A Bioengineering Analysis of Muscle and Joint Forces Acting in the Human Lower Limbs during Running

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Summary

A fully dynamic two dimensional bioengineering model of the human lower limbs has been produced and has been solved for the activity of running using an inverse dynamics approach. The model includes all the major contributory muscles and muscle groups in the lower limbs including the soleus and the tibialis anterior. The actions of all the muscles have been verified by coordinated electromyographic experimentation.

The analysis method of model production, data collection and data processing has been carried out using standard biomechanical practices, techniques and equipment and this allows comparison with other studies in similar fields. This equipment and technique, based around a Kistler force platform, a Locam high speed cinematographic camera, a magnetorestrictive digitising tablet and individually tailored bone models displays results reliable to within ± 3% per subject. This produces results for ground reaction forces, joint moments, limb angles, muscular tensions and joint reactions.

For a basic running speed of 4.47 m s^{-1} results indicate maximum joint moments of 150 Nm, 189 Nm and 214 Nm occur in the hip, knee and ankle respectively. Maximum mean peak muscle forces of 15.0 BW and 10.7 BW occur in the quadricep and tricep surea muscle group accordingly. The hamstring and shin groups display more modest values of 2.8 BW and 0.5 BW respectively. Muscular loading rates of 116 kN s^{-1} are recorded in the tricep surea group and 292 kN s^{-1} in the quadriceps which compares favourably with values expressed by Komi et al. (1985) who used a strain gauge implant into the human

body. Maximum joint reactions occur in the knee at 21.3 BW for the compressive component and 2.4 BW for the shear component.

Variations in these values were noted with changes in running style and speed. Rear foot strikers exhibit longer foot contact time ($p > 0.05$) and reduced loading rates when compared to front foot strikers. Increases in speed from 3.38 m s^{-1} to 4.47 m s^{-1} and to 5.36 m s^{-1} do not result in significant increases in compressive elements of the force system but significant increases in the shear force elements are noted ($p > 0.05$).

It is concluded from these findings that front foot strikers may be more susceptible to injury than rear foot strikers. Also it is hypothesised that it is not necessarily the high force values that result in athletic injury but rapid changes in training routine that result in the shear components of the lower limbs being overloaded. With this in mind, the design of running shoes maximising on grip rather than impact force protection is recommended.

Recommendations for further work include verification of the model via comparisons with known works in other activities, walking, squatting and weightlifting and investigations into the use of digital cameras to reduce analysis time.

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1. INTRODUCTION

1.1 Introduction

Athletics, as reported by the popular athletic journals, both as a spectator and participatory sport is enjoying a boom period. This is exhibited firstly by the trends in marathon running where in the period 1979-1984 the number of available UK marathons grew from 21 to 136. This popularity in marathon running appears to be waning with fewer entries except for the larger 'peoples' events such as the Great North Run, the London Marathon and the New York Marathon which was officially the largest 26.2 mile event in 1987 with 21,141 athletes completing the distance (Athletes Weekly Nov. 26th 1987). Although marathon events and entrants are seeing a fall in numbers other distances are experiencing an upturn with the total number of events over all distances up by 30% in the six month period March-August 1988 when compared to the same period in 1987. It is now estimated that 6% of the British Adult population run or jog for fitness; this is over 2½ million athletes and the most popular racing distances now appear to be 10 km and 10 mile events. Worldwide there is now estimated to be 42 million active athletes with 46% of these involved in road or cross country running.

Both competitive and recreational runners are commonly indulging in high mileage training; 70 km per week (44 miles/week) being the average figure for a range of club runners although 27% of these runners interviewed for Running Magazine (1988 March) were reported to be completing more than 96 km per week (60 miles per week).

The cyclical nature of running and hence the repetitive impacting of the foot on the ground (foot strike) exposes the human body, especially the lower limbs, to the possibility of injury. These injuries usually manifest themselves as 'overuse' injuries such as a stress fracture of the tibia where the repeated loading caused by foot strike continually stresses the tibia until the area of maximum stress fails under a fatigue syndrome. The possibilities of these injuries can be elevated by such factors as footwear, terrain (hills, cambered roads, mud or sand), asymmetrical or dysfunctional aspects of the muscles or bones of the lower limbs (flat feet or pigeon toed) or commonly the kinematics of the joints such as excessive pronation: Here, the calcaneus is rapidly transferred from an inverted position to an over everted position causing muscle imbalances throughout the lower leg which can lead to knee injuries such as chondramalacia patella (Clarke, et al. 1983b).

Conversely acute injuries such as muscle 'pulls', the tearing of muscle fibres, may occur during faster running due to the intensity of the activity indulged in. This intensity, created by the elevated speeds, is present from the need for the muscles in the lower limbs to contract in a shorter time and over a greater range; as speed increases, foot contact time or stance phase decreases and stride length increases. Foot/ground reaction forces are also higher due to a greater change in the body's kinetic and potential energy states, i.e. the body decelerates to foot strike then accelerates from the push off, or toe off, in both the vertical and horizontal direction (Mann, et al., 1980 and Agre, 1985).

Hence the implications for the occurrence of either an overuse or acute injury in the muscles and joints of the human body are that they can be influenced by such simple parameters as speed of gait, style, footwear and surfaces. Any poor combination of parameters can greatly elevate the possibility of contracting an injury and a study of running habits reported that during 1987 10% of the British public attempted either competitive or recreational running. Of these 4.4 million athletes only 2.5 million have continued to run regularly with the predominent cause for dropping out of the sport being injury (Running Magazine, 1987).

Of the injuries sustained most occur within the lower limbs, the knee joint being the most commonly injured area. All the major leg muscle groups cross the knee and it is one of the three major load bearing joints in the body (Paul, 1967). It is also complex in its geometry and articulation being a junction of four bones; femur, fibula, tibia and patella with complex three dimensional motion stabilised by four major ligaments at the end of its defined ranges. Around the knee joint muscular injury can occur to such groups as the hamstrings or quadriceps although the most commonly injured muscle group is the tricep surea or foot plantar flexion mechanism. Why this one particular muscle/muscle group is more prone to injury than other equally large muscle groups is unknown. Direct measurement of tension developed in individual muscles is not possible in living athletes and consequently non invasive techniques need to be developed.

Joint replacement developments have enabled force transducers to be implanted into living subjects but the life of these measuring devices and their calibration and/or validity is open to question. They are also not ideally suitable for use in sporting activities -

not withstanding the average age of the artificial joint's recipient.

Stress fractures are one form of bone injury occurring in athletes, less common are such injuries as Osgood Schlatter's Disease where a portion of bone is pulled away at the corresponding muscle (or tendon) insertion (Klafs & Arnheim, 1973). Again, stress and strain values cannot normally be obtained invasively.

For all the above mentioned factors cadaveric tests do provide data on, for example, ultimate tensile strength, bending stresses and strains to induce failure and strain related fatigue failure. These however do not tell us what is going on within the human body during a sporting activity and at present the only evasive technique open to researchers is via mathematical models. These models based upon sound anatomical characteristics enable the force system present in the muscles, joints and bones to be identified and evaluated. This technique of bio-engineering has become popular in analysis of all forms of human activity from chewing to skiing.

Simple bio-engineering models have been developed and used by such investigators as Smith (1975) who produced a below the knee model for the analysis of gymnastic activities. This simple model only contained three muscle groups and thus allowed quick if somewhat over simplified analysis to be performed. Seireg and Arvikar (1975) however developed a biomechanical model including all 31 lower limb muscles (single leg). As only a limited number of basic equations of motion exist for a unique solution a mainframe computer is needed to evaluate all possible solutions and a mathematical algorithm required to select the optimum result.

This optimum result may be based upon assumptions of gross human physiological measures. This might be the total work done by the muscles and although the model is comprehensive in its inclusion of all the lower limb muscles the results are still only estimates. This method classified as optimisation does not lend itself to sporting analyses where amongst other things a more rapid turn around of results is required and most sports biomechanics laboratories cannot repeat the results due to inadequate computing power.

A compromise biomechanical model must, therefore, be sought that enables relatively quick analyses to be performed without oversimplifying the human body. Thus, a biomechanical model is needed that sufficiently represents the human lower limbs, the muscles employed in running and the joint articulations present, and is simple enough in its use that its analysis method is both quick and repeatable, across biomechanics laboratories. This will thus involve using standard biomechanical techniques and equipment such as cine film, video, force platforms etc.

1.2 Aims & Objectives of this Study

The overall aim of this study is to develop an understanding of muscle and joint forces present in the human lower limbs during running. This may enable relationships between forces present, athletic activity and injuries occurring to athletes to be developed and possible preventative measures to be suggested. These aims will be fulfilled via the following objectives:-

1. To produce a biomechanical model to estimate the muscle and joint forces present during athletic activities.

2. To develop an analysis method, based on standard biomechanical laboratory equipment, that produces repeatable, accurate and meaningful muscle and joint force data.

3. To evaluate external performance parameters such as speed, footwear and surfaces and running style to assess risk factors associated with such athletic activities or conditions.

4. To enable recommendations to be made, based upon point 3 above, as how to minimise the likelihood of injury to certain populations of athletes.

5. To establish a data base of kinematic, kinetic and anthropometric data for future use in areas such as computer simulation.

Henceforth this project is a bioengineering investigation into muscle and joint forces acting in the human leg during running. Primary concern will be given to those areas that are most frequently injured namely the knee joint and surrounding anatomy. This data should be beneficial to the athlete, coach and medical or health care services and as such should be presented in a clear and easily interpretable manner.

2. REVIEW OF LITERATURE

2.1 Introduction

Before a detailed investigation can be undertaken into muscle and joint forces acting in the human lower limbs during running and their relationship to athletic injury, an understanding of the individual elements needs to be developed. What follows is a review of selective literature relating to the elements of injury in sport and athletics in particular; the effects of footwear and running style on the kinetics of the lower limbs; assessment of human forces with regard to data collection, transfer and analysis and a review of the few papers devoted to biomechanical assessment of muscle and joint forces in the lower limbs.

2.2 Injury

For the average middle and long distance athlete, typical training routines involve running 70 km per week (44 miles per week) which could consist of approximately 80,000 impacts of the foot on the ground (foot strike). As suggested in the introduction, these repeated impacts can result in 'overuse' injuries or assist in the development of 'acute' injuries and in general Krissoff et al. (1979) report that 66% of all runners encounter injury each year.

For a non contact/non impact sport running can be seen to have a high incidence of general sports injury when compared to other more explosive or complex sports. Table 2.2.1 shows how in two out of three studies of injuries encountered in various sports, running (the speed or level of achievement is not specified) has the highest rate; 61% and 69% from studies by Lloyd-Smith et al. (1985) and

TABLE 2.2.1 Sports/Activity Leading to Injury

Investigator	Activity[1]	% Sportsmen or Athletes Injured[2]
Lloyd-Smith et al., 1985[3]	Running	61
	Fitness Class	8
	Racquet Sport	7
Matheson et al., 1987	Running 33 km	69
	Fitness Class	8
	Racquet Sport	5
	Basketball	4
Kajula et al., 1986[4]	Soccer	21
	Running	13
	Volleyball	12
	Orienteering	8
Nicholl & Williams, 1983	Marathon Runners	18
	½ Marathon Runners	4
Watson et al., 1987	Sprinting	46
	Distance Running	17
	Other Track & Field	17
James et al., 1978	Running 78 km	60
Valmassey via Newell et al., 1984	Running 96 km	50
Maughan et al., 1983	Running 54 km	58
Koplan via Powell et al., 1986	Running 0-31 km	29
	" 32-47 km	38
	" 48-63 km	46
	" 64+ km	57
Taunton et al., 1982	Running 15-29 km	22
	" 30-49 km	33
	" 50-70 km	35
	" 70+ km	10
Blair et al., 1987	Running 1-32 km	19
	" 33-64 km	29
	" 65+ km	38

(1) = Running distance per week

(2) = All injuries included

(3) = Hip injuries only

(4) = Knee injuries only

Matheson et al. (1987) respectively. For a primarily two dimensional, non impact, 'healthy' pastime this is obviously both large and concerning. The other sports represented in the studies all involve impacts, contacts, three dimensional action or explosive actions.

A comparative study of running habits also indicates that as running distances completed each week increases so the likelihood of injury increases. Table 2.2.1 shows that as running distance increases in excess of 48 km per week (30 miles per week) then the risk of incurring an injury is on average greater than 40% and for weekly totals in excess of 64 km per week approximately 50% of all athletes can expect to contract an injury. These details are further supported by the study of Nicholl & Williams (1983) who found that 18% of Marathon runners encountered injury during their race whilst only 4% of those athletes completing a corresponding half-marathon reported injury.

Hence whilst it may be hypothesised that the more running undertaken the greater the athletes fitness becomes, the repeated impacting at foot strike is responsible for rendering the athlete liable to an overuse injury.

Maughan & Miller (1983) reported that 30% of the athletes entered for a marathon had incurred some form of injury during their preparation. 94% of these injuries occurred in the lower limbs and 32% affected the knee joint. Nichol & Williams (1983), found that running a marathon caused injury to 23% of the competitors with 50% of these involving the lower limbs. This trend is repeated throughout the literature and chondramalacia patella, a swelling around the knee cap from retropatellar grating or rubbing is responsible for the greatest number of individual injuries. Fiegel et al. (1980) describees the knee as "the most vunerable joint in the runners body" with its complex 'Screw home' three dimensional motion of glide (translation : antereo-postereo), rotation (flexion-extension) and twist (inversion-eversion). The location of the most common injuries sustained during running are collated in Table 2.2.2 with the knee joint averaging 30% of all injuries reported here and between 30% and 50% in studies by Grana (1985), Kajula (1986) and Newell et al. (1984). As can be seen the second most injured areas of the athletes frame is the lower leg with shin injuries - tibealis anterior compartment dysfunction (shin splints) and stress fractures and achilles tendon disorders - functional equinus (tight tricep surea) and tendonitis accounting for on average 8% of all running injuries each.

It is also interesting to note that despite the popularity of the sport and the increase in awareness of injury, its causes, preventions and cures the relationship between incidence and site of injury has not changed since 1978 as shown in Table 2.2.3. Here the data in Table 2.2.3 has been subdivided into pre and post 1982, the start of the athletic boom in this country, (excluding Krissoff et al., 1979) and shows that whilst there has been a slight decrease in the occurrence of injury in the knee, the Achilles tendon and the shin area it cannot be seen as a significant fall.

TABLE 2.2.2 Summary of Major Injury Sites in Athletes

Investigators	Number of Athletes	Major Injury Site(%)		
		Knee	Achilles Tendon	Shin
Blair et al., 1987	438	31	11	11
Clement et al., 1981	1650	42	8	11
Clough et al., 1987	502	26	5	2
Eggold, 1981	146	39	8	5
James et al., 1978	180	29	11	13
Krissoff et al , 1979	not stated	25	18	15
Lysholm et al., 1987	60	13	8	13
Maughan et al., 1983	497	32	11	6
Orava et al., 1979	1311	18	9	9
Pagliano et al., 1980	1077	30	6	4
Watson et al., 1987	257	20	2	20
Total Athletes (1)	6118			
	Average (1)	30	8	8

(1) Excludes Krissoff et al., Data

TABLE 2.2.3 Major Injuries Expressed as Pre & Post 1982, The Beginning of The British Running Boom

Date (1)	Number of Athletes	Major Injury Site %		
		Knee	Achilles Tendon	Shin
Pre 1982	4364	31.1	7.9	8.6
Post 1982	1754	27.7	7.8	8.4

(1) See References in Table 2.2.2

James et al. (1978) identified the reasons behind these injuries and classified them into three main areas as training errors, anatomical factors and shoes and surfaces. Training errors account for 60% of all injuries, a figure substantiated by other studies (as summarised in Table 2.2.4). These training errors normally manifest themselves as either overuse injuries caused by excessive mileage or sudden changes in normal routines, or acute injuries caused by intensive actions such as sprinting or hill running for example.

Any rapid change in training regime can result in different muscle groups or actions being called upon which might not otherwise be commonly utilised by the athlete. Different surfaces require different running actions, road running in the wet, grass, mud or gravel running may all strain those muscles, ligaments and tendons not normally called upon.

TABLE 2.2.4 Major Cause of Athletic Injury - Training Errors

Investigators	Percentage of Injuries Caused By Training Errors
Clement et al., 1980	50%
James et al., 1978	60%
Krissof et al., 1979	60%
Orava et al., ,1979	22%

An increase in speed can only usually be accommodated with an increased range of motion and cadance (Mann et al., 1986). Thus muscles are stretched beyond their familiar working range. This can also occur due to a change in footwear. Racing shoes, and spiked running shoes, built for lightness, offer less protection (heel lift) in the rear foot area by minimising or eliminating midsole cushioning. This results in a stretching and invariably a strain, of the Achilles tendon.

Hill running is specifically incorporated into training programs to 'overload' the lower limb muscles by working against gravity. If done in a sensible and progressive manner it is an effective and common training regime. If not undertaken carefully the 'overload' cannot be accommodated by the muscles, joints, ligaments and tendons and injury occurs. Downhill running is equally hazardous as continuous breaking may be required putting greater emphasis on the anti-gravity muscles such as the quadriceps muscle group. The anterior shin group may also be overworked during hill running, both down and up as the action of foot dorsi-flexion is greatly exaggerated.

A more in-depth study of the causes of running injuries shows that these causes are not as simple as summarised above with eight major catagories identified and presented in Table 2.2.5. Whilst these data are varied in both their findings and presentation the underlying trend shows that excessive mileage, shoe design and functional anatomy are primarily the cause of a great number of disorders. Excessive mileage has already been highlighted as a prime cause and athletic and running shoe design has become an exacting science and will be discussed later in Section 2.3.

Functional anatomy disorders however are even more diverse and complex and can be hereditary in nature or caused by previous injuries (over compensation) or from neuro-muscular dysfunction. Some of the most common of these disorders are collated in Table 2.2.5 and can commonly be termed bow legged (tibeal valgus), pigeon toed/knock kneed (tibeal varum), flat feet (pes planus), high arched feet (pes cavus) and pronated feet (subtalar varus) (Debrunner, 1982).

TABLE 2.2.5 Causes Identified for Running Injuries

		Investigators									
		Taunton, 1979	Newell et al., 1984	Lloyd-Smith et al., 1985	Smart et al., 1980	Lysholm et al., 1987	Matherson et al., 1987	James et al., 1987	Taunton et al., 1982	Armstrong, 1986	Dressendorfer et al., 1983
	Number of Athletes	N/S	329	204	N/S	60	320	180	N/S	N/S	12
Item	Cause										
1	Rapid changes in training		46%					N/S	25%		
2	Excessive Mileage (General overuse)	N/S	40%		N/S	72%		[7]29%			
3	Intensive workouts		32%		N/S			N/S			
4	Hill training		27%					N/S	15%		
5	Shoe design	N/S	23%		N/S	[4]8%			20%		
6	Race Related Training		22%								
7	Changes in surfaces		20%		N/S						
8	Functional Anatomy	N/S	[2]29%			[5]11%		[8]N/S			
	a tibeal varum			63%			20%	N/S	10%		
	b rear foot varum						74%	58%	12%		
	c forefoot varum						77%		17%		
	d tricep surea inflexibility								20%		
	e leg length discrepancies			31%				N/S			
	f rearfoot valgus		N/S								
	g genu varum						40%	N/S			
	h cavus foot							20%	15%		
9	Biochemical status	N/S								N/S	N/S

NOTES:
N/S = Non Specified
1 = Study of Achilles Tendon
2 = Includes (a) to (d) inclusive
3 = Study of Hip Injuries
4 = Shoe & Surface Interface
5 = Includes (d), (f) & (g)
6 = Study of Stress Fractures
7 = 60% overall training errors; Items 1-4 in table
8 = Includes (a), (b), (f) & (g)

TABLE 2.2.6 Common Cures to Athletic Injuries

Investigator	Activity	Injury	Prevention/Cures
Clement et al., 1984	Running	Achilles	Anti Pronation Shoes & orthotics. Flexibility and strengthening regimes.
Clement, 1974	"	Shin Splints	Greater shoe cushioning (less hardness)
Grana, 1985	"	Knee	Non change in routine/ intensity
Newell et al., 1984	"	Knee	Orthotics
Subotnik, 1985	"	Foot	Greater fitness
Smart et al., 1980	"	Achilles	Shoes, well fitting. Orthotics Avoiding training errors Flexibility & strengthening regimes
Taunton, 1979	"	All	Flexibility regimes Correct footwear Orthotics Medication(1) Correct training regimes
Viitaslou et al., 1983	"	Shin Splints	Shoes, correct sole hardness
James et al., 1978	"	All	Orthotics Reduced mileage Shoe change/modification Medication(1) Surgery
Dressendorfer, 1983	"	All	Reduced mileage Medication(1)
Taunton et al., 1982	"	Plantar Fasciitis	Correct training schedules Correct training surfaces Increase flexibility/ strength Correct shoes Orthotics
Taunton et al., 1987	"	Knee	Medication(1)/rest Orthotics/correct shoes Strengthening/flexibility Physiotherapy

(1) = Anti-inflammatory drugs and steroidal compounds for exmaple.

James et al. (1978) and Clement et al. (1980) reported that the three most common modes of treatment for injuries were rest, complete or partial; footwear modification and physiotherapy (including surgery). These can be expanded upon by scrutinising Table 2.2.6 which shows a wide variety of corrective measures can be and are being applied to athletic disorders. These varied preventions and cures show that there are no definitive treatments for specific injuries and that most injuries can be treated and/or prevented by careful training regimes and sportswear. As surgery and physiotherapy can be expensive and extensive in time lost to the sport, prevention has to be the primary aim of all athletes involved in "high mileage" training.

As outlined in Section 1, 4.4 million athletes in Great Britain ran in 1987. If this figure is related to the 66% injury rate proposed by Krissoff et al. (1979) then 2.9 million athletes will have incurred an injury during that 12 month period. This is obviously a large and alarming figure when consideration is given to the passive nature of the sport compared to other more explosive sports. In addition when the popularity of long distance running is compared to the increasing likelihood of injury with increasing mileage the effect on an otherwise healthy population can be viewed with concern.

These injuries, commonly occurring within the lower limbs and majoring around the knee joint can be seen to be caused by a multitude of reasons but have been categorised as training errors, anatomical factors and the foot/ground interface. Cures and treatments are equally diverse but correct training regimes are clearly shown as being an adequate form of preventative measure.

Thus it is this prevention via an understanding of injury that should be the aim of all athletes.

2.3 Factors Affecting Injury: Footwear & Running Style

Footwear modification, either by complete change of shoe or with the use of an orthosis, can alter various parameters of human gait cycle such as pressure patterns underneath the foot, foot/ground reaction forces, kinematic patterns of heel strike i.e. the control of excessive rearfoot movement, specifically eversion of the calcaneous, termed pronation. All these have been shown to be specifically beneficial in the treatment of running injuries (James et al., 1978; Clement et al., 1980; Scranton et al., 1982) as for instance the occurrence of chondromalacia patella has been correlated with excessive pronation. (Clarke et al., 1983b).

With this in mind the running shoe manufacturers have produced a wide range of running shoes with technological features to help reduce injury and optimise performance. The number of specialist running shoe companies has also proliferated and Running Magazine (No. 85, May 1988) report that there are 22 major shoe companies producing 213 different shoe models (spiked running shoes are excluded from this summary).

A summary of some of the major shoe companies selling in Great Britain illustrates the variety of technological features incorporated into running shoes. The Nike company had been incorporating pressurised gas for cushioning into some of their shoes midsoles, either full length or just in the heel wedge. Encapsulated in polyurethene (PU) this pressurised gas eventually escaped and has now been superceded by the use of Freon, an inert gas with larger molecules that are unable to escape through the urethene skin. The use of these "Air-Soles" has been claimed to retain 98 percent of its cushioning capability after 800 km (500 miles) of usage. This compares favourably with conventional materials such as ethylene vinyl acetate (EVA) or blown rubber compounds which have been shown to lose between 67% and 70% of their initial cushioning properties over 800 km of usage (Harrison et al., 1984 and Cook et al., 1985).

An alternative to "Air-Soles" has been the development by Asics Tiger with the "Gel" system in which a viscous silicon based gel is injected into pads in an EVA midsole unit. These pads, positioned under the heel and forefoot act like visco-elastic materials in absorbing shock but are substantially lighter than traditional materials.

Turntec have incorporated the best of both the above two systems, the lightness of the Freon based system and the increased cushioning of the silicon gel system to produce a brand named ZO^2 cushioning agent. This combination of silicon rubber and air, placed inside the sockliner at the heel is aimed at reducing impact forces only.

Reebok have developed an 'energy return system' based upon open ended Hytrel plastic tubes laid laterally across the middle at the rearfoot surrounded by compression moulded EVA. As the foot strikes the ground the tubes compress momentarily before springing back to their original shape.

The New Balance Encapsulation system combines the lightness of EVA with the durability and stability of PU. EVA, as shown by Harrison et al. (1984), can lose its cushioning properties quickly but the plastic resin of PU is an excellent shock absorber and does not lose incompatible compounds have hence been combined to produce a full length midsole material - an advantage to all styles of runners not just heel strikers or forefoot strikers.

All of the above features, aimed at reducing impact forces, are incorporated into the midsole. Avia however have used the outsole in their design of the Cantilever Sole which "flares" out on impact to absorb shock and increase stability. This being an extension of the Nike Waffle - Centre of Pressure outsole which provides small deformable studs all over the outsole.

Increasing cushioning is not always beneficial as a softer sole system can result in excessive pronation/supination which as stated above has been linked with knee injury (Clarke et al., 1983b). Hence combining cushioning with rear foot and/or total foot motion control has become a primary aim of all major running shoe manufacturers. This has been achieved with the use of varus or valgus wedges which when inserted into the shoe places the rear foot in an inverted or everted position prior to footstrike so as to control over pronation or over supination accordingly, also, within the running shoe a heel wedge is included to provide heel lift, or height variation between the fore- and rear-foot regions and to provide a heel flare to stabilise the rearfoot on foot strike, flare being the angle made between the heel and the ground when viewed posteriorly. This has also been shown to influence the amount of pronation, propulsion force and power developed during foot stance (Clark et al., 1983a).

Split density or composite outsoles and midsoles are also used in which softer material is provided over the impacting regions of the outsole but harder, more stable material is utilised over the remaining areas to produce the stability and performance response characteristics required by the athlete (Nigg, 1986 and Dickinson et al., 1985). See Plate 2.1.

The effects of these technological features on various kinematic parameters can be seen from various studies carried out over the last few years and specifically studies on the effects of foot ground reaction forces as measured by force platforms. These external forces are those that are usually responsible for the acute stresses placed on the body that leads to injury. Maximum foot/ground reaction forces here are in the region of three times the body's weight (3BW) (Cavanagh et al., 1980a) but are dependant upon running speed, footwear and foot strike characteristics. Table 2.3.1 summarises the findings of various investigations and indicates that the use of footwear, speed or dynamics of activity can influence the magnitude of foot/ground reaction forces.

The material used in the construction of athletic footwear has also been analysed both in vivo and in the laboratory. Studies by Clarke et al. (1983a) show that maximum foot/ground reaction forces can vary by 8% when comparing an "Air-sole" shoe with an EVA soled shoe with a Shore A hardness value of 45 (regarded as a hard sole - normal range: 25 = soft, 35 = medium, 45 = hard). A three percent variation was found between EVA hardnesses. Not only does the peak impact force alter with material used but the time at which it occurs with foot stance also varies, this peak being reached later with softer soled shoes (Clarke et al., 1983a; Nigg, 1986) thus reducing muscular loading rates. However no significant differences were found between conditions for overall support or contact time

PLATE 2.1. Running Shoe with Split Density or Composite Outsoles

TABLE 2.3.1 Comparison of Foot/Ground Reaction Forces in Various Activities

Investigators	Activity	Speed and/or Other Characteristic Values and Conditions	Max.Foot/Ground Reaction Forces in Body Weight Units
Alexander et al., 1975	Walking	$1.8m.s^{-1}$ N/D(4)	1.5
	Running	$3.9m.s^{-1}$ N/D(4)	2.9
	Jumping	Standing Take-Off N/D(4)	1.3
	Landing	From 0.81m.N/D(4)	1.5
Burdett, 1982	Running	$4.5m.s^{-1}$ N/D(4)	2.8
Cavanagh et al., 1980a	Running	$4.5m.s^{-1}$ FFS(1)	2.7
	Running	$4.5m.s^{-1}$ RFS(2)	2.8
Cavanagh et al., 1980b	Walking	Slow-Barefoot	900(5)
	Walking	Slow-Shod	1600(5)
	Jumping	Vertical-Barefoot	1400(5)
Clarke et al., 1982	Running	$2.7m.s^{-1}$ RFS(2) N/D(2)	2.8
Clarke et al., 1983a	Running	4.5m.s Soft Sole	2.7
	Running	4.5m.s Hard Sole	2.8
Draganich et al., 1980	Walking	Normal Speed-Barefoot	0.9
Elftman, 1939	Walking	N/S(3)-Barefoot	1.3
Frederick et al., 1981	Running	$3.4m.s^{-1}$	2.0
	Running	$3.8m.s^{-1}$ Barefoot	2.3
	Running	$4.5m.s^{-1}$ RFS(2)	2.9
Nigg et al., 1981	Running	$5.5m.s^{-1}$ RFS(2) N/D(4)	3.6(6)
Simon et al., 1981	Walking	N/S(3) Barefoot	1.3
	Walking	N/S(3) Shod	1.3
Smith, 1975	Landing	From 1.0m N/D(4)	4.0

(1) FFS = Front foot striker - First contact of foot with ground
(2) RFS = Rear " " " " " " " "
(3) N/S = Speed not specified
(4) N/D = Footwear not declared
(5) Values of force expressed as body weight/foot contact area.
(6) Values calculated for a 70 kg body weight.

possibly indicating that the human body had to work harder to maintain a constant velocity.

As mentioned earlier a softer shoe, whilst reducing impact forces, increases the amount of pronation present in the musculo-skeletal system and this has been correlated to injury. Clarke et al. (1983b) found that by varying the heel lift and heel flare rearfoot motion could be controlled and that hard soled shoes (Shore A hardness 45) provided the 'best' anti pronation results. Nigg and Morlock (1987) extended Clarke's work and suggested the use of a negative heel flare to reduce initial rearfoot motion and possibly reduce the likelihood of anterior medial compartment syndrome (shin splints).

The use of orthotics (soft, semi flexible or rigid) or inserts within existing shoes can reduce impact forces (Subotnik, 1983) and the angle of pronation from 11 degrees, when wearing a running shoe, to 7 degrees for the same shoe with an orthotic inserted (Bates et al., 1979). This use of an inserted device has been shown by Eggold (1981) and James et al. (1978) to be effective in the treatment of 'overuse' injuries - 40% and 78% of patients claimed 100% relief with their use respectively.

Not only can mechanics of the lower extremities be affected by footwear modification but physiological body measures such as the condition of the red blood cells responsible for oxygen transportation to the muscles, haemoglobin and haematocrit, can also be influenced. Softer soled running shoes reduce the deterioration of these physiological measures and as such they are more efficient in minimising the acute stresses placed on the various components of the lower extremities in endurance running (Falsetti et al., 1983). Taunton (1979) has also noted elevated uric acid levels as a gout-like state during endurance training in the development of tendonitis of the achilles. The use of softer soled shoes may help in limiting the increased turnover of tissue protein and red cells that lead to an increase in the purine pool in the blood, a forerunner of uric acid and hence lactic acid production (Smith et al., 1983).

Hence with a bewildering range of shoes available, a range of technological features "needed" to optimise performance and a wide fluctuation of price to be considered, choosing a running shoe can be a difficult process. Bates et al. (1980) tried to identify a "best" shoe but failed due to the complex nature of each athletes feet being anatomically and functionally different. With this in mind a number of attempts have been made to help the athlete choose his/her most suitable running shoe (Cavanagh, 1980 & 1987; Nigg, 1986). Appendix 1 gives one example of this advice.

Not only is the running shoe market complex but the running styles of athletes, as in any human kinematic activity, are widespread and a number of studies have tried to categorise the running action according to certain performance characteristics. Cavanagh and Lafortune (1980) in assessing ground reaction forces in distance running identified, from the subjects centre of pressure patterns under the shoe sole during the running action, groups of runners who could be classified as rear-foot, mid-foot or front-foot strikers i.e. depending upon where first contact was made with the ground. Of these three categories two are commonly cited: rear-foot from the above and fore-foot meaning any foot strike pattern not resembling a rear-foot case. Kerr et al. (1983) studied two races of 10 km and marathon duration and found that 80% of the participants were heel or rear foot strikers (the remainder classified as mid-foot strikers). The study also showed that as speed of gait increased then so did the percentage of mid-foot strikers.

The above two studies refer to distance running and recreational runners but little has been written about establishing categories for such groups as recreation runners, joggers, distance runners, experienced athletes, elite runners, sprinters etc. A running speed of 4.47 $m.s^{-1}$ (6 minutes per mile) seems to becoming the norm for biomechanical evaluation of distance runners (Cavanagh & Lafortune, 1980; Frederick et al., 1981; Clarke et al., 1983a; Burdett, 1982, Harrison et al., 1986 & 1987). However, in the study of rear foot kinematics a slower speed of around 3.8 $m.s^{-1}$ (7 minutes per mile) appears to becoming standard (Clarke et al., 1983b; Bates et al., 1980; Nigg & Morlock, 1987).

Other categories are less well defined and in some scientific studies running speeds are not recorded thus making verification difficult. The precise guidelines for jogging, running and sprinting need establishing as do the groups of joggers, recreational runners, distance runners and elite ahtletes. It may be suggested that distance runners are those who on average complete approximately 60 km per week or more and compete regularly for an athletic club over distances in excess of 1500 m (1 mile). Elite athletes may be considered as those who have competed for or at National level at their chosen event and are not just highly experienced due to running longevity.

The final category used in assessing athletes is, as mentioned above, their rear foot motion pattern i.e. pronation or supination. Here depending upon the amount of eversion of the calcaneus, a subject can be said to overpronate, or if excessive inversion or very little subtalar motion occurs. If the athlete contacts the ground first with the medial aspect of the foot then transfers his centre of pressure profile to the lateral aspect, he can be said to be a supinator (Cavanagh, 1980 & 1982; Nigg, 1986). These categories are often linked to anatomical dysfunctions such as pes planus (flat feet) for over pronation and pes cavus (rigid - high arched foot) for excessive supination (Nigg, 1986). Procedures for identifying or classifying arch indexes are now common (Cavanagh & Rodgers, 1987).

As suggested earlier, the load on the musculo-skeletal system is 'provided' by the foot/ground reaction forces and these have been shown to be dependant upon a number of factors. Shoe and orthotic design, running speed and style have all been seen to alter the forces presented to the human lower limbs. Hence in any experimental design these parameters must be controlled and a data base of information built up based on a standard shoe (and surface), speed and style (where this should include classification of experienced and non experienced athletes and illimination of any effects of fatigue). If a two dimensional study is to be performed then third dimension classifications can be ignored.

Once this data base of results has been established the effects of footwear, speed, style, experience, fatigue, sex etc., can be examined by varying them systematically and observing their influence on the musculo-skeletal kinetic system. Then the identification of possible causes of injury may be made.

2.4 Techniques for the Assessment of Human Kinetics

In the assessment of the musculo-skeletal system the standard data collection equipment consists of an optical recording device and a force recording medium. From these two basic processes the majority of biomechanical analysis can be performed.

Optical systems available to the researcher include high speed cinematography, optoelectronic movement systems, television based video systems and specific digital cameras.

High speed cinematography is the most commonly used by investigators, usually 16 mm but with 8 mm ('Super 8') becoming increasingly more viable due to its smaller size and the development of higher speed cameras. Smith (1975), Seedhom & Terayama (1976), Ariel (1974) and Burdett (1982) used cinematographic techniques with frequencies of between 60 and 100 Hz.

The use of cinematographic equipment allows joint positions to be marked directly onto the skin, which can lead to errors of between 50 and 100% (Smith, 1975) in the calculation of muscle and joint forces from incorrect moment arms. Conversely joint coordinates can be omitted and bone models reproduced from either individual radiographs (Seedhom & Terayama, 1976) or cadavers (Burdett, 1982). The position of the joint centres and muscular lines of action can then more subjectively be established. Even if the joint centre is obscured by another limb its position can easily be established with the aid of a bone model. Other advantages of cinematographic techniques over corresponding optical systems are resolution, cost and unrestrictive subject movement. However the required analysis method, digitising, along with the time taken to develop the film makes this a time consuming, labour intensive method.

Optoelectric movement systems (for example, Selspot, CODA3) and some television-video systems (VICON) use light sensitive receiving cameras and as such early systems could not be used in daylight but they do have the advantages of on line processing of data. Subjects also wear emitters or reflectors and this can restrict movement. They normally operate at a maximum of 50 Hz which may be too slow for dynamic sporting activities (Nissel & Mizrahi, 1988). Resolution may be ± 3 mm and processing time can be reduced to three hours. However set up times as large as six hours may be necessary per subject whilst costs are relatively high in comparison to other systems (Winter et al., 1972; Dainty & Norman, 1987; Grieve et al., 1975; Plagenhoeff, 1971).

Digital cameras, linked to microcomputers, have the advantage of being cheap ($300 at 1987 prices) and portable. However the frame rates are extremely low at 13 to 35 Hz and resolution poor at between 5 and 10 mm. They do, however, provide results within 20 s of an activity being completed (Vaughan et al., 1987).

Overall, although data processing time for cinematographic techniques is long, resolution, subject invasion, technological familiarity and cost make it a highly popular optical recording device.

Force recording devices are usually based on a force platform (or pressure platform) although the use of smaller transducer elements such as strain gauges have been used. Komi et al. (1987) report on the implant of a strain gauge into the human tricep surea (achilles) tendon to measure muscle forces. More commonly accelerometers are mounted onto external aspects of the human limbs in order to record accelerations from which the corresponding displacement, velocity and inertial force data can be extrapolated. To improve the response characteristics Light et al. (1980) attached these directly to the tibia (Pezzack, 1977; Dainty & Norman, 1987; Winter, 1979).

The need to attach accelerometers to bone to reduce interference or 'noise' created by skin movements makes the use of this technique difficult to justify to a large number of athletic volunteers. The attachment of wires to the subjects' limbs also restricts movement and should kinematic data be required further obstructions such as goniometers are needed. Force platforms offer none of these drawbacks and can be located under a walkway or running track out of sight of the subject. Thus the use of force platforms has proliferated and the majority of bioengineering studies involving gait analysis, or similar actions, have utilised the equipment. Smith (1975) - drop jumps, Burdett (1982) - ankle joint biomechanics in running, Ariel (1974) - weightlifting, Seedhom & Terayama (1976) - sitting to standing and Nissel & Mizrahi (1988) - step jumps all used force platforms in their studies.

Angular data can also be obtained via electrogoniometry or potentiometry either two dimensionally or three dimensionally. Traditional measurement devices have tended to be bulky and as such somewhat restrictive. Their use in sports biomechanics has seen limited use (Chao, 1980; Peat et al., 1976; Dainty & Norman, 1987; Winter, 1979). Nicol (1987) however has produced a flexible electrogoniometer which allows general unrestrictive movements thus being suitable for sports studies.

As suggested if high speed cine film has been used the data analysis is usually carried out with the aid of a digitising tablet linked directly to a computer and storage device. Commercially available digitisers vary in size and accuracy and/or resolution from for example, Grafpad II with a resolution of 0.7 mm and working area of 250 mm x 200 mm to a TDS HR 48 digitiser of 1.2 m x 0.9 m working area and resolution of 0.025 mm. Both these devices operate on a magnetorestrictive principle in comparison to a sonic digitiser such as the Grafpen GP6 which by design permits back projection (Kerwin, 1988). Front projection offers greatest flexibility of magnification and configuration (front or overhead) and magnetorestrictive devices are well suited to this. They also can be easier to use in comparison to the sonic versions where obstructions to the data collecting microphones need to be avoided.

Whilst using high speed film projected onto a digitiser a number of investigators have found the use of tailored bone models, cardboard cut outs of the subjects skeleton obtained from radiographs, to be particularly beneficial in the identification of muscle attachments, and hence lines of action (Seedhom & Terayama, 1976; Harrison, 1982). Burdett (1982) however used data from five cadavers, positioned in 10° intervals, to obtain average moment arm values in the study into ankle joint biomechanics. Ariel (1974) viewed 50 radiographs to establish standard knee joint and muscle moment arms in a weightlifting investigation. These two methods overcome the need to mark joint centres on the subject's limbs, inherently inaccurate due to skin movements, but the use of standardised or scaled data introduces errors and uncertainties. The use of individual bone models from each subject's own radiographs has been shown by Harrison (1982) to reduce knee joint force estimates by 40%.

Muscular activity can be identified by the use of electromyographic (EMG) techniques. Two common forms exist, surface electrodes and wire implanted electrodes. The implanted electrodes offer greater analysis power as they are smaller and hence can be used on the smaller muscles or used to reduce interference caused by other nearby muscles (Basmajian, 1978). Ethically, however, the implantation of these electrodes is open to question. Surface electrodes are commonly used to record the actions of the large superficial muscles (groups) such as gastrocnemius. Guidelines for standardisation of EMG testing routines and procedures are available in Dainty & Norman (1987).

A number of investigators have tried to correlate the level of activity recorded by the EMG signal with the force developed in the individual muscle but this cannot be substantiated (Jorge & Hull, 1986; Hof, 1987). This level of activity, produced after integrating the output signal, is useful for observing the role of the muscle in the gait cycle. For instance Elliott et al. (1979a) showed that the major muscle groups in the leg increased their activity by up to 46% for a speed increment of 1 m s^{-1} and these results correlate well with similar studies by van der Straaten et al. (1975); Sutherland (1966); Cappozzo et al. (1976) and the University of California (1953). Actual muscle force developed is not however a function of the level of activity and is difficult to determine, direct measurement is not possible and hence mathematical or mechanical modelling of muscle is needed.

From the recorded literature, equipment and techniques for the assessment of human kinetics are numerous. In the study of sports biomechanics however the range of available experimental tools is more limited. Higher frequency optical recording systems are needed and at present this can only be achieved with cinematographic equipment. This however also provides the greatest resolution, familiarity and non-invasiveness to subjects. Similar non-invasive force recording devices provide high frequency, high resolution and familiarity. Force platforms also provide an ideal partnership with cinematographic film.

Data analysis can also be greatly assisted and accuracy improved frontal projection onto a magnetorestrictive digitising tablet combined with individual bone models.

Finally, the modelling, data collection, processing and analysis can all be verified by the simple use of EMG with surface electrodes.

2.5 Loads in the Human Lower Limbs

Modelling of the mechanics of muscle is difficult due to the numerous ways they can function, this could be by concentric contraction (muscle shortening), eccentric contraction (muscle lengthening), isotonic (contraction under constant load), isokinetic (constant velocity) or isometric (constant length) as suggested by Elftman (1966). Muscle structure also alters the mechanics of muscle with pennate muscles, which attach over a skewed angle, 'pulling' indirectly between origin and insertion (Alexander and Vernon, 1975). Muscle strength, force and power are all dependant upon the speed of its contraction, for instance maximum power is obtained at 30% of maximum isometric strength, this strength being reported by Haxton (1944), in terms of maximum stress, to be 380 KPa, in the tricep surea, and by Alexander and Vernon (1975) to be 350 KPa, for the quadricep muscle group. Maximum force, like power, is obtained at submaximal muscle velocity. Morrison (1969) noted that maximum force was obtained at a muscle velocity of 0.15 $m\ s^{-1}$ when compared to other velocities of 0.375 $m\ s^{-1}$ and 0.5 $m\ s^{-1}$. Morrison hence produced length tension curves which subsequently used by Grieve et al. (1978); Abbott and Wilkie (1979); and Elliott et al. (1979b). These length tension curves however are not a function of the contractile characteristics of muscle mentioned earlier. The ultimate tensile strength (UTS) of muscle has also been found to be dependant upon the level of activity within it, passive muscle failing at lower force values when compared to stimulated muscle (Garrett et al., 1987). This UTS for striated muscle has been quoted as between 10 and 22 MPa. Tendons are also time or speed related as their ultimate tensile strength can vary from 34.6-147 MPa depending on rate of loading (Abrahams, 1967; Cronkite, 1936; Walker et al., 1964; Yamade et al., 1970).

The recruitment of individual muscles within a group is also not a directly determinable aspect of the mechanics of muscle with 'shunt' and 'spurt' muscles being recruited depending upon the level of activation within the group (Jackson et al., 1977). The use of springs and dashpots to mechanically model the muscles has also been heavily studied and is being constantly revised (Baildon et al., 1983; Hill, 1938).

The mechanics of muscle is hence a complex problem. Direct measurement of force within muscles is not possible, only activity and level of activity can be readily measured. Contractile mechanisms take various forms and are complicated by their speed and structure, or attachment modes, for each muscle. Activity of each of the muscles within a group is also difficult to quantify.

Bone and cartilage mechanics is equally complex. The critical limits quoted in the literature for features such as UTS generally do not qualify whether static, dynamic or quasistatic conditions were employed. Thus with bone and cartilage again having time dependancy characteristics dynamic impacts may appear to exceed 'known' biological limits (Nigg, 1986), both cortical and cancellous bone increases in strength and stiffness by over 50% with an increased rate of loading (Frankal & Nordin, 1980; Currey, 1975; McElhaney, 1966).

Considering bone on its own may also create misleading conclusions being drawn. The UTS of long bone and compact bone are quoted in the region of 100 MPa and 150 MPa respectively, (Nigg, 1986; Frankel & Nordin, 1980), however this does not consider the role of the cartilage in damping out any loading and spreading the contact area (Frankel & Nordin, 1980). Bone rarely fails in tension in sporting events, however, and the ultimate compressive strength of cortical (long) bone can be 50% greater (Reilly & Burstein, 1975).

Cartilage, the covering on the articulating surface of the bones provides the functional connections between the bones, within a synovial joint, and allows relative movement with minimal friction whilst dispersing the joint load over a larger area than would be otherwise available. The most important features of cartilage are the mechanical properties of collagen which constitutes 60% of cartilage. This material has a tensile stiffness in the region of

10^3 MPa and a UTS of between approximately 50 & 70 MPa (Kempson, 1973 in Frankel & Nordin, 1980) but is also rate dependant due to its low permeability to the proteoglycan gel present in the cartilage. Thus during rapid loading or impact there is insufficient time for the gel to be exuded and the cartilage structure sets almost elastically. For slow loading rates the proteoglycan gel has time to be exuded and a viscoelastic behaviour is exhibited (Frankel & Nordin, 1980).

The complex and exacting nature of the human body means that direct representation of an individual or group of structures cannot take place. Biomechanical modelling of the human body is hence needed to understand forces in the muscles and joints in locomotion. This biomechanical modelling must simplify the actions of the muscles, joints, and surrounding tissue, while adhering to the basic functional aspects of the human body outline above.

However, the main limitation of modelling biomechanically is that the many muscles and even muscle groups acting about each joint outnumber the limbs degrees of freedom hence making the system statically indeterminate. For example Seireg et al. (1975) identified 31 'important' muscles of the lower extremities (Table 2.5.1), and when added to three components of joint reactions at each of 3 joints this gives a total of 40 unknowns for a full three dimensional analysis.

This does not include the roles of the ligaments in the lower limbs as studied by Moeinzadeh et al. (1983), Blacharski et al. (1975) and Crowninshield et al. (1976). For a three segment model, with a full six degrees of freedom per segment, only 18 equations can be produced - hence the indeterminacy of the model.

TABLE 2.5.1 Important Muscles of the Lower Extremities

Muscle No.	Muscle name	Joint[(1)] Crossed
1	Gracilis	H, K
2	Adductor longus	H
3	Adductor magnus(adductor part)	H
4	Adductor magnus(extensor part)	H
5	Adductor brevis	H
6	Semitendinosus	H, K
7	Semimembranosus	H, K
8	Biceps femoris (long head)	H, K
9	Rectus femoris	H, K
10	Sartorius	H, K
11	Tensor fasciae latae	H, K
12	Gluteus maximus	H
13	Iliacus	H
14	Gluteus medius	H
15	Gluteus minimus	H
16	Biceps femoris (short head)	K
17	Vastus medialis	K
18	Vastus intermedius	K
19	Vastus lateralis	K
20	Gastrocnemius (medial head)	K, A
21	Gastrocnemius (lateral head)	K, A
22	Soleus	A
23	Tibialis anterior	A
24	Tibialis posterior	A
25	Extensor digitorum longus	A
26	Extensor hallucis longus	A
27	Flexor digitorum longus	A
28	Flexor hallucis longus	A
29	Peroneus longus	A
30	Peroneus brevis	A
31	Peroneus tertius	A

(1) H=Hip
K=Knee
A=Ankle

From Seireg & Arvikar (1975)

The indeterminacy can be increased if anatomical constraints are imposed such as condylar loading limitations (Minns, 1981), variations in friction/contact characteristics in joints with changes in leg angle (Ellis et al., 1980), stress distribution and transfer through bones (Chand et al., 1976) and the tangential path of some of the muscles (Jensen & Davy, 1975). Hence alternative solution routes are necessary and these involve mathematical optimisation which incorporate into the model additional assumptions to resolve the indeterminacy and uniquely apportion segment resultant forces to the anatomical structures.

Two basic techniques exist in linear optimisation algorithms or non-linear or inverse-non linear formulation. Non linear techniques have the advantage of "allowing more active muscles into the solution (i.e. predicting synergism) without the formulation of additional constraints" (Pederson et al., 1987). To overcome indeterminacy both processes utilise additional equations based upon physiological observations such as maximum permissible muscle stress, work done or joint force, these are termed objective functions and the results obtained are usually confirmed with electromyographic justification (Crowninshield et al. 1978).

Seireg & Arvikar (1973 & 1975) set out their five objective functions used in a linear programming gait analysis of the lower limbs as consisting of:
(a) Minimising the total muscle force, (b) minimising the work done by each muscle from the product of its force and change in length, (c) minimising the total vertical joint reactions, (d) minimising the total joint moments and (e) apportioning force in each muscle by its cross sectional area. In a similar study, Patriarro et al.

(1981) used a thermodynamic objective function for the mechanico-chemical power minimisation of muscular energy output and the physiological stress limit of each muscle expressed as force divided by cross sectional area. An et al. (1984) and Bean et al. (1988) both used a double linear programming technique with two objective functions of minimisation of the total muscular intensity expressed as a function of muscle stress and minimisation of the total joint forces. The minimisation of total muscular effort, expressed as a proportion of muscle size, was used by Penrod et al. (1974) in an analysis of the wrist. In simpler analyses Hardt (1978); Chao et al. (1973 & 1976); Nissan (1981) and Barbenel (1983) used the minimisation of the total muscle forces as their objective function. All the above studies used linear programming.

Non linear programming has been used by Crowninshield et al. (1978); Crowninshield & Brand (1981); Dul et al. (1984a & b); Pederson et al. (1987). Here, in order to give more solutions additional constraints are introduced and Crowninshield & Brand (1981) utilised the phenomnenon that muscle contraction endurance time relates inversely to contraction force raised to a power. They developed a numerical criterion for maximum endurance of musculoskeletal function. Dul et al. used a non linear muscle fatigue criterion involving muscle fibre type as well as muscle size and moment arm. Pederson et al. (1987) used both linear and non-linear techniques and their non-linear objective functions were expressed as (a) minimisation of the sum of muscle forces; (b) minimisation of the sum of muscle forces cubed; (c) minimisation of the sum of muscle stresses cubed and (d) relaxation of the equality constraints for non linear minimisation of the sum of muscle stresses cubed.

All the above methods of linear and/or non-linear analysis give widely differing results due to the inherently different problem formulation and models and differing gait input data and hence direct comparison is not always possible. The majority of the studies utilise stress limits of individual muscles which have been shown to be strain rate dependant (Yamada, 1970; Abrahams, 1967; Walker et al., 1964). This also means that cross sectional area data is needed for all muscles in the model (change of lengths of each muscle is also required in the study by Seireg & Arvikar (1973)). Obtaining this data for a number of subjects and the muscles involved is not experimentally feasible for large numbers of living subjects (neither is obtaining muscle length changes). EMG verification is also required and this provides its own problems especially for the deep muscles which will need a needle electrode technique to elicit results - an ethically questionable practice.

For the non-linear techniques binomial expansions with empirically derived constants and powers are often 'tailored' to produce an algorithm. The complexity and inherently non-linearity of the human neuromusculo-skeletal system makes the production of reliable and efficient algorithms a considerable problem. As this involves an extremely laborious and expensive task and needs an expert team of system scientists, numerical mathematicians, highly experienced programmers and a mainframe computer, it is unlikely to become a common approach to solving sports biomechanics problems (Hatze, 1983).

Other alternative solutions have included a mechanical model produced by Dostal & Andrews (1981) consisting of 27 elastic strings representing muscles and a Kelvin model of 2 sets of springs and dashpots supporting one mass to represent the lower limbs of the human body (Siegler et al., 1982).

If a non-simplistic study into the action of the muscles, and joints, of the lower limbs is required, common techniques are not available. Rigid analysis requires complex mathematics which often lose sight of the purely anatomical limitations of the human body. The results are only estimations, and the degree of estimation must be countered by the complexity of the analysis. Producing mechanical models is extending the boundaries of biomechanics into robotics, which is beyond the scope of this study.

The assessment of forces within the sporting domain hence needs to utilise standard biomechanical equipment to enable repeatability to be achieved. This will more often consist of a force recording device such as a force platform and an optical recording device as in high speed cinematography. Evaluation of internal or human forces, however, has been shown not to be such a simple case and the complex nature of muscle, tendon, ligament, bone and other tissue make modelling a complicated matter. Whilst sophisticated data collection and evaluation equipment and validation procedures such as EMG can improve accuracy and repeatability, simplification of the human musculo-skeletal system must also take place. This simplification must permit biomechanical modelling to be performed for a mathematical model based upon sound anatomical reasoning to be produced.

In an attempt to gain an insight into items such as muscle and joint forces, muscle and bone loading rates and joint moments many investigators have tried to represent the human lower limbs in a simplified way. This biomechanical modelling has been based upon the understanding of the kinematics of the segments of the lower limbs during locomotion, for instance joint rotations and contact points (Nissan, 1980). The human gait cycle is primarily two dimensional and it is from this aspect that the majority of kinematic investigations have specialised upon (Sammarco et al., 1973; Morrison, 1970; Mann et al., 1980; Frankel, 1971).

Once the kinematic guidelines are established biomechanical modelling may be undertaken. Whitsett (1963) and Hanavan (1964) replaced the bones of the human body by geometrically simple 'segments'. Paul (1967) gathered all the muscles into their major groups depending upon function and proposed that all the muscles within that group acted synchronously between a general origin and insertion. The pull or tension developed by a muscle group being independent of speed, contractile mode or structure.

These and various other simplifications have been used in a number of investigations to estimate muscle and joint forces in human activity. Most have relied on simplifying the biomechanical model to perform a Newtonian analysis on the lower limbs. This involves considering the joints as having non complex articulating surfaces with negligible frictional forces (transverse loads being absorbed by the ligaments). Antagonistic muscle actions are also overlooked with one major muscle group contributing all the force.

Analysis of muscle and joint forces in the lower limbs of the human is common in sedate activities such as walking, and has been extended to include studies into rising from a chair and from a deep squat. The stresses imposed upon the musculo-skeletal system in these activities are well established (Table 2.5.2). However, dynamic activities have been subjected to less scrutiny, with only a

TABLE 2.5.2 Muscle and Joint Force Summary

Investigators	Area of Study	Activity	Muscle Force In Body Weight Units			Joint Reactions[1] In Body Weight Units			
			Calf	Ham	Quad	Ankle	Knee	Hip	P-F
Ariel, 1974	Knee	Squatting[2]					8.9		
Bishop, 1977	Knee	Squatting			16.0		16.0		6.0
Bresler et al., 1950	Hip	Walking						2.5	
Burdett, 1982	Ankle	Running - 4.5 $m\ s^{-1}$	10.0			13.3			
Crowninshield et al., 1982	Hip	Walking						4.3	
Dahlkvist et al., 1982	Knee	Squatting	1.4	2.2	6.9		5.6		7.6
Ellis et al., 1979	Knee	Rise from Sit	3.0	2.4	3.3		3.1		2.5
Ellis et al., 1985	Knee	Rise from Sit		1.6	3.2		3.2		3.4
Harrington, 1976	Knee	Walking					3.5		
Inman, 1947	Hip	Walking						2.6	
Morrison, 1968	Knee	Walking	1.5	2.0	1.1		3.0		
Morrison, 1969	Knee	Walking - Level	2.0	2.4	1.1		3.4		
		Walking - Upstairs	0.6	1.4	3.4		4.3		
		Walking - Downstairs	1.2	0.7	3.0		3.8		
		Walking - Up Ramp	2.6	1.9	1.3		4.0		
		Walking - Down Ramp	-	1.5	3.3		4.0		
Paul, 1967	Joints	Walking					3.4	3.9	
Paul, 1974	Knee	Walking - Level					2.8		0.6
		Walking - Fast					4.3		1.5
		Walking - Upstairs					4.4		1.8
		Walking - Downstairs					4.9		2.9
		Walking - Up Ramp					3.7		1.6
		Walking - Down					4.4		2.6
Proctor et al., 1982	Ankle	Walking	2.5			3.9			
Rohle et al., 1984	Hips	Walking	1.4		3.4	3.2	4.5	6.1	1.1
Seedhom et al., 1976	Knee	Rise from Sit	0.5	1.5	1.2		2.5		2.4
Smith, 1975	Lower Limbs	1 Metre Drop Landing	6.1		16.5	7.4	24.4		
Williams et al., 1968	Hip	1-Leg Standing						6.0	
Capozzo, 1983a	Trunk	Walking	Intersegmental Pelvic Force					0.9	
Capozzo, 1983b	Trunk	Running	Vertebral Column Lumbar Force					5.4	

(1) Joint reactions are the direct compressive forces. P-F = Patella-Femoral reaction.

(2) With 200 kg weight.

few studies reported in the literature. Smith (1975) studied a 1 m drop jump while Ariel (1974) evaluated deep knee bends during weight lifting. Burdett (1982) analysed the ankle during running, but used cadavers to obtain average dimensions of the joint. The emphasis of the majority of these studies has been upon an identification of joint reactions and their relevance in prosthetic design and rehabilitation.

From the summary of results of various investigations (Table 2.5.2) it can be seen that in the dynamic activities of running and jumping (Smith, 1975; Burdett, 1982) muscle and joint forces are considerably higher than in the more sedate activities. Knee-joint reactions of 24.4 times the subjects body weight were recorded, compared to an average 3.6 BW reported in walking. Ankle-joint reactions of 13.3 BW were recorded by Burdett (1982) in running, compared to 3.9 BW cited by Proctor & Paul (1982) for walking. Peak muscle forces are similarly elevated: calf-muscle forces of 10.0 BW were found during running (Burdett, 1982), whilst studies of walking give an average value of 1.7 BW. The highest muscle forces quoted in the literature are those occurring in the quadriceps group i.e. 16.5 BW (Smith, 1975) during a drop landing or 16.0 BW (Bishop, 1977) during a squat exercise. Patella-femoral contact forces show a wide range of values possibly reflecting the complex nature of this joint or the infrequency of its scrutiny.

These muscle forces are more commonly represented by the net moment acting about the specific joint. Calculation of these moments can be determined directly from the foot-ground reaction forces and gross joint spatial co-ordinates and thus require no investigation into individual muscular activity. Table 2.5.3 shows results from various investigations into different forms of gait. The knee joint moments rise from on average, 53 Nm during walking to 203 Nm during running. The hip joint moment shows similar increases from walking to running but it exhibits a significant increase when the speed of gait increases to sprinting: 77 Nm for walking; 179 Nm for running; 497 Nm for sprinting. The ankle however shows the least change of all three joints.

The significance of these high muscle and joint forces and net joint moments acting particularly in and around the knee joint may be appreciated when viewed in conjunction with the injury history of athletes. Table 2.2.2 shows that the major injury site of running related injuries in the knee joint and even in these sedate activities of walking knee joint compressive forces of up to 5 BW are reported. For those studies that investigated more dynamic activities (Smith, 1975) these forces have increased significantly to 24.4 BW.

The second most injured component in the human lower limbs as reported in Table 2.2.2 is the achilles tendon and in the studies by Smith (1975) and Burdett (1982) tensions of 6.1 BW and 10.0 BW are reported respectively. These high forces could be the cause of these injuries.

TABLE 2.5.3 Summary of Joint Moments Recorded in the Literature

Investigators(1)	Activity	Maximum Joint Moments (Nm)(2)		
		Ankle	Knee	Hip
Cappozzo et al., 1976	Walking	-160	45	-85
Olney & Winter, 1985		-120	80	-
Paul, 1985		-100	-35	70
Winter, 1983	Jogging	175	212	-95
Alexander & Vernon, 1975	Running	200	200	-
Harrison et al., 1986		-174	-188	98
Miller, 1987		200	-225	260
Nicol & Harrison, 1985		-	200	-
Mann, 1981	Sprinting	-250	280	-475
Sprague & Mann, 1983		-	250	-520

(1) Moment sense is dependent upon investigation sign convention.

(2) Stance phase only

The shin muscle, the third most injured component in the human lower limbs is not reported in the studies summarised in Table 2.5.2. Its significance in the walking cycle being less than in running.

Nearly all those forces and moments are recorded for sedate, two dimensional activities and hence when more dynamic actions are considered the force values multiply. How these are then affected by such items as footwear, speed and style, which have been shown via Table 2.3.1 to affect the foot/ground reaction forces, are unknown. If the effects on speed on the footground reaction force (Table 2.3.1) are as marked as the effects upon muscle and joint forces then this could be the reasoning behind training errors being the predominant cause of injury (see Table 2.2.4).

2.6. Summary of Literature Review

Injury occurs to 66% of all runners and the most commonly injured site is the knee joint. Causes of these injuries have been shown to include anatomical factors and footwear conditions and hence modification of the footwear worn has been identified as an important method of rehabilitation and cure.

A full analysis of muscle and joint forces in running has not been undertaken, the nearest dynamic activities including those of Smith (1975) and Burdett (1982). These and other studies have in general omitted the shin muscle group from the model and as this is the third most injured component in the human lower limbs it must be included in any model.

It is proposed that biomechanical modelling will allow a study of muscle and joint forces in running to be carried out. Controlling such variables as speed and footwear will enable anatomical anomalies of the gait cycle to be identified, varying running speed and categorising the athletes into rear and front foot strikers will enable the influence of speed and style to be analysed in accordance with the corresponding injury profiles outlined in Sections 2.2 and 2.3.

Hence a biomechanical model needs to be developed that will allow a detailed analysis of muscle and joint forces in running and specifically how the magnitude, sense and loading rates are affected by such variables as fatigue, speed, footwear, surfaces and running style. This may be important in understanding the aetiology of running related injuries.

An evaluation technique will also have to be devised that is both accurate and repeatable as no one standard method is reported in the literature. This should be designed to use standard biomechanical equipment.

3. THEORETICAL ANALYSIS

3.1 Introduction

Direct measurement of forces acting in the lower limbs of the human body can only be made after surgical implantation of a transducer into the appropriate region. This is usually done during the installation of a prosthetic device into a joint, for example the hip. This has severe limitations as an experimental method particularly for the study of athletic activities. Therefore in order to analyse muscle forces and joint reactions throughout the whole of the lower limbs a mathematical model, based upon biomechanical principles must be used. The studies by Seedhom & Terayama (1976), Smith (1975) and Burdett (1982) are ideal examples of this technique.

The human musculo-skeletal system can be approximated to a spatial linkage system and solved as for any 'rigid body dynamic linkage' problem. From knowledge of the kinematics of the linkages, or limbs, the forces and moments present around the joints of the linkages may be obtained. In the human body these forces and moments are manifested as muscle forces and joint reactions. The dynamic inputs to the system are represented by D'Alemberts inertial forces and the foot/ground reaction forces. A Newtonian analysis, of summing forces horizontally and vertically and moments about points of contact permits these muscle forces to be estimated.

Before any analysis can be performed, biomechanical modelling of the musculo-skeletal system must be carried out for the human body to represent a spatial linkage system. This modelling takes the form of the assumptions and observations outlined below.

3.2 Biomechanical Modelling of the human body - General Assumptions and Simplifications

(1) Running is primarily a two-dimensional activity with ground reaction forces, net joint moments and motion in the third plane generally small. Nigg (1986) found vertical ground reaction forces in the region of 2000 N anterior-posterior reaction forces of ±200 N but medial-lateral (M-L) horizontal forces of only ±100 N. Thus these M-L forces are only in the order of 5% of the vertical component and are in agreement with studies by Cavanagh & Lafortune (1980).

Joint moments about the anterior-posterior axis, that is those causing internal and external rotation, are similarly small. For example Harrison and Nicol (1988) found knee joint moment of 200 Nm in the sagital plane but only 20 Nm in the frontal plane. Hence these 'rotational' moments are only 10% of the 'flexing' moments. These results are confirmed by Crowninshield et al. (1978) and Miller (1987).

Motion in the third small is also small. Rotation of the hip in presenting the leg to the ground for foot strike causes the action of pronation/supination at the ankle. Both of these actions are in general restricted to 5-10 degrees compared to 30° knee flexion and 40° hip flexion/extension (Mann & Hagy, 1980; Mann, 1982; Cavanagh, 1980).

To record these third dimension joint motions and segmental kinematics required two orthogonally placed optical recording devices. These need to be accurately calibrated and synchronised. This is a lengthy and complex process. (Greive et al., 1975; Dainty and Normal, 1987; Williams and Lissner, 1962). The collection time of the data, via digitising, is doubled and hence a lengthy data analysis process is further lengthened into an unwieldy and cumbersome technique. This does not lend itself to a sport analysis technique.

The use of a three dimensional model has also been criticised by Engberg (1987), Burdett (1982) and Denoth (1985) as being unnecessarily complex and costly in the nature of data capture and not yielding significantly more accurate results than a corresponding two dimensional model. With these considerations in mind a number of dynamic studies have utilised two dimensional models (Smith, 1975; Bishop, 1977; Ariel, 1974; Dahlkvist et al., 1982; Ellis et al., 1979,1985; Moeinzadeh et al., 1983) Hence for this study a two-dimensional model is employed.

(2) The muscles can be considered to act together in their major groups according to their main function (Paul, 1967). See Table 3.1.1 and Figure 3.1. They are also considered to act synchronously and initially at all times during the stance phase according to their primary functions (University of California, 1953). This technique has been used by numerous investigators in

TABLE 3.1.1 Muscle Grouping and Function[1]

Muscle Group Notation	Nomenclature Used in this Study	Contributory Muscle(s)	Muscle Group Primary Function
-	AT[2] . .	. . Achilles Tendon	Foot Plantar-Flexor
Gluteals	GL	Gluteal Maximus	Hip Extensor
Hamstrings	HM	Biceps Femoris Semimembranosus Semitendinosus	Hip Extensors and Knee Flexors
Ilio-Psoas	IP	Iliacus Psoas Major	Hip Flexors
-	PL[2] .	. . Patellar Ligament	Shank Extensor
Quadriceps	QU { VA RF. . .	Vasti Medialis Vasti Intermedialis Vasti Lateralis . . Rectus Femoris	Knee Extensors Hip Flexor and Knee Extensor
Shin	SH	Extensor Hallucis Longus Extensor Digitorum Longus Tibialis Anterior	Foot Dorsi-Flexors
Tricep Surea	GA. . . SO. . .	. . Gastrocnemius . . Soleus	Knee Flexor and Foot Plantar-Flexor Foot Plantar-Flexor

(1) Reference; Gray's Anatomy (1980) 36th edition

(2) Included for completeness

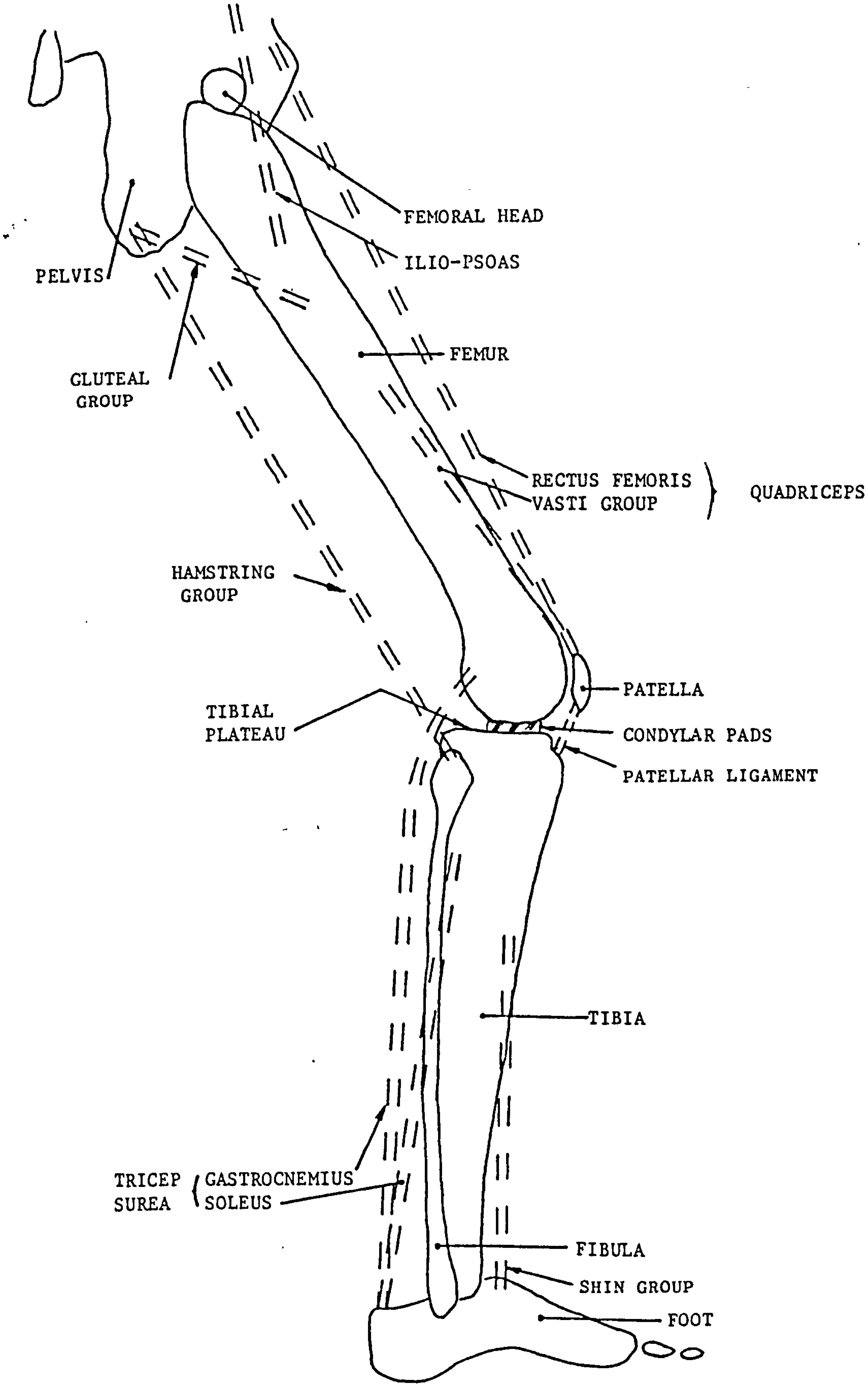


FIGURE 3.1 MAJOR ELEMENTS OF BIOMECHANICAL SYSTEM

biomechanics and has become standard modelling (Morrison, 1968; Williams & Svensson, 1968; Morrison, 1969; Seedhom & Terayama 1976; Dahlkvist et al., 1982 are some examples). The use of the muscles acting individually results in gross mathematical redundancy of the model and as such needs optimisation techniques to solve the system. As outlined in Chapter 2 this is not practical in sports biomechanics.

(3) The limbs of the lower extremities may be replaced by geometrically simple segments of known mass, centre of gravity and moment of inertia as presented by Whitsett (1963) and Hanavan (1964). These data have been updated by a number of investigators (Drillis & Contini, 1966; Contini, 1972; Chandler et al., 1975; and Reynolds et al., 1975) but the original works are still extensively used as they provide compatibility with other studies. As such a number of studies have used one or other of the above basic studies, for example Miller (1973) studied diving, Huston and Passerello (1971) analysed kicking and Smith (1975) investigated a drop jump.

(4) The joints of these lower limbs are represented by non complex articulating surfaces with point load contact. Although this is not at all correct, especially for the hip and knee joints (Smidt, 1973; Dowson & Wright, 1981) it is standard biomechanical practice and is used extensively in all areas of biomechanical modelling (Winter, 1979; Dainty & Norman, 1987; Plagenhoef, 1971; Miller & Nelson, 1973).

(5) The line of action or tension of a muscle or muscle group can be assumed to be one directly between the muscles origin and insertion unless the muscle is severely distorted by bony obstacles. In these cases a tangential direction of action from either the origin or the insertion may be assumed (Seedhom & Terayama, 1976). However the use of a tangential approach requires the collection of more data than a straight line approach and wide fluctuation in the assumed direction of tangency may occur (Dostal & Andrews, 1981). A straight line approach affords greater inter- and intra-investigator repeatability (Jensen & Davy, 1975). A tangential approach can result in a change of muscle moment arms of between 1 & 12%. However, this change in moment arms may be masked by the accuracy of determining and repeating tangential vectors, (Jensen & Davy, 1975). Thus a straight line appraoch has been greatly adopted by biomechaical researchers (Smith, 1975; Rohle et al., 1984; Dahlkvist et al., 1982; Morrison, 1968).

The selection of a single point to represent the insertion point of a number of muscles acting in a group and attaching over an area also causes problems. The angle of pennation, uniformity and cross sectional area of each of the muscles in the group complicates the matter but a simple centre of the cross sectional area of the muscular attachments is assumed to represent the effective centre of the force vector (Seedhom & Terayama, 1976; Jenson & Davy, 1975; Brand et al., 1982).

(6) Frictional forces within the joints may be neglected due to the minimal coefficient of friction existing within synovial joints. Any tangential forces are then assumed to be absorbed by the ligaments within that joint (Seedhom & Terayama, 1976). The coefficient of friction in a synovial joint is in the range 0.001 to 0.1 depending upon the friction mechanism considered (Hydrodynamic, Elastohydrodynamic or Boundary) (Dowson & Wright, 1982) and it has been shown that as load increased, the coefficient of friction decreases (Tabor, 1982; and O'Kelly et al., 1978 - Via Ellis et al., 1980) and as such, at the high impact loads sustained by the body in sport (see Table 2.5.2) the coefficient of friction may be less than those figures stated above. The neglection of friction forces is hence accepted by a wide range of biomechanical investigators (Morrison, 1967; Perry et al., 1975; via Ellis et al., 1979; Seedhom & Terayama, 1976; Ellis et al., 1979).

On the basis of these assumptions a dynamic biomechanical model can be produced as shown in Figure 3.2. Here D'Alemberts inertia forces are omitted for clarity only. If D'Alemberts forces are also included a Newtonian analysis can be performed. This enables a mathematical evaluation based upon the inversed dynamics principle to the model to be produced and gives rise to nine equations from the three major segments i.e. for each segment:

$$\Sigma \text{ Forces vertically} = 0$$

$$\Sigma \text{ Forces horizontally} = 0$$

$$\Sigma \text{ Joint Moments} = 0$$

(D'Alemberts inertia forces omitted for clarity only)

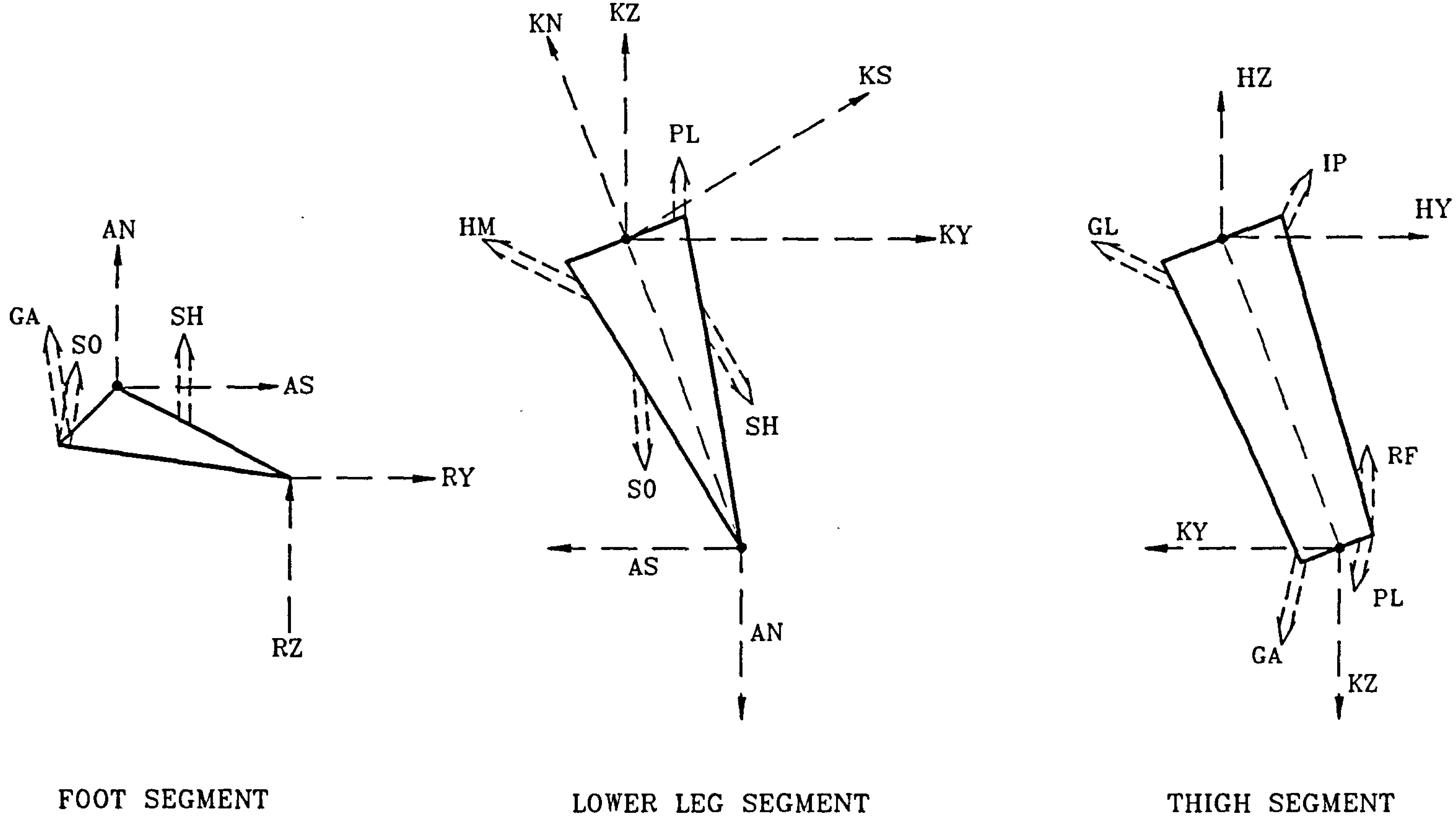


FIGURE 3.2 Primary Leg Model

Examinations of this biomechanical model shows that there are six joint reactions (vertical and horizontal components at each of the three joints) and eight muscle forces and one ligament force (PL), resulting in this biomechanical model being statistically indeterminant by a redundancy factor of six. Therefore further, or specific, assumptions need to be made to simplify the model.

3.3 Specific Biomechanical Modelling with Respect to Running

(1) The 'short' flexors of the hip - the ilio-psoas muscles may be ignored from the model due to their unimportance for the following reasons:-

The origins of the psoas muscles lie on the 12th thoracic vertebra and the 1st, 4th and 5th lumbar vertebra. The muscle crosses the hip joint very close to the capsular of the acetabulum and is only separated from the capsule by an iliac bursa. It attaches onto the femur at and below the lesser trochanter i.e. inside the thigh. The muscle not only flexes the hip but also rotates it.

The origin of the iliacus is the medial portion of the pelvis and it obtains fibres from the capsule of the hip, as it passes over the joint. It attaches below the lesser trochanter on the femur. Its action is to assist the psoas major and is primarily a postural muscle. (Gray's Anatomy, 1980).

Combined, to form the ilio-psoas muscles, the group have a small moment arm and non significant role in athletic events, and are only active during the swing or recovery phase in gait.

No activity was detected during the support phase by Le Ban et al., 1965 in an EMG study into the muscles role in the gait cycle. These muscles are rarely injured in running sports. Hence they can be eliminated from the biomechanical model due to their lack of contribution to hip or thigh flexion when compared to the stronger flexor muscles such as the quadriceps.

(2) Maximum knee flexion angles during the stance phase in running have been expressed as between 27° (Sinning & Forsyth, 1970) and 30° (Elliott & Blanksby, 1979a). For such angles of knee flexion Ellis et al. (1980) found that the ratio between tensions along the quadriceps tendon and the patella tendon was on average 0.97 for 30°, 0.99 for 20° and 1.00 for angles less than 15°. For this study it was assumed, therefore, that the tension developed in the quadriceps muscle group is fully transmitted to the patellar ligament.

Initially the patella is assumed to be segmentally a part of the thigh, such that the three vasti muscles (see Table 3.1.1) become internal forces, Seedhom & Terayama, 1976).

(3) Due to the similarity in the physiological cross sectional areas of the rectus femoris and the three vasti muscles (see Table 3.1.1), each individual muscle can be assumed to equally share the total force developed in the quadriceps, that being one quarter each. (Alexander &

Vernon, 1975; Seedhom & Terayama, 1976).

(4) The hip extensor moment can be assumed to be equally shared between the hamstrings and the gluteals, Seedhom & Terayama (1976).

(5) During the stance phase the foot possesses almost zero angular and linear accelerations and it also has a minimal mass value compared to the other masses in the biomechanical model. The inertia forces of the foot can hence be ignored (Harrison, 1982).

(6) Net muscular torques or joint moments can be approximated well through cinematographic techniques but the accuracy of joint reactions, being dependant upon input parameters of muscle moment arms, accelerations and anthropomorphic data, decrease with the increasing number of segments included in the analysis (Dainty & Norman, 1987; Smith, 1975). The acquisition of radiographic data of the hip area of the human subjects also presents ethical problems. With this in mind hip joint reactions are not calculated in this model, the hip joint centre is only used as a centre of rotation to take moments from. This thus reduces the number of equations available for evaluation to seven and the number of unknowns still exceeds this figure.

3.4 Model Development

The previous specific biomechanical modelling reduced the number of unknowns to nine but the corresponding number of equations available to produce solutions has been reduced to seven. Additional modelling was therefore undertaken to facilitate the production of results. A continuous process of model refinement was then undertaken as outlined in the following sections.

3.4.1 First Biomechanical Model

In line with models produced and presented by Seedhom & Terayama (1976) and Smith (1975) the redundancy of the model was accommodated by ignoring the role of the shin muscle group and the soleus muscle. This was initially justified in the following way.

(1) The shin muscles (see Table 3.1.1) are foremost dorsi-flexors of the foot and during the stance phase in running are inactive for the majority of that period. Their action is primarily involved in preparing the foot for landing (while in the swing phase of the running cycle) and there may only be a residual of force present in the shin muscles at heel strike (Proctor et al., 1982). The shin muscles are also responsible for inverting the foot (commonly termed pronation) and this action is held to be the cause of the majority of shin injuries (James et al., 1978). It could then be hypothesised that the shin muscles are active only in the third dimension during the stance phase of running.

(2) The tricep surea muscle group of the soleus and gastrocnemius can be assumed to share the moment of force produced at the ankle either equally (50% each) or non equally depending upon such factors as physiological cross sectional area, mathematical optimisation or (electromyographical) activity. For this part of the

study the soleus could be considered inactive in accordance with the study by Seedhom & Terayama (1976).

(3) Simple linear mathematical optimisation techniques were employed in order to accommodate the shin muscle group and the soleus muscle into the model. Primarily this consisted of sharing the ankle joint moment between the shin and calf muscle and then sharing the tricep surea plantar flexion force between the soleus and gastrocnemius muscles. Weighted values were attributed firstly to shin and then to the soleus and a computer program iterated through all possible combinations. This can be shown as follows,

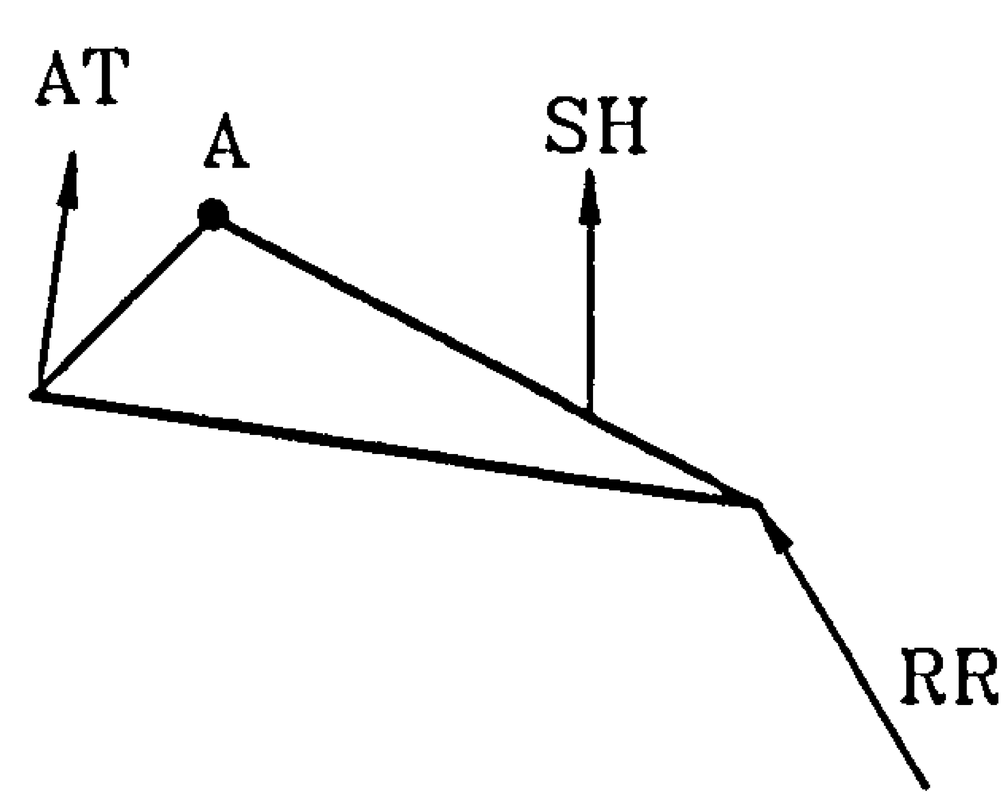


Fig. 3.3 Ankle Joint Biomechanics

RR = resultant ground reaction force

AT = achilles tendon or tricep surea muscle force -

GA = gastrocnemius; SO = soleus

SH = shin group muscle force

A = ankle joint centre of rotation

AM = ankle moment caused by RR

X_1, X_2 = moment arms of AT & SH

The ankle moment AM can be balanced by either AT or SH or any combination of the two. Hence let,

$AT.X_1 = y.AM$ & $SH.X_2 = (1-y)\ AM$ where $0 > y > 1$

Similarly the achilles tendon force AT can be shared by the gastrocnemius (GA) or the soleus (SO) in any combination such that,

$GA = z.AT$ & $SO = (1-z).AT$ where $0 > z > 1$

Hence by iterating through y & z and solving the full biomechanical model an appropriate solution was obtained based upon one of the following objective functions,

(a) minimum total muscle force (viz. minimum energy expenditure)

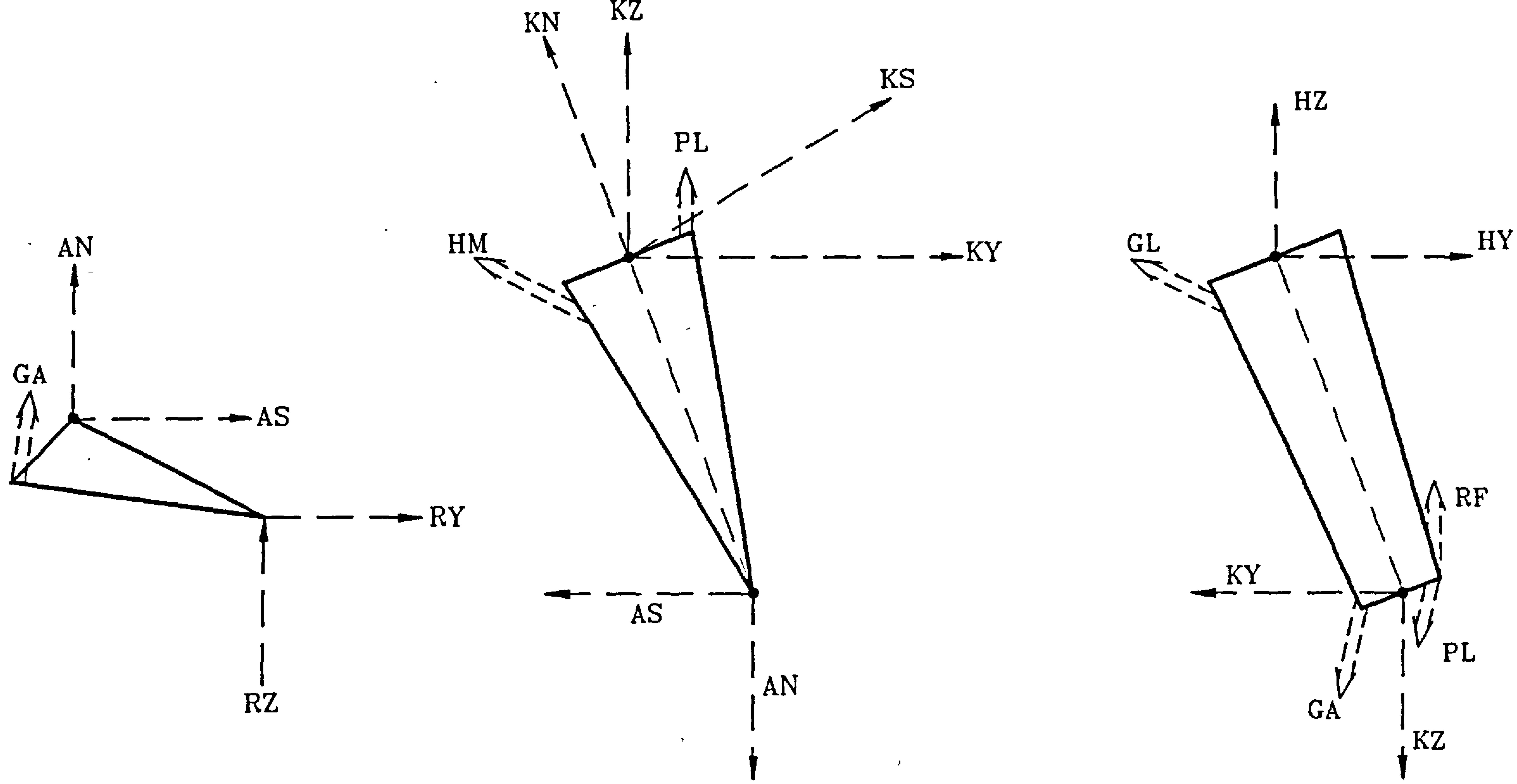
(b) minimum total joint force (viz. joint damage reduction)

(c) a combination of (a) & (b) (viz. efficiency of movement).

All proved inconclusive however and the only feasible solutions obtained were with the shin and soleus inactive as all the objective functions increased in value as value was apportioned to the two muscles in question and the hamstring forces turned negative.

(4) An Engineering analysis based on Virtual Work techniques was also investigated but this relied upon treating the foot as a deformable structure which did not conform with

(D'Alemberts inertia forces omitted for clarity only)



FOOT SEGMENT LOWER LEG SEGMENT THIGH SEGMENT

FIGURE 3.4 FIRST BIOMECHANICAL MODEL

the rigid body dynamics approach adopted here and thus had to be discarded.

Hence a simplistic approach was adopted with the shin and soleus omitted which permitted a primary analysis to be performed. Thus a dynamic biomechanical model was produced as presented in figure 3.4 (with D'Alemberts inertia forces omitted for clarity only) and was used to produce results presented in the study by Harrison (1982) and summarised by Harrison et al. (1982).

3.4.2 Second Biomechanical Model

The result produced by the model presented in 3.4.1 allowed refinement to take place and this consisted in the first instance of the inclusion of the tibialis anterior muscle to represent the shin muscle group. A study by Burdett (1982) showed that the dorsiflexion/inversion group of muscles were active for the first 10-20% of the support phase whilst running at 4.47 ms^{-1} and Proctor & Paul (1982) report similar findings for their anterior tibial group (average 25% activity during walking). This would appear to be verified by EMG studies carried out by Mero et al. (1987). This plantar or dorsi flexion moment is best represented by the moment created by the ground reaction forces giving either a negative or positive sense (see figure 3.3). Thus a positive moment is countered by the tricep surea muscles and a negative by the shin group (Harrison et al., 1986).

Hence with the shin muscle group represented in the model (figure 3.5) a secondary set of results could be calculated. This analysis

(D'Alemberts inertia forces omitted for clarity only)

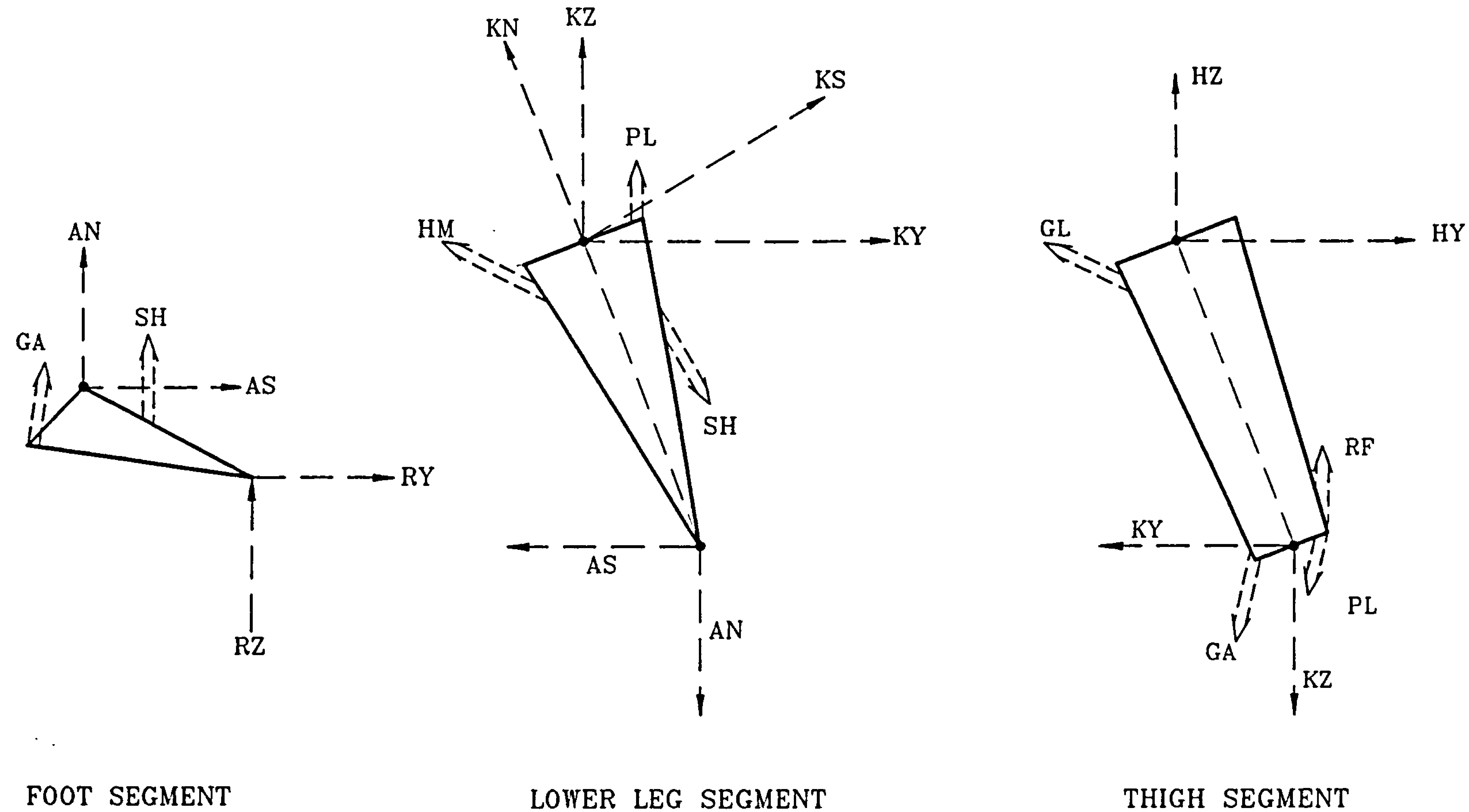


FIGURE 3.5 SECOND BIOMECHANICAL MODEL

was used by Harrison et al. (1986) and Harrison et al. (1988) and afforded good comparisons with the previous study and that of Burdett (1982).

3.4.3 Third Biomechanical Model

An inactive soleus has been favoured by numerous investigators Burdett (1982), Seedhom & Terayama (1976), Morrison (1968 & 1969) Paul (1967) and Smith (1975), but electromyographic studies show that it is active during the stance phase in running (Brandell, 1973). Hence the inclusion of the soleus into the biomechanical model was addressed in various ways following the unsuccessful optimisation trials reported in 3.4.1. Physiological cross sectional area (PCSA) measurement was not available to this investigation and thus force sharing based upon this criteria was eliminated. Sharing of the plantar flexion moment equally, as used with the hip moment sharing between the hamstrings and gluteals, was thus adopted. Results are presented with and without the soleus in section 5.4 and conclusions about the effectiveness are presented in section 6.6.

Initially, and in common with studies by Seedhom and Terayama (1976), Smith (1975) and Nissel & Mizrahi (1988) the patella was considered to be integral to the thigh. This permitted the evaluation of the model to be performed with the equations produced and thus comparison between phases and models is possible. If the patella is considered a separate segment, the rectus femoris muscle no longer acts upon the thigh segment. The vasti group of muscles, initial classified as totally internal to the thigh, would hence have to be included in the model. Thus direct comparisons between

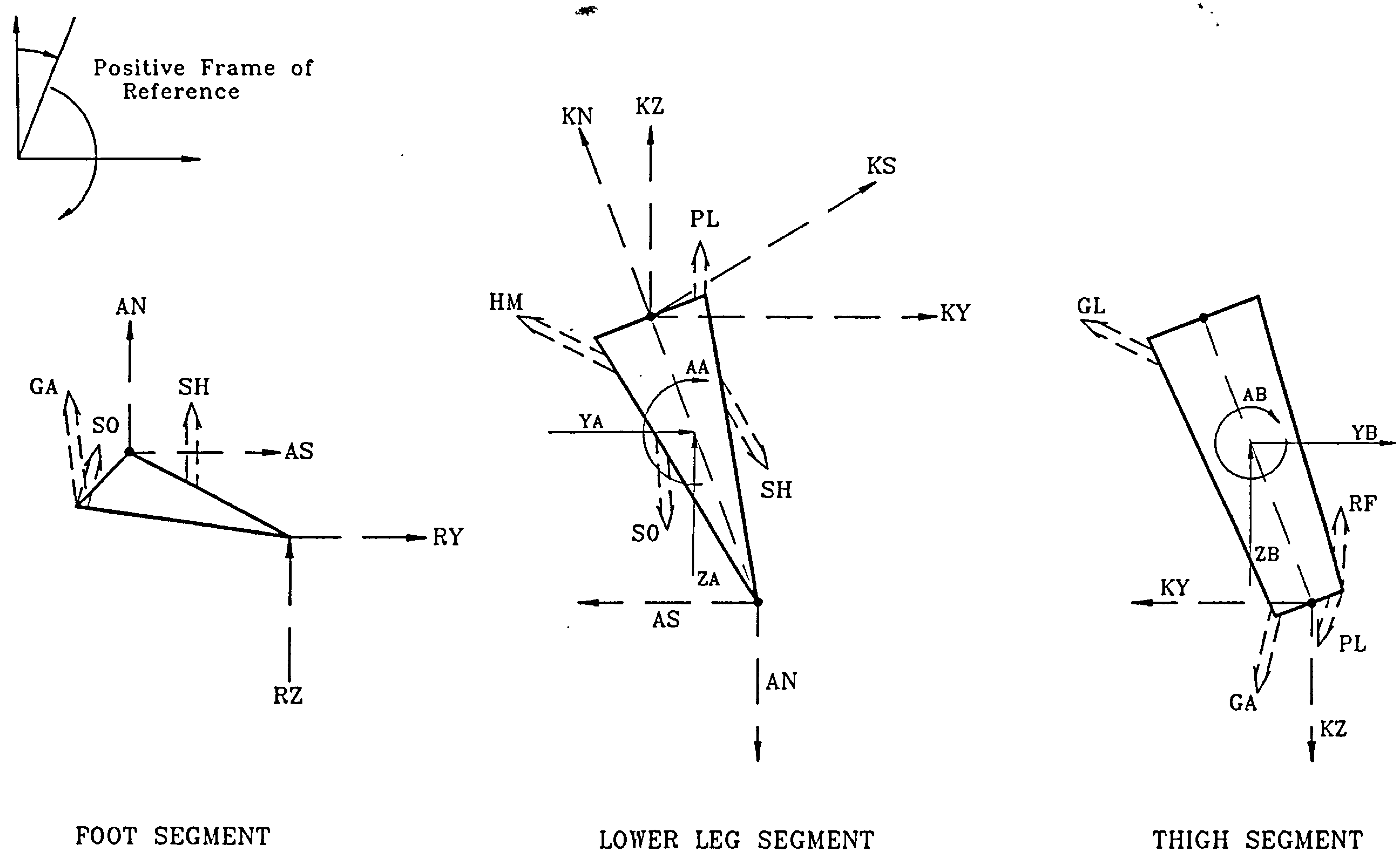


FIGURE 3.6 THIRD BIOMECHANICAL MODEL

the models developed here and those of, for example, Smith (1975), Seedhom and Terayama (1976) and Nissel & Mizrahi (1988) would not be possible.

Thus this model is shown in figure 3.6 and the calculation of the patella-femoral contact force is performed as shown in figure 3.7.

3.5 Verification of Model

Verification of the above model, muscle grouping and roles during the stance phase in running was carried out over a period of time on a number of subjects (as recommended by McIntyre & Robertson, 1987) using EMG techniques whilst undergoing treadmill running. A San-ei four channel polygraph was utilised with surface electrodes placed over various muscles. Co-ordination of the EMG traces with the beginning of stance phase was achieved with the aid of a contact switch built into the running shoes of the subject (Plate 3.1). For a range of speeds from 3.8 ms^{-1} to 5.4 ms^{-1} all the major muscle groups outlined for this study indicated activity during the stance phase. The shin muscles, represented by the tibialis anterior, displayed activity for approximately the first 5% of the stance phase; the tricep surea group then exhibited the dominant muscular activity. This compares favourably with the 9% reported by Burdett (1982).

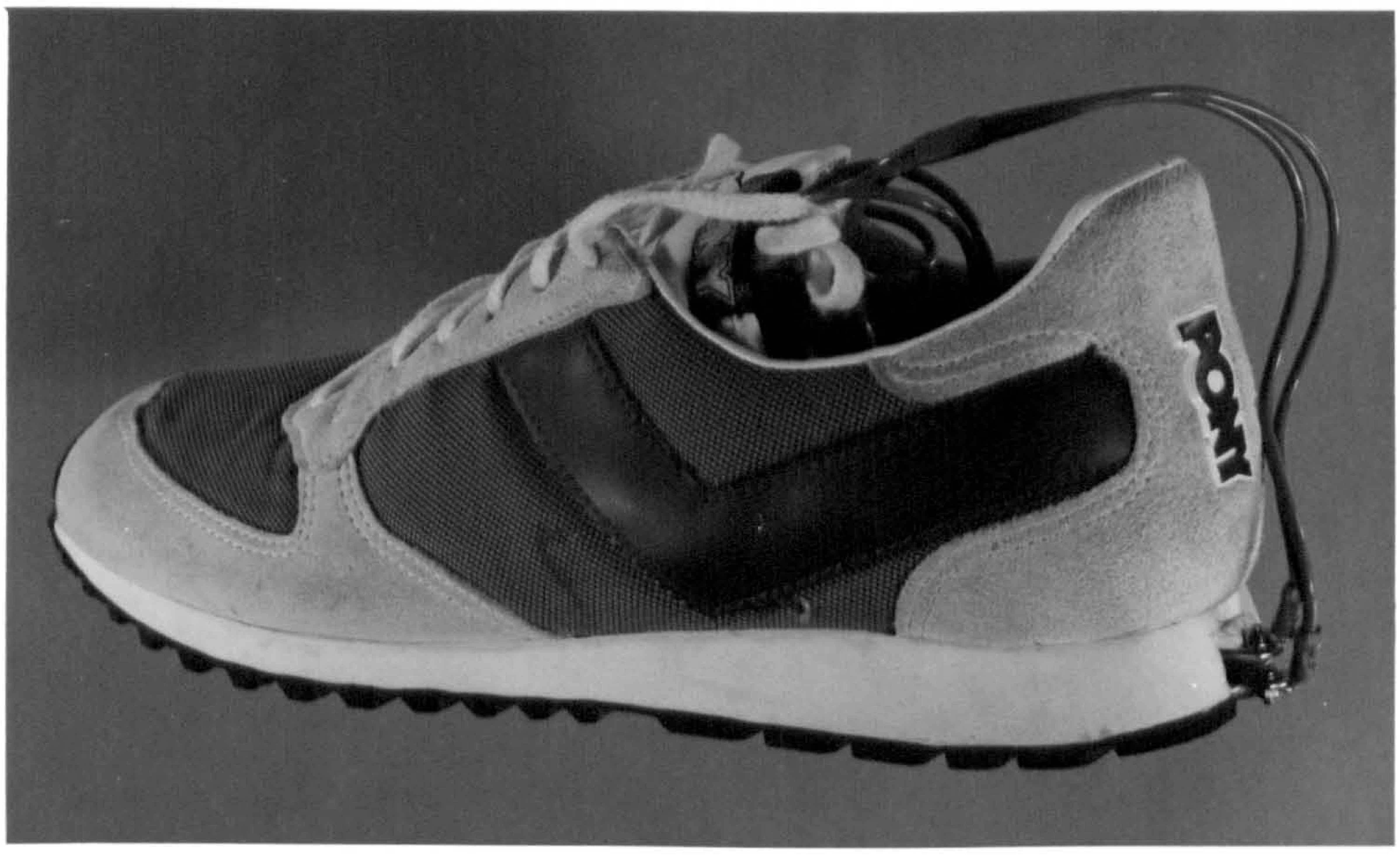
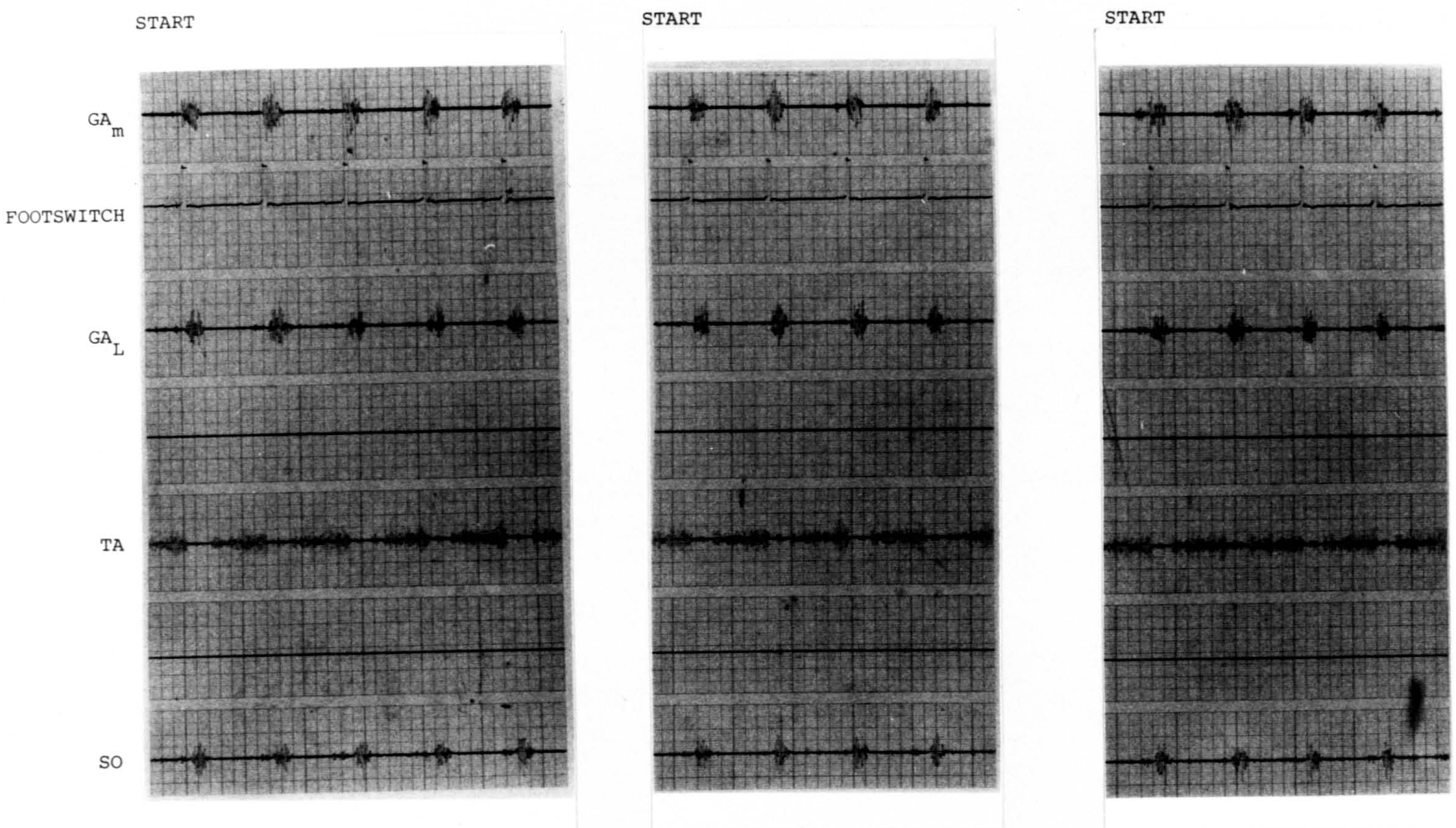


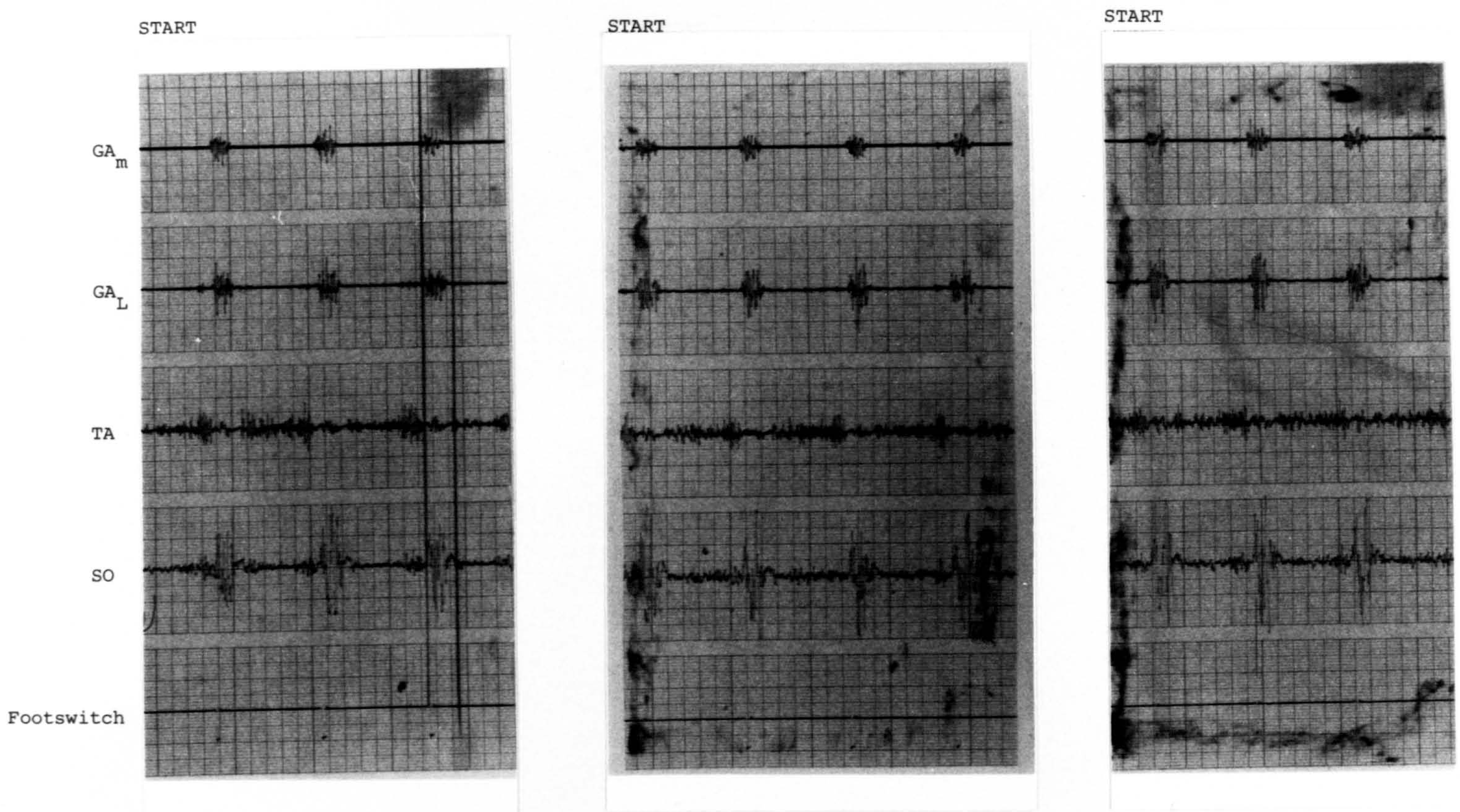
Plate 3.1 Running Shoe Adapted to Include a Contact Switch.

From Plates 3.2 it can be seen that of the major muscle groups represented in the bioengineering model the soleus and gastrocnemius are almost fully active. In phase two the gastrocnemius displays only 80% activity but in phase three it is fully active. However, both GA and SO do not display activity at heel strike. The tibialis anterior, representing the anterior shin muscles displays activity during the recovery section of the swing phase and during the first 5% of the stance phase following heel or foot strike. Of the upper limb muscles the hamstrings, as represented by bicep femoris, display full activity for the two faster speeds and 60% activity at



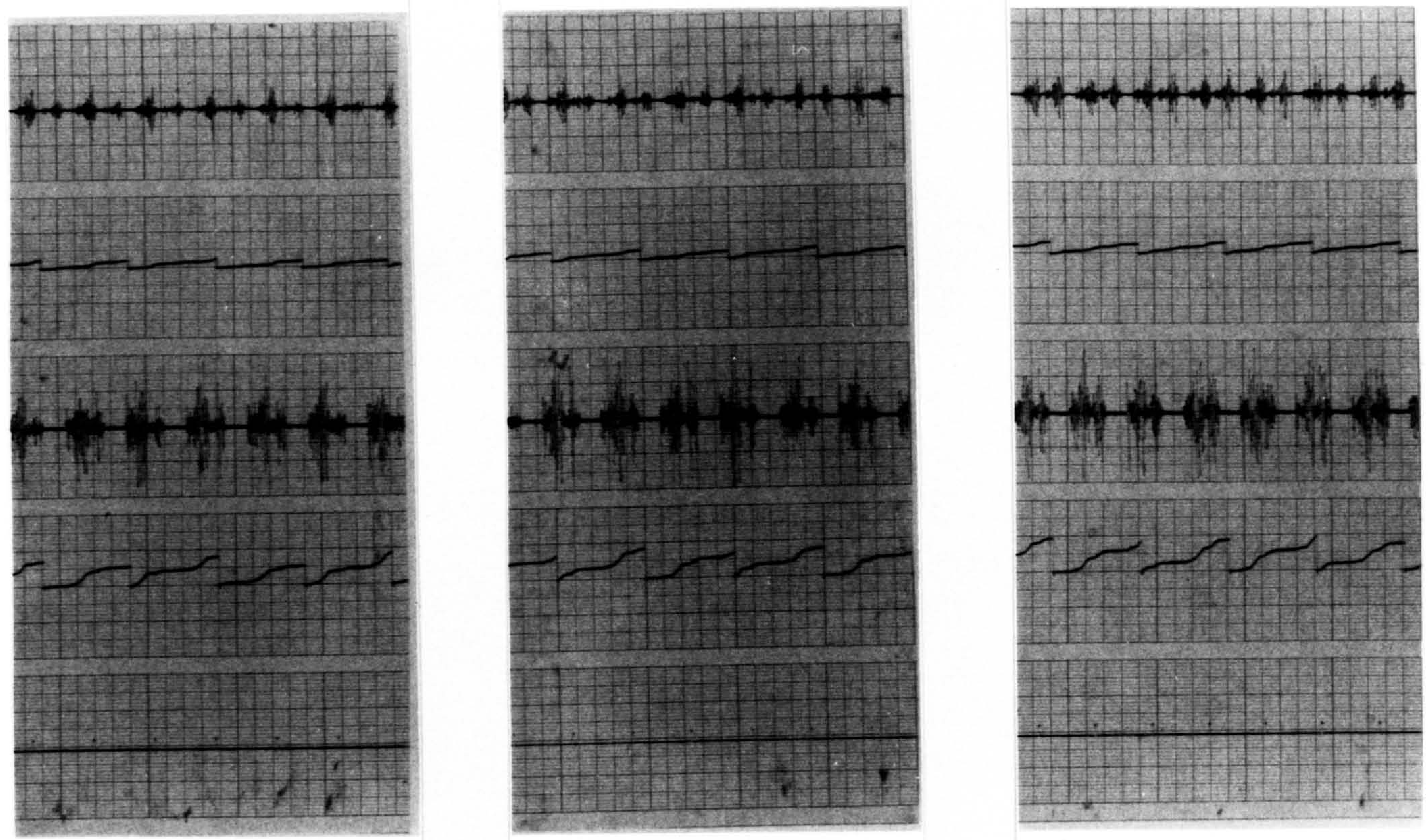
(a) 3.83 m s^{-1} (b) 4.47 m s^{-1} (c) 5.36 m s^{-1}

Plate 3.2.1. EMG Results (GA_m and GA_L refer to medial and lateral heads of gatrocnemius. Footswitch indicates foot strike)



(a) 3.83 $m\ s^{-1}$ (b) 4.47 $m\ s^{-1}$ (c) 5.36 $m\ s^{-1}$

Plate 3.2.2. EMG Results (GA_m and GA_L refer to medial and lateral heads of gastrocnemius. Footswitch indicates foot strike)



(a) 3.83 m s^{-1} (b) 4.47 m s^{-1} (c) 5.36 m s^{-1}

Plate 3.2.3. EMG Results (Footswitch indicates foot strike)

the slowest speed. This could partly explain why hamstrings are not frequently injured by long or middle distance athletes but suffer muscle pulls at higher velocities by sprinters. The quadricep muscle group, as represented by the rectus femoris, display activity, on average, 50%-60% of the stance phase following heel strike. It could be suggested that this was "built in" to the biomechanical model i.e. assigning zero force to the muscle group once 50%-60% of the stance phase had elapsed. The phasic activity of the vasti muscles is such that they may develop force throughout the stance phase (University of California, 1953). It was thus concluded that full activity of the quadricep muscle group was justified in the model.

The model, thus produced in section 3.4.3 can therefore be solved as for any rigid body dynamics problem with D'Alembert's inertia forces incorporated.

Analysis of Biomechanical Model

Foot segment (see Fig. 3.6).

Taking moments about the centre of rotation of the ankle joint,

$$\Sigma \text{ External moments} = \Sigma \text{ Internal moments} \tag{1}$$

and

Moments of ground reaction forces (RZ and RY) +
Moments of muscular balancing force (AT or SH) +
Moments of foot mass = Moments of foot inertia (2)

(see Table 3.1.2 for nomenclature).

Now, if the moment of ground reaction force is positive then the moment of muscular balancing force is provided by the shin muscle (SH). If the moment of ground reaction force is negative then the moment of muscular balancing force is provided by tricep surea muscles (AT). Foot mass and inertia are small compared to the ground reaction forces, RZ and RY, and hence can be ignored. Therefore, either

$$-RY(A1)+RZ(A2)+AT(A3)=0 \text{ and } SH=0 \tag{3}$$

or

$$-RY(A1)+RZ(A2)+SH(A4)=0 \text{ and } AT=0 \tag{4}$$

Now, summing forces vertically gives

$$\Sigma \text{ External forces vertically} = \Sigma \text{ Internal forces vertically} \tag{5}$$

$$AT[\cos(BO)]+SH[\cos(B4)]+AN+RZ-WF=ZO(MF) \tag{6}$$

where WF (foot weight) and ZO(MF)(foot inertia) are small and can be ignored. (Harrison, 1982).

Hence,

$$AN= -RZ-AT[\cos(BO)]-SH[\cos(B4)] \tag{7}$$

Similarly, summing forces horizontally gives

$$AS= -RY-AT[\sin(BO)]-SH[\sin(B4)] \tag{8}$$

Assuming equal sharing of the tricep surea force (see main text)

$$GA = 0.5\ AT \tag{9}$$

$$SO = 0.5\ AT \tag{10}$$

<u>Shank segment</u> (see Fig. 3.6)

Taking moments about the knee centre of contact,

$$\Sigma \text{ External moments} = \Sigma \text{ Internal moments} \tag{11}$$

$$PT(K1)+HM(K2)-WL(K3)+AS(K4)-AN(K5)-SH(K8)-SO(K7)=BF \tag{12}$$

where BF = shank inertial moment, and PT = 4RF (see main text).

Hence,

$$4RF(K1)+HM(K2)-WL(K3)+AS(K4)-AN(K5)-SH(K8)-SO(K7)=BF \tag{13}$$

Summing forces vertically gives,

$$\Sigma \text{ External forces vertically} = \Sigma \text{ Internal forces vertically} \tag{14}$$

PT[cos(B1)]+HM[cos(B2)]-SH[cos(B4)]-SO[cos(BO)]+KZ-AN-WL

= ML(ZA) (15)

where ML(ZA) = shank inertial forces and PT = 4RF.

Hence,

4RF[cos(B1)]+HM[cos(B2)]-SH[cos(B4)]-SO[cos(BO)]+KZ-AN-WL

= ML(ZA) (16)

Summing horizontal forces gives,

Σ External horizontal forces=Σ Internal horizontal

forces (17)

PT[sin(B1)]+HM[sin(B4)]-SH[sin(B4)]+SO[sin(BO)]+KY-AS

=ML(YA) (18)

where ML(YA) = shank intertial forces and PT = 4RF.

Hence,

4RF[sin(B1)]+HM[sin(B2)]-SH[sin(B4)]+SO[sin(BO)]+KY-AS

=ML(YA) (19)

<u>Thigh segment</u> (see Fig. 3.6)

Taking moments about hip centre of rotation,

Σ External moments = Σ Internal moments (20)

RF(H1)+GL(H2)-PT(H3)-GA(H4)-KZ(H5)+KY(H6)-WT(H7) = TF (21)

where TF = thigh inertial moment, PT = 4RF, and GL(H2) = HM(H9), i.e. hip extension equally shared between hamstring and gluteal muscles (see main text).

Hence,

RF(H1-4H3)+HM(H9)-GA(H4)-KZ(H5)+KY(H6)-WT(H7)=TF (22)

Therefore, by examining Equations 13,16,19 & 22 it can be seen that four unknowns exist, i.e. RF,HM,KZ and KY. These can be solved, which allows solutions for PT and GL to be derived.

Also, resolving along the femoral axis gives,

$$KN=KZ[\cos(B3)]+KY[\sin(B3)] \tag{23}$$

$$KS=KY[\cos(B3)]-KZ[\sin(B3)] \tag{24}$$

Patella Segment (see Fig. 3.7)

Resolving the force system at the patella gives,

$$PF=PL\sqrt{2[1-\cos(B1-B9)]} \tag{25}$$

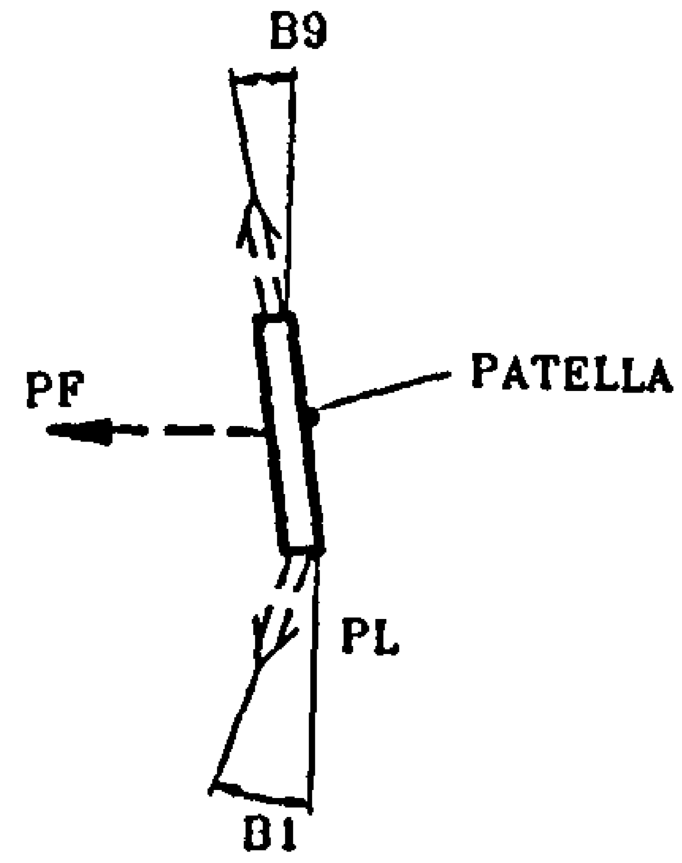


FIGURE 3.7 PATELLA-FEMORAL FORCE

TABLE 3.1.2 Nomenclature

Abbreviation	Item
RZ	Vertical foot - ground reaction force
RY	Horizontal foot - ground reaction force
AN	Vertical ankle-joint reaction
AS	Horizontal ankle-joint reaction
KZ	Vertical knee-joint reaction
KY	Horizontal knee-joint reaction
KN	Normal knee-joint reaction (parallel to lower leg centre line)
KS	Shear knee-joint reaction (perpendicular to lower leg centre line)
HZ	Vertical hip-joint reaction
HY	Horizontal hip-joint reaction
PF	Patella-femoral contact force
AT	Achilles tendon action
HM	Hamstring muscle action
GA	Gastrocnemius muscle action
GL	Gluteal muscle action
PT	Patellar ligament action
RF	Rectus femoris muscle action
SH	Shin muscle action
SO	Soleus muscle action
QU	Quadriceps muscle action
ZO	Vertical acceleration of foot
YO	Horizontal acceleration of foot
AO	Angular acceleration of foot
ZA	Vertical acceleration of lower leg

YA	Horizontal acceleration of lower leg
AA	Angular acceleration of lower leg
ZB	Vertical acceleration of thigh
YB	Horizontal acceleration of thigh
AB	Angular acceleration of thigh
WL,ML	Lower leg weight (N) and mass (kg)
WT,WL	Thigh weight (N) and mass (kg)
A1,A2...A4	Moment arms about foot centre of contact
K1,K2...K8	Moment arms about knee centre of contact
H1,H2...H9	Moment arms about hip centre of rotation
BO	Angle made by AT measured from vertical
B1	Angle made by PT, measured from vertical
B2	Angle made by HM, measured from vertical
B3	Angle made by lower leg, measured from vertical
B4	Angle made by SH, measured from vertical
B9	Angle made by thigh, measured from vertical

3.7 Discussion and Conclusions

A biomechanical model has been produced which allows calculation of five joint reaction components and seven muscle forces using a three segment model and an inverse dynamics approach to its solution. The model affords comparison with others produced in the field of bioengineering for example, Paul (1967); Williams & Svensson (1968); Seedhom & Terayama (1976); Burdett (1982); and Proctor & Paul (1982).

The roles of the muscles have been verified by EMG techniques and mathematical optimisation has been investigated, albeit unsuccessfully, in order to try to enhance the model.

Whilst other techniques exist for the investigation of the human locomotor system, the process of simplifying the body to produce a biomechanical model presents the sports bioengineer with a most flexible analysis method.

4. EXPERIMENTAL METHODS

4.1 Installation, Calibration and Modification of Equipment

A previous study by Harrison (1982) has outlined the basic data collection, transfer and analysis system. This used a Bolex 16 mm camera to film an athlete running across a Kistler force platform. Six components of force could then be recorded on an ultra-violet galvanometer. Hand measurements from the ultra-violet trace and the developed film were taken and then manually fed into a computer program on a microcomputer. The final output gave muscle forces and joint reactions corresponding to each frame of developed film for the foot in contact with the force platform (see figure 4.1).

The limitations of the above equipment configuration are outlined below.

1) Filming speed of the Bolex camera is a maximum of 64 frames per second. This needed to be increased as for the running speed of 4.47 $m.s^{-1}$ only, on average, nine frames of developed film showed the foot in contact with the platform.

2) The analogue signal trace output from the ultra-violet unit was not accurate enough due to the format of the actual trace (line thickness of 1 mm). Also the mode of operation of the electronic unit used to convert the electrical charges yielded by the quartz transducers within the force platform into proportional voltages and then into force, moment and displacement values was inadequate. This final conversion of voltages into moment and displacement data is obtained via the electronic unit's analog divider. It was specifically this unit that was unreliable because it only became operational

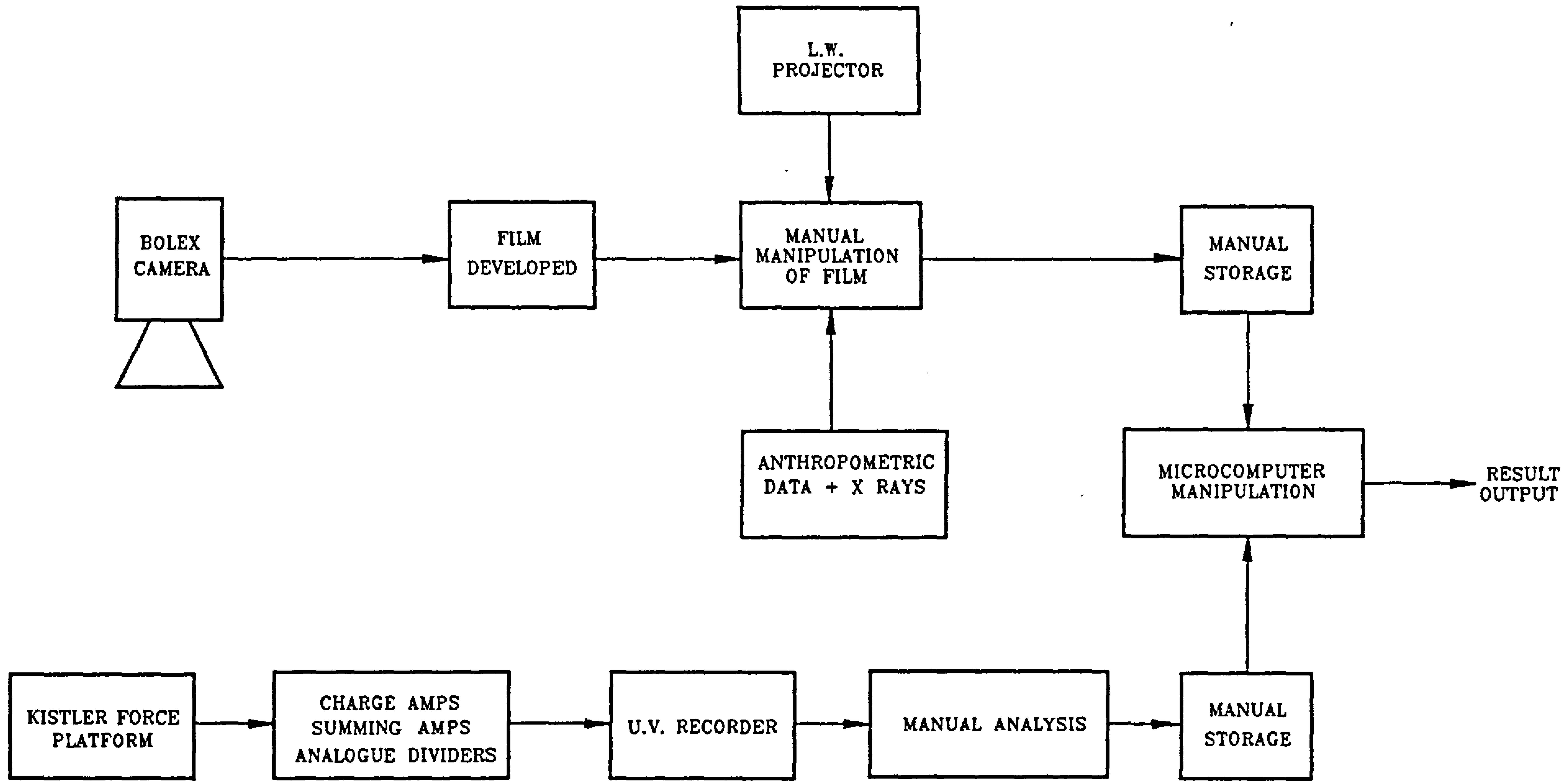
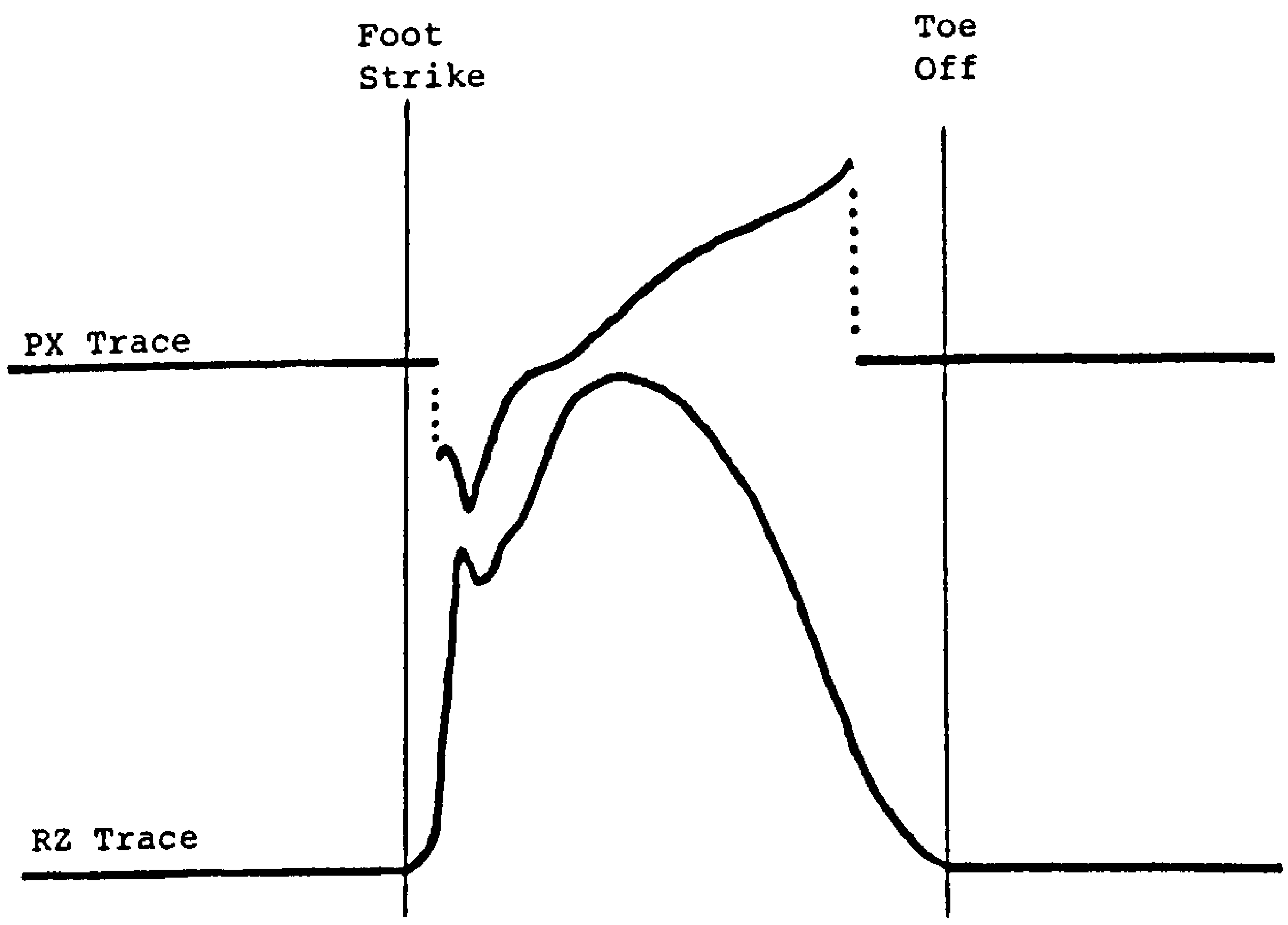


Figure 4.1 Primary Equipment.

above a threshold value of 10% of the sensitivity setting of the charge amplifiers. Thus the first and last portion of the "Centre of Pressure" data calculated by the analogue divider were given as zero, as illustrated below, which was unacceptable in establishing specifically the position of the force vector on foot strike.



UV Galvanometer Output

PX = Centre of Pressure in X Direction

RZ = Vertical Foot/Ground Force

3) The memory storage space of the microcomputer, Commodor 64 (PET), used for the data analysis was insufficient if a fully computerised data collection, transfer and analysis route was to be instigated.

4) Data collection, transfer and analysis time was generally lengthy, the process laborious and predominantly manual.

5) Synchronisation of the force platform results with film data was difficult due to the relatively few frames of film and knowledge of exact foot contact time on the force platform.

Hence refinement of the experimental equipment was needed, this included use of:-

1) A Locam high speed camera capable of filming at up to 500 frames per second (Hz).

2) A '3D Digital Design and Development' 8-channel, 12 bit, analog-to-digital converter (ADC) unit capable of operating between 1000 Hz for one channel and 125 Hz for all 8 channels.

3) A 'Summographic' digitising tablet interface to the DEC-20 mainframe computer. Operating size being 750 mm x 750 mm, with overhead projection (see Plate 4.1).

4) Commodore 64 PET Microcomputer analysis, bypassing the analogue divider in the control electronics of the force platform to calculate displacement (centre of pressure or point of application of force) data along with the recorded force values via the A.D.C. (see figure 4.2).

5) Microcomputer disc storage of data (stored on two discs each time to prevent data loss with corrupt disc/disc drive interface).

6) Data transfer from microcomputer disc to DEC-20 mainframe computer memory via a mainframe compatible microcomputer terminal and disc drive.

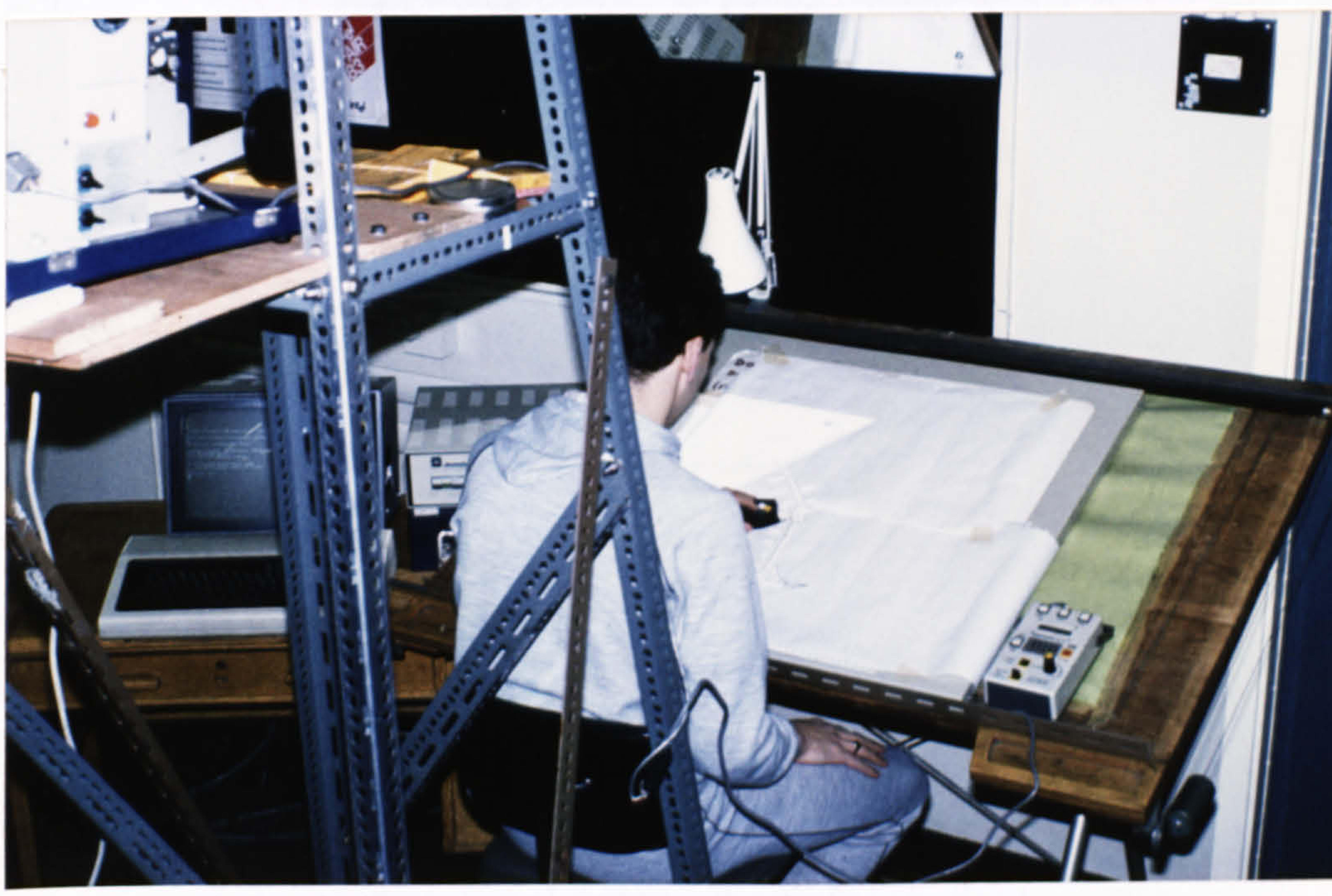


Plate 4.1. Initial Data Analysis Equipment

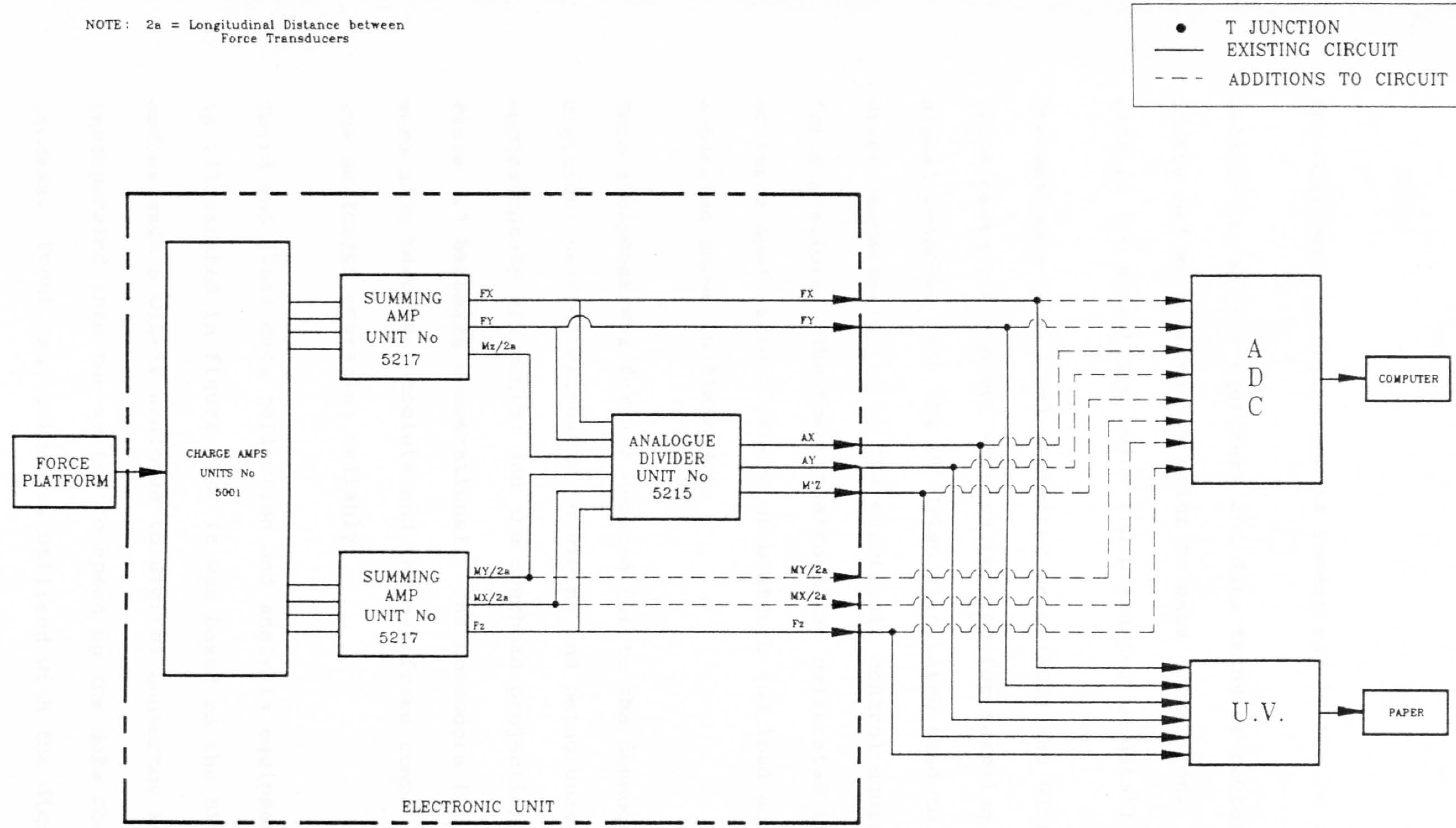


FIGURE 4.2

Modification of Force Platform's Electronic Unit to Facilitate Direct Control of AX & AY, Point of Application

7) DEC-20 mainframe manipulation of the two data sets with a polynomial cubic spline interpolation routine (Conte & Deboor, 1980)

The equipment configuration is summarised in Figure 4.3.

Calibration of the equipment and data transfer routes shown in figure 4.3 were necessary prior to data collection. Calibration details are summarised below and presented in detail in Section 4.3.

The Analogue to Digital Converter Unit (ADC) and Ultra-Violet Galvanometer (UV Galvo) were calibrated for sampling frequency and signal response with the UV Galvo also being used to determine the drift characteristics of the electronic control equipment of the force platform. The force platform was calibrated itself for point of force application, centre of platform and load accuracy using the apparatus shown in Plate 4.2.

This equipment was further updated due to the Summographics digitiser needing extensive servicing and being unavailable for approximately six months and the overhead projection frame shown in Plate 4.1 becoming unoperational. The Commodore Pet microcomputers were also becoming obsolete and the mainframe compatible terminal was no longer operating reliably.

Third and final data collection and analysis equipment configuration is illustrated in figure 4.4. It was based on the BBC microcomputer and as such a CIL-16 analogue to digital convertor was also incorporated into the system to speed up the data collection process. Front projection was utilised with the digitiser as being

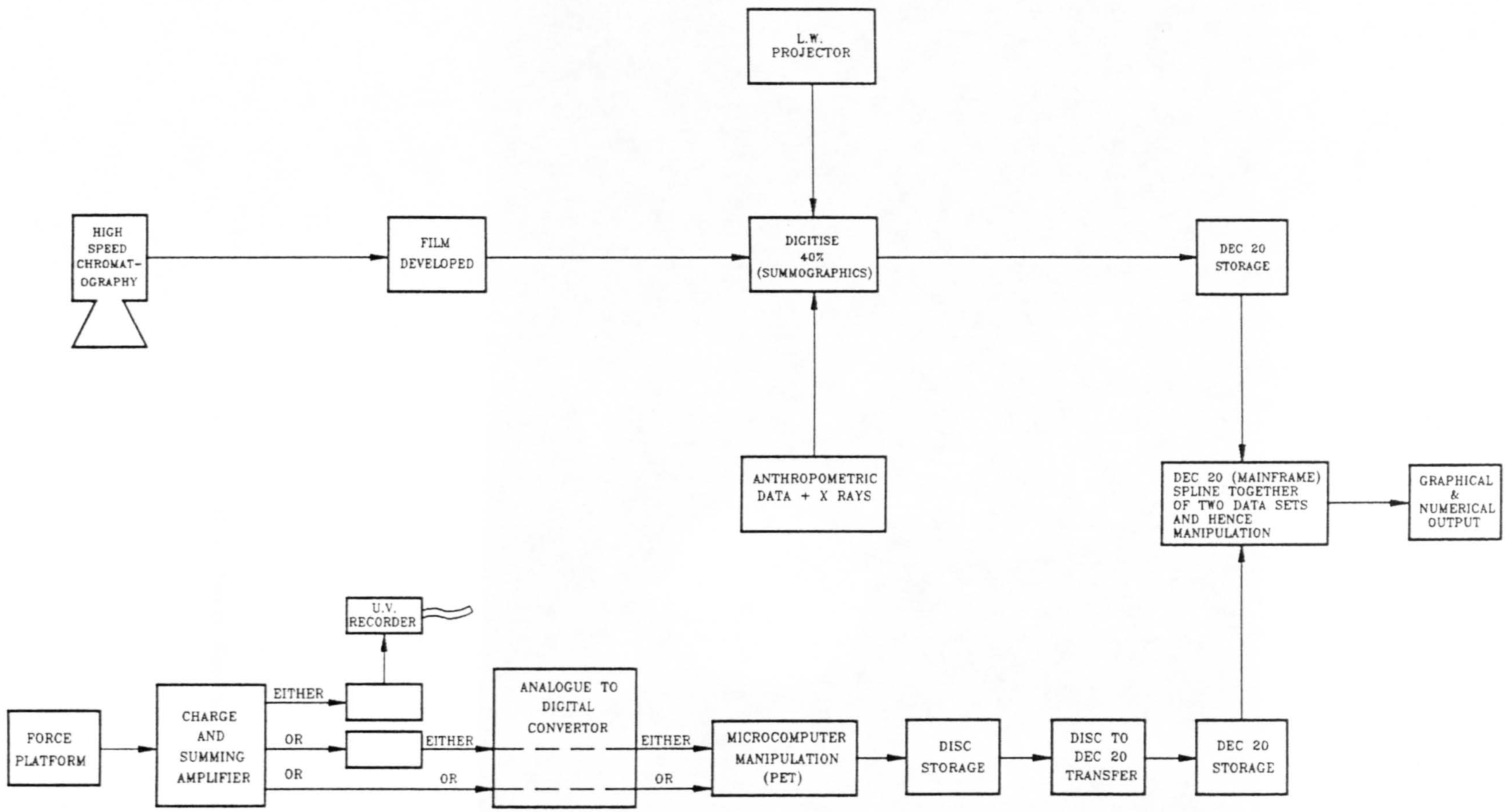


FIGURE 4.3 Data Collection, Transfer & Processing Route

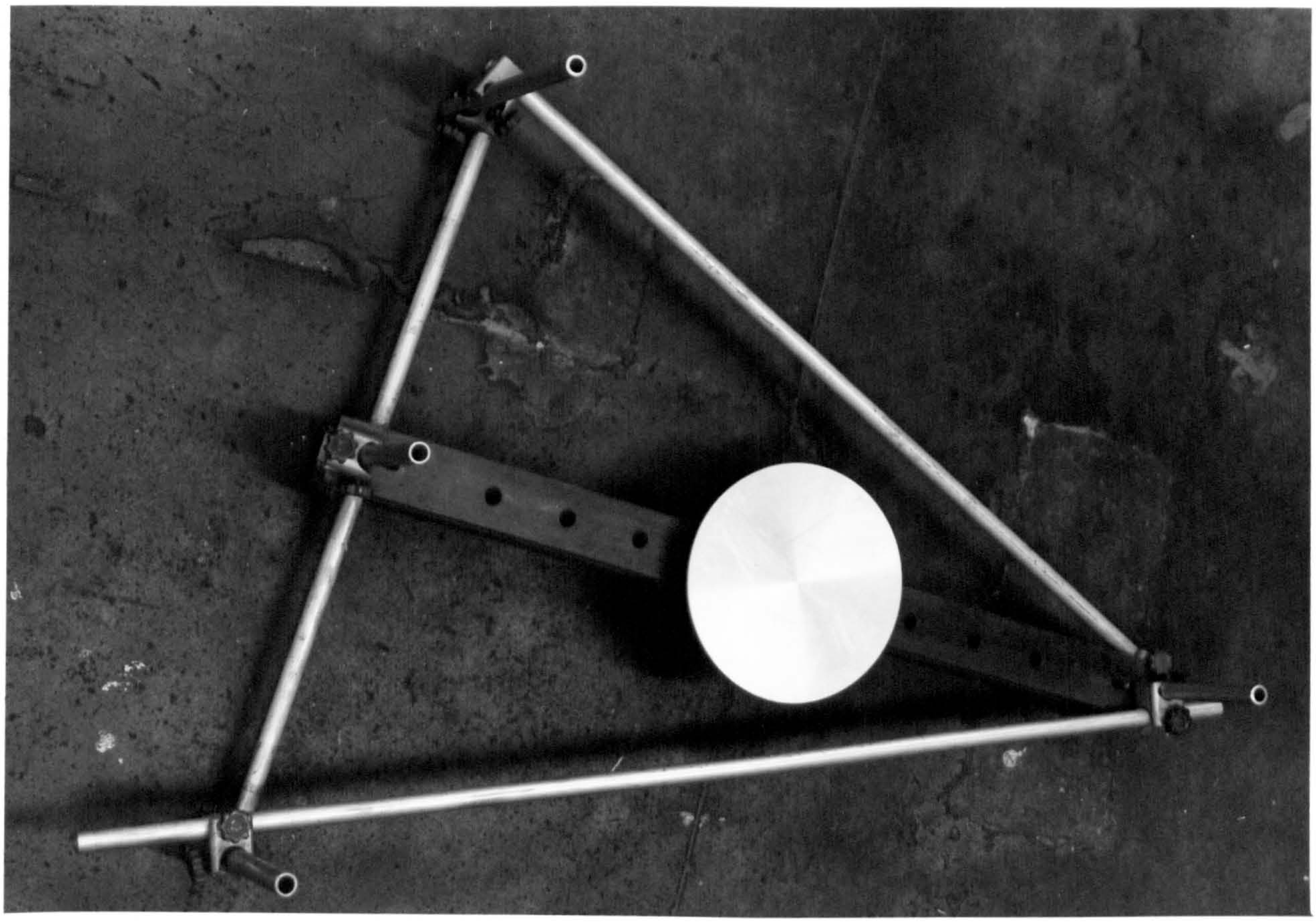


Plate 4.2. Point Loader Used to Calibrate Force Platform

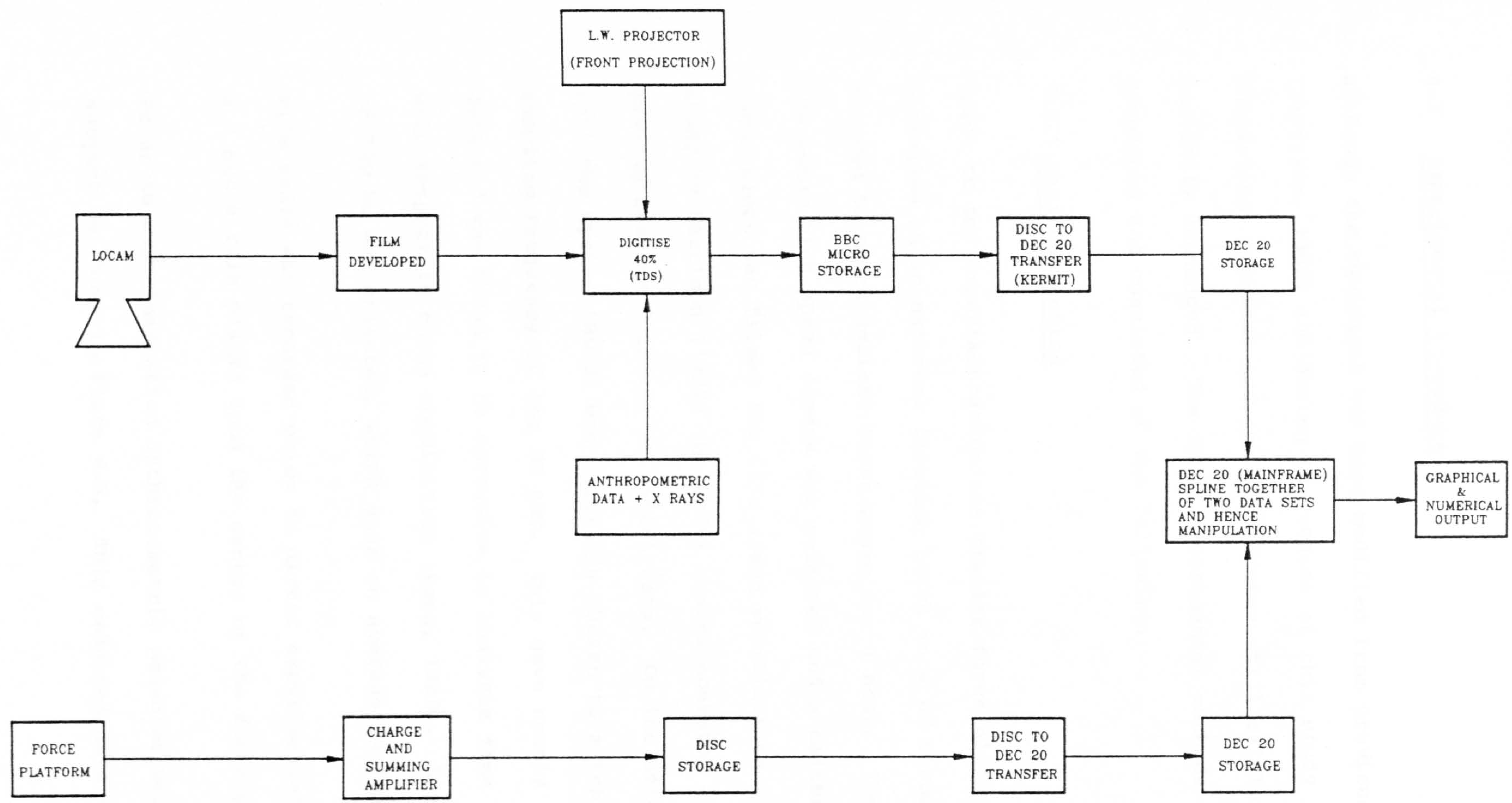


FIGURE 4.4 Final collection, transfer & Processing Route.

the most practicable methods available to the laboratory location of the tablet (see Plate 4.3).

4.2 Experimental Procedure

Although the equipment had been modified from previous studies (Harrison, 1982) and during the course of this study the basic data collection, transfer and analysis routes still existed and remained basically unchanged. The final experimental methods procedure was developed and consisted of the following.

4.2.1 Data Collection

Prior to the experimentation the Kistler force platform was installed in an external location berth in a 60 m roadway and allowed to reach ambient temperature for 1 hour. The film rate of the Locam high speed camera was selected and a calibrated 2 second sweep clock was filmed for film speed verification. In the initial study by Harrison (1982) the Bolex camera operated at 64 Hz. This gave on average 9 frames of useful data. On inclusion of the Locam into the system 180 Hz was initially chosen to match the selected sampling frequency of the 3D ADC. This gave nearly 40 frames of data and was found to be excessive in analysis time and erroneous with respect to noise acquisition (Lees, 1980). Hence a frame rate of 120 Hz was selected, which gave on average 24 frames.

Film scale was recorded prior to direct experimentation by recording a 1 metre rule placed upon the centre of the force platform.

Prior to experimentation anthropometric measures were taken off each subject as shown in Plate 4.5. This enhanced data from individual

Plate 4.3. Final Data Analysis Equipment

radiographs of the subjects lower limbs. Each athlete was then allocated a pair of standard running shoes and a one legged white tracksuit (this highlighted the support leg over the non support leg as they crossed during mid stance) and was weighed in the clothing they would run in. The athlete was then asked to warm up by undertaking as many trial runs as necessary over the platform under filming conditions to achieve the correct speeds and full right foot contact on the force platform (without altering their stride to accomplish this). 2 x 1000 watts flood lights illuminated the test area and background film identification numbers were used to co-ordinate film and force platform records.

Once familiarised, each subject ran along the 60 m roadway and timed themselves over a 10 m span (5 m either side of the force platform), this gave them immediate feedback of their performance. Trails outside a ± 10% velocity tolerance band or with partial or unnatural foot contact, as determined by subject feedback, visual observation or force trace anomalies, were rejected until upto 10 good trials had been obtained. On the subjects return up the roadway, if a successful trial had been noted, they advanced the background numbering system in recognition of success. The data could then be transferred to microcomputer disc storage and at the end of the regime transferred to DEC 20 mainframe storage.

Radiographs of the lower limbs of each subject were obtained to enable production of individual 'bone models' of the legs as shown in Plate 4.4. This was deemed necessary in order to identify joint contact points, muscle lines of action, and origins and insertions for each individual subject and so improve accuracy. The use of

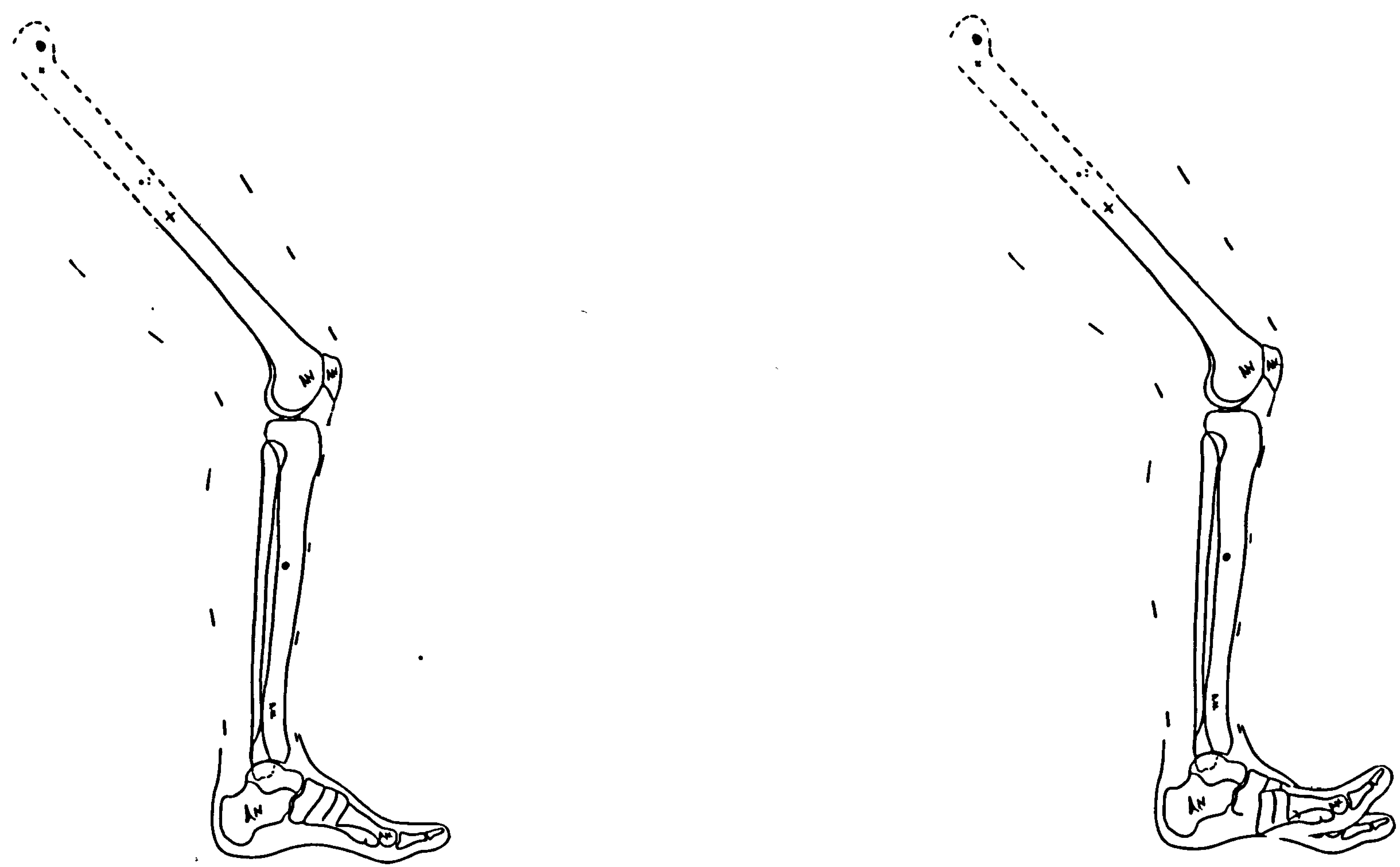
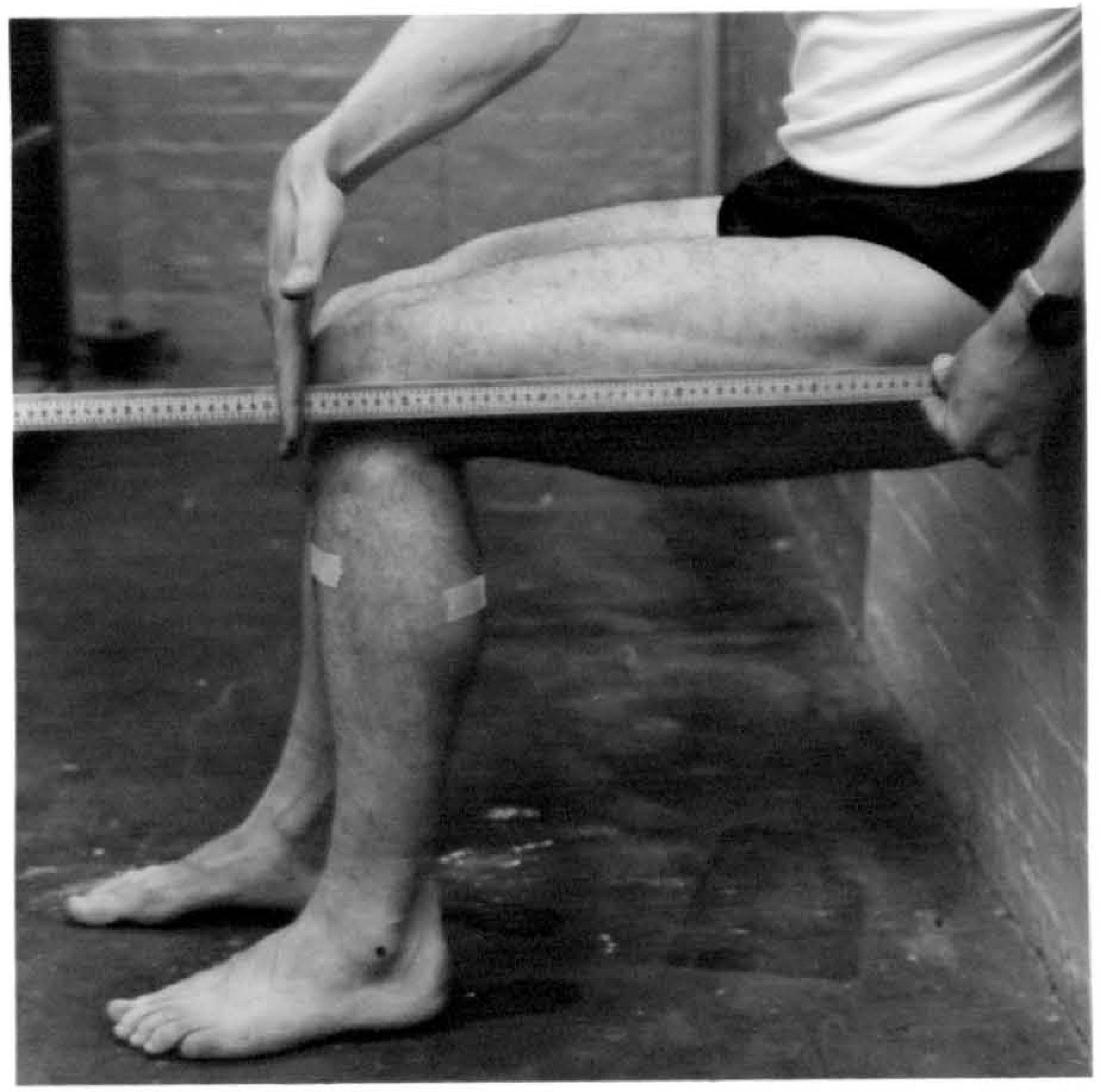


PLATE 4.4   Subject Bone Models Showing Foot in Two Positions



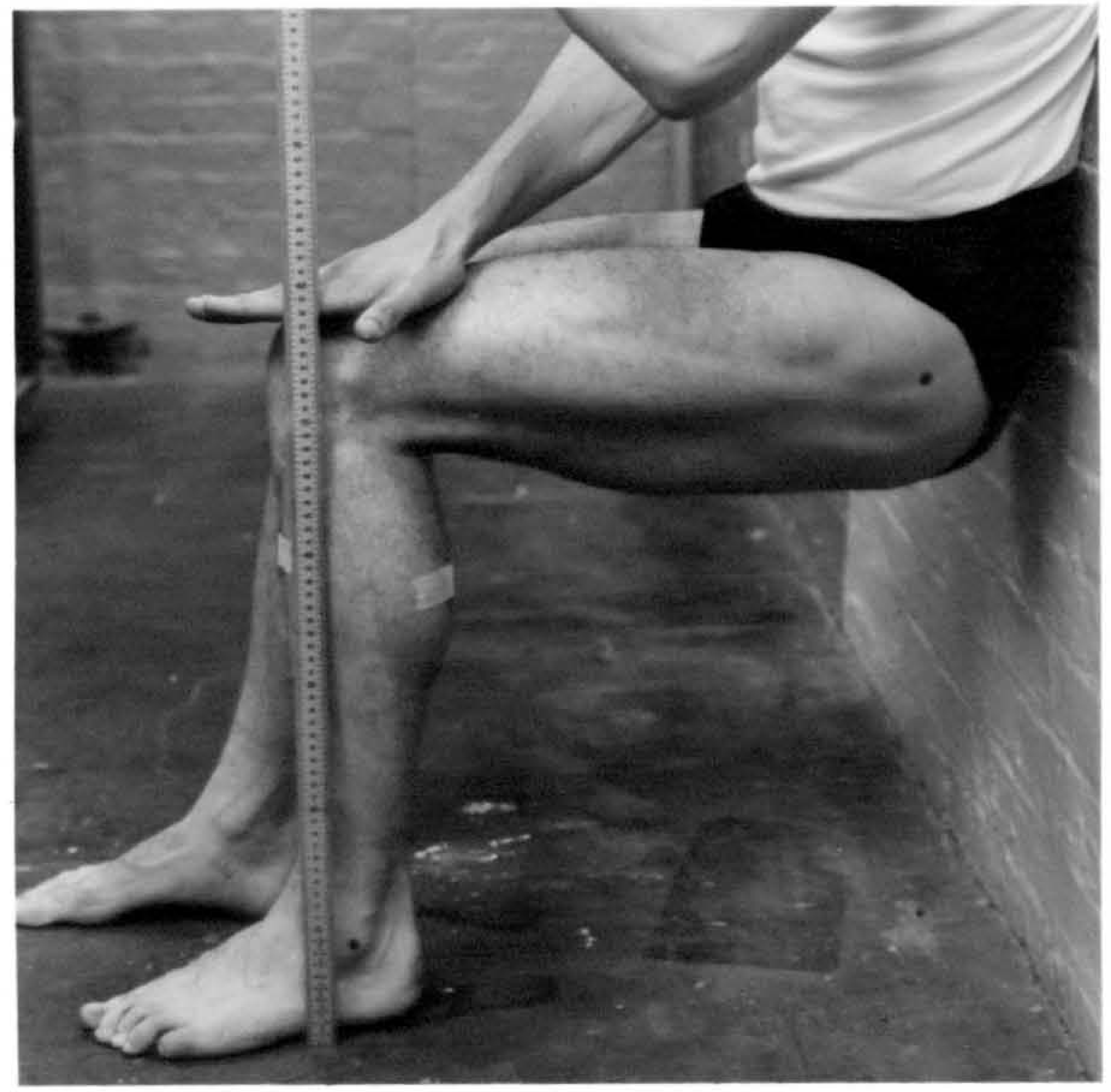


PLATE 4.5. Representative Collection of Anthropometric Data from Subjects Showing Spherically Ended Pins Strapped to Limbs

individual bone models has been shown to improve accuracy by reducing peak forces by 50% to more conservative and acceptable values (Harrison, 1982). This was performed after obtaining informed subject consent from each individual.

The production of these individual bone models was preceded by collection of anthropometric data from the subjects lower limbs. Here foot length, floor to top of femoral condyles and patella to greater trochanter (obtained by palpation) could be recorded (Plate 4.5). This was to enable accurate reconstruction of the bone models to take place. Spherically ended steel pins of known length (50 mm) were strapped vertically on the anterior and posterior aspects of the subject's limbs. This quantified the amount of divergence present with the radiographs and also permitted reconstruction of the developed radiograph films.

Two radiographs of the limbs were then taken in the positions indicated in figure 4.5 and the divergence of the radiographs measured from the developed film using the above pins. This divergence could also be verified from the anthropometric data recorded earlier.

Tracings were made, with a skin line included, from the reconstituted radiographs. The position of the hip was estimated because the complete leg could not be radiographed as this would have meant exposing the subjects genital area. This was achieved by establishing the position of the greater trochanter and adding the correction factor shown in figure 4.6.

The tracings were reduced to either 33% or 40% of real full size

FIGURE 4.5 Experimental Configuration for Establishment of Radiograms of Subject's Limbs

a) Height of Radiographic Lens

b) Positions Adopted

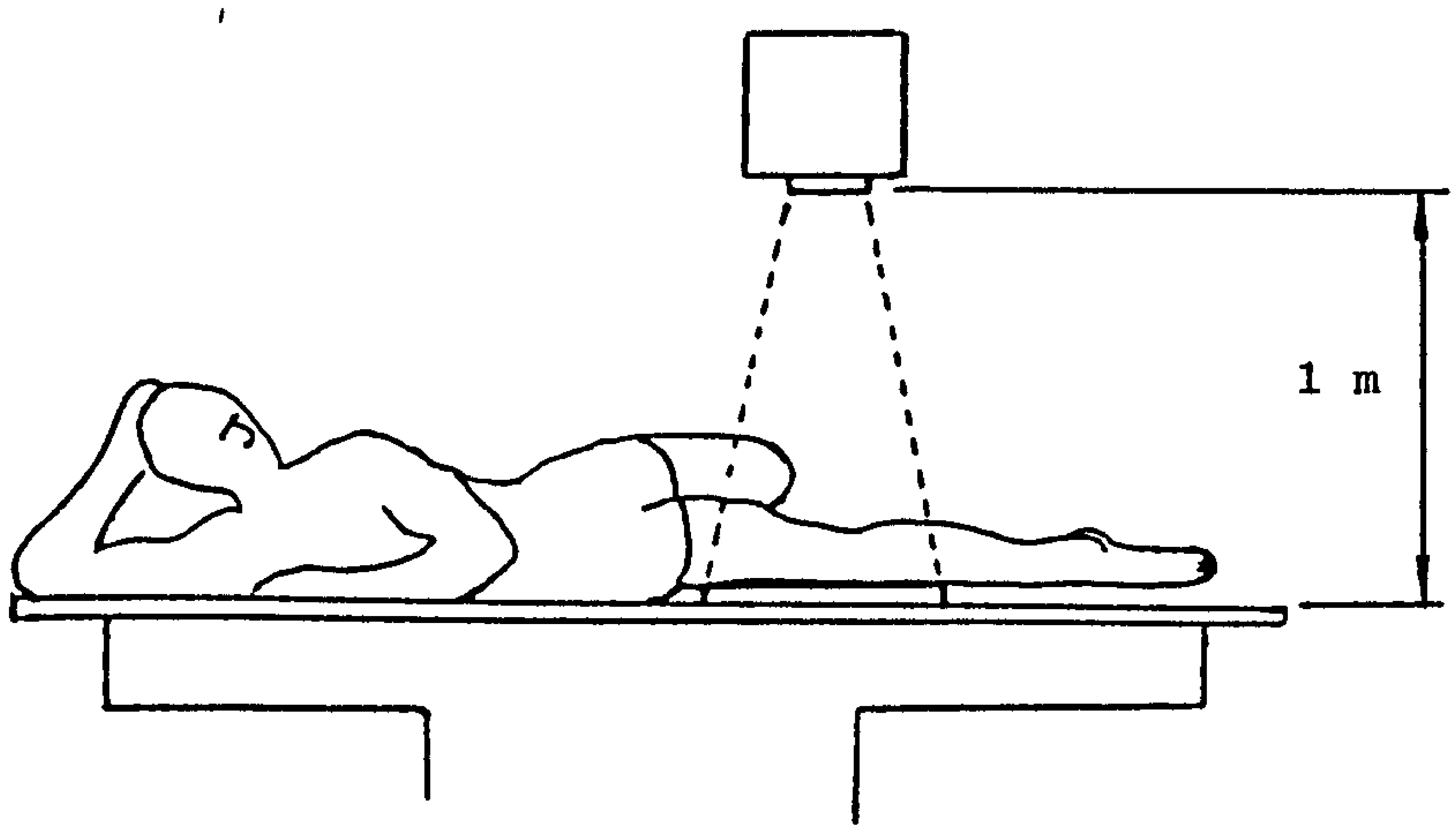


Fig. 4.5a

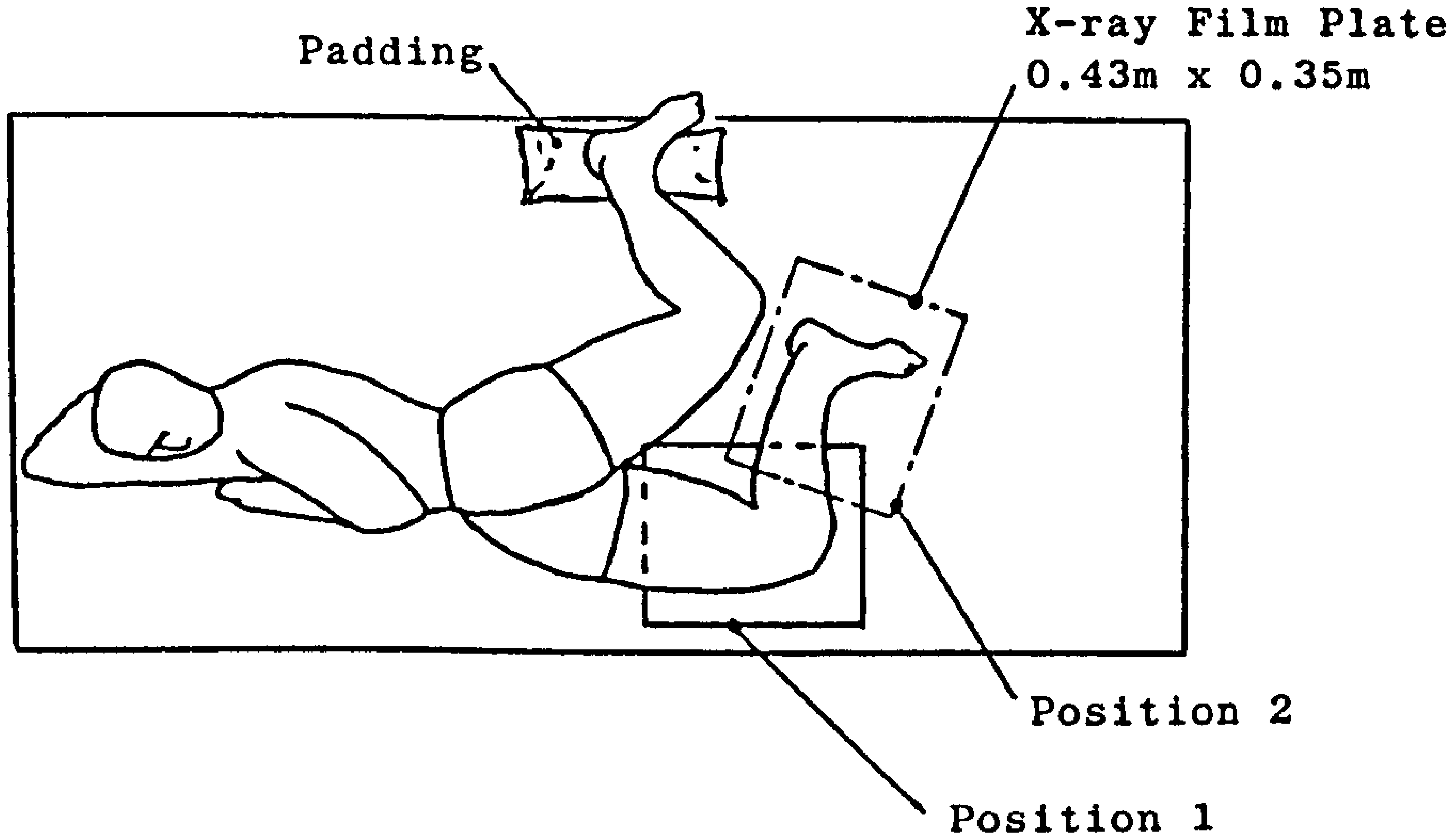
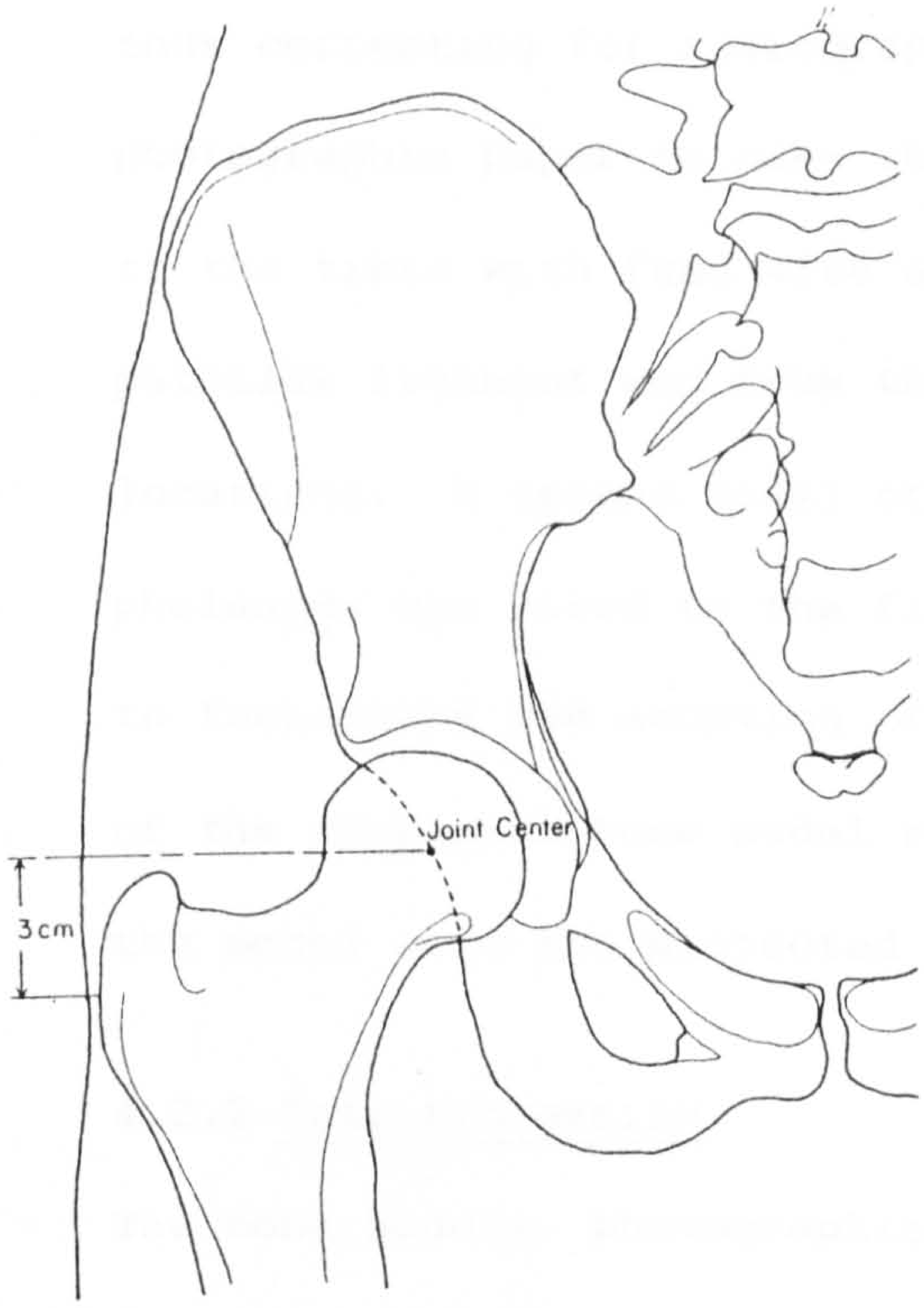
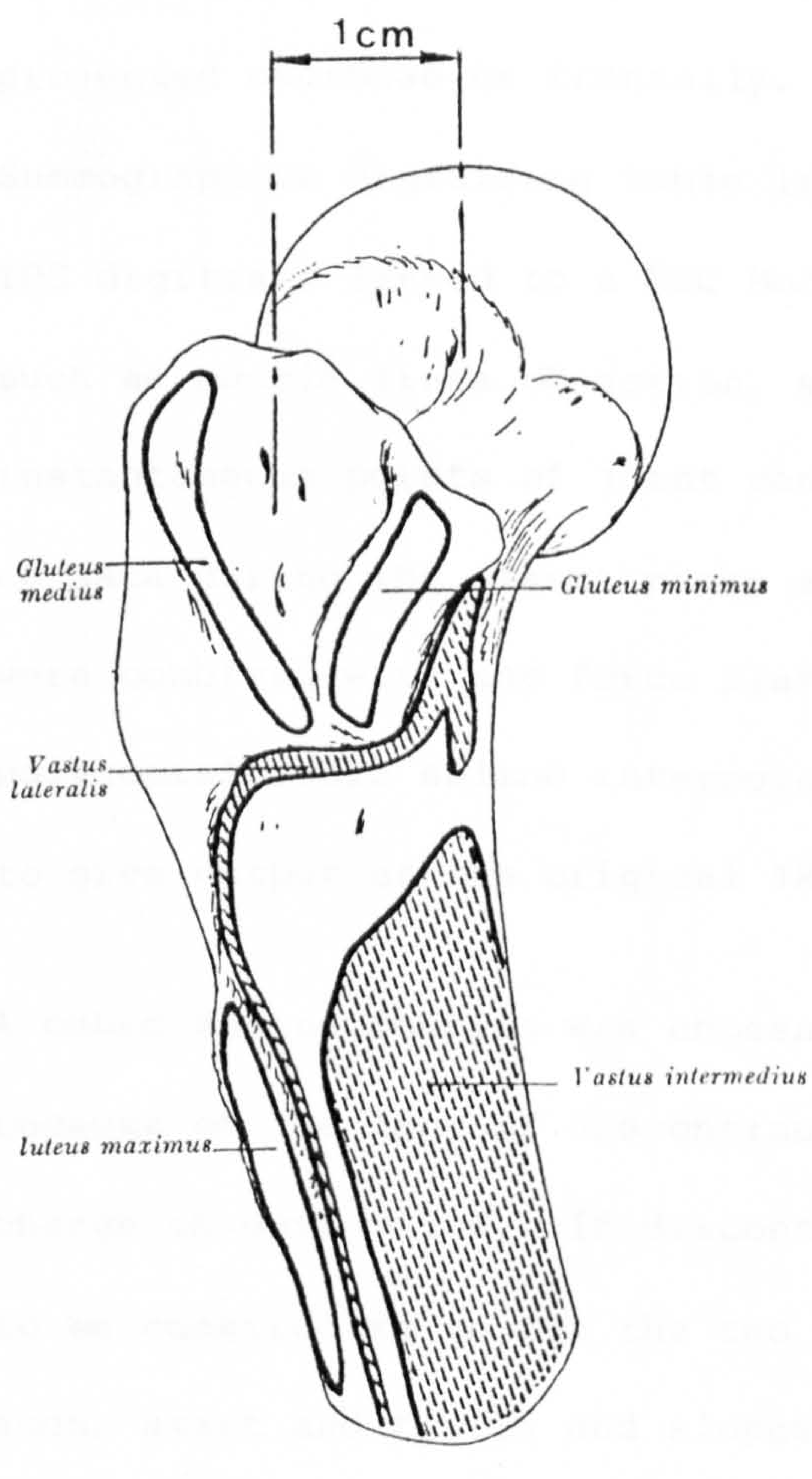


Fig. 4.5b



"The hip joint centre is located approximately 3cm above the most lateral bony prominence of the greater trochanter". From Patterns of Human Motion. Plagenhoef, S.1971.



Hip joint centre, lateral aspect is located 1cm from the "tip of femoral trochanter anterior to the most laterally projecting part of femoral trochanter". From Biomechanics of Human Motion. Williams, M. & Lissner, H.R. 1962. Sketch from Grays Anatomy 1980.

FIGURE 4.6. Hip Centre of Rotation in Relation to Greater Trochanter

thus correcting for radiograph divergence and cut from the photographic paper to make the bone models. The patella was wired to the tibia with fuse wire at a distance appropriate to the patellar ligament and from the corresponding origin and insertion locations. A second model of the foot, from the cuboid to the phalanges was wired to the first foot model, as shown in Plate 4.4, to facilitate the adoption of a flexible foot regime. Verification of the completed bone model could then be carried out by overlaying the model onto the projected cine film.

4.2.2 Data Processing

The bone models, photographically reduced to either 33% or 40% of full size, were combined with the developed cine film which was projected overhead or frontally, also at 33% or 40% full size onto a Summographics digitizing table linked to the DEC 20 computer or a TDS digitiser linked to a BBC Model B microcomputer. Salient points such as muscle lines of action, segmental centres of gravity and instantaneous points of joint contact were digitized for each frame of data during the stance phase as shown in figure 4.7. These data were combined with the force platform data using a piecewise polynomial cubic spline interpolation routine (Conte & Deboor, 1980) to give output at the original 180 Hz sampling frequency.

A cubic spline routine was chosen for its ease of implementation and because of the lack of discontinuities in the data due to sudden change in data value. If discontinuities exist then end slopes have to be constructed to fit the two halves of the data together. As such, start and finish end slopes were calculated from the first and last two data. All other data were then splined between these end

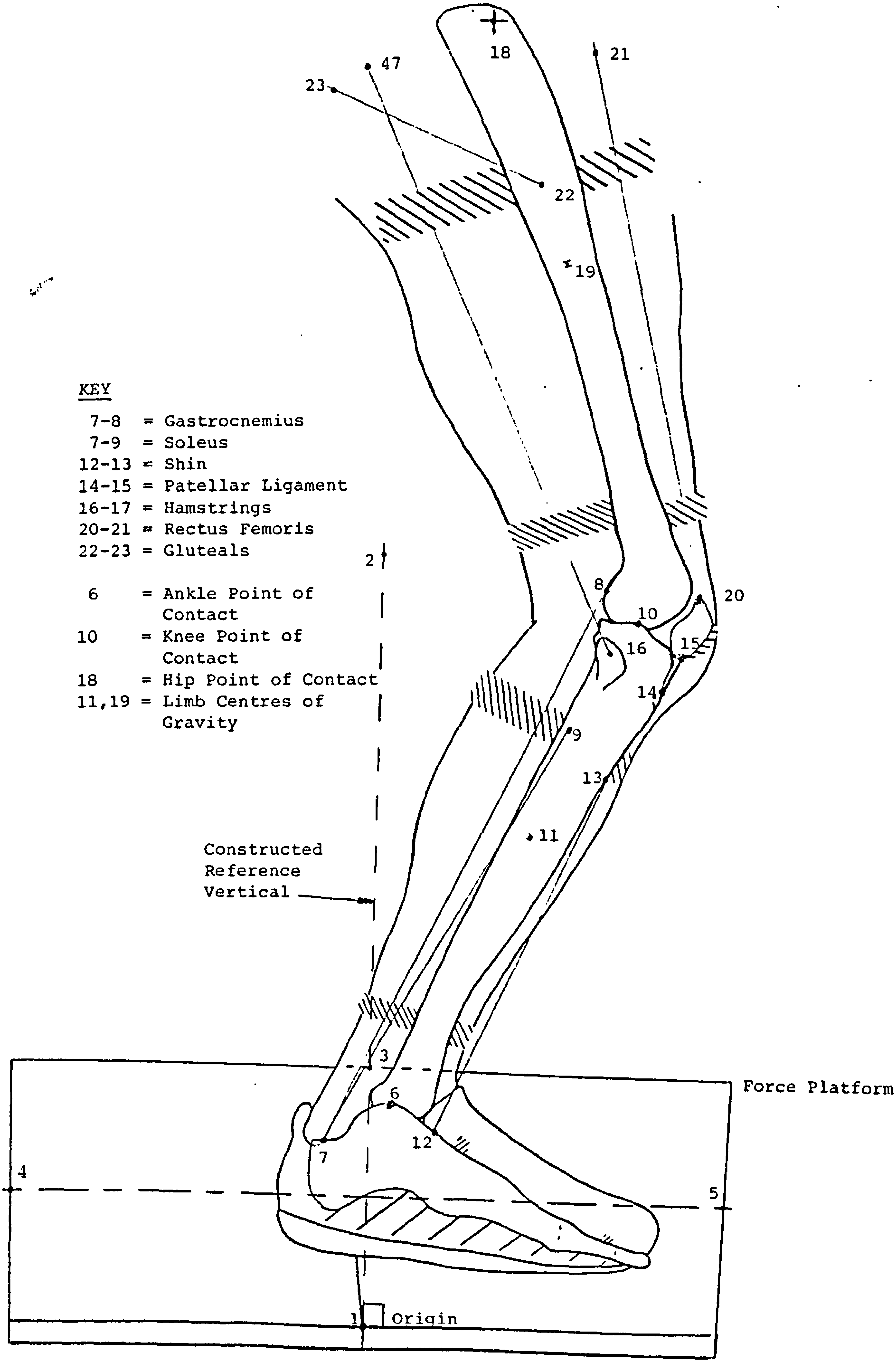
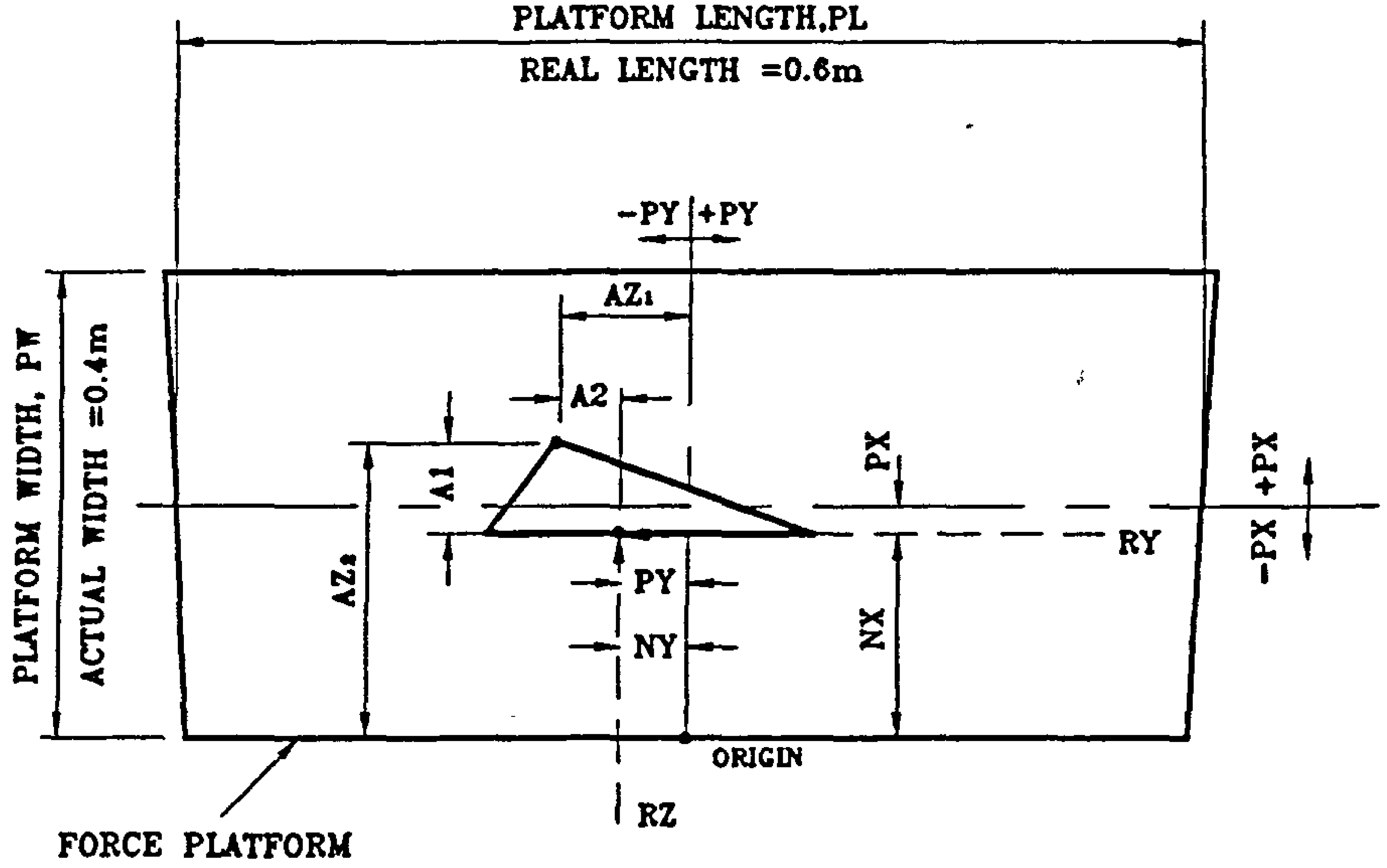


Figure 4.7 Points and Order of Digitisation

slopes.

With the two data sets constructed at the higher frequency of 180 Hz limb displacements and accelerations could be calculated and the data combined to calculate the relative position of the centre of pressure or point of force application on the force platform (Figure 4.8).

Following the calculation of the moment arms A1 and A2 all muscle and joint forces, joint moments and muscular loading rates were calculated as illustrated in the following computer flowchart (Computer program is included in Appendix A3).



PL,PW = FORCE PLATFORM DIMENSION OBTAINED FROM DIGITISED FILM IN SCALE UNITS

PX,PY = CENTRE OF PRESSURE VALUES RECORDED FROM FORCE PLATFORM IN ACTUAL UNITS(m)

NX,NY = SCALED CENTRE OF PRESSURE COORDINATES

AZ_1, AZ_2 = COORDINATES OF ANKLE CENTRE (FROM ORIGIN) AS DIGITISED IN SCALE UNITS

A1,A2 = REQUIRED MOMENT ARMS ABOUT ANKLE OF CENTRE OF PRESSURE OF FOOT/GROUND FORCES IN SCALE UNITS CONSISTENT WITH REMAINDER OF DIGITISED DATA

THUS

$$NX = \left(\frac{0.4}{2} + PX\right)\left(\frac{PX}{0.4}\right)$$

and

$$NY = PY \cdot \left(\frac{PL}{0.6}\right)$$

THEREFORE

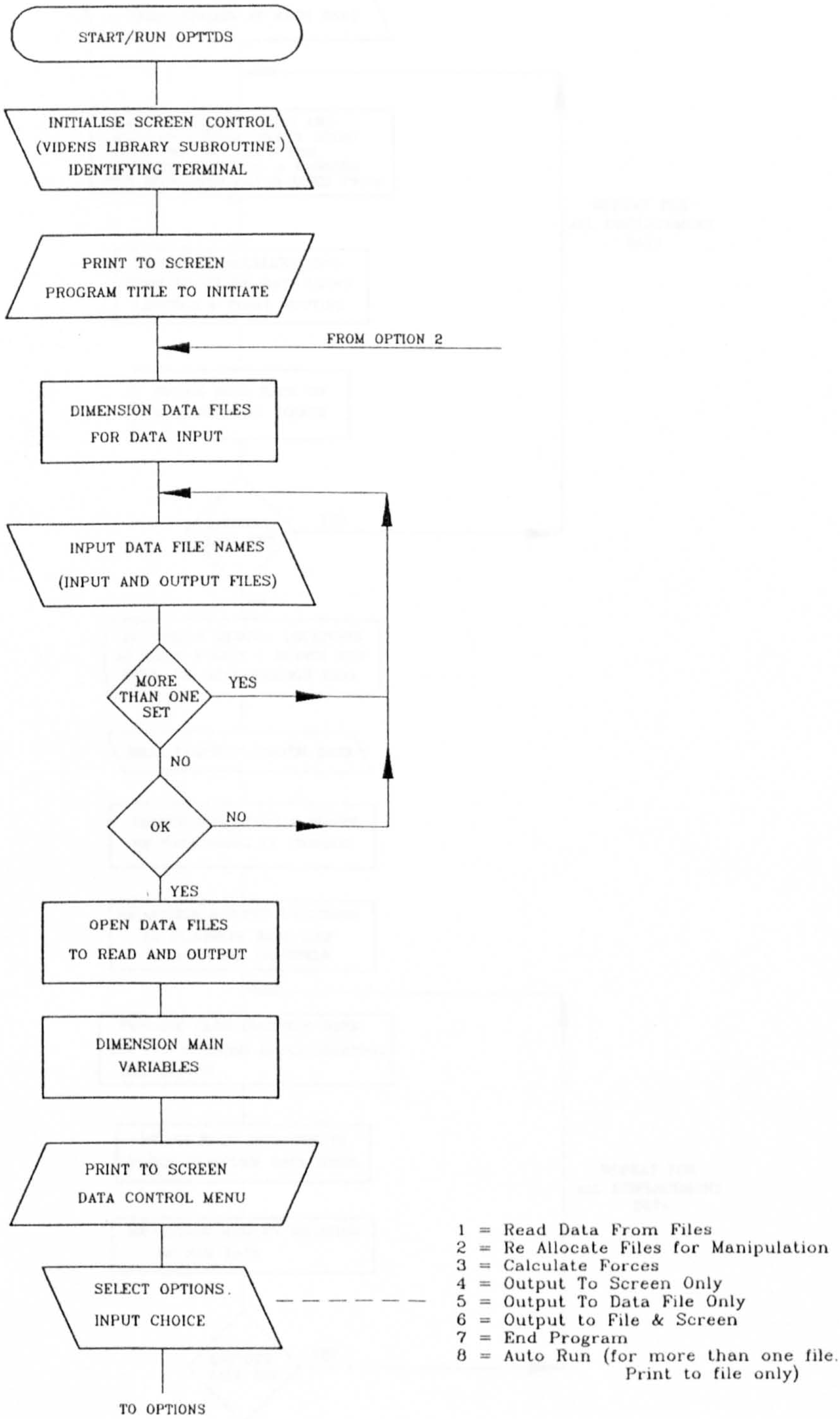
$$A1 = NX - AZ_1$$

and

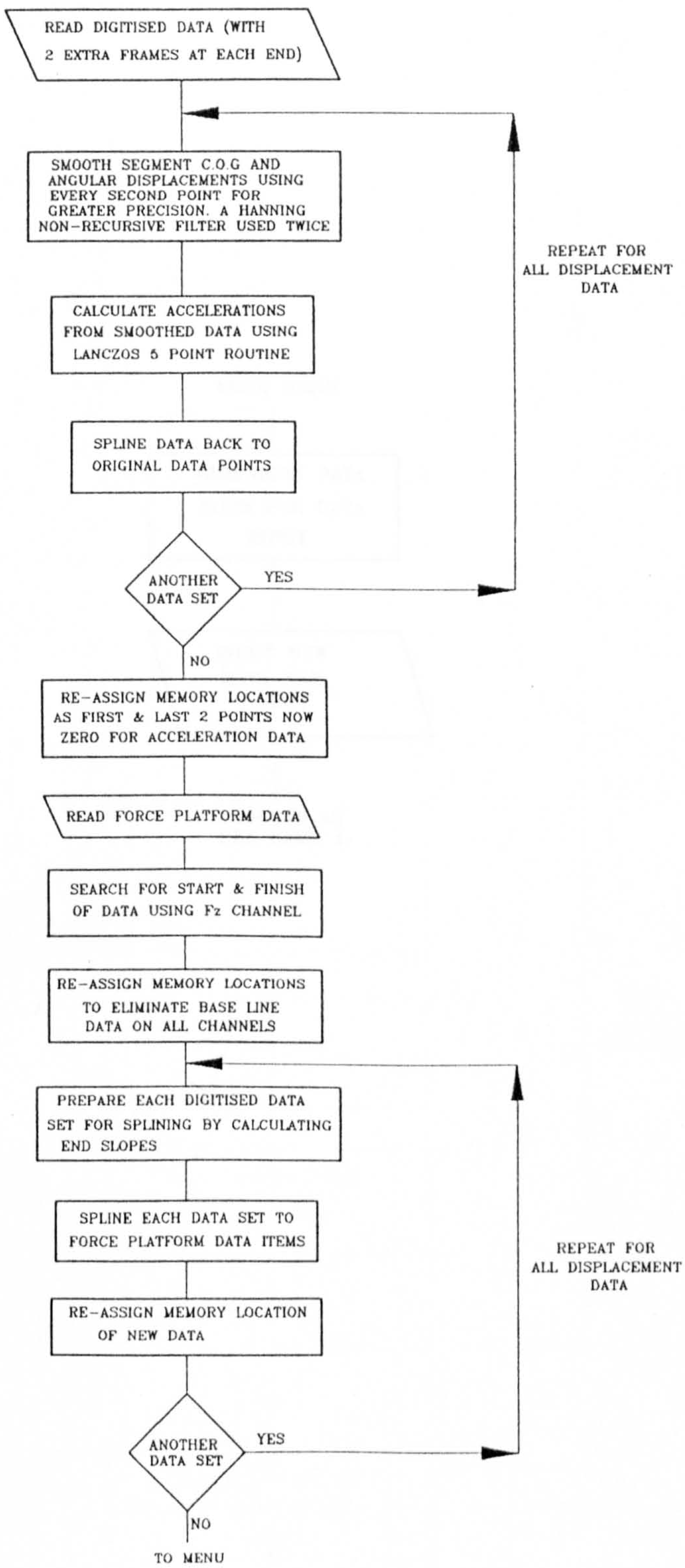
$$A2 = NY - AZ_2$$

FIGURE 4.8 CALCULATION OF THE CENTRE OF PRESSURE BY COMBINING FORCE & FILM DATA.

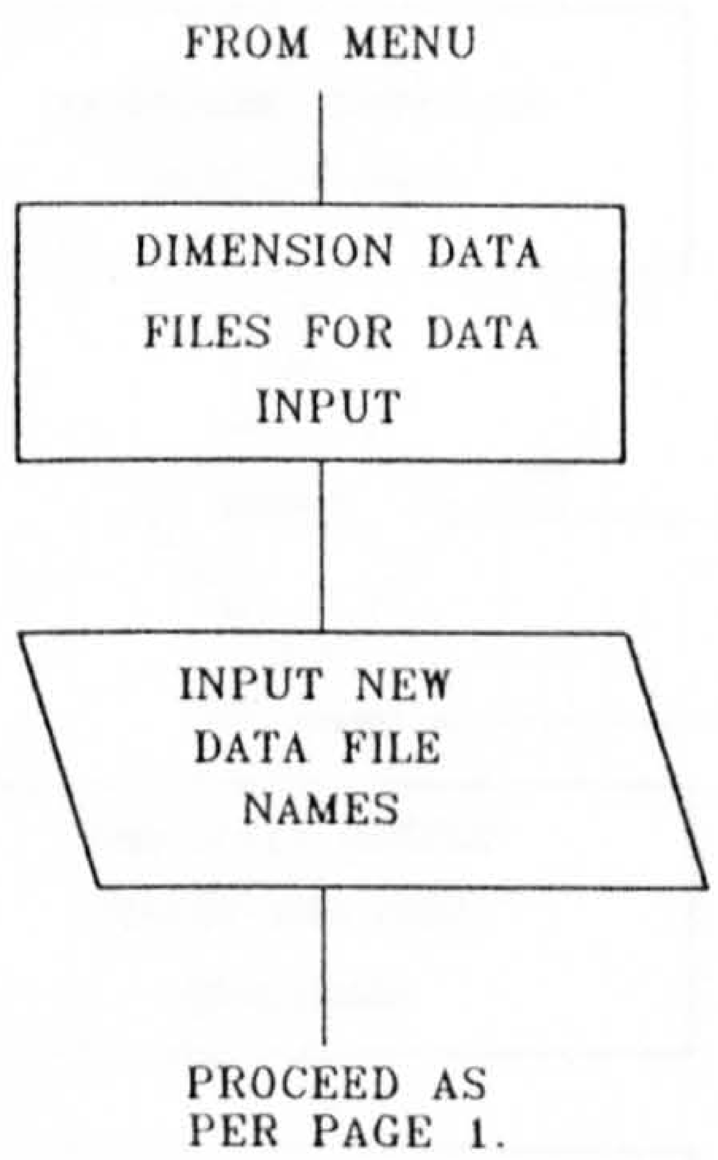
DATA ANALYSIS COMPUTER PROGRAM FLOWCHART :OPTTDS



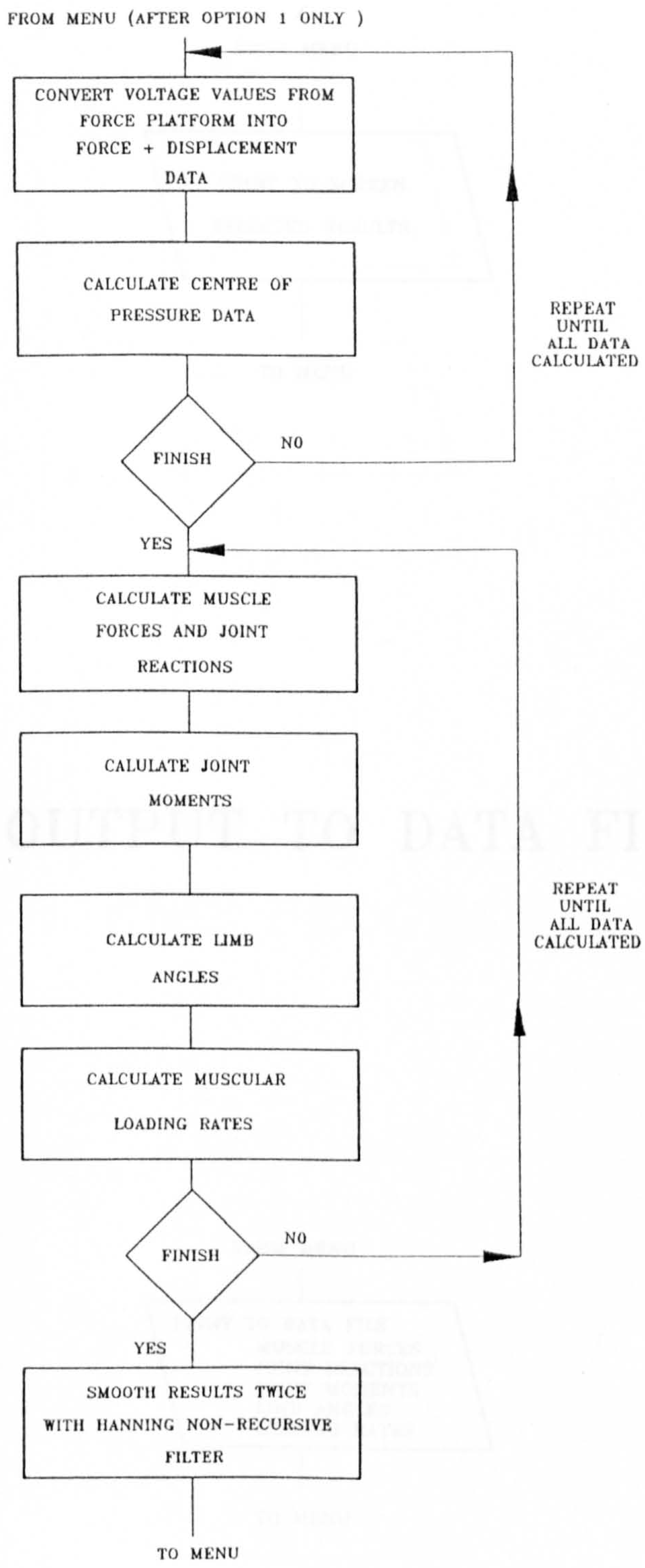
OPTION 1 : READ DATA FROM FILES



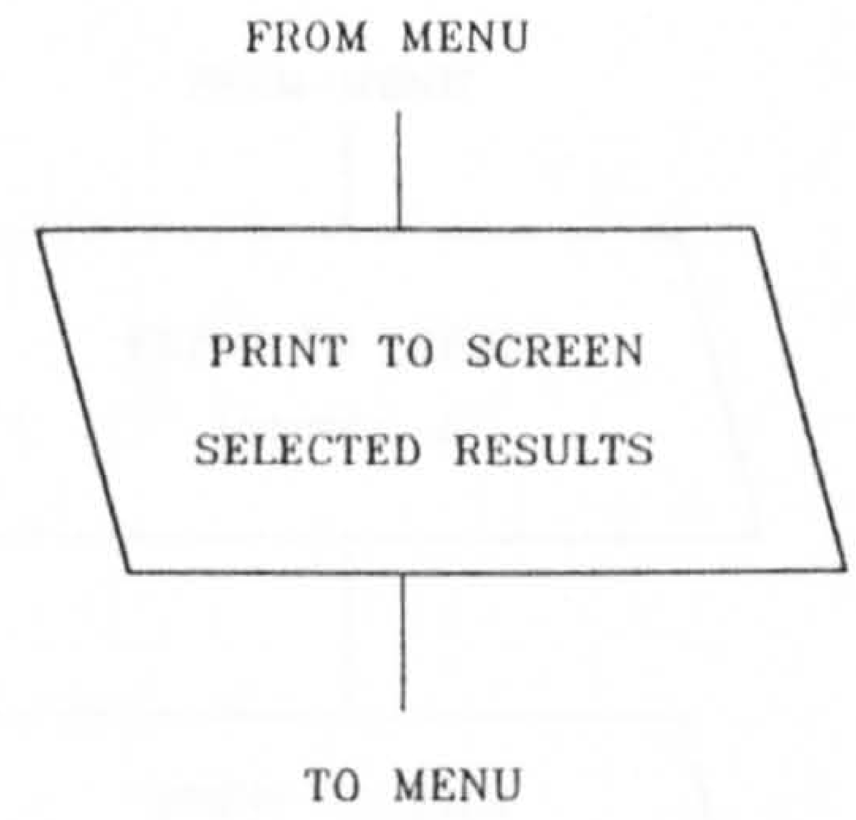
OPTION 2 : RE-ALLOCATE FILES FOR MANIPULATION



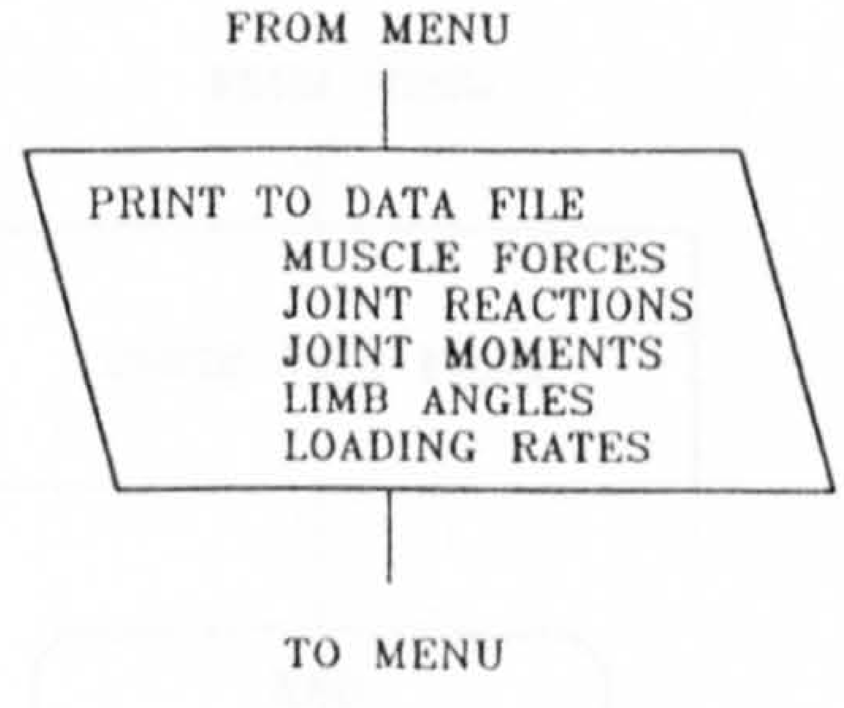
OPTION 3 : CALCULATE FORCES



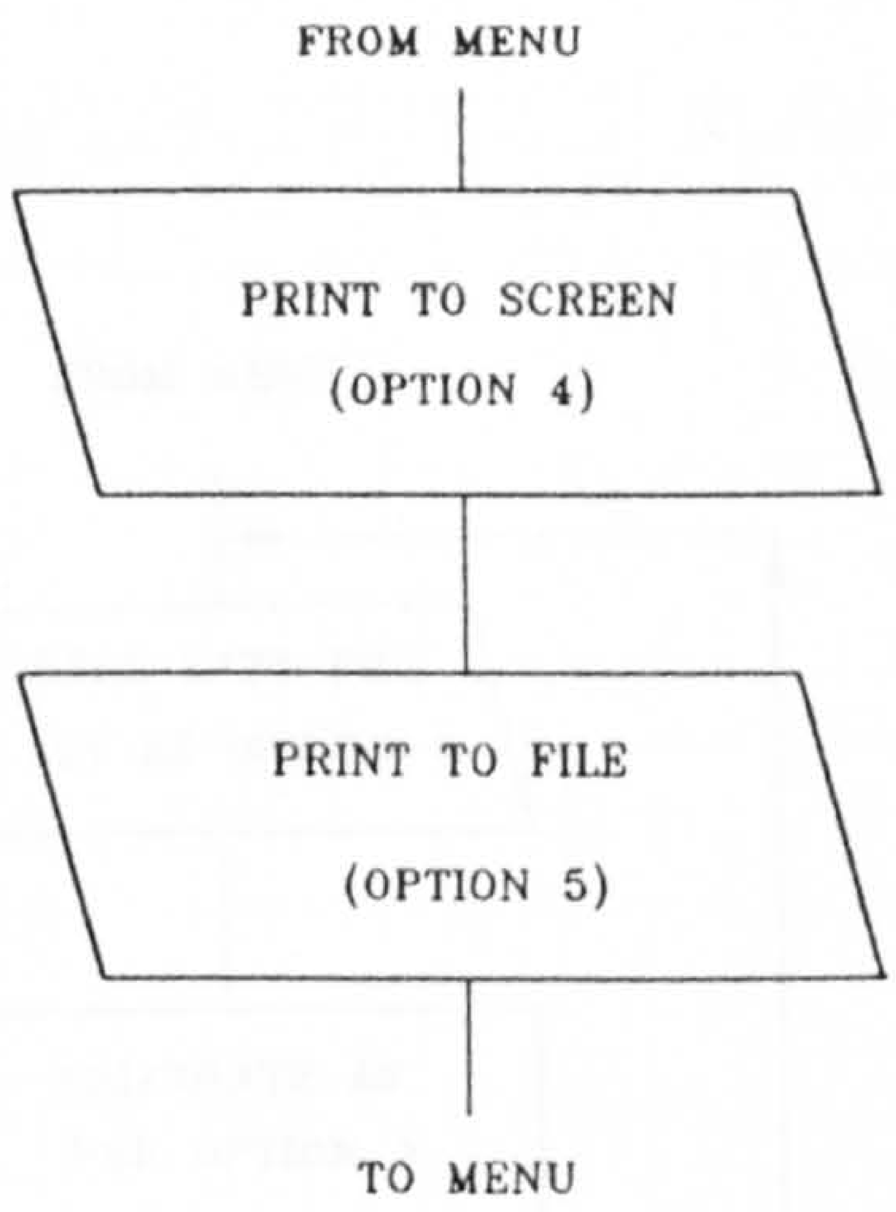
OPTION 4: OUTPUT TO SCREEN ONLY



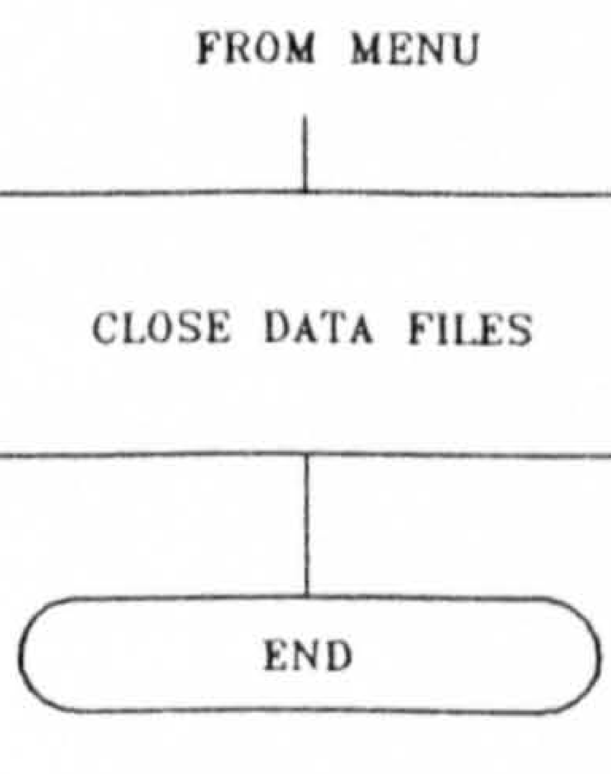
OPTION 5: OUTPUT TO DATA FILE ONLY



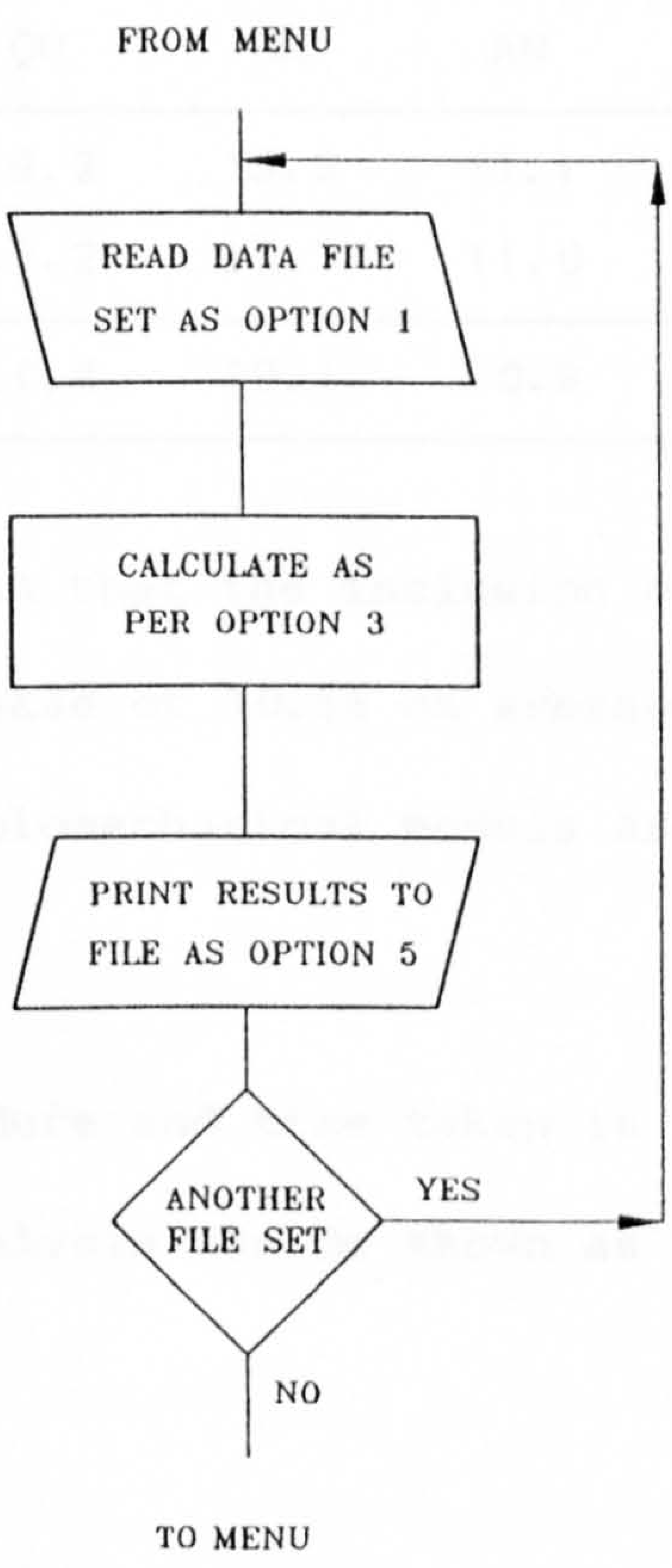
OPTION 6 : OUTPUT TO FILE & SCREEN



OPTION 7 : END PROGRAM



OPTION 8 : AUTO RUN (For more than one file. Print to file only)



4.2.3 Data Analysis

In a previous study (Harrison, 1982) data were analysed with and without D'Alemberts inertia forces to test the effect of using a dynamic model. For the primary biomechanical model presented in section 3.4.1 the following results were obtained.

TABLE 4.2.1 Comparison of Results from Static and Dynamic Models

	Selected Muscle & Joint Forces (BW)				
Model	QU	HM	AN	AS	KZ
Static	19.2	15.2	11.1	6.3	24.5
Dynamic	17.2	12.3	11.0	5.3	23.1
Difference(%)	10.4	19.1	0.9	15.9	5.7

Thus it can be seen that the inclusion of these inertia forces results in a decrease of 10.5% on average and as such all further results from the biomechanical models are for the fully dynamic case.

Overall the procedure and time taken in performing the data capture, processing and analysis can be shown as follows.

TABLE 4.2.2 Gross Data Analysis Procedure and Time Taken

Title	Details	Approximate Time Taken in Hours
Data Collection	Subjects running over platform whilst being filmed	1.5 per 30 runs
Radiograph Production	Anthropometric data collection and radiograph production	1 per subject
Bone Model Production	Reconstruction of radiography, photography and model making	1 per subject
Digitisation	Approximately 24 frames per run	1 per subject
Data Transfer	To Mainframe Computer	1 per 30 runs
Analysis	Splining, plotting and data storage	1 per 30 runs
	Average Time Taken Per Run	1.2 hours

As can be seen from Table 4.2.2 the average time per run for overall data analysis is 1.2 hours and as 339 runs were analysed throughout the study then a total analysis time of over 400 hours has been consumed in the collection of useful data.

To establish the repeatability of the data collected from these 339 data sets three investigations were undertaken. Here the repeatability of the data processing was evaluated, firstly to establish if there were any significant differences in digitising performance between days of the week and secondly to assess if there were similar differences during a single day.

In the first study film from one subject was digitised five times over a full working week, the time of day of that analysis being kept constant. The data were combined with its corresponding force data and three muscle and four joint force curves were produced. A 2-way anova statistical test was performed on these seven selected measures and significant differences between days were found for all items, levels of significance ranging from $p > 0.0005$ to $p > 0.04$ with an average value of $p > 0.01$. This suggests that digitisation of each subject's run should be carried out on a single day.

In the second study film from one subject was analysed twelve times in succession on one day. The digitised data was combined with force data and pertinent peak muscle and joint forces were selected before a statistical error analysis was performed (Bajpai et al., 1974). From these data the required sample size for a given accuracy level was established. A suggested target accuracy of ±3% was used here to establish this sample size. The following results were obtained.

TABLE 4.2.3 Selected Muscle or Joint Peak Force Data used in Repeatability Study

Item	Peak Range (N)	Mean Peak (N)	3% of Mean Peak (N)	Standard Deviation ±(N)
GA	5317-4967	5184	155	110
QU	6139-5532	5827	174	196
AZ	7138-6788	7004	210	110
KN	8060-7389	7695	230	222

For a 3% error, from the mean of each muscle and joint force (99.9% confidence level or 3.3 standard deviations), the following equation was used to establish the respective number of runs required (N_R);

$$N_R = \left(\frac{3.3 \times \text{Standard Deviation}}{3\% \text{ of Mean Peak}}\right)^2$$

These calculations gave the following results;

TABLE 4.2.4 Sample Sizes Required to give ±3% Repeatability

Item	N_R
GA	6
QU	14
AZ	3
KN	10
Average =	8

Overall, the number of runs required per subject to achieve a final repeatability figure of ±3% is eight (p = 0.1%). Similar calculations show that ten runs per subject would result in repeatability within ±2.6%.

Thus between seven and ten runs per subject were employed in this analysis and repeatability is suggested to be approximately ±3%. This figure is, however, not a representation of the model's validity which can only be achieved, as suggested in Section 3, with direct force measurements such as strain gauges implanted into joint replacements and transducers implanted into muscular-tendon complexes. Neither does this figure represent the reproductability between investigators using the same model and method. This can only be established when a number of independant investigators analyse the data separately.

Thus, the ±3% figure of reliability quoted here represents a level of skill and to some extent accuracy in digitising acquired by the single investigator used in this study. To assist the achievement of this required performance value between seven and ten runs per subject per speed were employed and each subject's data (per speed) was analysed on one day.

Table 4.3.1 Summary of Calibratory Work Performed on Data Collection System (See page 112)

Item		Details	Results
1.	(See Figure 4.2)	Kistler force platform's Electronic Unit adapted for use with either an ADC or a UV galvonometer	-
2.	(See Tables 4.3.2)	ADC frequency calibration	1.4% error reduced to 0.4% error (reading slow)
3.	(See Table 4.3.3)	UV Galvonometer frequency calibration	4% error noted (reading slow)
4.	(See Table 4.3.4)	UV Galvonometer drift calibration	35 N/min (worst case)
5.	(See Table 4.3.5 and Plate 4.2)	Kistler force platform point of load application (P.O.L.A.)(AX & AY) calibration	3.1% overall P.O.L.A(1) error noted
6.	(See Table 4.3.6)	Kistler force platform load calibration 0.989	1.05% underweighing noted regression =
7.		Kistler force platform load & point of load application recalibration after change amp. modification (12.5% load error noted)	2.0% overall P.O.L.A. error noted 0.8% underweighing noted regression = 0.997
8.		CIL-16 ADC & BBC B Computer Program verification	To establish 5 channel input of 240 samples, 185 Hz frequency and estimated sampling time of 1.3 seconds

(1) Point of load application.

4.3 Calibration of Equipment

4.3.1 Introduction

At each stage of installation and at various times during the testing regime calibration of all the equipment used in data collection was undertaken as summarised in Table 4.3.1. This included, for example, the two analogue to digital converters (ADC), the force platform and an ultra violet (UV) mirror galvonometer. A summary of the full list of calibrations follows.

4.3.2 Analogue to Digital Converter Calibration

Following modification of the force platform's electronic unit as outlined in Section 4.1 and presented in Figure 4.2 both the ADCs used were calibrated by inputting known signals. Results of these tests are presented in the following Table 4.3.2.

TABLE 4.3.2a Initial Calibratiuon of "3D" A.D.C.

Input Frequency = 50 Hz (Mains)
Selected Frequency of Sampling = 900 Hz

Trial	Output Frequency Recorded (Hz)	Error(%)
1	888.5	1.28
2	889.5	1.17
3	886.1	1.54
4	886.3	1.52
5	887.0	1.44
Mean Error		1.39%

TABLE 4.3.2b Final Calibration of "3D" A.D.C.

Input Frequency = 50 Hz (Mains)
Selected Frequency of Sampling = 900 Hz

Trial	Output Frequency Recorded (Hz)
1	50.36
2	50.40
3	50.02
4	50.08
Mean	50.21
Error	0.42%

Thus the results indicated that sufficient accuracy could be expected from equipment and data collection could proceed.

4.3.3 U.V. Galvonometer Calibration

This was performed in two separate experiments, firstly the frequency 'response' was calibrated then the drift characteristics established. For the frequency calibration a known input (National Grid 'Mains') signal was used and a four percent variation in output signal was noted as shown in Table 4.3.3.

TABLE 4.3.3 UV Galvo Frequency Calibration

Item	Frequency (Hz)	No. of Cycles	Sample Time (seconds)
Mains Input	50	72	1.44
UV Output	48(1)	72	1.5
Error	4%		

$$\text{Output Frequency} = \frac{\text{No of Cycles}}{\text{Sample Time}}$$

For the drift characteristics, the galvonometer was connected to the Kistler Force Platform to be used in this study. No load was applied and the amount of drift from a datum setting was noted after 10 minutes as recorded in Table 4.3.4.

TABLE 4.3.4. UV Galvo. Drift Test

UV Magnification Setting:- 1 v/cm
Platform Sensitivity:- 500 Channel Units/Volt[1]

Channel	UV Drift After 10 Minutes in cm	UV Drift in Channel Units After 10 minutes	UV Drift in Channel Units/Min
FZ	0.7 cm	350.0 N	35.0 N/min
FY	0.275 cm	137.5 N	13.75 N/min
FX	0.025 cm	1.5 N	1.25 N/min
AY	0.35 cm	1.75 cm	0.175 cm/min
AX	0.25 cm	1.25 cm	0.25 cm/min
MZ	0.075 cm	0.375 Nm	0.0375 Nm/min

(1) Channel Units = Newtons for PZ, FY, FX; cm for AY, AX and Nm for MZ

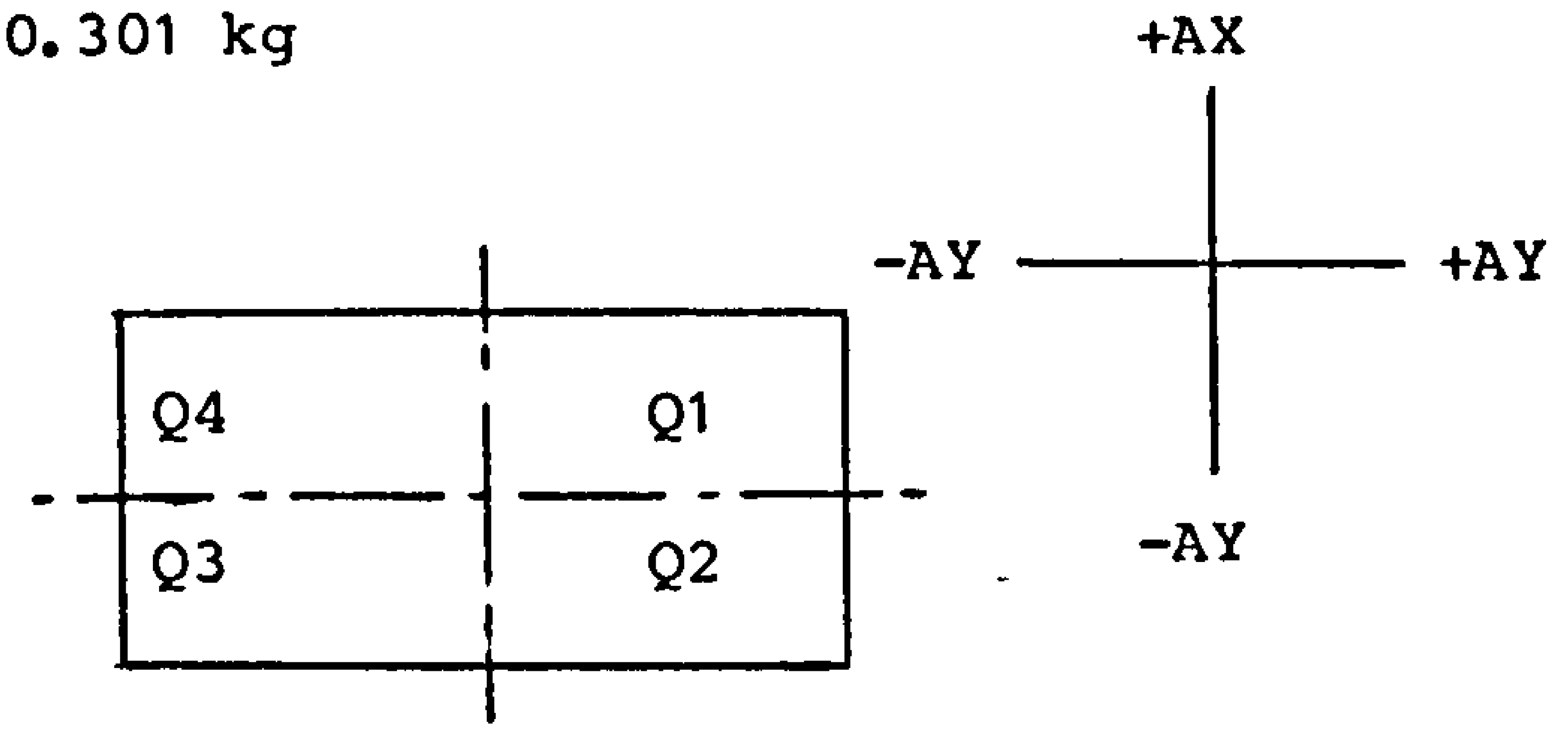
The worst channel could thus be considered to be FZ at 35 N/min but as the average time to perform each subjects run was only one minute, and resetting could be performed before each run, this was considered reasonable.The UV Galvonometer was only to act as back up to the main data recording equipment outlined in Section 4.1 and thus these for frequency and drift were acceptable.

4.3.4 Force Platform Calibration

With the Kistler Force Platform's electronic unit modified to calculate the point of application of force indirectly as outlined in Section 4.1 verification was performed on the platform, ADC and corresponding computer program. A tripod or point loading frame of known weight (Plate 4.2) was used to apply loads onto random locations on the force platform. The actual or measured and calculated results are presented in Table 4.3.5.

TABLE 4.3.5 Force Platform Positional Calibration

Weight used = 20 kg & 0.301 kg
= 199.15 N



Position	Measured values(mm) AX	AY	Program Results(mm) AX	AY	% Absolute Error AX	AY
Q1	42	184	41	181	2.4%	1.6%
Q2	-109	156	-108	159	0.9%	1.9%
Q3	-144	-243	-135	-234	6.3%	3.7%
Q4	79	-127	80	-132	1.3%	3.9%
Centre	0	0	1.4	0.6	-	-
Q2	-63	69	-60	72	4.8%	4.3%
Q2	-90	107	-87	110	3 3%	2.8%
Q2	-110	134	-108	138	1.8%	3.0%
Q2	-144	180	-138	182	4.2%	1.1%
Q2	-169	215	-162	212	4.1%	1.4%
Q2	-187	273	-183	263	2.1%	3.7%
			Mean error		3.12	2.75
			Overall error		2.95	

As shown an overall error of 2.9% in the calculation of the point of application of load was obtained.

Using the same equipment the load response of the force platform was tested by applying varying loads. The following results were obtained.

TABLE 4.3.6. Calibration of Force Platform's Load Response

Applied Load (kgs)	Recorded Load (kgs)	Error (%)
10	9.96	0.45
20	19.96	0.21
40	39.97	0.08
60	58.14	3.11
70	68.49	2.16
70	68.94	1.51
60	59.58	0.70
20	19.96	0.21

These results show a mean error of 1.05% with a regression slope of r = 0.989.

Both these results were found unacceptable and after modification to one of the force platform's electronic unit's charge amplifiers the point of load application and load response were re-assessed and the followed results obtained.

Point of Load Response Overall Error = 2.0%

Load Response Mean Error = 0.77%(Regression Slope = 0.997).

These results were deemed acceptable and data collection could take place.

5. RESULTS

Four phases of experimentation took place which corresponded to the development of the biomechanical models outlined in Section 3.4. Firstly sample muscle and joint forces were calculated from four subjects running at 4.47 ms^{-1} in order to establish the data collection, processing and analysis route. Using this as a foundation the second phase of experimentation evaluated different running styles of front- and rear-foot strikes also at 4.47 ms^{-1}. Two groups of seven athletes were employed here. In the third stage of this investigation, running speed was varied and nine athletes performed the experiment at the velocities of 3.83 ms^{-1}, 4.47 ms^{-1} and 5.36 ms^{-1}. The final phase concerned enhancing the bioengineering model with the inclusion of the soleus muscle and data was reanalysed from the third phase at all speeds for the same nine subjects.

All subjects were highly experienced athletes, members of local athletic clubs, who regularly completed 70 km per week in training and raced at distances from 800 m to marathon. Common elements across all regimes involved the same footwear for all subjects, warm up and familiarisation routines, film and force platform co-ordination and data analysis procedures. Results obtained for each phase varied in their content depending on, as suggested above, the model and the relevance to the object proposed. All graphs are presented with a time based abscissa starting at foot strike and terminating at take off and are mean curves from all subjects.

As reported in Section 4.2 all data were analysed with and without D'Alemberts inertia forces. The inclusion of these elements reduced the mean peak muscle and joint forces by an overall value of approximately 10% and hence all results are presented for a fully dynamic bioengineering model of the lower limbs. Significance between speeds, styles and model employed were tested for significance using pair t tests. These differences are presented with each result section.

5.1 Phase One: Establishment of Muscle and Joint Forces

The model outlined in Section 3.4.1 was used to establish the experimental procedure. Table 5.1.1 summarises the testing and model basis for these results and subject information is included in Table 5.1.2.

TABLE 5.1.1 Phase One Experimentation

Item	Data		
Speeds (ms^{-1})	5.36	4.47	3.83
Subjects	-	4	-
Runs/Speed(1)	-	29	-
Model Includes:			
SH, Shin	-	YES	-
SO, Soleus	-	NO	-

(1) Total runs for all subjects, average of 7 runs per subject.

Data are presented either graphically or numerically. Graphical data (Figs. 5.1.1 to 5.1.3) are mean curves from all subjects and a full summary of all the results is given in Tables 5.1.3 to 5.1.5.

Table 5.1.3 shows the mean peak joint moments for the four subjects and the overall mean for all subjects. Knee joint moments are the highest at 188 Nm while the maximum hip moment is the smallest at 98 Nm. These results afford good comparison with the published data presented in Table 2.5.3.

The muscle and joint forces generated within the human lower limbs are presented for all subjects in Table 5.1.4 in body weight units(BW), whilst Figs. 5.1.2 and 5.1.3 show the muscle and joint forces, respectively, in newtons. The largest of the mean peak muscle forces present is that of the quadriceps (QU) group at 22.0 BW (mean for all subjects). The gastrocnemius (GA) develops the next largest muscle force at 7.3 BW, whilst the shin muscle group (SH) develops only 0.8 BW during the first 9.3% of the stance phase as determined by the ankle joint moment. The hamstrings also exhibit a relatively low level of activity at 3.1 BW.

Joint forces are greatest at the knee. The compressive component exhibits a mean peak maximum of 33.0 BW and the shear component a maximum of 2.5 BW. The compressive force of the ankle joint is 9.0 BW whilst the shear component is 3.9 BW, but is of an opposite sense to the shear component of the knee joint.

The high muscle forces developed during the stance phase occur over a short period of time and, as can be seen from Fig. 5.1.2, peak values for the quadriceps muscle group are reached after approximately 0.09 s. This short time period gives muscular loading rates of on average 175 kN s^{-1} (as displayed in Table 5.1.5). The gastrocnemius muscle can be seen to exhibit a more gradual loading rate, and Komi et al. (1985) quoted values of 100 kN s^{-1} for an in vivo measurement of the loading rate of the achilles tendon.

TABLE 5.1.2 Subject Information

Subjects (n=4)	Age (years)	Height (m)	Weight (kg)
Mean	21.50	1.78	65.3
Standard deviation	±0.58	±0.66	±3.3
Range	21-22	1.72-1.87	62.5-69.8

TABLE 5.1.3 Summary of Joint Moment Results

Maximum joint moments (Nm)(mean of seven trials per subject)			
Subject number	Ankle	Knee	Hip
1	-155	-183	137
2	-169	-182	83
3	-198	-192	103
4	-172	-194	70
Total mean	-174	-188	98

TABLE 5.1.4 Mean Peak Muscle and Joint Forces in Body Weight (BW) Units for Seven Runs Per Subject

	Muscle forces[a]				Joint reactions				Ground[b]		
Subject Number	SH	GA	QU	HM	AN	AS	KN	KS	RZ	Foot contact time(s)	Shin activity time(s)[c]
1	0.5	7.1	21.9	2.6	-9.2	-3.5	-32.5	1.2	3.1	0.20	0.025
2	0.3	7.4	20.2	2.5	-9.2	-4.2	-30.8	1.8	2.9	0.21	0.018
3	1.4	6.3	25.1	3.7	-7.8	-3.6	-35.4	3.4	2.7	0.19	0.014
4	0.9	8.5	21.0	3.7	-10.0	-4.3	-33.5	3.6	2.8	0.19	0.017
Total mean	0.8	7.3	22.0	3.1	-9.0	-3.9	-33.0	2.5	2.9	0.20	0.018[d]

a See Table 3.1.2 for nomenclature

b Vertical foot - ground reaction force

c Time from first foot contact

d Equal to 9.3% of stance phase

TABLE 5.1.5 Summary of Maximum Quadriceps Muscular Loading Rate Results

Subject number	Maximum loading rate (kNs^{-1})
1	165
2	135
3	199
4	201
Total mean	175

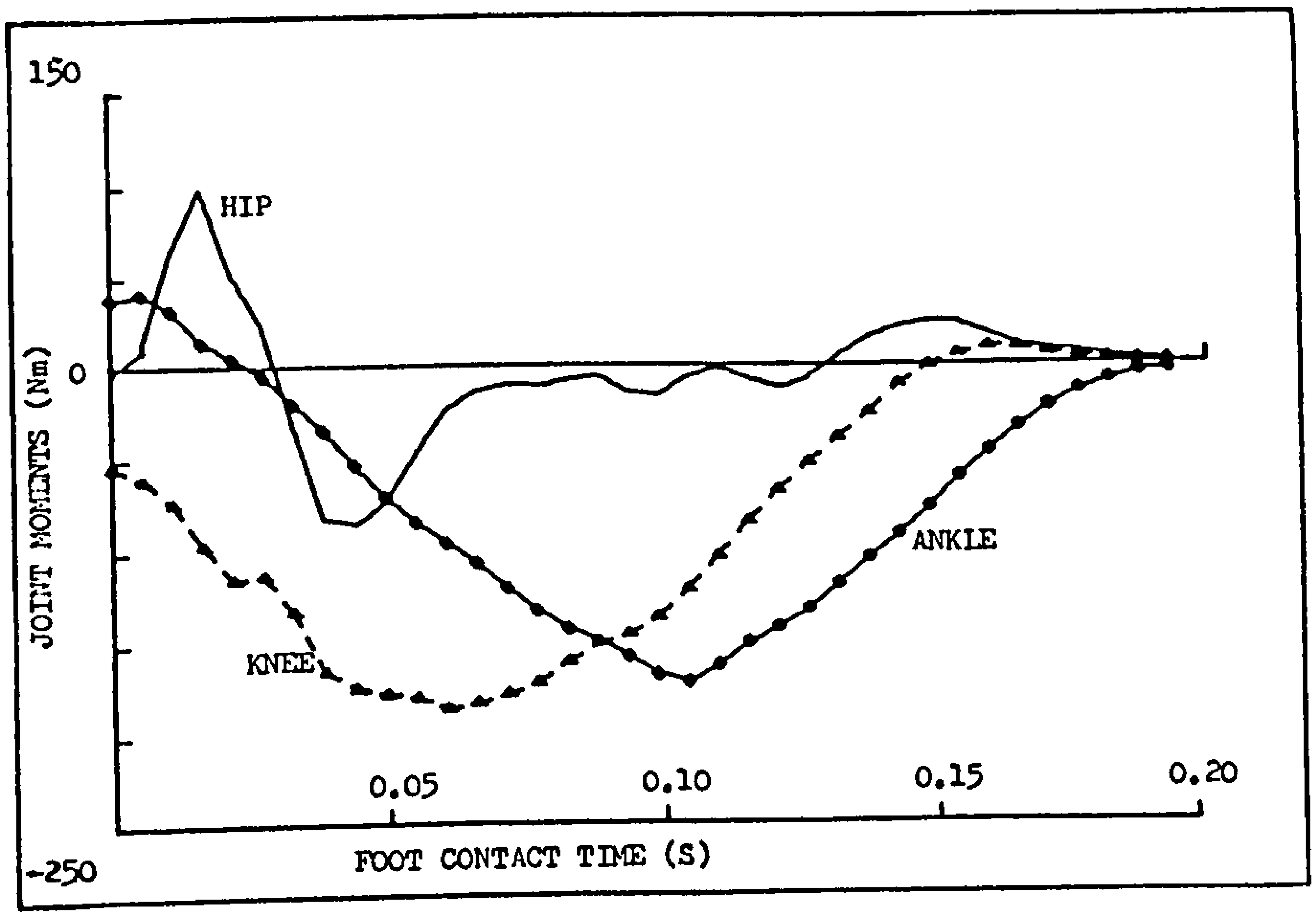


FIGURE 5.1.1. Net Joint Moments During Stance Phase

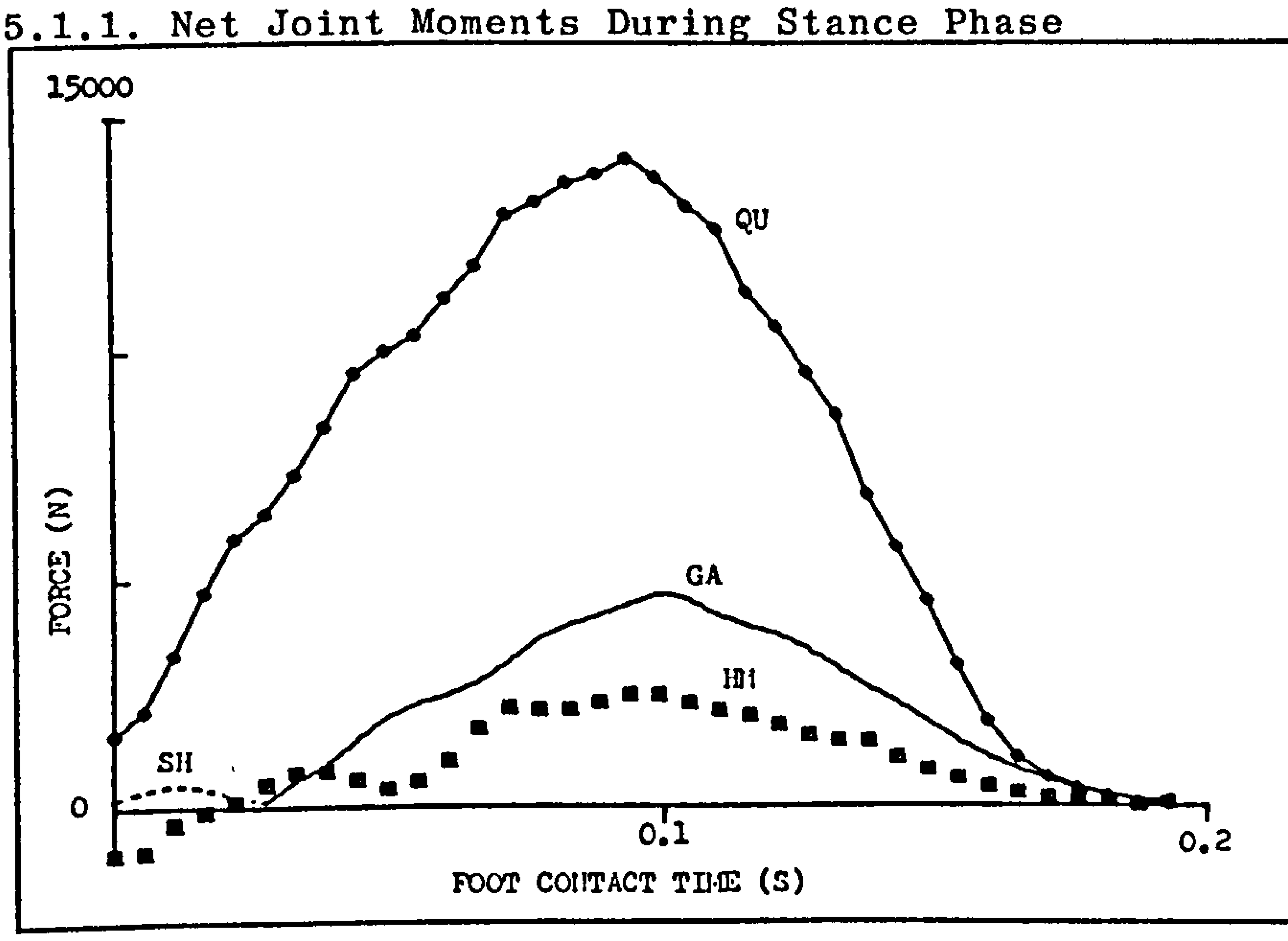


FIGURE 5.1.2. Mean Muscle Forces During Stance Phase for Four Subjects

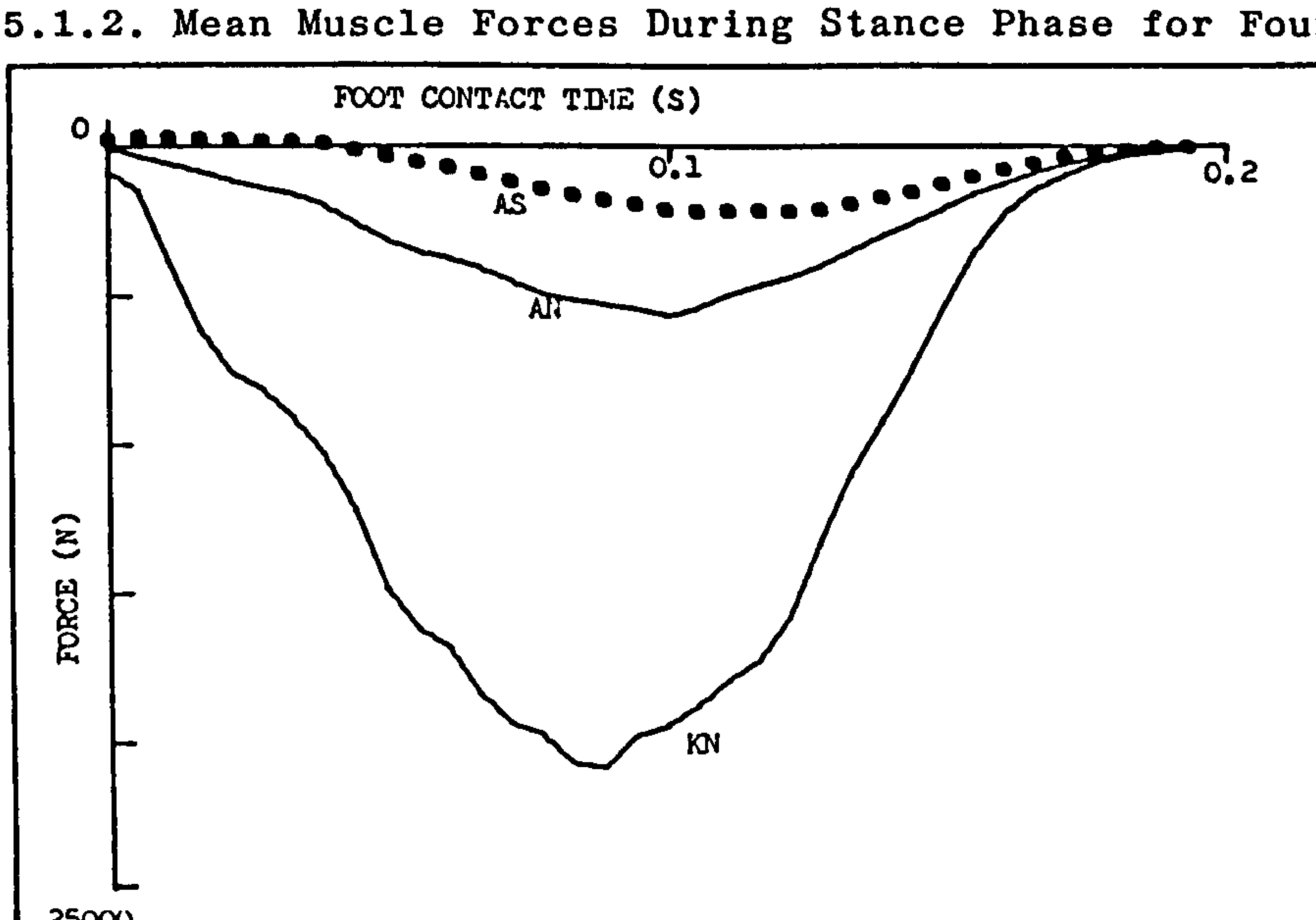


FIGURE 5.1.3. Mean Joint Forces During Stance Phase for Four Subjects

5.2 Phase Two: Front Foot Strikers versus Rear Foot Strikers

Fourteen highly experienced and well trained male middle distance athletes were selected on the grounds of their athletic performances. This group included one European Champion, one Welsh International, five National student representatives with the remainder being top class club representatives in the North West of England. Seven of these athletes exhibited rear foot strike patterns as identified by their foot/ground reaction force profile and their recorded foot contact centre of pressure pattern (Cavanagh & Lafortune, 1980). The remaining seven being grouped together into fore foot strikers. Table 5.2.1 summarises the experimental methodology and Table 5.2.2 displays subject data.

TABLE 5.2.1 Phase Two Experimentation

Item	Data		
Speeds ($m.s^{-1}$)	5.36	4.47	3.83
Subjects	-	14	-
Runs/Speed(1)	-	101	-
Model includes:			
SH, Shin	-	YES	-
SO, Soleus	-	NO	-

(1) Total runs for all subjects, average of 7 per subject

The results for the two groups are presented graphically in Figures 5.2.1 to 5.2.7 and are summarised in Table 5.2.3. Standard deviation bands are not included to assist comparison between groups.

Figures 5.2.1 to 5.2.3 show differences between the rear and front strikes in the muscle groups gastrocnemius, shin, quadriceps and hamstrings. For the gastrocnemius activity the rear foot group have a much more defined peak than the front foot group and this peak appears somewhat phase shifted. The converse is true for the quadriceps muscle group and the rear foot group shows initial muscular activity. More significant profile differences occur in the hamstrings although for both an initial peak, followed by a reduction in tension then a gradual increase to peak values is observed. The negative force values presented in the figure are minimal and would normally be viewed as representing a quiescent state.

The joint reaction forces are represented in Figures 5.2.4 to 5.2.7. Both groups profiles are very similar for the ankle joint shear and compressive components (Figures 5.2.4 and 5.2.5 respectively). More subtle differences occur in the knee joint compressive component (Figure 5.2.7), even though the peak values occur at the same time, the loading of the rear foot group begins earlier then levels out to a more 'gradual' maximum value. The knee joint shear components exhibit the greatest profile differences between groups as the rear foot group's profile is predominantly positive (posterior-anterior) whilst the front foot group's is almost entirely negative (anterior-posterior). The profiles exhibit similarity however in that they initially decrease to a minimum value, then increase to a maximum before fluctuating back to zero through the secondary peak.

All these results are summarised in Tables 5.2.3 to 5.2.6 with the significant differences between the groups peak muscle and joint forces identified in Table 5.2.3. Table 5.2.5 shows the shin muscle activity time for both groups in seconds and as a percentage of the total foot contact time. The front foot group's activity time is less than that for the rear group (8.8% compared to 10.5%), however, this was not found to be significant due to high intra group variance. Table 5.2.4 shows the foot/ground contact data for the experiment and here the greatest differences between groups is found in the foot contact times; the front foot strikers contact time being the shorter of the two, 0.19s compared to 0.20s.

The differences between groups which were found to be significant, are summarised in Table 5.2.6.

TABLE 5.2.2 Subject Information (N = 7 per foot strike group)

Foot Strike Characteristics F = Front Foot R = Rear Foot	Age (Years) F	Age (Years) R	Height (m) F	Height (m) R	Weight (kg) F	Weight (kg) R
Mean	20.1	19.9	1.76	1.80	68.9	69.9
Standard Deviation	1.8	1.3	0.05	0.04	5.6	8.1
Range	18-23	18-22	1.70-1.83	1.73-1.85	60.5-76.2	62.5-85.0

TABLE 5.2.3 Mean Peak Muscle and Joint Forces

Foot Strike Group	Muscle Forces SH	Muscle Forces GA	Muscle Forces QU	Muscle Forces HM	Joint Reactions AN	Joint Reactions AS	Joint Reactions KN	Joint Reactions KS(1)
Front	0.49	7.61	20.38	2.66	-9.34	-4.13	-31.02	1.98
Rear	0.74	7.60	20.65	2.93	-9.08	-4.13	-31.74	2.45
Signif. Differ.	$p < 0.1\%$	No significant differences						$p < 1\%$

(1) Modulus value reported here

TABLE 5.2.4 Foot/Ground Reaction Forces and Foot Contact Time

Group	Vertical Force (RZ) (BW)	Contact Time (CT) (S)
Front	2.90	0.193
Rear	2.83	0.202
Significant Difference	$p < 1\%$	$p < 0.1\%$

TABLE 5.2.5 Shin Activity Time in Seconds and as a Percentage of Overall Foot Contact Time

	Shin Activity Time in seconds		Shin Activity as Percentage of Foot Contact Time	
Group	$\bar{x}$	σ	Per Group	Overall
Front	0.017	±0.013	8.8	9.7
Rear	0.021	±0.011	10.5	

No significant differences found

TABLE 5.2.6 Summary of Features Found Significantly Different Between the Two Groups

Feature	SH Peak Force (BW)	SH activity time (s)	KS Peak Force (BW)	RZ Peak Force (BW)	CT (S)
Group:					
Front	0.49	0.017	1.98	2.90	0.193
Rear	0.74	0.021	2.46	2.83	0.202
Rear Group	higher	longer	higher	lower	longer
Significance $p <$	0.1%	None	1%	1%	0.1%

FIGURE 5.2.1. Gastrocnemius & Shin Muscle Force Mean Curves for Rear and Front Foot Strikers

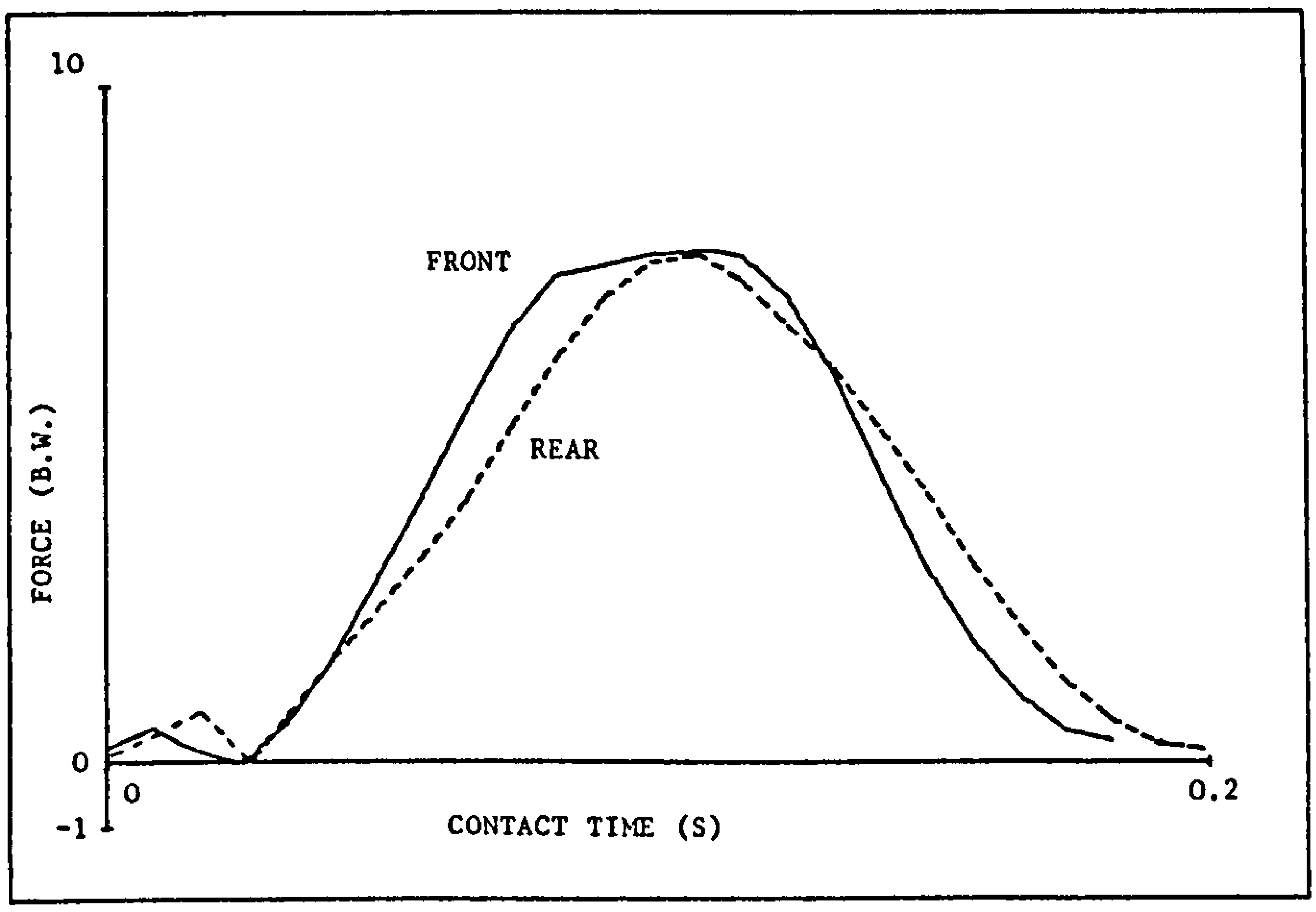


FIGURE 5.2.2. Quadricep Muscle Force Mean Curves for Rear and Front Foot Strikers

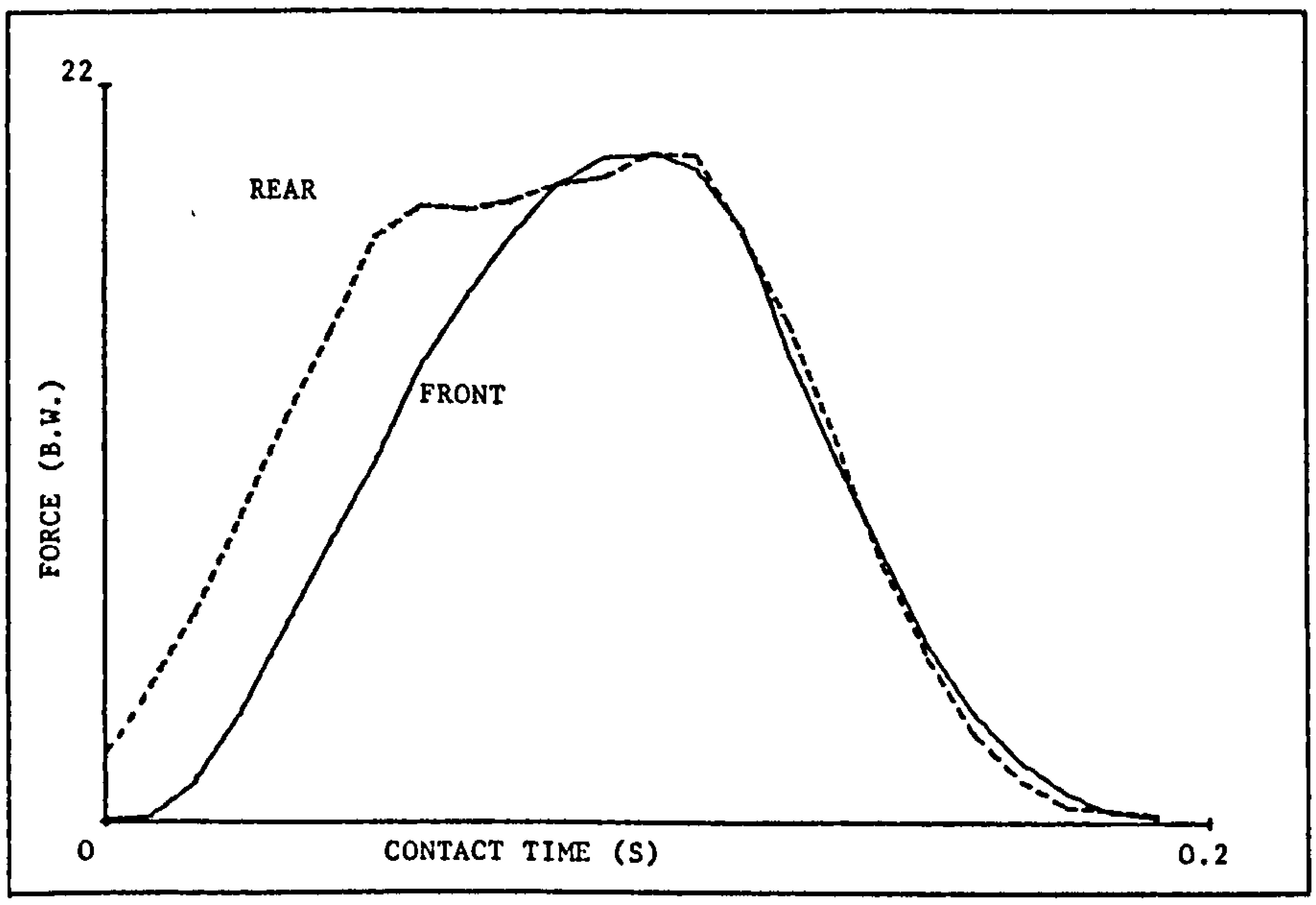


FIGURE 5.2.3. Hamstring Muscle Force Mean Curves for Rear and Front Foot Strikers

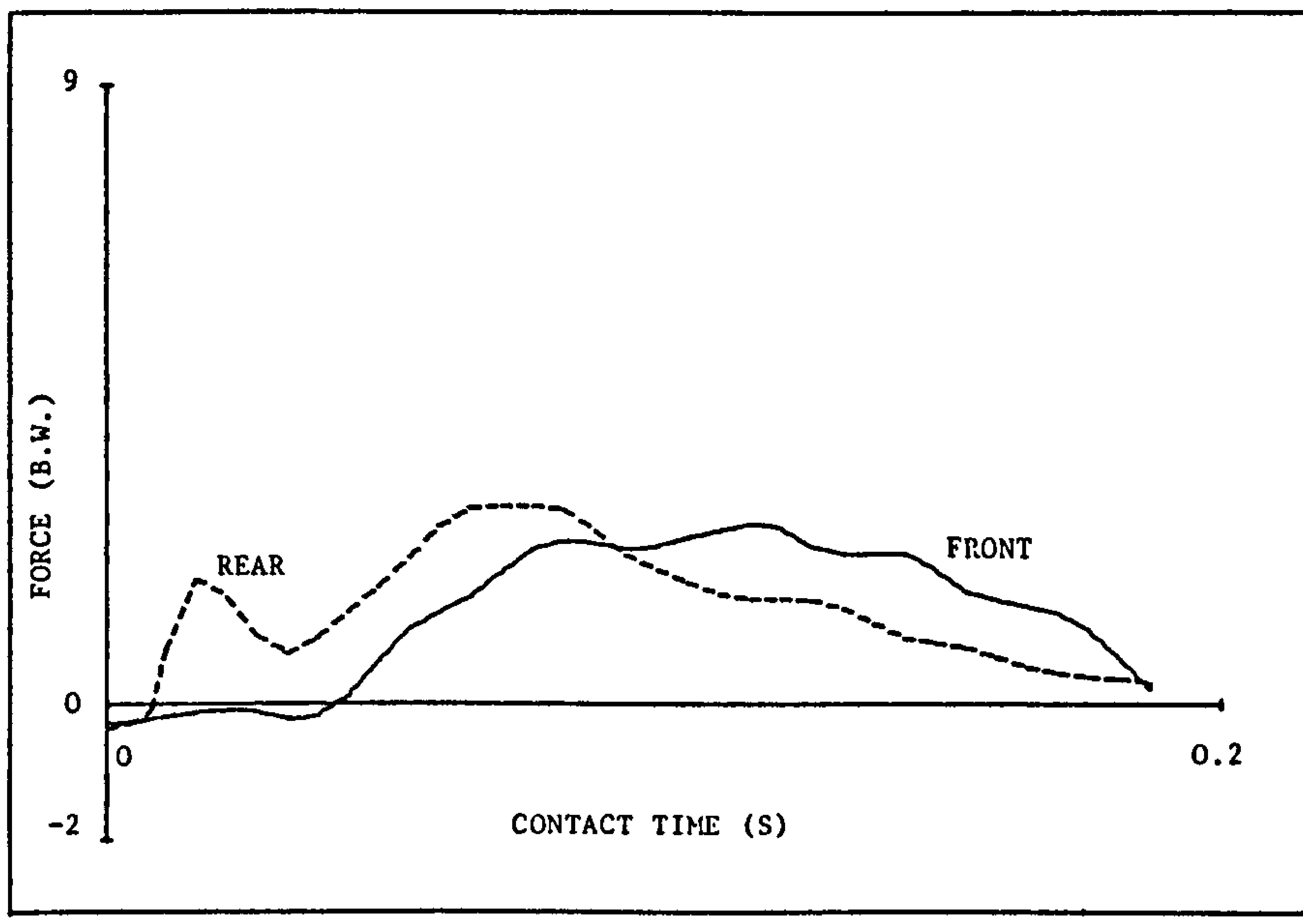


FIGURE 5.2.4. Ankle Joint Shear Reaction Mean Curves for Rear and Front Foot Strikers

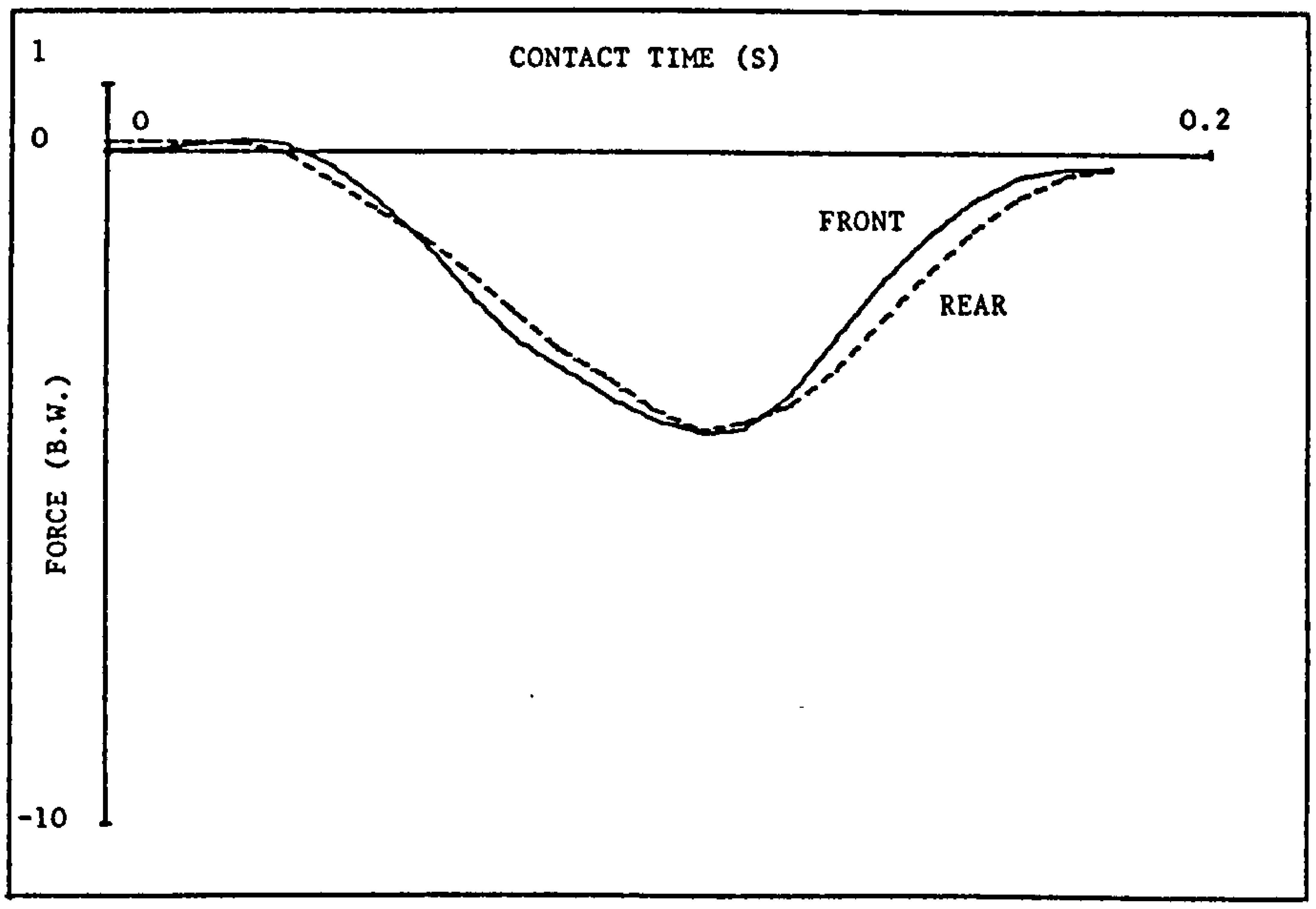


FIGURE 5.2.5. Ankle Joint Normal Reaction Mean Curves for Rear and Front Foot Strikers

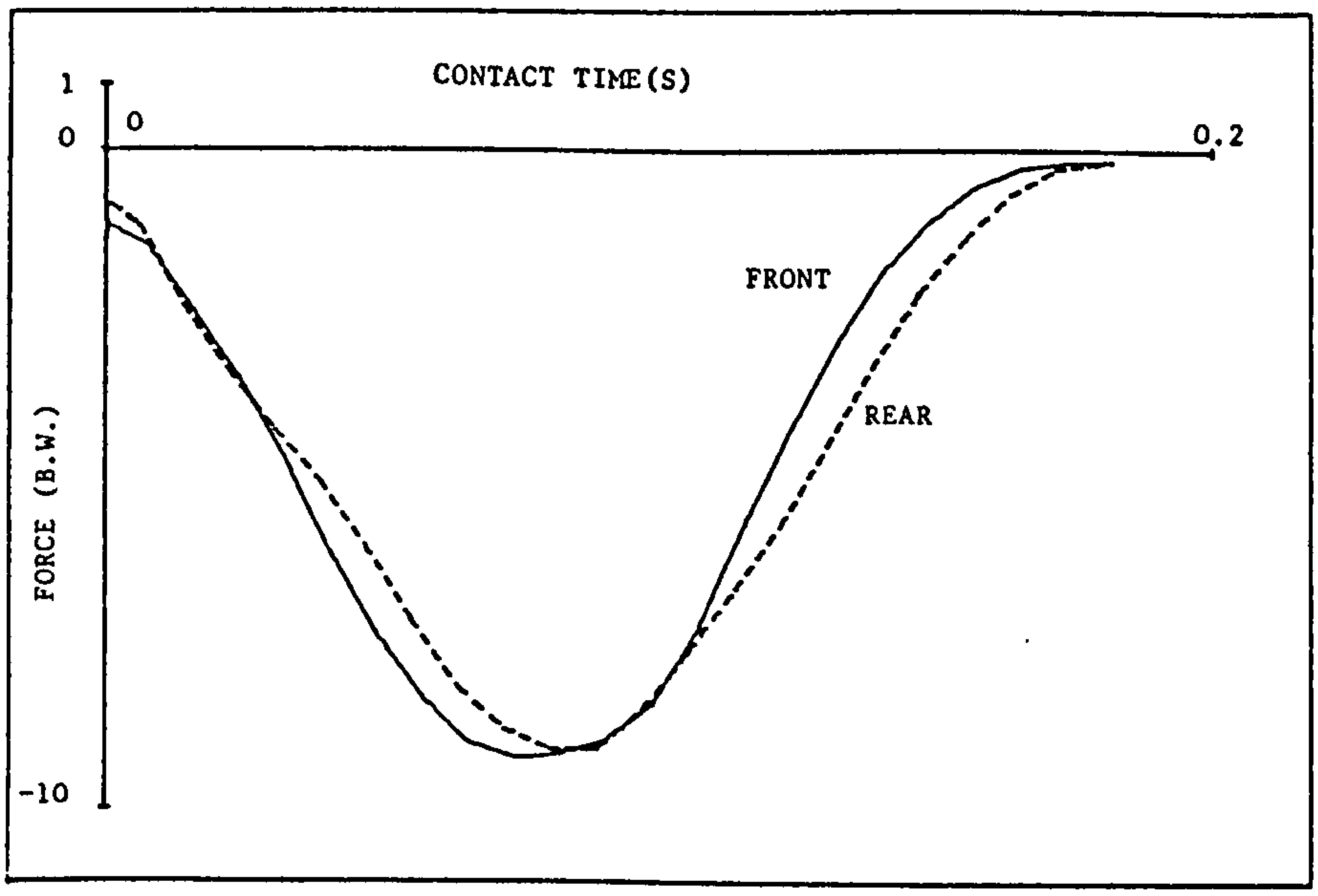


FIGURE 5.2.6. Knee Joint Shear Reaction Mean Curves for Rear and Front Foot Strikers

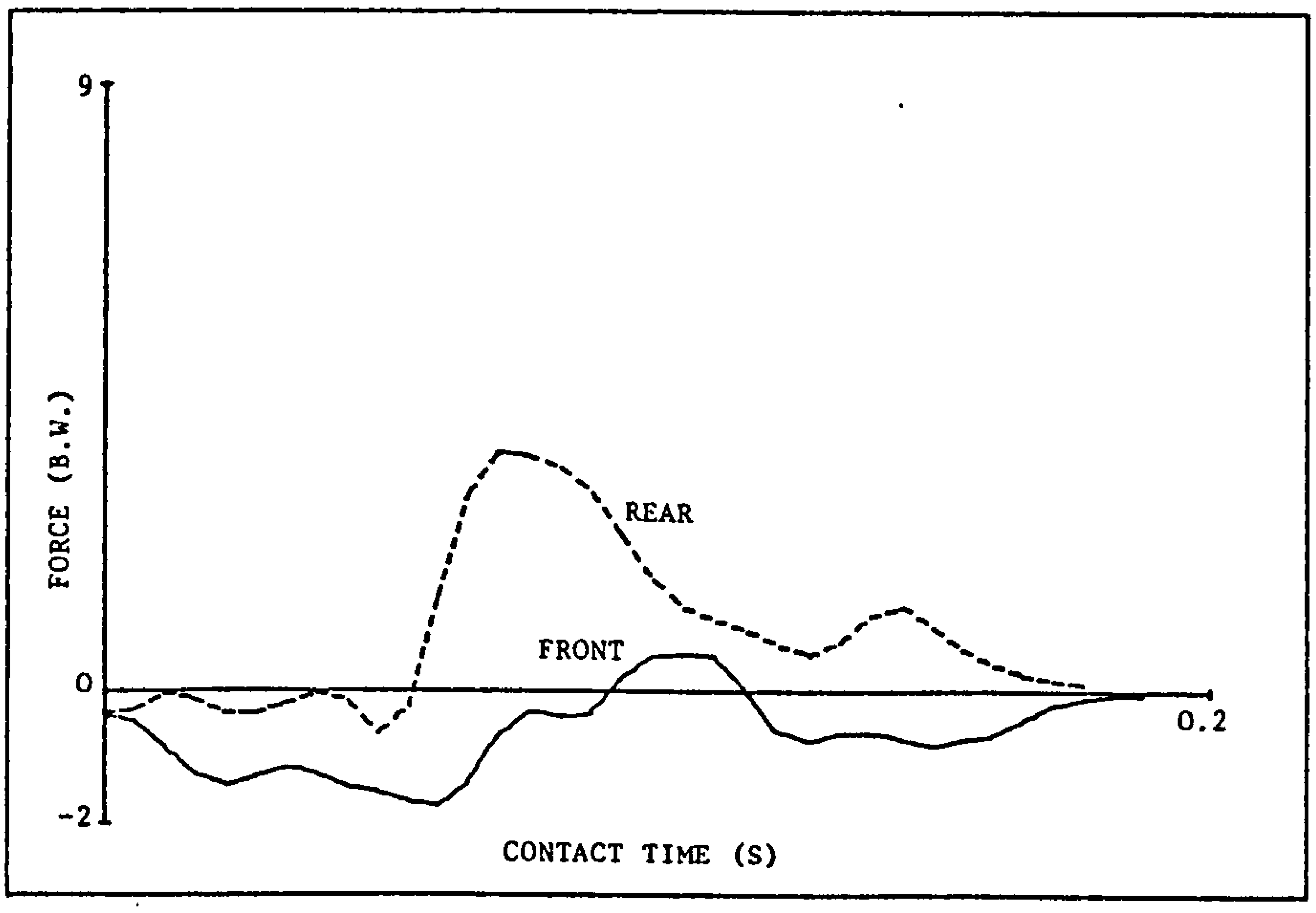
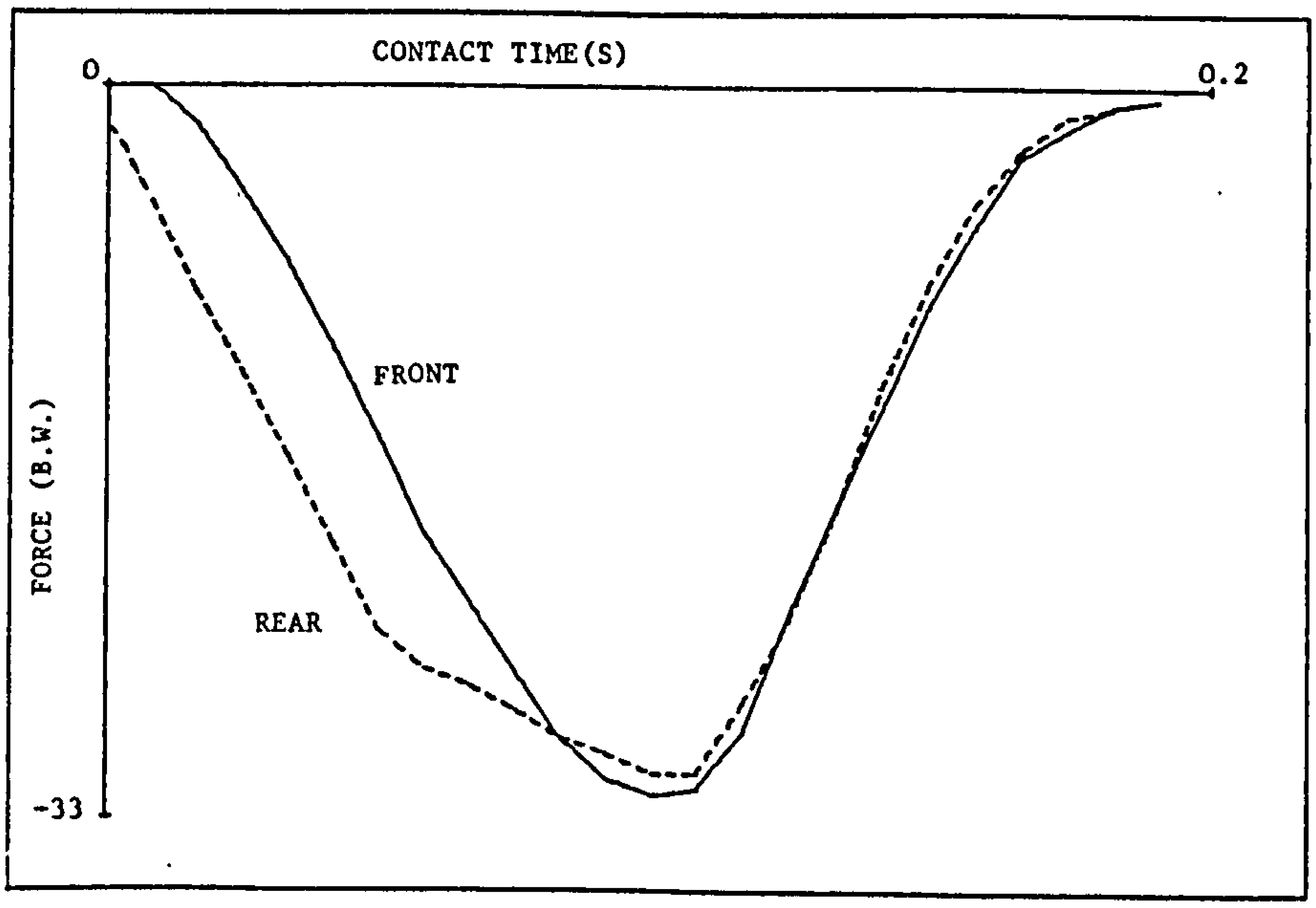


FIGURE 5.2.7. Knee Joint Normal Reaction Mean Curves for Rear and Front Foot Strikers



5.3 Phase Three: Analysis between Running Speeds

Three speeds representing a spectrum of middle distance racing and training velocities were adopted, these being, 5.36 ms^{-1} (five minutes per mile), 4.47 ms^{-1} (six minutes per mile) and 3.83 ms^{-1} (seven minutes per mile).

Five minutes per mile for 26.2 miles is near world record pace (equals 2 hours and 11 minutes) and gives a 31 minute 15 second ten kilometer time which is a respectable time for a good club runner.

Six minutes per mile is a typical middle distance training speed, and as outlined in Section 2.3 has become a standard biomechanical testing speed.

Seven minutes per mile for the full marathon distance would result in a finishing time of 3 hours and 3 minutes, a time achievable by most club marathon runners and a time bettered by 2,215 runners in the 1988 London Marathon (2 runners per second crossed the line during the finishing times of 2 hours 57 minutes and 3 hours.

Nine experienced subjects were employed as outlined in the summary presented in Tables 5.3.1 and 5.3.2.

TABLE 5.3.1 Phase Three Experimentation

Item	Data		
Speed ms^{-1}	5.36	4.47	3.88
Subjects	9	9	9
Runs per speed(1)	61	57	51
Model includes:			
SH, Shin	Yes	Yes	Yes
SO, Soleus	No	No	No

(1) Total runs for all subjects ie average of 6 per subject per speed

The results for the three speeds are presented graphically in Figures 5.3.1 to 5.3.5 and summarised in Tables 5.3.3 to 5.3.7. Graphical data is presented in newtons and tabular data in body weight (BW) units where appropriate.

Table 5.3.3 and Figure 5.3.1 show the joint moments during stance phase for the three speeds and as can be seen from the data there is little difference between the results at these speeds. The peak forces were not found to be significantly different with running speed and whilst the largest moment is reported in the ankle joint (plantar flexion) at -225 Nm at 5.36 ms^{-1} the largest hip moments, both flexion and extension, are reported at 185 Nm and -53 Nm for the slower speeds. Knee joint peak moments are completely unaffected by speed and can be seen to be -189 Nm for all velocities.

Graphically the hip can be seen to exhibit rapid flexion followed by extension before a final flexion prior to toe off. As speed of gait decreases this change over from flexion to extension take proportionally longer whilst the change back to flexion appears to remain constant in the stance phase cycle. The only significant differences found between speeds were recorded for 4.47 ms^{-1} and 5.35 ms^{-1} ($p < 0.02$).

The knee joint moment is somewhat smoother than the hip with a predominent extensor moment but representative flexor moment seen to be exhibited only for the two faster speeds. The maximum moment occurs later in the stance phase cycle as speeds decrease and would appear to be moving more in line with the ankle moment curve for the 3.83 ms^{-1} speed.

The ankle is the last of the joints to reach its peak, this peak corresponding in time to the peak propulsion ground reaction force. No significant differences were found with change in speed.

Mean peak muscle forces are displayed in Figures 5.3.2 and Table 5.3.4. The largest forces are reported in the quadricep muscle group (QU) at 18.2 BW whilst the gastrocnemius (GA) has a maximum value of 11.4 BW and the hamstrings (HM) 2.0 BW. The shin muscle (SH) group's activity is not clearly identified graphically but represents, at most, the first 0.016 seconds of the stance phase and has a maximum value of 0.44 BW. No significant differences were found between speeds for all peak muscle forces.

Graphically the quadricep force curves exhibit two steps during load application and one during load removal. These steps become less noticeable as speed decreases. Significant, however, is the reduction to almost zero after only approximately 65 to 70 percent of the stance phase. This compares favourably to the 50 to 60 percent activity recorded with EMG during model verification (see Section 3.5).

The hamstring muscle force, the lowest of all muscle groups in the model, exhibits an initial peak corresponding to the hip joint moment peak. Thereafter, apart from a secondary maximum at mid stance, possibly corresponding with the maximum knee extension moment, the hamstring force just fluctuates with a nominal positive value.

Apart from the faster speed, the gastrocnemius force curve could almost be described as a 'normal distribution' curve.

Figures 5.3.3 to 5.3.4 and Table 5.3.4 show joint reaction data for all speeds and maximum reactions are reported in the knee joint (normal, KN) at -31.6 BW (the negative representing a compressive sense) and the ankle joint (normal, AN) at -11.9 BW. The shear components for knee (KS) and ankle (AS) are a maximum of -2.1 and -7.2 respectively, here the negative represents the anterior-posterior sense. Whilst the general trend is a reduction in mean peak force with speed (KS is the only exception) the only significant differences were found for the ankle joint shear component, AS; p 0.001 for differences between 5.36 ms^{-1} and 3.83 ms^{-1} and p 0.10 for differences between 5.36 ms^{-1} and 4.47 ms^{-1}.

Graphically, as shown in Figures 5.3.3, AN exhibits a small step corresponding to the ground reaction force impact peak, and apart from the fastest speed also exhibits a small step during force reduction. The major peak is at mid stance and corresponding to the ground reaction propulsion peak. The AS force trace is posterio-anterior in sense for approximately the first 15% of the stance phase which would correspond to the braking action at heel strike.

The knee joint shear force fluctuates about the zero value with two noticeable peaks in the first third of the cycle. As speed decreases the curve shifts more negative such that the positive peak present at 5.36 ms^{-1} becomes negative.

Very little differences are detectable for the knee joint normal force (Figure 5.3.4a) with speed except for a stretching out of the cycle over the longer time base. Two steps in the curve are present, the first corresponding to the maximum foot/ground impact peak and the second during reaction reduction.

The patella-femoral contact force (PF) is reprsented in Figure 5.3.4b and as can be seen very little difference can be seen in the curves form. Two major, almost equal peaks exist, the second being the greater of the two, at -5.0 BW for the faster speeds. The first peak could correspond to the foot/ground impact peak and the second to the propulsion peak.

As can be seen from Figure 5.3.5 and Table 5.3.5 foot ground reaction force curves and peak values are similar for all speeds showing relatively high subject repeatability. The only significant differences found between all speeds were in the foot contact times (CT) and the horizontal ground reaction force (RY). Here contact time decreased from 0.225 seconds to 0.191 seconds for the overall increase in speed. All other foot/ground data reduced with speed but not significantly except for the peak propulsion vertical reaction force which did decrease significantly ($p < 0.10$) from 3.04 BW to 2.84 BW over the extremes of speed. Impact peaks and impact peak loading rates were highest for the 5.36 ms^{-1} speed at 2.66 BW and106 $kN.s^{-1}$ respectively.

Table 5.3.6 displays muscular loading rates for the tricep surea (TS) and quadricep (QU) groups occurring at heel strike. Maximum muscular loading rates were reported for the faster speed, 129 $kN.s^{-1}$ for TS and 340 $kN.s^{-1}$ for QU. These loading rates did not reduce significantly with speed, except at the two extremes, to 109 $kN.s^{-1}$ and 299 $kN.s^{-1}$ respectively.

Kinematic data is presented in Table 5.3.7. Here the position of the lower limbs at heel strike are represented by the amount of knee flexion and the angle of the shank, maximum knee flexion is also recorded. Significant differences were found between speeds for a number of aspects and generally as speed increases the amount of knee flexion at heel strike increases as does the shank angle. Maximum knee flexion is unaffected at between 38 and 40 degrees.

TABLE 5.3.2 Subject Information

	Age(years)	Height(m)	Weight(kg)
Mean	23.2	1.76	69.3
Standard Deviation	3.5	0.02	6.1
Range	20-28	1.72-1.80	60.6-81.4

TABLE 5.3.3 Mean Peak Joint Moments in Nm

Speed(ms^{-1})	Ankle	Knee	Hip Max	Hip Min
5.36	-225	-189	177	-46
4.47	-214	-189	150	-53
3.83	-218	-189	185	-48

TABLE 5.3.4 Mean Peak Muscle and Joint Forces in Body Weight (BW) units

Speed(ms^{-1})	Muscle Forces				Joint Reactions					Shin Activity Time(s)
	SH	GA	QU	HM	AN	AS	KN	KS	PF	
5.36	0.42	11.4	18.0	2.0	-11.9	-7.2	-31.6	-1.9	-5.0	0.013
4.47	0.40	10.7	18.2	1.8	-11.5	-6.4	-31.0	-2.0	-5.0	0.016
3.83	0.44	10.7	17.3	1.8	-11.6	-6.1	-30.3	-2.1	-4.8	0.015

TABLE 5.3.5 Foot/Ground Reaction Data

Speed (ms^{-1})	Peak Ground Reaction Forces(1) Vertical		Horizontal(2)	Impact Peak Loading Rate(1) ($kN.s^{-1}$)	Foot Contact Time (s)
	Impact	Propulsion	Propulsion		
5.36	2.66	3.04	-0.62	106	0.191
4.47	2.36	2.92	-0.55	89	0.210
3.83	2.16	2.84	-0.50	82	0.225

(1) Expressed in body weight units

(2) Negative sign indicates anterio-posterior sense

TABLE 5.3.6 Muscular Loading

Speed (ms^{-1})	Tricep Surea Group ($kN.s^{-1}$)	Quadricep Group ($kN.s^{-1}$)
5.36	129	340
4.47	116	333
3.83	109	299

TABLE 5.3.7 Kinematic Summary; Mean Leg Angles in Degrees, 180° Equals Full Extension

Speed	Knee Flexion(1) Angle at Heel Strike	Maximum Knee(1) Flexion Angle	Lower Leg at(2) Heel Strike
5.36	167	140	29
4.47	171	141	26
3.83	171	142	25

(1) Hence maximum knee flexion equals 38°

(2) Measured to the vertical.

TABLE 5.3.8 Summary of Features Found Significantly Different Between Speeds

See Tables 5.3.3 to 5.3.7 for individual data.

Speeds Compared ($m.s^{-1}$)	ITEM[1]								
	Joints	Moments	Loading Rates		Ground Reaction Data			Limb Angles[2]	
	AN	Hip	QU	TS	RZ	RY	CT	Knee	Lower Leg
	Significance Level ($p <$)								
3.83 to 5.36	0.001	-	0.05	0.10	0.10	0.001	0.001	-	0.001
for 5.36 ms^{-1}	Higher		Higher	Higher	Higher	Higher	Shorter		Greater[2]
3.83 to 4.47	-	-	-	-	-	0.10	0.001	-	-
for 4.47 ms^{-1}					Higher	Shorter			
4.47 to 5.36	0.10	0.02	-	-	-	0.10	0.002	0.05	0.001
for 5.36 ms^{-1}	Higher	Higher			Higher	Shorter	Greater[2]		Greater[2]

(1) See Nomenclature

(2) Limb flexion at heel strike

FIGURE 5.3.1. Net Joint Moments for Three Speeds

a) 5.36 m s^{-1}, b) 4.47 m s^{-1}, c) 3.83 m s^{-1}

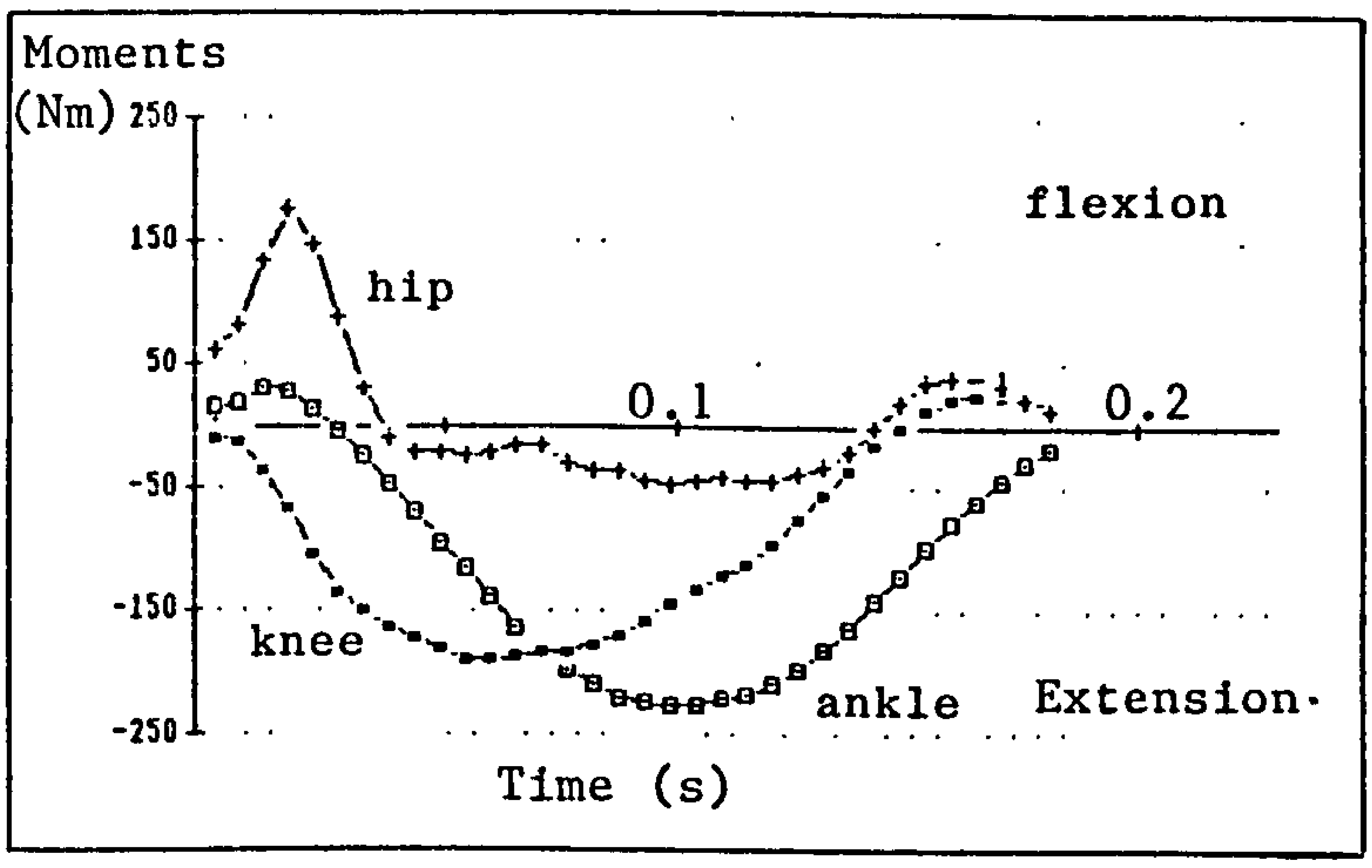


Fig. 5.3.1a

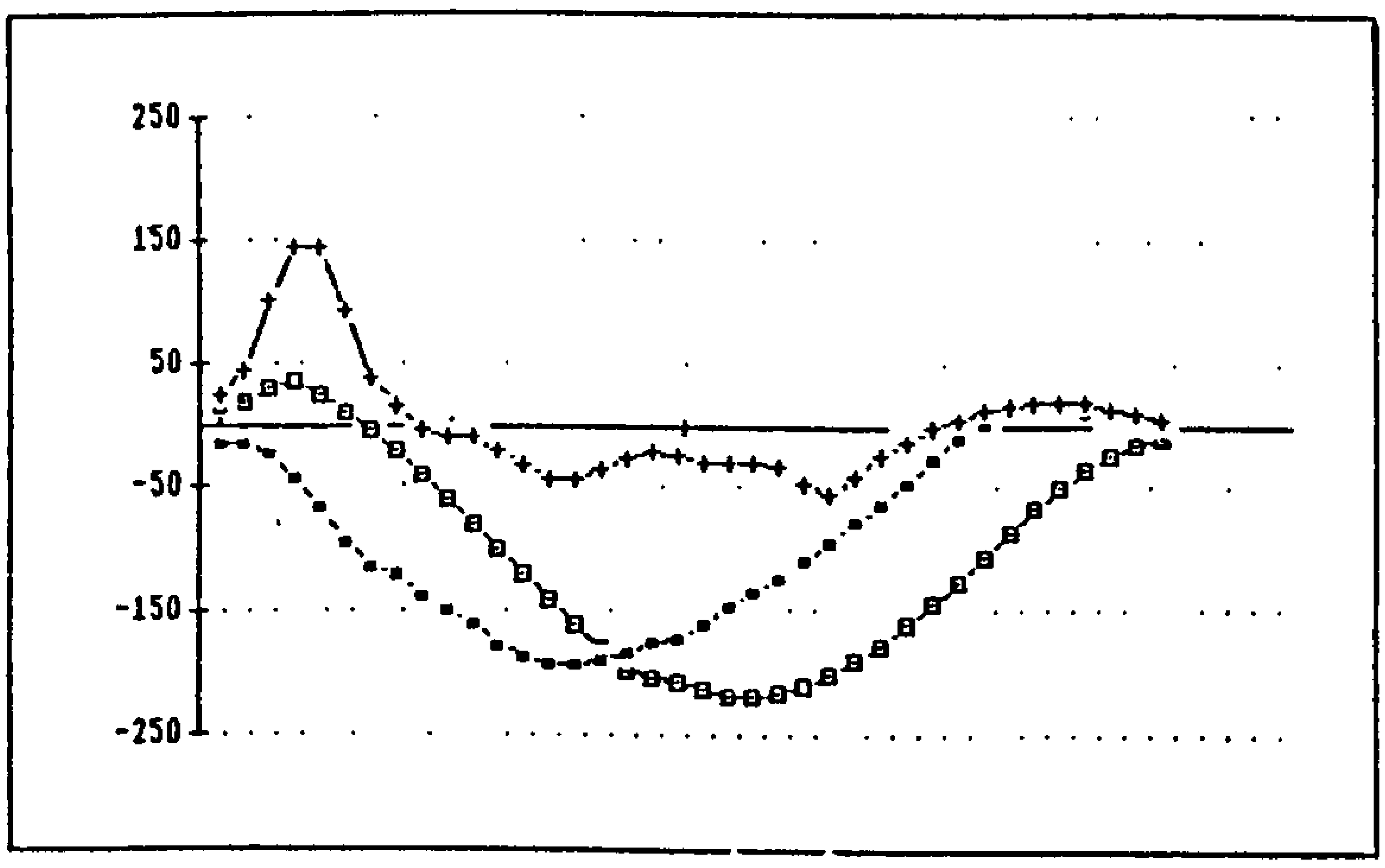


Fig. 5.3.1b

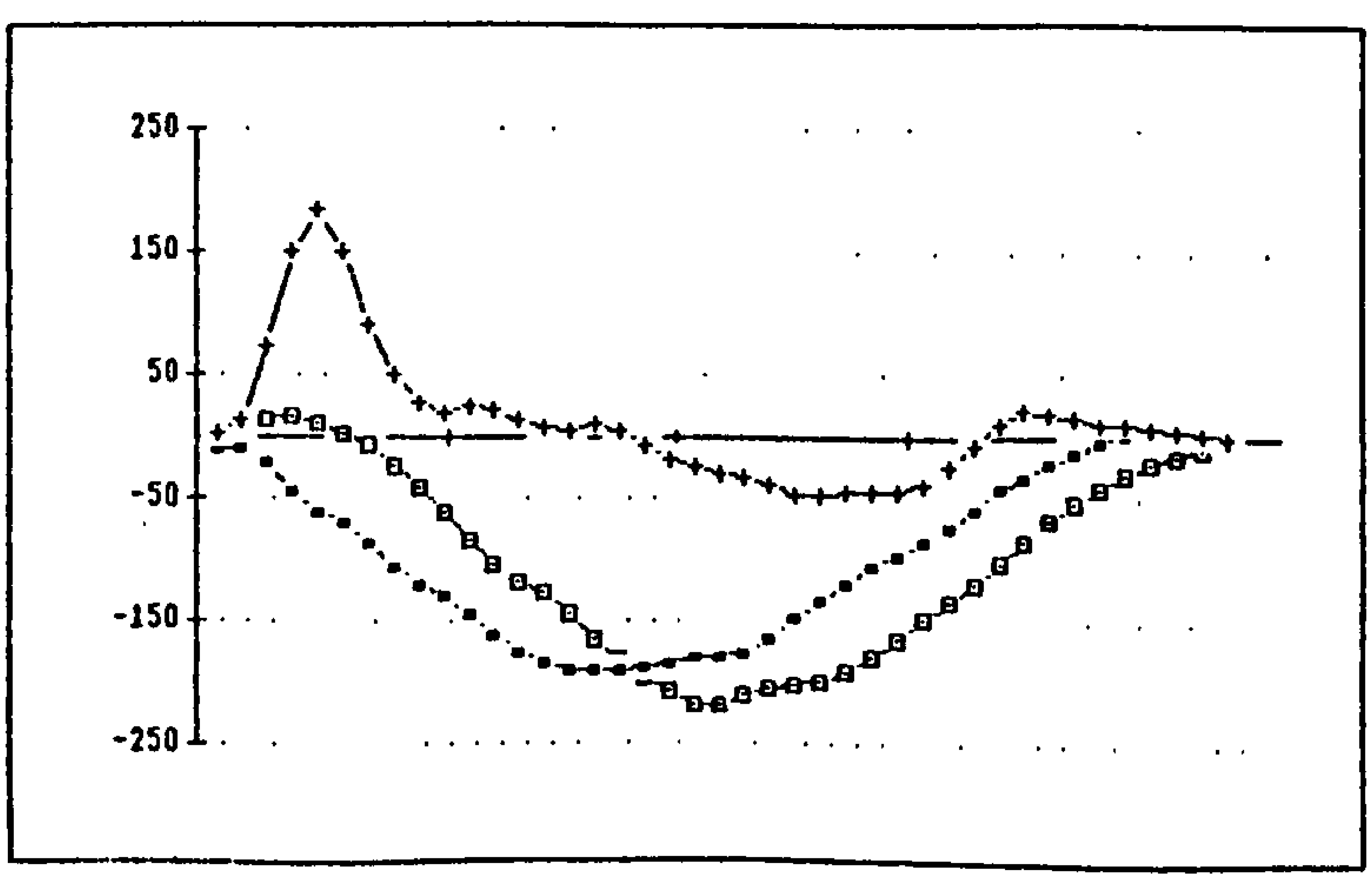


Fig. 5.3.1c

FIGURE 5.3.2. Gastrocnemius, Quadricep and Hamstring Muscle Forces for Three Speeds

a) 5.36 m s^{-1}, b) 4.47 m s^{-1}, c) 3.83 m s^{-1}

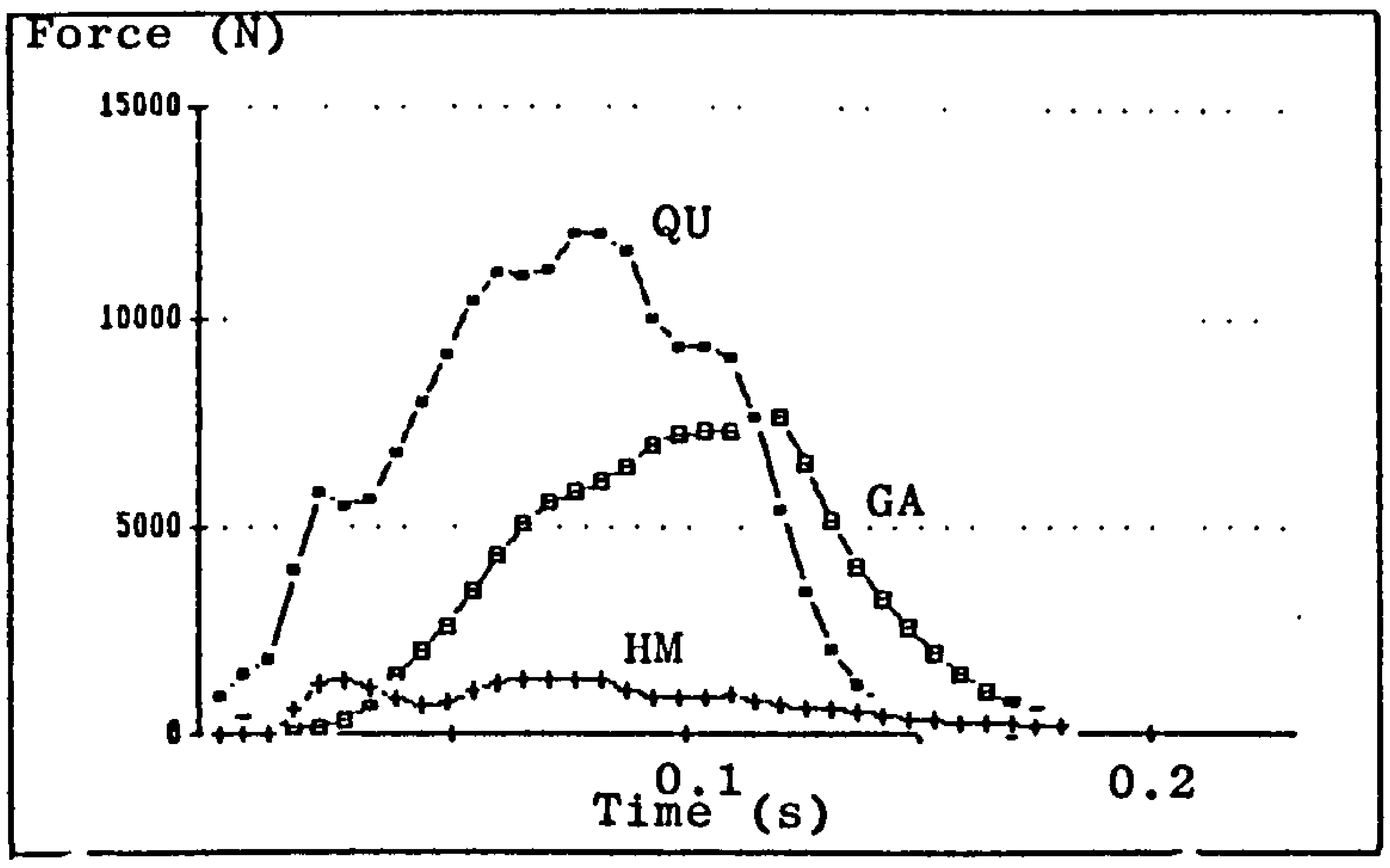


Fig. 5.3.2a

Fig. 5.3.2b

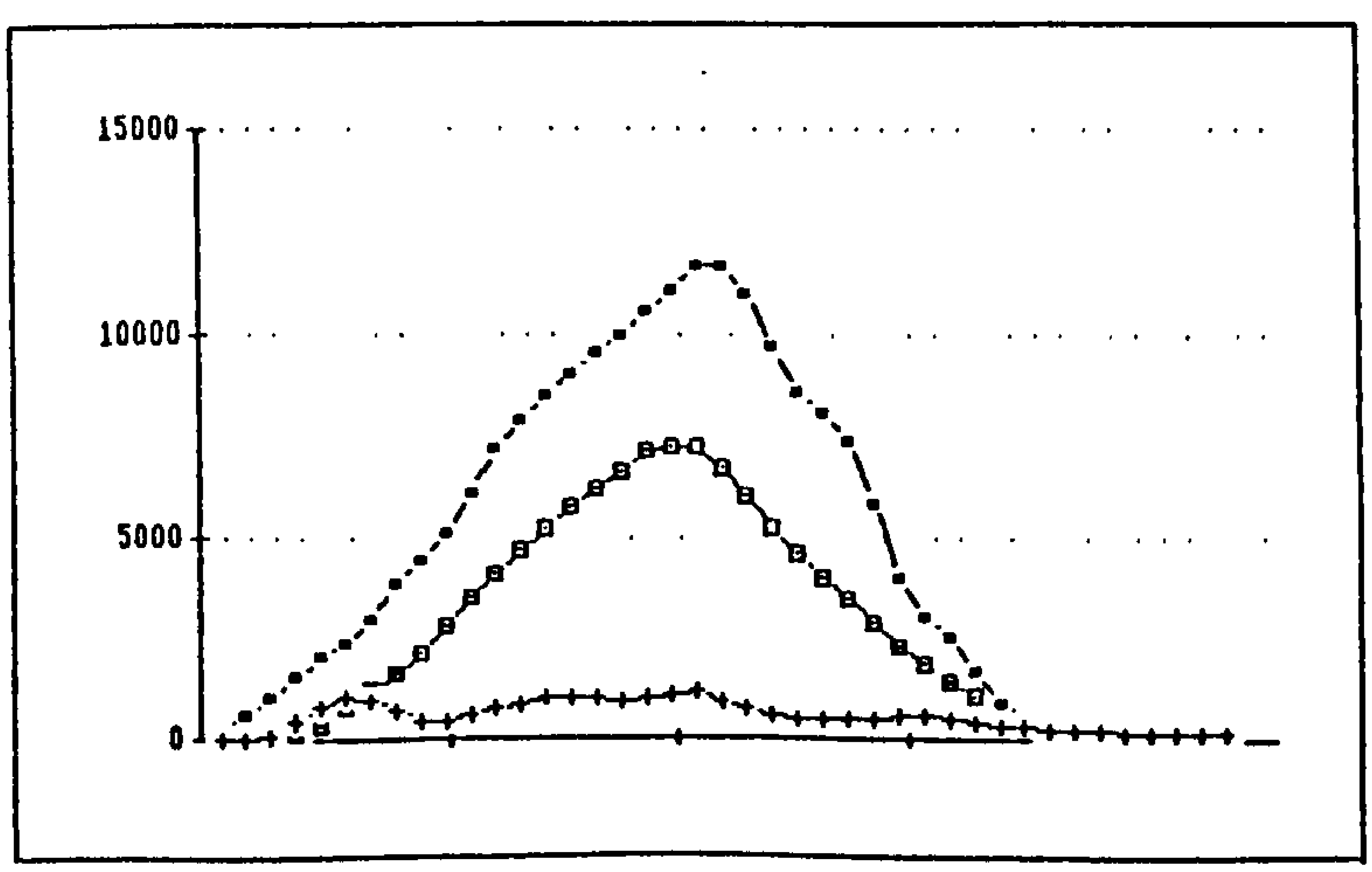


Fig. 5.3.2c

FIGURE 5.3.3. Ankle Joint, Shear & Normal, and Knee Joint Shear Reactions for Three Speeds

a) 5.36 m s^{-1}, b) 4.47 m s^{-1}, c) 3.83 m s^{-1}

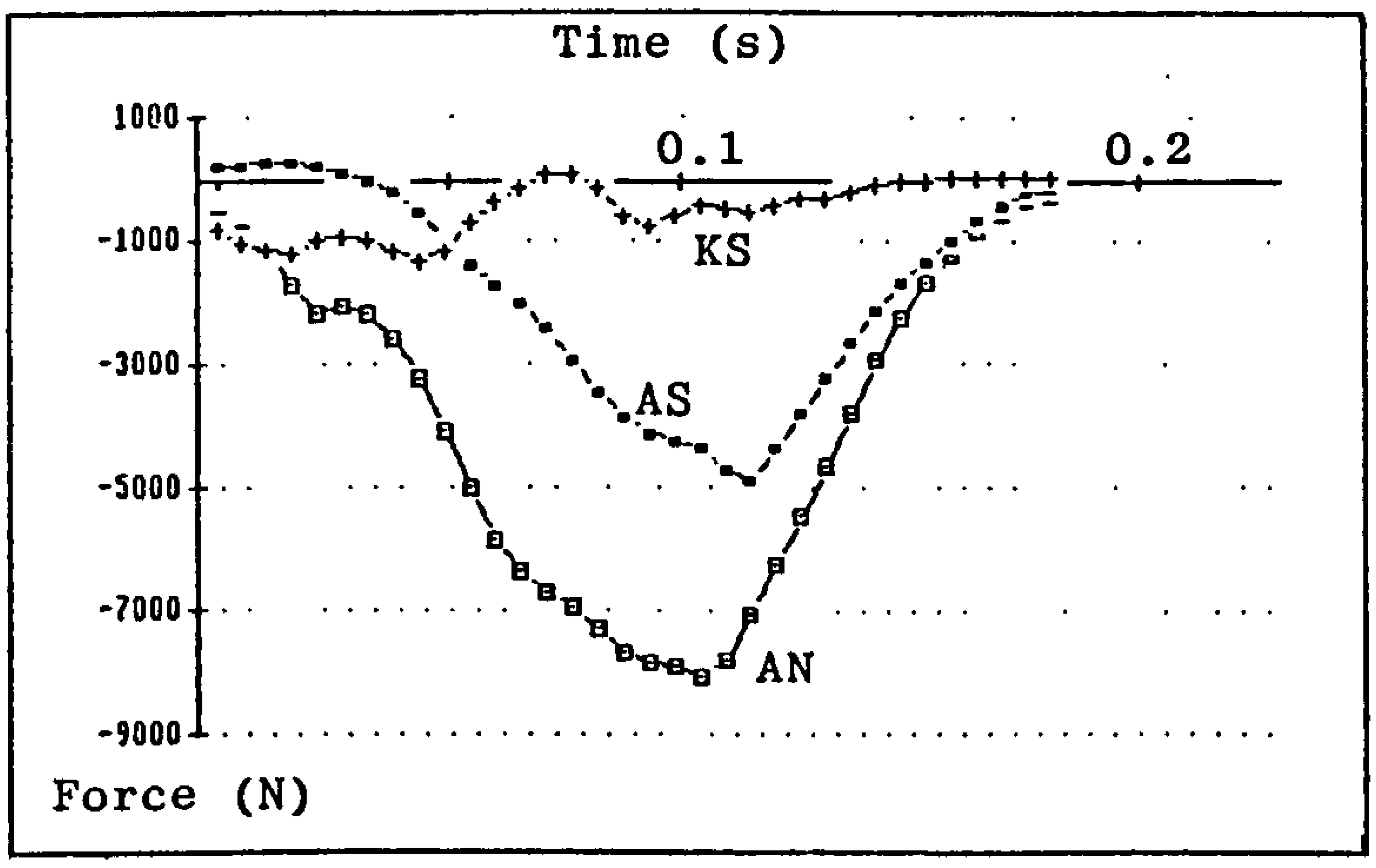


Fig. 5.3.3a

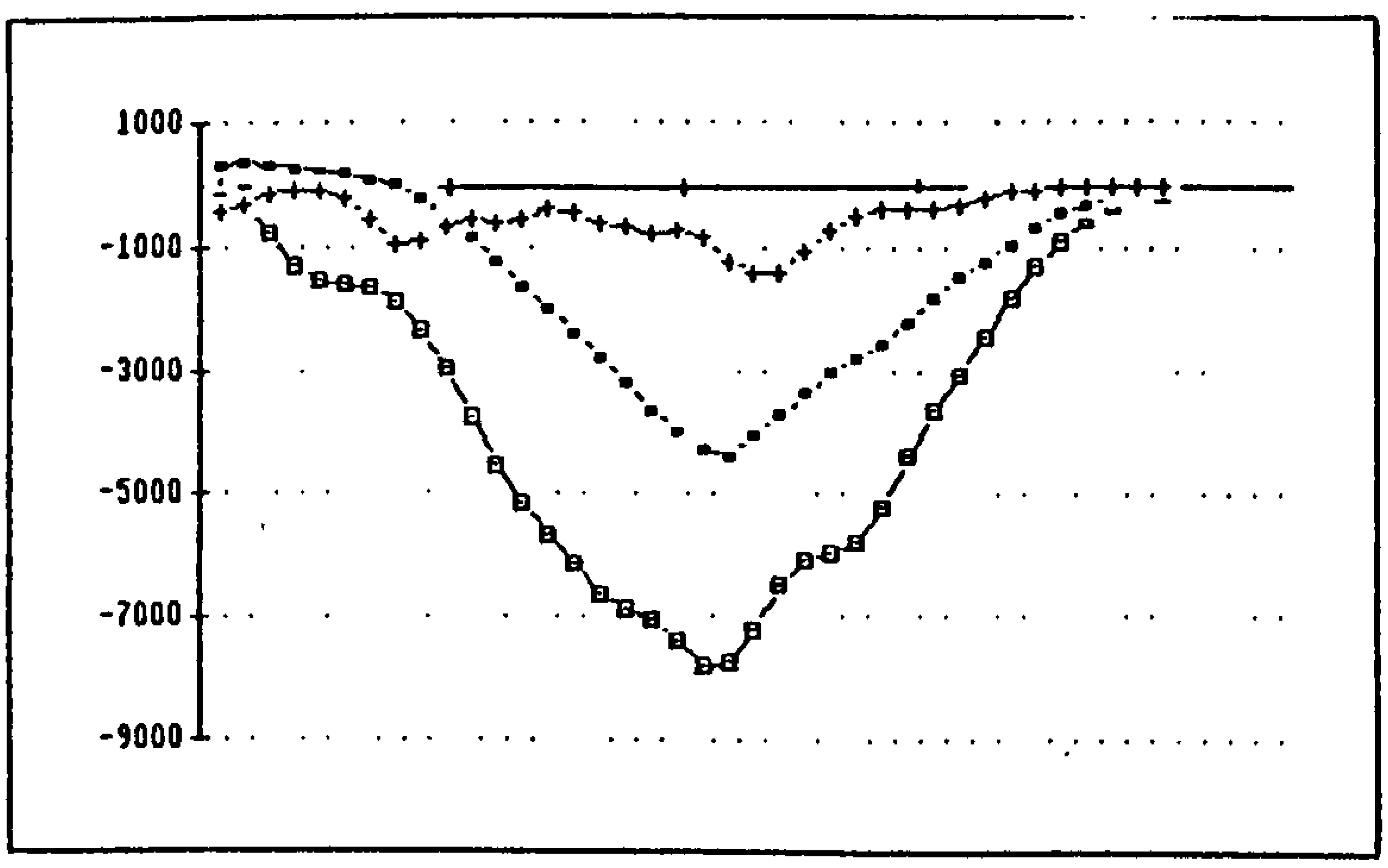


Fig. 5.3.3b

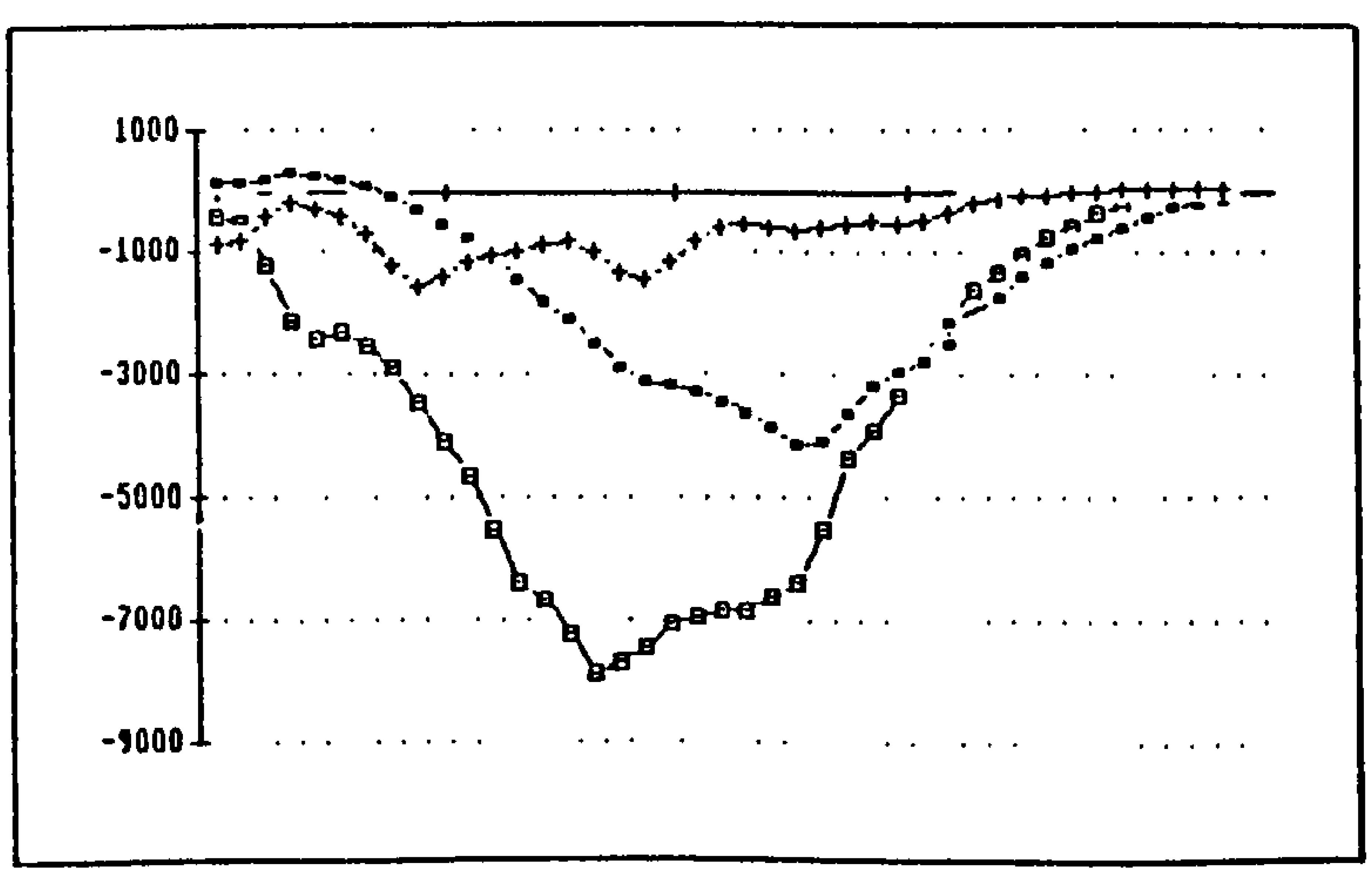


Fig. 5.3.3c

FIGURE 5.3.4. (a) Knee Joint, Normal and (b) Patella-Femoral Reactions for Three Speeds

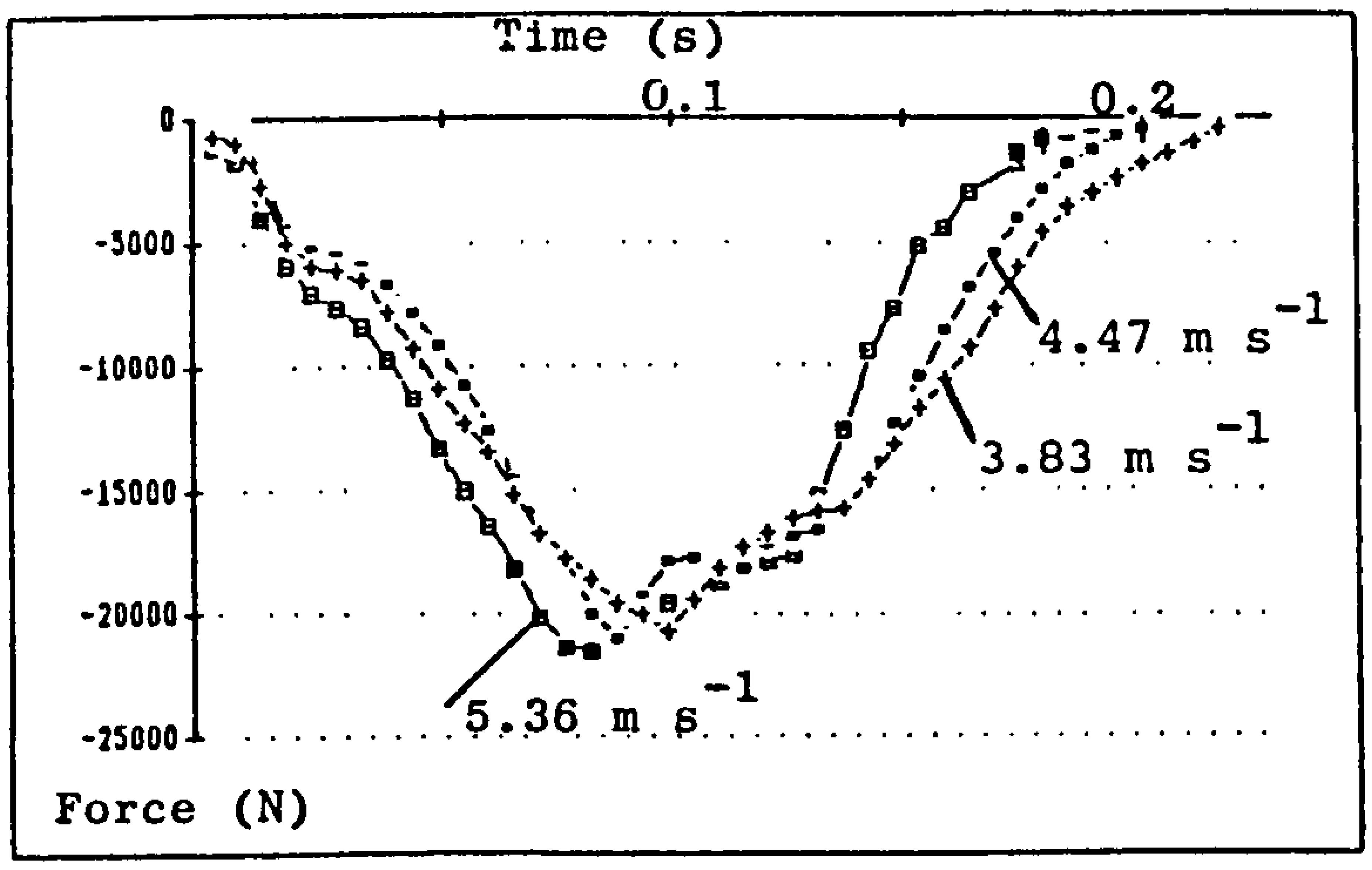


Fig. 5.3.4a

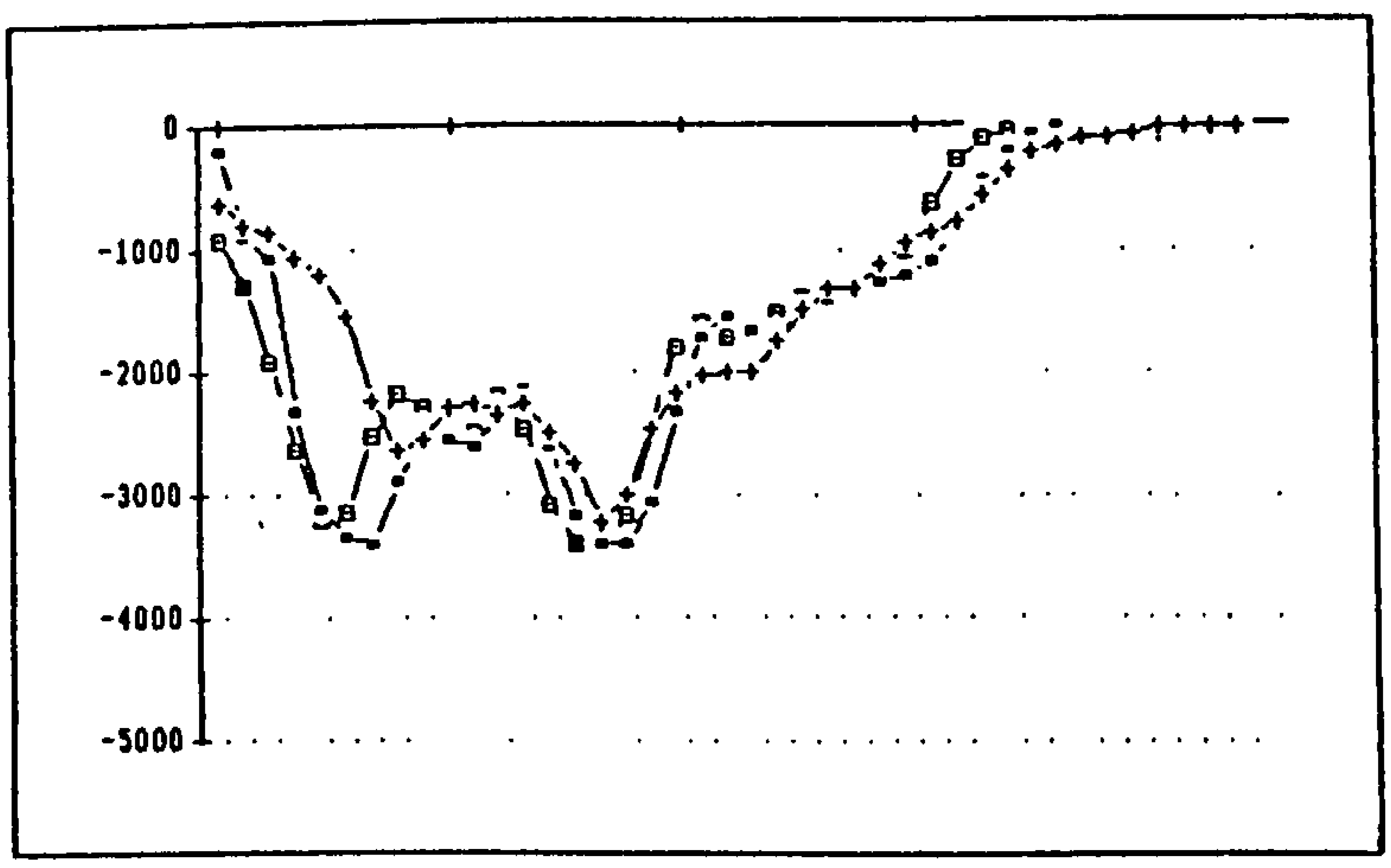
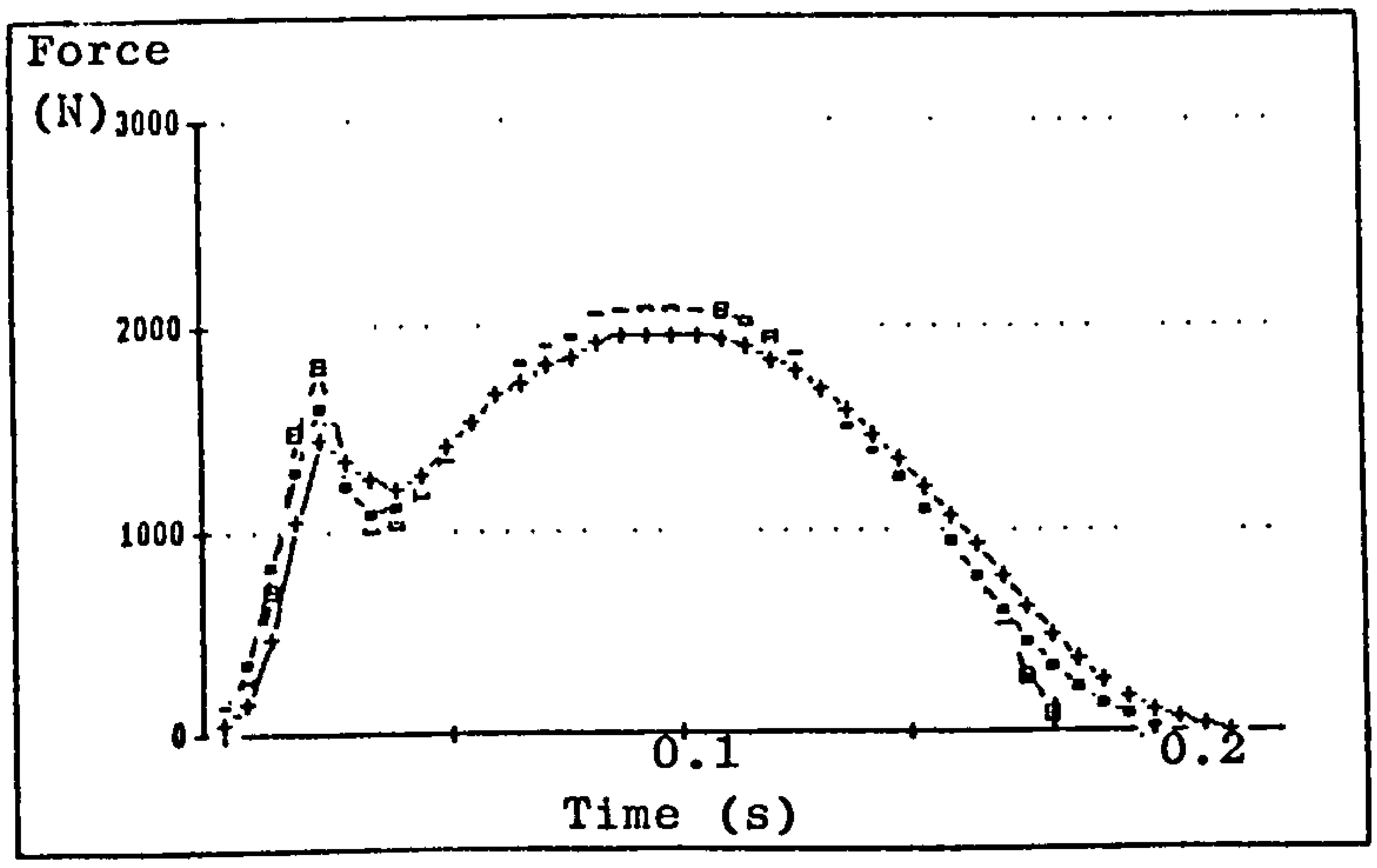


Fig. 5.3.4b

FIGURE 5.3.5. Vertical Ground Reaction Force Curves for Three Speeds (see Figure 5.3.6 for legend)



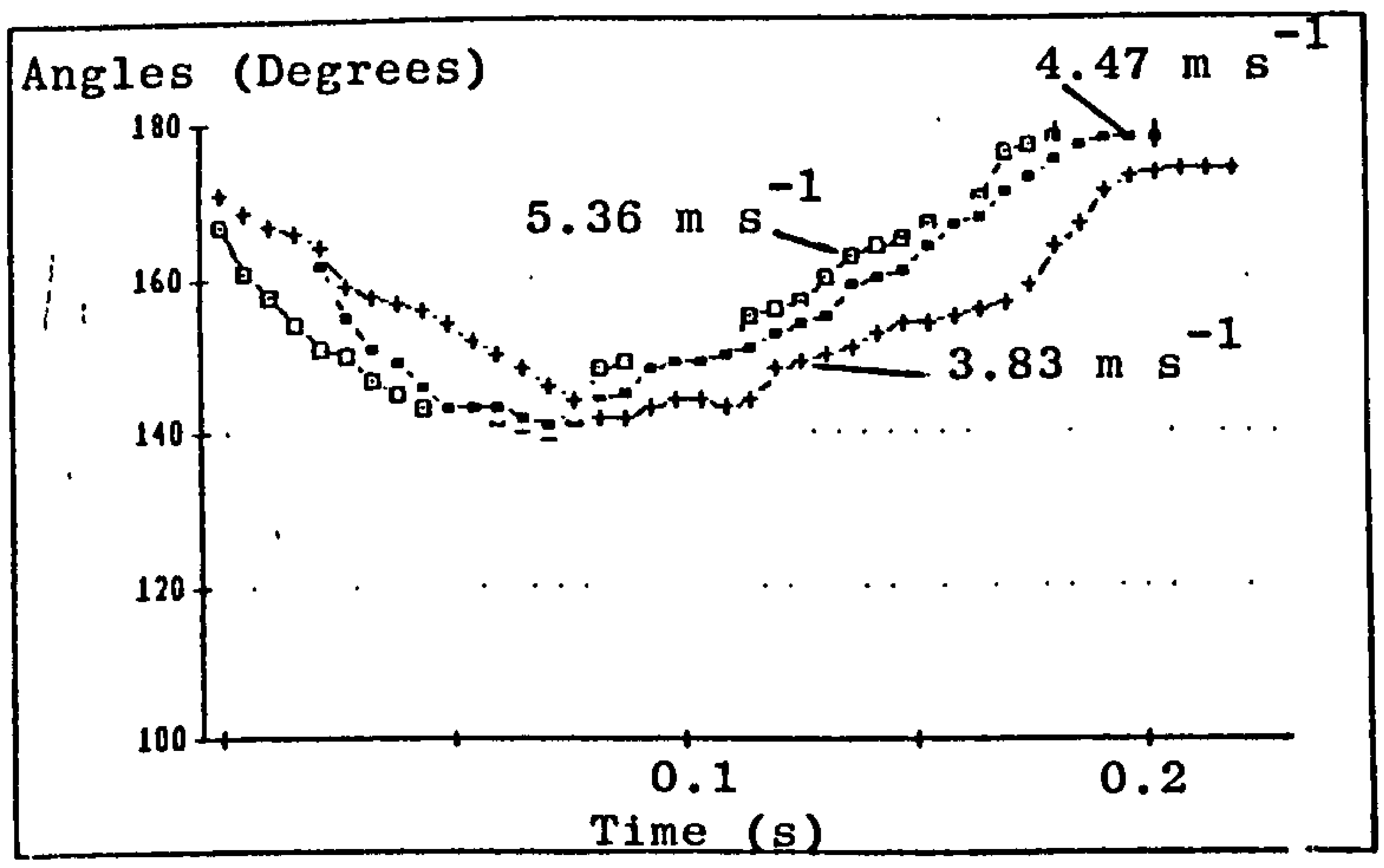


FIGURE 5.3.6 Knee Joint Angles During Foot Strike for Three Speeds

5.4 Phase Four: Model Enhancement

Using the subjects and testing regime outlined in Section 5.3 (see Table 5.3.2) the model was manipulated to include the soleus muscle. As the input data to the model was unaltered, ground reaction forces, kinematics and joint moments were unaffected and only data for muscle and joint forces and muscular loading rates is presented.

Table 5.4.1 Phase Four Experimentation

Item	Data		
Speed ($m.s^{-1}$)	5.36	4.47	3.83
Subjects	9	9	9
Runs per Speed(1)	30	57	25
Model Includes:			
SH, Shin	Yes	Yes	Yes
SO, Soleus	Yes	Yes	Yes

(1) Total runs for all subjects (Data reanalysed from Phase Three)

Comparison between speeds was not possible here due to the smaller sample sizes, and large inter- and intra-subject variences obtained (examples of these variences is shown in Table 5.4.2.). Similarly comparisons between models was not attempted for these two speeds due to the unequal sample sizes represented.

Figures 5.4.1 to 5.4.2 show the differences in results obtained for the enhanced model as against the data presented in Section 5.3 (referenced as the Normal Model). All results are for 4.47 ms^{-1} and only those values affected by the enhancement are displayed. Thus data from AN, AS joint reactions, foot/ground reactions, kinematics and joint moments are not tabulated.

As can be seen from figure 5.4.1a the gastrocnemius force has been halved but a slightly less smooth curve obtained. The curve also represents the soleus muscle force such that peak values of 5.35 BW were recorded for both instead of 10.7 BW previously solely attributed to the gastrocnemius. Tests for significance were not performed here.

The quadricep peak muscle force was reduced from 18.2 BW to 15.0 BW ($p < 0.001$) for the new model with a smoother force profile as displayed in figure 5.4.1b. Conversely the peak hamstring muscle force was increased with the inclusion of the soleus into the model, 1.8 BW to 2.8 BW ($p < 0.05$). This was caused by a much more severe initial loading peak occurring at heel strike (see Figure 5.4.1c). Thereafter the force level decreased almost to zero before increasing to a plateau prior to toe off. This later part of the force curve is almost a mirror image of the previously obtained data but is still relatively low, less than 1000 N, when compared to all other muscle forces.

For the knee joint reactions (Figure 5.4.2) the normal aspect KN is reduced from -31.0 BW to -21.3 BW ($p < 0.0005$) with a smoother force profile but the shear aspect KS is generally unaffected (-2.0 BW to -2.4 BW but non significant) except for a longer peak plateau around mid stance. The patella-femoral contact force is reduced with the inclusion of the soleus with the model, -5.0 BW to -4.1 BW ($p < 0.002$), the force traces are however very similar.

TABLE 5.4.2 Intra- & Inter- Subject Variability (at 4.47 ms^{-1})

Subject Number	Statistical Test	Muscle Force (B.W.)				Joint Reaction (B.W.)			
		SH	GA	QU	HM	AN	AS	KN	KS
1. (n=7)	Mean	0.05	9.8	18.2	1.7	-10.1	-6.2	-29.2	-2.0
	S.D.(±)	0.09	1.49	0.58	0.48	1.15	0.99	2.25	0.62
2. (n=8)	Mean	1.2	9.4	18.8	2.1	-10.3	-6.3	-31.2	-1.3
	S..D.(±)	0.35	0.53	0.94	0.33	0.42	0.56	1.16	0.20
3. (n=8)	Mean	0.4	11.9	14.3	1.3	-12.2	-7.0	-27.8	-2.2
	S.D.(±)	0.28	0.87	0.22	0.16	0.60	0.72	1.51	1.18
4. (n=7)	Mean	1.0	7.9	16.5	1.4	-9.2	-4.8	-26.1	-3.4
	S.D.(±)	0.29	1.37	1.65	0.42	1.15	0.83	2.93	0.97
Overall	Grand Mean	0.6	9.8	16.9	1.6	-10.5	-6.1	-29.0	-2.2
	Grand S.D.(±)	0.54	1.80	1.97	0.45	1.39	1.10	2.94	1.07

n = number of runs per subject

TABLE 5.4.3 Mean Peak Muscle and Joint Forces in Body Weight (BW) Units at 4.47 ms^{-1}

Model	Muscle Forces					Joint Forces					Shin Activity Time
SH	SH	GA	SO	QU	HA	AN	AS	KN	KS	PF	(S)
Without Soleus	0.40	10.7	-	18.2	1.8	-11.5	-6.4	-31.0	-2.0	-5.0	0.016
With Soleus	0.51	5.35	5.35	15.0	2.8	-11.5	-6.4	-21.3	-2.4	-4.1	0.015
Significant differences $p <$	N/S	-	-	0.001	0.05	N/S	N/S	0.0005	N/S	0.002	N/S

TABLE 5.4.4 Muscle Loading Rates in Body Weight Units

Model	Tricep S Group ($kN.s^{-1}$)	Quadricep Group ($kN.s^{-1}$)
Without Soleus	116	333
With Soleus	116	292
$p <$	-	0.10

FIGURE 5.4.1. a) Gastrocnemius, b) Quadricep and c) Hamstring Force Curves for Normal and Enhanced Model

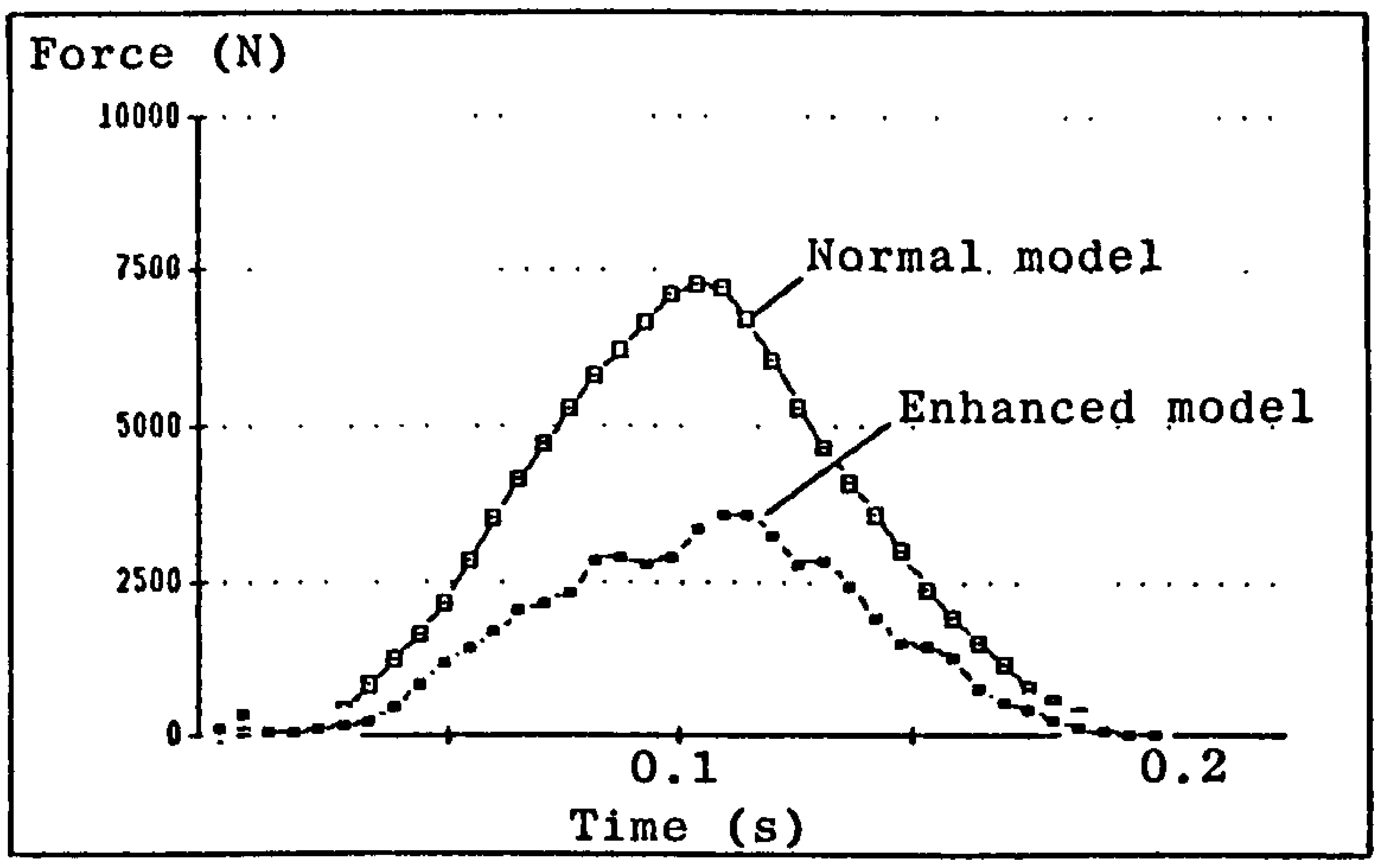


Fig. 5.4.1a

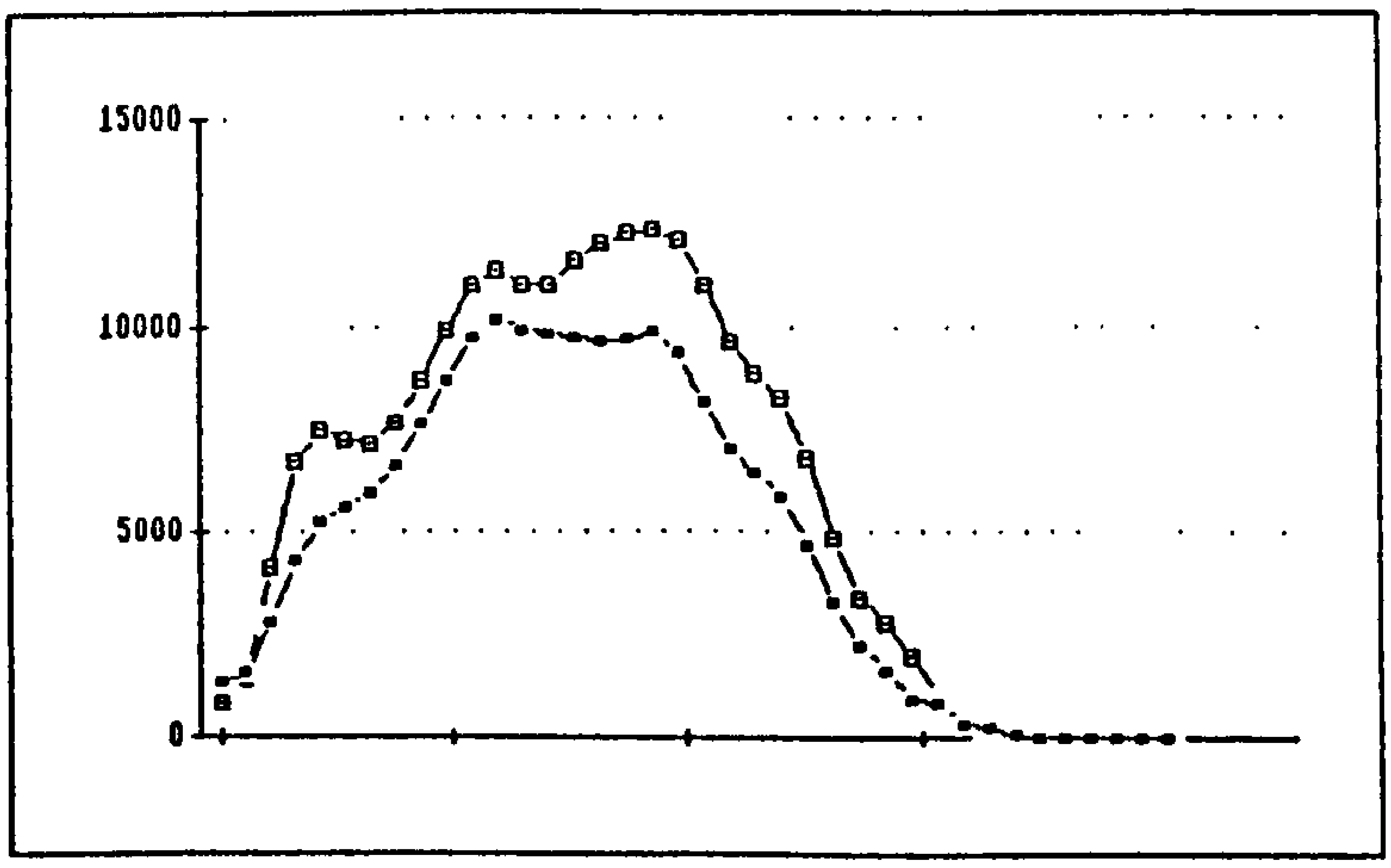


Fig. 5.4.1b

Fig. 5.4.1c

FIGURE 5.4.2. Knee Reaction Curves for Normal and Enhanced Model

a) kN, b) KS & c) PF

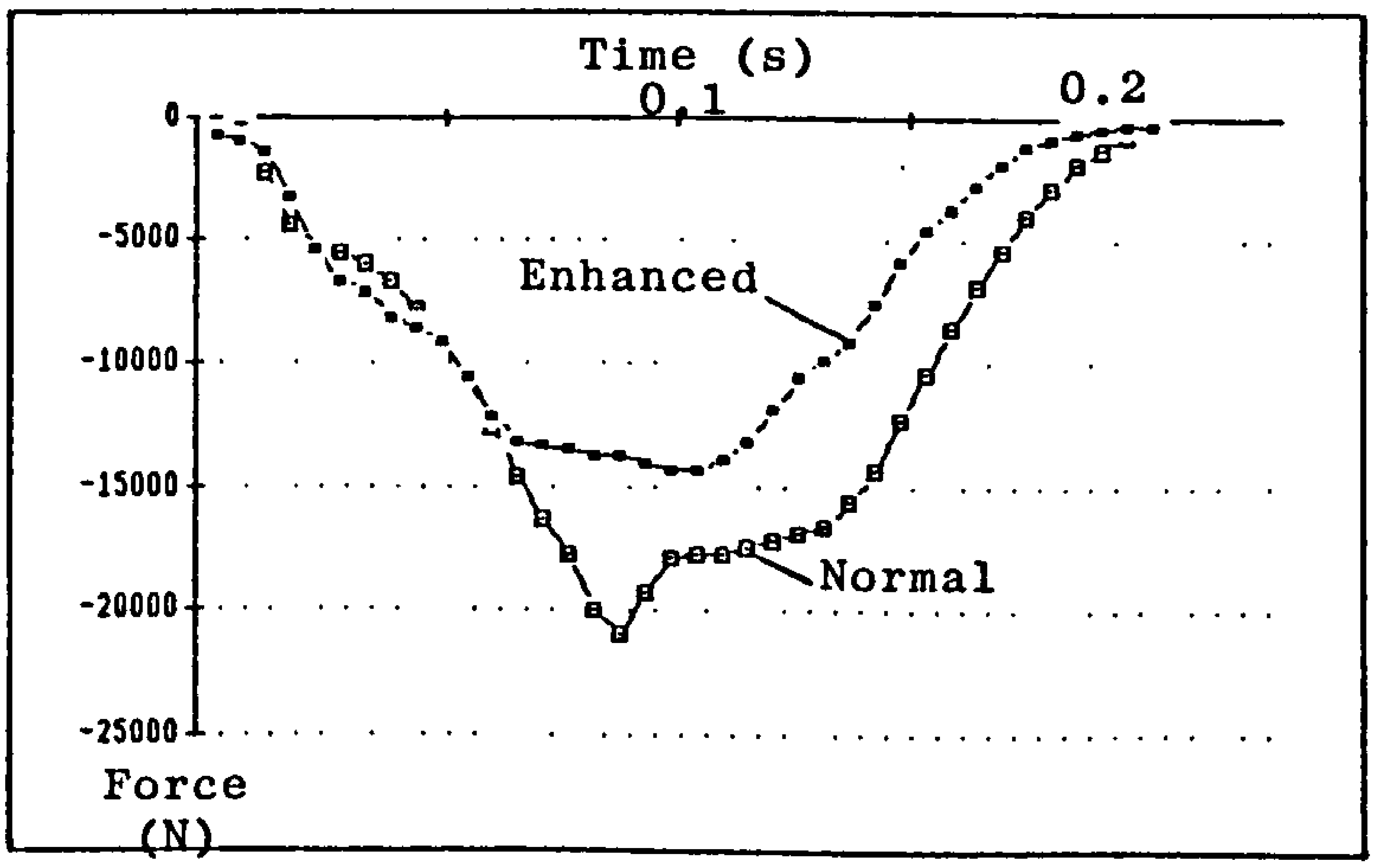


Fig. 5.4.2a

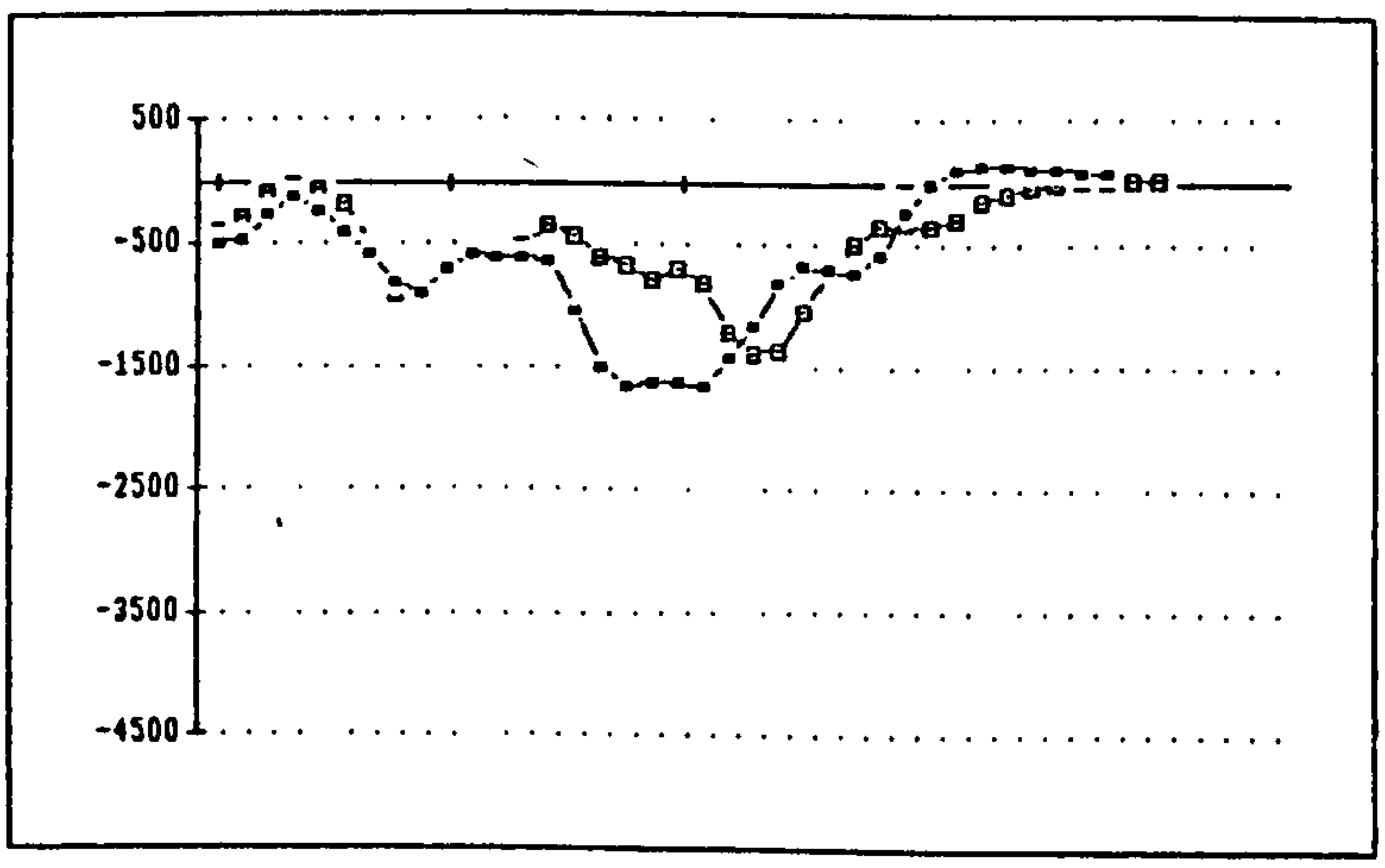


Fig. 5.4.2b

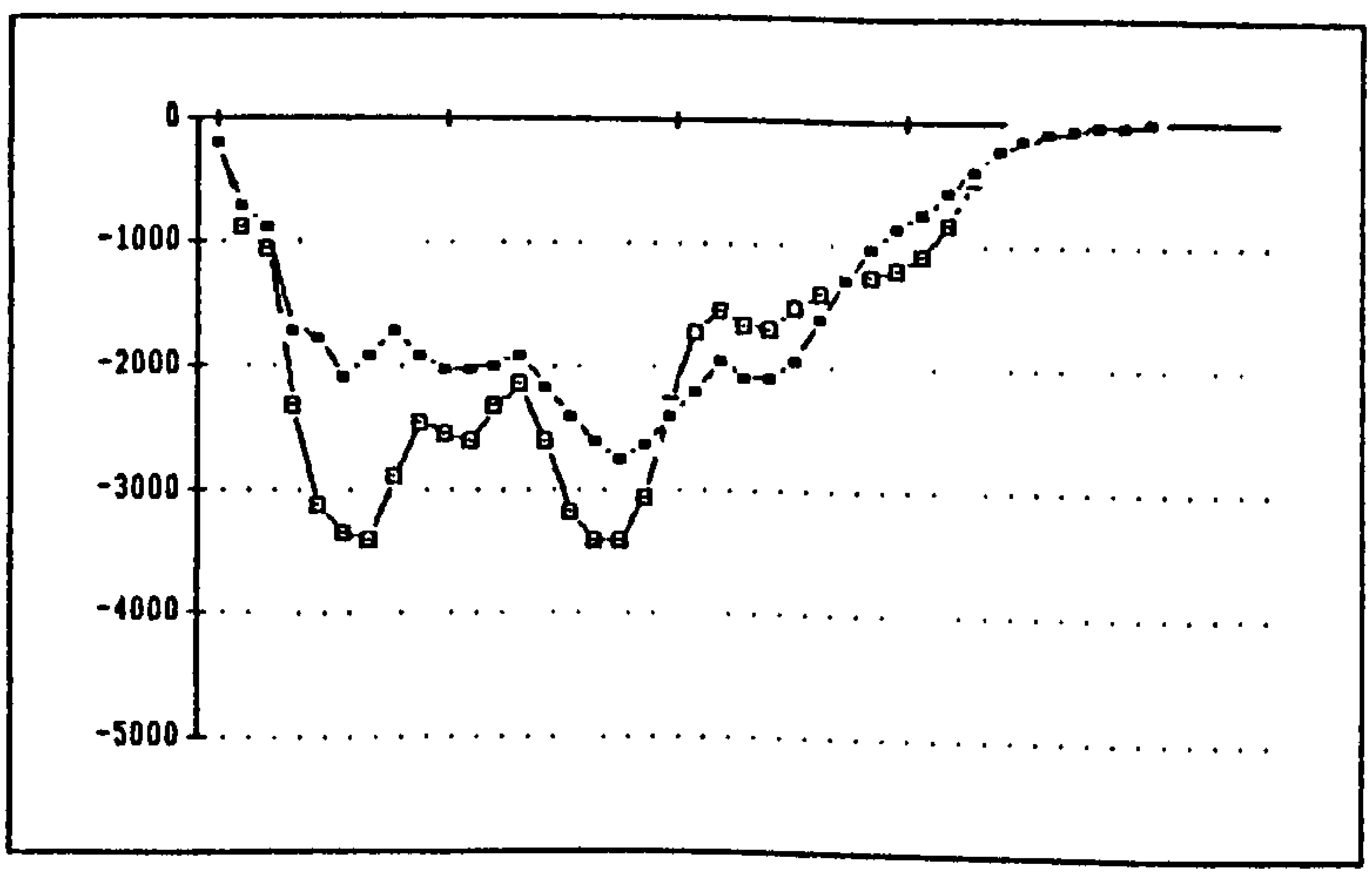


Fig. 5.4.2c

6. DISCUSSION

6.1 Phase One Model: Muscle and Joint Forces

The bioengineering model developed for this study includes a number of assumptions which need scrutiny before verification of the model can be established. The use of a straight-line approach to muscular modelling in preference to tangential modelling was chosen on the grounds of repeatability and comparability to other studies such as Smith (1975) and Burdett (1982) the only two dynamic studies reported in the literature. Data presented in section 3.5 indicate that the method of analysis used gave high repeatability, hence indicating the value of using this method.

A fully dynamic model is also needed as omitting D'Alemberts inertia forces caused the muscle and joint forces to be increased by 10.5% as shown in section 3. This compares with an increase in value of 2.2% found by Smith (1975) in his study of knee-joint forces in a drop jump.

The shin muscle group is commonly omitted from bioengineering models to reduce mathematical indeterminancy, but a study by Burdett (1982) showed the anterior leg muscles to be active for the first 9% of the stance phase. The force data here indicate activity for 9.3% and were verified by EMG (at approximately 5%), as were all the major muscle groups, whilst running on a treadmill at the same speed as used in the investigation.

The inclusion of the soleus may well influence the final results of this type of study and hence further analysis was performed later on in this investigation to include this muscle. Moment sharing based on physiological cross-sectional area provides the easiest solution, but mathematical optimization using a cost function of, say, minimum total muscle force or minimum total joint force would normally be an equally valid solution. Clear electromyographic analysis of soleus activity would assist in determining the role played by this muscle in both force generation and injury causation.

Results obtained from this bioengineering model displayed good agreement with other studies with regard to joint moments. Mean values of 174 Nm for the ankle, 188 Nm for the knee and 98 Nm for the hip are within the range of values reported in the literature and recorded in Table 2.5.3.

Quadriceps muscular loading rates were recorded at between 135 and 201 kN s^{-1} for the four subjects. The calculation is based upon the maximum loading rate value over 0.02 seconds obtained from each muscular force trace. It is not an average value calculated from the commencement of force development to the peak value. This could account for the differences recorded here, average of four subjects of 175 kN s^{-1}, with those of Komi et al. (1985) who reported a value of 100 kN s^{-1} using a strain gauge implant attached to the Achilles tendon. If this "averaging" technique on the Achilles tendon then a value of 62 kN s^{-1} is derived. This is hence more conservative than Komi's values.

Scrutiny of the muscle and joint force data obtained show that other dynamic activities studied display similar results. Smith (1975) found knee-joint compressive loads of 24.4 BW for a 1 m drop jump compared to 33.0 BW reported here. Smith also reported ankle compressive forces of 7.4 BW, quadriceps muscle forces of 16.5 BW and a calf muscle force of 6.1 BW. All these forces reported by Smith are approximately three-quarters of the values found in this study at 9.0 BW, 22.0 BW and 7.3 BW, respectively.

Burdett (1982), in his research into ankle biomechanics in running, found ankle-joint compressive forces of 13.3 BW and gastrocnemius muscle forces of 10.0 BW. Both are larger than those respective components reported here for the same running speed.

In comparison, forces in sedate activities, walking for example, are much lower. Knee joint compressive forces are commonly 3.8 BW to 3.5 BW and quadricep muscle forces in the range 1.1 to 3.4 BW (see Table 2.5.2 for full summary). Thus for a modest increase in speed from normal walking (approximately 1.3 m s^{-1}) to running at 4.47 m s^{-1} knee joint reactions and quadricep muscle forces both increase ten-fold. This is achieved with the aid of 'protective' running shoes and shows the injury potential of this activity. Running without adequate footwear or increasing these forces further may increase the likelihood of injury.

The largest muscle force reported here of 22.0 BW in the quadriceps muscle group could result in a tensile stress of 70 MPa existing within the quadriceps tendon (assuming a cross-sectional area of the tendon of 2 x 10^{-4} m^2). This is within the range of ultimate tensile strength (UTS) reported by Burdett (1982) to be between 34 and 147 MPa. However, these UTS tests were performed statically on cadavers and the strength of human tissue has been shown to be dependent upon strain rate (Abrahams, 1967). The brevity of the loading action occurring in running may well explain how the tendons accommodate such high forces without incurring physical damage.

Table 2.2.2 shows the incidence of running-related injuries as reported by various investigators and, as can be seen, the knee joint incurs the highest of all injuries (average of 28%) followed by Achilles tendon disorders (8%) and shin splints, usually tibialis anterior compartment dysfunction (8%). The results summarized in Table 5.1.4 show that the highest muscle and joint forces occur in and around the knee, and the second largest muscle force is in the gastrocnemius, i.e. Achilles tendon tension.

With such high muscle and joint forces in and around the knee, including patella-femoral contact force, it is not surprising that the knee joint incurs injury, especially in long-distance running. Knee joint injuries in this case are predominantly over-use injuries caused from either training errors, as highlighted in section 2.2, human biomechanical irregularities or the shoe-surface (James et al., 1978); any small change in running action, either from footwear, terrain or style, could be suggested to cause irregularities in the force patterns developed by the knee joint components and thus cause injury.

The second most injured part of the human body in running, the Achilles tendon, displays high muscular loading rates as previously outlined. For its small cross-sectional area, very high stress and strain rates are present and hence with extensive running (70 km per week on average for long-distance athletes) it is not surprising that the Achilles tendon 'breaks down'. Passive stretching and progressive strengthening (Nigg, 1986) routines are recommended to reduce the likelihood of such injuries (Clement & Taunton, 1980).

Shin splints, a term given to almost every anterior tibial leg pain, is used more specifically to identify tibialis anterior muscular compartment dysfunction, or swelling of the muscle causing it to rub against the tibia. Shin splints is also an over-use injury, but as can be seen from Table 5.1.4, SH has a low peak force value of 0.8 BW and is only active for the first 9% of the stance phase. It is suspected that variations in style, footwear and speed may have a significant effect upon the role of the shin muscles. Any small alteraton in these parameters may cause relatively large differences in this relatively quiescent muscle group and thus may be the cause of such injuries.

As suggested above, it is suspected that changes in speed, footwear, running style, terrain and fatigue may contribute to the occurrence of injury. Further studies were considered to investigate the effects of such parameters and how the muscle and joint forces are altered by them. It may then be possible to propose a strategy to minimize the likelihood of injury from these parameters. Investigation of all these variables was not be feasible in one test and as such a systematic approach of analysing one or two at a time was the most efficient approach.

6.2 Phase Two: Effects of Running Style

As suggested in section 6.1 the effects of speed, style and shoes may influence muscle and joint force profiles and one aspect was investigated here with running style being analysed. Two groups of runners, front and rear foot strikers, were identified by their foot ground reaction force traces.

As can be seen from Table 5.2.6 the significant difference between the two groups of runners occurs in shin muscle group (tibialis anterior) and knee joint shear force. Table 2.2.2 shows the occurence of injuries to runners and it can be seen that the knee joint is the most frequently injured aspect of the human lower limbs, the shin group being the third most injured aspect.

The second most injured site in the human lower limbs as can be seen from Table 2.2.2 is the Achilles tendon as served by the tricep surea muscle group. This muscle group can accommodate a large load capacity and is reflected by its relatively large cross sectional area (Alexander and Vernon, 1975). Any small changes occurring in this group, due to running style, may not be immediately noticeable and is possibly why no variation is reported here. Conversely the shin muscle group is anatomically smaller and hence "weaker" and contributes less to the gait cycle (Figure 5.2.1), any small changes here are significant (as shown) and as proposed in section 6.1 this could be why injury so readily occurs in this group.

The knee joint is one of the most complex structures in the human body with intricate movements in 3 planes needed for flexion and extension. It is also surrounded by or crossed by a number of large muscle groups as well as numerous smaller ligaments and muscles. Its ability to absorb vertical or compressive loading can be seen from Table 5.2.3 as being in the region of 31 BW (2 Tonne). Its ability to absorb anterior-posterior (or posterior-anterior) shear forces is questionable and can be considered due to its lack of protection offered by local soft tissue. This is represented in Figure 5.2.6 by the large differences between subject groups. In a similar context to the force in the anterior shin muscle group, large variations in this otherwise weak load bearing joint aspect may lead to joint, and ligament, injuries. Chondromalacia patella, or cartilage degradation may well occur occur as a combination of high compressive forces and fluctuations in shear forces.

Which group is more susceptible to injury from peak forces is not conclusive and foot contact time data may be significant in further understanding the aetiology of injury. The rear foot striker group has a longer foot contact time and lower vertical ground reaction force indicating a more gradual loading regime of the muscles and joints. This increased contact time is also represented with a longer shin activity time and a higher shin muscle peak force as might be expected due to the role of the tibialis anterior in the rear foot strikers' gait. Here the foot is held in dorsi-flexion longer by the anterior shin group. This also results in the foot being presented to the ground in an inverted position which can cause greater pronation.

It may be hypothesised therefore that rear foot strikers, with their reduced ground reaction force and increased foot contact time could be less susceptable to injury than front foot strikers. The reduced loading rates, produced by the longer foot contact time and reduced force peaks may be the cause of injury more than the actual muscle and joint force value. If this is correct, these front foot strikers may be more at risk. This could be clarified with a coordinated injury history and muscle and joint force profile and may be an area for sports injury clinics to consider in the future.

6.3 Phase Three: Analysis Between Running Speeds

Following the recommendations set down in 6.1 the next variable to be tested was speed with three different velocities being selected according to the guidelines established in 5.3. Here speeds of 3.83 m s^{-1}, 4.47 m s^{-1} and 5.36 m s^{-1} were employed. Further parameters were also evaluated in the analysis, namely lower limb angles, foot/ground impact data and minimum hip flexion moments.

Although intra-subject variations were minimised, larger inter-subject differences in muscle and joint force data were noted. This made comparisons between speeds difficult due to the high statistical variances recorded. These inter subject differences could be due to the unnatural speeds selected for evaluation. Some subjects found it difficult to run comfortably at 3.83 m s^{-1} and took longer to warm up at this speed than the 4.47 m s^{-1} velocity. Similarly some of the athletes, more comfortable at the slower speeds, due to their main athletic events being longer distance races (marathons), found it difficult to maintain the elevated pace of 5.36 m s^{-1} and needed extended recovery periods. This faster speed was also the last of the three to be undertaken by each subject and fatigue may have begun to effect their performance. However no subjective feedback was recorded on any of these matters.

The kinematic range displayed by the athlete in knee flexion (Figure 5.3.6) is similar to those reported in the literature and summarised in Table 6.3.1. As can be seen the results here resemble most closely the data presented by Nigg (1986), Nigg et al. (1987), Mero et al. (1987) and Sinning & Forsyth (1970). The maximum knee flexion angles reported here of 38°, in accordance with the data in the literature (Table 6.3.2.), tends to suport the modelling criteria of the force developed in the quadriceps being fully transmitted to the patellar ligament.

TABLE 6.3.1. Summary of Angular Kinematics Reported in the Literature

Investigators and Speed	Limb Angles in Degrees		
	Knee Flexion		Lower Limb
	Heel Strike	Max.	Heel Strike
Elliott & Blanksby, 1979a			
$3.5ms^{-1}$	156.2	NR	6.2
$3.5ms^{-1}$	156.3	NR	2.9
Elliott & Roberts, 1980			
$5.2ms^{-1}$	151.0-153.7	NR	0.5-3.1
Gamble et al., 1988			
$4.0ms^{-1}$	153-158	NR	1.0
Mann et al., 1985			
$3.4ms^{-1}$	142	120	NR
$4.5ms^{-1}$	137	118	NR
$6.3ms^{-1}$	150	125(1)	NR
Mero et al., 1987			
$9.7-10.3ms^{-1}$	146-152	138-142	1.0-4.0
Nigg, 1986 & Nigg et al, 1987			
$3ms^{-1}$	161.9-161.4	NR	NR
$4ms^{-1}$	158.9-158.6	NR	NR
$5ms^{-1}$	152.2-150.4	NR	NR
$6ms^{-1}$	144.0-144.2	NR	NR
Sinning & Forsyth, 1970			
$3.83ms^{-1}$	NR	153.5	NR
$4.47ms^{-1}$	NR	153.0	NR
$5.36ms^{-1}$	NR	153.2	NR

NR = Not Recorded
(1) = No peak maximum flexion exhibited

TABLE 6.3.2 Ratio of Tensions Along Quadricep Tendon and Patellar Ligament

Investigator	Angle of Knee Flexion (degrees) 0	10	20	30	40
Bishop & Denham, 1977	NR	NR	1.00	0.95	0.88
Buff et al., 1988	1.10	1.30	1.10	0.92	0.76
Ellis et al., 1980	1.00	1.00	0.98	0.95	0.91

NR = Not recorded

This maximum value of nearly 40° would result in a minimum ratio of tension along the quadriceps tendon to the force in the patellar ligament of 0.88 according to Bishop & Denham (1977), 0.76 according to Buff et al. (1988) and 0.91 according to Ellis et al. (1980). If the average knee flexion angle, during foot stance, is 20° (Figure 5.3.6), then the average ratio, as expressed in Table 6.3.2 would be close to unity.

The maximum knee flexion angle reduces as speed increases and this could be a protective mechanism that helps reduce forces at higher speeds by transferring loads to and from other bony structures. This is similar to the concept forwarded by Nigg (1986) with his 'equivalent mass' hypothesis.

Minimum hip joint moments were extrapolated from data during this study to elicit an understanding of hamstring injuries as the previous study (Phase 1, section 6.1) had shown an initial maximum and then proceeded to become negative. However no significant differences between speeds are reported from this value.

Impact loading rates were calculated from the ground reaction data but no significant differences between speeds were recorded. As suggested in section 6.2, muscular loading rates could be responsible for injury causation. Hence the lack of significant difference between running speeds for the ground reaction force impact loading rate is inconsistent with this hypothesis. However the sampling rate chosen for the force platform data capture of 180 Hz may not be high enough to extract this relevant information. Data smoothing and cubic spline reconstruction of the data may also rob the raw data of the impact peak characteristics. The loading rates at between 82 and 106 kN s^{-1} are generally lower than those recorded by Nigg (1986) who recorded 150 kNs^{-1} for barefoot running and values between approximately 70 kNs^{-1} and 270 kNs^{-1} for various speeds and footwear conditions. These lower loading rates could be due to this smoothing of the data. It could also be suggested from these results that the shoes worn by the subjects throughout the tests were thus providing significant impact force protection and could have outsoles with a Shore A hardness of approximately 25.

Very few significant differences between speeds can be reported throughout this analysis and out of the twenty four separate items considered for scrutiny for each speed Table 5.3.8 shows only two significant differences between the slowest speeds, 3.83 ms^{-1} to 4.47 ms^{-1} (7 min/mile to 6 min/mile); six significant differences between the faster speeds, 4.47 ms^{-1} to 5.36ms^{-1} (6 min/mile to 5 min/mile); and seven significant differences for the extremes of speed, 3.83 ms^{-1} to 5.36 ms^{-1} (7 min/mile to 5 min/mile).

Predominantly the shear force elements within the overall force system display significant differences across nearly all speeds. RY, the horizontal foot/ground propulsion force, increases significantly as speed increases and AY, the ankle joint reaction force shows significant increases across the higher speed comparisons. This ankle joint shear increase may be viewed alongside the increase in contact angle of lower limb which also increases significantly as speed similarly increases. The vertical ground reaction forces do not show respectively significant increases across the velocity range and it could be hypothesised that the athlete provides the extra drive required more by maximising his horizontal force profile, increasing his range of motion, all in a shorter time, without subjecting his body to elevated loads. This would require a flattening of the hips trajectory i.e. reduction of loss of potential energy with each stride.

Although a film re-analysis would be needed to establish this trajectory pattern, scrutiny of the knee flexion angles shows that for an almost constant amount of knee flexion (across the speeds, as is represented in Table 5.3.7) the hip must lower as the lower leg angle increases (see Figure 6.3.1 below) at heel strike.

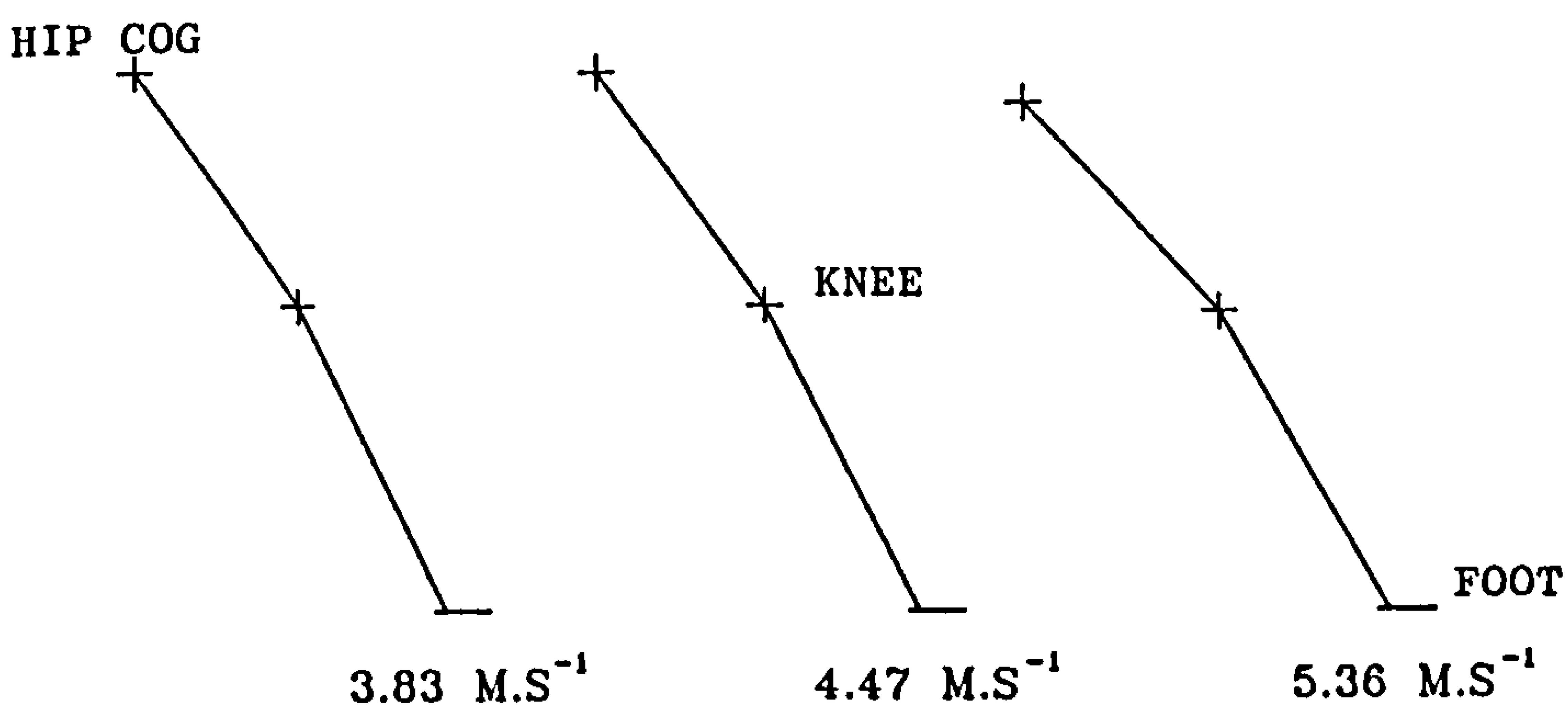


FIGURE 6.3.1 LIMB ANGLES AT HEEL STRIKE ILLUSTRATING LOWERING OF HIP & HENCE BODY COG

These increases in shear forces can also be seen to be significant when reference is made to the discussion on knee joint shear force (KS) variations with running style presented in Section 6.2. There it was hypothesised that large differences recorded in the KS value could be a factor in the cause of chondramalacia patella. Although the knee joint shear force KS (and the patella-femoral contact force) do not display significant differences in this study the overall trend tends to show that variations in style and speed affect the horizontal force elements specifically. As these values are, in general, small compared to the horizontal or normal elements in the force system they may have been easily overlooked in previous studies.

However, when considering the anatomical construction of the human long bones it can be seen that they are "designed" more to withstand normal forces than tangential forces. Soft tissue, and ligaments mostly, are all that the human lower limbs have to resist these lateral loads.

These facilities may be minimised by appropriate strengthening and stretching routines and running shoe companies may do well to maximise on grip rather than purely on impact force reduction and motion control if this hypothesis is confirmed.

James et al. (1978) and other investigators state that the most common cause of injury is training errors (see Section 2.1 & Table 2.2.4) and as discussed earlier are commonly reported as being rapid changes in training, intensive workouts and race related training. All these could be reflected in the significant differences between the two extreme speeds i.e. any rapid change of pace from the 7 min. per mile pace to 5 min. per mile pace will produce elevated force components, specifically the muscular loading rates (see Table 5.3.8 for summary).

Combined with the increases in force elements with increased speed, athletes tend to utilise racing footwear or spiked shoes to undertake intensive workouts or races. These lighter shoes also offer less protection to the human locomotor force system and often have a construction with, for example, less heel lift thus stretching the tricep surea muscle group over a larger working range (Plate 6.1). This combination could be responsible for increasing injury potential.

In conclusion it can be seen that an analysis between three speeds representing 7, 6 & 5 minute per mile running has produced in general inconclusive results and it could be suggested that a wider spread of speeds could be selected. These could be a jog, run and sprint as used by Mann et al. (1986) or speeds of 3 ms^{-1} (9 min per mile), 4.5 ms^{-1} (6 min per mile) and 7 ms^{-1} (3.75 min per mile). These speeds represent a four hour marathon, an extremely popular finishing time in the 1988 London Marathon for the slower speed. Average training velocities for most club and elite athletes for the median speed and world record pace for one mile for the fastest speed. However, finding subjects who can run naturally and comfortably at all speeds may be difficult and analysing three different groups of athletes, one group at each speed, may not show why changes in routine and speed elevates athletes injury potential.

A re-analysis with larger groups, more runs per speed and slightly wider speeds may be the most profitable way of both reducing variability and establishing concrete proposals of the effects of speed on injury.

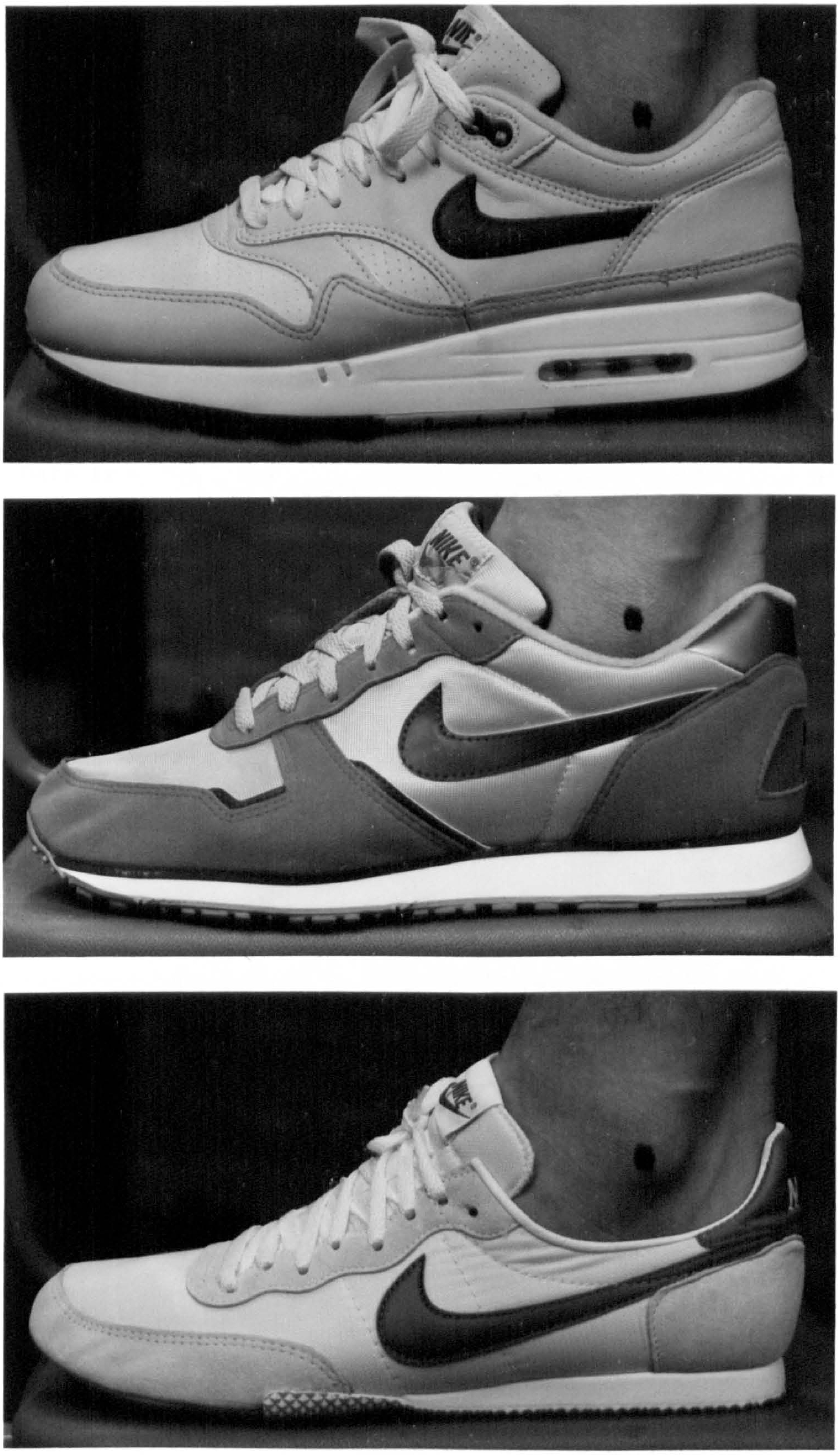


PLATE 6.1. Sample of Running Shoes Showing Varying Amounts of Heel Lift

6.4 Phase Four Model: Model Enhancement

The three previous analyses have been performed without the inclusion of the soleus muscle in the tricep surea group. The results presented in Section 5.4 were obtained by re-analysing the data from Section 5.3 with the soleus included in the bioengineering model. Hence the results discussed here are in comparison to the data presented and discussed in phase 3 (Sections 5.3 and 6.3).

Data from the three speeds were analysed, but at the two extremes of velocity data were corrupted, and hence only 25 and 31 runs were successfully analysed for the respective speed of 5.36 m s^{-1} and 3.83 m s^{-1}. In comparison 61 and 51 runs were analysed in phase 3 for these respective speeds. Combined with the large inter-subject differences (see Table 5.4.2) this made comparison between models difficult and no significant differences were recorded. Hence data from these speeds have been omitted from this discussion.

Whilst some studies, Smith (1975); Nissel & Mizrahi (1988) and Seedhom & Terayama (1976) have excluded the soleus from the model for various reasons Ellis et al. (1979) and Bobbert et al. (1986a & b) have included the muscle into their biomechanical models. EMG scrutiny clearly shows that the soleus should be included in the model and hence a method of inclusion had to be sought. The choice of moment sharing about the ankle is hence consistent with the studies of Dahlkvist et al. (1982), the specific biomechanical modelling performed in 3.2 (point 4) and the EMG validation performed and reported in Section 3.5.

The results obtained reduce the largest muscle group force and joint reaction, QU & KN by substantial amounts from 18.2 BW to 15.0 BW and -31.0 BW to -21.3 BW respectively. The hamstring muscle force is increased, this however is only in the order of 11%, from 1.8 BW to 2.8 BW, and the sharp peak force trace displayed in Figure 5.4.1c is more consistent with the type of injury sustained by the hamstring i.e. acute muscle 'pulls'. The quadriceps muscle force curve is also consistent with the EMG recordings in that the force developed reduces significantly after approximately 60% of the stance phase. The reduction of the peak quadricep muscle force and knee joint normal reaction gives results similar to those found by Smith (1975) of 16.5 BW and 24.4 BW respectively, and may be considered to be more realistic overall.

Hence the model presented here produces results more conservative in nature but also consistent with other results in this study and other models and investigations. This model with the soleus included, according to a joint moment sharing criteria, should now be adopted for all further analyses.

6.5 Comparison between all Phases

In total 339 runs were analysed across all the phases of this study and Table 6.5.1 (and Figure 6.5.1) shows the consistency of the results for 227 of these runs (all those at 6 minute per mile pace).

The consistency of the GA result in phase three, performed over 57 runs is relatively high:- mean value 10.7 BW, standard deviation $\pm$1.5 BW. This consistency could be due to the refinements made to the subjects bone models where a one segment rigid foot model was replaced with a two segment flexible foot model. This allowed

TABLE 6.5.1 Comparison of Key Elements Between Phases of the Study for 4.47 $m.s^{-1}$

Phase	n(3)	Muscle Forces(1)				Joint Reactions(1)				Ground Contact Data		
		SH	GA	QU	HM	AN	AS	KN	KS(7)	RZ(1)	CT(2)	SAT(2)
1	28	0.8	7.3	22.0	3.1	-9.0	-3.9	-33.0	2.5	2.9	0.20	0.018
2a	42(4)	0.5	7.6	20.4	2.7	-9.3	-4.1	-31.0	2.0	2.9	0.19	0.017
2b	43(5)	0.7	7.6	20.6	2.9	-9.0	-4.1	-31.1	2.5	2.8	0.20	0.021
3	57	0.4	10.7	18.2	1.8	-11.5	-6.4	-31.0	2.0	2.9	0.21	0.016
4	57	0.5	5.4(6)	15.0	2.8	-11.5	-6.4	-21.3	2.4	2.9	0.21	0.015

(1) Mean Body Weight units

(2) Seconds

(3) Total number of runs by subjects

(4) Front foot strikers

(5) Rear foot strikers

(6) Soleus also equals 5.4 BW

(7) Modulus values reported here

FIGURE 6.5.1. Comparison of a) Total Muscle Force, b) Total Joint Force and c) Total Muscle and Joint Force in Body Weight Units. Modulus values used in calculation of totals. Data from Table 6.5.1. Tricep Surea value used for Phase 4. Mean & Deviations presented.

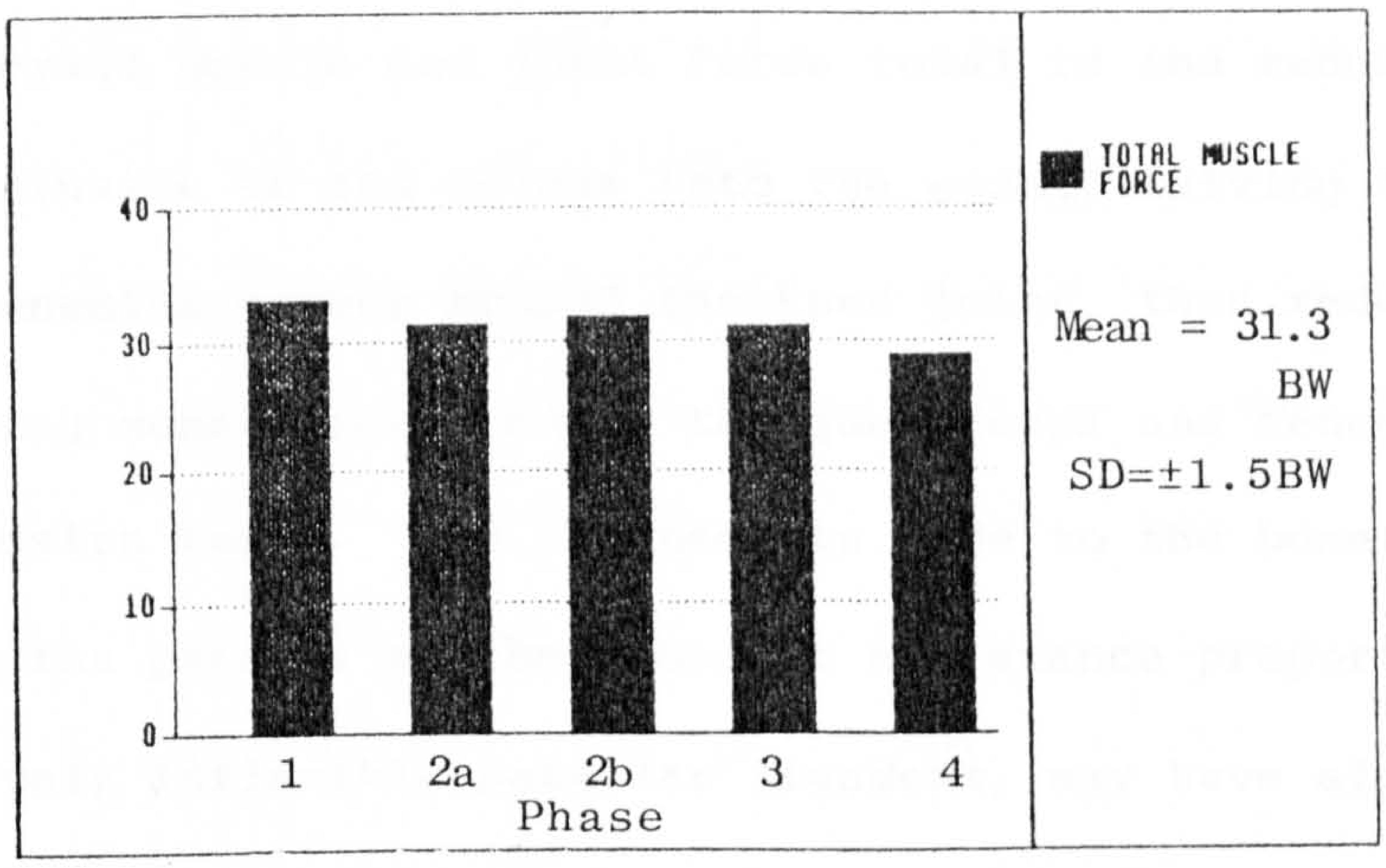


Fig. 6.5.1a

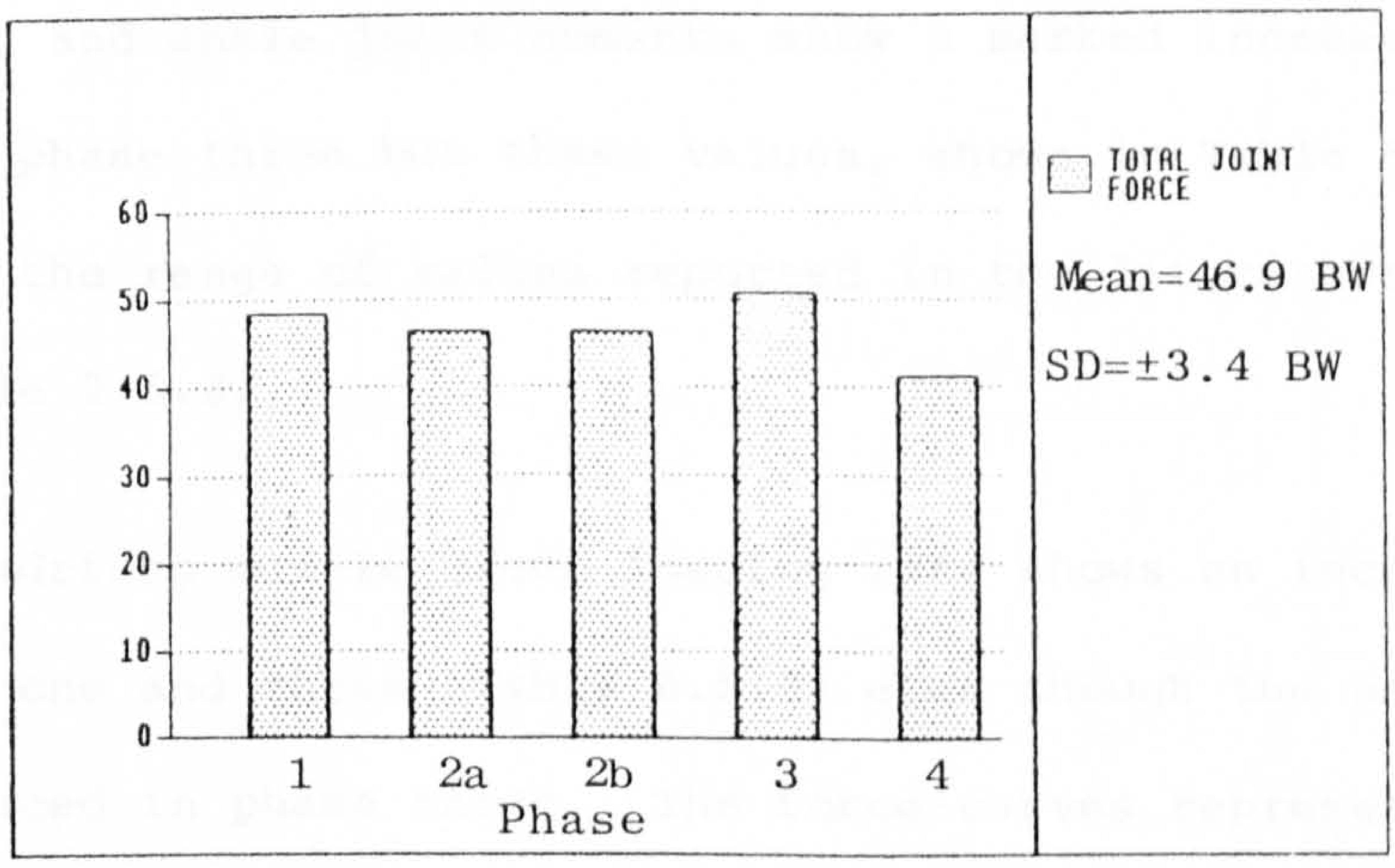


Fig. 6.5.1b

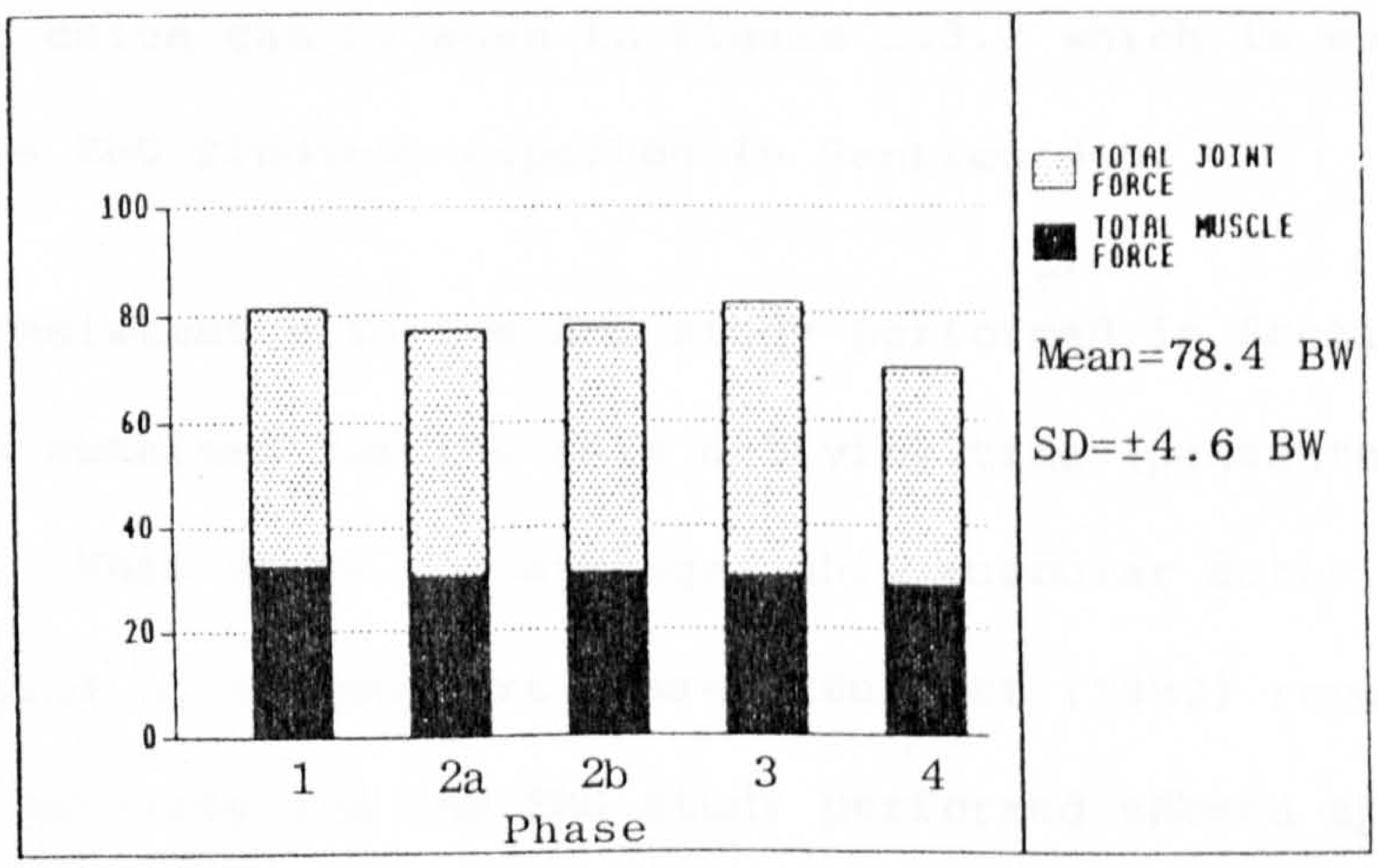


Fig. 6.5.1c

accurate placement of the bone model when the metatarso-phalangeal joint is in extension (see Plate 4.4).

After the soleus was included in the model the total muscle force remained relatively constant and the predominant factor in reducing the overall muscle and joint force total is the reduction in KN. The inclusion of the soleus into the model, halving the gastrocnemius moment around the knee joint, thus reduces the balancing moment required by the quadriceps and hence the joint compression force. The refinements made to the bone models, in wiring the patella to the tibea at a distance proportional to the relatively inflexible patellar ligament, may have also helped in making the forces around the knee more consistent.

The hip and ankle joint moments show a marked increase from phase one to phase three but these values, shown in Table 6.5.2 are still within the range of values reported in the literature and reviewed in Table 2.5.3.

The quadricep muscle group loading rate shows an increase across phases one and three (Table 6.5.3) even though the peak force for QU is reduced in phase three. The force curves representing these items are shown in Figures 5.1.2 and 5.3.2, the 'sharpness' of the loading curve can be seen in Figure 5.3.2 which is more consistent with the EMG findings reported in Section 3.5.

Also consistent with the EMG study performed in Section 3.5 are the results obtained for the shin activity time (presented in Table 6.5.1). This shows, on average, shin muscular activity for the first 8.6% of the support phase. Burdett (1982) reported 9% shin muscle activity and the EMG study performed showed approximately 5%

activity.

The larger number of analyses performed in phases 3 and 4 should make these the most consistent results of the study. The differing number of samples and large inter-subject differences makes statistical comparisons across phase difficult however. Different subjects, running styles, experience and data collection, processing, analysis equipment and routines also reduces the viability of statistical tests between phases. For these reasons these tests were not performed. In general, however, a marked degree of consistency can be seen from Figure 6.5.1 and within phase repeatability has been quoted at ± 3%.

TABLE 6.5.2 Comparison of Joint Moments Between Phases of Study for 4.47 $m.s^{-1}$

Phase	Joint Moments (Nm)		
	Ankle	Knee	Hip
1	174	188	98
3	214	189	150

TABLE 6.5.3 Comparison of Quadricep Muscular Loading Rate Between Phases of Study for 4.47 $m.s^{-1}$ in $KN.s^{-1}$

Phase	Loading Rate
1	175
3	333
4	292

6.6 Model and Method Verification

6.6.1 Model

Although there is no method for establishing the validity of this bioengineering model, Section 3 outlines the reasoning behind the modelling performed for this study. The first part looks at the basic biomechanical techniques used, most of which are standard techniques employed in this field of engineering. Representing the running action as a two dimensional activity is consistent with a number of studies involved in athletic and similar sports (Jumping - Smith, 1975; Weightlifting - Ariel, 1974; Cycling - Nordeen & Cavanagh, 1976; Skiing - Quigley & Chaffin, 1971; Walking - Morrison, 1968). Grouping the muscles according to their major function, segmenting the body into simple shapes and neglecting friction in synovial joints have all become standard techniques. Considering the joints to be represented by non complex articulating surfaces with point load contact is also becoming standard but care must be taken in identifying the knee joint centre of contact. This is different from the centre of rotation (or instantaneous centre of rotation) which cannot be used in the bioengineering model. It introduces two more variables into the system if moments are taken about the centre of rotation i.e. KN & KS.

The use of straight lines between muscle origins and insertions in preference to tangential representations has been discussed by Jenson & Davy (1975) who state that the straight line approach accords greater repeatability. Scrutiny of Figure 6.5.1 shows that the results presented throughout the four phases of testing display relatively high consistency. The average error within a phase of testing is that quoted in Section 4.2 where a repeatability study showed an average error of ±3% from experimentation. These different phases were performed during a three year period, on different equipment and under different lighting conditions thus confirming this hypothesis.

For the specific modelling performed in Section 3.2 the exclusion of the ileo-psoas muscles can be further justified by the force generated by the quadricep muscle group. These short weak flexors of the hip could easily be absorbed into this QU muscle force value.

The results from the kinematic study performed in 5.3 show maximum knee flexion angles to be 38 degrees. According to Ellis et al. (1980) this would result in a ratio of 0.91 for the tensions between quadricep tendon and the patellar ligament. As this is only of a short duration and varies for each subject and speed therefore adoption of a ratio of unity is justified.

Apportioning the force developed in the tricep surea group between the soleus and the gastrocnemius could have been performed on the grounds of physiological cross sectional area. This would have been consistent with the rationale for equally splitting the quadriceps muscle group up into four equally potent muscles. Data from Alexander & Vernon (1975) could have been used but this would have been inconsistent with other studies by for example Seedhom & Terayama (1976). It was consistent with the sharing of the hip moment by the gluteals and hamstring muscle groups. The inclusion of the soleus was justifiable on the grounds of the EMG results (Section 3) which showed across two separate testing regimes activity continuing almost throughout the stance phase. Similar EMG results validate the inclusion of the shin muscle group as described in Section 6.5. This also keeps the model consistent with the only other dynamic bioengineering model used in the analysis of running (Burdett, 1982).

Whilst these reasonings justify the muscles omitted or included in the model they do not validate the levels of force recorded by the model per se. Only surgical implantation will enable this, and the work currently being performed by Komi and his associates is likely to lead to ratification. Komi et al. (1985) have already reported muscular loading rates in the achilles tendon of 100 kNs^{-1} using a surgically implanted strain gauge. Results for this study show muscular loading rates in the tricep surea group of being between 109 $kN.s^{-1}$ and 129 $kN.s^{-1}$ for the three speeds investigated in Section 5.3. This helps to validate that section of the model to a great extent and further results from these investigators will be eagerly awaited and recent publications from their study group have elicited achilles loadings in the cycling domain (Komi et al., 1987; Gregor et al., 1987).

The use of a simple inverse dynamics approach has been used but as suggested in the Review of Literature other techniques exist in mathematical optimisation and these have been rejected for the reasons cited in Section 2. During the study other techniques were also investigated within the engineering and mathematical domain.

These possible solutions included using the Principle of Virtual Work (Case & Chilver, 1971) to determine if a relationship between deflection of the foot and the work done displacing it, and hence the forces in the shin and tricep surea muscles existed. This proved unsuccessful as did an attempt to calculate the forces in the muscles of the lower limb from knowledge of their tension/length characteristics (Elftman, 1966; Morehouse & Miller, 1976). A further possible solution involved calculating the total upper body linear and angular accelerations to determine their inertial forces. This produced twelve equations with twelve unknowns but the twelve by twelve mathematical matrix produced was 'ill-conditioned' in that wide fluctuations in the solutions were obtained for very small changes in input parameters. This method was hence rejected

The only method hence available via computing power, knowledge and experience was that of simplification or reduction such that "an originally indeterminate problem is simplified until it becomes determinate" Herzog (1985). Hence all results produced here are from this simplification technique of biomechanical modelling and the data is within the range or expected range of data presented in the literature. Whilst most of these studies have been more static in nature than dynamic those involving more dynamic activities show good correlation with data produced here (Burdett, 1982; Ariel, 1974; Nissel & Mizrahi, 1988).

6.6.2 Methodology

A number of the techniques used in this study may not be considered standard biomechanical testing or analysis regimes but have been adopted to improve the models accuracy.

Firstly the number of runs per subject, per speed or style, was initially chosen in accordance with the findings of Bates et al. (1980) which established expected error bands for corresponding number of repeat trials. They reported that repeatable data could only be expected if the number of trials exceeded eight. The repeatability study performed here showed an expected accuracy of ±3% if seven runs were utilised. This compares favourably with the 50-100% errors reported by Smith (1975) who however only used one subject and does not report the number of repeat trials employed.

Where different speeds were to be analysed the subjects started at the slower velocities and progressed up the speed range, warming up to each in the process. This reduced any effects of fatigue and problems caused by undertaking 'fast' speeds were thus minimised. Adequate rest between trials was accommodated by allowing the subject to participate in the experimental procedure. The subject recorded his own time over the 10 m span and reported if that was acceptable within the given tolerance. He also reported a subjective analysis of the trial such that he had not stuttered during the approach run or landed with a non standard foot strike. If he and the experimenter felt happy he advanced the background co-ordination number for the next trial and returned to the start line ready for the next run. This involvement increased subject enthusiasm and thus performance.

Bone models, of various sizes, have been used by a number of investigators, Seedhom & Terayama (1976), Smith (1975), Ellis et al. (1979) and Dahlkvist et al. (1982). 40% full size were employed in this study due to the availability of corresponding digitisers and film projection equipment. Full size models have been used by Harrison (1982) to show how model viability could be improved when compared to using standard anthropometric data obtained from cadavers or the literature as used in studies by, for example, Ariel (1974), Bishop & Denham (1977), Burdett (1982), Proctor & Paul

(1982). Brand et al. (1982) also used data from three cadavers to produce a biomechanical model but expressed possible errors of 20% in establishing muscle moment errors using this 'standardising technique'. Thus these errors can be minimised by utilising bone models tailored for each subject from their own radiographs.

These radiographs need to be accurately reproduced however and the double precision technique adopted in this study (using pins of known length and cross referencing with subject anthropometric measurements), enables errors from divergent rays and non lateral exposures to be corrected for.

If the use of individual radiographs to produce bone models is advisable then so should individual EMG recordings. These could be used to either build up a full data base or standardised gait pattern for the group to enable an algorithm for each subjects muscular activity pattern to be included in the mathematical model. This may be required due to the findings of McIntyre & Robertson (1987) and Van der Straaten et al. (1975) who report variations in EMG recordings between subjects. For this study the production of a footswitch built into every subjects running shoe to co-ordinate footstrike and EMG recordings was not feasible and the employment of three experienced subjects is in partial acknowledgement of this requirement.

Hence methodological repeatability has been established using careful experimental techniques in accordance with practices recorded in the literature above.

6.7 Forces in Relation to Injury

Data presented in Table 2.5.2 shows the magnitude of forces estimated in the human lower limbs during various activities. Only Ariel (1974); Burdett (1982); Nisell & Mizrahi (1988) and Smith (1975) could be considered dynamic or sporting in nature. Only Smith's study reports muscle and joint forces of the magnitude reported in this study, the highest joint force reported there was 24.4 BW in his study of a 1 m deep landing. None of these studies present kinematic data, joint moments, loading rates and muscle and joint forces together and as such direct comparison is difficult. However, the underlying facts that derive from this study are the high magnitude of all the kinetic parameters in a somewhat safe and healthy pastime. Even squatting with 200 kg (or approximately three times body weight) only produces knee joint compression forces of 8.9 BW (Ariel, 1974).

If a subject was asked to repeat that squatting activity, or the vertical jump from one metre as in Smith's study, 80,000 times during one week, it could be expected that some subjective resistance might be met. That, however, is the nature of the activity with running i.e. an average runner completing 70 km per week can expect to impact the ground 80,000 times. The injurious nature of athletics should thus not be taken too lightly.

This injurious nature is obviously reflected in the history profile of injured athletes as presented in Table 2.2.2. The major injury sites of athletes are reported as being the knee joint, achilles tendon and the shin and comparison of this data with the summary of data presented in Table 6.5.1 shows the relationships between these forces and injuries.

The knee joint compressive force is the highest joint force and the muscular forces around the knee are similarly high (QU and GA) thus helping to create the high joint force. The knee joint shear forces KS display relatively low magnitudes of approximately 2.0-2.5 BW but this value has a very high range of values across all subjects suggesting that this is in need of further investigation and these variations have been discussed in Section 6.2. It could also be suggested that quadricep tendon strains might manifest themselves as knee injuries due to the close proximity of this tendon, and its related patellar ligament. This could be the reason for the lack of quadricep muscle/tendon injuries when the muscle force generated is greater than that in the tricep surea group.

These two muscle groups also have similar physiological cross sectional areas and thus should develop similar tensions. According to Alexander & Vernon (1975), the ratio of cross sectional areas of the quadriceps to the tricep surea group is 1.3:1 which is similar to the tension developed by them in Phase 4 of this study. The ratio recorded there is 1.4:1 i.e. the quadriceps equal 15.0 BW and the tricep surea equal 10.7 BW.

For the shin muscle group, tension developed is only between 0.4 BW and 0.8 BW across all phases of testing. This muscle group is the third most commonly injured despite its low force value although stress fractures in the shin bones, tibia and fibula are also often grouped as 'shin splints'. The physiological cross sectional area reported by Alexander & Vernon is approximately one eighth of that of the tricep surea group which would be in line with the force data presented in Phase 1 of this study (SH = 0.8 BW and GA = 7.3 BW). As reported through the EMG study and the biomechanical data the muscle is not used greatly in the gait cycle and hence any sharp variations in its usage could render it liable to trauma. This combined with the muscles limited space both at the front of the shin, and at its posterior aspect where it is bordered closely by bone makes development of the muscle/muscle group difficult. Here intra muscular pressure in the muscles compartment increases creating "circulatory disturbances within the muscle ischemia and delayed venous outflow". The tight fascial lining of the muscle compartment is thus too rigid to permit swelling of the muscle during exercise. (Wallensten & Eriksson, 1982).

Overuse injuries in the hamstring muscles are rare in middle and long distance running as reported in Table 2.2.2. and this is reflected in the peak values recorded throughout all phases of testing and summarised in Table 6.5.1. The phase four testing regime results in a quadricep to hamstring force ratio of 5.3:1 (QU = 15.0 BW, HM = 2.8 BW) which does not agree with the findings of Morris et al. (1983) and Agre (1985) who report isokinetic ratios of 1.2:1 to 1.6:1 in cross country athletes and values of between 1.1:1 to 1.8:1 for sprinters. The nature of the isokinetic test may not be representative of the running action thus nullifying the validity of this comparison.

During the stance phase the hamstrings display EMG activity for approximately the first 60% producing hip extension and resisting extension of the knee by concentric muscle contraction. There is also a sharp hip flexion moment (Figure 5.3.1) and force development (Figure 5.4.1) within the hamstring. It is this rapid recruitment of muscle fibres that could lead to the majority of muscular strains that generally occur more in sprinting than middle distance running (Agre, 1985 and Mann, 1982).

This effect of elevated speed, making the muscles contract in a shorter period of time and over a great joint range could be one of the main causes of athletic injury. James et al. (1978) proposed that training errors were the cause of the greatest percentage of running injuries. These training errors could manifest, themselves as rapid changes in routine, interval training, hill running and race related training. All these involve an increase in speed and/or joint kinematics and as shown in Section 5.4 peak muscle and joint forces increase within a decreased stance phase. Thus the loading rates, rather than the peak forces, may be the prime causes of these injuries, especially if the body in not accustomed to the activity.

Running style has also been shown to influence the muscle and joint force profiles (Section 5.2). Here, as above, the effects of decreased foot contact time results in increased loading rates for the front foot striker rendering him more liable to injury. It would, however, have been interesting in this phase to evaluate knee angles during this study to see if greater knee flexion was adopted by either group to accommodate increased forces. Nigg (1986) proposed that increased knee flexion at foot strike results in a reduction of the effective body mass and thus reduces impact forces.

This analysis was performed in phase three and knee flexion angles at impact, and at maximum condition increased as speed increased (these increases were non significant however). The rationale behind this increase in flexion could be similar to that proposed by Nigg and further investigation could well produce more significant results.

It could be however that the physiological or mechanical limits of the knee joint's ability to absorb shock is reached (perhaps psychologically) and other structures of the musculo-skeletal system contribute more. This could be considered with the changing knee angles discussed earlier or the development of eversion (pronation) where the outside of the foot is used more and provides a softer landing. Hip joint moments are also elevated at the faster speed and thus this joint could be absorbing more of the applied load.

Similarly the muscles may be recruiting more fibres to accommodate this force and protect the joint and the increased muscular loading rates may be representative of that.

Conversely it could be suggested that whilst increasing cadence from walking speed to running speed produces significant increases in muscle and joint forces, variations in running speed do not affect them accordingly. Thus the change from a double support phase in walking to a single support phase (and non-support phase) in running results in a limit in impact forces presented to the skeletal system. Analysis of a speed at which subjects are comfortable both walking and running may assist in this hypothesis.

Throughout this investigation the muscle and joint force elements which were found to be significantly different, between subjects speeds and styles, have been the shear force elements of RY, KS and AS. The impact and propulsion peaks of ground reaction forces, muscle forces and joint reactions have exhibited less variation than might have been expected. It should therefore be the shear or anterior-posterior force elements that should be further discussed and evaluated.

This is particularly important when considering the construction of running shoes. Much research has been directed towards reducing the impact forces acting on the lower limbs during running and as outlined in the review of literature in Section 2.3 various shoe companies have approached this in different ways. Pronation and supination has also been the subject of a great number of studies and has had an effect on the development of shoe design. Both these areas have been catered for by evaluation of running shoe midsole/outsole hardness, shock absorption and deformation. These developments have seen an increase in heel lift or toe spring (Nigg, 1986) to accommodate greater cushioning as shown below in Figure 6.7.1 and in Plates 6.1.

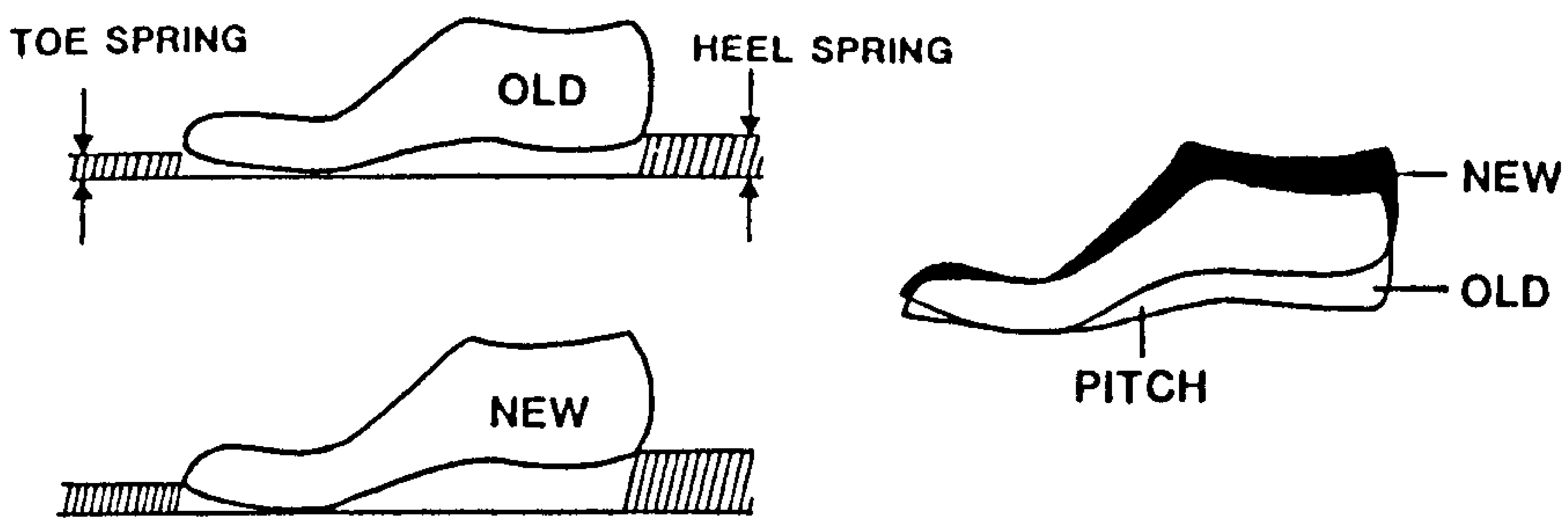


Figure 6.7.1 Heel and Toe Spring with Increased Heel Height (From Nigg, 1986)

This increase in heel lift has allowed cushioning systems such as air encapsulated midsoles (Plate 6.1) to be incorporated. With this increased cushioning greater motion control (pronation) has been needed and this has been achieved in some cases by the incorporation of different density midsoles or outsoles into the shoes as shown in Plate 2.1.

No research has been directed (reported) towards the effects of these technological advancements upon kinematics or kinetics with special respect to the shear force elements. The increase in heel height could be hypothesised to increase knee flexion angle, which according to Nigg 1986 reduces the effective body mass and thus reduces impact forces:- this would be a favourable trade off. However it could cause the athlete to become more of a front foot striker by presenting the toe area to the floor earlier and this has been suggested to increase the likelihood of injury.

This larger heel lift can also create problems when an athlete transfers from training shoes to racing shoes or spikes as these lighter, less protected shoes as shown in Plate 6.1. have less of a heel lift. This causes the tricep surea group to be stretched at each impact by an average one centimeter; repeated continuously over a 10 km, ½ marathon or further running distance it will cause obvious damage and possibly trauma. The use of the same style shoe for racing and training is thus recommended to reduce this risk.

The significance of the anterior-posterior shear forces must also be viewed in conjunction with the grip offered by running shoes. The impact and push off phases of the gait cycle are accompanied by sliding of the shoe over the ground, thus creating wear patterns (Cavanagh, 1980). If insufficient grip is proffered by the shoe for the corresponding terrain then the horizontal ground reaction forces, RY, may be significantly altered and this could lead to athletic injury. This could manifest itself in the anterior shin muscles if insufficient grip is present at heel strike as the tibialis anterior is stretched beyond its normal range. The ligaments in and around the knee may also be over stretched causing muscular or skeletal imbalances which create chondramalacia patella. The hamstrings may also be over exerted causing trauma and the EMG and force profiles produced here display a major peak early in the gait cycle. Thus overloading this group could lead to hamstring muscle 'pulls'.

Other areas where the running shoe can influence muscle and joint forces can be found with respect to the 'life' of a shoe. Harrison & Lees (1984) found that running shoes with EVA midsoles can lose up to 40% of that resistance after approximately 1200 km of use and Cook et al. (1985) found that a similar amount of shock absorption was lost after only 800 km. In both cases tread and upper appearance was good giving no indication of loss of performance. If the average athlete is completing 70 km per week, on average, 14 weeks of usage can be expected from a running shoe with EVA midsoles. The effects on the musclo-skeletal system of this change in cushioning is unknown and in need of further investigation. As outlined in the review of current literature the major running shoe manufacturers have introduced alternative materials into the midsoles to improve the 'life' of their running shoes. This may be gas or liquid encapsulated midsoles or other energy return systems.

The effects of all the various parameters involved in the construction of a running shoe can only be evaluated with the use of a biomechanical model (Nigg, 1986). The biomechanical model developed here will allow these effects to be studied so as to provide information on the reduction of the likelihood of injury.

Although prevention through an understanding of muscle and joint forces and their reliance upon such parameters as speed, footwear terrain and style can be sought, common preventative measures exist. Agre (1985) expresses the common causes of hamstring injuries as "poor flexibility, inadequate muscle strength and endurance, dys-synergic muscle contraction during running, insufficient warm up and stretching prior to exercise, awkward running style and a return to activity before complete rehabilitation following injury". This could also relate to any muscle and joint injury and similarly the treatments suggested by Agre are correspondingly valid for all injuries: "..the best treatment for hamstring injuries is prevention, which should include training to maintain and/or improve strength, flexibility, endurance, co-ordination and agility".

The use of a biomechanical model to develop a data base of information on muscle and joint force values in fit athletes may help to identify any abnormalities/anomalies in other athletes that could be leading to a muscular-skeletal injury. A longitudinal study of athletes with and without a history of injury may assist in the data base development.

6.8 Further Work

Although a biomechanical model has been produced further work with this model, and method, may well provide interesting data. For example the effects of footwear upon musculo-skeletal forces can now be evaluated. Hence varying parameters such as grip (co-efficient of friction of the outsole via materials or tread patterns); heel lift (training versus racing shoes); age (effects of usage and mileage on loss of cushioning); construction (board, slip or combination lastings); materials (EVA, blown rubber, air soles) and orthotic inserts may produce results which will provide advice on the selection and the design of running shoes to minimise injury.

Similarly the effects of sports surfaces can be evaluated to assess both the kinematics and the kinetics of running on different terrain. Hard surfaces such as tarmacadam and concrete are easily tested, as are synthetic surfaces such as synthetic running tracks. Topping a force platform with softer materials such as artificial turf is also feasible if the remainder of the runway is also covered. However, analysis of natural grass and other organic surfaces presents problems and will need careful scrutiny before force platforms can be topped with such terrain. The use of accelerometers cross calibrated against known force platform data may be of use here.

As suggested in Section 6.3 analysis between speeds of a greater range than employed in this study (3.83, 4.45 and 5.36 $m.s^{-1}$) must be undertaken to elicit the effects of velocity on the musculo-skeletal force system. Such speeds as walking (1.3 $m.s^{-1}$), jogging (3 $m.s^{-1}$), running (6 $m.s^{-1}$) and sprinting (in excess of 7 $m.s^{-1}$). If closer speeds are employed it is suggested that the subject group size should exceed the nine used here or the number of runs performed by all the subjects exceeds the 60 (on average) performed here.

Agre (1985) suggests that one cause of injury is inadequate muscle endurance and as such having the athletes perform the data collection test both with and without the effects of fatigue may also elicit significant results. Quantifying the amount of fatigue may however present problems and as the length of an average test was 20 minutes for 10 runs at each speed the athlete may either recover or tire depending upon level of fitness. This test may thus lend itself to analysis through accelerometers and treadmill running whilst physiological monitoring of, for example, oxygen uptake levels takes place.

A longitudinal study of athletes with and without injury histories and studies of elite and non elite or novice athletes may also be enlightening as an insight into such factors as running style and muscle strength, endurance, agility and fitness level. All these above mentioned studies will however be longwinded in nature due to the number of trials in need of analysis for repeatability and the nature of the data collection and analysis (film techniques). As such cross correlation of a number of variables may reduce the overall workload.

Whilst no literature exists of muscle and joint force analyses in the lower limbs during running, verification of the biomechanical model produced here is difficult. Employment of this model (and method) by other independent investigators will help validate the model to some extent. Although not in the running domain the model has been used by Nel (1985); Brandon (1986); Sharples (1987) and Thomas (1988) in independent analysis of different aspects of weight lifting and squatting exercises. Their results conform with results obtained by other experimenters using separately developed biomechanical models (Ariel, 1974). This can also be seen as approaching full verification of this model.

Further verification may also need to incorporate individual EMG testing and the production of algorithms derived from the EMG traces of a muscle's activity or non activity. This would probably require a larger computer memory than is presently available to this department if a microcomputer based system is sought. The rapid development of microcomputers should however make this feasible.

The use of individual EMG profiles would produce a need for individual shoes incorporating a foot switch. the shoe/footswitch utilised in this study does not lend itself to use by athletes with varying foot sizes. It is also capable of only being used with rear foot strikers and as such a more flexible footswitch is in need of development that can be transferred to any shoe.

A final refinement to the model and method could be developed due to the need to provide rapid feedback of results to athletes, coaches and physiotherapists. The present method of analysis is slow due to the use of high speed cinematographic techniques (developing of the film etc.) and the slow nature of digitisation (each run takes approximately one hour to digitise carefully). It is possible thus that feedback to the subjects may take longer than five days which may not be acceptable. The development of on-line systems with digital cameras may assist in reducing this turnaround time.

7. CONCLUSIONS

A bioengineering analysis of muscle and joint forces acting in the human lower limbs has been performed to allow a scrutiny of the gait cycle involved in running to take place. This analysis produces results for ground reaction forces and loading rates, joint moments, limb angles, muscular tensions and loading rates and joint reactions. From this an understanding of the relationships between athlete activity and injury has been developed.

A biomechanical model has been produced. It is a fully dynamic one which includes all the major muscles within the human lower limbs including the soleus muscle and the anterior shin group. The model has been solved by an inverse dynamics approach to solving a rigid body simplification of the human lower limbs. Standard biomechanical assumptions have been employed which will allow replication by other investigators and allows comparison with other studies and models in related fields.

An analysis method based on standard biomechanical laboratory equipment and techniques has been developed. This method, combined with the biomechanical model produces results repeatable to within ± 3% per subject per speed run. The equipment used was basically a Kistler force platform, a Locam high speed cinematographic camera, a magnetorestrictive digitising tablet, individual tailored bone models, electromyography and foot switches. All the primary analysis was performed with the aid of laboratory microcomputers with the final cubic spline combining of the data being performed on a DEC 20 mainframe computer.

This study has shown maximum joint reactions to be present in the knee, the mean compressive or normal force component being 21.3 times body weight (BW) when the final biomechanical model is employed. The highest muscle force can be found in the quadriceps group at 15.0 BW whilst the tricep surea group exhibit 10.7 BW and the hamstrings 2.8 BW. The shin muscle group, active for approximately the first seven percent of the stance phase displays a low level of activity of 0.5 BW. Other joint reactions include 2.4 BW for the shear component at the knee, 4.1 BW for the patella-femoral reaction and 11.5 BW and 6.4 BW for the ankle joint compressive and shear elements respectively.

These results, along with the other kinetic and kinematic data presented in Sections 5.1 to 5.4 provide the first fully dynamic bioengineering analysis of muscle and joint forces acting in the human lower limbs during a sporting activity.

Evaluation of external performance parameters such as speed and style have been undertaken with rear foot strikers being suggested as less likely to incur injury than front foot strikers due to their longer foot contact time and more gradual loading rates of both muscle and joint.

Elevated speeds increase these muscle and joint forces also but generally modestly and the speeds employed here, 7, 6 and 5 minutes per mile pace do not show significant differences in the majority of peak values. It is concluded that it is not necessarily the higher velocities that are more likely to cause injury but rapid or sudden changes in training that elevate the forces present in the lower limbs above values they are accustomed to.

More significant differences occur within the anterio-posterior aspect and the muscular loading rates and it is suggested that it is these forces that are more likely to cause injury than purely the vertical impact forces. These vertical impact or drive off forces do not increase in proportion to the increase in speed from walking and running. It is suggested that the body reaches a force limit and by altering the kinematics of lower limb angles the maximum forces are shared by all the load bearing joints.

From these conclusions, recommendations for the minimisation of the likelihood of injury are:

1. Design of running shoes to maximise on grip and therefore minimise sudden variations in the anterio-posterior or shear forces.
2. Careful selection of running footwear to maximise on grip (from (1) above) and to minimise changes in heel lift between training and racing shoes.
3. Avoiding sudden changes in training routine, speed or terrain, to minimise the training error injury risk. Gradual introduction to these regimes must be considered (i.e. progressive strengthening).
4. The incorporation of flexibility and strengthening programs to assist (3) above.
5. Selecting shoes purely on their shock absorbing properties should be avoided.

A wealth of data has now been amassed through this study which can be used in various ways. Limb accelerations, centre of pressure profiles, other limb kinematics (angles), muscle length changes and EMG profiles can all be analysed from the data base created here and may form a basis for further work.

8. RECOMMENDATIONS FOR FURTHER WORK

Before any more parameters such as footwear and surfaces are analysed the biomechanical model and method should be further verified. An analysis of the walking gait would enable the model and method to be compared against the standard reference works of Morrison (1968,1969), Paul (1967,1974) and Harrington (1976).

The reliability of the data analysis procedure should also be evaluated by allowing a number of independent experimenters to perform the analysis. A level of reliability, in comparison to the ± 3% figure for repeatability, may then be presented.

Validation of the model should also be sought either by direct experimentation (force transducer implantation) or via EMG to force processing as attempted by Hof and his co-workers (Hof, 1987).

Once the validity, reliability and repeatability of the procedures have been established, further parameters may be considered. The effects of footwear (construction and age), surfaces, fatigue, level of fitness and injury history may all produce useful results. Deliberately altering the subject's running action, by causing slippage, stuttering or over-stretching may highlight other factors associated with running style.

Finally, the data collection, processing and analysis time should be reduced from approximately the five days now required to one day. This may be achieved with the aid of high speed digital video cameras linked to micro-computers. This would reduce the waiting time for the cinematographic film - the prime cause of the long turnaround time.

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APPENDICES

APPENDIX A1

ATHLETE SHOE & INJURY FACT SHEETS

A1.1 Buying a New Running Shoe

Due to the popularity of the sport specialised running shoe companies and general sport shoe manufacturers are attempting to capitalise on the upsurge in athletic footwear demand. As such the athlete is now faced with a bewildering selection of brands and models of running shoes to choose from. But which is the best or correct shoe for a particular athlete? Outlined below is an 8 point guide to buying a new pair of running shoes.

POINT	OBSERVATION TO MAKE	CONCLUSION
1.	LOOK AT YOUR OLD SHOES	
	The heel counters turn in	You pronate excessively and need a strong heel counter.
	The heel counters turn out	Your foot is too rigid - avoid antipronation devices.
	The uppers are badly stretched	You need wider shoes
	Hole in toe box	You need bigger toe box
	Midsole & Wedge hard and brittle	Shoes are too old
	Uneven compression on midsole and wedge	Need a 'soldier' shoe
2.	LOOK AT YOUR OWN BODY	
	You often get blackened toenails	Need bigger toe box
	Toes or forefoot goes numb when running	Need wider shoes
	Body weight higher than most	Avoid light shoes
	Feet different sizes	Try both shoes before buying
3.	CONSIDER YOUR INJURY HISTORY	
	Knee injuries	You need narrower shoe with snug heel counter
	Shin splints	Shoes need more grip
	Achilles tendon problems	Shoe needs heel lift and reasonable heel flare
	General ankle pains	Shoe needs reasonable heel flare and snug fitting heel counter
	See common causes of injury fact sheet attached.	
	CONSIDER YOUR RUNNING STYLE	
	Most outsole wear is at heel	Rearfoot striker - good heel protection needed
	Most outsole wear is at mid/forefoot	Forefoot striker - don't need shoe with antipronation device

CONSIDER YOUR TRAINING REGIME	
Running surface predominantly: Road	Impact properties are important: racing type O.K
Running surface predominantly: Grass etc.	Grip properties more important: waffle needed
GO TO GOOD RUNNING SHOE SHOP	
Choose a well stocked specialised shop	They have most shoes
Avoid mail order	Unless you are reordering
Take this guide with you	For help
TRY AS MANY SHOES ON AS POSSIBLE	
Try both shoes	In case of differences
Check quality	In & out stitching glueing of sole
Check function	ie. they stand straight to start with
WHEN BOUGHT YOUR SHOES	
Treat them well	Clean and dry
Run them in	Like a car, for longer life
Check points 1-5 regularly	Discard if not happy - don't wait for injury
Discard when cushioning lost	See running shoe life face sheet attached

Conclusion

In conclusion, "you get what you pay for", so buying the best you can afford is usually very good policy. No shoe is the best for everyone,but there will be one that suits you. With the aid of the above checklist you should be able to purchase a pair of running shoes that will give you many miles of trouble free running.

A1.2 Common Causes of Injury

Although a very healthy segment of the population, athletes are continually filling doctors surgerys with complaints that would be tolerated by almost any other member ofthe public but are viewed a serious handicap to the runner, and with the increase in popularity of running the number of athletic injuries is rising. Outlined below are the common causes of these injuries and guidelines to help athletes avoid contracting them.

COMMON CAUSES		
Cause	**Comments**	**Preventative Measures**
Training Errors	60% of injuries caused by excessive mileage.	Less miles, more speed is better.
	20% of injuries caused by other errors i.e.	
	(i) Rapid change in schedule: body not used to.	Change training gradually
	(ii) Intense session: insufficient warm up	Warm up well
	: Body too cold	Keep warm
	: Body not recovered	Have easy day after hard one
	(iii) Hill running: Requires different muscle parts	Start gradually
Anatomical factors	Excessive pronation: weak ankles	Strengthen ankles and get shoe with anti pronation device
	Poor flexibility: muscle tightness	Do plenty of stretching
	flat or high arched feet: need orthotics	
	Different leg lengths: need orthotics	See specialist
Shoes	Poor cushioning can = joint injury Poor grip = shin splints Poor fit = knee pain	Buy new shoes, good fitting or have own resoled to improve cushioning/grip

With a small amount of thought and time an athlete can reduce the possibility of contracting injury. Overall flexibility and strengthening exercises and correctly chosen footwear coupled to thoughtful training routines can lead to an enjoyable, injury free, athletic career.

A1.3 Running Shoe Life

Average club runners are recording in excess of 40 miles per week in training and thus in a matter of twelve weeks nearly 500 miles can have been completed. But how long does or should a running shoe last? When does the protection afforded by the midsole etc., begin to deteriorate? How long will the tread be effective? What should be done when the shoes are at the end of their life? Below are some answers.

Figure 1 shows tread wear rates for two running shoes tested at Liverpool Polytechnic and as can be seen shoe B has far better wear characteristics, a life in excess of 1500. Shoe A appears to have a life of 1000 miles and it is usually at this stage when no tread is left on the shoe that it is discarded. In contrast the

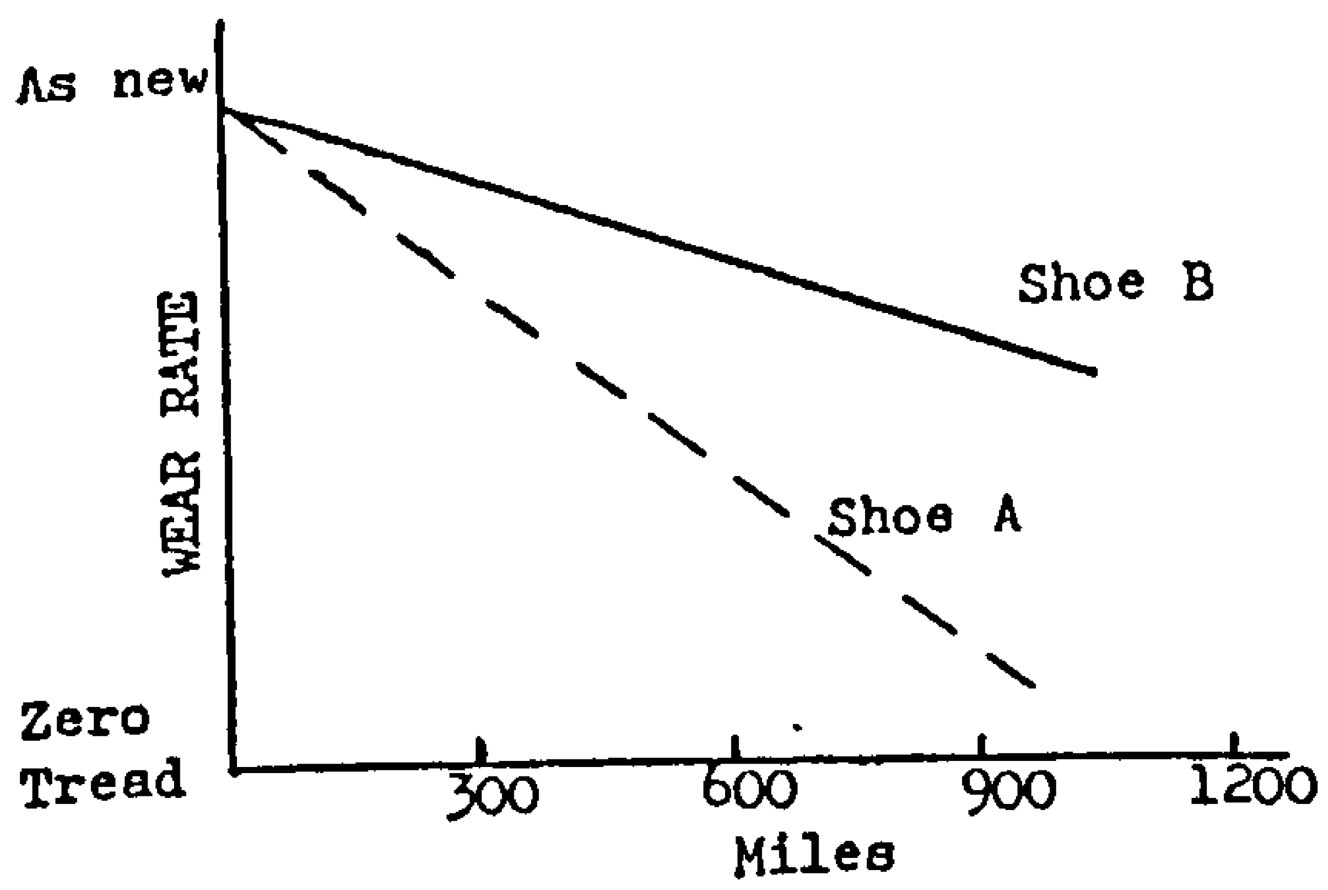

Figure 1

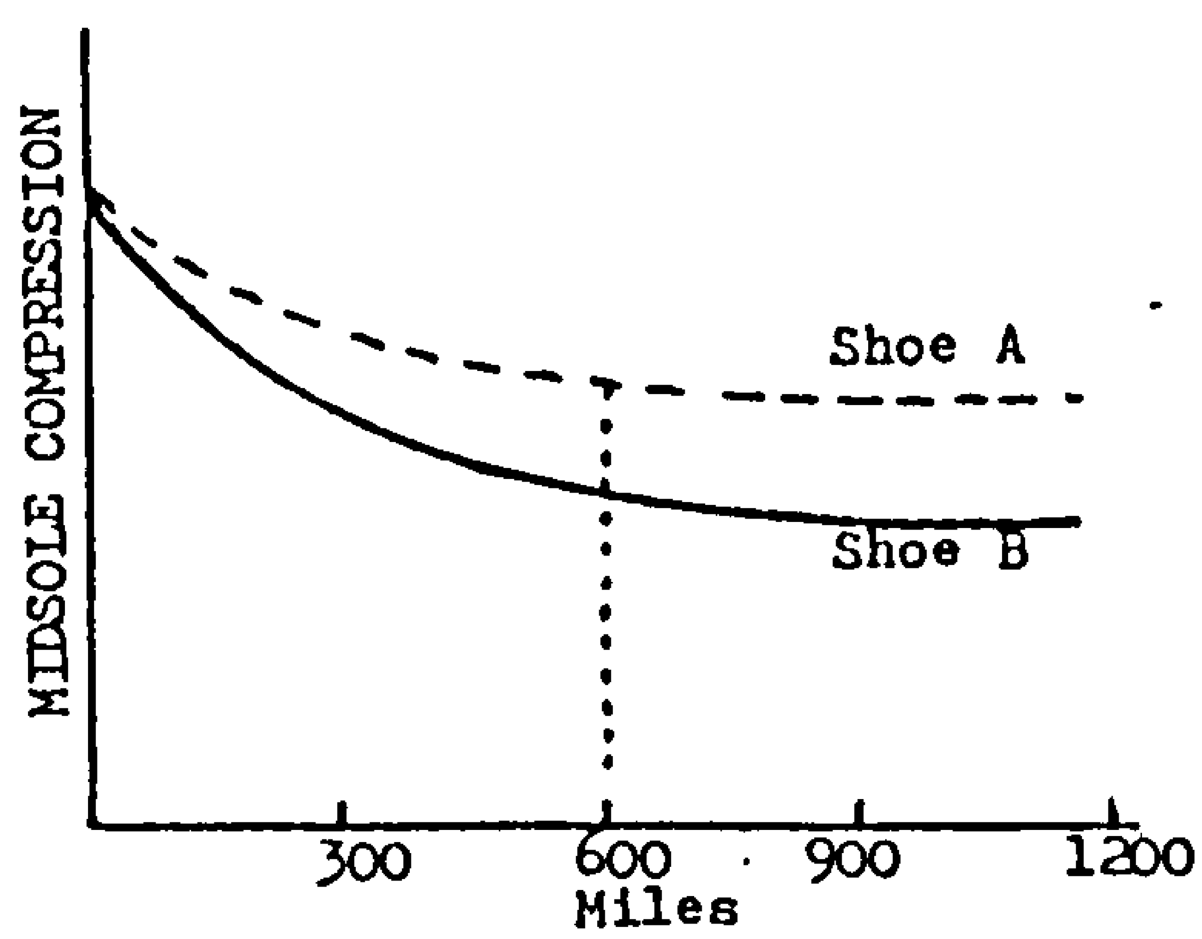

Figure 2

compression set results (figure 2), which indicate how much the midsole of the shoe have compressed from their original thickness, show that after only about 600 miles shoe A is fully compressed and therefore providing very poor cushioning or protection to the body. Hence the shoe lives can be said to be in the region of 600-800 miles governed by their cushioning.

In conclusion it can be seen that the life of a running shoe should not be governed by its external appearance, in this case tread depth, and should be discarded or have new midsoles/outsoles fitted after approximately 600 miles. If the price of an average training shoe is taken to be £40 then this equates to 6 pence per mile: very economic in relation to other forms of entertainment.

APPENDIX A2. Sample Results.

------------ RESULTS FOR TEXT.TXT FILE --------

------------ BODY WEIGHT= 799.515 N. ------------

1	GA/SH	RF	PT	HM	GL	RZ	RY
1	228.476	268.6972	1074.789	953.2175	685.9243	92.08027	15.15219
2	318.3642	409.9648	1639.859	1243.027	1172.979	364.4497	39.87143
3	596.4264	1000.259	4001.037	2280.147	2239.334	1069.558	-45.4252
4	613.3696	1362.301	5449.206	2514.128	2565.402	1989.816	-176.0404
5	448.1711	1441.833	5767.331	1689.774	1805.62	2397.663	-254.4706
6	461.4117	1755.417	7021.667	981.7544	1077.948	2280.343	-423.3854
7	540.4631	1912.472	7649.888	537.4715	584.2975	1749.185	-454.2082
8	931.5237	1872.323	7489.293	333.9551	361.6909	1394.265	-340.8354
9	1498.581	1979.958	7919.833	309.5362	342.0902	1613.076	-427.963
10	2055.924	2234.225	8936.899	297.4972	328.0779	1850.503	-581.1613
11	2705.707	2648.038	10592.15	258.431	266.3619	1942.514	-428.1156
12	3339.519	3061.354	12245.42	279.2994	273.0455	2248.147	-410.568
13	3824.416	3091.758	12367.03	287.5855	284.3195	2419.961	-316.7265
14	4234.379	3006.723	12026.89	397.4846	401.7979	2528.604	-257.675
15	4695.199	2964.965	11859.86	526.6301	532.6049	2571.481	-203.6588
16	5288.85	2864.337	11457.35	454.6618	456.5109	2570.871	-159.4084
17	5953.019	2728.39	10913.56	357.217	353.4355	2522.805	-124.1606
18	6609.837	2664.451	10657.8	389.8477	374.0651	2486.184	-117.7519
19	7211.461	2652.486	10609.95	409.4462	378.6368	2441.323	-94.71109
20	7578.072	2515.634	10062.54	470.0238	437.6188	2354.043	-22.6896
21	7644.638	2220.204	8880.816	621.2085	611.5153	2287.973	21.4083
22	7750.602	1999.731	7998.925	718.2362	732.1112	2225.717	54.51987
23	7941.973	1830.228	7320.911	830.0508	835.8923	2132.791	94.34531
24	7913.589	1559.223	6236.891	1028.308	1017.099	2012.399	154.4649
25	7561.148	1280.725	5122.899	1143.69	1137.769	1869.882	207.1078
26	6987.029	1083.373	4333.494	1071.031	1087.556	1724.466	237.6253
27	6305.817	912.6202	3650.481	894.745	912.6509	1565.011	254.8678
28	5481.211	699.8012	2799.205	679.7429	683.873	1374.276	288.7423
29	4561.147	453.674	1814.696	475.4283	473.8622	1174.997	306.2899
30	3720.304	248.5375	994.1499	318.9911	321.1628	974.1909	292.557
31	3031.236	131.1568	524.6272	200.9187	207.3846	770.1808	280.3499
32	2458.123	110.3147	441.2587	118.7692	123.4306	576.6994	250.5953
33	1935.71	140.1643	560.6571	100.4852	99.05455	415.8718	213.9742
34	1443.133	162.9616	651.8464	125.4934	115.9323	291.8178	168.8082
35	1018.34	157.8733	631.4933	144.9953	131.9402	197.6711	129.5931
36	676.5327	134.2796	537.1183	149.7302	138.3523	127.6332	89.15732
37	414.948	104.7268	418.9071	159.6147	147.9562	81.85688	56.80669
38	227.2377	75.8186	303.2744	182.4056	166.5731	47.21943	29.03769
39	115.6378	54.80898	219.2359	202.0505	182.0743	17.92255	14.23667
40	82.03273	47.2176	188.8704	204.394	183.9481	2.053411	2.029634
40							

J	SL	AZ	AY	KZ	KY	KN	KS
1	0	-543.0023	199.4027	-596.5451	-450.0934	-538.6497	-501.3868
2	0	-740.2169	289.3452	-1049.816	-643.01	-970.2737	-744.2315
3	0	-1407.283	605.5623	-3166.5	-1392.557	-3019.762	-1641.128
4	19.73542	-2101.354	677.8592	-5558.521	-1582.931	-5428.868	-1797.411
5	98.04069	-2414.927	481.8291	-7455.583	-1391.771	-7443.641	-1233.351
6	185.9178	-2443.653	403.1015	-8942.208	-1976.693	-9075.278	-1239.321
7	255.0778	-2287.608	325.8067	-9222.164	-2546.579	-9493.412	-1274.136
8	401.2842	-2526.971	195.5644	-8907.417	-2813.094	-9281.762	-1114.502
9	747.9616	-3130.482	55.34351	-9411.44	-3260.113	-9913.105	-995.9461
10	1027.577	-3788.365	-144.0489	-10639.22	-3958.054	-11318.02	-908.5001
11	1352.744	-4556.872	-465.0703	-12553.69	-5139.292	-13543.59	-933.2428
12	1669.729	-5327.586	-817.7575	-14399.77	-6425.176	-15742.83	-1020.179
13	1912.2	-5908.345	-1146.154	-14869.05	-6766.552	-16316.15	-781.3276
14	2117.187	-6350.448	-1466.535	-14965.24	-6739.124	-16402.32	-317.9819
15	2347.599	-6707.369	-1824.781	-15086.66	-6943.348	-16606.86	-10.88283
16	2644.425	-7243.768	-2251.129	-14684.8	-7304.666	-16408.68	-129.7946
17	2976.529	-7738.05	-2719.204	-14010.01	-7618.301	-15953.28	-348.7057
18	3304.918	-8202.684	-3189.259	-13672.46	-7934.922	-15810.68	-399.3971
19	3605.73	-8603.873	-3638.042	-13556.73	-8349.578	-15922.17	-473.2523
20	3789.036	-8796.161	-3965.325	-12835.84	-8495.695	-15383.05	-720.9332
21	3622.319	-8738.607	-4117.654	-11446.99	-8131.476	-14011.94	-980.2533
22	3875.301	-8706.333	-4286.166	-10517.79	-7768.718	-13037.95	-1018.565
23	3970.967	-8742.417	-4480.041	-9911.913	-7375.383	-12321.98	-904.702
24	3956.794	-8600.12	-4515.747	-8869.28	-6574.381	-11012.33	-781.4481
25	3780.574	-8147.203	-4400.348	-7715.449	-5679.532	-9559.503	-616.1264
26	3493.514	-7454.987	-4211.279	-6815.862	-5015.381	-8450.704	-420.3083
27	3157.924	-6673.462	-3930.955	-5962.168	-4451.459	-7434.382	-303.4108
28	2740.605	-5796.613	-3507.242	-4960.928	-3707.645	-6190.51	-209.6102
29	2280.574	-4856.752	-2979.091	-3888.387	-2762.845	-4770.978	-61.53169
30	1860.152	-3979.984	-2477.039	-2925.109	-1900.979	-3488.906	81.22149
31	1515.618	-3231.198	-2048.705	-2139.918	-1228.726	-2461.755	185.3131
32	1229.062	-2596.246	-1669.395	-1519.203	-706.88	-1656.562	261.4083
33	967.8552	-2021.334	-1321.711	-1025.797	-305.7827	-1024.501	314.281
34	721.5064	-1492.208	-999.8152	-646.7404	-43.83822	-564.2858	320.9733
35	509.17	-1050.062	-711.7661	-377.7176	76.42678	-275.4771	271.8056
36	336.2064	-703.5507	-471.7352	-207.159	116.2931	-114.096	210.7938
37	207.474	-437.5499	-289.5907	-125.713	117.4274	-46.74241	166.9025
38	113.5122	-242.7878	-160.3134	-92.73275	83.66245	-36.19348	120.2319
39	57.31239	-123.0355	-81.73779	-76.80638	46.29718	-41.24399	80.06483
40	40.22993	-87.17619	-57.33181	-72.21666	33.51686	-43.90367	66.72011
40							

I	AM1	KM1	PM1	PF	KAN	AAN
1	0.140631	15.57277	100.7861	489.3097	168.6216	157.0191
2	0.513149	15.29773	126.4886	746.7903	164.6698	155.808
3	15.00622	-3.335725	195.8538	1741.839	162.1275	155.1795
4	12.24992	-42.94778	235.6032	2209.717	159.7504	155.083
5	-1.338996	-101.1816	196.7402	2069.139	155.0579	154.8605
6	-11.57357	-161.6466	91.94371	2277.149	150.5089	154.3065
7	-17.98561	-193.2416	-0.8411915	2322.55	144.9855	153.1554
8	-33.31466	-206.855	-61.14628	2063.875	143.608	154.6053
9	-54.48269	-235.4733	-116.2049	1918.037	143.8009	157.2082
10	-74.06461	-258.6485	-130.3692	2020.807	141.6333	157.1644
11	-95.98795	-267.0979	-96.36026	2491.792	139.405	156.7791
12	-117.7323	-277.693	-65.20342	2981.432	137.0028	157.6087
13	-138.8667	-277.076	-32.54477	2859.583	137.7049	159.6728
14	-157.9527	-274.2617	-3.764068	2472.362	138.2853	160.9561
15	-174.6626	-269.9412	14.25357	2276.806	135.676	160.5399
16	-191.3551	-255.9767	14.41937	2336.704	136.0307	162.2365
17	-208.3465	-239.3022	0.1016644	2308.241	138.1573	165.2697
18	-223.8153	-225.9354	-11.87887	2127.162	138.3799	167.1834
19	-236.3606	-209.1544	-8.47903	2155.574	137.6842	167.6903
20	-244.9917	-187.2746	-6.756981	2356.532	137.4657	168.706
21	-250.3365	-164.6533	-18.8229	2432.887	141.0987	172.6265
22	-253.5165	-142.0144	-26.55008	2361.35	143.9171	175.6845
23	-253.7416	-114.842	-25.67471	2148.102	143.8748	176.8389
24	-249.8111	-81.37797	-26.50285	1783.242	147.7455	180.48
25	-242.2943	-52.37622	-29.33089	1388.069	153.8736	185.7531
26	-230.5867	-34.16414	-24.9111	1102.728	154.7737	188.2325
27	-212.9027	-19.3139	-8.353279	975.6585	154.0297	188.8558
28	-191.2568	-1.07446	16.17986	846.6774	154.5096	190.1095
29	-168.6651	18.74109	38.9372	592.7352	157.4888	192.1784
30	-144.9664	34.63437	53.87109	323.2851	160.2698	194.1531
31	-120.8144	45.9259	63.689	111.6774	163.4962	197.6688
32	-97.58736	52.42751	70.37752	-47.29592	166.2343	200.1023
33	-76.92684	52.69427	70.60159	-169.1965	168.8507	202.0309
34	-56.17281	47.25567	63.33979	-231.18	170.2697	203.796
35	-42.10998	39.19326	51.5641	[illegible]	172.4573	205.7575
36	-28.84119	29.55895	37.18322	-198.329	177.994	209.5838
37	-18.65931	19.4516	23.33633	-178.134	180.5889	211.0005
38	-10.19771	11.08291	12.65701	-141.7774	181.125	211.4204
39	-5.220149	5.932585	6.976933	-103.581	179.9661	212.6543
40	-3.605366	4.254372	5.075177	-90.22922	180.475	212.1996

QUADS. MAX LOADING RATE...= 503554

CALF MAX LOADING RATE.....= 93464.12

APPENDIX A3. Computer Program

```
00090 GOTO 200
00100 PRINT" ************************************************************************"
00110 PRINT" ************************************************************************"
00120 PRINT" ********                                                        *********"
00130 PRINT" ********       OPTTDS  (SPLINE FORCES) PROGRAM                 *********"
00140 PRINT" ********       RESULTS GIVE MUSCLE & JOINT FORCES, JOINT MOMENTS *********"
00141 PRINT" ********       & MAXIMUM LOADING RATE FOR QUADS. & ACHILLES PLUS *********"
00143 PRINT" ********       PATELLA-FEMORAL CONTACT FORCE AND               *********"
00145 PRINT" ********       KNEE & HIP ANGLES OF FLEXION .   LIMB ACCELS    *********"
00147 PRINT" ********       ALSO CALCULATED BUT NOT PRINTED OUT.            *********"
00160 PRINT" ********                                                        *********"
00170 PRINT" ************************************************************************"
00180 PRINT" ************************************************************************"
00182 SLEEP 12
00185 GOTO 695
00190 REM
00200 REM ============================================================
00210 REM ==========      DEFINE SCREEN CONTROL      ===================
00220 REM ============================================================
00240 COMMON (V.AREA) V.HOMES(2)=4,V.ERASS(2)=3,V.ERALS(2)=3,TERMTYP%,V.CLRS(2)=7
00245 ESCS=CHRS(27) \ TERMTYP%=2%
00250 V.HOMES(1)=CHRS(13)+ESCS+"[H" \ V.HOMES(2)=CHRS(13)+CHRS(30)+CHRS(0)+CHRS(0)
00255 V.ERASS(1)=ESCS+"[J" \ V.ERASS(2)=ESCS+'Y'+CHRS(0)
00260 V.ERALS(1)=ESCS+"[K" \ V.ERALS(2)=ESCS+'T'+CHRS(0)
00265 V.CLRS(1)=ESCS+"[2J"+ESCS+"[H"
00270 V.CLRS(2)=CHRS(26)+CHRS(0)+CHRS(0)+CHRS(0)+CHRS(0)+CHRS(0)+CHRS(0)
00275 OPEN "TTY:" AS FILE #1
00280 X=NOECHO(1%)
00285 PRINT #1,ESCS;"Z";
00290 SLEEP 1%
00295 LINPUT #1,"HIT RETURN";TTS
00300 TTS=LEFTS(EDITS(TTS,32%),3)
00310 IF NOT(TTS=(ESCS+"[?")) THEN 333
00320 TERMTYP%=1%
00333 X=ECHO(1%) \ CLOSE #1
00340 DEF FN.CLRS=V.CLRS(TERMTYP%)
00350 DEF FN.HOMES=V.HOMES(TERMTYP%)
00360 DEF FN.ERALS=V.ERALS(TERMTYP%)
00370 DEF FN.ERASS=V.ERASS(TERMTYP%)
00380 DEF FN.GOTOXYS(X%,Y%)
00390 IF TERMTYP%=2% THEN FN.GOTOXYS=CHRS(13)+ESCS+'='+CHRS(Y%+32)+CHRS(X%+32) \ FNEXIT
00400 FN.GOTOXYS=CHRS(13)+ESCS+'['+STRS(Y%+1)+';'+STRS(X%+1)+'H'
00410 FNEND
00460 REM ============================================================
00470 REM ==========      END OF SCREEN CONTROL        =================
00480 REM ============================================================
00490 REM
00500 REM
00510 REM
00520  REM ***********************************************************
00530 REM
00540 REM INITIALISE & OPEN FILES AND DIMENSION VARIABLES
00550 REM
00560 REM ***********************************************************
00565 DIM FL1S(20),FL2S(20),FL3S(20)                  \REM AUTO RUN FILE COUNTER
00570 PRINT FN.CLRS
00580 REM
00590 REM ==============================================================
00600 REM ================   INITIALISE PROGRAM   =====================
00610 REM ==============================================================
```

```
00620 REM
00625 GOTO 100
00695 PRINT FN.CLR$
00700 PRINT \ PRINT "----------------------------------------------------------------------"
00710 PRINT \ PRINT "FIRST YOU MUST IDENTIFY THE TWO GROUPS OF DATA FILES FROM  ----------"
00720 PRINT \ PRINT "WHICH DATA IS TAKEN (THIS OPENS THE DATA FILES FOR READING) ----------"
00730 PRINT \ PRINT "A NEW GROUP OF FILES  MUST BE OPENED ALSO FOR THE RESULTS  ----------"
00740 PRINT \ PRINT "TO BE SENT TO (SEE BELOW FOR THIS NEW NAME)              ---------------"
00750 PRINT \ PRINT "- - - - - - - - - - - - - - - - - - - - - - - - - - - - - - - - - - - -"
00752 PRINT \ INPUT "HENCE INPUT NO. OF FILES TO BE READ       ";FRN
00754 FOR FRC=1 TO FRN
00756 PRINT \ PRINT "FILE SET NO. ";FRC
00757 PRINT "$$$$$$$$$$$$$$$$$$"
00760 PRINT \ INPUT "HENCE INPUT @DIGITISE@ DATA FILE (IDS.BBC)";FL1$(FRC)
00770 PRINT \ INPUT "AND INPUT @CIL-16@ DATA FILE (IDS.BPC)    ";FL2$(FRC)
00780 PRINT \ INPUT "AND INPUT A RESULTS 'TEXT' FILE (IDS.TXT) ";FL3$(FRC)
00790 PRINT "                                            *****************"
00800 PRINT \ INPUT "IS THIS ACCEPTABLE ? Y/N . . . . . . . . . .";AS$
00810 IF AS$="Y" THEN GOTO 830
00820 GOTO 760
00830 NEXT FRC
00835 FRC=0
00840 REM ============================================================
00850 REM ============      OPEN DATA FILES     ========================
00860 REM ============================================================
00862 FRC=FRC+1
00864 IF FRC>FRN GOTO 1310
00866 FL1$=FL1$(FRC) \ FL2$=FL2$(FRC) \ FL3$=FL3$(FRC)
00870 REM
00880 OPEN FL1$ FOR INPUT AS FILE #1,ACCESS READ
00890 OPEN FL2$ FOR INPUT AS FILE #2,ACCESS READ
00900 OPEN FL3$ FOR OUTPUT AS FILE #3
00910 MARGIN#3,132
00920 REM
00930 REM ============================================================
00940 REM ===========     DIMENSION VARIABLES     =======================
00950 REM ============================================================
00960 REM
00970 DIM FX(150),FY(150),FZ(150),MX(150),MY(150)        \REM F.P. INPUT
00980 DIM RX(100),RY(100),RZ(100)                      \ REM F.P. ACTUAL DATA
00990 DIM AZ1(100),AZ2(100),PL(100),PW(100)             \ REM F.P. COORDINATES
01000 DIM PX(100),PY(100)                             \ REM F.P. PTS.OF APPC.
01010 REM
01020 DIM A1(100),A2(100),A3(100),A4(100)
01030 DIM K1(100),K2(100),K3(100),K4(100)
01040 DIM K5(100),K6(100),K7(100),K8(100)               \ REM MOMEMNT ARMS
01050 DIM H1(100),H2(100),H3(100),H4(100)
01060 DIM H5(100),H6(100),H7(100),H8(100),H9(100)
01070 REM
01080 DIM B0(100),B1(100),B2(100)
01090 DIM B3(100),B4(100)                              \ REM ANGLES
01100 DIM B5(100),B6(100),B9(100)
01110 REM
01120 DIM AZ(100),AY(100),KZ(100)
01130 DIM KY(100),KS(100),KN(100),PF(100)               \ REM JOINT REACTIONS
01140 DIM GA(100),HM(100),RF(100),SL(100)
01150 DIM GL(100),SH(100),PT(100),AT(100)               \ REM MUSCLE FORCES
01160 REM
01170 DIM ZA(100),ZB(100),AA(100)
01180 DIM AB(100),YA(100),YB(100)                     \ REM ACCEL. DATA
```

```
01190 DIM RA(100),DA(100)                                    \ REM ACCEL. VARIABLES
01200 REM
01210 DIM XI(100),XB(100),DG(100)
01220 DIM DM(150),VC(150),VB(150)                            \ REM SPLINE VARIABLES
01230 DIM C(4,150)
01240 REM
01250 DIM ID$(25,2)                                        \ REM GENERAL ARRAY STR
01260 DIM AM2(100)                                          \ REM ANKLE FORCING MOMENT
01270 DIM AM1(100),KM1(100),PM1(100)                         \ REM JOINT MOMENTS
01280 DIM AAN(100),KAN(100),HAN(100)                         \ REM JOINT ANGLES
01290 PRINT \ GOSUB 10500
01300 IF SE=8 GOTO 1620
01310 PRINT FN.CLR$
01320 REM
01330 REM ==============================================================
01340 REM ============     DATA CONTROL MENU     =======================
01350 REM ==============================================================
01360 REM
01370 PRINT "                    DATA CONTROL MENU"
01380 PRINT "                 ************************"
01390 PRINT \ PRINT "READ DATA FROM FILES. . . . . . . . . . . . . . . . . . .1"
01400 PRINT \ PRINT "RE-ALLOCATE FILES FOR MANIPULATION . . . . . . . . . . . 2"
01410 PRINT \ PRINT "CALCULATE FORCES. . . . . . . . . . . . . . . . . . . . .3"
01420 PRINT \ PRINT "OUTPUT TO SCREEN ONLY. . . . . . . . . . . . . . . . . . 4"
01430 PRINT \ PRINT "OUTPUT TO DATA FILE ONLY. . . . . . . . . . . . . . . . .5"
01440 PRINT \ PRINT "OUTPUT TO FILE & SCREEN. . . . . . . . . . . . . . . . . 6"
01450 PRINT \ PRINT "END PROGRAM . . . . . . . . . . . . . . . . . . . . . . .7"
01455 PRINT \ PRINT "AUTO RUN (FOR MORE THAN ONE FILE. PRINT TO FILE ONLY)..8"
01460 PRINT \ PRINT \ PRINT
01470 INPUT "ENTER SELECTION + @CR@ ";SE
01480 IF SE<1 THEN GOTO 1470
01490 IF SE>8 THEN GOTO 1470
01495 PRINT FN.CLR$
01500 ON SE GOTO 1520,550,3870,5570,5150,5150,10510,1620
01510 REM
01520 REM ==============================================================
01530 REM ===                  READ DATA FROM FILES  ==================
01540 REM ==============================================================
01550 REM
01560 PRINT \ PRINT \ GOSUB 10500
01570 PRINT FN.CLR$
01580  PRINT "-------------------------------------------------------------"
01590  PRINT "READING DATA FROM FIRST FILE . THIS IS:-";FL1$
01600 PRINT  "AND CALCULATING ACCELERATION DATA"
01610  PRINT "-------------------------------------------------------------"
01620 INPUT #1,SN$
01630 INPUT #1,SD$
01640 INPUT #1,FO1
01650 INPUT #1,IC1
01670 FOR K=1 TO IC1\INPUT#1,ZA(K)\NEXT K
01672 FOR K=1 TO IC1\INPUT#1,YA(K)\NEXT K
01674 FOR K=1 TO IC1\INPUT#1,B3(K)\NEXT K
01676 FOR K=1 TO IC1\INPUT#1,ZB(K)\NEXT K
01678 FOR K=1 TO IC1\INPUT#1,YB(K)\NEXT K
01680 FOR K=1 TO IC1\INPUT#1,B4(K)\NEXT K
01682 FOR K=1 TO IC1\INPUT#1,K7(K)\NEXT K
01684 FOR K=1 TO IC1\INPUT#1,K8(K)\NEXT K
01686 FOR K=1 TO IC1\INPUT#1,AZ1(K)\NEXT K
01688 FOR K=1 TO IC1\INPUT#1,AZ2(K)\NEXT K
01690 FOR K=1 TO IC1\INPUT#1,PW(K)\NEXT K
```

```
01692 FOR K=1 TO IC1\INPUT#1,PL(K)\NEXT K
01694 FOR K=1 TO IC1\INPUT#1,A3(K)\NEXT K
01696 FOR K=1 TO IC1\INPUT#1,A4(K)\NEXT K
01698 FOR K=1 TO IC1\INPUT#1,K1(K)\NEXT K
01700 FOR K=1 TO IC1\INPUT#1,K2(K)\NEXT K
01702 FOR K=1 TO IC1\INPUT#1,K3(K)\NEXT K
01704 FOR K=1 TO IC1\INPUT#1,K4(K)\NEXT K
01706 FOR K=1 TO IC1\INPUT#1,K5(K)\NEXT K
01708 FOR K=1 TO IC1\INPUT#1,K6(K)\NEXT K
01710 FOR K=1 TO IC1\INPUT#1,H1(K)\NEXT K
01712 FOR K=1 TO IC1\INPUT#1,H2(K)\NEXT K
01714 FOR K=1 TO IC1\INPUT#1,H3(K)\NEXT K
01716 FOR K=1 TO IC1\INPUT#1,H4(K)\NEXT K
01718 FOR K=1 TO IC1\INPUT#1,H5(K)\NEXT K
01720 FOR K=1 TO IC1\INPUT#1,H6(K)\NEXT K
01724 FOR K=1 TO IC1\INPUT#1,H7(K)\NEXT K
01726 FOR K=1 TO IC1\INPUT#1,H8(K)\NEXT K
01728 FOR K=1 TO IC1\INPUT#1,H9(K)\NEXT K
01730 FOR K=1 TO IC1\INPUT#1,B0(K)\NEXT K
01732 FOR K=1 TO IC1\INPUT#1,B1(K)\NEXT K
01734 FOR K=1 TO IC1\INPUT#1,B2(K)\NEXT K
01735 FOR K=1 TO IC1\INPUT#1,B5(K)\NEXT K
01736 FOR K=1 TO IC1\INPUT#1,B6(K)\NEXT K
01738 FOR K=1 TO IC1\INPUT#1,B9(K)\NEXT K
01780 REM============================================================
01790 REM== SMOOTH DISPLACEMENTS THEN CALCULATE ACCELERATION DATA ======
01800 REM============================================================
01810 REM =======   DATA PREPERATION PRIOR TO ACCELERATION CALCS. ====
01820 REM ==========================================================
01830 ASV=2   \   REM  ARRAY STEP VALUE
01840 REM REASSIGN DATA FOR 'ASV' POINT ACCEL CALCS.
01850 IC2=INT((IC1-1)/ASV)+1
01860 FOR I=1 TO IC2
01870 AA(I)=B3(1+(I-1)*ASV)
01880 AB(I)=B4(1+(I-1)*ASV)
01890 ZA(I)=ZA(1+(I-1)*ASV)
01900 YA(I)=YA(1+(I-1)*ASV)
01910 ZB(I)=ZB(1+(I-1)*ASV)
01920 YB(I)=YB(1+(I-1)*ASV)
01930 NEXT I
01940 IC=IC2-4    \         ZC=IC1                  \   ZG=IC1-4
01950 REM ==========================================================
01960 REM ==== DATA PREP FOR SMOOTH.,ACCEL. & FIRST SPLINE FIT  =====
01970 REM ==========================================================
01980 REM
01990 REM ==========================================================
02000 REM ===============         FOR  ZA        ====================
02010 REM ==========================================================
02020 FOR I=1 TO IC2 \ DA(I)=ZA(I) \ RA(I)=0 \ NEXT I
02030 GOSUB 2660
02040 GOSUB 2780
02050 FOR I=1 TO IC \ C(1,I)=RA(I+2) \ XI(I)=I*ASV/FO1 \ NEXT I
02060 C(2,1)=(RA(3)-RA(4))*FO1/ASV \ C(2,IC)=(RA(IC)-RA(IC-1))*FO1/ASV
02070 GOSUB 5880
02080 FOR I= 1 TO ZG \ ZA(I)=VC(I) \ NEXT I
02090 REM ==========================================================
02100 REM =============          FOR  YA        =====================
02110 REM ==========================================================
02120 FOR I=1 TO IC2 \ DA(I)=YA(I) \ RA(I)=0 \ NEXT I
02130 GOSUB 2660
```

```
02140 GOSUB 2780
02150 FOR I=1 TO IC \ C(1,I)=RA(I+2) \ NEXT I
02160 C(2,1)=(RA(3)-RA(4))*FO1/ASV \ C(2,IC)=(RA(IC)-RA(IC-1))*FO1/ASV
02170 GOSUB 5880
02180 FOR I=1 TO ZG \ YA(I)=VC(I) \ NEXT I
02190 REM ========================================================
02200 REM =============          FOR   ZB       =====================
02210 REM ========================================================
02220 FOR I=1 TO IC2 \ DA(I)=ZB(I) \ RA(I)=0 \ NEXT I
02230 GOSUB 2660
02240 GOSUB 2780
02250 FOR I=1  TO IC \ C(1,I)=RA(I+2) \ NEXT I
02260 C(2,1)=(RA(3)-RA(4))*FO1/ASV \ C(2,IC)=(RA(IC)-RA(IC-1))*FO1/ASV
02270 GOSUB 5880
02280 FOR I=1 TO ZG \ ZB(I)=VC(I) \ NEXT I
02290 REM ========================================================
02300 REM ==============       FOR  YB       =======================
02310 REM ========================================================
02320 FOR I=1 TO IC2 \ DA(I)=YB(I) \ RA(I)=0 \ NEXT I
02330 GOSUB 2660
02340 GOSUB 2780
02350 FOR I=1 TO IC \ C(1,I)=RA(I+2) \ NEXT I
02360 C(2,1)=(RA(3)-RA(4))*FO1/ASV \ C(2,IC)=(RA(IC)-RA(IC-1))*FO1/ASV
02370 GOSUB 5880
02380 FOR I=1 TO ZG \ YB(I)=VC(I) \ NEXT I
02390 REM ========================================================
02400 REM ==============       FOR  AA     =========================
02410 REM ========================================================
02420 FOR I=1 TO IC2 \ DA(I)=B3(I) \ RA(I)=0 \ NEXT I
02430 GOSUB 2660
02440 GOSUB 2780
02450 FOR I=1 TO IC \ C(1,I)=RA(I+2) \ NEXT I
02460 C(2,1)=(RA(3)-RA(4))*FO1/ASV \ C(2,IC)=(RA(IC)-RA(IC-1))*FO1/ASV
02470 GOSUB 5880
02480 FOR I=1 TO ZG \ AA(I)=VC(I) \ NEXT I
02490 REM ========================================================
02500 REM ==============      FOR  AB     ==========================
02510 REM ========================================================
02520 FOR I=1 TO IC2 \ DA(I)=B4(I) \ RA(I)=0 \ NEXT I
02530 GOSUB 2660
02540 GOSUB 2780
02550 FOR I=1 TO IC \ C(1,I)=RA(I+2) \ NEXT I
02560 C(2,1)=(RA(3)-RA(4))*FO1/ASV \ C(2,IC)=(RA(IC)-RA(IC-1))*FO1/ASV
02570 GOSUB 5880
02580 FOR I=1 TO ZG \ AB(I)=VC(I) \ NEXT I
02590 REM ===================================================================
02600 IC=ZG     \    ZG=0     \      ZC=0       \     REM RESET FOR SECOND FILE
02610 REM ===================================================================
02620 REM
02630 GOTO 2910
02640 REM ========================================================
02650 REM ==============SUBROUTINE --- SMOOTHING DISPLACEMENTS=======
02660 REM ========================================================
02670 HC=0
02680 HC=HC+1
02690 FOR I=2 TO (IC2-1)
02700 DA(I)=0.25*(DA(I-1)+DA(I+1))+0.5*DA(I)
02710 NEXT I
02720 DA(1)=0.5*(DA(1)+DA(2)) \ DA(IC2)=0.5*(DA(IC2-1)+DA(IC2))
02730 IF HC=1 THEN GOTO 2680
```

```
02740 RETURN
02750 REM ================ END OF SMOOTHING SUBROUTINE ================
02760 REM
02770 REM ==============================================================
02780 REM ===============SUBROUTINE --- ACCELERATION===================
02790 REM
02800 REM =============== A LANCZOS 5 POINT ROUTINE ===================
02810 REM ==============================================================
02820 FOR I=3 TO (IC2-2)
02830 R1=-1*(DA(I-1)+DA(I+1))-2*(DA(I)-DA(I-2)-DA(I+2))
02840 RA(I)=R1*FO1*FO1/(7*9)
02850 NEXT I
02860 RETURN
02870 REM ===============END OF ACCELERATION ROUTINE================
02880 REM
02890 REM
02900 REM ========================================================
02910 REM --------------- NOW REASSIGN MEMORY LOCATIONS ------------
02920 REM ========================================================
02930 FOR I=1 TO IC1 \ A3(I)=A3(I+2) \ NEXT I
02940 FOR I=1 TO IC1 \ A4(I)=A4(I+2) \ NEXT I
02950 FOR I=1 TO IC1 \ PW(I)=PW(I+2) \ NEXT I
02960 FOR I=1 TO IC1 \ PL(I)=PL(I+2) \ NEXT I
02970 FOR I=1 TO IC1 \ AZ1(I)=AZ1(I+2) \ NEXT I
02980 FOR I=1 TO IC1 \ AZ2(I)=AZ2(I+2) \ NEXT I
02990 FOR I=1 TO IC1 \ K1(I)=K1(I+2) \ NEXT I
03000 FOR I=1 TO IC1 \ K2(I)=K2(I+2) \ NEXT I
03010 FOR I=1 TO IC1 \ K3(I)=K3(I+2) \ NEXT I
03020 FOR I=1 TO IC1 \ K4(I)=K4(I+2) \ NEXT I
03030 FOR I=1 TO IC1 \ K5(I)=K5(I+2) \ NEXT I
03040 FOR I=1 TO IC1 \ K6(I)=K6(I+2) \ NEXT I
03050 FOR I=1 TO IC1 \ K7(I)=K7(I+2) \ NEXT I
03060 FOR I=1 TO IC1 \ K8(I)=K8(I+2) \ NEXT I
03070 FOR I=1 TO IC1 \ H1(I)=H1(I+2) \ NEXT I
03080 FOR I=1 TO IC1 \ H2(I)=H2(I+2) \ NEXT I
03090 FOR I=1 TO IC1 \ H3(I)=H3(I+2) \ NEXT I
03100 FOR I=1 TO IC1 \ H4(I)=H4(I+2) \ NEXT I
03110 FOR I=1 TO IC1 \ H5(I)=H5(I+2) \ NEXT I
03120 FOR I=1 TO IC1 \ H6(I)=H6(I+2) \ NEXT I
03130 FOR I=1 TO IC1 \ H7(I)=H7(I+2) \ NEXT I
03140 FOR I=1 TO IC1 \ H8(I)=H8(I+2) \ NEXT I
03150 FOR I=1 TO IC1 \ H9(I)=H9(I+2) \ NEXT I
03160 FOR I=1 TO IC1 \ B0(I)=B0(I+2) \ NEXT I
03170 FOR I=1 TO IC1 \ B1(I)=B1(I+2) \ NEXT I
03180 FOR I=1 TO IC1 \ B2(I)=B2(I+2) \ NEXT I
03190 FOR I=1 TO IC1 \ B3(I)=B3(I+2) \ NEXT I
03200 FOR I=1 TO IC1 \ B4(I)=B4(I+2) \ NEXT I
03210 FOR I=1 TO IC1 \ B5(I)=B5(I+2) \ NEXT I
03220 FOR I=1 TO IC1 \ B6(I)=B6(I+2) \ NEXT I
03240 FOR I=1 TO IC1 \ B9(I)=B9(I+2) \ NEXT I
03250 FOR I=1 TO IC1 \ XI(I)=I/FO1  \ NEXT I
03260 REM
03270 IF SE=8 GOTO 3322
03280 PRINT "------------------------------------------------------------"
03290 PRINT "THAT DATA READ AND ACCELERATION DATA CALCULATED"
03300 PRINT "NOW READING DATA FROM SECOND FILE WHICH IS:-":FL2$
03310 PRINT "------------------------------------------------------------"
03320 INPUT #2,ZC \ INPUT#2,PS \ INPUT#2,FO2 \ INPUT#2,AW
03330 INPUT #2,BL1 \ INPUT #2,BL2 \ INPUT #2,BL3
03340 INPUT #2,BL4 \ INPUT #2,BL5
```

```
03350 IL=0.128 \ IT=0.265
03360 MF=0.0145*BW \ WF=MF*9.81
03370 ML=0.0465*BW \ WL=ML*9.81
03380 MT=0.0990*BW \ WT=MT*9.81
03390 FOR I=1 TO ZC \ INPUT#2,FX(I) \ NEXT I
03400 FOR I=1 TO ZC \ INPUT#2,FY(I) \ NEXT I
03410 FOR I=1 TO ZC \ INPUT#2,FZ(I) \ NEXT I
03420 FOR I=1 TO ZC \ INPUT#2,MX(I) \ NEXT I
03430 FOR I=1 TO ZC \ INPUT#2,MY(I) \ NEXT I
03440 REM
03450 REM ==========================================================
03460 REM SELECT START AND FINISH POINTS FROM DATA (USING FZ CHANNEL)
03470 REM ==========================================================
03480 REM
03490 FOR I=1 TO ZC
03500 IF ABS(FZ(I)-BL3)>80 THEN ZD=I \ GOTO 3520
03510 NEXT I
03520 FOR I=ZD TO ZC
03530 IF ABS(BL3-FZ(I))<20 THEN ZE=I-1 \ GOTO 3545
03540 NEXT I
03545 IF SE=8 GOTO 3570
03550 PRINT "START VALUE =.....",ZD
03560 PRINT "FINISH VALUE=.....",ZE
03570 ZG=ZE-ZD+1
03575 IF SE=8 GOTO 3600
03580 PRINT \ PRINT "START TO FINISH VALUES (INCLUSIVE) = ";ZG
03590 REM
03600 REM ==========================================================
03610 REM ========  REALLOCATION OF MEMORY STORES ====================
03620 REM ==========================================================
03630 REM
03640 FOR K=1 TO ZC \ FX(K)=FX(K+ZD-1) \ NEXT K
03650 FOR K=1 TO ZC \ FY(K)=FY(K+ZD-1) \ NEXT K
03660 FOR K=1 TO ZC \ FZ(K)=FZ(K+ZD-1) \ NEXT K
03670 FOR K=1 TO ZC \ MX(K)=MX(K+ZD-1) \ NEXT K
03680 FOR K=1 TO ZC \ MY(K)=MY(K+ZD-1) \ NEXT K
03690 REM
03700 REM ==========================================================
03710 REM               DATA READY FOR SPLINE FITTING
03720 REM ==========================================================
03730 IF SE=8 GOTO 3810
03740 PRINT \ PRINT \ GOSUB 10500
03750 PRINT \ PRINT "ALL DATA READ AND ACCEPTED"
03760 PRINT \ PRINT "ADC PLATFM DATA ITEMS =  ";ZG
03770 PRINT \ PRINT "FILM DIG0D DATA ITEMS =  ";IC
03780 IF ZG=IC THEN PRINT "SPLINE CURVE FITTING NOT REQUIRED"
03790 PRINT \ INPUT "PRESS @O@ + @CR@ TO CONTINUE",AS$
03800 PRINT FN.CLR$
03810 REM
03820 IF ZG<>IC THEN GOSUB 5680
03830 REM THIS IS THE SPLINE SUBROUTINE
03860 IF SE=8 GOTO 3920
03870 PRINT " ============================================================="
03880 PRINT "                         CALCULATIONS"
03890 PRINT " ============================================================="
03900 REM
03910 REM
03920 REM *********************************************************
03930 REM ******   CONVERSION OF VOLTAGES TO FORCE DATA   ***********
03940 REM *********************************************************
```

```
03950 REM
03960 SF=PS*10/32768                    \    REM SCALE FACTOR
03970 FOR I=1 TO ZG
03980 RZ(I)=(FZ(I)-BL1)*SF
03990 RX(I)=(FX(I)-BL2)*SF
04000 RY(I)=(FY(I)-BL3)*SF
04010 MX(I)=(MX(I)-BL4)*SF
04020 MY(I)=(MY(I)-BL5)*SF
04030 PX(I)=9.999999 \ PY(I)=9.999999 \ REM SAFTY CHECK FOR ZERO DIVISION
04040 IF RZ(I)=0 THEN GOTO 4070
04050 PX(I)=(-0.054*RX(I)-0.24*MY(I))/RZ(I)
04060 PY(I)=(-0.054*RY(I)+0.24*MX(I))/RZ(I)
04070 RY(I)=-RY(I)
04080 NEXT I
04090 REM
04100 REM ************************************************************
04110 REM            CALCULATION OF M & J FORCES
04120 REM ************************************************************
04130 MLRAT=0 \  MLRQU=0            \ REM MAX LOADING RATE INDICATION
04140 FOR I =1 TO ZG
04150 NX=(0.2+PX(I))*PW(I)/0.4
04160 A1(I)=NX-AZ2(I)
04170 NY=PY(I)*PL(I)/0.6
04180 A2(I)=NY-AZ1(I)
04190 AM1=(RY(I)*A1(I)-RZ(I)*A2(I))
04200 IF AM1>0 THEN GOTO 4220
04210 AT(I)=AM1/A3(I) \ AM2(I)=AT(I) \GA(I)=0.5*AT(I) \ SL(I)=0.5*AT(I) \ GOTO 4230
04220 SH(I)=AM1/A4(I) \ AM2(I)=SH(I)
04230 AY(I)=-RY(I)-AT(I)*SIN(B0(I))-SH(I)*SIN(B6(I))
04240 AZ(I)=-RZ(I)-AT(I)*COS(B0(I))-SH(I)*COS(B6(I))
04250 BF=ZA(I)*ML*K3(I)-YA(I)*ML*K6(I)+IL*AA(I)
04260 TF=ZB(I)*MT*H7(I)-YB(I)*MT*H8(I)+IT*AB(I)
04270 C1=BF+WL*K3(I)-AY(I)*K4(I)+AZ(I)*K5(I)+SH(I)*K8(I)+SL(I)*K7(I)
04280 C2=ML*YA(I)+AY(I)-(C1*SIN(B2(I))/K2(I))-SH(I)*SIN(B6(I))+SL(I)*SIN(B5(I))
04290 C3=ML*ZA(I)+WL+AZ(I)-(C1*COS(B2(I))/K2(I))+SH(I)*COS(B6(I))+SL(I)*COS(B5(I))
04300 C4=TF+WT*H7(I)+GA(I)*H4(I)-C1*H9(I)/K2(I)
04310 C5=4*COS(B1(I))-4*K1(I)*COS(B2(I))/K2(I)
04320 C6=4*SIN(B1(I))-4*K1(I)*SIN(B2(I))/K2(I)
04330 C7=H1(I)-4*H3(I)-4*K1(I)*H9(I)/K2(I)
04340 C8=C3*H5(I)-C2*H6(I)+C4
04350 C9=C7+C5*H5(I)-C6*H6(I)
04360 REM
04370 RF(I)=C8/C9
04380 PT(I)=4*RF(I)
04385 PF(I)=PT(I)*SQR(2*(1-COS(B1(I)-B9(I))))
04390 KY(I)=C2-RF(I)*C6
04400 KZ(I)=C3-RF(I)*C5
04410 HM(I)=(C1-4*RF(I)*K1(I))/K2(I)
04420 GL(I)=HM(I)*H9(I)/H2(I)
04430 KN(I)=KZ(I)*COS(B3(I))+KY(I)*SIN(B3(I))
04440 KS(I)=KY(I)*COS(B3(I))-KZ(I)*SIN(B3(I))
04450 REM ============== CALCULATE JOINT MOMENTS ==========================
04460 AM1(I)=AM1
04470 KM1(I)=RZ(I)*(A2(I)+K5(I))-RY(I)*(K4(I)+A1(I))
04480 PM1(I)=RZ(I)*(H5(I)+K5(I)+A2(I))-RY(I)*(H6(I)+K4(I)+A1(I))
04490 REM ============== CALCULATE 'LEG' ANGLES ===========================
04500 HAN(I)=(3.14159+B4(I))*57.296
04510 KAN(I)=(3.14159+B4(I)-B3(I))*57.296
04530 REM ============== CALCULATE MAX MUSCLE LOADING RATE ================
04540 REM ==============        FOR CALF & QUADS.           ================
```

```
04545 IF I<2 GOTO 4580
04546 IF I>(ZG/2) GOTO 4580
04550 LPOU=(PT(I)-PT(I-2))*FO2/3
04555 IF LROU>MLROU THEN MLROU=LROU
04560 LRAT=(AM2(I)-AM2(I-2))*FO2/3
04570 IF LRAT>MLRAT THEN MLRAT=LRAT
04580 NEXT I
04595 IF SE=8 GOTO 4640
04600 PRINT " ******************************************************************"
04610 PRINT " ********   SMOOTHING ROUTINE. A HANNING NON RECURSIVE   *******"
04620 PRINT " ********   FILTER TWICE. FROM KB(GPA) PROGRAM ON PET.   *******"
04630 PRINT " ******************************************************************"
04640 SM1=.25 \ SM2=0.5 \ SM3=0.25
04650 HC=0
04660 HC=HC+1
04670 FOR I=2 TO (ZG-1)
04680 AM2(I)=SM1*(AM2(I-1)+AM2(I+1))+SM2*AM2(I)
04690 GA(I)=SM1*(GA(I-1)+GA(I+1))+SM2*GA(I)
04695 SL(I)=SM1*(SL(I-1)+SL(I+1))+SM2*SL(I)
04700 SH(I)=SM1*(SH(I-1)+SH(I+1))+SM2*SH(I)
04710 RF(I)=SM1*(RF(I-1)+RF(I+1))+SM2*RF(I)
04720 PT(I)=SM1*(PT(I-1)+PT(I+1))+SM2*PT(I)
04725 PF(I)=SM1*(PF(I-1)+PF(I+1))+SM2*PF(I)
04730 HM(I)=SM1*(HM(I-1)+HM(I+1))+SM2*HM(I)
04740 GL(I)=SM1*(GL(I-1)+GL(I+1))+SM2*GL(I)
04750 AZ(I)=SM1*(AZ(I-1)+AZ(I+1))+SM2*AZ(I)
04760 AY(I)=SM1*(AY(I-1)+AY(I+1))+SM2*AY(I)
04770 KZ(I)=SM1*(KZ(I-1)+KZ(I+1))+SM2*KZ(I)
04780 KY(I)=SM1*(KY(I-1)+KY(I+1))+SM2*KY(I)
04790 KS(I)=SM1*(KS(I-1)+KS(I+1))+SM2*KS(I)
04800 KN(I)=SM1*(KN(I-1)+KN(I+1))+SM2*KN(I)
04810 AM1(I)=SM1*(AM1(I-1)+AM1(I+1))+SM2*AM1(I)
04820 KM1(I)=SM1*(KM1(I-1)+KM1(I+1))+SM2*KM1(I)
04830 PM1(I)=SM1*(PM1(I-1)+PM1(I+1))+SM2*PM1(I)
04840 NEXT I
04850 AM2(1)=SM2*(AM2(1)+AM2(2)) \ AM2(ZG)=SM2*(AM2(ZG-1)+AM2(ZG))
04860 GA(1)=SM2*(GA(1)+GA(2)) \ GA(ZG)=SM2*(GA(ZG-1)+GA(ZG))
04865 SL(1)=SM2*(SL(1)+SL(2)) \ SL(ZG)=SM2*(SL(ZG-1)+SL(ZG))
04870 SH(1)=SM2*(SH(1)+SH(2)) \ SH(ZG)=SM2*(SH(ZG-1)+SH(ZG))
04880 RF(1)=SM2*(RF(1)+RF(2)) \ RF(ZG)=SM2*(RF(ZG-1)+RF(ZG))
04890 PT(1)=SM2*(PT(1)+PT(2)) \ PT(ZG)=SM2*(PT(ZG-1)+PT(ZG))
04895 PF(1)=SM2*(PF(1)+PF(2)) \ PF(ZG)=SM2*(PF(ZG-1)+PF(ZG))
04900 HM(1)=SM2*(HM(1)+HM(2)) \ HM(ZG)=SM2*(HM(ZG-1)+HM(ZG))
04910 GL(1)=SM2*(GL(1)+GL(2)) \ GL(ZG)=SM2*(GL(ZG-1)+GL(ZG))
04920 AZ(1)=SM2*(AZ(1)+AZ(2)) \ AZ(ZG)=SM2*(AZ(ZG-1)+AZ(ZG))
04930 AY(1)=SM2*(AY(1)+AY(2)) \ AY(ZG)=SM2*(AY(ZG-1)+AY(ZG))
04940 KZ(1)=SM2*(KZ(1)+KZ(2)) \ KZ(ZG)=SM2*(KZ(ZG-1)+KZ(ZG))
04950 KY(1)=SM2*(KY(1)+KY(2)) \ KY(ZG)=SM2*(KY(ZG-1)+KY(ZG))
04960 KN(1)=SM2*(KN(1)+KN(2)) \ KN(ZG)=SM2*(KN(ZG-1)+KN(ZG))
04970 KS(1)=SM2*(KS(1)+KS(2)) \ KS(ZG)=SM2*(KS(ZG-1)+KS(ZG))
04980 AM1(1)=SM2*(AM1(1)+AM1(2)) \ AM1(ZG)=SM2*(AM1(ZG-1)+AM1(ZG))
04990 KM1(1)=SM2*(KM1(1)+KM1(2)) \ KM1(ZG)=SM2*(KM1(ZG-1)+KM1(ZG))
05000 PM1(1)=SM2*(PM1(1)+PM1(2)) \ PM1(ZG)=SM2*(PM1(ZG-1)+PM1(ZG))
05010 IF HC=1 THEN GOTO 4660
05015 IF SE=8 GOTO 5230
05020 PRINT \ PRINT \ PRINT
05030 GOSUB 10500 \ PRINT
05050 REM ==========================================================
05060 PRINT "               **********  CONE CALCULATIONS  **********"
05070 PRINT "               **********  FITTED SPLINE      **********"
```

```
05080 PRINT "              ++++++++++  SMOOTHED DATA      ++++++++++"
05090 REM ============================================================
05100 REM
05110 PRINT \ GOSUB 10500
05120 PRINT \ INPUT "PRESS 'O' +  'CR' TO CONTINUE  ",AS$
05130 GOTO 1300
05140 REM
05150 REM ==============================================================
05160 REM =========    OUTPUT (TO FILE & TO SCREEN)     =================
05170 REM ==============================================================
05180 REM
05190 REM **************************************************
05200 REM ********** OUTPUT TO FILE FIRST *****************
05210 REM **************************************************
05220 REM
05230 PRINT#3,ZG
05240 LL$="-------------------------------------------------------------------------------------------------"
05250 PRINT#3,LL$
05260 PRINT#3,"                                    ------------- RESULTS FOR ";FL3$;" FILE ---------"
05261 PRINT#3,"                                    ------------- BODY WEIGHT= ";BW*9.81;" N. --------
05270 PRINT#3,LL$
05280 PRINT#3
05290 PRINT#3,LL$
05300 PRINT#3," I "," GA/SH "," RF "," PT "," HM "," GL "," RZ "," RY "
05310 PRINT#3,LL$
05320 FOR I=1 TO ZG \ PRINT#3,I,AM2(I),RF(I),PT(I),HM(I),GL(I),RZ(I),RY(I) \ NEXT I
05330 PRINT#3,ZG
05340 PRINT#3,LL$
05350 PRINT#3," I "," SL "," AZ "," AY "," KZ "," KY "," KN "," KS "
05360 PRINT#3,LL$
05370 FOR I=1 TO ZG \ PRINT#3,I,SL(I),AZ(I),AY(I),KZ(I),KY(I),KN(I),KS(I) \ NEXT I
05380 PRINT#3,ZG \ PRINT#3,LL$
05390 PRINT#3," I "," AM1"," KM1"," PM1"," PF "," KAN"," HAN"
05400 PRINT#3,LL$
05410 FOR I=1 TO ZG \ PRINT#3,I,AM1(I),KM1(I),PM1(I),PF(I),KAN(I),HAN(I)  \ NEXT I
05420 PRINT#3,LL$
05430 PRINT#3,"QUADS. MAX LOADING RATE...=        ",MLRQU
05435 PRINT#3,"CALF MAX LOADING RATE.....=        ",MLRAT
05440 PRINT#3,LL$
05450 REM
05460 REM ============================================================
05470 PRINT \ PRINT "DATA WRITTEN TO FILE ";FL3$ \ SLEEP 5
05480 REM ============================================================
05490 REM
05500 PRINT \ PRINT \ PRINT
05510 IF SE=5 THEN GOTO 1300
05512 IF SE=8 GOTO 862
05520 REM
05530 REM ============================================================
05540 REM ============      RESULTS TO SCREEN      ====================
05550 REM ============================================================
05560 REM
05570 PRINT \ PRINT "++++++++++   RESULTS   ++++++++++"
05580 PRINT \ PRINT "+++++++++++++++++++++++++++++++++"
05590 PRINT "GA/SH";TAB(25);"RF";TAB(50);"HM"
05600 FOR I=1 TO ZG \ PRINT AM2(I);TAB(25);RF(I);TAB(50);HM(I) \ NEXT I
05610 PRINT \ PRINT "------------------------------------------------------------------"
05620 PRINT \ PRINT "AZ";TAB(25);"KZ";TAB(50);"KS"
05630 FOR I=1 TO ZG \ PRINT AZ(I);TAB(25);KZ(I);TAB(50);KS(I) \ NEXT I
05640 PRINT \ PRINT "------------------------------------------------------------------"
```

```
05650 PRINT \ INPUT "PRESS 'O' + 'CR' TO CONTINUE";ASS
05660 GOTO 1300
05670 GOTO 10510
05680 IF SE=5 GOTO 5820
05690 REM ************************************************************ \ GCSUB 10500
05700 PRINT "****************** SPLINE CURVE FITTING SUBROUTINE **************"
05710 REM ************************************************************ \ GCSUB 10500
05720 REM
05730 REM *******************************************************************
05740 REM REFERENCE 'ELEMENTARY NUMERICAL ANALYSIS: AN ALGIORITHMIC APPROACH'
05750 REM BY-CONTE & DE BOOR. PUB MC.GRAW HILL THIRD EDITION
05760 REM PAGES 284-293....'PEICEWISE-POLYNOMIAL CUBIC SPLINE INTERPOLATION'
05770 REM *******************************************************************
05780 REM
05790 REM
05800 REM !!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!
05810 REM ^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^
05820 REM GO TO DATA PREPERATION SUBROUTINE
05830 GOSUB 6540
05840 REM !!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!
05850 REM ^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^^
05860 RETURN
05870 REM *********************************************************************
05880 REM ********************** CALCULATE SPLINE *************************
05890 REM *********************************************************************
05900 REM
05910 DG(1)=1 \ DM(1)=0
05920 FOR M=1 TO IC
05930 DM(M)=XI(M)-XI(M-1)
05940 DG(M)=(C(1,M)-C(1,M-1))/DM(M)
05950 NEXT M
05960 ZF=IC-1
05970 FOR M=2 TO ZF
05980 C(2,M)=3*(DM(M)*DG(M+1)+DM(M+1)*DG(M))
05990 DG(M)=2*(DM(M)+DM(M+1))
06000 NEXT M
06010 FOR M=2 TO ZF
06020 G=-DM(M+1)/DG(M-1)
06030 DG(M)=DG(M)+G*DM(M-1)
06040 C(2,M)=C(2,M)+G*C(2,M-1)
06050 NEXT M
06060 FOR M=ZF TO 2 STEP-1
06070 C(2,M)=(C(2,M)-DM(M)*C(2,M+1))/DG(M)
06080 NEXT M
06090 REM =============================================================
06100 REM =========           CALCCF PART          =====================
06110 REM =============================================================
06120 FOR I=1 TO ZF
06130 DX=XI(I+1)-XI(I)
06140 D1=(C(1,I+1)-C(1,I))/DX
06150 D3=C(2,I)+C(C,I+1)-2*D1
06160 C(3,I)=(D1-C(2,1)-C3)/DX
06170 C(4,I)=C3/(DX*DX)
06180 NEXT I
06190 REM =============================================================
06200 REM ==========          PCUBIC PART          ====================
06210 REM =============================================================
06220 XB=X1(1)
06230 ISP=0
06240 FOR I=1 TO ZC
```

```
06250 ISP=ISP+1
06260 IF XB<XI(I) THEN GOTO 6320
06270 GOTO 6280
06280 FOR J=1 TO ZF
06290 IF XB<XI(J+1) THEN GOTO 6370
06300 NEXT J
06310 J=N
06320 FOR J=I-1 TO 1 STEP-1
06330 IF XB<XI(J) THEN GOTO 6370
06340 GOTO 6370
06350 NEXT J
06360 J=1
06370 I=J
06380 DX=XB-XI(I)
06390 PC=C(1,I)+DX*(C(2,I)+DX*(C(3,I)+DX*C(4,I)))
06400 VC(ISP)=PC   \   VB(ISP)=XB
06410 XB=XB+(IC/(FO1*ZG))
06420 IF ISP=ZC THEN GOTO 6440
06430 NEXT I
06440 RETURN
06450 REM ============================================================
06460 REM =======   END OF SUBROUTINE FOR SPLINE   ====================
06470 REM ============================================================
06480 REM
06490 REM
06500 REM
06510 REM
06520 REM
06530 GOTO 10510
06540 REM
06550 REM
06560 REM ===========================================================
06570 REM ========= . SUBROUTINE FOR DATA PREPERATION FOR SPLINING  ===
06580 REM
06590 REM ============================================================
06600 REM
06610 REM
06620 REM*********************************************
06630 REM*********       FOR A3      ********************
06640 REM*********************************************
06650 REM
06660 C(2,1)=(A3(2)-A3(1))*FO1 \ C(2,IC)=(A3(IC)-A3(IC-1))*FO1
06670 FOR I=1 TO IC \ C(1,I)=A3(I) \ NEXT I
06680 GOSUB 5880
06690 FOR I=1 TO ZG \ A3(I)=VC(I)  \ NEXT I
06700 REM
06710 REM
06720 REM*********************************************
06730 REM*********       FOR A4      ********************
06740 REM*********************************************
06750 REM
06760 C(2,1)=(A4(2)-A4(1))*FO1 \ C(2,IC)=(A4(IC)-A4(IC-1))*FO1
06770 FOR I=1 TO IC \ C(1,I)=A4(I) \ NEXT I
06780 GOSUB 5880
06790 FOR I=1 TO ZG \ A4(I)=VC(I)  \ NEXT I
06800 REM
06810 REM
06820 REM*********************************************
06830 REM*********       FOR PW      ********************
06840 REM*********************************************
```

```
06850 REM
06860 C(2,1)=(PW(2)-PW(1))*FQ1 \ C(2,IC)=(PW(IC)-PW(IC-1))*FQ1
06870 FOR I=1 TO IC \ C(1,I)=PW(I) \ NEXT I
06880 GOSUB 5880
06890 FOR I=1 TO ZG \ PW(I)=VC(I)  \ NEXT I
06900 REM
06910 REM
06920 REM*************************************************
06930 REM*********      FOR PL      **********************
06940 REM*************************************************
06950 REM
06960 C(2,1)=(PL(2)-PL(1))*FQ1 \ C(2,IC)=(PL(IC)-PL(IC-1))*FQ1
06970 FOR I=1 TO IC \ C(1,I)=PL(I) \ NEXT I
06980 GOSUB 5880
06990 FOR I=1 TO ZG \ PL(I)=VC(I)  \ NEXT I
07000 REM
07010 REM
07020 REM*************************************************
07030 REM*********      FOR AZ1      *********************
07040 REM*************************************************
07050 REM
07060 C(2,1)=(AZ1(2)-AZ1(1))*FQ1 \ C(2,IC)=(AZ1(IC)-AZ1(IC-1))*FQ1
07070 FOR I=1 TO IC \ C(1,I)=AZ1(I) \ NEXT I
07080 GOSUB 5880
07090 FOR I=1 TO ZG \ AZ1(I)=VC(I)  \ NEXT I
07100 REM
07110 REM
07120 REM*************************************************
07130 REM*********      FOR AZ2      *********************
07140 REM*************************************************
07150 REM
07160 C(2,1)=(AZ2(2)-AZ2(1))*FQ1 \ C(2,IC)=(AZ2(IC)-AZ2(IC-1))*FQ1
07170 FOR I=1 TO IC \ C(1,I)=AZ2(I) \ NEXT I
07180 GOSUB 5880
07190 FOR I=1 TO ZG \ AZ2(I)=VC(I)  \ NEXT I
07200 REM
07210 REM
07220 REM*************************************************
07230 REM*********      FOR K1      **********************
07240 REM*************************************************
07250 REM
07260 C(2,1)=(K1(2)-K1(1))*FQ1 \ C(2,IC)=(K1(IC)-K1(IC-1))*FQ1
07270 FOR I=1 TO IC \ C(1,I)=K1(I) \ NEXT I
07280 GOSUB 5880
07290 FOR I=1 TO ZG \ K1(I)=VC(I)  \ NEXT I
07300 REM
07310 REM
07320 REM*************************************************
07330 REM*********      FOR K2      **********************
07340 REM*************************************************
07350 REM
07360 C(2,1)=(K2(2)-K2(1))*FQ1 \ C(2,IC)=(K2(IC)-K2(IC-1))*FQ1
07370 FOR I=1 TO IC \ C(1,I)=K2(I) \ NEXT I
07380 GOSUB 5880
07390 FOR I=1 TO ZG \ K2(I)=VC(I)  \ NEXT I
07400 REM
07410 REM
07420 REM*************************************************
07430 REM*********      FOR K3      **********************
07440 REM*************************************************
```

```
07450 REM
07460 C(2,1)=(K3(2)-K3(1))*FO1 \ C(2,IC)=(K3(IC)-K3(IC-1))*FO1
07470 FOR I=1 TO IC \ C(1,I)=K3(I) \ NEXT I
07480 GOSUB 5880
07490 FOR I=1 TO ZG \ K3(I)=VC(I)  \ NEXT I
07500 REM
07510 REM
07520 REM**************************************************
07530 REM*********      FOR K4      ***********************
07540 REM**************************************************
07550 REM
07560 C(2,1)=(K4(2)-K4(1))*FO1 \ C(2,IC)=(K4(IC)-K4(IC-1))*FO1
07570 FOR I=1 TO IC \ C(1,I)=K4(I) \ NEXT I
07580 GOSUB 5880
07590 FOR I=1 TO ZG \ K4(I)=VC(I)  \ NEXT I
07600 REM
07610 REM
07620 REM**************************************************
07630 REM*********      FOR K5      ***********************
07640 REM**************************************************
07650 REM
07660 C(2,1)=(K5(2)-K5(1))*FO1 \ C(2,IC)=(K5(IC)-K5(IC-1))*FO1
07670 FOR I=1 TO IC \ C(1,I)=K5(I) \ NEXT I
07680 GOSUB 5880
07690 FOR I=1 TO ZG \ K5(I)=VC(I)  \ NEXT I
07700 REM
07710 REM
07720 REM**************************************************
07730 REM*********      FOR K6      ***********************
07740 REM**************************************************
07750 REM
07760 C(2,1)=(K6(2)-K6(1))*FO1 \ C(2,IC)=(K6(IC)-K6(IC-1))*FO1
07770 FOR I=1 TO IC \ C(1,I)=K6(I) \ NEXT I
07780 GOSUB 5880
07790 FOR I=1 TO ZG \ K6(I)=VC(I)  \ NEXT I
07800 REM
07810 REM
07820 REM**************************************************
07830 REM*********      FOR K7      ***********************
07840 REM**************************************************
07850 REM
07860 C(2,1)=(K7(2)-K7(1))*FO1 \ C(2,IC)=(K7(IC)-K7(IC-1))*FO1
07870 FOR I=1 TO IC \ C(1,I)=K7(I) \ NEXT I
07880 GOSUB 5880
07890 FOR I=1 TO ZG \ K7(I)=VC(I)  \ NEXT I
07900 REM
07910 REM
07920 REM**************************************************
07930 REM*********      FOR K8      ***********************
07940 REM**************************************************
07950 REM
07960 C(2,1)=(K8(2)-K8(1))*FO1 \ C(2,IC)=(K8(IC)-K8(IC-1))*FO1
07970 FOR I=1 TO IC \ C(1,I)=K8(I) \ NEXT I
07980 GOSUB 5880
07990 FOR I=1 TO ZG \ K8(I)=VC(I)  \ NEXT I
08000 REM
08010 REM
08020 REM**************************************************
08030 REM*********      FOR H1      ***********************
08040 REM**************************************************
```

```
08050 REM
08060 C(2,1)=(H1(2)-H1(1))*F01 \ C(2,IC)=(H1(IC)-H1(IC-1))*F01
08070 FOR I=1 TO IC \ C(1,I)=H1(I) \ NEXT I
08080 GOSUB 5880
08090 FOR I=1 TO ZG \ H1(I)=VC(I)  \ NEXT I
08100 REM
08110 REM
08120 REM*********************************************************
08130 REM*********      FOR H2      ******************************
08140 REM*********************************************************
08150 REM
08160 C(2,1)=(H2(2)-H2(1))*F01 \ C(2,IC)=(H2(IC)-H2(IC-1))*F01
08170 FOR I=1 TO IC \ C(1,I)=H2(I) \ NEXT I
08180 GOSUB 5880
08190 FOR I=1 TO ZG \ H2(I)=VC(I)  \ NEXT I
08200 REM
08210 REM
08220 REM*********************************************************
08230 REM*********      FOR H3      ******************************
08240 REM*********************************************************
08250 REM
08260 C(2,1)=(H3(2)-H3(1))*F01 \ C(2,IC)=(H3(IC)-H3(IC-1))*F01
08270 FOR I=1 TO IC \ C(1,I)=H3(I) \ NEXT I
08280 GOSUB 5880
08290 FOR I=1 TO ZG \ H3(I)=VC(I)  \ NEXT I
08300 REM
08310 REM
08320 REM*********************************************************
08330 REM*********      FOR H4      ******************************
08340 REM*********************************************************
08350 REM
08360 C(2,1)=(H4(2)-H4(1))*F01 \ C(2,IC)=(H4(IC)-H4(IC-1))*F01
08370 FOR I=1 TO IC \ C(1,I)=H4(I) \ NEXT I
08380 GOSUB 5880
08390 FOR I=1 TO ZG \ H4(I)=VC(I)  \ NEXT I
08400 REM
08410 REM
08420 REM*********************************************************
08430 REM*********      FOR H5      ******************************
08440 REM*********************************************************
08450 REM
08460 C(2,1)=(H5(2)-H5(1))*F01 \ C(2,IC)=(H5(IC)-H5(IC-1))*F01
08470 FOR I=1 TO IC \ C(1,I)=H5(I) \ NEXT I
08480 GOSUB 5880
08490 FOR I=1 TO ZG \ H5(I)=VC(I)  \ NEXT I
08500 REM
08510 REM
08520 REM*********************************************************
08530 REM*********      FOR H6      ******************************
08540 REM*********************************************************
08550 REM
08560 C(2,1)=(H6(2)-H6(1))*F01 \ C(2,IC)=(H6(IC)-H6(IC-1))*F01
08570 FOR I=1 TO IC \ C(1,I)=H6(I) \ NEXT I
08580 GOSUB 5880
08590 FOR I=1 TO ZG \ H6(I)=VC(I)  \ NEXT I
08600 REM
08610 REM
08620 REM*********************************************************
08630 REM*********      FOR H7      ******************************
08640 REM*********************************************************
```

```
08650 REM
08660 C(2,1)=(H7(2)-H7(1))*F01 \ C(2,IC)=(H7(IC)-H7(IC-1))*F01
08670 FOR I=1 TO IC \ C(1,I)=H7(I) \ NEXT I
08680 GOSUB 5880
08690 FOR I=1 TO ZG \ H7(I)=VC(I)  \ NEXT I
08700 REM
08710 REM
08720 REM*********************************************
08730 REM*********      FOR H8      *********************
08740 REM*********************************************
08750 REM
08760 C(2,1)=(H8(2)-H8(1))*F01 \ C(2,IC)=(H8(IC)-H8(IC-1))*F01
08770 FOR I=1 TO IC \ C(1,I)=H8(I) \ NEXT I
08780 GOSUB 5880
08790 FOR I=1 TO ZG \ H8(I)=VC(I)  \ NEXT I
08800 REM
08810 REM
08820 REM*********************************************
08830 REM*********      FOR H9      *********************
08840 REM*********************************************
08850 REM
08860 C(2,1)=(H9(2)-H9(1))*F01 \ C(2,IC)=(H9(IC)-H9(IC-1))*F01
08870 FOR I=1 TO IC \ C(1,I)=H9(I) \ NEXT I
08880 GOSUB 5880
08890 FOR I=1 TO ZG \ H9(I)=VC(I)  \ NEXT I
08900 REM
08910 REM
08920 REM*********************************************
08930 REM*********      FOR B0      *********************
08940 REM*********************************************
08950 REM
08960 C(2,1)=(B0(2)-B0(1))*F01 \ C(2,IC)=(B0(IC)-B0(IC-1))*F01
08970 FOR I=1 TO IC \ C(1,I)=B0(I) \ NEXT I
08980 GOSUB 5880
08990 FOR I=1 TO ZG \ B0(I)=VC(I)  \ NEXT I
09000 REM
09010 REM
09020 REM*********************************************
09030 REM*********      FOR B1      *********************
09040 REM*********************************************
09050 REM
09060 C(2,1)=(B1(2)-B1(1))*F01 \ C(2,IC)=(B1(IC)-B1(IC-1))*F01
09070 FOR I=1 TO IC \ C(1,I)=B1(I) \ NEXT I
09080 GOSUB 5880
09090 FOR I=1 TO ZG \ B1(I)=VC(I)  \ NEXT I
09100 REM
09110 REM
09120 REM*********************************************
09130 REM*********      FOR B2      *********************
09140 REM*********************************************
09150 REM
09160 C(2,1)=(B2(2)-B2(1))*F01 \ C(2,IC)=(B2(IC)-B2(IC-1))*F01
09170 FOR I=1 TO IC \ C(1,I)=B2(I) \ NEXT I
09180 GOSUB 5880
09190 FOR I=1 TO ZG \ B2(I)=VC(I)  \ NEXT I
09200 REM
09210 REM
09220 REM*********************************************
09230 REM*********      FOR B3      *********************
09240 REM*********************************************
```

```
09250 REM
09260 C(2,1)=(B3(2)-B3(1))*F01 \ C(2,IC)=(B3(IC)-B3(IC-1))*F01
09270 FOR I=1 TO IC \ C(1,I)=B3(I) \ NEXT I
09280 GOSUB 5880
09290 FOR I=1 TO ZG \ B3(I)=VC(I)  \ NEXT I
09300 REM
09310 REM
09320 REM***************************************************
09330 REM*********       FOR B4       **********************
09340 REM***************************************************
09350 REM
09360 C(2,1)=(B4(2)-B4(1))*F01 \ C(2,IC)=(B4(IC)-B4(IC-1))*F01
09370 FOR I=1 TO IC \ C(1,I)=B4(I) \ NEXT I
09380 GOSUB 5880
09390 FOR I=1 TO ZG \ B4(I)=VC(I)  \ NEXT I
09400 REM
09410 REM
09420 REM***************************************************
09421 REM*********       FOR B5       **********************
09422 REM***************************************************
09423 REM
09424 C(2,1)=(B5(2)-B5(1))*F01 \ C(2,IC)=(B5(IC)-B5(IC-1))*F01
09425 FOR I=1 TO IC \ C(1,I)=B5(I) \ NEXT I
09426 GOSUB 5880
09427 FOR I=1 TO ZG \ B5(I)=VC(I)  \ NEXT I
09428 REM
09520 REM***************************************************
09530 REM*********       FOR B6       **********************
09540 REM***************************************************
09550 REM
09560 C(2,1)=(B6(2)-B6(1))*F01 \ C(2,IC)=(B6(IC)-B6(IC-1))*F01
09570 FOR I=1 TO IC \ C(1,I)=B6(I) \ NEXT I
09580 GOSUB 5880
09590 FOR I=1 TO ZG \ B6(I)=VC(I)  \ NEXT I
09600 REM
09610 REM
09720 REM***************************************************
09730 REM*********       FOR B9       **********************
09740 REM***************************************************
09750 REM
09760 C(2,1)=(B9(2)-B9(1))*F01 \ C(2,IC)=(B9(IC)-B9(IC-1))*F01
09770 FOR I=1 TO IC \ C(1,I)=B9(I) \ NEXT I
09780 GOSUB 5880
09790 FOR I=1 TO ZG \ B9(I)=VC(I)  \ NEXT I
09800 REM
09810 REM
09820 REM***************************************************
09830 REM*********       FOR ZA       **********************
09840 REM***************************************************
09850 REM
09860 C(2,1)=(ZA(2)-ZA(1))*F01 \ C(2,IC)=(ZA(IC)-ZA(IC-1))*F01
```

```
09870 FOR I=1 TO IC \ C(1,I)=ZA(I) \ NEXT I
09880 GOSUB 5880
09890 FOR I=1 TO ZG \ ZA(I)=VC(I)  \ NEXT I
09900 REM
09910 REM
09920 REM**************************************************
09930 REM*********       FOR YA       **********************
09940 REM**************************************************
09950 REM
09960 C(2,1)=(YA(2)-YA(1))*FO1 \ C(2,IC)=(YA(IC)-YA(IC-1))*FO1
09970 FOR I=1 TO IC \ C(1,I)=YA(I) \ NEXT I
09980 GOSUB 5880
09990 FOR I=1 TO ZG \ YA(I)=VC(I)  \ NEXT I
10000 REM
10010 REM
10020 REM**************************************************
10030 REM*********       FOR ZB       **********************
10040 REM**************************************************
10050 REM
10060 C(2,1)=(ZB(2)-ZB(1))*FO1 \ C(2,IC)=(ZB(IC)-ZB(IC-1))*FO1
10070 FOR I=1 TO IC \ C(1,I)=ZB(I) \ NEXT I
10080 GOSUB 5880
10090 FOR I=1 TO ZG \ ZB(I)=VC(I)  \ NEXT I
10100 REM
10110 REM
10120 REM**************************************************
10130 REM*********       FOR YB       **********************
10140 REM**************************************************
10150 REM
10160 C(2,1)=(YB(2)-YB(1))*FO1 \ C(2,IC)=(YB(IC)-YB(IC-1))*FO1
10170 FOR I=1 TO IC \ C(1,I)=YB(I) \ NEXT I
10180 GOSUB 5880
10190 FOR I=1 TO ZG \ YB(I)=VC(I)  \ NEXT I
10200 REM
10210 REM
10220 REM**************************************************
10230 REM*********       FOR AA       **********************
10240 REM**************************************************
10250 REM
10260 C(2,1)=(AA(2)-AA(1))*FO1 \ C(2,IC)=(AA(IC)-AA(IC-1))*FO1
10270 FOR I=1 TO IC \ C(1,I)=AA(I) \ NEXT I
10280 GOSUB 5880
10290 FOR I=1 TO ZG \ AA(I)=VC(I)  \ NEXT I
10300 REM
10310 REM
10320 REM**************************************************
10330 REM*********       FOR AB       **********************
10340 REM**************************************************
10350 REM
10360 C(2,1)=(AB(2)-AB(1))*FO1 \ C(2,IC)=(AB(IC)-AB(IC-1))*FO1
10370 FOR I=1 TO IC \ C(1,I)=AB(I) \ NEXT I
10380 GOSUB 5880
10390 FOR I=1 TO ZG \ AB(I)=VC(I)  \ NEXT I
10400 REM
10410 REM
10420 REM ============================================================
10430 REM ==========        END OF DATA PREPERATION      ================
10440 REM ============================================================
10450 RETURN
10460 REM
```

```
10470 REM
10480 REM
10490 REM
10500 PRINT  "*************************************************************************" \ RETURN
10510 REM ======================================================================
10520 REM ==========          CLOSE DATA FILES AND END PROGRAM   ==========
10530 REM ======================================================================
10540 CLOSE#1 \ CLOSE#2 \ CLOSE#3
10550 END
```

APPENDIX A4. Publications Related to this Study

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HARRISON, R.N., LEES, A., McCULLAGH, P.J.J. (1982). "Forces in Running" Proceedings of British Association of Sports Scientists. Crystal Palace.

HARRISON, R.N. LEES, A. (1984). "A Longitudinal Study of Running Shoe Wear Properties". Proceedings of British Association of Sports Scientists Study Group, Sheffield.

HARRISON, R.N., LEES, A., McCULLAGH, P.J.J., ROWE, W.B. (1986). "A Bioengineering Analysis of Human Muscle & Joint Forces in the Lower Limb During Running". Journal of Sports Sciences. 4. pp.201-218.

HARRISON, R.N., LEES, A., McCULLAGH, P.J.J., ROWE, W.B. (1987). "A Biomechanical Analysis of Muscle & Joint Forces Acting in the Human Leg During Running". In Biomechanics X-B edited B. Jonsson. Human Kinetics.

HARRISON, R.N., LEES, A., McCULLAGH, P.J.J., ROWE, W.B. (1986). "The Effects of Foot Strike on Muscle & Joint Forces Generated in the Human Lower Limb During Middle Distance Running of Highly Trained Athletes. Biomechanics in Sport. M.E.P.