Mechanical energy transformations and energy expenditure in running man

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MECHANICAL ENERGY TRANSFORMATIONS AND ENERGY EXPENDITURE

IN RUNNING MAN

..............................................

by

Martyn Robert Shorten

A Doctoral Thesis

Submitted in partial fulfilment of the requirements for the award of the degree of Doctor of Philosophy of Loughborough University of Technology

1st October 1984

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PREFACE

Unless otherwise indicated by acknowledgments, footnotes or references to published literature, the work contained herein is that of the author. Parts of this study have been published as follows:

Published Papers


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***

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***

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ABSTRACT

It has been suggested that the lower oxygen consumption of some running athletes may be caused by differences in "running style".

In an initial study of treadmill running, segmental potential and kinetic energy changes were determined using a three-dimensional fifteen-segment rigid body model of the human body. Energy expenditure was determined by expired air analysis. The more economic running patterns were characterised by variations in total body energy of lower amplitude and greater exchange of energy within and between body segments.

The analytical procedures were developed in several ways. An automated system for the breath by breath monitoring of respiratory function and energy expenditure was developed. Since expired air analysis only enables the direct measurement of the aerobic component of energy expenditure, the validity of a commonly used method for the detection of the "anaerobic threshold" from respiratory responses was investigated. The validity of this indirect method was not supported.

A generalised energy analysis procedure was developed, allowing constraints on passive energy exchange to be varied.

A method for the determination of the elastic compliance of the knee extensor muscles was devised and used to incorporate a strain energy component into the energy analysis. In a further analysis of ten athletes, energy storage in the elastic components of the knee extensors was found to be significant during the support phase of the running stride. The inclusion of the elastic components resulted in a significant reduction of the magnitude of changes in the whole body energy curve even though the sum of the absolute changes in the partitioned energy components increased.

It was found that there is some correspondence between the magnitude of passive energy transfers and the "economy" of a running style. Also, muscle elasticity appears to act as an energy conserving mechanism during the support phase, reducing both the amount of work and the work-rate required of the extensor muscles. The additional energy transfers due to elastic energy storage may account for the unusually high efficiency values previously reported for running.
Chapter 1

INTRODUCTION
In 1914, Bernarr MacFadden, offered the following advice on form to aspiring runners:

"The novice should not go plunging, diving, stumbling forward with head hanging downward. Nor should he go high-stepping along with head thrown back, gazing into the celestial spaces above him. The various members of the body should not flop and fly about aimlessly, but should be kept steady, though without stiffness."

(MacFadden's Encyclopaedia of Physical Culture, Volume II, p 932)

Seventy years later, it could be argued that our understanding of the fundamental biomechanical principles involved in skillful running has only slightly improved.

Running is a very complex movement pattern. At the same time, it is a highly evolved skill, acquired instinctively at an early age. Perhaps because it is such a primitive movement, running athletes have in the past paid little attention to skill development. Athletes in "technical" sports strive continuously to improve the effectiveness and efficiency of their movement patterns. Runners train for "fitness" rather than technical proficiency. Science too has paid more attention to the physiological and metabolic responses and adaptations of man to running than to any physical or biomechanical consideration.

In recent years, interest in the biomechanics of the running gait has increased. Coincidentally with the rise of competitive track and field athletics as a major sporting and media event, running has become more popular as a recreation. At the top level, sophisticated training methods have reduced the margins between winning and losing to small fractions of a percentage point. Differences in running style that were previously thought to be of only marginal importance, now take on greater significance as the margins become smaller. At the popular level, increased participation, stress injury rates and running shoe sales have all stimulated interest in the mechanisms underlying the way people run.

The simple question being asked is "Why can some people run faster than others?". The complete answer no doubt lies deeply and inextricably hidden in the complex interactions between the physical, physiological, psychological and social factors that regulate human behaviour and
performance. This study is an investigation of one aspect of the mechanical factors influencing running performance.

There is some evidence to suggest that one of the factors influencing athletic performance in middle and long distance running events is related to some "economy of movement" quantity. It has been noted that some athletes use more oxygen than others at a particular running speed. The maximum rate at which oxygen can be consumed is an important limiting factor on performance so if one athlete uses less oxygen at a given speed, he may well have an advantage over his rivals. This has been demonstrated by experiments (Conley and Krahenbuhl, 1980; Sjodin and Schele, 1982) in which the performances of athletes were found to be correlated with "running economy", expressed in terms of the oxygen consumed at a particular running speed.

As a preliminary to this study, the 1500m performances and oxygen consumption measurements of 35 middle distance athletes were collated. All the athletes had been tested in the Sports Science Laboratory at Loughborough University during 1979 and were of varying performance standards. The results of this survey are illustrated in Figure 1.1, which shows the relationship between measurements of maximum oxygen uptake ($VO_2\text{max}$), energy expenditure at a running speed of 4.47m.s$^{-1}$ (10 mph) and 1500m performance time. $VO_2\text{max}$ is a measure of the capacity of the athlete to use oxygen ("aerobic capacity"), in this diagram expressed in terms of the amount of energy that this oxygen could release in the muscles. Energy expenditure at 4.47m.s$^{-1}$ was used as a measure "running economy" and in Figure 1.1 is expressed as the quantity of energy used by the athlete to run a distance of 1 metre. The less energy the athlete uses, the more "economic" he is. In common with other experiments, the survey suggests that differences in "running economy" have an effect on performance capacity. 1500m performances were in general faster the higher the athlete's aerobic capacity. In addition the faster athletes were also more "economic". The general trend was for athletes with greater economy to be faster than those with a similar $VO_2\text{max}$ but lower economy.

It has been suggested that differences in "running economy" may be a result of differences in the running "style" or "mechanical efficiency" of athletes. If this is the case, then a more "economic" running style, i.e. one that requires a lower oxygen uptake at a given speed, may be of considerable advantage. A number of experiments have shown that energy expenditure during walking and running can be manipulated by modifying the gait pattern (Zarrugh et al, 1974, Zarrugh and Radcliffe, 1978; Cavanagh
Figure 1: Relationship between aerobic capacity, running economy and 1500m time in a group of 35 athletes (See text).
and Williams, 1982).

The purpose of this study was to investigate the relationships between the mechanics of the running gait and the energy expenditure it incurs.

"Efficiency" and "economy of movement" are popular concepts among observers of sports; more often used to describe an aesthetic quality than a mechanical quantity. It is possible, however, to observe the human body as one would a machine and to attempt to measure its "efficiency" in the same way. In this analogy, the body is seen as an energy converter, using oxygen to burn food-fuel and hence produce heat, movement and work.

The use of "energy" as a basis for observation is an appropriate one because both the fuel consumed by the human machine—food and oxygen—and its products—work and movement—can be quantified in terms of the physical units of energy.

Any muscular activity requires energy in the form of chemical substrates to be released and transformed into mechanical work and movement. The higher the muscular work rate involved in the activity, the greater the rate at which the energy must be supplied. When the activity is light or moderate, enough energy can be supplied by "aerobic" processes which use oxygen to break down the chemical substrates. Under these circumstances, measurements of an athlete's oxygen consumption can be used as an indicator of the energy demands imposed on the muscles by exercise. At higher exercise intensities, the oxygen demand in the muscles exceeds the cardio-respiratory system's ability to supply oxygen and the additional energy requirement is met by "anaerobic" metabolism. This breakdown of energy substrates without the use of oxygen, incurs a "debt", however, which must later be repaid by extra oxygen consumption. Not all the energy stored in the chemical substrates is converted into muscle activity—some of it is "wasted" as heat, another form of energy.

Movement of the body can also be quantified in terms of energy. One conclusion of the very first scientific study of the biomechanics of human locomotion (Marey and Demeny, 1885a) was that there are periodic fluctuations in the body's velocity and that, as a consequence, the energy of the body varies as it moves. As muscles contract to move the body's limbs, they bring about changes in its kinetic and potential energy. Kinetic energy is directly related to the speed with which an object moves, and potential energy changes as the object moves with or against gravity. Similarly, anything the body does to apply forces to other objects, doing
mechanical work to move them, can be expressed in terms of energy changes in the external object, and measured as a physical quantity.

Thus, from one point of view, human activity can be seen as a series of changes and transformations between energy forms.

This study of the "efficiency" and "economy of movement" is, more specifically therefore, an analysis of energy transformations in running man, and in particular, an examination of the relationships between mechanical energy changes in the body during movement and energy expenditure.

A number of sub-problems are also developed:

In the past, a wide variety of methods have been used to analyse the "mechanical work" and "efficiency" of walking and running man. The results obtained are heavily dependent on the way in which energy transformations are described, i.e. on the choice of energy model. Thus one section of this study (Chapter 5) is devoted to the development of a generalised energy analysis procedure.

Another section (Chapter 6) describes the development of an automated system for analysing the air expired by exercising athletes. This system was then used to evaluate the validity of commonly used methods of detecting the onset of anaerobic metabolism "anaerobic threshold" from respiratory responses to an incremental exercise test (Chapter 7). The detection of a significant contribution of anaerobic metabolism to energy expenditure is important to this study because any energy released anaerobically cannot be directly quantified.

Since it has previously been suggested, but not demonstrated, that the storage of energy in the stretched elastic components of muscle and connective tissue may be significant in running, a further sub-study was undertaken to measure the elastic properties in the knee extensor muscles (Chapter 8). This enabled the contribution of elastic energy storage to be examined in a further study of treadmill running (Chapter 9).

The choice of "energy" as a focal point for this study is made more appropriate by the suggestion that the body naturally optimises its movement pattern in order to minimise energy expenditure - several experiments have shown that the body's naturally selected walking or running pattern at a given speed is near an energy expenditure minimum (Zarrugh et al, 1974, Zarrugh and Radcliffe, 1978; Cavanagh and Williams, 1982).
"... nature doesn’t care very much how an individual walks. Nature’s goal is to get us from here to there most efficiently, with the least expenditure of energy. This is accomplished by giving us a small sinusoidal displacement and oscillating our velocity in the line of progression. In this way, we can convert potential energy to kinetic energy and save about half the energy that would be required if this did not occur. Anything that upsets this mechanism or disturbs this beautiful, smooth displacement pattern causes an increase in energy expenditure."

V.T. Inman (1968)
Chapter 2

REVIEW OF LITERATURE
2:1 INTRODUCTION

This section is a review of three areas of the scientific literature, each of which is of general relevance to this study. These concern:

1: Running kinematics
2: The oxygen cost of running
3: "Mechanical work" and the "efficiency" of running

Further, more specific reviews of literature have been included in Chapters 6, 7 and 8.

2:1 RUNNING KINEMATICS

2:1.0 Summary

The general features of the human running gait are briefly described, together with the systematic changes in the temporal features of the running stride that occur as running speed is increased. Although the running gait has a characteristic pattern, there is inter-individual variation that can be brought about by anatomical differences, fatigue and training.

There are differences between the gait patterns adopted when running overground and when running on a treadmill, but these are generally small and may be masked by inter-individual variation if subjects are well trained. It is clear that some period of habituation to treadmill running is required.

As far as the energy analysis of treadmill running is concerned, it is necessary to invoke Galilean invariance if direct comparisons with overground running are to be made. The possibility of energy exchange between treadmill and athlete must also be considered.

2:1.1 The Running Cycle

The human biped running gait is cyclical in nature. To assist discussion of running patterns, it is useful to describe the different phases of the gait cycle in general terms. (Figure 2.1)

Unlike the walking gait there is no period of double support and the running step cycle can be divided into two generalised phases - the 'support phase' and the 'flight phase'. A support period exists while
Figure 2.1: The running cycle
either foot is in contact with the ground, the flight phase when neither foot is in contact.

A support phase begins at 'foot contact'. This point in the cycle is equivalent to 'heel contact' in the walking gait but in the running gait it is not necessarily the heel which first makes contact with the ground (Cavanagh and Lafortune, 1980). The support phase begins with a short period of amortisation during which the knee flexes and muscle activity is primarily concerned with establishing the stability of the contact (Elliott and Blanksby, 1979a, 1979b). This is followed by vigorous extension of the support leg. The effect of this extension is to propel the body forwards and upwards. The characteristic pattern of the foot-ground reaction force during the support phase (Cavanagh & Lafortune, 1980) develops at an early age (Fortney, 1983). The support phase ends with the foot of the support leg leaving the ground. If the whole plantar surface of the foot has been in ground contact then "toe-off", the point when the foot leaves the ground, is preceded by "heel off" as the heel is raised. The support phases of the two legs are distinguished by reference to "left support phase", "right support phase", etc.

The flight phase of the stride cycle is the period of non-support between alternate support phases. During this period, and the following support phase, the leg which has previously been in support becomes the "recovery leg". Characteristically, the recovery leg is flexed at the knee and recovered to a forward position. The recovery action is primarily affected by hip flexion until the middle of the contralateral support phase (Dillman, 1974). Leg extension at the knee begins at the point of initial deceleration of the thigh (Elliott, 1977) and the thigh reaches its maximum forward position at contralateral toe-off (Slocum and James, 1968).

The division of the total stride cycle time into support and flight phases varies with running speed. Typically, a single running step at distance running speeds lasts about 0.3-0.4 seconds. At maximum running speeds, the step time may be as short as 0.2-0.25s. Although some have suggested that the duration of the support phase increases slightly with running speed (Dillman, 1975) results of the type reported by Kurakin (1973) are more generally accepted, i.e. the duration of the support phase decreases significantly as running speed increases. The overall reduction in stride time is largely due to this effect. In turn, the reduction in support time is largely due to a shorter extension period (Tsujino, 1966; cited by Dillman, 1975).
2:1.2 Stride length, cadence and running speed

"Cadence" is defined as the rate at which foot contacts occur. To avoid confusion between overground and treadmill locomotion, "stride length" is defined running speed (or treadmill speed) divided by cadence.

Increased running speed can be achieved by increasing stride-length or cadence or a combination of both. In general, for overground running, increases in speed are brought about by increases in stride-length up to a certain level, beyond which speed increases are predominantly brought about by higher cadences (Cavagna et al., 1965a; Kurakin, 1972; Saito et al., 1974, Dillman, 1975, Luhtanen and Komi, 1977). Similar relationships have been reported for treadmill running (Hogberg, 1952a; Matsui et al., 1970; Hoshikawa et al., 1973; Sinning and Forsyth, 1970). There is a tendency for better runners to have longer strides and lower cadence at a given velocity (Deshon and Nelson, 1964; Dillman, 1975).

2:1.3 Inter-individual variation in the running cycle.

Clearly, individual differences in anatomical, anthropometric and physiological factors lead to substantial differences, between individuals, in the running stride. Hoffman (1971, 1972) found variations in stride length and cadence in male and female "world class" athletes to be correlated with leg length and stature, for example. The running pattern of the individual itself can vary with running speed and terrain. Fatigue causes changes in velocity, stride length and cadence, and is associated with a greater time spent in the support phase (Bates and Haven, 1974).

Adrian and Kreighbaum (1973) noted differences in head and upper body posture between runners in the "fresh" state (after 10-12 miles of running) and when "tired" (after 28-30 miles). There is evidence that the variability of individual running patterns is large. Slocum and Bowerman (1962) are able to describe three distinct running patterns associated with three individual champion athletes for example. It is suggested (Brandell and Williams, 1974) that inter-individual variation can mask statistical differences between treadmill and overground locomotion. Similarly, Bates et al. (1983) found inter-subject variability in the ground reaction force during running to be greater than variation caused by different shoe types.

2:1.4 Kinematic differences between trained and untrained runners.

Although the running gait is highly evolved, it appears to be a
trainable skill. Nelson and Gregor (1976) note significant changes in the stride pattern of athletes in training over a four year period. Elliott et al (1981) trained two groups of active non-runners over an eight week period using continuous running and games-type interval training. At the end of the training regime, both groups produced more efficient and more repeatable gait patterns during a 50 minute submaximal run. Maximum sprinting speed increased by 10%, the difference being attributed to a higher cadence achieved post-training. The different training regimes had slightly different effects but in general, the post-training gait showed a more vertical position of the leg at foot contact and a greater extension at the end of the support phase.

This increase in extension at the end of the support phase and also greater knee flexion during the earlier part of support are characteristics of better performers (Miura et al, 1970; cited by Dillman, 1975).

Kurakin (1972) suggests that "better" runners spend less time in support than "poor" runners. The durations of both knee flexion and knee extension decrease with increasing speed.

After the foot leaves the ground, hip extension continues momentarily. (Slocum and James, 1968; Dillman, 1971; Miura et al, 1973) and this is also, apparently, a feature of an effective running style - good runners maintain hip extension end then recover the leg rapidly (Miura et al, 1973). In direct contrast, Saito et al (1974) found that the trained runner flexes at the hip earlier in the recovery phase than the untrained runner. The amount of knee flexion in recovery increases with running speed with better runners showing the most knee flexion (Sinning and Forsyth, 1970). The elevation of the knee in front of the body at the end of flexed recovery also increases with running speed with trained subjects showing a greater overall range of movement (Saito et al, 1974; Hoshikawa et al, 1973).

Although it appears possible to distinguish kinematically between trained and untrained runners, the distinction between "good" and "elite" athletes seems to be more subtle. Cavanagh et al (1977) made a comprehensive kinematic comparison of samples of the two groups but found only minor differences, principally a greater stride length and greater bilateral symmetry in the "elite" athletes.
2:1.5 Overground versus treadmill running.

The continuous monitoring of overground running is made difficult by the necessity to move laboratory apparatus at speeds of up to 10 m.s⁻¹. It is experimentally convenient therefore to study athletes running on a treadmill rather than overground. It is possible, however, that the laboratory treadmill environment produces a different running action to that observed under normal conditions, limiting the extent to which extrapolations can be made from the laboratory to the real world. There is, for example, no wind resistance on the treadmill and on motorised treadmills the motor may represent an additional source of work in the man-treadmill system. Also, any external reference point used for spatial measurements becomes fixed with respect to the athlete when he runs on a treadmill with potential implications for biomechanical analyses.

A number of kinematic differences between overground and treadmill running have been observed. In a study of 16 athletes running at three different speeds in the range 3.55 to 6.4 m.s⁻¹, Nelson et al (1972) found the support phase of the running cycle to be longer in treadmill running. This increased contact time was due to an increased tendency for foot placement in front of the body’s centre of mass. The velocities of vertical centre of mass displacement were generally lower and less variable. The differences between overground and treadmill running became more distinct at higher running speeds. Nelson’s results show that the direction of change of kinematic variables is generally the same for both types of running but that the magnitude of change may be different (Dillman, 1975). Elliott and Blanksby (1976a) found no significant differences in stride length and cadence between overground and treadmill running at speeds in the range 3.3 – 4.78 m.s⁻¹ (male subjects), but differences did occur at higher speeds, in the range 4.82 – 6.82 m/s.

Dal Monte et al (1973) also found the amplitude of centre of mass displacements to be reduced in treadmill running. In contrast, however, their study of 3 subjects running at speeds in the range 4.2 to 5.6 m.s⁻¹ found that the differences between the two types of running became smaller as speed increased. Nelson’s subjects had a greater stride length on the treadmill at a given speed while Dal Monte’s subjects had a smaller stride length. Elliott and Blanksby (1976a) also found stride lengths on the treadmill to be smaller, as did Brandell and Williams (1974) for treadmill walking. The conflict between these results is not irreconcilable. Brandell and Williams (1974) found that their experimental between-subject variance of kinematic variables was greater than the between-groups variance. In
other words, the differences between overground and treadmill walking was not as great as the differences between individual subjects. The same might well be true of running, given the variation in individual running patterns.

Dal Monte et al (1973) also failed to find any outstanding differences between the two modes of running. Most importantly, their subjects were extensively treadmill trained, with over 30 hours treadmill experience. Wall & Charteris (1980) studied the long-term habituation process of naive subjects to the task of treadmill walking. Stride to stride variation in kinematic parameters reduced sharply in the first 30s of the first exposure and was followed by a longer more gradual learning process. All the major adaptations of stride patterns occurred during the first 4 ten minute exposures. While these latter results cannot be directly extrapolated to running, they do indicate the order of time required to make the perceptual adjustment between normal overground locomotion and treadmill locomotion.

The forward foot placement recorded by Nelson et al (1972) is commonly observed in athletes running on a motorised treadmill. Figure 2.2 shows computer-drawn representations of an athlete running on a treadmill at 4.47 m.s⁻¹ using data from Chapter 9 of this study. In this typical example, the forward placement of the foot at ground contact is clearly seen. The longer support period results in a need to recover the leg more quickly. Such a foot placement would cause retardation and therefore incur a penalty if used when running overground. On a motorised treadmill, the moving belt assists the rearward movement of the foot and no such penalty is incurred. On an unmotorised treadmill the reverse tends to be true - foot contact is displaced rearwards with respect to the centre of mass and contact with the belt tends to be shorter. In this latter case, the athlete is driving the treadmill belt against a resistance rather than being driven by it.

Dal Monte et al (1973) found no outstanding differences between the energy cost of running at a given speed on the track and on a motorised treadmill. This is unusual since one would expect the work rate on a treadmill to be less in view of the absence of air resistance and the potential contribution of the treadmill motor.

Pugh (1970,1971) measured oxygen uptake during track and treadmill running and estimated the additional oxygen cost required to overcome air resistance. This energy cost for a track runner, was found to be 7.5% of the total energy cost at middle distance running speeds and 13% at sprinting speeds. The energy expenditure of an athlete running in the wake
Figure 2.2: Movements of the arm and leg during a treadmill running stride showing forward placement of the foot at ground contact.
of another (1 metre behind) was reduced by 6.5% at middle distance speeds. At world record sprinting speeds (~10 m.s\(^{-1}\)), the power required to overcome air resistance has been calculated as ranging between 250 and 370 W (Shanebrook and Jaszczak, 1976). Since aerodynamic drag force is proportional to the square of running velocity, the work rate of the drag force is proportional to velocity cubed. A runner could therefore be expected to dissipate only 30 – 40 W at a speed of 5 m.s\(^{-1}\). At an efficiency of 36% for running against an additional horizontal impending load (Lloyd and Zacks, 1972) the aerodynamic drag at 5 m.s\(^{-1}\) should incur a metabolic energy cost of 80-100 Watts, about 10% of total net energy expenditure at this speed. On the treadmill, overall air resistance is much less and one would expect the net energy cost to be reduced, therefore. Reduced energy expenditure was not observed in treadmill runners by Dal Monte at al (1973). Ralston (1960) reports a 10% lower energy expenditure for treadmill walking when compared with walking on an asphalt road. In this case, lack of air resistance would only account for a reduction of the order of 2 Watts (< 1% of net energy expenditure) however, on the basis of Shanebrook and Jaszczack’s results.

The treadmill itself may also contribute to the energy transformations during running. Woolley and Winter (1979) report no significant differences between the magnitudes of mechanical work in subjects walking overground and those walking on a treadmill. Between and within segment energy transfers were significantly higher overground however. Winter et al (1980) measured fluctuations in the electrical current to a motorised treadmill on which subjects were walking. Motor current rose sharply, during heel contact and fell to a minimum during plantar flexion before toe off. It would thus appear that the treadmill was absorbing work done during heel strike by forces tending to slow the treadmill down. Conversely, during the extension phase, the walker’s movement tends to speed up the motor and the current requirement is lower. While the overall power output of the treadmill may be no greater than when under static load (this appears to be the case in Winter’s examples) and no net contribution is made to the energy of the subject, there is evidence to suggest that some exchange of energy may occur between the subject and the treadmill during the walking stride.

This problem creates part of the need for a wider consideration of the way in which the analysis of treadmill locomotion should be approached, especially when energy analyses are involved.
Newton's fifth corollary states:

"Bodies enclosed in a given space have the same motions relative to one another, whether that space be at rest or be moving in a straight line without rotation"

[Translated from Newton's "Philosophiae naturalis principia mathematica" (1687), after Feather (1970)]

The laws of classical mechanics are thus formulated in such a way that forces will always be the same regardless of whether the frame of reference is at rest or moving at constant velocity with respect to the observer. This is the principle of Galilean invariance. While the forces in a static frame of reference are not different from those in a reference frame undergoing rectilinear motion at constant velocity the same is not true of potential and kinetic energies, which are dependant on the coordinate system used.

Ingen Schenau (1980) has considered this problem in relation to treadmill locomotion. The kinetic energies of the limbs in the horizontal plane depend on whether the selected reference frame is static in relation to the earth's surface or is considered to move with the treadmill belt. In order to reconcile these differences with the laws of energy conservation, changes in the momentum of the man-treadmill-earth system in each reference frame must also be taken into consideration. These calculations can be eliminated by selecting a reference frame which moves with the surface against which the subject is moving.

Since the different reference frames can yield different kinetic energy values, it is important that any study of treadmill locomotion involving energy calculations should clearly state whether the reference frame used was moving or static with respect to the earth. If a reference frame that moves with the treadmill belt is selected, kinetic energies can be compared directly with those calculated for overground running from a static camera. As far as energy exchange between the subject and the treadmill are concerned (Winter et al, 1980), it is equally true, theoretically that some energy exchange occurs between man and earth during overground walking and running. The changes in the velocity of the earth are immeasurably small however. Analytical difficulties should arise, therefore, only if the the speed of the treadmill belt varies substantially when the foot is in contact with it. This problem is minimised when a
large, rigid treadmill with very powerful motor is used.

It would appear that while there may be differences between overground and treadmill running kinematics they are small enough to be outweighed by the advantages of having experimental subjects quasi-static in the laboratory. If, as is often the case, only one stride is to be analysed in detail, the problems of recording the movement itself are not greatly reduced by using a treadmill. At the same time, treadmill running is probably better represented by a single stride sampled from a continuous treadmill run than overground running is represented by a stride taken from a sequence where the subject is running over a short distance or is constrained to making contact with a force platform. Where continuous recording of biomechanical variables is required, or where steady state expired gas collections and blood samples are to be made simultaneously, the treadmill becomes a practical and manageable simulator. Most importantly, a laboratory treadmill allows running speed to be controlled with some precision. Williams and Cavanagh (1983) compromised by measuring oxygen consumption on a treadmill and making biomechanical measurements on the same subjects running overground. Similarly, Luhtanen and Komi (1978) determined the work rates of athletes running on a treadmill, but added calculated terms for the work that would theoretically have been required to overcome air resistance had the athletes been running on the track. If there are differences in the energy costs of the two modes of running at a given speed then experimental arrangements of this type could lead to confusion.
THE OXYGEN COST OF RUNNING

2.2.0 Summary

Over a large part of the normal range, the oxygen cost of running is a rectilinear function of running speed. There is inter-individual variation in the oxygen cost of running at a particular speed, but much of this variation is eliminated when body-weight adjusted oxygen uptakes are used. The residual variance in oxygen cost has been described as inter-individual differences in "running economy". There is some evidence to suggest that differences in running economy may be a determinant of running performance.

Changing gait patterns by imposing fixed cadences produces significant changes in oxygen consumption at a given speed. Differences in running economy may therefore be due, at least in part, to differences in "running style".

2.2.1 The oxygen cost of running

Some investigators have reported a curvilinear relationship between the oxygen cost of treadmill running and running velocity (Boje, 1944; Knuttgen, 1961). While non-linearities may occur at very low running speeds and at high running speeds where anaerobic metabolism makes a significant contribution to the energy supply, the oxygen cost of running is now generally accepted as being a rectilinear function of running speeds over much of the velocity range. (e.g. Christensen and Hogberg, 1950; Hogberg, 1952a, 1952b, Margaria et al, 1963; Davies and Thompson, 1979; Williams and Nute, 1983). Figure 2.3 shows for example the oxygen uptake responses of a single subject (an Olympic finalist at 5000m) over a range of running speeds from 6 mph to 14 mph (2.7 to 6.3 m.s\(^{-1}\)) determined using the procedures described in Chapter 3. In the range 8 - 13 mph (3.6-5.8 m.s\(^{-1}\)) the data are well described by a rectilinear function.

There is considerable inter-individual variation in oxygen uptake at a given running speed but much of this variation is accounted for by expressing the oxygen cost relative to body weight. All studies of groups of subjects inevitably report some variation in the body-weight corrected oxygen cost of running at a given speed however, coefficients of variation (standard deviation/mean * 100%) of the order of 5-6% being typical (Howley and Glover, 1974; Keren et al, 1981; Williams and Nute, 1983). Even when data is expressed in this body-weight corrected form there is a residual variance in "running economy" therefore (e.g. Costill and Fox, 1969; Daniels, 1974; Daniels et al, 1977; Farrell et al, 1979; McMiken and
Figure 2.3: Example of oxygen consumption over a range of running speeds

Data from a single subject, an international 1500m/5000m runner.
Daniels, 1976; Pollock, 1977). Sjodin and Schele (1979) for example report a range of 15mls.kg\(^{-1}\).min\(^{-1}\) in the recorded values of oxygen uptake at a running speed of 15 kph (4.2 m.s\(^{-1}\)). Williams and Nute (1983) report only a 5.5 ml.kg\(^{-1}\).min\(^{-1}\) range in 10 subjects at a similar speed, but this still represents about 10% of the mean oxygen uptake at that speed. The relatively low correlations (0.73 to 0.84) between oxygen consumption and running velocity in different groups of athletes reported by Mayhew et al. (1979) also indicate large inter-individual variation.

There is evidence to suggest that these variations in submaximal oxygen consumption influence performance. Sjodin and Schele (1982) report a high correlation (p < 0.05) between "running economy" (expressed as the oxygen requirement at a speed of 15kph) and 5000m performance. On the other hand, Williams and Nute (1983) found no such relationship between running economy and half-marathon performance; and Davies and Thompson (1979) report that 98% of the between-subjects variance in marathon performance could be accounted for by differences in maximum oxygen uptake (VO\(_{2}\)max) and the proportion of VO\(_{2}\) max at which the run was performed.

In view of the importance of VO\(_{2}\)max as a limiting factor on running, and since variations in VO\(_{2}\) are greater than variations in running economy, it is important to try and distinguish the effects of different running economies from the effects of different VO\(_{2}\) max values alone. Conley and Krahnenbuhl (1980) compared 12 highly trained athletes of similar ability. At each of three steady running speeds, "running economy" was highly and significantly correlated with performance in a 10km race. Overall, 65% of the variance in race performance was accounted for by variations in running economy (submaximal oxygen uptake at a set speed). Significantly, the subjects for this experiment had similar VO\(_{2}\) max values. Therefore, while VO\(_{2}\)max may be an important primary determinant of performance capacity in the population as a whole, it appears that among highly trained athletes of similar ability and VO\(_{2}\) max, running economy is a significant determinant of performance.

### 2:2.2 Biomechanical influences on oxygen cost

Differences in "running style" may influence running economy. Costill and Fox (1969), for example, suggest that inter-individual differences in the oxygen cost of running at submaximal speeds may be partly due to variations in "mechanical efficiency". Similarly, Ariyoshi et al. (1979) consider differences in "running technique" to be the basis for the lower oxygen consumption of elite marathon runners that has been reported in a
number of studies (e.g. Brandsford and Howley, 1977; Costill and Winrow, 1970; Fox and Costill, 1972).

These variations in "efficiency" might equally be the result of differences in the efficiency with which energy is released at the metabolic level as of differences in the "mechanical efficiency" of running styles. However, there is evidence to suggest that the way in which a person runs or walks at a given speed influences the oxygen cost of the activity. This can be shown by varying cadence at a fixed walking or running speed. Hogberg (1952a, 1952b) and Knuttgen (1961), both using a single subject, demonstrate that running at stride-lengths greater or smaller than that freely selected results in a greater oxygen cost. Increasing stride length incurred a greater penalty than increasing cadence. Cotes and Meade (1960) report similar effects for walking.

Cavanagh and Williams (1982) studied the effects of varying cadence on the oxygen cost of running in a group of 10 trained male runners. Individual relationships between relative stride length (expressed as a percentage of leg length) were well fitted by a quadratic function with a minimum near the self-selected stride length. Intra-individual differences in oxygen consumption produced by small changes in stride length were small but the oxygen cost rose significantly as stride lengths moved further away from the optimum. Zarrugh et al (1974), Zarrugh and Radcliffe (1978) and Zarrugh (1981) similarly report the effects of varying cadence and stride length on the oxygen cost of treadmill walking. Again, at a given speed, each subject had an optimal stride length at which energy expenditure is a minimum. Variations in the stride length and cadence away from this optimum result in an increased oxygen cost.

On an inter-individual basis with athletes running at self-selected cadences, Alexander and Theissen (1983) found no significant relationship between stride length, cadence and oxygen consumption.

In both walking and running, an individual's self-selected stride-length/cadence combination is near to that producing the minimum energy expenditure at a given running speed. This suggests that either there has been adaptation to the chosen stride length OR that successful minimum energy optimisation of the gait pattern to physiological and musculo-skeletal capacities and limitations has occurred. Nelson and Gregor (1976) studied a group of athletes in training over a four year period. Self-selected stride lengths changed significantly during that period suggesting the possibility of biomechanical adaptation to the changes in
2:3 MECHANICAL WORK AND THE EFFICIENCY OF RUNNING

2:3.0 Summary

The known overall efficiency of the conversion of chemical energy to mechanical work in muscle is in the range 20-30%. In biomechanically simple activities, such as cycling against a frictional load the measured efficiency of the body in performing work is of a similar magnitude.

Attempts to determine an "efficiency" value for walking and running analyses have led to a wide variety of methods being employed.

Point mass models of the human body fail to take into account the energy of reciprocating limbs, and the rotational component of kinetic energy cannot be calculated. Similarly, intra- and inter- segmental transfers cannot be studied using these models. The use of multiple link-segmental models permits a more complete energy analysis to be performed, but the calculation of the mechanical work done by the body is dependent on the assumptions made about between segment energy transfer. The energy models on which these "pseudo work" analyses are based have not been clearly defined, and some important energy components, including elastic energy storage in muscle and connective tissue and heat generation have not been included in the analysis.

Depending on the methods selected, mechanical efficiencies for running in the range -120% to +200% have been determined.

2:3.1 Work, energy and power

A force $F$ acting on a body over distance $s$ does work $W$

$$ W = \int F \, ds \quad \text{Eqn 2.1} $$

The work done is equivalent to a change, $\Delta E$ in the total energy, $E$, of the body. The mechanical work done on a simple body can therefore be determined
by analysis of the forces acting, or by examination of the changes in the energy of the body.  

"Negative work" is a term often used to describe the work quantity when the change in energy is negative. Strictly speaking, energy is a scalar quantity and the reference to negative work therefore represents a change in reference point. The reference system does "positive work", but any work done when the system is resisting external forces is termed "negative work". Thus the term "negative work" appears in the literature in association with, for example, running downhill, eccentric muscle contractions and downward changes in body total energy curves.

"Power flow" is another term that is commonly used (e.g. Quanbury et al, 1975) and which may cause confusion. Power is the rate at which work is done, and is equivalent to rate of change of energy or "energy flow". Quanbury et al (1975) calculate the "Instantaneous Power" of the lower leg as the rate of change of the total segment energy. Similarly, "Power flow to the shank" is calculated from the product of joint forces and moments with their respective velocities and angular velocities. "Power" and "Power flow" in this case have the same units and ultimately refer to the same quantity calculated by different routes. Thus "work" or "energy flow" might be an appropriate substitution for "Power flow".

2:3.2 Efficiency

Efficiency is a calculated ratio relating the usefulness of a process to some cost function. "Mechanical efficiency" is normally used to indicate the efficiency of a system in which there are energy inputs and outputs. The ratio in this case is that of useful work or energy output to energy input.

The efficiency with which muscles do work can be expressed in a wide variety of ways. Each step or process in an energy conversion pathway may be given an efficiency, based on the proportion of the useful energy available at the beginning of the process that is still usefully available when the step is completed. For example, only a proportion of the energy released by oxidation of carbohydrates and fats in muscle is usefully conserved in ATP. The actual efficiency of this "phosphorylative coupling" process (Whipp and Wasserman, 1969a) is in the range 40-60% (Lehninger, 1970). The next step in the conversion of food-stuff energy into mechanical work is the generation of work using the energy conserved in ATP. The highest estimate of the efficiency of this process, the "contraction coupling" efficiency (Whipp and Wasserman, 1969a) is 50% (Stainsby et al, 1980). Thus the overall efficiency of muscle in converting the free
energy of oxidised foodstuffs into muscular work, the product of the phosphorlyative and contraction coupling efficiencies is of the order of 20-30%. The mechanical efficiency of human movement cannot be expected to exceed this value unless there is some external energy input into the system from a source other than muscle.

Efficiencies for human movement, generally based on the ratio of external work accomplished to the energy expended, have been expressed in a number of ways.

"Gross efficiency" generally refers to the ratio of the total external work done to total energy expenditure.

"Net efficiency" refers to the ratio of external work accomplished to the "net" energy expenditure, where net energy expenditure is the total energy expenditure of the body less some baseline value. When the baseline reflects basal metabolism, the efficiency ratio is generally called "net efficiency". The term "work efficiency" is used when the energy expenditure is calculated as the difference between that incurred in working and non-working conditions. "Delta efficiency" as the ratio of an increment in external work to the associated increment in energy expenditure and "instantaneous efficiency" as the ratio dW/dE. (See for example Gaesser and Brooks, 1975).

Other efficiency terms have been used, usually as a result of some partitioning of the total work or energy expenditure; or by distinguishing between different pathways in the transformation of energy to work. Thus "net aerobic efficiency" (Norman et al, 1976) refers to the ratio of external work to the net energy expenditure incurred via aerobic, as opposed to anaerobic, respiration. Similarly, "positive work efficiency" and "negative work efficiency" are based on a partitioning of the work output term of the overall efficiency ratio.

There is some debate as to whether the use of base-line subtractions in efficiency calculations produces valid estimates of overall muscle efficiency. Stainsby et al (1980) suggest that the base-line corrections used in typical efficiency calculations are likely to lead to error since the baseline changes as work rate increases. Higher work rates involve greater rates of ventilation and blood transport and hence greater "base-line" energy expenditure.

There has, since the earliest days of biomechanics research, been considerable attention given to the determination of a mechanical efficiency ratio for the human body as it undertakes a range of activities. In the case of the human "machine", the calculation of the denominator of
the efficiency term, the energy "used" by the body during a period of activity, can be readily determined by indirect calorimetry from measurements of oxygen consumption. It is in the determination of the numerator, the work output of the human system, that analytical difficulties arise.

2:3.3 Energy in - work out models

In simple cycling activities on cycle ergometers it is possible to directly measure energy expenditure by indirect calorimetry and to measure directly the work done against an external frictional load. "Efficiency" can be calculated from the ratio of the work output to the energy expenditure. Such experiments generally yield efficiencies similar to the overall efficiency of muscle (20 - 30%). Faria et al (1982) for example report gross efficiencies of cycle ergometry in the range 18 to 22%. Gaesser and Brooks (1975) similarly report gross efficiencies for cycling activity of up to 24.4%.

The use of the simple energy expenditure - work output calculation of efficiency works well with cycle ergometer exercise. With running and walking however, the "work output" is much less easily defined. Also, in the case of cycle ergometry, the work done in moving the body is small compared with that done against the external load and there is little exchange between potential and kinetic energies. Kinematic variations do have some influence on the calculated efficiency, as shown by the variations produced by changes of pedalling speed (Gaesser and Brooks, 1975; Faria et al, 1982) but these are probably small compared with those that occur during the running gait.

2:3.4 Point-mass models

The simplest mathematical treatment of human body mechanics models the body as a point mass located at its centre of mass. The energy of the system is thus calculated as the sum of the potential and kinetic energies of the mass. Margaria (1938) and Margaria et al (1965) computed the body's mechanical work rate from the change in the average potential energy of the body which was assumed to be progressing at constant speed. Thus subjects walking or running uphill do "positive work", those running downhill do "negative work" and no work appears to be done during locomotion on a level surface. Cotes and Meade (1960) calculated work from the "vertical lift per step", i.e intra-cycle potential energy changes which is marginally more sensible. Dean (1965) calculated both potential and kinetic energy changes
during walking. Cavagna et al (1963, 1964, 1976) used a force platform method (Cavagna 1975) to determine the apparent kinetic and potential energy of the mass centre during walking and running and calculated the work done from changes in the total (kinetic + potential energy).

Gerston et al (1969) assumed that energy changes of the body could be determined directly from the kinematics of the anatomical centre of mass and calculated work done in level walking from the output of an accelerometer attached to the torso of their subjects. Similarly, Lukin et al (1967) determined the motion of the centre of mass and hence it's potential and kinetic energy by means of displacement transducers attached by strings to a point near the subjects' anatomical centre of mass. These authors would appear to have equated a particular anatomical location with the dynamically varying centre of mass of the body. Therefore, although they have used the point mass model, its dynamics have not been correctly determined, since the velocities and displacements calculated relate to the torso, not the centre of mass.

Fukunaga et al (1977, 1980) calculated the "external mechanical power output" of eight sprinters using a force platform and a cinéphotographic analysis to determine the trajectory of the body's centre of mass. Again, the work was calculated on the basis of a point mass model and divided into the "work done against gravity" (i.e. producing potential energy changes) and "work done by forward velocity change" (i.e. producing horizontal kinetic energy changes). Kinetic energy changes in the vertical direction were ignored.

Since kinetic energy is not a vector quantity, the motion of the centre of mass alone is insufficient to calculate the total kinetic energy of the body. Reciprocal movements of the limbs, which are the rule rather than the exception in human bipedal locomotion produce energy changes which are not reflected in the movement of the centre of mass. Margaria (1938) ignores kinetic energy changes completely and in no case can a point-mass model be used to calculated the rotational component of kinetic energy.

In order to overcome the problem of energy changes due to reciprocal motion, Cavagna et al (1964) and Cavagna and Kaneko (1977) included additional kinetic energy terms in the calculation of the total mechanical energy of runners. These kinetic energies were calculated from the velocity of each limb relative to the centre of mass of the body. Smith (1975) has demonstrated that this calculation underestimates the body's total kinetic energy. On the basis of models of this type, Cavagna and Kaneko report mechanical efficiencies of 0.35 to 0.40 for walking and 0.4 to 0.80 for
running at a range of speeds. Margaria's average potential energy method (Margaria, 1938; Margaria et al, 1963) yields efficiencies of zero for level walking and running. The same method gives a calculated efficiency for downhill running of up to -1.18 (-118%). The thermodynamic details of this unusual observation have yet to be published.

Applying typical energy expenditure values to Fukunaga et al's "external mechanical power output" data gives efficiencies in the range 0.6 to 0.8. Since the transformation of chemical energy into mechanical work by muscle has a maximal overall efficiency of the order of 0.25 (Stainsby et al, 1980), these high efficiency values infringe the conservation of energy principle. Even efficiencies of the order of 25% (such as the gross efficiency of 23.9% for level walking reported by Ralston and Lukin, 1969) imply that the frictional and other energy losses are negligible. This error is undoubtedly due to the inadequacy of the point-mass models in describing the energy transformations that occur during locomotion.

2:3.5 Link segmental energy models

A more complete picture of the work-energy transformations during human locomotion is likely to emerge if individual segmental motions are considered. A link-segmental model defines the human body as a series of linked rigid bodies, each rigid body being given the inertial properties of a human limb or body segment. Such models may vary in resolution from four or five representative segments (Miller 1971) to seventeen segments (Hatze, 1977). By calculating the kinetic energies of each rigid segment of a link segmental model a more reliable estimate of the total kinetic energy can be made since the cancelling effects of reciprocal motion are eliminated and the rotational component of the kinetic energy of each segment can be included in the total.

Fenn (1930a, 1930b) used cinephotographic data to calculate the potential and kinetic (translational + rotational) segmental energies of a sprinter. He then calculated the mechanical work from the sum of the potential and kinetic energy increases of all the segments. Performed without the aid of electronic digitisers and digital computers, Fenn's analysis represents an Herculean achievement, especially in view of the fact that energy analyses based on the much simpler point-mass model were still being published 50 years later (e.g., Fukunaga et al, 1980). His method, however, takes no account of the passive transfer between potential and kinetic energy and as a result the mechanical work rate recorded is high (in excess of 2000W).
For the purposes of a link-segmental energy analysis, the total mechanical energy of the \( i \)th body segment at any instant, \( t \), is defined as the sum of its potential and translational and rotational kinetic energies:

\[
E_{i,t} = m_i g h_{i,t} + \frac{1}{2} I_i \omega_{i,t} + \frac{1}{2} m_i v_{i,t}^2 \tag{Eqn 2.2}
\]

where

- \( E_{i,t} \) = total energy of the \( i \)th segment at time \( t \)
- \( m_i \) = segment mass
- \( v_{i,t} \) = velocity of the segment centre of mass
- \( h_{i,t} \) = vertical position of the segment centre of mass
- \( I_i \) = segmental moment of inertia
- \( \omega_{i,t} \) = angular velocity of the segment about its mass centre
- \( g \) = acceleration due to gravity

The total energy of the body, \( E_B,t \) at any time, \( t \), is defined as the sum of the segmental energies:

\[
E_{B,t} = \sum_{i=1}^{5} E_{i,t} \tag{Eqn 2.3}
\]

The rotational component of segmental kinetic energy has often been ignored in energy analyses, largely because a point-mass system has no rotations. Chapman (1979) has recorded the segmental rotational kinetic energies of two female sprinters, however.

Norman et al (1976) reawakened interest in the mechanical efficiency of running by introducing the idea of "pseudo-work". Using a planar, 12-segment link-segmental model, these authors estimated the "total pseudo-mechanical work", (TPW) as the sum of the absolute changes in the instantaneous partitioned mechanical energies of each segment:

\[
W_n = \sum_{i=1}^{5} \sum_{j=1}^{n} (|\Delta PE_{i,j}| + |\Delta KE_{i,j}| + |\Delta RE_{i,j}|) \tag{Eqn 2.4}
\]

where

- \( \Delta PE_{i,j} \) = change in the potential energy of \( i \)th segment during the \( j \)th time interval
\[ \Delta KE_{i,j} = \text{change in the translational component of kinetic energy} \]
\[ \Delta RE_{i,j} = \text{changes in the rotational component of kinetic energy}. \]

The mechanical efficiency of level treadmill running was calculated in a variety of ways by combining different parts of the energy sum with physiological energy expenditures calculated in three different ways. The values reported range between 12.3 and 34%. The pseudowork method attributes all energy changes to muscular work, including passive energy transfers. This leads to fluctuations in the total body energy curve of a runner that are three to four times greater than those in the sum of the partitioned (but not absolute) values for the energy changes. Norman et al.'s data must therefore be questioned, since they indicate much lower levels of total pseudowork than can be reasonably expected. Professor Norman (pers. comm.) has confirmed that this is the case.

Gregor and Kirkendall (1978) applied Norman et al.'s method to an analysis of the efficiency of three female marathon runners at four submaximal treadmill running speeds. A planar, 11-segment kinematic model was used to determine instantaneous segmental energy changes. "Energy input" was determined from net oxygen consumption. The average "efficiency" thus determined was 25.8%. The average efficiency was highest in the most experienced runners and lowest in the least experienced runners. These efficiency values broadly agree with those reported by Norman et al. (1976). It must again be noted that these efficiencies are very similar to the efficiency values reported using point-mass models. However, Eqn 2.4, if applied correctly, would be expected to produce much greater "energy output" values than the point mass model, and hence much higher efficiencies.

Luhtanen and Komi (1978, 1980) performed a similar mechanical energy analysis of 6 male athletes running on a treadmill at speeds from 3.9 to 9.3 m.s\(^{-1}\). "Positive" work was distinguished as the sum of the positive increases in partitioned energies of Norman's pseudowork equation and "Negative" work as the sum of negative changes. (Over the whole running cycle, assuming the stride to start and end in the same kinematic state, the totals of "positive" and "negative" work calculated by this method should be equal. Luhtanen and Komi (1978) report "an average value of 1" for the ratio of positive to negative work.) The total pseudo work rate increased quadratically with running speed from 11.4 J.s\(^{-1}.kg\(^{-1}\) at 3.9 m.s\(^{-1}\) to 23.7 J.s\(^{-1}.kg\(^{-1}\) at 9.3 m.s\(^{-1}\). Luhtanen and Komi did not measure
the energy expenditure of their subjects and did not therefore report efficiency values. Assuming a net energy expenditure of 0.8 kJ.s\(^{-1}\) for an average subject at running at 3.9 m.s\(^{-1}\) gives a total pseudo work rate of 931 W. Typically, net energy expenditure at this running speed is of the order of 750 Watts, suggesting an "efficiency" in excess of 120%. This figure, though thermodynamically meaningless, is a true reflection of the energy changes determined by the pseudowork method.

The main problem with this new approach to the calculation of the body's mechanical energy is, again, that no account is taken of the possibility of passive energy transfer within and between segments. Energy changes brought about by the passive, pendular swinging of a limb in a gravitational field are attributed to muscular work. For example, the passive transfer of 1 Joule of potential energy to 1 Joule of kinetic energy results in no net change in the energy of the whole body but nevertheless results in 2 Joules of "pseudowork" being done. Thus while Norman et al's (1976) pseudowork equation may calculate a value for the total magnitude of kinematic energy changes, these changes cannot be attributed to muscular effort alone. Attempts to calculate the "efficiency" of human movement using pseudo-work calculations are futile therefore. Nevertheless, Norman et al's (1976) paper is significant in that it stimulated a great deal of new interest in the energy analysis of human locomotion.

Winter (1979a) produced another "new" definition of the mechanical work done in human movement, based on an extension of the instantaneous energy calculations reported by Winter et al (1976). In this case, the total body energy curve was calculated as the sum of the segmental energies (Eqn 2.3) and the work done as the sum of the changes in this total body energy curve:

\[
W_{wb} = \sum_{i=1}^{n} \Delta E_{b,i} \quad \text{Eqn 2.5}
\]

where

\[
n = \text{total number of time intervals in the gait cycle.}
\]

The "work" done during the gait cycle is thus calculated by summing positive and negative changes in the whole body energy curve. Increments in the energy curve are attributed to "positive work" and decrements to
"negative work". The work done to produce kinetic energy changes and that done in producing potential energy changes were separated by partitioning the terms of equation 2.2. This method of determining work has been applied in studies of both walking (Pierrynowski et al, 1980; Woolley and Winter, 1979) and running (Williams, 1980). Since only net changes in the total energy of the body are considered, passive energy transfers do not contribute to \( W_{wb} \) as they do to Norman et al's (1976) work estimate, \( W_n \).

Winter (1979a) calculates the average discrepancy between work measurements based on point-mass as opposed to link-segmental models to be 17% in normal walking, a difference which is likely to be greater in running where the magnitude of segmental energy changes is greater.

2.3.6 Estimates of energy transfer

\( W_n \) represents the summed energy changes in the body during a gait cycle with no correction for energy transfer. \( W_{wb} \) is calculated from the net total energy curve of the body and excludes all possible energy exchange between and within segments from the work total. On this basis, Pierrynowski et al (1980) assume that the difference between \( W_n \) and \( W_{wb} \) represents the energy changes attributable to passive energy transfer, and compute the total intra-stride energy transfer within and between segments (\( T_{wb} \)):

\[
T_{wb} = W_n - W_{wb}
\]  

Eqn 2.6

A further pseudowork term was calculated by summing the changes in the energy of the body in such a way that within-segment energy exchange but not between-segment energy exchange were eliminated from the total:

\[
W_w = \sum_{i=1}^{s} \sum_{j=1}^{n} |\Delta E_{i,j}|
\]  

Eqn 2.7

where

\[
E_{i,j} = \text{the change total energy of the ith segment during the jth time interval}
\]

\[
W_w = \text{is the total work done, assuming all possible within segment energy transfer.}
\]

Thus the total within-segment energy transfer (\( T_w \)) and total between segment energy transfer (\( T_b \)) were determined by subtraction:
\[ T_b = W_w - W_{wb} \quad \text{Eqn 2.8} \]
\[ T_w = T_{wb} - T_b \quad \text{Eqn 2.9} \]

Pierrynowski et al (1980) performed this analysis on six male subjects walking on a treadmill at an average speed of 1.54 m.s\(^{-1}\). A mean net energy expenditure for the six subjects of 255W was reported. Average values of \( W_n \), \( W_w \), and \( W_{wb} \) were 501W, 340W and 165W respectively. The average rate of total within and between segment energy exchange was therefore calculated as 335W. Thus some 67% of the observed energy changes in walking can be attributed to energy transfer. If no transfer is permitted in the model, (i.e. if \( W_n \) is taken to indicate the work-rate) an efficiency of 196% would be determined. Even when \( W_{wb} \) is assumed to indicate the work rate (i.e. all possible passive energy transfers are accounted for) an average efficiency of 64.7% is indicated. This efficiency value is again, rather high. Pierrynowski et al (1980) account for this by assuming that muscle performs positive work and negative work with different efficiencies. The \( W_{wb} \) total includes equal quantities of "positive" and "negative" work. Assuming negative work is performed three times as efficiently as positive work gives positive and negative work efficiencies of 43% and 130% respectively. The negative work efficiency in excess of 100% is rationalised by the suggestion that "negative work" includes the dissipation of energy as heat by passive viscosity at no metabolic cost. The high "negative work efficiency" is therefore a combination of a "metabolic negative work efficiency" of less than 100% and a "passive negative work efficiency" which is theoretically infinite.

Ito et al (1983) applied Pierrynowski et al's method to 3 marathon runners running on a treadmill at speeds in the range 1.94 to 6.11 m.s\(^{-1}\). The calculated mechanical efficiency was on average, 55% and was fairly constant over the range of speeds.

The calculations described by Pierrynowski et al (1980) depend on implied assumptions about the nature of positive and eccentric work. It is also assumed that energy is freely and totally transferred between segments at no metabolic cost and that there is no contribution to energy changes made by elastic energy storage in muscles and tendons. No reason is given for the choice of a 1:3 positive:negative work efficiency ratio and it is assumed therefore that the decision is arbitrary.
Williams and Cavanagh (1983; also Williams, 1980) investigated more closely and more critically the assumptions made in the calculation of mechanical work rates using "pseudowork" and related techniques. These authors use the methods described by Pierrynowski et al (1980), with some additional constraints, to calculate mechanical work rates and energy transfers in athletes running overground at a speed of ~3.57 m.s$^{-1}$. They describe the following equation:

\[
\frac{c \cdot \text{TNEG}}{d} = (1-a)(1-b) \cdot \text{TPOS} + \text{PTOT} \quad \text{Eqn 2.10}
\]

where

- \( \text{TPOS} \) = "Total positive power" assuming complete within segment energy exchange
- \( \text{TNEG} \) = "Total negative power"
- \( a, b, c, d \) = correction factors
- \( \text{PTOT} \) = The total adjusted mechanical power output

\( \text{TPOS} \) and \( \text{TNEG} \) are partitioned from the \( W_{\text{wb}} \) total and are therefore approximately equal in magnitude but of opposite sign. The coefficients \( a, b, c, d \) represent specific assumptions, as follows:

- \( a \): the fraction of \( \text{TPOS} \) attributable to between segment energy transfer
- \( b \): the fraction of \((1-a) \cdot \text{TPOS}\) attributable to elastic energy storage
- \( c \): the fraction of \( \text{TNEG} \) due to eccentric contraction
- \( d \): The negative:positive work efficiency ratio

By varying the values of the coefficients, it is possible to test the effect of different assumptions on the calculated "power output". In addition to the three transfer conditions (\( W_n, W_w \) and \( W_{\text{wb}} \)) described previously, Williams (1980) used two further conditions in which between segment energy transfers are constrained (1) to exchanges between adjacent segments and (2) to exchanges between segments of the same limb and to the trunk, but not across the trunk to another limb. These variations on between segment transfer give four possible values of the coefficient \( a \).
Alternative values for the other coefficients were also applied. The assumptions concerning energy transfer constraints and positive-negative work efficiency were found to have more influence on PTOT than those concerning elastic energy storage and eccentric work.

When total transfer was assumed, the group of subjects with the lowest mechanical work rates ($W_{wb}$) also had the lowest energy expenditure. The opposite was true if no transfer was assumed. The net mean energy expenditure rate for all 31 subjects was 44% when the default assumption set was applied.

In general link-sequential energy analyses have been restricted to potential and kinetic energies. Luhtanen and Komi (1980) and Ito et al (1983) used a simple model of the elasticity of the lower limbs to calculate variations in the "apparent spring constant" of the body. No attempt was made to incorporate the effects of elastic energy storage into the kinematic model, although a crude estimate of "stored elastic energy" was made.

2:3.7 **Calculation of the "power flow" at joints.**

Elftman (1940) reanalysed Fenn's data, calculating joint forces and muscle moments directly and hence the work done by muscles themselves. Elftman used the inverse dynamic approach, based on D'Alembert's principle (Elftman 1939a, 1939b) which has since formed one of the foundation stones of biomechanics. This method has been the basis of many biomechanical analyses since. Elftman's calculations produced a value for the work rate very similar to that determined by Fenn, but about 750 Watts of this was accounted for by "distribution of energy between the parts of the body", which is roughly equivalent to passive energy transfer.

Quanberry et al (1975) demonstrate two methods of calculating the "instantaneous power" of individual body segments. The first is by summation of the potential, translational kinetic and rotational kinetic energies of the segment, the second by calculation of "power flow" to and from the limb. The calculated joint reaction forces were multiplied by the velocity of the joint to give "power flow due to joint force". Similarly, the product of the estimated knee joint moment and knee joint angular velocity were used to calculate the "power flow due to tendon moment". The "instantaneous power" can then be estimated by integration of power flow. The two sets of measurements, when applied to the motion of the shank during walking were in close agreement. A similar approach was used by
Chapman and Caldwell (1983a,b) to study energy transformations in the swing leg during running.

Cappozzo et al. (1975) calculated the segmental energies of walking subjects using a seven segment model in which the legs were considered as three segments each, and the head-arms-trunk system were treated as a single unit. Again, they used the dual approach of calculating energy transformations from segmental centre of mass kinematics and from estimations of joint reaction forces and moments.

Robertson and Winter (1980) calculate "muscle moment power" and "joint force power" as a means of examining the energy transformations between segments during walking. Joint reactions and joint moments were calculated using D'Alembert's principle. "Joint force power" is essentially the product of the joint reaction force and the velocity of the joint centre. "Muscle moment power" is, similarly, the product of the calculated muscle moment and the angular velocity of the limb.

Chapman and Caldwell (1983a,b) used this approach in an analysis of the recovery leg during running. Winter (1983) reports a similar analysis of the lower limb for the whole walking cycle.

The main disadvantage of this technique is that it is necessary to determine the forces at the extremities of the body. When the extremity is not in contact with a surface, the forces are zero. During the support phase of walking and running however, ground reaction forces must be determined. Thus the technique cannot be readily applied to treadmill running, or if it is applied, must be restricted to the flight phase. The analysis is also more heavily dependant on the accurate determination of joint centre locations for good results. Kinematic data must be differentiated twice to obtain accelerations in order to calculate forces and moments. Double differentiation is inherently more error prone and noise-amplifying than the single differentiation (for velocities) in an energy analysis. Similarly, the inadequacy of the link-segmental model in describing the skeletal structure and joints of the torso prevent the analysis from being sensibly applied to the whole body.
DISCUSSION

It is clear from reference to published research that the oxygen cost of running is not a fixed function of running speed, but that there are differences in "running economy" between individuals. There is also evidence that in groups of athletes of similar ability and aerobic capacity, running economy is a significant determinant of performance.

The kinematics of the running gait are also subject to inter-individual variation. Although the human running gait has a characteristic pattern, there is individual variation. Systematic differences between treadmill and overground; skilled and non-skilled; and fresh and fatigued states have been reported. Also, although highly evolved, the running gait pattern appears to adapt to both short-term and long-term training.

A tentative link between "running economy" and the kinematics of the gait pattern has been established by experiments in which the oxygen cost of running at a particular speed was manipulated by imposing fixed cadences on the subjects.

Taken as a whole, this evidence would appear to support the suggestions made by a number of researchers that differences in running economy are at least partly determined by differences in "running style" or "mechanical efficiency".

Attempts to measure the "efficiency" of the walking or running gait have generally led to calculations of efficiency that are higher than the known overall efficiency of muscle, which suggests that the models are not a complete representation of the energy transformations that occur during the running or walking stride. Estimates of based point-mass models tend to underestimate the total body mechanical energy. More recently, the development of "pseudowork" models has enabled more detailed examinations of potential and kinetic energy changes to be performed, but has also added to the variety of methods available for calculating "mechanical efficiency".

Since the body appears to optimise its movement pattern in order to minimise energy expenditure, the relationships between energy expenditure and movement would appear to be worthy of further investigation. This study is therefore concerned with the relationships between mechanical energy changes in the human body during the running gait and energy expenditure. Both movements and oxygen consumption can be quantified in
terms of "energy", so a mechanical energy analysis would appear to be an appropriate tool for the study of "movement" in this instance. Given the variety of energy models that have been used to measure "mechanical work" and "efficiency", and the wide variety of results which have been obtained, it is proposed to restrict the energy analysis to examination of "energy changes".

From a practical viewpoint, the use of a treadmill to simulate overground running for experimental purposes has numerous advantages. There are however, some differences in the gait pattern between treadmill and overground running. These differences are relatively small, however and are reduced after a short period of treadmill training. While the study of treadmill running may limit the extent to which results can be applied to athletes on the track, the selection of one or the other mode of running for experimental purposes does not prevent study of the movement mechanisms common to both activities.
Chapter 3

CINEPHOTOGRAPHIC ANALYSIS OF TREADMILL RUNNING
This section describes the methods used to perform cinephotographic analyses of athletes running on a treadmill.

The analysis was based on a 15-segment rigid body model of the human body which is implemented for both two and three dimensional analyses. The inertial characteristics of the model were determined from published cadaver data in conjunction with anthropometry of the experimental subjects.

A comparison of the error introduced into the film data by different joint centre estimation procedures was performed and an improvement in the procedure made by a small adjustment to the way in which body surface markers were placed on the subjects.

The frequency distributions of the signal and noise components of a typical data set from film of a runner were used to establish appropriate minima for sampling rates and filter cut-off frequencies.

Spatial calibration was achieved by a least squares approximation to the direct transformation between image and object space coordinates, based on a control point distribution in object space.

Preliminary processing of the film data involved filtering and the determination of time derivatives of joint centre and extremity loci. In the three dimensional analysis, these calculations were performed in the frequency domain by manipulation of Fourier coefficients. In the two dimensional analyses, a fourth order, zero-lag Butterworth digital filter was used for smoothing and derivatives were calculated by finite difference techniques.

The preparation of a data set describing the motion of the link-segmental model was completed by the calculation of segmental centre of mass and angular kinematics.

The use of cinephotography as a means of recording human movement has a number of advantages over other methods. Probably the most important of these are that data for the kinematics of the whole body can be recorded simultaneously, and that the apparatus does not directly intrude upon the subject. The main disadvantages are that the analysis of film data can be prolonged; and that the forces acting on the body can only be determined
indirectly, by inverse dynamic inference.

This section describes the cinephotographic analysis procedures used in the experiments on treadmill running described elsewhere in this study. As a result of the experience gained from the first of these two studies, together with changes in the availability of computer power, the numerical procedures used in the two studies are substantially different. The principles of analysis however remained unchanged and are described in this section. Details specific to one or other of the two studies have been included in Chapters 4 and 9.

The basic requirement of film recordings for the purpose of biomechanical analysis is that information be recorded in such a way that it can be applied to some appropriate model of the human body and used to describe its motion. The model is a necessity imposed by the mechanical and dynamic complexity of the human body and represents the simplifying assumptions that must be allowed in order to make an analysis possible.

The movements of a subject are recorded on film and combined with anthropometric and inertial data to define the movements of a link-segmental model. The kinematics and dynamics of the link-segmental model are then analysed as an approximation to those of the subject.

It is essential that the film can be fully "calibrated" so that the real space-time coordinates of points of interest can be determined from the recorded image. The transposition of data from the film to the model is also a non-trivial problem. Numerical procedures must then be selected to calculate the kinematics of the model with due recognition of the noise content of the original data.

3:2 A LINK SEGMENTAL MODEL OF THE HUMAN BODY

The link-segmental model (e.g. Huston and Passerello, 1971) is a commonly used tool in biomechanics research. Variations on the theme of the link segmental model are almost as numerous as the people who have used it. In general, the model consists of a representation of the human body as a series of rigid segments, hinged at joints which represent the anatomical joints of the human body. The inertial properties of the model are chosen to reflect those of a particular individual or group of subjects where possible. These may be determined by means of an inertial model or by using anthropometric data from cadavers recorded in the literature. With inertia models each segment is assumed to be of uniform density and of simple geometric shape (e.g. Hanavan, 1964,1966; Whitsett, 1963,1964). The centre
of mass and other inertial properties of each segment is then calculated from its geometry and density.

The number of segments included in the link-segmental model has varied considerably, depending on the application. A typical model has fifteen segments if the hands are included as separate segments and the torso is modelled as separate thoracic and abdominal segments. (Hanavan, 1964, 1966; Aleshinski and Zatsiorsky, 1978). The use of a single torso segment and the inclusion of hands into the forearm segments gives a 12 segment model. In some instances, the whole of the head, arms and trunk are modelled as a single "HAT" segment, giving a seven segment model (Onyshko and Winter, 1980). Hatze's (1977) 17-segment model included separate shoulder segments.

The orientation of the model in space is determined from recordings of the kinematics of real subjects, normally by means of photography, but also with the use of TV systems (Winter et al, 1972) and, more recently, optoelectronic methods (Lindholm, 1974; Cappozzo et al, 1983).

For the purposes of this study, a 15-segment rigid body model of the human body was defined for use in the analysis of athletes running on a treadmill. The inertial properties of the model were defined by a combination of published cadaveric data and anthropometric measurements on individual subjects. The kinematics of the model were calculated using two and three dimensional cinephotographic methods.

3:2.1 A 15-segment rigid body model

The link-segmental model used consisted of 15 segments. These segments are defined in Table 3.1 and illustrated in Figure 3.1.

The endpoints of each segment are defined by the proximal and distal joint centres or extremities, which themselves are determined by reference to anatomical landmarks. The definition and determination of the segmental endpoints are considered in more detail in section 3:4. In the case of the abdomen, the lower extremity was defined as the mid-point between the two hip joint centres. Each foot was defined by four points: the ankle joint centre, the heel extremity, the toe extremity and the metatarso-phalangeal joint. Two alternative models of the hand were used depending on whether the subject being studied ran with the fingers extended or with the fist clenched. In the first case, the extremity of the hand was defined as the tip of the third finger and in the second case as the third metacarpo-phalangeal joint.
Table 3:1: Configuration of a 15 segment rigid body model of the human body.

<table>
<thead>
<tr>
<th>SEGMENT</th>
<th>PROXIMAL EXTREMITY</th>
<th>DISTAL EXTREMITY</th>
</tr>
</thead>
<tbody>
<tr>
<td>No.</td>
<td>Label</td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>Head centroid</td>
<td>Top of head</td>
</tr>
<tr>
<td>2</td>
<td>Thorax centroid at 10th rib</td>
<td>Neck centroid</td>
</tr>
<tr>
<td>3</td>
<td>Abdomen centroid at 10th rib</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>Left upper arm L shoulder</td>
<td>L elbow</td>
</tr>
<tr>
<td>5</td>
<td>Left lower arm L elbow</td>
<td>L wrist</td>
</tr>
<tr>
<td>6</td>
<td>Left hand L wrist</td>
<td></td>
</tr>
<tr>
<td>7</td>
<td>Left thigh L hip</td>
<td>L knee</td>
</tr>
<tr>
<td>8</td>
<td>Left shank L knee</td>
<td>L ankle</td>
</tr>
<tr>
<td>9</td>
<td>Left foot L ankle</td>
<td>L toe and heel*</td>
</tr>
<tr>
<td>10</td>
<td>Right upper arm R shoulder</td>
<td>R elbow</td>
</tr>
<tr>
<td>11</td>
<td>Right lower arm R elbow</td>
<td>R wrist</td>
</tr>
<tr>
<td>12</td>
<td>Right hand R wrist</td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>Right thigh R hip</td>
<td>R knee</td>
</tr>
<tr>
<td>14</td>
<td>Right shank R knee</td>
<td>R ankle</td>
</tr>
<tr>
<td>15</td>
<td>Right foot R ankle</td>
<td>R toe and heel*</td>
</tr>
</tbody>
</table>

* See text
Figure 3.1: Configuration of a 15 segment link-segmental model of the human body
3:2.2 Implementation of the link-segmental model.

For the purposes of analysis, a generalised link-segmental model was used. The analytical software was designed to act on any arbitrarily configured model with any number of segments. The configuration of the model was defined by a table of references between joint centre and extremity inputs and segment definitions. Thus a segment was defined by references to its proximal and distal extremities and the segmental analysis performed by means of references rather than by explicit programming of a particular model. Thus the software written for analysis of a 15-segment model of running might equally well be applied to a 7-segment model of a cyclist with minimal adaptation. The FORTRAN IV subroutine used to define the relationships between data sets and model segments is listed in Appendix C.

3:3 INERTIAL PROPERTIES OF THE RIGID BODY MODEL

The most important of the parameters required to define the inertial properties of each segment of the rigid body model are:

1. mass
2. centre of mass location
3. moments of inertia

Ideally, the parameters selected for the model should reflect as closely as possible the "true" inertial properties of an individual experimental subject. Direct measurements are not generally possible on living subjects, however, so some estimation process must be used. Gamma mass scanning (Brooks and Jacobs, 1975) and computed tomography (Rodrique and Gagnon, 1980) of the limbs can yield values for the masses, centres of mass and moments of inertia of body segments, but these techniques are not widely available. The three estimation processes commonly used involve (i) inertial models of the human body (ii) segment-zone volume measurements (iii) estimation based on the directly measured inertial parameters of cadavers.

An inertial model represents each segment of the body as a geometric solid of uniform shape such as a cylinder, a sphere, or the frustrum of a cone. The dimensions of the solid are determined by measuring a living subject. Segment volumes can be calculated from the geometry of the solid and assuming a value for density (Dempster, 1955; Drillis and Contini, 1966). The mass and centre of mass of the segment can also be estimated. Alternatively, segmental masses from cadaver data may be substituted...
(Miller and Morrison, 1975). The most widely used inertial model of the human body has been the Hanavan model (Hanavan 1964, 1966) (a fifteen segment model) and its derivatives.

Segment-zone methods involve the subdivision of body segments into a number of "zones" (Katch et al, 1974). Each zone is modelled as a simple geometric shape, the volume of which is calculated from measurements of length and circumference from the living subject. Jensen (1976, 1978) has extended the segment-zone method to its logical conclusion by dividing (theoretically) each segment of the body into elliptical slices 2cm thick. The major and minor radii of each ellipse are determined from photographs of the subject in the anthropometric position. Given density data for each segment, the properties of each can be calculated. Summation over a number of slices yields the mass, centre of mass, and moments of inertia, for each segment or for the whole body.

Probably the most ubiquitous method of estimating the inertial properties of a human body model is to base the estimates on the measured properties of human cadavers. There are a number of reasons why cadaveric data is unsuitable for predicting the inertial properties of living humans. Pre- and post-mortem wasting of the body, post-mortem redistribution of fluids, changes brought about by freezing or embalming of the cadavers and tissue and fluid losses during segmentation all contribute to produce differences between living and cadaveric segments. On the other hand, it is suggested (Clauser et al, 1969) that the differences between the living and cadaveric material are likely to occur throughout the body, not just in localised parts of segments. If this is the case then relationships between measurements within the cadaver would be very similar to those in the living, even though absolute values may change. A further problem is that cadaveric samples usually represent a sample of "available material" rather than a representative sample of the young athletic male population. Ultimately, in the absence of techniques to make direct measurements on living subjects, the use of cadaveric data is unavoidable.

For the purposes of this study, a number of sources of anthropometric data were used to determine the inertial properties of the link-segmental model.

3.3.1 Anthropometric data (1)

In a preliminary study, segmental masses, centres of mass and moments of inertia were estimated from the cadaveric data of Dempster (1955) and Barter (1957).
Barter (1957) collated the cadaver data of Braune and Fischer (1889), Fischer (1906) and Dempster (1955), establishing regression equations for the prediction of segment masses from body weight (Table 3.2). The regression equations are based on the dissected segmental masses of a total of 12 cadavers. Previously, the data had been reported and used in the form of direct ratios between segment mass and body weight.

The choice of Barter's equations was based on the observation that the mean body mass of the cadavers included in the equations (68.75 kg) was similar to that of the athletic subjects used in this study. This is an important consideration in view of the fact that some cadaver samples from other ethnic groups are totally unrepresentative of the population being sampled in this study. Mori and Yamamoto (1959) and Fujikawa (1963) for example, report results from cadaveric samples with mean body masses of 34.6kg and 50.3kg respectively.

Segmental masses were obtained by substituting a subject's body mass into the regression equations. Torso mass was calculated as the difference between the mass of the head and the mass of (head + torso) estimates. The calculated mass of the torso was then distributed between the thoracic and abdominal segments in the proportions 0.35 : 0.65 (Miller and Morrison, 1975). Inevitably, the total body mass computed as the total of the estimated segmental masses does not equal the measured body mass of the total. Segmental mass estimates were therefore adjusted to give a model with a mass equal to the total mass of the subject by means of the simple correction:

\[ m_e = m_o \times \frac{M}{M_e} \]  

(3.1)

where \( m_e \) = corrected segment mass  
\( m_o \) = original estimate of segment mass  
\( M \) = measured whole body mass  
\( M_e \) = sum of estimated segmental masses

Segmental centres of mass were assumed to coincide with some point on the straight line joining the proximal and distal joint centres. The proportionate distance of the centre of mass along the line was assumed to be the same as the mean value reported by Dempster (1955) (Table 3.2).

Segmental moments of inertia (about the segment centre of mass) were assumed to be the same as the mean values for 30 subjects calculated by Miller and Morrison (1975) on the basis of Barter's cadaver data applied to
Table 3.2: *Estimates of body segment parameters*

Segment mass = $a_{(body mass)} + b$

**SEE** = standard error of estimate of segment mass  
**cg** = distance of segment centre of mass from proximal joint  
as a fraction of segment length (after Dempster, 1955)

<table>
<thead>
<tr>
<th>Segment</th>
<th>a</th>
<th>b</th>
<th>SEE</th>
<th>cg</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head &amp; torso</td>
<td>0.470</td>
<td>+5.45</td>
<td>1.45</td>
<td></td>
</tr>
<tr>
<td>Head &amp; neck</td>
<td>0.079</td>
<td>0.00</td>
<td></td>
<td>0.433</td>
</tr>
<tr>
<td>Upper arm</td>
<td>0.040</td>
<td>-0.66</td>
<td>0.23</td>
<td>0.436</td>
</tr>
<tr>
<td>Lower arm</td>
<td>0.020</td>
<td>-0.12</td>
<td>0.23</td>
<td>0.430</td>
</tr>
<tr>
<td>Hand</td>
<td>0.005</td>
<td>+0.16</td>
<td>0.09</td>
<td>0.494</td>
</tr>
<tr>
<td>Upper leg</td>
<td>0.090</td>
<td>+0.73</td>
<td>0.82</td>
<td>0.433</td>
</tr>
<tr>
<td>Lower leg</td>
<td>0.055</td>
<td>-0.43</td>
<td>0.36</td>
<td>0.433</td>
</tr>
<tr>
<td>Foot</td>
<td>0.010</td>
<td>+0.34</td>
<td>0.14</td>
<td>0.433</td>
</tr>
</tbody>
</table>
the Hanavan inertia model. Since the kinetic energy of rotations about the longitudinal axis of segments in the human running gait is very small, no attempt to measure these rotations was made and no estimates of the moment of inertia of the segments about that axis were made. A single value for the moments of inertia about the other two axes through the segmental centre of mass were selected on the base that the segments are generally symmetrical about the longitudinal axis.

Miller and Morrison (1975) in fact calculated two sets of moment of inertia data, one by applying cadaveric data from Barter (1975) to the Hanavan model and the other by applying data from Clauser et al (1969). The results were statistically significantly different for all segments apart from the foot, illustrating just how difficult it is to obtain reliable moment of inertia data. In this study of the energy changes in running, a certain amount of error in the estimation of segmental moments of inertia can be tolerated, since the energy of segments rotating around their own centres of mass is relatively small when compared with the potential and translational kinetic energy components.

3:3.2 Anthropometric data (2)

In a further study of running mechanics (Chapter 9), a more comprehensive approach to the generation of segmental parameters was adopted, based on the cadaver data of Clauser et al (1969) and Chandler et al (1975).

Clauser et al (1969) made a comprehensive study of 13 carefully selected male cadavers with a mean body mass of 66.52kg (+8.7 s.d.). A total of 73 anthropometric measurements were made in addition to direct measurements of segmental masses, volumes and centre of mass locations. A step-wise regression method was used to determine predictive equations between the inertial parameters and the anthropometric data. The predictive equations are therefore based on the two or three best predictors of each inertia parameter.


The use of predictive equations of this type has greater face validity than the use of a simple proportional distribution of mass, since the use of a wide range of anthropometric data in addition to body mass and stature can be expected to account for more of the inter-individual variation in body type. Both Chandler et al (1975) and Clauser et al (1969) report
correlation coefficients and standard errors of the estimate for their predictive equations.

For the purposes of this study, data from Clauser et al (1969) was used to estimate the segmental masses of the 15-segment model. Moments of inertia and centre of mass location were taken from Chandler et al (1975).

A total of 31 anthropometric measurements were taken from each subject to be modelled (Table 3.3). Circumferences were measured using a narrow steel tape. Heights, bone breadths and bone lengths were measured using a Harpenden anthropometer. Full details of each measurement procedure are recorded in appendices to Clauser et al (1969) and Chandler et al (1975). As before, segmental masses were adjusted to correct for any discrepancy between predicted and measured total body mass (Eqn 3.1)

The running shoes of each subject were weighed and their masses added to those of the feet.

The location of the centre of mass of the hand was estimated in two ways, depending on whether the subject ran with the hand in an open (fingers extended) or closed (fist clenched) position. Figure 3.2 illustrates the basis of this estimation. The hand was modelled as three subunits of equal size and mass. For the open hand, the distal extremity was defined as the tip of the third finger and the centre of mass assumed to be midway between the proximal and distal segmental endpoints. For the closed hand, the distal extremity was defined as the third metacarpal-phalangeal joint and centre of mass assumed to be located at a distance from the proximal joint centre equivalent to 0.58 of the segmental length.

The predictive equations were implemented in a FORTRAN IV computer programme which is listed in Appendix D.
Table 3.3: Anthropometric measurements use in the prediction of segmental inertial parameters.

Details of each measurement procedure are reported in Clauser et al (1969), Appendix D and Chandler et al (1975), Appendix C.

All measurement were recorded in cm.

<table>
<thead>
<tr>
<th>Whole body</th>
<th>Bone breadths</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Mass</td>
<td>13. Head</td>
</tr>
<tr>
<td>2. Stature</td>
<td>14. Elbow</td>
</tr>
<tr>
<td></td>
<td>15. Wrist</td>
</tr>
<tr>
<td></td>
<td>16. Hand</td>
</tr>
<tr>
<td></td>
<td>17. Knee</td>
</tr>
<tr>
<td></td>
<td>18. Bi-spinous</td>
</tr>
<tr>
<td>Circumferences</td>
<td></td>
</tr>
<tr>
<td>3. Head</td>
<td>Lengths</td>
</tr>
<tr>
<td>5. Upper thigh</td>
<td>20. Acromion-radiale</td>
</tr>
<tr>
<td>7. Ankle</td>
<td>22. Foot</td>
</tr>
<tr>
<td>8. Arm (axilla)</td>
<td></td>
</tr>
<tr>
<td>9. Forearm</td>
<td>Heights</td>
</tr>
<tr>
<td>10. Wrist</td>
<td>23. Chin/neck intersect</td>
</tr>
<tr>
<td>12. Hand</td>
<td>25. Trochanter</td>
</tr>
<tr>
<td></td>
<td>26. Tibiale</td>
</tr>
<tr>
<td></td>
<td>27. Sphyrion</td>
</tr>
<tr>
<td></td>
<td>28. Lateral malleolus</td>
</tr>
<tr>
<td></td>
<td>Derived measurements</td>
</tr>
<tr>
<td></td>
<td>29. Trunk length</td>
</tr>
<tr>
<td></td>
<td>30. Thigh length</td>
</tr>
<tr>
<td></td>
<td>31. Calf length</td>
</tr>
<tr>
<td></td>
<td>32. Shoe mass (kg)</td>
</tr>
</tbody>
</table>
Figure 3.2: Simple model for the estimation of the location of the hand when (a) open and (b) closed.
3:4 BODY SURFACE MARKERS, SKIN MOVEMENTS AND DIGITISING ERRORS

Film analysis of human movement is based on the transformation of the coordinate data from film images of the body to parameters describing the loci of the inter-segmental joints of a linked rigid-segment model. Given a link-segmental model that assumes perfect, unicentric joints this problem reduces to the estimation of the position of the theoretical joint centre from the photographic image. Markers attached to the skin of the subject are commonly used to assist this estimation process. The markers are placed over a point considered to represent the theoretical joint centre or are placed on the body in such a way that the joint centre's position can be estimated from them. During a movement, however, the instantaneous centre of a real joint is likely to move in relation to anatomical landmarks. Similarly, the position of the skin markers relative to anatomical markers will change.

A careful, experienced operator can repeatedly digitise a small clear mark to within 0.01cm but this accuracy is rarely achieved consistently when digitising projected images. "Noise" is therefore introduced into the data by the digitising process.

This section describes an analysis of the errors associated with digitising film images and the estimation of joint centre positions.

3:4.1 Positioning of body surface markers

Markers were attached to an athlete's skin at selected points on the body. The markers consisted of a 3 cm diameter black disc with a central 1 cm diameter white target area. During a preliminary investigation, markers were attached to the the right side of the body in the following positions:

1. Neck : Vertically, half way between the inferior surface of the ear-lobe and the upper surface of the external extremity of the clavicle. Horizontally, midway between the dorsal and ventral surfaces of the neck.
2. Shoulder : Overlying the greater tuberosity of the humerus
3. Elbow : Overlying the proximal extremity of the radius
4. Wrist : Overlying the dorsal surface of the semilunar bone of the carpus.
5. Rib: Overlying the 10th rib, midway between the ventral and dorsal surfaces of the thorax.

6. Hip: Overlying the greater trochanter

7. Knee: Midway between the tuberosity of the lateral femoral condyle and the styloid process of the fibula.

8. Ankle: Overlying the lateral malleolus of the fibula

9. MTP joint: On the outer surface of the shoe, overlying the fifth metatarso-phalangeal joint

The positions of the bony landmarks were determined by palpation. Other landmark locations were estimated without the aid of markers as follows:

10. Top of head: Upper surface of the skull, midway between the anterior and posterior surfaces.

11. MCP joint: 3rd metacarpo-phalangeal joint.

12. Toe: Distal extremity of the foot, midway between the upper and lower surfaces of the shoe.

13. Heel: Posterior extremity of the shoe, 1 cm above the level of the upper surface of the sole. (i.e. in the approximate position of the distal extremity of the calcaneum.

3.4.2 Digitising errors

Film of the athlete running on a treadmill was digitised using an SAC Graf-Pen model GP6 sonic digitiser. Extremity and joint centre positions were estimated from 0.14 scale projected film image by three different methods:

M0: Joint centre/extremity position estimated without the aid of joint-markers. (Top of head, MCP joint, Toe, Heel)

M1: Joint centre estimates using markers based on bony landmarks as guides. (Shoulder, ankle, wrist)

M2: Joint centre position assumed to be that of a marker placed on the skin. (Neck, elbow, hip, knee, MTP joint)
A single frame was digitised 50 times and the means and standard deviations of the estimated coordinates of the joint centres and extremities were calculated.

Table 3.4 shows the standard deviation of the fifty samples of each location. This variation is also illustrated in Figure 3.3 where the size of the circle surrounding each location is proportional to the distribution of the data sample about the mean value. This variation is not an estimate of any systematic error in determining the absolute location of the joint centre or extremity, but is an indication of the random error introduced by the digitising process. The digitiser itself is repeatable to 0.1 mm when the cursor is fixed. Any additional variation can be attributed to variation in the placement of the digitiser cursor by the operator.

The overall resultant standard deviations for the three methods M0, M1 and M2 were 0.37 mm, 0.32 mm and 0.19 mm respectively. Where the location of the point digitised is solely dependant on the position of the centre of a body marker (M2), the repeatability of the digitising process approaches that of the digitiser itself. Where the location is estimated purely "by eye" (M0), the repeatability is lowest. The most variable estimates were those of the position of the top of the head and the wrist "joint centre", with standard deviations of the order of 0.4 mm (image units). When this error is scaled up to full size, it represents a standard deviation of 3mm for the estimate of the position of the location in real space.

These errors are relatively small compared with the 2m * 2m extent of the calibrated space that is being digitised. Their effect would be reduced proportionately as the size of the projected image being digitised is increased. The errors do however, introduce a random "noise" component into the digitised data which would be amplified considerably when the body landmark coordinate time-series are differentiated. Random noise of mean peak to peak amplitude 0.3 cm (life size coordinates) at a sampling frequency of 50Hz for example, gives a noise amplitude of 150 mm.s^-1 in the unfiltered first derivative. It is important, therefore that appropriate filtering techniques are employed before differentiation is performed.

Since the position of a segment is defined by two endpoints, errors in digitising the location of joint centres and extremities may be compounded when segment lengths and segmental centres of mass are computed. Table 3.5 shows the results of computing segmental lengths from the 50 sets of joint centre and extremity data shown in Table 3.4 and Figure 3.3. This illustrates the effect of random digitising errors on the computation of
Table 3.4: Variation in estimates of joint centre and extremity coordinates.

Standard deviations of 50 samples of X and Y coordinates and resultant deviation, R.

<table>
<thead>
<tr>
<th>Location</th>
<th>Method</th>
<th>X</th>
<th>Y</th>
<th>R</th>
</tr>
</thead>
<tbody>
<tr>
<td>Too of head</td>
<td>M0</td>
<td>0.40</td>
<td>0.15</td>
<td>0.43</td>
</tr>
<tr>
<td>MCP joint</td>
<td>M0</td>
<td>0.29</td>
<td>0.23</td>
<td>0.37</td>
</tr>
<tr>
<td>Toe</td>
<td>M0</td>
<td>0.25</td>
<td>0.19</td>
<td>0.31</td>
</tr>
<tr>
<td>Shoulder</td>
<td>M1</td>
<td>0.25</td>
<td>0.17</td>
<td>0.30</td>
</tr>
<tr>
<td>Ankle</td>
<td>M1</td>
<td>0.24</td>
<td>0.37</td>
<td>0.29</td>
</tr>
<tr>
<td>Wrist</td>
<td>M1</td>
<td>0.32</td>
<td>0.20</td>
<td>0.38</td>
</tr>
<tr>
<td>Neck</td>
<td>M2</td>
<td>0.14</td>
<td>0.11</td>
<td>0.18</td>
</tr>
<tr>
<td>Elbow</td>
<td>M2</td>
<td>0.17</td>
<td>0.12</td>
<td>0.21</td>
</tr>
<tr>
<td>Hip</td>
<td>M2</td>
<td>0.07</td>
<td>0.13</td>
<td>0.15</td>
</tr>
<tr>
<td>Knee</td>
<td>M2</td>
<td>0.13</td>
<td>0.17</td>
<td>0.21</td>
</tr>
</tbody>
</table>
Figure 3.3: Error in the estimation of joint and extremity coordinates.

Each circle has a radius equal to 25 times the standard deviation of fifty estimates from the same frame.
(See Table 3.4)
Table 3.5: Variation in apparent segment length due to errors in joint centre estimation

Mean segment length, standard deviation and coefficient of variation of 50 samples.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mean length (image mm)</th>
<th>s.d.</th>
<th>C of V %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>30.5</td>
<td>0.3</td>
<td>1.1</td>
</tr>
<tr>
<td>Thorax</td>
<td>53.1</td>
<td>0.4</td>
<td>0.7</td>
</tr>
<tr>
<td>Abdomen</td>
<td>24.4</td>
<td>0.4</td>
<td>1.6</td>
</tr>
<tr>
<td>Upper arm</td>
<td>39.0</td>
<td>0.6</td>
<td>1.5</td>
</tr>
<tr>
<td>Lower arm</td>
<td>36.4</td>
<td>0.6</td>
<td>1.6</td>
</tr>
<tr>
<td>Hand</td>
<td>14.0</td>
<td>0.7</td>
<td>4.9</td>
</tr>
<tr>
<td>Thigh</td>
<td>59.0</td>
<td>0.4</td>
<td>0.6</td>
</tr>
<tr>
<td>Shank</td>
<td>61.0</td>
<td>0.4</td>
<td>0.7</td>
</tr>
<tr>
<td>Foot</td>
<td>29.8</td>
<td>0.8</td>
<td>2.8</td>
</tr>
</tbody>
</table>
segmental lengths. On average, the standard deviation of the 50 independently calculated segmental lengths, from the same frame of film, was 1.7% of the mean value. In the case of the thigh, a long segment with well marked extremities, the variation in the apparent length is very small. In the case of the hand, a small segment with ill-defined extremities, the error can be large. In this instance, the coefficient of variation over 50 samples approached 5%.

If the centre of mass of each segment is estimated from the locations of the segmental endpoints, then the random error in the raw data contributes to the error in the segmental centre of mass estimate. Since, with simple anthropometric models, the location of the segmental centre of mass is computed as a constant proportion of the segmental length, the relative error in centre of mass estimation will be similar to that for segmental lengths.

3.4.3 Skin movements

While it is more reliable to digitise a clearly marked point than to estimate the position of a joint centre, marks placed on the skin are unlikely to remain fixed relative to the centre of rotation of a joint over its full range of movement. The instantaneous centre of rotation of a real joint moves in relation to the skeletal references over the range of movement. Similarly, the skin and the skeleton move in relation to one another. Since these two motions are not simply related, the location of a joint marker and the location of the joint centre it is supposed to mark will move independently of one another.

A simple way of estimating the degree of skin movement is to measure the distance between the markers at opposing ends of segments, which should, ideally, remain constant during a movement. In reality, one effect of the independent motion of joint centre and superficial marker is that the digitised length of a segment will vary systematically. This variation is in addition to any random digitising error.

Table 3.6 shows the variability of the length of selected segments during a running stride. In the upper limb, much of the variation is due to changes in the angle between the limb segment and the sagittal plane. This variation is therefore reduced considerably when a three dimensional analysis is used.

In the lower limb, almost 90% of the variance in apparent thigh length is due to relative motion of the knee and hip markers, rather than to random digitising error.
Table 3.6: Variation in the apparent length of model segments during a running stride.

Mean, standard deviation and coefficient of variation of 36 samples

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mean length (image mm)</th>
<th>s.d.</th>
<th>C of V %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>30.3</td>
<td>0.4</td>
<td>1.4</td>
</tr>
<tr>
<td>Thorax</td>
<td>52.2</td>
<td>1.2</td>
<td>2.4</td>
</tr>
<tr>
<td>Abdomen</td>
<td>25.7</td>
<td>0.6</td>
<td>2.5</td>
</tr>
<tr>
<td>Upper arm</td>
<td>37.1</td>
<td>1.6</td>
<td>4.3</td>
</tr>
<tr>
<td>Lower arm</td>
<td>32.5</td>
<td>4.3</td>
<td>13.2</td>
</tr>
<tr>
<td>Hand</td>
<td>12.7</td>
<td>2.1</td>
<td>16.3</td>
</tr>
<tr>
<td>Thigh</td>
<td>56.5</td>
<td>1.1</td>
<td>2.0</td>
</tr>
<tr>
<td>Shank</td>
<td>58.4</td>
<td>1.4</td>
<td>2.3</td>
</tr>
<tr>
<td>Foot</td>
<td>27.5</td>
<td>0.8</td>
<td>3.0</td>
</tr>
</tbody>
</table>
Figure 3.4 shows the apparent change in the length of the thigh (i.e. the measured distance between the hip and knee markers) during the running stride cycle. The data in this graph has been smoothed using a cubic spline technique. The range of values in the raw data, before smoothing, was 5.44 to 5.84 image cm. Thus the range of apparent lengths of the thigh is some 7% of the overall length. Any estimate of the location of the centre of mass of the segment will vary accordingly.

This data suggests that there is some argument in favour of using surface markers to assist the estimation of the position of a joint centre or extremity, rather than to use the position of a single marker alone as an absolute reference. The ideal circumstance would arise if it were possible to place a finite set of markers around a joint in such a way that it were always possible to combine and transform the marker positions to establish the coordinates of the instantaneous centre of rotation of the joint. If such a system could be achieved, however the link-segmental model would still be limited by the assumptions of rigid segments, unicentric joints, and by errors in the estimation of the inertial properties of the segments. Nonetheless, some improvement in the internal consistency of the model, (in the sense that the digitised data should reflect the fact that the model segments are defined as having constant length) may be possible if multiple markers are used to assist a visual estimation of the position of joint centres.

3.4.4 An alternative approach to the estimation of hip and knee joint centre locations.

In view of the distortion in apparent length of the thigh and shank brought about by relative motion of joint centres and surface marks, alternative methods for estimating the position of the joint centres of the hip and knee were devised.

Hip markers:

Three markers were placed over the hip; the first was placed overlying the greater trochanter with the subject standing upright with the feet together. The subject then lifted the foot of leg being marked onto a chair or stool in order to bring about stabilised flexion at the hip of approximately 90%. Typically, palpation shows that the positions of the original marker and the greater trochanter diverge by several centimetres when the hip is flexed. A second marker was placed over the the greater trochanter in this flexed position. Similarly a third marker was placed
Figure 3.4: Change in the apparent length of the thigh during a running stride.
over the greater trochanter with the hip in a fully flexed position. Thus the three markers indicate the position of the greater trochanter when the hip joint is in flexed, extended and neutral positions.

Knee markers:

The alternative approach to marking the knee consisted of two markers, one placed overlying the lateral femoral condyle and the other overlying the styloid process of the fibula.

Results:

The digitising procedures described above were repeated with the alternative hip and knee marker systems. The positions of the hip and knee joint centres were estimated from the positions of the markers. A single frame of film was digitised 50 times in order to estimate the variance due to random errors in digitising and estimating the joint centre. The hip, knee and ankle markers were also digitised over the same 36 frames of film of the athlete running on a treadmill, as before.

Table 3.7 shows the variation in the position of the hip and knee markers determined 50 times from the same frame of film. The resultant standard deviation is greater than that found when a single joint marker is assumed to represent a joint centre. While the magnitude of the random digitising error is increased by the estimation process, the error in the determination of segment lengths (Table 3.8) is much less than previously (Table 3.5). Figure 3.5 shows the variation in the apparent length of the thigh, determined as the distance between the hip and knee joint centre positions estimated using the alternative marker system. The systematic error due to relative motion of the surface markers and the joint centre is much reduced.

The alternative, multiple marker joint centre estimation process was selected for use in later film analyses of runners. While the random error in the estimation of the joint centre location is increased slightly by the alternative method, systematic error due to the relative movement of surface markers and bony landmarks is substantially reduced, giving results that are more appropriate to the assumptions inherent in the link-segmental model.
Table 3.7: Variation in estimates of hip and knee joint centre coordinates (multiple marker method).

Standard deviations of 50 samples of X and Y coordinates and resultant deviation, R.

<table>
<thead>
<tr>
<th>Location</th>
<th>Method</th>
<th>X</th>
<th>Y</th>
<th>R</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td>M1</td>
<td>0.23</td>
<td>0.30</td>
<td>0.38</td>
</tr>
<tr>
<td>Knee</td>
<td>M1</td>
<td>0.19</td>
<td>0.21</td>
<td>0.28</td>
</tr>
</tbody>
</table>
Table 3.8: Variation in the apparent length of model thigh and shank during a running stride (multiple marker method).

Mean, standard deviation and coefficient of variation of 36 samples

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mean length (image mm)</th>
<th>s.d.</th>
<th>C of V %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thigh</td>
<td>56.2</td>
<td>0.6</td>
<td>1.1</td>
</tr>
<tr>
<td>Shank</td>
<td>57.9</td>
<td>0.7</td>
<td>1.2</td>
</tr>
</tbody>
</table>
Table 3.8: Variation in the apparent length of model thigh and shank during a running stride (multiple marker method).

Mean, standard deviation and coefficient of variation of 36 samples

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mean length (image mm)</th>
<th>s.d.</th>
<th>C of V %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thigh</td>
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<td>0.6</td>
<td>1.1</td>
</tr>
<tr>
<td>Shank</td>
<td>57.9</td>
<td>0.7</td>
<td>1.2</td>
</tr>
</tbody>
</table>
Figure 3.5: Change in the apparent length of the thigh during a running stride

(a) original marker system
(b) modified marker system
3:5 **SIGNAL AND NOISE COMPONENTS OF CINEPHOTOGRAPHIC DATA**

3:5.1 Introduction

Digitised film data is "noisy". The data contains, in addition to information about the movements of the body, an amount of noise or random error brought about by minor distortions in the projected image, frame to frame image shift, graininess in the film itself, and operator error when the film is digitised. It is obviously desirable to maximise the information content of the data and at the same time minimise the noise content. Consideration must be given therefore to the way in which the information is sampled and the way in which the noise may be filtered. This is particularly important when data is to be differentiated since differentiation has a greater amplifying effect on the higher (noise) frequencies in the data than it does on the lower (signal) frequencies.

A wide range of filtering techniques is available for reducing the noise content of numerical data providing the range of signal frequencies and noise frequencies are largely distinct. The filtering problem thus reduces to the determination of an appropriate sampling frequency for recording information, and the selection of a cut-off frequency for the filter.

The selection of an appropriate sampling frequency and hence minimum camera frame speed requires careful consideration. In general, the higher the sample rate, the more high frequency information that will be recorded. There is however, a finite upper boundary to the frequency spectrum of the human movement, and much higher sampling frequencies will add only noise to the recorded information. In the case of film data, there is a small random variance due to the error in the operator's positioning of the digitiser cursor in relation to the film. This noise will always add to the data at the frequencies up to the maximum recordable by the selected sampling frequency (i.e. half the sampling frequency). Of equal importance, the man-hours and computer-time required to analyse film records increase at least proportionately with the sampling frequency. It is therefore desirable to select the minimum sampling frequency that will enable the information content of the movement to be determined to the required resolution. The minimum sampling frequency can be determined the Nyquist sampling theorem (Bendat and Piersol, 1971) which requires that the sampling frequency be at least twice the maximum frequency of interest. In order to give adequate separation in the frequency domain between the signal and the digitising noise a ratio in excess of two may be required,
Winter et al (1974) recorded walking subjects using a television technique. In excess of 99% of the information content of marker trajectories was contained in the first 7 harmonics. However, this data concerns walking rather than running subjects, the maximum frequency content of the recorded information was limited to 50Hz by the TV technique, and the data from the TV system appears to be somewhat more noisy than that collected using film techniques. Winter’s results are not therefore directly applicable to the requirements of this study, therefore.

This section describes a preliminary investigation of the signal and noise content of typical film data of a running subject.

3.5.2 Methods
The signal and noise properties of film records of treadmill running were examined using a sample data set. A subject was filmed in the sagittal plane using an "Hicam" high-speed cine camera while running on a motorised treadmill at a speed of 10 mph (4.47 m.s⁻¹). A camera speed of 200 frames per second was selected. The motion of a marker placed on a shoe of the subject, over the 5th metatarso-phalangeal joint was recorded from the film using an SAC model GP7 sonic digitiser.

The duration of the sample stride cycle was 0.66 seconds, giving 133 coordinate pairs in the data set. While this motion has the largest amplitude of any in treadmill running, it also has the highest velocities and high accelerations. The sample was assumed to contain both signal and noise components (SN).

A second "noise only" data set (N) was determined by repeatedly digitising a marker on a single frame of film in a manner similar to that adopted when digitising film records.

The example data set is illustrated in Figure 3.6.

The harmonic content of the sample data sets was determined by Fourier analysis of the raw data. The fundamental period was assumed to be the duration of the example stride cycle, i.e 0.66 seconds.
Figure 3.6: Sample data set (SN) used in the analysis of signal and noise characteristics of kinematic data

(a) locus of foot marker
(b) x and y components.
Any time series can be approximated by a Fourier series of the form

\[ x = a_0/2 + \sum_{i=1}^{n} a_i \cos(2\pi it/T) + \sum_{i=1}^{n} b_i \sin(2\pi it/T) \]  

Eqn 3.2

where

- \( x = f(t) \) is the cyclical function being approximated
- \( T \) is the period of the fundamental cycle
- \( a_i, b_i \) are the \( n \) harmonic coefficients.

The \( i \)th harmonic represents a frequency component of \( i/T \) Hz with an amplitude of \( \sqrt{a_i^2 + b_i^2} \). The coefficients \( a \) and \( b \) are determined from the following equations:

\[ a_i = -\frac{2}{N} \sum_{j=1}^{N} x_j \cos(2\pi jt/T) \]  

Eqn 3.3a

\[ b_i = -\frac{2}{N} \sum_{j=1}^{N} x_j \sin(2\pi jt/T) \]  

Eqn 3.3b

Where \( N \) is the number of items in the sample data set.

The significance of the contribution of a particular harmonic component can be determined by examining its effect on the residual sum of squares of the fit of the expanded Fourier series with respect to the raw data.

Let \( S_n \) be the residual sum of squares of the fit of a Fourier series in \( n \) harmonics. The variance due to the \( n \)th term of the series is given by

\[ V_1 = S_n - S_{n-1} \]  

Eqn 3.4

and has 1 degree of freedom.
The overall residual variance, \( V_2 = S_n \) has \( (N-n-1) \) degrees of freedom. The F-ratio \( V_1/V_2 \) was used to determine the significance of reduction in the overall residual variance by the addition of the \( n \)th coefficient.
The effect of adopting different sampling frequencies was determined by analysis of subsets of the original SN and N data. By selecting every second third, fourth, etc., frame of the original data set, data at effective sampling frequencies of 100Hz, 67Hz, 50Hz, 50Hz, 33Hz and 16Hz were obtained, enabling comparisons to be made.

3.5.3 Results

Figure 3.7 shows the normalised frequency/amplitude characteristics of the SN and N samples. The noise amplitude was fairly constant over the range up to 100Hz. Signal+noise amplitudes become constant and similar to the noise level at frequencies above about 10Hz in the x-component of the SN data and 15Hz in the y-component.

Tables 3.9 (a and b) show the results of an analysis of residual variance on the first 10 harmonics of the SN sample data. The critical value of F at the 0.05 probability level is 3.9201. Therefore, in neither x nor y components did the 4th and higher harmonics contribute significantly to the reduction of residual variance. This is further illustrated in Table 3.10 which shows the proportional contribution of each harmonic term to the overall variance of the fitted Fourier series. In both cases, more than 99.9% of the total sum of squares about the mean value is contained at harmonic frequencies of 6 Hz and less.

The effect of differentiation on the frequency characteristics of the y components of the SN and N data sets is shown in Figure 3.8. The derivatives were determined by direct algebraic differentiation of the approximating series in the frequency domain. There is a disproportionate amplifying effect on the higher frequencies of the derivative which result in the amplitudes becoming significant, in relation to the signal component, at frequencies above about 10Hz. This effect is even more pronounced in the second derivative.

Figure 3.9 shows a surface representing the interactive effect of sampling frequency and harmonic order of the approximating Fourier series on the root mean square residual of the fit onto the x-component of the SN sample. At harmonic orders of 4 and above, and sampling frequencies of 25Hz and above, the residual is largely unaffected by different combinations of sampling frequency and harmonic order. At combinations of high harmonic orders and low sampling frequencies, the residual increases since there are not enough data points to correctly compute the higher harmonics. The
Figure 3.7: Normalised frequency characteristics of signal and noise (SN) and noise only (N) data samples

(a) x component
(b) y component
Table 3:9a: Analysis of residual variance of the Fourier approximation of the x-component of the SN sample data set (see text).

<table>
<thead>
<tr>
<th>h</th>
<th>V2</th>
<th>V1</th>
<th>df</th>
<th>F</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>*1000</td>
<td>*10000</td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>13274.140</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>321.330</td>
<td>12952.8100</td>
<td>131</td>
<td>40.31</td>
</tr>
<tr>
<td>2</td>
<td>84.195</td>
<td>237.1354</td>
<td>130</td>
<td>2.82</td>
</tr>
<tr>
<td>3</td>
<td>2.525</td>
<td>81.6705</td>
<td>129</td>
<td>32.34</td>
</tr>
<tr>
<td>4</td>
<td>2.412</td>
<td>0.1135</td>
<td>128</td>
<td>0.05</td>
</tr>
<tr>
<td>5</td>
<td>1.738</td>
<td>0.6745</td>
<td>127</td>
<td>0.39</td>
</tr>
<tr>
<td>6</td>
<td>1.356</td>
<td>0.3825</td>
<td>126</td>
<td>0.28</td>
</tr>
<tr>
<td>7</td>
<td>1.3366</td>
<td>0.0205</td>
<td>125</td>
<td>0.02</td>
</tr>
<tr>
<td>8</td>
<td>1.190</td>
<td>0.1465</td>
<td>124</td>
<td>0.12</td>
</tr>
<tr>
<td>9</td>
<td>1.046</td>
<td>0.1445</td>
<td>123</td>
<td>0.14</td>
</tr>
<tr>
<td>10</td>
<td>1.043</td>
<td>0.0035</td>
<td>122</td>
<td>0.01</td>
</tr>
</tbody>
</table>

h = harmonic order of Fourier series
V2 = residual variance
V1 = variance attributable to last term in the series
df = degrees of freedom of V2
F = F-ratio, V1/V2
Table 3:9b: Analysis of residual variance of the Fourier approximation of the \( y \)-component of the SN sample data set (see text).

<table>
<thead>
<tr>
<th>( h )</th>
<th>( V_2 )</th>
<th>( V_1 )</th>
<th>df</th>
<th>( F )</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(*1000)</td>
<td>(*1000)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>5313.860</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>155.687</td>
<td>5158.173</td>
<td>131</td>
<td>33.13</td>
</tr>
<tr>
<td>2</td>
<td>4.798</td>
<td>150.889</td>
<td>130</td>
<td>31.44</td>
</tr>
<tr>
<td>3</td>
<td>4.017</td>
<td>0.781</td>
<td>129</td>
<td>0.19</td>
</tr>
<tr>
<td>4</td>
<td>3.116</td>
<td>0.901</td>
<td>128</td>
<td>0.28</td>
</tr>
<tr>
<td>5</td>
<td>2.627</td>
<td>0.489</td>
<td>127</td>
<td>0.19</td>
</tr>
<tr>
<td>6</td>
<td>2.319</td>
<td>0.308</td>
<td>126</td>
<td>0.03</td>
</tr>
<tr>
<td>7</td>
<td>1.710</td>
<td>0.609</td>
<td>125</td>
<td>0.36</td>
</tr>
<tr>
<td>8</td>
<td>1.550</td>
<td>0.160</td>
<td>124</td>
<td>0.13</td>
</tr>
<tr>
<td>9</td>
<td>1.470</td>
<td>0.080</td>
<td>123</td>
<td>0.05</td>
</tr>
<tr>
<td>10</td>
<td>1.422</td>
<td>0.488</td>
<td>122</td>
<td>0.34</td>
</tr>
</tbody>
</table>

\( h \) = harmonic order of Fourier series

\( V_2 \) = residual variance

\( V_1 \) = variance attributable to last term in the series

\( df \) = degrees of freedom of \( V_2 \)

\( F \) = \( F \)-ratio, \( V_1/V_2 \)
Table 3.10: Proportional contribution of each harmonic of the best fitting Fourier series to $x$ and $y$ components of the SN sample data

<table>
<thead>
<tr>
<th>$h$</th>
<th>freq</th>
<th>$p_x$</th>
<th>$p_y$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Hz</td>
<td>%</td>
<td>%</td>
</tr>
<tr>
<td>1</td>
<td>1.51</td>
<td>97.58</td>
<td>97.07</td>
</tr>
<tr>
<td>2</td>
<td>3.03</td>
<td>1.79</td>
<td>2.92</td>
</tr>
<tr>
<td>3</td>
<td>4.54</td>
<td>0.62</td>
<td>0.07</td>
</tr>
<tr>
<td>4</td>
<td>6.06</td>
<td>&lt;0.01</td>
<td>0.01</td>
</tr>
<tr>
<td>5</td>
<td>7.58</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td>6</td>
<td>9.09</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td>7</td>
<td>10.60</td>
<td>&lt;0.01</td>
<td>0.01</td>
</tr>
<tr>
<td>8</td>
<td>12.12</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td>9</td>
<td>13.63</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td>10</td>
<td>15.15</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
</tr>
</tbody>
</table>

$h$ = harmonic order of Fourier series

freq = equivalent harmonic frequency

$p_x$ and $p_y$ are determined from the proportion of the total sum of squares residual about the mean value of the raw data accounted for by each harmonic term.
Figure 3.8: Effect of differentiation on the frequency characteristics of the \( y \) components of signal + noise (SN) and noise only (N) data samples.
Figure 3.9: Effect of sampling frequency ($f_s$) and harmonic order ($h$) on the goodness of fit of a Fourier series to sample kinematic data ($R$).
"plateau" of the surface probably represents the residual "noise" content of the SN data.

3.5.4 Discussion

The results of this study show that the majority of the information contained in the sample data set occurs at frequencies up to 4th harmonic (~6 Hz.). The movement of the foot during the treadmill running stride is characteristically a high-amplitude low frequency movement, however and these results may not be representative of the frequency characteristics of, say, parts of the torso where the fundamental cycle frequency is twice that of the foot. James and Brubaker (1973) calculated the contributions of the first 7 harmonics of a Fourier series fit to the motion of hip and head markers during a running stride. Over 98% of the information in the data was described by the first four harmonics and over 99.8% by the first five. This result confirms that the kinematics of human body movements are predominantly below 6Hz in the frequency domain.

In the example used here, the noise due to digitising became significant, in relation to the "signal" at frequencies above about 10 Hz. If the noisy data is differentiated directly, the contribution of the higher noise frequencies becomes significant. In the second derivative, noise frequencies would undoubtedly dominate the data, and the true second derivative of the signal component would be completely masked by it. It is important, therefore, that the higher noise frequencies be filtered out before the data is differentiated. On the basis of the results described above, a low-pass filter with a cut-off frequency in the range 5-10Hz would appear appropriate. Other researchers studying the kinematics of running have come to the same conclusion. Cut-off frequencies of 5Hz (Williams and Cavanagh, 1983; Ito et al, 1983), 7Hz (Norman et al, 1976) and 8Hz (Winter, 1983) have been reported.

Since the information content of human movement data is predominantly below 6Hz in the frequency domain then a sampling rate of 12 Hz is all that is needed to satisfy the minimum requirements of the Nyquist sampling theorem. However, it is generally recommended that a sampling rate some 4-5 times the minimum is used (Lees, 1980). In theory, therefore, a sampling frequency of 50 Hz is adequate to record the kinematics of a runner. This was confirmed by the comparison of the effects of different combinations of sampling frequency and harmonic order of the approximating Fourier series on the goodness of fit of the approximation to the original 200Hz sample (Figure 3.9). No significant reduction in the root mean square residual
occurred at sample frequencies above 25Hz and harmonic orders above the 4th. Camera frame rates of 100Hz and above have often been used in cinephotographic studies of running (Williams and Cavanagh, 1983; Luhtanen and Komi, 1978, 1980; Norman et al, 1976). Winter (1983) used a sample rate of 50Hz to record the kinematics of jogging subjects. Ito et al (1983) used a camera rate of 50Hz to record running at speeds of up to 6.1 m.s⁻¹ and Chapman and Caldwell (1983) used a camera rate of 65 frames per second to record athletes at running speeds of up to 9.5 m.s⁻¹, which is probably the highest recorded treadmill running speed.

The advantages of using higher camera rates include the reduction of blur in cameras with fixed shutters. Also, digitising noise is introduced into the data at higher frequencies, giving greater separation from the signal component of the data in the frequency domain. The only problems with higher camera frequencies occur if this higher frequency noise is not adequately filtered before the data is differentiated.

On the basis of these results, it is concluded that satisfactory recording of the kinematics of treadmill running, at the speeds to be used in this study, can be obtained with sampling frequencies of 50Hz and above; and that the raw data should be low-pass filtered with a cut-off frequency in the range 5-10Hz before differentiation.
3.6 SPATIAL CALIBRATION

3.6.1 Introduction

The determination of real-space coordinates from film images is a non-trivial problem, especially when multi-camera systems are used for three-dimensional analyses. The cameras and projection system jointly transform points in the real world into points on a set of projected images. The general calibration problem is to determine an inverse transformation in order to calculate coordinates in the real-world reference frame from image coordinates. Primarily, there is a need for the scale and orientation of the images with respect to a real world coordinate system. In addition, the image data can be non-linearly distorted by lenses in the camera and projection system.

This section describes the methods used to determine "real-world" coordinates from film images, in the cinephotographic experiments described in this study.

Gagnon et al (1983) solve for the three orientation angles of ice skates from the image, in a single camera, of three orthogonal rods attached to the skates. This method is not generally applicable to all three-dimensional motions for which a minimum of two cameras are normally required.

Duquet et al (1971) describe a method for determining the three dimensional coordinates of points in object space from images provided by two cameras set at right angles to one another. Miller and Fetak (1973) developed a geometric correction for perspective errors in images from three cameras. Again the camera positions had to be precise—in this case, the cameras needed to be positioned so that their optical axes were in the horizontal plane and intersected at a single point. Other systems where the camera positions are not fixed, rely on the calculation of camera positions from the images of known coordinates in object space. The characteristics of the camera are determined by minimising the least squares fit of the known calibration points to the theoretical predictions from the camera model (e.g. Woltring, 1975).

Andriacchi et al (1979) corrected for calibration and image distortion by calculating a line passing through the object point, a calibration plane and the camera focal point. The three-dimensional coordinates of an object point are determined from the orientations of two such lines from two or more cameras, intersecting at the object point. Huntingdon et al (1979) apply a similar method to the determination of the three-dimensional
angular orientation of the knee in walking subjects, using three cameras. Again, the transformation involves the calculation of the parameters of the camera model.

Abdel-Aziz and Karara (1971) describe a "direct linear transformation" between object and image space coordinates without the need to make an initial approximation of the camera parameters. This method and its derivatives (e.g. Van Gheluwe, 1978) are now in common usage in three dimensional biomechanical analyses.

All these methods are based on a set of known "control points" in object space, which permit the transformation between image and object space to be calculated. It is possible to eliminate the need for precise knowledge of the positions of control points by a method known as "the bundle method of analytical photogrammetry" (Brown, 1969) or "analytical self-calibration" (Kenefick et al, 1972). Even when modified for human movement analysis (Woitring, 1980), this method has a high degree of complexity.

In view of the known inaccuracies of joint centre determination, and the relative motion of body surface markers, very high precision calibration methods are not required in studies of human movement. Object coordinate errors of a few millimetres are undoubtedly within the limits on resolution imposed by joint centre estimation. The resolution of estimates of the anthropometric and inertial properties of the link-segmental model itself have a much lower relative precision than this.

For the purposes of this study a simplified direct transformation between image and object space coordinates was devised, based on least squares approximation methods. Primary considerations in the development of the method were that it should be computationally simple, should be adaptable to both two- and three-dimensional analyses, and that some estimate of the precision of the predicted object space coordinates should be available.
3.6.2 A general film calibration method.

The location of a point in three dimensional space is defined by the set of Cartesian coordinates \((X,Y,Z)\). The same point is represented in image space by the set \((x_i,y_i,z_i, \ i=1,I)\) where \(I\) is the number of simultaneously recorded images of the object point. For a two camera measurement system, the image coordinate set can be reduced to the set \((x,y,z)\) where \(x\) and \(y\) are recorded from one camera image and \(z\) from the second.

Given conventional cameras such that each point in object space maps to a unique point in each image space; each member of the image coordinate set (say the generalised coordinate \(q\)) is a function of the object coordinates:

\[
q = f(X,Y,Z) \quad \text{Eqn 3.5}
\]

Similarly, each coordinate in the object space may be modelled as some function of the image space coordinates:

\[
Q = F(x,y,z) \quad \text{Eqn 3.6}
\]

Calibration of the image data therefore involves the determination of the transformation function \(F\) for each object space coordinate. The approach adopted here involves:

1: The determination of an appropriate general form for the approximation to \(F\).

2: For each individual camera setup, the determination of the parameters of a specific form of \(F\) using from the real and image coordinates of known points in the object space.

While the camera parameters are not explicitly stated in this formulation, they are implicit in the transformation function.

3.6.3 General form of the transformation function

The basic transformations brought about by the camera are those of translation, rotation and magnification of the reference axes. These bring about linear distortion of the image of the form

\[
Q_{3D} = a_0 + a_1x + a_2y + a_3z \quad \text{Eqn 3.7}
\]
where $Q_{3D}$ is a generalised Cartesian coordinate in a 3 dimensional reference frame and $a_0 \ldots a_3$ are constants.

There may, however, be non-linearities in the image coordinates, brought about by spherical distortion in camera and projection lenses. Also, if the cameras are not aligned orthogonally with the planes of the object coordinate system and similarly if the projected image is not aligned correctly with the plane of the projection screen, the object reference space will be trapezoidal in the image space.

These linear effects can be modelled by assuming the coefficients $a_0 \ldots a_3$ are themselves linear functions of $(x,y,z)$.

Thus equation 3.7 becomes

$$Q_{3D} = (b_0 + b_1x + b_2y + b_3z) + x(b_4 + b_5x + b_6y + b_7z) + y(b_8 + b_9x + b_{10}y + b_{11}z) + z(b_{12} + b_{13}x + b_{14}y + b_{15}z)$$

where $b_0 \ldots b_{15}$ are constants.

Simplifying and combining coefficients:

$$Q_3 = F(x,y,z) = c_0 + c_1x + c_2y + c_3z + c_4xy + c_5yz + c_6xz + c_7x^2 + c_8y^2 + c_9z^2$$

where $c_0 \ldots c_9$ are constants.

In the two-dimensional case, with one image set $(x,y)$, Eqn 3.9 reduces to the form

$$Q_{2D} = F(x,y) = c_0 + c_1x + c_2y + c_3xy + c_4x^2 + c_5y^2$$

Equation 3.9 has 10 unknowns which may be determined with knowledge of the real and image coordinates of 10 or more control points. The two dimensional transformation has 6 unknowns and could be solved given a minimum of 6 unknown control points.
3.6.4 Least squares solution of the transformation function

While it is possible to solve for the unknowns in the three dimensional calibration transformation equation with a minimum of 10 known control points in object space, this is not generally advisable. Firstly, in order to avoid the need for extrapolation in circumstances where the transformation may have been fitted to local distortion, control points should be widely distributed throughout the object space. Where a large volume is to be calibrated, 10 control points may well be insufficient to satisfactorily encompass the whole volume. More importantly, there will always be error in the determination of the object space coordinates of control points. It is therefore advisable to determine a least squares solution to the transformation problem using more than the minimum number of control points.

Assuming a transformation between the a Cartesian object space coordinate, Q and its image coordinates (x,y,z), of the form of Equation 3.9; and given the image and object space coordinates of N control points (N > 9), it is required to determine the set of coefficients c0...c9 which minimise the residual sum of squares

\[ R = \sum_{i=1}^{N} (Q_i - F(x_i, y_i, z_i))^2 \]  \hspace{1cm} \text{Eqn 3.11}

The partial derivatives of R with respect to the coefficients c0 ... c9 are

\[ \frac{\partial R}{\partial c_0} = -2 \sum (Q_i - F(x_i, y_i, z_i)) \]  \hspace{1cm} \text{Eqns 3.12a - 12j}

\[ \frac{\partial R}{\partial c_1} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-x_i) \]
\[ \frac{\partial R}{\partial c_2} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-y_i) \]
\[ \frac{\partial R}{\partial c_3} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-z_i) \]
\[ \frac{\partial R}{\partial c_4} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-x_i y_i) \]
\[ \frac{\partial R}{\partial c_5} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-y_i z_i) \]
\[ \frac{\partial R}{\partial c_6} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-x_i z_i) \]
\[ \frac{\partial R}{\partial c_7} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-x_i^2) \]
\[ \frac{\partial R}{\partial c_8} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-y_i^2) \]
\[ \frac{\partial R}{\partial c_9} = -2 \sum (Q_i - F(x_i, y_i, z_i))(-z_i^2) \]

When R is a minimum, the partial derivatives \( \frac{\partial R}{\partial c_s} \), s = 0,1,...,9 are all equal to zero.
Expanding $F(x,y,z)$ in equations 3.12 and equating the partial derivatives to zero gives the set of normal equations for the least squares solution.

In matrix form, the Normal equations are:

$$
\begin{bmatrix}
N \sum x & \sum y & \sum z & \sum xy & \sum yz & \sum xz & \sum x^2 & \sum y^2 & \sum z^2 \\
\sum x & \sum x^2 & \sum xy & \sum xz & \sum x^2y & \sum x^2z & \sum x^3 & \sum x^2y & \sum x^2z \\
\sum y & \sum xy & \sum y^2 & \sum yz & \sum x^2y & \sum y^2z & \sum y^3 & \sum x^2y & \sum y^2z \\
\sum z & \sum xz & \sum yz & \sum z^2 & \sum x^3 & \sum y^3 & \sum z^3 & \sum x^3z & \sum y^3z \\
\sum xy & \sum x^2y & \sum xy^2 & \sum x^2yz & \sum x^3y & \sum x^2y^2 & \sum x^3y^2 & \sum x^3yz & \sum x^2y^2 \\
\sum yz & \sum y^2z & \sum x^2yz & \sum y^3z & \sum y^2z & \sum x^3yz & \sum y^3z & \sum x^3y^2 & \sum x^2y^2 \\
\sum xz & \sum x^2z & \sum x^3z & \sum x^3yz & \sum x^3y^2 & \sum x^3y^2 & \sum x^3y^3 & \sum x^3y^2 & \sum x^3y^3 \\
\sum x^2 & \sum x^3 & \sum x^2y & \sum x^2z & \sum x^3y & \sum x^3z & \sum x^2y^2 & \sum x^2y^2 & \sum x^2y^2 \\
\sum y^2 & \sum y^3 & \sum y^2z & \sum y^3z & \sum y^2z & \sum y^3z & \sum y^2z & \sum y^2z & \sum y^2z \\
\sum z^2 & \sum z^3 & \sum x^2yz & \sum z^3 & \sum x^2yz & \sum z^3 & \sum x^2yz & \sum x^2yz & \sum x^2yz \\
\sum x^2z & \sum y^2z & \sum x^2yz & \sum x^3z & \sum x^2yz & \sum y^3z & \sum x^3z & \sum x^3z & \sum x^3z \\
\end{bmatrix}
\begin{bmatrix}
C_0 \\
C_1 \\
C_2 \\
C_3 \\
C_4 \\
C_5 \\
C_6 \\
C_7 \\
C_8 \\
C_9 \\
\end{bmatrix}
= \begin{bmatrix}
\sum \Theta \\
\sum \Phi_x \\
\sum \Phi_y \\
\sum \Phi_z \\
\sum \Phi_{xy} \\
\sum \Phi_{yz} \\
\sum \Phi_{xz} \\
\sum \Phi_{xy}^2 \\
\sum \Phi_{yz}^2 \\
\sum \Phi_{xz}^2 \\
\end{bmatrix}
$$

Eqn 3.13

In abbreviated form

$$
\begin{bmatrix}
A & \boldsymbol{U} = \boldsymbol{S}
\end{bmatrix}
$$

where $\boldsymbol{E}$ = matrix of equations

$\boldsymbol{U}$ = array of unknowns

$\boldsymbol{S}$ = array of solutions

These simultaneous equations can be solved by conventional numerical methods (e.g. Gaussian elimination; Bajpai et al., 1974). Independent solutions for $X, Y$ and $Z$ object coordinates ($U(X)$, $U(Y)$ and $U(Z)$) are obtained from the same equation matrix, but with appropriate values of $\Theta$ substituted in the array of solutions.
Once calculated, the three sets $U(X)$, $U(Y)$ and $U(Z)$ can be transformed into a single transformation matrix, permitting direct transformation of film coordinates to object space coordinates thus:

$$
\begin{bmatrix}
X \\
Y \\
Z
\end{bmatrix} =
\begin{bmatrix}
1 & C_{0x} & C_{0y} & C_{0z} \\
X & C_{1x} & C_{1y} & . \\
Y & C_{2x} & C_{2y} & . \\
Z & . & . & . \\
x & . & . & . \\
y & . & . & . \\
z & . & . & . \\
x^2 & . & . & C_{7z} \\
y^2 & . & C_{7y} & C_{8x} \\
z^2 & C_{9x} & C_{9y} & C_{9z}
\end{bmatrix}
$$

Eqn 3.15

On a mainframe computer or other facility with a large available memory and a library of matrix manipulation software, an entire film sequence can be transformed in one macro-manipulation of the form:

$$
\begin{bmatrix}
X_1 \\ Y_1 \\ Z_1
\end{bmatrix} =
\begin{bmatrix}
1 & X_1 & Y_1 & Z_1 & X_1Y_1 & Y_1Z_1 & X_1^2 & Y_1^2 & Z_1^2
\end{bmatrix}
$$

$$
\begin{bmatrix}
C_{0x} & C_{0y} & C_{0z} \\
C_{1x} & C_{1y} & . \\
C_{2x} & C_{2y} & . \\
. & . & . \\
. & . & . \\
. & . & . \\
. & . & . \\
C_{7z} & . & . \\
C_{7y} & C_{8x} & . \\
C_{9x} & C_{9y} & C_{9z}
\end{bmatrix}
$$

Eqn 3.16

Such a manipulation becomes one statement of a high level programming language when matrix subroutines are available. When mini-computers are being used, available memory or operating system limitations may require the matrix transformations to be programmed explicitly and the image coordinates to be transformed sequentially. An example computer programme to perform these tasks for the two-dimensional transformation is included in Appendix E.
3.6.5 Implementation

The method described above has been used for both two and three dimensional calibration of film of treadmill runners. In both cases, the control points were established by the construction of rigid frames. Also, fixed points on the treadmill frame itself were established and marked for use as control points. The positions of the control points in relation to a datum point on the treadmill base were measured, the two dimensional case with plumb-line, set-square and metre rule. In the three dimensional case, three reference points were established on the bed of the treadmill, and the distance from each reference point to each calibration point measured with a steel tape. Given the three dimensional coordinates of the three reference points, \((x_1, y_1, z_1), (x_2, y_2, z_2), (x_3, y_3, z_3)\) and the distance of the unknown control point \((X, Y, Z)\) from each of the reference points \((L_1, L_2, L_3)\), the following simultaneous equations can be derived:

\[
\begin{align*}
(X-x_1)^2 + (Y-y_1)^2 + (Z-z_1)^2 &= L_1^2 \\
(X-x_2)^2 + (Y-y_2)^2 + (Z-z_2)^2 &= L_2^2 \\
(X-x_3)^2 + (Y-y_3)^2 + (Z-z_3)^2 &= L_3^2
\end{align*}
\]

Eqs 3.17a - 17c

There are thus three equations in three unknowns which can be solved for the position coordinates \((X, Y, Z)\) of the control points.

When film of treadmill runners was recorded, images of the control points were recorded before the experiment commenced and then the calibration frame was removed. The three dimensional transformation was performed using 24 control points spread over a volume 2m x 2m x 1m. The two dimensional transformation was implemented using 12 control points distributed over an area of 2m x 2m. The calibration process was repeated every time the cameras were moved. When the film was analysed, the image coordinates of the control points were recorded and the calibration transformation matrix was determined by numerical solution of Equation 3.13 using a Gaussian elimination method. For three-dimensional analyses, the calibration transformation matrix and the transformed coordinates were determined using a computer programme written in ALGOL 68, running on an ICL 1904S mainframe computer. Full use was made of the facilities available for direct manipulation of matrices with the available system library procedures. The two dimensional transformation was implemented using a
FORTRAN IV programme on a PDP11-03 laboratory mini-computer. A listing of this latter programme has been included in Appendix E.

3.6.6 Determination of error in film calibration.

The goodness of fit of the transformation function to the calibration data can be determined by examination of the residual variance. The determination of the calibration function is essentially a multiple regression procedure. Coefficients of multiple determination, multiple correlation coefficients and the standard error of the estimate can be determined in the normal way, as for a multiple regression problem (See Cohen and Holliday, 1982; pp. 104-110). Typically the the coefficients of multiple determination are in excess of 0.99. A value of less than this generally indicates some error in the determination of the image and object coordinate data. Similarly, highly significant coefficients in the higher order terms of the calibration function are a sign that the data entered into the calibration procedure is not correct.

These statistical procedures can be used to ensure that a valid transformation matrix has been determined, and to calculate the residual variance not accounted for by the calibration procedure. However, this does not give a measure of the error in the determination of the coordinates of unknown points in object space. This error can be determined by using the calibration transformation matrix and image coordinates to estimate the positions of a second set of control points in object space. The second set of control points must of course, be independent of the control points used to determine the calibration matrix. Table 3.11 shows the mean and standard deviation of the difference between measured and estimated coordinates of the object space control points. The two dimensional calibration produced better reproduction than the three dimensional analysis, probably due to the greater complexity and greater difficulty in establishing control point positions in the latter. Of the three axes, positions in the lateral horizontal axis (y) were the least reliable, with differences of up to 1cm between the measured positions of additional control points and the estimate of their position from film images.
Table 3.11: Errors in film calibration

Mean and standard deviation of absolute differences between measured and estimated positions of control points in object space (cm)

<table>
<thead>
<tr>
<th>Axis</th>
<th>Mean</th>
<th>s.d.</th>
</tr>
</thead>
<tbody>
<tr>
<td>X</td>
<td>0.51</td>
<td>0.10</td>
</tr>
<tr>
<td>Y</td>
<td>0.51</td>
<td>0.19</td>
</tr>
<tr>
<td>Z</td>
<td>0.61</td>
<td>0.12</td>
</tr>
</tbody>
</table>

3 dimensional

\[ n = 60 \]

2 dimensional

\[ n = 117 \]
3:7 TIME CALIBRATION

The frame rates of cameras used in this study were calibrated by filming the display of an electronic timer/counter running in timer mode with the clock display incrementing every 1/1000th of a second. The time displayed on each frame of film from the moment the camera was switched on, for a period of ten seconds, were recorded and plotted as a function of frame number. Thus the constancy of the frame rate and the time required to reach constant running speed could be established. All three cameras examined achieved a constant frame rate after a short period which was maintained for the remainder of the ten second sample period. In no case was the measured frame rate equal to the nominal frame rate. The nominal and actual frame rates and run up times of each camera are recorded in Table 3.12. In this table, 'run up' time is the time required to reach a constant frame rate, with an 50% safety margin included. None of the frames recorded during this period of a film sequence were used for analysis.

3:7.1 Correction of non-linearities in camera timebase

In the case of the Bolex H16 camera used for two dimensional analyses, a small non-linearity in the frame interval was detected. This amounted to a variation in the measured frame rate of approximately 3 frames per second between the end of the run-up period and the end of the 10 second film sequence. While this represents a discrepancy of only 0.25 frames per second over the duration of a typical stride, it is possible for the frame interval of sample strides taken from different parts of the film sequence to differ by up to 5%.

When a particular piece of film was analysed, the starting point of the recorded data in the film sequence was noted. This enables the time-base of the frames to be established from the calibration runs on the camera.

For convenience all film sequences were normalised to a common, time base using a Newtonian interpolation algorithm designed for use with data collected at unequal intervals. The Newtonian interpolation method effectively calculates a local approximation of the data time-series by a polynomial function of given order. The polynomial is fitted analytically to the minimum number of points by an analytical procedure rather than by a least squares fit, so the frequency characteristics of the data are largely unaffected by the interpolation routine, providing the interpolation interval is similar to the mean interval of the raw data.
Table 3.12: Camera speed calibration results.

<table>
<thead>
<tr>
<th>Camera Type</th>
<th>Motor Type</th>
<th>Frame rate Nominal Hz</th>
<th>Frame rate Recorded Hz</th>
<th>'Run up' time s</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bolex EBM</td>
<td>Electric</td>
<td>50</td>
<td>52</td>
<td>1.4</td>
</tr>
<tr>
<td>Bolex EBM</td>
<td>Electric</td>
<td>50</td>
<td>52</td>
<td>1.5</td>
</tr>
<tr>
<td>Bolex H16</td>
<td>Clockwork</td>
<td>64</td>
<td>68</td>
<td>1.8</td>
</tr>
</tbody>
</table>
The algorithm used to perform the interpolation was that described by Khabaza (1966, pp 209-216), using recursively defined divided differences. A FORTRAN IV implementation of the algorithm is listed in Appendix F.

A further possible use as for the interpolation algorithm is the generation of data for the locations of body markers that are intermittently obscured from the view of a sagittal plane camera. For example, the arms obscure the rib marker for part of the running cycle and in some subjects can also obscure the hip marker momentarily.

3:7.2 Synchronisation

For three dimensional analysis, two Bolex EBM Electric cameras were placed on approximately orthogonal axes with respect to the centre of the object space; one recording movements in approximately the frontal plane and the other recording movements in approximately the sagittal plane. The calibration system used (section 3:6) does not require the cameras to be placed at any particular configuration, but the best results will be obtained when the cameras are approximately orthogonal, since the overlap in the information recorded by the two cameras is minimal, giving a well-conditioned calibration matrix. In order to be able to synchronise images from the two cameras, two clocks were placed in the field of view of the sagittal plane camera. One was a 3-second sweep clock with a single hand some 40cm long. The "second clock" consisted of an electric motor synchronised to the 50 cycles AC in the main electricity supply. The motor was geared down to drive a clock hand some 10cm in diameter with a sweep period of 1/6th of a second. A mirror was arranged in such a way that a reflected image of the clocks could also be seen in the frontal plane camera. Examination of the images of the clocks from both views enabled the film recordings to be post-synchronised to the nearest frame, the position of the subject acting as an additional check that an error of greater than three seconds had not been made.
3:8 DIGITISING PROCEDURES

3:8.1 Digitising the film

For each subject treadmill running trial that was recorded on film, one whole stride from foot contact to foot contact of the same leg was digitised for analysis.

The film was projected using a Vanguard M16C projection head mounted on a PCD projection cabinet. An SAC 'Graf-Pen' GP6 sonic digitiser was mounted on the front of the projection cabinet to enable the image coordinates of joint centres and extremities to be recorded.

A number of strides from the film sequence were superficially examined to determine the duration of each stride, and a stride from the middle of the sequence was selected on the basis that the stride time was not atypical. The position of the selected stride in the film sequence was recorded.

Before digitising commenced, the appropriate calibration frame was selected and the control points digitised for use with the calibration procedures described in section 3:6. Four points on the fixed background of the calibration frame were also digitised. The points selected were clearly visible in each frame to be digitised and as near to the four corners of the image as possible. These additional "fixed" image points were subsequently digitised in every data frame to enable any 'frame-shift' of the film in the projector to be detected and corrected for.

The film was digitised frame by frame, beginning with 5 frames before foot contact and ending 5 frames after the next foot contact on the same foot. The additional frames permit the use of numerical differentiation and filtering methods without distortion of the data of interest.

In the case of three dimensional analyses, the additional film from the frontal view of the same stride was digitised in the same manner. Only the positions of those joint centres and extremities on the side of the body recorded in the sagittal plane film were digitised.

All digitising was performed "off-line", with coordinate data being stored on a Sykes Commstor II independant 7" flexible diskette storage unit. This raw data was transferred to an ICL 1904S mainframe computer for three dimensional analysis using ALGOL 68R software or to a DEC PDP11/03 laboratory computer for two-dimensional analysis using FORTRAN IV software.
3.9 KINEMATICS OF THE SEGMENTAL MODEL

3.9.1 Introduction

Raw coordinate data from the digitised film was organised into a more manageable form on computer backup storage. The raw data was translated using the segmental model definition criteria into a matrix of joint centre-time coordinates, and stored on a diskette in direct access format together with other information about the number of frames digitised, timebase information, subject and experimental details and anthropometric data.

In the case of the two dimensional analysis, the data from film was recorded using a camera with a slightly non-linear time-base. While the time interval between frames is constant to within measurable limits over the duration of a given running stride, strides taken from different parts of a stride sequence may have slightly different time-intervals. For this reason, a 3rd order Newtonian interpolation was performed on the raw data at this stage in order to adjust all data to normalised sample frequency of 50 Hz (See section 3.8 and Appendix F).

The two and three dimensional analyses of treadmill running employed in two experiments documented in this study were accomplished by means of different analytical methods using different computers. The differences between the two implementations of the analytical principles are in part due to the lessons learned from the first set of analyses and in part due to the smaller amounts of data and lesser complexity involved in the two dimensional analyses.

For the two dimensional analyses, computer time and memory capacity were not limiting factors. Data analysis was therefore performed in the time domain using numerical filtering and differentiation methods.

For the three dimensional analysis, the coordinate time-series of joint centres and extremities were approximated by Fourier series of 10 harmonics. As far as possible, all calculations were performed in the frequency domain in order to minimise computer time and memory requirements. Each coordinate was therefore represented by a set of harmonic coefficients, determined by Fourier analysis. By fitting a function to kinematic data, some smoothing automatically occurs, and differentiation to calculate velocities and accelerations can be performed directly on the coefficients of the best fitting equation. Aleshinsky and Zatsiorsky (1978) used a least squares polynomial approximation for this purpose. The use of a Fourier series as the fitting function has the
particular advantage that the effects of calculations on the signal and noise characteristics of the data can be directly monitored in the frequency domain. A number of researchers have previously used Fourier series in the analysis of human motion data (Chao et al, 1983; Cappozzo et al, 1975; Zarrugh and Radcliffe, 1979; Jackson, 1979; Sutherland et al, 1980; Jacobs et al, 1972; James and Brubaker, 1975; Alexander and Jayes, 1980; Cappozzo and Gazzini, 1983). Cappozzo et al (1975) describe a general computing method for the analysis of human movement based on Fourier series. There are a variety of ways in which the fourier analysis of a time series can be performed. If the data points are not equally spaced in time, special procedures apply. (Cappozzo and Gazzini, 1983). Where the data to be approximated is not naturally cyclical, artificial cycles can be generated to permit harmonic analysis (Alexander and Jayes, 1980). In this study, the kinematics of treadmill running is a naturally cyclical phenomenon, and the samples (frames of film) are equally spaced, enabling a simple harmonic analysis to be used.

3:9.2 Three dimensional data analysis in the frequency domain

Let the locus of a generalised spatial coordinate q(t), over one running gait cycle, be represented by the time series \( (q_1, q_2, ..., q_n) \), where \((q_1, q_{j-1})\) are equally spaced in time for all values of \(j\). q(t) can be approximated in the frequency domain by the harmonic series

\[
q(t) = a_0/2 + \sum_{i=1}^{H} a_i \cos(2\pi i t / T) + \sum_{i=0}^{H} b_i \sin(2\pi i t / T) \tag{Eqn 3.19}
\]

where \(H\) is the order of the harmonic series

\(T\) is the total cycle time (fundamental period)
The harmonic coefficients \(a_0 \ldots a_p\) and \(b_0 \ldots b_p\) are calculated from the summations:

\[
\begin{align*}
  a_i &= \sum_{n=1}^{2n} q_n \cos(2\pi it_j / T) & \text{Eqn 3.20a} \\
  b_i &= \sum_{n=1}^{2n} q_n \sin(2\pi it_j / T) & \text{Eqn 3.20b}
\end{align*}
\]

Initially, coefficients for the first 10 harmonic orders were calculated. Thus the locus of a single joint centre or extremity landmark in three dimensional space is described by the three sets of coefficients:

\[
\begin{align*}
  x &= (a_{x0}, a_{x1}, \ldots a_{x10}; b_{x1}, b_{x2}, \ldots, b_{x10}) \\
  y &= (a_{y0}, a_{y1}, \ldots a_{y10}; b_{y1}, b_{y2}, \ldots, b_{y10}) & \text{Eqn 3.21a - 21c} \\
  z &= (a_{z0}, a_{z1}, \ldots a_{z10}; b_{z1}, b_{z2}, \ldots, b_{z10})
\end{align*}
\]

Simple arithmetic and algebraic manipulations of the time series can be performed in the frequency domain. Addition or subtraction of two coordinate time series is achieved by adding or subtracting the harmonic coefficients. Multiplication by a simple factor is achieved by multiplication of each component in the coefficient set by the factor.

Filtering of the raw data was achieved by truncating the harmonic series. Analysis of the signal and noise components of typical kinematic data shows that most information content of the coordinate data time series appears in the frequency domain at frequencies below 6Hz. The duration of fundamental period \(T\) of the running cycle was typically 0.7 seconds. The \(i\)th harmonic of the Fourier series represents a frequency component of \(i/T\) Hz. Thus successive harmonics represent frequency components of approximately 1.43 Hz, 2.85 Hz, 4.29 Hz, 5.71 Hz, 8.57 Hz, etc., respectively. Previous analysis has (Section 3:5) has shown that a combination of a 50 Hz sample rate and 4 harmonics (containing frequency components up to between 5.56 and 6.25 Hz, depending on the fundamental period) gives satisfactory reproduction of data collected at 200Hz. Loci with periods equal to the fundamental period (i.e. on the limbs) were filtered by truncation of harmonics above the fourth. Loci with cycle periods equal to half the fundamental period (i.e on the head and torso where there are two movement cycles between successive ground contacts by
the same foot) were filtered by truncation at the fifth harmonic.

The calculation of the derivatives of the time series can also be performed directly on the coefficients of the harmonic series. The first and second derivatives of equation 3.19 are

\[ q'(t) = \sum_{i=1}^{H} (i w_i) (-a_i \sin(w_i t) + b_i \cos(w_i t)) \]

Eqn 3.22

\[ q''(t) = -\sum_{i=1}^{H} (i w_i)^2 (a_i \cos(w_i t) + b_i \sin(w_i t)) \]

Eqn 3.23

where \( w_i = 2\pi i / T \)

The calculation of the harmonic coefficients of a derivative are thus accomplished by the direct transformation of coefficients:

\[ a_i \leftarrow w_i \cdot b_i \]

\[ b_i \leftarrow -w_i \cdot a_i \]

Eqns 3.24a - 24b.

which can be repeated as many times as required. In this case, only the first two derivatives were calculated, and only first derivatives used in energy analyses.

3.9.3 Two dimensional analysis in the time domain.

As an alternative to the global fit of Fourier series and polynomials to a whole coordinate time series, local approximations to the series may be used. The greatly increased number of calculations involved in making successive local approximations no longer presents a computational problem with the advent of fast and powerful computers.

The use of cubic and quintic spline functions to make local approximations of time-series and calculate derivatives (Dierckx, 1975; McLaughlin et al, 1977; Soudan and Dierckx, 1979; Wood and Jennings, 1979) has recently gained favour. A spline function is a set of piecewise polynomials of order n joined at points called "knots" in such a way as to have n-1 continuous derivatives. Th derivatives of cubic splines have a tendency to approach zero at the endpoints, rendering them unsuitable for approximations to cyclical data. Also, preliminary trials with cubic spline approximations to typical kinematic data from film of treadmill runners revealed that the form of the derivatives was highly dependant on the
subjective positioning of knots. It was therefore decided to differentiate coordinate time series by finite difference methods after having previously filtered the raw data.

While a simple multiple point moving average has a filtering effect on a time series, its effects on the signal and noise components of the raw data are not easily measurable. More sophisticated numerical filtering techniques are available (Radar and Gold, 1967) which enable the ratio between data sample and filter cutoff frequency ($f_s/f_c$ ratio) to be predetermined. Pezzak et al (1977) have shown that once the raw data is filtered, reliable derivatives can be obtained by simple finite difference techniques (Khabaza, 1966). Pezzack et al's (1977) example data has been widely used for comparisons of filtering and derivative determining techniques (Soudan and Dierckz, 1979; Wood and Jennings, 1979), despite the fact that the data is notably smooth to begin with (Lanshammar, 1982).

For the purposes of this study a low pass, second order Butterworth digital filter (Gold and Radar, 1969, pp 48-97) was selected. The filter has the general recursive form

$$Q_i = Aq_i + 2Aq_{i-1} + Aq_{i-2} + Bq_{i-1} + Cq_{i-2} \quad \text{Eqn 3.26}$$

where $q$ and $Q$ are raw and filtered coordinates respectively. The coefficients $A$, $B$ and $C$ are selected in order to give a particular $f_s/f_c$ ratio. Winter (1979b, p36) and Lees (1980) give tables of coefficients for the second order Butterworth filter over a range of $f_s/f_c$ ratios.

For this particular second order filter, there is a 90° phase lag at the cut-off frequency, giving rise to phase distortion in the filtered data throughout the frequency range. To cancel this effect, the filtered data can be filtered again in reverse time giving the effect of a fourth order, zero lag filter with a sharper cut-off than the original second order filter. The cutoff frequency for the 4th order filter is 0.802 that of the second order primary filter.

For the purposes of this study, a primary second order filter with an $f_s/f_c$ ratio of 6 was selected, giving an effective cutoff frequency of 6.68 Hz when applied as a 4th order zero lag filter to data sampled at 50Hz. The specific form of the primary second order filter with $f_s/f_c = 6$ is:

$$Q_i = 0.1551q_i + 0.3101q_{i-1} + 0.1551q_{i-2} + 0.52020q_{i-1} - 0.24040q_{i-2} \quad \text{Eqn 3.26}$$

First and second derivatives of the kinematic time-series were
determined by finite differences methods applied directly to the filtered data. Five point velocity and acceleration routines (Lees, 1980) were used for this purpose. The five point velocity algorithm was

\[ q_t' = q_{t-2} - 8q_{t-1} + 8q_{t+1} - q_{t+2} \]  

Eqn 3.27

The five point Lanczos acceleration algorithm (Lanczos, 1957) was

\[ q_t'' = 2q_{t-2} - q_{t-1} - 2q_t - q_{t+1} + 2q_{t+2} \]  

Eqn 3.28

3.9.4 Data reflection to non-visible segments.

Even with two cameras simultaneously recording movements in the frontal and sagittal planes, much of one side of the body of a treadmill runner is not visible. There is therefore no data available to describe the movements of one half of the link-segmental model unless some assumptions are made.

The cyclical pattern of normal walking and running gaits is generally bilaterally symmetrical. Certainly, in the case of the groups of athletes used in this study no significant intra-individual differences were found between the left and right step times. Bilateral symmetry was therefore assumed and used to calculate the movements of the "invisible" half of the link-segmental model. Joint centre loci for the non-visible half were generated by introducing a phase shift of one half of the stride cycle time into data for the visible half. Where the number of frames in the sequence was even, preventing a direct substitution from visible data in one frame to non-visible data in another frame half a cycle away, intermediate points were calculated by second order Newtonian interpolation (Appendix F).

3.9.5 Segmental centre of mass calculations

The locus of the centre of mass of each segment of the kinematic model was calculated from the data describing the motion of its endpoints. The centre of mass was assumed to be on a straight line joining the proximal and distal joint centres of the segment, at some proportionate distance, c, from the proximal joint centre determined from cadaver data (Section 3.3).

\[ q_{cm} = q_p + (Q_d - Q_p) \times c \]  

Eqn 3.29

where \( Q \) represents any cartesian coordinate

\( Q_{cm} \) is the coordinate of the centre of mass
\( Q_p \) is the coordinate of the proximal joint centre
\( Q_d \) is the coordinate of the distal joint centre.

Segmental centre of mass velocities were calculated using the differentiation procedures previously described (Sections 3:9.2 and 3:9.3)

3:9.6 Segmental angular orientation calculations

If angular changes about the longitudinal axis of a segment are ignored, the angular orientation of a segment in three-dimensional space can be defined by two angular displacements about two orthogonal axes passing through the segment centre of mass. In a two-dimensional analysis, only a single angle is required. Given the coordinates of the endpoints of a segment in the plane orthogonal to a particular axis, the tangent of the angular displacement about that axis can be simply calculated. The only practical problem arises when the angular orientation of the segment passes through the reference axis which may cause the appearance of a sudden jump in the calculated segment angle from 0 to \( 2\pi \) radians or vice versa. This computational problem is overcome by calculating a number of alternative representations of the angle at intervals of \( 2\pi \) and selecting that representation which has the smallest algebraic distance from the angle calculated during the previous time-period.

Angular velocities were calculated using the differentiation procedures previously described (Section 3:9.2 and 3:9.3)

* * * *

The procedures described above yield a set of processed data describing the motion of a link segmental model of an athlete running on a treadmill. This data represents the basic database on which the energy analyses described in Chapters 3, 4 and 9 are based.
Chapter 4

MECHANICAL ENERGY CHANGES AND THE ENERGY COST OF TREADMILL RUNNING
4:0 SUMMARY

This section describes an experimental investigation of the relationships between potential and kinetic energy changes in the body during a running stride, energy exchange between and within segments and energy expenditure in four athletes running on a motorised treadmill at speeds of 3.6 to 5.8 m.s\(^{-1}\). A total of twenty trials were analysed.

Segmental potential and kinetic energies were determined using a three-dimensional link-segmental analysis. Intra-stride changes in the energy of the body were determined with no allowance for energy transfer and with various energy transfer constraints imposed upon the model. Oxygen consumption was determined by expired air analysis and used to determine energy expenditure.

For each transfer condition, net energy expenditure was more highly correlated with the magnitude of changes in the whole body energy curve (\(W_{wb}^+\)) than with running speed per se. When energy expenditure and \(W_{wb}^+\) were corrected for both subject body weight and running speed, there was a low but statistically significant correlation between the two. There was also a significant correlation between the ratio of \(W_{wb}^+\) and the magnitude of within and between segment energy exchange. The more economic running patterns were therefore characterised by total body energy changes of smaller amplitude and by a greater degree of passive energy transfer.

Examination of example data provides evidence for the validity of the calculation of energy exchange within segments and between adjacent segments. The assumption of unrestricted between-segment energy transfer used by Pierrynowski et al (1980) and others is not supported however, and it is suggested that constraints need to be applied to the energy model in order to restrict the possible between segment energy exchanges.

It is also noted that the segmental energy model does not provide a complete analysis of the energy transformations during running. In particular, the possibility of energy storage in the stretched elastic components of muscle and connective tissue have been generally ignored.

4:1 INTRODUCTION

Costill and Fox (1969) suggest that individual differences in the oxygen cost of running at submaximal speeds may be partly due to variations in "mechanical efficiency". Similarly, Ariyoshi et al (1979) consider...
differences in running technique to be the main basis for the lower oxygen consumption of elite marathon runners that has been observed by a number of authors (e.g. Brandsford and Howley, 1977; Costill and Winrow, 1970; Fox and Costill, 1972). If this is the case then a more 'economic' running style, i.e. one that requires a lower oxygen uptake at a given speed, may be of considerable advantage.

There have been few studies of the relationship between segmental mechanical energy changes and the oxygen cost of running. While there have been measurements of energy expenditure and the "Total Pseudomechanical Work" done by elite athletes running at range of speeds (Norman et al., 1976; Gregor and Kirkendall, 1978; Luhtanen & Komi, 1978, 1980), these analyses made no allowance for energy transfer in the mechanical energy calculations. Also, the accuracy of the results presented by the first two of these studies has been questioned. Williams (1980) reported a range of energy measurements, including energy transfer rates, for a group of athletes running overground at 3.57m.s⁻¹.

The purpose of this study was to examine the relationships between the magnitudes of energy changes in the body, passive mechanical energy transfers and energy expenditure in athletes running at a range of sub-maximal speeds.

4.2 METHODS

Four highly trained male middle distance runners were used as subjects for this experiment. All were actively competing at National or International level during the period of the experiment. Two of the athletes later competed at the 1984 Olympic Games of Los Angeles, one reaching the final of the 1500m, the other achieving fourth place in the 5000m.

Prior to the experiments, the subjects made several visits to the laboratory in order to familiarise themselves with running on a motorised treadmill and with the experimental apparatus. Each subject had a minimum of 30 minutes treadmill running experience before any measurements were made. The experimental series consisted of six separate treadmill runs of six minutes duration at speeds of 3.58, 4.02, 4.47, 4.92, 5.36 and 5.81 metres per second. The trials were conducted on separate days in increasing order of treadmill speed.
4:2.1 Oxygen uptake

During the last minute of each trial, expired air was collected in order to determine oxygen consumption. The subject breathed through a light-weight valve, inspiring from atmosphere. Expired air was conducted via lightweight tubing, through a 25L capacity mixing chamber, to a 600L capacity Tissot spirometer. A potentiometer attached to the spirometer enabled the volume of its contents to be recorded continuously on a chart recorder.

The sampling probe of a Centronic RGA007 mass spectrometer was connected to the mixing chamber. Gas samples were withdrawn from the mixing chamber at an average rate of 20 ml.min⁻¹ for analysis by the mass spectrometer which was previously calibrated using known gas mixtures. The concentrations of oxygen and carbon-dioxide in the expired air in the mixing chamber were output continuously and plotted on a chart recorder, together with volume information from the spirometer. Barometric pressure, room temperature, temperature of the expired air and saturated water vapour pressure were also recorded. Ventilation rate (V̇e), oxygen uptake (VO₂), carbon dioxide output (VCO₂) and the gas exchange ratio (R, equal to VO₂/VCO₂) were calculated from the expired air measurements using the Haldane transformation (See section 6:2.2).

4:2.2 Gas exchange ratio and the energy equivalent of oxygen uptake

The effective calorific value of the oxygen consumed is dependant upon the proportions of energy from carbohydrate and fat metabolism. When carbohydrate is oxidised, equal molar quantities of oxygen and carbon dioxide are exchanged. When fat is oxidised, however, more oxygen is consumed than carbon dioxide is produced, giving a theoretical gas exchange ratio (R) of the order of 0.70. The exact gas exchange ratio for fat oxidation depends on the extent to which the fat is saturated. Normally, energy will be derived from both carbohydrate and fat sources. Thus the calorific equivalent of a litre of oxygen consumed can vary between 19.6 and 21.1 kJ.L⁻¹ as R varies between 0.7 and 1. For the purposes of this study, the energy equivalent of oxygen consumption was determined by substituting a subject's oxygen uptake and carbon dioxide output into the equations for the oxidation of fat and carbohydrate. These equations were then solved simultaneously for the total quantity of energy released (Consolazio et al, 1963).

A value of R greater than 1.0 was assumed to indicate a contribution
to the energy supply from anaerobic sources. Excess hydrogen ions in the blood are buffered by the bicarbonate system:

$$H^+ + HCO_3^- \rightleftharpoons H_2CO_3 \rightleftharpoons H_2O + CO_2$$

Thus the metabolic acidosis associated with anaerobic glycogenolysis and lactic acid production in the muscles results in the increased production of CO₂ from the dissociation of carbonic acid. Due to this unknown anaerobic contribution to energy supply, three trials in which the recorded R value was greater than 1 were discarded.

4:2.3 Biomechanical analysis

During the last minute of each trial, the subjects were filmed in the sagittal and frontal planes by two Bolex EBM electric camera with a nominal frame rate of 50 Hz (Section 3:7). The film image coordinates of body landmarks were calibrated against a set of known points distributed in the field of view and converted to three dimensional object space coordinates by means of the transformation method described in Section 3:6. One further trial was discarded from the study on the basis of poor film quality, leaving a total of twenty subject trials to be reported here. One complete stride from each trial was analysed using a 15 segment linked rigid body model of the human body (Section 3:2). The inertial properties of the model were estimated using cadaver data from Barter (1957) and Dempster (1955), on the basis of each individual subject's body weight (Section 3:3.1). Differentiation and filtering of the film data were performed in the frequency domain. Raw position-time data from the digitised film were reduced to sets of harmonic coefficients by Fourier analysis. Initially, the first ten harmonics were computed for each landmark in each dimension. Having estimated the noise content of the data by analysis of residual variance (Section 3:5), the data was filtered by truncation of the higher harmonics, giving an effect of a low-pass filter with a cutoff frequency of 5-6 Hz. The velocities of the segmental centres of mass were determined by analytical differentiation of the truncated Fourier series (Section 3:9.2).

4:2.3 Energy analysis

The energy analysis used in this study is similar to that proposed by Norman et al (1976), Winter (1979) and Pierrynowski et al (1980) (Sections 2:3.5 and 2:3.6). The total energy of a segment at any time was calculated as the sum of its potential, translational kinetic and rotational kinetic energies. However, since kinematic energy changes alone do not provide a complete energy model, the analysis has been restricted to an examination
of observable energy changes and energy transfers during the running stride. Assuming a treadmill running stride to have identical starting and ending states, the net changes in any energy parameter over a stride cycle will be zero. Within a cycle therefore, the magnitudes of the sums of positive and negative changes in any energy parameter are equal and given by

$$W^+ = |W^-| = \frac{1}{2} \sum_{i=1}^{n} |\Delta e_{i,e}| = \frac{1}{2} W$$  

Eqn 4.1

where

$$\Delta e_{i,e} = \text{the change in an energy parameter during the ith time interval}$$

$$n = \text{the number of time intervals}$$

$$W^+ = \text{sum of the positive changes in the energy parameter during a running cycle}$$

$$W^- = \text{sum of the negative changes in the energy parameter during a cycle.}$$

Under Winter's (1979) scheme (Section 2.3.5), $W^+$ and $|W^-|$ would be termed positive and negative work respectively. It should be re-emphasised, however, that these values relate to changes in the level of a given kinematic energy parameter and not necessarily the work done by muscles. For the purposes of this study, $W^+$ was used as a measure of the magnitude of changes in the total body energy curve, this curve having been derived with the following energy transfer constraints imposed on the segmental model:

1. No transfer allowed between any component energies ($W_a^+$)
2. Transfer permitted between the component energies of a segment but not between segments (i.e. within segment transfer only) ($W_w^+$)
3. Transfer permitted within segments and between segments of the same limb ($W_l^+$)
4. Transfer permitted within and between all segments ($W_{w+l}^+$)

Conditions 1, 2 and 4 are essentially those described by Pierrynowski et al (1980) for the calculation of energy transfer between and within segments of the body during walking (Section 2.3.6, Equations 2.6, 2.8 and 2.9). Condition 2 was calculated on a similar basis. The body was divided into five sections (four limbs and the torso), each with three segments. Energy transfer was permitted between the segments of each section but not
between sections. Thus

\[ W_i = \sum_{i=1}^{n} \sum_{j=1}^{5} | \Delta Q_{i,j} | \]  

Eqn 4.2

where \( \Delta Q_{i,j} \) is the change in the sum of all the energy components in the \( j \)th section during the \( i \)th time period. The magnitudes of energy transfers of different types were calculated using the same principle of subtraction as the method described by Pierrynowski et al. (1980)

Within and between segment transfer: \( T_{wb} = W_{n^+} - W_{wb^+} \)

Between segment transfer: \( T_b = W_{n^+} - W_{wb^+} \)

Within segment transfer: \( T_w = T_{wb} - T_b \)

In addition:

Within limb transfer: \( T_{b1} = W_{n^+} - W_{b1^+} \)  

Eqn 4.4

Between limb transfer: \( T_{w1} = T_{wb} - T_{b1} \)  

Eqn 4.3

The use of magnitudes (\( W^* \)) to describe energy changes has the advantage that calculations of energy transfer will be exactly half those computed by Pierrynowski et al's methods. This is in fact a better reflection of the magnitude of energy transfers, since Pierrynowski's calculations include a unit of passive energy exchange twice; firstly as a decrement in one energy term and secondly as an increment in another.

4.2.5 Comparison of two and three dimensional energy analyses

The human running gait is dominated by movements of the limb in the sagittal plane. Movements of limbs in the frontal plane are of small amplitude and usually of low velocity. It is possible therefore that kinetic energy changes due to the translation of body segments along the transverse axis and due to rotation about segmental axes normal to the frontal plane are small. If this is the case then it may be possible to obtain adequate information about the energy transformations during running from film of the sagittal plane only. Two dimensional and three dimensional energy analyses were compared directly using data from four trials, one from each subject, all running at a speed of 4.47 m.s\(^{-1}\). The two dimensional analysis was performed using the three-dimensional system but with all but harmonic 0 (i.e. the mean value) of the z-component
(transverse component) set to zero after calibration and image reconstruction. The energy components in the third dimension could have been calculated directly, but this would have given no indication of the contribution of these components to energy transfer.

4:2.6 Comparison of different methods of calculating "mechanical work" and efficiency

It has been noted (Section 2:4) that there has been a wide variety of methods employed to calculate the "mechanical work" and "efficiency" of human walking and running gaits and an equal variety of values for efficiency have been reported. While it is suspected that these differences are due to methodological differences, and differences in the ways in which the body is modelled as a system, there remains the possibility that some of the variation is due to inter-subject differences. A sample of the methods reported in the literature was applied to the same data from one subject running at 4.47 m.s⁻¹ in order to determine the variation in reported efficiency values due to methodological differences alone.

"Efficiency" can be simply defined as the ratio of mechanical work output to energy expenditure.

Energy expenditure can be expressed in a number of ways, depending on the base-line used for "resting" energy expenditure in net efficiency calculations. For the purposes of this comparison, energy expenditure was expressed in two ways; firstly as the total recorded energy expenditure rate (Gross energy expenditure) and secondly as the total energy expenditure rate less that recorded with the subject standing at rest (net energy expenditure).

The "Mechanical work" done by the body has also be calculated in a variety of ways depending on how changes in the total body energy are calculated. The following six definitions of "work rate" were selected for comparison.

AVPOT: the rate of change in the average potential energy of the body assumed to be moving at constant speed. (Margaria, 1938, 1963)

POT: "vertical lift per step", the magnitude of intra-cycle total body potential energy changes (Cotes and Meade, 1960).

CG: changes in the total potential and kinetic energy of a point mass model (Cavagna et al, 1963).

CG+R: changes in the total potential and kinetic energy of the centre of mass plus the kinetic energy of the limbs relative to the centre of mass.
TPW : Sum of the absolute changes in the instantaneous partitioned
energies of each segment (Norman et al, 1976)

WwB : Sum of the positive changes in the total body energy curve
calculated as the sum of the segmental potential and kinetic
energies. (Winter, 1979; positive work component)

Ignoring manipulations of the relative efficiencies of positive and
negative work (Winter, 1979; Pierrynowski et al, 1980; Williams and
Cavanagh, 1983) these example methods permit twelve different calculations
of efficiency, six of "gross efficiency" and six of "net efficiency".

4:3 RESULTS

Figure 4.1 shows examples, from one subject, of total body (kinetic +
potential) energy variations during a running stride. In addition to the
overall increase in the average energy level at higher running speeds, the
amplitude of intra-stride changes in the energy curve also increases. The
peaks in the energy curves occur shortly after toe off, when the kinetic
energy of the recovery leg is greatest.

The results of the kinematic energy analysis are summarised in figures
4.2 and 4.3. Figure 4.2 shows the mean values of W* at each speed for each
of the transfer constraints applied to the model. W* was approximately
linearly related to running speed under each transfer condition. The rates
of increase of W0+, Ww+ and Ww+ with running speed were similar to one
another but greater than that of Ww+. Mean rates of W+ and calculated
energy transfer rates are shown in Table 4.1. Figure 4.3 shows the mean
magnitudes of within-segment, between-segment, within limb, between-limb
and total energy transfer at each running speed. Within-segment and
within-limb transfer rates were similar across the whole range of running
speeds studied. Increases in the total within and between segment transfer
with running speed were almost entirely due to increases in the
between-segment energy transfer. The proportion of the total energy
exchange due to within segment transfer decreased slightly and that
attributable to between segment transfer increased slightly with running
speed. The total energy exchange, T wb, remained a fairly constant
proportion of Ww+ over the range of running speeds, having a mean value of
72.2% (+1.7 s.d.)

If no constraints are applied to energy transfer, then most (more than
90%, on average) of the apparent between segment energy transfer is due to
Figure 4.1: Intra stride changes in total (potential+kinetic) energy at a range of running speeds (Subj. 4).
Table 4.1: Mean rates of kinematic energy change and energy transfer

<table>
<thead>
<tr>
<th>Treadmill speed m/s</th>
<th>3.58</th>
<th>4.02</th>
<th>4.47</th>
<th>4.92</th>
<th>5.36</th>
<th>5.81</th>
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<td>n</td>
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<td>4</td>
<td>4</td>
<td>4</td>
<td>3</td>
<td>2</td>
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<td>wk+</td>
<td>856</td>
<td>1635</td>
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<td>1357</td>
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<td>1779</td>
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<tr>
<td>Ww+</td>
<td>715</td>
<td>845</td>
<td>1045</td>
<td>1174</td>
<td>1331</td>
<td>1608</td>
</tr>
<tr>
<td>Wwl+</td>
<td>662</td>
<td>745</td>
<td>981</td>
<td>1119</td>
<td>1277</td>
<td>1536</td>
</tr>
<tr>
<td>Wwb+</td>
<td>240</td>
<td>280</td>
<td>334</td>
<td>376</td>
<td>430</td>
<td>505</td>
</tr>
<tr>
<td>Twb</td>
<td>615</td>
<td>756</td>
<td>864</td>
<td>981</td>
<td>1074</td>
<td>1295</td>
</tr>
<tr>
<td>Tw</td>
<td>140</td>
<td>191</td>
<td>154</td>
<td>183</td>
<td>163</td>
<td>191</td>
</tr>
<tr>
<td>Tb</td>
<td>475</td>
<td>565</td>
<td>710</td>
<td>798</td>
<td>910</td>
<td>1103</td>
</tr>
<tr>
<td>Tbl</td>
<td>422</td>
<td>472</td>
<td>646</td>
<td>743</td>
<td>857</td>
<td>1032</td>
</tr>
<tr>
<td>Twl</td>
<td>193</td>
<td>281</td>
<td>218</td>
<td>258</td>
<td>217</td>
<td>262</td>
</tr>
</tbody>
</table>
Figure 4.2: Magnitude of intra-stride energy changes under different energy transfer conditions (mean ± sd).
Figure 4.3: Magnitude of intra stride energy transfer rates over a range of running speeds (mean ± sd).
energy exchange between segments of different limbs.

4:3.1 Mechanical energy changes and energy expenditure

Figure 4.4 shows the relationship between net energy expenditure and running speed for the twenty trials examined. Energy expenditure and running speed were well correlated \((r = 0.83, p < 0.001)\) and each subject shows the characteristic linear rise in energy expenditure as speed increases. There are, however, significant inter-subject differences in the energy expenditure - speed relationships.

For each transfer condition, energy expenditure was more highly correlated with \(W^+\) than with running speed \((W_{r^+}, r=0.87; W_{w^+}, r=0.88; W_{wkb^+}, r=0.86; W_{wb^+}, r=0.92; p < 0.001\) in each case). This is illustrated in figure 4.5 where energy expenditure is plotted against \(W_{wb^+}\) for each trial. Compared with figure 4.4, inter-subject variation is reduced.

In figures 4.4 and 4.5, energy expenditure and \(W^+\) are not corrected for body-weight or running speed. These factors both contribute to the shared variance and are largely responsible for the high correlations shown. Figure 4.6 shows the relationship between energy expenditure and \(W_{wb^+}\) when both are corrected for subject body weight and running speed. The correlation between the two is low \((r = 0.45)\) but achieves statistical significance \((p < 0.05)\).

The relationship between energy expenditure and the magnitude of changes in the body total energy curve was significantly related to the overall within and between-segment transfer rate. The ratio of \(W_{wb^+}\) to energy expenditure and total transfer rate \((T_{wb}, \text{corrected for both bodyweight and running speed})\) were significantly correlated \((r = 0.69, p < 0.01), \text{Figure 4.7}).

4:3.2 Examples of energy transfer

1: Within segment energy transfer:

Figure 4.8a shows the changes in the total and component energies of the left thigh during a single stride of one subject running on a treadmill at 4.02 m.s\(^{-1}\). The dominant peak in the total energy curve is due to the large increase in the kinetic energy of the thigh during the recovery phase. The level potential energy is significant throughout the stride but varies little when compared with the translational kinetic energy component. The potential energy changes are consistent with the "pendular" motion of the thigh during the support and recovery phases. The rotational component
of kinetic energy is relatively insignificant throughout the stride, never contributing more than 2% of the total segment energy. There is some evidence for within segment transfer of energy between the energy components. Figure 4.8b shows percentage of the summed absolute changes in the thigh energy components (equivalent to \( W_n^+ \) for the segment) attributable to passive energy transfer between potential and kinetic energy. During the early part of the recovery phase, as the thigh swings forward, up to 70% of the rise in segmental kinetic energy is attributable to the corresponding fall in potential energy. Similarly, as the recovery leg swings forwards and upwards, up to 40% of the gain in potential energy can be attributed to losses in kinetic energy.

2: Between segment energy transfer

There is also evidence for the transfer of energy between segments of the same limb. As an example, energy changes in the segments of the left leg will be considered (Figure 4.9). The total energy of the leg is highest during the recovery phase, during which all three segments are being moved rapidly in the forward direction and lifted with respect to the potential energy reference point. The energy of the calf and foot components continues to rise after the energy of the thigh has reached a peak and begun to decline as it decelerates before floor contact. This suggests the possibility of energy transfer between the thigh and the calf and foot segments. Indeed, at this stage of recovery, up to 80% of the total changes in leg segment energies can be attributed to passive energy transfer between segments, the energy lost by the thigh being gained by the calf and foot segments (Figure 4.9b).

3: Energy transfer between limbs.

If it is assumed that energy transfer between limbs is unrestricted then some interesting results arise, particularly in relation to energy exchange between reciprocating limbs. For example, consider the possibility of energy transfer between the left and right legs (Figure 4.10). The reciprocating action of the legs is such that fluctuations in the sum of their energies are small, even though each leg shows large changes during the stride cycle. Thus the energy exchange between the legs, if assumed unrestricted, is very large in magnitude (Figure 4.9b). In fact, under this assumption, almost all the changes in the energy of the two limbs at any particular time can be attributed to passive energy exchange between limbs.
4:3.3 Estimates of mechanical work and efficiency

Table 4.2 records the "mechanical work" and "efficiency" values for a single subject running on a treadmill at 4.47m.s⁻¹; calculated using six, different, methods published in the literature. The same work estimates are illustrated in Figure 4.11. The "average potential energy method (Margaria et al., 1938, 1963) yields a work output of zero since the subject is running on a level treadmill. The methods of Cotes and Meade (1960) and Cavagna et al 1963 yeild positive values for mechanical work, the first of these including only the potential energy of a point mass model and the second both the potential and kinetic energy components of the point mass model. The addition of the relative segmental motions to the kinetic energy term (CG+R, Cavagna et al, 1964) increase the work estimate further but underestimate the work value obtained if absolute segmental energies are included in the total body energy curve (WWB). If the magnitudes of the energy increments of all the component energies of each segment are summed to give a work measurement (TPW, Norman et al, 1976) then the apparent efficiency of the conversion of metabolic energy into mechanical work is well in excess of 100%.

4:3.4 Comparison of two and three dimensional energy analyses

The observed differences between three-dimensional and two-dimensional energy analyses were small in the four cases examined. Mean values of energy change and energy transfer magnitudes are shown in Table 4.3. In all cases, the 2D estimate of mechanical energy changes was lower than the 3D estimate by 3-5%. There were no significant differences in the relative magnitudes of total within and between segment energy transfers (Twb/Wn%) determined by the 2D and 3D analyses.
Figure 4.4: Energy expenditure and treadmill running speed in four trained middle distance runners.
Figure 4.5: Intra-stride energy changes, assuming total within and between segment energy transfer, and energy expenditure in four trained middle distance runners.
Figure 4.6: Intra-stride energy changes, assuming total within and between segment energy transfer, and energy expenditure. After correction for subject bodyweight and running speed.
Figure 4.7: Mechanical energy changes per unit energy expenditure and total energy transfer rate.
Figure 4.8: (a) Intra stride changes in the total and component energies of the left thigh (Subj. 3, 4.02 m/sec) (b) Within segment energy transfers (The positive direction is transfer from potential to kinetic energy).
Figure 4.9: (a) Intra-stride changes in the total and segmental energies of the left leg
(b) Magnitude of between segment energy exchange.

(Subj. 3, 4.02 m/sec)
Figure 4.10: (a) Intra-stride changes in the total energies of the left and right legs (Subject 3, 4.02 m/sec).
(b) Between limb energy exchanges, assuming unrestricted energy transfer.
(c) Relative between limb energy exchange.
Table 4.2: Mechanical work rates and efficiencies calculated by different methods.

Subject 3 running at 4.47 m/s⁻¹
See text for details of methods

<table>
<thead>
<tr>
<th>METHOD</th>
<th>Work rate J/s</th>
<th>Eff. gross %</th>
<th>Eff. net %</th>
</tr>
</thead>
<tbody>
<tr>
<td>AVPOT</td>
<td>0</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>POT</td>
<td>85</td>
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<td>10.9</td>
</tr>
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<td>189</td>
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<td>24.3</td>
</tr>
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<td>283</td>
<td>32.9</td>
<td>36.4</td>
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<td>1283</td>
<td>149.2</td>
<td>165.1</td>
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<tr>
<td>WWB</td>
<td>325</td>
<td>37.8</td>
<td>41.8</td>
</tr>
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</table>
Figure 4.11: "Mechanical work" estimated using six different methods described in the text.

Subject 3 running at 4.47 m/sec.
Net and gross energy expenditure are also indicated.
Table 4.3: Comparison of the results of 2D and 3D energy analyses

Treadmill speed: 4.47 m.s\(^{-1}\)

<table>
<thead>
<tr>
<th>Subject</th>
<th>1</th>
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<td>338</td>
<td>310</td>
<td>359</td>
<td>321</td>
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<tr>
<td>Δ</td>
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<td>-3.4%</td>
<td>-4.6%</td>
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<td>Twb 3D</td>
<td>710</td>
<td>1008</td>
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<tr>
<td>Twb/Wn 3D</td>
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<td>+0.5</td>
<td>-0.3</td>
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</table>
4:4 DISCUSSION

Segmental energy analyses of running are a relatively recent innovation and available comparisons of the data presented here with previously published work are therefore limited. Data from other studies is presented together with the mean values from this study in Figure 4.12. Norman et al (1976) present results for calculations of $W_n$ on 3 subjects running at speeds in the range 2.7 to 5.4 m.s$^{-1}$, but the accuracy of these results has been questioned. Gregor and Kirkendall (1978) present results of similar magnitude. Values of $W_n^*$ extracted from the results of Luhtanen and Komi (1978, 1980) are shown as is the mean value of $W_n^*$ reported by Williams (1980) for a group of athletes running overground at an average speed of 3.57 m.s$^{-1}$.

4:4.1 Two dimensional vs three dimensional energy analyses.

The observed differences between two and three dimensional analyses of the same data sets were significant but relatively small in magnitude (3-5%). Williams (1984) reports small differences (~1%) between displacements and velocities calculated by two dimensional and three dimensional analyses which is consistent with the differences of less than 5% in energy parameters observed here. Given the greater complexity of the three dimensional analysis, there are practical reasons why the two-dimensional analysis is to be preferred. The relationships between the measured energy parameters were not affected by the analytical method. While it must be accepted that some information will be lost if a two dimensional analysis is used, these results suggest that the two dimensional analysis is adequate for energy analyses of running and walking where movements in the sagittal plane predominate.

4:4.2 Mechanical energy changes and energy expenditure

The results of this experiment suggest that differences in the magnitude of energy changes during the running stride account for some of the inter-individual differences in energy expenditure at a particular running speed. Thus it is possible that differences in "running style" may contribute to differences in "running economy".

Energy expenditure was more highly correlated with the magnitude of energy fluctuations ($W^*$) during the running stride than with running speed (Figures 4.4 and 4.5). This would suggest that the energy changes are a more appropriate measure of the work requirement of the running task than
Figure 4.12: Values of $W_n^+$ from the results of this study compared with data from other sources.

- Norman et al (1976)
- Gregor and Kirkendall (1978)
- Williams and Cavanagh (1983)
- This study
running speed alone. However, these results do not allow for the effects of individual differences in bodyweight.

Where the energy expenditure requirements of tasks involving the movement of the whole body are concerned, it is normal practice to correct oxygen consumption and energy expenditure results for each subject's body weight. Wyndham et al (1971), for example, found oxygen consumption in subjects walking both on the road and on the treadmill to be highly correlated \( (r = 0.76 \text{ to } 0.96) \) with bodyweight. In relatively static exercise, on a cycle ergometer for example, the bodyweight correction does not normally produce significant reductions in the between-subjects variance of energy expenditure or oxygen uptake. The magnitudes of kinetic and potential energy changes are both directly dependent on the mass of the body and it's segments. The reduction in inter-individual differences in energy expenditure when expressed as a function of \( W_{wb}^+ \) may simply be to account for individual mass differences therefore. Also, the high correlation between \( W_{wb}^+ \) and energy expenditure is in large part a result of the shared variance brought about by the inclusion of trials at a range of running speeds, to which both \( W_{wb}^+ \) are both linearly related.

The correction for body weight normally applied by physiologists to oxygen uptake and energy expenditure measurements implies an attempt to adjust for differences in the amount of work required to perform a particular task that are brought about simply by the shift in the mass of the system being moved. In this case energy expenditure (corrected for differences in running speed) were well correlated with subject body weight \( (r=0.77, \ p < 0.01) \). Also, the results of this study show that the amplitude of mechanical energy changes rises linearly with increments in running speed (Figure 4.2). This observation therefore supports the validity of the use of body-weight correction in physiological measurements of oxygen consumption as far as treadmill running is concerned.

Even when differences in body weight were taken into account, there was still some residual variation in energy expenditure that was not accounted for, however. When \( W_{wb}^+ \) and energy expenditure are adjusted for subject body weight and running speed, there is a low but significant correlation between the two (Figure 4.6, \( r = 0.45, \ p < 0.05 \)) suggesting that biomechanical differences between individual running patterns, expressed in terms of the amplitude of the total body energy curve, can account for at least part of this variance. These results are consistent with those of Williams and Cavanagh (1983) who found small differences
between the magnitudes of mechanical changes and energy transfers of runners allocated to three groups on the basis of "physiological efficiency". Although these authors found no statistically significant differences between the groups, there was a trend for the more "efficient" runners to have a smaller magnitude of changes in the total body energy curve and a greater total energy transfer rate.

Since it has been suggested that a lower oxygen cost of running is associated with smaller fluctuations in total body energy, it is useful to consider how passive energy transfers may contribute to the reduction in amplitude of the energy curve. There was a significant relationship between total (within + between) segment energy transfer and the ratio of $W_{wb}^+$ to energy expenditure (Figure 4.7, $r = 0.69$, $p < 0.01$). Movements produced by muscle action incur a metabolic cost. When movements are brought about by passive energy exchange this cost is eliminated. Thus the magnitude of passive energy exchange could be a factor in reducing the requirement for "active" energy input.

The ratio $W_{wb}^+/W_{ex}$ is effectively the same calculation as the "positive work efficiency" described by Winter (1979). In these terms, Figure 7 suggests that greater magnitudes of passive energy transfer result in greater "efficiency". It cannot be claimed that $W_{wb}^+$ is a direct measure of the mechanical work output of muscle however, and the energy system of which this ratio is the efficiency has not actually been defined. As has been demonstrated, the calculation of "efficiency" values for human movement is largely a question of choice of method (Figure 4.11, Table 4.2).

4:4.3 Within and between segment energy transfer

Since the values of $W_n$, $W_w$ and $W_{wb}$ calculated by Pierrynowski et al (1980) include both positive and negative changes in their respective energy curves, the calculated values for the total energy transfer include the transfer of a given energy unit twice, firstly as a decrement in one energy term and secondly as an increment in another energy term. This is not the case in this study where only the magnitudes of energy changes are considered, and the calculated values of energy transfer are half of those calculated by Pierrynowski's original method.

The potential and kinetic energies calculated for one trial of subject 3 have been examined in more detail for evidence of within and between segment energy transfer.
As an example, the component energies of the thigh demonstrate the opportunity for within segment energy transfer between potential and kinetic components (Figure 4.8). During the recovery phase, part of the increase in the kinetic energy of the thigh can be attributed to the loss of potential energy. Such energy exchange occurs in pendular motions under the influence of gravity. Potential and kinetic energy are mathematical constructs rather than biological entities, and there is no possibility that muscular work is specifically transformed into one energy component or the other. The pendular motions of the limbs are no different from the pendular motions of inanimate objects under the influence of gravity in this respect. The calculation of within-segment energy transfer would appear to be a realistic one, therefore.

The exchange of energy between segments has also been demonstrated, using the segments of the leg as an example (Figure 4.8). At one stage of the recovery phase, up to 80% of the increase in the energy of the lower limb segments (calf and foot) can be attributed to a loss in the energy of the thigh. Again, the energy exchanges observed mimic the behaviour of inanimate systems under the influence of gravity. Phillips et al (1983) have demonstrated that proximal thigh motion has a significant effect on distal shank motion, due to the effects of intersegmental reactions alone. In the example used here, the energy exchange would appear to be initiated by muscle action, the passive energy transfer to the lower limb being the result of the deceleration of the thigh. It could be suggested therefore that "passive energy transfer" actually incurs a physiological energy cost. However, the muscle action can be regarded as bringing about changes in the total energy of the system, with passive energy exchange being a secondary consequence of the interaction between muscle moments, limb configuration and gravity. Also, muscles can act as passive, possibly elastic connectors between segments. Energy exchange is always likely to occur between objects connected in this way regardless of the nature of the primary energy source.

In this study, it was found that much of the apparent (calculated) energy exchange between segments was due to the reciprocal motions of different limbs. If energy exchange between segments is assumed to be unrestricted (i.e. energy exchange between any segment and any other is permitted), then the motion of the legs appears to be largely the result of passive energy transfer (Figure 4.9). At any stage of the running cycle, increases in the total energy of one leg are associated with decrements of similar magnitude in the energy of the other. Without constraints on the
energy transfer calculation, (Pierrynowski et al, 1980) these alternating motions are attributed to between segment energy transfer. Unlike the previous two examples, this transfer cannot be mimiced using inanimate models. Therefore, if this energy exchange actually occurs, muscle must play an active role in coordinating the motion, even if the source of energy is passive exchange between segments. It is probable that such between segment exchanges do not actually occur in the magnitudes suggested by the calculation, however. Firstly, it has been assumed that no loss of energy occurs. In a mechanically "perfect" rigid body model, this may be reasonable, but in the real human body, movements and energy transformations are likely to be damped by soft tissue and body fluids. Secondly, any possible transmission of energy from one coordinated motion to a coordinated motion at another extremity without motion or muscle activity in intermediate segments is highly improbable since in such circumstances, the coordinated motions must be assumed to be the result of random events. By calculation, the reciprocal motion of two extremities, say the hands, may appear to be the result of energy transfer between the extremities when in fact EMG records may show that both are due to muscular action. Even with muscle activity contributing to the coordination of the movements, some of the possible energy transfer between limbs is likely to be dissipated by the damping effect of the viscera, other soft tissues and body fluids.

There is, therefore, evidence for the passive transfer of energy between the component energies of a segment and between adjacent segments, but the assumption of unrestricted between segment energy transfer implicit in the calculation of \( W_{wb}^+ \) requires further examination. The calculation of \( T_{wb} \) represents the maximum possible magnitude of within and between segment energy transfer due to the passive exchange of potential and kinetic energy. Clearly some constraints need to be applied to the calculation of between segment energy transfer if a realistic assessment is to be made, but what these constraints should be is, at the moment, a matter of speculation.

4:4.4 Limitations of the kinematic energy model.

If \( T_{wb} \) represents the maximum possible passive energy transfer within and between segments then \( W_{wb}^+ \) must be the minimum possible magnitude of mechanical energy changes that must be attributed to muscular work. Thus the minimum estimate of the "efficiency" of muscle in producing work from
chemical energy would be, on average, 0.35, if the positive increments in the total body (potential + kinetic) energy curve are attributed to muscular effort.

Given that the overall average efficiency of isolated muscle is of the order of 0.25 it would appear that the "pseudowork" methods described in the literature do not give an exact reflection of muscular work or "efficiency".

Since potential and kinetic energy are not the only forms that energy may take in the body, the link-segmental energy models currently in use are incomplete. Until recently, it was assumed that only potential and kinetic energy changes were significant in locomotion. Most importantly, the possibility of strain energy storage in the elastic components of muscle and tendon have been generally ignored. Elastic energy storage has been shown to be significant in simple knee-bending activities (Thys et al., 1982) and in the locomotion of kangaroos (Dawson and Taylor, 1973; Alexander and Vernon, 1973). A number of authors have suggested that such energy storage may be significant in running (Cavagna et al., 1971; Cavagna and Kaneko, 1977; Luhtanen and Komi, 1980), but elastic energy storage has not been specifically included in an energy model of walking or running.

The simple energy models in current usage are also ill-defined in the sense that the boundaries of the energy system, energy flows and the constraints on passive energy transfer are not described.

While it may not establish a value for the "efficiency" of muscle, an energy model with discrete boundaries, in which elastic energy storage is included as a component, may permit a more consistent analysis of the energy transformations during human movement.

46 CONCLUSIONS

The results of this study lead to the following conclusions:

1: Some of the between-subjects variance in the energy cost of running can be accounted for by the magnitude of variations in the total body (potential + kinetic) energy.

2: The ratio of energy expenditure to intra-stride energy changes is significantly related to the magnitude of the total possible exchange of energy between and within body segments.
3: While there are small differences between the results of two and three-dimensional energy analyses of running, a two dimensional analysis is adequate for most purposes.

4: There is evidence to support the validity of the calculation of energy transfer within segments and between segments of the same limb, but there is a need to apply constraints on the possible between segment transfers if a sensible calculation of between segment energy exchange is to be made.

5: The "pseudowork" models of Norman et al (1976), Winter (1979) and Pierrynowski et al (1980) over-estimate the mechanical work done by muscle. The models are incomplete in that a number of energy components, in particular the storage of strain energy in stretched muscle and connective tissue are not included in the analysis.
Chapter 5

A GENERAL ENERGY ANALYSIS PROCEDURE
5:0 **SUMMARY**

A procedure is developed for the analysis of an energy system defined in term of a set of subsystems between which energy exchange can occur. The procedure is based on a generalisation of previously published methods but allows energy components other than segmental potential and kinetic energy to be included in the energy model. Also, the procedure enables a variety of constraints to be applied to energy exchange.

5:1 **INTRODUCTION**

The energy analyses of human gait proposed by Norman et al (1976), Winter (1979) and Pierrynowski et al (1982) have been found deficient in a number of respects. The most important of these limitations are:

1: The analysis is restricted to segmental potential and kinetic energy changes only. No provision is made for the inclusion of other energy components. In particular, the possible contribution of strain energy storage in muscle and connective tissue has not been considered.

2: The calculation of between segment energy transfer assumes that energy may be freely transferred from any part of the body to any other, thus permitting energy exchanges in the model that may not be physically possible.

3: The energy model on which calculations of "efficiency" are based is not completely defined. The boundaries of the energy system have not been explicitly declared, heat production has not been considered and no consistent procedure for the treatment of "negative work" has been devised.

The basic requirements of any development of the energy analysis are therefore that it should allow additional energy components to be included, and should allow constraints to be applied to the possible energy exchanges within and between segments. Also, any developments should allow the energy components and the flows between them to be clearly defined. Without such definitions, attempts to interpret the results of the energy analysis in terms of "mechanical work" and "efficiency" measurements can only lead to further confusion.

In this section, the energy analyses previously described are developed into a generalised form in order to remove some of the
limitations described above. It is not intended to define a complete energy model of the human body. This would be a task of enormous complexity, and would undoubtedly result in a model that could not be evaluated experimentally. Instead, the purpose of this section is to develop a framework within which the energy analysis can be developed, and to initiate that development by extending the energy models currently in use.

5.2 GENERALISATION OF THE KINEMATIC ENERGY MODEL

On the basis of the first law of thermodynamics, it is possible to determine an energy balance equation for any system with defined boundaries. The simplest form of non-flow energy equation is:

\[ e_{in} - e_{out} = \Delta e \]  
Eqn 5.1

where \( e_{in} \) and \( e_{out} \) represent net exchanges of heat and work across the system boundary and \( \Delta e \) is the net change in the internal energy of the system (Figure 5.1). In the case of the human body in motion, \( \Delta e \) must be taken to represent the sum of the transformations between a variety of energy forms, including the following:

1. Mechanical energies:
   - Potential and kinetic energy of body segments
   - Strain energy due to deformation of elastic structures.

2. Chemical energies:
   - Free energy conserved as ATP and phosphocreatine
   - Higher energy stores (glycogen, fat, etc.)

3. Heat
   - Heat generation by muscle activity, viscosity, friction at joints, etc.
   - Heat losses in expired air and due to perspiration.

4. Mass transfers
   - Inspiration and expiration of air; gas exchange.
   - Intake of food and water
   - Loss of perspiration and waste materials.

As far as this study is concerned, it is the transformations between
Figure 5.1: A simple energy system.
the internal mechanical and chemical energy components that are of particular interest, and the energy model must therefore be defined with greater resolution by division of the total energy term into its component parts.

In the simple model, the magnitude of the net energy exchanges across the system boundary is |\( \Delta E \)|. Given a more complex system consisting of \( n \) components each described by equations of the form of Equation 5.1 and between which energy exchange can occur (Figure 5.2) then the net energy exchange between the system and the environment, across its boundary is

\[
\Delta E = \sum_{i=1}^{n} \Delta e_i = \Delta \sum_{i=1}^{n} e_i
\]

Eqn 5.2

If all the energy exchanges between components balance, there is no net input or output of energy. Otherwise \( \Delta E \) represents the net energy exchange across the system boundary, irrespective of exchange between the component energies. The magnitude of total energy exchanges (TEE) in the system, including those across the outer boundary can be calculated by the sum:

\[
TEE = \sum_{k=1}^{n} |\Delta e_k|
\]

Eqn 5.3

This sum includes the exchange of any given unit of energy twice; firstly as an increment in one term and secondly as a decrement in another energy component. Thus the true magnitude of the total energy exchanged is

\[
TEE_{corrected} = \frac{1}{2} \sum_{k=1}^{n} |\Delta e_k|
\]

Eqn 5.4

Over a period of \( t \) time intervals, the energy exchanges can be summed, giving:

\[
TEE = \frac{1}{2} \sum_{i=1}^{t} \sum_{k=1}^{n} |\Delta e_k|
\]

Eqn 5.5

This simple analysis can be extended to any system with known, bounded energy components. Consider a generalised energy system consisting of \( U \) elements (e.g. limbs or body segments) with the \( j \)th element having \( n_j \) possible energy components between which exchange can occur. The total energy of element \( j \) at any instant \( t \) is given by the sum of its components:
Figure 5.2: Simple energy systems combined to form a more complex system.
where \( e_{j,k,t} \) is the energy of the \( k \)th component of the \( j \)th element at time \( t \). The total energy of a group of elements is given by their sum:

\[
E_{j,t} = \sum_{k=1}^{n_j} e_{j,k,t}
\]

Eqn 5.6

Since each element can be treated as a component of a higher element, it is possible to devise a hierarchical system (Figure 5.3) with each group of elements having a boundary across which energy exchanges can occur and be observed.

During any time period, \( \Delta t \), the change in the total energy, \( E_{j,t} \) of the \( j \)th element is equal to the energy crossing its boundary irrespective of energy exchanges between its components. Thus the magnitude of energy exchanges between components can be calculated by comparing the sum of the absolute changes in the components with any change in the element total, thus:

\[
|T|_{\Delta t} = \frac{1}{2} \left( \sum_{k=1}^{n_j} |\Delta e_{j,k}| - |\Delta E_{j,t}| \right)
\]

Eqn 5.8

where \( |T|_{\Delta t} \) is the magnitude of the energy transfers within the system during the time period \( \Delta t \).

The magnitude of energy exchange within a system of \( U \) elements during \( t \) time periods is therefore given by

\[
|T| = \frac{1}{2} \left\{ \sum_{i=1}^{t} \sum_{j=1}^{t} \left( \sum_{k=1}^{n_j} |\Delta e_{i,j,k}| - |\Delta \left( \sum_{k=1}^{n_j} e_{i,j,k} \right) | \right) \right\}
\]

Eqn 5.9

i.e. it is assumed that the magnitude of energy exchanges between components within the system boundary can be calculated by comparing
Figure 5.3: A generalised hierarchical energy system.

Circles represent boundaries of elements containing components between which energy exchange may occur.
changes in the individual component energies with changes in the system
total, the change in this total being equal to the sum of the absolute
changes in the energy components, less any exchange between them. In this
way, energy exchange within the system is calculated as the sum of all the
observed energy exchanges less any exchanges across the system boundary.

In chapter 4, a procedure was defined for calculating the magnitude of
energy changes in a system undergoing cyclical change, in particular
running or walking man. The net changes in any energy parameter over a
cycle is zero. Over a complete cycle, therefore, the magnitudes of positive
and negative changes in an energy parameter are equal and given by

\[ W^+ = |W^-| = \sum_{i=1}^{n} |\Delta Q_i| = \frac{1}{n} W \]  

Eqn 5.10

where \( \Delta Q_i \) is the value during time interval \( i \) of the energy quantity \( Q \)
which varies cyclically with a period of \( n \) time intervals. \( W^+ \) is thus the
magnitude of the positive changes in the energy curve during the cycle, and
\( W^- \) the magnitude of the energy changes. Using the formulation described
here, the same quantity is given by subtraction of equations 9 and 5:

\[ W^+ = |T| - TEE \]

The generalised energy transfer equation (Eqn 5.9) can only be used to
calculate the magnitude of energy exchanged within a system boundary. The
energy exchanged between two parts of a system can be calculated by
observing the same events with different internal boundaries. Given the
system can be divided into two parts, \( P \) and \( Q \), then the magnitude of energy
exchanges between \( P \) and \( Q \) can be calculated by subtraction:

\[ |T|_{P<Q} = |T|_{(P+Q)} - |T|_{(Q)} \]  

Eqn 5.12

where \( |T|_{P<Q} \) is the energy exchanged between \( P \) & \( Q \)
\( |T|_{(P+Q)} \) is the energy exchange within the whole \( (P+Q) \) system
\( |T|_{(Q)} \) is the energy exchange within \( Q \) alone.

Equations 5.9 and 5.12 are in effect, a generalisation of the transfer
calculation proposed by Pierrynowski et al (1980), giving numerical results
exactly half of those produced by Pierrynowski et al 's method if the model
is configured with body segments as "units" and segmental potential and kinetic energies as the "components" of each unit. The generalisation has a number of advantages, however. Firstly, the energy system can be arbitrarily configured to model any number of energy components of different types. Secondly, constraints on energy exchange can be implicitly defined when the configuration of the internal energy boundaries of the model is selected. As before, within and between unit energy exchange can be distinguished by applying different energy boundaries to the same observations and calculating the difference between the two energy transfer magnitudes thus obtained. Essentially, energy analysis using the generalised model involves:

1: Definition of the energy components of the system and the relationships between them.
2: Selection of the internal boundaries to enclose measurable energy units. Implicit in the definition of the boundaries are the constraints on energy exchange and the selection of the exchanges which are to be observed.

5.3 EXAMPLES

The generality of the proposed energy analysis method can be illustrated by means of a number of examples, each representing an alternative interpretation of an energy model previously described in the literature.

5.3.1 The point-mass energy model

The energy model implicit in the point mass model of the human body (Figure 5.6) has two components; kinetic energy and potential energy, calculated from the displacement and velocity of the body's centre of mass. When the model is used to calculate "efficiency" it is implicitly assumed that a proportion of the chemical energy released by muscle is transformed into mechanical energy, the rest being dissipated as heat. Only potential and kinetic energy are explicitly modelled, however, so chemical energy input and heat production must be included in the energy exchange across the system boundary (Figure 5.6). The change in the total mechanical energy of the system during the time interval $\Delta t$ is

$$\Delta \text{(mechanical energy)} = \Delta (PE + KE)$$

Eqn 5.13
Figure 5.4: Mechanical energy changes implied in the point-mass analysis.
and the magnitude of the energy exchange between potential and kinetic energy is given by application of equation 5.9

\[ |T| = \frac{1}{2} (|\Delta PE| + |\Delta KE| - |\Delta(PE+KE)|) \]  
Eqn 5.14

### 5.3.2 The "pseudowork" model

The pseudowork model of Norman et al (1976) can also be configured. (Figure 5.5). The model assumes that no energy transfers are possible either between segmental components or between segments. Defining segments as "units" and segmental potential, translational kinetic and rotational kinetic energies as "components", application of equation 5.5 without correction yields Norman et al's (1976) equation for "total pseudo work" (Equation 2.4, section 2.3.5). Since no allowance is made for energy exchange, it is implied that the total energy exchange thus calculated all occurs across the system boundary, and application of equation 5.9 to calculate energy exchange within the system will always give a zero result.

### 5.3.3 The unrestricted transfer model

The method of calculation of mechanical work proposed by Winter (1979a) considers only the sum of changes in the total energy of the system. This model can thus be configured in the same way as that for the "pseudowork" model, but with energy exchange permitted between all segmental components and between all segments (Figure 5.6). The total energy exchange in the system over a gait cycle, including that across the outer boundary is given by application of equation 5.5. Again the system is modelled as \(s\) segmental units, each with \(c = 3\) components:

\[ TEE = \frac{1}{2} \sum_{i=1}^{t} \sum_{j=1}^{s} \sum_{k=1}^{c} |\Delta e_{i,j,k}| \]  
Eqn 5.15

Similarly applying equation 5.10 gives

\[ |T| = \frac{1}{2} \sum_{i=1}^{t} \left( \sum_{j=1}^{s} \sum_{k=1}^{c} |\Delta e_{i,j,k}| - |\Delta \sum_{j=1}^{s} \sum_{k=1}^{c} e_{i,j,k}| \right) \]  
Eqn 5.16

The energy exchanged across the system boundary is obtained by subtraction:
Figure 5.5: Mechanical energy changes implied in the "pseudo-work" analysis
Figure 5.6: Unrestricted energy transfer: mechanical energy changes implied in the calculation of $W_{\text{mb}}$. 
This is equivalent to the "positive work" component of Winter's (1979) work calculation; i.e. is equal to half the "total internal work done during a given time" (Winter, 1979; p.80, Eq 4). It is also noted that equation 5.16, the calculation of total energy transfer exchanges within the system boundary is equivalent to the difference between the pseudowork result (Norman et al, 1976) and that of Winter (1979a). It thus directly mirrors the calculation of total within and between segment energy transfers ($T_{wb}$) used by Pierrynowski et al (1980).

5:3.4 Partial (within segment only) energy exchange configuration

Pierrynowski et al (1980) also describe an energy calculation ($W_{w}$, equation 2.7, section 2:3.6) performed with the assumption that there is energy exchange within the body's segments but not between segments (Figure 5.7). Applying equations 5.5 and 5.9 gives

$$\text{TEE} = \frac{1}{2} \sum_{i=1}^{c} \sum_{j=1}^{c} \sum_{k=1}^{c} \left| \Delta e_{i,j,k} \right|$$

(as before) and

$$|T| = \frac{1}{2} \sum_{i=1}^{c} \sum_{j=1}^{c} \sum_{k=1}^{c} \left( \Delta \sum_{i=1}^{n} e_{i,j,k} \right) - \Delta \left( \sum_{i=1}^{c} e_{i,j,k} \right)$$

By subtraction,

$$\text{TEE} - |T| = \frac{1}{2} \sum_{i=1}^{c} \sum_{j=1}^{c} \sum_{k=1}^{c} \left| \Delta \left( \sum_{i=1}^{n} e_{i,j,k} \right) \right|$$

which is equivalent to Pierrynowski et al's calculation of $W_{w}$. Applying the subtraction principle (equation 5.12) to the results of equations 5.16 and 5.19 allows calculation of "within segment transfer" ($T_{w}$) on the same basis as that proposed by Pierrynowski et al (1980).
Figure 5.7: Mechanical energy changes with partially restricted (within segment only) energy exchange.
5:4 CONSTRAINTS ON BETWEEN-SEGMENT ENERGY TRANSFER

The generalised model enables constraints to be applied between-segment energy transfer in the moving human body. The generalised transfer equation is expressed in terms of an energy structure consisting of units containing components between which energy exchange can occur. Assumptions regarding energy exchange between segments are therefore represented by internal boundaries within the energy system, provided that where energy exchange is allowed, it is assumed unrestricted. If, for example, it is assumed that energy exchange can occur only between segments of the same limb, then the limbs must be separated by boundaries within the energy system. This example is illustrated in Figure 5.8. Equations 5.5 and 5.9 can be applied as before, with the system defined as having 5 units, each containing 9 energy components.

The case where energy can be exchanged, but not freely, between segments is less straightforward. However it is possible to apply such constraints by limiting the change in an energy component that can occur during a given time period. This could be done in two ways:

1: Assuming each possible energy exchange pathway has a maximum capacity, it is possible to calculate the maximum rate at which the value any given energy component can increase or decrease. Thus in the calculation of total energy exchange (Eqn 5.5) and within-boundary energy transfer (Eqn 5.9) the values of $e_{i,j,k}$ can be limited by a computation of the form

$$
\Delta e_{i,j,k} \leftarrow \max(\Delta e_{i,j,k} , \Delta e_i) ; \quad \Delta e_{i,j,k} > 0 \tag{5.21a}
$$

$$
\Delta e_{i,j,k} \leftarrow \min(\Delta e_{i,j,k} , \Delta e_0) ; \quad \Delta e_{i,j,k} < 0 \tag{5.21b}
$$

where $\Delta e_i$ and $\Delta e_0$ are the maximum absolute energy flows into and out of the component in a given time period and the operators $\max$ and $\min$ yield the highest and lowest members of a set respectively.

2: If it is to be assumed that the actual rate of energy exchanges via a particular pathway is a simple proportion or some function of the unrestricted energy exchange potential then the exchange can be modelled by applying the transformation
Figure 5.8: Mechanical energy changes with partially restricted (within limb) energy exchange
\[ \Delta e_{i,j,k} \leftarrow L(\Delta e_{i,j,k}) \]  

Eqn 5.22

where \( L() \) is the modifying function.

**5.5 DISCUSSION**

The energy analysis procedures methods developed here are a simple extension of those previously reported in the literature (Norman et al., 1980; Winter, 1979a; Pierrynowski et al., 1980). The generalisation enables analysis of an energy model defined in terms of a set of energy subsystems between which energy exchange may occur. There is no restriction on the form which component energies may take, provided that their relationships with the other parts of the system can be defined. Models in which energy exchanges are restricted or constrained to particular pathways are also dealt with.

An alternative method is to calculate energy exchanges using previously published methods, but to apply factors to the results in order to adjust the change in the total energy on the basis of assumptions concerning between segment energy transfer, positive and negative work efficiency ratios, elastic energy storage, etc. (Williams and Cavanagh, 1983; Eqn 2.10 section 2:3.6). This method is also based on a recognition of the need to consider additional energy components and possible constraints on energy exchange. However a method whereby the energy components, energy flows and constraints are defined in the energy model before analysis is probably a more sound basis for future development.

Although a procedure for the analysis of an energy model has been defined, the methods described here do not assist in the construction of an appropriate energy model of the human body. Notably, while methods have been developed to enable the analysis of a human body mechanical energy system in which between-segment energy transfers are constrained, the question of what these constraints should be remains unanswered. Of more importance from a practical viewpoint therefore is that the analytical framework described here permits the potential contributions of other energy components, in particular strain energy, to be assessed.
Chapter 6

AN AUTOMATED SYSTEM FOR THE BREATH BY BREATH ANALYSIS OF RESPIRATORY FUNCTION DURING EXERCISE.
6:0 SUMMARY

A system for the continuous measurement of ventilation, gas exchange and heart rate in exercising human subjects is described. Expired air was analysed in real time, breath by breath, using a mass spectrometer and pneumotachograph under computer control. The system features fully automatic calibration and facilities for the storage, tabulation and graphical display of data. A "trend" subsystem enables the user to reduce sampling resolution if required.

The method was shown to give good agreement with traditional Douglas Bag techniques.

6:1 INTRODUCTION

The measurement of the respiratory function and heart rate responses to exercise is fundamental to research in Sports Medicine and exercise physiology and is of increasing importance in the assessment of a variety of pathological conditions. There is a need therefore for fast, automated instrumentation systems that enable these measurements to be made routinely, with maximum flexibility and a minimum of operator intervention.

Traditionally, measurements have been made on expired air collected in Douglas bags or meteorological balloons. The volume of air collected is measured by evacuation through a gas meter and its composition determined by chemical analysis or by means of oxygen and carbon dioxide analysers. Such methods can be tedious and only yield the average values of ventilation and gas exchange for the 30 of 60 second period of the gas sample collections.

For the purposes of this study, it is required to make both physiological and biomechanical measurements simultaneously on exercising subjects. There is therefore an advantage in terms of laboratory management if some of this data can be collected automatically.

This chapter describes the design and validation of a flexible and fully automated system for the continuous measurement of ventilation, gas exchange and heart rate during exercise. Expired air was analysed breath by breath, and numerical results were displayed in real time. The main advantages of the system are that calibration, data acquisition and the storage, retrieval and display of measured parameters are under computer control.
6:2 REVIEW OF LITERATURE

6:2.1 Traditional Methods

In order to measure ventilation, oxygen uptake and carbon dioxide output during exercise it is necessary to be able to determine the volume and composition of expired air.

Traditional methods of determining respiratory function parameters during exercise are based on the collection of expired air in Douglas Bags (Douglas, 1911) or Meteorological balloons (Daniels, 1971). Volumes of expired air are determined by evacuation of the bags through dry gas meters or into spirometers and the composition of the gas is determined by chemical analysis (Haldane, 1912; Scholander, 1947; Lloyd, 1958). The reliability of sampling and analysis by chemical methods is within 0.02% when the apparatus is in good condition and operated efficiently (Shephard et al, 1966).

More recently, the use of chemical gas analyses has been superseded by the introduction of electronic $O_2$ and $CO_2$ meters. These instruments have been demonstrated to be both accurate and reliable by a number of studies (Shephard et al, 1966; Sodal et al, 1968; Benade et al, 1970; Auchincloss et al, 1966).

6:2.2 The Haldane Transformation

Given a known volume of air of known composition, the volume of air inspired has generally been calculated by the Haldane transformation (Haldane, 1912). Due to gas exchange and variations in tidal volume, the volumes of air expired and inspired are not necessarily equal. The Haldane transformation corrects for this effect by assuming that the amount of Nitrogen exhaled is equal to the amount of Nitrogen inspired. Thus the inspired ventilation is estimated from the expired ventilation as:

\[ \dot{V}_i = \dot{V}_e * \frac{F_e N_2}{F_i N_2} \]

where
- $\dot{V}_i$ = inspired ventilation rate
- $\dot{V}_e$ = expired ventilation rate
- $F_e N_2$ = % Nitrogen in inspired air
- $F_i N_2$ = % Nitrogen in expired air
Oxygen uptake (VO₂) and Carbon dioxide production (VCO₂) are therefore calculated as:

\[
\dot{V}O_2 = \dot{V}_I \times F_{I}O_2 - \dot{V}_E \times F_{E}O_2 \\
\dot{V}CO_2 = \dot{V}_I \times F_{I}CO_2 - \dot{V}_E \times F_{E}CO_2
\]

where

\begin{align*}
F_{I}O_2, F_{E}O_2 & = \text{inspired, expired oxygen concentrations} \\
F_{I}CO_2, F_{E}CO_2 & = \text{inspired, expired Carbon dioxide concentrations}
\end{align*}

\(F_{E}N_2\) is not normally measured directly since only \(F_{E}O_2\) and \(F_{E}CO_2\) are easily determined by chemical and electronic methods. \(F_{E}N_2\) is therefore assumed to be 100% \(-(F_{E}O_2 + F_{E}CO_2)\). Similarly, the gaseous composition of inspired air is usually assumed to be that of the International Standard Atmosphere.

The validity of this procedure has been questioned by suggestions that there is a net production of expired Nitrogen during steady state exercise. Costa (1960) proposes a hypothetical metabolic pathway whereby Nitrogen might be excreted from the intestine via the lungs. Cizzik et al (1972a, 1972b) and Cizzik and Johnson (1972a, 1972b) found that 9 subjects produced an average excess of 217 ml N₂/min post absorptive and 509 ml N₂/min following a 67% protein meal. In the extreme case, Nitrogen production of this magnitude would cause a 13% error in the calculation of oxygen uptake by the Haldane transformation using traditional open circuit techniques.

Several studies refute the suggestion that Nitrogen production is significant however. Wilmore and Costill (1973) found Nitrogen retention in three of six subjects undergoing steady state exercise on a motor-driven treadmill. The total Nitrogen retention was less than 1% of the total volume of Nitrogen inspired and had no effect on subsequent calculations using the Haldane transformation. Fox & Bowers (1973) found a statistically insignificant Nitrogen uptake in a study of 20 sets of measurements on 5 fasted, resting subjects. Most conclusively of all, Luft et al (1973) made a total of 53 paired comparisons of oxygen uptake determined by the Haldane transformation and by direct measurement of \(V_I\). No significant difference was found between the two methods. The conclusion drawn by all these studies is that if Nitrogen uptake or production does occur, it is not significant enough to invalidate the calculation of \(\dot{V}O_2\) and \(\dot{V}CO_2\) by the Haldane transformation.
6:2.3 Semi-automated and automated systems

A number of automated and semi-automated systems for the analysis of expired air in human subjects have been described previously. (e.g. Poole & Maskell, 1975; Kappagoda et al, 1974; McCain & Donald, 1975). These systems require extensive manual calibration and are only capable of recording average oxygen uptakes over a period of time.

Wilmore & Costill (1974) measured ventilation directly by collecting expired air continuously in three anaesthesia bags connected to the subject via a three way valve. The valve system cycled in such a way that at any one time, one of the three bags is being filled by the subject, one is being sampled by gas analysers and the other is being evacuated by a vacuum pump into a Tissot spirometer. The system is described as "semi-automated" since the valves are cycled by hand. Also, ventilation volumes, $F_eO_2$ and $F_eCO_2$ were read directly from the instrument by an operator and fed into a programmable calculator by hand. The system enabled the collection of samples of as little as 15 seconds duration however, and the results of the Haldane transformation were available within about 10 seconds of the end of the sample collection.

Beaver et al (1973) describe a system in which analogue signals from a differential pressure pneumotachograph and fast-response $O_2$ and $CO_2$ analysers and a heart rate meter were recorded by a 16-bit digital laboratory computer. The use of a pneumotachograph enables breath by breath measurements to be made since expiratory flow rates are measured directly. The system was calibrated manually. A high correspondance was found between the results obtained using the automated system and those obtained from conventional methods. Similar but less sophisticated systems had been previously described by Durfrey and Leeming (1966), Cardus and Newton (1970) and Elliott et al (1970).

The Beckman "Metabolic Measurement Cart" (MMC) described by Wilmore et al (1976) consists of Beckman fast response $O_2$ and $CO_2$ analysers with a mixing chamber and biased flow turbine flow meter linked to a programmable calculator. The system has been shown to cross-validate closely with traditional methods (Wilmore et al, 1976) but requires manual calibration of the gas meters and breath by breath analysis is restricted to the measurement of $F_eO_2$ and $F_eCO_2$.

The use of mass spectrometers to analyse the composition of expired air (Hickson et al, 1978; Cerretelli et al, 1979; Hagberg et al, 1980) during exercise testing is a recent one, largely due to the high cost of these instruments. The advantage of a mass spectrometer is that the gas
sample can be analysed for FeN2 and FeArgon as well as FeO2 and FeCO2. It is also possible, using gas dilution techniques (whereby an inert tracer gas is introduced into the expire flow at a known rate) to calculate ventilation and gas exchange by mass spectrometry alone (Davies and Denison, 1979; Sawin et al, 1974).

6:3 GENERAL PRINCIPLES

The primary components of the system are a mass spectrometer (Centronic MGA 200) and a pneumotachograph (Mercury type F1000L) interfaced to a laboratory computer (DEC PDP11/03). The subject breathes through a lightweight, low resistance breathing valve (Jakeman & Davies, 1979) which has been modified to accept a probe from the mass spectrometer. The expire flow of the breathing valve is connected by light-weight, flexible tubing to the pneumotachograph. Heart rate is monitored by an ECG monitor/display unit (Albury Instruments Ltd.). Overall control of the system was achieved by application related software written in MACRO-11 assembly language and FORTRAN. Analogue signals from the measuring instruments are monitored continuously. During each expiration incoming data is stored in computer memory. During each inspiration, data from the last expiration is analysed and the results of calculations are displayed on the terminal screen. A "trend" system allows the recorded data to be "replayed", summarised and displayed in tabular or graphical form.

6:4 HARDWARE

The main components of the system are shown in Figure 6.1 and illustrated in schematic form in Figure 6.2

6:4.1 The Mass spectrometer

The Centronic MGA 200 is a fast response, multichannel gas analyser designed for use in the clinical environment (e.g. Gothard et al, 1980). The main component of the unit is a quadropole mass spectrometer which can detect gases and fragments of gases in the molecular weight range 2-200 atom mass units (amu).

A small fraction of the gas sample diffuses into the analyser at a low pressure (10^-7mmHg) where it is ionised. A variable electrostatic field, generated by the quadropole head is used to select ions of a
Figure 6.1: (Plate) An automated system for the breath by breath analysis of respiratory function during exercise.

Main components:

A: Breathing valve + gas sampling probe
B: Heated pneumotachograph + thermistor
C: Flow/temperature unit
D: MGA 200 Mass spectrometer
E: Flow meter for calibration
F: DEC VT100 console terminal
G: ECG monitor and heart rate meter
H: DEC PDP11/03 computer + gas control unit

Not shown: Printer
Calibration gas bottles
Figure 6.2: Schematic diagram of the system illustrated in Figure 6.1 showing functional relationships between main components.
particular mass/charge ratio. The selected ions bombard an electron multiplier which acts as both detector and amplifier. For every ion entering the electron multiplier, about $2^{14}$ electrons emerge. This electron stream is proportional to the concentration of the selected ion in the sample. Additional circuitry enables the selectivity of the quadrupole analyser to be scanned or switched rapidly, thus permitting the generation of mass spectra or the virtually simultaneous monitoring of up to eight gases on separate output channels.

Under normal operating conditions, each channel of the mass spectrometer is "tuned" to a particular peak in the mass spectrum and calibrated manually with standard gas mixtures. It is possible however to bypass the manual tuning and calibration hardware in such a way that the device can be placed directly under computer control.

According to the manufacturer's literature, the mass spectrometer has a response time of less than 100 ms, linearity of $>99\%$, drift of less than 1% per 24 hours and a resolution equivalent 0.03 mol% of Argon. These data refer to a new machine. Performance is likely to fall off as consumable components deteriorate. Hallback, Karlsson and Ekblom (1980) compared analyses of oxygen and carbon dioxide concentrations using chemical analytical methods and the MGA 200. The MGA 200 was found to be linear in the range of normal ambient air breathing and the differences between the two methods were of the same order as the variability within the standard Haldane chemical analysis technique.

6.4.2 Gas sampling

Gas samples are directed to the MGA 200 inlet by means of 2.5 metres of 0.25 internal diameter polyethylene capillary tubing. The transit time of the gas sample in the capillary can be approximated by the equation

$$T_t = \pi d^2 l/4s$$

where

- $T_t$ = transit time
- $d$ = capillary diameter
- $l$ = capillary length
- $s$ = sample flow rate

Given a typical sample flow rate of 20 mls.min$^{-1}$, and the capillary described above, the transit time is of the order of 0.4 seconds. Factors other than capillary volume and sample flow rate have an effect on the
transit time. Surface frictional resistance to flow and the presence of water vapour in the capillary would tend to increase the transit time.

The transit time of the sample presents no methodological problem with this system since all incoming signals from a range of analogue devices are synchronised by the computer (See below, section 6:6.1). However, any absorption by water vapour in the capillary while the gas sample is in transit may effect its composition. Also, lighter gases tend to maintain a sharper concentration profile in the capillary due to faster radial diffusion (Taylor, 1953). This distortion would tend to increase with greater transit times. Davies and Denison (1979), however, have found that 30m long capillaries did not introduce any important distortions into the gas concentration profile in a range of respiratory measurements. The effects of water vapour and the Taylor effect on gas composition can therefore be considered negligible in a 2.5 metre capillary.

6:4.3 Breathing valve

The construction and performance of the breathing valve (Figure 6.3a) used in this study have been fully described by Jakeman & Davies (1979). This particular valve has the advantage of light weight (65 gm). It is therefore possible with a flanged rubber mouthpiece for subjects to carry the valve without additional support. Also, the low resistance of the valve system is important since high resistance in the breathing circuit can reduce $V_{O_2}\text{max}$ and limit endurance (Cerretelli et al, 1969; Stemler and Craig, 1977).

6:4.4 Gas sampling probe

The breathing valve described by Jakeman & Davies (1979) was modified by the addition of a gas sampling probe (Figure 6.3b) connected directly to the gas sampling capillary. The probe was designed to allow respiratory gases to be sampled at the mouth and also to permit the introduction of calibration gases into the mass spectrometer without dismantling the probe from the breathing valve. The gas sampling capillary was terminated in a short length (3 cm) of stainless steel tubing in the probe. The sampling capillary was encased in a second capillary connected to the calibration gas inlet. A small hole between the two capillaries was normally sealed by a valve rubber. Positive pressure in the calibration gas capillary opens the valve rubber seal, allowing calibration gas to pass through.
Figure 6.3a: Salford HPL breathing valve (after Jakeman and Davies, 1979) modified by the addition of a gas sampling probe.
Figure 6.3b: Schematic diagram of gas sampling probe
(1) during data acquisition
(2) during calibration
6:4.5 Pneumotachograph

The pneumotachograph used in this system was a standard unit (Mercury Type F1000L) of the differential pressure transducer type with a peak flow capacity of 1000 L.min⁻¹. Expired air from the breathing valve is conducted to the pneumotachograph by lightweight, flexible, 1.75 inch diameter tubing (Falconia Ltd.). To assist the generation of a linear flow through the pneumotachograph, the tubing was held in line with the inlet port of the pneumotachograph by the suspending arm (See Figure 6.1).

During early trials of the system, a tendency for water vapour to condense in the pneumotachograph was noted. The condensation was prevented by adding a 5W heating wire to the casing of the unit. The pneumotachograph was further modified by the addition of thermocouple to enable gas temperature to be determined continuously at the point of flow measurement.

The electronic components of the flow transducer and the transducer circuitry for the thermocouple are housed in a single unit referred to as the "flow/temperature unit."

6:5 INTERFACING AND SOFTWARE

The computer used for this system was as a standard DEC PDP11/03 with twin floppy disk drives and 32K RAM. The computer was interfaced to the mass spectrometer and other transducers by an eight channel multiplexed analogue to digital converter, digital to analogue converters and switched inputs and outputs. The unit to perform these functions was designed and built by Centronic Ltd., and installed in the computer cabinet.

The system software falls neatly into two parts. The first of these was written mainly in MACRO-11 assembly language and is responsible for all interface control, data acquisition and calibration. Most of this "low-level" software was driven under interrupt control. The second part was written in FORTRAN IV and performs calculations, high level functions, data storage retrieval and analysis in addition to interacting with the user.

User commands to the system are input at the computer console keyboard. At each phase of the operation, the next command is selected from one of a hierarchy of menus displayed on the terminal screen. Messages, instructions and warnings of hardware failure and operator error are also displayed on the console screen.
**DATA ACQUISITION**

Analogue signals output by the flow/temperature unit, mass spectrometer and ECG unit are sampled continuously, under computer control. The eight channels of the analogue to digital converter are switched every cycle of the real time clock. Thus all eight channels are scanned every eight clock cycles. In fact there are only seven active input channels, receiving signals representing:

1: Flow rate  
2: Temperature  
3: % Nitrogen in gas sample  
4: % Oxygen  
5: % Argon  
6: % Carbon dioxide  
7: Heart rate

During the eighth clock cycle, when there is no data to collect, the data from the seven previous inputs are added to a sample buffer for later processing, and the channel buffer is cleared for the next cycle.

As each of the 4 gas measurements are input from the mass spectrometer, control signals change the mass value to be analysed by the mass spectrometer and if necessary, change its gain.

This sampling cycle repeated until a period of 0.04 seconds has elapsed at which time the sum of all the inputs on each channel during the preceding 0.04 seconds are transferred to a specially nominated area of computer memory which is referred to as "the queue". In addition to the sums of the seven input channels, the information stored in the queue includes the number of sample collections that have been made, and a "breath phase indicator". This indicator is determined by close monitoring of expiratory flow. As expiratory flow rises above a certain threshold (10 L.min⁻¹) a "start expire" signal is passed to the queue. Thereafter an "expiring" indicator is passed until the expire flow falls below the threshold level at which point an "end expire" indicator is passed.

The sampling period is adjusted automatically depending on the amount of available space in the queue. This allows the FORTRAN application programme to use data from the queue at a different rate from that of data collection. Data acquisition starts with a sample period of 0.04 seconds i.e. ADC inputs are summed over this period before being passed to the
If the application software does not process data quickly enough then the amount of data in the queue increases. If this accumulation of data was allowed to continue then the queue would become full and data would be lost. For this reason, every time the remaining space in the queue is halved, the sample period is doubled, thus halving the rate at which the queue fills. Conversely, if the space available in the queue doubles, the sample period is halved, giving twice as many samples to be queued. The system allows compression or expansion of data samples to be repeated as many as four times consecutively. This means that a queued sample may represent from one to 16 collected samples. Under most circumstances this facility was not required, except at very high breathing frequencies when the inspiratory period is brief. Even under these conditions, the sample period was normally doubled only once to 0.08s.

6:6.1 Synchronisation

Although the mass spectrometer has a response time of less than 0.1s to a step change in gas concentration there is a response lag due to the transit time of the sample between the probe and the instrument. There is therefore a requirement to achieve some form of synchronisation between flow and gas concentration signals, since the gas data arrives at the computer some 300-400 milliseconds after the flow data. This synchronisation is achieved by buffering incoming flow data. The size of the queue is determined by the sample rate and the required synchronisation delay time. Incoming flow and temperature data are stored in the buffer and the values at the head of the flow stack are used in the sample. Thus flow and temperature data are held in the buffer for the required period of time until the appropriate gas concentration data arrives from the mass spectrometer.

6:7 DATA PROCESSING

The first phase of data processing is programmed as part of the interrupt driven data acquisition software. This phase is simply an averaging process whereby the last set of queued data is divided by the number of samples to give a mean value for the sample period.

Also at this point, the values of gas concentrations from the mass spectrometer are normalised to a total of 100%. The simple numerical procedure performs the function of an "automatic summing circuit" (Scheid et al, 1971) and compensates for the effects of variations in the sample
water vapour, probe resistance and mass spectrometer gain that can introduce "drift" into the mass spectrometer output (Gibson, 1978; Auchtincloss et al; 1970). As each normalised sample is returned to the queue, gas concentrations, flow and temperature measurements are output to digital to analogue converter ports so that, if required, the intra-breath variation in these parameters can be plotted on a chart recorder in real time (Figure 6.4) the system also permits the display of intra breath data in numerical form on the computer console screen.

Further processing of the input data signals is performed by high level application-related software. The high level routines run "continuously" while the low level data acquisition is interrupt driven. Thus while data is being collected during expiration, most of the processor time is taken up by data acquisition. During inspiration, there is no data to acquire and the high level subroutines are able to act on the queued data.

Data is analysed one breath at a time. For the purposes of breath by breath analysis, an expiration was defined as beginning or ending when respiratory flow rises above or falls below 10 L.min⁻¹ and is required to have a minimum duration of 0.1 seconds. The minimum expired volume that can constitute a "breath" is 0.17L therefore. These thresholds are generally sufficient to allow the rejection of spurious breaths caused by swallows, coughs, valve flutter, etc. and their values can be adjusted to suit particular requirements. The information contained in any sub-threshold expire is not lost, but is included in the next "true" breath.

Flow is integrated over each expiration to give tidal volume. Each sample is corrected to standard temperature before summation into the integral to allow for intra-breath temperature variations. Gas partial volumes are determined as the integral of the product of flow and gas concentration. Gas concentrations in inspired air are assumed to be equal to atmospheric values, as measured immediately post calibration. Oxygen uptake and carbon dioxide output are calculated using the Haldane transformation and all volumes are corrected for saturated water vapour pressure and reduced to standard temperature and pressure.
Figure 6.4: Variation in expiratory flow rate, expired air temperature and expired air concentrations of oxygen and carbon dioxide recorded during a single breath at rest.
The following parameters are computed and displayed for each breath:

<table>
<thead>
<tr>
<th>PARAMETER</th>
<th>Symbol</th>
<th>UNITS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elapsed time</td>
<td></td>
<td>seconds</td>
</tr>
<tr>
<td>breath count</td>
<td></td>
<td></td>
</tr>
<tr>
<td>breath duration</td>
<td></td>
<td>seconds</td>
</tr>
<tr>
<td>breathing frequency</td>
<td>( f )</td>
<td>breaths/min</td>
</tr>
<tr>
<td>Ventilation</td>
<td>( \dot{V}E )</td>
<td>L.min(^{-1})</td>
</tr>
<tr>
<td>Expired ( O_2 )</td>
<td>( F_e O_2 )</td>
<td>%</td>
</tr>
<tr>
<td>Expired ( CO_2 )</td>
<td>( F_e CO_2 )</td>
<td>%</td>
</tr>
<tr>
<td>Oxygen uptake</td>
<td>( \dot{V}O_2 )</td>
<td>L.min(^{-1})</td>
</tr>
<tr>
<td>Oxygen uptake</td>
<td>( \dot{V}O_2 )</td>
<td>mls.kg(^{-1}).min(^{-1})</td>
</tr>
<tr>
<td>Carbon dioxide output</td>
<td>( VCO_2 )</td>
<td>L.min(^{-1})</td>
</tr>
<tr>
<td>Carbon dioxide output</td>
<td>( VCO_2 )</td>
<td>mls.kg(^{-1}).min(^{-1})</td>
</tr>
<tr>
<td>Gas exchange ratio</td>
<td>( R )</td>
<td></td>
</tr>
<tr>
<td>Ventilatory equivalent</td>
<td>( V_e \cdot \dot{V}O_2^{-1} )</td>
<td></td>
</tr>
<tr>
<td>Heart rate</td>
<td>( HR )</td>
<td>beats.min(^{-1})</td>
</tr>
</tbody>
</table>

The respiratory function measurements are calculated using standard equations (Consolazio et al, 1963).

As it is calculated, breath by breath data is stored on diskette for later review and analysis.

Under normal operating conditions, the processing of data queued during a given expiration is completed and the results displayed during the next inspiration.

6:8 CALIBRATION

The measuring instruments incorporated into the system are calibrated automatically by the computer. During calibration, an air blower is connected via a 45m\(^3\) min\(^{-1}\) capacity flow meter (GEC Marconi, Series 2000, Type 47X) to the breathing valve in place of the mouthpiece (Figure 2). With the blower switched on by the computer, the indicated flow is compared with the analogue output of the pneumotachograph to determine the calibration constant for flow. The flow transducer zero is determined electronically, and the thermocouple is calibrated against room air
6:8.1 Mass Spectrometer Tuning

The computer system tunes the mass spectrometer by scanning the whole mass spectrum of the atmosphere (Figure 6.5). The largest peak in the spectrum is found and assumed to be that of Nitrogen at a mass/charge ratio of 28. If the Nitrogen peak is not of sufficient magnitude for reliable results, an error message is presented and the calibration is abandoned. Once the mass spectrometer head voltage required to elicit the Nitrogen peak is determined, the other important peaks can be found. The four peaks used by the system are:

<table>
<thead>
<tr>
<th>Gas fraction</th>
<th>Mass/charge ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nitrogen</td>
<td>28</td>
</tr>
<tr>
<td>Oxygen</td>
<td>32</td>
</tr>
<tr>
<td>Argon</td>
<td>40</td>
</tr>
<tr>
<td>Carbon dioxide</td>
<td>44</td>
</tr>
</tbody>
</table>

These peaks can be clearly seen in the typical mass spectrum of air shown in Figure 6.5.

6:8.2 Mass spectrometer calibration

The mass spectrometer is calibrated using 100% Argon to determine zeros values and a standard mixture of Nitrogen, Oxygen, Argon and Carbon Dioxide. Standard gas mixtures are always cross-checked against other standards using both electronic O₂ and CO₂ analysers and mass spectrometry, the original standard having been assayed for O₂ and CO₂ using Lloyd-Haldane technique (Lloyd, 1958).

The flow of calibration gases to the mass spectrometer probe is regulated by electromagnetic valves under computer control. Positive pressure in the calibration gas capillary (Figure 6.4) opens the valve rubber seal covering an inlet port in the probe and allowing calibration gas to pass through. A delay between calibration gas collection and data sampling allows gas conditions to stabilise.

The calibration is checked by comparing the measured composition of a sample of room air with standard values. If these values are within predetermined tolerance limits, the system accepts the calibration and data acquisition is enabled; otherwise calibration is rejected and an error
Figure 6.5: The mass spectrum of normal atmosphere.

Main peaks:
- 28 Nitrogen
- 32 Oxygen
- 40 Argon
- 44 Carbon dioxide
- 18 Water vapour
- 16 Oxygen (double charged)
- 14 Nitrogen (double charged)
- 29 Nitrogen ($^{14}$N$^{15}$N)

mass/charge ratio
message is displayed. Normally, the automatic calibration sequence is completed within two minutes of initiation. It has been found that once calibrated, the system can remain stable for up to three hours, but to maximise reliability, the system is normally recalibrated between exercise tests or at hourly intervals when in continuous use.

6:9 **TREND SUBSYSTEM**

In many experimental situations, breath by breath sampling resolution of respiratory function measurements is not required and, indeed, can cause confusion. Some data reduction process is necessary, therefore. Under "steady-state" conditions, the mean rates of ventilation, oxygen uptake and other parameters can be approximated by averaging the breath by breath values. Hagberg et al (1980) for example, smoothed breath by breath data by numerically averaging, after each five second period, the next breath with the two breaths preceding and following it. Since the values being averaged in this case are rates, however, the true mean rate will only be equal to the average calculated in this way if the sample time base is regular, i.e. if breaths are of precisely equal duration. Thus a simple averaging procedure can give erroneous results.

For the purposes of this study analytical software was developed which allows breath by breath data to be integrated over any time base to give correct values for ventilation and gas exchange. For each sample period, the system records tidal volume, partial volumes of Nitrogen, Oxygen, Carbon dioxide and Argon, and the duration of the sample. These values can be summed over any time period to give the total ventilation and total partial volume of each gaseous component of the expired air, from which $F_{E}O_2$ and $F_{E}CO_2$ can be calculated by simple division.

During on-line, breath by breath analysis the time limits of the summation are given by the start and end of a breath. Previously collected data can be recalled, however, and reintegrated over any time period. In this way, large quantities of data can be reduced to a more manageable form without loss of accuracy. Figure 6.6 shows the effect of this computation on typical breath by breath oxygen uptake data collected during a five minute treadmill run. Figure 6.7 and Table 6.1 show examples of "trended" results from a 6-minute "steady state" exercise test of the type described in Chapter 9.
Figure 6.6: Oxygen uptake recorded before during and after a five minute steady state exercise test.

(A) breath by breath data
(B) 15 second trend
Figure 6.6: Oxygen uptake recorded before during and after a five minute steady state exercise test.

(C) 30 second trend
(D) 60 second trend.
Figure 6.7: Trended results (30s) of a six minute exercise test.

(A) Subject at rest

(B) Mouthpiece removed

(C) 6 minute treadmill run at 10 mph

(D) Recovery
Table 6.1: Trended results (30s) of a six minute exercise test.

(A) Subject at rest
(B) Mouthpiece removed
(C) 6 minute treadmill run at 10 mph
(D) Recovery
6:10 VALIDATION

6.10.1 Methods

The described system was compared with traditional methods by connecting Douglas bags to the outlet of the pneumotachograph. Thus expired air, having passed through the automated system, was collected for one minute periods while subjects walked or ran on a motorised treadmill at a range of constant speeds. The subjects were male and female undergraduate and postgraduate Physical Education students of mixed running ability.

A sample of known volume was removed from each Douglas Bag and its composition determined using a paramagnetic oxygen analyser (Taylor Servomex Ltd, Type OA272) and an infra red Carbon dioxide analyser (Mine Safety Appliances Ltd., Lira model 303). The remaining volume of air collected in each Douglas bag was measured by evacuating the bags through a Parkinson Cowan dry gas meter which had previously been calibrated against a 600L capacity Tissot spirometer. All volumes were corrected to standard temperature and pressure.

$\dot{V}_E$, $F_{\text{E}O_2}$ and $F_{\text{E}CO_2}$ thus determined were compared directly with results recorded by the automated system, reintegrated by the trend subsystem over the one minute period of the Douglas bag collection. In a first series of experiments, 57 comparisons of $\dot{V}_E$ measurements were made. A later series consisted of a further 44 comparisons of $\dot{V}_E$, $\dot{V}_{\text{O}_2}$ and $\dot{V}_{\text{CO}_2}$.

6.10.2 Results

$\dot{V}_E$ measured by the Douglas bag method (DB) and the automated system (AS) for all 101 trials are plotted in Figure 6.8. The two methods of determining $\dot{V}_E$ are highly correlated ($r=0.998$). Student's t test for correlated means showed no significant difference between $\dot{V}_E$ determined by the two methods ($t=0.176$). Regression analysis showed that the relationship between $\dot{V}_E$ Determined by the two methods lies close to the line of identity. ($\text{AS} = 0.996 \times \text{DB} + 0.24$).

The correlation between the two sets of $\dot{V}_{\text{O}_2}$ measurements (Figure 6.9) was 0.988, with a Student's t test for correlated means showing no significant difference between the two methods ($t=-1.64$).

In the case of $\dot{V}_{\text{CO}_2}$ measurements, similarly, a product moment correlation coefficient of 0.994 was determined and a Student's t test for correlated means showed no significant difference between the two methods.

In all three cases, the mean absolute differences between measurements made by the two systems (1.0 L/min $\dot{V}_E$: 0.938 mls/kg/min $\dot{V}_{\text{O}_2}$ and 1.08 mls/kg/min $\dot{V}_{\text{CO}_2}$) were of the same order as the resolution of the Douglas
Figure 6.8: Minute ventilation ($\dot{V}_E$) recorded by Douglas Bag (DB) and automated (AS) methods.
Figure 6.9: Minute oxygen uptake (\(\dot{V}O_2\)) recorded by Douglas Bag (DB) and automated (AS) methods.

\[ r = 0.988 \]
Figure 6.10: Minute carbon dioxide output ($\mathring{\text{VCO}_2}$) recorded by Douglas Bag (DB) and automated (AS) methods.
breath by breath measurement procedure. There is the possibility however that there are random errors which are averaged out by integration. Variations in end expiratory volume for example can cause $\dot{V}_{O_2}$ and $\dot{V}_{CO_2}$ values measured at the mouth to be more dispersed about the mean value. Breath by breath variations in end expiratory lung volume of ± 360 mls have been reported (Wessel et al, 1979). These variations invalidate the use of the Haldane transformation at the breath by breath level since it is based on the assumption that $\dot{V}_1N_2$ and $\dot{V}_E N_2$ are equal. (This problem would tend to be exaggerated if averaging procedures of the type described by Hagberg et al (1980) are used in preference to the integration procedure used in the system described here.) Notwithstanding this effect, a number of breath by breath systems have used the Haldane transformation (Beaver et al, 1973 for example) even when the exercise conditions being studied are non-steady state (Auchincloss et al, 1968; Hagberg et al, 1973) and good correspondences with traditional methods have been reported. Given that variations in end expiratory volume are random, and that there is no net uptake or production of Nitrogen under normal exercise conditions (Luft et al, 1973; Fox and Bowers, 1973; Wilmore and Costill, 1973) it is reasonable to assume that, over a number of breaths, variations in end expiratory volume even out, the sums of $\dot{V}_1N_2$ and $\dot{V}_E N_2$ equalise and the Haldane transformation becomes acceptable. Given that the mean absolute difference between ventilation measured by the Douglas bag method and that measured by the automated system were of the order of 1 L.min⁻¹, then an end expiratory volume of 0.32L (the outer limit of the range determined by Wessel et al, 1979) would be lost in a typical 20 second sample.

In conclusion, it would seem reasonable to accept results from the trend system integrated over time periods less than that used in the validation procedure, providing the integrated sample contains a large enough volume of expired air to render variations due small changes in end expiratory volume negligible.
Chapter 7

AN EVALUATION OF THE MODELS USED TO DETERMINE THE "ANAEROBIC THRESHOLD" FROM RESPIRATORY RESPONSES TO INCREMENTAL EXERCISE.
The purpose of this study was to examine the validity of the models used to determine the "anaerobic threshold" non-invasively from the respiratory responses to incremental exercise.

Ten male subjects underwent an incremental treadmill exercise test to exhaustion. Ventilation and gas exchange were measured continuously using an automated, fast response system. Two-part and three-part multiple linear models and a continuous model were fitted to $\dot{V}_E$, $\dot{V}CO_2$ and $\dot{V}E\cdot\dot{V}O_2^{-1}$ responses by least squares regression techniques. Root mean square residuals were used to compare the goodness of fit of each model. Approximations to the first and second derivatives of ventilation and gas exchange functions were used to assist in the detection of inflection points in the response data.

The continuous model was found to provide an equally good or better representation of the respiratory responses than either the two-part or three-part linear model. Inflections in all the measured parameters occurred synchronously in each subject at a mean relative workload of 71.6% $\dot{V}O_2\max$. The buffering effect of the cardiovascular system and "noise" due to variations in tidal volume and breath duration make the determination of "instantaneous" inflection points impractical, however.

It is concluded that the method of determining the anaerobic threshold by applying multiple linear models to respiratory responses to incremental exercise has not been adequately validated.

**INTRODUCTION**

In Man, the total body oxygen uptake response to light exercise is linearly related to the work rate. Under heavy exercise conditions, the oxygen demand in the muscles exceeds the cardio-respiratory system's ability to supply oxygen and the additional energy requirement is met by anaerobic glycogenolytic metabolism, with a concomitant increase in lactic acid production.

While it is possible to estimate the metabolic energy released via aerobic pathways from measurements of oxygen uptake and carbon-dioxide production, the release of energy by anaerobic metabolism cannot be directly quantified. It is therefore necessary to have some method of
determining the exercise intensity at which the contribution of anaerobic metabolism becomes significant enough to invalidate the calculation of energy expenditure from gas exchange measurements.

Blood lactate concentrations can be determined by enzymatic assay of small blood samples (Olsen, 1971) and indicate the net balance between lactate production and removal in muscle. However it would be more convenient, in experiments where ventilation and gas exchange are being measured, if the onset of anaerobic metabolism (the "anaerobic threshold") could be detected non-invasively by examination of respiratory responses.

A number of published reports have described the use of inflections or "breakpoints" in the ventilatory and gas exchange responses to incremental exercise tests as a means of determining the anaerobic threshold. The rationale behind these methods relates the stimulation of the respiratory control centres to the increase in the concentration of hydrogen ions and bicarbonate ions in the blood associated with lactic acidosis.

This chapter describes an experiment in which statistical analyses of the respiratory responses to incremental exercise were employed to evaluate the models used to determine the anaerobic threshold; and to examine the nature of inflections in ventilatory and gas exchange responses.

Figure 7.1 shows a typical set of ventilatory and gas exchange responses to an incremental exercise test. Inflections in graphs of this type are commonly used to determine, non-invasively, the onset of anaerobiosis during incremental exercise ("anaerobic threshold"). How the idea of a "threshold" or "non-linear increase" is applied to \( V_E \) or \( V\text{CO}_2 \) data is shown schematically in Figure 7.2. The data illustrated in this figure may represent either \( V_E \) or \( V\text{CO}_2 \) responses. A departure from the general trend of the increasing response is modelled by two lines (Figure 7.2a). Diagrams of this type are frequently seen in published studies of the anaerobic threshold. When the increments in work-rate are linear with time, the ventilatory responses are normally plotted as a function of time (e.g. Wasserman et al, 1973; Caiozzo et al, 1982). When the change in work rate or oxygen uptake is non-linear, then the responses are commonly treated as functions of oxygen uptake (e.g.Orr et al, 1982). For the purposes of this study, a model of this type is referred to as a "multiple linear model" and the junction point of the lines is termed a "knot". The position of the knot, i.e. the point of inflection is used to determine the anaerobic threshold. Figure 7.2b shows an extension to a three-part multiple linear model, where it is assumed that there are two independant inflection points.
Figure 7.1: Example respiratory responses to the incremental exercise test used in this study.
Figure 7.2: Schematic form of (a) 2 part and (b) 3 part multiple linear models as applied to $V_E$ or $VCO_2$ responses to incremental exercise.
Naimark et al. (1964) and Wasserman and McIlroy (1964) found that changes in the gas exchange ratio, R, reliably reflected arterial blood lactate and bicarbonate concentrations in patients with ischaemic heart disease. In a study of 85 normal but predominantly sedentary subjects, Wasserman et al. (1973) identified the anaerobic threshold as the point at which a non-linear increase in minute ventilation (\(\dot{V}_e\)) occurred, associated with a non-linear increase in carbon-dioxide production (\(\dot{V}CO_2\)), an increase in end tidal CO\(_2\) and an increase in R as work rate increased during an incremental cycle ergometer test. It was suggested that the point at which \(\dot{V}_e\) becomes non-linear is, of itself, an adequate indicator of the anaerobic threshold.

Non-linear increases in \(\dot{V}_e\) and \(\dot{V}CO_2\) together with the occurrence of an abrupt increase in the concentration of oxygen in the expired air (\(F_{\text{EO}_2}\)) were used by Davis et al. (1976) as the criteria for the anaerobic threshold in subjects undergoing three different incremental three different incremental exercise tests. The non-invasive measurements were found to correlate highly (r=0.95) with the onset of a systematic rise in blood lactate concentrations. A test-retest reliability of only 75% was also reported for the non-invasive method, however, suggesting large intra-individual variability.

More recently, abrupt increases in the ventilatory equivalent for oxygen (\(\dot{V}_e.\dot{V}O_2^{-1}\)) have also been used for anaerobic threshold detection. (Davis et al., 1979; Davis et al., 1981; Moritani et al., 1981; Orr et al., 1982). Reinhard et al. (1976), Yoshida et al. (1981) and Caiozzo et al. (1982) have also reported high correlations (of the order of 0.9) between the anaerobic threshold detected by an increase in blood lactate and by non-invasive respiratory function measurements of this type. Caiozzo et al. (1982) report that the threshold detected by the inflection in the \(\dot{V}_e.\dot{V}O_2^{-1}\) response during incremental exercise was the most highly correlated with that determined from blood lactate concentrations and also had the highest test-retest reliability of all the indices examined.

Skinner and McLellan (1980) distinguish two inflection points in the transition between aerobic and anaerobic metabolism. The first (the "aerobic threshold"), it is suggested, occurs when oxygen demand is between 40% and 60% of the subjects maximum oxygen uptake (\(\dot{V}O_2\text{max}\)) and is
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Skinner and McEllan (1980) distinguish two inflection points in the transition between aerobic and anaerobic metabolism. The first (the "aerobic threshold"), it is suggested, occurs when oxygen demand is 40% and 60% of the subjects maximum oxygen uptake (\( VO_2\text{max} \)) and is
been published in full. Unfortunately, the method has only been validated against subjective determinations of the anaerobic threshold. Also, implicit in Orr's method is the assumption that the multiple linear model is intrinsically correct; i.e. that it is an appropriate representation of the experimental data.

In view of the substantial subjectivity in the determination of the anaerobic threshold from respiratory function measurements, it is not surprising that some controversy has arisen concerning the validity of the method. Firstly, the relationship between blood lactate concentrations and ventilatory thresholds have been reconsidered. Hagberg et al (1981) have shown that McArdle's syndrome patients who lack the enzyme myophosphorylase and are therefore incapable of independant glygogenolysis still produce non-linear increases in ventilatory responses to an incremental work test, even though they are not producing lactic acid.

Other research has shown that the anaerobic threshold as determined by respiratory measurements and as determined by blood lactate measurements can be manipulated independantly of one another. The subjects studied by Davis and Gass (1979) performed two successive incremental exercise tests. The second test began while blood lactate levels were still elevated following the first test. The characteristic non-linear increase in ventilatory responses occurred at a the same work rate in the first and second tests regardless of whether blood lactate levels were rising or falling. Hughes et al (1982) found that glycogen depletion brought about a downward shift in the ventilatory threshold and an upward shift lactate threshold, compared with the subjects' normal glycogen status. In the same study, it was found that changing the pedalling rate at which a cycle ergometer incremental work test is conducted can also cause the lactate and ventilatory thresholds to shift independantly.

7:3 PURPOSE OF EXPERIMENT

The purposes of this investigation were:

1: To compare the validity of the commonly used two-part and three-part multiple linear models of respiratory responses to an incremental exercise test with that of a continuous model.

2: To examine statistically the nature of inflection in ventilatory and gas exchange responses to incremental exercise.
7.4.3 Fitting multiple linear models to respiratory responses

Two-part (L2) and three-part (L3) linear models (Figure 7.2) were fitted to the $V_e \cdot VCO_2$ and $V_e \cdot VO_2-1$ data collected from each subject during the running phase of the exercise test. In order to determine the best fitting set of lines, an exhaustive search of all possible combinations of junction points or knots was implemented.

The L2 and L3 models were constrained in three ways. Firstly, all solutions in which the criterion of concavity applied by Orr et al (1982) (See section 7:2) was not true were rejected. Secondly, solutions in which knots lie within three sample periods of the endpoints of the data or within three sample periods of one another were not examined. This ensures that each line-segment of the model contains at least three data points. Thirdly, in order to achieve a model in which the regression lines actually met at the junction points, the second and third line segments of the model were calculated using a modified least squares technique in which the least squares solution is constrained by the requirement that, at the junction point, the equation of the line has the same value as that of the previous line segment (Section 7:5.1).

For each combination of knots, a regression line for the data in each linear section was calculated using the normal least squares method for the first line segment and the constrained method for subsequent line segments. The goodness of fit of the model generated by each combination was obtained by calculating the root mean square (RMS) residual of the raw data about the fitted lines using the formula

$$\text{RMS residual} = \sqrt{\frac{\sum (f(x_i) - y_i)^2}{n}}$$  \hspace{1cm} \text{Eqn. 7.1}

where

- $n$ is the number of raw data points being fitted
- $x_i$, $y_i$ are the values of the $i$th data point.
- $f(x_i)$ is the calculated value of $y$ at $x_i$ from the regression equation.

The best fitting combination was selected as that which minimised the residual.
In particular, the following assumptions, on which non-invasive procedures for the determination of the anaerobic threshold are based were examined:

A1: Multiple linear models of respiratory responses to incremental exercise are a more appropriate representation than a continuous, hyperbolic model.

A2: There are instantaneous inflections or "breakpoints" in the respiratory responses to an incremental exercise test.

A3: There is a statistically significant delay between the inflections in $\text{F}_{\text{E}O_2}$ and $\text{F}_{\text{E}CO_2}$ responses to incremental exercise on the basis of which the "aerobic threshold" and the "anaerobic threshold" can be distinguished.

7:4 MATERIALS AND METHODS

7:4.1 Experimental Protocol

Ten healthy male volunteers whose physical characteristics are shown in Table 7.1 underwent an incremental work test on a motorised treadmill. The running abilities of the subjects were mixed, but all were regularly participating in some form of running training at the time of the experiment.

The incremental test began with the subject walking on the treadmill at 1.6 m.s\(^{-1}\) (3.0 mph, 4.8 kph). The treadmill speed was increased by 0.22 m.s\(^{-1}\) (0.5 mph, 0.8 kph) every minute until the subject became fatigued and could no longer continue. The subjects were not instructed to commence running at a particular speed but in each case the transition occurred at the third or fourth speed increment. The maximum speed attained and the performance time of each subject are also recorded in Table 7.1.

7:4.2 Recording of respiratory function data

Ventilation and gas exchange were recorded continuously, throughout the test using the automated, breath by breath system described in Chapter 6. The following parameters were recorded for later analysis: tidal volume, breathing frequency $f$, $V_e$, $F_{E}O_2$, $F_{E}CO_2$, $VO_2$, $VCO_2$, $V_e$. $VO_2^{-1}$, $R$ and heart rate.

The trend system (Section 6:8) was used to reintegrate the stored raw data over consecutive 20 second sample periods.
7.4.4 Fitting a continuous model to respiratory responses

A continuous model was also applied to the raw data by fitting a curve of the form:

\[ y = a_0 + a_1 \exp(qx) + a_2 \exp(-qx) \]

Eqn. 7.2

with \( a_0, a_1, a_2 \) and q constants (Figure 7.5).

If q is known, then the least squares solution of Eqn.7.2 can be determined by standard methods. With q unknown, a slightly different procedure was adopted. Upon fitting a model of this type, with preselected values of q, to data collected during a preliminary study, it was found that as q increases from near zero, the RMS residual of the least squares solution to the equation falls to a minimum value and then rises again. Therefore the model was fitted to the data collected in this study with a range of values of q. An equal interval search procedure (Bajpai et al., 1974, page 379) for the value of q giving the minimum residual was implemented, and q was determined to a resolution of 0.1. The value of q which yielded the minimum residual, and the associated values of \( a_0, a_1 \) and \( a_2 \) were taken as the least squares solution. RMS residuals calculated using Eqn 7.1 were retained for comparison with those obtained using the multiple linear models.

7.4.5 Comparison of models

Each model was fitted to \( \dot{V_E}, \dot{VCO}_2 \) and \( \dot{V_E}.\dot{VO}_2^{-1} \) data expressed as functions of time and to \( \dot{V_E} \), and \( VCO_2 \) expressed as functions of \( VO_2 \).

The overall goodness of fit of the L2, L3 and C models was compared by Student's t test for correlated means performed on the means of the RMS residuals for all ten subjects.

7.4.6 Examination of inflection points.

\( \dot{V_E}, \dot{VCO}_2, \dot{FE}_2, \dot{FECO}_2 \) and \( \dot{V_E}.\dot{VO}_2^{-1} \) responses measured during the whole of the incremental exercise test were examined statistically with the objective of locating the onset of any significant changes in the response gradient.

It would be possible using numerical differentiation techniques to examine the first derivative of any respiratory response, and hence reveal any changes in the response gradient. This approach is not entirely appropriate in this instance however. The L2 and L3 models specifically imply the existence of instantaneous inflections or "breakpoints", an
assumption which would not be reflected by numerical differentiation. Also, indicators of the statistical significance of any change in the response gradient, rather than the actual values of the derivatives per se, are required.

To this end, two functions, which reflect the first and second derivatives of the original response function were devised for this purpose. The first, termed the "t-function" locates significant changes in the value of a parameter and the second, "termed the "θ function", reflects the change in the gradient of a parameter on the assumption that an instantaneous change in the response gradient has occurred at a given point in time.

7:4.6.1 t-function

A Student's t-test was performed on the means of ten sample points on either side of each sample point in the recorded data. A significant positive or negative value of t thus indicates a significant rise or fall in the value of the parameter concerned. Typically, FeO₂ and Ve.VO₂⁻¹ rise and FeCO₂ falls systematically as work intensity increases during an incremental exercise test to exhaustion. (Figure 7.1). The point beyond which the rise or fall in some parameter P remained significant at the 0.01 probability level (t*(P)) was used as an indicator of the onset of these sudden changes or systematic increases. Figure 7.3 illustrates an example of this analysis applied to a typical set of Ve.VO₂⁻¹-time data recorded during an incremental exercise test.

7:4.6.2 θ function.

The relative amplitude of any change in the response gradient of a parameter, P, at a sample point x was defined as the angle θPₓ between the least squares regression lines fitted to 10 sample points on either side of x (Figure 7.4). Maxima in the function θPₓ were used to indicate peak inflections.
Figure 7.3: The value of the t-function of a parameter P at time x is defined as the Student's t statistic for the difference between the means of two samples on either side of x (Samples A and B). In this instance a sample size of 10 has been selected.
Figure 7.4: The relative magnitude of inflection in a parameter P at time $x$ is defined as the angle $\theta_{P,x}$ between the linear least squares regression lines through samples on either side of the given point, $x$. In this instance a sample size of 10 has been selected.
7:5 MATHEMATICAL PROCEDURES

7:5.1 Calculation of the least squares regression line subject to the constraint that the equation of the line must pass through a given point.

A line segment of the L2 and L3 models is required to be of the form:

\[ y = a_0 + a_1 x \]  \hspace{1cm} \text{Eqn. 7.3}

where \( a_0 \) and \( a_1 \) are constants.

To find the line of best fit to a set of \( n \) data points \( (x_i, y_i) \) the residual sum of squares, \( S \), of the raw data about the fitted line must be minimised, i.e. it is necessary to determine the minimum value of the function

\[ S = \sum_{i=1}^{n} (y_i - a_1 x_i - a_0)^2 \]  \hspace{1cm} \text{Eqn. 7.4}

If the equation of the line is constrained to pass through the point \( (x_c, y_c) \) then the least squares solution to equation 7.3 must be found subject to the constraint

\[ y_c = a_0 + a_1 x_c \]  \hspace{1cm} \text{Eqn. 7.5}

Combining the constraint with equation 7.3 gives

\[ y = a_1 (x - x_c) + y_c \]  \hspace{1cm} \text{Eqn. 7.6}

The best fit is found by minimising the residual

\[ S = \sum_{i=1}^{n} (y_i - a_1 (x_i - x_c) - y_c)^2 \]  \hspace{1cm} \text{Eqn. 7.7}

Setting \( q_i = x_i - x_c \) and taking the partial derivative of \( S \) with respect to \( a_1 \) gives

\[ \frac{\partial S}{\partial a_1} = 2 \sum_{i=1}^{n} (y_i - a_1 q_i - y_c)(-q_i) \]  \hspace{1cm} \text{Eqn. 7.8}
Thus $S$ is a minimum when

$$\sum_{i=1}^{n} (y_i - a_1 q_i - y_c)(-q_i) = 0$$

Eqn. 7.9

Expanding gives

$$-\sum_{i=1}^{n} y_i q_i + a_1 \sum_{i=1}^{n} q_i^2 + \sum_{i=1}^{n} q_i y_c = 0$$

Eqn. 7.10

or

$$a_1 = \frac{\sum y_i q_i - q_i y_c}{\sum q_i^2}$$

Eqn. 7.11

The value of $a_0$ when $S$ is minimum is then determined by substitution in the constraint equation, Eqn.7.5. A BASIC computer programme to calculate best fitting two-part and three-part linear models is listed in Appendix A.

7.5.2 Least squares fit to the continuous model

The continuous model takes the form of Eqn.7.2. Assuming $q$ is known and all values of $x$ have been multiplied by $q$, the residual sum of squares to be minimised is

$$S = (y_i - a_0 - a_1 \exp(x_i) - a_2 \exp(-x_i))^2$$

Eqn. 7.12

This is a minimum when the partial derivatives of $S$ with respect to $a_0$, $a_1$ and $a_2$ are zero.

i.e.

$$\sum (y_i - a_0 - a_1 \exp(x_i) - a_2 \exp(-x_i))(-1) = 0$$

$$\sum (y_i - a_0 - a_1 \exp(x_i) - a_2 \exp(-x_i))(-\exp(x_i)) = 0$$

$$\sum (y_i - a_0 - a_1 \exp(x_i) - a_2 \exp(-x_i))(-\exp(-x_i)) = 0$$

Eqn. 7.13
Expanding and expressing in matrix form gives the following simultaneous equations:

\[
\begin{bmatrix}
  n & \Sigma \exp(x_i) & \Sigma \exp(-x_i) \\
  \Sigma \exp(x_i) & \Sigma (\exp(x_i))^2 & \Sigma \exp(x_i)\exp(-x_i) \\
  \Sigma \exp(-x_i) & \Sigma \exp(x_i)\exp(-x_i) & \Sigma (\exp(-x_i))^2
\end{bmatrix} \begin{bmatrix}
a_0 \\
a_1 \\
a_2
\end{bmatrix} = \begin{bmatrix}
\Sigma y \\
\Sigma y_1\exp(x_i) \\
\Sigma y_1\exp(-x_i)
\end{bmatrix}
\]  

Eqn. 7.14

These equations can be solved by standard methods to give \(a_0\), \(a_1\) and \(a_2\). In this case, the equations were solved using a Gaussian elimination algorithm (Bajpai et al., 1974, pp 223-237). A BASIC programme to execute these procedures is listed in Appendix B.
RESULTS

The highest value of oxygen uptake registered during any 20 second sample period of the incremental exercise test to exhaustion was recorded as the subject's maximum oxygen uptake (\( \text{VO}_{2\text{max}} \)). These values of \( \text{VO}_{2\text{max}} \) are recorded in Table 7.1, as are the total duration of the test (performance time) and the treadmill speed attained.

7:6.1 Comparison of multiple linear and continuous models

Figure 7.5 shows examples of the L2 and L3 and C models fitted to \( \dot{V}_E \) data expressed as a function of \( \dot{V}_O_2 \). Table 2 shows the mean RMS residuals of the L2, L3 and C models applied to \( \dot{V}_E \), \( \dot{V}_{CO_2} \) and \( \dot{V}_E \dot{V}_{O_2}^{-1} \) responses of each of the 10 subjects.

In every case examined, the L3 model provided a significantly better fit (i.e. gave a smaller residual) than the L2 model.

When \( V_E \) and \( V_{CO_2} \) were expressed as functions of \( \dot{V}_O_2 \) there was no significant difference between the goodness of fit of the L3 and C models. This was also the case when the L3 and C models were fitted to \( \dot{V}_{CO_2} \) data expressed as a function of time. However, the continuous model gave a better fit (\( p < 0.01 \)) than either the L2 or L3 when \( \dot{V}_E \) and \( \dot{V}_E \dot{V}_{O_2}^{-1} \) were expressed as functions of time.

These results fail to support the assumption (A1) that multiple linear models of respiratory responses to incremental exercise are a more appropriate representation than a continuous, hyperbolic model.

7:6.2 Analysis of inflection functions

Figure 7.6 shows the results of the application of the t-function applied to the \( \dot{V}_E \dot{V}_{O_2}^{-1} \) response of a single subject. The transition from one response gradient to another is clear, and appears to occur over a period of time. In this example, which is typical, the transition period recorded by the t-function extends over some four minutes, and hence some four different running speeds. Part of this apparent transition period is due to the smoothing effect of the function. The actual inflection point was therefore recorded as the point at which the difference between two successive sets of 10 samples exceeded a value of 2.78. This level is equivalent to a probability of less than 0.01 that the difference between the samples is due to chance.

A significant fall in \( F_{ECO_2} \) and significant rises in \( F_{EO_2} \) and \( \dot{V}_E \dot{V}_{O_2}^{-1} \) were detected using this method. The significant drop in \( F_{ECO_2} \) occurred, on
### Table 7.1: Subjects' physical characteristics and performance in an incremental treadmill work test.

<table>
<thead>
<tr>
<th>Sub.</th>
<th>Age</th>
<th>Mass</th>
<th>Perf. time</th>
<th>Speed</th>
<th>VO2max attained</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>yrs</td>
<td>kg</td>
<td>min</td>
<td>kph</td>
<td>ml/kg/min</td>
</tr>
<tr>
<td>1 :</td>
<td>36</td>
<td>72.10</td>
<td>15.0</td>
<td>16.8</td>
<td>60.5</td>
</tr>
<tr>
<td>2 :</td>
<td>30</td>
<td>68.60</td>
<td>17.0</td>
<td>18.4</td>
<td>56.0</td>
</tr>
<tr>
<td>3 :</td>
<td>28</td>
<td>70.65</td>
<td>14.8</td>
<td>16.8</td>
<td>58.0</td>
</tr>
<tr>
<td>4 :</td>
<td>34</td>
<td>82.00</td>
<td>12.8</td>
<td>15.2</td>
<td>49.7</td>
</tr>
<tr>
<td>5 :</td>
<td>26</td>
<td>75.25</td>
<td>19.0</td>
<td>20.0</td>
<td>58.2</td>
</tr>
<tr>
<td>6 :</td>
<td>44</td>
<td>74.35</td>
<td>16.0</td>
<td>17.6</td>
<td>53.4</td>
</tr>
<tr>
<td>7 :</td>
<td>26</td>
<td>82.10</td>
<td>15.0</td>
<td>16.8</td>
<td>55.9</td>
</tr>
<tr>
<td>8 :</td>
<td>26</td>
<td>64.15</td>
<td>16.0</td>
<td>17.6</td>
<td>57.5</td>
</tr>
<tr>
<td>9 :</td>
<td>25</td>
<td>74.95</td>
<td>20.0</td>
<td>20.8</td>
<td>71.0</td>
</tr>
<tr>
<td>10 :</td>
<td>22</td>
<td>65.95</td>
<td>19.0</td>
<td>20.0</td>
<td>64.2</td>
</tr>
<tr>
<td>MEAN</td>
<td></td>
<td>73.01</td>
<td>16.4</td>
<td>18.0</td>
<td>58.5</td>
</tr>
<tr>
<td>s.d.</td>
<td></td>
<td>6.04</td>
<td>2.4</td>
<td>1.78</td>
<td>6.3</td>
</tr>
</tbody>
</table>
TABLE 7.2: RMS residuals for each of three models fitted to $\dot{V}_E$, $\dot{V}CO_2$ and $\dot{V}_E \cdot \dot{V}O_2^{-1}$, data plotted as functions of (a) time and (b) $\dot{V}O_2$

Means for ten subjects ± standard deviations.

### (a)

<table>
<thead>
<tr>
<th>MODEL</th>
<th>L2</th>
<th>L3</th>
<th>C</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\dot{V}_E$</td>
<td>6.01</td>
<td>4.89</td>
<td>4.24*</td>
</tr>
<tr>
<td></td>
<td>+1.61</td>
<td>+1.47</td>
<td>+1.27</td>
</tr>
<tr>
<td>$\dot{V}CO_2$</td>
<td>0.194</td>
<td>0.140</td>
<td>0.140</td>
</tr>
<tr>
<td></td>
<td>+0.051</td>
<td>+0.051</td>
<td>+0.058</td>
</tr>
<tr>
<td>$\dot{V}_E \cdot \dot{V}O_2^{-1}$</td>
<td>1.97</td>
<td>1.17</td>
<td>1.03*</td>
</tr>
<tr>
<td></td>
<td>+0.36</td>
<td>+0.280</td>
<td>+0.31</td>
</tr>
</tbody>
</table>

### (b)

<table>
<thead>
<tr>
<th>MODEL</th>
<th>L2</th>
<th>L3</th>
<th>C</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\dot{V}_E$</td>
<td>4.97</td>
<td>4.62</td>
<td>4.61</td>
</tr>
<tr>
<td></td>
<td>+1.68</td>
<td>+1.19</td>
<td>+1.83</td>
</tr>
<tr>
<td>$\dot{V}CO_2$</td>
<td>0.110</td>
<td>0.102</td>
<td>0.106</td>
</tr>
<tr>
<td></td>
<td>+0.034</td>
<td>+0.035</td>
<td>+0.035</td>
</tr>
</tbody>
</table>

*Students t test for correlated means shows a significant difference (P<0.01) between L2 and C.

In each case, the residuals for L3 are significantly lower than for L2 (P<0.01).
TABLE 7.3: Relative oxygen uptake at inflection points in respiratory responses.

Mean, standard deviation and range for ten subjects, expressed as % \( VO_2 \) max.

<table>
<thead>
<tr>
<th>Inflection parameter</th>
<th>Mean ( %VO_2 ) max</th>
<th>s.d.</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 L3 ( VO_2 ), knot 1</td>
<td>61.7</td>
<td>6.5</td>
<td>44.8 - 70.5</td>
</tr>
<tr>
<td>2 L3 ( VO_2 ), knot 2</td>
<td>71.1</td>
<td>6.7</td>
<td>63.0 - 84.3</td>
</tr>
<tr>
<td>3 t* ( F_EO_2 )</td>
<td>73.0</td>
<td>6.2</td>
<td>64.1 - 86.7</td>
</tr>
<tr>
<td>4 t* ( F_ECO_2 )</td>
<td>73.9</td>
<td>8.3</td>
<td>62.1 - 94.1</td>
</tr>
<tr>
<td>5 t* ( \dot{V}_E.\dot{VO}_2^{-1} )</td>
<td>72.6</td>
<td>7.9</td>
<td>64.1 - 86.7</td>
</tr>
<tr>
<td>6 ( \dot{V}_E ) (peak)</td>
<td>70.2</td>
<td>6.0</td>
<td>63.0 - 84.2</td>
</tr>
<tr>
<td>7 ( \dot{V}CO_2 ) (peak)</td>
<td>68.7</td>
<td>7.1</td>
<td>63.0 - 84.2</td>
</tr>
</tbody>
</table>
TABLE 7.4: Intercorrelations between inflection points.

Key:

<table>
<thead>
<tr>
<th>No.</th>
<th>Inflection parameter</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>L3 ( \dot{V} \text{CO}_2 ), knot 1</td>
</tr>
<tr>
<td>2</td>
<td>L3 ( \dot{V} \text{CO}_2 ), knot 2</td>
</tr>
<tr>
<td>3</td>
<td>( t \times \dot{E} \text{O}_2 )</td>
</tr>
<tr>
<td>4</td>
<td>( t \times \dot{E} \text{CO}_2 )</td>
</tr>
<tr>
<td>5</td>
<td>( t \times \dot{E} \cdot \dot{V} \text{O}_2-1 )</td>
</tr>
<tr>
<td>6</td>
<td>( \dot{V} \text{E} ) (peak)</td>
</tr>
<tr>
<td>7</td>
<td>( \dot{V} \text{CO}_2 ) (peak)</td>
</tr>
</tbody>
</table>

Correlation matrix:

<table>
<thead>
<tr>
<th></th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td></td>
<td>0.220</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>0.220</td>
<td></td>
<td>0.372</td>
<td>0.513</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>0.372</td>
<td>0.513</td>
<td></td>
<td>0.196</td>
<td>0.412</td>
<td>0.641</td>
</tr>
<tr>
<td>4</td>
<td>0.196</td>
<td>0.412</td>
<td>0.641</td>
<td></td>
<td>0.360</td>
<td>0.985</td>
</tr>
<tr>
<td>5</td>
<td>0.360</td>
<td>0.985</td>
<td>0.706</td>
<td>0.810</td>
<td>0.419</td>
<td>0.769</td>
</tr>
<tr>
<td>6</td>
<td>0.419</td>
<td>0.810</td>
<td>0.769</td>
<td>0.619</td>
<td>0.716</td>
<td></td>
</tr>
<tr>
<td>7</td>
<td>0.189</td>
<td>0.783</td>
<td>0.701</td>
<td>0.698</td>
<td>0.646</td>
<td>0.845</td>
</tr>
</tbody>
</table>

Correlations greater than 0.632 are significant at the 0.05 level.
Correlations greater than 0.765 are significant at the 0.01 level.
Figure 7.5: Example of $V_e$ data from this study with best fitting models.

(a) 2 part multiple linear model
(b) 3 part multiple linear model
(c) continuous model
Figure 7.6: Successive Student's t-tests used to determine the statistical significance of any inflection in example $\dot{V}_E \cdot \dot{V}_{O_2}^{-1}$ data from this study. The point of inflection is determined as the point where t rises above the critical value for $p=0.01$.

(A) original data
(B) plot of t-function for adjacent groups of two samples
Figure 7.7: Magnitude of apparent inflection in example $V_e$ response from this study determined by the $\theta$-function.

(A) original data

(B) $\theta$-function for adjacent groups of 10 samples.
average, just after a significant rise in \( F_EO_2 \) (Table 7.2). The mean difference between the two samples, equivalent to a change in the relative work rate of 1% of \( VO_2_{max} \) is within the error of the measurement system and does not approach statistical significance (Student's \( t \) for correlated means = 0.48 WITH 9 d.f.).

Therefore, with the methods used in this investigation, a delay between inflections in \( F_EO_2 \) and \( F_ECO_2 \) responses was not detected and assumption A3 was not supported.

Figure 7.7 shows the results of the application of the \( B \)-function applied to the \( V_E \) responses of a single subject. A peak in the function occurs, in this case, at an oxygen uptake equivalent to 63% of \( VO_2_{max} \). This maximum in the \( B \)-function can also be used to indicate the point of most significant inflection. Table 7.3 shows the relative work-rates at which peaks in the \( B \)-function occurred in the \( V_E \) and \( VCO_2 \) responses of all the subjects studied. This table shows clearly that the changes in the gradient of \( V_E \) and \( VCO_2 \) responses were closely linked (\( r = 0.85 \)). In eight of the ten subjects, the peaks in \( V_E \) and \( VCO_2 \) occurred during the same 20 second sample period, as indicated by identical relative oxygen uptake values.

Table 7.3 shows the means, standard deviations and ranges of the seven selected including the positions of the knots in the L3 model of \( VCO_2 \) responses. Table 7.4 shows the intercorrelations between these 7 inflection indices. The positions of the second knot in the L3 model of \( VCO_2 \) correlated highly (\( r = 0.810 \)) with peaks in the \( B \)-functions of \( V_E \) responses. The positions of inflections in \( V_E \), \( VCO_2 \), \( F_EO_2 \), \( F_ECO_2 \) and \( V_E VO_2^{-1} \) were all highly intercorrelated.

**7:7 DISCUSSION**

Recent evidence calls into question the use of inflections in respiratory function measurements as a means of determining the onset of anaerobic metabolism. On the other hand, these methods are widely used and published and are lent some face validity by experiments in which a high correlation is found between ventilatory threshold measurements and the appearance of lactic acid in the blood. Little attention has been paid to the actual methods by which the points of inflection are determined. Notably, it has not been established that the breakpoints, i.e. instantaneous changes in response gradient, are statistically recordable and repeatable events, and at no time has the suitability of the multiple
linear model, as opposed to a continuous model, been questioned.

The application of the t-, and s-functions, which behave in a similar manner to first and second derivatives, to the respiratory responses recorded during an incremental exercise test clearly reveal significant changes in the responses and in the response gradients. It is clear from the high intercorrelations between inflections in $V_e$, $VCO_2$, $FeO_2$, $FeCO_2$ and $Ve.VO_2^{-1}$ responses that changes in the response gradient of the different respiratory function parameters occurred more or less synchronously in each subject. The observation that the changes occurred synchronously does not necessarily mean that they occurred instantaneously however.

Since a 20s sample period was selected when treating the data collected in this study, any analyses of "instantaneous" responses are limited by the resolution of the system. In this case, any change that takes less than 20 seconds to complete cannot be distinguished from an instantaneous response. The changes in response gradient detected by the use of s- and t-functions appear to have occurred over a period of several minutes, however (Figures 7.6, 7.7). The duration of these transition periods is partly determined by the s- and t-functions themselves, since they have a filtering effect on the raw data. This smoothing is essential due to the presence of "noise" in the recorded respiratory function data. In addition to any random error due to the measurement system itself, there are apparently random variations in tidal volume and breath duration from breath to breath. The s- and t-functions filter out much of this variation by taking the mean value or mean slope over a number of points. Since each function uses 10 data points on either side of the particular point being examined, a true instantaneous change would be made to appear to extend over 9 time periods (i.e. 3 minutes) in the case of the t-function. In the case of the s-function, the instantaneous transition from one response gradient to another would be reflected in the function for 20 sample periods. The position of an instantaneous inflection is accurately determined by the peak value of s, however. The overall effect of this behaviour is that if any instantaneous changes in responses do occur, they will either be masked by the natural noise in the subjects' breathing pattern or by the filtering procedure used to eliminate the noise. Since this will probably be true for any expired air measurement technique, models of respiratory responses that depend on the detection of these instantaneous changes are unlikely to succeed.
The use of discontinuous models to describe the respiratory responses to incremental exercise presents the experimenter with a number of further difficulties:

Firstly, the assumption of discontinuities or "breakpoints" prevents a sensible analysis of the first and second derivatives of the response time-series which would otherwise permit direct measurement of changes in the response gradient. This problem is exaggerated by measurement systems in which data is collected intermittently (for example for 15 seconds during each minute of the exercise test) rather than continuously, since intermittent sampling may tend to introduce apparent discontinuities into the observations.

Secondly, there is a danger of "over-fitting". Mathematically it is possible to fit any model to any set of data provided that the data has at least one more point than the model has independent parameters. The fewer data points there are per model parameter, the more likely it is that a good fit will be found. When using multiple regression models it is generally recommended that there should be 10 or more raw data points for each independent variable. The L3 model, with 6 independent variables therefore requires 60 data points if there is to be no tendency to overfit. Similarly, the L2 and C models require of the order of 40 data points. Orr et al (1982) record that

"... We have generally not been successful in fitting lines to data sets containing fewer than 20 points. Most data sets analysed have contained approximately 50 points with some containing up 80 data points."

In this case, the sample size quoted for "most data sets" are satisfactory. The computer algorithm used by Orr et al was unable to fit a sensible multiple linear model to sets of data containing less than 20 data items. Unfortunately, a human operator is quite capable of making a subjective fit to fewer data points than are necessary for a statistically valid fit.

Thirdly, and perhaps most importantly, the L2 and L3 models are based on the assumption that there are discontinuities or "breakpoints" in the respiratory responses. Ipso facto, the inflections can always be found when subjective judgements are made.

It is not possible, with the methods used in this investigation, to
distinguish between an "instantaneous" change in respiratory responses and a continuous transition from one response gradient to another. It is only possible to determine that a change in the response gradient has occurred and to indicate the point at which the rate of change from one gradient to another was greatest. It is noted that response rates at the level of cellular respiration cannot be directly reflected in the ventilatory and gas exchange measurements made at the mouth due to the buffering effect of the cardiovascular and respiratory systems and due to the cyclical nature of breathing itself. Thus the concept of an "instantaneous" change in the response gradient of a respiratory function may not be a useful one.

Also, the results of this study suggest that the multiple linear model may not be the most appropriate representation of respiratory responses to incremental exercise. In no case did the discontinuous three-part linear regression model with 6 independant variables provide a better fit to the respiratory responses examined than a continuous hyperbolic model with only 4 independant variables. In two instances, when \( \dot{V}_E \)-time and \( \dot{V}CO_2 \)-time data were considered, the continuous model provided a significantly better fit than the L3 model. In a preliminary examination of the data presented here (Shorten & Williams, 1982) the three part linear model was found not to improve significantly on an even simpler, quadratic model with only three parameters.

Undoubtedly, changes in the response gradients do occur but it may be more appropriate to describe these changes in terms of a continuous transition rather than a distinct breakpoint.

The results of this study bring into question the validity of the methods used to detect the anaerobic threshold by analysis of respiratory responses alone. The derivation of a more appropriate model for the detection of the anaerobic threshold will depend on the substantiation of a direct relationship between metabolic events in muscle and changes in respiratory events. Recent studies (Hagberg, 1981; Davis et al, 1981; Hughes et al, 1982) suggest that the relationship between lactic acidosis and ventilatory thresholds is not consistent and the development of such a model in a simple form is therefore unlikely.

On the basis of these results, the detection of the onset of lactic acidosis should not be based upon the analysis of inflections in respiratory responses alone.

The direct measurement of blood lactate would appear to be a more appropriate method of detecting anaerobiosis, since lactic acid is a direct product of anaerobic metabolism. Blood lactate levels are determined by the
sum of lactate production and lactate removal in the muscle. Radiotracer studies of lactate kinetics in animals (Eldridge, 1974; Donovan and Brooks, 1983) have demonstrated that lactate production in muscle cannot be predicted from blood lactate concentrations. Blood lactate measurements cannot therefore be used to quantify the extent of anaerobic metabolism in muscle tissue. However, the presence of significant levels of lactate in the blood does provide direct evidence of a contribution to energy supply from sources which are not measurable by indirect calorimetry.
Chapter 8

ELASTIC COMPLIANCE OF THE HUMAN KNEE EXTENSORS IN VIVO
SUMMARY

The storage and subsequent recovery of strain energy in the elastic components of muscle may contribute to the enhancement of the efficiency and effectiveness of performance.

In order to study these effects in detail, it is necessary to establish a method for determining the elastic properties of a muscle group. Of the methods described in the literature, none actually determine an elastic compliance-force function for a particular muscle group in the lower body. The techniques which have been applied to the upper body satisfactorily appear difficult to adapt to the lower body.

A method is developed based on measurement of the rise in tension at the onset of an isometric muscle contraction. Studies on isolated muscle have shown that the force-velocity function of such a contraction is not the same as that in an isotonic contraction. Evidence is presented that the two force-velocity functions are related in some undefined way. This is sufficient to eliminate the force-velocity relationship from the compliance calculations, provided two or more contractions with differing added series compliances are performed.

Results are presented for 10 athletic male subjects. Mean angular equivalent compliance (G, N·m⁻¹) and knee extension moment (Mk, Nm) were related by the equation

\[ G = 0.021 \exp(-0.0133M_k) \]

During a typical isometric contraction, 70 J of strain energy were stored in the elastic component and the SEC was extended by 10% of the muscle length. This energy storage is significant when compared with typical levels of segmental potential and kinetic energy during running.
INTRODUCTION

Measurements of the mechanical efficiency of running based on potential and kinetic energy changes alone appear to be greater than the efficiency of transformation of energy substrates into mechanical energy in muscle. This is the case even when all possible transfers of energy between observable segmental kinetic and potential energy are taken into account.

In order to reconcile these observations with the laws of thermodynamics, it is necessary to determine the source of the additional mechanical energy that appears over and above that released by muscles. For an athlete running on a treadmill at 4.5 m.s\(^{-1}\) these "additional" energy changes amount to about 100 J.s\(^{-1}\).

Some care has been taken to eliminate possible energy inputs from anaerobic metabolism and while they may still make a contribution, it is unlikely to be of a high enough magnitude to account for the observed discrepancy.

The external work done by gravity is accounted for when between segment energy transfers are eliminated from the work total.

A further possibility worthy of consideration is that some transfer of energy within and between body segments has not been accounted for in energy analyses. In particular, the storage and recovery of energy in stretched elastic components of muscle and connective tissue may act in this way.

There is substantial evidence to suggest that muscle behaves elastically under certain conditions and that the overall mechanical efficiency of the body may be enhanced as a consequence. Previously, elastic energy storage has not been directly included in energy analyses of whole body movements. Such an analysis requires a knowledge of the elastic properties of muscle which can be used to calculate the strain energy component.

The purpose of the following investigation was therefore to determine the elastic properties of a specific human muscle group, the knee extensors, with a view to estimating the magnitude of elastic energy storage and recovery during running.
8:2 REVIEW OF LITERATURE

8:2.1 Muscle elasticity

During eccentric contraction of skeletal muscle, the contractile proteins are developing tension but at the same time are being stretched by an external force. This force therefore does work on the contracting muscle. The work done may, in theory, be stored temporarily in any elastic materials in the muscle and connective tissues that are stretched by the force. Assuming low hysteresis this stored energy may be recovered when the external force ceases to act and the stretched materials recoil, exerting a contractile force which may assist a subsequent active concentric contraction of the muscle.

Fenn (1930, 1957) has suggested that the usefulness of energy storage in muscle would be negligible due to the inability of muscle to retain the energy without continuous contraction. Similarly Elftman (1944, 1966) was of the opinion that any energy stored during an eccentric contraction would be dissipated as heat and would therefore be of no value. More recent experimental evidence conflicts with these views however.

8:2.2 Elasticity and muscular performance enhancement

Many of the experimental studies of elastic energy storage during human movement have compared performance in standing vertical jumps performed with and without a preliminary counter-movement. Cavagna et al (1971) found that the effect of the counter-movement was to increase the 'power output' of the subsequent jump. Asmussen and Bonde-Petersen (1974a) reported increased performances with the countermovement but also studied the variation in performance of subjects jumping down onto a force-platform from various heights and immediately jumping vertically upwards. The increased performance in the drop jumps was attributed to the reutilisation of strain energy stored in the elastic components as the muscle undergoes a forceful stretch during the landing.

In addition to stretching the elastic components of muscle, a counter movement may also initiate a stretch reflex which increases the myoelectric stimulus to the muscle. In some cases this effect may also contribute to enhanced performance (Bosco et al, 1982a).

The effects of preliminary movements on vertical jump performance are qualitatively similar to the reported effects of pre-stretching on isolated muscle preparations. Cavagna et al (1975) demonstrated that the force production or power output of frog sartorius muscle is enhanced if
contraction immediately follows an active stretch. Stretching in this case has the effect of increasing the work done during the subsequent contraction. This phenomenon is attributed to elastic energy storage, but may also be due, in part, to a direct effect on the contractile mechanism (Cavagna et al., 1968; Bergel et al., 1972).

Other evidence suggests that the elastic behaviour of muscle may also enhance the efficiency with which some movements are performed by allowing energy to be reutilised rather than dissipated as heat (Asmussen & Bonde-Petersen, 1974b). Changing the rhythm with which repetitive rebound jumps are performed and hence the extent to which stored elastic energy can be regained, has been shown to substantially reduce the oxygen cost of the activity by 22% (Thys et al., 1972; Asmussen and Bonde-Petersen, 1974a).

It is also interesting to note that the oxygen consumption of kangaroos is fairly constant over a wide range of hopping speeds (Dawson & Taylor, 1973); a phenomenon which seems to be accounted for by the animal's high capacity for elastic energy storage and transfer (Alexander & Vernon, 1975; Morgan et al., 1978).

8:2.3 Contractile behaviour of muscle

Fenn & Marsh (1935) and Hill (1938) described the mechanical behaviour of muscle in terms of three functionally distinct components. This three component model has formed the basis of many of the later studies concerned with describing the mechanical properties of muscle and is still commonly used when a simple description of muscle mechanical behaviour is required. The three components of the model are a contractile component (CC), an elastic component in series with the contractile component (Series elastic component, SEC) and an elastic component in parallel with the contractile component (Parallel elastic component, PEC).

The contractile component represents the active force and shortening generating processes within the muscle, described by characteristic force-velocity (Hill, 1938) and force-length (Gordon et al., 1966) relationships. Hill's (1938) characteristic equation for the force-velocity function of muscle describes this relationship in the hyperbolic form:

\[(P + a)(v + b) = \text{constant} \tag{Eqn 8.1}\]

where

- \(P\) = force generated by the CC
- \(v\) = velocity of contraction of the CC
a = a constant with the dimensions of force
b = a constant with the dimensions of velocity

The constant value of this term can be determined by considering the isometric case where \( v \) is zero and \( P = F_0 \), the maximum (isometric) tension that the contractile component can generate under the prevailing conditions of muscle length, stimulation, temperature, etc.

Another equation, originally proposed by Fenn & Marsh (1935) but later modified by Aubert (1956) describes the same relationship in exponential form:

\[
v = B \log\left(\frac{A}{(P+F)}\right) \quad \text{Eqn 8.2}
\]

or

\[
P = A \exp\left(-\frac{v}{B}\right) - F \quad \text{Eqn 8.3}
\]

where \( A \) (which is equal to \( F_0 + F \)) and \( B \) are constants with the same dimensions as \( a \) and \( b \) of Hill's equation. \( F \) represents a small internal frictional or compressional force tending to resist shortening.

Both Hill's and Aubert's equations are empirical in nature and cannot be considered to structurally model the muscle even though, in general, the observed properties of muscle are well described by either set of equations. In practice the values of the constants are derived by fitting experimental data to either equation. The models apply only to concentric contractions at a given muscle length. The constants \( a \), \( b \), \( A \), \( B \) and \( F \) cannot reasonably be interpreted in terms of any directly measurable muscle parameter although \( a \) would appear to relate to the cross-sectional area of the muscle and \( b \) to its length (Stern, 1974).

Hill's equation has been shown to adequately describe the force-velocity behaviour, at a given length, of a wide variety of mammalian and vertebrate muscles. (e.g. Ritchie, 1964; Close & Hoh, 1967; Thompson, 1961; Bahler et al, 1968; Katz, 1939; Hill, 1964). There is also evidence that Hill's equation is obeyed under conditions of variable force and velocity (Hill, 1938; Jewell & Wilkie, 1958; Parmeley et al, 1970) and when the muscle is only partially activated (Mashima et al, 1972). However, some modification of the simple form of the equation has been necessary to describe variations in contraction velocity over a range of muscle lengths. (See below).

Of particular importance for the purposes of this study are the results of Ralston et al (1949), Wilkie (1950), Komi (1973) and Thorstensson et al (1976) which show that Hill's equation can adequately
describe voluntary contractions of human muscle in vivo.

In addition to the characteristic force-velocity relationship, the contractile force of maximally activated muscle shows, independently of contraction velocity, systematic variation with muscle length. The relationship between maximum isometric tension and length is approximately parabolic in form with peak isometric tension occurring at around the "resting" length of the muscle. At extreme extension and extreme contraction, the muscle is unable to generate any tension. According to the sliding filament model of muscle contraction (Huxley, 1957) maximum isometric tension is a function of the overlap between the actin and myosin myofibrils, since this overlap determines the number of cross-bridges that can make contact between the filaments and contribute to tension generation. Gordon et al (1966) have confirmed the relationship between the tension-length curve and the geometry of the actin and myosin filaments.

The same form of tension-length relationship has been demonstrated in a variety of muscles including human muscle in vivo (Ralston et al, 1947).

Studies concerned with the force-velocity relationship of muscle have in general been performed under controlled, near isometric conditions: muscle lengths are usually held near the optimum (l₀) and not allowed to vary significantly during the experiment. Similarly, force-length determinations are based on isometric contractions where any potential influence of velocity variations is eliminated. Clearly, however there is likely to be interaction between the two phenomena.

Abbot and Wilkie (1953) demonstrated that velocity changes during a contraction in which the muscle was allowed to shorten in the range below l₀ could only be described by Hill's equation if the instantaneous value of maximum tension as a function of length, P₀(l) was substituted for the constant P₀ (This form may be described as the "length-corrected form" of Hill's equation).

Gordon et al (1966) further observed that the decline in velocity at lengths below l₀ was greater than would be suggested by Abbott and Wilkie's (1953) correction and that Hill's equation would only fit the force-velocity behaviour at lengths greater than l₀ if the constant a was allowed to decrease in proportion to P₀(l) (i.e. the ratio a/P₀, which completely determines the shape of the force-velocity function described by Hill's equation, is held constant). Similarly, Bornhorst and Minardi, (1970) suggest that a should be allowed to increase systematically as the
muscle shortens below 10.

It is also possible that the previous force history of a contraction may have an effect on a muscle's subsequent behaviour (Jewell and Wilkie, 1958). Bahler et al (1968), for example, suggest that if the initial length of the muscle is varied, then the velocity of contraction is a function, also, of the duration of stimulation.

While Hill's equation does not precisely describe the force-velocity behaviour of muscle under continuously varying conditions, Stern (1975) reports that the effects of the corrections proposed by Bornhorst and Minardi (1970) and Bahler et al (1968) are "quantitatively very small" and "of no significance with regard to the general pattern of results or their interpretation."

9:2.4 Series elasticity

The series elastic component of the three component model represents the mechanical properties of non-rigid structures in series with the contractile apparatus. These elastic structures include both intra and extra-sarcomeric compliances within the myofibrils and that of the tendon and connective tissues if whole in situ muscle systems are being modelled. The tendon and connective tissue represent a 'passive' series elastic component whose properties are largely unaffected by the active contractile behaviour of the muscle.

Although often modelled as one or two elastic structures, the observable series elasticity of a muscle-tendon system is anatomically distributed. A major part of this elasticity resides in the tendon which may be considered a 'passive' elastic component since the tendon stiffness is largely independant of the active contractile components of the system. Mammalian tendon, if assumed perfectly elastic, has a Youngs modulus of around 1.2 GN/m². Morgan (1977) found cat soleus tendon to have an average compliance of 0.059 mm/N for example (Morgan, 1977). As a result, at a given muscle tension, the stretch of the tendon was approximately equal to that in the muscle fibres. This is also true of frog sartorius, where approximately half of the series compliance lies in the pelvic tendon and the rest is distributed among the muscle fibres. (Jewell and Wilkie, 1958). In contrast, the stretch in Wallaby tendon may be eight times that of the associated muscle fibres (Morgan et al, 1978). In intact human muscle in vivo (Ralston et al, 1947), the contribution of tendons to series elasticity has been shown to vary with the gross anatomy of the muscle.
When the forearm flexors are stretched, the passive tension in tendon is sufficient to prevent the occurrence of a minimum point in the overall (active + passive) force-length relationship. In the case of the pectoralis major however, the series elasticity of the flat, broad tendon is small.

Some of the observed series elasticity of muscle is distributed within the contractile apparatus itself. When a previously resting muscle becomes active, the tension must initially be borne by the contractile proteins. It is unlikely that these structures are ideally rigid. In particular, the junction of the oppositely polarised actin filaments in the Z-disc is a likely region of compliance (Jewell & Wilkie, 1958). In addition, the coiled protein filaments of actin and myosin themselves may be compliant, as may the cross-bridge and cross-bridge attachments themselves.

The observation that the character of the series elastic component is independent of muscle length (Jewell and Wilkie, 1958) suggests that the elastic properties are extrasarcomeric (i.e. are to be found in a region of the contractile apparatus not directly involved in the production of tension). Such regions include the Z-disks (Szent Gyorgi, 1953) and the linkages between the terminal sarcomere and the tendon (Bendall, 1969). It is now accepted, however, that the observed elasticity is not entirely extrasarcomeric but that it can be attributed, at least in part, to the cross-bridges (Huxley & Simmons, 1971; Fung, 1970). Huxley and Simmons (1971) have calculated that each cross-bridge may stretch 15nm. Jewell and Wilkie (1958) found the stretch to be less than this whereas Rack and Westbury (1974) determined a higher value of 30nm. It has been suggested that intrasarcomeric compliance acts as a passive recovery mechanism to balance active processes during an isometric contractions (Schreiner, 1976).

8:2.5 Force dependency of series elasticity

In intact muscle, increases in total force production are achieved by increasing motor unit recruitment. If each motor unit or fibre is considered to possess an independent SEC which is only effectively engaged when the fibre is recruited, then an ancillary effect of increased recruitment is to increase the number of SECs lying in parallel with one another. Thus it would be expected that the observed series stiffness of a whole muscle will be a function of the tension generated, with the value at zero force equal to the "passive" stiffness. This hypothesis is supported by observations on series stiffness:force relationships in isolated heart muscle Sonnenblick (1964), cat soleus (Morgan et al., 1977) and rat soleus.
(Parmley et al, 1970). Morgan et al (1977) found the stiffness of isometrically contracting cat soleus muscle to be a linear function of force alone, independent of both length and stimulus rate. Thus the series stiffness or compliance of a muscle should be described as a function of muscle tension.

Since muscle stiffness increases with tension and muscle compliance, the inverse of stiffness, therefore decreases. Wilkie (1950) found the relationship between compliance and tension in the human elbow flexors in situ to be a negative exponential in the range 0-1000N. Goubel et al (1971), Goubel and Pertuzon (1973) and Cnockaert et al (1978), using a range of experimental procedures found the compliance-force function of the same muscle group to be linear only in the log-log domain in the muscle tension range 500-1500N.

8:2.6 Damping of the series elastic component

Hill (1950) in proposing the existence of series elasticity within the contractile apparatus referred to these elastic components as "undamped". Recent experiments suggest that both the contractile component and the series elastic component behave as if lightly damped however (Wells, 1967).

Woledge (1961) determined the viscous damping of isolated frog sartorius to be in the range 200-500 dynes per cm/s. Bahler (1967) reported a similar value of 300 dynes per cm/second for isolated rat gracilus anticus.

In the intact animal or human, the connective and other tissue surrounding the muscle will have a damping effect additional to that of muscle tissue in isolation. Wells (1965), for example, reports a viscous damping coefficient of $1.3 \times 10^4$ dynes per cm/second for rat tibialis anterior in situ. In humans, the passive angular damping coefficient of the structures surrounding the human elbow joint varies between 0.06 and 0.27 Nm/radian in the mid-range of elbow flexion angles (Hayes & Hatze, 1977).

8:2.7 Parallel elasticity

The main site of parallel elasticity in isolated muscle would appear to be the connective tissue sheath (Banus & Zetlin, 1938). However, The sarcoplasmic reticular vesicles within the fibre itself also make a significant contribution (Bendall, 1969; p135). Some evidence suggests that the cross-bridges themselves may contribute to the parallel elasticity of the resting muscle (Alexander & Johnson, 1965; Hill, 1968).

In the intact body, other tissues surrounding the joint across which a

\*1 dyne per cm/second = $10^{-3}$ N/s
muscle acts also contribute to the apparent elasticity and under these circumstances it is difficult to distinguish, for measurement purposes, between the contributions of the passive series components (e.g. tendon) and the so-called parallel component. In general, therefore, measurements have been made of the overall passive resistance to motion of the structures spanning a joint either at a qualitative level for clinical assessment (Foley, 1961) or with the aim of measuring the lumped passive torques at different joint angles. (Tardieu et al, 1977; Hayes and Hatze, 1977; Nashold, 1976)

Hayes and Hatze (1977) measured both the viscous damping and the elastic components of the passive torque around the elbow joint in three living subjects. The passive elastic torque function was found to be linear in the middle one-third of the range of motion but non-linear outside this range. The average value for the passive angular stiffness in this middle part of the range was 1.3 Nm/radian.

The PEC of the three component model can be modelled as an elastic component parallel to both the CC and the SEC or parallel to the SEC alone. Both models fit experimental data well (Jewell and Wilkie, 1958), suggesting that the CC, SEC and PEC of the three-component model can be treated as undistributed elements, that is, the mechanical locations and linkages of the PEC and SEC need not be defined.

8:2.8 The measurement of muscle elasticity in humans.

The experimental procedures that have been adopted to determine the elastic properties of the human body in vivo can be divided into two broad categories: those which aim to determine the local compliance or stiffness function of a particular muscle group and those which attempt to determine an overall or "global" stiffness equivalent for the body.

In the "global" category may be grouped the methods proposed by Cavagna (1970), Greene & McMahon (1979), Aruin et al (1979) and Luhtanen and Komi (1980).

Cavagna (1970) modelled the body as a damped mass-spring system and determined its properties from the damped harmonic oscillations of the ground reaction force produced by trained subjects landing on a force platform with the knees locked and the calf muscles held in sustained contraction. The damping coefficient of the model system is determined from the exponential decrease of the amplitude of the oscillations and the stiffness of "the elastic structures" is calculated from the frequency of
oscillation when the damping coefficient and body mass are known. This method yielded an average damping coefficient in five subjects of 58.17 kg per m/s (571 Ns/m) and an average stiffness of 3.80 kg/mm (37.3 kN/m) for two legged landings. The equivalent values for single legged landings (converted to SI units) were 474 Ns/m and 24.3 kN/m respectively.

Greene and McMahon (1979) similarly modelled the body as a damped mass-spring system. In this case, however, the properties of the system were determined by studying the interaction between this 'internal' system and an external mass-spring system in the form of a spring-board. The equation for the resonant frequency of the two-mass, two-spring system thus created was solved for the stiffness of the 'internal' spring. The stiffness thus measured was found to vary with knee angle and with the additional load borne by the subjects. The average stiffness thus determined for five subjects with both feet in contact with the spring board was 37.6 kN/m. This value is remarkably close to the average value reported by Cavagna (1970). Although a slight trend for the measured stiffness to increase with increasing weight was observed, these changes were not considered significant.

Aruin et al (1978) determined the global viscoelastic characteristics of the muscles acting at the ankle by means of another "vibration" method. Subjects stood on a vibrating platform with the knees locked and the ankles held in an extended position by voluntary effort. As with the method of Greene and McMahon (1979) the stiffness is calculated from the resonant frequency of the combined man-vibrator system. The stiffnesses thus recorded varied between 14.9 kN/m and 18.8 kN/m.

Alexander and Vernon (1975) describe a method of calculating the apparent elastic stiffness of hopping kangaroos from the motion of the whole body centre of mass and the point of contact with the ground. This method has been adapted by Luhtanen & Komi (1980) to calculate the apparent stiffness of running humans. The leg is modelled as a spring extending from the point of contact of the foot with the ground to the body centre of mass. The period of ground contact is divided into two phases ('concentric' and 'eccentric') by the point at which the height of the centre of mass of the body is a minimum. The path of the centre of mass is assumed to be an harmonic wave in each phase. The ground reaction force vector is assumed to pass through the centre of mass and is also modelled as a harmonic wave. The apparent stiffness of the lower body in each phase is calculated as the ratio of the change in the estimated force and the change in the length of the "leg". The "concentric" stiffness was found to
be constant over a range of speeds whereas the "eccentric" stiffness was generally higher, increasing with increasing running speed. Average values for six running subjects were reported as 0.133 N/mm/kg body mass (10.87 kN/m) and 0.645 N/mm/kg body mass (52.70 kN/m) for the "concentric" and "eccentric" phases respectively.

Ito et al (1983) used the same method to examine the apparent spring constants of athletes at a range of treadmill running speeds. The "concentric" spring constant increased from 0.08 to 0.11 N.mm⁻¹kg⁻¹ and the "eccentric" spring constants increased from 0.18 to 0.39 N.mm⁻¹kg⁻¹ as treadmill speed increased from 7 to 22 kph.

Many of the methods that have been used to determine directly the visco-elastic properties of a particular muscle group in situ are similar in principle to those used in isolated muscle preparations. The 'quick-release' method (Goubel & Pertuzon, 1973; Cnockaert et al, 1978) for example is comparable to that used in experiments on the elasticity of isolated vertebrate skeletal muscle preparations (Wilkie, 1956; Jewell and Wilkie, 1958). In these latter experiments the load on the tetanised muscle is suddenly reduced and the initial rapid contraction attributed to the shortening of the elastic component of the muscle. By means of repeated measurements, at different loadings, it is possible to map out the relationship between the change in muscle tension and the change in muscle length. Thus the stiffness or compliance characteristics of the muscle can be calculated over a range of contraction forces.

When this technique is applied to human muscle in situ (Goubel and Pertuzon, 1973; Cnockaert, 1978), the tethered limb is released during a voluntary contraction of its flexors or extensors. This release, as in isolated muscle, results in the shortening of the elastic component which is suddenly unloaded. The force-extension ratio is in this case determined from recordings of joint angle and angular acceleration during the period between release and the onset of reflex EMG activity in the antagonists. The muscle compliance of the elbow flexors thus calculated for five subjects was found to be well modelled as power function of force.

These results also confirmed that muscle compliance behaves independently of muscle length as measurements made at elbow flexion angles in the range 30-105 degrees all yielded similar results.

At lower muscle tensions, in the range 30-300N, the same research group (Goubel et al, 1971) made measurements of elbow flexor compliance by means of a different experimental procedure in which subjects made
voluntary, small amplitude, unidirectional flexions ("inertia movements") of the forearm in a horizontal plane, both with and without a small additional load. A typical recording of elbow flexor and extensor EMGs during such a movement exhibits a silent period during which neither muscle group is active. The length-tension changes during the silent period are attributed to the series elastic component and the compliance calculated from the ratio of changes in calculated muscle length and changes in calculated muscle tension over a range of tensions.

A further method reported by the same group (Cnockaert et al, 1978) uses the same apparatus and recording techniques as the "quick-release" and "inertia movement" methods but, in this case, voluntary flexion movements were performed against a range of additional loads. These "loaded movements" exhibit an isometric phase, the duration of which is a function of load. The extension of the SEC at a given load, and hence compliance, was calculated from the initial acceleration of the limb at the end of the isometric phase and the duration of that phase.

Wilkie (1950) calculated the compliance-tension function of the elbow flexors in five subjects from an independently determined force-velocity function and the rate of change of tension during the onset of an isometric tetanic contraction using the relation:

\[ \frac{v}{G} = \frac{dP}{dt} \]  

where \( G \) is muscle compliance, \( v \) is the velocity of shortening of the contractile component and \( dP/dt \) is the rate of change of muscle tension, \( P \), with respect to time; all determined at a given value of \( P \).

This equation is derived by equating the velocity of shortening, \( v \), with the velocity of extension of the series elastic component \((G)\frac{dP}{dt})\) and is discussed in further detail below. Values of elbow flexor compliance, expressed at the hand, at a tension of 67% of maximum isometric tension varied in five subjects from 0.5 to 3.6 cm/megadyne, with an average value of 1.38 cm/megadyne. Wilkie (1950) also reports data for the elbow flexor compliance of one subject over a range of tension values. Compliance falls with increasing tension, in an approximately negative exponential manner from a passive compliance (compliance at zero tension) of 2 cm/megadyne (expressed at the hand).

Despite the variety of methods described in the literature, there is
apparently no record of any measurements of the elastic properties of specific muscle groups in the human lower limbs.

The "local" methods of the French group (Cnockaert et al, 1978) for the determination of the elbow flexor and extensor compliance relate closely to the methods used on isolated muscle preparations and appear to have good cross-validity. It is envisaged however, that there would be some difficulty in applying these methods to the lower limb due to the much greater forces and masses involved.

The method used by Luthanen and Komi (1980) and others seems particularly removed from commonly used muscle models. The lower limb is treated as a single spring with a single simple elastic constant. No attempt is made to anatomically locate the elastic energy stored. Also, unlike Alexander and Vernon’s (1975) original proposal, no correction is made for active joint flexion and extension. Thus any bending movement of the joints which changes the distance between the centre of gravity and the point of contact with the ground is attributed to elastic extension and contraction of the "apparent spring". The implied model in this technique appears to be that of the analogy between the running action and a bouncing ball (Cavagna, 1978). While this image is a very graphic one, its applications are limited.

The quasi-static methods used by Cavagna (1970), Greene and McMahon (1979) and Aruin et al (1979) give more acceptable results since joint movements are limited when the elastic stiffness is measured. The stiffness thus measured cannot be attributed to a particular muscle group or anatomical structure, however, as is a simple "elastic constant" rather than an elastic compliance-force function for a particular muscle group over the full range of muscle contraction forces.

The method proposed by Wilkie (1950) appears to give slightly different results from those of Cnockaert et al (1978), but the difference could well be within the range of inter-subject variation. The determination of a force-time curve for the isometric contraction ("isometric myogram") presents few practical difficulties, but a force-velocity function for the muscle must also be established. The method would appear to be adaptable to the extensor muscles of the lower limb, however. Possible developments of methods of this type are given further consideration in the following sections.
8.3 MUSCLE ELASTICITY AND THE RISE IN MUSCLE TENSION AT THE ONSET OF AN ISOMETRIC CONTRACTION.

8.3.1 Simple model of the isometric myogram.

Given a simple mechanical model of muscle in which the contractile component is considered to act in series with a single elastic component then the tension-time curve at the onset of an isometric contraction (the "isometric myogram", Figure 8.1) should be fully determined by the force-velocity function of the contractile component and the stress-strain relation on the series elastic component (Hill, 1938).

Assuming, that the velocity of contraction of the contractile component and the rate of extension of the elastic component are equal in a contraction in which the net length of the muscle does not change (i.e. an isometric contraction); then, instantaneously,

\[
\frac{dx}{dt} = \frac{dP}{dt} \frac{dx}{dP}
\]

Eqn 8.5

where \(dx/dt\) is the velocity of contraction of the contractile component, \(dx/dP\) is the compliance of the series elastic component and \(dP/dt\) is the rate of rise of tension, the first derivative of the isometric myogram.

This equation has been used to calculate the force-velocity function isometric myogram and stress-strain function in a wide range of muscle preparations (Parmley & Sonnenblick, 1967; Parmley et al, 1970). Also, given the isometric myogram and the force-velocity function it is possible to calculate a stress-strain relationship (Wilkie, 1950; Eqn 8.4).

Also, given any arbitrary force-velocity and stress strain functions, the form of the isometric myogram can be calculated by iterative integration of equation 8.5 thus:

\[
t = \int_{P=0}^{P=P_c} \frac{dP}{\frac{dx}{dt} \cdot \frac{dP}{dx}}
\]

Eqn 8.6

MacPherson (1953) describes a method for calculating the force-velocity function of a muscle from two isometric contractions, based
Figure 8.1: Rise in muscle tension at the onset of an isometric contraction; the "isometric myogram" (schematic).
on the simple model of muscle implied in equation 8.5. As an intermediate step in this calculation, MacPherson derives an equation for series compliance based on the assumption that the contractile component of the muscle has a fixed force-velocity relation and that the compliance-force function is also fixed:

$$\frac{dx}{dP} = G_e \cdot \frac{d^2\xi}{dt^2} \left[ \frac{dP}{dt} - d\xi \right]$$

Eqn 8.7

where subscripts $o$ and $c$ refer to contractions with and without a known added compliance, $G_e$. This equation enables the compliance-force function of a muscle to be calculated from the force-time curves of two isometric contractions, one a normal contraction, and the other a contraction with a known added compliance in series with the muscle. Chapman and Harrower (1977) for example, apply the equation to the calculation of the compliance-force function of rat gastrocnemius. The method has not been applied to human muscle, however.

While MacPherson's method would seem to be easily applied to human muscle groups in vivo, there is experimental evidence to suggest that there are differences between the force-velocity function of muscle in isometric and isotonic conditions. If this is the case, then force-velocity and compliance-force functions calculated from isometric contractions may not apply under non-isometric conditions.

8:3.2 Differences between isometric and isotonic force-velocity functions

Jewell & Wilkie (1958) made independent determinations of the isometric myogram, force-velocity relation and stress-strain function of isolated frog sartorius muscle. An estimate of the isometric myogram was then calculated from the force-velocity and non-linear stress-strain relations by integration of equation 8.5. A discrepancy was found between the observed and estimated isometric myograms, the estimated values of $dP/dt$ again being higher. These results are further supported by the results of Farmley and Sonnenblick (1967) and Farmley et al (1970) who found the observed speed of contraction to be less than that estimated from equation 8.5 in preparations of cat papillary muscle, rat soleus, and cardiac muscle of the invertebrate, Limulus polyphemus. Similarly, McCrorey et al (1966) also found the series compliance of cat tenuissimus muscle, as estimated from equation 8.5, to be somewhat higher than that determined directly by the controlled release method (Hill, 1950).
While it is apparent that the isometric tension rises more slowly the greater the series compliance, these published results suggest that the simple model expressed by equation 8.5 is in fact an oversimplification and that the force-velocity behaviours of muscle under isotonic and isometric conditions are different. Jewell & Wilkie (1958) express the view that the difference is in some way due to the the previous history of the muscle having an effect on the instantaneous velocity. Evidence for such an effect has been described by Bahler et al (1968) for example, who found that if the initial length of the muscle is varied then the velocity of shortening is also a function of some variable related to the duration of stimulation. The results of Cavagna et al (1967,1975) and Cavagna & Citterio (1974) also suggest that the previous history of the muscle may affect the force-velocity behaviour - in this case rapid stretching of the muscle appears to affect the subsequent behaviour of the contractile component as well as the series elastic component. The discrepancy may also be accounted for, in part, by the fact that at a particular load and length, the velocity of the contractile material as a whole is less in an isometric contraction than in an isotonic one (Close,1972).

Parmley et al (1970) studied rat soleus and Limulus heart muscle preparations and found the contraction velocity calculated from the isometric myogram to be substantially less than than that determined directly from isotonic measurements. The calculated velocities did however, coincide with those recorded during the initial "rapid rise" phase of the isotonic contraction. This, it is suggested, shows that equation 8.5 does have some "physiological reality", with the discrepancy perhaps being accounted for by some difference in the cross-bridge bond position or angle between isometric and isotonic contractions.

8.3.3 Observations on the difference between isometric and isotonic force-velocity functions

Parmley et al (1970) describe the isotonic and isometric force-velocity functions of the cardiac muscle of the Horseshoe crab, Limulus. Figure 8.2 is a reproduction of their diagram (Figure 7, page 548, Parmley et al, 1970) showing the observed isotonic force-velocity curve and that calculated by an equation of the form of Eqn 8.5. The two curves have been digitised and compared at intervals of 1g force in the range 1 to 7g force. The two curves are found to be highly correlated ($r > 0.99$). On average the isotonic velocity at a given force is 1.95 times the isometric velocity at the same force. Dividing the isotonic curve by a
Figure 8.2: Force velocity curves of Limulus heart muscle

- isotonic
- isometric

both redrawn from Parmley et al (1970), Figure 7.

◊ isotonic curve times 1/2
Figure 8.3: Force velocity curves of cat tenuissimus muscle.

- isotonic
- isometric

both redrawn from Parmley et al (1970), Figure 3A

△ isotonic curve times 2/3
factor of 2 gives a good approximation to the calculated isometric force-velocity function (Figure 8.2). The suggestion that the discrepancy is described by a simple factor may of course be purely coincidental.

Parmley et al (1970) also record isotonic and calculated isometric force-velocity functions for a cat papillary muscle preparation. The data displayed in their diagram (Figure 3A, page 548, Parmley et al) has been treated in similar fashion to that described above for the Limulus data, in the range 0.5 to 6.5g afterload. Again, the two curves were found to be highly correlated ($r > 0.99$). On average, the ratio of isotonic velocity to isometric velocity at a given force was 1.5 (standard deviation 0.09). Multiplying the isotonic velocity by a factor of $2/3$ gives a good approximation to the calculated isometric force-velocity function (Figure 8.3).

Data for rat soleus was also presented by Parmley et al (1970). The resolution of their diagram (Figure 5C, page 548, Parmley et al, 1970) is such that it is not possible to retrieve data from the figure with reasonable accuracy. It would appear, however, that the difference between the isotonic and isometric force-velocity curves could not be accounted for by a simple multiplicative factor.

Figure 8.4a shows the isometric and isotonic force-velocity functions of frog sartorius muscle, redrawn from Figure 7 of Jewell and Wilkie (1958). The "calculated" isometric myogram was determined by iterative integration of Equation 8.6, using Jewell and Wilkie's (1958) data for stress-strain and force-velocity functions. The "observed" isometric myogram is the rise in tension recorded by Jewell & Wilkie (1958).

Examination of the relationship between the observed and calculated isometric myograms reveals that the calculated and observed times to achieve a particular tension differ by a factor of 2 (Figure 8.4b). Thus the observed isometric myogram is well modelled by iterative integration of equation 8.6 with the force-velocity function divided by a factor of two. This function is plotted in Figure 8.4c together with the original data from Figure 8.4a for comparison.

Similarly, McCrory et al (1966) calculated the compliance of cat tenuissimus and found that at a given tension the compliance values calculated from the isometric contraction were approximately 2 times greater than those obtained by a quick release method.

In four of the five cases examined, the isometric and isotonic force-velocity functions differ, to a first approximation, by a simple
Figure 8.4: Comparison of observed and calculated isometric myograms of frog sartorius muscle.

(a) Isometric myograms from Jewell & Wilkie (1958);
   (i) observed
   (ii) calculated from equation 8.6
Figure 8.4: Comparison of observed and calculated isometric myograms of frog sartorius muscle.

(b) Observed time to reach a given force as a function of the calculated time to reach a given force.
Figure 8.4: Comparison of observed and calculated isometric myograms of frog sartorius muscle.

(c) Isometric myograms:
* observed (Jewell & Wilkie, 1958)
* calculated from Equation 8.12 with velocities divided by 2
ratio. In the fourth case, a full examination was not possible. It must be stressed that these observations are limited. The data examined was retrieved from published reports, and the endpoints of the curves were excluded from the investigation.

The evidence of Parmley et al (1970) showing that the isometric force-velocity function is equal to that of the initial "rapid rise" phase of the velocity of isotonic contraction suggests that the difference is due to some change in cross-bridge configuration. Under such circumstances it is possible that changes in cross-bridge geometry would cause contraction velocity to change in a simple way. Without repeating the isolated muscle experiments it is unwise to speculate on the extent to which this observation can be generalised. The evidence strongly suggests however, that the isometric and isotonic force-velocity functions of muscle are related by a simple function.

8:4 FURTHER ANALYSIS OF THE SIMPLE MODEL OF ISOMETRIC CONTRACTION

On the basis of a simple model of muscle, it is possible to calculate a compliance-force function for a muscle from the force-time curves of two isometric contractions. It has been shown however, that there is some discrepancy between the isometric and isotonic force-velocity functions of isolated muscles. Examination of published data suggests that the relationship between the isometric and isotonic force-velocity functions is a simple one. While the differences between the two types of contraction appear to invalidate the calculation of dynamic force-velocity functions from isometric contractions in the manner proposed by MacPherson (1953), it has not been demonstrated that the intermediate step, calculation of a compliance-force function, is affected in the same way.

8:4.1 Calculation of muscle compliance from two isometric contractions.

Consider a simple model of muscle action in which a contractile component is considered to act in series with an elastic component (Figure 8.5a) and some viscous damper.

For a maximal isometric contraction, the following assumptions are made:
1. The velocity of contraction of the contractile component in an isometric contraction is linked in some way to the isotonic force-velocity function and can be represented by the undefined
Figure 8.5: Elementary series viscoelastic muscle model.

(a) muscle model only
(b) muscle + added compliance

■ contractile component
♩ series elastic component
□ series viscosity
∧ added compliance
function \( V() \).

2. The series elastic component has a compliance function, \( G(P) \), a function of tension, \( P \) that is independent of muscle length. (As demonstrated by Jewell & Wilkie (1958) for frog sartorius and by Goubel & Pertuzon (1973) for the intact human elbow flexors in vivo.)

3. The series elastic component is damped, with an effective damping coefficient governed by some function of force and velocity, \( H(P,V()) \). The simple model of contraction dynamics ignores the possible effects of any damping on the series elastic component but such damping is known to occur and has been estimated as being in the range 0.2 - 0.5 Ns/m for isolated frog sartorius and rat gracilis muscles (Woledge, 1961; Bahler et al, 1967).

4. The tension developed in the contractile component is a function of contraction velocity alone. This assumption is embodied in Hill's equation (equation 8.1) which has been shown to describe the force-velocity function of a wide range of vertebrate muscles.

5. The amount of shortening of the contractile component during the contraction is small enough to allow \( P_o \), the maximal isometric tension to be assumed constant.

Equating the velocity of shortening of the contractile component \( V() \) to the velocity of extension of the series visco-elastic component gives

\[
V() = G(P). \frac{dP}{dt} + \frac{P}{H(P,V())}
\]

Eqn 8.8

If a compliance, \( G_o \) is added in series with the system (Figure 8.5b) then, assuming the effect of the added compliance on the overall length change of the muscle is negligible:

\[
V()_o = G(P_o). \frac{dP_o}{dt} + G_o \frac{dP_o}{dt} + \frac{P_o}{H(P,V())}
\]

Eqn 8.9

where \( P_o \) is the tension generated with the added compliance.
Referring to assumption 6, above, then at a given tension, \( P = P_0 \), the velocities \( V_0 \) and \( V_2 \) may be equated, giving by subtraction

\[
G(P) = G_0 \frac{dP_0}{dt} = \frac{dP}{dt} - \frac{dP_0}{dt} \tag{Eqn 8.10}
\]

This equivalent to MacPherson's (1953) equation, developed as an intermediate stage in the calculation of a force-velocity function. In this case, however, the equation has been derived on the basis that there is some simple but unknown relationship between the isometric and isotonic force-velocity relationships and that some simple series viscosity also acts to resist the motion of the SEC. On the basis of a simple model of muscle therefore, the calculation of a compliance from isometric contractions is independant of muscle viscosity and any first order differences between isometric and isotonic force-velocity behaviour.

Thus the compliance function, \( G(P) \), can be calculated by comparison of the isometric myograms obtained directly and with the addition of a known compliance in series with the contracting muscle.

The main advantage of this method, compared to that of Wilkie (1950) is that the force-velocity function of the muscle group under consideration need not be known. No assumption is made about force-velocity behaviour of the contractile component other than that it is the same for both contractions. Secondly, the effects of simple viscous damping are eliminated in the subtraction. Other effects common to both contractions may be eliminated in a similar way.

**8.4.2 The effect of added series compliance on the isometric myogram**

The general effect of adding further series compliance to the contractile system on the onset of an isometric contraction is to reduce the rate of rise of tension (Katz, 1939; Wilkie, 1950).

Figure 8.6 shows the effect of various added compliances on the isometric myogram calculated from equation 8.6 (assuming no damping) using data for the in vivo human biceps from Wilkie (1950, subject DW).

When the added compliance is small, the force-time histories of the
two contractions are very similar and it may be justifiable to assume that simple effects of the muscle's "previous history" (Jewell & Wilkie, 1958) may also be eliminated. The nature of these effects is, however, unknown.

At higher added compliances, the model muscle fails to reach a constant isometric tension. As illustrated in Figure 8.6, the muscle-spring system begins to oscillate. The simulations illustrated in Figure 8.6, based on equation 8.6 with muscle parameters from Wilkie (1950) were achieved by means of a simple computer programme which is listed in Appendix G.

8:4.3 The general case for two added compliance trials

Given a second added compliance, Gb in series with the contracting muscle:

\[ V(t) = G(P_b) \frac{dP_b}{dt} + G_b \frac{dP_b}{dt} + \frac{P_b}{H(P, V)} \]

Eqn 8.11

where \( P_b \) is the tension generated with the second added compliance, Gb.

Subtracting equation 8.9 at a given velocity \( V(t) = V(t) \) gives

\[ G(P) = \frac{G_b \cdot dP_b - G_a \cdot dP_a}{dP_a - dP_b} \]

Eqn 8.12

Thus the compliance function could be determined by comparison of two isometric contractions, both with known added compliance in series with the muscle. Equating either of the added compliances \( G_a \) and \( G_b \) to zero gives equation 8.10.

8:4.4 Correction for series compliance in the measuring apparatus.

Any compliance or "give" in the measuring apparatus used to record the isometric myogram will act in series with the contractile component of the model, and can be considered as being additional to any known added
Figure 8.6: Rise in tension ($P$) with time ($t$) at the onset of an isometric contraction with added compliances of

1. 0
2. $0.5 \times 10^{-2}$ m/N
3. $1.0 \times 10^{-2}$ m/N
4. $1.3 \times 10^{-2}$ m/N

Calculated from equation 8.12 assuming the force-velocity function of the muscle to obey Hill's equation (Equation 8.1) with $P_0 = 30$, $a = 12$ and $b = 18$; and SEC stiffness of $3P + 20$

(See Appendix G)
compliance. Given a measuring apparatus series compliance of \( g \), added to the known series compliances in Eqn 8.12 gives

\[
G(P) = \frac{G_b + g}{dt} \frac{dP_a}{dt} - \frac{(G_a + g) . dP_a}{dt} \frac{dP_a}{dt}
\]

which simplifies to

\[
G(P) + g = \frac{G_b . dP_a}{dt} - \frac{G_a . dP_a}{dt}
\]

in the general case and to

\[
\frac{dP_a}{dt} = \frac{G_a . dP_a}{dt} - \frac{dP_a}{dt}
\]

when only one added compliance is used. Thus any error due to compliance in the measuring system acting in series with the contractile component of the muscle can be corrected for by subtraction.
8.5 EXPERIMENTAL PROCEDURE

The methodology developed in the previous section was implemented to enable the compliance-force relationship of the knee extensors of human subjects to be determined. Knee extension force-time curves and m. Vastus lateralis EMG were recorded during the onset of a maximal voluntary isometric contraction. The recordings were repeated with springs of known compliance added in series with the force-measuring system. Comparison of the recordings made with and without the added compliances enables the compliance-force function to be determined.

8.5.1 Subjects

The results presented here were recorded as part of a series of experiments, including those described in Chapter 9. Briefly, the subjects were all male athletes in current training who volunteered to take part in the experimental series. The results presented here relate to the sub-group of ten athletes who completed the whole series of experiments, described in Chapter 9.

8.5.2 Apparatus

A rigid table with a steel box frame and hardwood top (Figure 8.7) was adapted for the purposes of this experiment. A strain gauge force-transducer (Pioden Controls, Type UF2) was mounted horizontally on oblong-section steel tube cross-members which were in turn clamped to two square-section steel tubing uprights bolted to the original frame. The clamping system enabled the height of the force transducer and its distance from the end of the table to be adjusted over a range of about 0.6m vertically and 0.3m horizontally.

A heavy duty canvas ankle strap was attached to the force transducer by a light chain and rigid snap links (Figure 8.8) in such a way that the chain could quickly be removed and replaced by a spring of the same length.

The preamplifier of one channel of an EMG recording system was also mounted on the table frame.

The voltage output from the conditioning unit of the force transducer was passed via one channel of an 8-channel, 12-bit analogue to digital converter to a CBM4032 microcomputer. Amplified EMG signals were full-wave rectified and low-pass filtered by means a simple rectifier/capacitor circuit before analogue to digital conversion and transmission to the CBM4032.
Figure 8.7: (Plate) Arrangement of apparatus for the measurement of the elastic properties of the knee extensors.

A: Rigid table
B: Force transducer
C: Chain link
D: Canvas ankle strap
E: EMG electrodes

Figure 8.8: (Plate) Alternative arrangements of the linkage between the subject's ankle strap (D) and the force transducer (B). The chain linkage (C) can be replaced by a spring (F) to give added compliance.
8.5.3 Calibration

The force transducer was calibrated by static loading with standard masses in the range 0 to 30 kg. The transducer was found to be linear. A standard resistance in the transducer's signal conditioning unit was found to give a consistent signal from day to day when applied across the bridge circuit. Once a linear calibration equation for the transducer had been determined by static loading, the load equivalent to the standard resistance was calculated. Thereafter, the unit was calibrated by application of zero load to the transducer and by application of the standard resistance (equivalent to a standard load) across the bridge circuit. During the data collection period, the calibration of the standard resistance was checked daily against static loading.

The elastic stiffnesses of the added compliances was checked by static loading. Three of the four springs used for this purpose were manufactured to specification by Clarkes Springs Ltd. of Leicester. The springs were suspended vertically from the rigidly mounted force transducer. Loads in the range 0 to 30 kg were in turn suspended from the spring. A calibrated electronic displacement transducer was mounted beside the spring in such a way as to measure the displacement of its endpoint from the original, zero load position. Between 100 and 252 independent measurements were made on each spring. The average product moment correlation between load and spring extension was greater than 0.96 in each case. The compliance of each spring was determined as the slope of the least squares regression line best fitting a graph of extension against load. Spring stiffness was calculated as the inverse of compliance. The properties of each of the 4 springs used in this study are summarised in Table 8.1.

The compliance in the measuring apparatus was determined by mounting the table used for this purpose on two supports in such a way that the axis of the force transducer was vertical. The chain links and canvas ankle strap normally used in the experimental procedure were attached to the force transducer in the normal way. A weight carrier was suspended from the canvas ankle strap. A calibrated displacement transducer was mounted below the weight carrier to measure its deflection under load. A total mass of 30 kg was loaded onto the weight carrier and the deflection of the displacement transducer recorded by computer. This procedure was repeated 10 times. The mean deflection of the measurement system, assuming the weight carrier to be rigid was 0.7 cm under a load of 294 Newtons which is
<table>
<thead>
<tr>
<th>Spring</th>
<th>Nominal Measured</th>
<th>Measured Stiffness</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>stiffness compliance</td>
<td>Stiffness</td>
</tr>
<tr>
<td></td>
<td>lb/inch</td>
<td>$10^{-4}$m/N</td>
</tr>
<tr>
<td>1</td>
<td>0.93</td>
<td>10750</td>
</tr>
<tr>
<td>2</td>
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<td>7090</td>
</tr>
<tr>
<td>3</td>
<td>21</td>
<td>5200</td>
</tr>
<tr>
<td>4</td>
<td>23</td>
<td>4130</td>
</tr>
</tbody>
</table>
equivalent to a compliance of $2.4 \times 10^{-5}$ m.N$^{-1}$. On the basis of equation 8.14 this value was subtracted from all the muscle compliances determined using the methods described above.

8.5.4 Experimental protocol

After being verbally familiarised with the requirements of the experiment, subjects were weighed and bipolar EMG surface electrodes attached to the skin above the distal portion of the m. Vastus lateralis of the right leg. An earthed electrode was also attached to the skin above the patella.

Subjects were then instructed to lay supine on the experimental table with both lower legs hanging over the table end (Figure 8.7) with hands resting loosely across the chest and with the head supported by a low pillow. A small foam pad was placed behind the knee at the end of the table. This had the effect of raising the knee slightly, aligning it horizontally with the hip, and also increased the subjects' comfort.

The canvas ankle strap linked to the force transducer was attached to the subject's leg and the level of the force transducer was raised or lowered until the line of action of the tether was horizontal. The perpendicular lever arm between the approximate knee joint centre and the tether was then measured with a metre rule.

At this point, the force transducer was calibrated as previously described.

The subject was instructed to respond to the command "go" by exerting maximum force against the tether, to attempt to attain maximum effort as quickly as possible and to maintain maximum effort until instructed to relax (after 1-2 seconds). Each subject was required to rehearse this procedure using the left (non-experimental) leg until both he and the experimenter were confident that the subject was fully familiarised with the requirements of the test.

Each subject then performed five trials with a rest interval of two minutes between each trial. The first trial was conducted with a chain link between the ankle strap and the force transducer (Figure 8.8). During subsequent trials, the chain was replaced by springs of compliance 0.93 * $10^{-4}$ m/N, 1.42 * $10^{-4}$ m/N, 1.91 * $10^{-4}$ m/N and 2.41 * $10^{-4}$ m/N respectively (Figure 8.8).

The outputs of the force transducer and EMG recording system were sampled at 100Hz during each trial by the analogue to digital converter under computer control. Digitised input data was stored in computer memory.
and displayed graphically on the terminal screen for examination by the experimenter. On the basis of the force and rectified, filtered EMG records, trials in which the subject's effort did not appear to be sustained and maximal were discarded and repeated. (Approximately 6% of all trials were required to be repeated on this basis). Force and EMG data collected during all trials were recorded on computer diskette for later analysis.

8.5.5 Analysis of data

The force-time records (isometric myograms) of each subject for each of the five trials were recalled from diskette into computer memory and displayed (Figure 8.9).

In some cases, trials performed with the more compliant springs in series with the force transducer demonstrated significant oscillations in the force-time trace at higher force levels (Figure 8.10). Normally, oscillation of the spring had been observed and noted during the measurement sequence. All recordings of this type, whether noted during the measurement sequence or not, were discarded.

The remaining trials for each subject were then analysed as follows:

1. Each force-time record was digitally filtered using a fourth order, zero-lag Butterworth filter at a cut-off frequency of 25Hz.
2. The first time derivative of the filtered force-time record was determined using a five-point numerical differentiation algorithm (Figure 8.11).
3. The values of the first derivative at intervals of 1% of maximum isometric tension, $P_0$, were determined over the full range of tensions by 2nd order Newtonian interpolation (Appendix F; Figure 8.12).
4. Knee extensor compliance, $G$ was calculated at each 1% interval by comparison of records obtained with and without added compliance using equation 8.15, Figure 8.13.

8.5.6 Error handling

Several important factors influence the reliability of the values of $G(P)$ at the extremes of the range of tensions. In particular:

1. Since the value of $G(P)$ is dependant on the difference $(dP/dt)_0 - (dP_a/dt)_a$, at a given tension level, the detrimental effects of experimental error and noise on data quality are greatest when this difference is small and therefore less reliably determined. Such circumstances occur at tension levels near zero and at tensions
Figure 8.9: Example knee extensor isometric myogram recording. 

(a) Tension in restraining tether 
(b) EMG linear envelope from Vastus lateralis.
Figure 8.10: Example knee extensor isometric myogram recording with added compliance of $2.42 \times 10^{-4}$ m/N, showing oscillations observed with high added compliances (c.f. Figure 8.6).
Figure 8.11: Example first time derivative ($dP/dt$) of knee extensor isometric myograms:

- normal
- with added compliance of $0.93 \times 10^{-4}$ m/N
Figure 8.12: Example first time derivative (dP/dt) of knee extensor isometric myograms, expressed as a function of tension (P).

- normal
- with added compliance of $0.93 \times 10^{-4}$ m/N
Figure 8.13: Knee extensor compliance ($G$) as a function of tension ($P$), both expressed at the tethering point of the force transducer. Same data plotted on linear and logarithmic scales. Compliance values at tensions less than 10% $P_0$ and greater than 90% $P_0$ are discarded.
approaching maximum isometric tension.

2. The maximum frequency response of the force transducer may be a limiting factor in determining the high rates of change of the force at the onset of the isometric contraction.

3. At the initiation of the contraction the implied assumption that the myoelectric activity during both contractions is maximal is least valid.

4. The greater the added compliance, \( G_a \), the greater the violation of the assumption that the muscle activity is isometric.

In view of the possible introduction of error from these sources, the following precautions were taken:

1. A single added compliance trial was selected for comparison being that performed at the lowest added compliance which yielded differences between \( (dP/dt)_o \) and \( (dP/dt)_a \) of greater than 20% of \( (dP/dt)_o \) at the point of maximum rate of change. Data from one subject was discarded because no added compliance trial met this criterion.

2. Calculated values of \( G \) for tensions in the ranges 0-10% of \( P_0 \) and 90-100% of \( P_0 \) were discarded.

The remaining values of \( G \) were fitted by a least squares regression to an exponential function of the form:

\[
G = k \cdot \exp(qP)
\]

where \( P \) is the tension,
\( k \) and \( q \) are constants.

8:5.7 Normalisation of compliance values

The compliance due to the series elastic component thus calculated is considered to act at the point of application of force at the restraining tether, not at the muscle. Since the lever arm of the tether about the knee-joint centre varies between subjects, some normalisation of the compliance data is necessary if between subject comparisons are to be made. Furthermore, the compliance data needs to be expressed in some convenient standard form if it is to be used in conjunction with other biomechanical analyses. The actual "muscle compliance" can be calculated from the lever ratio between the points of attachment of the muscle and the tether about the knee joint centre. Variation in the length of the muscle is magnified
by the lever arm and the muscle force is reduced by the same ratio. The
muscle compliance would therefore be given by the formula:

$$G_m = G \frac{(l_m)^2}{(l_t)^2}$$  \hspace{1cm} \text{(Eqn 8.16)}

where

- $G_m$ is the series compliance expressed at the muscle attachment,
- $G$ is the series compliance expressed at the restraining tether,
- $l_m$ is the lever arm of the muscle about the knee joint centre
- $l_t$ is the lever arm of the restraining tether about the knee joint centre.

The accuracy of this correction is highly dependant on the accuracy
with which the knee-joint centre and the line of action of the equivalent
knee extensor muscle group can be determined.

An alternative approach to normalisation, which is less susceptible to
error of this kind, is to calculate the 'angular compliance' effective
about the knee joint as a function of the extensor torque about that joint.
This correction is dependant only on the lever arm of the tether about the
knee joint. Since this is a much larger quantity than the lever arm of the
muscle, the relative error due to inaccurate estimation of the location of
the knee-joint centre is smaller. Since a force of $P$ Newtons at the tether
is equivalent to a moment of $P.1_t$ about the knee joint centre and an
apparent extension of $x$ metres at the tether is equivalent to an angular
displacement of $x/l_t$ radians at the knee-joint centre, a series compliance
of $x/P$ m/N expressed at the tether is equivalent to an "angular
compliance" of $(x/l_t)/(P.1_t)$ radians per Newton-metre. Thus normalisation
can be achieved by the transformation:

$$g(T) = \frac{G(P)}{c}$$  \hspace{1cm} \text{(Eqn 8.17)}

where

- $g(T)$ is the angular compliance about the joint, a function of the
  joint moment, $T$;
- $G(P)$ is the series compliance about the joint centre, expressed at
  the restraining tether
- $c$ is the "angular correction factor", equal to $(l_t)^2$.

Angular equivalent muscle models have also been used by Hof et al (1983).
and Baildon and Chapman (1983a).

Expressing the results in the angular form has a further advantage when compliance data is used in conjunction with a biomechanical analysis of movement. The calculation of muscle forces requires detailed anthropometric data defining the geometry of joints and muscle attachments, and can involve the use of time-consuming numerical optimisation techniques. The alternative approach, that of restricting the analysis to angular equivalents, eliminates these requirements.

8.6 RESULTS

Figures 8.9, 8.10, 8.11 and 8.12 are derived from recordings made on a single subject during the course of this experiment.

Table 8.2 shows the maximum recorded values of the isometric knee extension moment exerted about the approximate knee joint centre by each subject, together with the subjects' body masses and the value of the "angular correction factor" (Equation 17). Also in this table are recorded the coefficients $k$ and $q$ of the exponential curve of the form of equation 8.15a which best fitted the compliance-force data for each subject. Angular compliance can be determined by expanding this function and applying the correction factor.

Table 8.3 records the angular compliance for each subject as a function of the relative knee moment, expressed as a percentage of the maximum isometric extension moment, $M_{ko}$. Values are recorded for 10, 25, 50, 75 and 90% of $M_{ko}$. These data are plotted in Figure 8.14. The mean and standard deviation of these data are also plotted in Figure 8.14, with the curves extrapolated from the best fitting line to include angular compliance values at 0% and 100% of $M_{ko}$. These curves show a characteristic exponential fall in compliance with increasing knee extension moment. The variation between subjects appears to be reduced as knee moment increases. In fact the coefficient of variation of the compliance values about the mean value increases from 55% at 10% $M_{ko}$ to 83% at 90% $M_{ko}$. Much of this increased variation appears to be due to the qualitatively different compliance-moment function determined for subject 2. Re-examination of the original recordings shows no abnormalities in the measurement procedure or isometric myogram recordings for this subject.

Figure 8.15 records the mean angular compliance for all ten subjects as a function of knee moment in the range 0 to 150Nm. The compliance values
are displayed on a logarithmic scale. Table 8.4 shows angular compliance values for individual subjects at knee moments of 0, 50, 75, 100, and 125 Nm. All the values for 0 Nm knee moment have been extrapolated from the exponential fit to data in the range 10% to 90% $M_{ko}$. Where $M_{ko}$ for a particular subject did not reach the level in the appropriate column of the table, a value has been omitted from the table.

The recorded value of angular compliance at a knee extensor moment of 100 Nm was not significantly correlated with either body mass or stature ($r = -0.025$ and -0.234 respectively).

Also recorded in Figure 8.15 for comparative purposes are compliance values for the human elbow flexors from Wilkie (1950) and Goubel and Pertuzon (1973), converted to angular equivalent units. The published values for the elbow flexors are of a similar magnitude to those recorded here for the knee extensors in the range of joint moments where the data sets overlap.
Table 8.2: Mass, maximum observed isometric knee extension moment, angular correction factor and coefficients of the best fitting knee-extension compliance-force function.

<table>
<thead>
<tr>
<th>Subj</th>
<th>M</th>
<th>M_{ko}</th>
<th>c</th>
<th>log(k)</th>
<th>q * 1000</th>
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<tr>
<td></td>
<td>kg</td>
<td>Nm</td>
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<tr>
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<td>122</td>
<td>0.164</td>
<td>-5.97</td>
<td>-13.90</td>
</tr>
<tr>
<td>s.d</td>
<td>3.40</td>
<td>21</td>
<td>0.019</td>
<td>0.94</td>
<td>5.14</td>
</tr>
</tbody>
</table>
Table 8.3: Angular compliances at a range of relative knee moments.

<table>
<thead>
<tr>
<th>Subj.</th>
<th>Relative knee moment (% $M_\text{kne}$)</th>
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<tr>
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<td>9</td>
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<tr>
<td>s.d.</td>
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Table 8.4: Angular compliances at a range of knee extension moments

All compliance values are * 1000

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Figure 8.14: Angular compliance of the knee extensors as a function of relative knee extension moment for 10 individual subjects.
Figure 8.15: Mean knee extensor angular compliance (-----) and standard deviation (••••) from this study with data for the elbow flexors adapted from Wilkie (1950) (----) and Goubel and Pertuzon (1973). (○○)
8:7 DISCUSSION

8:7.1 Methodology

The muscle model on which the calculation of compliance-force functions has been based is not a comprehensive one. No account is taken of parallel elasticity or viscosity. The force-velocity relationships and series viscosity of the model have not been defined. While a more rigorous mathematical treatment of the visco-elastic properties of muscle might yield a more sophisticated model, some compromise must be drawn between mathematical elegance and experimental practicability. In this case, the simplification is based on a number of assumptions, which are generally supported by published research.

The assumption that the muscle contractions monitored were isometric is obviously violated by the addition of springs into the measurement system, although, as described above, the experimental protocol was designed to obtain satisfactory results with the minimum of added compliance. In the case of the combination of the softest spring used in the experiment with the largest force applied against it, an extension of the spring of 9.6 cm would occur. This upper limit was not actually achieved in practice. More typically, the trial used to calculate compliance would involve an force of 300N against a spring of compliance 1.42*10^{-4} m.N^{-1} giving an extension of the spring of the order of 4 cm. This extension is equivalent to some 0.1 radians angular equivalent extension at the knee joint. Since the knee extensors have an angular equivalent length of the order of 12 radians, the error due to the added compliance represents less than 1% of the total muscle length.

It was not been possible to directly measure or control muscle activation. The effect of variations in muscle activation is therefore unknown, but other experimental evidence suggests that in similar contractions, any error introduced by differing stimulus rates will be small. MacPherson and Wilkie (1954) have demonstrated that the relationship between tension and rate of change of tension in frog sartorius (Figure 8.12) is independent of the stimulus applied to the muscle. It has also been demonstrated that the compliance-force function of cat soleus does not vary when the stimulus pattern is changed (Morgan, 1977).

The methodology used in this study has a number of practical weaknesses. In particular, the range of knee moments over which a compliance value can be calculated directly is restricted between the approximate limits of 10 - 90% of the maximum isometric moment, at a joint
angle of approximately 90 degrees. Outside of this range, values can only be determined by extrapolation. This is likely to be necessary as soon as dynamic activities are considered, since the moment generated by the knee extensors is likely to be substantially outside this range when eccentric contractions and concentric contractions at knee angles other than 90° are considered. Without making measurements of compliance under dynamic conditions, it is not possible to say whether this extrapolation is acceptable. However, observation has supported the theory that muscle stiffness is a linear function of increasing muscle tension, and that compliance is therefore well represented by an exponential decay as tension or its angular equivalent increases. An exponential function of the form of Equation 8.15a fits the observations better than a straight line or a polynomial up to order 3. It may not be unreasonable therefore, to extrapolate a short way beyond the knee moments measured using the exponential model. Extrapolation to other joint angles is also supported by Goubel and Pertuzon’s (1973) observation that elbow flexor compliance is independent of joint angle. Similarly, Jewell and Wilkie (1958) found the elastic stiffness of frog sartorius to be independent of the muscle’s length. While Greene and McMahon (1979) report that the dynamic stiffness of the lower limb changes as knee angle changes, their methodology does not calculate the compliance-force for a particular muscle group and the variation is as likely to be due to changes in muscle tension and orientation as it is to changes in muscle length. Hayes and Hatze (1977) found the passive stiffness of the structures about the elbow to vary with joint angle. The passive elastic torque increased at the extremes of the range of movement due to the stretching of connective tissues, but was fairly constant in the middle part of the range of motion. If a similar phenomenon applies to the knee then the passive stiffness can be assumed fairly constant over a wide range of joint angles.

The main advantage of the measurement technique described here is that it produces a function for the compliance of a muscle group that can be directly related to a specific model of the muscle, and can be easily applied to perform calculations on typical biomechanical kinematic models of movement. The procedure is non-invasive, quickly executed and there is no requirement to determine a force-velocity function for the muscle group being studied.
8:7.2 Comparisons with published data

It is difficult to make comprehensive comparisons of the data presented here with that recorded elsewhere in published literature. The results of the type used by Cavagna (1970), Greene and McMahon (1979) and Luhtanen and Komi (1980) cannot be related to the elastic properties of a particular muscle group. Wilkie (1950) and the French group (Cnockaert et al, 1978) present results for the elbow flexors, but there is no comparable data for the knee extensors. Comparison of the results of this study with those published for the elbow flexors (Figure 8.15) shows that the angular equivalent compliances are of a similar order of magnitude where the range of joint moments overlaps. This comparability between different muscle groups supports the idea that active series elasticity is a function of the number of active cross bridges (and hence force). Some variability between muscle groups is to be expected where there are different tendon-length/muscle length ratios, different angles of pennation, attachment geometry and varying force-velocity characteristics.

8:7.3 Inter subject variability

The recorded angular equivalent compliance data shows considerable inter-subject variability. The magnitude of this variability at a given knee extension moment gives a coefficient of variation of the order of 50% on average. Since Wilkie (1950) and Cnockaert et al (1978) present data for only one subject, and for a different muscle group, no comparison of inter subject variability is possible. If subject 2 is ignored, then there appears to be a trend for the inter-subject variability of the angular compliance to be reduced as knee extension moment increases. This may be due to the existence of greater natural variation in the anatomy and elastic properties of the passive structures surrounding the knee joint than there is the elastic properties of muscle. The influence of the passive element would be reduced as the muscle tension increases. There appears to be an interaction between muscle speed type (as indicated by the proportions of "fast" and "slow" muscle fibres) and speed of movement which effects individual energy storage capacity in rebounding activities (Bosco and Komi, 1982b). It is possible therefore that individual differences in muscle fibre type distributions influence the elastic properties of the muscle. Certainly, the isometric myograms produced by "slow" and "fast" muscles differ in a way that is similar to the effect of added compliance (Wells, 1965; Buller & Lewis, 1965).
8:7.4 Passive vs active elastic components

The methodology adopted here does not distinguish between the 'active' component of series elasticity (that which varies with the muscle tension) and the 'passive' component (that due to tendon and connective tissue). It is possible to extrapolate from the recorded compliance data to a knee moment of zero and hence establish a value for the 'passive' component of compliance (Chockaert et al, 1978). What this represents is probably not the passive series compliance however, since the tendons can only be stretched if there is some tension on the muscle itself. If anything, the extrapolated value at zero muscle tension represents some parallel component of elasticity. Depending on whether the Maxwell of Voight model is used to arrange the parallel and series elastic components in relation to the contractile component, the resting tension can be considered to be borne by the parallel component alone or by the series and parallel components in combination. In view of the fact that there no significant difference between the fit of the Voight and Maxwell models to experimental observations of muscle behaviour (Jewell and Wilkie, 1958) and that the parallel and series elastic components are in fact anatomically distributed, it probably better to refer to the extrapolated compliance value at zero tension as the "passive" compliance rather than try to attribute it to either series or parallel components.

8:7.5 Energy storage during the isometric contraction.

The amount of energy stored in the elastic components during an isometric contraction can be calculated as the work done against series elasticity. During a small time interval, the apparent extension of the SEC is given by the product $dM_k \cdot G(M_k)$ and the work done, equal to the change in energy stored, $dE$, is given by

$$dE = M_k \cdot G(M_k) \cdot dM_k$$  

Eqn 8.18

where $G(M_k)$ is the angular equivalent compliance at a moment of $M_k$.

These equations can be iterated to calculate energy storage and extension of the SEC during the isometric contraction. Figure 8.16 shows the results of one such calculation using typical data from this study. As muscle tension rises (Figure 8.16a), the compliance of the elastic components decreases (Figure 8.16b) and their stiffness (the inverse of compliance) therefore increases. Fifty percent of the energy storage occurs in the
first 0.2 second of the contraction and a total of 70 J of strain energy is stored over the course of the contraction, assuming no losses due to viscous damping and relaxation.

In this example, the total stretch in the angular equivalent SEC was approximately 1.2 radians. This is not equivalent to 1.2 radians of knee flexion, but to a movement of the muscle subtending an angle of 1.2 radians at the joint centre. The total angular equivalent length of the knee extensors is of the order of 12 radians, so the proportional stretch of the SEC is approximately 10%. This estimate compares favourably with the proportional SEC stretch reported by Wilkie (1950) for the human elbow flexors in vivo. Bahler (1968) reports a 7% normalised stretch for rat gracilis anticus muscle and McCrorey et al (1966) report 6% for the cat tenuissimus.

These results clearly demonstrate the capacity of the knee extensors to store energy. The potential levels of strain energy storage in this single muscle group are significant in terms of segmental potential and kinetic energy levels. At an extension moment of 50% of $M_{\text{keo}}$, the knee extensors store 30J of strain energy in the example used above and shown in Figure 8.16. This is equivalent to the kinetic energy of a typical whole leg travelling at a velocity of 2.5 ms$^{-1}$.

8.7.6 Energy dissipation

When the isometrically contracting muscle relaxes, any stored strain energy is dissipated as heat. During a dynamic contraction, the stored energy can be usefully regained if the SEC is allowed to contract and do work. Under these circumstances, energy losses will be determined by a combination of relaxation, dissipation by viscous forces and hysteresis of the elastic components.

Reductions in stored energy due to relaxation of muscle tension are accounted for by equation 8.18. Losses due to the physical viscoelastic relaxation of the elastic structures is not accounted for, however. Aruin et al (1979) determine that the time constant of this relaxation is approximately 5.9 seconds; i.e.

$$E' = E \exp(-t/T)$$

Eqn 8.19

Where

- $E'$ = Retained energy
- $E$ = Energy stored at time zero
Figure 8.16: Example angular equivalent parameters of the knee extensors during the onset of an isometric contraction.

$M_k$: extension moment, $\text{Nm}$

$G$: angular compliance, $\text{N}^{-1}\text{m}^{-1}$

$C$: SEC extension, radians

$E$: SEC strain energy, $\text{J}$
Figure 8.17: Example calculated strain energy stored in the SEC of the knee extensors during an isometric contraction.

A: Simple model (Equation 8.24)
B: Assuming energy dissipation (Equation 8.25)
Figure 8.17 shows the effect of including this energy loss in the iteration of equation 8.18. The energy dissipated during the first 0.5 seconds of the contraction is only a small proportion of the total energy stored. Larger losses occur if the isometric contraction is maintained, however, giving the elastic structures more time to relax. In the course of the running action, where muscles contract for only short periods, these losses can be safely ignored.

* * * * *

The object of the study described in this Chapter was to develop a method for the measurement of elastic energy storage in a particular muscle group. The method devised is based on a highly simplified model of muscle and a considerable number of assumptions have been made in order to implement a muscle-model based technique in humans in vivo. Insofar as comparisons can be made, however, the results agree well with published literature and in the absence of any other published information of this type the "estimates" reported here acquire a greater value than would otherwise be the case.
Chapter 9

ELASTIC ENERGY STORAGE, MECHANICAL ENERGY CHANGES AND ENERGY EXPENDITURE DURING TREADMILL RUNNING
**SUMMARY**

The relationship between energy expenditure and mechanical energy changes was investigated in 14 male athletes running on a motorised treadmill at 4.47 m/s\(^{-1}\). In addition, the strain energy stored in the elastic components of the knee extensors was calculated for ten of the subjects. Energy analyses both including and excluding the strain energy components were compared.

Energy expenditure was significantly correlated with the magnitude of changes in total body energy and there was further evidence for a relationship between the magnitude of passive energy transfers and economy of movement.

Calculated energy storage in the knee extensors was significant during the support phase of the running stride, reaching a mean peak value of 66 J.

The inclusion of strain energy components in the energy model resulted in a significant reduction of the calculated changes in whole body energy by 70 J even though the sum of the absolute changes in the partitioned energy components was increased by 13.6%.

The observed changes in strain energy storage support the suggestion that muscle elasticity acts as an energy conservation mechanism, not only reducing the mechanical work requirement but also reducing the rate at which work must be done.

**INTRODUCTION**

The kinematic energy models that have previously been applied to human gait patterns (Norman et al, 1976; Winter, 1979; Pierrynowski et al, 1980) have only included analysis of changes in segmental potential and kinetic energy components.

The storage and subsequent recovery of strain energy in stretched muscle and connective tissue appear to enhance both the effectiveness and the efficiency of rebounding movements (Asmussen and Bonde-Petersen, 1974; Thys et al, 1972) and it has frequently been suggested that the potential contribution of elastic energy storage to mechanical energy changes in running is likely to be significant, especially at higher running speeds (Cavagna et al, 1971; Luhtanen and Komi, 1980; Cavagna and Kaneko, 1981; Williams and Cavanagh, 1983).

Despite this speculation regarding the magnitude of elastic energy storage during the running stride, the phenomenon has not previously been
the subject of direct experimental measurements. Such direct measurement can only be made if the elastic properties and force-time history of a muscle group is known. Luhtanen and Komi (1980) calculate an "apparent spring constant" for runners based on the model that Alexander and Vernon (1975) originally applied to kangaroos, but as the authors themselves concede, "the interpretation of the real meaning of the apparent spring constant has to be made with caution" (See section 8:2.8).

The main purpose of this study was to investigate the contribution of elastic energy storage to mechanical energy exchange during the running stride by direct determination of energy stored in the series elastic components of the knee extensor muscles. The study also provided an opportunity to further investigate the relationships between mechanical energy changes and energy expenditure in running.

9:2 METHODS

The subjects for this experiment were 14 male athletes of varying performance standards from good club level to international standard, selected from a group of volunteers. The physical characteristics of each subject are shown in Table 9.1. The only selection criteria applied were that the subjects should be actively competing in and training for middle distance running competition, and that they were willing to give their informed consent to participation in the whole series of measurements.

Of the original 14 subjects, four failed to complete the whole experimental series as a result of injury or absence.

Measurements of energy expenditure and cinemographic recordings for biomechanical analysis were made during a six minute run on a motorised treadmill at a constant speed of 4.47 m.s⁻¹.

Prior to the experiments, each subject made a number of visits to the laboratory in order to become familiar with treadmill running and to become accustomed to experimental apparatus.

Before any experimental measurements were made, each subject had spent at least 30 minutes running on the treadmill at various speeds, of which at least ten minutes were spent with a mouthpiece for expired air collection in position.
Table 9.1: Subject characteristics.

Body fat was estimated from skinfold measurements

VO_{2\max} was recorded during an incremental level running test described in the text.

* indicates those subjects who did not complete whole series of experiments.

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9:2.1 Respiratory function measurements
Throughout the treadmill run, expired air was analysed continuously, 
breath by breath using the automated system described in Chapter 6. For the 
purposes of this study, the "trend" system (Section 6.9) was used to 
calculate respiratory function parameters over twenty second sample 
equivalents. As previously described (Section 4:2.2) oxygen uptake and 
carbon dioxide output measurements were used to calculate energy 
expenditure.
An additional sample of expired air was taken prior to the trial in 
order to determine resting levels of energy expenditure.
Prior to, and immediately following each trial capillary blood samples 
were obtained from the runners thumb. These samples were later assayed for 
lactic acid using a modification of Olsen's fluorimetric procedure (Olsen, 1971).
Each subject's maximum oxygen uptake was measured in a separate 
treadmill test on another occasion. The test began with the subject running 
on the motorised treadmill at a speed of 3.58 m.s⁻¹ (8 mph). The treadmill 
speed was increased by 0.22 m.s⁻¹ (0.5 mph) every minute until the subject 
became exhausted and could no longer continue. Respiratory responses were 
recorded throughout the test using the automated system described in 
Chapter 6. Breath by breath data were reintegrated over to give sample 
equivalents of 20s duration. The highest 20s sample value of oxygen uptake 
measured during the test was recorded as the subjects maximum oxygen 
uptake.

9:2.2 Biomechanical analysis
During the last minute of each trial, the subjects were filmed in the 
sagittal plane using a Bolex H16 camera at a measured frame rate of 68 
frames/sec (Section 3:7). The digitised film image coordinates of body 
landmarks were calibrated against a set of known points distributed in the 
field of view and converted to two-dimensional object space coordinates 
using the transformation method described in Section 3:6. Coordinate data 
were filtered using a fourth order, zero lag, Butterworth filter with a 
cutoff frequency of 6.7Hz and differentiated using finite difference 
methods (Section 3:6). One complete stride from each subject was analysed 
using a planar, 15 segment, linked rigid body model of the human body 
(Section 3:2).
The inertial properties of the model were determined using the

* See Acknowledgements
regression equations of Clauser et al (1969) and Chandler et al (1975) (Section 3:3.2). A total of 31 anthropometric measurements were made on each subject in order to enable the calculation of inertial parameters from the regression equations which are based on anthropometry of cadaver samples. While there was some difference in the ages of the cadaver samples and the subjects of this study, the heights and weights of the experimental group and the cadaver samples of Clauser et al (1969) and Chandler et al (1975) are similar (Table 9.1b).

9:2.3 Elastic energy storage in the knee extensor muscles

The procedures used to calculate an elastic compliance function for the knee extensors have been described in Chapter 8. The compliance measurements described in section 8:6 were performed on a subgroup of 10 of the subjects of this study. Given an experimentally determined function for the elastic compliance-knee moment relationship of the knee extensors, the change in the energy stored in the stretched elastic component can be calculated as the work done against series elasticity. During a small time interval, the apparent extension of the series elastic component is given by the product dM_k.G(M_k) and the work done, equal to the change in energy stored, dE, is given by

\[ dE = M_k.G(M_k).dM_k \]

where \( G(M_k) \) is the angular equivalent compliance at a knee moment of \( M_k \). These calculations can be summed over a number of time intervals to calculate changes in strain energy storage.

Given that the compliance function of each subject is known, it is necessary to determine the net muscle moment acting at the knee throughout the running stride. The total muscle moment at the knee is the sum of the actions of all the muscles that cross it. To solve for the individual contributions of each muscle is a numerically indeterminate problem, and it is therefore necessary to make some assumptions regarding load-sharing and to optimise the solution (Chao and Rim, 1973; Seireg and Arkivar, 1975; Hardt, 1978). For the purposes of this study, a single net muscle moment about the knee was calculated. When the sign of this resultant represents an extension moment it is assumed to be entirely due to the action of an equivalent knee extensor muscle.
9.2.4 Calculation of inter-segmental forces and moments:

The net forces and moments acting at the ankles and knees resolved by analysis of the 15 segment model. The calculations are based on the application of D'Alembert's principle to the free body diagram of a generalised segment (Figure 9.1) and is restricted to the analysis of forces, moments and accelerations in the sagittal plane. Given the distal reaction forces and moments, the accelerations of the centre of mass and the inertial properties of the segment, the equations of motion of the segment can be solved for the reaction forces and moment at the proximal joint (notation as Figure 9.1):

(1) Summing forces in the horizontal direction:

\[ R_{x_0} = m_a x - R_{xa} \]  
\[ \text{Eqn 9.2} \]

(2) Summing forces in the vertical direction:

\[ R_{y_0} = m_a y + mg - R_{ya} \]  
\[ \text{Eqn 9.3} \]

(3) Taking moments about the segment centre of mass:

\[ M_o = I_a - M_a - R_{xa}(y-y_a) - R_{yd}(y-y_d) + R_{ya}(x-x_a) + R_{yd}(x-x_d) \]  
\[ \text{Eqn 9.4} \]

The reaction forces at the distal end of the next segment in the proximal direction are equal and opposite to those calculated at the proximal joint of the first segment. The proximal reactions of the next segment can therefore be calculated in the same way. On this basis, given the reaction forces at the extremities of the link-segmental model, and its kinematics, it is possible to calculate the forces and moments at any joint. This "link-segmental chain" approach to the analysis was implemented as a FORTRAN IV computer programme (Appendix H).

As far as treadmill running is concerned, when the body is not in contact with the ground or other surface, the reaction forces at all extremities are assumed to be zero and therefore known. During the support
phase however, the ground reaction is not known.
 Due to the inaccuracies in kinematic and anthropometric data, and the simplifications in the dynamic model, the calculations of joint forces and moments are generally more inaccurate the further the joint is from an extremity where the forces are known.

9.2.5 Estimation of ground reaction forces.

Since the experimental trials were performed on a treadmill in order to facilitate the collection of expired air, it was not possible to directly measure ground reaction forces during the support phase of the running stride. Without this or equivalent information, calculation of knee moments during the support phase are not possible. An estimate of the ground reaction forces in the sagittal plane was therefore made using the acceleration vector of the whole body centre of mass. Assuming the ground reaction force to be the only force acting on the body during the support phase:

\[ F_x = M \cdot \frac{d^2X}{dt^2} \quad \text{Eqn 9.5a} \]
\[ F_y = M \cdot \frac{d^2Y}{dt^2} \quad \text{Eqn 9.5b} \]

where \( F_x \) and \( F_y \) are the horizontal and vertical components of the ground reaction force, \( M \) is the mass of the whole body and \((X,Y)\) is the location of the whole body centre of mass. The whole body centre of mass coordinates were calculated by summation from segmental masses and centres of mass.

\[ X = \left[ \sum_{i=1}^{S} m_i x_i \right] / M \quad \text{Eqn 9.6a} \]
\[ Y = \left[ \sum_{i=1}^{S} m_i y_i \right] / M \quad \text{Eqn 9.6b} \]

where \((x_i, y_i)\) are segmental centre of mass coordinates of a \( s \)-segment model and \((m_i)\) are segmental masses.

Usually, these equations would be used to determine the dynamics of
the centre of mass of the body from force-platform records (e.g. Cavagna, 1975). The reverse procedure, used in this instance, is inherently more inaccurate since it is based on numerically differentiated noisy film data, a simplified model of the human body and estimates of the masses and inertial properties of the body segments.

A similar procedure for resolving joint forces and moments (in this case for a 5-segment model) is described by Plagenhoef (1968).

Some assumptions must also be made about the centre of pressure of the ground reaction force in relation to the lower surface of the foot, since this represents the effective point of application of the reaction force vector. Cavanagh and Lafortune (1980) record ground reaction forces and centre of pressure patterns for a group of athletes running at the same speed as that used in this study. Some of these subjects were variously classified as "fore-foot strikers", "mid-foot strikers" or "rear-foot strikers", depending on which part of the foot first made contact with the ground. The adaptations to treadmill running are such that, almost without exception, athletes first make ground contact with the rear-part of the foot. The movement of the centre of pressure of Cavanagh and Lafortune's "rear-foot strikers" during the support phase followed a characteristic pattern. For the purposes of this study, this pattern was modelled by a simple quadratic function (Figure 9.2) relating the distance of the centre of pressure from the heel marker as a proportion of the length of the foot to the duration of the support phase:

\[ X_c = X_r + (X_f - X_r) \cdot \sqrt{\frac{t}{T}} \]

**Eqn 9.7**

where \( X_c \) is the estimated horizontal position of the centre of pressure

\( X_r, X_f \) are the horizontal positions of the rear-foot and fore-foot markers respectively.

\( t \) is the time elapsed in the support phase

\( T \) is the total duration of the support phase.

Thus the centre of pressure was assumed to move rapidly towards the forefoot during the early support phase, progressing more slowly in the region of the metatarsal heads. Figure 9.3 is an example of the centre of pressure model applied to an actual data set from this study.

Thornton-Trump and Daher (1975) used a similar approach to calculate hip and knee joint reaction forces in walking subjects from
cinephotographic data but did not directly compare the results with measurements based on direct measurement of ground reaction forces. Lamb and Stothard (1978) compared body centre of mass velocities calculated from force-platform and cinephotographic records and found a high correlation (r=0.96) between values calculated from integration of force traces and from differentiation of cinephotographic data in conjunction with a segmental model. This suggests that an acceptable approximation of the low frequency components of the ground reaction force can be made from kinematic data.

9:2.6 Energy analysis

Mechanical energy transformations during the running stride were determined by analysis of the link-segmental model. The methods described in section 4:2.4 and Chapter 5 were used to calculate $W_n^+, W_w^+$ and $W_{wb}^+$. Energy transfers within and between segments ($T_w, T_b, T_{wb}$), assuming no constraints on between segment energy transfer were calculated using the generalised method described in Chapter 5. In the absence of constraints, the calculated values of within and between segment energy transfer represents the upper limit of the total possible energy exchange within the body.

The energy analysis was repeated with two additional energy components included in the model. These represented the strain energy stored in the left and right knee extensor equivalent muscle groups, calculated using equation 9.1, above. For convenience, the energy model was configured with the strain energy components as additional components of the thigh segments. Since free exchange of energy between segments was assumed, however, the actual location of the strain energy components within the energy model has no effect on the calculation of $W_n^+, W_{wb}^+$, or $T_{wb}$. 
Figure 9.2: Assumed relationship between contact time and centre of pressure position during the support phase of running.
Figure 9.3: Example of centre of pressure model applied to example cinephotographic data.
9:3 RESULTS

9:3.1 Respiratory function.

An example of the respiratory responses recorded during the steady state exercise protocol used in this experiment has been recorded in Section 6.9, Figure 6.7 and Table 6.1. As a further example, the accumulated energy expenditure data from the same subject is shown in Figure 9.4. The mean oxygen uptake for 14 subjects during the steady state run was 49.3 mls.kg\(^{-1}\).min\(^{-1}\) (+ 2.8 s.d.). On average this represented 77% of the subjects' VO\(_2\) max recorded during the incremental treadmill test. The VO\(_2\)max results from the incremental test have been recorded in Table 9.1. It should be noted that VO\(_2\)max recorded during a level running test is generally lower than that which would be recorded during an incremental test in which the gradient of the treadmill is increased.

The calculated rate of energy expenditure during the steady state running test and at rest are recorded in Table 9.2. The net energy expenditure values in Table 9.2 were calculated by subtraction of the resting value from the gross value.

With one exception, blood lactate concentrations rose during the six minute steady state run (Table 9.2). The average rise was small (1.75 mmol), suggesting that the source of energy during the run was predominantly aerobic. In no case was a gas exchange ratio of greater than one recorded. In the case of subject 4, blood lactate rose from a resting level of 0.66mmol to 4.15mmol during the course of the test. It is generally accepted that a blood lactate level above 4 mmol indicates a significant contribution to energy expenditure from anaerobic sources, and the data from this subject was therefore not included in the energy analyses. Post-exercise blood lactate concentrations were not significantly correlated with either gross or net "aerobic" energy expenditure (r = 0.238 and r = 0.229 respectively.). Nor was ventilation rate significantly correlated with either energy expenditure measurement (r=0.556 and 0.506 respectively for all 14 subjects).

9:3.2 Mechanical energy changes and energy expenditure.

Figure 9.5 illustrates a typical movement pattern recorded on film from this study. The diagram shows joint centre positions from one side of the body after digital filtering, normalisation to a 50Hz timebase by Newtonian interpolation and adjustment by galilean invariance principle to a static reference frame.
Figure 9.6 shows the changes in the total body (kinetic + potential energy) during the running stride. This data is from the 10 subjects who completed the whole experimental series and has been normalised to a standard stride duration. The calculated magnitudes of energy change and energy transfer within and between segments for 13 subjects are recorded in Table 9.3.

Net energy expenditure was highly correlated with energy change magnitudes (\(W_{n+}, r=0.834\); \(W_{w+}, r=0.861\); \(W_{wb+}, r=0.813\), Figure 9.7a). When net energy expenditure and \(W_{wb+}\) (the magnitude of changes in the whole body energy curve) were both corrected for bodyweight, the two were still significantly correlated (Figure 9.7b; \(r = 0.774, p < 0.01\)).

9:3.3 Strain energy storage.

The mean and standard error of the vertical (\(F_y\)) and antero-posterior (\(F_x\)) components of the estimated ground reaction force histories of 10 subjects is shown in Figure 9.8. Again, these data were normalised to a standard stride duration before averaging. The estimate of the vertical component is both qualitatively and quantitatively similar to those reported by Cavanagh and Lafortune for athletes running overground at the same speed, and qualitatively similar to that reported by Alexander and Jayes for a single subject running at 3.6 m.s\(^{-1}\). The mean of the estimated antero-posterior components differs slightly from that reported by Cavanagh and Lafortune (1980) however in two main respects. Firstly, the average magnitude of the estimated antero-posterior component is approximately \(2/3\) of the magnitude of that reported for overground running. Secondly, in some subjects there is a small negative component shortly after foot contact which is not recorded for overground running. This additional feature may be due to the forward position of the foot at contact, which is a characteristic of treadmill running (Figure 2.2).

The mean and standard error of the calculated net muscle moment (10 subjects) is shown in Figure 9.9. The support phase of the running stride was characterised by a large extension moment at the knee. In some cases, a small flexion moment of short duration was observed, just after foot contact. This small flexion moment is not normally observed in overground running and was associated with the small period of negative antero-posterior ground reaction. During the recovery phase, the net moment at the knee tends to first extend and later flex the knee joint. During the last part of the cycle, shortly before foot contact, the knee moment tends to extend the joint.
These knee moment data were combined with the knee extensor angular compliance-moment functions of each of ten subjects to estimate the energy stored in the series elastic component (SEC) of the equivalent knee extensor muscle (Figure 9.10). Estimated energy storage in the SEC was significant during the support phase. Stored energy increased during the first 100ms of support, reaching a mean peak value of 66 J (±44 J s.d.). This energy was lost from the SEC during the extension of the knee joint during latter part of the support phase. These changes are generally opposed to those that occur in the total body potential and kinetic energy during the stride cycle (Figure 9.6), suggesting the possibility of transfer between the body's potential and kinetic energy and strain energy components.

9:3.4 Energy analysis with strain energy components.

A second energy analysis of 10 subjects was carried out with the strain energy storage in knee extensor muscle equivalents included as two additional components. The mean total body energy curve with strain energy components included is shown in Figure 9.11. The magnitudes of energy changes and energy exchanges calculated with this model configuration are shown in Table 9.4. Student's t tests for correlated means show that there are significant differences between all the measured energy parameters shown in Tables 9.3 and 9.4 with the probability of the differences being due to chance less than 0.01 in each case. These data are compared in Figure 9.12. The inclusion of the knee extensor strain energy components in the kinematic energy model resulted in a statistically significant reduction in the magnitude of changes in whole body energy ($W_{wb*}$) by 70± 35 J s$^{-1}$ even though the sum of the absolute changes in the partitioned energy components ($W_{n*}$) was increased by 14%. These changes are reflected by the increased values for within segment ($T_w$), between segment ($T_b$) and hence total possible ($T_{wb}$) energy transfer rates.

Table 9.5 shows the mean values of the summed change magnitudes for all energy components in the model together with total energy exchange for each segment ($W_{n*}$), the sum of changes in the segmental total energy ($W_{n*}$) and within segment transfer rates ($T_w$) for each segment. As previously, these data are corrected for varying stride durations by expression as joules per second. Changes in the strain energy component are of comparable magnitude with potential and translational kinetic energy terms, and exceed changes in all recorded segmental rotational kinetic energy components.
9.3.5 Energy expenditure and energy transfers

In those subjects who completed the whole series of experiments, the ratio of \( \text{Wwb}^+ \) to energy expenditure was not significantly correlated with total energy transfer, \( \text{Twb} \) (\( r = -0.134 \)), but was correlated significantly with the magnitude of within-segment energy exchange (\( r = 0.79 \), \( p < 0.01 \); Figure 9.13a). This was also true when \( \text{Wwb}^+ \) was calculated with strain energy components included in the energy model (\( r = 0.666 \), \( p < 0.05 \); Figure 9.13b).
Table 9.2: Energy expenditure and lactate production during a six minute run at 4.47 m.s⁻¹.

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Table 9.3: Whole body (kinetic + potential) energy change magnitudes and energy exchange

All units are J s⁻¹

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Table 9.4: Whole body energy change magnitudes and energy exchange including knee extensor strain energy components

All units are J.s⁻¹

T is the results of a Student's t-test for correlated means between the data in this table and that in Table 9.3 (no strain energy components). \( P < 0.01 \) in all cases.

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Table 9.5: Summed changes in component energy levels during the running stride at 4.47 m.s\(^{-1}\)

Mean ± s.d for 10 subjects
All units are J.s\(^{-1}\)

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SEU

80.7

65.2
### CENTRALIC EXERTS MEDICINE TRED SYSTEM

**SAMPLE:** 3009 SECONDS

**TEST DETAILS**

- **DATE:** 21-OCT-92
- **TIME:** 03:51:40
- **SYNCH. DELAY TIME:** 360 MSECNS

**TEST NUMBER:** 2

**SUBJECT NAME:** JAN

**BODY WEIGHT:** 95.2

**WORKLOAD:** SPEED (KPH): > 10.0

**GRADIENT (DEG.):** > 0.0

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</tbody>
</table>

Figure 9.4: Example of accumulated energy expended during a six minute run at 4.47 m.s⁻¹.
Figure 9.5: Treadmill running at 4.47 m/s$^{-1}$
Figure 9.6: Total body (potential + kinetic) energy during the running stride.

(Mean + SEM for 10 subjects)
Figure 9.7: $W_{wb}^+$ and energy expenditure.

(a) uncorrected and

(b) corrected for subject body weight.

[Mean values from 3D study, Chapter 4]
Figure 9.8: Estimated ground reaction forces for the left foot.

(Mean ± SEM for 10 subjects.)
Figure 9.9: Net muscle moment at the left knee.

(Mean ± SEM for 10 subjects.)
Figure 9.10: Strain energy storage in the left knee extensor muscles during the running stride.

(Mean ± SEM for 10 subjects.)
Figure 9.11: Total energy of the body during a running stride, including strain energy components.

(Mean ± SEM for 10 subjects.)
Figure 9.12: Comparision of energy changes calculated
(a) without
and (b) including strain energy component for the knee extensors.
Figure 9.13: Mechanical energy changes per unit energy expenditure and within segment energy transfer rate
(a) potential and kinetic energy model
(b) model including knee extensor strain energy.
Figure 9.14: Energy change magnitudes recorded at a running speed of 4.47 m.s\(^{-1}\) (10 mph).
(a) 14 subjects from this study.
(b) 4 subjects from the three dimensional study recorded in Chapter 4.
9:4 DISCUSSION

Figure 9.14 compares the mean mechanical energy changes recorded in this study with those trials performed at the same running speed from three-dimensional analysis reported in Section 4:3.1. The mean values of mechanical energy change magnitudes were all greater in this study; \( W_n^+ \) by 12\%, \( W_w^+ \) by 11\% and \( W_{wb}^+ \) by 28\%. Differences between the mean values of energy expenditure recorded at running speeds of 4.47 m.s\(^{-1}\) in the two studies differed by 2\%. The mean values of net energy expenditure and \( W_{wb}^+ \) lie within the distribution of those from this study both when expressed in J.s\(^{-1}\) and when corrected for subject body weight (Figure 9.7). However, in view of the different methodologies used to gather the two sets of data, they have not been combined.

9:4.1 Mechanical energy changes and energy expenditure

When the results from the previous study were corrected for running speed and body weight, a low but significant correlation was found between \( W_{wb}^+ \) and net energy expenditure (Figure 4.6). In this study, however, where running speed was not an additional confounding factor, the relationships between mechanical energy changes and energy expenditure have been more clearly established as being independant of running speed and body weight. These results therefore add further weight to the suggestion that differences in the magnitude of energy changes during the running stride can account for some of the inter-individual variations in energy expenditure at a particular running speed.

In contrast to the previous study, no significant relationship was found between the magnitude of total possible energy transfers (\( T_{wb} \)) and the ratio of \( W_{wb}^+ \) to energy expenditure. However, as Figure 9.13 shows, there was a significant relationship when within segment transfer alone was considered. This again suggests that the magnitude of passive energy transfers contributes to the economy with which a movement is performed.

Strain energy storage in the knee extensors

On the basis of the models used in this study to calculate muscle compliance and joint moments during the support phase of running, significant magnitudes of strain energy storage in the knee extensor muscles during the running stride were recorded. Figure 9.15 shows the observed relationship between elastic energy storage, knee extension moment and knee flexion velocity. Estimated stored energy increased as the extensor muscles contracted eccentrically during "weight-reception" at the
Left knee, support phase, $4.47 \text{m.s}^{-1} \{n=30\}$

Figure 9.15: Strain energy storage, knee extension moment and knee flexion velocity during the support phase of running. (Mean of 10 subjects.)

[Size of filled circle is proportional to stored energy]
onset of the support phase. This energy is lost from the strain energy component during the subsequent concentric contraction of the muscle.

The changes in the strain energy component are significant when compared with potential and kinetic energy components of the energy model. In all cases, the magnitude of changes in strain energy were greater than those in any of the segmental rotational kinetic energy components, and of similar magnitude to the changes in the potential and translational kinetic energies of the heavier segments.

The effects of including an additional energy component of this magnitude in the energy analysis are considerable. The whole body energy curves calculated with and without the inclusion of strain energy components (Figures 9.6 and 9.11) are compared on Figure 9.16. Changes in the strain energy component (Figure 9.16a) correspond inversely to changes in the total potential and kinetic energy of the body during the support phase (Figure 9.16b) so that the amplitude of the energy curve is reduced when the strain energy components are included (Figure 9.16c). The most notable effect of this is that the calculated total energy of the body begins to increase from the moment of foot contact, rather than falling to a minimum value around mid-support before increasing.

Energy exchange calculations, assuming the possibility of free transfer between the elastic energy component and the potential and kinetic energy of the rest of the body are considerably changed by the possibility of exchange with this additional energy component (Table 9.5). The magnitude of total energy exchange is increased, but the net magnitude of energy changes ($W_{exch}$) is reduced due to the increase in energy transfer.

The results reported here suggest that elastic energy storage in the anti-gravity muscle groups may act as a mechanical energy conserving mechanism during the support phase of the running stride. If potential and kinetic energy changes alone are considered, then a quantity of the body's energy must be dissipated (and therefore lost as heat) during weight-reception at the onset of the support phase. The muscle must then do work to reestablish the body's higher level of mechanical energy during the subsequent extension. If however, this "excess energy" at the onset of the support phase is conserved in the elastic components of eccentrically contracting muscles that are acting to resist the downward motion of the body during weight reception, the mechanical work required during subsequent extension is reduced in magnitude.

Of equal importance, such a mechanism would also act to reduce the rate at which the work must be done. There is evidence to suggest that the
Figure 9.16: (A) Estimated strain energy storage in the left knee extensor muscles. (Mean ± SEM for 10 subjects.)

(B) Total energy of the body excluding knee extensor strain energy components. (Mean ± SEM for 10 subjects.)

(C) Total energy including and excluding strain energy components for the left and right knee extensors.
speed and work rate at which a given task is performed effect the efficiency with which the work is done. Gaesser and Brooks (1975), conclude from their experiments on work output and energy expenditure during cycling activities, that the effective efficiency of the muscle either remains constant or falls as work rate increased. Similarly, efficiency was found to fall considerably as the pedalling speed at a particular work rate was increased.

Also, the energy exchange calculations are based on the assumption that energy stored in the elastic components can be fully recovered upon 'recoil'. Some of the stored energy will be lost due to stress relaxation in the elastic components, but this is likely to be small during the short time periods involved here (Aruin et al., 1979; See Figure 8.17, Section 8.7.6). Alternatively, relaxation of the contractile component itself may result in the dissipation of the stored energy. This mechanism however, involves the assumption that the losses in the total kinetic and potential energy during weight reception are dissipated as heat. Without more detailed calorimetric measurements, it is not possible to determine whether the strain energy is regained as useful energy or lost as heat. The evidence of the enhanced performance of jumping tasks when muscle stretching pre-movements are permitted (Asmussen et al., 1974; Thys et al., 1972) suggests that at least a proportion of the energy is usefully recovered.

While these results clearly suggest a significant contribution by series elasticity to energy transformations in the body, it must be noted that the strain energy values have been calculated from compliance measurements based on a simplified model of muscle elasticity and on knee moment calculations during the support phase based on calculated rather than directly measured ground reaction force and centre of pressure data. Thus the absolute magnitudes of the strain energy values recorded here may be questioned.

However, it must be assumed that the calculated values of elastic energy storage are many times greater in magnitude than the "real" values if the elastic behaviour of muscle is to lose significance in relation to changes in the rotational component of kinetic energy and to potential and kinetic energy changes in the lighter segments (Table 9.5). Also it must be borne in mind that the data presented here are from one group of antigravity muscles only. During the support phase of running, the function of the hip extensors and plantar flexors are, in general terms, similar to that of the knee extensors; the eccentric contraction of the
weight reception phase being followed by concentric contraction as the leg extends. Co-contraction of extensor muscle groups and others during the support phase is readily demonstrated by examination of EMG patterns (Brandell, 1973; Elliot and Blanksby, 1979a, 1979b). Each of these muscle groups is potentially able to behave elastically in the same way as the knee extensors. Therefore, it is to be expected that the actual total level of strain energy storage is considerably greater than that in the knee extensors alone.

Even at magnitudes much less that those suggested by the data reported here, the strain energy components would significantly change the qualitative pattern of the whole body energy curve. Measurements of "mechanical work", "power output" and "efficiency" based on whole body energy calculations including only potential and kinetic energies are likely to be unreliable, therefore. Most importantly, if possible energy conservation by elastic effects is ignored, then the amount of work done by muscle and hence muscle efficiency are likely to be overestimated. The addition of strain energy components into the energy model significantly reduced the mean ratio of $W_{me}^+$ to energy expenditure from $0.43 \pm 0.06$ s.d to $0.34 \pm 0.07$ ($t=6.81, p \ll 0.001$) which suggests that the inclusion of strain energy components in the energy model of human gait may bring about the calculation of smaller "efficiency" values, more in line with those recorded in isolated muscle.

Although elastic energy storage appears to act as an energy conserving mechanism, the magnitude of energy storage in the knee extensors was not significantly correlated with net energy expenditure, nor with the change in the value of $W_{me}^+$ brought about by the addition of strain energy components to the energy model. A significant correlation ($r = 0.623, p < 0.05$) was found between muscle elastic stiffness at a knee moment 150 Nm and net energy expenditure per unit body weight, but this and other high correlations between individual compliance values and energy expenditure in the data reported here are heavily influenced by the extreme compliance values recorded in one subject at higher muscle moments (See Figure 8.14, Section 8.6). Certainly possible relationships of this type merit further investigation.
9:4.3 Conclusions

In conclusion, the results of this study provide further evidence to suggest that some of the inter-individual variation in energy expenditure when running at a given speed are due to differences in "running style", as reflected in the magnitudes of energy changes in the body during the running stride. Additional evidence to suggest a role for passive energy transfer magnitudes in the determining movement economy was also found.

Estimates of energy storage in the elastic components of the knee extensor muscles suggest that strain energy stored in this way can make a significant contribution to overall energy exchange in the body. "Mechanical efficiency" estimates based on potential and kinetic energies alone should be treated with caution, therefore. Assuming, as the evidence suggests, that a proportion of this energy can be recovered, elastic energy storage acts as an energy conserving mechanism and also to reduce the rate at which the muscle must perform work in concentric contractions following eccentric contractions.
Chapter 10

GENERAL DISCUSSION
**Summary**

The results of this study suggest that variations in the "running economy" of athletes may be partly accounted for by biomechanical differences. Greater economy (lower energy expenditure) is associated with body energy curves of smaller amplitude and a greater degree of passive energy exchange.

Elastic energy storage appears to be a significant "energy saving" mechanism, which can also account for the unusually high "efficiencies" that have previously been reported for running activities.

**Energy Transformations in the Human Body**

Figure 10.1 illustrates in schematic form the energy transformations in running man with which this study has been primarily concerned. During exercise, the main flow of energy in the human body is from the chemical energy of metabolic substrates to mechanical work and heat; "movement" in energetic terms being characterised by large changes in the mechanical energy of the body parts.

The chemical energy released by muscle contraction is depicted as flowing to a strain energy component. This is a convenient way of describing the relationship between muscle contraction and movement. The first effect of contraction is to take up slack in and stretch the myofibrils. If the contraction results in movement then the energy becomes visible in the form of kinetic and gravitational potential energy. Otherwise, in an isometric contraction, the energy is taken up in stretching elastic structures within and surrounding the muscle and is dissipated as heat on relaxation.

"Movement" is depicted as involving interaction between potential, kinetic and strain energy. Work may be done on some external system, resulting in a net energy output from the body. Similarly, external forces may act on the body, resulting in an energy input into the system.

In this simple model, the energy degraded to heat by muscle relaxation and friction at joint is depicted as a heat loss, lumped with the heat generated by metabolism.
FIGURE 10.1 Some energy transformations in running man
ENERGY EXPENDITURE CALCULATIONS

For the purposes of this study, the chemical energy released via the aerobic pathway has been estimated from expired air analysis, with every litre of oxygen consumed by a human being equivalent to between 19.6 and 21.1 kJ of energy depending on the relative proportions of fat and carbohydrates acting as substrates. The contribution of anaerobic metabolism to the energy supply cannot be directly quantified. Analysis of the anaerobic component of energy expenditure has therefore been restricted to attempts to determine whether or not it is present, in order to confirm or otherwise the validity of energy expenditure calculations from the aerobic component alone. This limitation on energy expenditure measurements restricts analyses of the type described here to the lower half of the human function range.

In Figure 10.1, a quantity of the energy released from "chemical" sources is shown as being "lost" as heat. Included in these "losses" is the energy consumed by basal metabolism and life-support functions, which go on whether the body is exercising or not. In chapters 4 and 9, "Net energy expenditure" was calculated by subtracting the energy expenditure determined while the subject was at rest from the value determined during exercise. This subtraction assumes that the energy required to support the body's systems remains constant during exercise. This is clearly not the case. As exercise intensity increases, the energy consumed by skeletal muscles increases, but at the same time the "support cost" also increases. There is an additional energy requirement imposed by the higher ventilation rates and heart rates, for example.

The quantity of energy required to maintain basal functions will vary considerably between the onset of exercise and full post-exercise recovery. The base-line for net energy expenditure calculations at any given time is therefore not known. Comroe reports the work done in breathing as only 0.8W at rest, increasing to over 40W at ventilation rates of 200 L.min⁻¹ (Comroe, 1974, pp 140-141). In this study, energy expenditure at a given running speed was not significantly correlated with ventilation rate, suggesting that inter-individual differences in "running economy" cannot be explained simply in terms of ventilatory efficiency.

In a group of athletes running at the same, moderate running speed, errors in the calculation of net energy expenditure are likely to be of similar magnitude in each subject, and small when compared with the gross energy expenditure value.
However, at any particular running speed, the estimate of basal metabolism from resting measurements is likely to be too small. The energy used by working muscles will therefore be overestimated and the apparent efficiency of the generation of mechanical work from chemical energy will therefore be underestimated.

10:3 MECHANICAL WORK AND THE "EFFICIENCY" OF MOVEMENT.

J.P. Stapp described the problem confronting the would-be analyst of human body mechanics:

"This fifty litre rawhide bag of gas, juices, jellies, gristle and threads moveably suspended on more than 200 bones presided over by a cranium and worst of all living, presents a challenge to discourage a computer into incoherence"

Stapp (1971)

Not surprisingly, any analysis or model applied to the human body can be shown to be inadequate. Those who have attempted to analyse energy and "efficiency" seem to have been particularly afflicted, however. To calculate the kinetic and potential energies of the body's moving parts is comparatively straightforward. Although the calculations are based on an inanimate rigid body approximation to the human body there is no fundamental reason why the approximation should not be a good one. Work and energy are equivalent quantities and the energy changes involved in moving the body's parts are measurable. Yet there has been no consistent estimate of the work being done by the muscles as a man moves.

Attempts to determine the work output and efficiency of muscle from biomechanical analyses of movement have led to a wide range of methods and an equal diversity of work and efficiency values. In general, the methods used to determine the work done by the whole body examine, implicitly or explicitly, the changes in an energy curve. The energy curve has been calculated in a number of ways including, for example, from the kinematics of the centre of mass or from the sum of segmental energies.

When such methods are applied to a cyclical gait pattern on level terrain, an immediate problem arises in that the sum of the changes in any energy curve over an idealised gait cycle will always be zero, since the system starts and ends the cycle in the same configuration. While this
result may be mathematically correct in that no net external work is done and all energy expended by the athlete is dissipated as heat, its usefulness is strictly limited. The tendency, therefore, is to examine the "internal work" done within the gait cycle and to distinguish between positive and negative changes in the energy curve in order to achieve a non-zero result. Thus the "pseudo work" methods of Norman et al (1976), Winter (1979) and Pierrynowski et al (1980) sum the absolute values of segmental energy changes. The "efficiencies" calculated by these methods are, however, unrealistic when compared with the known mechanical efficiency of muscle preparations in vitro.

Over a given gait cycle the positive and negative changes in the energy curve and hence "positive work" and "negative work" are equal and oppositely signed, with a magnitude equal to one half of the pseudo work total. Pierrynowski et al (1980) suggest that "A more realistic assessment of [efficiency] can be realised if we assume different efficiencies for positive and negative work". Pierrynowski assumed that the muscle performs "negative work" with two or three times the efficiency of "positive work", and hence calculates separate efficiencies for the positive and negative phases. Williams and Cavanagh (1983) have demonstrated that the assumption of relative positive and negative work efficiencies is critical to the final outcome. It is clear that muscles use less energy in resisting an external force of a given magnitude than they do when applying a force of the same magnitude. Under eccentric contraction conditions, human muscle can produce much higher forces than during concentric or isometric contractions (Komi, 1973). This may be due to the assistance of passive viscosity in resisting the external force. Similarly, the external force may play a role in assisting the recovery of cross-bridges, hence reducing energy requirement. For cycling exercise, Abbott et al (1952) found the oxygen cost of positive work to be between 2.4 and 5.2 times that of negative work, depending on the cycling speed. The assumption of a single simple ratio for the correspondence between positive and negative work efficiencies is not supported, therefore.

That there is some variation in the methods used to calculate "work" is self evident. Similar possibilities exist for the determination of Energy expenditure, in the form of various baseline activities. Given some film of a runner and some oxygen uptake measurements it would theoretically be possible to calculate over 250 different efficiency values for the movement pattern using different combinations of work measurements, energy expenditure measurements and other assumptions described in the recent
literature.

Overall, it would appear unlikely that it will become possible to make definitive statements about muscular work and muscle efficiency with the kinematics-based mechanical energy models currently available. This is not to say that the energy analyses of this type should not be used, nor that they cannot contribute to our understanding of human movement. On the contrary, the link-segmental energy analysis gives a far more accurate reflection of the energy changes in the body than any of the methods that have preceeded it. The model includes individual segmental component energies, including the rotational component of kinetic energy, which have often been ignored.

Clearly, however, the link-segmental energy analysis should be used carefully. The purpose of most previous energy analyses of human motion has been to calculate some value for the "mechanical work rate", "power" output or "efficiency" of movement. In the absence of a clearly defined and complete model of all the energy transformations in the human body and some means of quantifying them, such measurements are not possible.

Ultimately, these attempts to calculate a value for "muscle efficiency" may be futile, since the mechanisms underlying a particular mode of human movement are unlikely to be well described by a single number. In any case, the mechanical efficiency of muscle is unlikely to be rigourously defined by a single value. Work and heat generation in the active muscle vary with muscle tension, muscle length and the velocity of contraction of the muscle (Bendall, 1962, pp 149-162). During an isometric contraction for example, only heat and no external work is generated, giving a work efficiency of zero. The "efficiency" with which the muscle does work is, therefore, a variable quantity.

The failure of currently used potential-kinetic energy models to yield satisfactory values for the efficiency of muscle is itself revealing.

On average, only about 25% of the energy released in skeletal muscle is converted into mechanical energy, the rest being lost as heat. Since energy can be neither created nor destroyed, any efficiency value for the transformation of chemical energy into some other energy form cannot exceed this basic overall efficiency of muscle. As has been shown (Chapters 2 and 4) calculations of muscle efficiency from biomechanical analyses of running humans are generally well in excess of this value. If all the positive increments in all measureable potential and kinetic energy components of a running man are summed, the total rate of change in a running man is greater than the measurable energy expenditure rate. Thus, if attributed to
the action of the muscles (as implied in the calculation of "Total Pseudo-Work", Norman et al., 1976), an efficiency in excess of 100% is derived. Obviously, this is because some of the movement brought about by energy exchange between kinetic and potential energy and by passive transfer of momentum (and hence energy) between segments have been included in the work total.

There is evidence for passive energy transfer within a segment during movement. Human limbs behave no differently from inanimate objects under the influence of gravity in this respect. When within segment energy exchange is subtracted from the pseudowork total, however, the summed energy changes are still high with respect to the rate of chemical energy expenditure implied in energy expenditure measurements and if attributed to muscular work, would imply a muscular efficiency of approximately 100%.

As has been previously discussed the calculation of the passive energy exchange between segments presents a number of problems. The assumptions concerning which between segment energy transfers are allowable will obviously have a considerable effect on estimates of the mechanical work done by muscles during a running stride. Williams and Cavanagh (1983) suggest the application of a factor to the "mechanical power" calculation, representing the fraction of the total positive work attributable to between segment energy transfer. Their comparison of four different factors is similar to the four different energy transfer conditions considered in this study, and no conclusion about the "real" contribution of between segment energy transfer can be drawn. Varying the restrictions on energy transfer can have a significant influence on the outcome of an energy analysis applied to running. The large differences between calculated values of $W_n^+$ (assuming no energy transfer) and $W_{wb}^+$ (assuming all possible within and between segment energy transfers) are evidence of this. While analytical procedures of the type proposed in Chapter 5 of this study allow the analysis of a clearly defined energy model with constrained energy exchange, they do not assist in the development of the energy model itself. The accuracy of any between segment energy transfer computation is therefore questionable, since the selection of transfer constraints is largely arbitrary.

Despite the limitations on analysis of between segment energy transfer, it is possible to calculate the upper bound of the possible potential and kinetic energy exchange. If calculations are made with a model allowing energy to be exchanged freely between any energy component (as implied in the calculation of the $W_{wb}^+$) then all possible passive
transfers of potential and kinetic energy within and between body segments are eliminated from the energy change total. \( T_{wb} \) therefore represents the maximum possible magnitude of passive potential-kinetic energy exchange, and \( W_{wb}^+ \) the minimum magnitude of energy changes that must be attributed to muscular activity if only potential and kinetic energy changes are considered. It can be shown however, that the calculation of \( W_{wb}^+ \) can result in reciprocal motions of limbs and body segments, brought about by muscle action, being attributed to passive energy transfer. In reality, some constraint should be applied to between segment energy transfer, and the calculation of \( W_{wb}^+ \) therefore underestimates the energy changes that must be attributed to muscular effort in the potential-kinetic energy model.

In Chapter 9 of this study, measurements of ten athletes running at 4.47 m.s\(^{-1}\) gave average values of 427 J/s and 964 J/s \( W_{wb}^+ \) and net energy expenditure respectively. These calculations imply an efficiency for muscle of approximately 44\%, if the changes in the energy curve measured by the calculation of \( W_{wb}^+ \) are all attributed to muscular work. Furthermore, it has already been noted that the net energy expenditure calculated as the difference between active and resting energy expenditure measurements is probably an overestimate of the energy consumption of the muscles alone. Similarly, \( W_{wb}^+ \) is an underestimate of the increments in the total body energy that cannot be accounted for by the action of passive potential-kinetic energy transfers. Thus the apparent muscular efficiency of 44\%, (which is already much higher than the range of values calculated from direct observation of muscle tissue) must be considered to underestimate the efficiency of muscle implied in the link segmental analysis of potential and kinetic energies.

Ironically, while the competitive athlete or coach may be concerned with improving the efficiency of the running style, the theoretical problem posed here is how to reduce the calculated mechanical efficiency of runners in order to reconcile the mechanical energy analysis with the known energetic capacity of muscle tissue.

Assuming that all passive potential and kinetic energy exchange is accounted for by the methods used here, the possibility that there are other energy components involved in energy exchange processes must be investigated. If such energy components exist, then it is possible that there are additional passive energy exchanges which have not been accounted for.
Muscle tissue has known elastic properties. Measurements of the elastic compliance of the knee extensors of a group of athletes (Chapter 8) has shown that this muscle group alone has a substantial capacity for the storage of strain energy. While these measurements are based on a simple model of muscle, the results obtained are quantitatively similar to those reported for other muscle groups, in the range of muscle tensions common to both (Figure 8.15). Also, although the model is highly simplified, it predicts the qualitative effect that high added series compliances result in oscillation of the muscle-spring system (Figures 8.6 and 8.10).

Chapter 9 describes a study in which compliance data for the knee extensors were included in energy analyses of running athletes. The energy stored in the knee extensors during the support phase of the running stride was found to be significant. Again, the calculation was based on a simplified model. Ground reaction forces during the support phase were calculated from centre of mass dynamics rather than measured directly. These calculations gave results which were both qualitatively and quantitatively similar to direct measurements reported in the literature, however. In any case, the estimate of elastic energy storage would have to be many orders of magnitude too large in order to become insignificant with respect to the magnitudes of the smaller segmental potential and kinetic energy components during the same activity.

If the strain energy components calculated for the knee extensors are included in the energy analysis, the magnitude of changes in the total body energy curve are reduced. The main implication of this effect for measurements of "mechanical work" is that the energy changes attributable to muscular action are reduced. When only knee extensor elasticity was included in the analysis, the average apparent net aerobic efficiency of ten runners was reduced from 43% to 34%.

This suggests that muscle elasticity can act as an energy transfer mechanism of the type required to reconcile measurements of the apparent efficiency of movement with that of isolated muscle. Indeed, muscle elasticity probably represents the only energy system capable of accounting for the discrepancy. The inclusion of the elastic properties of only one extensor muscle group in the analysis reduces the apparent efficiency substantially, by attributing to passive elastic energy exchange energy changes that have previously been attributed to muscular work.

If other extensor groups in the lower body have similar energy storage
capacities, then an analysis of the type used in Chapter 9 would result in an apparent efficiency of similar magnitude to that measured in isolated muscle and in other activities, such as cycling, where the opportunity for high magnitude mechanical energy transfers is limited.

Given the substantial contribution that muscle elasticity can make to mechanical energy transformations during dynamic gross activities such as running, measurements of "mechanical work" and "efficiency" based on potential and kinetic energy measurements alone should be treated with extreme caution.

10:5 **THE ROLE OF SERIES ELASTICITY**

In the energy analysis of runners recorded in Chapter 9, the inclusion of strain energy components in the energy model reduced the magnitude of energy changes attributable to muscle action. Thus muscle elasticity appears to act as an energy conserving mechanism. Instead of the body losing a substantial quantity of its mechanical energy during the first part of the support phase, the total (potential + kinetic + elastic) energy of the body begins to increase from the moment of foot contact. The extensor muscle of the legs are contracting vigorously at this stage, resisting the downward motion of the body under gravity. These eccentric muscle contractions appear to result in some of the energy that has to be dissipated being taken up by the elastic components. Concentric contraction of the extensor group rapidly follows the eccentric contraction, and recoil of the stretched series elastic components of the extensor muscles would assist in this action, effectively restoring the stored energy to the moving parts of the body. An important effect of this behaviour is that the amount of work that must be done by the muscles is reduced when a concentric contraction follows an eccentric contraction. In this sense, running appears to be similar to other rebounding activities where elastic energy storage has been shown to enhance both the effectiveness and the efficiency of movement (Thys et al, 1972; Asmussen and Bonde-Petersen, 1974).

Perhaps of equal importance is the observation that, during the support phase of running, elastic energy storage has the effect of reducing the rate at which the muscles must work during leg extension.

Also, the overall contraction rate of a muscle contractile component combined with a stretched elastic component is greater than that of the contractile component alone. The effective force-velocity function of the
muscle is shifted in favour of increased force at a given velocity, therefore.

10:6 **RUNNING ECONOMY**

In both the studies of running biomechanics reported here (Chapters 4 and 9), evidence was found of a relationship between the relative oxygen cost of running at a particular speed ("running economy") and mechanical energy parameters.

In both cases, when the effects of running speed and bodyweight were eliminated energy expenditure was significantly correlated with the magnitude of changes in the whole body total (potential + kinetic) energy during the running stride. Thus it would appear that the more economical runners are saving muscular effort by minimising fluctuations in the body’s energy.

Similarly, in both studies, the ratio of energy expenditure to \( W_{\text{wb}} \), considered by some to be a measure of "positive work efficiency", was significantly correlated with a measure of passive energy exchange. Use data collected over a range of speeds, this ratio was significantly correlated with total within and between segment energy exchange (\( T_w \)) assuming unrestricted transfer. In the study conducted at one speed \( \gamma \) (Chapter 9), the ratio was highly and significantly correlated with the magnitude of within segment energy exchange alone. This discrepancy can probably be explained by the variety of running speeds employed in the first study, since within-segment energy exchange remains fairly constant over a range of running speeds. Correction for variations in running speed is therefore likely to mask any link between within-segment energy exchange and other measurements. However, in a previous report of part of the presented in Chapter 4 (Shorten et al, 1981) energy expenditure was significantly correlated with relative within-segment energy transfer, \( T_w \), expressed as a percentage of \( W_n \); \( r=-0.48, p<0.05 \).

Two interactions between mechanical energy changes and energy expenditure have been observed, therefore. Firstly, a greater magnitude of changes in the body’s energy is associated with higher energy expenditure. Secondly, passive energy exchange tends to reduce the energy requirement.

Elastic energy storage in extensor muscle groups represents a further possible "energy saving" mechanism. While it has been shown that the general effect of muscle elasticity is to reduce the muscular work
requirement, no relationships between the magnitude of elastic energy storage in the knee extensors and energy expenditure were found. A significant correlation was found between a measure of muscle elastic stiffness and energy expenditure (Section 9:4.2) but this was not considered reliable due to the abnormal distribution of the data. The lack of findings does not eliminate the possibility that the magnitude of elastic energy storage in the support phase is also a determinant of running economy, however, since this analysis has been restricted to one muscle group only, and has been limited to a small subject group.

In broad terms, the energy cost of a particular work task can be considered as being dependent upon the amount of work done in performing the task and on the efficiency with which chemical energy in muscle is converted to work. The energy expenditure for a given performance requirement may therefore be reduced in two ways; firstly by an increase in the metabolic energy conversion efficiency and secondly by a reduction in the mechanical work required to complete the task. As far as more "economic" runners are concerned, the observed relationships between the magnitude of intra-stride energy changes in the total body energy curve, energy exchange and energy expenditure during running are consistent with the second of these mechanisms.

10:7 CONCLUSIONS

The main experimental findings of this study are that some of the interindividual variance in the energy expenditure of athletes running at the same speed can be accounted for by differences in the gait pattern; that energy storage in the elastic components of leg extensor muscles is significant and that muscle elasticity appears to act as an energy conservation mechanism during the support phase. While these experimental results may reveal a little of the mechanism underlying successful performance on the running track, their direct application to the athlete is less easily seen. An athletics coach might interpret these results in terms of "relaxation", "range of motion" or "smoothness". Certainly lack of flexibility may limit the extent of passive energy exchange and movements involving rapid changes in mechanical energy may increase the physiological energy demand. But how does one instruct an athlete to "maximise within and between segment energy exchange" or to "reduce the magnitude of fluctuations in the whole body energy curve"?
An eminent coach, Geoff Dyson, once wrote:

"Speaking generally, all athletes are best left unaware of the exact nature of their movements and need only sufficient detail to correct faults, satisfy curiosity and inspire confidence."

(Dyson, 1962)

Perhaps, as MacFadden says:

"The various members of the body should not flop and fly about aimlessly ..."
Over a large part of the normal range, the body-weight adjusted oxygen cost of running is a rectilinear function of running speed. At any particular running speed, however, there is some inter-individual variation in oxygen consumption which, it has been suggested, may be due to differences in "running style". If this is the case, then the "economy" of a running style, may be a significant determinant of athletic performance.

Recently, the mechanical work done by walking and running humans has been determined by examination of the changes in the kinetic and potential energies of body segments, the magnitude of these changes being used as an estimate of the work done (Norman et al, 1975; Winter, 1979; Pierrynowski et al, 1980; Williams and Cavanagh, 1983).

The purpose of this study was to examine the relationships between mechanical energy transformations in the body and the oxygen cost of the running gait.

The relationship between the oxygen cost of running at submaximal speeds and mechanical energy changes was investigated in a group of trained athletes. Subjects were filmed in two planes while running on a treadmill at speeds from 3.6 to 5.8 m.s\(^{-1}\). Segmental potential and kinetic energy changes were determined using a three-dimensional fifteen-segment rigid body model of the human body. Intra-stride changes in the mechanical energy of the whole body were determined with no allowance for energy transfer and with various energy transfer constraints imposed on the model. Oxygen consumption was determined by expired air analysis and used to estimate energy expenditure. For each energy transfer condition, net energy expenditure was more highly correlated with the magnitude of intra-stride energy changes than with running speed per se. The more economic running patterns were characterised by higher within-segment energy transfers.

It was also found that the calculation of the mechanical work done by the athletes is largely dependant on the choice of energy model. Using a range of previously published methods of calculating mechanical work, the nominal efficiencies for an athlete running at 4.47 m.s\(^{-1}\) varied between 0 and 120%. Even when all possible within and between segment energy transfers are accounted for, efficiency, estimated at 40%, is notably higher than the known overall efficiency of muscle. This confirms the suggestion that the energy models in common usage are either incomplete or poorly defined.

A generalised energy model is developed which allows constraints on
passive energy transfer to be varied and also allows the additional energy components to be included in the energy analysis. In particular the storage of energy in stretched muscle and connective tissue has not previously been included in the energy analysis.

The methods used to record energy expenditure were also reconsidered. Since expired air analysis only allows the calculation of energy released via aerobic metabolism, it is necessary to detect the onset of a significant contribution to energy expenditure via anaerobic, glycogenolytic pathways. A simple method of detecting the "anaerobic threshold" on the basis of inflections ("break-points") in the ventilation and gas exchange responses to incremental exercise (Wasserman et al, 1973) is in common usage. An automated system for the determination of breath by breath respiratory responses, based on a mass spectrometer linked to a laboratory computer was developed and used to examine more closely the significance of these inflections in ten subjects performing an incremental treadmill exercise test. Two-part and three-part multiple linear models were fitted to ventilation, carbon dioxide output and ventilatory equivalent for oxygen responses to an incremental treadmill running test by least squares regression. The goodness of fit of the models was compared. Approximations to the first and second derivatives of ventilation and gas exchange parameters were used to assist in the detection of inflection points in the response data.

The continuous model was found to provide an equally good or better representation of the respiratory responses than either the two-part or three-part linear model, and it was concluded that the method of determining the anaerobic threshold by applying multiple linear models to respiratory responses has not been adequately validated. Therefore, the onset of anaerobic metabolism was determined by measurement of blood lactate accumulation, which is attributable to anaerobic glycogenolysis.

In an attempt to estimate the magnitude of the contribution of elastic energy storage to the overall energy transformations in running, a method for the determination of the elastic compliance of the knee extensor muscles was developed; based on the measurement of the rise in tension at the onset of an isometric muscle contraction. The mean angular equivalent compliance of ten athletic male subjects was found to be a negative exponential function of knee extension moment. It was calculated that during a typical isometric contraction, some 70J of strain energy is stored
in the stretched muscle system. This magnitude of energy storage is significant when compared with the typical levels of segmental potential and kinetic energy storage during running. It is therefore suggested that elastic energy storage may make a significant contribution to energy exchanges during running, and may be of sufficient magnitude to account for the excessively high values of "efficiency" recorded in the literature.

A two-dimensional energy analysis of ten athletes running on a treadmill at 4.47 m.s\(^{-1}\) was performed, with a strain energy component for the knee extensors included in the energy model. Estimated energy storage in the elastic components of the knee extensors was found to be significant during the support phase of the running stride. Energy storage increased during the first 100 ms of support, reaching a mean peak value of 66 joules. This energy was lost from the strain energy components during the subsequent knee extension. The inclusion of the elastic components resulted in a significant reduction of the magnitude of changes in the whole body energy curve even though the sum of the absolute changes in the partitioned energy components increased by. These changes are reflected by increases in the within-segment and between-segment energy transfers. The apparent net aerobic efficiency was reduced from 43% to 34% by the inclusion of the strain energy component.

It was found that there is some correspondence between the magnitude of passive energy transfers and the "economy" of a running style. Lower energy expenditure was associated with a whole body energy curve of smaller magnitude and greater passive energy exchange. Elastic energy storage in muscle appears to act as an "energy saving" mechanism during the support phase of the running stride and may account for the abnormally high efficiencies that have been reported for the running gait.
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APPENDIX A

BBC BASIC to find the best-fitting 2-part and 3-part multiple linear models to a set of data

These routines are specific to a particular format of data. PROCreaddata which reads data into the arrays X & Y and sets the value of N to the number of XY pairs can be adjusted to suit any data format.

10 REM Programme to find best 2-part and 3-part multiple linear models.

40 MODE7
50 DIM X(100), Y(100)

70 PRINTCHR$141" Spot the Anaerobic Threshold"
80 PRINTCHR$141" Spot the Anaerobic Threshold"
90
100 PROCreaddata
110 PRINT"Sorting...";
120 PROCsort:PRINT"Sorted"
130 RMSMIN=9999:RMSMINK1=9999
140 FOR knot1=3 TO N-4
150 FOR knot2=knot1+3 TO N-3
160 PROCmodel(knot1,knot2)
170 IF RMS(RMSMIN THEN PROCmin
180 IF knot1=3 AND RMS<RMSMINK1 THEN PROCmink1
190 NEXT
200 NEXT
210
220 PROCsaveresults
225 PROCdisplayresults
230 REPEAT:UNTIL INKEY(-99)
240 MODE4
250 PROCplot
260 STOP
270
280 DEF PROCmodel(k1,k2)
290 REM Calculate best fitting model with knots at k1 and k2
300 @i:=&10
310 PRINT"Knots at ";k1;" ,";k2
320 @i:=&20309
330 residual=0
340 PROCregr(k1,k2)
350 A1=A:Bl=B
360 residual=residual+MS
370 PROCregrconstr(k1,k2,X(k1),A1+B1*X(k1))
380 A2=A:B2=B
390 IF B2(Bl AND k1<k2 THEN ENDPROC
400 residual=residual+MS
410 PROCregrconstr(k2,N,X(k2),A2+B2*X(k2))
420 A3=A:B3=B
430 RMS=SQR(residual/N)
440 PRINT;"RMS residual = ";RMS
450 PRINT
460 ENDPROC
470
480 DEF PROCregr(a,b)
485 REM Standard linear least squares regression.
490LOCAL i
500 n=b-a+1
510 sumx=0
520 sumy=0
530 sumxsq=0
540 sumysq=0
550 sumxy=0
560 FOR i=a TO b
570  sumx=sumx+x(i)
580  sumy=sumy+y(i)
590  sumxsq=sumxsq+x(i)*x(i)
600  sumysq=sumysq+y(i)*y(i)
610  sumxy=sumxy+x(i)*y(i)
620  NEXT
630  B=(n*sumxy - sumx*sumy)/(n*sumxsq - sumx*sumx)
640  A=sumy/n - B*sumx/n
650
660 REM residual
670 MS=0
680 FOR i=a TO b
690  x=X(i)
700  yreal=Y(i)
710  ycalc= A + B*x
720  MS=MS+(ycalc-yreal)^2
730  NEXT
740 PRINT A,B
750 ENDPROC
760
770 DEF PROCreaddata
780 INPUT "X Data file", file$ = chan=OPENIN(file$)
790 INPUT#chan,N
800 FOR i=1 TO N
810  INPUT#chan, X(i)
820  NEXT
830 CLOSE#chan
834 INPUT "Y Data file", file$ = chan=OPENIN(file$)
838 INPUT#chan,ndata
840 FOR i=1 TO N
850  INPUT#chan, Y(i)
860  NEXT
870 CLOSE#chan
880 ENDPROC
890
900 DEF PROCmin
902 REM Record best solution to L3 model so far.
910 IF B2<B1 OR B3>B2 THEN ENDPROC
920 VDU7
930 RMSMIN=RMS
940 A1min=A1
950 A2min=A2
960 A3min=A3
970 B1min=B1
980 B2min=B2
990 B3min=B3
1000 junc1=knot1
1010 junc2=knot2
1020 ENDPROC
1030
1040 DEF PROCmink1
1045 REM Record best solution to L2 model so far.
1050 IF B3<B2 THEN ENDPROC
1060 VDU7
1070 RMSMINK1=RMS
1090 A2mink1=A2
1100 A3mink1=A3
1110 B1mink1=B1
1120 B2mink1=B2
1130 B3mink1=B3
1140 junk1=knot2
1150 ENDPROC
1160
1170 DEF PROC displayresults
1180 CLS
1190 PRINTCHR$141 "Spot the Anaerobic Threshold"
1200 PRINTCHR$141 "Spot the Anaerobic Threshold"
1210 PRINT "FILE >" ;file$'
1220 PRINT "Single knot:"
1230 PRINT "RMSmin >" ;RMSMINK1
1240 PRINT "at >" ;junk1
1250 PRINT "Coeffs:"
1260 PRINT A2mi\nm, B2mi\nkl
1270 PRINT A3minkl, B3minkl
1280 PRINT "Two knots"
1290 PRINT "RMSmin >" ;RMSMIN
1300 PRINT "at >" ;junc1,junc2
1310 PRINT "Coeffs:"
1320 PRINT A1mi\nm, B1mi\n
1330 PRINT A2mi\nm, B2mi\n
1340 PRINT A3mi\nm, B3mi\n
1350 ENDPROC
1360
1370 DEF PROC plot
1380 FOR I=1 TO N
1390 x=X(I)*200
1400 y=Y(I)*5+100
1410 PLOT69,x,y
1420 NEXT
1430
1440 x=X(junc1)*200
1450 y=A1mi\nm+X(junc1)*B1mi\n
1460 y=y*5+100
1470 MOVE x,y
1480
1490 x=X(junc1)*200
1500 y=A1mi\nm+X(junc1)*B1mi\n
1510 y=y*5+100
1520 DRAW x,y
1530
1540 y=A2mi\nm+X(junc1)*B2mi\n
1550 y=y*5+100
1560 MOVE x,y
1570
1580 x=X(junc2)*200
1590 y=A2mi\nm+X(junc2)*B2mi\n
1600 y=y*5+100
1610 DRAW x,y
1620
1630 y=A3mi\nm+X(junc2)*B3mi\n
1640 y=y*5+100
1650 MOVE x,y
1660
1670 x=X(N)*200
1680 y=A3mi\nm+X(N)*B3mi\n
1690 y=y*5+100
1700 DRAW x,y
1710 ENDPROC
DEF PROC sort
If data has a VO₂ on the x-axis rather than time, input data may not
be in ascending order of x. This procedure sorts it out.

LOCAL C%, F%, G%, I%, T
G% = N
REPEAT
G% = G% DIV 2
C% = N - G%
REPEAT
F% = FALSE
FOR I% = 1 TO C%
F% = I% + G%
IF X(I%) > X(F%)
THEN T = X(I%):
X(I%) = X(F%):
X(F%) = T:
T = Y(I%):
Y(I%) = Y(F%):
Y(F%) = T:
F% = TRUE
NEXT
UNTIL F% = FALSE
UNTIL G% <= 1
ENDPROC

DEF PROC regr constr(a,b,xc,yc)
REM Linear least squares regression constrained to pass through
REM the point (xc, yc)
LOCAL i
n = b - a + 1
sumqsq = 0
sumqyc = 0
sumyc = 0
sumsq = 0
sumqyc = 0
sumx = 0
sumy = 0
FOR i = a TO b
x = X(i)
y = Y(i)
q = x - xc
sumx = sumx + x
sumy = sumy + y
sumsq = sumqsq + q * q
sumqyc = sumqyc + q * yc
sumyc = sumyc + y * q
NEXT
B = (sumyc - sumqyc) / sumsq
A = yc - B * xc
REM residual
MS = 0
FOR i = a TO b
x = X(i)
yreal = Y(i)
ycalc = A + B * x
MS = MS + (ycalc - yreal)^2
NEXT
PRINT A, B
ENDPROC
APPENDIX B

BBC BASIC programme to fit continuous model to respiratory function data.

This programme calculates a best fitting curve of the form

\[ y = U(1) + U(2) \exp(x) + U(3) \exp(-x) \]

given \( N \) \( xy \) data pairs

```basic
140 DIM X(100), Y(100)
150 DIM A(5,5), S(5), U(5)
160 DIM a(5)
170 DEF FNy(x) = U(1) + U(2) * EXP(x) + U(3) * EXP(-x)
180 MODE7
190 PRINT "Exponential regression on VE:VO2 data"
200 PROCreaddata
210 PRINT "Sorting ...";
220 PROCsort
230 PRINT "Sorted"
240 PROCsetupmatrices
250 PROCgaussian_elimination
260 PROCresidual
270 PROCprint
280 REPEAT: UNTIL INKEY(-99)
290 MODE4
300 PROCplot
310 STOP
320 REM ===============
330
340 DEF PROCreaddata
350 INPUT "X Filename", file$
360 chan=OPENIN(file$)
370 INPUT#chan, NDATA
380 FOR I=1 TO NDATA
390 INPUT#chan, X(I)
400 NEXT
410 CLOSE#chan
420 INPUT "Y Filename", file$
430 chan=OPENIN(file$)
440 INPUT#chan, ndata
450 FOR I=1 TO ndata
460 INPUT#chan, Y(I)
470 NEXT
480 CLOSE#chan
490 ENDPROC
500
510 DEF PROCsetupmatrices
520 FOR I=1 TO NDATA
530 x=X(I)
540 a(1)=1
550 a(2)=EXP(x)
560 a(3)=EXP(-x)
570 y=Y(I)
580 FOR i=1 TO 3
590 FOR j=1 TO 3
600 A(i,j)=A(i,j)+a(i)*a(j)
610 NEXT
620 NEXT
```

```
620 \[ S(i) = S(i) + y \cdot a(i) \]
630 \[ U(i) = 0 \]
640 \[ \text{NEXT} \]
650 \[ \text{NEXT} \]
660 \[ N=3 \]
670 \[ \text{ENDPROC} \]
680
690 \[ \text{DEF PROC residual} \]
700 \[ MS=0 \]
710 \[ \text{FOR } I=1 \text{ TO } NDATA \]
720 \[ x=X(I) \]
730 \[ yreal=Y(I) \]
740 \[ ycalc=FNy(x) \]
750 \[ MS=MS+(yreal-ycalc)^2 \]
760 \[ \text{NEXT} \]
770 \[ RMS=SGR(MS/NDATA) \]
780 \[ \text{ENDPROC} \]
790
800 \[ \text{DEF PROC print} \]
810 \[ \text{PRINT } "\text{FILE}:\"};file$ \]
820 \[ \text{PRINT } "a(1) ="};U(1) \]
830 \[ \text{PRINT } "a(2) ="};U(2) \]
840 \[ \text{PRINT } "a(3) ="};U(3) \]
850 \[ \text{PRINT } "\text{RMS =}"};RMS. \]
860 \[ \text{ENDPROC} \]
870
880 \[ \text{DEF PROC plot} \]
890 \[ \text{FOR } I=1 \text{ TO } NDATA \]
900 \[ x=X(I) \cdot 200 \]
910 \[ y=Y(I) \cdot 5+100 \]
920 \[ \text{PLOT69},x,y \]
930 \[ \text{NEXT} \]
940 \[ x=X(I) \]
950 \[ y=FNy(x) \]
960 \[ \text{MOVE } x \cdot 200,y \cdot 5+100 \]
970 \[ \text{FOR } I=2 \text{ TO } NDATA \]
980 \[ x=X(I) \]
990 \[ y=FNy(x) \]
1000 \[ \text{DRAW } x \cdot 200,y \cdot 5+100 \]
1010 \[ \text{NEXT} \]
1020 \[ \text{ENDPROC} \]
1030
1040 \[ \text{DEF PROC gaussian elimination} \]
1050 \[ \text{REM Solution of a set of } N \]
1060 \[ \text{REM linear equations by} \]
1070 \[ \text{REM Gaussian elimination.} \]
1080 \[ \text{REM ================} \]
1090
1100 \[ \text{REM } A(,) \text{ } N*N \text{ matrix of equations} \]
1110 \[ \text{REM } S() \text{ Solutions to equations} \]
1120 \[ \text{REM } U() \text{ Unknowns} \]
1130
1140
1150 \[ \text{REM 1: Subtract multiples of rows} \]
1160 \[ \text{REM from each other to introduce} \]
1170 \[ \text{REM zeros into the lower} \]
1180 \[ \text{REM triangular part of the matrix} \]
1190
1200 \[ \text{FOR } I1=1 \text{ TO } N-1 \]
1210 \[ M=I1 \]
1220 \[ w=A(I1,I1) \]
1230 \[ \text{FOR } I2=I1+1 \text{ TO } N \]
1240  IF ABS(A(I2,I1)) > W THEN W=ABS(A(I2,I1)):M=I2
1250  NEXT
1260
1270  IF M=I1 THEN 1380
1280
1290  FOR I2=I1 TO N
1300    W=A(I1,I2)
1310    A(I1,I2)=A(M,I2)
1320    A(M,I2)=W
1330  NEXT
1340  W=S(I1)
1350  S(I1)=S(M)
1360  S(M)=W
1370
1380  FOR I2=I1+1 TO N
1390    W=-A(I2,I1)/A(I1,I1)
1400  FOR I3=I1+1 TO N
1410      A(I2,I3)=A(I2,I3) + W*A(I1,I3)
1420  NEXT
1430  S(I2)=S(I2) + W*S(I1)
1440  NEXT
1450  NEXT
1460
1470 REM 2: Find U by back substitution
1480  U(N)=S(N)/A(N,N)
1490  FOR I1=N-1 TO 1 STEP -1
1500    U(I1)=S(I1)
1510  FOR I2=I1+1 TO N
1520    U(I1)=U(I1)-U(I2)*A(I1,I2)
1530  NEXT
1540  U(I1)=U(I1)/A(I1,I1)
1550  NEXT
1560 ENDPROC
1570
1580 DEF PROCsort
1590 LOCAL C%,F%,G%,I%,T
1600  G%=NDATA
1610 REPEAT
1620  G%=G% DIV 2
1630  C%=NDATA-G%
1640 REPEAT
1650  F%=FALSE
1660  FOR I%=1 TO C%
1670    P%=I%+G%
1680    IF X(I%) > X(P%)
1690      THEN T=X(I%):
1700      X(I%)=X(P%):
1710      X(P%)=T:
1720      T=Y(I%):
1730      Y(I%)=Y(P%):
1740      Y(P%)=T:
1750      F%=TRUE
1760  NEXT
1770 UNTIL F%=FALSE
1780 UNTIL G%<=1
1790 ENDPROC
APPENDIX C

FORTRAN IV subroutines defining the configuration of a two-dimensional 15 segment rigid-body model of the human body.

SUBROUTINE MODEL

LINK SEGMENTAL MODEL CONFIGURATION

Sets up configuration of link-segmental model of the human body, and is used to define the relationships between model parameters and the input data. This routine can be modified to set up a variety of link-segmental models.

NJNT, IPROXS, IDISTS

IPROXJ, IDISTJ

JNTNAM, SEGNAM

SNAM, MNAM

NFRAME, NPOINT

NLAG

DELTAT, VEL

MARK

IREFL, LON, LOFF

IRON, IROFF

The following data blocks define a model configuration...
DATA NSEG/15/ +NJNT/22/

DATA MNAM/' HS-GRMON','E 2/15- ','A /

DATA SEGnam/' HEAD ','THORAX ','ABDOMEN,'

DATA JNTNAM/' VERTEX ','NECK ','WAIST '

DATA IPROXJ/2,3,3,5,6,7,9,10,11,14,15,16,18,19,20,

DATA IDISTJ/-1,1,9,2,3,5,6,10,11,14,15,16,18,19,20,

DATA Prototype:
The model is configured by defining the joints at
the proximal & distal ends of each segment in
IPROXJ & IDISTJ and the segments proximal & distal
to each joint in IPROXJ. 'Proximal' &
'Distal' are with respect to a joint centre labeled
'origin'.

Special cases:
0 = origin central joint centre; if it exists
-1 = extremity
-9 = illegal
90...99 = special cases where the required joint or
segment cannot be uniquely defined.

DATA IPROXS/1,2,0,3,2,4,5,6,7,8,90,9,

DATA IDISTS/-1,1,9,2,3,5,6,10,11,14,15,9,9,9,9,9,9,9,

DATA IPROXJ/2,3,3,5,6,7,9,10,11,14,15,16,18,19,20,

DATA IDISTJ/1,9,2,3,5,6,7,8,10,11,13,15,16,17,19,20,22,

DATA Reflection characteristics

DATA IREFL/1/+LON/5/+LOFF/13/+IRON/14/+IROFF/22/

Direct correspondences between body landmarks
and model Joint centres.

DATA MARK/1,5,10,-9,-8,-7,-6,-9,11,12,13,15,14,

Model configuration:
The model is configured by defining the joints at
the proximal & distal ends of each segment in
IPROXJ & IDISTJ and the segments proximal & distal
to each joint in IPROXJ. 'Proximal' &
'Distal' are with respect to a joint centre labeled
'origin'.

Special cases:
0 = origin central joint centre; if it exists
-1 = extremity
-9 = illegal
90...99 = special cases where the required joint or
segment cannot be uniquely defined.
Display model identification
WRITE(7,100) (NAM(I), I=1,3), NSEG, NJNT

Get details for individual analysis
RETURN

100 FORMAT(' MODEL CONFIGURATION > ' , 3A8,
1     I4,' Segs.', I4,' Jnts.' )

END
APPENDIX D

FORTRAN IV programme and subroutines to calculate the inertial parameters of a link-segmental model of the human body based on individual anthropometric measurements.

PARAM() contains the set of anthropometric measurements listed in Table 3.3

Segmental inertial characteristics are calculated and stored in the COMMON block SEGDAT

BLOCK DATA contains regression equations and other data from cadaver studies (Clauser et al., 1969; Chandler et al., 1975)
READ(5,142,ERR=30) IHTYP
IF((IHTYP.NE.1).AND.(IHTYP.NE.2)) GOTO 30

C Compute segmental masses
BMSS=0.0
DO 10 ISEG=1,15
  WRITE(7,9001) SEGNAM(ISEG)
  SMASS(ISEG)=0.0
DO 12 J=1,3
  JJ=IPCMIJ,ISEG)
  XX=CVMJ,ISEG)
  YY=PARAH(JJ)
  SMMAG(ISEG)=SMMAG(ISEG)+YY*XX
  SMMAG(ISEG)=SMMAG(ISEG)+CVM4,ISEG)
  WRITE(7,902) CVM4,ISEG)
PAUSE 'FOR DEBUG . Press return to continue'
BMSS = BMSS+SMAG(ISEG)

C Compute segmental moments of inertia
DO 16 ISEG=1,15
  XX=1000*PARAM(1)*FM0IC(ISEG)+CM0IC(ISEG)
  SMOIC(ISEG) = XX/1000000000.0

C Display segmental data
WRITE(7,117)(SUBNIAM(I),I=1,3),PARAH(1),PARAH(2)/100
WRITE(7,118)
WRITE(7,120)
DO 14 I=1,15
  WRITE(7,122) I,SEGNAM(I),SMASS(I),SMASS(I)/BMSS*PARAH(1),
  SMOIC(I)
  SMMAG(I)=SMMAG(I)/BMSS*PARAH(1)
  WRITE(7,124) BMSS,PARAH1)

C Add shoe mass to mass of foot
SMAG(9)=SMAG(9)+PARAH(32)
SMAG(15)=SMAG(15)+PARAH(32)
WRITE(7,132) PARAH(32)

C Compute data for segment centre positions
N.B. a negative result indicates that a special
procedure applies,
DO 18 ISEG=1,15
  SCLN(ISEG)=0.0
DO 20 I=1,3
  JJ=IPCGY(J,ISEG)
  XX = CVCGY(J,ISEG)
  YY = PARAH(JJ)
  SCLN(ISEG)=SCLN(ISEG) + XX*YY
  SCLN(ISEG) = SCLN(ISEG)+CVCGY(4,ISEG)
  ILEN=ISELN(ILEN)
  XLEN=PARAH(ILEN)
  IF(ILEN.EQ.-1) XLEN = -1.0
SLEN(ISEG) = XLEN
SCLN(ISEG) = SCLN(ISEG)/XLEN
SCLN(1) = 0.464
SCLN(2) = 0.5
SCLN(3) = 0.5
SCLN(4) = 0.5
SCLN(5) = 0.5
SCLN(6) = 0.6
SCLN(7) = 0.6
SCLN(8) = 0.6
SCLN(9) = 0.6
SCLN(10) = 0.6
SCLN(11) = 0.6
SCLN(12) = 0.6
SCLN(13) = 0.6
SCLN(14) = 0.6
SCLN(15) = 0.6

IF(IHTYP.EQ.1) GOTO 19
C

APPROXIMATION OF MASS CENTRE OF FOLDED HAND
SCLN(6) = 0.67
SCLN(12) = 0.67

PAUSE 'Press RETURN to continue'
GOTO 21
C

APPROXIMATION OF MASS CENTRE OF OPEN HAND
SCLN(6) = 0.45
SCLN(12) = 0.45

PAUSE 'Press RETURN to continue'

DISPLAY SOME MORE RESULTS
WRITE(7,117) (SUBNAM(I),I=1,3),PARAM(1),PARAM(2)/100
WRITE(7,130)
DO 22 I=1,15
WRITE(7,122) I,SEGNAM(I),SMASS(I),SLEN(I),SCLN(I)*100
CONTINUE

WRITE data to file
WRITE(7,126)
CALL AOUT(5,ADFILE,6)

CALL AOUT

WRITE(7,116)
STOP 'C ANTHRO J'


D 901 FORMAT(' ',A8)
D 902 FORMAT(12X,F8.3)
D 906 FORMAT(' ',3A8, ' ',F6.2,' ',Kg,' ',F6.3,' ',mJ')
D 907 FORMAT(' ',SEGMENT SEGMENT SEGMENTAL MASSES I(yy)')
D 908 FORMAT(' ',SEGMENT actual corrected')
D 909 FORMAT(' ',SEGMENT MASS SLEN SCLEN')
0110 122 FORMAT(I4,2X,A8,2X,F8.3,2X,F8.3,2X,F10.5)
0111 124 FORMAT(/6X,' TOTAL ',F8.3,2X,F8.3)
0112 126 FORMAT('/ Anthrometric data output file >',$)
0113 132 FORMAT('/ Shoe mass of ',F6.3,' kg now added to feet')
0114 140 FORMAT('/ Hand open (1) or closed (2) >',$)
0115 142 FORMAT(I4)
0116  END
SUBROUTINE GETPAR(IL, IU)

This subroutine inputs data into the array PARAM between the limits L and U, giving an appropriate prompt for each parameter.

IMPLICIT INTEGER(I-N)
REAL*8 PARNAM(33)
COMMON /ANTPAR/ PARNAM(33)
DATA PARNAM/'Mass ', 'Stature ', 'Head ', 'Chest',
  'U Thigh ', 'Calf ', 'Ankle ', 'Arm/axil', 'Forearm ',
  'Wrist ', 'Arch ', 'Hand ', 'Head ', 'Elbow ',
  'Wrist ', 'Hand ', 'Knee ', 'Bi-spin.', 'Head ',
  'Acro-rad', 'Rad-styl', 'Foot ', 'Y/neck ', 'S-stern',
  'Trochant', 'Tibiale', 'Spherion', 'Lat.Tail', 'Trunk ',
  'Thigh ', 'Calf ', 'Shoemass', 'Head ht.'/

DO 10, I=IL, IU
WRITE(7,100) I, PARNAM(I)
READ(5,102) PARAM(I)
10 CONTINUE
RETURN
100 FORMAT(I4,'
',A8,',',$)
102 FORMAT(F8.3)
Data block containing all the information required to compute segmental parameters from the anthropometric measurements.

Segment names:

- REAL8 SEGNAME(15), SUBNAME(3)
- COMMON / NAMES / SEGNAME, SUBNAME
- DATA SEGNAME / HEAD, THORAX, ABDOMEN, LL ARM, L HAND, LU LEG, LL LEG, L FOOT, RL ARM, RL HAND, RU LEG, RL LEG, R FOOT

Segmental masses:

Using Clavser et al's data, the mass of each segment is a function of up to three parameters and 4 constants. The parameter codes are held in IPCM and the constants in CVM for each of 15 segments.

DATA IPCM / 3, 1, 0:
1 1.9010; 1.9010; 1.8220; 10.920; 10.1515; 2.1515; 1.920; 1.42; 1.590; 10.1515; 1.590; 0.6257; 1.722;

DATA CVM / 0.104, 0.015, 0.0, -2.189; 0.0171, 0.0, 0.0, 0.1283, 0.0, 0.0, 0.0, 0.0, 0.1283, 0.0, 0.0, 0.0; 0.007, 0.092, 0.0, 0.108, 0.0, 0.0; 0.029, 0.075, 0.031, 0.081, 0.0, 0.0; 0.074, 0.138, 0.0, 0.0, 0.0; 0.111, 0.047, 0.074, 0.0, 0.0; 0.003, 0.048, 0.027, 0.0, 0.0;
Segm~ntal moments of inertia:

Using Chandler et al's data, the moment of inertia of each segment about its centre of mass is a function of body weight and 2 constants. The constants are held in FMOIC and CMOMIC.

``` FORTRAN
0008 COMMON /HOLDAT/ FMOIC(15), CMOMIC(15)
0009 DATA FMOIC/1, 676, 0.0, 0.0, 1.352, 1.397, 0.129, 23.633, 5.341, 0.391,
          1 1.3221, 1.397, 0.129, 23.633, 5.341, 0.391/
0010 DATA CMOMIC/54818.0, 1055000.0, 4381000.0, 49572.0, -26562.0, -850.0,
          1 -319070.0, 44739.0, 4959.0, 49572.0, -26562.0, -850.0, -319070.0,
          2 44739.0, 4959.0/
```

Segm~ntal centre of mass locations:

This section contains data from Clauser et al (1969) relating the location of segmental centres of mass to body landmarks and anthropometric measurements. IPCCGX and IPCCGY contain pointers to anthropometric variables used to determine the X and Y locations of the centre of mass. CVCGX and CVCGY contain ISLEN contains references to initial values of segment length for use in calculation of SCLEN the associated constants.

``` FORTRAN
0011 COMMON /CHDAT/ IPCCGX(3,15), IPCCGY(3,15), CVCGX(4,15), CVCGY(4,15)
0012 COMMON /LEN/ ISLEN(15)
0013 DATA IPCCGX/3,13,0,
          1 0.0, 0.0,
          1 0.0, 0.0,
          1 0.0, 0.0,
          1 0.0, 0.0,
          1 0.0, 0.0,
          1 22.7, 28,
          1 0.0, 0.0,
          1 0.0, 0.0,
          1 0.0, 0.0,
          1 22.7, 28/
```
### FORTRAN IV V02.04

#### DATA IPCCGY/3,0,0:

<table>
<thead>
<tr>
<th>Line</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0,0,0</td>
</tr>
<tr>
<td>2</td>
<td>20,8,14</td>
</tr>
<tr>
<td>3</td>
<td>21,15,0</td>
</tr>
<tr>
<td>4</td>
<td>0,0,0</td>
</tr>
<tr>
<td>5</td>
<td>25,17,0</td>
</tr>
<tr>
<td>6</td>
<td>26,17,0</td>
</tr>
<tr>
<td>7</td>
<td>11,0,0</td>
</tr>
<tr>
<td>8</td>
<td>20,8,14</td>
</tr>
<tr>
<td>9</td>
<td>21,15,0</td>
</tr>
<tr>
<td>10</td>
<td>0,0,0</td>
</tr>
<tr>
<td>11</td>
<td>25,17,0</td>
</tr>
<tr>
<td>12</td>
<td>26,17,0</td>
</tr>
<tr>
<td>13</td>
<td>11,0,0</td>
</tr>
</tbody>
</table>

#### DATA CVCGX/0.238,-0.576,3.376,0.0:

<table>
<thead>
<tr>
<th>Line</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>2</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>3</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>4</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>5</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>6</td>
<td>0.153,0.137,0.444,1.403</td>
</tr>
<tr>
<td>7</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>8</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>9</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>10</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>11</td>
<td>0.0,0,0,0,0</td>
</tr>
<tr>
<td>12</td>
<td>0.153,0.137,0.444,1.403</td>
</tr>
</tbody>
</table>

#### DATA CVCGY/0.293,0.0,0.0,-5.573:

<table>
<thead>
<tr>
<th>Line</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.0,0.0,0,0,0</td>
</tr>
<tr>
<td>2</td>
<td>0.0,0.0,0,0,0</td>
</tr>
<tr>
<td>3</td>
<td>0.329,-0.259,2.827,-6.168</td>
</tr>
<tr>
<td>4</td>
<td>0.440,0.761,0.0,-5.645</td>
</tr>
<tr>
<td>5</td>
<td>0.0,0.0,0,0,0</td>
</tr>
<tr>
<td>6</td>
<td>0.214,0.902,0.0,-11.6</td>
</tr>
<tr>
<td>7</td>
<td>0.309,-0.558,0.0,5.786</td>
</tr>
<tr>
<td>8</td>
<td>0.325,0.0,0,0,0</td>
</tr>
<tr>
<td>9</td>
<td>0.214,0.902,0.0,-11.6</td>
</tr>
<tr>
<td>10</td>
<td>0.309,-0.558,0.0,5.786</td>
</tr>
<tr>
<td>11</td>
<td>0.325,0.0,0,0,0</td>
</tr>
</tbody>
</table>

#### DATA ISLEN/33,-1,-1,20,21,-1,30,31,27,20,21,-1,30,31,27:

#### END
APPENDIX E

FORTRAN IV implementation of the spatial calibration system for film analysis described in section 3.6.

Given calibration image and real coordinates, the programme sets up the simultaneous equations described in section 3.6.4 (Equations 3.13 and 3.14), with the equations in A(,), the solutions in S() and the unknowns in U(I). These equations are then solved by Gaussian elimination (SUBROUTINE GAUSS) to calculate the unknowns. This process is repeated for both x and y coordinates of the calibration data, giving the calibration matrix. The goodness of fit of the calibration matrix is determined by SUBROUTINE COFD.

The second part of the main programme takes a set of raw film data from a file and applies the calibration to it, putting the calibrated data in another file.

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```fortran
PROGRAM SC2

C TWO DIMENSIONAL SPATIAL CALIBRATION
C
C Martin Shorten, Sports Council Research Group,
C Dept. of Physical Education & Sports Science,
C Loughborough University of Technology.
C
C This programme calculates the transformation matrix between the digitised coordinates of
C an image on file and its real coordinates in two-dimensional space and performs
C the transformation.
C
C This version includes correction for non-linearity.
C
IMPLICIT INTEGER(I-N)
LOGICAL Hybrid
REAL S(10,10),S(10),U(10)
REAL SDEV(10),BETA(10),CORR(10)
COMMON /CMP/ SDEV,BETA,CORR,CMP
COMMON /SMD/ A,B,U,M,N
COMMON /FIL/ NFIF,NPOINT,HLAG
COMMON /FBUF/ X(25,100),Y(25,100)
COMMON /CAL/ CALX(20),CALY(20),CALRX(20),CALRY(20)
COMMON /SOL/ SOL(S,N)
COMMON /SHT/ SHT(S,N),SDEV
C
CALX,CALY,Calibration image coords
CALRX,CALRY,Calibration real coords
HCAL, Number of calibr'n coords
SOL, Output transformation matrix
X,Y, Buffers for file coordinates
C
IXAXIS=1
IXAXIS=2
C
WRITE(7,100)
READ(5,102) Hybrid
C
If response is X then skip matrix calculations
IF(NY.EQ.78) GOTO 23
If response is not Y then set another one
IF(NY.EQ.87) GOTO 12
WRITE(7,104)
GOTO 10
C
PHASE 1: SOLVE LEAST SQUARES CALIBRATION SYSTEM
C
Open calibration data files
```
FORTRAN IV

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0025 WRITE(7,110) 
0026 CALL ASSIGN(8.',+1.',RDO','CC',+1) 
0027 WRITE(7,144) 
0028 CALL ASSIGN(9.',+1.',RDO','CC',+1) 
  C Read calibration data 
0029 DO 14=1,NCAL 
0030 READ(9,116) CALX(I),CALY(I) 
  CONTINUE 
0031 14 
0032 READ(9,116) CALX(I),CALY(I) 
  CONTINUE 
0033 16 
0034 CALL CLOSE(9) 
0035 CALL CLOSE(8) 
0036 C Set up data matrix for x-axis calibration 
0037 CALL CLRMAT 
0038 CALL SETMAT(IAXIS) 
0039 WRITE(7,146) 
0040 CALL PRIMAT 
0041 CALL HOLDIT 
  C Solve system by Gaussian elimination 
0042 CALL GAUSS 
0043 C Display equations, solutions and unknowns 
0044 WRITE(7,142) 
0045 CALL PRIMAT 
0046 DO 20=1,N 
0047 SOL(IAXIS,I)=U(I) 
  C Transfer solution 
0048 DO 18=1,N 
0049 SOL(IAXIS,I)=U(I) 
  C Compute coefficient of multiple determination for x 
0050 CHDX = 0 
0051 CHD = 0 
0052 Restore data matrix 
0053 CALL CLRMAT 
0054 CALL SETMAT(IAXIS) 
0055 Restore unknowns 
0056 DO 185=1,N 
0057 U(I)=SOL(IAXIS,I) 
  C Calculate variances, beta weights & intercorrelations 
0058 WRITE(7,136) ((SDEV(J)*SDEV(J)),J=1,N) 
0059 WRITE(7,138) (BETA(J),J=1,N) 
0060 WRITE(7,140) (CORR(J),J=1,N) 
  C CALL HOLDIT 
0061 C Set up data matrix for y-axis calibration 
0062 CALL CLRMAT 
0063 CALL SETMAT(IYAXIS) 
0064 C Solve system by Gaussian elimination 
0065 CALL GAUSS 
0066 C Display equations, solutions and unknowns 
0067 WRITE(7,144) 
0068 CALL PRIMAT 
0069 DO 20=1,N 
0070 SOL(IYAXIS,I)=U(I) 
  C Compute coefficient of multiple determination for Y 
0071 U(I)=SOL(IYAXIS,I) 
0072 CALL COFD 
0073 CMDY = CMD 
0074 WRITE(7,136) ((SDEV(J)*SDEV(J)),J=1,N) 
0075 WRITE(7,138) (BETA(J),J=1,N) 
0076 WRITE(7,140) (CORR(J),J=1,N) 
  C CALL HOLDIT 
0077 C Calculate root mean square residual 
0078 RMSX=0.0 
0079 RMSY=0.0 
0080 RMS=0.0 
0081 DO 22=1,NCAL 
0082 XI=CALX(I) 
0083 YI=CALY(I) 
0084 XR=CALRX(I) 
0085 YR=CALRY(I) 
0086 XC = TV(IAXIS,XI,YI) 
0087 YC = TV(IAXIS,XI,YI) 
0088 RMSX=RMSX + (XC-XR)**2 
0089 RMSY=RMSY + (YC-YR)**2 
0090 CONTINUE 
0091 RMSX=RMSX/NCAL 
0092 RMSY=RMSY/NCAL 
0093 CEND
FORTRAN IV

0094 RMS=Sort(RMS/NCAL)
0097 CMUX = 1 - (1-CMBY)/(NCAL-1)/(NCAL-N)
0098 SEX = S55EV*SORT(1-CMUX)
0099 WRITE(7,120) RMSX,CMUX,CMUX,CMUX,SEY,SEY
0099 1 RMSX,CMUX,CMUX,CMUX,SEY,SEY
0100 WRITE(new calibration by file)
0100 WRITE(7,122)
0101 CALL ASSIGN(B,'-1,' 'NEW', 'CC' +1)
0102 WRITE(B,123)((SOL(I,J),X=J,1,6),(I=1,2)
0103 CALL CLOSE(B)

PHASE 2: TRANSFORM PHOTOGRAPHIC IMAGE

0104 OPEN(file data file)
0104 WRITE(7,124)
0105 READ(102) HY
0106 IF (HY. E=78) GOTO 50
0107 IF (HY. E=78) GOTO 26
0108 WRITE(7,104)
0109 GOTO 24
0112 WRITE(7,130)
0113 CALL ASSIGN(B,'-1,' 'RDO', 'CC' +1)
0114 READ(B,123)((SOL(I,J),X=J,1,6),(I=1,2)
0115 CALL CLOSE(B)

TRANSFORM FILE DATA

0117 DO 28,J=1,NPOINT
0118 DO 28,J=1,NFRAME
0119 XI=X(I,J)
0120 YI=Y(I,J)
0121 XI=TV(IYAXIS+1,YI1)
0122 YC=TV(IYAXIS,XI,1)
0123 X(I,J)=X
0124 Y(I,J)=Y
0125 CONTINUE

OPEN OUTPUT FILE FOR TRANSFORMED COORDINATES

0126 WRITE(7,128)
0127 CALL ASSIGN(B,'-1,' 'NEW', 'CC' -1)
0128 WRITE(B,108) NFRAME,NPOINT
0129 DO 30,J=1,NPOINT
0130 DO 30,J=1,NFRAME
0131 30 WRITE(B,123) X(I,J),Y(I,J)
0132 CALL CLOSE(B)

0133 CONTINUE
SUBROUTINE GAUSS

SUBROUTINE TO SOLVE A SET OF N LINEAR ALGEBRAIC EQUATIONS
BY THE CONVENTIONAL GAUSSIAN ELIMINATION ALGORITHM

A N x N matrix containing equations
S Array containing solutions to equations
U Array containing unknowns

IMPLICIT INTEGER(I-N)
COMMON /SIMED/ A(10+10),S(10),U(10),M,N

STAGE 1: Subtract multiples of rows from each other
to systematically introduce zeros into the lower
triangular part of the matrix

DO 10 II=1,N-1

Select the element (H) with the largest absolute
magnitude (W) in the current column below the diagonal
M = II
W = ABS(A(II+II))
DO 20 II=II+1,N
IF(ABS(A(II+II)),LT,W) GOTO 20
W = ABS(A(II+II))
M = II
20 CONTINUE

If pivotal element (H) is not in row II then rows
M and II need to be interchanged; W is used as working
space. The corresponding elements of B are also interchanged
IF(M.EQ.II) GOTO 30
DO 40 II=II+1,N
W = A(II+II)
A(II+II) = A(M+II)
A(M+II) = W
40 CONTINUE
W = S(II)
S(II) = S(M)
S(M) = W
30 CONTINUE

Eliminate elements in lower part of current column
DO 50 II=II+1,N
DO 60 II=II+1,N
A(II+II+II) = A(II+II+II) + W * A(II+II+II)
70 CONTINUE
S(II+II) = S(II) + W * S(II)
50 CONTINUE

STAGE 2: Determine values of U by back substitution

U(N) = S(N)/A(N,N)
SUBROUTINE CDFD

Calculate variances, beta weights and intercorrelations for spatial calibration

IMPLICIT INTEGER(-N)
LOGICAL NY
REAL A(10,10),S(10),U(10)
REAL SDEV(10),BETA(10),CORR(10)
COMMON/CDFD/ SDEV,BETA,CORR,CHDI
COMMON/SIMED/ A,S,U,N
COMMON/CALIB/ NCAL,CALX(20),CALY(20),CALRX(20),CALRY(20)
COMMON/SUMSQ/ SUMSSO,SSDEV

C SDEV Standard deviation of independent variables
C BETA Beta weights
C CORR Pearson R between dependent and independent variable
C SUMSSO Sum of squares of dependent variable
C SSDEV Standard deviation of dependent variable

SSDEV = SORT((SUMSSO:S(I))*S(I))/SIGMA(A(I,1:1))/SUM(A(I,1:1))
WRITE(7,901) SUMSSO,SSDEV

901 FORMAT(//,S10,2X,F8.3,S10,2X,F8.3)
DO 17 I=1,N
SDEV(I) = SORT((A(I,1):A(I,1)):SIGMA(A(I,1))/A(I,1:1))
BETA(I) = SDEV(I)/SSDEV
TOPLIM = (A(I,1):S(I)) - A(I,1:1)
BOTLIN = A(I,1:1):A(I,1:1)
BOTLIN = SORT(BOTLIN)
CORR(I) = 0.0
IF(BOTLIN.GT.0.0) CORR(I) = TOPLIM/BOTLIN
CND = BETA(I)*CORR(I)
RETURN
END

FUNCTION TV(IAXIS,X,Y)
IMPLICIT INTEGER(-N)
COMMON /SOL/SOL(IAXIS)
C TV = transformed value
TV = 0.0
TV = TV + X*SOL(IAXIS)
TV = TV + Y*SOL(IAXIS)
TV = TV + X*X*SOL(IAXIS)
TV = TV + X*Y*SOL(IAXIS)
RETURN
END
SUBROUTINE PRIHAT
IMPLICIT INTEGER(1-N)
REAL A(10,10), S(10), U(10)
COMMON /SIMEQ/ A,S,U,N,N
WRITE(7,100)
WRITE(7,102) (S(I), (A(I,J), J=1,N), I=1,N)
WRITE(7,104) (U(I), I=1,N)
100 FORMAT(' CURRENT MATRIX:', 6F8.3)
RETURN
END

SUBROUTINE FILMIN
IMPLICIT INTEGER(1-N)
COMMON /HFRAME/HFRAME,NPOINT, NLAG
COMMON /BUFF/ X(25,100), Y(25,100)
COMMON /CALIX/ CALIX(20), CALRY(20)
COMMON /SOL/ SOL(2,6)
COMMON /SIMEQ/ SIMEQ, SSDEV
DIMENSION EL(6)
N=6
SIMEQ=0
DO 1 II=1, N
10 EL(II)=I
DO 2 II=2, N
20 EL(II)=EL(II-1)+EL(II-2)
CONTINUE
RETURN
END

SUBROUTINE SETHAT(IXY)
C Displays SIMEQ data space
IMPLICIT INTEGER(1-N)
REAL A(6*6)
C SIMEQ (6*6) matrix of solutions in A
C and (18) array of solutions in B
C SUMSSO = sum of squares of the solutions
IMPLICIT INTEGER(1-N)
REAL A(10,10), S(10), U(10)
COMMON /SIMEQ/ A,S,U,N,N
COMMON /BUFF/ X(25,100), Y(25,100)
COMMON /CALIX/ CALIX(20), CALRY(20)
COMMON /SOL/ SOL(2,6)
COMMON /SIMEQ/ SIMEQ, SSDEV
DIMENSION EL(6)
N=6
SIMEQ=0.0
DO 10 II=1, N
10 EL(II)=I
DO 20 II=2, N
20 EL(II)=EL(II-1)+EL(II-2)
CONTINUE
RETURN
END

SUBROUTINE HLDIT
WRITE(6,100)
PAUSE ' Press RETURN to continue'
RETURN
END
APPENDIX F

FORTRAN IV implementation of a variable order Newtonian interpolation algorithm for unequal intervals. The mathematical principles of this method are described by Khabaza (1965, pp 207-217).

```fortran
SUBROUTINE INTERP(IORDER,RMIN,RMAX)

NEWTONIAN INTERPOLATION FOR IRREGULAR INTERVALS
This subroutine computes a series with a regular time-base from one with an irregular time base using the Newtonian interpolation formula for 1st - 5th order divided differences. The routine is used for the correction of uneven camera speed, and for the normalisation of running stride data to a common realative timebase. In this version data is transfered to the routine via the working buffers, SX and SY.

IMPLICIT INTEGER(I-N)
DIMENSION XIN(100),YIN(100),BUFF(100)
COMMON /FILM/ NFRAME,NPOINT,LAG
COMMON /TIME/ DELTAT,VEL
COMMON /NEAR/ XNEAR(6),YNEAR(6)
COMMON /SEARCH/ DIFF3(100),IMIN,VALMIN
COMMON /ETC/ NDATA,NEWT,X
COMMON /WORK/ SX(100),SY(100)

XIN     Input time base
YIN     Irregular time-series, replaced by regular one.
IORDER  Order of interpolation polynomial
RMIN,RMAX Range of output series
RINT    Interval of output series
NDATA   Number of points in input series

NDATA = NFRAME
RINT = DELTAT
Transfer data from working buffer to XIN,YIN
DO 25 I=1,100
   XIN(I)=SX(I)
   YIN(I)=SY(I)
25

Determine integer range for loop
ISTART = 1
IEND = IFIX((RMAX-RMIN)/RINT) + 1

Start main loop
DO 90,J=ISTART,IEND
   X = RMIN + RINT*(J-1)
90
Find nearest XIN's to X
DO 10,J=1,NDATA
   DIFFS(J) = ABS(XIN(J) - X)
```

```fortran
C C C C C C C C C C C C C C C C C C C C C C C
```

```fortran
C C C C C C C C C C C C C C C C C C C C C C C
```

```fortran
C C C C C C C C C C C C C C C C C C C C C C C
```
SUBROUTINE GETMIN

Find smallest value in DIFFS

IMPLICIT INTEGER(I-N)
COMMON /SEARCH/ DIFFS(100),IMIN,VALMIN
COMMON /ETC/ NDATA,YNEWT,X

VALMIN = 9999999.9
IMIN = 0
DO 10 J=1,NDATA.
        IF (DIFFS(J).GE.VALMIN) GOTO 10
10     VALMIN =DIFFS(J)
        IMIN = J
CONTINUE
DIFFS(IMIN) = 9999999.9 RETURN
END

DO 20,J=1,6 CALL GETMIN
IF(J.NE.1).AND.(XNEAR(J-1).EQ.XIN(IMIN))) GOTO 15
XNEAR(J) = XIN(IMIN)
YNEAR(J) = YIN(IMIN)
CONTINUE

Perform interpolation from nearest points
CALL NEWTON(ORDER)
YOUT = YNEWT + YNEAR(1)
BUFF(I) = YOUT
CONTINUE

Transfer result
DO 92 I=1,IEND SY(I) =BUFF(I)

RETURN
END
SUBROUTINE NEWTON(IORDER)

Perform 1st order Newtonian interpolation

IMPLICIT INTEGER(I-N)
COMMON /XYIN/ XIN(100),YIN(100)
COMMON /XNEAR/ XNEAR(6),YNEAR(6)
COMMON /ETC/ NDATA,YNEWT,X

WRITE(7,100)
IA = 1
NEWT = 0.0
DO 10 I = IA,IR
10 YNEWT = YNEWT + VAL * DIVDIF
15 IA = IA + 1
20 CONTINUE

WRITE(7,100)
RETURN DO FORMAT(5 NEWTON )
FUNCTION DD1(IA,IR)  
IMPLICIT INTEGER(I-N)  
COMMON/XEAR/ XNEAR(6),YNEAR(6)  
WRITE(7,100) IA,IR  
DD1 = (YNEAR(IR) - YNEAR(IA))/(XNEAR(IR) - XNEAR(IA))  
WRITE(7,200) DD1  
RETURN  
D100 FORMAT(' DD1(',I2,','I2,') ..,'$)  
D200 FORMAT(F8.3)  
END

FUNCTION DD2(IA,IR)  
IMPLICIT INTEGER(I-N)  
COMMON/XEAR/ XNEAR(6),YNEAR(6)  
WRITE(7,100) IA,IR  
DD2 = DD1(IA+1,IR) - DD1(IA,IR+1)  
DD2 = DD2/(XNEAR(IR) - XNEAR(IA))  
WRITE(7,200) DD2  
RETURN  
D100 FORMAT(' DD2(',I2,','I2,') ..,'$)  
D200 FORMAT(F8.3)  
END

FUNCTION DD3(IA,IR)  
IMPLICIT INTEGER(I-N)  
COMMON/XEAR/ XNEAR(6),YNEAR(6)  
WRITE(7,100) IA,IR  
DD3 = DD2(IA+1,IR) - DD2(IA,IR+1)  
DD3 = DD3/(XNEAR(IR) - XNEAR(IA))  
WRITE(7,200) DD3  
RETURN  
D100 FORMAT(' DD3(',I2,','I2,') ..,'$)  
D200 FORMAT(F8.3)  
END

FUNCTION DD4(IA,IR)  
IMPLICIT INTEGER(I-N)  
COMMON/XEAR/ XNEAR(6),YNEAR(6)  
WRITE(7,100) IA,IR  
DD4 = DD3(IA+1,IR) - DD3(IA,IR+1)  
DD4 = DD4/(XNEAR(IR) - XNEAR(IA))  
WRITE(7,200) DD4  
RETURN  
D100 FORMAT(' DD4(',I2,','I2,') ..,'$)  
D200 FORMAT(F8.3)  
END
FUNCTION DD5(IA,IR)
C Fifth order divided differences
C
IMPLICIT INTEGER(1-N)
COMMON/XNEAR/ XNEAR(6),YNEAR(6)
WRITE(7,100) IA,IR
DDS = DD4(IA+1,IR) - DD4(IA,IR-1)
DDS = DDS/(XNEAR(IR) - XNEAR(IA))
WRITE(7,200) DDS
RETURN
D100 FORMAT( DDS('',I12,' ',I12,' ')..,$)
D200 FORMAT(F8.3)
END
APPENDIX G

BBC BASIC programme to implement a simple model of an isometric contraction (Figure 8.5)

100 REM SIMPLE MODEL OF MUSCLE
110 REM ======================
120 130 REM Parameters for Hill's equation
140 REM based on data for the human elbow
150 REM flexors from Wilkie (1950),
160   J Physiol 110:249-280
170   F0=30
180   a=120
190   b=180
200   P=0
210   t=0
220 230 REM graphics
240   MODE4
250   xscale=10000
260   yscale=20
270   yoffset=100
280 290 REM simulate isometric myogram for contractions
290   with various added series compliances (8a)
300   PROCmyogram(0)
310   PROCmyogram(0.5E-2)
320   PROCmyogram(1.0E-2)
330   PROCmyogram(1.3E-2)
340   STOP
350
360 [Procedure definitions overleaf]
DEF PROC myogram(Ga)
REM Procedure to iterate simple isometric muscle model by
iterative integration of the equation based on the
equation:
\[
dP \quad dP \quad dx \\
--- \quad --- \quad --- \\
dt \quad dx \quad dt
\]
370 P=0
380 t=0
390 dt=0.001
400 MOVE 0,yoffset
410 REPEAT
420 dPdt = FNdPdx(P)*FNdxdt(P)
430 P=P+dPdt*dt
440 DRAW (t+dt/2)*xscale,P*yscale + yoffset
450 t=t+dt
460 UNTIL t>0.15
470 ENDPROC
480
490 REM Stiffness function
500 DEF FNdPdx(P)=3*P+20
510
520 REM Force-velocity function
[including added compliance]
530 DEF FNdxdt(P)=b*(P0-P)/(P+a) - Ga*dPdt
540 END
APPENDIX H

FORTRAN IV programme to calculate joint reaction forces and moments of a link segmental model

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PROGRAM GNOINE

LINK SEGMENTAL ANALYSIS OF RUNNING (4) : FORCES AND MOMENTS

Martyn Shorten, Sports Council Research Group, LUT

This program estimates forces and moments at the joints of a GNOME generated link-segmental model of running using the data base files created by GNOME's 1 & 2.

IMPLICIT INTEGER(I-N)

COMMON /HDIM/ HDIM
COMMON /MODEL/ NJNT,IPROX(25),IDISTS(25),
SEGAT,SMAS(25),SMEN(25)

COMMON /FNAME,TFILE,FFILE,ADFILE,AFILE,BFILE,
JNTNUM(25),SEGNAM(25),MNAM(3),EFILE

COMMON /JNAM/,SEGNAME,SEGNAME,MNAME,MNAME

COMMON /FILTER/ NFRAME,NPOINT,
COMMON /TIME/ DELTAT,VEL
COMMON /ICHAN/ FNAME,ICHAN,NREC,LENREC,IREC
COMMON /OCHAN/ FNAME,ICHAN,KCHAN
COMMON /IFILM/ NNAME,NPOINT,NLAG
COMMON /TIME/ DELTAT,VEL
COMMON /IOCHAN/ JCHAN,KCHAN
COMMON /WORK/ XIXIOO),SY(100)

NEW VARIABLES:

RXD,RYD Distal forces on current segment
XAP,XMD Proximal forces
XYD Location of distal joint centre
XP,YP Proximal & distal moments
XTOE Location of proximal joint centre

/

/AD/ Anthroponometric data of current seg
XTOE() Location, velocity & accel. of current seg
XHEE() "Toe" x-coordinate, frame by frame
XCONT Number of frames of foot contact
CONT Current position in foot contact sequence
/STORE/ SFX(1),FFY() Ground forces[frame by frame
FFX() X-coord. of point of contact[frame by frame
SFX Buffer for calculated forces and moments

COMMON /FORCE/ RXD,RYD,XMD,RFXP,RYXP:XMP
COMMON /CG/ CGX,CGY,YX,YV,AX,AY
COMMON /POS/ XDP,YPD,XYP,YP
COMMON /ANG/ THET,OMEGA,ALPHA
COMMON /AD/ GMAS,SMAS,SMEN,SMOC
COMMON /CONT/ NCONT,ICONT,FFX(50),FFY(50),FFX(50)
COMMON /FOOT/ XTOE(30),XHEE(50)
Fortran IV

COMMON /STORE/ XFORC(25,50),YFORC(25,50),XMOM(25,50)

Default file names

DATA FNAME/'COORD' /
DATA BFILE/'ACCEL' /
DATA ADFILE/'ANTHRO' /

GRAV = 9.80665
PI = 3.141593

WRITE(7,100)
WRITE(7,102) ! Print headings
READ(5,104) IPR = 5-IPR ! Request output device, either TT or LP
WRITE(7,130) ! Input duration of foot contact
CALL MODEL ! Configure segmental model
CALL ADIN ! Read anthropometric data
IDISTJ(2) = 2 ! Link neck to thorax
ICHAN=2
KCHAN=4
NREC=100
LENREC=256
NITEMS=128
CALL SETDA(FNAME,ICHAN,NREC,LENREC,IREC)! Set up direct access file
CALL SETB(KCHAN,NREC,LENREC,IREC)! Set up direct access file
CALL HDIN(ICHAN)! Read header from file
MIDDLE = (NFRAME+1)/2 ! RF contact frame
CALL EGRF ! Estimate ground forces
DO 10 IFRAME = NLAG-1,NFRAME-2
10 (1) Calculate forces on central segments
CALL SETDIS(0.0,0.0,0.0,0.0)! Assign distal forces and
CALL SAFT(IFRAME,12,0.0,0.0,0.0,0.0)! moment, assuming no contact
CALL SETSEG(1FRAME,9)! Recall data for left foot
! Branch if no contact
IF(IFRAME.GT.NFLAG+NCOUNT) GOTO 71
CALL SETDIS(FFX(IFRAME),FFY(IFRAME),0,0) ! Assign ground forces
YD = 0.0
CALL SETDIS(FFX(IFRAME),12,FFY(IFRAME),0,0) ! Assign point of contact
ANGLE = (PI/2.0) - ATAN2(XP-XD,YP)
CALL RESOLV
CALL SAVE(IFRAME,11,-RXP,-RYP,-XMP)
CALL SETDIS(-RXP,-RYP,-XMP)
CALL CHAIN(IFRAME,6,7)
CALL SETDIS(FFX(IFRAME),FFY(IFRAME),0,0)
CALL SAVE(IFRAME,21,FFX(IFRAME),FFY(IFRAME),0,0)
YD = 0.0
CALL SETDIS(FFX(IFRAME),12,FFY(IFRAME),0,0)
ANGLE = (PI/2.0) - ATAN2(XP-XD,YP)
CALL RESOLV
CALL SAVE(IFRAME,20,-RXP,-RYP,-XMP)
CALL SETDIS(-RXP,-RYP,-XMP)
CALL CHAIN(IFRAME,14,15)
! Save reactions on thorax
CALL SETDIS(0.0,0.0,0.0)
CALL CHAIN(IFRAME,1,1)
FXNECK = -RXP
FYNECK = -RYP
XHNECK = -XMP
CALL SETDIS(0.0,0.0,0.0)
CALL CHAIN(IFRAME,6,4)
FXLSHO = -RXP
FYLSHO = -RYP
XHLSHO = -XMP
CALL SETDIS(0.0,0.0,0.0)
CALL CHAIN(IFRAME,12,10)
FXRSHO = -RXP
FYRSHO = -RYP
XHRSHO = -XMP
CALL SETDIS(RX,RY,XM)
CALL CHAIN(IFRAME,2,2,0)
! Calculate reaction at BB
(2) Thoracic/abdominal forces assume co-linear shoulders
EX = FXHSHO+FXLSHO+FXNECK
RY = FYHSHO+FYLSHO+FYNECK
XM = YMXSHO+YMXLSHO+YMNECK
CALL SETDIS(RX,RY,XM)
CALL CHAIN(IFRAME,2,2,0)
! Calculate reaction at BB
SUBROUTINE CHAIN(IFRAME,ISEG1,ISEGN)

This subroutine performs a force analysis of a sequence of linked rigid segments given the forces and moments at one end of the sequence and inertial characteristics and kinematic data. Analysis starts at the proximal end of ISEG1 and ends at the distal end of ISEGn.

IMPLICIT INTEGER(I-N)
COMMON /MODEL/ NJNT,IPROXG(25),IDISTS(25),!
SEGIPROXJ(25),IDISTJ(25)
COMMON /FORCE/RXD,RYD,XMD,RXP,RYP,XMP

Loop through segments: calculating reaction forces and moments

DO 10 ISEG=ISEG1,ISEGN
    IFR=IFR-1
    CALL SETSEG(IFRAME,ISEG)
    CALL RESOLV
    CALL SAVE(IFRAME,IPROXJ(ISEG),RXD,-RYP,-RYP,-RYP)
    CALL CLOSE(IFRAME)
    CALL CLOSE(ISERCH)
    CALL FMOUT
    STOP 'GNOME4'
    CONTINUE
    DUMP RESULTS TO DISK
    STOP 'GNOME4'

FORMAT(* MS-GNOME//: C JOINT FORCES & MOMENTS */)
FORMAT(* Output to */ 1. Terminal */ 2. Printer */ */ */ )
FORMAT(* MS-GNOME//: C ; Subject >/25A) FORMAT(*
C
SUBROUTINE CHAIN(IFRAME,ISEG1,ISEGN)

CALL CLOSE(IFRAME)
! Close database files

CALL FMOUT
! Dump results to disk
STOP 'GNOME4'

CONTINUE

CALL FMOUT
! Dump results to disk
STOP 'GNOME4'

CONTINUE

CONTINUE

CONTINUE

CONTINUE
SUBROUTINE RESOLV

Resolve joint forces and moments

IMPLICIT INTEGER(I-N)

COMMON /FORCE/ RXD,RYD,XMD,RXP,RYP,XMP

COMMON /POS/ XD,YD,XP,YP

COMMON /ANG/ THETA,OMEGA,ALPHA

COMMON /AD/ QMASS,QLEN,QCLEN,QDQIC

GRAV = 9.80665

RETURN

END

SUBROUTINE SETDIS(RX,RY,XM)

Assi~n distant forces and moment

COMMON /FORCE/ RXD,RYD,XMD,RXP,RYP,XMP

RXD = RX

RYD = RY

XMD = XM

RETURN

END
SUBROUTINE SETSEG(IFRAM, ISEG)

Fetches anthropometric, position and acceleration data for segment ISEG

COMMON /NDIM/ NDIM
COMMON /MODEL/ NJNT,IPROX(25),IDIST(25), INTHAM(25), SEGHAM(25), MNAME(3), EFILE
COMMON /NAMES/ JNTNAM, SEGNAM, MNAM, SNAM
COMMON /SEGDAT/ SNAME(25), SLEN(25), SCLEN(25), SMOIC(25), ADFILE
COMMON /FNAME/ HNAME(40), NP, INT, NLAG
COMMON /TIME/ DELTA(=1000), VEL
COMMON /FILM/ SMNAME(128), NITEMS
COMMON /WORK/ 3X(100), 5Y(100)
COMMON /FORCE/ RXD, RYD, XMD, XRP, YRP, XMP
COMMON /CS/ CGX, CGY, VX, VY, AX, AY
COMMON /ANG/ THETA, OMEGA, ALPHA
COMMON /TMOIC/ SMASS(25), SLEN(25), SCLEN(25), SMOIC(25), ADFILE

JP = IPROX(ISEG) ! Find proximal & distal joints
JD = IDIST(ISEG)
QMASS = MASS(ISEG) ! Assign segment mass...
QMOIC = SMOIC(ISEG) !...moment of inertia about cs

CALL DAPULL(ICHAN, IFRAME) ! Coordinate data for this frame
YP = DABUFF(2+2*JP) ! Proximal joint centre position
XP = DABUFF(2+2*JP-1)
YP = DABUFF(2+2*JP-1) ! Proximal joint centre position
XD = DABUFF(2+2*JD-1)
CST = DABUFF(2+2*NJNT+2*ISEG) ! Segment CG position
CGX = DABUFF(2+2*NJNT+2*ISEG-1)
THETA = 3.142 - DABUFF(2*(1+NJNT+NSEG)+ISEG) ! Angular orientation

CALL DAPULL(KCHAN, IFRAME) ! Accelrn data for this frame
AY = DABUFF(2+2*NJNT+2*ISEG) ! Segment CG acceleration
AX = DABUFF(2+2*NJNT+2*ISEG-1)
ALPHA = -DABUFF(2*(1+NJNT+NSEG)+ISEG) ! Angular acceleration

RETURN
END
SUBROUTINE FMOUT

Dump force & moment data to disk

IMPLICIT INTEGER(I-N)
COMMON /MODEL/ NJNT,IPROXS(25),IDISTS(25),
NSEG,IPROXJ(25),IDISTJ(25)
COMMON /FILM/ NFRAME,NPOINT,NLAG
COMMON /TIME/ DELTAT,VEL
COMMON /STORE/ XFORC(25,50),YFORC(25,50),XMOM(25,50)
COMMON /NAME/ ICAN,HREC,LENREC,IREC
COMMON /DATABUFF/ DABUFF(NITEMS),NITEMS
COMMON /WORK/ SX(100),SY(100)

References to 'joints' at which forces & moments are zero
i.e. extremities not in contact with the ground

DIMENSION IZREF(9)
DATA IZREF/1,8,12,13,17,22,23,24,25/

ICHAN=3
NREC=100
LENREC=50
NITEMS=128
CALL SETDA('FORCES ',ICHAN,NREC,LENREC,IREC) ! Assign file
! Write header
DO 20 I=1,NFRAME
CALL HDOUT(ICHAN)
! Output data frame by frame
DO 21 J=1,NJNT
XMOM(IZREF(J),I)=0.0
XFORC(IZREF(J),I)=0.0
DO 10 IY=2,IY+3,NJNT
DABUFF(IY)=XMOM(J,I)
DABUFF(IY+1)=YFORC(J,I)
IFR=I-NLAG-1
TIM=IFR*DELTAT
DABUFF(IFR+2)=TIM
DO 10 IY=IY+2,IY+3
DABUFF(IY)=DABUFF(IY+2)
10 CONTINUE
CALL DAPUSH(ICHAN,I)
! Send DABUFF() to file
21 CONTINUE
CALL CLOSE(ICHAN)
! Close direct access file
RETURN
END
SUBROUTINE EGRF

Estimation of ground reaction force

COMMON /NDIM/ NDIM
COMMON /MODEL/ NJNT,IPROXS(25),IDISTS(25),
COMMON /NAME/ NJNT,IPROXS(25),IDISTS(25),
COMMON /TIME/ DELTAT,VEL
COMMON /DA/ FNAME,ICHAN,NREC,LENREC,IREC
COMMON /DADUFF/ DABUFF(128),NITEMS
COMMON /CONT/ ICONT,FFXI(50),FFY(50),FXP(50)
COMMON /FOOT/ SX(100),SY(100)

REAL*6 FNAME,TBFILE,FFILE,AFILE,BFILE,
COMMON /JNTNAM(25),SEGNAM(25),MNAM(3),EFILE
COMMON /NAMES/,JNTNAM,SEGNAM,MNAM,NAM
COMMON /SEGDAT/ SMASS(25),SLEN(25),SCLEN(25),SMOIC(25),ADFILE
COMMON /FNAME,ICHAN,NREC,LENREC,IREC
COMMON /BASE/ JCHAH,KCHAN
COMMON /&NAME/,IPROXS,IPROXS,IPROXS,IPROXS,IPROXS

COMMON /CONT/ ICONT,FFXI(50),FFY(50),FXP(50)
COMMON /FOOT/ SX(100),SY(100)

Segemental CG's in current frame:
DIMENSION SCGX(15),SCGY(15)

GRAV = 9.80665
MIDDLE = (NFRAME+1)/2
JUMP = MIDDLE-NLAG

Main loop calculating centre of mass of whole body and foot contact position for each frame
DO 30 IFR=1,NFRAME

CALL DAPULL(ICHAN,IFR)
DO 12 I=1,NSEG
SCGX(I) = DABUFF(2+2*NJNT+2*I-1)
SCGY(I) = DABUFF(2+2*NJNT+2*I)
CONTINUE

DO 14 I=1,NSEG
FFX(IFR) = FFX(IFR)*SMASS(I)*SCGX(I)
FFY(IFR) = FFY(IFR)*SMASS(I)*SCGY(I)
RMASS = RMASS + SMASS(I)
CONTINUE

Whole body CG (x)
Whole body CG (y)
Whole body mass

C
Estimated centre of mass position

\[ FFX(IFR) = FFX(IFR)/BMASS \]

\[ FFY(IFR) = FFY(IFR)/BMASS \]

\[ \text{WRITE}(7,202) IFR,FFX(IFR),FFY(IFR) \]

\[ \text{FORMAT('CG(x,y)'), 'F3.5') } \]

It is assumed that the point of contact moves rapidly from the rear or mid foot to the fore foot during the early part of contact. This is modelled by assuming the ratio of \( L/L = (\text{heel-FXP } x\text{-distance})/(\text{heel-toe } x\text{-distance}) \) to be a function of the ratio \( t/T = (\text{time since contact})/(\text{total contact time}) \) of the form \( L/L = \text{SORT}(t/T) \)

\[ \text{XTOE contains horizontal position of toe (MTP joint)} \]
\[ \text{XHEEL contains horizontal position of heel} \]

\[ LR = 0 \]
\[ \text{Initialise LR flag to left} \]

\[ \text{IF(IFR.GT.MIDDLE) LR=1} \]
\[ \text{Test for right foot contact} \]

\[ \text{ICONT} = IFR + (\text{NLAG-MIDDLE})*LR \]
\[ \text{Position in contact sequence} \]

\[ \text{IF(TSHELL.LT.0.0) TSHELL = 0.0} \]
\[ \text{t} \]

\[ \text{IF(IFRIFR) = XHEEL(icont) + (XTOE(icont)-XHEEL(icont))*SQRT(T)} \]

Calculate accelerations of centre of gravity by finite differences and store in workspace

\[ \text{DO 32 IFR = 2,NFRAME-1} \]

\[ \text{SX(IFR) = (FFX(IFR-1) - 2*FFX(IFR) + FFX(IFR+1)) / (DELTAT*DELTAT)} \]

\[ \text{SY(IFR) = (FFY(IFR-1) - 2*FFY(IFR) + FFY(IFR+1)) / (DELTAT*DELTAT)} \]

Calculate forces acting on body centre of mass

\[ \text{DO 34 IFR = 2,NFRAME-1} \]

\[ \text{FFX(IFR) = BMASS * SX(IFR)} \]

\[ \text{FFY(IFR) = BMASS * (SY(IFR) + GRAV)} \]

Assuming symmetry, average ground forces

\[ \text{XDIFF = 0.0} \]
\[ \text{Zero accumulators for test of symmetry assumption} \]

\[ \text{YDIFF = 0.0} \]

\[ \text{DO 36 IFR = NLAG+1,NLAG+NCONT} \]

\[ \text{DO 36 IFR = NLAG+1,NLAG+NCONT} \]

\[ \text{FXBAR} = (\text{FFX(IFR-1)} + \text{FFX(IFR+1)))/0.5 \]

\[ \text{Average X force} \]

\[ \text{FYBAR} = (\text{FFY(IFR-1)} + \text{FFY(IFR+1)))/0.5 \]

\[ \text{Average Y force} \]

\[ \text{XSUM} = XSUM + FXBAR \]

\[ \text{Accumulate sums for simple test of asymmetry} \]

\[ \text{YSUM} = YSUM + FYBAR \]

\[ \text{XDIFF} = (\text{FFX(IFR-1)} - \text{FFX(IFR+1)))/2.0 \]

\[ \text{YDIFF} = (\text{FFY(IFR-1)} - \text{FFY(IFR+1)))/2.0 \]

\[ \text{Assuming symmetry, average ground forces} \]

\[ \text{DO 36 IFR = NLAG+1,NLAG+NCONT} \]

\[ \text{FXBAR} = (\text{FFX(IFR-1)} + \text{FFX(IFR+1)))/0.5 \]

\[ \text{Average X force} \]

\[ \text{FYBAR} = (\text{FFY(IFR-1)} + \text{FFY(IFR+1)))/0.5 \]

\[ \text{Average Y force} \]

\[ \text{XSUM} = XSUM + FXBAR \]

\[ \text{Accumulate sums for simple test of asymmetry} \]

\[ \text{YSUM} = YSUM + FYBAR \]

\[ \text{XDIFF} = (\text{FFX(IFR-1)} - \text{FFX(IFR+1)))/2.0 \]

\[ \text{YDIFF} = (\text{FFY(IFR-1)} - \text{FFY(IFR+1)))/2.0 \]

\[ \text{Assuming symmetry, average ground forces} \]

\[ \text{DO 36 IFR = NLAG+1,NLAG+NCONT} \]

\[ \text{FXBAR} = (\text{FFX(IFR-1)} + \text{FFX(IFR+1)))/0.5 \]

\[ \text{Average X force} \]

\[ \text{FYBAR} = (\text{FFY(IFR-1)} + \text{FFY(IFR+1)))/0.5 \]

\[ \text{Average Y force} \]

\[ \text{XSUM} = XSUM + FXBAR \]

\[ \text{Accumulate sums for simple test of asymmetry} \]

\[ \text{YSUM} = YSUM + FYBAR \]

\[ \text{XDIFF} = (\text{FFX(IFR-1)} - \text{FFX(IFR+1)))/2.0 \]

\[ \text{YDIFF} = (\text{FFY(IFR-1)} - \text{FFY(IFR+1)))/2.0 \]
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0067  FFX(IFR) = FXBAR
0068  FFY(IFR) = FYBAR
0069  FFX(IFR+JUMP) = FXBAR
0070  FFY(IFR+JUMP) = FYBAR
0071 36 CONTINUE

0072  FFX(NFRAME-NLAG+1) = FFX(NLAG+1)
0073  FFY(NFRAME-NLAG+1) = FFY(NLAG+1)

C Report asymmetry

0074  WRITE(7,100) XDIFF/XSUM*100, YDIFF/YSUM*100
0075  100 FORMAT(/', 'FORCE ASYMMETRY:',
1  , 'X ', F6.1, ' %',
1  , 'Y ', F6.1, ' %')

C

0076  RETURN
0077  END
SUBROUTINE FOOT

Fetch toe and heel positions from file for use in estimation of point of application of estimated ground reaction force. N.B. Heel data has not been assigned to GNOME database and therefore needs some working on... I promise to do it properly next time!

IMPLICIT INTEGER(I-N)

CALL ASSIGN(8,'TIME.DAT',8,'RDO','CC',1) ! Open timebase file
DO 10 I=1,NFRAME
READ(8,101) SX(I)
10 CONTINUE
CALL CLOSE(8) ! Close timebase file

CALL ASSIGN(8,'FLMT.DAT',8,'RDO','CC',1) ! Open film data file and read. As the file is read, the x-coordinate of the heel marker is picked out and saved in XHEEL(I)
DO 16 I=1,NFRAME
READ(8,106) DUMMY
READ(8,106) XHEEL(I)
16 CONTINUE
READ(8,106) DUMMY
CALL CLOSE(8) ! Close film data file

Perform Newtonian interpolation of heel position onto timebase.
ORDER = 3 ! = 3rd order interpolation
TSTART = SX(NFRAME-NLAG)+4*DELTAT ! Start and ... end of interpolation =
DO 18 I=1,NFRAME
SY(I) = XHEEL(I)
18 CONTINUE
CALL INTERP(ORDER,TSTART,TEND) ! Go and do the biz...
DO 19 I=1,NFRAME
XHEEL(I) = SY(I)
19 CONTINUE
CALL DAPULL(ICHAN,I)
XTOE(I) = XHEEL(I)
CALL DAPULL(ICHAN,I)
RETURN ! Now fetch left MTP data from COORD file and store in XTOE(I)
END
APPENDIX I

Energy analyses of the link segmental model

FORTRAN IV

PROGRAM GNOME3

LINK SEGMENTAL ANALYSIS (3): ENERGY ANALYSIS

Martyn Shorten, Sports Council Research Group, LUT

This programme performs an energy analysis of the link segmental model and data established GNOME'S 1 & 2. Segmental energy data is output to the existing file 'ENERGY.DAT'. By default, the programme calculates within and between segment energy transfers with no constraints applied to the model.

0014  IMPLICIT INTEGER(1-N)
0035  COMMON /NDIM/ NDIM
0036  COMMON /MODEL/ NJNT,IPROX(25),IDISTS(25),
0037    ISSEG,IPROXI(25),IDISTJ(25)
0038  REAL 8 FNAME,TBFILE,FFILE,AFFILE,BFFILE,
0039    JNTNAM(25),SEGNAM(25),MNAM(3),EPFILE
0033  LOGICAL NY,SNAM(25)
0062  COMMON /NAMES/ JNTNAM,SEGNAM,MNAM,SNAM
0063  COMMON /SEGDAT/ SMASSC25!,SLEN(25),SCLENSI,SMOIC(25),ADFILE
0064  COMMON /FILT/ NFRAME,NPOINT,NLAE
0065  COMMON /TIME/ DELTA,TVEL
0066  COMMON /DNAME/ FNAME,ICHAN,NREC,LENREC,IREC
0067  COMMON /DARUFF/ DARUFF(128),NITEMS
0068  COMMON /WORK/ SX(100),SY(100)
0091  COMMON /SYSTEM/
0092  COMMON /VC/ ENERGY(15,3,100)
0093  COMMON /NU/ NUHI
0094  COMMON /CSUM/ CSUM(15,3),USUM,SSUM
0095  COMMON /W/ WW,WV,WWB,WWU(15),TW,TWB,TWU(15)
0096  DIMENSION NNN(3),FE(25),TKE(25),RKE(25),ESEG(25)
0097
0098  GRAV = 9.81183
0099  ! - standard g
0100  DATA FNAME/'COORD' / ! Filtered coordinates
0101  DATA AFFILE/'VELOC' / ! Velocities
0102  DATA EFILE/'ENERGY' / ! Segmental energy data
0103  DATA ADFILE/'ANTHRO' / ! Anthropometric data
0104  CALL ADIN
0105  WRITE(7,100) ! Print header
0106  CALL ADIN ! Read anthropometric data
CALL SETDA(FNAME,ICHAN,NREC,LENREC,IREC)
CALL SETDA(AFNAME,IVCHAN,NREC,LENREC,IREC)
CALL SETDA(ENFILE,IECHAN,NREC,LENREC,IREC)

CALL MODEL
CALL HDIN(ICHAN)
CALL HDOUT(ICHAN)

Calculate energies frame by frame
DO I=1,NFRAME
PEBOD=0.0
TKEBOD=0.0
RKEBOD=0.0
EBOD=0.0

1. Segment potential energy
CALL DAPULL(ICHAN,I)
DO 12 J=1,NSEG
ESEG(J)=0.0
IY=2+2*NJT+2*NSEG+J
PE(J)=SMASS(J)*GRAU*DABUFF(IY)

12 PE(J)=SMASS(J)*GRAU*DABUFF(IY)

2. Segment kinetic energies
CALL DAPULL(IVCHAN,I)
DO 14 J=1,NSEG
IV=2+2*NJT+2*NSEG+J
IX=IX-1
VX=DABUFF(IX)
VY=2+2*NJT+2*NSEG+J
IY=DABUFF(IY)

14 VY=DABUFF(IY)

VX=UVEL
V=SQRT(VX*VX+VY*VY)
TKE(J)=0.5*SMASS(J)*V*V
RKE(J)=0.5*SMOIC(J)*OMEGA*OMEGA

3. Accumulate energy sums
ESEG(J)=PS(J)+TKE(J)+RKE(J)
PEBOD=PEBOD+PE(J)
TKEBOD=TKEBOD+TKE(J)
RKEBOD=RKEBOD+RKE(J)
EBOD=EBOD+ESEG(J)

Copy component energies to matrix
ENERGY(I,J,1)=PE(J)
ENERGY(I,J,2)=TKE(J)
1. Define each segment as a unit
2. Each segment has three components
3. Each frame = a time unit
4. Compute energy exchange
5. Results to terminal
6. Results to printer

STOP 'ENERGY J'

FORMAT(// 'MS-GNOME/1: ENERGY J')
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PROGRAM Gnomes

LINK SEGMENTAL ANALYSIS OF RUNNING (5)
(Strain energy storage in the knee extensors)

Martin Shorten, Sports Council Research Group, LUT

This programme performs an energy analysis of the
link segmental model and data established GNOME's 1 &
2. Segmental energy data is retrieved from the existing
file 'ENERGY.DAT' created by GNOME3. This programme is
specific to the model configuration GNOME 2/15A, the 15
segment model of running.

A standard analysis on the unconstrained system is
followed by the addition of a strain energy component for
the knee extensors and a second energy exchange analysis.
The strain energy component is calculated by integrating an
angular compliance function for the knee extensors with
respect to the knee moments calculated by GNOME4.

IMPLICIT INTEGER(I-N)
COMMON /MODEL/ HJNT,IPROX(25),IDISTS(25),
1 NSEG,IPROXJ(25),IDISTJ(25)
REAL* FNAME,TFI,EFILE,ADFILE,AFILE,BFILE,1
JNTNAM(25),SEGNAME(25),ANAM(3),EPFILE
LOGICAL* NY,SNAM(25)
COMMON /NAMES/ JNTNAM,SEGNAME,ANAM,SNAM
COMMON /SEGBAT/SMASS(25),SLEN(25),SMLN(25,FMOD(25),ADFILE
COMMON /FILM/ NFILE,NPOINT,NLAG
COMMON /TIME/ DELTAT,VEL
COMMON /DA/ NNAME,KCHAN,NREC,KHERE,IREC
COMMON /DABUFF/ DABAFF(128),NITEMS
COMMON /WORK/ 3X(100),SY(100)

Energy system data blocks. ** Note new dimensions;
COMMON /SYSTEM/ NC(16),ENERGY(16,4,50)
COMMON /POSS/ WU(25),WU(16,3),USUM(16),SSUM
COMMON /W/ WN,WW,WWB,WWU(16),TW,TWB,TWU(16)
COMMON /EBOD/ EBO(50)

DIMENSION GG(50); Total body energy
! Compliances

Default file names
DATA FNAME/COORD
DATA AFILE/VELOC
DATA DFILE/ENERGY
DATA BFILE/FORCES
DATA AFILE/ANTHR0
ICHAN=2
ICHAN=3
ICHAN=4
ICHAN=4
NREC=100
LENREC=56
 ITEMS=128

WRITE(7,100) ! Print header
WRITE(7,102) ! Input compliance function
READ(5,104) GA
WRITE(5,104) GB
WRITE(7,108) ! Angular correction factor
READ(5,104) ACF
GA = EXP(GA)
GB = GB/1000

CALL MODEL ! Configure link-segmental model
CALL ADIM ! Read anthropometric data
CALL ENIN ! Retrieve energy data from file

NU=NSEG ! Define each segment as a unit
DO J=1,NSEG
NDJ(J)=3
NT=NFRAME-NLAG
CALL DELTA
CALL EPR(7)
CALL EPR(6)

CALL SEH1A('FORCES','IFCHAN',NREC,LENREC,IREC) ! Assign force file
DO 30 I=1,NFRAME
CALL DAPULL(IFCHAN+I)
SX(I) = DABUFF(32)
SY(I) = DABUFF(59)
CONTINUE
CALL CLOSE(IFCHAN) ! Close force file

CALL SETDA('FORCES',IFCHAN,NREC,LENREC,IREC) ! Assign force file
DO 32 IFR = NLAG,NFRAME-NLAG+1
P = SX(IFR)
PO = SX(IFR-1)
IF(P.GT.0.0) PO = 0.0
IF(P.LT.0.0) GOTO 31
31 DP = -P
32 Q = EXP((GB*P) * GA/ACF
33 G(IFF) = 0
35 DX = 0.6*DP

Calculate energy changes in right knee extensors

\[ E = 0.0 \]

DO 42 IFR = NLAG, NFRAME-NLAG+1

PO = SY(IFR)

IF(IFPO.LT.0.0) E = E + PO ! if P < 0.0)

IF(IFPO.GT.0.0) GOTO 41

\[ E = E + \text{EXP}(\text{GG}(P)*\text{GAG}/\text{ACF})*(P-PO) \]

CALL ERESET

Initial value of \( E \) is unknown. Thus integration may lead to apparently negative energies. To make sense of this, the routine ERESET finds the minimum value of \( E \), sets it to zero and resets the other values relative to it.

Display intermediate results & copy to file

CALL ASSIGN(IFCHAN, 'SEDAT.DAT', 9, 'NEW', 'CC', 1)

CALL ASSIGN(IFCHAN, 'SEDAT.DAT', 9, 'NEW', 'CC', 1)

CALL ERESET

Recalculate energy model

\[ G = \exp(bF) \]

Calculate energy exchange

Results to terminal

Results to printer

STOP 'C ENERGY I'

This programme includes analysis of elastic'

\[ G \] is compliance, \( F \) is muscle force and \( a \) and \( b \) are constants. This function is handled in the more useful angular form given by'

\[ G(T) = 0(F)/c \]

where \( G \) is the angular compliance, \( F \) is the muscle moment, \( T \) is the angular correction factor, \( c \) is the angular correction factor, \( a \) and \( b \) are the natural moment

END
MECHANICAL ENERGY CHANGES AND TRANSFERS WITHIN A GENERALISED ENERGY SYSTEM

The system consists of \( U \) subunits (e.g., segments or limbs), the \( J \)th subunit having \( N(J) \) energy components between which transfers can occur. It is assumed that the magnitude of the energy transfer between the components of a system can be determined by comparing changes in the component energies with changes in the system total, this latter being equal to the sum of the absolute changes in the energy components plus any transfer between them.

\[
\begin{align*}
& \text{N}\mu \quad \text{NUMBER OF ENERGY SUBUNITS} \\
& \text{NC}(J) \quad \text{NUMBER OF ENERGY COMPONENTS IN Jth SUBUNIT} \\
& \text{NT} \quad \text{NUMBER OF TIME INTERVALS} \\
& \text{ENERGY} \quad \text{ENERGY DATA (Has dimensions(U,MAX(NJ),T+1)} \\
& \text{DELTA} \quad \text{ENERGY CHANGES (Dimensions as ENERGY)} \\
& \text{CSUM} \quad \text{COMPONENT ENERGY CHANGE SUMS} \\
& \text{USUM} \quad \text{SUBUNIT ENERGY CHANGE SUMS} \\
& \text{SSUM} \quad \text{SYSTEM ENERGY CHANGE SUM} \\
& \text{WU} \quad \text{SUM OF CHANGES IN A UNIT (NO TRANSFER)} \\
& \text{TW} \quad \text{TOTAL WITHIN UNIT TRANSFER} \\
& \text{TB} \quad \text{TOTAL BETWEEN UNIT TRANSFER} \\
& \text{TU} \quad \text{TOTAL TRANSFER WITHIN & BETWEEN} \\
& \text{TWU} \quad \text{TRANSFER WITHIN EACH UNIT}
\end{align*}
\]

**SUBROUTINE DELTA**

Calculate delta values from raw energy data and sum component energy changes

IMPLICIT INTEGER(I-N)

COMMON /FILM/ NFRAME, NPOINT, NLAG
COMMON /SYSTEM/ NC(16), ENERGY(16,4,50)
COMMON /SIZE/ N\mu, NT
COMMON /SUMS/ CSUM(16,4), USUM(16), SSSUM
COMMON /W/ WH, WW, WWB, WWU(16), TW, TB, TWB, TWU(16)

\[
\begin{align*}
& \text{Initialise sums} \\
& \text{DO } J=1, N\mu \quad \text{USUM(J)} = 0.0 \\
& \text{DO } K=1, NC(J) \quad \text{CSUM(J,K)} = 0.0 \\
& \text{WW} = 0.0 \\
& \text{WH} = 0.0 \\
& \text{WWB} = 0.0
\end{align*}
\]

Allow for dead frames in frame counter

\[
\begin{align*}
& \text{DO } 10 \quad \text{I=1, NLAG} \\
& \text{STOT} = 0.0 \\
& \text{DO } 20 \quad \text{J=1, N\mu} \\
& \text{UTOT} = 0.0
\end{align*}
\]
DO 30 K=1,NC(J)

DELTA = ENERGY(J,K,1) - ENERGY(J,K,1-1)

UTOT = UTOT + DELTA

STOT = STOT + DELTA

30 CONTINUE

CSUM(J,K) = CSUM(J,K) + ABS(DELTA)

USUM(J) = USUM(J) + ABS(UTOT)

$S$UM = $S$UM + ABS(STOT)

C Assign component sums

DO 40 J=1,NU

WWU(J) = 0.0

30 CONTINUE

CSUM(J,K) = CSUM(J,K)/2

WU(J) = WU(J) + CSUM(J,K)

50 CONTINUE

USUM(J) = USUM(J)/2

WW(J) = WW(J) + WU(J)

CONTINUE

CONTINUE

USU = USU/2

Calculate overall transfers

TW = WW - WWB

30 RETURN

END
SUBROUTINE EPRIIPR(IPR)

Subroutine to display data generated by 'EDELTA'. IPR defines output channel.

IMPLICIT INTEGER(I-N)
REAL*8 JNTNAM(25), SEGNAM(25), MNAM(3)
LOGICAL SNAM(25)

COMMON /SYSTEM/ NC(16), ENERGY(16, 4), 50
COMMON /SUMS/ CSUM(16, 4), USUM(16), SSUM
COMMON /W/ WN, WW, WWB, WWU(16), TW, TB, TWB, TWU(16)

WRITE mechanical energy changes to channel IPR
WRITE(IPR, 100) (SNAM(I), I=1, 25)
100 FORMAT('MECHANICAL ENERGY CHANGES (BY UNIT) : ', 25A, '
UNIT ELEM 1 ELEM 2 ELEM 3 ELEM 4 WN TW WW')

DO 10 I=1, NU
WRITE(IPR, 200) I, (CSUM(I, J), J=1, 4), WWU(I), TW(I), USUM(I)
10 CONTINUE

WRITE(IPR, 300) WN, WW, WWB, TW, TB, TWB, TWU
300 FORMAT('MECHANICAL ENERGY CHANGES (OVERALL) : ')

RETURN
END
SUBROUTINE ERESET

Find minimum stored energy & set to zero, resetting all other values relative to it.

IMPLICIT INTEGER(I-N)

COMMON /FILM/ NFRAME, NPOINT, NLAG
COMMON /SYSTEM/ NC(16), ENERGY(16,4,50)
DIMENSION A1(50), A2(50)

DO 20 LEG=0,1
ISEG = 7+LEG*6

DO 10 IFR=1,NFRAME
E = ENERGY(ISEG,4,IFR)
E = E-EMIN
IF (E.LT.0.0) E=0.0
ENERGY(ISEG,4,IFR) = E
CONTINUE
20 CONTINUE

DO 30 IFR=1,NFRAME
Al(IFR) = ENERGY(7,4,IFR)
DO 40 IFR=2,NFRAME-1
SA1 = A1(IFR-1)+A1(IFR)+A1(IFR+1)
ENERGY(7,4,IFR) = SA1/3
DO 50 IFR=1,NFRAME
A2(IFR) = ENERGY(13,4,IFR)
DO 60 IFR=2,NFRAME-1
SA2 = A2(IFR-1)+A2(IFR)+A2(IFR+1)
ENERGY(13,4,IFR) = SA2/3
CONTINUE
30 CONTINUE
RETURN
END