be relatively insignificant compared to the gravity forces (Bajd et al, 1982).
6. The sitting-down procedure is the reverse to the standing-up procedure (Petrofsky et al, 1985).

4.3 Kinematic Analysis

The assumption that the standing-up procedure is a quasi-static movement implies that for every position assumed by the body, it is in a state of equilibrium. Thus, the internal joint moments required to stabilize the hip, knee and ankle joints are calculated by projecting the line of force through the various segment centres of gravity (centre of mass), vertically downwards, and multiplying this force vector by the horizontal distance to the centre of joint rotation.

4.3.1 Force and Muscle Torque Equations

Each segment, of the four-linked-segment system, shown in Figure 4.3, was considered individually in a sequential analysis, which began with the foot, and then progressed to the calf (lower leg), thigh and upper body (HAT), shown in Figure 4.4. The free body diagram of the segments, Figure 4.3, indicate their segmental weights directed vertically down from their respective mass centres. The resultant muscle forces crossing each joint is the vector sum of all the individual muscle forces, both agonists and antagonists, which act across a particular joint. These resultant muscle forces will cause a torque about their respective joints. They can be replaced by a moment acting about the joints (couple) and an equivalent force acting at the joint. Finally, the equivalent muscle force components are combined to form the resultant X and Y force components at the joint.

For uniformity in the free body diagrams and facilitation of computer modelling, the following sign notations were adopted:

1. Lines of action of forces directed upwards or to the right: Positive.

Lines of action of forces directed downwards or to the left: Negative.

2. Anti clockwise moments : Positive.

Clockwise moments : Negative.

3. The resultant muscular torque and resultant force components at the proximal end of a segment were regarded as be negative and those at the distal end as positive in the free body diagram.

4. Angles alpha, beta and phi were expressed as the angle subtended by two limb segments.

5. The symbols used were :

$W(i)$: segmental weight.

$T(i)$: joint torque generated by the resultant muscle torque.

$F_x(i), F_y(i)$: x and y components of the combined resultant muscle and joint reaction forces.

With reference to Figure 4.4, by summing the forces at the proximal and distal ends of each segment, in both the horizontal and vertical directions the following force equations resulted (see Appendix A):

$$F_{KY} = -GRF + W_f + W_c \quad (1)$$

$$F_{HY} = -GRF + W_f + W_c + W_t \quad (2)$$

$$GRF = W_{HAT} + W_f + W_c + W_t \quad (3)$$

Also, by summing the moments acting about a joint for each of the segments, and equating them to zero for equilibrium, the following

moment equations were obtained (see Appendix A).

$$T_A = -GRF * pGRF + W_f * x_{foot} \quad (4)$$

$$T_K = T_A - (F_{ay} * calf_l * \cos\alpha) - (W_c * 0.42 * calf_l * \cos\alpha) \quad (5)$$

$$T_H = T_K + (F_{ky} * thigh_l * \sin(90 + \alpha - \beta)) + W_t * 0.39 * thigh_l * \sin(90 + \alpha - \beta) \quad (6)$$

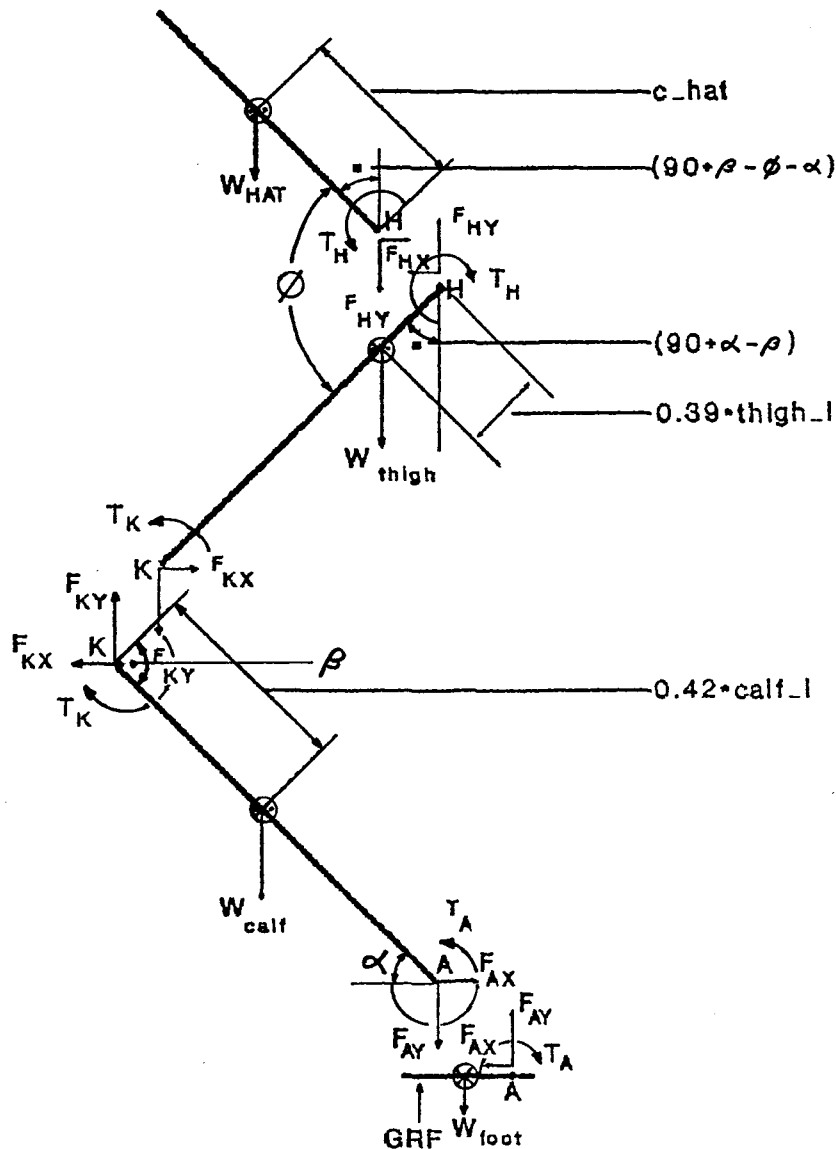


Figure 4.4
Combination of joint reaction and muscle force components
into resultant components at the joint.

4.3.2 Calculation of pGRF

The position of the ground reaction force (pGRF), was mathematically obtained by taking moments for all the segments about a fixed point and equating them to zero for equilibrium. Moments were taken for the body, represented as a stick figure in Figure 4.3, about the Y (vertical) axis with the origin at the lateral malleolus. This resulted in the following equation :

$$pGRF = 1/GRF*(x_{foot}*W_f+x_{calf}*W_c+x_{thigh}*W_t+x_{hat}*0.5*W_{HAT}) \quad (7)$$

where : x_{foot} , x_{calf} , x_{thigh} and x_{hat} are the horizontal distances between the centre of gravities for the respective limb segments and the position of the lateral malleolus, as indicated in Figure 4.3.

Referring to Figure 4.3 :

$$x_{foot} = 0.44*foot_l-foot_p \quad (8)$$

$$x_{calf} = 0.58*calf_l*\cos \alpha \quad (9)$$

$$\begin{aligned} x_{thigh} &= 0.39*thigh_l*\sin(90+\alpha-\beta) \\ &\quad -(thigh_l*\sin(90+\alpha-\beta)-calf_l*\cos\alpha) \\ &= -0.61*thigh_l*\sin(90+\alpha-\beta)+calf_l*\cos\alpha \end{aligned} \quad (10)$$

$$\begin{aligned} x_{hat} &= c_{hat}*\sin(90+\beta-\phi-\alpha)-(thigh_l*\sin(90+\alpha-\beta) \\ &\quad = c_{hat}*\sin(90+\beta-\phi-\alpha) - thigh_l*\sin(90+\alpha-\beta) \\ &\quad +calf_l*\cos\alpha \end{aligned} \quad (11)$$

Where: $W_f = W_{foot}$: the weight of the foot

$W_c = W_{calf}$: the weight of the calf

$W_t = W_{thigh}$: the weight of the thigh

W_{HAT} is the combined weight of the head, upper limbs and trunk.

4.4 Kinetic Analysis

As in the kinematic approach, the human body was modelled as a four-linked-segment model. The equations for hip, knee and ankle torque derived for the kinematic approach, were applicable for this approach.

Referring to Figure 4.3, by taking moments for the body about the Y-axis, with the origin at the lateral malleolus, and equating to zero for equilibrium, resulted in the following equation:

$$-pGRF*GRF+x_{foot}*W_f+x_{calf}*W_c+x_{thigh}*W_t+x_{hat}*0.5*W_{HAT}=0 \quad (12)$$

As can be seen, Equations 7 and 12 are mathematically the same. However, in the kinematic approach pGRF is mathematically obtained whereas, in the kinetic approach, pGRF is experimentally obtained. Both the kinematic and kinetic approaches resulted in equations with numerous unknown variables thus reducing them to an indeterminate system. Many of these variables however, include the segmental weights, mass centres and lengths, position of the ground reaction forces, and the hip, knee and ankle angles, which can either be determined experimentally or estimated from the work of previous researches.

4.5 Introduction to the Kinematic and Kinetic Studies

In the previous sections, the development of the mathematical model was described. It was shown that the joint torques could be calculated from two aspects: kinematically and kinetically. There were however, certain unknown variables which could not be calculated and it was thus necessary to obtain these variables experimentally.

For this purpose a number of experiments were conducted on ten able-bodied subjects, three males and seven females. Two separate studies were conducted on each subject. The first one required the use of a digital camera system and video recording equipment. The aim of these experiments was to establish the values for alpha, beta and phi required for the kinematic part of the mathematical model.

The second study required the use of a segmented force plate and video recording equipment to generate values for pGRF (position of the ground reaction force under the foot) and corresponding alpha values. Since one of the aims of this project was to develop a mathematical model of the lower limb which would predict the joint torques for the standing-up and sitting-down procedures, it is thus reasonable to assume that the results of these experiments, which were conducted on able-bodied people, can be taken as representative of standing up procedures involving spinal cord injured persons.

4.6 Establishing a Suitable Standing-Up Procedure

One of the criteria of a suitable standing-up procedure is that the movement, and consequently, muscle contraction, which the spinal cord injured person must perform to achieve standing, should be as natural as possible. As rising from a seated to a standing position is something most people perform numerous times during a day, it is reasonable to assume that the body performs this task with minimal energy expenditure and a smooth motion. Factors which are difficult to take into account are, variations in anatomical structure from one person to another, and variations in the characteristic movements ascribed to muscles.

Taking the above points as guidelines to establish a suitable standing-up procedure, a number of experiments were conducted to achieve these aims. The aim of this subdivision is, to present the methodology of the experiments performed together with the results obtained.

4.7 Experiment 1 - Data Capturing Using the Digital Camera System and Video Camera

4.7.1 Aim

The aim of this experiment was to obtain kinematic data of the standing-up procedure, relating to the hip, knee and ankle angles for use in solving the torque Equations 4,5, and 6.

4.7.2 Method

First a suitable standing-up procedure had to be established. This relied heavily on the experience and knowledge of various people. Discussion with physiotherapists, occupational therapists and doctors, together with a method of standing called the "Alexander Method", lead to an acceptable procedure.

An important factor was the initial positioning of the feet and chair height as shown in Figure 4.5. The chair height used was one which complied with that of the standard folding wheelchair height as per SABS 1060 namely, 480-520mm, (SABS, 1985).

The standing-up process was performed with the arms folded and resting on the chest. The movement was initiated by flexion of the hip joint until the centre of gravity of the whole body was

Technology, 2809 East Columbia Road, Suite 100, Chicago, IL 60640. The subject was seated in front of the ankle. Once this position was assumed, hip flexion ceased and was basically followed by extension of the hip, knee and ankle joints to achieve lift off (initial raising from the seat) and eventually upright posture. The whole procedure took approximately four seconds, with the actual standing up sequence approximately two seconds.

In Section 4.3.1 (page 57), three equations (equations 4, 5 and 6) were developed which in predicting the torque generated at the hip, knee and ankle joints for the standing-up procedure, require the angles at these three joints.



Figure 4.5
Subject in the sitting position.

It was thus necessary to measure these angles. The system used to capture data was a "micronEye" digital camera (Micron

Technology, 2805 East Columbia Road, Boise, Idaho, USA) which interfaces directly with a personal micro-computer. The camera has a 10mm wide lens and is housed in a specially constructed casing together with a spotlight and reflective mirror as shown in Figure 4.6. Retro-reflective material (Scotchlite from 3M) was affixed to table tennis balls which were used to mark anatomical points on the subject being tested. Markers were affixed to the lateral trunk superior to the iliac crest, the greater trochanter, the lateral epicondyle of the femur and lateral malleolus, as shown in Figure 4.7. The subject being tested was first shown a video recording of the desired standing up procedure. She/he then carried out a number of trials to gain confidence and also practice the motion before data was captured. Data was captured for subjects rising from the seated to standing position. The sampling frequency was 13 Hz and 50 frames of data were captured, thus allowing 3.8 secs for the whole sequence. Software, written in TURBO PASCAL 4, developed by Vaughan, (Nissan et al, 1988 (b)), enabled 2-D data to be captured on disk, processed and displayed on the screen. Due to lack of laboratory space, the camera had to be placed on its side to capture data. Consequently, the digitized data which gave the X,Y co-ordinates of each point could not be used in this form. They were thus first swapped around and then converted to angles using the Procedure 'calcmicroneyeangles' and Function 'angval' and 'len' (see Appendix G) which could either be listed to the screen or printer.

Whilst data was being captured by the "micronEye, the process was simultaneously recorded on video tape. The reason for this was to ease the selection of the best standing-up sequence for analysis, after consulting with the various professional

personnel previously mentioned. Stick figure diagrams, graphically show the standing-up procedure, (see Appendix D).



Figure 4.6
The "micronEye" camera with the computer.



Figure 4.7
Subject showing position of markers.

4.7.3 Results

The results for two of the subjects, a 20 year old female and 22 year old male, as presented in Tables 4.1, 4.2, 4.3 and 4.4 linearly depict the digitized "micronEye" data and corresponding angles ϕ , β and α represented by angles 1, 2 and 3 respectively. For Tables 4.1 and 4.2 the following apply:

Markers 1, 2, 3 and 4 were placed on the subject's trunk (the lateral aspect of the tenth rib), hip (the lateral aspect of the greater trochanter), knee (the lateral epicondyle of the femur) and ankle (lateral malleolus). The fifth marker, a fictitious one, was generated by extending the fourth marker 50mm horizontally. Each X, Y coordinate represents the position of the markers in millimeters in relation to the top left hand corner (0; 0) measurements increase horizontally to the right and vertically downwards.

These angles were then read from a file into the Kinematic routine of the program thus enabling the hip, knee and ankle torques to be calculated.

From analysis of Tables 4.2 and 4.4, it is interesting to note that the knee-joint does not hyperextend in the upright posture. This is the posture which Petrofsky et al (1985) recommended, warning that hyper-extension of the knee-joint can lead to Charcot's joints. Contrary to Petrofsky et al (1985), other researches, (Kralj et al, 1980, Bajd et al, 1983), have actually recommended that the knee-joint be slightly hyper-extended, to enhance stability of the joint.

Subject = Beth Harley
 Date = 31 July 1988
 Comment = standing

Marker 1	Marker 2	Marker 3	Marker 4	Marker 5
X Y	X Y	X Y	X Y	X Y
673,1603	619,1764	260,1941	361,2326	311,2326
671,1603	617,1760	260,1941	361,2326	311,2326
666,1601	617,1758	260,1941	361,2326	311,2326
656,1600	612,1764	260,1941	361,2326	311,2326
645,1600	611,1759	256,1942	361,2326	311,2326
635,1598	606,1756	256,1942	363,2328	313,2328
624,1598	604,1756	255,1946	361,2326	311,2326
614,1597	600,1754	255,1946	361,2326	311,2326
600,1593	595,1752	255,1946	363,2328	313,2328
583,1592	590,1745	255,1946	363,2328	313,2328
565,1591	583,1743	255,1946	363,2328	313,2328
534,1577	568,1737	249,1947	363,2328	313,2328
506,1568	545,1715	243,1946	363,2328	313,2328
484,1549	526,1712	232,1951	362,2330	312,2330
459,1533	509,1687	223,1949	363,2328	313,2328
441,1516	489,1681	215,1951	362,2330	312,2330
421,1506	465,1668	200,1953	362,2330	312,2330
400,1489	445,1657	196,1953	362,2330	312,2330
387,1475	426,1652	189,1953	362,2330	312,2330
372,1458	413,1631	187,1955	362,2330	312,2330
359,1444	398,1625	187,1955	362,2330	312,2330
352,1430	386,1605	187,1951	362,2330	312,2330
345,1417	379,1600	189,1949	362,2330	312,2330
336,1401	362,1578	198,1946	363,2328	313,2328
328,1384	355,1570	198,1937	363,2328	313,2328
324,1373	350,1556	209,1937	361,2326	311,2326
322,1361	339,1547	209,1927	363,2328	313,2328
317,1347	332,1542	218,1925	363,2328	313,2328
310,1342	329,1532	226,1923	363,2328	313,2328
311,1333	322,1519	232,1918	365,2326	315,2326
306,1325	319,1514	243,1918	368,2326	318,2326
307,1321	316,1511	255,1909	368,2326	318,2326
304,1317	317,1506	260,1904	368,2326	318,2326
305,1313	317,1494	266,1899	369,2324	319,2324
307,1312	318,1492	275,1897	368,2326	318,2326
308,1306	317,1491	280,1897	368,2326	318,2326
308,1306	319,1489	288,1895	371,2324	321,2324
311,1305	317,1488	294,1895	371,2324	321,2324
311,1305	321,1486	300,1895	371,2324	321,2324
312,1302	322,1488	309,1893	371,2324	321,2324
314,1301	330,1486	314,1893	371,2324	321,2324
315,1299	330,1486	317,1891	371,2324	321,2324
320,1300	331,1484	322,1890	371,2324	321,2324
322,1299	332,1486	325,1893	371,2324	321,2324
322,1299	336,1484	328,1895	371,2324	321,2324
324,1297	334,1484	334,1890	371,2324	321,2324
322,1299	331,1484	334,1890	372,2322	322,2322
326,1296	334,1484	339,1891	373,2323	323,2323
326,1296	334,1484	341,1889	373,2323	323,2323
326,1296	338,1483	345,1890	373,2323	323,2323

Table 4.1
 List of "micronEye" data for a 20 year old female subject.

Subject = Beth Harley
Date = 31 July 1988

Frame 1	Angle 1 :	135.100	Angle 2 :	101.625	Angle 3 :	75.278
Frame 2	Angle 1 :	135.883	Angle 2 :	102.201	Angle 3 :	75.278
Frame 3	Angle 1 :	134.597	Angle 2 :	102.497	Angle 3 :	75.278
Frame 4	Angle 1 :	131.837	Angle 2 :	102.075	Angle 3 :	75.278
Frame 5	Angle 1 :	129.349	Angle 2 :	101.918	Angle 3 :	74.614
Frame 6	Angle 1 :	128.562	Angle 2 :	102.381	Angle 3 :	74.386
Frame 7	Angle 1 :	125.573	Angle 2 :	102.806	Angle 3 :	74.312
Frame 8	Angle 1 :	124.373	Angle 2 :	103.380	Angle 3 :	74.312
Frame 9	Angle 1 :	121.422	Angle 2 :	103.735	Angle 3 :	74.084
Frame 10	Angle 1 :	118.397	Angle 2 :	105.067	Angle 3 :	74.084
Frame 11	Angle 1 :	115.135	Angle 2 :	105.866	Angle 3 :	74.084
Frame 12	Angle 1 :	111.331	Angle 2 :	106.584	Angle 3 :	73.253
Frame 13	Angle 1 :	112.313	Angle 2 :	109.944	Angle 3 :	72.526
Frame 14	Angle 1 :	114.702	Angle 2 :	110.096	Angle 3 :	71.050
Frame 15	Angle 1 :	114.315	Angle 2 :	112.133	Angle 3 :	69.681
Frame 16	Angle 1 :	118.534	Angle 2 :	113.331	Angle 3 :	68.787
Frame 17	Angle 1 :	121.975	Angle 2 :	113.895	Angle 3 :	66.791
Frame 18	Angle 1 :	124.735	Angle 2 :	115.993	Angle 3 :	66.171
Frame 19	Angle 1 :	129.318	Angle 2 :	117.124	Angle 3 :	65.354
Frame 20	Angle 1 :	131.785	Angle 2 :	119.994	Angle 3 :	64.894
Frame 21	Angle 1 :	135.237	Angle 2 :	122.252	Angle 3 :	64.894
Frame 22	Angle 1 :	139.142	Angle 2 :	125.244	Angle 3 :	65.167
Frame 23	Angle 1 :	140.900	Angle 2 :	127.021	Angle 3 :	65.583
Frame 24	Angle 1 :	147.465	Angle 2 :	132.526	Angle 3 :	66.573
Frame 25	Angle 1 :	148.635	Angle 2 :	133.891	Angle 3 :	67.074
Frame 26	Angle 1 :	151.657	Angle 2 :	138.404	Angle 3 :	68.682
Frame 27	Angle 1 :	155.900	Angle 2 :	140.082	Angle 3 :	68.952
Frame 28	Angle 1 :	158.851	Angle 2 :	143.580	Angle 3 :	70.190
Frame 29	Angle 1 :	159.701	Angle 2 :	146.612	Angle 3 :	71.299
Frame 30	Angle 1 :	163.737	Angle 2 :	149.131	Angle 3 :	71.906
Frame 31	Angle 1 :	165.669	Angle 2 :	152.481	Angle 3 :	73.059
Frame 32	Angle 1 :	168.680	Angle 2 :	156.136	Angle 3 :	74.852
Frame 33	Angle 1 :	168.049	Angle 2 :	157.634	Angle 3 :	75.730
Frame 34	Angle 1 :	169.006	Angle 2 :	159.212	Angle 3 :	76.376
Frame 35	Angle 1 :	170.385	Angle 2 :	161.694	Angle 3 :	77.803
Frame 36	Angle 1 :	172.066	Angle 2 :	163.191	Angle 3 :	78.422
Frame 37	Angle 1 :	172.432	Angle 2 :	164.849	Angle 3 :	79.105
Frame 38	Angle 1 :	175.042	Angle 2 :	166.652	Angle 3 :	79.834
Frame 39	Angle 1 :	173.930	Angle 2 :	167.604	Angle 3 :	80.566
Frame 40	Angle 1 :	174.704	Angle 2 :	169.818	Angle 3 :	81.761
Frame 41	Angle 1 :	172.856	Angle 2 :	170.077	Angle 3 :	82.392
Frame 42	Angle 1 :	173.647	Angle 2 :	170.972	Angle 3 :	82.838
Frame 43	Angle 1 :	175.054	Angle 2 :	172.324	Angle 3 :	83.593
Frame 44	Angle 1 :	175.949	Angle 2 :	172.828	Angle 3 :	83.871
Frame 45	Angle 1 :	174.671	Angle 2 :	173.154	Angle 3 :	84.269
Frame 46	Angle 1 :	176.896	Angle 2 :	175.070	Angle 3 :	85.070
Frame 47	Angle 1 :	177.551	Angle 2 :	175.293	Angle 3 :	84.964
Frame 48	Angle 1 :	178.429	Angle 2 :	176.304	Angle 3 :	85.507
Frame 49	Angle 1 :	178.663	Angle 2 :	176.776	Angle 3 :	85.745
Frame 50	Angle 1 :	177.238	Angle 2 :	177.237	Angle 3 :	86.264

Table 4.2
List of "micronEye" angles for a 20 year old female subject.

Subject = Allan Horn
Date = 15 July 1988
Comment = standing up

Marker 1	Marker 2	Marker 3	Marker 4	Marker 5
X Y	X Y	X Y	X Y	X Y
474,1292	393,1428	65,1552	155,1917	105,1917
473,1293	393,1428	63,1552	155,1917	105,1917
473,1293	395,1429	65,1552	155,1917	105,1917
473,1293	395,1429	66,1553	153,1918	103,1918
471,1293	395,1432	66,1553	155,1917	105,1917
465,1293	395,1434	66,1553	155,1917	105,1917
461,1292	395,1434	66,1553	155,1917	105,1917
452,1292	395,1434	66,1553	155,1917	105,1917
442,1291	395,1429	66,1553	153,1918	103,1918
433,1291	391,1429	67,1555	153,1918	103,1918
418,1292	387,1427	64,1553	153,1918	103,1918
402,1290	383,1426	63,1555	153,1918	103,1918
384,1290	375,1421	60,1555	153,1918	103,1918
363,1289	369,1414	59,1555	155,1922	105,1922
344,1288	362,1413	57,1556	153,1923	103,1923
327,1273	353,1399	52,1555	153,1921	103,1921
314,1269	343,1384	52,1555	153,1921	103,1921
296,1249	333,1369	49,1553	153,1918	103,1918
280,1230	322,1355	46,1555	152,1921	102,1921
265,1222	309,1348	43,1555	153,1918	103,1918
252,1205	296,1333	40,1556	153,1923	103,1923
237,1185	287,1318	35,1556	153,1918	103,1918
227,1164	274,1304	34,1555	153,1921	103,1921
212,1145	261,1293	32,1555	153,1921	103,1921
201,1130	247,1282	32,1555	153,1918	103,1918
185,1111	234,1262	32,1555	153,1921	103,1921
171,1094	222,1255	34,1555	153,1918	103,1918
160,1073	210,1239	34,1552	153,1923	103,1923
151,1055	199,1231	35,1551	153,1921	103,1921
141,1047	190,1218	36,1545	153,1921	103,1921
132,1030	179,1215	40,1540	153,1921	103,1921
126,1015	172,1202	46,1538	153,1921	103,1921
119,1005	163,1192	54,1536	153,1918	103,1918
113, 990	157,1183	61,1535	155,1920	105,1920
107, 985	150,1172	65,1534	157,1918	107,1918
104, 981	146,1172	69,1533	159,1922	109,1922
100, 967	141,1172	73,1531	159,1919	109,1919
97, 964	137,1165	80,1530	161,1918	111,1918
92, 960	132,1159	87,1530	161,1918	111,1918
90, 954	126,1152	93,1528	161,1918	111,1918
87, 948	124,1150	96,1528	161,1918	111,1918
87, 946	117,1152	101,1520	162,1921	112,1921
85, 943	115,1150	106,1520	164,1924	114,1924
84, 942	110,1150	112,1519	162,1923	112,1923
79, 942	106,1150	115,1517	162,1921	112,1921
78, 941	104,1149	115,1517	161,1918	111,1918
78, 941	95,1149	113,1516	163,1922	113,1922
74, 941	91,1147	114,1516	162,1923	112,1923
72, 941	88,1147	115,1514	163,1922	113,1922
69, 943	82,1149	109,1515	162,1921	112,1921

Table 4.3
List of "micronEye" data for a 22 year old male subject.

Subject = Allan Horn
Date = 15 July 1988

Frame 1	Angle 1 :	141.358	Angle 2 :	96.897	Angle 3 :	76.157
Frame 2	Angle 1 :	141.074	Angle 2 :	96.504	Angle 3 :	75.897
Frame 3	Angle 1 :	140.102	Angle 2 :	96.586	Angle 3 :	76.157
Frame 4	Angle 1 :	140.351	Angle 2 :	97.373	Angle 3 :	76.695
Frame 5	Angle 1 :	138.902	Angle 2 :	96.667	Angle 3 :	76.361
Frame 6	Angle 1 :	136.331	Angle 2 :	96.270	Angle 3 :	76.361
Frame 7	Angle 1 :	134.640	Angle 2 :	96.270	Angle 3 :	76.361
Frame 8	Angle 1 :	131.645	Angle 2 :	96.270	Angle 3 :	76.361
Frame 9	Angle 1 :	129.335	Angle 2 :	97.373	Angle 3 :	76.695
Frame 10	Angle 1 :	127.762	Angle 2 :	97.965	Angle 3 :	76.823
Frame 11	Angle 1 :	124.437	Angle 2 :	97.747	Angle 3 :	76.430
Frame 12	Angle 1 :	119.742	Angle 2 :	98.111	Angle 3 :	76.214
Frame 13	Angle 1 :	116.909	Angle 2 :	98.807	Angle 3 :	75.777
Frame 14	Angle 1 :	111.861	Angle 2 :	99.708	Angle 3 :	75.358
Frame 15	Angle 1 :	107.052	Angle 2 :	100.491	Angle 3 :	75.438
Frame 16	Angle 1 :	105.374	Angle 2 :	101.776	Angle 3 :	74.507
Frame 17	Angle 1 :	106.282	Angle 2 :	104.937	Angle 3 :	74.507
Frame 18	Angle 1 :	105.649	Angle 2 :	107.095	Angle 3 :	74.099
Frame 19	Angle 1 :	107.129	Angle 2 :	109.570	Angle 3 :	73.762
Frame 20	Angle 1 :	108.477	Angle 2 :	111.033	Angle 3 :	73.195
Frame 21	Angle 1 :	111.957	Angle 2 :	113.906	Angle 3 :	72.889
Frame 22	Angle 1 :	112.829	Angle 2 :	115.404	Angle 3 :	72.082
Frame 23	Angle 1 :	117.779	Angle 2 :	118.241	Angle 3 :	71.975
Frame 24	Angle 1 :	120.744	Angle 2 :	120.609	Angle 3 :	71.721
Frame 25	Angle 1 :	124.632	Angle 2 :	123.396	Angle 3 :	71.672
Frame 26	Angle 1 :	127.402	Angle 2 :	127.093	Angle 3 :	71.721
Frame 27	Angle 1 :	130.439	Angle 2 :	129.905	Angle 3 :	71.929
Frame 28	Angle 1 :	134.208	Angle 2 :	132.852	Angle 3 :	72.215
Frame 29	Angle 1 :	137.752	Angle 2 :	135.304	Angle 3 :	72.361
Frame 30	Angle 1 :	139.010	Angle 2 :	137.561	Angle 3 :	72.753
Frame 31	Angle 1 :	142.687	Angle 2 :	140.224	Angle 3 :	73.427
Frame 32	Angle 1 :	145.471	Angle 2 :	143.762	Angle 3 :	74.354
Frame 33	Angle 1 :	149.005	Angle 2 :	147.766	Angle 3 :	75.461
Frame 34	Angle 1 :	152.011	Angle 2 :	151.014	Angle 3 :	76.277
Frame 35	Angle 1 :	153.863	Angle 2 :	153.182	Angle 3 :	76.491
Frame 36	Angle 1 :	155.831	Angle 2 :	155.006	Angle 3 :	76.999
Frame 37	Angle 1 :	157.836	Angle 2 :	156.802	Angle 3 :	77.537
Frame 38	Angle 1 :	160.025	Angle 2 :	159.440	Angle 3 :	78.223
Frame 39	Angle 1 :	161.815	Angle 2 :	162.202	Angle 3 :	79.115
Frame 40	Angle 1 :	164.887	Angle 2 :	165.038	Angle 3 :	80.035
Frame 41	Angle 1 :	165.581	Angle 2 :	166.295	Angle 3 :	80.486
Frame 42	Angle 1 :	169.239	Angle 2 :	168.970	Angle 3 :	81.385
Frame 43	Angle 1 :	170.383	Angle 2 :	170.295	Angle 3 :	81.734
Frame 44	Angle 1 :	172.975	Angle 2 :	173.198	Angle 3 :	82.969
Frame 45	Angle 1 :	174.126	Angle 2 :	174.727	Angle 3 :	83.348
Frame 46	Angle 1 :	174.665	Angle 2 :	175.060	Angle 3 :	83.379
Frame 47	Angle 1 :	178.192	Angle 2 :	175.751	Angle 3 :	82.946
Frame 48	Angle 1 :	178.678	Angle 2 :	176.805	Angle 3 :	83.304
Frame 49	Angle 1 :	179.645	Angle 2 :	177.338	Angle 3 :	83.207
Frame 50	Angle 1 :	179.352	Angle 2 :	176.704	Angle 3 :	82.524

Table 4.4
List of "micronEye" angles for a 22 year old male subject.

4.8 Experiment 2 - Data Capturing Using a Force Plate and Video Camera

4.8.1 Aim

The aims of this experiment were, to obtain kinetic data, for the standing-up procedure, relating the position of the ground reaction forces, in relation to the lateral malleolus, to the corresponding ankle angles. In addition, to verify the claim that standing-up is a quasi-static process (Bajd et al, 1982).

4.8.2 Method

The kinetic approach, (Section 4.4, pg59) resulted in equation 7:

$$-pGRF*GRF+x_{foot}*W_f+x_{calf}*W_c+x_{thigh}*W_t+x_{hat}*0.5*W_{HAT}=0 \quad (7)$$

As mentioned, the unknown variables were the ankle, knee and hip angles, alpha, beta and phi respectively, and pGRF. From Appendix E an equation relating beta and phi was obtained thus reducing the number of unknown variables to three, namely alpha, phi and pGRF. If the values for two of the three variables could be found, then the third could be found iteratively. One value which could be found, by means of a forceplate, was pGRF. Figure 4.8 is a typical plot of ankle, knee and hip angles against time as obtained from the "micronEye". From the plot of ankle angle against time it can be seen that after lift-off, where the value for alpha is a minimum, the relationship between alpha and time is basically linear. Thus by using this linear relationship in conjunction with the values obtained for pGRF, from the forceplate the number of unknown variables is reduced to one which, in turn, could be found by using an iterative process. A

segmented force plate, which is described in Appendix F, was used in conjunction with the "micronEye" camera, Appendix D, to obtain values for pGRF and alpha simultaneously. The subject under study, placed one foot centrally on the force plate with the other foot next to, but not on the force plate, and then stood up in the same manner as for the previous experiments.

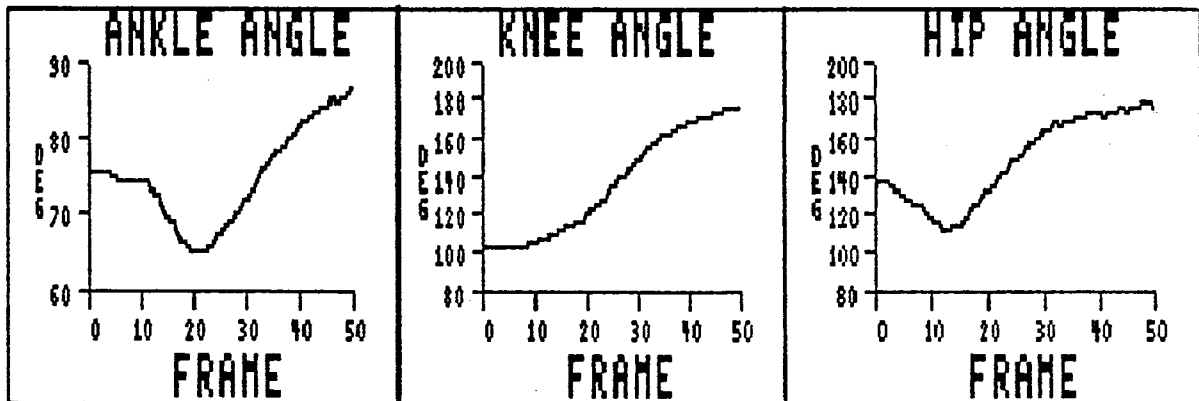


Figure 4.8
Plot of ankle, knee and hip angles against time.

4.8.3 Results

The results presented are for the same male subject used in the previous experiment. Figure 4.9 is a plot of the ground reaction force, in newtons, as a function of time, measured whilst the person was rising from the seated to the standing positions. Points (a) to (b) are for the period of sitting to "just off" the chair. The overshoot is probably due to inertia forces. Points (b) to (c) is correction of the overshoot. Points (c) to (d) are for the rest of the standing up process. The line joining (c) to (d) is basically horizontal which indicates no acceleration, thus supporting the assumption that standing-up is a quasi-static process (Bajd et al, 1982).

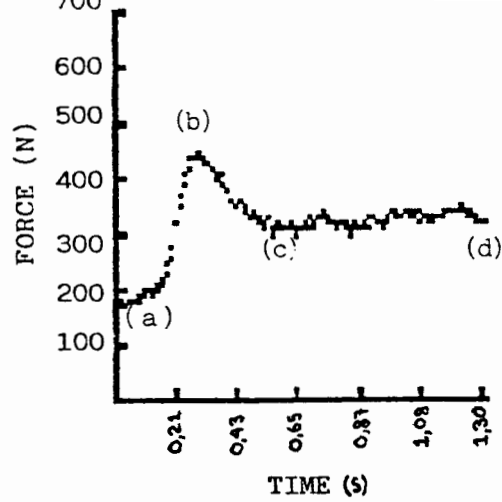


Figure 4.9
Plot of ground reaction forces for one foot
as a function of time.

Tables 4.5 and 4.6 are tables of the positions and corresponding values of GRF, where T is the sample number, X and Y are the coordinates of the foot on the force plate. Knowing the X,Y coordinates of the foot on the force plate, the actual value required for input into the program is calculated from formula 13, where CM is the centre of mass, for the whole body, obtained from the force plate.

$$P_{grf} = CM - (x_{heel} + p_{heel}) \quad (13)$$

Relating pGRF to its corresponding angle, alpha, was found to be an impractical task. The discrepancy arose due to the difference in sampling frequencies of the two measuring systems. The force plate sampled at a rate of 25Hz whereas the "micronEye" sampled at a rate of 13Hz. At the outset it seemed that a discrepancy of 1 or 2 Hz was negligible. However, numerous attempts to match up the two sets of results resulted in joint torque values which did not resemble those obtained from the "kinematic" approach. One of the major difficulties was in establishing the exact position of "lift-off" on an angle against time graph.

T	X	Y	T	X	Y	T	X	Y
0	-16069	14824	1	196	82	2	158	59
3	155	61	4	150	62	5	152	108
6	148	74	7	144	81	8	142	82
9	152	72	10	143	78	11	144	77
12	143	75	13	142	87	14	142	95
15	138	81	16	146	93	17	137	98
18	133	88	19	128	102	20	134	91
21	128	97	22	127	96	23	122	106
24	122	103	25	115	109	26	113	107
27	108	117	28	105	112	29	103	118
30	102	116	31	109	116	32	105	119
33	102	117	34	106	119	35	104	120
36	104	117	37	105	117	38	111	120
39	113	117	40	113	119	41	118	120
42	120	124	43	115	116	44	120	121
45	123	118	46	122	114	47	126	119
48	127	112	49	130	120	50	132	116
51	130	115	52	133	119	53	139	123
54	144	115	55	139	122	56	134	124
57	138	113	58	143	121	59	147	119
60	147	125	61	147	115	62	147	118
63	150	122	64	146	117	65	148	114
66	146	129	67	148	117	68	149	120
69	153	126	70	152	125	71	150	113
72	144	108	73	148	120	74	141	116
75	143	114	76	142	125	77	143	118
78	146	118	79	143	113	80	142	119
81	146	119	82	147	120	83	148	130
84	144	114	85	144	121	86	149	127
87	152	124	88	150	111	89	152	127
90	152	117	91	150	118	92	145	109
93	144	113	94	144	117	95	144	115
96	141	128	97	143	113	98	139	118
99	141	115	100	141	111	101	143	116
102	142	112	103	142	124	104	139	121
105	142	116	106	140	121	107	139	121
108	148	110	109	142	120	110	143	117
111	150	118	112	148	121	113	146	115
114	139	124	115	140	125	116	139	116
117	139	111	118	138	120	119	134	115
120	140	112	121	137	121	122	135	116
123	143	117	124	136	118	125	133	122
126	134	117	127	130	114	128	135	120
129	134	114	130	139	117	131	139	116
132	139	126	133	138	122	134	133	122
135	134	128	136	136	118	137	137	117
138	140	117	139	136	123	140	138	118
141	135	122	142	134	119	143	138	117
144	136	121	145	133	128	146	131	124
147	133	125	148	132	120	149	130	123

Table 4.5
Table of the positions of the ground reaction forces.

T FORCE		T FORCE		T FORCE		T FORCE		T FORCE	
0	177	1	167	2	168	3	169	4	183
5	183	6	178	7	185	8	186	9	179
10	186	11	196	12	197	13	200	14	189
15	199	16	214	17	210	18	216	19	235
20	246	21	258	22	283	23	317	24	348
25	366	26	387	27	411	28	425	29	440
30	444	31	476	32	450	33	445	34	441
35	431	36	426	37	418	38	421	39	405
40	406	41	406	42	389	43	383	44	376
45	365	46	356	47	354	48	354	49	360
50	346	51	344	52	334	53	331	54	338
55	328	56	327	57	325	58	324	59	314
60	326	61	309	62	304	63	314	64	316
65	315	66	308	67	306	68	307	69	316
70	314	71	313	72	305	73	311	74	322
75	307	76	328	77	311	78	312	79	324
80	327	81	333	82	336	83	329	84	323
85	330	86	317	87	325	88	313	89	318
90	317	91	320	92	307	93	301	94	312
95	314	96	322	97	312	98	310	99	317
100	312	101	332	102	330	103	327	104	319
105	323	106	316	107	310	108	316	109	328
110	337	111	327	112	342	113	333	114	328
115	337	116	338	117	342	118	331	119	335
120	343	121	322	122	342	123	329	124	329
125	316	126	319	127	332	128	335	129	332
130	333	131	338	132	342	133	342	134	340
135	340	136	341	137	348	138	342	139	337
140	332	141	331	142	341	143	324	144	329
145	325	146	316	147	324	148	317		

Table 4.6
Table of the values of the ground reaction forces (in newtons).

CHAPTER 5

ELECTROMYOGRAPHIC STUDIES

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The gathering and interpreting of electromyographic signals requires a knowledge of the technique used for the collection and processing of the data.

5.1 Surface Electrodes

In general, surface electrodes can be used for superficial muscles for the study of gross muscle function. The function of an electrode is to monitor the electromyographic signal by converting ionic currents in the body to normal electrical currents in the recording equipment. When applying surface electrodes it is important that the skin-electrode impedance be kept as low as possible. This is achieved by lightly rubbing the skin surface with emery paper and then cleaning it with alcohol swabs. A limiting factor was the signal attenuation caused by large amounts of subcutaneous fat. The only way around this problem was to select lean subjects for electromyographic studies. The electrodes used were the commercially available, pre-gelled disposable, silver/silver chloride ones used in electro-cardiography. They were 1.5 cm in diameter.

5.2 Amplifier

The amplifier used in electromyography serves the following functions:

- (i) provides isolation between the source of signals, namely the subject and the recording instrumentation.
- (ii) converts the current signal to a voltage signal.
- (iii) provides distortionless reproduction of the signal.
- (iv) provides voltage gain.
- (v) reduces noise.

An eight channel amplifier, designed by Hewitson, (1987), was used to record electromyographic signals. Each channel consisted of four elements. The first of these was an instrumentation amplifier with a differential input, high common mode rejection ratio with gain set at 100, and a very high input impedance. The latter was important because the magnitude of signals being recorded is very small (0.5 - 1000 microvolt) (Soderberg and Cook, 1984). The second, third and fourth elements were a high pass filter (cut off frequency of 25 Hz and a gain of 10), a full wave rectifier (unity gain), and finally a low pass filter (cut off frequency of 30Hz and a gain of (-2)). The purpose of these three elements was to remove direct current and movement artefact from the signal. Total gain per channel was 2000 which typically gave a 2V maximum rectified output. The amplifier was powered by two rechargeable nickel-cadmium cells.

5.3 Isolation Amplifier

When mains-operated equipment is being used, special care must be taken to ensure that no lethal currents flow from the equipment through the person to ground, in case of equipment failure. There must, therefore, be no direct electrical connection between the input circuitry and any mains-powered circuitry (Soderberg and Cook, 1984). The available amplifier did not provide this isolation and an opto-coupled isolation amplifier had to be used in conjunction with it (Hefftner, 1986).

5.4 Analogue-to-Digital Conversion and Data Storage

After the electromyographic signal has been filtered and amplified it is still in the raw (analogue) form. The signal can

be displayed in this form by means of an oscilloscope or recorded on a chart recorder. However, storage of the signal would be a problem. A more convenient method, viz. sampling the electromyographic signals using a computer, was the method used in these experiments. To accomplish this, the electromyograph signal first had to be converted from an analogue to a digital form, by using a 16 channel, 12 bit analogue-to-digital converter card. A card with a dynamic input range of -5 volts to +5 volts was commercially available under the trade name "PC 26" from Eagle Electronics in Cape Town, and fitted into one of the extension slots of a personal computer. Figure 5.1 is a diagrammatic picture of the recording system.

Finally, software was written to control the operation of the analogue-to-digital converter which allowed the simultaneous recording of up to four channels with sampling rates of 300 Hz per channel. The recording data was stored in files on diskette for later analysis.

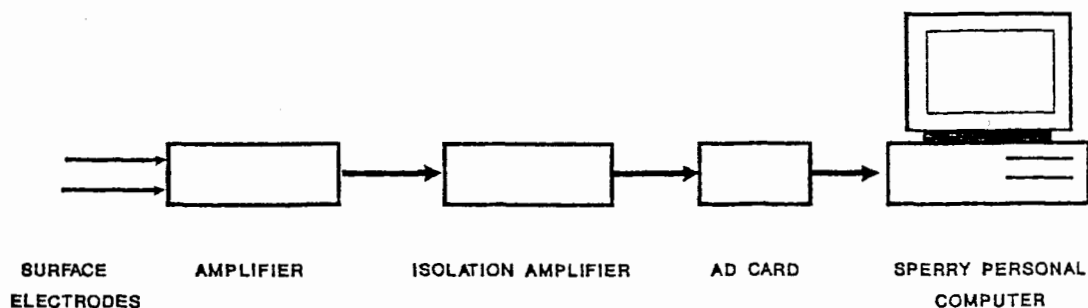


Figure 5.1
Schematic representation of the electromyographic
recording equipment.

All experiments were performed on the same ten subjects used for the kinematic and kinetic studies. The site for electrode placement was established using the guide to motor nerve points in 'Electrophysical Agents in Physiotherapy' (Wadsworth and Cnanmugam, 1980) and also the Rancho Los Amigos handbook (Benton et al,1981). The surface was prepared by initially shaving to remove the hairs, rubbed with emery paper to remove excess surface keratin, and finally wiping clean with alcohol swabs. The equipment used has been described earlier in Chapter 5.

5.5 Aim and Method

The muscles monitored were gluteus maximus, biceps femoris, rectus femoris, vastus medialis, gastrocnemius and tibialis anterior. The electrode locations were individually checked by maximally contracting each muscle. Figure 5.2 shows the subject with the electrodes attached. Electromyograms were recorded for the following three movements:

- (i) Standing-up from a seated position.
- (ii) Upright standing.
- (iii) Sitting-down from a standing position.

These movements were repeated a number of times to establish a uniform motion before electromyographic data of the superficial muscles of the lower limb was captured. Once the correct timing was established, four sets of data were captured for analysis. A sampling frequency of 300Hz was used. Sampling time was four seconds which resulted in 1200 data points. Further filtering of the data was performed using the filtering facilities of the software.

5.6 Results

As mentioned previously, there were ten subjects. The results for each subject will, however, not be discussed individually. This would be an extremely lengthy and tedious task. Instead the general trends of the electromyographic data will be given and mention of the variations will be made. It must also be noted that the fact that there are anatomical differences between people could probably account for some of these variations.

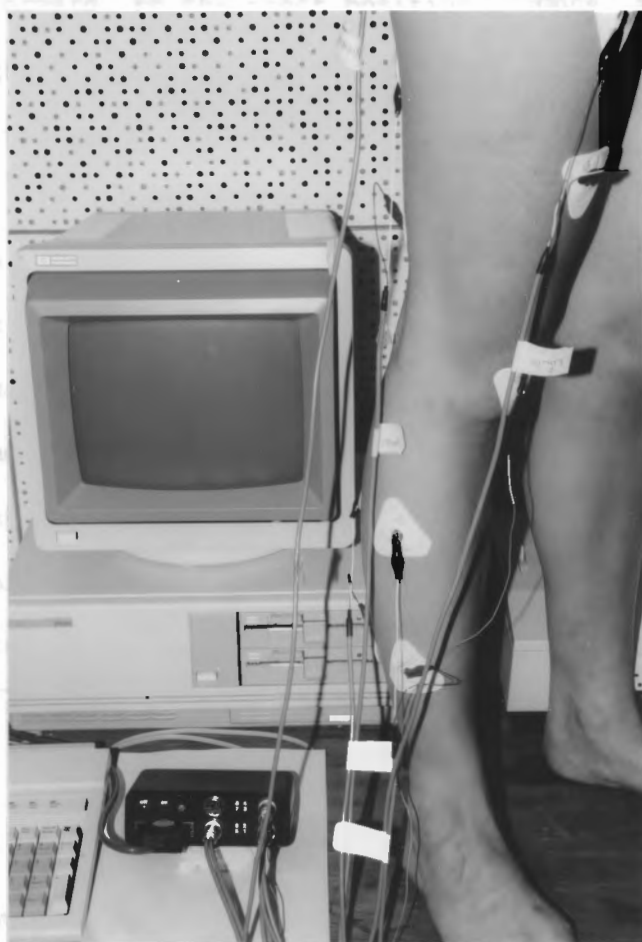


Figure 5.2
Subject standing showing the EMG equipment.

5.6.1 Standing-up

Figure 5.3 is a sample of the electromyograms obtained, in this case of the rectus femoris muscles. Figure 5.4 diagrammatically

represents the general trends of the electromyographic data for the rising-up to standing position activity.

Initial movement was initiated by a forward pivoting of the upper body. For the initial stages of this movement, no muscle activity was observed. Obviously there had to be some muscle contraction which would depend on the initial sitting posture. If the posture was initially one with the trunk leaning beyond the erect position, then iliopsoas would probably contract to bring it forward to the erect position. Once in the erect position, gravity would be likely to take over, accompanied by the eccentric "playing out" action of the erector spinae muscles. Soon after passing the erect position, muscle activity was observed in biceps femoris of all the subjects, indicating an antagonistic function to stabilize the hip joint. Just prior to lift-off, muscle activity was observed in vastus medialis, rectus femoris, and gastrocnemius, with the exception of two subjects, in which activity in vastus medialis was only recorded after lift-off. A little later muscle activity was also observed in tibialis anterior.

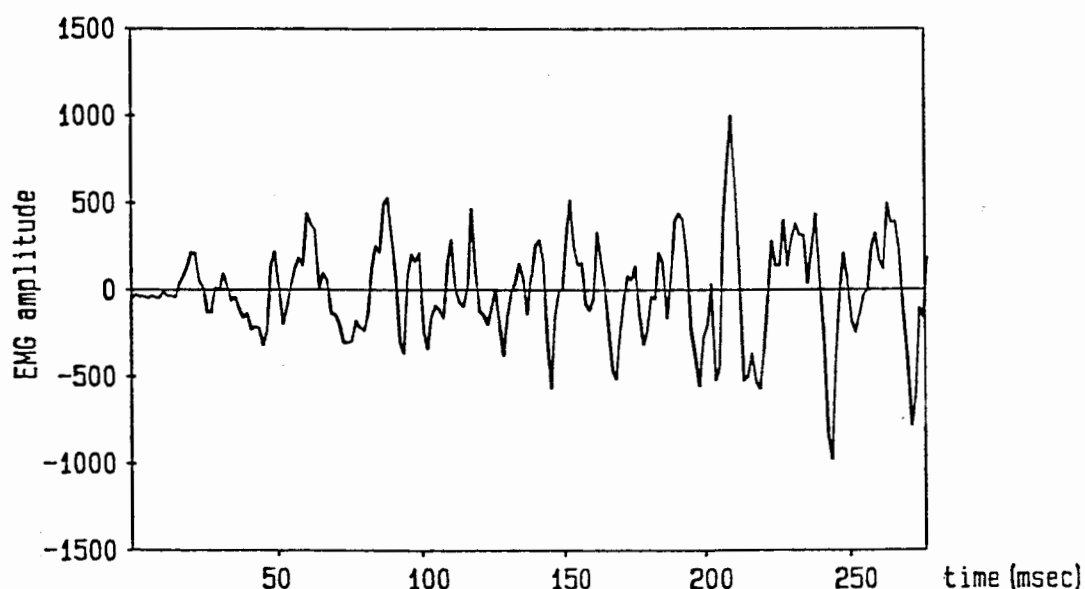


Figure 5.3
EMG of the rectus femoris muscle during
the standing-up procedure.

In all but one subject, no activity was observed in gluteus maximus throughout the whole standing-up sequence. This finding was in contrast to that of Kelley et al, (1976) who recorded slight muscle activity in gluteus maximus. Basmajian in his book "Muscles Alive" (Basmajian, 1978 page 241), states that gluteus maximus is only active when heavy or moderate efforts were made in the movement classically ascribed to gluteus maximus. Probably activity in gluteus maximus would thus be recorded if standing up were to be executed quickly, necessitating rapid acceleration. In determining the activities of the gluteus medius muscle, Soderberg and Dostal, (1978) found that all three segments (anterior, middle and posterior) were active during the standing-up and sitting-down movements, with more activity during standing-up. Thus, even though gluteus medius is primarily a hip abductor muscle, it appears that it also stabilises the hip joint during these movements.

Activity increased to a maximum for vastus medialis, rectus femoris, gastrocnemius, tibialis anterior and biceps femoris at lift-off, with the exception of vastus medialis in the two subjects previously mentioned. Activity then gradually tapered off as the upright standing position was neared, with the most activity being recorded in rectus femoris, followed by vastus medialis. With the exception of three subjects, tibialis anterior activity ended before upright posture was assumed. This was probably because the centre of gravity for the whole body was in front of the lateral malleolus at this position, (Appendix F) and in the terminal part of standing the ankle movement is plantar flexion. For one of the above mentioned three subjects, tibialis anterior was active throughout. For the other two, tibialis anterior activity ended a little later.

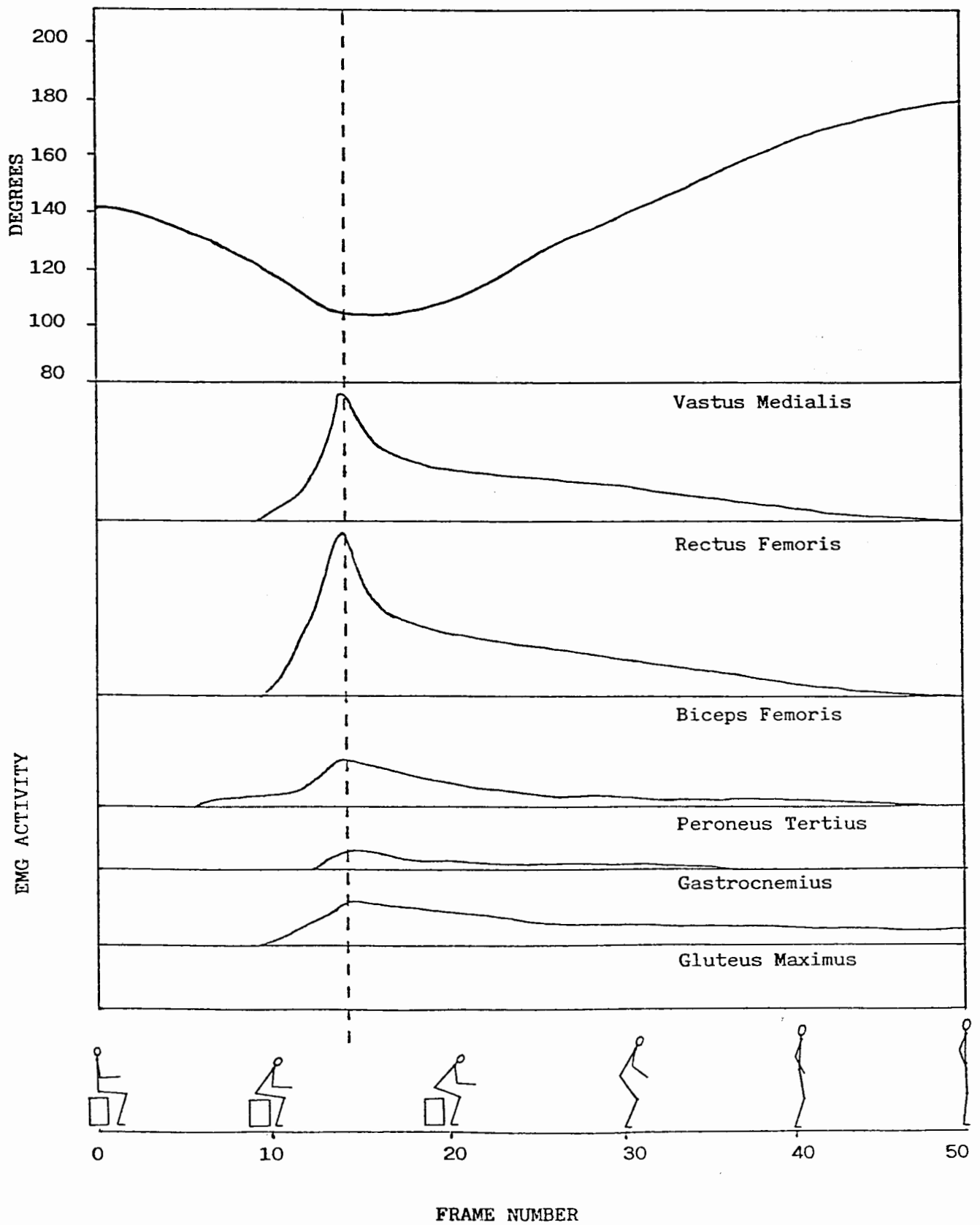


Figure 5.4
EMG activity for a typical subject
during the standing-up procedure.

Gastrocnemius, on the other hand, was active throughout standing-up, thus supporting the hypothesis that plantar flexion occurs at the ankle at the terminal part of standing.

5.6.2 Standing

To maintain equilibrium in the standing position with the least energy expenditure a posture is assumed in which the line through the centre of gravity should pass each joint in such a way as to lock it, ie. behind the hip, in front of the knee and in front of the ankle.

5.6.2.1 Posture of the lower limbs

When standing in the relaxed position no significant activity was recorded for the tibialis anterior muscle. One exception was the same subject for whom muscle activity, for tibialis anterior, was recorded throughout standing-up. A possible reason for these findings was that this person had tight tendons of Achilles. On the other hand, gastrocnemius (and probably soleus) was found to be active for all the subjects all the time. This was because the line of gravity actually fell in front of the knee and ankle joints. These findings were in general agreement with those of Basmajian and Bentzon, (1954), Kobayashi and Matsui, (1976), Williams and Svensson, (1968) and Soderberg and Dostal, (1978). Basmajian and Bentzon, (1954) found activity in tibialis anterior when the person leant backwards, with no activity in gastrocnemius and soleus; the opposite was found when leaning forwards. This would be expected as leaning either forwards or backwards would result in a compensatory activity in the muscles to maintain equilibrium.

Contrary to the findings of Kralj and Vodovnik, (1977(b)) very little activity was found in the quadriceps muscles, whereas biceps femorus was generally slightly but constantly active. No activity was found in the gluteus maximus muscles. Basmajian and Bentzon, (1954) had similar findings. Instead they found that gluteus medius and tensor fascia latae were active. Since tensor fascia latae is a hip flexor and abductor, and gluteus medius an abductor and, to some extent, a rotator, it would seem that they both are stabilising the hip joint. Basmajian in his book "Muscles Alive", (page 238), explains that the iliopsoas muscle also remains constantly active in the upright posture acting as a vital ligament to prevent hyper-extension of the hip joint in upright standing.

5.6.3 Sitting Down

To initiate sitting-down, the knee joint has to unlock. This function is performed by the popliteus muscle which is a deep muscle and can therefore not be monitored using surface electrodes.

The sitting-down procedure was the reversal of the standing-up procedure. The control of sitting down comes from eccentric contractions of the quadriceps muscles (rectus femoris, vastus medialis and lateralis). One could suspect this to cause both greater fatigue and force generation per fibre. This was not found to be the case from the data obtained as there was no general decrease in the frequency of the electromyograms. This finding also supports the work of Basmajian, (1978) who found, when doing a series of electromyographic studies of the quadriceps femoris muscle, that concentric contraction caused

more activity than eccentric contraction. From this one could deduce that when a muscle is shortening, it is more active than when lengthening under load (antagonist action). Thus it would seem that sitting-down requires less muscle force than standing-up. It must be noted, however, that these experiments were neither strenuous, nor of sufficient duration to cause the muscles to fatigue.

SECTION III
FINAL MODEL DEVELOPMENT

CHAPTER 6

FINALISATION OF THE MATHEMATICAL MODEL

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In Chapter 4, the kinematic and kinetic analyses resulted in equations containing a number of unknown variables. It was established that some of these variables could be obtained using the work of previous researchers, whilst others had to be determined experimentally or calculated mathematically.

In Section II, various experiments were conducted to establish various values and combinations for alpha, beta, phi and, for the kinetic analysis, pGRF.

The results of these experiments, together with those of previous researchers' work, have been incorporated into the above mentioned equations and are presented in this chapter.

6.1 Kinematic Analysis

In Chapter 4 the following equations were derived:

$$F_{KY} = -GRF + W_f + W_c \quad (1)$$

$$F_{HY} = -GRF + W_f + W_c + W_t \quad (2)$$

$$GRF = W_{HAT} + W_f + W_c + W_t \quad (3)$$

$$T_A = -GRF * pGRF + W_f * x_{foot} \quad (4)$$

$$T_K = T_A - F_{ay} * thigh_l * \cos\alpha - (W_c * 0.42 * thigh_l * \cos\alpha) \quad (5)$$

$$T_H = T_K + F_{ky} * thigh_l * \sin(90 + \alpha - \beta) + W_t * 0.39 * thigh_l * \sin(90 + \alpha - \beta) \quad (6)$$

$$pGRF = 1/GRF * (x_{foot} * W_f + x_{calf} * W_c + x_{thigh} * W_t + x_{hat} * 0.5 * W_{HAT}) \quad (7)$$

The segmental lengths were obtained by recording the necessary anthropometrical data. For this purpose a table, (Appendix B.3, Table B.1) was drawn up, on which the subject's various

anthropometrical data was recorded. Guidelines for measuring these values were obtained from the work of Chandler et al, (1975). These are included in a help file in the program. Using the work of Chandler et al, (1975) and the regression equations of Chandler and Vaughan (Vaughan, 1980), the segmental weights and their corresponding centre of gravity were calculated as in Appendices B and C.

Referring to Equation 3:

$$GRF = W_{HAT} + W_f + W_c + W_t \quad (3)$$

For this model bilateral symmetry was assumed which implied that W_{HAT} is equally divided between each lower limb. Thus Equation 3 becomes :

$$GRF = 0.5 * W_{HAT} + W_t + W_c + W_f \quad (7)$$

Assuming no horizontal acceleration occurs, the line of action of the ground reaction force (GRF) is directed vertically upwards, through the centre of mass for the whole body. This is a fair assumption as, in standing-up, most of the movement occurs in the vertical plane as is evident from a plot of the position of centre of mass for the standing up process (Figure 6.1), and also from the work of Kelley et al, (1976).

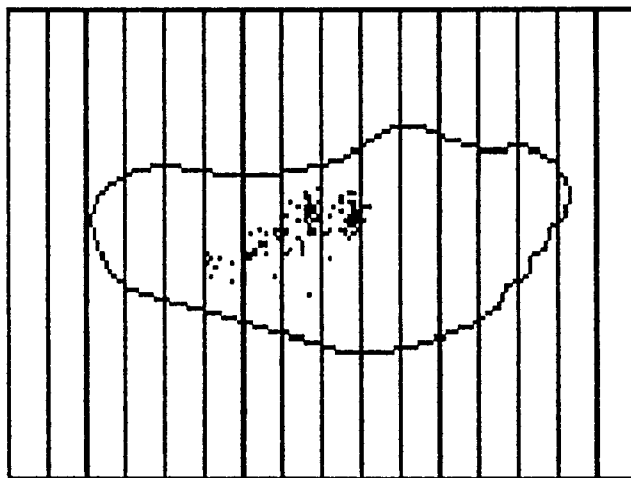


Figure 6.1
Plot of resultant centre of mass as $f(t)$.

Referring back to Equation 11, of Section 4.3.2 (page 58), it contains the value c_{hat} which is calculated as follows. From Figure 4.4 (page 57), c_{hat} is the position of the centre of mass for HAT, as measured from the hip joint for any angle ϕ . From Appendix C, Equation C8, the centre of mass for HAT, (cg_{hat}), measured from the base of the foot, was calculated to be :

$$cg_{\text{hat}} = (1/m[9])*(m[1]cg_{\text{head}}+m[2]*cg_{\text{trunk}}+2*(m[3]*cg_{\text{upperarm}}+m[4]*cg_{\text{forearm}}+m[5]*cg_{\text{hand}}) \quad (9)$$

From Appendix C, Equation C9, c_{hat} is:

$$c_{\text{hat}} = cg_{\text{hat}} - (lat_{\text{mal}}+calf_1+thigh_1) \quad (10)$$

6.1.1 Calculation of the Hip, Knee and Ankle Angles

Angles α , β and ϕ were obtained experimentally using the "micronEye" which was developed by Vaughan (Nissan et al, 1988,(b), Appendix D). Data was captured for subjects rising from a seated to a standing position with their arms folded on their chests. The "micronEye" returned the X-Y coordinates for the four markers placed on the lateral malleolus, lateral epicondyle, greater trochanter and lateral trunk, superior to the iliac crest. Table 4.1 (Section 4.7.3, page 66), is an example of the data generated by the subroutine, 'listmicroneyedata' using the X-Y coordinates obtained from the "micronEye". The hip, knee and ankle angles, for each frame, were calculated from the X-Y coordinates obtained from the "micronEye" using the functions 'angval' and 'len'. The calculations were based on the grounds that any angle in 2 dimensional space requires a set of three co-linear points to describe it. Figure 6.2 shows three co-linear

points A,B and C, forming a triangle. If the lengths of the three sides are known, then it is possible, using the Cosine rule, Equation 11, to determine each of the three angles of the triangle :

$$A^2 = B^2 + C^2 - 2*B*C*\cos A \quad (11)$$

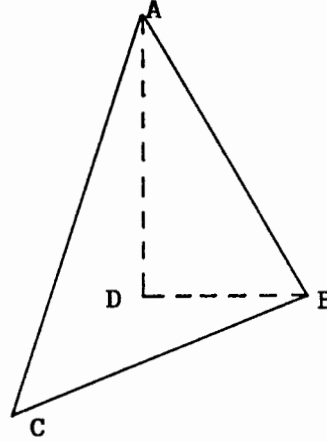


Figure 6.2
Diagram of triangle ABC.

The lengths of the triangle were then determined as follows. By taking any two points of the triangle, A and B, a 90 degree triangle is constructed by creating a third point, D, from these two points. This is done by taking either the x or y coordinates of point A and then the opposite coordinate of point B. The lengths of the two sides forming the 90 degree angle were obtained by subtracting the relevant coordinates of the third point, D, from the corresponding coordinates of points A and B. The length AB is then obtained from the relationship relating the three sides of a triangle, namely :

$$(AB)^2 = (AD)^2 + (BD)^2$$

This process is repeated for points B and C until all the lengths are known. These are then substituted into Equation 11 to calculate the required angles. By substituting these angles, together with the preceding established variables, into Equation

7, the corresponding value for pGRF for each set of joint angles was calculated. Finally, by substituting Equation 7 together with the necessary variables into Equations 4, 5 and 6, the corresponding ankle, knee and hip torques were calculated.

6.2 Kinetic Analysis

The derivations of Equations 1 to 6 in Section 4.3.1, (page 55), and of Equations 8 to 10 of Section 6.1, (page 89), are applicable to the kinetic analysis. Rearranging Equation 7, resulted in Equation 12, containing four variables; alpha, beta, phi and pGRF:

$$-pGRF*GRF+x_{foot}*W_f+x_{calf}*W_c+x_{thigh}*W_t+x_{hat}*0.5*W_{HAT}=0 \quad (12)$$

From the work of Kelley et al, (1976), a relationship, Equation 13, between beta and phi was obtained using the method of least squares (Appendix E).

$$\beta = 30.0149834 + 1.0880786 * \phi - 0.006618382 * \phi^2 + 0.000026807 * \phi^3 \quad (13)$$

However, as explained in Appendix E, using this equation resulted in erroneous values. Consequently, "micronEye" data was used to obtain the following relationship between beta and phi :

$$\beta = -19369.400377 + 725.30394266 * \phi - 10.650018262 * \phi^2 + 0.077098914 * \phi^3 - 0.000275425 * \phi^4 + 0.0000003892 * \phi^5 \quad (14)$$

Substituting for beta into Equation 12 reduced the number of variables to three namely, alpha, phi and pGRF.

Using the forceplate, "micronEye" and video camera, data depicting the position of GRF with respect to time was

obtained (Appendix F, and Tables 4.5 and 4.6, pages 73/4). Lift-off was established when GRF approximately equaled half body weight. The corresponding angle, α , was obtained using data from the "micronEye".

This relationship between α and pGRF reduced the number of unknown variables in Equation 12 to one, namely ϕ . ϕ was calculated by using an iterative process. This was done by entering a value for β into Equation 12, and adjusting ϕ until a particular value of ϕ satisfied the equation. Having established all four variables for a particular position in standing-up, these were substituted into Equations 4, 5 and 6 to calculate the corresponding hip, knee and ankle joint torques.

6.3 Hardware and Software Selection

The software language chosen was TURBO PASCAL V4.0 (Borland International Incorporated California) implemented on an IBM compatible personal computer. Pascal was chosen as it is more versatile than either BASIC or FORTRAN and offers a full library of graphics routines, namely the TURBO GRAPHIX TOOLBOX (Borland International). TURBO PASCAL V4.0 is also the software language at which the author is most proficient. It also allows the programmer to compile certain sections of a program into what is termed a UNIT. Thus, if more than one model is developed for testing, the model subroutines and variable declarations can be compiled into individual units allowing the programmer to select the required model and compile time with ease. The graphics card chosen was an IBM colour graphics card (CGA, 640 x 200 pixels), as the need for the greater pixel resolution of the IBM monochrome graphics card (720 x 350), was not required.

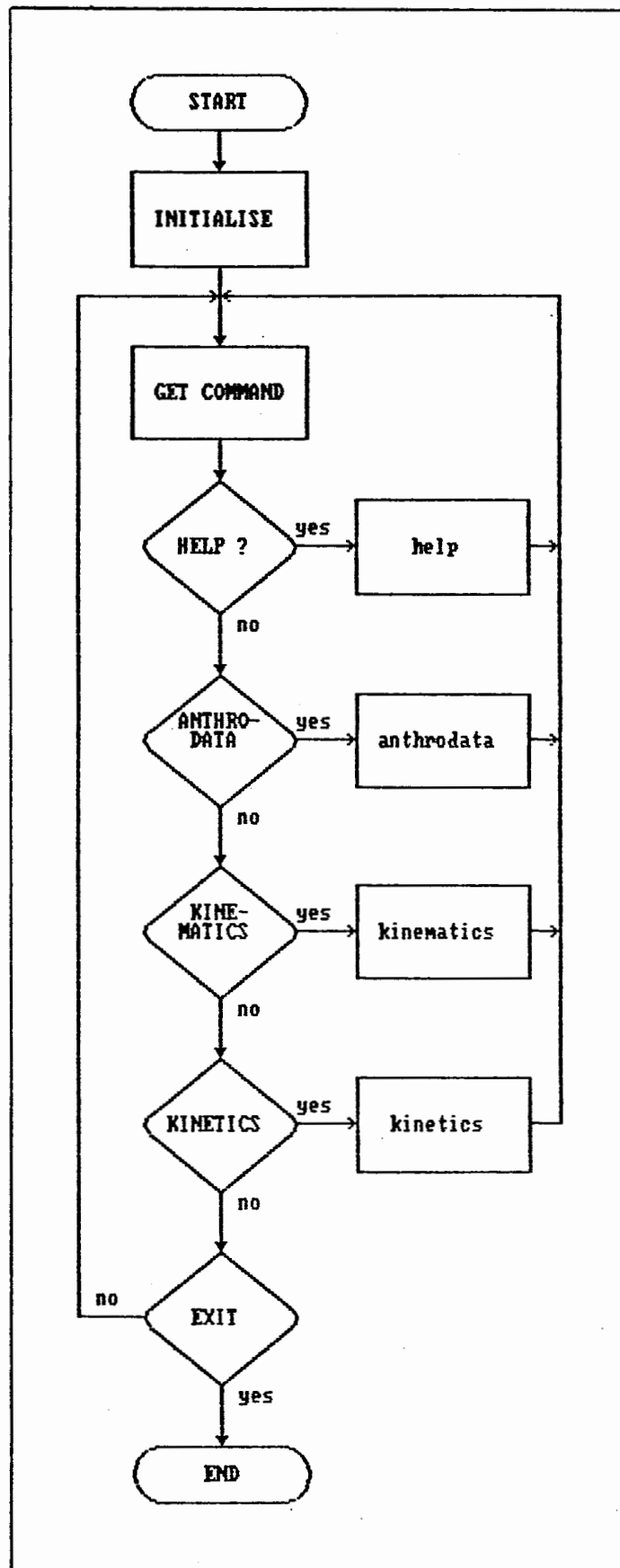


Figure 6.3
The main program flow diagram.

6.4 The Main Program Outline

Figure 6.3 shows a flow chart of the mathematical model program. A full listing of the program is given in Appendix I. The listing contains comment lines that explain the function of each subroutine. The main program is at the end of the listing. The program starts off by first checking to ascertain if the correct graphics card is being used and if not, the program defaults. The help screens, anthropometrical data and "micronEye" data files are then loaded. The help file is also opened and displayed only when required by typing "?". The anthrodata file is then opened and the subject's required anthropometrical data is loaded. If there is any editing to be done the program transfers control over to the anthrodata sub-routine until the operator has completed the desired editing. The program then continues and the operator has two choices of subroutines depending on the manner in which data was obtained. Once either kinematics or kinetics has been chosen the program transfers control to either of these subroutines. After the operator has completed the necessary calculations, the current files can either be saved or 'dumped' before terminating the program. A full description of these subroutines are given in Appendix H.

6.5 Operation of the Mathematical Model

All commands to the model are entered via the keyboard using the alpha-numeric and cursor keys. The cursor keys are used to move a selection bar around on the screen to select topics in the help screen and together with the alpha-numeric keys to enter or edit the data in the anthropometrical data files. Other keys that are used are the return/enter key for editing, spacebar to toggle

between screens and the escape key to return from a subroutine back to the main program.

6.5.1 The help facility

The help facility may be called up at any time during the running of the model by depressing the ? key. If the model is running at the time, the process is halted, the present process is saved, and the help menu displayed. The selection bar may be used to select a help topic in the top menu by moving it with cursor keys. The corresponding help text will then be displayed in the lower part of the screen. The escape key is depressed to return to the simulator. The original menu is restored and the model continues on from where it was interrupted.

6.5.2 Starting the mathematical model

The model hardware requirements are:

- IBM PC or compatible.
- IBM screen (CGA graphics card or equivalent).
- 20 Mbyte of memory (RAM).
- One floppy disk drive (360 Kbyte).
- Epson graphics printer or equivalent - optional.

The program disk must have the following files on it:

- ANGLE 4.EXE - Main program.
- UCTAS.HLP - Help file (UNIT).
- 4 x 6.FON - Graphics small font (UNIT).
- 12 x 6.FON - Graphics large font (UNIT).

RESTAND ANTHROPOMETRIC PROGRAM
JULY 88 UCT BIOENG

? : Help 1 : Anthrodata 2 : kinematics 3 : kinetics Esc OR 9 : Quit

Figure 6.4
Main menu screen.

ANTHRODATA PROCEDURE

1 : load from file 2 : save to file 3 : edit/view data 4 : finished

Figure 6.5
Anthrodata menu screen.

KINEMATICS PROCEDURE

1:anthrodata 2:microneyedata 3:plotdata Esc OR 9:Quit

Figure 6.6
Kinematics menu screen.

LOAD ANTHRO DATA PROCEDURE

Select anthropometrical data file

b:\ Search = b:*.ant			
ALLAN.ANT	ALLAN.ANT	ALLAN6.ANT	BETH.ANT
CAROL.ANT	CAROL1.ANT	CAROL2.ANT	DAVE.ANT
DAVE1.ANT	DAVE2.ANT	DAVE3.ANT	DAVE5.ANT
DAVE6.ANT			

Move with ↑ ↓ ← →, ← to select, Esc to quit

Figure 6.7
Load anthrodata menu screen.

To run the program, place the disk in the current drive and type the program name RESTAND followed by return. All commands must be followed by return, unless stipulated otherwise. The main menu screen will be displayed, shown in Figure 6.4. Should you

want to enter new anthropometrical data, type 1. If there is no floppy disk in drive B then the program will issue a warning 'File error or disk drive error!!'. For the usual case, the menu screen shown in Figure 6.5 will be displayed. However, should the user not wish to open up a new anthropometrical data file then either 2 or 3 is typed. This will result in the menu screen in Figure 6.6 being displayed. Type 1 to load the anthrodata file. As above, the program will display the menu screen to the screen. As can be seen, there are four options which are all self explanatory. To load from file, type 1 and the menu screen, shown in Figure 6.7, will be displayed. An anthropometrical data file is selected by moving the selection bar to the desired file using the cursor keys and then typing return. The program then returns to the main menu screen.

If any editing was done, this has to be saved by typing 2. The program will request a file name. Should the file name already exist (for new files only) then the warning 'File error or disk drive error!!' will appear on the screen and then return to the previous mode to enable the operator to retype the file name. After this has been completed type 4 and the program is returned to the KINEMATICS PROCEDURE screen.

Type 2 to load the microneye data file. The program will display a menu screen, from which a file is selected in the same manner as for an anthropometrical data file. Having selected the required files the program gives the operator the option to look at the "micronEye" data (Y/N?) on the screen. If Y is entered then the option to list the data either to the screen or printer is given by writing the command 'List to screen or printer (S/P) ?'. The data is then listed to whatever is chosen.

The same applies to the hip, knee and ankle angles. The program then returns to the KINEMATICS PROCEDURE menu. Typing 3 results in plotting of the data for the particular subject. There are two screens and toggling between them is achieved via the spacebar. On completion, typing ESC returns back to the KINEMATICS PROCEDURE MENU, then ESC or 9 is then typed to quit the model.

SECTION IV

USING THE MODEL TO STUDY STANDING UP AND SITTING DOWN

CHAPTER 7

RESULTS OF SIMULATION EXPERIMENTS

The results obtained from the experiments conducted to obtain kinematic, kinetic and electromyographic data for the standing-up procedure has been presented and discussed in Chapters 4 and 5. The results for two subjects, a 22 year old male, and a 20 year old female weighing 70.3kg and 50.4kg respectively, obtained from the mathematical simulation model, for both the kinematic and kinetic approaches, are presented here.

The anthropometrical data, listed in Tables 7.1 and 7.2 are the data which the model simulation program requires to calculate the various torques. This feature of the model allows the user to determine the torque requirements for each subject individually. A description of how to measure the various anthropometrical data is included in a "help file" in the program.

The results for the standing-up procedure, based on the measured anthropometric and experimental "micronEye" data for the above subjects, are presented in Figures 7.1, 7.2, 7.3 and 7.4. These are printouts obtained from the kinematics part of the simulation program. Because of the difficulty experienced in obtaining force plate data, (see Section 4.8.3, page 71), the data for the Kinetics approach was obtained from calculated values of pGRF and the values for alpha were obtained from the "micronEye". The results for the standing-up procedure, based on this method for for the above subjects are presented in Figures 7.6 and 7.7 which, as above, are printouts obtained from the Kinetics part of the simulation program.

Figures 7.1 and 7.3 are plots of joint torques and angles against frame number for the male and female subjects respectively. For the above figures the independent variable, the frame number, is

obtained from the "micronEye". The sampling frequency of the micronEye is 13Hz, thus each frame represents 1/13 sec (about 0.08 secs). For the torque against frame number (time) plots, the torque axes are divided up into positive and negative values as there is both flexion and extension of the joints during the standing-up procedure. The joint angle against time plots are the result of using the "micronEye" data for each subject only.

Referring to Figures 7.2 and 7.4, these are, with the exception of the plot of pGRF against ankle angle, a combination of Figures 7.1 and 7.3. The first three, Figures 7.2 and 7.4 (a), (b), (c), are plots of the ankle torque against ankle angle and likewise for the knee and hip. From these plots a change in the direction of joint torque sign is generally accompanied by a change in direction of the graph. For example, Figure 7.2 (a) is the plot of ankle torque against α . It begins with a torque value of approximately 20Nm and an α of 75 degrees. This reduces to 0Nm after which the torque becomes negative and is accompanied by a change in direction. The same applies to Figure 7.2 (c). However, Figures 7.2 (b) and 7.4 (b) are exceptions where there is a very slight change in torque sign, but no subsequent change in slope direction. A possible explanation for this is that the knee joint is hyperextended in the final stages of the standing-up procedure.

Figures 7.2 (d) and 7.4 (d), are plots of ϕ (hip angle) against β (knee angle). The initial stages are dominated by the change in ϕ (flexion of the hip joint) with no significant change in β . A change in slope indicates "lift-off", after which the plot of ϕ against β is virtually linear. Figures 7.2 (e) and 7.4 (e) are plots of pGRF against the ankle angle. The plot only starts from lift-off and, as can be seen, pGRF

PARTICIPANTS ANTHROPOMETRIC DATA

PARTICIPANTS NAME :Allan_Horn DATE (dd/mm/yy) :13/7/88
 BIRTHDATE (dd/mm/yy) :13/1/66 SEX (M or F) :m

(all measurements are in metres besides mass in kg)

TOTAL BODY MASS	:70.300	TOTAL HEIGHT	: 1.759
MASTOID TO VERTEX LENGTH	: 0.171	HEAD CIRCUMFERENCE	: 0.565
TORSO LENGTH	: 0.610		
CHEST DEPTH	: 0.233	CHEST BREADTH	: 0.341
BALL OF HUMERUS TO RADIALE LENGTH	: 0.288	BICEPS CIRCUMFERENCE	: 0.278
BALL OF HUMERUS TO VERTEX LENGTH	: 0.305	FOREARM CIRCUMFERENCE	: 0.271
STYLION TO METACARPAL III LENGTH	: 0.107	RADIALE TO STYLION LENGTH	: 0.250
HAND BREADTH	: 0.084	HAND DEPTH	: 0.029
THIGH LENGTH	: 0.393	MID-THIGH CIRCUMFERENCE	: 0.526
CALF LENGTH	: 0.426	CALF CIRCUMFERENCE	: 0.360
LATERAL MALLEOLUS HEIGHT	: 0.069	FOOT BREADTH	: 0.101
BACK OF FOOT TO LAT MAL LENGTH	: 0.062	FOOT LENGTH	: 0.251

Table 7.1

Printout of a 22 year old male subject's anthropometrical data.

PARTICIPANTS ANTHROPOMETRIC DATA

PARTICIPANTS NAME :Beth_Harley DATE (dd/mm/yy) :11/06/88
 BIRTHDATE (dd/mm/yy) :16/11/67 SEX (M or F) :F

(all measurements are in metres besides mass in kg)

TOTAL BODY MASS	:50.400	TOTAL HEIGHT	: 1.607
MASTOID TO VERTEX LENGTH	: 0.155	HEAD CIRCUMFERENCE	: 0.539
TORSO LENGTH	: 0.625		
CHEST DEPTH	: 0.206	CHEST BREADTH	: 0.308
BALL OF HUMERUS TO RADIALE LENGTH	: 0.282	BICEPS CIRCUMFERENCE	: 0.232
BALL OF HUMERUS TO VERTEX LENGTH	: 0.315	FOREARM CIRCUMFERENCE	: 0.212
STYLION TO METACARPAL III LENGTH	: 0.190	RADIALE TO STYLION LENGTH	: 0.235
HAND BREADTH	: 0.081	HAND DEPTH	: 0.030
THIGH LENGTH	: 0.275	MID-THIGH CIRCUMFERENCE	: 0.475
CALF LENGTH	: 0.382	CALF CIRCUMFERENCE	: 0.337
LATERAL MALLEOLUS HEIGHT	: 0.074	FOOT BREADTH	: 0.096
BACK OF FOOT TO LAT MAL LENGTH	: 0.039	FOOT LENGTH	: 0.245

Table 7.2

Printout of a 20 year old female subject's anthropometrical data.

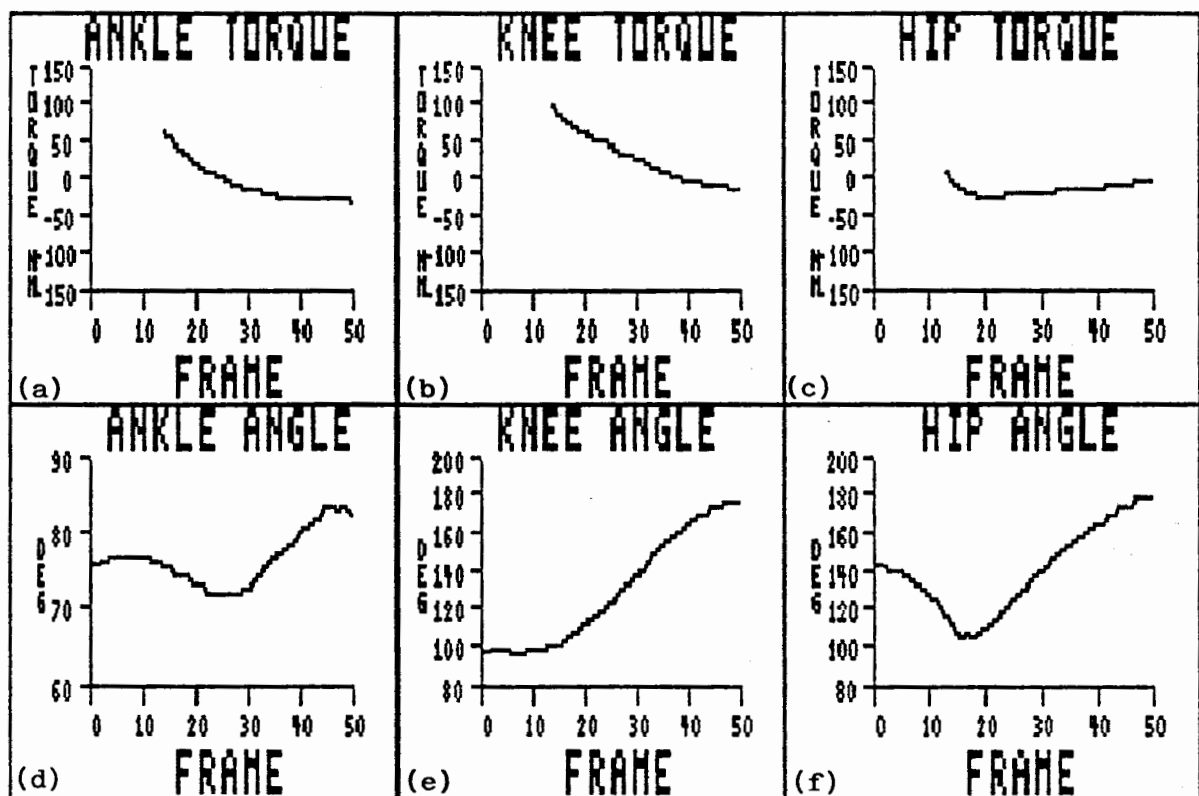


Figure 7.1
 Joint torque and angle against time plots for a 22 year old male subject obtained from the kinematic approach.

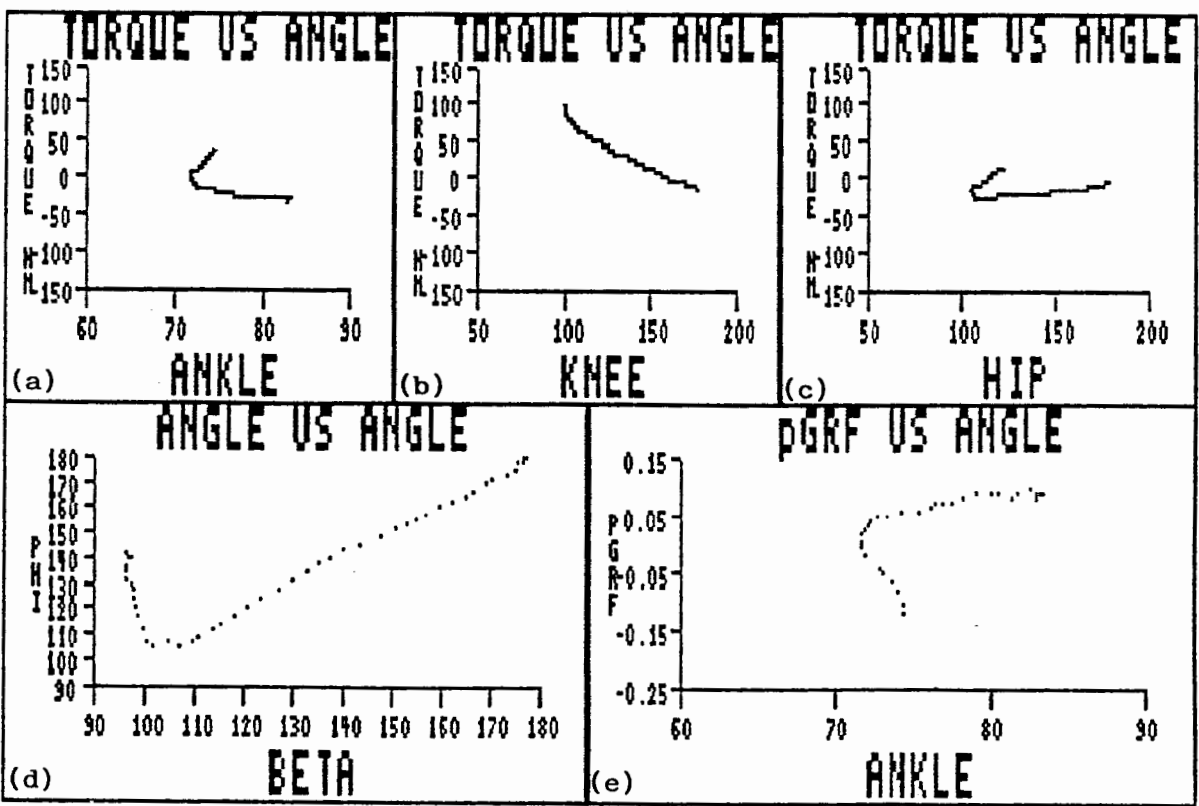


Figure 7.2
 Combination of joint torque, angle and pGRF plots for a 22 year old male subject obtained from the kinematics procedure.

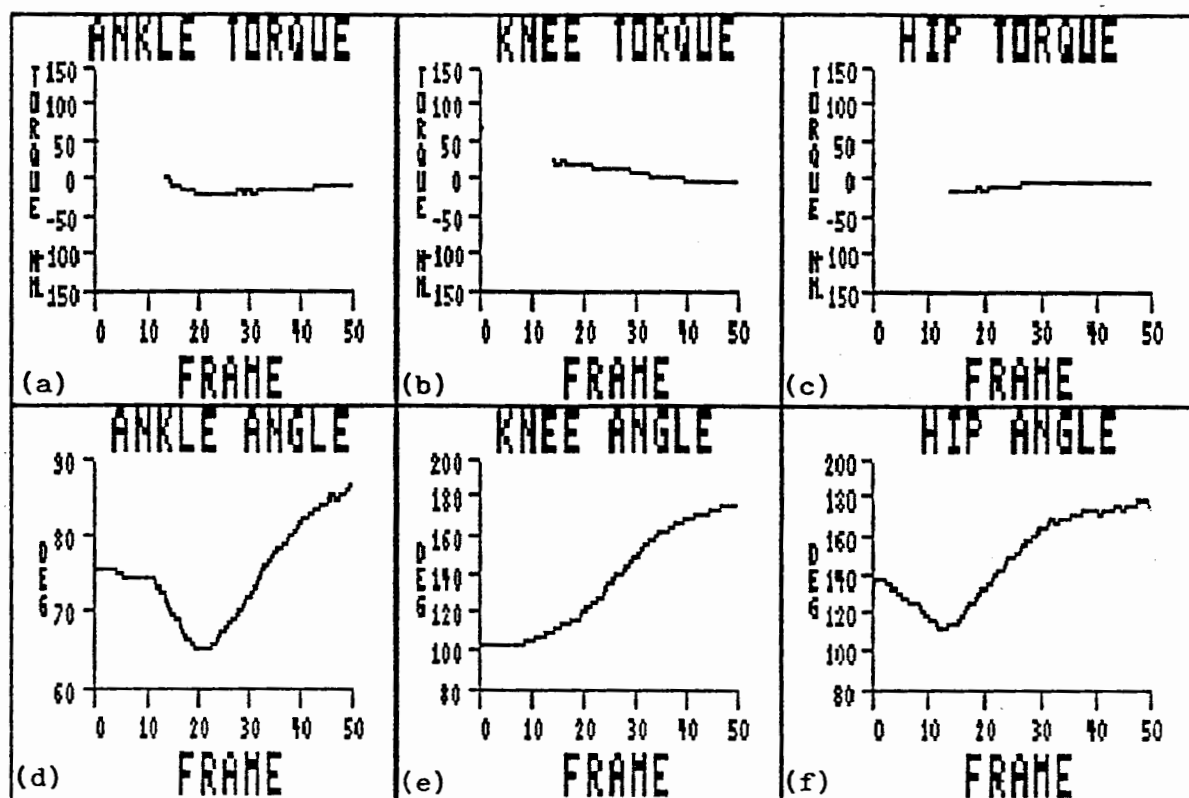


Figure 7.3
Joint torque and angle against time plots for a 20 year old female subject obtained from the kinematic approach.

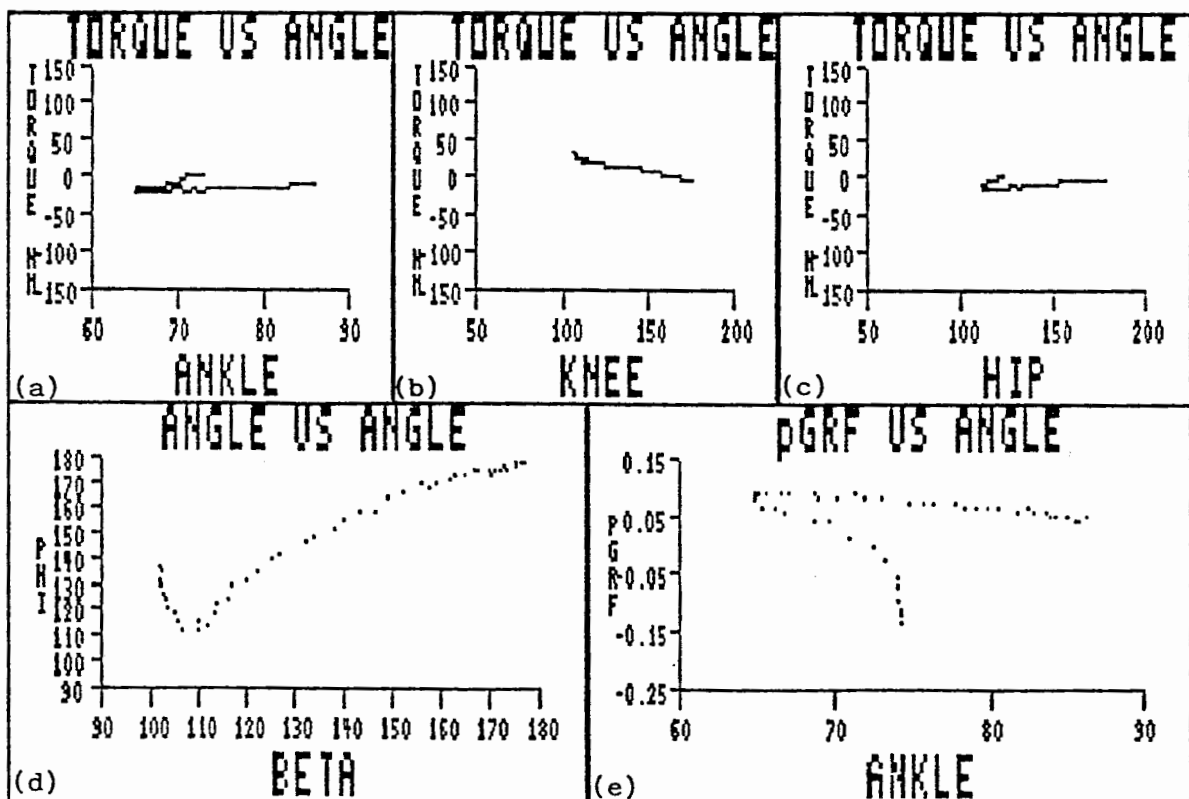


Figure 7.4
Combination of joint torque, angle and pGRF plots for a 20 year old female subject obtained from the kinematics procedure.

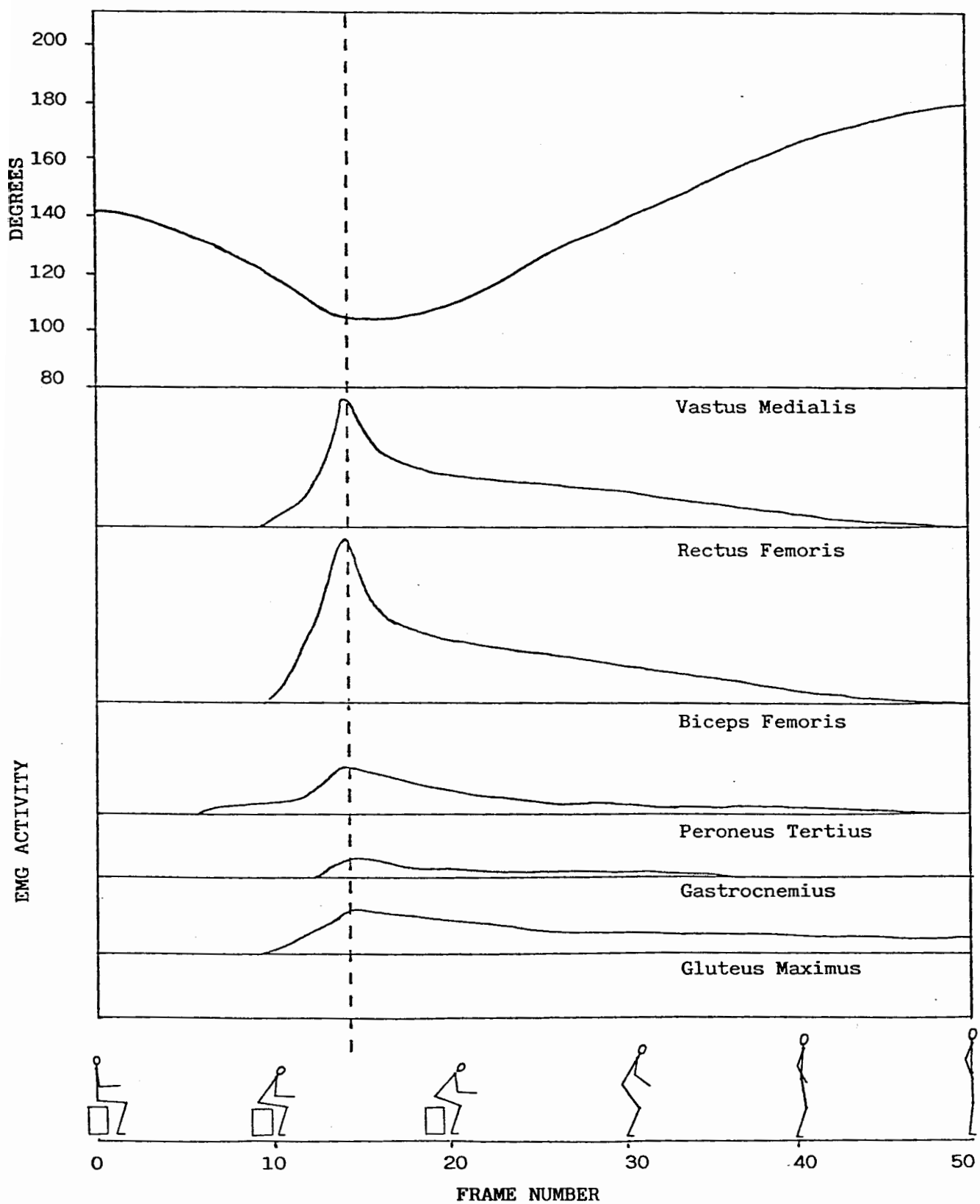


Figure 7.5
EMG activity for a typical subject
during the standing-up procedure.

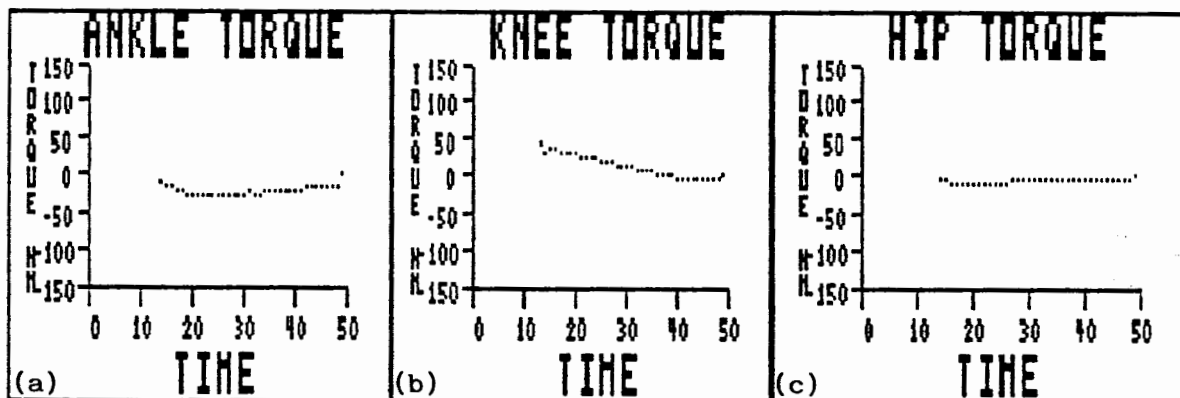


Figure 7.6
Joint torque against time plots for a 22 year old male
subject obtained from the kinetics procedure.

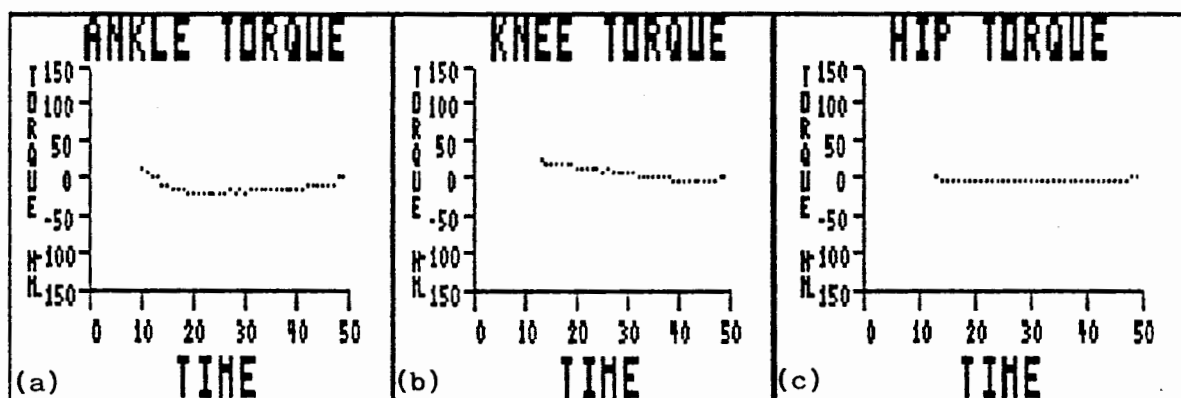


Figure 7.7
Joint torque against time plots for a 20 year old female
subject obtained from the kinetics procedure.

rapidly becomes positive and remains positive for the rest of the sequence. This indicates that the centre of gravity for the body remains in front of the ankle joint for almost the whole standing-up procedure.

Figure 7.5 is the general representation of EMG activity for all ten subjects used in this study. Stick figures indicate sequential body positions as they relate to the EMG activity. To enable comparison between the recorded EMG activity and the plots obtained, the graph of Hip Angle (ϕ) against Frame No. (time) is included.

Finally, Figures 7.6 and 7.7 are plots of joint torques and angles against time for the male and female subjects respectively obtained from the kinetics approach. In comparing Figures 7.6 and 7.7 to Figures 7.1 and 7.3, there are similarities in the general trends of the values obtained. For example, in the plot of ankle torque against time for the male subject, both torques decrease from a maximum, although the decrease is more evident in Figure 7.1 (a). The same applies for the knee and, to a lesser extent, hip torques. In comparing the actual values between the two methods, the maximum torque values are evidently lesser in the kinetic approach. This is probably because of the method used to obtain these values. In comparing the torque values between the male and female subjects for the kinetics, there is a definite difference in values, with the heavier male recording higher torques than the lighter female.

CHAPTER 8

SUMMARY, DISCUSSION, CONCLUSIONS AND RECOMMENDATIONS

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8.1 Summary

An extensive literature survey was conducted to try and establish the present state regarding mathematical models of the lower limb. This survey first covered mathematical models of the musculo-skeletal system in general with the emphasis being on those of the lower limb. A common problem experienced in modelling the lower limb is the indeterminacy of the muscle-force equations. Various attempts to solve this problem were presented in the literature, with varying degrees of success.

The second section of the literature survey dealt with the use of functional neuromuscular stimulation (FNS) to achieve standing-up in persons with spinal cord injuries. The problems experienced with the use of orthotic aids to achieve standing-up and even walking, for such people, were highlighted in that most people found them clumsy and difficult to use, and eventually discarded them. On the other hand, the main advantages of FNS are its easy implementation and use of the person's own muscles. FNS is also plagued by disadvantages; the main one being the choice of electrodes together with their inherent advantages and disadvantages.

A brief review of the muscles of the lower limb was given. Since surface electrodes were going to be used, only the superficial muscles were covered. This was followed by a thorough study of the electrophysiology of nerves and skeletal muscles. This chapter also included reviews of the surface electromyogram.

The mathematical model of the lower limb was developed viewing the human body as a system of four homogenous rigid bodies.

Knowing the various segmental weights and their respective centres of gravity together with the assumption that the standing-up process is a quasi-static one (Bajd et al, 1982), it was possible to develop the model by taking moments about a fixed point, the ankle joint, and equating them to zero. The problem was approached from both the Kinematic and Kinetic aspects. Various unknown variables such as the ankle, knee and hip angles were needed to complete the model, and a number of experiments were conducted to establish them. The results of these studies are presented in Chapter 4. These variables were implemented into the mathematical model and values for the hip, knee and ankle torques were subsequently obtained.

Electromyograms were taken in an attempt to establish which of the superficial muscles were active during the standing-up and sitting-down procedures. The results of these studies are presented in Chapter 5.

8.2 Discussion

For the standing-up procedure the initial movement is initiated by a forward pivoting of the upper body. As discussed in Chapter 5, this movement is most likely due to a combination of gravity and the eccentric "playing-out" action of the erector spinae muscles. This forward pivoting results in the reduction of the hip angle ϕ , which is evident in Figure 7.1 (f). ϕ continues to reduce to a minimum value at which lift-off occurs, after which the hip joint starts to extend. The forward pivoting of the upper body also causes the centre of gravity for the whole body (pGRF) to move forward as shown in Figure 7.2 (e). Figure 7.1 (d) also shows dorsiflexion of the ankle joint. The

initial dorsiflexion, ie. up to lift-off, would most likely be a result of the linkage system of the lower limb and not of muscle activity. However, after lift-off, the ankle joint still continues to dorsiflex. This is because the body is still moving forward as is evident from the plot of pGRF against alpha (Figure 7.2 (e)). This continues till pGRF is approximately zero, ie. acting through the lateral malleolus, after which the ankle begins to plantarflex.

The torque curves for the three joints of the lower limb are generally smooth; the stepped nature of the curves being a result of the curve plotting routine. As can be seen, there are no torque values for the three joints up to lift-off. Understandably, there would be no torque generated in either the knee or ankle joints as the movements there are passive. The hip is the only joint where it would be necessary for a torque to be generated as the upper body is not supported by the chair, assuming the person is not leaning against the back-rest. This torque could be obtained by taking moments about the hip joint for the upper torso only. Hip flexion, however, is normally obtained by using the upper limbs in conjunction with gravity and the erector spinae muscles. Consequently there would be no need to electrically stimulate either the hip flexor or erector spinae muscles to initiate hip flexion. This is also evident as there was no recorded EMG activity in the rectus femoris muscle till just before lift-off. Just after lift-off there was a slight positive torque which immediately thereafter became negative and increased to a maximum shortly afterwards. Just prior to lift-off, a rapid increase in the EMG's of rectus femoris, vastus medialis, and, to a lesser extent, biceps femoris, was observed. These large, rapid increases are in response to the sudden

transfer of body weight from the chair to the lower limbs. The activity in biceps femoris also show that there was some eccentric action to assist in stabilising of the hip and knee joints.

The knee joint torque started off with a maximum which was also accompanied by a rapid increase in the EMG activity, in both the rectus femoris and vasti medialis muscles, peaking at lift-off with the former showing a higher maximum. These large increases were also in response to the sudden transfer of body weight from the chair to the lower limbs. As for the hip joint, there was also an accompanying increase in EMG activity prior to lift-off, indicating eccentric contraction to stabilise the knee joint.

The ankle joint torque at lift-off was positive indicating that initially, dorsiflexion accompanied the forward movement of pGRF as discussed above. The torque then became negative, indicating plantarflexion of the ankle joint which corresponds to the increasing ankle angle shown in Figure 7.1 (d).

For the rest of the procedure, as the knee and hip joints approach complete extension, joint torques reduce smoothly to approximately zero. The EMG activity of rectus femoris, vastus medialis and biceps femoris also followed the same trend with essentially no activity at the end of the movement. This indicates an easy standing position. In the case of the ankle joint, both the joint torque and gastrocnemius muscle activity leveled off but did not reduce to zero, indicating that the upright standing posture was maintained by the ankle plantarflexor muscles.

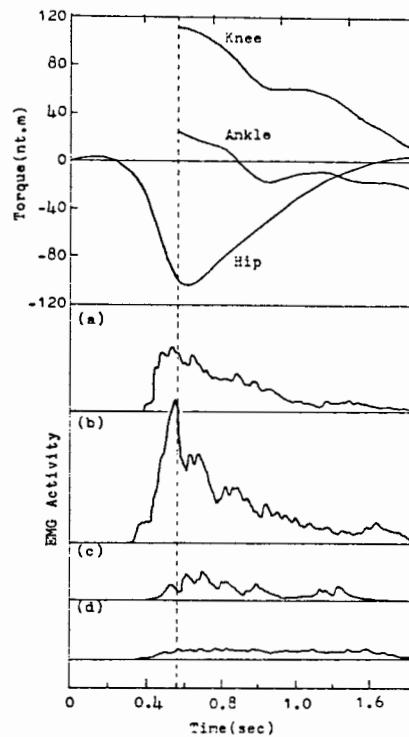


Figure 8.1

A reprint from Kelley et al, (1976, Fig. 3, pg 130) showing the relationship between body position, joint torques, and EMG activity for a typical subject.

In comparing the torque/time plots with those of Kelley et al, (1976), reproduced in Figure 8.1, a similarity in the general trends can be seen. Initially there is a negative hip torque which reduces to zero. Both the knee and ankle joint torques start off with a maximum and then reduce to a minimum. The actual torque values obtained however, do not correspond. The maximum knee torque, for example, is off by a factor of about 4. Unfortunately no reference as to their subjects' weight or sex is given thus not enabling any comparison between the torque values obtained. A comparison of the results obtained with those of Bajd et al, (1982), is more favorable in that, although no reference as to the subjects' sex, age or weight is given, the results obtained for the maximum knee torque correlate well. Bajd et al, (1982) quoted a value of 30Nm whereas the maximum

knee torque obtained for the female and male subjects, occurring at approximately frame 14, is 25Nm and 35Nm respectively.

The above discussion has, on the whole, been based on the results, with regards to the torque values etc., of the male subject. Comparing the two sets of values, it can be seen that the above discussion does apply to the results obtained for the female subject. An interesting deviation is in the minimum value obtained for the ankle angle with a difference of about ten degrees. A possible reason for this is that she was more supple than the male subject. There were no major differences noted in either the torque or motion patterns obtained for all ten subjects. The actual torque values however, definitely varied according to size, with higher values obtained for the heavier subjects. No obvious sex-related trends in either the torque, motion or EMG patterns were noted, indicating that only the anthropometrical data of a spinal cord injured person needs to be accounted for when determining the torque values required for the standing-up and sitting-down procedures.

Figures 7.6 and 7.7 are plots of joint torques against time obtained from the kinetics part of the simulation model. There is some similarity between the shapes of the curves obtained and those obtained from the kinematics procedure. The torque values obtained are, however, generally lower. As previously mentioned, this part of the program did not work out as planned, as the data obtained from the forceplate and "micronEye" did not correlate exactly. The reason was the differences in the sampling frequencies of the two systems. Using the data obtained from the "micronEye" and the calculated pGRF values, resulted in the hip, knee and torque versus time plots shown in Figures 7.6 and 7.7.

The reason for including this second option was to increase the versatility of the model. The operator could then run the program by either entering data obtained from the "micronEye" (or similar digital system), or from a forceplate, to predict the joint torques generated whilst standing-up or sitting-down.

8.3 Conclusions

8.3.1 Conclusions relating to the mathematical model

A mathematical model which predicts the torques generated at the hip, knee and ankle joints, during the standing-up process, has been developed. According to Petrofsky et al, (1986) the sitting-down process is the reversal of standing-up. This was found to be the case using EMG (Chapter 8) and also "micronEye" studies (Appendix E).

The model requires the input of various anthropometrical data, together with data either of the various hip, knee and ankle angles, or of the position of the ground reaction force, with respect to the lateral malleolus, together with the corresponding ankle angles. In its present form, the model requires a certain level of experimental expertise to operate, in that either "micronEye" or forceplate data is required to be inputted. This could be overcome by carrying out extensive experimental work to obtain data, perform a statistical analysis of the data and thus obtain a standard set of data which could be stored on disk and called up whenever required (Nuzik et al, 1986). The actual model itself is easy to operate with clear concise instructions. Included is a help file to aid in the gathering of anthropometrical data.

Noting that the torques obtained are, in general, in agreement with those of previous researchers, it can be concluded that the model does present acceptable torque values for the standing-up and sitting-down procedures.

8.3.2 Conclusions relating to the standing-up model

To establish a standing-up model in paraplegics one must employ the principles used in normal standing-up procedures. This, on the whole, is a reasonable assumption. Consequently, experimental work was performed on able-bodied people to establish an acceptable standing-up sequence. The data obtained using the "micronEye" compares favorably with that of Nuzik et al, (1986) who did extensive work in establishing their sit-to-stand movement pattern. The data obtained for the EMG studies also compared favorably with that of Kelley et al's, (1976), except for a discrepancy in the activity of the gluteus maximus muscle. However, the results obtained were also in agreement with Basmajian, (1978). Soderberg and Dostal (1978) concluded that the main hip extensor muscle for standing-up is the gluteus medius instead of gluteus maximus.

It can thus be concluded that the muscle-activity sequence depicted in Figure 7.5, and the hip, knee and ankle angles, do depict an acceptable standing-up sequence.

It must be noted, however, that the act of normal standing-up and sitting-down is extremely complex and that the simple model, presented in this thesis, right through to the extremely complex models of human movement, have all fallen short of exactly modelling these movements. Exact modelling is further

complicated by the variability in muscle geometry from one person to another. Thus the restoration of completely normal standing-up and sitting-down in spinal cord injured people is an extremely complex and formidable task.

8.4 Recommendations for future work

It has been concluded that the mathematical model developed can be used successfully to predict torques generated at the hip, knee and ankle joints during the standing-up and sitting-down procedures. However, although successful, the model and methods used to demonstrate this conclusion are not optimal. Thus, further study must be directed into this field before a final system which can ultimately be linked up to FNS control, can be defined. Recommendations regarding the directions of such future work are presented in this section. These are divided into two categories, namely those dealing with the model itself, and those dealing with the establishment of a standing-up sequence.

(1) Investigation of an alternative method to obtain hip, knee and ankle angle data. The present system makes use of a digital camera to capture data. Goniometric equipment may be however, more useful. It would be appropriate to determine whether this system is in fact superior to the "micronEye" system.

(2) Investigation of an alternative modelling approach. It has been noted that the major obstacle regarding mathematical modelling of the lower limb was the one of indeterminacy. This came about because of the numerous muscles in the lower limb. This could be overcome by the fact that, for this project, only the surface muscles are being stimulated

electrically to achieve standing-up and sitting-down. This greatly reduces the number of unknown equations, possibly to a more manageable number. Thus more research needs to be directed in this area and then compared to the present system for analysis.

(3) Obtaining a standard set of angle data. It has been noted above that a standard set of angle data will greatly enhance the present mathematical model. Thus more work needs to be directed at obtaining such data. It is envisaged that the model will be used to predict the torques generated for persons of different age, sex and anthropometrical form. Consequently emphasis must be placed in obtaining standard sets of data which will cover these differences.

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DERIVATION OF TORQUE EQUATIONS

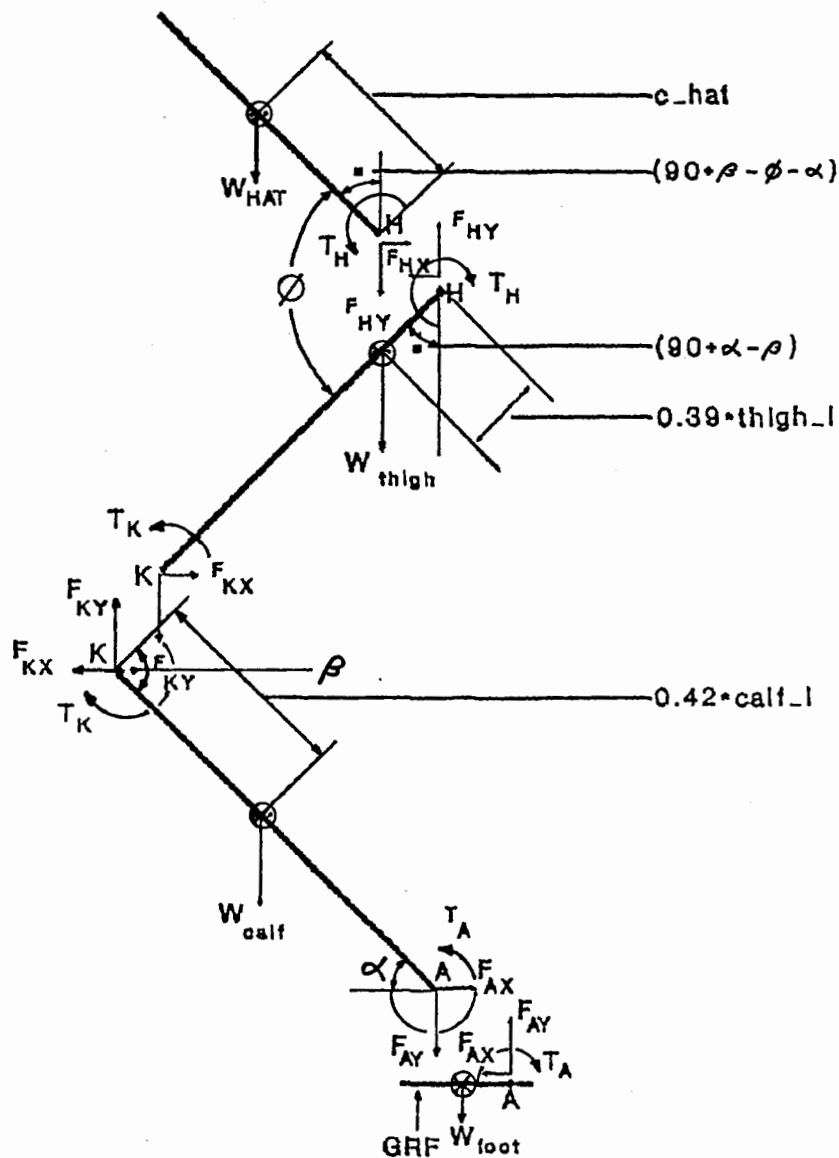


Figure A.1
Combination of joint reaction and muscle force components
into resultant components at the joints.

Since movement for the body has been restricted to a single X-Y plane, three equations of equilibrium can be derived by summing the forces in the two orthogonal directions of the X-Y plane, and by summing the moments of force about a given point, per segment. Referring to Figure A.1, the three equations of equilibrium for the foot, calf and thigh were obtained as follows:

Foot

Sum of the forces in the horizontal (x plane) and vertical (y plane) planes are zero for equilibrium.

$$\Sigma F = 0 :$$

$$-F_{ax} = 0$$

$$F_{ay} + GRF = W_f$$

$$\text{ie: } F_{ay} = -GRF + W_f$$

Sum of the moments of force about the ankle are zero for equilibrium.

$$\Sigma M_A = 0 :$$

$$-T_A - GRF * p_{GRF} + W_f * x_{foot} = 0$$

$$\text{ie: } T_A = -GRF + W_f * x_{foot}$$

The same conditions apply for the calf and thigh.

Calf

$$\Sigma F = 0 :$$

$$F_{ax} - F_{kx} = 0$$

$$\text{ie: } F_{ax} = F_{kx} = 0$$

$$-F_{ay} - W_c + F_{ky} = 0$$

$$\text{ie: } F_{ky} = F_{ay} + W_c$$

$$\Sigma M_K = 0 :$$

$$T_A + Fax * calf_1 * Sina - Fay * calf_1 * Cosa - W_c * 0.42 * calf_1 * Cosa - T_K = 0$$

$$T_K = T_A + Fax * calf_1 * Sina - Fay * calf_1 * Cosa - W_c * 0.42 * calf_1 * Cosa$$

Thigh

$$\Sigma F = 0 :$$

$$F_{kx} - F_{hx} = 0$$

$$\text{ie: } F_{kx} = F_{hx} = 0$$

$$-F_{ky} - W_t + F_{hy} = 0$$

$$\text{ie: } F_{hy} = F_{ky} + W_t$$

$$\Sigma M_H = 0 :$$

$$\begin{aligned} T_K + F_{kx} * thigh_1 * \cos(90 + \alpha - \beta) \\ + F_{ky} * thigh_1 * \sin(90 + \alpha - \beta) \\ + W_t * 0.39 * thigh_1 * \sin(90 + \alpha - \beta) - T_H = 0 \end{aligned}$$

$$\begin{aligned} \text{ie: } T_H = T_K + F_{kx} * thigh_1 * \cos(90 + \alpha - \beta) \\ + F_{ky} * thigh_1 * \sin(90 + \alpha - \beta) \end{aligned}$$

From the above equations for the foot, calf and thigh, the following equations were obtained:

$$Fax = F_{kx} = F_{hx} = 0 \quad (A1)$$

$$Fay = -GRF + W_f \quad (A2)$$

$$\begin{aligned} F_{ky} &= Fay + W_c \\ &= -GRF + W_f + W_c \end{aligned} \quad (A3)$$

$$\begin{aligned} F_{hy} &= F_{ky} + W_t \\ &= -GRF + W_f + W_c + W_t \end{aligned} \quad (A4)$$

$$T_A = -(GRF * p_{GRF}) + (W_f * x_{foot}) \quad (A5)$$

$$\begin{aligned} T_K &= T_A - (Fay * calf_1 * Cosa) \\ &\quad - (W_c * 0.42 * calf_1 * Cosa) \end{aligned} \quad (A6)$$

$$\begin{aligned} T_H &= T_K + (F_{ky} * thigh_1 * \sin(90 + \alpha - \beta)) \\ &\quad + (W_t * 0.39 * thigh_1 * \sin(90 + \alpha - \beta)) \end{aligned} \quad (A7)$$

APPENDIX B

DETERMINATION OF SEGMENT WEIGHTS AND CENTRE OF MASSES

B.1 Introduction

In the standing-up procedure the muscles acting across a joint generate a torque which opposes the torque generated by the body weight acting about that joint. To determine these torques thus requires knowledge of the weight of the various body segments (segmented weights) and their centres of gravity.

Chandler et al, (1975) presented anthropometric data on fourteen body segment masses, centres of mass and moments of inertia of six dismembered male cadavers. They also presented regression equations, based on total body mass, for calculating the fourteen segmented masses and principal moments of inertia.

The Chandler/Vaughan regression equations (Vaughan, 1980), which were derived using the anthropometric data of Chandler et al, (1975) and the related body segments from the study of Hanavan, (1964) are as follows:

Body segment mass :

$$M = a_1M + a_2L^3 + a_3$$

where : a_1 to a_3 are regression coefficients

M : total body mass

L : a dimension of a body segment based on the particular geometric shape chosen to represent the various body segments.

B.2 Determination of Segment Mass

The required anthropometric measurements of a subject were taken

for use in the Chandler/Vaughan regression equations and entered into the program (Procedure "anthrodata"). Table B.1 is a list of the anthropometric data taken. Measurements were based on descriptions of anthropometric dimensions taken from the work of Chandler et al, (1975). These descriptions have been included in a help file in the computer program.

The Chandler/Vaughan regression equations are as follows:

$$\begin{aligned} m(1) &= 0.0221 * M + 63.76 * \text{mass_ver} * \text{head} * \text{head} - 0.742 \\ m(2) &= 0.2791 * M + 413.6 * \text{torso} * \text{chest_d} * \text{chest_b} - 3.881 \\ m(3) &= 0.0096 * M + 14.51 * \text{hum_rad} * \text{biceps} * \text{biceps} + 0.812 \\ m(4) &= 0.0172 * M + 12.89 * \text{rad_sty} * \text{forearm} * \text{forearm} - 0.286 \\ m(5) &= 0.0023 * M + 1274 * \text{sty_met} * \text{hand_b} * \text{hand_d} - 0.038 \\ m(6) &= 0.1032 * M + 12.76 * \text{thigh_l} * \text{thigh_c} * \text{thigh_c} - 1.023 \\ m(7) &= 0.0226 * M + 31.33 * \text{calf_l} * \text{calf_c} * \text{calf_c} + 0.016 \\ m(8) &= 0.0083 * M + 254.5 * \text{foot_l} * \text{lat_mal} * \text{foot_b} - 0.065 \end{aligned}$$

where : $m(*)$ represents the masses, in kilograms, of the various body segments, being:

- $m(1)$: mass of the head
- $m(2)$: mass of the trunk
- $m(3)$: mass of the upperarm
- $m(4)$: mass of the forearm
- $m(5)$: mass of the hand
- $m(6)$: mass of the thigh
- $m(7)$: mass of the calf
- $m(8)$: mass of the foot

B.3 Determination of Center of Mass

Using the anthropometrical data from Table B.1, the various centres of masses for the body segments were calculated using the

NAME.....	DATE.....
DATE OF BIRTH.....	SEX.....
Anthropometric Measurement	Values
Total height.....	
Total body mass.....	
Mastoid to vertex length.....	
Head circumference.....	
Torso length.....	
Chest depth.....	
Chest breadth.....	
Ball of humerus to radiale length.....	
Ball of humerus to vertex length.....	
Biceps circumference.....	
Forearm circumference.....	
Stylian to metacarpal III length.....	
Radial to stylian length.....	
Hand breadth.....	
Hand depth.....	
Thigh length.....	
Mid-thigh circumference.....	
Calf length.....	
Calf circumference.....	
Lateral malleolus height.....	
Foot breadth.....	
Back of foot to lateral malleolus length.....	
Foot length.....	

Table B.1
Table of anthropometrical data required for model.

$P_{cg \text{ head}} = P_{vertex} + 0.66 * (P_{mastoid} - P_{vertex})$
 $P_{cg \text{ trunk}} = P_{chin} + 0.52 * (P_{pubic} - P_{chin})$
 $P_{cg \text{ upperarm}} = P_{shoulder} + 0.51 * (P_{elbow} - P_{shoulder})$
 $P_{cg \text{ forearm}} = P_{elbow} + 0.41 * (P_{wrist} - P_{elbow})$
 $P_{cg \text{ hand}} = P_{wrist} + 0.52 * (P_{pip3} - P_{wrist})$
 $P_{cg \text{ thigh}} = P_{hip} + 0.39 * (P_{knee} - P_{hip})$
 $P_{cg \text{ calf}} = P_{knee} + 0.42 * (P_{ankle} - P_{knee})$
 $P_{cg \text{ foot}} = P_{heel} + 0.44 * (P_{toe} - P_{heel})$

where : P_{vertex} etcetera, are the positions of the anatomical part, measured in meters from the base of the foot. The exception is in P_{heel} and P_{toe} which represents the position of the back of the heel and front of the first toe in an axis perpendicular to the above values. These are diagrammatically represented in Figure B.1.

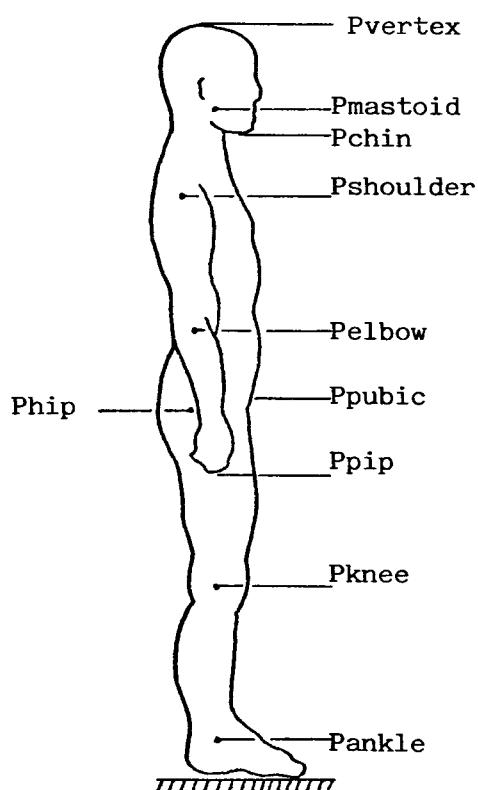


Figure B.1
 Diagrammatic representation of the model showing the anatomical positions.

APPENDIX C

DETERMINATION OF CENTRE OF MASS OF THE UPPER BODY

For the purposes of this model the head, upper limbs and trunk (HAT) were considered as one segment. It was thus necessary to calculate the centre of mass for HAT.

Figure C.1 illustrates, diagrammatically, the model standing in the sagittal plane with arms folded on the chest.

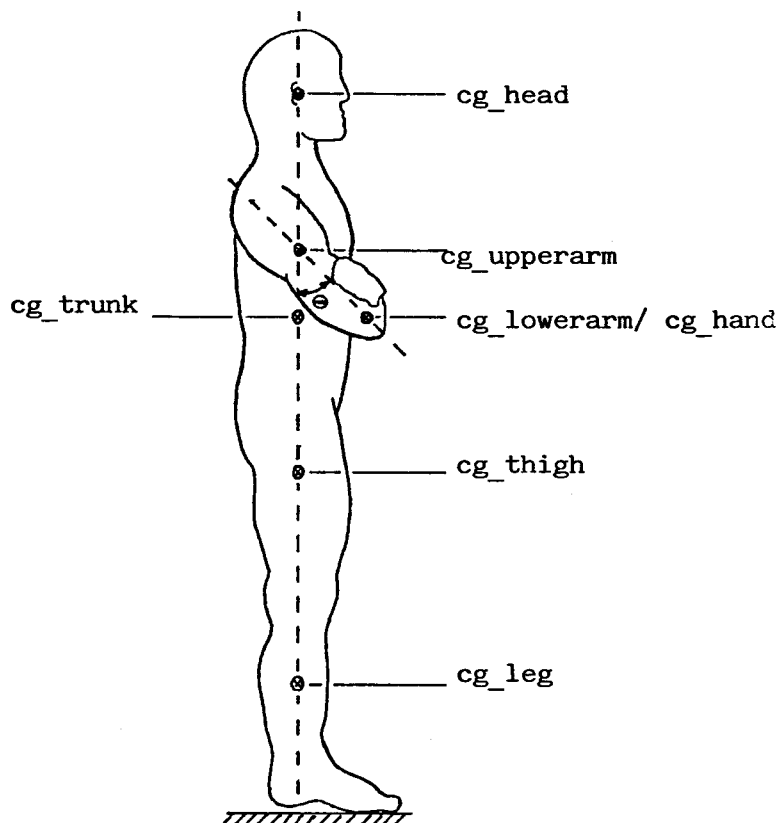


Figure C.1
Diagrammatical representation of model showing
the various segmental centres of masses.

Theta θ , is the angle subtended between a line drawn vertically down through the ball of the humerus and a line drawn through the ball of the humerus and the lateral epicondyle of the elbow joint. The various body segment lengths and their respective centres of mass were calculated in Appendix B.

$$\sin \theta = [0.5 * (\text{chest-d} + \text{forearm}/\pi)] / \text{hum_rad}$$

$$\text{Therefore } \theta = \arcsin((\text{chest_d} + \text{forearm}/\pi)/(2 * \text{hum_rad})) \quad (\text{C1})$$

$$\text{Pcg_head} = \text{tot_hgt} - 0.66 * (\text{mass_ver}) \quad (\text{C2})$$

$$\text{Pcg_trunk} = \text{lat_mal} + \text{calf_1} + \text{thigh_1} + 0.48 * \text{torso_1} \quad (\text{C3})$$

$$\text{Pcg_upper_arm} = \text{tot_hgt} - (\text{hum_ver} + 0.51 * \text{hum_rad} * \cos \theta) \quad (\text{C4})$$

$$\text{Pcg_lower_arm} = \text{tot_hgt} - (\text{hum_ver} + \text{hum_rad} * \cos \theta) \quad (\text{C5})$$

$$\text{Pcg_hand} = \text{tot_hgt} - (\text{hum_ver} + \text{hum_rad} * \cos \theta) \quad (\text{C6})$$

where: tot_hgt, lat_mal etcetera are the distances, in meters,
between the various body prominances.

The centre of gravity of HAT (cg_hat) was calculated by taking moments about the x axis, with zero being at the intersect between the perpendicular line through the lateral malleolus and the horizontal line at the base of the feet, (Figure C.1).

Sum of the moments about the x (horizontal) axis :

$$\begin{aligned} & m[1] * \text{cg_head} + m[2] * \text{cg_trunk} + 2 * (m[3] * \text{cg_upperarm} + \\ & m[4] * \text{cg_forearm} + m[5] * \text{cg_hand}) = m[9] * \text{cg_hat} \end{aligned} \quad (\text{C7})$$

From which :

$$\begin{aligned} \text{cg_hat} = & (1/m[9]) * (m[1] * \text{cg_head} + m[2] * \text{cg_trunk} \\ & + 2 * (m[3] * \text{cg_upperarm} + m[4] * \text{cg_forearm} \\ & + m[5] * \text{cg_hand}) \end{aligned} \quad (\text{C8})$$

Hence, from Figure C.1 :

$$\text{c_hat} = \text{cg_hat} - (\text{lat_mal} + \text{calf_1} + \text{thigh_1}) \quad (\text{C9})$$

APPENDIX D

DIGITAL CAMERA SYSTEM

The system used to capture data was a "micronEye" digital camera (Micron Technology, 2805 East Columbia Road, Boise, Idaho, USA) which interfaces directly with most personal micro-computers. The camera has a 10mm wide lens and is housed in a specially constructed casing together with a spotlight and reflective mirror as shown in Figure D.1. Retro reflective material (Scotchlite from 3M) was affixed to table tennis balls which were used to mark the subject being tested/monitored. Software, written in TURBO PASCAL 4, was developed by Vaughan (Nissan et al, 1988(b)) which enabled 2-D data to be captured on disk, processed and displayed to screen. This digitized data which gave the X,Y co-ordinates of each point could not be used in this form.



Figure D.1.
The "micronEye" camera with computer.

Subject = Beth Harley
Date = 31 July 1988
Comment = standing

Marker 1	Marker 2	Marker 3	Marker 4	Marker 5
X Y	X Y	X Y	X Y	X Y
673,1603	619,1764	260,1941	361,2326	311,2326
671,1603	617,1760	260,1941	361,2326	311,2326
666,1601	617,1758	260,1941	361,2326	311,2326
656,1600	612,1764	260,1941	361,2326	311,2326
645,1600	611,1759	256,1942	361,2326	311,2326
635,1598	606,1756	256,1942	363,2328	313,2328
624,1598	604,1756	255,1946	361,2326	311,2326
614,1597	600,1754	255,1946	361,2326	311,2326
600,1593	595,1752	255,1946	363,2328	313,2328
583,1592	590,1745	255,1946	363,2328	313,2328
565,1591	583,1743	255,1946	363,2328	313,2328
534,1577	568,1737	249,1947	363,2328	313,2328
506,1568	545,1715	243,1946	363,2328	313,2328
484,1549	526,1712	232,1951	362,2330	312,2330
459,1533	509,1687	223,1949	363,2328	313,2328
441,1516	489,1681	215,1951	362,2330	312,2330
421,1506	465,1668	200,1953	362,2330	312,2330
400,1489	445,1657	196,1953	362,2330	312,2330
387,1475	426,1652	189,1953	362,2330	312,2330
372,1458	413,1631	187,1955	362,2330	312,2330
359,1444	398,1625	187,1955	362,2330	312,2330
352,1430	386,1605	187,1951	362,2330	312,2330
345,1417	379,1600	189,1949	362,2330	312,2330
336,1401	362,1578	198,1946	363,2328	313,2328
328,1384	355,1570	198,1937	363,2328	313,2328
324,1373	350,1556	209,1937	361,2326	311,2326
322,1361	339,1547	209,1927	363,2328	313,2328
317,1347	332,1542	218,1925	363,2328	313,2328
310,1342	329,1532	226,1923	363,2328	313,2328
311,1333	322,1519	232,1918	365,2326	315,2326
306,1325	319,1514	243,1918	368,2326	318,2326
307,1321	316,1511	255,1909	368,2326	318,2326
304,1317	317,1506	260,1904	368,2326	318,2326
305,1313	317,1494	266,1899	369,2324	319,2324
307,1312	318,1492	275,1897	368,2326	318,2326
308,1306	317,1491	280,1897	368,2326	318,2326
308,1306	319,1489	288,1895	371,2324	321,2324
311,1305	317,1488	294,1895	371,2324	321,2324
311,1305	321,1486	300,1895	371,2324	321,2324
312,1302	322,1488	309,1893	371,2324	321,2324
314,1301	330,1486	314,1893	371,2324	321,2324
315,1299	330,1486	317,1891	371,2324	321,2324
320,1300	331,1484	322,1890	371,2324	321,2324
322,1299	332,1486	325,1893	371,2324	321,2324
322,1299	336,1484	328,1895	371,2324	321,2324
324,1297	334,1484	334,1890	371,2324	321,2324
322,1299	331,1484	334,1890	372,2322	322,2322
326,1296	334,1484	339,1891	373,2323	323,2323
326,1296	334,1484	341,1889	373,2323	323,2323
326,1296	338,1483	345,1890	373,2323	323,2323

Table D.1
List of "micronEye" for a 20 year old female subject.

Subject = Beth Harley
Date = 31 July 1988

Frame 1	Angle 1 :	135.100	Angle 2 :	101.625	Angle 3 :	75.278
Frame 2	Angle 1 :	135.883	Angle 2 :	102.201	Angle 3 :	75.278
Frame 3	Angle 1 :	134.597	Angle 2 :	102.497	Angle 3 :	75.278
Frame 4	Angle 1 :	131.837	Angle 2 :	102.075	Angle 3 :	75.278
Frame 5	Angle 1 :	129.349	Angle 2 :	101.918	Angle 3 :	74.614
Frame 6	Angle 1 :	128.562	Angle 2 :	102.381	Angle 3 :	74.386
Frame 7	Angle 1 :	125.573	Angle 2 :	102.806	Angle 3 :	74.312
Frame 8	Angle 1 :	124.373	Angle 2 :	103.380	Angle 3 :	74.312
Frame 9	Angle 1 :	121.422	Angle 2 :	103.735	Angle 3 :	74.084
Frame 10	Angle 1 :	118.397	Angle 2 :	105.067	Angle 3 :	74.084
Frame 11	Angle 1 :	115.135	Angle 2 :	105.866	Angle 3 :	74.084
Frame 12	Angle 1 :	111.331	Angle 2 :	106.584	Angle 3 :	73.253
Frame 13	Angle 1 :	112.313	Angle 2 :	109.944	Angle 3 :	72.526
Frame 14	Angle 1 :	114.702	Angle 2 :	110.096	Angle 3 :	71.050
Frame 15	Angle 1 :	114.315	Angle 2 :	112.133	Angle 3 :	69.681
Frame 16	Angle 1 :	118.534	Angle 2 :	113.331	Angle 3 :	68.787
Frame 17	Angle 1 :	121.975	Angle 2 :	113.895	Angle 3 :	66.791
Frame 18	Angle 1 :	124.735	Angle 2 :	115.993	Angle 3 :	66.171
Frame 19	Angle 1 :	129.318	Angle 2 :	117.124	Angle 3 :	65.354
Frame 20	Angle 1 :	131.785	Angle 2 :	119.994	Angle 3 :	64.894
Frame 21	Angle 1 :	135.237	Angle 2 :	122.252	Angle 3 :	64.894
Frame 22	Angle 1 :	139.142	Angle 2 :	125.244	Angle 3 :	65.167
Frame 23	Angle 1 :	140.900	Angle 2 :	127.021	Angle 3 :	65.583
Frame 24	Angle 1 :	147.465	Angle 2 :	132.526	Angle 3 :	66.573
Frame 25	Angle 1 :	148.635	Angle 2 :	133.891	Angle 3 :	67.074
Frame 26	Angle 1 :	151.657	Angle 2 :	138.404	Angle 3 :	68.682
Frame 27	Angle 1 :	155.900	Angle 2 :	140.082	Angle 3 :	68.952
Frame 28	Angle 1 :	158.851	Angle 2 :	143.580	Angle 3 :	70.190
Frame 29	Angle 1 :	159.701	Angle 2 :	146.612	Angle 3 :	71.299
Frame 30	Angle 1 :	163.737	Angle 2 :	149.131	Angle 3 :	71.906
Frame 31	Angle 1 :	165.669	Angle 2 :	152.481	Angle 3 :	73.059
Frame 32	Angle 1 :	168.680	Angle 2 :	156.136	Angle 3 :	74.852
Frame 33	Angle 1 :	168.049	Angle 2 :	157.634	Angle 3 :	75.730
Frame 34	Angle 1 :	169.006	Angle 2 :	159.212	Angle 3 :	76.376
Frame 35	Angle 1 :	170.385	Angle 2 :	161.694	Angle 3 :	77.803
Frame 36	Angle 1 :	172.066	Angle 2 :	163.191	Angle 3 :	78.422
Frame 37	Angle 1 :	172.432	Angle 2 :	164.849	Angle 3 :	79.105
Frame 38	Angle 1 :	175.042	Angle 2 :	166.652	Angle 3 :	79.834
Frame 39	Angle 1 :	173.930	Angle 2 :	167.604	Angle 3 :	80.566
Frame 40	Angle 1 :	174.704	Angle 2 :	169.818	Angle 3 :	81.761
Frame 41	Angle 1 :	172.856	Angle 2 :	170.077	Angle 3 :	82.392
Frame 42	Angle 1 :	173.647	Angle 2 :	170.972	Angle 3 :	82.838
Frame 43	Angle 1 :	175.054	Angle 2 :	172.324	Angle 3 :	83.593
Frame 44	Angle 1 :	175.949	Angle 2 :	172.828	Angle 3 :	83.871
Frame 45	Angle 1 :	174.671	Angle 2 :	173.154	Angle 3 :	84.269
Frame 46	Angle 1 :	176.896	Angle 2 :	175.070	Angle 3 :	85.070
Frame 47	Angle 1 :	177.551	Angle 2 :	175.293	Angle 3 :	84.964
Frame 48	Angle 1 :	178.429	Angle 2 :	176.304	Angle 3 :	85.507
Frame 49	Angle 1 :	178.663	Angle 2 :	176.776	Angle 3 :	85.745
Frame 50	Angle 1 :	177.238	Angle 2 :	177.237	Angle 3 :	86.264

Table D.2
List of "micronEye" angles for a 20 year old female subject.

The digitized data was converted to angles using the Procedure 'calcmicroneyeangles' and Functions 'angval' and 'len' and could be listed either to the screen or printer.

Tables D.1 and D.2 are tables of digitized microneye data and the corresponding angles alpha, beta and phi. These angles were then read from a file into the the program thus enabling the hip, knee and ankle torques to be calculated.

Whilst data was being captured by the "micronEye" digital camera, the process was simultaneously being videoed. This was to ease the selection of the best standing up sequence for analysis in collaboration with various professional personell in the field of gait analysis.

Four markers were positioned onto the lateral malleolus, lateral epicondyl of the knee, greater trochanter and on the lateral trunk superior to the iliac crest. The subject being tested was first shown a video recording of the standing up procedure. She/he then carried out a number of trials to gain confidence and also practice the motion before data was captured. Data was captured for subjects rising from the seated to standing position. The sampling frequency was 13 Hz and 50 frames of data was captured thus allowing 3.8 secs for the whole sequence. Stick figure diagrams, Figure D.2. graphically depicts the standing-up procedure. The lines were achieved by joining the lateral trunk, greater trochanter, lateral epicondyl and lateral maleolus markers respectively.

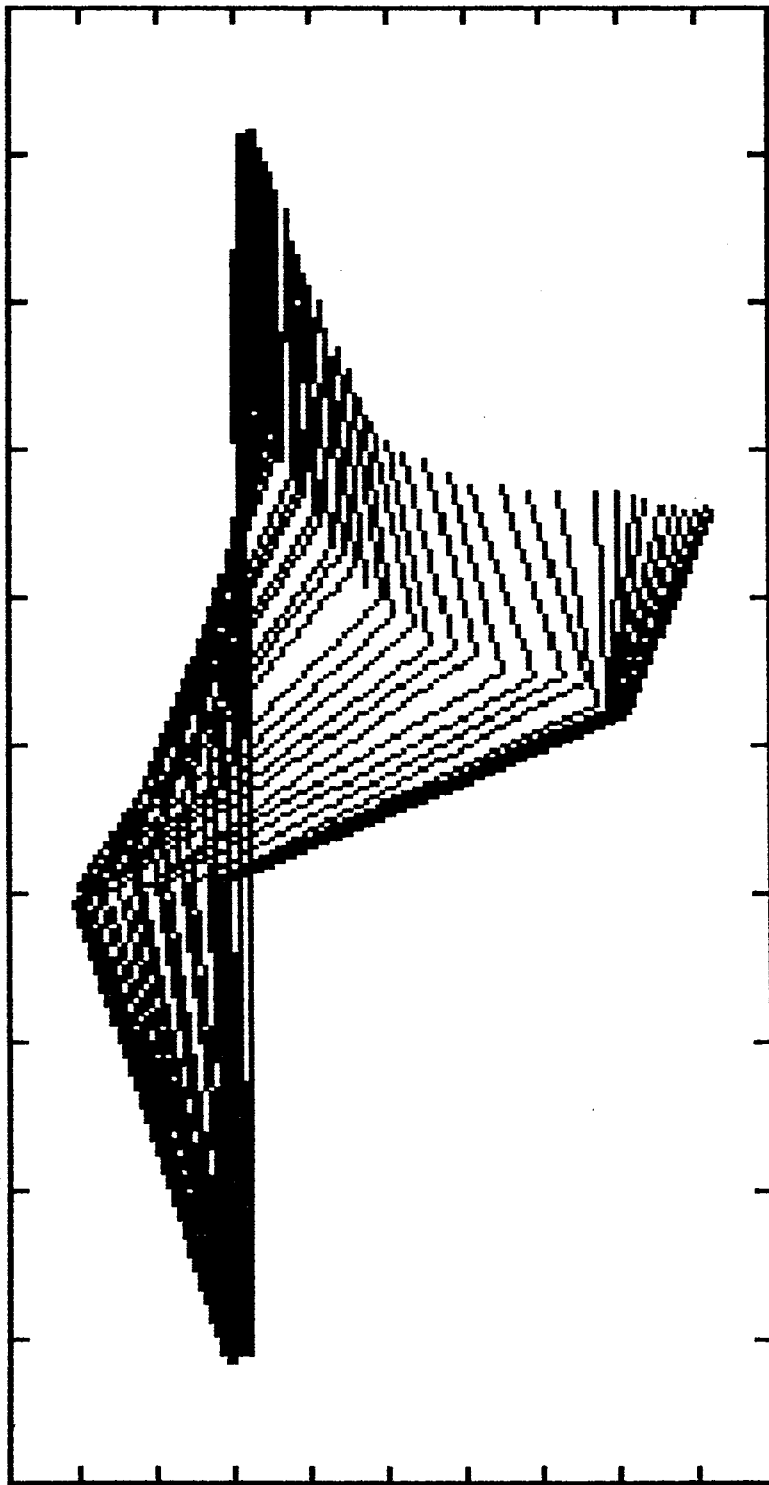


Figure D.2
Stick figure diagram.

APPENDIX E

DERIVATION OF EQUATION RELATING BETA TO PHI

From the work of Kelley et al, (1976), knee angle versus hip angle plots demonstrated generally linear characteristics (Figure E.1). This linear relationship between beta and phi was also achieved using the "micronEye" (Figure E.2), for from lift-off to upright standing. Curve (b) of Figure E.1 was initially used to obtain the relationship between knee angle and hip angle. The method of least squares (Holman and Gajda, 1978) was used to obtain an equation to describe curve (b) which related the knee angles (beta) to the hip angles (phi). The values for the knee angles (independent variable) and the corresponding hip angles (dependent variable) were obtained from curve (b) of Figure E.1.

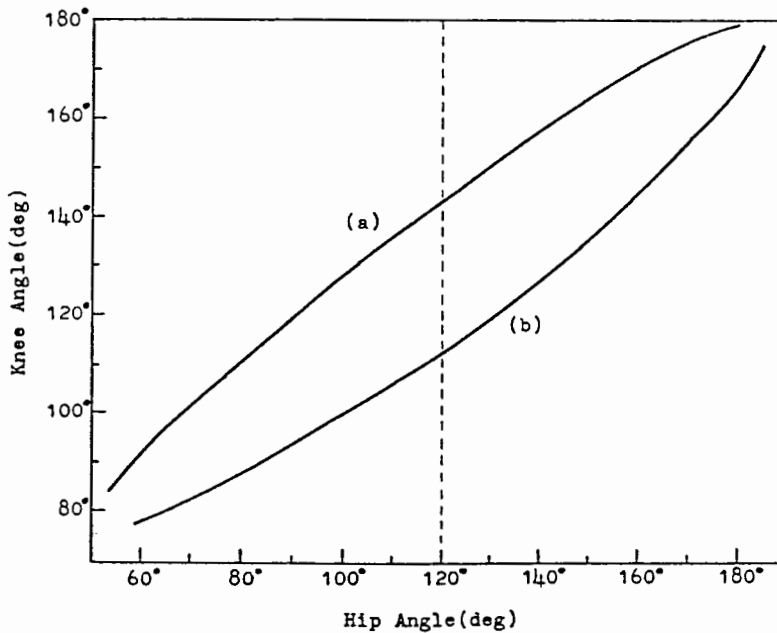


Figure E.1

A reprint from Kelley et al (1976, Fig 4, pg 132) of the plot of knee angle versus hip angle for an atypical case (a), and a typical case (b).

Inputting these values into a program, "Curvefit" written to obtain the constants a, b, c and d of the equation:

$$y = a + bx + cx^2 + dx^3$$

resulted in Equation E1.

$$\beta = 30.0149834 + 1.088078633 * \phi - 0.006618382 * \phi^2 + 0.000026807 * \phi^3 \quad (E1)$$

However, on implementing this equation into the program "Restand", (procedure 'Kinetics' and function 'beta') erroneous values for the torque were obtained in comparison to those obtained using the procedure 'Kinematics'.

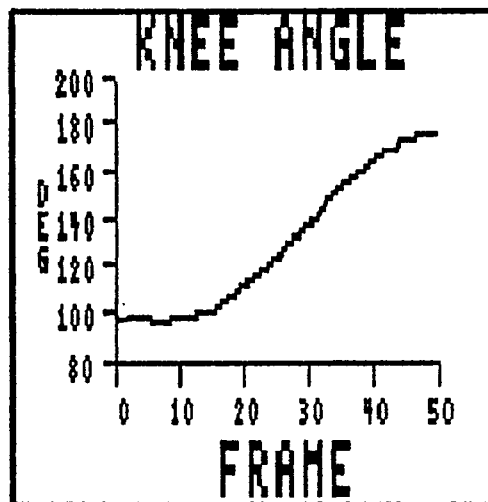


Figure E.2
Plot of knee angle (beta), versus hip angle (phi).

On careful inspection of Kelley et al's (1976) data, and that obtained from the "micronEye", revealed a large discrepancy between the hip and knee angles. The video recordings taken of the subjects performing the standing up procedure were subsequently then studied to specifically observe the variations in the hip and knee angles. The results of the study agreed with the data obtained using the "micronEye". The probable reason for the discrepancy between the two studies was the difference in chair heights. For Kelley et al's (1976) study the subject's thigh appears to be horizontal to the ground. This was not the

case for this study. The height of the chair was chosen to comply with the seat height for wheelchairs complying to the South African Bureau of Standards, (SABS).

Consequently "micronEye" data was used to obtain a mathematical relationship between the knee and hip angles using the program "curvefit" and data from a subjects' file, BETH7E. Substituting these values into "curvefit" resulted in Equation E2.

$$\beta = -19369.400377 + 725.30394266 * \phi - 10.650018262 * \phi^2 \\ + 0.077098914 * \phi^3 - 0.000275425 * \phi^4 + 0.0000003892 * \phi^5 \text{ (E2)}$$

On using this equation in the program "restand", acceptable torque values were obtained.

APPENDIX F

SEGMENTED FORCE PLATE

The segmented force plate (SFP), Figure F.1, used was one which was originally devised and built by Solomon and Manley (Manley and Solomon, 1979) and further developed by Nissan et al (1988 (b)). The system is installed in a walkway laboratory at the Princess Alice Orthopaedic Hospital in Retreat, Cape Town, to enable dynamic investigation of the foot.



Figure F.1
General view of the segmented force plate.

The SFP is made up of 16 transparent perspex beams mounted at each end by strain gauges which record only vertical forces when stood on. The strain gauges are linked up to a multiplexed strain-gauge amplifier which is used to calculate the resultant force and its point of application for each beam. Two video cameras are used in conjunction with the SFP. One captures the medical aspects of the foot and the position of the swinging foot. The other produces an image of the plantar surface of the foot whilst in contact with the SFP. Information from the strain

guage amplifier is fed into a personal computer which, together with a specially designed package enables one to analyse the data captured by the force transducers. The sampling frequency of the SFP is 50 Hz.

There are two programs written to capture data from the SFP. The first, DATAGAT, gathers data from the SFP and saves it, together with personal data and the heel, toe X,Y coordinates, onto a floppy disc. The second, DATAPRESS, enables one to present and analyse the results. For further details concerning the software, the reader is referred to Nissan et al, (1988 (a)).

The subject under study, (Figure F.2) placed one foot centrally on the force plate with the other foot next to it but not on the force plate, and then stood up, as described in Appendix D.



Figure F.2

Subject showing foot placement on the forceplate.

APPENDIX G

DESCRIPTION OF THE MAIN PROGRAM SUBROUTINES

G.1 The kinematics and kinetics subroutines

Figure G.1 is the flowchart for the kinematics subroutine. Once the kinematics subroutine has been opened two menu lines are written on the screen. The first line is the name of the subroutine, 'KINEMATICS PROCEDURE', the second is a list of operations. The subroutine then executes code in a loop until the quit option is chosen. If one of the command keys has been pressed, then the command is processed by calling the appropriate subroutines from within the CASE statement. Once the exit option has been chosen the graphics screen is closed and the text mode restored.

Figure G.2 shows the flowchart for the kinetics subroutine. As can be seen this subroutine follows exactly the same format as the kinematics subroutine and therefore warrants no further discussion.

G.2 The "anthrodata" subroutine

Figure G.3 is the flowchart for the anthrodata subroutine. As for the previous subroutines, once the anthrodata subroutine has been opened, two menu lines are written to screen. The first being the name of the subroutine, 'ANTHRODATA PROCEDURE', and the second is a list of operations for the operator to from choose. The subroutine then executes code in a loop until the quit option is chosen. The following is a brief description of the various subroutines which can be called from within the CASE statement by

pressing the appropriate command key.

G.2.1 Help file

This file contains information regarding the anthropometrical data. It consists of a number of screens, through which the operator can either page down or up, with descriptions of the anthropometrical terms used in the anthrodata file, and how to measure them.

G.2.2 Load anthrofile

This subroutine allows the operator to select a file containing the subject's anthropometrical data to be used in either the kinematics or kinetics subroutines.

G.2.3 Save anthrofile

This subroutine allows the operator to save the subjects' anthropometric data file with the option to rename it.

G.2.4 Edit anthrofile

This subroutine allows the operator to either edit existing files or to open a new data file. A menu appears on the screen with instructions as to how to edit the file. On completion of editing, the quit option is chosen which then returns the operator to either kinematics or kinetics subroutines.

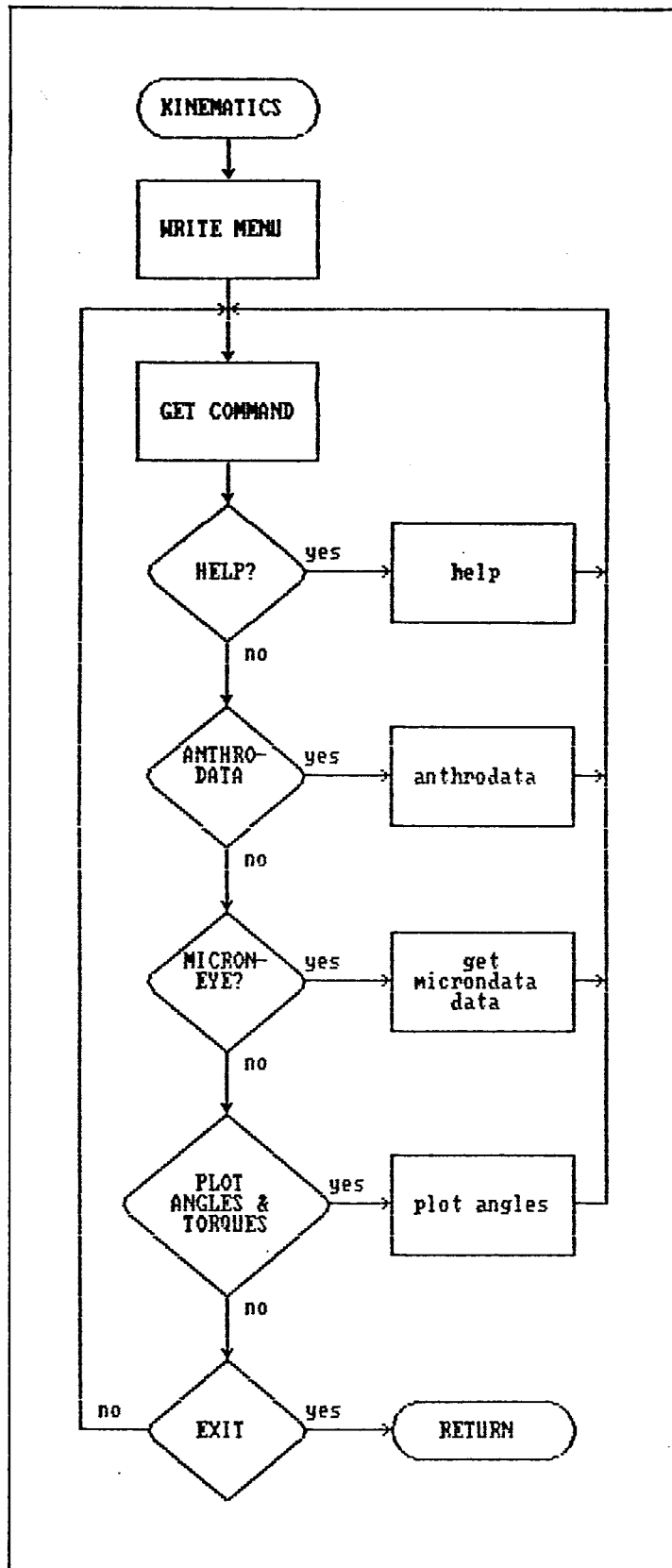


Figure G.1
The kinematics subroutine flowchart.

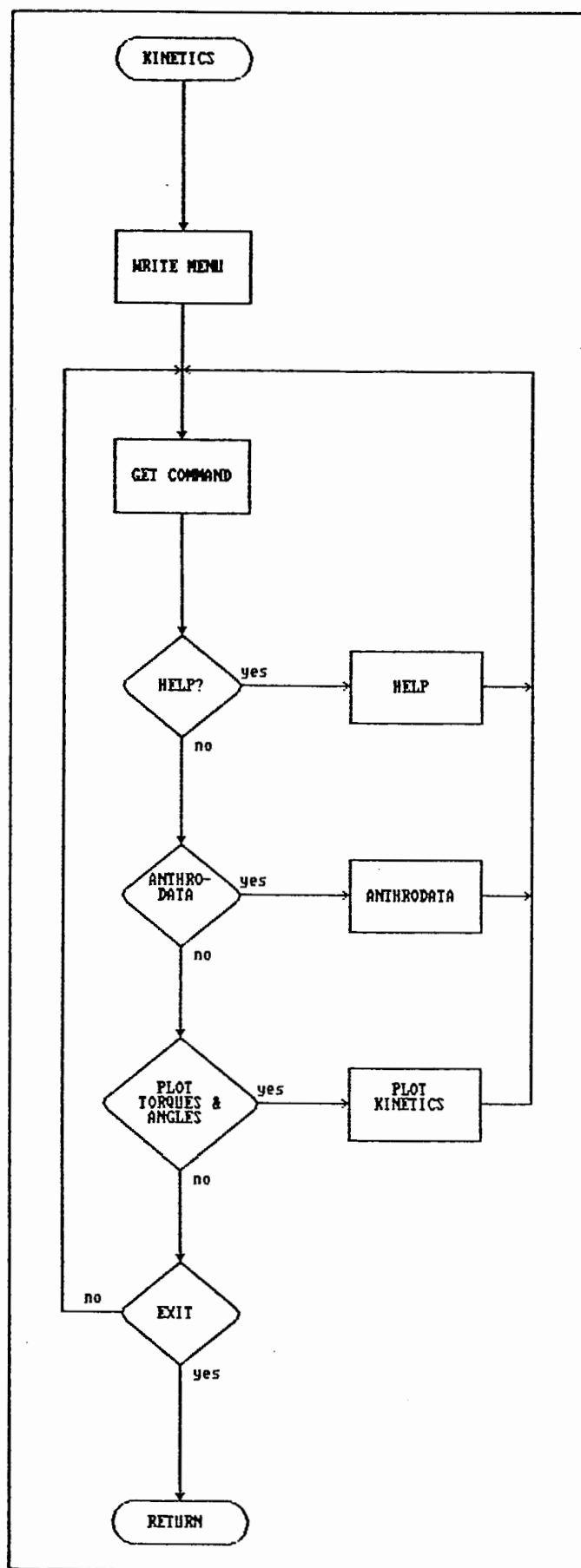


Figure G.2
The kinetics subroutine flowchart.

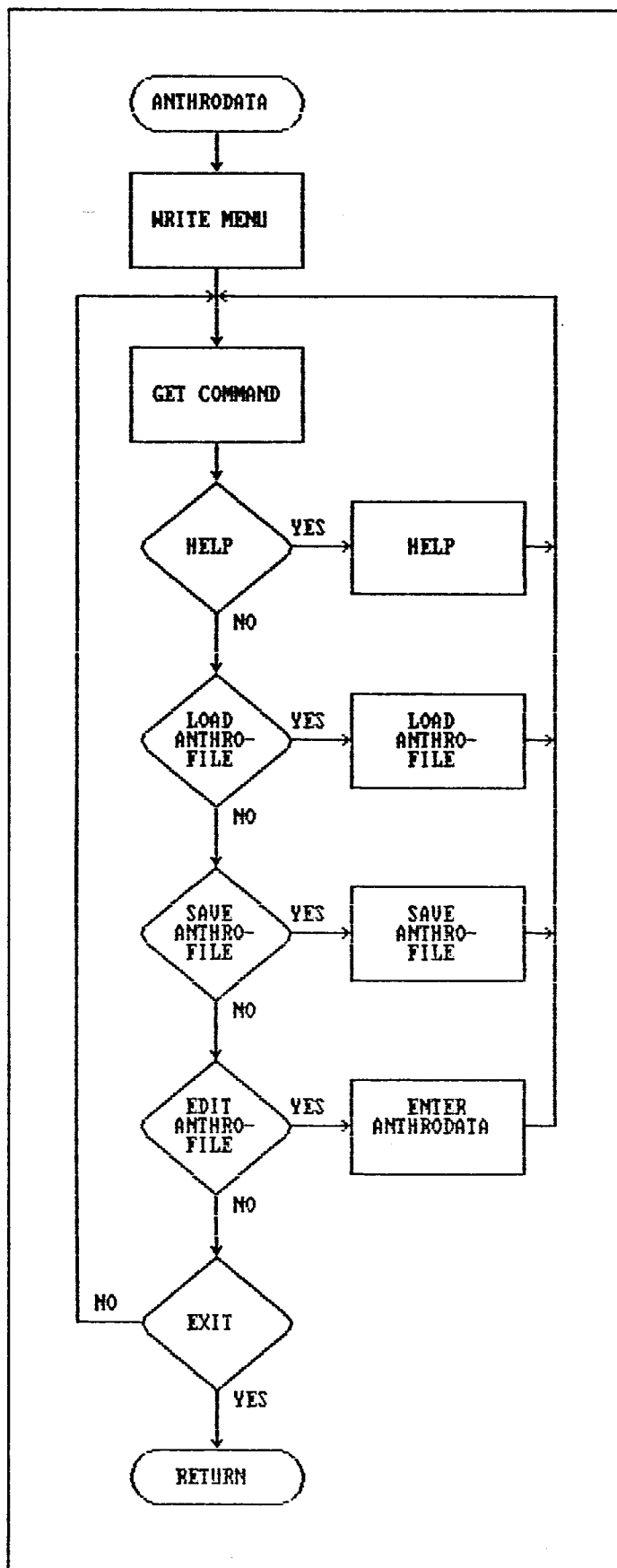


Figure G.3
The anthrodata subroutine flowchart.

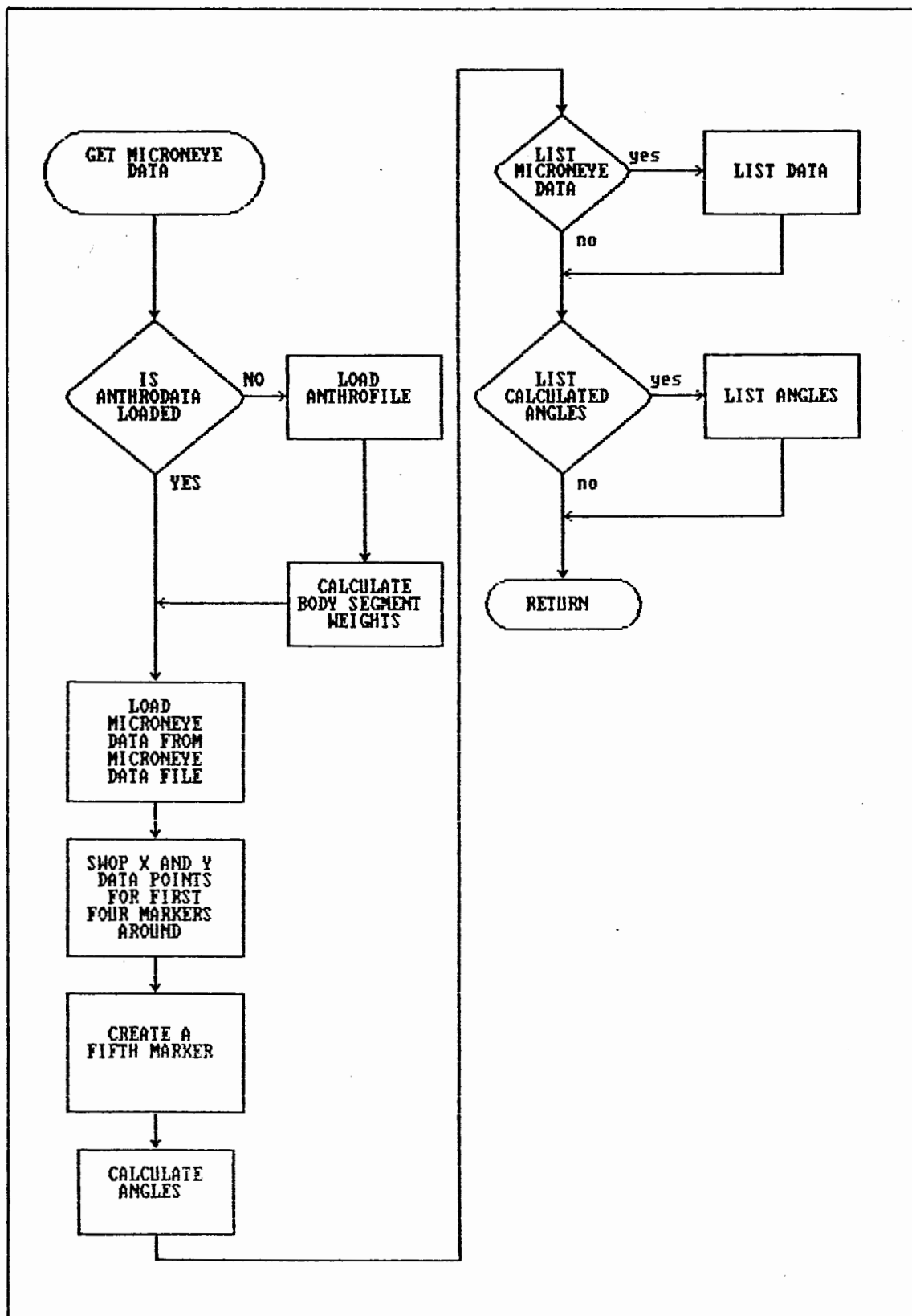


Figure G.4
The getmicroneyedata subroutine flowchart.

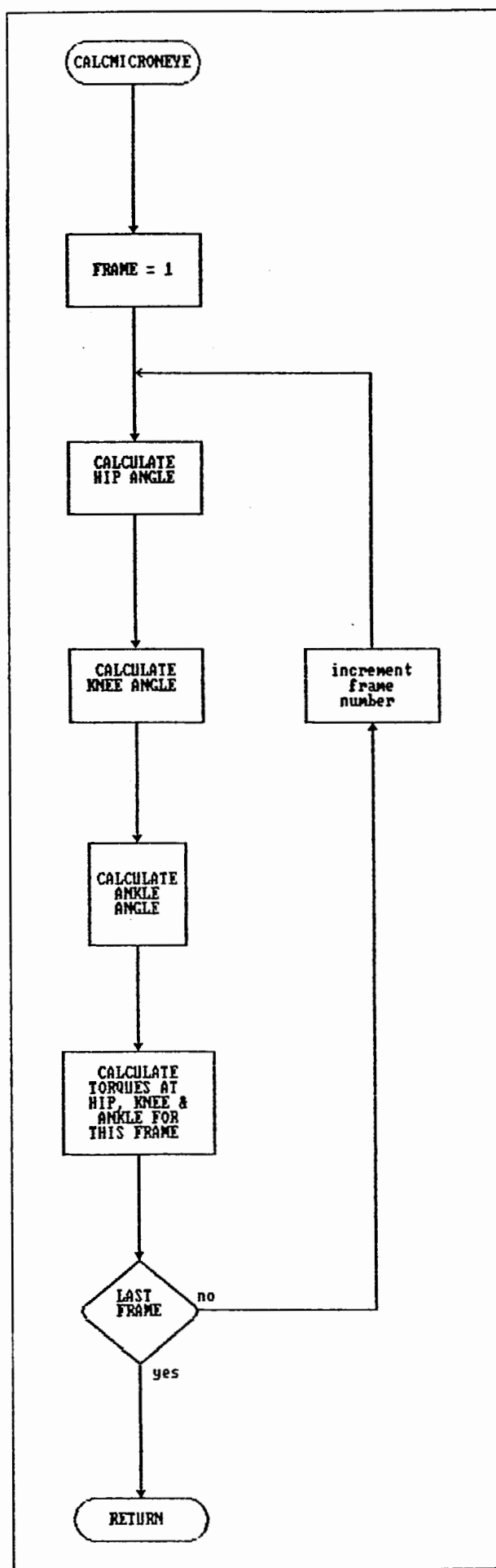


Figure G.5
The calcmicroneye angles subroutine flowchart.

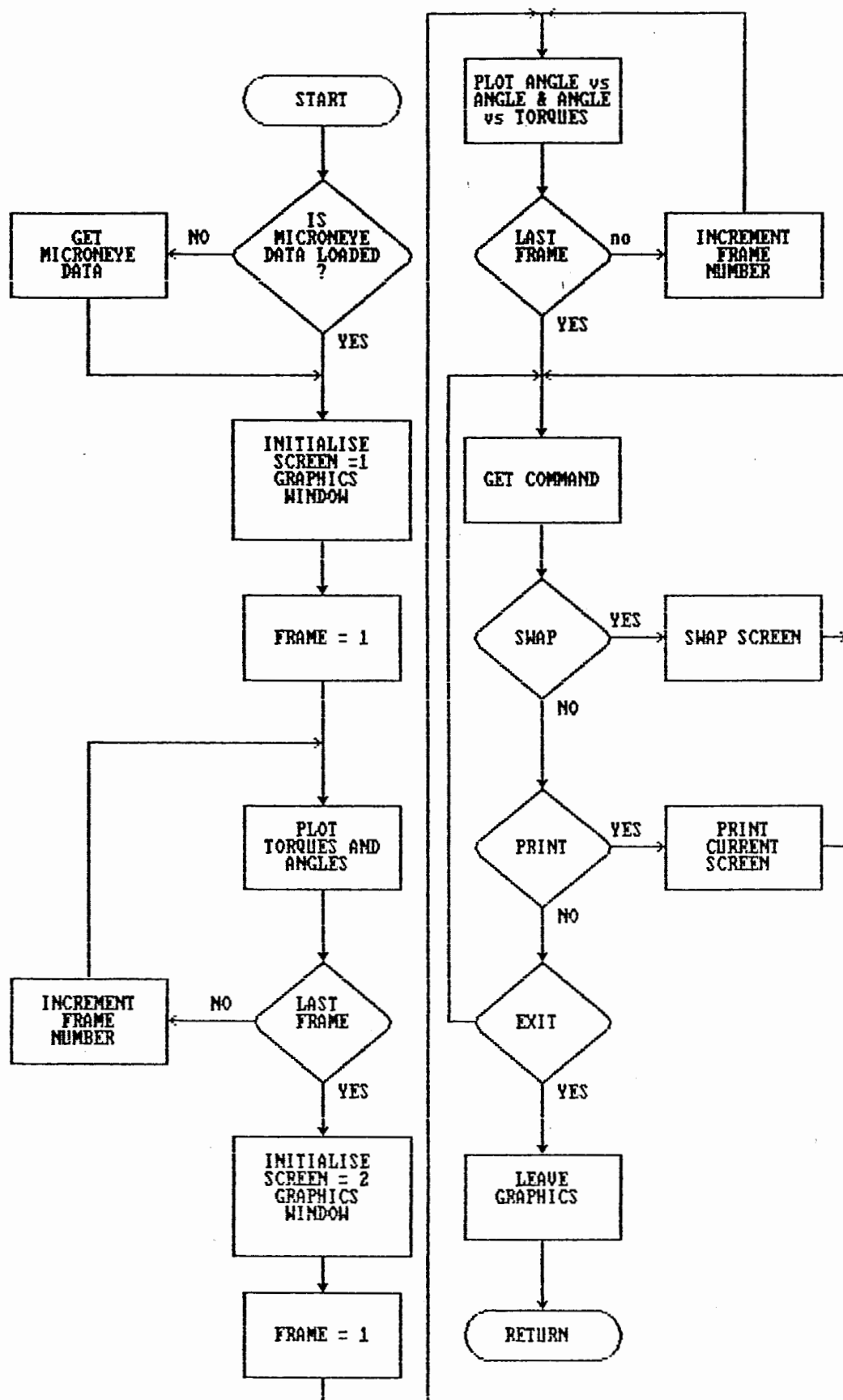


Figure G.6
The plotangles subroutine flowchart.

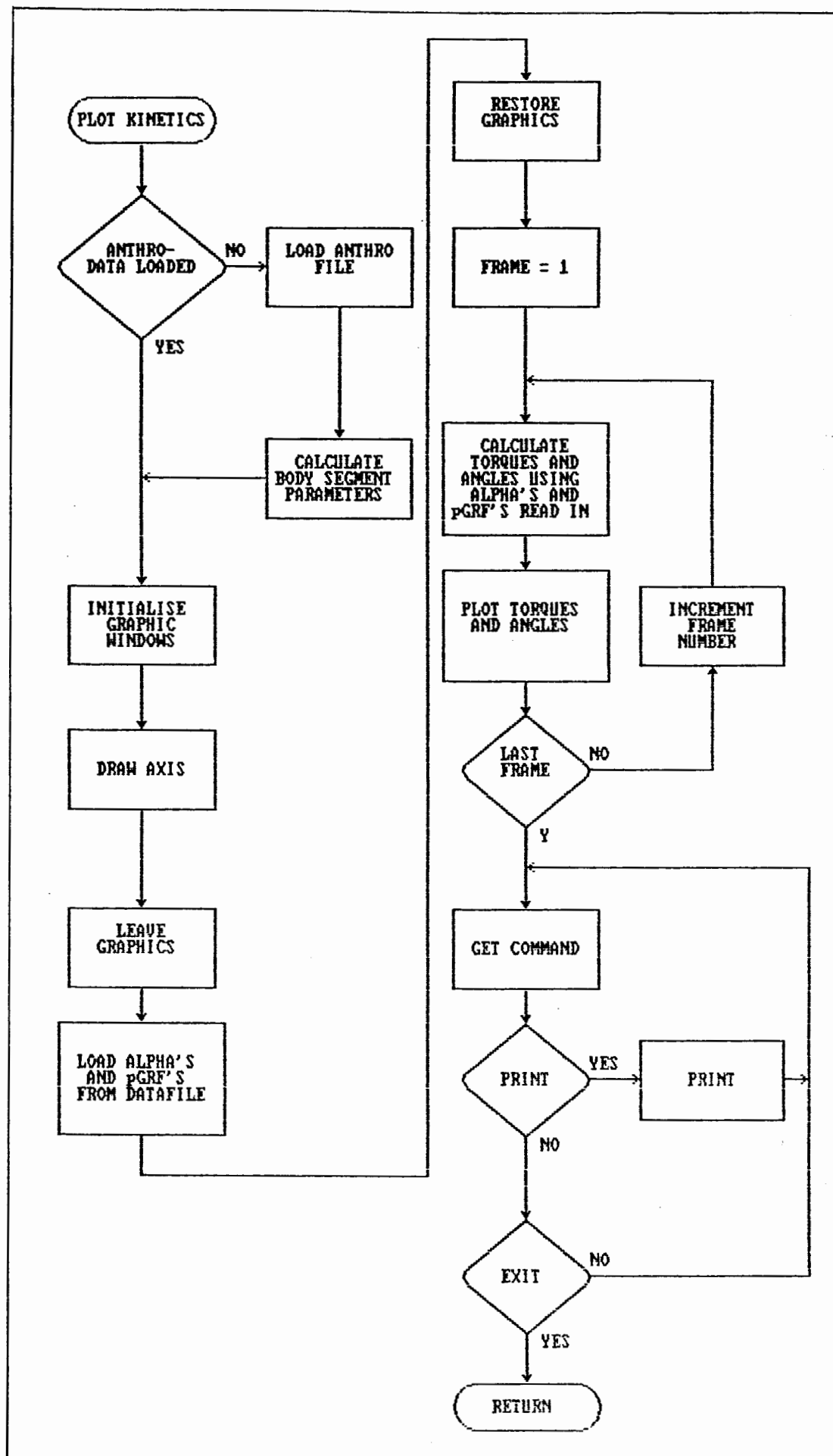


Figure G.7
The plotkinetics subroutine flowchart.

G.3 The "getmicroneyedata" subroutine

Figure G.4 shows the flowchart for the `getmicroneyedata` subroutine. The first operation that is performed is a check to determine whether the operator has loaded the `anthrodata` file. If not then the subroutines `loadanthrodata` and `calcbodysegparameter` are called. The subroutine `calcbodysegparameter` utilizes the anthropometrical data to calculate the various body segment weights. This is followed by the loading of the "micronEye" data file from a floppy diskette. Owing to the nature of the experiments (see Chapter 4) used to obtain the "micronEye" data, the camera had to be laid on its side thus rotating the x and y axis through 90 degrees. Consequently these had to be swopped around and a fifth marker was also created to enable the calculation of the ankle angles.

The program then proceeds to calculate the hip, knee and ankle angles, corresponding to the "micronEye" data using the subroutine `calcmicroneye angles`. The corresponding torques are then calculated using the calculated angles and returned to the `getmicroneye data`. Finally, the operator is given the options to list both the micronEye and corresponding angles data to screen and to make a hardcopy, before finally returning to the `kinematics` subroutine.

G.4 The "calcmicroneyeangles" subroutine

Figure G.5 shows the flowchart for this subroutine. This is the subroutine which is called by the `getmicroneye` subroutine. It initialises frame one of the micronEye and utilises the functions `angval` and `len` to calculate the hip, knee and ankle angles. The

function `len` calculates the distance between two points `x,y` obtained from the `micronEye` data, by making use of a right angled triangle and the geometric relationship between the sides of the triangle. The function `angval` (angle value) utilises the Cosine rule :

$$A^2 = B^2 + C^2 - 2BC \cos A$$

for triangles to calculate the required angles before returning to the calling routine. The subroutine `calctorques` is then called and the variable parameters `alpha`, `beta` and `phi` are passed to it. It then calculates the corresponding ankle, knee and hip torques. The subroutine executes the model equations, '`numberframesrec`', (in this case fifty times), before returning to the calling routine.

G.5 The "`plotangles`" subroutine

Figure G.6 shows the flow chart for the `plotangles` subroutine. The subroutine first checks to ensure that the operator has loaded the `microneye` data file. If not, the subroutine `getmicroneye` data is called. The next operation that is performed is to enter graphics mode. `TURBO GRAPHICS TOOLBOX` allows one to define graphics windows and then to define a world, (UNITS), coordinate system within each window. Turbo graphics allows graphics to be drawn on two screens: the visible screen and the invisible memory screen. All drawing is done on the invisible screen. The pop-up graph axes are drawn and the graph window saved on the window stack for future retrieval.

Screen one is initialised and the subroutine then plots graphs of the torques and angles against frame number (time)

'numberframesrec' times. Once completed screen two is initialised and, as above, graphs of angle against angle and angle against torque are plotted. Whilst the graphs are being plotted the menu line 'PLOTING CURVES' is written to the screen. On completion, the menu line 'P to print screen, Space to toggle screens. Any other to quit' is written below the graphs. If a command key has been pressed then the command is executed by calling the appropriate subroutines from within the CASE statement. Once the quit option has been chosen the graphics screen is cleared and the text mode restored.

Figure G.7 shows the flowchart for the plotkinetics subroutine which is called by the kinetics subroutine. The format for this subroutine is similar to that of the plotangles subroutine and hence no description as to its operation will be given.

APPENDIX H

A DESCRIPTION OF THE ROUTINES RESIDING IN PRE-COMPILED TURBO V4.0 UNITS THAT ARE CALLED BY THE SIMULATOR PROGRAM

H.1 The turbo graphix toolbox units

This graphics package consists of three units that contain all the basic graphics routines:

Gdriver - contains graphics card information. In the case of the model it has been set up to talk to a cga or monochrome graphics card.

modkern - contains the basic graphics routines.

Gwindow - contains routines for manipulating windows.

The cga card has the following characteristics:

Screen size in graphics mode = x - 0 to 639

y - 0 to 199

0,0, is the top left corner
in screen coordinates.

H.1.1 Initializing graphics

To initialise and leave graphics mode, the following routines are called:

initgraphic; - initialise the graphics system

leavegraphic; enter textmode. Defined windows and worlds are preserved.

entergraphic; re-enter graphics mode after a leavegraphic has been called. Old windows and worlds can still be used.

H.1.2 Windows and worlds

Windows are defined in screen coordinates as follows:

`definewindow (win,x1,y1,x2,y2):` where win is the window number and the x1, y1 = the top left coord and x2, y2 = the lower right.

NB!! The x screen coordinates can only be declared at positions that are a multiple of 8, ie. divide actual screen position by 8 and round off to nearest integer for x values. eg. screen x position 121 is declared as 15 in `definewindow` procedure.

Windows can have a world coordinate system assigned to them as follows:

First define a world system as:

`defineworld (world,x1,y1,x2,y2);` where world is the world number, x1, y1 refer to the lower left world coordinate and x2, y2 the top right.

eg. If one wants to plot data that ranges from -5 to 6 over the axis and from 0 to 0.5 over the y axis in a window which covers the whole graphics screen then proceed as follows:

`definewindow (win,0,0,90,299);` window over whole screen

NB! $x2 = 90 = \text{position } 720/8$

`defineworld (world,-5,0,6,0.5);` define world coordinates

`selectworld (world);` select the world

`selectwindow (win);` select the window

Data is now plotted in this window according to the world coordinate system world.

H.1.3 A list of routines used by the model

Routines used by the model are as follows:

initgraphic;

leavegraphic;

entergraphic;

definewindow (win,x1,y1,x2,y2);

defineworld (world,x1,y1,x2,y2);

selectwindow (win);

selectworld(world);

drawborder; - draw a border around the current window

setbackground (0); - clear the current window

setlinestyle(style); - select a linestyle 0-255.

0 = solid line - the rest are a bit mapped

drawtext(x,y,size,string); - draw text string at world coordinate x,y of size. The 4X6 size font is used.

drawline(x1,y1,x2,y2); - draw a line from world coordinate x1, y1 to x2, y2. Uses the correct linestyle.

storewindow(win); - save a window on the window stack for later retrieval.

restorewindow(win,x,y); - restore a saved window from the window stack. x,y is an offset in that it is ignored if they are negative.

copyscreen; - copy the inactive screen to the active one

swapscreen;	- swap the inactive screen with the active one
selectscreen (screen);	- select the visual or the ram screen as the active screen screen = 0 is the visual screen, 1 = the ram screen All plotting and drawing takes place in the active screen.
gotoxy (x,y) :	- send the cursor to text coordinate x,y. x = 1 to 80 y = 1 to 25 0,0 = top left
WRITE OR Writeln (data);	- write text at current text coordinate. Uses 12X9 size font. Data can be number or string. If the global variable, invglb, is TRUE then the text is written in the inverse colour.

H.2 The axes unit

The axes unit is used in conjunction with the TURBO GRAPHIX TOOLBOX. Available routines are:

PROCEDURE drawaxes (win, wrld	: INTEGER;
xtick, ytick	: REAL;
xaxis, yaxis,	
title	: string;
xlen, xpl, ylen, ypl	: INTEGER;
id	: BOOLEAN;)

The parameters passed to drawaxes are defined as:

win - predeclared window to draw graph in
wrld - predeclared world (ie 0,0 to 10,100)
xtick,
ytick - tick density along x and y axis
xaxis,
yaxis,
title - axis titles and main title of graph
xlen,
ylen,
xpl,
ylen,
ypl - field length and dec places of axis numbers

* Note drawaxes will only work in graphics mode setup by
turbographics toolbox.

PROCEDURE idgraph (win, number, size : INTERGER);

* If id was set TRUE during drawaxes call, then idgraph will
identify up to 6 variables whose names must be in the array
lineid. The names are drawn in the space left under the x axes.
Each line number is identified with a unique character every
maxcount points that are plotted. Idgraph draws these characters
next to the name as well.

PROCEDURE plotxy (x,y : REAL;
 line,
 linestyle : INTEGER);

* Plots line of linetype from ox,oy to x,y. If oxyinit[line]
is false then saves x,y in ox,oy for that line and sets
oxyinit[line] true.

```
PROCEDURE plotpoints (x,y : REAL; line, linetype : INTEGER );  
* Plots line of linetype from ox,oy to x,y. If oxyinit[line] is  
false then saves x,y in ox,oy for that line and sets  
oxyinit[line] true. If linecount[line] is > than maxcount, it is  
reset to zero and a character that identifies that line is drawn  
at x,y as well.
```

H.3 Various other routines

The following routines are also called by the main simulator program. The following list indicates what the routines do and which UNIT they reside in:

UNIT GENERAL

Beep (frequency) - sounds a tone of frequency hertz for 250ms

UNIT 10

Message (y,string); - centre the message string on line y but
clear the line first

Center (y,string); - same as above but does not clear the line

UNIT CONVERT

sin, cos, tan - converts an angle into radians

H.4 The help unit

The help unit contains the following three routines:

```
inithelp (filename);  
help (filename);  
closehelp
```

A call to inithelp will load the help file specified by filename.

If the file is not found, the program is terminated with an appropriate error message. If there is not enough memory to load the help file into the program, it is halted with an error message.

If help is called before inithelp has been called, the subroutine calls inithelp itself and then continues. Help saves the contents of the existing text screen, and displays a menu of help topics on the top 10 lines of the screen. The operator moves from topic to topic with the cursor keys. The text describing the currently selected topic is displayed in the lower 15 lines of the screen. The operator can return to the calling routine by pressing escape whereupon the old screen will be restored.

The format of the help file is as follows:

line 1 = number of help screens to be used
line 2 = first line of top screen. Help line ie. esc, cursor
 directions
line 3 = blank
line 4 = first line with four topics of sixteen characters each
 to
line 10 = last line of four topics
line 11 = blank

The rest of the file consists of the help screens each consisting of exactly fourteen lines of text. An editor such as PC WRITE can be used to edit the file.

APPENDIX I

LISTING OF PROGRAM RESTAND

```
uses dir, crt, dos, general, io, math, convert, hlp, axes,
      cgaDriver, cgakern, cgaWindow, printer;
```

```
CONST hip = 1;
      knee = 2;
      ank = 3;
```

```
CONST a1 = 0;
      a2 = 25;
```

```
TYPE kintype = (alpha,abeta,aphi,tankle,tknee,thip);
      artype = ARRAY[aalpha..thip] OF REAL;
```

```
TYPE
```

```
  Data = RECORD
    expose_timeREC :Integer;
    send_timeREC   :Integer;
    NumberFramesREC :Integer;
    NumberMarkersREC :Integer;
    SubjectREC      :String[20];
    DateREC         :String[20];
    CommentREC      :String[40];
    XcoordREC       :array[1..5,1..50] of Real;
    YcoordREC       :array[1..5,1..50] of Real;
  END;
```

```
TYPE
```

```
  participant_record = RECORD
```

{This record is the anthropometric data of the participant. The values are used to calculate the segment masses which in turn are used to calculate the hip, knee and ankle torques required for standing up.}

```
    name      : string;
    date      : string;
    bdate     : string;
    sex       : string;
    mass      : REAL;    {body mass in kgs}
    mas_ver   : REAL;    {distance from the vertex of the head to the mastoid}
    head      : REAL;    {head circumference}
    torso     : REAL;    {distance from the chin to the symphysis pubis}
    chest_d   : REAL;    {chest depth}
    chest_b   : REAL;    {chest breadth}
    hum_rad   : REAL;    {distance from the humerus to the radiale of the arm}
    hum_ver   : REAL;    {distance from the humerus to the vertex of the head}
    biceps    : REAL;    {biceps circumference}
    rad_sty   : REAL;    {distance from the radiale to the stylium of the arm}
    forearm   : REAL;    {forearm circumference}
    sty_met   : REAL;    {distance from the stylium to the tip of metacarpal III}
    hand_b    : REAL;    {hand breadth}
    hand_d    : REAL;    {hand depth}
    thigh_l   : REAL;    {distance from the greater trochanter to the lateral epicondyle}
    thigh_c   : REAL;    {thigh circumference}
    calf_l    : REAL;    {distance from the lateral epicondyle to the lateral malleolus}
    calf_c    : REAL;    {calf circumference}
    foot_l    : REAL;    {distance from the heel to the tip of the first toe}
    lat_mal   : REAL;    {distance from the lat malleolus to the sole of the foot}
    foot_b    : REAL;    {foot breadth}
    tot_hgt   : REAL;    {total height}
    foot_p    : REAL;    {distance from the heel to the lateral malleolus}
  END;
```

```

    anglearray = ARRAY[1..3,1..50] OF REAL; {ankle,knee & hip angles for 50 frames}

CONST angleflag : BOOLEAN = FALSE;
      g          = 9.81;
      pi         = 3.1415927;

VAR
  DataFile      : FILE OF Data;
  DataRecord    : Data;
  angles        : anglearray;
  torques       : ARRAY[1..4,1..50] OF REAL; {1=hip, 2=knee, 3=ankle, 4=pGRF}
  frec         : filerecord;
  x,y           : INTEGER;
  ch            : CHAR;
  participant_info : participant_record;
  participant_file : FILE OF participant_record;
  micronfile,
  anthrofile    : string;
  OK,
  glbesc,
  anthroflag    :BOOLEAN;
  i,j,
  pntr          : INTEGER;
  M             : Array[1..9] of REAL; {these are for the various segments}
  x_foot,x_calf,x_thigh,x_hat,          {horizontal distance from cg of seg masses to lat mal}
  c_hat,          {distance of cg of trunk from hip joint}
  GRF,            {ground reaction force in newtons}
  cg_head,        {distance of center of gravity of head,trunk,upperarm}
  cg_trunk,        {lowerarm, hand and HAT (head arms and trunk) measured}
  cg_upperarm,     {from the foot}
  cg_lowerarm,
  cg_hand,
  cg_HAT          : REAL;
  Fay,Fky,Fhy    : REAL; {vertical forces at the ankle, knee and hip joints}
  p_grf          : REAL; {distance of GRF from lat mall}
  arr            : artype;

{-----}

PROCEDURE listmicroneyedata;

VAR x,y : INTEGER;
    ch  : CHAR;

{This subroutine lists the data obtained from the micron-eye camera in their
x,y coordinates. Markers 1,2,3,4 & 5 correspond to the positions of the
markers on the lat trunk, greater trochanter, lat epicondyle, lat malleolus
& a point, extended horizontally from the lat malleolus, respectively}

BEGIN
  message(14,'List to screen or printer (S/P) ?');
  ch := readkey;

  IF upcase(ch) <> 'P' THEN
    ASSIGN(lst,"")
  ELSE
    ASSIGN(lst,'PRN');

  REWRITE(lst);
  clrscr;
  WITH datarecord DO

```

```

BEGIN
  { WRITELN(lst,'Exposetime = ',expose_timerec);
    WRITELN(lst,'Send time = ',send_timeREC);
    WRITELN(lst,'Frames = ',numberframesrec);
    WRITELN(lst,'Markers = ',numbermarkersrec);}
  WRITELN(lst,'Subject = ',subjectrec);
  WRITELN(lst,'Date = ',daterec);
  WRITELN(lst,'Comment = ',commentrec);
  WRITELN(lst);
  WRITELN('Hit a key to list data');
  CH := readkey;
  CLRSCR;
  WRITELN(lst,' Marker 1 | Marker 2 | Marker 3 | Marker 4 | Marker 5 |');
  WRITELN(lst,' X Y | X Y | X Y | X Y | X Y |');
  FOR y := 1 TO numberframesrec DO
    BEGIN
      FOR x := 1 TO numbermarkersrec DO
        BEGIN
          WRITE(lst,xcoordrec[x,y]:4:0,',',ycoordrec[x,y]:4:0,' | ');
        END;
      WRITELN(lst);
    END;
  END;
  close(lst);
  message(25,'Hit a key to continue');
  ch := READKEY;
  clrscr;
END;

{-----}

PROCEDURE listmicroneyeangles;

VAR ch : CHAR;
    fr,
    angle : INTEGER;

BEGIN
  message(14,'List to Screen or Printer (S/P)..>');
  CH := readkey;
  IF upcase(ch) <> 'P' THEN
    ASSIGN(lst,'')
  ELSE
    ASSIGN(lst,'PRN');
    REWRITE(lst);
    clrscr;
    WITH datarecord DO
      BEGIN
        WRITELN(lst,'Subject = ',subjectrec);
        WRITELN(lst,'Date = ',daterec);
        WRITELN;
        FOR fr := 1 TO numberframesrec DO
          BEGIN
            WRITE(lst,'Frame ',fr:3);
            FOR angle := 1 TO 3 DO
              WRITE(lst,' Angle ',angle,' : ',deg(angles[angle,fr]):8:3);
            WRITELN(lst);
          END;
        END;
      END;
    message(25,'Hit a key to return to menu');
    ch := readkey;
  
```

```

clrscr;
END;

```

```

{-----}

```

```

FUNCTION angval(x1,y1,x2,y2,x3,y3 : REAL) : REAL; {angle value}

```

```

{this function returns the angle between three points using the
cosine rule for triangles}

```

```

VAR a,b,c,
    temp : REAL;

```

```

{-----}

```

```

FUNCTION len(xa,ya,xb,yb : REAL) : REAL;

```

```

{this function returns the length between a and b by creating a
90 degree triangle & using the geometric relationship relating
the sides of a triangle. It is local to function angval only. }

```

```

VAR temp : REAL;

```

```

BEGIN
    temp := sqrt(sqr(xa-xb) + sqr(ya-yb));
    len := temp;
END;

```

```

{-----}

```

```

BEGIN
    a := len(x1,y1,x3,y3);
    b := len(x1,y1,x2,y2);
    c := len(x2,y2,x3,y3);
    temp := arccos((sqr(b) + sqr(c) - sqr(a))/(2*b*c));
    angval := temp;
END;

```

```

{-----}

```

```

PROCEDURE calcbodysegparameters; { body segmental parameters }

```

```

VAR
    theta : REAL;
    r      : REAL;
    ch     : CHAR;

```

```

{This subroutine calculates the segmental masses and the combined
center of gravity for the head, upper limbs & trunk, HAT, using the
regression equations of Chandler et al(1975), as in Appendix A}

```

```

{ stipulate segment numbers
  1 Head
  2 Trunk
  3 Upper arm
  4 Forearm
  5 Hand
  6 Thigh
  7 Calf
  8 Foot

```

9 HAT}

BEGIN

CLRSCR;

WITH participant_info DO

BEGIN {calculate segment mass}

```
m[1] := 0.0221*mass + 63.76 * mas_ver * head * head - 0.742;
m[2] := 0.2791*mass + 413.6 * torso * chest_d * chest_b - 3.881;
m[3] := 0.0096*mass + 14.51 * hum_rad * biceps * biceps + 0.812;
m[4] := 0.0172*mass + 12.89 * rad_sty * forearm * forearm - 0.286;
m[5] := 0.0023*mass + 1274 * sty_met * hand_b * hand_d - 0.038;
m[6] := 0.1032*mass + 12.76 * thigh_l * thigh_c * thigh_c - 1.023;
m[7] := 0.0226*mass + 31.33 * calf_l * calf_c * calf_c + 0.016;
m[8] := 0.0083*mass + 254.5 * foot_l * lat_mal * foot_b - 0.065;
```

{adjust the segment masses so that their sum equals total body mass}

r := m[1]+m[2]+2*(m[3]+m[4]+m[5]+m[6]+m[7]+m[8]);

{converting segment masses to segment weights}

FOR i:= 1 TO 8 DO

BEGIN

m[i] := m[i]*mass*g/r;

END;

{cal combined COG,in saggital plane, arms folded on chest}

```
thetha := arcsin((chest_d+forearm/pi)/(2*hum_rad));
m[9] := m[1]+m[2]+2*(m[3]+m[4]+m[5]); {weight of HAT}
cg_head:= tot_hgt - 0.66*mas_ver;
cg_trunk:= lat_mal + calf_l + thigh_l + 0.48*torso;
cg_upperarm:=tot_hgt-(hum_ver+0.51*hum_rad*cos(thetha));
cg_lowerarm:=tot_hgt-(hum_ver+hum_rad*cos(thetha));
cg_hand:=tot_hgt-(hum_ver+hum_rad*cos(thetha));
cg_HAT:= (1/m[9])*(cg_head*m[1]+cg_trunk*m[2]
+2*(cg_upperarm*m[3]+cg_lowerarm*m[4]+cg_hand*m[5]));
x_foot := 0.44*foot_l-foot_p;
```

GRF:= m[6]+m[7]+m[8]+0.5*m[9];

Fay := -GRF + m[8];

Fky := -GRF + m[8] + m[7];

Fhy := -GRF + m[8] + m[7] + m[6];

c_hat := cg_hat -(thigh_l + calf_l + lat_mal);

END;

END;

{-----}

FUNCTION CALCPGRF(alpha,beta,phi : REAL) : REAL;

{calculates distance of ground reaction force from lateral malleolus}

VAR

z : REAL;

temp : REAL;

BEGIN

WITH participant_info DO

BEGIN

```

z      := thigh_l*sin((pi/2) + alpha - beta) - calf_l*cos(alpha);
x_foot := 0.44*foot_l - foot_p;
x_calf := 0.58*calf_l*cos(alpha);
x_thigh := 0.39*thigh_l*sin((pi/2) + alpha - beta) - z;
x_hat   := c_hat*sin((pi/2) + beta - phi - alpha) - z;
calcpgrf := (x_foot*m[8] + x_calf*m[7] + x_thigh*m[6] + x_hat*0.5*m[9])/(GRF);
END;
END;

```

```

{-----}

```

```

PROCEDURE calctorques(alpha,beta,phi      : REAL;
                     VAR tor_ankle,tor_knee,tor_hip,
                           pgrf          : REAL);

```

```

{calculates the torques at the hip joints: requires anthro,bsp,Pgrf}

```

```

BEGIN

```

```

    WITH participant_info DO
    BEGIN
        pgrf      := calcpgrf(alpha,beta,phi);
        tor_ankle := (-GRF)*Pgrf + m[8]*x_foot;
        tor_knee  := tor_ankle - m[7]*0.42*calf_l*cos(alpha)
                        - Fay*calf_l*cos(alpha);
        tor_hip   := tor_knee + Fky*thigh_l*sin((pi/2) + alpha - beta)
                        + m[6]*0.39*thigh_l*sin((pi/2) + alpha - beta);
    END;
END;

```

```

{-----}

```

```

PROCEDURE calcmicroneyeangles;

```

```

{This subroutine calculates the hip, knee and ankle angles for each
frame of x,y coordinates obtained from the microneye camera data }

```

```

VAR angle : INTEGER;
    fr      : integer;

```

```

BEGIN

```

```

    fillchar(angles,sizeof(angles),0);
    WITH datarecord DO
    BEGIN
        FOR fr := 1 TO numberframesrec DO
        BEGIN
            FOR angle := 1 TO 3 DO
            BEGIN
                CASE angle OF
                    hip :angles[angle,fr]:=
                        angval(xcoordrec[1,fr],ycoordrec[1,fr],
                              xcoordrec[2,fr],ycoordrec[2,fr],
                              xcoordrec[3,fr],ycoordrec[3,fr]);
                    knee :angles[angle,fr] :=
                        angval(xcoordrec[4,fr],ycoordrec[4,fr],
                              xcoordrec[3,fr],ycoordrec[3,fr],
                              xcoordrec[2,fr],ycoordrec[2,fr]);
                    ank  :angles[angle,fr] :=
                        angval(xcoordrec[3,fr],ycoordrec[3,fr],
                              xcoordrec[4,fr],ycoordrec[4,fr],
                              xcoordrec[5,fr],ycoordrec[5,fr]);
                END;
            END;
        END;
    END;

```

```

        END;
        END;
        calctorques(angles[3,fr],angles[2,fr],angles[1,fr],
                    torques[3,fr],torques[2,fr],torques[1,fr],torques[4,fr]);
    END;
END;
END;

{-----}

PROCEDURE loadanthrofile ; FORWARD;

{-----}

PROCEDURE getmicroneyedata;

VAR fr,
    point : INTEGER;
    temp : REAL;

BEGIN
    IF NOT(anthroflag) THEN {can not carry on till anthrodata has been loaded}
    BEGIN
        loadanthrofile;
        calcbodysegparameters;
    END;
    clrscr;
    center(2,'Select microneye data file');
    REPEAT
        selectfile('b:*.dat',frec);
        IF length(frec.name)=0 THEN
            BEGIN
                gotoxy(1,25);
                WRITE('^G');
                clreol;
                message(25,'File error or Disk drive error!!!');
            END;
        UNTIL length(frec.name) > 0;
        micronfile := frec.name;
        ASSIGN(datafile,frec.name);
        RESET(datafile);
        READ(datafile,datarecord);
        CLOSE(datafile);
        clrscr;
        {Now swop x and y data around and set 5th marker}
        message(10,'Swopping data points');
        WITH datarecord DO
            BEGIN
                numbermarkersrec := 5;
                FOR fr := 1 TO numberframesrec DO
                    BEGIN
                        FOR point := 1 TO 5 DO
                            BEGIN
                                temp := xcoordrec[point,fr];
                                xcoordrec[point,fr] := ycoordrec[point,fr];
                                ycoordrec[point,fr] := temp;
                                {swap x y data around}
                            END;
                        END;
                        xcoordrec[5,fr] := xcoordrec[4,fr] - 50;
                        {creating a 5th marker}
                        IF xcoordrec[5,fr] < 0 THEN xcoordrec[5,fr] := 0;
                        ycoordrec[5,fr] := ycoordrec[4,fr];
                    END;
                END;
            END;

```

```

END;

message(10,'Calculating angles and torques');
calcmicroneyeangles;
angleflag := TRUE;
message(10,"");

message(12,'List microneye data (Y/N)..>');
ch := readkey;
IF upcase(ch) = 'Y' THEN listmicroneyedata;

message(12,'List calculated angles (Y/N)..>');
ch := readkey;
IF upcase(ch) = 'Y' THEN listmicroneyeangles;
END;

{-----}

PROCEDURE plotangles;

VAR fr,i : INTEGER;
    xsize,
    ysize : REAL;
    quit : BOOLEAN;

BEGIN
    IF NOT(angleflag) THEN getmicroneyedata;
    initgraphic;
    definewindow(1,0,10,(xmaxglb DIV 3) - 1,ymaxglb DIV 2);
    definewindow(2,xmaxglb DIV 3,10,(2*xmaxglb DIV 3) - 1,ymaxglb DIV 2);
    definewindow(3,2*xmaxglb DIV 3,10,xmaxglb,ymaxglb DIV 2);
    definewindow(4,0,ymaxglb DIV 2,(xmaxglb DIV 3) - 1,ymaxglb - 10);
    definewindow(5,xmaxglb DIV 3,ymaxglb DIV 2,(2*xmaxglb DIV 3) - 1,ymaxglb - 10);
    definewindow(6,2*xmaxglb DIV 3,ymaxglb DIV 2,xmaxglb,ymaxglb - 10);
    definewindow(7,0,ymaxglb DIV 2,(xmaxglb DIV 2) - 1,ymaxglb - 10);
    definewindow(8,xmaxglb DIV 2,ymaxglb DIV 2,xmaxglb,ymaxglb - 10);

    defineworld(1,0, - 150,50,150);
    defineworld(2,0,80,50,200);
    defineworld(3,0,60,50,90);

    FOR i := 1 TO 5 DO axesdrawn[i] := FALSE;

    gotoxy(30,10);
    WRITELN('PLOTING CURVES');
    selectscreen(2);
    drawaxes(1,1,10,50,'FRAME','TORQUE NM','ANKLE TORQUE',2,0,2,0,false);
    drawaxes(2,1,10,50,'FRAME','TORQUE NM','KNEE TORQUE',2,0,2,0,false);
    drawaxes(3,1,10,50,'FRAME','TORQUE NM','HIP TORQUE',2,0,2,0,false);
    drawaxes(4,3,10,10,'FRAME','DEG','ANKLE ANGLE',2,0,2,0,false);
    drawaxes(5,2,10,20,'FRAME','DEG','KNEE ANGLE',2,0,2,0,false);
    drawaxes(6,2,10,20,'FRAME','DEG','HIP ANGLE',2,0,2,0,false);

    gotoxy(1,1);
    WRITELN('Files are '+anthrofile+' and '+micronfile);
    gotoxy(1,25);
    WRITELN('      P to print screen. Space to toggle screens. Any other to quit      ');

    FOR i := 1 TO 3 DO
    BEGIN

```

```

    ox[i] := 0;
    oy[i] := torques[i,1];
END;
    ox[4] := 0;
    ox[5] := 0;
    ox[6] := 0;
    oy[4] := deg(angles[1,1]); {4 = hip}
    oy[5] := deg(angles[2,1]); {5 = knee}
    oy[6] := deg(angles[3,1]); {6 = ankle}

FOR fr := 2 TO 50 DO
BEGIN
    selectworld(1);
    selectwindow(1);
    plotxy(fr,torques[3,fr],3,0); {3 = ankle torque}
    selectwindow(2);
    plotxy(fr,torques[2,fr],2,0); {2 = knee torque}
    selectwindow(3);
    plotxy(fr,torques[1,fr],1,0); {1 = hip torque}

    selectworld(3);
    selectwindow(4);
    plotxy(fr,deg(angles[3,fr]),6,0); {6 = ankle angle alpha}
    selectworld(2);
    selectwindow(5);
    plotxy(fr,deg(angles[2,fr]),5,0); {5 = knee angle beta}
    selectwindow(6);
    plotxy(fr,deg(angles[1,fr]),4,0); {4 = hip angle phi}
END;

swapscreen;
clearscreen;

defineworld(1,50,-150,200,150);
defineworld(2,90,90,180,180);
defineworld(3,60,-150,90,150);
defineworld(4,60,-0.25,90,0.15);

FOR i := 1 TO 8 DO axesdrawn[i] := FALSE;

drawaxes(1,3,10,50,'ANKLE','TORQUE NM','TORQUE VS ANGLE',2,0,2,0,false);
drawaxes(2,1,50,50,'KNEE','TORQUE NM','TORQUE VS ANGLE',2,0,2,0,false);
drawaxes(3,1,50,50,'HIP','TORQUE NM','TORQUE VS ANGLE',2,0,2,0,false);
drawaxes(7,2,10,10,'BETA','PHI','ANGLE VS ANGLE',2,0,2,0,false);
drawaxes(8,4,10,0.10,'ANKLE','PGRF','pGRF VS ANGLE',2,0,1,2,false);

gotoxy(1,1);
WRITELN('Files are '+anthrofile+' and '+micronfile);
gotoxy(1,25);
WRITELN(' P to print screen, space to toggle screens. Any other to quit ');

FOR i := 1 TO 3 DO
BEGIN
    ox[i] := deg(angles[i,1]); {1 = hip; 2 = knee; 3 = ankle}
    oy[i] := torques[i,1];
END;

FOR fr := 2 TO 50 DO
BEGIN
    selectworld(3);
    selectwindow(1);

```

```

{IF torques[4,fr] >= 0 THEN}
plotxy(deg(angles[3,fr]),torques[3,fr],3,0);
selectworld(1);
selectwindow(2);
{IF torques[4,fr] >= 0 THEN}
plotxy(deg(angles[2,fr]),torques[2,fr],2,0);
selectwindow(3);
{IF torques[4,fr] >= 0 THEN}
plotxy(deg(angles[1,fr]),torques[1,fr],1,0);
selectworld(2);
selectwindow(7);
drawpoint(deg(angles[2,fr]),deg(angles[1,fr]));
selectworld(4);
selectwindow(8);
drawpoint(deg(angles[3,fr]),torques[4,fr]);
END;

```

```

selectscreen(1);
quit := FALSE;
REPEAT
  ch := readkey;
  CASE ch OF
    ' ' : swapscreen;
    'P','p' : hardcopy(false,1);
    ELSE quit := true;
  END;
UNTIL quit;
leavegraphic;
END;

```

{-----}

```

PROCEDURE getparm(x,y : INTEGER; VAR parm : REAL);

```

```

VAR ch : CHAR;
    esc2 : BOOLEAN;

```

```

BEGIN
  gotoxy(x,y);
  WRITE(parm:6:3);
  gotoxy(x,y);
  ch := readkey;
  CASE ch OF
    #13,'+' : BEGIN
      cursor_size(cursor_small);
      gotoxy(x,y);
      WRITE(' ');
      gotoxy(x,y);
      readreal(parm,esc2);
      cursor_size(cursor_large);
      gotoxy(x,y);
      WRITE(parm:6:3);
      pntr := pntr + 1;
      IF pntr > 27 THEN pntr := 1;
    END;
    #27 : glbesc := true;
  END; {Case}
END;

```

{-----}

```
PROCEDURE getstr(x,y : INTEGER; VAR parm : string);
```

```
VAR ch : CHAR;
    esc2 : BOOLEAN;
```

```
BEGIN
```

```
    gotoxy(x,y);
    WRITE(parm);
    gotoxy(x,y);
    ch := readkey;
    CASE ch OF
        #13,'+' : BEGIN
            cursor_size(cursor_small);
            gotoxy(x,y);
            WRITE(' ');
            gotoxy(x,y);
            readstr(parm,esc2);
            cursor_size(cursor_large);
            gotoxy(x,y);
            WRITE(parm);
            pntr := pntr + 1;
            IF pntr > 27 THEN pntr := 1;
```

```
        END;
```

```
        #27 : gibesc := true;
```

```
    END; {Case}
```

```
END;
```

```
{-----}
```

```
PROCEDURE loadanthrofile;
```

```
BEGIN
```

```
    clrscr;
    center(1,'LOAD ANTHRO DATA PROCEDURE');
    center(4,'Select anthropometrical data file');
    REPEAT
        selectfile('b:*.ant',frec);
        IF length(frec.name)=0 THEN
            BEGIN
                gotoxy(1,25);
                WRITE('^G');
                clreol;
                message(25,'File error or Disk drive error!!!');
```

```
            END;
```

```
    UNTIL length(frec.name) > 0;
```

```
    anthrofile := frec.name;
```

```
    ASSIGN(participant_file,frec.name);
```

```
    RESET(participant_file);
```

```
    READ (participant_file,participant_info);
```

```
    CLOSE (participant_file);
```

```
END;
```

```
{-----}
```

```
PROCEDURE saveanthrofile;
```

```
VAR exist,ok,esc : BOOLEAN;
```

```
    ch : CHAR;
```

```
FUNCTION writefile : BOOLEAN;
```

```
VAR ok : BOOLEAN;
```

```
ch : char;
```

```
BEGIN
```

```
{ $! - }
```

```
    REWRITE (participant_file);
```

```
    WRITE (participant_file,participant_info);
```

```
{ $! + }
```

```
    ok := (IOresult = 0);
```

```
    IF ok THEN
```

```
        BEGIN
```

```
            CLOSE (participant_file);
```

```
        END
```

```
    ELSE
```

```
        BEGIN
```

```
            center (16,'File write error. Hit any key to try again');
```

```
            ch := readkey;
```

```
        END;
```

```
        writefile := ok;
```

```
END;
```

```
{ ----- }
```

```
BEGIN
```

```
    ok := FALSE;
```

```
    REPEAT
```

```
        CLRSCR;
```

```
        PUTS (10,3,'ENTER NAME OF FILE TO SAVE : ');
```

```
        readstr anthrofile,esc;
```

```
        IF NOT(esc) THEN
```

```
            BEGIN
```

```
                anthrofile := 'b:' + anthrofile;
```

```
                ASSIGN(participant_file,anthrofile);
```

```
                { $! - } RESET(participant_file) { $! + };
```

```
                exist := (IOresult = 0);
```

```
                IF exist THEN
```

```
                    BEGIN
```

```
                        CLOSE(participant_file);
```

```
                        center (16,'Filename already exists. Overwrite (y/n)? ..>');
```

```
                        ch := readkey;
```

```
                        IF ch IN ['y','Y'] THEN ok := writefile;
```

```
                    END
```

```
                ELSE
```

```
                    ok := writefile;
```

```
            END
```

```
        ELSE
```

```
            OK := TRUE;
```

```
    UNTIL OK;
```

```
END;
```

```
{ ----- }
```

```
PROCEDURE enteranthrodata;
```

```
VAR ch : CHAR;
```

```
BEGIN
```

```
    cursor_size(cursor_large);
```

```
    WITH participant_info DO
```

```
        BEGIN
```

```
            CLRSCR;
```

```

center (1,'PARTICIPANTS ANTHROPOMETRIC DATA');
PUTS (2,3,'PARTICIPANTS NAME : ');GOTOXY (21,3); WRITE(name);
PUTS (45,3,'DATE (dd/mm/yy) : '); GOTOXY (62,3); WRITE(date);
PUTS (2,4,'BIRTHDATE (dd/mm/yy) : '); GOTOXY (24,4); WRITE(bdate);
PUTS (45,4,'SEX (M or F) : '); GOTOXY (59,4); WRITE(sex);
center(6,'(all measurements are in metres besides mass in kg)');
PUTS (2,8,'TOTAL BODY MASS : '); GOTOXY (37,8);WRITE(mass:6:3);
PUTS (45,8,'TOTAL HEIGHT : ');GOTOXY (72,8);WRITE(tot_hgt:6:3);
PUTS (2,9,'MASTOID TO VERTEX LENGTH : '); GOTOXY (37,9);WRITE(mas_ver:6:3);
PUTS (45,9,'HEAD CIRCUMFERENCE : ');GOTOXY (72,9);WRITE(head:6:3);
PUTS (2,10,'TORSO LENGTH : ');GOTOXY (37,10);WRITE(torso:6:3);
PUTS (2,11,'CHEST DEPTH : ');GOTOXY (37,11);WRITE(chest_d:6:3);
PUTS (45,11,'CHEST BREADTH : ');GOTOXY (72,11);WRITE(chest_b:6:3);
PUTS (2,12,'BALL OF HUMERUS TO RADIALE LENGTH : ');GOTOXY (37,12);WRITE(hum_rad:6:3);
PUTS (45,12,'BICEPS CIRCUMFERENCE : ');GOTOXY (72,12);WRITE(biceps:6:3);
PUTS (2,13,'BALL OF HUMERUS TO VERTEX LENGTH : ');GOTOXY (37,13);WRITE(hum_ver:6:3);
PUTS (45,13,'FOREARM CIRCUMFERENCE : ');GOTOXY (72,13);WRITE(forearm:6:3);
PUTS (2,14,'STYLION TO METACARPAL III LENGTH : ');GOTOXY (37,14);WRITE(sty_met:6:3);
PUTS (45,14,'RADIALE TO STYLION LENGTH : ');GOTOXY (72,14);WRITE(rad_sty:6:3);
PUTS (2,15,'HAND BREADTH : ');GOTOXY (37,15);WRITE(hand_b:6:3);
PUTS (45,15,'HAND DEPTH : '); GOTOXY (72,15);WRITE(hand_d:6:3);
PUTS (2,16,'THIGH LENGTH : ');GOTOXY (37,16);WRITE(thigh_l:6:3);
PUTS (45,16,'MID - THIGH CIRCUMFERENCE : ');GOTOXY (72,16);WRITE(thigh_c:6:3);
PUTS (2,17,'CALF LENGTH : ');GOTOXY (37,17);WRITE(calf_l:6:3);
PUTS (45,17,'CALF CIRCUMFERENCE : ');GOTOXY (72,17);WRITE(calf_c:6:3);
PUTS (2,18,'LATERAL MALLEOLUS HEIGHT : ');GOTOXY (37,18);WRITE(lat_mal:6:3);
PUTS (45,18,'FOOT BREADTH : ');GOTOXY (72,18);WRITE(foot_b:6:3);
PUTS (2,19,'BACK OF FOOT TO LAT MAL LENGTH : ');GOTOXY (37,19);WRITE(foot_p:6:3);
PUTS (45,19,'FOOT LENGTH : ');GOTOXY (72,19);WRITE(foot_l:6:3);
center(25,'Move with cursor. Select with + OR return. Esc when finished');
glbesc := FALSE;
pntr := 1;
REPEAT
CASE pntr OF
1 : getstr(21,3,name);
2 : getstr(62,3,date);
3 : getstr(24,4,bdate);
4 : getstr(59,4,sex);
5 : getparm(37,8,mass);
6 : getparm(72,8,tot_hgt);
7 : getparm(37,9,mas_ver);
8 : getparm(72,9,head);
9 : getparm(37,10,torso);
10: getparm(37,11,chest_d);
11: getparm(72,11,chest_b);
12: getparm(37,12,hum_rad);
13: getparm(72,12,biceps);
14: getparm(37,13,hum_ver);
15: getparm(72,13,forearm);
16: getparm(37,14,sty_met);
17: getparm(72,14,rad_sty);
18: getparm(37,15,hand_b);
19: getparm(72,15,hand_d);
20: getparm(37,16,thigh_l);
21: getparm(72,16,thigh_c);
22: getparm(37,17,calf_l);
23: getparm(72,17,calf_c);
24: getparm(37,18,lat_mal);
25: getparm(72,18,foot_b);
26: getparm(37,19,foot_p);
27: getparm(72,19,foot_l);

```

```

END; {Case}
IF keypressed THEN
BEGIN
  ch := readkey;
  CASE ord(ch) OF
    uparrow : pntr := pntr - 2;
    dntarrow : pntr := pntr + 2;
    ltarrow : pntr := pntr - 1;
    rtarrow : pntr := pntr + 1;
  END;
  IF pntr < 1 THEN pntr := 27;
  IF pntr > 27 THEN pntr := 1;
END
ELSE
BEGIN
  CASE ch OF
    #27 : glbesc := TRUE;
  END; {Case}
END;
UNTIL glbesc;
END;
cursor_size(cursor_small);
END;

```

{-----}

```

PROCEDURE anthrodata;

```

```

VAR fin : BOOLEAN;
    ch : CHAR;

```

```

BEGIN
  fin := FALSE;
  REPEAT
    clrscr;
    center(1,'ANTHRODATA PROCEDURE');
    message(25,1:load from file 2:save to file 3:edit/view data 4:finished');
    ch := readkey;
    CASE ch OF
      '?' : help('anthro.hlp');
      '1' : BEGIN
        loadanthrofile;
        anthroflag := TRUE;
        END;
      '2' : saveanthrofile;
      '3' : enteranthrodata;
      #27,'4' : fin := TRUE;
    END;
  UNTIL fin;
  IF anthroflag THEN calcbodysegparameters;
END;

```

{-----}

```

PROCEDURE kinematics;

```

```

VAR ch : CHAR;

```

```

BEGIN
  REPEAT
    clrscr;

```

```

message(1,'KINEMATICS PROCEDURE');
message(25,'1:anthrodata 2:microneyedata 3:plotdata Esc OR 9:Quit');
CH := READKEY;
CASE ch OF
    '?' : help('anthro.hlp');
    '1' : anthrodata;
    '2' : getmicroneyedata;
    '3' : plotangles;
END; {Case}
UNTIL (ch = '9') OR (ord(ch) = 27);

END;

{-----}

FUNCTION beta(phi : REAL) : REAL;

VAR dphi : REAL;

{This function is the result of plotting a graph of beta vs phi & then
using the method of least squares to obtain an equation for the curve}

BEGIN
    dphi := deg(phi);
    beta := rad(-19369.4003770000+725.3039426200*dphi-10.6500182620*dphi*dphi+
        0.0770989140*dphi*dphi*dphi-0.0002754250*dphi*dphi*dphi*dphi+
        0.0000003892*dphi*dphi*dphi*dphi*dphi);

    IF phi = 0 THEN beta := 0;

END;

{-----}

FUNCTION zero(p_grf,alpha,phi : REAL) : REAL;

VAR
    z      : REAL;
    tbeta  : REAL;
    ch     : CHAR;

{This function is the result of taking moments about the ankle
for each body segment & equating them to zero for equilibrium}

BEGIN
    tbeta := beta(phi);
    WITH participant_info DO
    BEGIN
        z      := thigh_l*sin((pi/2)+alpha-tbeta)-calf_l*cos(alpha);
        x_foot := 0.44*foot_l-foot_p;
        x_calf := 0.58*calf_l*cos(alpha);
        x_thigh := 0.39*thigh_l*sin((pi/2)+alpha-tbeta)-z;
        x_hat  := c_hat*sin((pi/2)+tbeta-phi-alpha)-z;
        zero   := -(GRF*p_grf)+x_foot*m[8]+x_calf*m[7]+
            x_thigh*m[6]+x_hat*0.5*m[9];
    END;

END;

{-----}

```

```
FUNCTION calcphi(alpha,p_grf : REAL) : REAL;
```

```
VAR tphi : REAL;
    error : BOOLEAN;
```

```
BEGIN
```

```
    tphi := rad(180);
    error := FALSE;
    IF zero(p_grf,alpha,tphi) > 0 THEN
    BEGIN
        WHILE zero(p_grf,alpha,tphi) >= 0 DO
        BEGIN
            tphi := tphi - rad(5);
            IF tphi < rad(80) THEN error := TRUE;
        END;
        IF NOT(error) THEN
        BEGIN
            WHILE zero(p_grf,alpha,tphi) < 0 DO
                tphi := tphi + rad(0.5);
            WHILE zero(p_grf,alpha,tphi) > 0 DO
                tphi := tphi - rad(1/60);
            calcphi := tphi;
        END
        ELSE
            calcphi := 0;
        END
    ELSE
    BEGIN
        calcphi := 0;
    END;
END;
```

```
{-----}
```

```
PROCEDURE torquecalc(alpha,beta,phi : REAL;
    VAR tor_ankle,tor_knee,tor_hip : REAL;
    p_grf : REAL);
```

```
{calculates the torques at the hip joints: requires anthro,bsp,Pgrf}
```

```
BEGIN
```

```
    IF (beta=0) AND (phi=0) THEN
    BEGIN
        tor_ankle := 0;
        tor_knee := 0;
        tor_hip := 0;
    END
    ELSE
    WITH participant_info DO
    BEGIN
        tor_ankle := (-GRF)*p_grf + m[8]*x_foot;
        tor_knee := tor_ankle - m[7]*0.58*calf_l*cos(alpha)
            -Fay*calf_l*cos(alpha);
        tor_hip := tor_knee + Fky*thigh_l*sin(pi/2+alpha-beta)
            +m[6]*0.39*thigh_l*sin(pi/2+alpha-beta);
    END;
END;
```

```
{-----}
```

```
PROCEDURE calcangpgprf(alpha,p_grf : REAL);
```

```

VAR ialpha      : INTEGER;
    ch          : CHAR;
    esc         : BOOLEAN;

```

```

BEGIN

```

```

    arr[aalpha] := rad(alpha);
    arr[aphi]  := calcphi(rad(alpha),p_grf);
    arr[abeta] := beta(arr[aphi]);
    torquecalc(rad(alpha),arr[abeta],arr[aphi],
               arr[tankle],arr[tknee],arr[thip],
               p_grf);

```

```

END;

```

```

{-----}

```

```

PROCEDURE str_to_num(VAR st : string;  sep : CHAR;
                    VAR i  : REAL; VAR error : BOOLEAN);

```

```

{This subroutine converts the first number in a string of numbers
 to a real and truncates the string by removing first number.
 Returns error if string empty or number format incorrect}

```

```

VAR tstr  : string;
    x,y,j : INTEGER;

```

```

BEGIN

```

```

    error := FALSE;
    IF length(st) > 0 THEN
        BEGIN
            x := 1;
            WHILE (st[x] = ' ') OR (st[x] = sep) DO x := x + 1;
            st := copy(st,x,length(st)-x+1);
            y := pos(sep,st);
            IF y <> 0 THEN
                BEGIN
                    tstr := copy(st,1,y-1);
                    st := copy(st,y,length(st) -y+1);
                END
            ELSE
                BEGIN
                    tstr := st;
                    st[0] := CHAR(0);
                END;
            i := 0;
            VAL(tstr,i,j);
            IF j <> 0 THEN
                BEGIN
                    error := TRUE;
                    i := 0;
                END;
            END
        ELSE
            BEGIN
                i := 0;
            END;
        END;
    END;

```

```

{-----}

```

```

PROCEDURE plotkinetics;

```

```

VAR
  ialpha,iphi : INTEGER;
  alp,p_grf : REAL;
  i          : INTEGER;
  quit,error : BOOLEAN;
  st         : string;
  delta     : REAL;
  tfil      : TEXT;

BEGIN
  IF NOT(anthroflag) THEN
    BEGIN
      loadanthrofile;
      calcbodysegparameters;
    END;
    initgraphic;
    definewindow(1,0,10,(xmaxglb DIV 3) - 1,ymaxglb DIV 2);
    definewindow(2,xmaxglb DIV 3,10,(2*xmaxglb DIV 3) - 1,ymaxglb DIV 2);
    definewindow(3,2*xmaxglb DIV 3,10,xmaxglb,ymaxglb DIV 2);
    definewindow(4,0,ymaxglb DIV 2,(xmaxglb DIV 3) - 1,ymaxglb - 10);
    definewindow(5,xmaxglb DIV 3,ymaxglb DIV 2,(2*xmaxglb DIV 3) - 1,ymaxglb - 10);
    definewindow(6,2*xmaxglb DIV 3,ymaxglb DIV 2,xmaxglb,ymaxglb - 10);
    definewindow(7,0,ymaxglb DIV 2,(xmaxglb DIV 2) - 1,ymaxglb - 10);
    definewindow(8,xmaxglb DIV 2,ymaxglb DIV 2,xmaxglb,ymaxglb - 10);
    definewindow(9,0,0,xmaxglb,ymaxglb);

    horgrid := FALSE;
    vertgrid := FALSE;

    defineworld(1,0,-150,50,150);
    defineworld(2,0,90,50,180);
    defineworld(3,0,0,50,90);

    FOR i := 1 TO 5 DO axesdrawn[i] := FALSE;

    drawaxes(1,1,10,50,'TIME','TORQUE NM','ANKLE TORQUE',2,0,2,0,false);
    drawaxes(2,1,10,50,'TIME','TORQUE NM','KNEE TORQUE',2,0,2,0,false);
    drawaxes(3,1,10,50,'TIME','TORQUE NM','HIP TORQUE',2,0,2,0,false);
    drawaxes(4,3,5,10,'TIME','ALPHA','ANKLE ANGLE',2,0,2,0,false);
    drawaxes(5,2,5,10,'TIME','BETA','KNEE ANGLE',2,0,2,0,false);
    drawaxes(6,2,5,10,'TIME','PHI','HIP ANGLE',2,0,2,0,false);
    { drawaxes(7,2,10,10,'TIME','BETA','KNEE ANGLE',2,0,2,0,false);
      drawaxes(8,2,10,10,'TIME','PHI','HIP ANGLE',2,0,2,0,false);}

    gotoxy(1,1);
    WRITELN('Files are '+anthrofile+' and '+micronfile);
    gotoxy(1,25);
    WRITELN('      P to print screen. Space to toggle screens. Any other to quit      ');

    storewindow(9);
    leavegraphic;
    clrscr;
    center(1,'LOAD ALPHA-PGRF FILE');
    center(4,'Select alpha-pgrf data file');
    REPEAT
      selectfile('b:*.apg',frec);
      IF length(frec.name)=0 THEN
        BEGIN
          gotoxy(1,25);

```

```

        WRITE(^G);
        clreol;
        message(25,'Opps! File error or Disk drive error!!!');
    END;
UNTIL length(frec.name) > 0;
ASSIGN(tfil,frec.name);
RESET(tfil);
entergraphic;
restorewindow(-9,0,0);

WITH participant_info DO
BEGIN
    i := 0;
    WHILE NOT(eof(tfil)) DO
    BEGIN
        READLN(tfil,st);
        str_to_num(st,'alp,error);
        IF error THEN WRITE(^G);
        str_to_num(st,'p_grf,error);
        IF error THEN WRITE(^G);
        calcangppgrf(alp,p_grf);
        selectworld(1);
        selectwindow(1);
        drawpoint(i,arr[tankle]);
        selectwindow(2);
        drawpoint(i,arr[tknee]);
        selectwindow(3);
        drawpoint(i,arr[thip]);
        selectworld(3);
        selectwindow(4);
        drawpoint(i,deg(arr[aalpha]));
        selectworld(2);
        selectwindow(5);
        drawpoint(i,deg(arr[abeta]));
        selectwindow(6);
        drawpoint(i,deg(arr[aphi]));
        i := i + 1;
    END;
END;
CLOSE(tfil);

quit := FALSE;
REPEAT
    ch := readkey;
    CASE ch OF
        ' ' : swapscreen;
        'P','p' : hardcopy(false,1);
        ELSE quit := true;
    END;
UNTIL quit;
leavegraphic;
END;

{-----}

PROCEDURE kinetics;

VAR {ialpha      : INTEGER;}
    ch           : CHAR;

BEGIN

```

```

REPEAT
  clrscr;
  message(1,'KINETICS PROCEDURE');
  message(25,'1:anthrodata 2:angldata 3:plotdata ESC or 9:Quit');
  CH := READKEY;
  CASE ch OF
    '1' : anthrodata;
    '2' : ;
    '3' : plotkinetics;
  END; {case}
  UNTIL (ch = '9') OR (ord(ch) = 27);
END;

{-----}

VAR fr : INTEGER;

BEGIN
  horgrid := FALSE;
  vertgrid := FALSE;
  anthroflag := FALSE;
  anthrofile := "";
  micronfile := "";
  inithelp('anthro.hlp');
  REPEAT
    clrscr;
    message(1,'RE STAND ANTHROPOMETRIC PROGRAM');
    message(2,'JULY 88 UCT BIOENG');
    message(25,'?:Help 1:Anthrodata 2:kinematics 3:kinetics Esc OR 9:Quit');
    CH := READKEY;
    CASE ch OF
      '?' : help('anthro.hlp');
      '1' : Anthrodata;
      '2' : kinematics;
      '3' : kinetics;
    END; {Case}
    UNTIL (ch = '9') OR (ord(ch) = 27);
    clrscr;
    closehelp;
  END.

```

REFERENCES

- Amis AA, Dowson D, Wright V.
1980.
Elbow joint force predictions for some strenuous isometric actions.
J Biomechanics; 13 :765-778.
- An KN, Kwak BM, Chao EY, Morrey BF.
1984(a).
Determination of muscle and joint forces: A new technique to solve the indeterminate problem.
ASME Transactions, J Biomechanical Engineering; 106 :364-367.
- An KN, Takahashi K, Harrigan TP, Chao EY.
1984 (b).
Determination of muscle orientations and moment arms.
ASME Transactions, J Biomechanical Engineering; 106 :280-282.
- Bajd T, Kralj A.
1980.
Simple kinematic gait measurements.
J Biomedical Engineering; 2 :129-132.
- Bajd T, Kralj A, Sega J, Turk R, Benko H, Strojnik P.
1981.
Use of two channel functional electrical stimulator to stand paraplegic patients.
Physical Therapy; 61(4) :526-527.
- Bajd T, Kralj A, Turk R.
1982.
Standing-up of a healthy subject and a paraplegic patient.
J Biomechanics 15; 1 :1-10.
- Bajd T, Kralj A, Turk R, Benko H, Sega J.
1983.
The use of a four-channel electrical stimulator as an ambulatory aid for paraplegic patients.
Physical Therapy; 63(7) :1116-1120.
- Bajd T, Trnkoczy A.
1979.
Attempts to optimise functional electrical stimulation of antagonistic muscles by mathematical modelling.
J Biomechanics; 12 :921-928.
- Basmajian, JV.
1978.
Muscles Alive. Their functions revealed by electromyography.
p. 241, 4th ed. Baltimore : Williams and Wilkins.
- Basmajian JV, Bentzon JW.
1954.
An electromyographic study of certain muscles of the leg and foot in the standing position.
Surg. Gynec. and Obst.; 98 :662-666.
- Benton LA, Baker LL, Bowman BR, Waters RL.
1981.
Functional Electrical Stimulation - A practical clinical guide.
Rancho Los Amigos. Rehabilitation Engineering Center. Rancho Los Amigos Hospital. Downey, California.

Bresler B, Frankel JP.

1950.

The forces and moments in the leg during level walking.

ASME Transactions; 72 :27-36.

Braun Z, Mizrahi J, Najenson T, Graupe D.

1985.

Activation of paraplegic patients by FES: Training and biomechanical evaluation.

Scand J Rehab Med; 12 :93-101.

Brindley GS, Polkey CE, Rushton DN.

1978.

Electrical splinting of the knee in paraplegia.

Paraplegia; 16 :428-435.

Burdett, Habasevich, Pisciotto, Simon SR.

1985.

Biomechanical comparison of rising from two types of chairs.

Physical Therapy; 65(8).

Chandler RF, Clauser CE, McConville JT, Reynolds HM, Young JW.

1975.

Investigation of inertial properties of the human body.

AMRL-TR-74-137, Wright Patterson Air Force Base, Ohio.

Chao EYS.

1986.

Biomechanics of the human gait.

Frontiers in biomechanics. Edt. Schmid-Schonbein GW, Woo SL, Zweifach BW, Springer Verlag, New York.

pp 225-244 In: Schmid-Schonbein GW, Woo SL, Zweifach BW eds. Frontiers in Biomechanics.

New York : Springer Verlag.

Chao EYS, An KN.

1978.

Graphical interpretation of the solution to the redundant problem in biomechanics.

ASME Transactions, J Biomechanical Engineering; 100 :159-167.

Chao EYS, Rim K.

1973.

Application of optimisation principles in determining the applied moments in human leg joints during gait.

J Biomechanics; 6 :497-510.

Chao EYS, Kwan R.

1973.

Application of optimisation principles in determining the applied moments in human leg joints during gait.

J Biomechanics; 6 :497-510.

Chizeck HJ, Marsolais EB, Kobetic R.

1984.

Closed-loop controller design of neuroprosthetics for functional walking in paralyzed patients.

Paper presented at 9th World Congress of the international

Crowninshield RD.

1978.

Use of optimization techniques to predict muscle forces.

ASME Transactions, J Biomechanical Engineering; 100 :88-92.

Crowninshield RD , Brand RA.

1981.

A physiologically based criterion of muscle force prediction in locomotion.

J Biomechanics; 14(11) :793-801.

Crowninshield RD, Johnston RC, Andrews JG, Brand RA.

1978.

A biomechanical investigation of the human hip.

J Biomechanics; 11 :75-85.

Davies DV, Davies F (Editors).

1962.

Gray's Anatomy.

pp 679-712, 23rd ed. Longmans, Green and Co Ltd.

De Luca CJ.

1979.

Physiology and mathematics of myoelectric signals.

IEEE Trans Bio Eng; 26(6), :313-325.

Ghista DN, Toridis TG, Srimivasan TM.

1976.

Human gait analysis: determination of instantaneous joint reactive forces, muscle forces, and the stress distribution in bone segments - II.

Biomedisinsche Technik 21, 3.

Guyton A.

1981.

Textbook of medical physiology.

WB Saunders Company.

Hardt DE.

1978.

Determining muscle forces in the leg during normal human walking - an application and evaluation of optimization methods.

ASME Transaction, J Biomedical Engineering; 100 :72-78.

Hart DL, Stobbe TJ, Till CW, Plummer RW.

1984.

Effect of trunk stabilization on quadriceps femoris muscle torque.

Physical Therapy; 64(9) :1375-1380.

Havanan EP.

1964.

A Mathematical Model of the human body.

AMRL Technical Report, 64-108, Wright Patterson Air Force Base, Ohio.

Hefftner, G.

1986.

The electromyogram as a control signal for functional electrical stimulation.

MSc Dissertation, Dept Biomedical Engineering, University of Cape Town Medical School, South Africa.

- Hewitson B.
1987.
Electromyograph amplifier.
Internal Publication, Dept Biomedical Engineering, University of
Cape Town Medical School, South Africa.
- Hogan N.
1976.
A review of the methods of processing EMG for use as a
proportional control signal.
Biomedical Engineering; 11 :81-86.
- Hogan N, Mann RW.
1980.
Myoelectric signal processing: Optimal estimation applied to
electromyography - Part II: Experimental demonstration of
optimal myoprocessor performance.
IEEE Transactions, Biomedical Engineering; 27(7) :396-410.
- Holman JP, Gajda WJ.
1978.
Experimental methods for engineers.
pp 74-77, 3rd ed. Tokyo: McGraw-Hill Kogakusha, Ltd.
- Isakov E, Mizrahi J, Najenson T.
1986.
Biomechanical and physiological evaluation of FES-activated
paraplegic patients.
J Rehabilitation Research and Development; 23(3); :9-19.
- Jaeger RJ.
1986.
Design and stimulation of closed-loop electrical stimulation
orthosis for restoration of quiet standing in paraplegia.
J Biomechanics; 19(10) :825-835.
- Kao C, Bunge R, Reier F.
1983.
Spinal Cord Reconstuction.
New York : Raven Press.
- Kelley DL, Dainis A, Wood GK.
1976.
Mechanics and muscular dynamics of rising from a seated position.
pp 127-133 In: Komi PV, ed. Biomechanics V-B. Proceedings of the
5th International Congress of Biomechanics, Jyvaskyla, Finland.
Baltimore MD. University Park Press.
- King AI.
1984.
A review of biomechanical models.
J Biomechanics; 106 :97-104.
- Kobayashi K, Matsui H.
1976.
Stability of standing posture against an external force.
pp 121-126 In: Komi PV, ed. Biomechanics V-B. Proceedings of the
5th International Congress of Biomechanics, Jyvaskyla, Finland.
Baltimore MD. University Park Press.

Komi PV.

1973.

Relationship between muscle tension, EMG and velocity of contraction under concentric and eccentric work.
pp 596-601 In: Desmeelt JE ed. New developments in Electromyography and Clinical Neurophysiology.

Komi PV, Buskirk ER.

1970.

Reproductibility of electromyographic measurements with inserted wire electrodes and surface electrodes.
Electromyography; 4 :357-367.

Kralj A, Bajd T, Turk R.

1980.

Electrical stimulation providing functional use of paraplegic patient muscles.
Med Progr Technol; 7 :3-9

Kralj A, Bajd T, Turk R, Benko H.

1986.

Posture Switching for prolonging Functional Electrical Stimulation Standing in Paraplegic patients.
Paraplegia; 24 :221-230.

Kralj A, Bajd T, Turk R, Krajnik T, Benko H.

1983.

Gait restoration in paraplegic patients: A feasibility demonstration using multichannel surface electrode FES.
J Rehab Res Dev; 20 :3-20.

Kralj A, Jaeger RJ.

1982.

Posture switching enables prolonged standing in paraplegic patients functionally electrically stimulated.
5th Annual conference on Rehabilitation Engineering , Houston, Texas.

Kralj A, Vodovnik L.

1977(a).

Functional electrical stimulation of the extremities: Part 1.
J Medical Engineering and Technology : 12-15.

Kralj A, Vodovnik L.

1977(b).

Functional electrical stimulation of the extremities: Part 2.
J Medical Engineering and Technology: 75-80.

Kwatny E, Thomas DH, Kwatny HG.

1970.

An application of signal processing techniques to the study of myoelectric signals.
IEEE Transactions, Biomedical Engineering; 17 :303-312.

Liberson NT, Holmquest H, Scott D Dow M.

1961.

Functional electrotherapy: Stimulation of the peroneal nerve synchronized with the swinging phase of the gait of hemiplegic patients.
Arch Phy Medical Rehabilitation; 1 (42) :101-105.

- Lombard WP.
1903.
The action of two-joint muscles.
American Physical Education Review; 8 :141-145.
- Lowery LL, Johnson F.
1985.
Empirical verification of a linked-segment model of rising from a chair.
pp 266-271 In: Whittle M, Harris D, eds. Biomechanical Measurement in Orthopaedic Practice; Oxford Science Publications.
- Manley MT, Solomon E.
1979.
The clinical assessment of the normal and abnormal foot locomotion.
Prosthetics Orthotics Int; 3(2) :103-110.
- Marsolais EF, Kobetic R.
1982.
Functional walking of paralyzed patients by means of electrical stimulation.
pp 60 In: Proceedings of the 7th Annual Conference on Rehabilitative Engineering.
Houston.
- Marsolais EB, Kobetic R.
1987.
Functional electrical stimulation for walking in paraplegia.
J Bone and Joint Surgery; 69-A(5) :728-733.
- Milner-Brown HS, Stein RB.
1975.
The relation between the surface electromyogram and muscular force.
J Physiology; 246 :549-569.
- Milner-Brown HS, Stein RB, Lee RG.
1975.
Synchronization of human motor units : Possible roles of exercise and supraspinal reflexes.
Electroenceph Clin Neurophysiology; 38 :245-254.
- Milner-Brown HS, Stein RB, Yemm R.
1973 (a).
Changes in firing rate of human motor units during linearly changing voluntary contractions.
J Physiology; 230 :371-390.
- Milner-Brown HS, Stein RB, Yemm R.
1973 (b).
The contractile properties of human motor units during voluntary isometric contractions.
J Physiology; 228 :285-306.
- Morrison JB.
1968.
Bioengineering Analysis of Force Actions Transmitted by the knee joint.
Biomedical Engineering; 3 :164-170.

Mortimer JT, Magnusson R, Peterson I.
 1970.
 Conduction velocity in ischemic muscle; Effect on EMG frequency spectrum .
 American J Physiology; 219 :1324-1329.

Murray MP, Drought AB, Kory RC.
 1964.
 Walking Patterns of Normal Men.
 Journal of Bone and Joint Surgery; 46A :336.

Nissan M, Davis BL, Vaughan CL.
 1988 (a).
 Segmented force plate (SFP) software package for clinical use.
 Computer Methods and Programs in Biomedicine; 26 :193-198.

Nissan M, Vaughan CL, O'Connor J.
 1988 (b).
 The micronEye motion monitor : a new tool for class and laboratory demonstrations.
 Physiology Education; 23 :38-41.

Nuzik S, Lamb R, Van Sant A, Hirt S.
 1986.
 Sit-to-stand movement pattern.
 Physical Therapy; 66(11) :1708-1713.

Paul JP.
 1967.
 Forces transmitted by joints in the human body.
 181(3) : 1-15 In : Proceedings of the Institution of Mechanical Engineers.

Pauwels F.
 1980.
 Biomechanics of the locomotor apparatus.
 pp 193-222 Berlin, Heidelberg, New York : Springer-Verlag.

Peckham HP.
 1987.
 Functional electrical stimulation: Current status and future prospects of applications to the neuromuscular system in spinal chord injury.
 Paraplegia; 25 :279-288.

Peckham PH, Mortimer JT, Marsolais DB.
 1976.
 Alteration in the force and fatiguability of skeletal muscles in quadriplegic humans following exercise induced by chronic electrical stimulation.
 Clinical Orthopedics; 114 :326.

Person RS.
 1974.
 Rhythmic activity of a group of human motoneurons during voluntary contractions of a muscle.
 Electroenceph Clin Neurophysio; 36 :585-595.

Petrofsky JS, Phillips CA, Douglas R, Lawson P.
1986.

A computer-controlled walking system: The combination of an orthosis with functional electrical stimulation.
J Clinical Engineering; 11(2) :121-123.

Petrofsky JS, Phillips CA, Larson P, Douglas R.
1985.

Computer synthesized walking. An application of orthosis and functional electrical stimulation (FES).
J Neurological and Orthopaedic Medicine & Surgery; 6(3) :219-230.

Redfield R, Hull ML.
1986.

Prediction of pedal forces in bicycling using optimization methods.
J Biomechanics; 19(7) :523-540.

Rosman N, Spira E.
1974.

Paraplegic use of walking braces.
Arch Phys Medical Rehabilitation; 55(7) :310-314.

Rydell NW.
1966.

Forces acting on the femoral head prosthesis.
Acta Orthop. Scand. Suppl. 88.

Seireg A, Arvikar RJ.
1973.

A mathematical model for evaluation of forces in lower extremities of the musculo-skeletal system.
J Biomechanics; 6 :313-326.

Seireg A, Arvikar RJ.
1975.

The prediction of muscular load sharing and joint forces in the lower extremities during walking.
J Biomechanics; 8 :89-102.

Soderberg GL, Cook TM.
1984.

Electromyography in Biomechanics.
Physical Therapy; 64(12) :1813-1820.

Soderberg GL, Dostal WF.
1978.

Electromyographic study of three parts of the gluteus medius muscle during functional activities.
Physical Therapy; 58(6) :69-72.

South African Bureau of Standards.
1985.

Standard specifications for wheelchairs (folding type).
SABS 1060.

Trnkoczy A, Bajd T, Malezic M.
1976.

A dynamic model of the ankle joint under functional electrical stimulation in free movement and isometric conditions.
J Biomechanics; 9 :509-519.

Vaughan, CL.

1980.

An optimization approach to closed loop problems in biomechanics.
Iowa: Physical Education, University of Iowa, Ph.D. dissertation.

Vodovnic L.

1981.

Therapeutic effects of FES of extremities.
Med and Biol Eng Comput ; 19 :470.

Vodovnik L, Stanic U, Kralj A, Acimovic R, Gracanin F et al.

1977.

Functional electrical stimulation in Ljubijana; Projects,
Problems, Perspectives.

pp 39-53 In: Hambrecht FT, Reswick JB eds. Functional Electrical
Stimulation: Applications in neural prosthesis.

New York and Basel. Marcel Dekker Inc.

Wadsworth H, Chanmugan APP.

1980.

Electrophysical agents in physiotherapy.
pp 285-291, Marrickville : Science Press.

Williams JF, Svensson NL.

1968.

A force analysis of the hip joint.
Bio-Medical Engineering: 365-370.

Winter, DA.

1979.

Biomechanics of Human Movement.
pp 127-148, New York : John Wiley and Sons.

Winters JM, Bagley AM.

1987.

Biomechanical Modelling of muscle-joint systems.
IEEE Engineering in Medicine and Biology magazine; 1987 :17-21.