BIOMECHANICAL ANALYSIS OF ISOKINETIC KNEE EXTENSION

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In my father's memory

and in memory of A.Vasaras and Dr C. Foster

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Isokinetic dynamometry is the assessment of dynamic muscle function during isolated joint movements performed at constant angular velocity. The optimal muscular loading (resistive moment equivalent to muscular moment) and control of the angular velocity during isokinetic movements resulted in widespread applications in the areas of muscle testing and rehabilitation. The assessment of muscle function from the resistive moment developed by the isokinetic dynamometer however, is affected by several mechanical and methodological problems. The present study considers the main problems of isokinetic dynamometry that influence muscular performance, measurement of isokinetic parameters, assessment of muscle function and development dynamic joint forces.

The effects of visual feedback on isokinetic maximum torque and reciprocal muscle group ratio of the knee extensors and flexors at a slow (1.04 rad·s^{-1}) and a fast (3.14 rad·s^{-1}) angular velocity of movement were examined using the gravity corrected resistive moment of the dynamometer as the real-time visual feedback source. This elicited a significant increase in the maximum torque output of both muscle groups at the slow angular velocity. There was no significant improvement at the fast angular velocity. The knee flexor/extensor torque ratio was not effected by visual feedback or angular velocity of movement. It was concluded that visual feedback can improve muscular torque output under certain restrictions on velocity and range of movement. Visual feedback is therefore essential during maximum voluntary activation tests in isokinetic dynamometry.

The angular velocity development and maintenance during isokinetic knee extension was examined at preset angular velocities of 0.52, 1.57, 2.62 and 3.67 rad.s^{-1} using a computerised AKRON isokinetic dynamometer. Angular velocity was determined from differentiation of the angular position-time data after optimal smoothing using a low pass digital filter. Maximum torque was determined from the part of the movement with the angular velocity within $\pm$ 10% of the preset velocity. The mean maximum torque ranged from 264.7 ($\pm$43.8) Nm at 0.52 rad·s^{-1} to 198.8 ($\pm$27.9) Nm at 3.67 rad·s^{-1}. During the initial acceleration period the velocity of the limb exceeded the preset velocity by an average of 145%, 44%, 29% and 18% at the four preset velocities respectively. The constant velocity period ranged from 63.7% at 0.52 rad.s^{-1} to 40.3% of the total movement at 3.67 rad.s^{-1}. These results indicate that the angular velocity during isokinetic movements using isokinetic dynamometers, fluctuates even after the initial acceleration period and appropriate correction methods are required before the measurement of isokinetic parameters and the assessment of dynamic muscle function.

The measurement dynamic joint forces during isokinetic knee extension requires the determination of a biomechanical model of the knee. The anatomical parameters required for this model are patellar tendon (PT) moment arm, tibial plateau-tibial axis angle and PT tendon-tibial axis angle. These parameters were determined *in vivo* during knee extension using videofluoroscopy. Image distortion in videofluoroscopy however requires appropriate non-linear correction methods, in order to obtain accurate biomechanical quantitative measurements. For this purpose an algorithm for two-dimensional coordinate reconstruction and non-linear distortion correction using a polynomial method was developed. The measurement error obtained using an image intensifier - video system was 0.246 $\pm$0.111 mm over a 180 mm x 180 mm field of view. Five males (mean age 20.8 $\pm$years, mass 79.2 $\pm$7.2 kg and height 179 $\pm$ 3.2 cm) without knee joint injury history participated in the study. The mean PT moment arm at full extension was 33.81 $\pm$3.44 mm, increased to a maximum of 39.87 $\pm$2.4 mm at 0.78 rad of knee flexion and decreased to 33.63 $\pm$4.01 mm at 1.57 rad. The PT-tibial plateau angle was 1.96 $\pm$0.12 rad at full extension and decreased linearly to 1.53 $\pm$0.05 rad at 1.57 rad of knee flexion. The mean angle between the tibial plateau and the tibial long axis was 1.48 $\pm$0.04 rad.

The muscular and tibiofemoral contact forces during isokinetic knee extension were examined at angular velocities ranging from 0.52 rad·s^{-1} to 3.66 rad·s^{-1}. The maximum moment (mean $\pm$SD) ranged from 226.20 $\pm$39.52 Nm at 0.52 rad·s^{-1} to 166 $\pm$27.56 Nm at 3.66 rad·s^{-1}. These differences were significant ($F_{3,12}$=17.9, $p<0.05$) and subsequent *post hoc* tests revealed that the significant differences were between the moments at 0.52 rad·s^{-1} and 2.62-3.66 rad·s^{-1}. The maximum muscular force ranged from 7.55 $\pm$0.49 times body weight (BW) at 0.52 rad·s^{-1} to 5.72 $\pm$0.94 BW at 3.66 rad·s^{-1}. The compressive tibiofemoral force ranged from 7.53 $\pm$0.49 BW at 0.52 rad·s^{-1} to 5.68 $\pm$0.91 BW at 3.66 rad·s^{-1} and the shear tibiofemoral force from 0.94 $\pm$0.48 BW to 0.83 $\pm$0.35 BW respectively. These differences were significant for both maximum muscular force ($F_{3,12}$=13.7, $p<0.05$) and compressive tibiofemoral force ($F_{3,12}$=13.57, $p<0.05$). Differences between the shear forces at the different angular velocities were not significant ($F_{3,12}$=0.64, $p>0.05$).

These results indicate that the forces developed during maximal isokinetic knee extension are significantly reduced relative to other dynamic activities and therefore isokinetic dynamometry is a safe and effective method for muscle function assessment, training and rehabilitation, provided that appropriate correction methods for the mechanical and methodological errors are implemented.

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CHAPTER 1

INTRODUCTION

Isokinetics is defined as the dynamic muscular activation during isolated joint movements performed at constant angular velocity (Thistle *et al.*, 1967; Hislop and Perrine, 1967; Perrine, 1968). Isokinetic movements require the use of a special dynamometer that controls and maintains the angular velocity of movement constant, by providing a resistive force that is equivalent to the muscular forces applied to the dynamometer. It must be emphasised that the term constant velocity refers to the velocity of the limb and not the linear velocity of muscular activation (concentric or eccentric) (Hinson *et al.*, 1979). The optimal muscular loading (resistive moment equivalent to muscular moment) and control of the angular velocity during isokinetic movements resulted in widespread applications in the areas of muscle testing and rehabilitation (Baltzopoulos and Brodie, 1989).

The measurement of the muscular moment from the equivalent resistive moment developed by the isokinetic dynamometer is affected by several mechanical and methodological problems (Winter *et al.*, 1981; Sapega *et al.*, 1982). The resistive moment is affected by the gravitational forces if the movement is taking place in the sagittal or frontal planes (Herzog, 1988). The resistive moment is also affected by the inertial forces during the initial acceleration phase of the movement (Sapega *et al.*, 1982). Methodological problems that affect muscular output are subject positioning, motivation and isolation of other muscular activity (e.g. Knoeppel, 1985; Hald and Bottjen, 1987). Several procedures for the correction of mechanical and methodological errors have been developed (e.g. Nelson and Duncan, 1983; Bemben *et al.*, 1988), but the validity of some procedures has been questioned (Sinacore *et al.*, 1983; Murray, 1986).

Another important consideration in isokinetic dynamometry is the resultant muscular and joint forces developed during constant angular velocity movements. This is important for rehabilitation because the muscular and resistive moments developed are maximal throughout the range of movement. Despite the plethora of studies for the examination of muscle function in normal and pathological conditions (for a review see Baltzopoulos and Brodie, 1989), only a limited number of studies examined the resultant muscle forces and intersegmental loading during isokinetic movements (Wickiewicz *et al.*, 1984; Nisell *et al.*, 1989).

The present study considers the main problems of isokinetic dynamometry that influence muscular performance, measurement of isokinetic parameters, assessment of muscle function and development of muscle and joint forces.

Chapter 2 is a review of the literature, examining the development and present state of research on the methodological and mechanical limitations of isokinetic dynamometry.

The most significant methodological problem that affects the measurement of maximum torque output and consequently the assessment of muscle function is visual feedback of muscular output. Chapter 3 examines the effects of visual feedback on maximal voluntary activation during isokinetic knee assessment.

The development and maintenance of constant angular velocity is another important aspect of isokinetic dynamometry. This condition (constant angular velocity) is fundamental to the isokinetic principle and the assessment of muscle function during velocity controlled movements. Chapter 4 examines the kinematics of isokinetic knee extension and the filtering requirements for the measurement of angular velocity from angular position-time data.

The study of joint forces during isokinetic knee extension requires the measurement of joint parameters (e.g. joint centre of rotation, muscle moment arm etc). Chapter 5 describes the development and validation of a system for the measurement of joint motion using videofluoroscopy. Chapter 6 describes the measurement of knee joint parameters using the above videofluoroscopy system.

Chapter 7 examines the muscle and joint forces during isokinetic knee extension at different angular velocities of movement using a two dimensional biomechanical model and knee joint parameters from Chapter 6.

Chapter 8 summarises the conclusions of the above studies and includes recommendations for isokinetic testing and rehabilitation.

An important consideration in the above studies was the definition of the terms adopted. A number of inappropriate or incorrect terms are used in the areas of biomechanics and muscle physiology (Cavanagh, 1988). It is therefore essential at this point to clarify and explain the definition of several controversial terms.

One of the most inaccurate terms in relation to muscular action is "muscular contraction", because the word contraction (derived from the Latin "contrahere" - "to draw together") is associated with shortening (Cavanagh, 1988). Expressions such as "eccentric contraction" or "isometric contraction" are therefore unacceptable. Cavanagh (1988) suggested use of the term "muscle action" but reported:

"...Any change in terminology, especially one so widespread as replacing "muscle contraction" with "muscle action" will inevitably meet resistance on the grounds that if the existing terms were good enough for our distinguished predecessors-including several Nobel laureates (Hill, 1949; Szent-Gyorgi, 1960)-then they should be good enough for us. "

The term proposed in the above report however ("muscle action"), is also usually associated with movement and terms such as "isometric action" are also inappropriate. The term proposed and adopted in the present thesis is "activation". This term covers not only the concept of movement (both at external and internal -sarcomere- levels) but also the concept of muscular activity without external movement (isometric). The derived terms have also been adopted (maximum voluntary activation instead of maximum voluntary contraction, isokinetic activation instead of isokinetic contraction etc).

Another incorrect term in relation to the resultant forces and moments during knee joint movement is "knee forces and moments" (Paul, 1985). The term "intersegmental force" is adopted in the present thesis to describe the contact forces exerted at the tibiofemoral joint.

The terms "torque" and "moment" both describe the turning effect of a force applied on a rigid body (Spiegel, 1980). The two terms are used interchangeably in the present thesis.

This thesis is presented using the "alternative format" proposed by Thomas and Nelson (1990). The main difference from the conventional format of a thesis is that the main Chapters of the thesis describing the experimental procedures and the discussion of the results are presented as complete manuscripts in journal publication form.

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CHAPTER 2

REVIEW OF THE LITERATURE

Baltzopoulos, V.

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ISOKINETIC DYNAMOMETRY

The introduction of isokinetic dynamometry (Thistle *et al.*, 1967; Hislop and Perrine, 1967; Perrine, 1968; Moffroid *et al.*, 1969), allowed the accurate assessment of muscle function in dynamic conditions. During isokinetic isolated joint movements the limb is attached to the input arm of an isokinetic dynamometer that controls the angular velocity of movement. The angular velocity is preset and the resistive mechanism of the dynamometer is activated when the angular velocity of the system accelerated by the examined muscle group, attains the level of the preset velocity. The resistive force is variable and equivalent to the muscular force applied to the dynamometer and consequently the limb angular velocity is maintained constant throughout the range of movement (ROM).

Muscular force and consequently joint moment is affected by joint position because of different biomechanical properties of the musculoskeletal system and the presence of any pathological conditions. During isokinetic movements the resistive moment developed by the dynamometer is equivalent to the muscular moment at different joint positions, providing optimal loading in dynamic conditions. The muscular moment is therefore measured from the equivalent resistive moment developed by the isokinetic dynamometer. Specific aspects of muscle function (e.g. maximum force and muscular endurance) are assessed using the maximum resistive torque, measured over different testing periods. This also allows accurate examination of muscle mechanics (force-velocity, force-length relationship) and assessment of muscle function in pathological conditions (muscle-joint injury) and rehabilitation. (e.g. Caiozzo *et al.* 1981; Parker 1982; Osternig *et al.* 1983; Grimby 1985).

OPERATION OF ISOKINETIC DYNAMOMETERS

Isokinetic dynamometers operate using either electromechanical (e.g. CYBEX) or hydraulic components (e.g. AKRON). The resistive mechanism in these dynamometers is passive and therefore resistance is developed only as a reaction to the applied muscular torque. This allows concentric muscular activation only. More recently, electromechanical dynamometers with active mechanisms have been developed (e.g. LIDO, KINCOM) that allow both concentric and eccentric muscular activation during constant velocity movements (Farrel and Richards, 1986; Francis and Hoobler, 1987). These dynamometers drive the limb at the preset angular velocity, irrespective of the muscular capabilities. The SPARK system (Seger *et al.*, 1988) allows the assessment of concentric and eccentric muscle function during isokinetic (constant velocity), linear acceleration, deceleration or a combination of the above movement modes. The operation of the resistive mechanism and the control of the angular velocity and acceleration in the above systems requires the use of appropriate microcomputer systems (e.g. Gransberg and Knutsson, 1983). The use of microcomputers also facilitates the application of gravitational and inertial correction methods and the provision of visual feedback of muscular output during the test (Richards and Cooper, 1982; Osternig *et al.*, 1982; Sapega *et al.*, 1983; Baltzopoulos and Brodie, 1989). The computation of several parameters derived from muscular torque data (e.g. mechanical work, reciprocal and bilateral muscle group ratios), is also simplified

(Potash *et al.*, 1983; Baltzopoulos and Brodie, 1989). Data analysis time and measurement error are also improved allowing accurate and time efficient assessment of muscle function, offering immediate muscular performance feedback.

ISOKINETIC PARAMETERS.

The maximum resistive moment during isokinetic movements is an indicator of the maximum muscular force exerted in dynamic conditions. Various testing protocols have been used for the assessment of maximum torque in isokinetic conditions using different number of repetitions. Maximum torque is usually evaluated from 2-6 maximal repetitions and is defined as the maximum single torque measurement during these repetitions (Sawhill *et al.* 1982; Jenkins *et al.*, 1984; Dibrezzo *et al.*, 1985; Baltzopoulos *et al.*, 1988). Another method for the assessment of maximum muscular force is to use the mean moment from a number of repetitions (Morris *et al.*, 1983; Yates and Kamon, 1983; Patton and Duggan, 1987). The biomechanical properties of the musculoskeletal system at different joint angles affect the muscular moment output at different angular positions (Thorstenssson *et al.*, 1976; Osternig *et al.*, 1983). The mean moment computed from resistive moment measurements at different joint positions is not an appropriate measure of muscle function, because there is no information about the joint position and therefore, the force-length relationship of the activated muscle group.

The function of reciprocal muscle groups is examined using the ratio of the maximum muscular moment developed by the muscle groups (e.g. knee flexors-extensors). It is an indication of normal joint function and stability and therefore an important parameter in isokinetic assessment of muscle function (Goslin and Charteris, 1979; Knight, 1980; Campbell and Glenn, 1982; Housh *et al.*, 1984). The reciprocal muscle group ratio depends on the physiological-biomechanical properties of the muscle groups during reciprocal isokinetic movements and can be used as rehabilitation-training target in order to restore normal joint function (Grace *et al.*, 1984; Grimby, 1985; Grimby *et al.*, 1980; Klopfer and Greij, 1988).

LIMITATIONS OF ISOKINETIC DYNAMOMETRY

Mechanical Problems

During isokinetic movements in the sagittal or frontal planes, muscular activation is affected by the gravitational forces of the limb-input arm system. The resistive moment developed by the dynamometer is equivalent to the resultant of the muscular and gravitational moments (Winter *et al.*, 1981; Herzog, 1988). The effect of gravitational forces on isokinetic parameters has been investigated and significant errors in the measurement of muscular torque, muscle group torque ratio and force-velocity relationship have been reported (Appen and Duncan, 1986; Fillyaw *et al.*, 1986). For example it has been shown that although there is an increase in the uncorrected knee flexor/extensor ratio with an increase in the angular velocity of movement, (indicating a decline in knee extensor activity), the gravity-corrected ratios remain relatively constant with increasing angular velocity (Appen and Duncan, 1986). The gravitational error depends on the muscular capabilities relative to the mass involved in the movement and the angular position. For typical knee extension movements, the maximum torque measurement error reported ranged from 11.7% at 2.09 rad·s^{-1} to 24.3% at 4.19 rad·s^{-1} (Herzog, 1988). Appen and Duncan (1986) reported that the gravitational error in the measurement of knee flexor/extensor ratio ranged from 18.5% at 1.05 rad·s^{-1} to 37.7% at 5.24 rad·s^{-1}. It is evident that isokinetic muscular moment measurements without appropriate correction for the effect of gravitational forces result in erroneous assessment of muscle function.

A simple method for gravity correction is the recording of the gravitational torque generated by the weight of the limb-lever arm system at a specific angular position within the ROM, while the limb is allowed to fall passively against the resistance of the dynamometer (Nelson and Duncan, 1983). This procedure must be performed at the minimum angular velocity or at isometric conditions if possible in order to avoid inertial effects. The limb must be completely relaxed during the passive fall in order to avoid errors in the measurement of the gravitational torque. In practice it is suggested that several trials are performed to ensure complete muscular inactivity during this procedure. In order to reduce measurement errors, the gravitational torque should be recorded close to its maximum position (i.e. around the horizontal axis). The gravitational torque throughout the ROM is then computed as a function of angular position and this correction factor is added to the dynamometer torque recording produced by muscle groups opposed by gravity (knee extensors during movements on the sagittal plane) or subtracted from the recorded torque produced by muscle groups facilitated by gravity (knee flexors).

The resistive moment is equivalent to the muscular moment applied to the dynamometer, provided the angular velocity of the system is constant and equal to the preset velocity. In the initial period of the movement from a stationary position until the development of the preset angular velocity, the limb-input arm system is accelerated by the activated muscle group(s), as there is no resistive moment exerted by the dynamometer. Consequently, when the preset velocity is attained by the system, a resistive moment is developed by the dynamometer in order to decelerate the limb-input arm system ("torque overshoot"). The magnitude of this resistive moment exceeds the muscular moment and it is most prominent at slow angular velocities (Sapega *et al.*, 1982). During this initial period the resistive dynamometer moment is not equivalent to the muscular moment and appropriate correction methods are essential before the assessment of muscle function (Murray, 1986).

Analog electrical filters have been used for the filtering of the resistive moment (Sapega *et al.*, 1982; Bemben *et al.*, 1988). The application of electrical filters (equivalent to a first order, non-recursive digital filter) is not appropriate however, because these filters affect the amplitude of the resistive moment throughout the ROM and introduce a phase shift in the signal (Sinacore *et al.*, 1983). The most appropriate correction method is the computation of the muscular moment during the initial period from the angular acceleration of the system and moment of inertia data. The measurement of kinematic parameters from position-time data is therefore essential. The angular position data however, contain random error resulting from the analog to digital conversion, requiring appropriate filtering methods before differentiation and measurement of the kinematic parameters (Murray, 1986; Murray and Harrison 1986).

A number of different methods for data filtering and differentiation are available and their applications in biomechanical analysis have been previously examined (e.g. Wood, 1982). Piecewise cubic splines (Reinsch, 1967) and digital filters (Winter, 1974; Vaughan, 1982) have been applied in isokinetic analysis (Osternig *et al.*, 1983; Murray, 1986) despite the problems associated with the implementation of these methods (Pezzack *et al.*, 1977; Phillips and Roberts, 1983; Smith, 1989).

Methodological Problems

Constant velocity movements are rarely performed during every day or sporting activities and therefore it is important to familiarise the subjects with this mode of dynamic activation before the test. The complete spectrum of the test angular velocities must be included in the familiarisation session. Standardised instructions should be given to the subjects explaining the main principles of isokinetic testing (variable resistance, constant velocity) and the importance of maximum muscular effort during the tests. The maintenance of maximum effort throughout the range of movement by both muscle groups during reciprocal movements (e.g. knee extension-flexion) must be specifically emphasised.

Appropriate stabilisation of the subject ensures that the recorded torque is generated by the examined muscle group without contribution from any other muscular action (Patteson *et al.*, 1984; Knoeppel, 1985). During a typical knee extension-flexion movement for example the opposite leg, waist, chest and arms should be stabilised with appropriate belts. The examined limb must be secured to the input arm of the dynamometer in order to prevent injury and avoid resistive torque overshoot from impacts between the limb and the input arm (Herzog, 1988). The most comfortable position for the attachment of the input arm is at a distal limb position, although the shear joint force is increased (Johnson, 1982; Lavin and Gross, 1990). During rehabilitation of ligamentous injuries, a proximal input arm position reduces the shear joint force and consequently the stress on the ligaments supporting the joint in the shear direction (Johhson, 1982; Nisell *et al.*, 1989). Mechanically, the torque output is not affected by the position of the input arm on the limb (i.e. the moment arm of the resistive force). The uncomfortable feeling of a proximal input arm position on the limb however (e.g. pressure on the tibia tuberosity during knee extension) can influence muscular torque output (Taylor and Casey, 1986). The position of the subject must be standardised in order to avoid variations in the length and function of the examined muscle group (e.g. hip angle during knee extension) (Knoeppel, 1985; Barr and Duncan, 1988). The axis of rotation of the dynamometer should be aligned with the joint axis. Although the instantaneous axis of rotation of a joint is difficult to establish, an approximation using anatomical landmarks is essential. Misalignment of joint and dynamometer axes affects the measurement of muscular torque, although the error for normal subjects and small deviations is negligible ($<2\%$) (Herzog, 1988).

Visual feedback (VF) of muscular output during isokinetic testing has a significant effect on the maximum torque output (Figoni and Morris, 1984; Hald and Bottjen, 1987). The magnitude of this effect depends on the angular velocity of movement. Figoni and Morris reported that VF improved maximum torque output by 12% at 0.26 $rad \cdot s^{-1}$ but there was no improvement at 5.27 $rad \cdot s^{-1}$. Hald and Bottjen reported improvements of 6% and 3% at 1.05 and 3.14 $rad \cdot s^{-1}$ respectively. It is important therefore to provide visual feedback of the muscular performance during isokinetic tests especially during maximum torque assessment at slow angular velocities. The real-time display of the torque output on a computer monitor or the analogue recorder can be used as visual feedback sources. It is important to give detailed instructions to the subject on the interpretation of the different visual feedback sources and the performance target.

JOINT FORCES DURING ISOKINETIC MOVEMENTS

Isokinetic dynamometry has widespread applications in the assessment and rehabilitation of dynamic muscle function. The majority of these applications are in the areas of knee function assessment, training and injury rehabilitation. The muscle and intersegmental forces developed however, have been examined by a limited number of studies (Wickiewicz *et al.*, 1984; Nisell *et al.*, 1986; Nisell *et al.*, 1989).

Johnson (1982) and Lavin and Gross (1990) examined the effects of a modified input arm for the CYBEX dynamometer on tibial translation during isometric conditions only. It was concluded that by reducing the moment arm of the resistive dynamometer force, the shear joint force and tibial translation are reduced (Malone, 1986; Timm, 1986). Nisell *et al.* (1989) reported that the maximum tibiofemoral compressive force during isokinetic knee extension was 9 times body weight (BW) and was not affected significantly by altering the moment arm of the resistive force. The maximum shear force was 1 BW and was reduced using a proximal position for the attachment of the input arm.

The examination of the muscular and tibiofemoral contact forces during isokinetic knee extension is important, because the load on the ligaments and muscles at different angular velocities can be estimated, preventing training or rehabilitation-induced injuries.

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CHAPTER 3

THE EFFECTS OF VISUAL FEEDBACK ON MAXIMUM MUSCULAR ACTIVATION DURING ISOKINETIC KNEE EXTENSION-FLEXION MOVEMENTS.

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ABSTRACT

The purpose of this study was to examine the effects of visual feedback on isokinetic maximum torque and reciprocal muscle group ratio of the knee extensors and flexors at a slow (1.05 $rad \cdot s^{-1}$) and a fast (3.14 $rad \cdot s^{-1}$) speed of movement. The real-time gravity corrected torque output, used as the visual feedback source, elicited a significant ($p < 0.05$) increase in the maximum torque output of both muscle groups (8% and 6% respectively) at the slow speed. There was no significant improvement at the fast speed of movement. The knee flexor/extensor ratio was not effected by visual feedback or speed of movement. It was concluded that visual feedback of the torque output can improve maximum voluntary activation in isokinetic dynamometry under certain restrictions on speed and range of movement.

INTRODUCTION

Accurate and objective assessment of muscle function is essential in both injury rehabilitation and the development of strength for specific purposes. Isokinetic dynamometry is widely used in such assessment because it offers the capability of providing variable resistance that corresponds to the muscular forces and constant preselected speed of movement (Grimby 1985; Jenkins *et al.*, 1984; Parker, 1982).

The main isokinetic parameters used in the assessment of muscle function are maximum torque and reciprocal muscle group ratio. Maximum torque is defined as the highest muscular torque value from 2-6 maximal repetitions (Jenkins *et al.* 1984; Baltzopoulos *et al.* 1987). Reciprocal muscle group ratio is the maximum torque quotient of two reciprocal muscle groups (e.g., hamstrings/quadriceps (h/q) ratio). This ratio is an indication of reciprocal joint support and stability, and its accurate measurement is important for injury prevention and rehabilitation (Campbell and Glenn 1982). The measurement of the isokinetic parameters, however, is affected by gravitational and inertial forces during the test. Accurate assessment of muscle function requires appropriate correction methods. The effect of gravitational forces on the measurement of several isokinetic parameters has previously been investigated and significant measurement errors have been reported (Winter *et al.*, 1981; Appen and Duncan, 1986). The importance of gravity correction for intrasubject comparisons, however, has been questioned because the effect of the gravitational forces is uniform over the same experimental procedures (Hald and Bottjen, 1987). This approach is valid only if the torque is recorded at a constant, predetermined angular position (i.e. similar gravitational force), using the same experimental procedure in all tests.

The development of the constant preset speed is another potential methodological problem in isokinetic data analysis. A finite period of time is necessary for the development of the preset speed. This acceleration period increases with increasing preset speed. The initially overspeeding limb is decelerated to the level of the preset speed by the resistive mechanism of the dynamometer. The torque overshoot that is frequently observed in the beginning of the movement, represents this resistive torque and must not be interpreted as muscular torque. (Sapega *et al.*, 1982).

Other sources of variability in isokinetic testing include the positioning and stabilisation of the subject on the dynamometer, rest periods between tests at different angular velocities, test instructions, and motivation during the test. Accurate assessment of muscle function and valid comparisons of isokinetic data require standardised testing and measurement protocols implementing appropriate correction methods and maximising voluntary muscular activation.

Visual feedback (VF) of the muscular torque output during isokinetic testing is a source of variability in the measurement of isokinetic parameters and consequently in the assessment of muscle function. (Riggsbee, 1983; Figoni and Morris, 1984; Hald and Bottjen, 1987). Riggsbee (1983) suggested that using the analogue recorder as a visual feedback source can improve patient response during isokinetic testing, although no experimental data to support this hypothesis were reported at that time. Subsequently, however, Figoni and Morris (1984) reported that VF improved the maximum torque output of both quadriceps and hamstrings by 12% at a slow speed of movement (0.26 $rad \cdot s^{-1}$), but there was no improvement at a fast speed (5.24 $rad \cdot s^{-1}$). Hald and Bottjen (1987) reported that VF improved the maximum torque output of both muscle groups by 6% at 1.05 $rad \cdot s^{-1}$ and approximately 3% at 3.14 $rad \cdot s^{-1}$. The analogue torque recorder of the isokinetic dynamometer was used as the VF source in the above studies. It is evident from these results that VF has a significant effect on torque output. The magnitude of this effect depends on the angular speed of movement. The isokinetic parameters, however, were not corrected for the effect of gravitational and inertial forces although Figoni and Morris (1984) measured the maximum torque after the first torque peak in order to avoid interpretation of the torque overshoot artifact as muscular torque. Therefore, it is not clear, whether VF affects torque output or this effect is a methodological artifact.

The purpose of this study was to examine the effect of VF on maximum torque and reciprocal muscle group ratio. The gravity corrected real-time display of the torque output was used as the VF source.

METHODS

Instrumentation

An AKRON isokinetic dynamometer was used to measure muscular torque. This system permits isolated joint testing at a constant, preset speed of movement that can be set independently for reciprocal muscle groups. The dynamometer was interfaced with an Intel 82086 based microcomputer for data collection and analysis (Baltzopoulos and Brodie 1989). The gravitational torque throughout the ROM was registered before the test. The torque data during the test were corrected for gravity and displayed on the monitor in

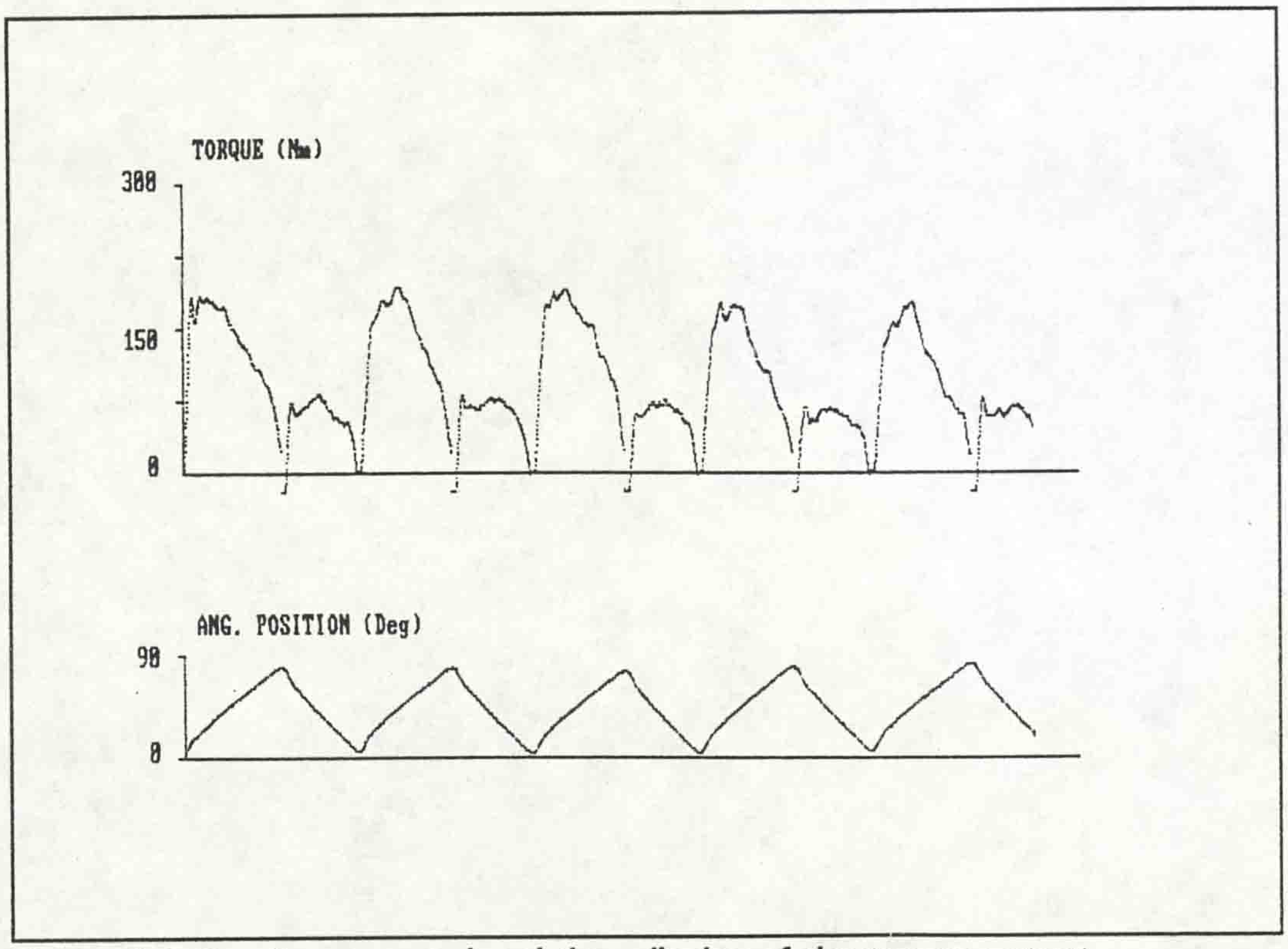

Figure 3.1 Gravity corrected real-time display of the torque output.

real-time (Baltzopoulos and Brodie 1989) (Fig. 3.1). Data from the constant angular velocity periods only were used to determine the isokinetic parameters.

The above computer system is described in detail by Baltzopoulos and Brodie (1989).

Subjects

Ten healthy males without any history of joint injury gave informed consent and volunteered to participate in this study. The subjects had a mean age of 25.8 ±1.7 years, mass 69.9 ±4.1 kg, and averaged 177.7 ±5.9 cm in height. The right leg was used in all the tests.

Procedures

A pilot study indicated that the effect of VF was similar for male and female subjects supporting previous findings that the effect of VF on maximum torque is not sex related (Peacock *et al.*, 1981). The testing protocol consisted of five maximal reciprocal repetitions of the knee extensors and flexors, since development of the maximum torque requires two to six repetitions (Baltzopoulos *et al.*, 1987). The test was performed at a slow (1.05 $rad{\cdot}s^{-1}$) and a fast (3.14 $rad{\cdot}s^{-1}$) speed of movement with and without visual feedback. The range of movement (ROM) for all tests was form 1.57 to 0.52 rad of knee flexion. The tests were completely randomized, and rest periods of five minutes were given before the tests. A familiarisation and warm-up period was given five minutes before the test. The tests were performed with the subjects seated and the thigh, pelvis and non-involved foot stabilised with appropriate belts. The axis of rotation of the dynamometer was aligned with the most prominent point of the lateral femoral condyle.

During all tests the computer monitor was positioned one metre away from the subject at eye level. All subjects were instructed to carefully observe the monitor. During the VF tests, real-time display of the gravity- corrected muscular torque was provided. During the no-VF tests, the monitor was blank. All subjects were given written, standardised instructions to work as hard and fast as possible against the resistance of the dynamometer during the tests and to try to overcome the torque curves from the previous repetitions displayed on the computer monitor.

Gravity-corrected torque data from constant speed periods only were used to calculate the following isokinetic parameters: 1) Maximum torque in extension (quadriceps) and flexion (hamstrings), 2) knee angle of maximum torque, and 3) hamstrings/quadriceps ratio. The study was designed to examine muscular torque differences between VF and no-VF condition, speeds of movement (slow-fast), and muscle groups (quadriceps-hamstrings).

Data Analysis

Differences between the VF conditions, speeds of movement and muscle groups were analysed using a three-factor (2 X 2 X 2) repeated measures ANOVA test, with muscular torque measurements as the performance variable. Significance was accepted at the 95% probability level.

RESULTS

There was a main effect for visual feedback ($F_{1,9} = 13.6$, $p < 0.05$), speed of movement ($F_{1,9} = 148.1$, $p < 0.05$), and muscle group ($F_{1,9} = 135.5$, $p < 0.05$). The torque measurements are presented graphically in

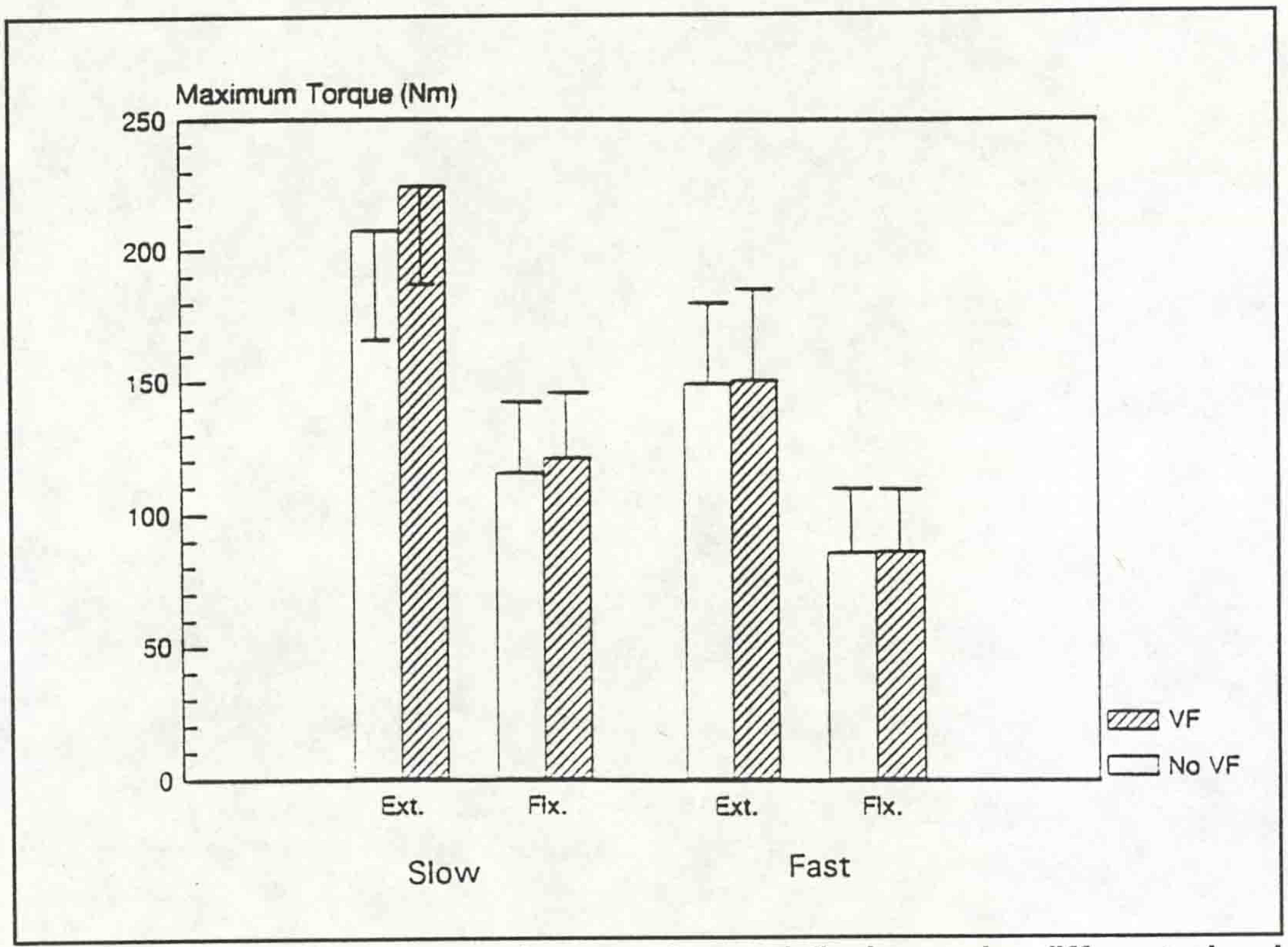

Figure 3.2. Maximum torque in extension and flexion under different visual feedback and angular velocity conditions.

Fig.3.2. At the slow speed, visual feedback improved the torque output by 8% and 6% in extension and flexion, respectively. There was no improvement at the fast speed. A two-factor (VF-speed) repeated measures ANOVA test was used to analyse the hamstrings/quadriceps ratio. There were no significant effects. The mean ±SD of the hamstrings/quadriceps ratio measurements are presented in Table 3.1.

Table 3.1. Mean ± SD of the hamstrings/quadriceps ratio measurements at different VF conditions and speeds of movement.

$1.05\ rad \cdot s^{-1}$		$3.14\ rad \cdot s^{-1}$	
No VF	VF	No VF	VF
0.57 ±0.10	0.54 ±0.06	0.58 ±0.78	0.58 ±0.08

This ratio was approximately 0.57 under all conditions of VF and speed of movement.

DISCUSSION

The mean extension torque obtained in this study under the VF condition is approximately 4% higher than respective values in previous VF studies (Figoni and Morris, 1984; Hald and Bottjen, 1987). The flexion torque, however, is lower, with differences ranging from 13% to 23%. These differences may be attributed to individual differences and the fact that no gravity correction procedures were used in previous studies.

The constant velocity periods of the movement were determined by measuring the angular velocity from the torque-position data (Baltzopoulos and Brodie 1989). Torque data from the isokinetic part of the movement only were used for subsequent analysis. This method does not affect the amplitude and phase of the torque signal (Sinacore *et al.*, 1983) and ensures that the torque overshoot during the initial acceleration period is not interpreted as muscular torque.

Visual feedback had a positive effect on the maximum torque of both muscle groups (quadriceps-hamstrings) at the slow speed only. Similar findings have been reported by Figoni and Morris (1984), although the increase was

12% for both muscle groups, compared to 8% and 6% for extensors and flexors in the present study. Hald and Bottjen (1987) reported a significant increase of 3% at the fast speed and 6% at the slow speed. Statistical significance however was determined using a series of t-tests and not an ANOVA design. It is evident that muscular torque output presented as extrinsic, concurrent, visual feedback (Schmidt, 1982), has a positive, motivational effect on muscular activation. The subjects were required to process the visual feedback and adjust their muscular response during the ongoing repetition. The temporal constraints of the movement may therefore explain the absence of VF effect at the fast speed test (Figoni and Morris, 1984). During the slow speed test (1.05 $rad \cdot s^{-1}$), the 1.05 rad ROM was completed in approximately one second. During the fast speed test, however, the movement time was approximately 300 ms. The reaction time to visual information is approximately 160 to 180 ms (Schmidt, 1982). This suggests that during the slow speed test, approximately 80% of the movement time was available for information processing and response adjustment. On the contrary, during the fast speed test, any response occurs during the last third of the movement when the muscles were beyond their optimal anatomical position for maximum torque production. The effectiveness of VF in isokinetic dynamometry appears to be related to the movement time of a particular testing procedure as a function of range and speed of movement. A positive effect of VF on maximum torque during a fast speed test is possible, provided that the ROM is appropriately set to allow the post-feedback muscular response to occur before the optimal anatomical position of the activated muscle group (e.g. shoulder extension-flexion) is reached. In isokinetic dynamometry however anatomical and apparatus constraints may limit the ROM and therefore the effect of visual feedback.

The effectiveness of visual feedback is also influenced by its precision and accuracy (Schmidt, 1982). With the present isokinetic system, the real-time display of torque and angular position could be presented in different forms. To enhance precision, the display was modified to present the torque output only, over the entire computer monitor (Fig. 3.3), without information

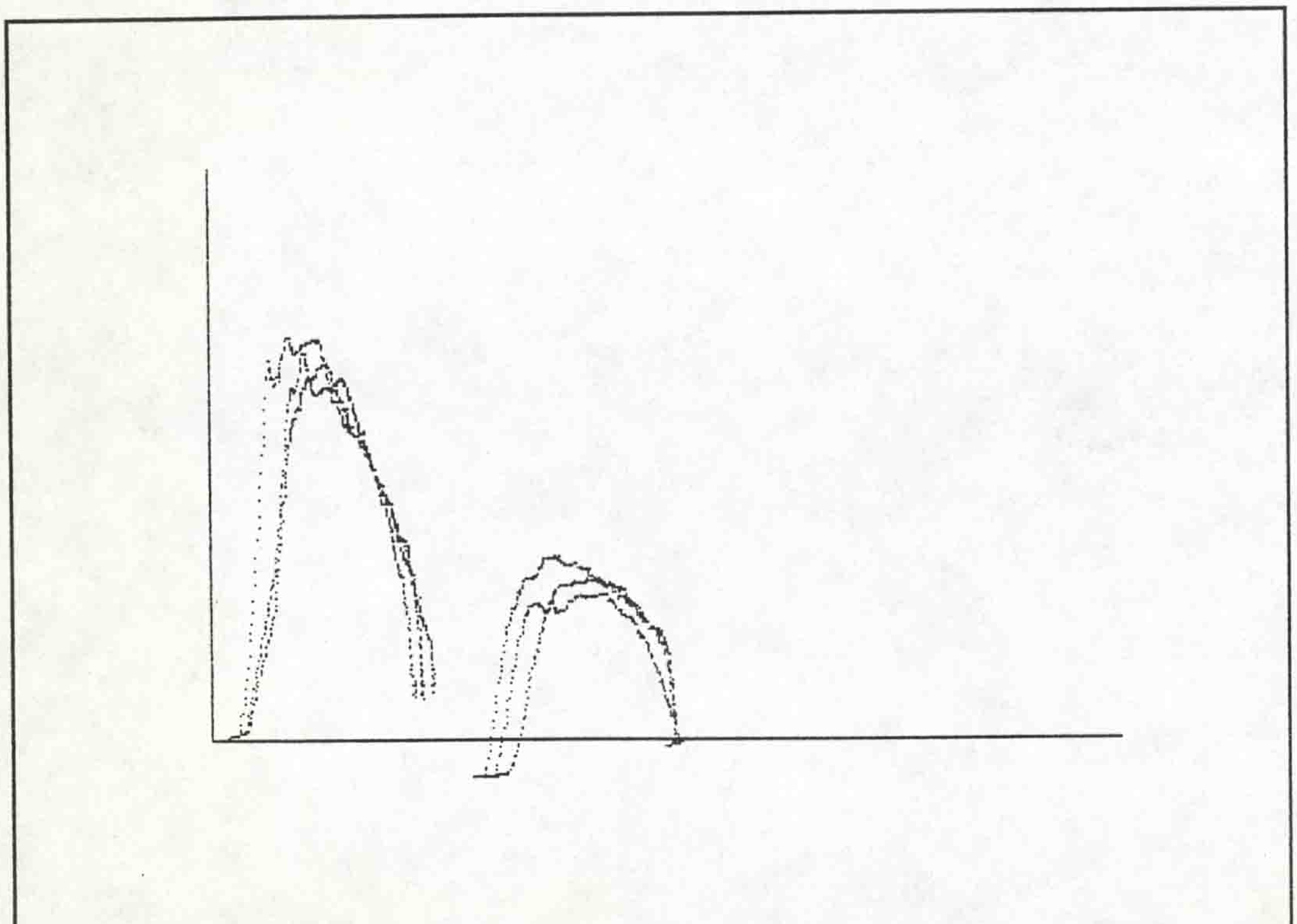

Figure 3.3. Simplified real-time display of torque output used for visual feedback.

that could distract the attention of the subjects (e.g., axes legends, angular position etc). Furthermore the torque output from the different repetitions is superimposed allowing easier comparisons with the previous repetitions during the test.

The hamstring/quadriceps torque ratio was approximately 0.56 at both speeds. An increase of this ratio with increasing speed has been reported previously (Davies *et al.*, 1981; Wyatt and Edwards, 1981). The results of this study, however, support recent findings (Appen and Duncan, 1986) that this increase is a gravitational artifact and the gravity-corrected ratio remains relatively constant with increasing speed. Despite a significant increase in the maximum torque under the VF condition, there is no significant difference in the hamstring/quadriceps ratio, since the maximum torque increase was similar for both muscle groups.

CONCLUSIONS

Within the limitations of the present study, visual feedback elicited: 1) a significant increase in the maximum torque output which was similar for both muscle groups tested; therefore, no effect on the reciprocal muscle group ratio was observed, and 2) an effect that depends on the movement time of a particular testing procedure as a function of range and speed of movement.

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CHAPTER 4

KINEMATIC ANALYSIS OF ISOKINETIC KNEE EXTENSION

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ABSTRACT

The purpose of this study was to examine angular velocity development and muscular joint moment output during isokinetic knee extension. The movement was performed by eight healthy subjects (mean age 22.37 ±3.4 years, mass 76.87 ±7.22 kg and height 181.5 ±4.28 cm), at preset angular velocities of 0.52, 1.57, 2.62 and 3.67 rad.s^{-1} using a computerised Akron isokinetic dynamometer. Angular velocity was determined from differentiation of the displacement time data after optimal smoothing using a low pass digital filter. Maximum joint moment was determined from the part of the movement with the angular velocity within ± 10% of the preset velocity. The mean maximum joint moment ranged from 264.7 ±43.8 Nm at 0.52 rad·s^{-1} to 198.8 ±27.9 Nm at 3.67 rad·s^{-1}. During the initial acceleration period the velocity of the limb exceeded the preset velocity by an average of 145%, 44%, 29% and 18% at the four preset velocities respectively. The constant velocity period ranged from 63.7% at 0.52 rad.s^{-1} to 40.3% of the total movement at 3.67 rad.s^{-1}. These results indicate that the constant velocity assumption during isokinetic movements even after the initial acceleration period are not valid. Appropriate correction methods are required before the measurement of isokinetic parameters and the assessment of dynamic muscle function.

INTRODUCTION

Isokinetic dynamometry has widespread applications in the areas of muscle testing and rehabilitation (Baltzopoulos and Brodie, 1989a; Grimby, 1985) because of its unique features. These include optimal muscle loading by providing variable resistance, equivalent to the muscular forces throughout the range of movement (ROM) and preselected, controlled angular velocity of movement. There are several mechanical problems however associated with the measurement of muscular joint moment using isokinetic dynamometers (Winter *et al.*, 1981; Sapega *et al.*, 1982; Murray and Harrison, 1986), affecting the accuracy and validity of the isokinetic parameters and consequently the conclusions about dynamic muscle function.

The effect of gravitational forces on the isokinetic parameters has been investigated and significant errors in the measurement of maximum moment, reciprocal muscle group ratio and moment-velocity relationship have been reported (Winter *et al.*, 1981; Appen and Duncan, 1986; Fillyaw *et al.*, 1986). Appropriate procedures and correction methods have been developed, however, that are easily implemented and eliminate the gravitational error from the measurement of the isokinetic parameters (Nelson and Duncan, 1983; Baltzopoulos and Brodie, 1989b).

The development and maintenance of the preset angular velocity is another potential methodological problem in isokinetic dynamometry (Sapega *et al.*, 1982; Murray and Harrison, 1986). The initial acceleration period is followed by a period of successive deceleration-acceleration phases after the activation of the resistive mechanism when the angular velocity of movement attains the level of the preset velocity. The resistive moment required to decelerate the system to the level of the preset velocity appears in the resistive moment output of the dynamometer as a prominent "overshoot" (Sapega *et al.*, 1982). Despite this inertial artifact in the dynamometer's resistive moment, the measurement of the temporal development of muscular force is an important factor of muscle mechanics and accurate measurement of the kinematic and kinetic parameters of the movement during this period is important for the examination of muscle function. The constant angular velocity (or "isokinetic") period is followed by a deceleration period at the end of the movement. Most of the isokinetic systems commercially available are interfaced to microcomputers. Moment and angular position data are obtained in digital form. Thus the kinematic parameters of the movement can be obtained from differentiation of position-time data.

The joint moment and angular position data however contain noise from both electrical interference and analogue to digital conversion, subsequently requiring the use of appropriate smoothing techniques for the accurate measurement of the kinematic parameters of the movement. Several smoothing and differentiation methods (e.g. digital filters, spline functions and fourier series) have been applied in other areas of biomechanics research (e.g. Pezzack *et al.*, 1977; Wood, 1982). However, only a limited number of studies examined their applications in isokinetic dynamometry in an effort to overcome the mechanical problems associated with the measurement of joint moment during isokinetic loading (Murray 1986; Murray and Harrison, 1986). These applications, however, use gravitational loading and there is no detailed examination of the effects of angular velocity fluctuations at different preset velocities during muscular loading. This is essential for the measurement of isokinetic parameters and consequently the assessment of muscle function.

The purpose of this study was the examination of the kinematic parameters of an isokinetic knee extension movement. The specific objectives of the study were to examine: 1) the development and maintenance of angular velocity at different preset velocities during muscular loading and 2) the implementation of digital filters for isokinetic data smoothing.

METHODS

Instrumentation

A knee extension movement was examined using an AKRON isokinetic dynamometer (Akron, Norfolk, England), connected to an Intel 82086 based microcomputer for data collection and analysis (Baltzopoulos and Brodie, 1989b). The maximum sampling rate for moment and angular position data from the dynamometer is 320 Hz. The 8 bit digital moment signal provides a resolution of 1.176 Nm in the 0-300 Nm measurement scale and 2.35 Nm in the 0-600 Nm measurement scale. The resolution of the angular position signal is 0.0087 rad (0.5 degrees). The reliability and validity of this system have been previously examined (Baltzopoulos, 1988; Baltzopoulos and Brodie, 1989b)

Subjects and Procedures

Eight males without any history of lower extremity joint injury signed informed consent and volunteered to participate in this study. The subjects had a mean age of 22.37 $\pm$3.4 years, mass 76.87 $\pm$7.22 kg and averaged 181.5 $\pm$4.28 cm in height. The right limb was used in all tests.

The testing protocol consisted of five maximal reciprocal repetitions of the knee extensors at 0.52, 1.57, 2.62 and 3.67 $rad \cdot s^{-1}$. The ROM for all tests was from 1.57 to 0.35 rad (90-20 degrees) of knee flexion. An audio signal indicated the limits of the ROM but the movement was not mechanically restricted to this range. The tests were completely randomised and rest periods of 5 min were given between the tests. A familiarisation and warm-up period was given before the testing session. The tests were performed with the subjects seated and the thigh, pelvis and non-involved foot secured with appropriate belts. The axis of rotation of the dynamometer was aligned with the most prominent point of the lateral femoral condyle. Dynamic calibration may be affected by inertial artifacts and in order to avoid measurements errors, only static calibration was performed using gravitational loading. The gravitational moment of the limb-input arm system throughout the ROM was determined before each test by a passive fall, and the raw joint moment data were corrected for gravity and displayed on the computer monitor in real-time (Baltzopoulos and Brodie, 1989b).

All subjects were given written, standardised instructions to work as hard and as fast as possible against the resistance of the dynamometer and to try to overcome the joint moment curves from the previous repetitions, displayed on the computer monitor during the test. This procedure was followed because visual feedback can improve maximum moment output significantly during slow speed tests (Baltzopoulos *et al.*, 1991).

Data Analysis

Angular displacement data were filtered using a second order Butterworth filter of the form:

$$f_i := a_0 \cdot r_i + a_1 \cdot r_{i-1} + a_2 \cdot r_{i-2} + b_1 \cdot f_{i-1} + b_2 \cdot f_{i-2}$$

where a and b are the filter coefficients determined by the sampling to cutoff frequency ratio and r_i and f_i the raw and filtered data at time i respectively. A second filtering of the data in the reverse direction of time results in a fourth order, zero phase shift filter (Winter *et al.*, 1974; Vaughan, 1982).

The optimal cutoff frequency of the filter was determined by comparing the variance in the residuals with the variance in the raw data (Lesh *et al.*, 1979). The mean number of data points in a complete knee extension movement (sampled with the maximum sampling frequency and averaged over subjects and angular velocities) was 501.7 $\pm$273.2. For this purpose the variance in the raw data was computed from 500 samples of the angular position with the input arm unloaded and locked at 10 random positions in the ROM. The optimal frequency was determined by filtering the data using different cutoff frequencies until the difference between the variance in the raw and filtered data was minimum. The mean optimal cutoff frequency was a positive function of angular velocity and ranged from 2.75 Hz at 0.52 $rad \cdot s^{-1}$ to 4.88 Hz at 3.67 $rad \cdot s^{-1}$. After the determination of the optimal cutoff frequencies, the sampling frequencies used for subsequent data collection and measurement of the angular velocity were according to the Nyquist theorem in order to prevent "aliasing" of the signal (Wood, 1982). The mean sampling frequency ranged from 36.1 $\pm$6.7 Hz at 0.52 rad·s^{-1} to 98.5 $\pm$14.1 Hz at 3.67 rad·s^{-1}.

The angular position series was extended using 30 extra points in order to avoid distortion of the filtered signal (Smith, 1989). The extra data points were collected from the dynamometer as part of the experimental procedure and were not artificially generated. Angular velocity was computed from the filtered data using a five point difference formula (Burden and Faires, 1985).

Isokinetic Parameters.

The angular velocity of movement was computed for the complete ROM. The maximum moment was determined from the part of the movement where the angular velocity was within $\pm$10% of the preset velocity. Knee joint angle at the maximum moment position was also determined. The "isokinetic" part of the movement (i.e. angular velocity within $\pm$10% of the preset velocity) was expressed as a percentage of the total movement during the maximum moment repetition.

RESULTS.

The mean maximum resultant moment ranged from 264.7 $\pm$43.8 Nm at 0.52 rad·s^{-1} to 198.8 $\pm$27.9 Nm at 3.67 rad·s^{-1} (Fig. 4.1). The knee angle at the maximum joint moment position shifted from 1.08 rad of knee flexion at 0.52 rad·s^{-1} to 0.79 rad at 3.67 rad·s^{-1} (Fig. 4.2). During the initial acceleration period the angular velocity of the system exceeded the preset velocity by an average of 145%, 44%, 29% and 18% at the four preset angular velocities respectively (Fig. 4.3). The part of the movement with the angular velocity within ±10% of the preset velocity ranged from 63.7% at 0.52 rad·s^{-1} to 49.3% of the total movement at 3.67 rad·s^{-1} (Fig. 4.4).

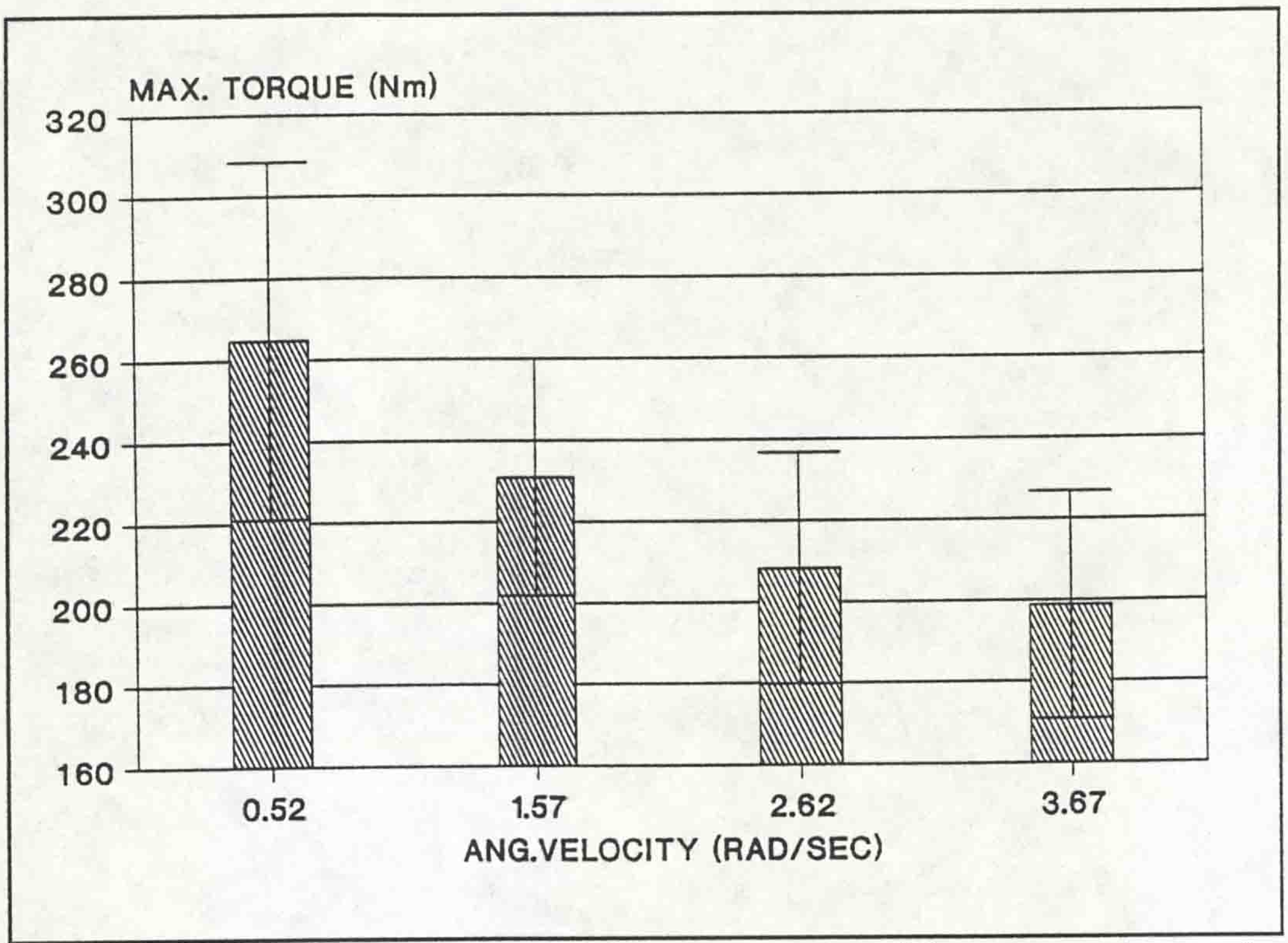

Figure 4.1. Maximum resultant joint moment at different preset angular velocities measured from the constant velocity part of the movement.

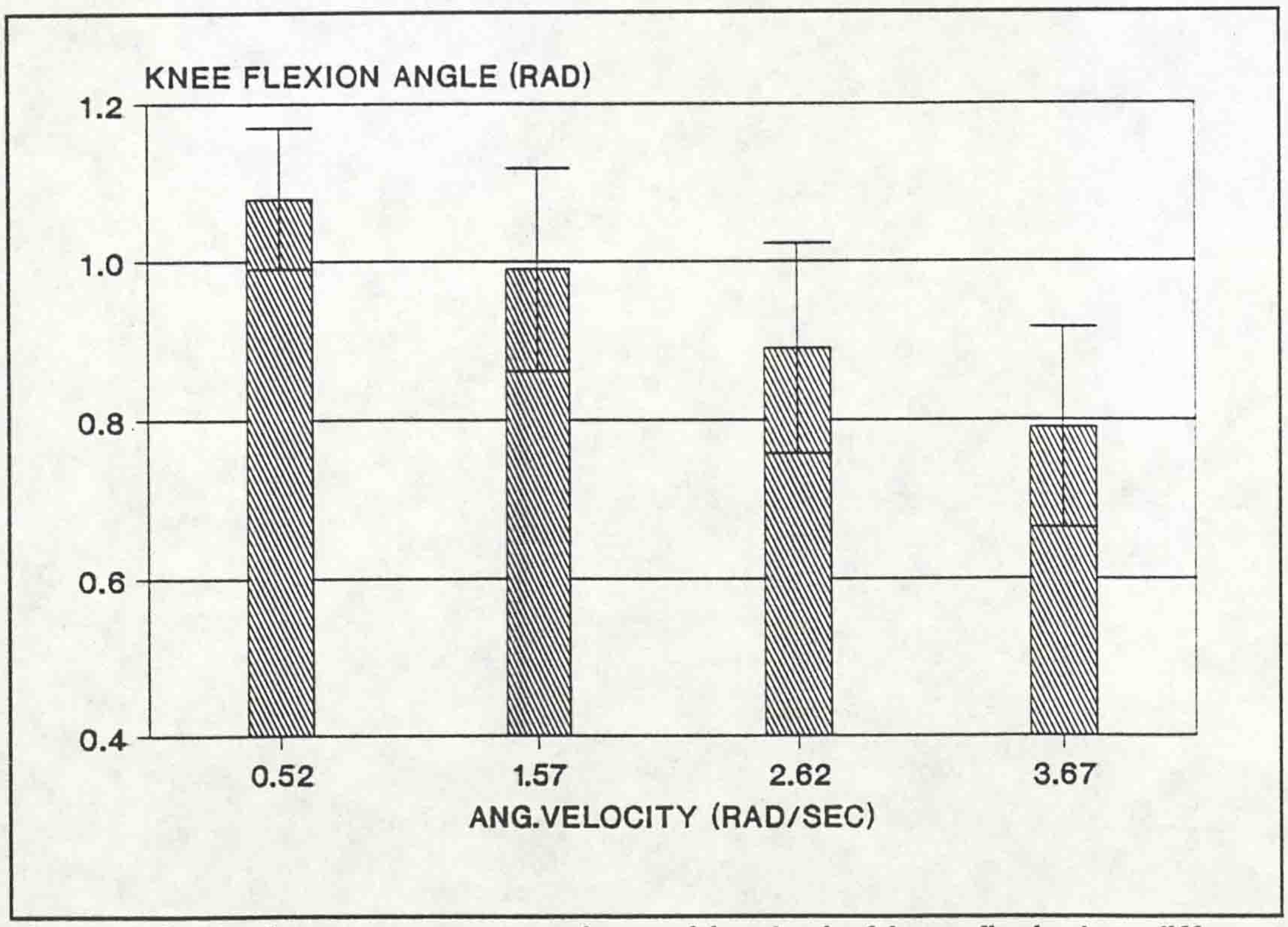

Figure 4.2. Maximum moment angular position (rad of knee flexion) at different preset angular velocities.

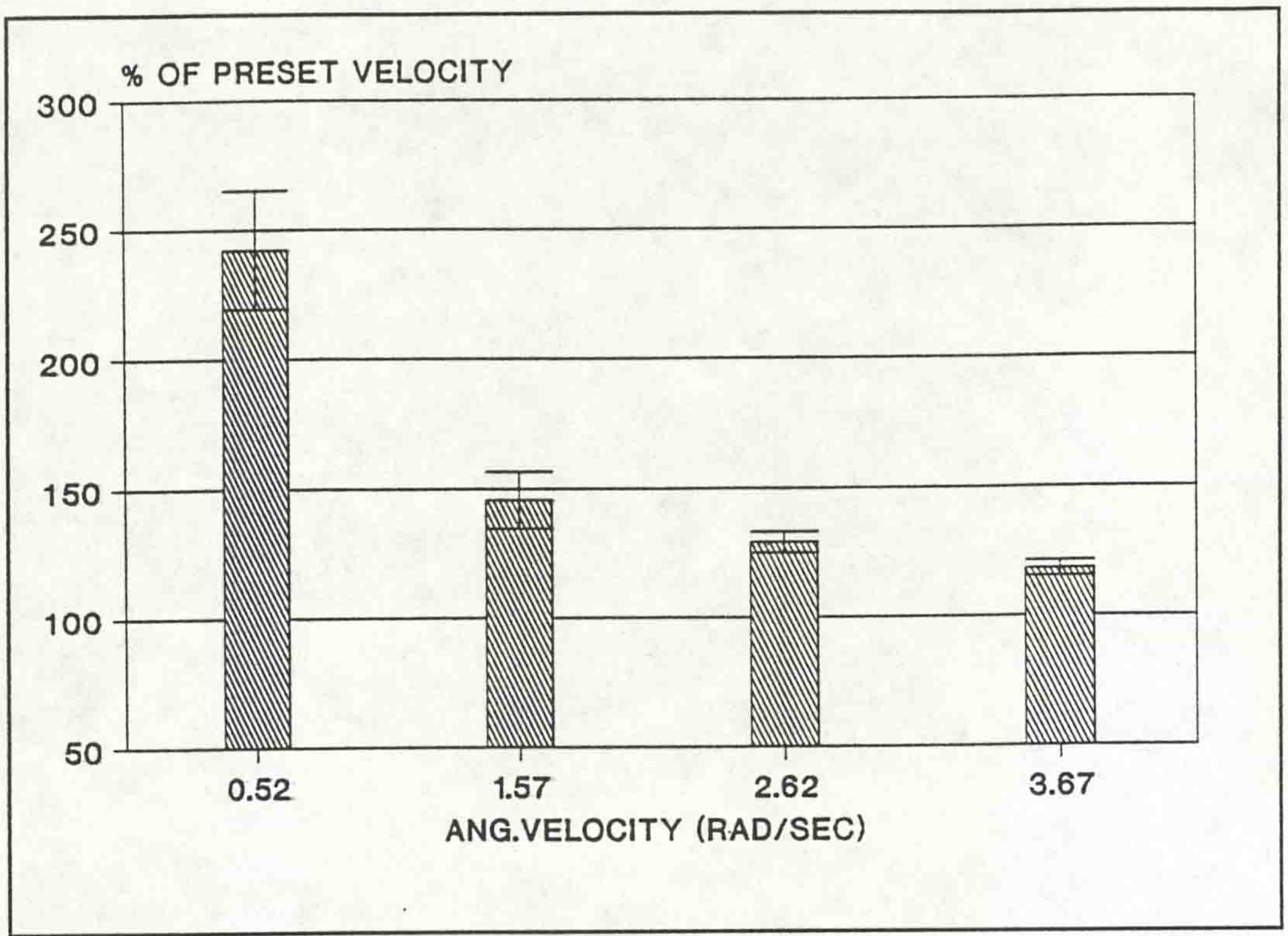

Figure 4.3. Maximum angular velocity relative to preset velocity during the initial acceleration period.

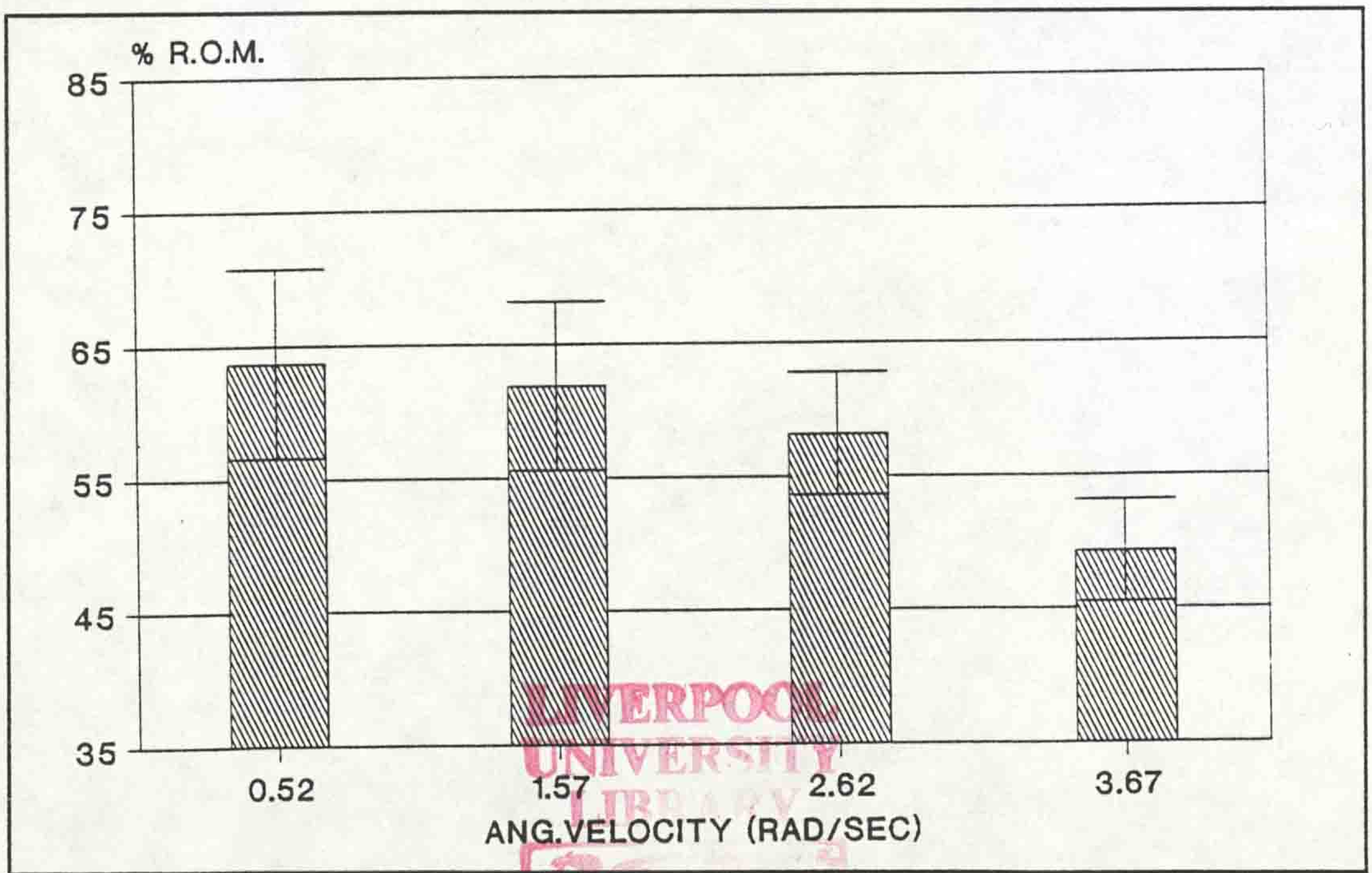

Figure 4.4. The part of the movement with the angular velocity ± 10% of the preset velocity expressed as a percentage of the total range of movement.

DISCUSSION

Angular velocity development.

The results of this study indicate that the constant velocity assumption during isokinetic dynamometry is not valid throughout the ROM and appropriate correction methods are required before the assessment of muscle function. Limitations in the operation of the resistive mechanism of the dynamometer

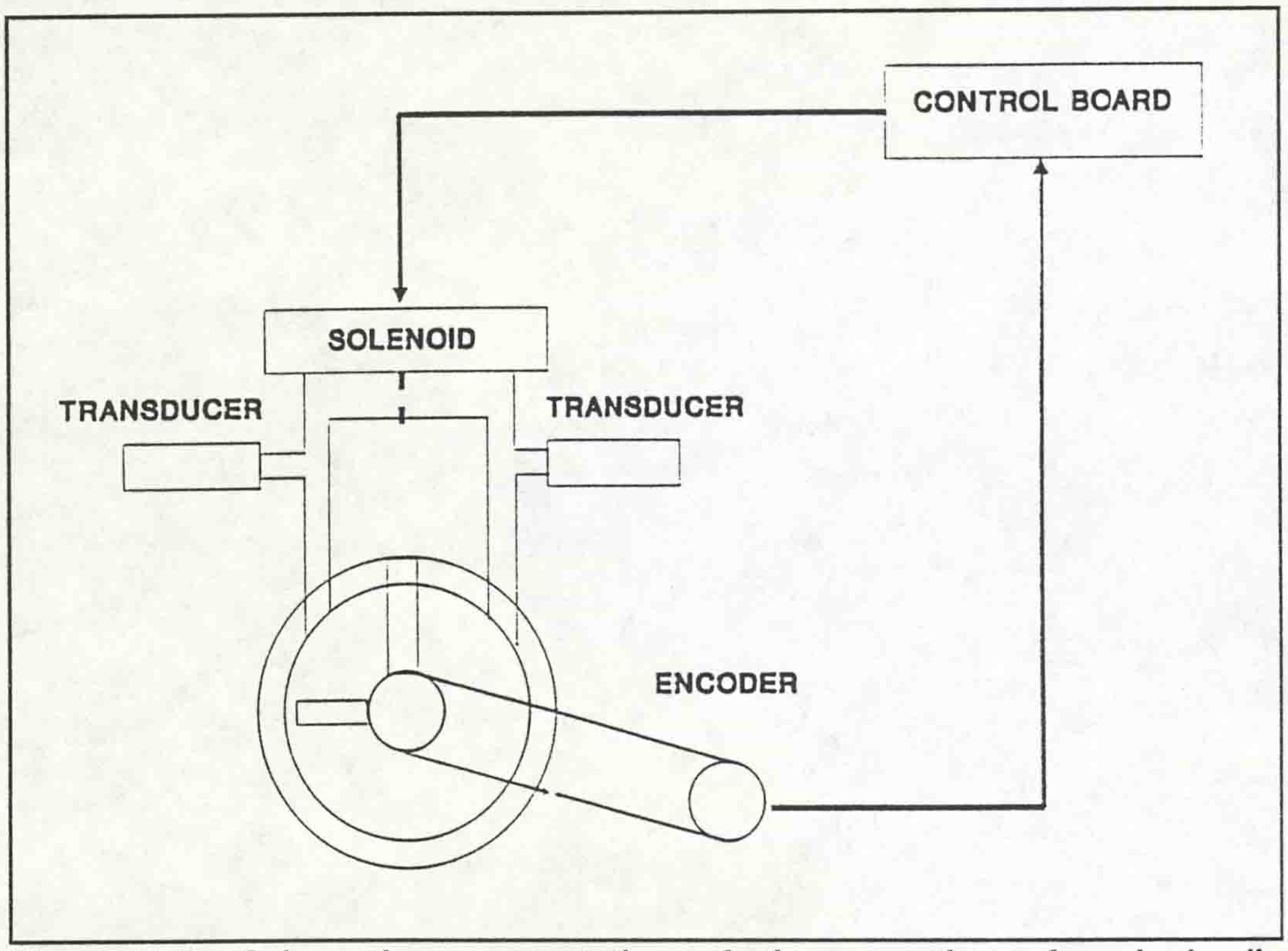

Figure 4.5. Schematic representation of the operation of a hydraulic dynamometer.

generate the inertial artifacts in the initial period of the movement. Fig. 4.5 illustrates the operation of a hydraulic dynamometer. The liquid flow through the solenoid valve - and therefore the resistive force - is adjusted according to the level of the angular velocity of the system relative to the preset angular velocity. Electro-mechanical dynamometers operate on a similar principle using electric motors instead of hydraulic components. An optical encoder monitors the angular velocity of the system and the resistive mechanism is activated only after the velocity of the system attains the level of the preset velocity (Fig. 4.6). During this initial period, the system is accelerated by the activated

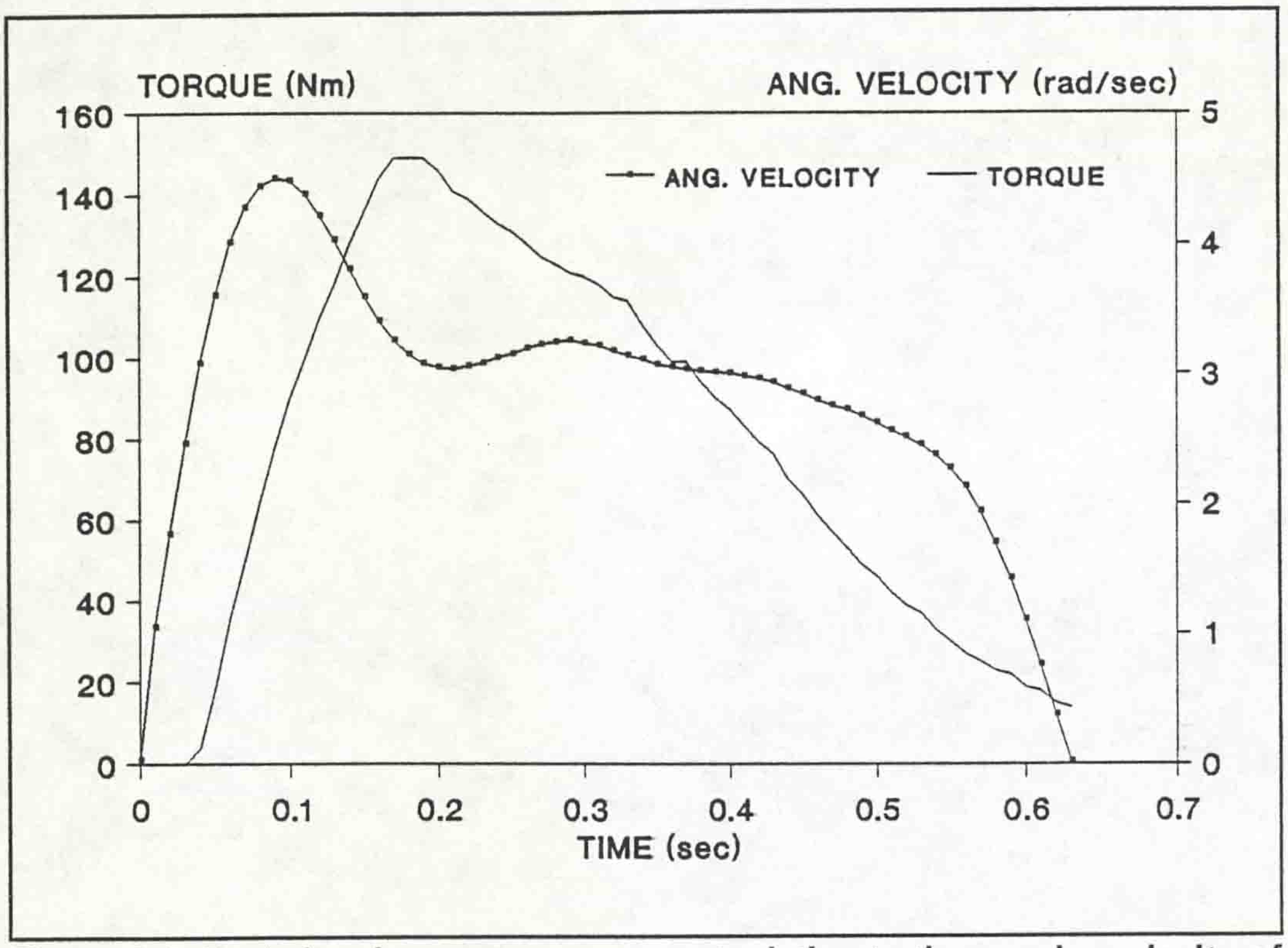

Figure 4.6. Resistive dynamometer moment relative to the angular velocity of the limb during a knee extension at a preset angular velocity of 2.62 rad·sec^{-1}.

muscle group since there is no resistive force applied. The duration of this period and the magnitude of the resistive moment required to decelerate the limb to the level of the preset velocity, depends on the muscular capabilities of the subject and the preset angular velocity. Furthermore the response of the system to muscle generated angular velocity fluctuations is critical for the maintenance of angular velocity at an approximately constant level throughout the ROM.

Filtering Requirements

In order to overcome measurement errors resulting from the inertial "overshoot", accelerometry techniques (Winter *et al.*, 1981) and analog electrical filters have been used for moment signal smoothing (Sapega *et al.*, 1982; Sinacore *et al.*, 1983). Different cutoff frequencies ("damping settings") have been suggested depending on the angular velocity of movement and the testing conditions (Bemben *et al.*, 1988). The implementation of such electrical filters however introduces a phase shift in the moment-time signal and affects its amplitude throughout the range of movement (Sinacore *et al.*, 1983). Despite these limitations, several isokinetic parameters (e.g. initial moment development and time to peak joint moment) for the analysis of muscular moment development during the initial part of the movement have been based on unfiltered or filtered resistive moment data. The inadequacy of the above method is evident since the resistive moment output during the initial acceleration period (whether filtered or unfiltered) represents the resistive moment developed by the dynamometer and not the actual muscular moment accelerating the system. Furthermore this resistive moment output of the dynamometer is delayed until the preset velocity is attained by the moving limb (Fig. 4.6). Consequently any conclusions about the mechanical properties of the muscle based on uncorrected dynamometer moment output during the initial acceleration period should be reexamined.

The resultant joint moment during this period should only be approximated from moment of inertia and angular acceleration data derived from either direct acceleration measurements (Winter *et al.*, 1981) or differentiation of the displacement-time data. The noise contained in the unfiltered data however requires appropriate smoothing procedures before differentiation (Fig. 4.7).

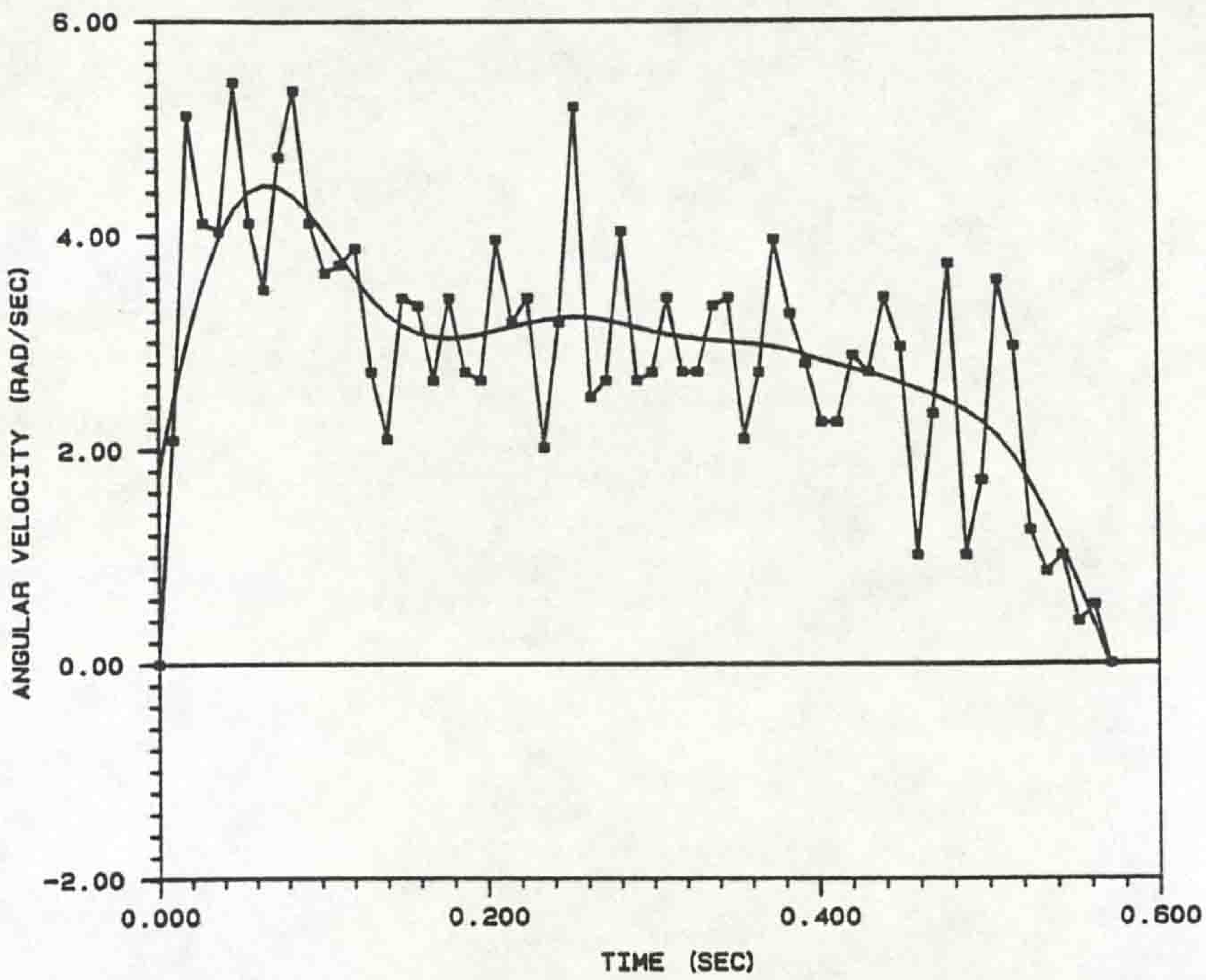

Figure 4.7. Measurement of limb angular velocity during isokinetic knee extension by differentiation of the position-time data without prior filtering. The noise in the data results in meaningless angular velocity measurements.

The applications and limitations of several methods for data smoothing and differentiation in biomechanics (including digital filters, fourier series and spline functions) have been extensively examined (e.g. Wood, 1982), although the applications are mainly in the areas of film and video analysis. Murray (1986) implemented spectral analysis of the joint moment and angular position data during isokinetic inertial loading, followed by low-pass digital filtering. The optimal cutoff frequency was a positive function of the preset angular velocity. Osternig *et al.* (1983) used cubic splines for the measurement of angular velocity from position-time data during muscular loading. It was reported that the constant velocity period ranged from 92% at 0.87 rad·s^{-1} to 76% of the

ROM at 6.98 rad·s^{-1} during a knee extension-flexion movement. The smoothing factor, the method used to determine it and the criteria to determine constant velocity, however, were not reported. The constant velocity periods in the present study (Fig. 4.4) differ considerably from those reported by Osternig *et al.* (1983). These differences may be attributed to the filtering methods used, the operation of the different types of isokinetic dynamometers (electromechanical-hydraulic) and the method implemented to determine constant angular velocity.

An important factor for the choice of an appropriate filtering method is computation time. In a clinical environment time efficient analysis of the isokinetic data and immediate feedback after the test are important. A typical examination of five consecutive repetitions (e.g. knee extension-flexion) at 0.52 rad·s^{-1} with a 1 rad ROM and a sampling frequency of 100 Hz, requires the analysis of approximately 2000 data points. The computation of angular velocity using iterative methods (e.g. spline functions) is not time efficient and therefore not appropriate for the analysis of such extensive data sets. The digital filter implemented in the present study is a computationally efficient method (Vaughan, 1982) allowing kinematic data analysis and performance feedback immediately after the test.

Isokinetic Parameters

The maximum resultant joint moment during knee extension at different preset angular velocities is in agreement with previous studies using gravity corrected data. (Prietto and Caiozzo, 1989). The decrease in moment output with increasing angular velocity (joint moment-velocity relationship) has been mainly attributed to different neurological activation patterns of motor units during different contractile velocities (Barnes, 1980).

Angular velocity fluctuations were present throughout the ROM and therefore the identification of the "isokinetic" part of the movement is not straightforward (Fig. 4.8-4.11).

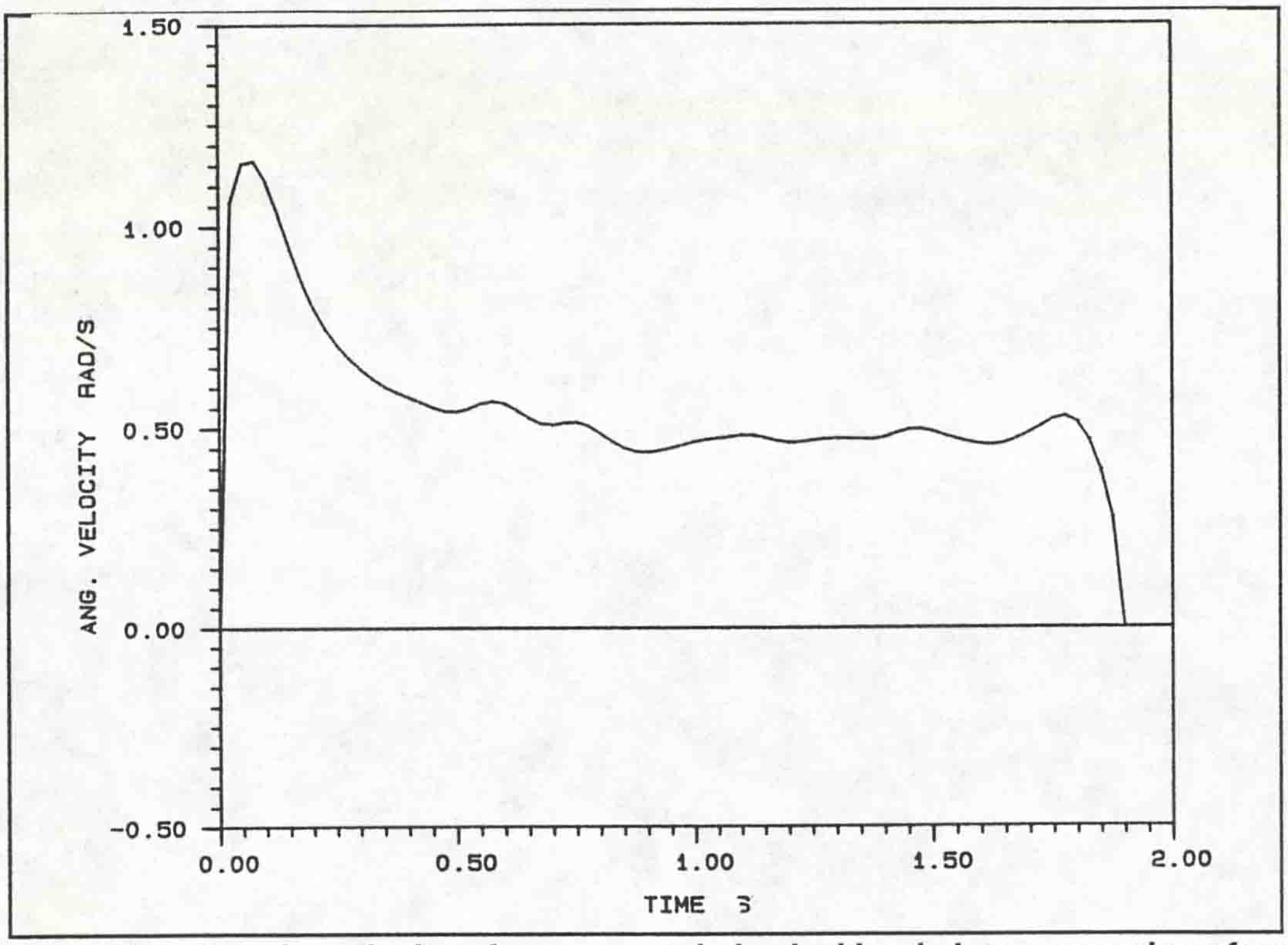

Figure 4.8. Angular velocity of movement during isokinetic knee extension after optimal filtering at a preset angular velocity of 0.52 rad·s^{-1}.

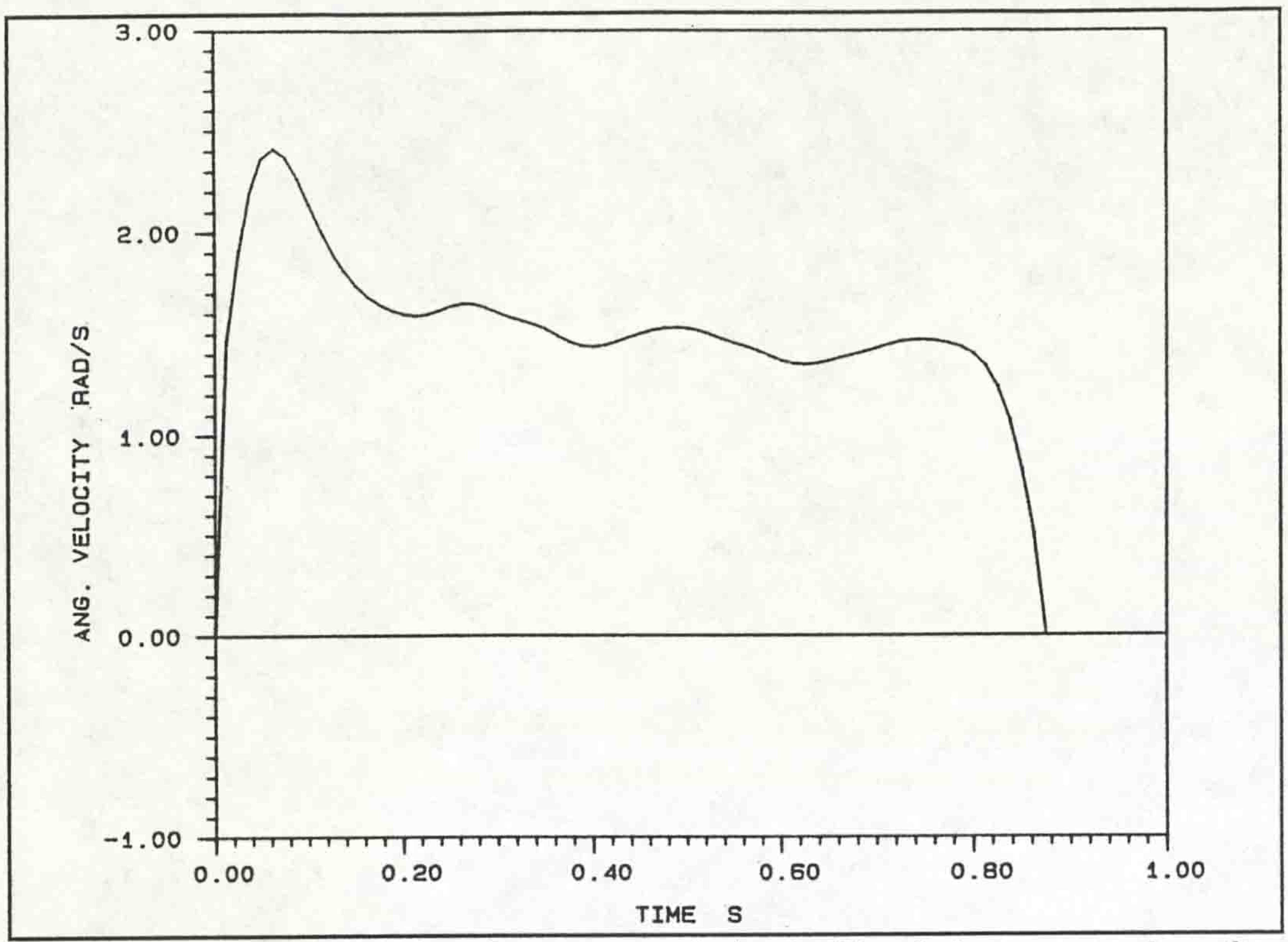

Figure 4.9. Angular velocity of movement during isokinetic knee extension after optimal filtering at a preset angular velocity of 1.57 rad·s^{-1}.

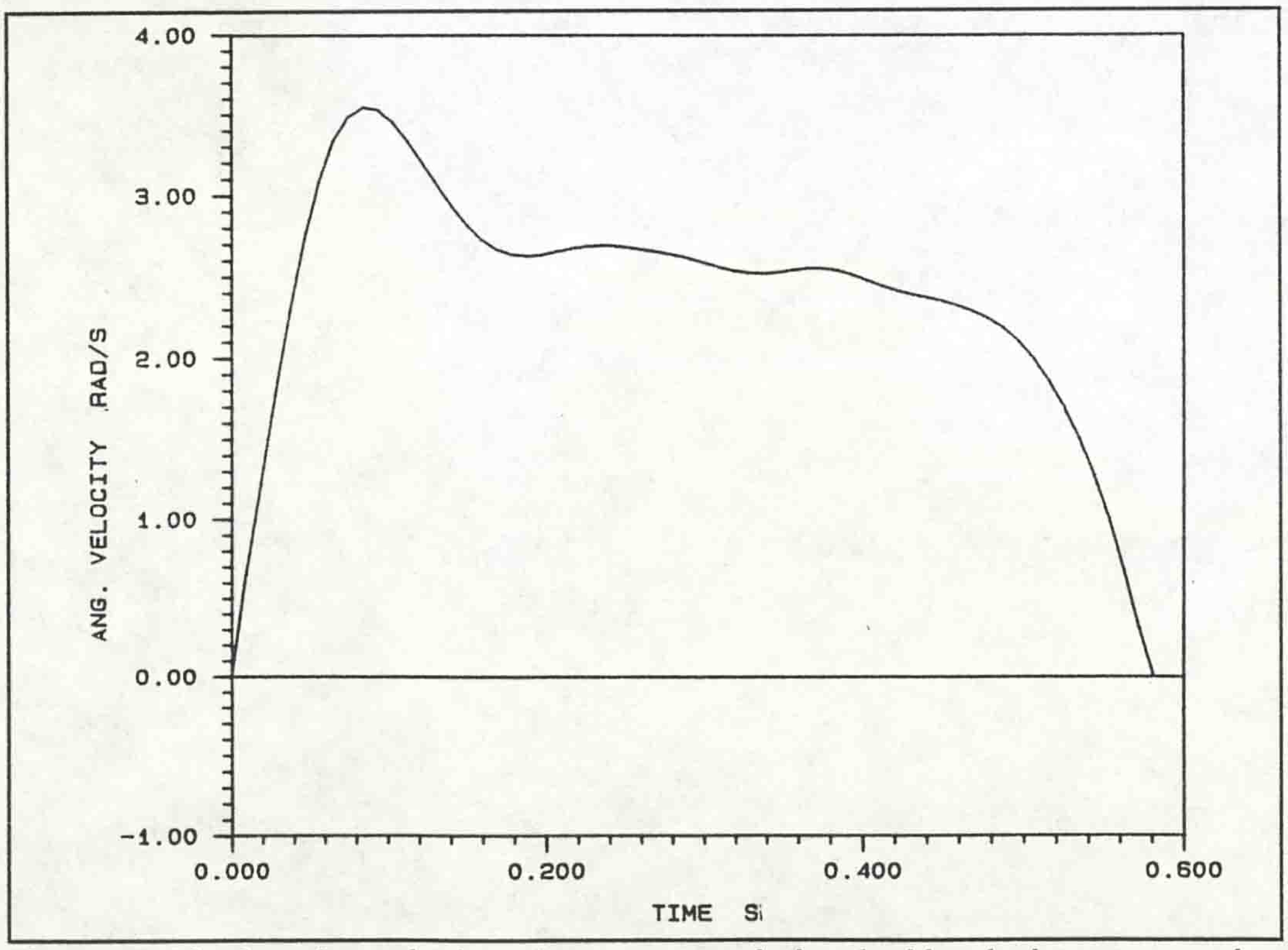

Figure 4.10. Angular velocity of movement during isokinetic knee extension after optimal filtering at a preset angular velocity of 2.62 rad·s^{-1}.

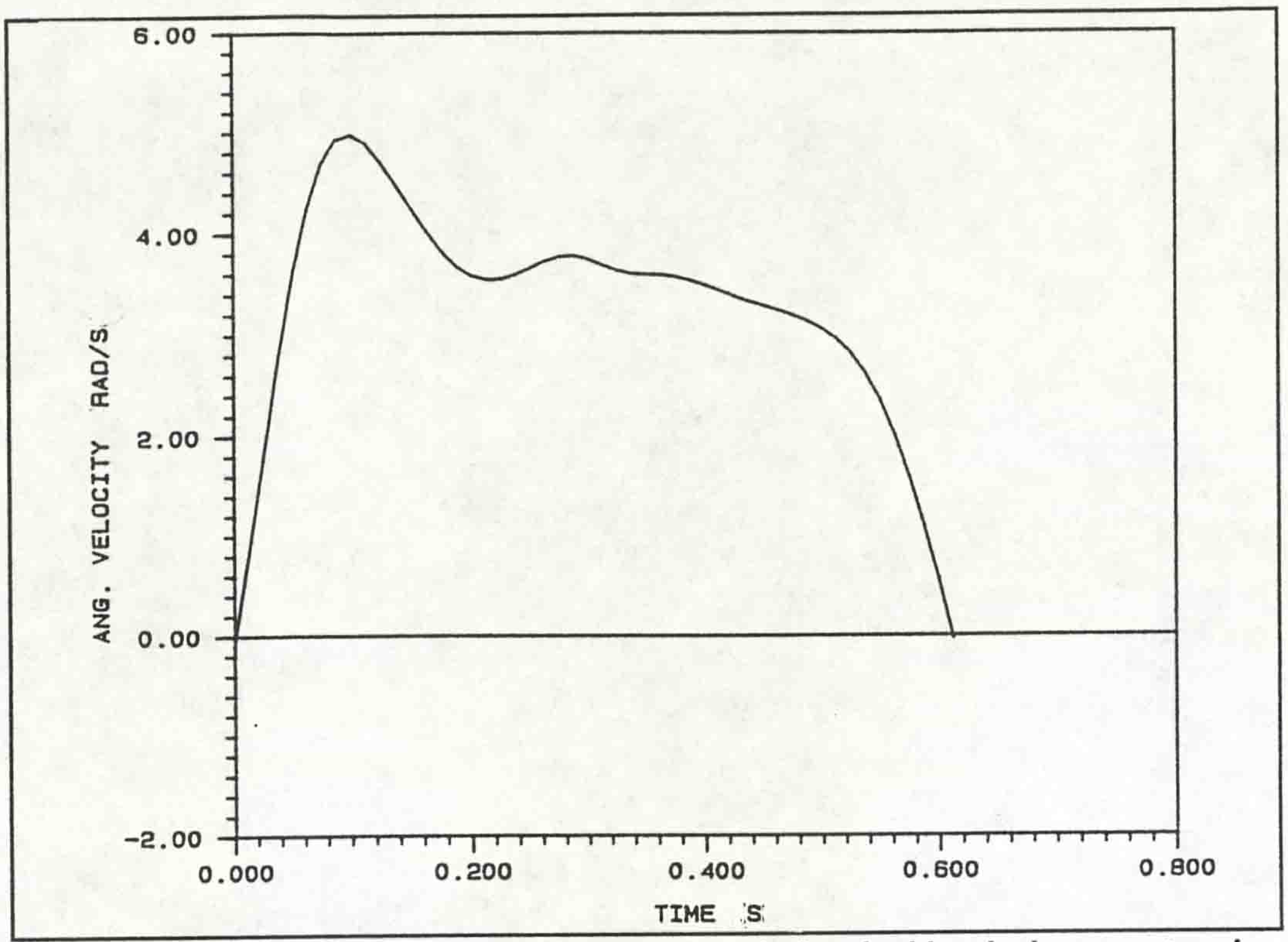

Figure 4.11. Angular velocity of movement during isokinetic knee extension after optimal filtering at a preset angular velocity of 3.66 rad·s^{-1}.

A range of ±10% of the preset angular velocity was considered an appropriate limit for the fluctuation of the angular velocity during muscular loading although for inertial loading this limit must be reduced. Furthermore, different applications (e.g. rehabilitation testing) may require a different range. This limit also depends on the type of the dynamometer, the operation of the resistive mechanism and the muscular ability of the subject. Angular velocity monitoring is also important because subjects with muscular and ligamentous injuries may not be able to perform the movement at the level of the preset velocity. Furthermore angular velocity development and fluctuations during the movement can be used for the assessment of muscle function during rehabilitation of muscle and joint injuries.

CONCLUSIONS

The conclusions within the limitations of the present study are:

1. The constant velocity assumptions during isokinetic muscular loading are not valid even after the initial acceleration period. The measurement of angular velocity therefore is essential before the assessment of muscle function.

2. Measurement of kinematic parameters from dynamometer data requires the use of appropriate smoothing methods.

3. Digital filters are appropriate for isokinetic data smoothing. The sampling and optimal cutoff frequencies are positive functions of the angular velocity.

4. Conclusions about muscle function based on the uncorrected joint moment output of the dynamometer during the initial acceleration period must be reexamined.

5. In order to allow valid comparisons and interpretations of published data appropriate gravitational and inertial correction methods must be implemented before the assessment of muscle function.

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CHAPTER 5

MEASUREMENT OF JOINT KINEMATICS USING VIDEOFLUOROSCOPY

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ABSTRACT.

Image distortion in video and image intensifier X-ray systems requires appropriate non-linear distortion correction methods, in order to obtain accurate biomechanical quantitative measurements. This paper presents an algorithm for two-dimensional coordinate reconstruction and non-linear distortion correction using a modified polynomial method. The measurement error obtained using an image intensifier - video system was 0.246 ± 0.111mm over a 180mm x 180mm field of view. Applications of the method include motion analysis using video and X-ray fluoroscopy systems with non-linear image distortion problems.

NOMENCLATURE

x_p, y_p	video reference system coordinates of object plane points
X_p, Y_p	object plane coordinates computed from x_p, y_p
X', Y'	global known coordinates of calibration point
x, y	video reference system coordinates of calibration points
n	maximum number of calibration points
k	number of proximal calibration points ($k \leq n$)
m	number of control points for measurement error determination
a, b	polynomial coefficients for computation of X_p, Y_p from x_p, y_p.
X	design matrix of the least-squares problem
N, M	maximum size of matrix **X** (N×M)
X'	vector of X' coordinate of calibration points
U	column orthogonal matrix for singular value decomposition of **X**
V	orthogonal matrix for singular value decomposition of **X**
$\boldsymbol{\sigma}$	diagonal matrix of singular values
w	weight function of calibration point
d	distance between two points in the video reference system
e	error of two-dimensional reconstruction after image distortion correction

INTRODUCTION

Roentgen stereophotogrammetric analysis (RSA) allows accurate three-dimensional skeletal measurements (Selvik, 1989) and has widespread applications in the areas of joint kinematics, joint stability, skeletal fractures, prosthetic implant fixation-loosening and skeletal growth (DeLange *et al.*, 1990; Kärrholm, 1989; Meijer *et al.*, 1989; Huiskes *et al.*, 1985; Lippert *et al.*, 1982). RSA is based on conventional static X-ray filming with exposure to relatively large doses of radiation. Furthermore, the examination of dynamic conditions in biomechanics and orthopaedics requires multiple static exposures at different joint angles, increasing the radiation dosage and limiting the applications in pathological conditions, for ethical reasons.

On the contrary, videofluoroscopy, using image intensifier (II) video systems, allows the acquisition of X-ray images of movements, recorded on video or film for further processing, reducing significantly radiation exposure (Kärrholm 1989). The application of II-video systems in biomechanical, quantitative research is limited however, despite their widespread use in clinical examination and diagnosis (Bell 1990). The main limitation of II systems is the optical distortion of the X-ray image. Appropriate correction methods are essential for accurate, quantitative analysis of the recorded movement.

Image distortion occurs during the different stages of the X-ray process using II systems (Fig. 5.1). The distance between object plane and II screen leads to perspective error. This type of error is minimised by placing the limb close to the II screen. Correction for perspective error is relatively straightforward using simple geometrical methods once the distances between X-ray source, object plane and II screen are known (Büchi *et al.*, 1990). The X-

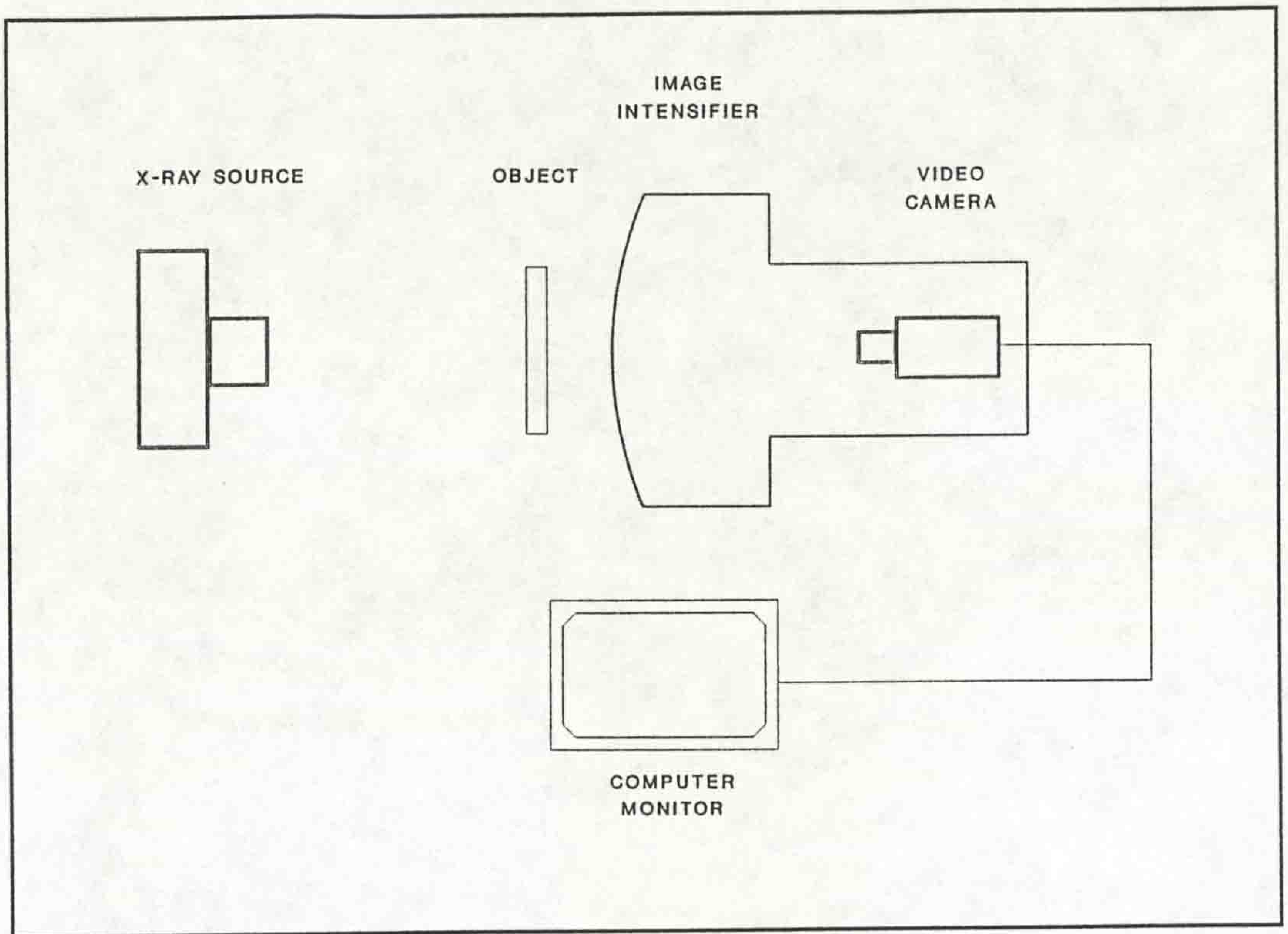

Figure. 5.1. The components of an image intensifier-video system.

ray projection of the limb on the II phosphor surface is further distorted because of the curvature of the screen (Fig. 5.1). The distortion is maximal at the periphery and minimal at the centre of the II screen (pin-cushion distortion). The television system used for the analysis of the video images may further introduce a combination of barrel, trapezoidal and non-linear distortion (Wallace and Johnson, 1981). Accurate analysis of the X-ray image therefore requires the use of appropriate distortion correction methods.

Wallace and Johnson (1981) developed a correction method based on a geometrical model. The position of any point is identified relative to four calibration points forming a square and the corrected position is computed from the distortion in the position of the calibration points using a simple geometrical model. Chakraborty (1987) separated II distortion into view dependent (VD) and view independent (VI) components. VD distortion resulting from the curvature of the II screen, is corrected using a geometrical model. With this method, the curvature of the II screen, if not available from manufacturer's data, is approximated using a simple calibration method. VI distortion resulting from the digital transformation of the image, is corrected using a linear least-squares procedure. Measurement error however was not reported in the above studies. The main limitation of the above methods is that the linear correction models used are not adequate for correction of non-linear distortion in II-video systems.

The purpose of this study was to develop a simple and efficient microcomputer-based method for II distortion correction in order to allow accurate, two dimensional, quantitative analysis of X-ray video records. Special emphasis was given to the implementation of the method using a microcomputer system and the simplification of the calibration procedure, in order to facilitate application of this method in a clinical environment by operators not familiar with the mathematical principles of analytic photogrammetry.

METHODS

Instrumentation

A SIEMENS PANTOSKOP/EXPLORATOR X-ray unit with a SIRECON television unit was used in the present study. The X-ray image was recorded on a Sirecord video cassette recorder. Analysis of the video tapes was performed using a SONY U-matic system connected to an IBM PS/2 (30/286) microcomputer (Fig. 5.2). The resolution of the graphics adapter of this system is 640 $\times$ 480 picture elements (pixels). Video X-ray records were displayed on

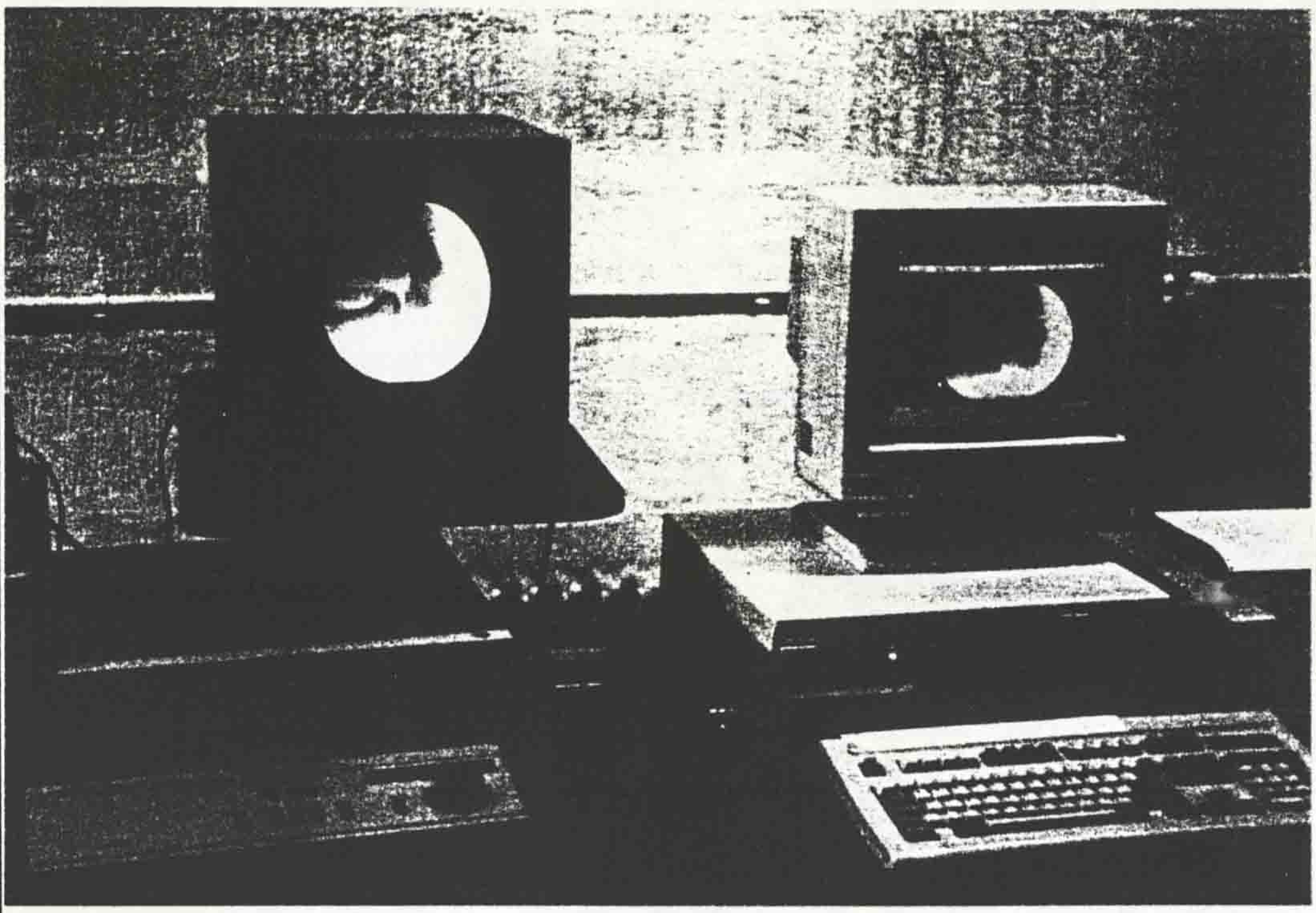

Figure 5.2. Video and microcomputer system for videofluoroscopy analysis

the computer monitor and were manually digitised using the computer's graphics cursor.

Mathematical model.

The global coordinates X_P, Y_P of any object plane point, are computed from the corresponding video reference system coordinates x_P, y_P using the polynomial functions:

$$X_P = a_1 + a_2 x_P + a_3 y_P + a_4 {x_P}^2 + a_5 x_P y_P + a_6 {y_P}^2 \quad (5.1)$$

$$Y_P = b_1 + b_2 x_P + b_3 y_P + b_4 {x_P}^2 + b_5 x_P y_P + b_6 {y_P}^2 \quad (5.2)$$

In order to determine the coefficients $a_1,\ldots,a_6$, n calibration points ($n \geq 6$) on the object plane are required with known $(X'_1, Y'_1, \ldots, X'_n, Y'_n)$ global coordinates and corresponding $(x_1, y_1, \ldots, x_n, y_n)$ video system coordinates. This method has previously been used for motion analysis using optoelectronic systems (Andriacchi *et al.*, 1979).

Assuming X_i is an approximation of the X'_i coordinate, then the sum of squares of the deviations of the n calibration points is the X coordinate error function

$$E(X) = \sum_{i=1}^{n} (X_i - X'_i)^2 \tag{5.3}$$

or

$$E(X) = \sum_{i=1}^{n} (a_1 + a_2x_i + a_3y_i + a_4x^2_i + a_5x_iy_i + a_6y^2_i - X'_i)^2$$

These equations can be expressed in matrix form as $|\mathbf{X}\cdot\mathbf{a}-\mathbf{X'}|^2$, where

$$\mathbf{X} = \begin{bmatrix} x_1 & y_1 & x_1^2 & x_1y_1 & y_1^2 \\ x_2 & y_2 & x_2^2 & x_2y_2 & y_2^2 \\ \cdot & & & & \\ \cdot & & & & \\ \cdot & & & & \\ x_n & y_n & x_n^2 & x_ny_n & y_n^2 \end{bmatrix},$$

the design matrix of the problem, $\mathbf{a} = [a_1,...,a_6]^T$ and $\mathbf{X'} = [X'_1,...,X'_n]^T$.

The six coefficients are determined by minimising the error function E(X). There are several techniques for the determination of the coefficients that minimise this function. Andriacchi *et al.*, (1979) used the normal equations derived from the error function. E(X) is a function of $a_1,...,a_6$ and will therefore have a minimum when $\partial E(X)/\partial a_j = 0$, or

$$\partial \sum_{i=1}^{n} (a_1 + a_2x_i + a_3y_i + a_4x^2_i + a_5x_iy_i + a_6y^2_i - X'_i)^2 / \partial a_j = 0$$

for $j = 1...6$

The above derivative conditions provide six equations that are linear in the unknown coefficients $a_1,...,a_6$ (normal equations):

$$Na_1 + (\Sigma x_i)a_2 + (\Sigma y_i)a_3 + (\Sigma x_i^2)a_4 + (\Sigma x_i y_i)a_5 + (\Sigma y_i^2)a_6 = \Sigma X_i$$

$$(\Sigma x_i)a_1 + (\Sigma x_i^2)a_2 + (\Sigma x_i y_i)a_3 + (\Sigma x_i^3)a_4 + (\Sigma x_i^2 y_i)a_5 + (\Sigma x_i y_i^2)a_6 = \Sigma x_i X_i$$

$$(\Sigma y_i)a_1 + (\Sigma x_i y_i)a_2 + (\Sigma y_i^2)a_3 + (\Sigma x_i^2 y_i)a_4 + (\Sigma x_i y_i^2)a_5 + (\Sigma y_i^3)a_6 = \Sigma y_i X_i$$

$$(\Sigma x_i^2)a_1 + (\Sigma x_i^3)a_2 + (\Sigma x_i^2 y_i)a_3 + (\Sigma x_i 4)a_4 + (\Sigma x_i^3 y_i)a_5 + (\Sigma x_i^2 y_i^2)a_6 = \Sigma x_i^2 X_i$$

$$(\Sigma x_i y_i)a_1 + (\Sigma x_i^2 y_i)a_2 + (\Sigma x_i y_i^2)a_3 + (\Sigma x_i^3 y_i)a_4 + (\Sigma x_i^2 y_i^2)a_5 + (\Sigma x_i^2 y_i^3)a_6 = \Sigma x_i y_i X_i$$

$$(\Sigma y_i^2)a_1 + (\Sigma x_i y_i^2)a_2 + (\Sigma y_i^3)a_3 + (\Sigma x_i^2 y_i^2)a_4 + (\Sigma x_i^2 y_i^3)a_5 + (\Sigma y_i^4)a_6 = \Sigma y_i^2 X_i$$

and can be expressed in matrix form as $(\mathbf{X}^T \cdot \mathbf{X}) \cdot \mathbf{a} = \mathbf{X}^T \cdot \mathbf{X}'$

Solution of this system for $a_1,\ldots,a_6$ using standard methods of computational linear algebra (e.g. Gauss-Jordan elimination), determines the polynomial in (5.1) for the computation of the global X_P coordinate of any point on the calibration plane from the digitised video coordinates x_P, y_P.

However, the use of normal equations for the solution of least-squares problems is not always appropriate as these equations are often ill-conditioned (matrix is either singular or very close to singular). Singular value decomposition (SVD) is an appropriate technique for the solution of an overdetermined system with numerical stability problems (Wilkinson and Reinsch, 1971). In brief, instead of forming the normal equations, SVD decomposes the N × M design matrix **X** (N = n, M = 6 in the present model) into a product of an N × M column orthogonal matrix **U**, an M × M diagonal matrix $\boldsymbol{\sigma}$ with positive or zero elements and the transpose of an N × N orthogonal matrix **V** (Wilkinson and Reinsch, 1971):

$$\mathbf{X} = \mathbf{U} \cdot \boldsymbol{\sigma} \cdot \mathbf{V}^T$$

where

$$\mathbf{U}^T \cdot \mathbf{U} = \mathbf{V}^T \cdot \mathbf{V} = 1 \text{ and } \boldsymbol{\sigma} = \mathrm{diag}[\sigma_1,\ldots,\sigma_6]$$

The elements of matrix $\boldsymbol{\sigma}$ $(\sigma_1,\ldots,\sigma_6)$ are the non-negative square roots of the eigenvalues of $\mathbf{X}^T \cdot \mathbf{X}$ (singular values).

The solution of the above system is the coefficients vector **a** that minimise $|\mathbf{X}\cdot a-\mathbf{X}'|^2$:

$$a = V \cdot \sigma^{-1} \cdot U^T \cdot X'$$

The SVD algorithm in Wilkinson and Reinsch (1971) was implemented in the present study.

An important assumption of this polynomial method is that image distortion is uniform over the entire field of view (FOV). The coefficients **a** are therefore determined from calibration points distributed throughout the FOV. Although this assumption may be valid for film systems and central projections this is not the case with video systems and II screens.

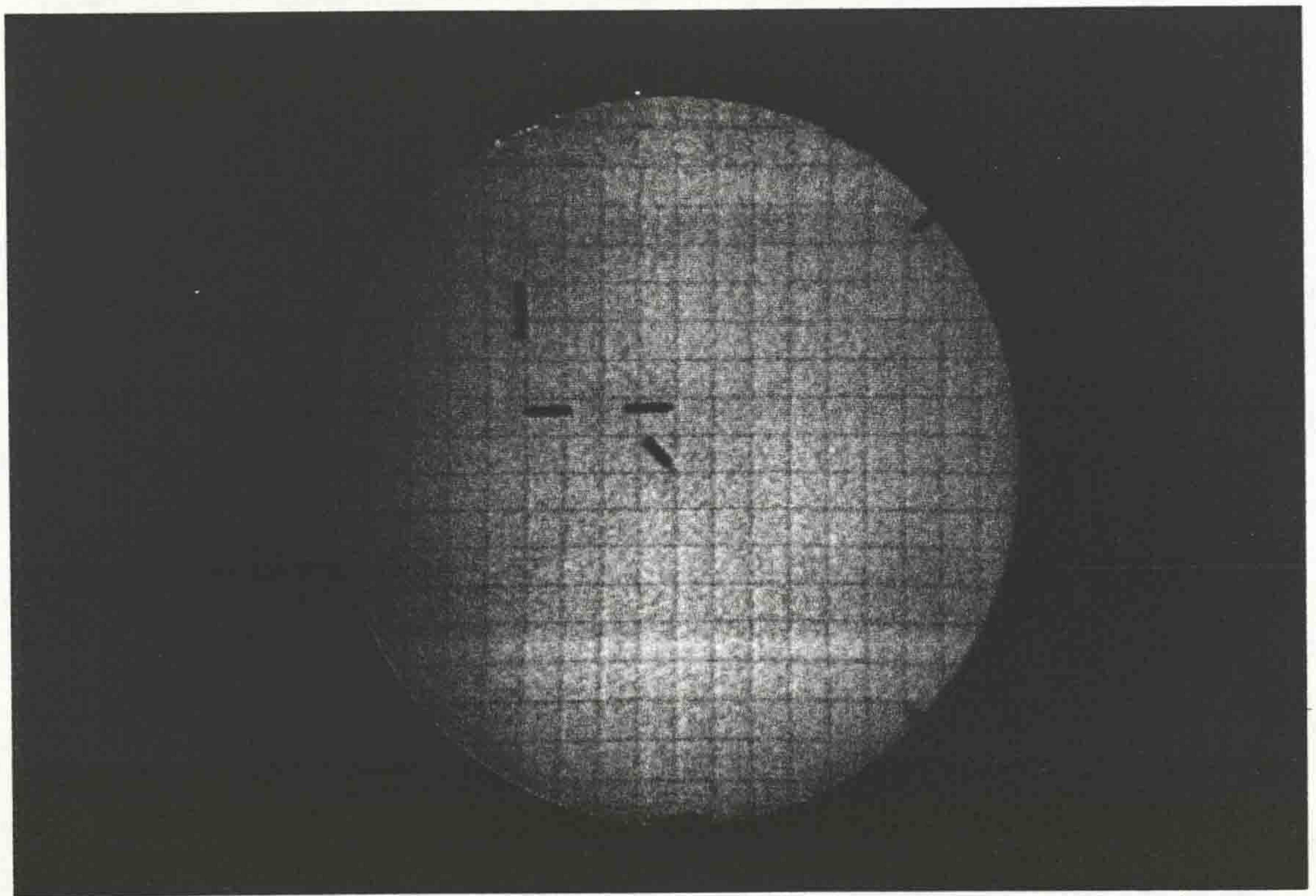

Figure 5.3. The calibration points displayed on the computer monitor. Distortion is minimal in the centre and maximal in the periphery of the screen.

Fig. 5.3 illustrates that image distortion is not uniform throughout the FOV. Reconstruction of the object plane coordinates in the centre of the screen, (where the distortion is minimal), from coefficients determined using calibration points in the periphery of the screen (where the distortion is maximal), will produce unacceptable errors. A modification of the above method is therefore essential for use in II-video systems, in order to determine **a** using only proximal calibration points from an area with uniform distortion. This was accomplished by introducing non-negative weight functions, $w_i(x_P,y_P)$, for each calibration point X_i,Y_i $(i=1,\dots,n)$ that depend on the relative distance between the video coordinates x_i,y_i of the calibration points and the video coordinates x_P,y_P of any point in the object plane (Lancaster and Šalkauskas, 1986). By introducing a distortion uniformity (DU) constant d, the weight functions were determined as

$$w_i(x,y)=\begin{cases}1, & \text{if } [(x_P-x_i)^2+(y_P-y_i)^2]^{1/2} \le d & (5.4)\\ 0, & \text{if } [(x_P-x_i)^2+(y_P-y_i)^2]^{1/2} > d & (5.5)\end{cases}$$

By introducing the weight functions for each calibration point, the error function E(X) in (5.3) becomes

$$E(X)=\sum_{i=1}^{n} w_i(X_i - X'_i)^2 \qquad (5.6)$$

The coefficients **a** are determined from k calibration points $(k \le n)$ that satisfy the condition in (5.4), so that image distortion within the area covered by the k calibration points is approximately uniform. Distortion uniformity depends mainly on II systems and X-ray angulation and therefore different applications require appropriate adjustment of the DU constant.

A similar procedure is followed in order to determine the coefficients $b_1,\dots,b_6$ for the computation of the Y coordinate.

Calibration

The calibration structure consisted of stainless steel wires (0.254mm in diameter) mounted on perspex glass of 5mm thickness and forming 10 mm squares. The co-planar calibration points were located on the intersection of

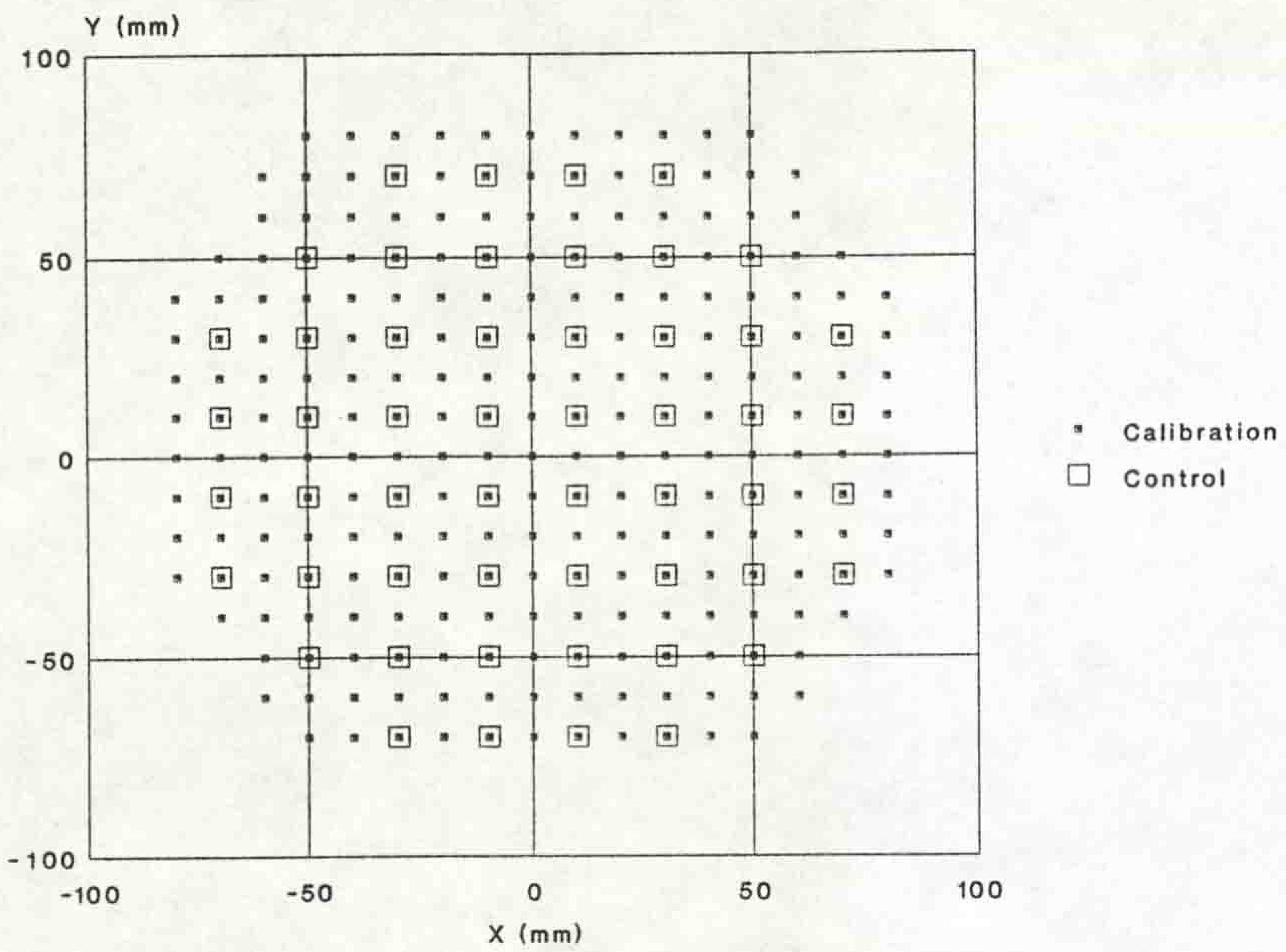

Figure 5.4. The position of the calibration and the control points.

the wires in the corners of the squares (Fig. 5.4). The calibration structure was placed perpendicular to the central X-ray beam to coincide with the object plane at a distance of 100 mm from the II screen (Fig. 5.1). A plastic container filled with water was placed in front of the calibration structure in order to simulate soft tissue radiation deflection. The angulation of the central X-ray beam was 0 rad. The (FOV) at the object plane was approximately 180 mm × 180 mm. The two dimensional coordinates of the calibration points were digitised and stored for the determination of the polynomial coefficients in (5.1) and (5.2) using the above method.

Error Analysis

Measurement error was defined as the root mean square error in reconstructing the two dimensional coordinates of m control points (m = 52) on the calibration structure distributed throughout the FOV (Fig. 5.4):

$$\bar{e} = [m^{-1} \cdot \sum_{i=1}^{m} (X_{Pi} - X'_{Pi})^2 + (Y_{Pi} - Y'_{Pi})^2]^{1/2} \tag{5.7}$$

where X'_{Pi}, Y'_{Pi} the known global object plane coordinates of the control points and X_{Pi}, Y_{Pi} the global coordinates computed using the polynomial models in equations (5.1) and (5.2). Measurement error was examined using a total number of 240 calibration points distributed throughout the FOV. In order to examine DU, each control point was reconstructed using three different sets of k calibration points ($k \leq n$) according to (5.4), covering an area equal to 20%, 40% and 60% of the FOV. Measurement error differences using different DU constants were examined using one-way ANOVA.

RESULTS

The measurement error with a total of 240 calibration points was 0.253 mm (± 0.127 mm), 0.246 mm (±0.111 mm) and 0.272 mm (±0.121 mm) using only proximal calibration points covering approximately 20%, 40% and 60% of the FOV respectively. These differences however were not statistically significant ($F_{2,102} = 2.43$ $p > 0.05$). Fig. 5.5 illustrates the distribution of the measurement error in the 52 control point locations in the FOV. The correlation

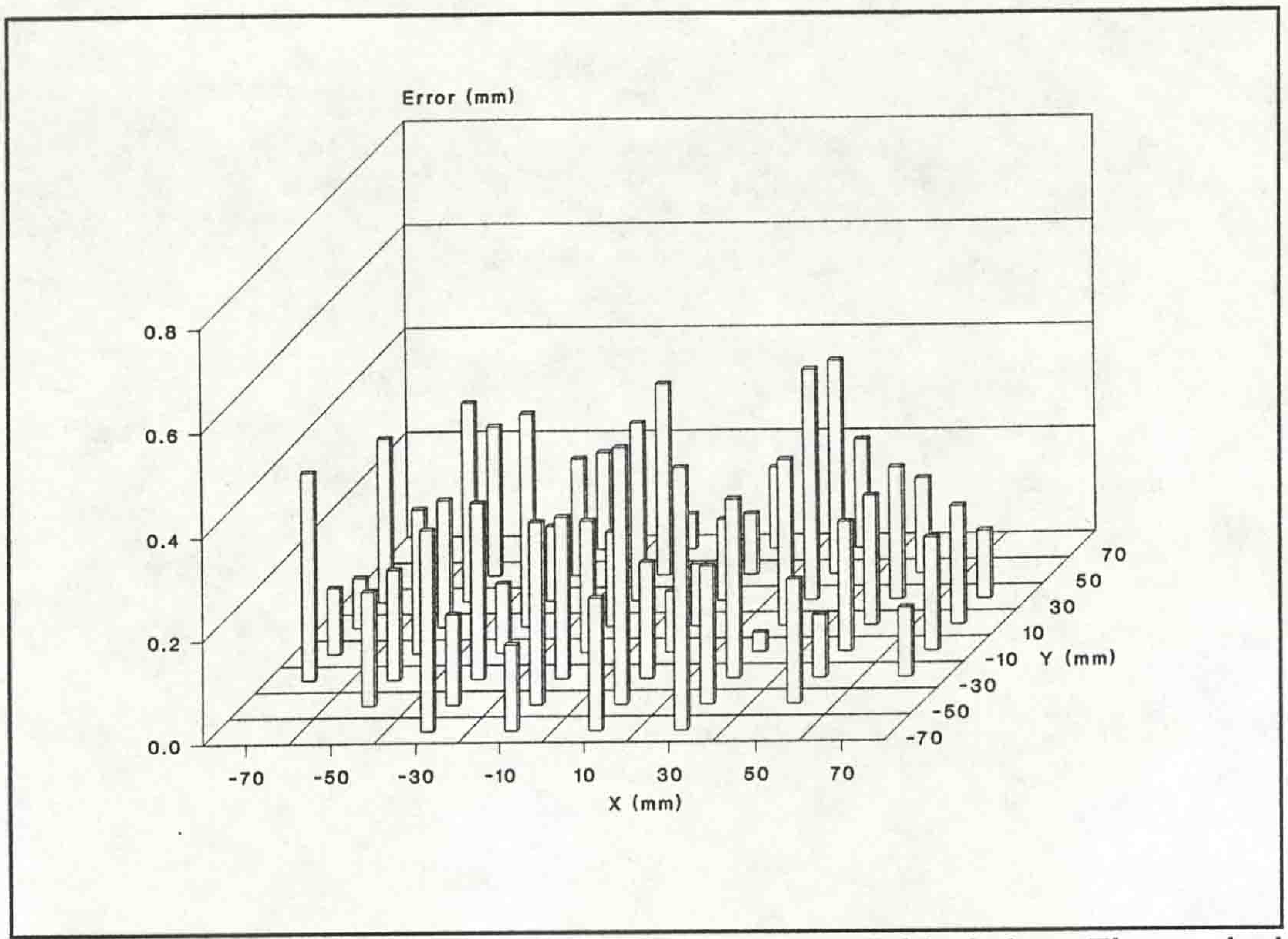

Figure 5.5. Distribution of measurement error in the field of view. The vertical bars represent the absolute error in the 52 control point locations.

coeeficients between measurement error and distance from the origin of the reference system were r=0.13, r=-0.02 and r=-0.08 for the three DU constants respectively. These results indicate that measurement error is independent of the distance of the control points from the centre of the screen.

DISCUSSION

Roentgen stereophotogrammetry is a very accurate method for X-ray analysis (Kärrholm, 1989). Huiskes *et al.* (1985) in a study of three-dimensional geometry of articular surfaces using RSA, reported measurement error ranging from 42 μm to 492 μm. This was approximately 0.01%-0.1% of the FOV respectively. Selvik (1989) also reported measurement error of 0.08% (98 μm). The main factor for the reduction in measurement error using RSA is the accuracy of the digitising process. Huiskes *et al.* (1985) for example used different types of digitisers with accuracy ranging from 20 μm to 100 μm. The resolution of the digitising system therefore was approximately 1/3400-1/17000. The X-ray film is also magnified by approximately 15 times, using special video cameras (Selvik, 1989; Kärrholm, 1989). The combination of digitising accuracy, image quality and film magnification are the main factors for the superior accuracy of RSA systems compared with videofluoroscopy. Digitising accuracy in II-video systems is limited by the resolution of the video system which is approximately 1/500.

Measurement error in previous studies of II distortion correction was not reported. (Wallace and Johnson, 1981; Chakraborty, 1987). Cholewicki *et al.* (1991) in a study of vertebral kinematics using videofluoroscopy reported measurement error of 0.33 mm (0.25% of the FOV). The distortion correction method used was modified from Wallace and Johnson (1981). The minimum measurement error in the present study was 0.246 $\pm$0.11 mm or 0.13% with a maximum of 240 calibration points. The calibration method in the present study is time efficient and easily implemented using a microcomputer. Furthermore, no additional information concerning the curvature of the II screen or the relative position of the II system components (X-ray source, object plane, II screen) is required for the correction of image distortion. Although the present algorithm was developed mainly for X-ray analysis using II-video systems, it is also applicable without modification in movement analysis using film, video or optoelectronic systems (Andriacchi *et al.*, 1979). The measurement error is better than that reported in other studies of movement analysis using film and video systems (e.g. Kennedy *et al.*, 1989).

The most widely used method for coordinate reconstruction in these studies is the direct linear transformation (DLT) (Abdel-Aziz and Karara, 1971). Measurement error using the DLT method depends mainly on the number and distribution of the calibration points. Hatze (1988) reported measurement error of approximately 0.24% using 30 calibration points, Wood and Marshall (1986) 0.23% using 30 points, Stokes (1984) 0.27% using 10 points and Dapena *et al.* (1982) 0.5% using a modified DLT method with 15 points. Andriacchi *et al.* (1979) reported measurement error ranging from 0.25%-0.31% using a similar polynomial method with 10-29 calibration points. Kennedy *et al.* (1989) compared DLT reconstructions using video and film systems and reported errors of 0.29% and 0.24% respectively. It is evident that measurement error is significantly reduced using the modified polynomial method presented in this study by increasing the number of calibration points.

Another problem with the use of the DLT method is that non-linear distortion usually present in video systems is not corrected using the basic DLT equations. Although there are DLT models for non-linear distortion correction (Abdel-Aziz and Karara, 1971), they are not widely used in biomechanical analysis.

Image distortion was minimal in the present study because the X-ray projection was central. In medical imaging applications however, requiring X-ray angulation (e.g. Lippert *et al.*, 1982), the image distortion is further increased (Chakraborty 1987) and therefore distortion correction methods are essential for quantitative analysis.

Measurement error is also affected by image quality. The sharpness of the image is different as the curvature of the II screen alters the focusing distance. Moreover, the persistence of the phosphor screen produces blurring during rapid movements, although image quality is acceptable during slow movements (Cholewicki *et al.*, 1991). Manual digitising, operator experience and video resolution constraints also affect measurement error. Recent image processing and digital enhancement techniques can improve image quality and facilitate automation of digitising process and reduction of measurement error (Breen *et al.*, 1989).

The minimum number of calibration points required depends on the number of unknown coefficients in the polynomial models (1) and (2). Measurement error is reduced significantly using more calibration points (Andriacchi *et al.* 1979), but computation time is increased.

The modification of the method in order to avoid non-uniform distortion errors also increases computation time. The coordinates of each digitised point are reconstructed from a different set of coefficients **a** and proximal calibration points. This requires solution of the overdetermined system of equations in (3) and therefore execution of the SVD algorithm twice (X,Y) for each digitised point.

For two dimensional analysis this method assumes that the object and calibration planes coincide. If the examined limb is moving during the X-ray process then it must be ensured that this movement is taking place on the calibration plane or otherwise variable perspective error will be introduced in the recorded image. The calibration points must cover the entire FOV since this method is useful for interpolation only. Any digitised points outside the calibration points area will result in erroneous reconstructed coordinates.

Two dimensional analysis of movement using video systems requires the placement of the video camera perpendicular to the plane of movement. The digitised video coordinates are transformed to global coordinates using simple ratio methods without considering the non linear distortion present in the image. Accurate reconstruction procedures such as DLT, are mainly used in three-dimensional analysis and require the construction of an accurate three dimensional calibration structure. The distortion correction and calibration procedure for two-dimensional analysis described in the present study requires the filming of a simple two dimensional structure and allows the placement of the video camera at any position relative to the plane of movement.

The calibration procedure for two dimensional analysis using II-video systems is also simple and requires only the digitisation of the X-ray image of the calibration structure placed in the object plane. Once the calibration points are digitised (in the video reference system) and the polynomial coefficients in equations (5.1) and (5.2) are determined, the distorted position of any digitised point is corrected and transformed in the global coordinate system.

CONCLUSIONS

A simple and efficient system for image distortion correction has been described. The applications of the system include two dimensional movement analysis using video systems and medical imaging using II systems, where accurate spatial information and quantitative analysis of the image is required. This is particularly important when X-ray angulation and therefore image distortion is increased.

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CHAPTER 6

BIOMECHANICAL ANALYSIS OF KNEE EXTENSION USING VIDEOFLUOROSCOPY

V. Baltzopoulos

ABSTRACT

The purpose of this study was the measurement of patellar tendon (PT) moment arm, tibial plateau-tibial axis angle and PT tendon-tibial axis angle during knee extension using videofluoroscopy *in vivo*. These parameters allow the determination of a two dimensional biomechanical model of the knee for the measurement of muscle and joint forces during isokinetic knee extension. Five males (mean age 20.8 ±years, mass 79.2 ±7.2 kg and height 179 ± 3.2 cm) without knee joint injury history participated in the study. The mean PT moment arm at full extension was 33.81 ±3.44 mm, increased to a maximum of 39.87 ±2.4 mm at 0.785 rad of knee flexion and decreased to 33.63 ±4.01 mm at 1.57 rad. The PT-tibial plateau angle was 1.97 ±0.12 rad at full extension and decreased linearly to 1.53 ±0.05 rad at 1.57 rad of knee flexion. The mean angle between the tibial plateau and the tibial long axis was 1.48 ±0.04 rad.

INTRODUCTION

The knee is the largest and the most complex joint in the human body and its normal function is essential for general mobility and participation in sporting activities. Movement is taking place predominantly in the sagittal plane (extension-flexion) with a range of movement (ROM) of approximately 2.5 rad. The movement in the transverse plane (internal-external rotation) and frontal plane (adduction-abduction) is restricted because of the shape of the tibial and femoral condyles and obstruction from soft tissue. Because the predominant movement of the knee is extension-flexion in the sagittal plane, three dimensional biomechanical models for the examination of knee function are not essential and acceptable results can be obtained using simplified two dimensional models (Moeinzadeh, 1983; Wongchaisuwat *et al.*, 1984; Yamaguchi and Zajac, 1989; Nisell *et al.*, 1986). The determination of a biomechanical model for the examination of knee extension requires the measurement of several anatomical parameters using either cadaveric joints (Grood *et al.*, 1984; Wickiewicz *et al.*, 1984; van Eijden *et al.*, 1985) or medical imaging methods such as radiography and magnetic resonance imaging (Smidt, 1973; Nisell *et al.*, 1986; Soudry *et al.*, 1986).

One of the most important parameters for the examination of muscle and joint forces in knee extension using a biomechanical model is the patellar tendon (PT) moment arm (Smidt, 1973; Nisell *et al.*, 1986). Its measurement in isokinetic analysis is essential because it allows the computation of the muscular force during knee extension from external moment measurements using isokinetic dynamometers (Johnson, 1982; Nisell *et al.*, 1989). Different definitions and measurement methods have been used however for the PT moment arm: a) Measurement of the extensor muscle force required to maintain an external flexing moment in cadaveric joints (Kaufer, 1971; Wendt and Johnson, 1985), b) The PT moment arm is defined as the perpendicular distance between the instantaneous centre of rotation (ICR) of the knee joint and the PT (Smidt, 1973; Soudan *et al.*, 1979) and is measured using lateral X-rays of the knee joint in different flexion positions, c) The tibiofemoral (TF) contact point is used instead of the ICR for the determination of the PT moment arm (Lindahl and Movin, 1967; Reilly and Martens, 1972; Bishop, 1977, Ellis *et al.*, 1979; Nisell *et al.*, 1986). This method was also used in the present study. The advantage of this method is that the moment arm of the tibiofemoral (TF) contact forces (compressive and shear) is negligible and therefore the moments produced by these forces can be omitted from a model for the computation of the resultant joint moment (Nisell *et al.*, 1989).

The purpose of this study was the measurement of PT moment arm, tibial plateau-tibial axis angle and PT-tibial axis angle during knee extension using the radiographic technique described in Chapter 5. These parameters allow the determination of a two dimensional biomechanical model of the knee for the measurement of muscle and joint forces during isokinetic knee extension.

METHODS

Instrumentation

A SIEMENS PANTOSKOP/EXPLORATOR X-ray unit with a SIRECON television unit was used in the present study. Further details of the system are described in Chapter 5.

Subjects

Five males without knee joint injury signed informed consent and volunteered to participate in this study. The experimental procedure (recording of the movement using an X-ray II video system) was approved by the Ethical Committee of the University Hospital. The anthropometric data of the subjects are presented in Table 6.1.

Table 6.1. Anthropometric measurements (mean ± SD)

Age	20.80 ± 3.89 years
Height	179 ± 3.2 cm
Mass	79.25 ± 7.17 kg
Epicondyle Width	9.1 ± 1.6 cm
Shank Length	45.2 ± 2.1 cm

Procedures

Knee extension was performed in front of the II screen on the sagittal plane. The upper leg was secured in a special supporting base in order to eliminate movement of the upper leg during knee extension. The lateral side of the joint remained in contact with the II protective plate throughout knee extension in order to eliminate movement in the frontal plane and therefore perspective error. The distance from the patella midpoint to the II protective plate was recorded with the subject in that position. After the recording of the extension movement the calibration structure (see Chapter 5) was placed parallel to the protective plate in order to allow distortion correction, reconstruction of the two dimensional coordinates and measurement of the anatomical parameters. The distance between calibration structure and II plate was equal to the distance between patella midpoint and II plate.

Ideally the X-ray recording of the movement must take place during the isokinetic test in order to measure accurately the effects of muscular activation and tibia-femur translation on the anatomical parameters (Johnson, 1982; Lavin and Gross, 1990). The spatial arrangements of II units and isokinetic dynamometers however do not allow the simultaneous recording of the movement. In order to reduce this error, manual resistance was applied to the shank during the X-ray recording, near the point of the input arm attachment.

The subjects were instructed to perform consecutive knee extension-flexion movements covering the complete ROM of the joint at a very slow speed in order to avoid deterioration of X-ray image quality. On average two complete extension-flexion sequences from each subject were recorded on video with an approximate exposure time of five seconds. Special radiation protection robes were used during the experimental procedure.

Data Reduction

Ten frames at approximately 0.17 rad (10 degrees) of knee flexion intervals were digitised by an experienced operator. The TF contact point was determined as the minimum distance between the femoral condyles and the tibial plateau. Thirty arbitrary points on the contours of the femoral condyles at intervals of approximately 2-5 mm were digitised. The images of the contours of the two femoral condyles were digitised separately because the two condyles are not congruent. The same procedure was followed for the medial and lateral sides of the tibial plateau. The two dimensional coordinates were computed using the distortion correction and reconstruction procedure described in Chapter 5. The femoral condyles and the tibial plateau digitised coordinates were then fitted with a cubic B-spline. (Rankin, 1989). This procedure allowed the modelling of the femoral condyles and the tibial plateau using piecewise cubic polynomials. The smoothed two dimensional coordinates derived from the B-splines were then used to determine the two midpoints of the minimum distances between the two femoral condyles and the lateral and medial tibial plateaus.

The TF contact point was determined as the midpoint of the line

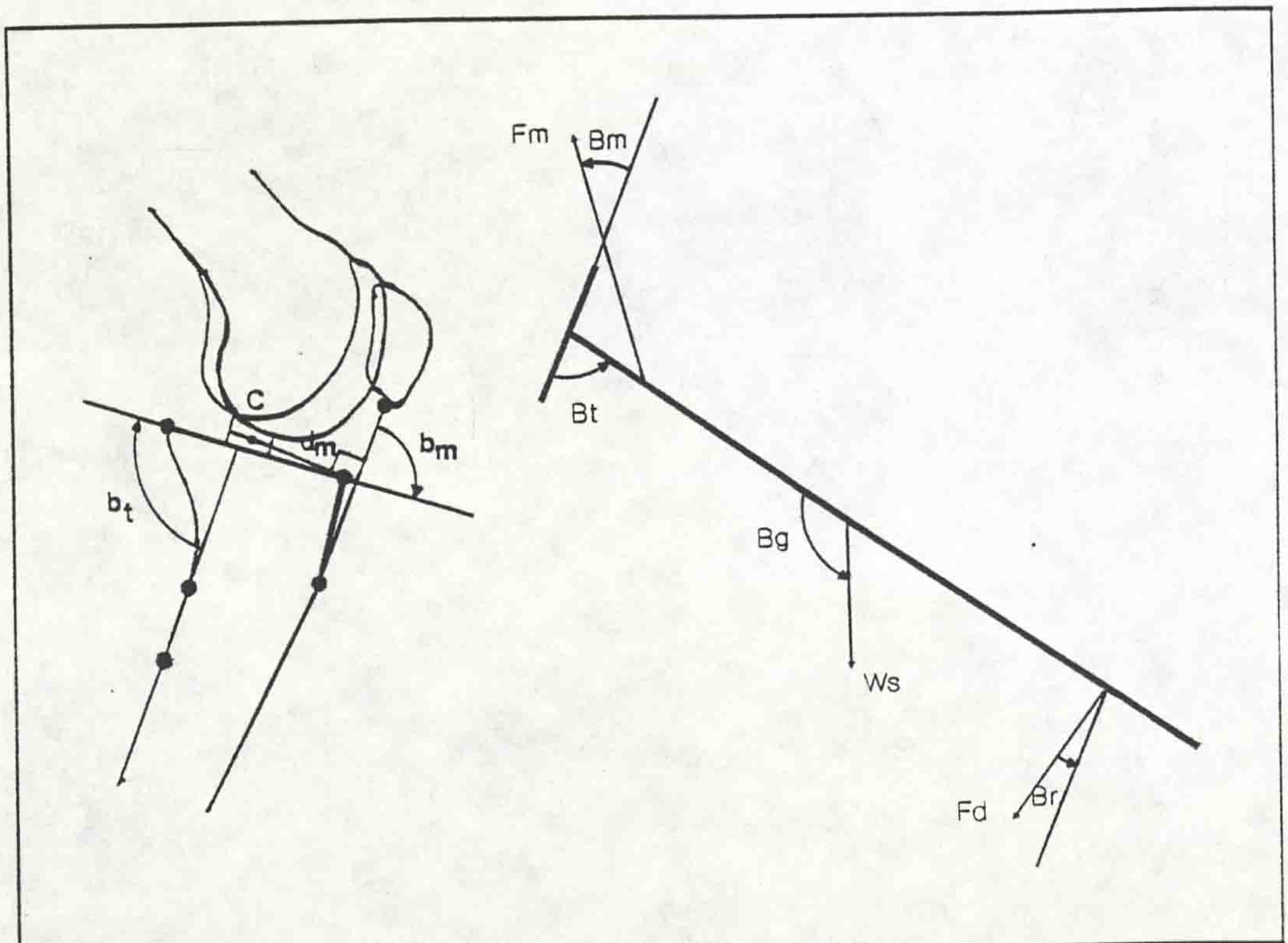

Figure 6.1. Definition of anatomical parameters (C: TF contact point, b_m: PT-tibial plateau angle, d_m: PT moment arm, b_t: tibial plateau-tibial axis angle).

between these two points (Fig. 6.1). The patellar tendon was modelled as a line segment using the coordinates of the patella and tibial tuberosity. A similar procedure was followed for the determination of the tibial plateau using the coordinates of the anterior and posterior border of the tibial plateau. The femoral and tibial long axes were determined from two arbitrary points on the posterior borders of the bones, approximately 5-7 mm from the femoral condyles and the tibial plateau respectively. Full extension was determined as the maximum angle computed between the long axes of the femur and tibia. Knee flexion angles were subsequently computed using this offset.

Knee flexion angle, PT moment arm and PT-tibial plateau angle data were smoothed using a Butterworth digital filter (Winter *et al.*, 1974), in order to eliminate digitising error. The variance of the digitising process was computed by repeated digitisation of an arbitrary frame. The smoothing factor was subsequently determined by comparing the variance of the digitising process with the variance in the residuals using different smoothing factors, until the difference between the two variances was minimal (Winter *et al.*, 1974; Lesh *et al.*, 1979). The smoothed data were then interpolated using cubic spline interpolation procedures (Reinsch, 1967). PT moment arm and PT-tibial plateau angles for every 0.087 rad (5 degrees) of knee flexion were then computed from the cubic polynomials.

RESULTS

Anatomical Parameters

The mean PT moment arm at full extension was 33.81 $\pm$3.44 mm, increased to a maximum of 39.87 $\pm$2.4 mm at 0.785 rad of knee flexion and decreased to 33.63 $\pm$4.01 mm at 1.57 rad (Fig. 6.2). The PT-tibial plateau angle was 1.967 $\pm$0.12 rad at full extension and decreased linearly to 1.53

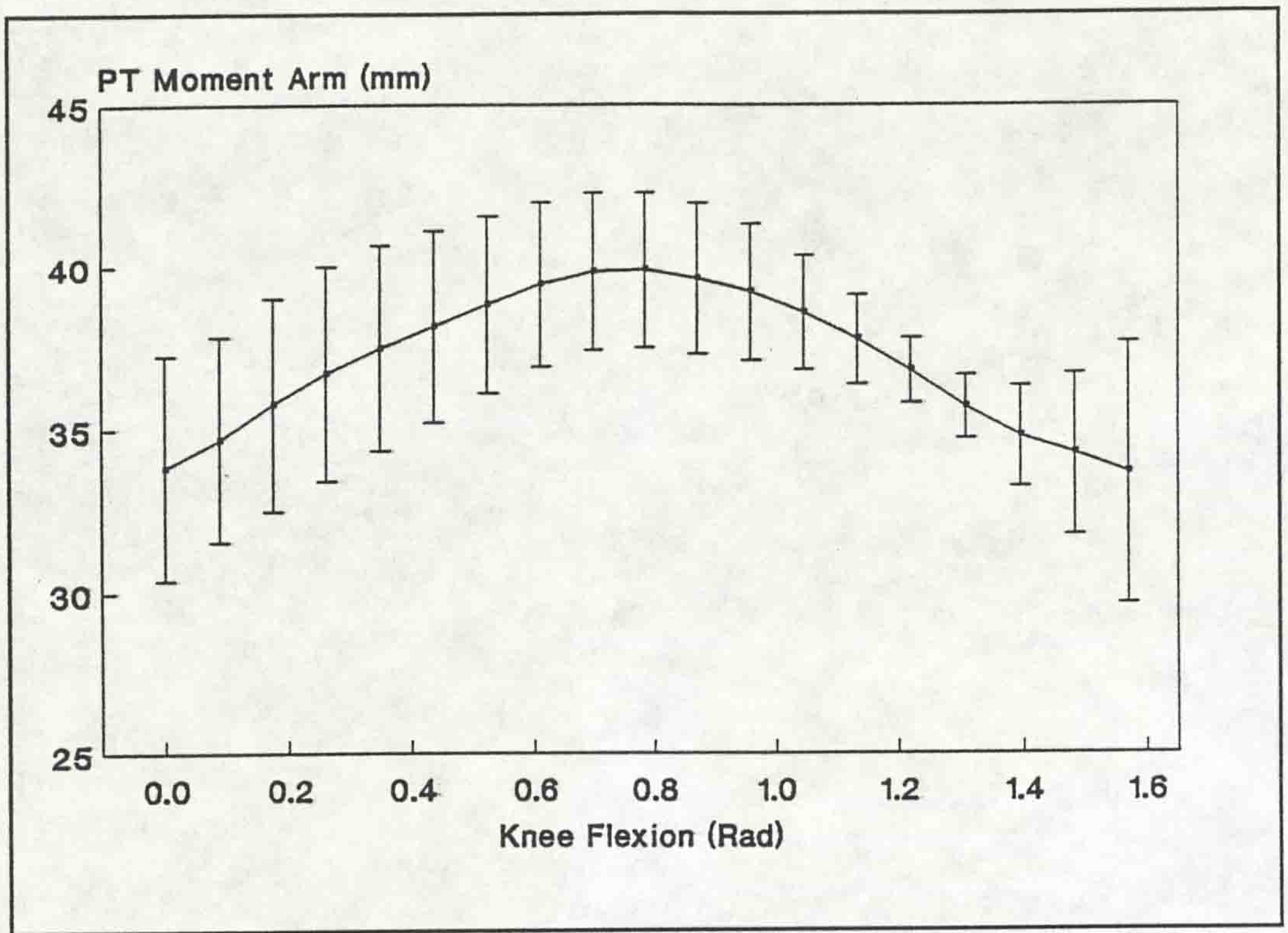

Figure 6.2. PT moment arm at different knee flexion angles.

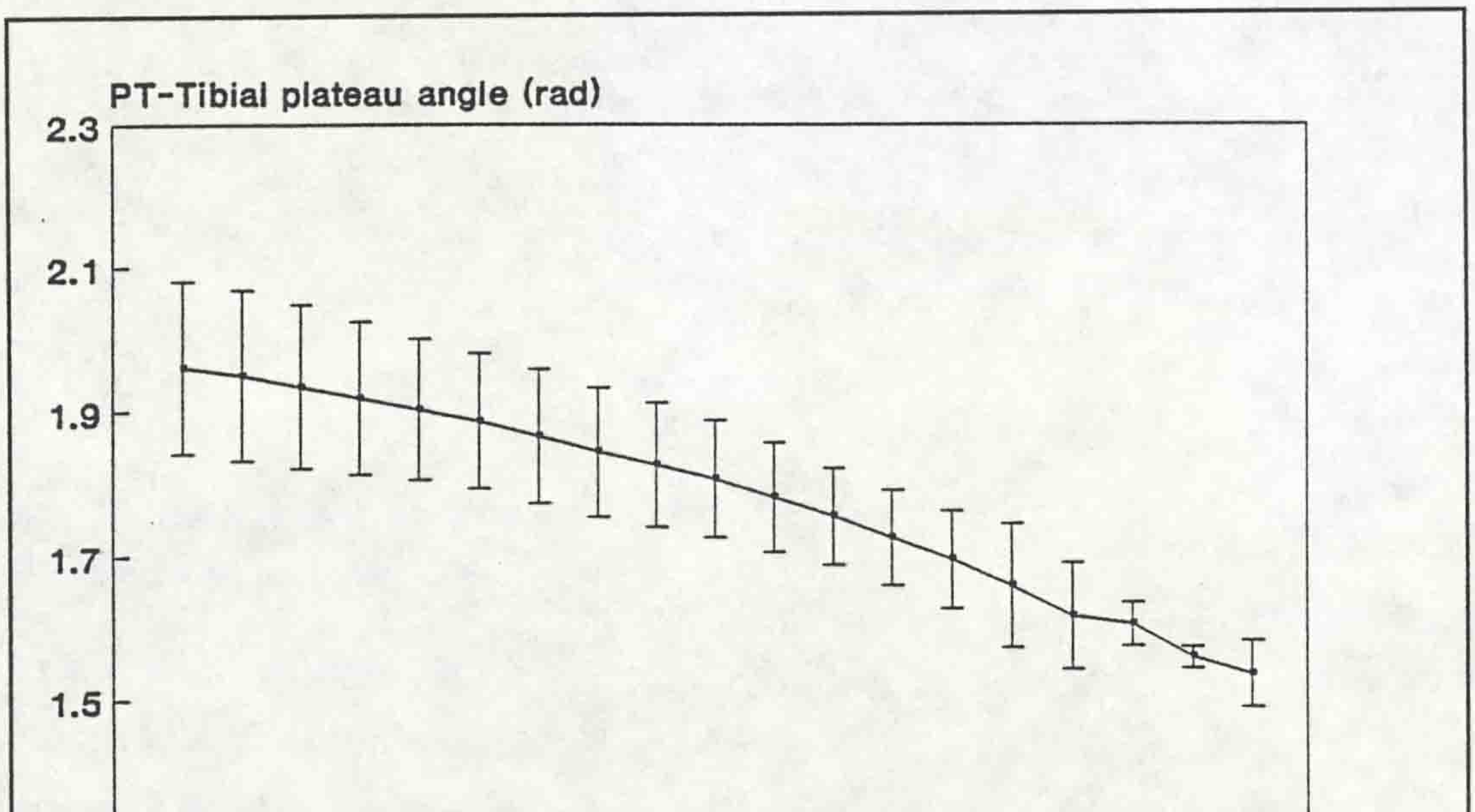

Figure 6.3. PT-Tibial plateau angle at different knee flexion angles.

±0.05 rad at 1.57 rad of knee flexion (Fig. 6.3). The mean angle between the tibial plateau and the tibial long axis (Fig. 6.1) was 1.48 ±0.04 rad.

Reliability

The reliability of the digitising process using this system was examined by digitising a single frame ten times. The knee flexion angle, PT moment arm and PT-tibial plateau angle were computed in each repeated frame (Fig. 6.4). The variance in the digitised data was considered as the variance of the

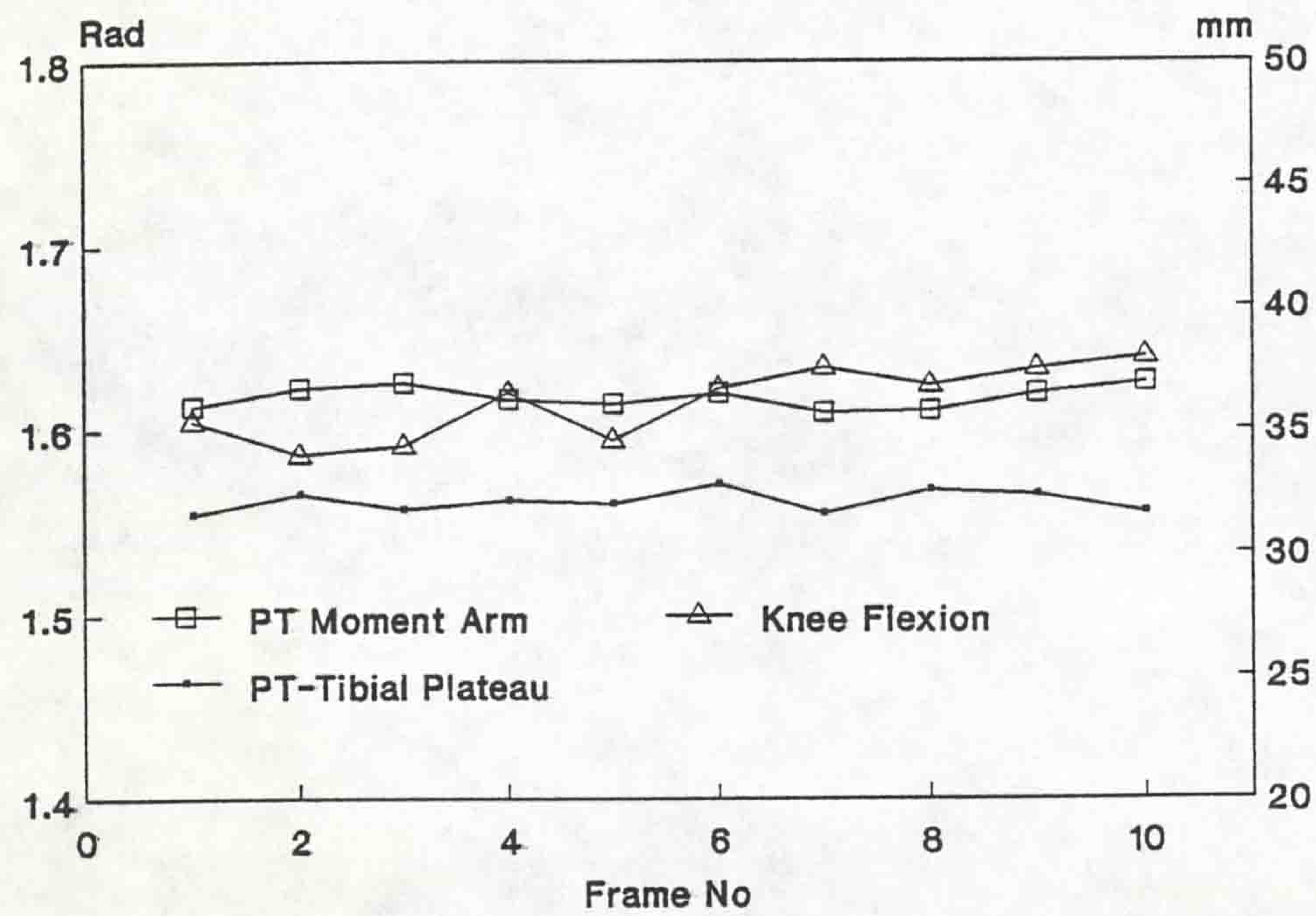

Figure 6.4. Repeated measurements (n = 10) of anatomical parameters.

digitising process because all the points were stationary in time and therefore no signal was present in the data (Winter *et al.*, 1974).

DISCUSSION

The PT moment arm-knee flexion angle relationship is similar to that reported in recent studies using radiographic techniques (Nisell *et al.*, 1986) and mathematical modelling of the knee joint (e.g. Yamaguchi and Zajac, 1989) (Fig. 6.5). The main difference in previous studies that examined the PT moment arm (Smidt, 1973; Kaufer, 1971) is its length near full extension. The

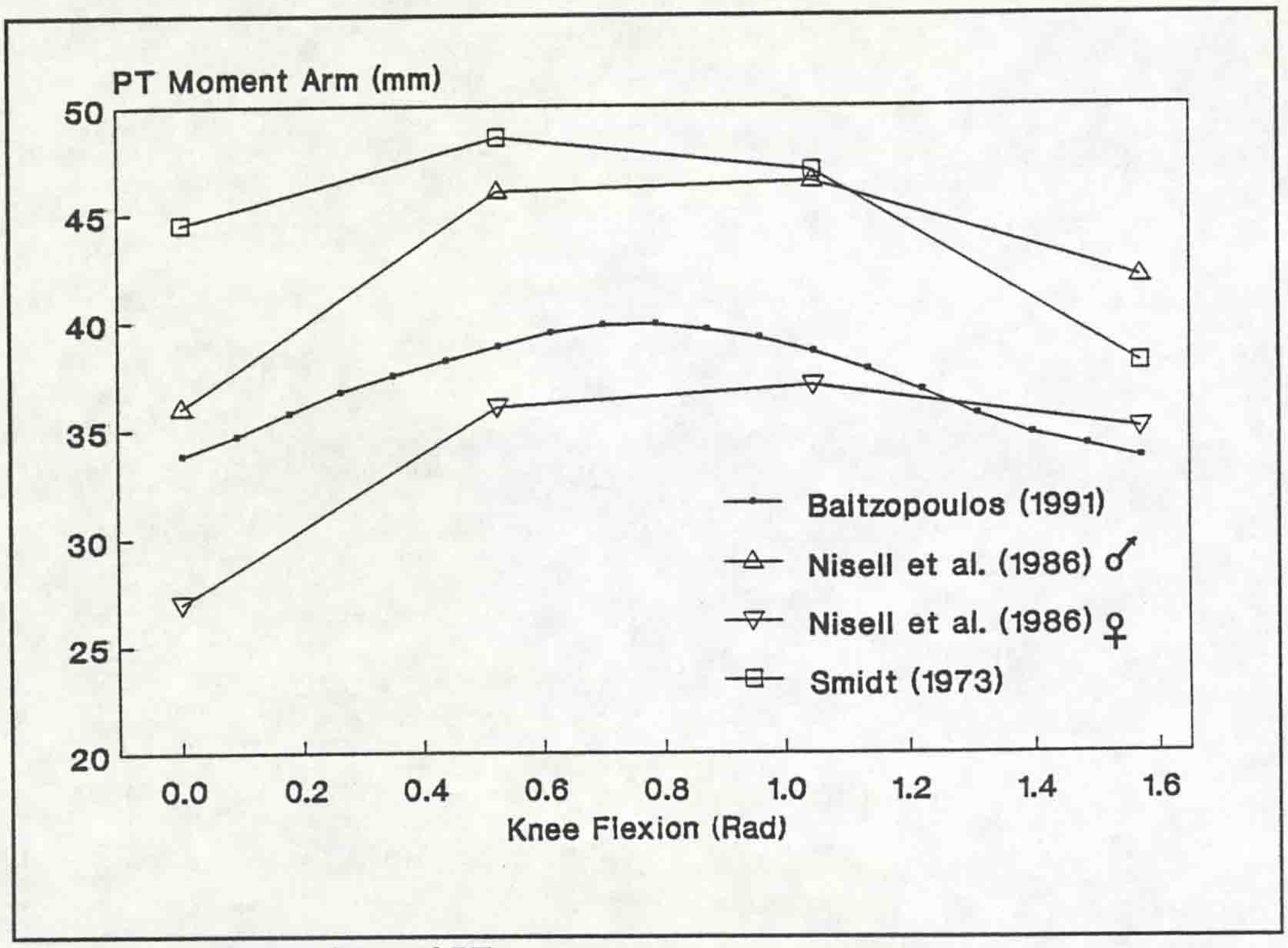

Figure 6.5. Comparison of PT moment arm measurements.

results of the present study agree with recent findings (Nisell *et al.*, 1986; Yamaguchi and Zajac, 1989) that show a decrease of the moment arm near full extension and therefore a mechanical disadvantage of the knee extensors at this position. Smidt (1973) reported only a small decrease at full extension (Fig. 6.5). Kaufer (1971) reported that the maximum moment arm length was at full extension. On the contrary the results of the present study indicate that PT moment arm is at its maximum at approximately 0.70-1.05 rad (40-60 degrees) of knee flexion. These findings are in agreement with the measurement of the maximum resultant joint moment using isokinetic dynamometers.

The PT-tibial plateau angle at different positions of knee flexion was previously measured only by Nisell *et al.*, (1986) and was defined as the angle between the PT and an axis perpendicular to the tibial plateau instead of the tibial plateau axis used in the present study. By adjusting the data for this offset (1.57 rad), a similar relationship between PT-tibial plateau angle and knee flexion angle was also observed in the present study (Fig. 6.6). These results

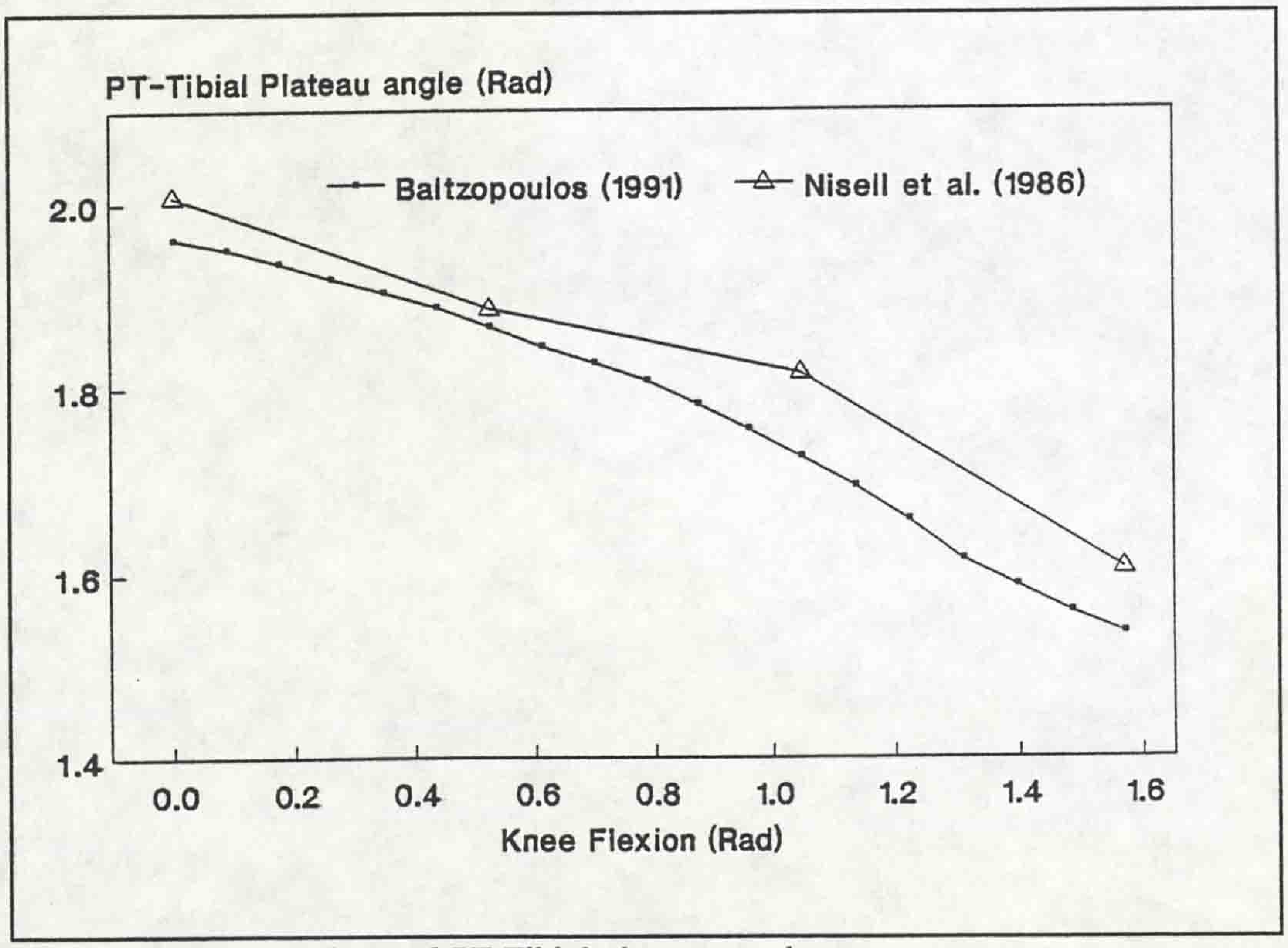

Figure 6.6. Comparison of PT-Tibial plateau angle.

indicate that for knee flexion angles between 0-1.57 rad, knee extensor activity translates the tibia in an anterior direction relative to the femur. For knee flexion angles above 1.57 rad however, a posterior translation of the tibia is indicated.

The tibial plateau-tibial axis angle was approximately 1.48 rad. This indicates a posterior slope of the tibial plateau relative to the tibial axis. The measurement of this angle is essential for the computation of the shear and compressive TF forces and similar findings have been reported previously (Nisell *et al.*, 1986).

The results of the reliability measurements indicate that the variance in the computation of the PT-tibial plateau angle was minimal. The opposite was observed for the measurement of knee flexion angle. These results reflect the limitations of the digitising process. The PT-tibial plateau angle is determined from distinct points that are usually well defined in the X-ray recording (anterior, posterior tibial plateau, patella apex and tibial tuberosity) and therefore digitising error is minimal. The knee flexion angle and the PT moment arm however are determined from arbitrary points (see Methods), that are not well defined, increasing digitising error.

CONCLUSIONS

Anatomical parameters of the knee joint required for biomechanical models have been determined during knee extension *in vivo* using videofluoroscopy. The main findings within the limitations of the present study are that PT moment arm is maximum at 0.785 rad of knee flexion and decreases approximately 15% near full extension. PT-tibial plateau angle is maximal at full extension and decreases linearly with knee flexion.

Measurement reliability of the above parameters depends on the anatomical points included in the digitising process.

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CHAPTER 7

MUSCULAR AND TIBIOFEMORAL CONTACT FORCES DURING ISOKINETIC KNEE EXTENSION

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ABSTRACT

The purpose of this study was the examination of muscle and tibiofemoral (TF) contact forces during isokinetic knee extension at angular velocities ranging from 0.52 rad·s^{-1} to 3.66 rad·s^{-1}, using a two dimensional biomechanical model. Five males (mean age 20.8 ±years, mass 79.2 ±7.2 kg and height 179 ± 3.2 cm) without knee joint injury history participated in the study. The maximum moment (mean ±SD) ranged from 226.20 ±39.52 Nm at 0.52 rad·s^{-1} to 166 ±27.56 Nm at 3.66 rad·s^{-1}. These differences were significant ($F_{3,12}=17.9$, $p<0.05$) and *post-hoc* tests revealed that the significant differences were between the moments at 0.52 rad·s^{-1} and 2.62-3.66 rad·s^{-1}. The maximum muscular force ranged from 7.55 ±0.49 times body weight (BW) at 0.52 rad·s^{-1} to 5.72 ±0.94 BW at 3.66 rad·s^{-1}. The compressive tibiofemoral force ranged from 7.53 ±0.49 BW at 0.52 rad·s^{-1} to 5.68 ±0.91 BW at 3.66 rad·s^{-1} and the shear tibiofemoral force from 0.94 ±0.48 BW to 0.83 ±0.35 BW respectively. These differences were significant for both maximum muscular force ($F_{3,12}=13.7$, $p<0.05$) and compressive tibiofemoral force ($F_{3,12}=13.57$, $p<0.05$). Differences between the shear forces at the different angular velocities were not significant ($F_{3,12}=0.64$, $p>0.05$). These results indicate that the forces developed during maximal isokinetic knee extension are significantly reduced relative to other dynamic activities and therefore isokinetic dynamometry is a safe and effective method for muscle function assessment, training and rehabilitation.

NOMENCLATURE

$\mathbf{M_d}$	resistive dynamometer moment.
$\mathbf{F_s}$	force exerted by the limb on the input arm.
$\mathbf{d_d}$	moment arm of $\mathbf{F_s}$ around the dynamometer axis of rotation.
$\mathbf{W_d}$	gravitational force of the input arm.
$\mathbf{d_{wd}}$	distance between dynamometer axis of rotation and input arm centre of mass (CM).
I_d	moment of inertia of input arm around the axis of rotation of the dynamometer.
α_d	angular acceleration of input arm.
$\mathbf{M_m}$	knee joint resultant moment.
$\mathbf{F_d}$	force exerted by the input arm on the leg as a reaction to $\mathbf{F_s}$.
$\mathbf{d_s}$	moment arm of $\mathbf{F_d}$ around the knee joint axis of rotation.
$\mathbf{W_s}$	gravitational force of the shank-foot system.
$\mathbf{d_{ws}}$	distance between knee joint axis of rotation and shank-foot system CM.
I_s	moment of inertia of shank-foot system around the knee joint axis of rotation.
α_s	angular acceleration of the shank-foot system around the knee joint axis.
$\mathbf{F_m}$	knee extensor muscular force.
$\mathbf{F_c}$	tibiofemoral compressive force.
$\mathbf{F_s}$	tibiofemoral shear force.
$\mathbf{a_r}$	radial acceleration of shank-foot system CM.
$\mathbf{a_t}$	tangential acceleration of shank-foot system CM.
b_r	angle between $\mathbf{F_d}$ and tibial plateau.
b_m	angle between $\mathbf{F_m}$ and tibial plateau.
b_g	angle between $\mathbf{W_s}$ and tibial plateau.
b_t	angle between $\mathbf{a_t}$ and tibial plateau.
m	mass of shank-foot system.

INTRODUCTION

Isokinetic dynamometry allows muscular activation during isolated joint movements at a controlled, constant angular velocity. The resistive moment developed by an isokinetic dynamometer is variable and equivalent to the resultant joint moment apart from the initial and final parts of the ROM, providing optimal muscular loading. Because of these unique features, isokinetic dynamometry has widespread applications in rehabilitation of muscular and ligamentous injuries (for a review of the applications see Baltzopoulos and Brodie, 1989).

In general, the resistive moment developed by the isokinetic dynamometer is not equal to the resultant joint moment (Winter *et al.*, 1981; Sapega *et al.*, 1982; Herzog, 1988). Implementation of appropriate correction methods however (Nelson and Duncan, 1983; Herzog,1988), allows the measurement of the resultant joint moment and assessment of muscle function using isokinetic dynamometers.

Despite the application of isokinetics in the assessment of dynamic muscle function both in normal and pathological conditions, only a limited number of studies examined the forces developed during isokinetic movements at different angular velocities (Wickiewicz *et al.*, 1984; Nisell *et al.*, 1989);

Johnson (1982) introduced the Johnson Anti-Shear Accessory (JASA) in order to reduce the anterior shear TF contact force during knee extension and flexion movements using the CYBEX dynamometer. It was suggested that the typical input arm for knee extension movements, with the contact point near the malleoli, produces considerable shear force at the knee and consequently overstresses repaired ligaments. The JASA is composed of two freely pivoting

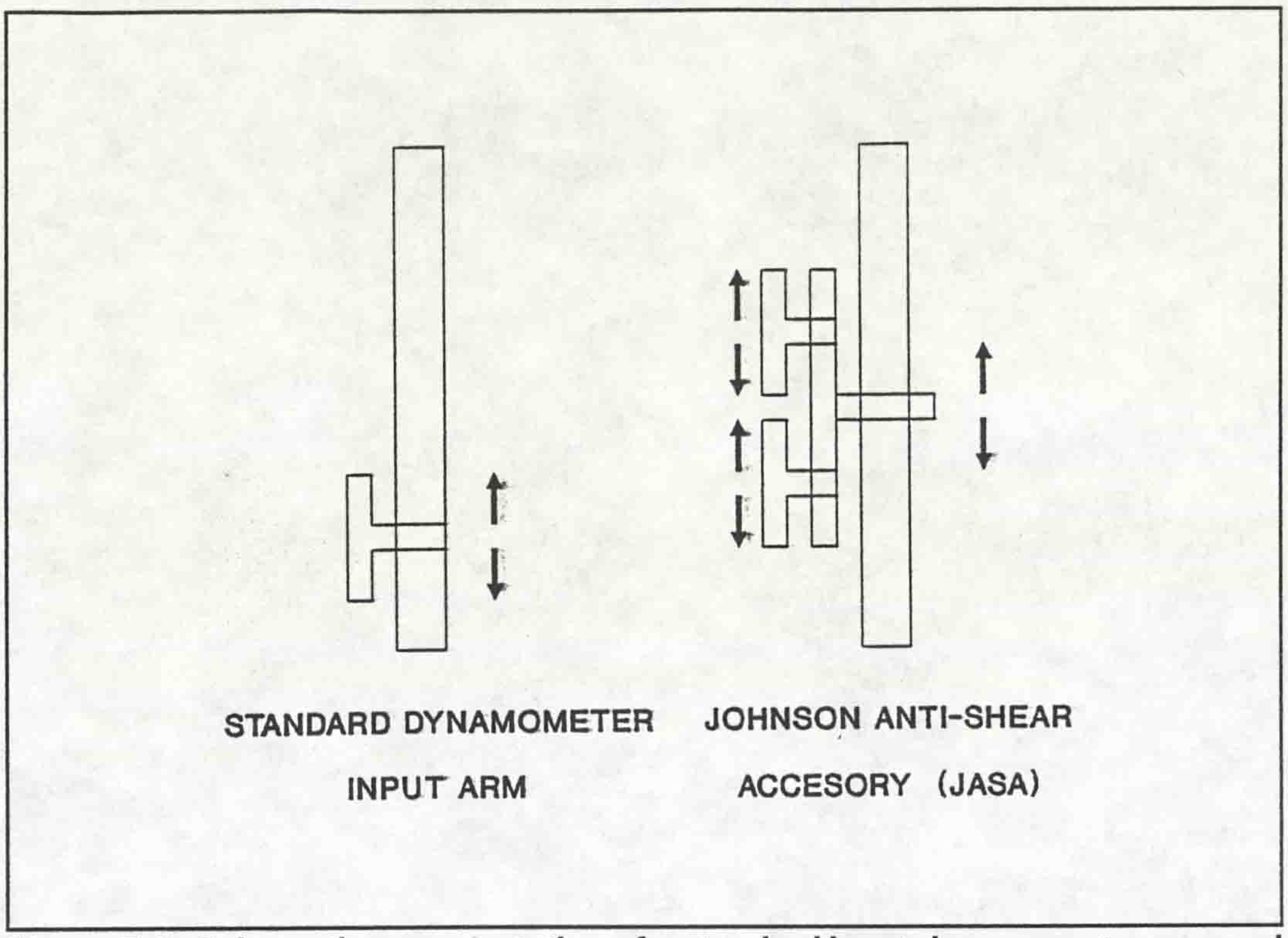

Figure 7.1. Schematic representation of a standard input dynamometer arm and the Johnson Anti Shear Accessory (JASA).

resistance pads which are secured to the shank at a proximal and a distal position (Fig. 7.1). The pads are connected by a bar that also pivots freely at a fulcrum attached to the input arm of the dynamometer. Positioning the fulcrum at a proximal position increases shank-input arm contact force and reduces anterior shear force. It was reported that JASA reduced the anterior shear force by approximately 7%-87% using reduced resistive moment arms, and improved the sagittal alignment between the femur and the tibia in a patient with anterior knee laxity. It was therefore suggested to use this device in clinical practice, during rehabilitation and testing of patients with ligamentous injuries.

Lavin and Gross (1990) examined tibial translation relative to the femur and input arm contact force during isometric knee extension and flexion at an angle of 0.785 rad of knee flexion, using the JASA and the standard CYBEX input arm. The results of this study are in agreement with Johnson (1982), indicating reduced tibial translation but increased tibial-input arm contact force using the JASA.

The main limitations of the above studies is that tibial translation was examined in isometric conditions only at specific knee angles (Johnson, 1982; Lavin and Gross, 1990). Furthermore, Lavin and Gross (1990) examined only the shank-input arm contact force without considering the tibiofemoral contact forces.

Wickiewicz *et al.* (1984) examined the torque output at a specific knee joint position during knee extension-flexion movements at angular velocities ranging from 0-5 rad·s^{-1}. It was reported that the maximum muscular force developed was 3.67 kN in extension and 4.63 kN in flexion. The isokinetic data however, were not corrected for the effect of gravitational forces.

To date, only Nisell *et al.* (1989) examined both muscular and tibiofemoral contact forces during isokinetic knee extension. Eight healthy male subjects performed an isokinetic knee extension movement on a CYBEX II dynamometer at 0.52 and 3.14 rad·s^{-1}. Two different positions of the resistance pad (proximal and distal) were used during the test at 3.14 rad·s^{-1}. The ROM was from 1.57 to 0 rad of knee flexion. The dynamometer torque output was corrected for the effects of gravitational forces using a passive fall of the shank-foot-input arm system (Nelson and Duncan, 1983). The acceleration of the dynamometer was controlled by a computer system (Gransberg and Knutsson 1983). In order to avoid the impact forces during the activation of the resistive mechanism, the initial and final 0.08 rad (5 degrees) in the ROM were excluded from the analysis. A two dimensional biomechanical model of the knee joint was used for the measurement of the tibiofemoral shear and compressive forces in the sagittal plane (Nisell *et al.*, 1986). The anthropometric measurements of the subjects (age, mass, height and femoral epicondyle width) in the study by Nisell *et al.* (1989) were similar to the data in the study by Nisell *et al.* (1986). The anatomical data for the model including tibiofemoral contact point and patellar tendon (PT) moment arm, were also assumed to be similar without any direct radiographic measurements. The maximum tibiofemoral compressive and shear forces developed were 9 BW and 1 BW respectively. It was also reported that proximal positioning of the input arm affected the shear but not the compressive tibiofemoral force.

The main limitation of these studies is that the muscle and joint forces were examined at a specific angle without any gravitational or inertial corrections (Wickiewicz *et al.*, 1984) or at a preselected part of the movement only (Nisell *et al.*, 1989). The exclusion of the initial and final parts of the movement in order to avoid the resistive moment developed by the dynamometer (torque overshoot) is an appropriate procedure for the accurate measurement of the resultant joint moment and the assessment of muscle function (e.g. Murray, 1986). In order to examine the joint forces however, the resistive dynamometer moment, both during the acceleration and constant velocity period of the movement must be included in the analysis. Failure to include the resistive moment in the initial period may underestimate the TF contact forces.

The purpose of this study was the examination of the muscular and tibiofemoral contact forces during isokinetic knee extension at angular velocities ranging from 0.52 $rad \cdot s^{-1}$ to 3.66 $rad \cdot s^{-1}$, using a two dimensional biomechanical model. The inertial forces and the resistive force developed by the dynamometer throughout the ROM were included in the model.

METHODS

Instrumentation

An AKRON isokinetic dynamometer interfaced to a microcomputer was used in the present study for the assessment of isokinetic knee extension. The operational details of the system have been presented in Chapter 4.

Subjects

Five males without knee joint injury signed informed consent and volunteered to participate in this study. The anthropometric measurements of the subjects have been presented in Table 6.1.

Procedures

Knee extension was examined using the above described system at angular velocities of 0.52, 1.52, 2.62 and 3.66 $rad \cdot s^{-1}$. The details of the experimental procedure have been presented in Chapter 4.

Biomechanical Model

The free body diagrams of the input arm and the shank-foot segment are shown in Fig. 7.2 and 7.3 respectively. The use of the TF contact point as the origin of the knee joint has been discussed in Chapter 6. The planar angular equation of the input arm in the sagittal plane about a frontal axis through O is the following:

$$\Sigma M = I_d \alpha_d \tag{7.1}$$

or (see Fig. 7.2)

$$-M_d + F_s \times d_d - W_d \times d_{wd} = I_d \alpha_d \tag{7.2}$$

and therefore

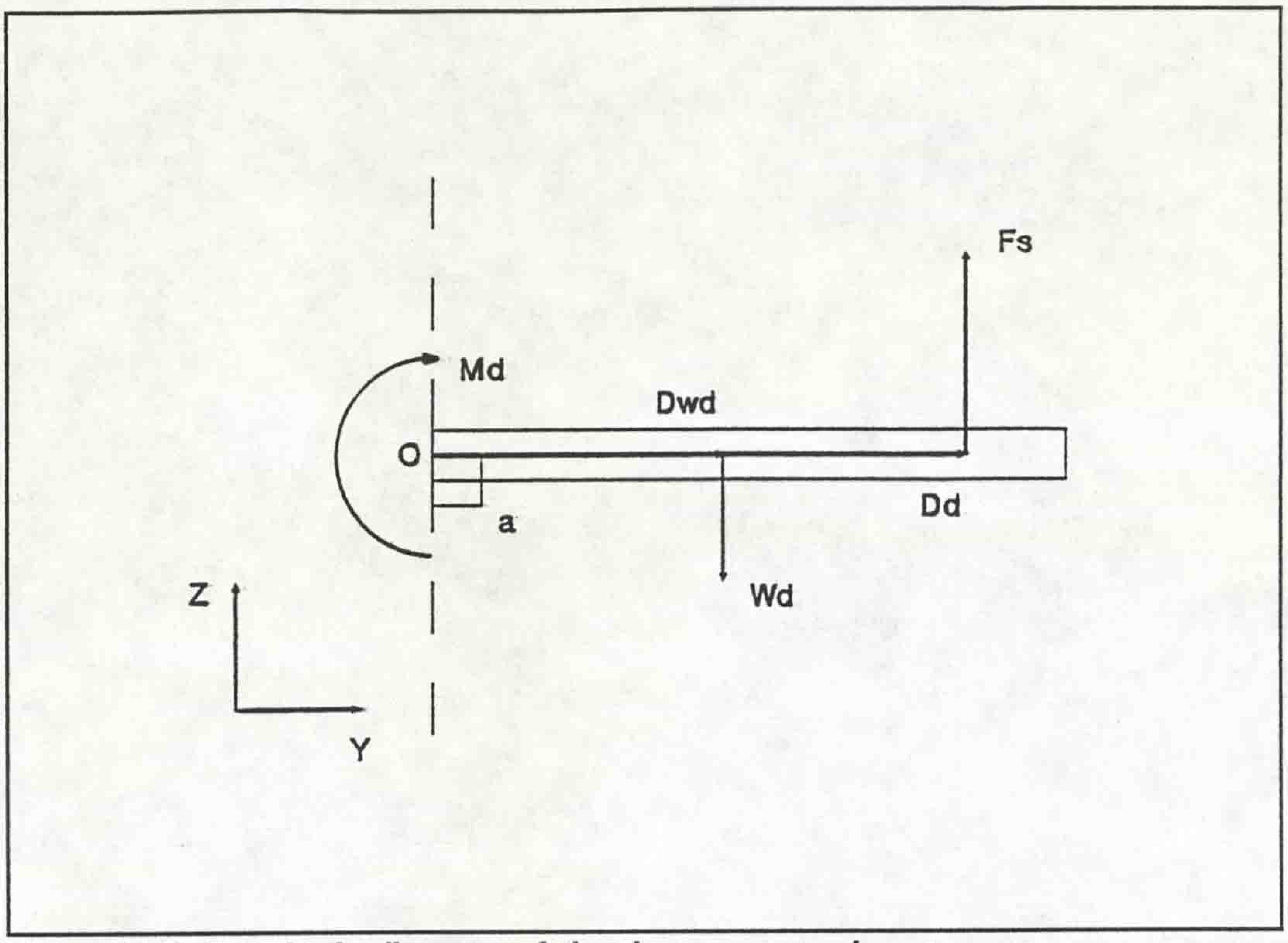

Figure 7.2. Free body diagram of the dynamometer input arm.

$$F_s \times d_d = I_d \alpha_d + M_d + W_d \times d_{wd} \tag{7.3}$$

$F_s \perp d_d$ and (7.3) becomes

$$F_s d_d = I_d \alpha_d + M_d + W_d \times d_{wd} \tag{7.4}$$

Similarly, the planar angular equation of the shank-foot segment in the sagittal plane about a frontal axis through the centre of the knee joint (C) is the following:

$$M_m - F_d \times d_s - W_s \times d_{ws} = I_s \alpha_s \tag{7.5}$$

$F_d \perp d_s$ and (7.5) becomes

$$F_d d_s = M_m - W_s \times d_{ws} - I_s \alpha_s \tag{7.6}$$

By definition, F_s and F_d have equal magnitudes (action-reaction). Assuming also that the axes of the input arm and the shank-foot segment and points C and O coincide, then $d_s = d_d$ and consequently $F_s d_d = F_d d_s$.

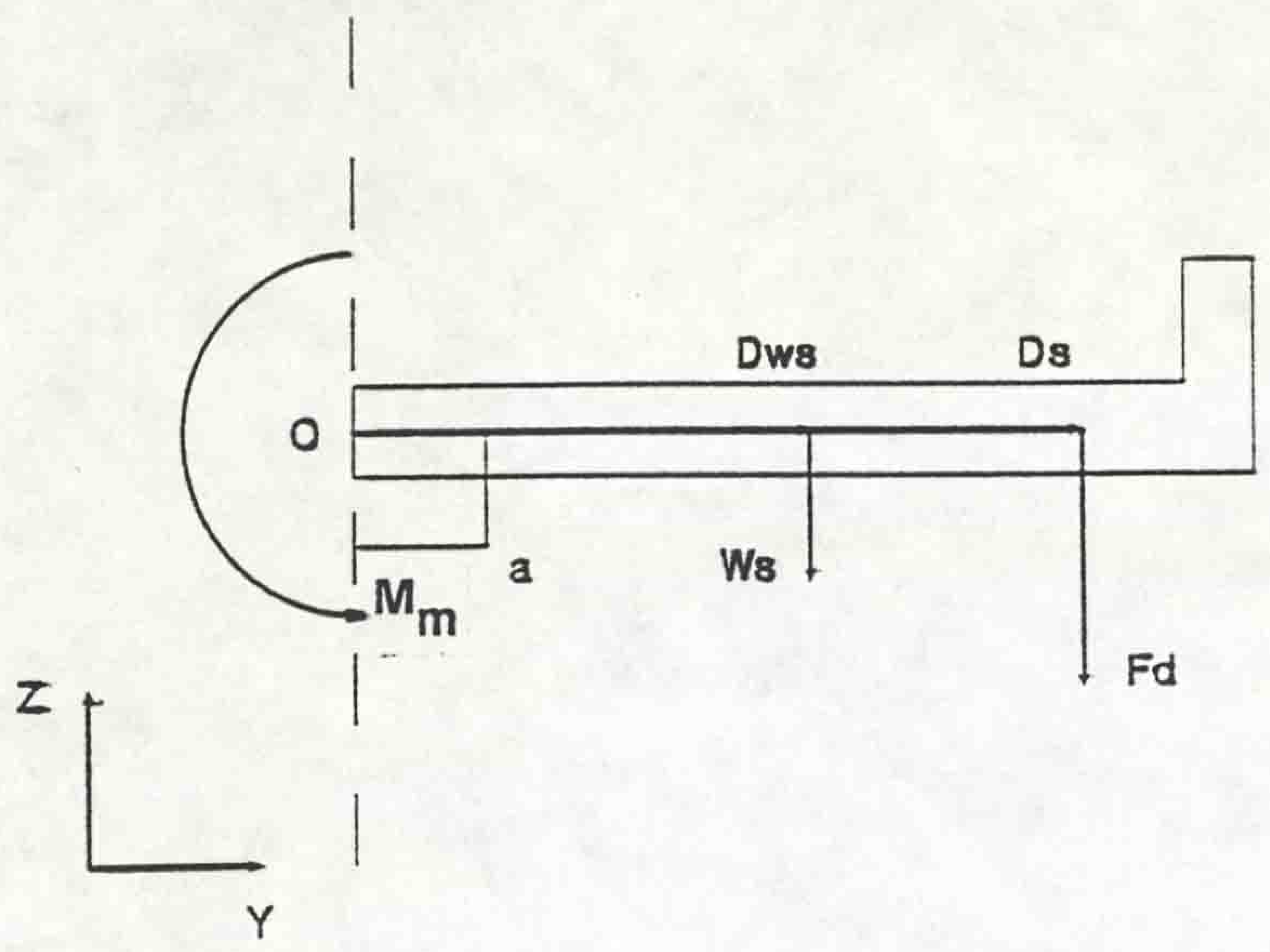

Figure 7.3. Free body diagram of the shank-foot system.

Substituting $F_d d_s$ in equation (7.6) with the equivalent $F_s d_d$ from (7.4) and rearranging the terms, equation (7.6) becomes

$$M_m - I_d\alpha_d - M_d - W_d \times d_{wd} - W_s \times d_{ws} = I_s\alpha_s \qquad (7.7)$$

and therefore

$$M_m = M_d + I_d\alpha_d + I_s\alpha_s + W_d d_{wd} + W_s d_{ws} \qquad (7.8)$$

This equation describes the relationship between the resultant joint moment (**M_m**) and the moment recorded by the isokinetic dynamometer (**M_d**) (Herzog, 1988). **M_m** is not equal to **M_d** and the terms in equation (7.8) represent the effects of the gravitational and inertial forces.

If the points C and O do not coincide because of misalignment errors, then **$d_s \neq d_d$** and equation (7.8) becomes

$$M_m = (M_d + I_d\alpha_d + W_d \times d_{wd}) \cdot (d_s/d_d) + I_s\alpha_s + W_s \times d_{ws} \qquad (7.9)$$

In the present study it was assumed that $d_s = d_d$ and therefore equation (7.8) was used. Herzog (1988) reported that misalignment error has a minimal effect on the measurement of $\mathbf{M}_m$ (1%-2%). However, the accurate measurement of d_s from external anatomical landmarks or markers using film cameras must be questioned. Given the magnitude of these differences it is possible that the error reported resulted from skin movement and digitising inaccuracies in locating the knee joint marker. Accurate measurement of the relative position of points C and O and consequently d_s, requires the use of radiography during the movement. The difficulties associated with this procedure however have been discussed in Chapter 6.

The moment of the gravitational forces was computed and $\mathbf{M}_d$ was corrected using the procedure described in Chapter 4. I_d was determined mathematically by modelling the input arm of the dynamometer as two rectangular blocks (Beer and Johnston, 1972). I_s was determined using anthropometric data for the moment of inertia of the shank-foot system (Dempster, 1955). Angular velocity and acceleration of the system were computed from the angular position-time data using the smoothing and differentiation method described in Chapter 4.

The kinematic parameters of the shank-foot system were assumed to be equal to the kinematic parameters of the input arm. This assumption is not valid if the shank-foot system is not secured properly and is moving relative to the input arm of the dynamometer. Furthermore, the interface between shank and input arm is not rigid, allowing relative movement during impacts. Herzog (1988) reported that the error resulting from relative movement between limb and input arm was minimal at 2.09 $rad \cdot s^{-1}$ (<1%) but increased to 4% at 4.19 $rad \cdot s^{-1}$. In the present study the shank-foot was secured to the input arm and for the range of angular velocities examined, it was assumed that the kinematic parameters of the shank-foot system and input arm were equal.

Once M_m and M_d are determined, the muscular (F_m) and resistive force (F_d) acting on the shank-foot segment can be computed as $F_m = M_m/d_m$ and $F_d = M_d/d_d$.

The patellar tendon moment arm d_m was measured using the procedure described in Chapter 6. M_m is the resultant joint moment. In the present study however, M_m was attributed to the activation of the knee extensors only. This is a valid assumption because muscular activation of the knee flexors is minimal during isokinetic knee extension (Osternig *et al.* 1983; Nisell *et al.*, 1989). The contribution of soft tissue forces in the resultant joint moment is also negligible compared to the muscular forces exerted during isokinetic knee extension (Dowson and Wright, 1981; Nisell at al., 1989).

The tibiofemoral contact forces were examined by considering the free body diagram of the shank-foot segment. The linear equation of motion in the direction of the compressive force is

$$F_c - F_d \sin b_r - F_m \sin b_m + W_s \sin b_g = m a_r \sin b_t - m a_t \cos b_t \tag{7.10}$$

and therefore the compressive force is

$$F_c = F_d \sin b_r + F_m \sin b_m - W_s \sin b_g + m a_r \sin b_t - m a_t \cos b_t \tag{7.11}$$

The linear equation of motion in the direction of the shear force is

$$F_s + F_d \cos b_r - F_m \cos b_m + W_s \cos b_g = m a_r \cos b_t + m a_t \sin b_t \tag{7.12}$$

and therefore the shear force is

$$F_s = -F_d \cos b_r + F_m \cos b_m - W_s \cos b_g + m a_r \cos b_t + m a_t \sin b_t \tag{7.13}$$

Data Analysis

Maximum moment, muscular and TF contact force differences at the different angular velocities were examined using one way ANOVA test. The same procedure was followed for the angular position (knee flexion angle) of the maximum moment and force measurements at the different angular velocities. In order to ensure that the assumption of homogeneity of variance was not violated because of the limited number of subjects ($N=5$), Cochran's tests (Dixon and Massey, 1969) were performed for all sets of data (Appendix II). The results of these tests indicated that there were no significant differences between the variances and therefore homogeneity of variance was accepted and ANOVA tests were performed.

RESULTS

The maximum moment (mean $\pm$SD) ranged from 226.20 $\pm$39.52 Nm at 0.52 rad·s^{-1} at to 166 $\pm$27.56 Nm at 3.66 rad·s^{-1} (Fig. 7.4). These differences were overall significant ($F_{3,12}=17.9$, $p<0.05$) and subsequent Tukey tests revealed that the significant differences were between the moments at 0.52 rad·s^{-1} and 2.62-3.66 rad·s^{-1}. The maximum muscular force ranged from 7.55 $\pm$0.49 BW at 0.52 rad·s^{-1} to 5.72 $\pm$0.94 BW at 3.66 rad·s^{-1}. (Fig. 7.5). The compressive tibiofemoral force ranged from 7.53 $\pm$0.49 BW at 0.52 rad·s^{-1} to 5.68 $\pm$0.91 BW at 3.66 rad·s^{-1} (Fig. 7.6) and the shear tibiofemoral force from 0.94 $\pm$0.48 BW to 0.83 $\pm$0.35 BW respectively (Fig. 7.7). These differences were significant for both maximum muscular force ($F_{3,12}=13.7$, $p<0.05$) and compressive tibiofemoral force ($F_{3,12}=13.6, p<0.05$).

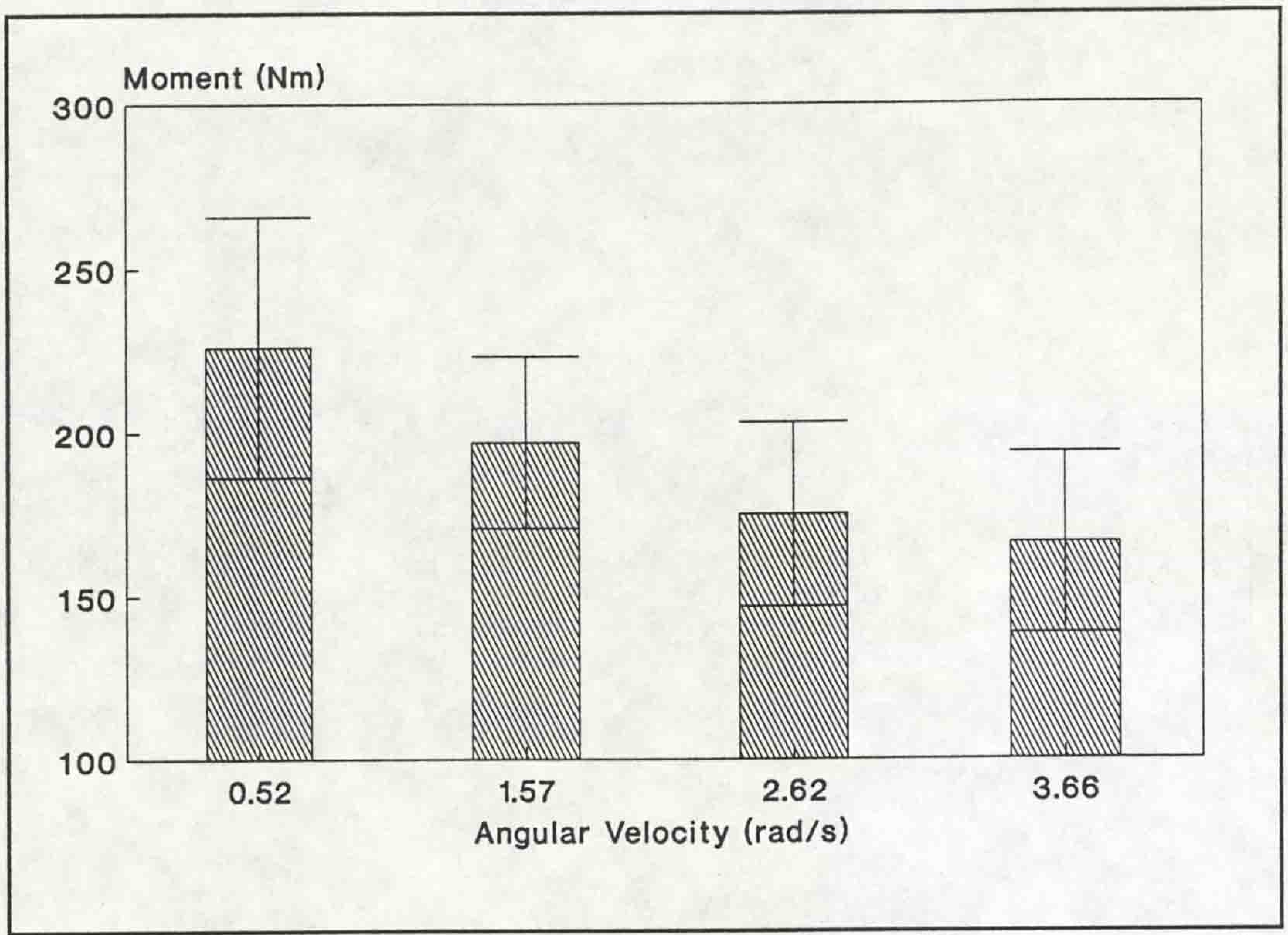

Figure 7.4. Maximum resultant moment (mean ± SD) at different preset angular velocities.

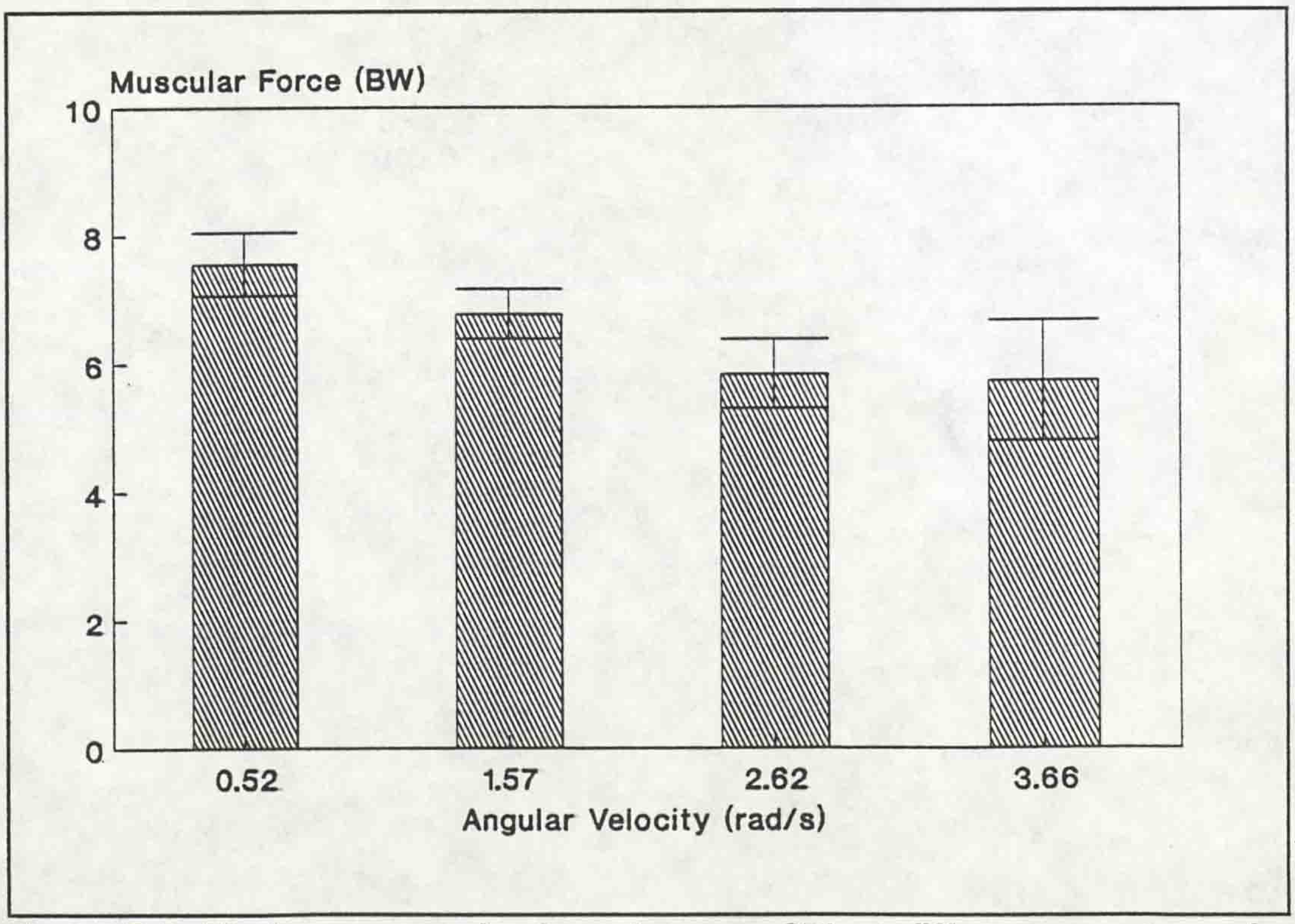

Figure 7.5. Maximum muscular force (mean ± SD) at different preset angular velocities.

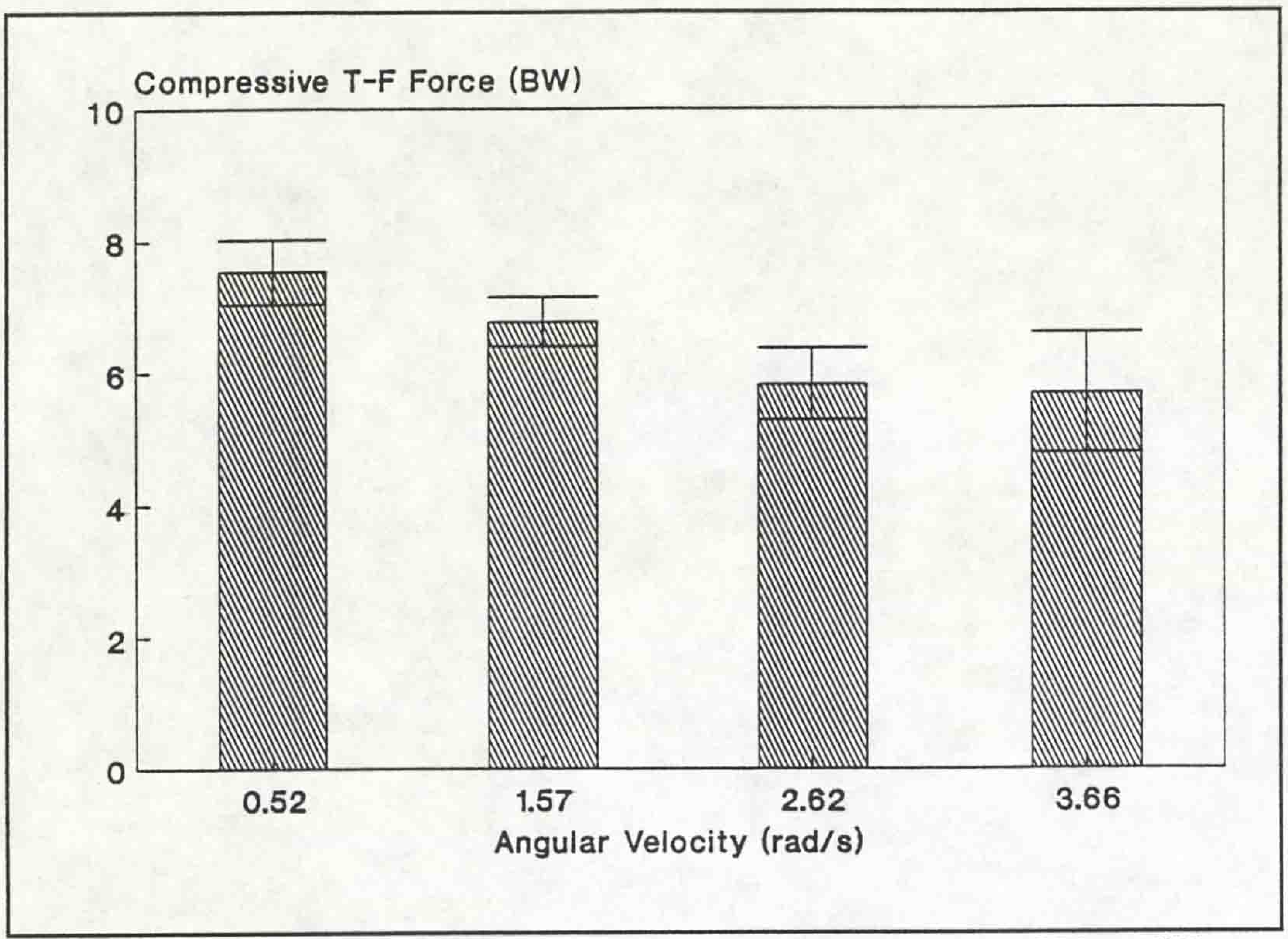

Figure 7.6. Maximum tibiofemoral compressive force (mean ± SD) at different preset angular velocities.

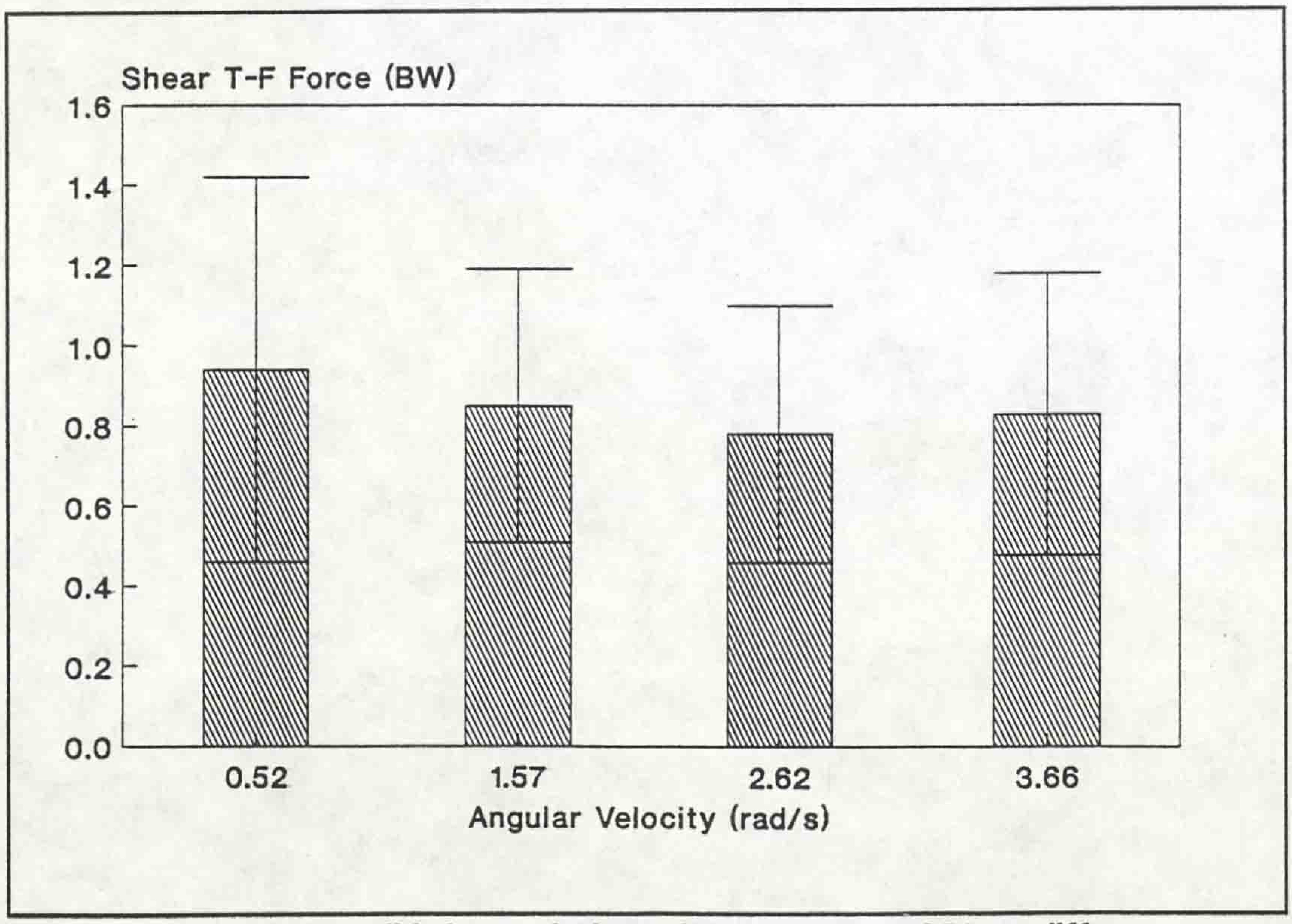

Figure 7.7. Maximum tibiofemoral shear force (mean ± SD) at different preset angular velocities.

Differences between the shear forces at the different angular velocities were not significant ($F_{3,12}=0.64$, $p>0.05$).

The angular position of the maximum moment, muscular force, compressive and shear tibiofemoral forces at the different angular velocities are shown on Table 7.1.

Table 7.1. Knee flexion angle (rad) of the maximum moment, muscular force, compressive and shear tibiofemoral forces at different preset angular velocities.

	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
Resultant Moment	1.12 ±0.16	1.24 ±0.17	1.23 ±0.12	1.02 ±0.14
Muscular Force	1.41 ±0.16	1.40 ±0.17	1.17 ±0.09	1.01 ±0.26
TF Compressive Force	1.41 ±0.16	1.40 ±0.17	1.22 ±0.07	1.01 ±0.25
TF Shear Force	0.54 ±0.12	0.65 ±0.15	0.68 ±0.19	0.61 ±0.21

These differences however were not significant.

DISCUSSION

The muscular and tibiofemoral contact forces during isokinetic knee extension are computed using the two dimensional biomechanical model developed in the present study, from measurements of the resistive dynamometer moment, the kinematic parameters of the movement and joint anatomical data. Inertial forces were included in this model in order to estimate the muscular and TF contact forces throughout the ROM. Previous studies examined isokinetic knee extension using static models (Nisell *et al.*, 1989).

The resistive force developed by the dynamometer throughout the ROM was also included in the model, in order to estimate the T-F contact forces both during isokinetic conditions and during the development of the resistive moment required to decelerate the system.

The moment developed by the knee flexors was not included because the accurate measurement of the force exerted by the knee flexors during isokinetic knee extension is not possible. Furthermore, Osternig *et al.* (1983) reported that the activity of the knee flexors during isokinetic knee extension, estimated from EMG measurements, is minimal.

A standard dynamometer input arm with the attachment point at a distal position on the shank was used in the present study and the effects of a proximal position on the force measurements were not examined. The reduction of the shear joint force using a proximal attachment position (JASA) has been previously reported during both isometric (Johnson, 1982; Lavin and Gross, 1990) and isokinetic knee extension (Nisell *et al.*, 1989). Johnson (1982) reported that during an isometric knee extension of 39.2 Nm at approximately 1.22 rad of knee flexion, the anterior shear force was 188.3 N using a standard input arm with a moment arm of 0.4 m. It was calculated that the anterior shear force was reduced to 174.6 N and 24.5 N using a JASA with moment arms of 0.35 m and 0.15 m respectively. These results however were based on PT moment arm normative data adapted from Smidt (1973) and not on direct anthropometric measurements. A pilot study was also undertaken to determine if this device altered the sagittal alignment of the tibia and femur in a subject with severe anterior laxity. Mediolateral roentgenograms of the subject's knee were taken near terminal extension at three loading conditions:

a) no external load, b) maximal quadriceps activation with a standard input arm positioned distally and c) maximal activation using the JASA and the fulcrum at the most distal position. It was reported that the tibia subluxated approximately 1 mm and 4 mm using the JASA and a standard input arm respectively. The anterior knee force was increased from 69.5 N to 371.8 N using the JASA and the standard input arm respectively.

Translation of the tibia during maximal activation knee extension affects the measurement of the anatomical parameters (TF contact point, PT moment arm, PT-tibial axis angle), required for the measurement of muscular and TF contact forces. A translation of approximately 4 mm using a standard input arm (Johnson, 1982) will underestimate the muscular force for example by 10%. In an attempt to reduce this error in the present study, a manual resistive force was applied to a distal position on the shank during the X-ray process (see Chapter 6).

From a mechanical point of view, the muscular moment during isokinetic knee extension remains the same irrespective of the position of the input arm on the limb and therefore it was suggested that the JASA exerts a stabilizing effect on the knee and reduces anterior shear without altering muscular performance. Recent studies however have shown that muscular moment during isokinetic extension is significantly reduced using the JASA (Taylor and Casey 1986; Nisell *et al.*, 1989). This reduction has been attributed to the increase of the shank-input arm contact force, inducing subliminal pain or other inhibiting influences (Nisell *et al.*, 1989). The reduction of the maximum muscular moment may be appropriate in pathological conditions. During maximum voluntary activation tests in normal conditions however, a distal attachment position of the input arm must be used in order to maximise muscular output.

The muscular and TF contact forces were decreased with increasing angular velocity (Fig. 7.4). Similar findings have been reported by the only other study that examined both muscular and TF contact forces during isokinetic knee extension (Nisell *et al.*, 1989). The joint forces estimated during other dynamic activities range from 2.8-4.9 BW during various walking movements (Morrison, 1969; Ellis *et al.*, 1979; Paul, 1974) to 16-33 BW during jumping and running activities (Bishop, 1977; Smith, 1975; Harrison *et al.*, 1986).

Wickiewicz *et al.* (1984) examined the torque output of the knee extensors and flexors at angular velocities ranging from 0 to 5 $rad \cdot s^{-1}$. The muscular torque was recorded at a constant angle only without considering the maximum torque. The recording angle was 0.52 rad of knee flexion during extension and 1.22 rad during flexion. The torque was converted into force by estimating the moment arm from two cadaveric preparations (Wickiewicz *et al.*, 1983). The maximum force observed was 3675 N and 4631 N for the knee extensors and flexors respectively. The subjects participating in this study were 8 males and 4 females ranging from 20 to 38 years of age but no further anthropometric data of the subjects are available. The magnitude of the muscular forces developed therefore was approximately 6-8 BW. The torque data in the above study however were not corrected for the effect of gravitational and inertial forces that could affect the magnitude of the estimated muscular forces of the knee extensors and flexors. Nisell *et al.* (1989) estimated the tibiofemoral forces developed during isokinetic knee extension. The maximum torque recorded was 181 Nm at 1.13 rad of knee flexion during the fast speed test and 284 Nm at 1.22 rad of knee flexion during the slow speed test. The maximum torque at the fast speed was significantly reduced with the resistance pad placed proximally although it was recorded at the same joint position. The magnitudes of the tibiofemoral compressive and patellar tendon forces were similar throughout the ROM with a maximum of approximately 9 BW recorded at 1.22 rad of knee flexion. The magnitude of these forces however was not reduced significantly with the resistance pad placed in a proximal position. The maximum tibiofemoral shear force ranged from 0.5 BW in a posterior direction to 1 BW in an anterior direction and remained approximately constant from 1.47 to 0 rad of knee flexion. The anteriorly directed tibiofemoral shear force was decreased considerably by placing the resistance pad in a proximal position.

CONCLUSIONS

This study examined the muscle and tibiofemoral (TF) contact forces during isokinetic knee extension at angular velocities ranging from 0.52 $rad \cdot s^{-1}$ to 3.66 $rad \cdot s^{-1}$, using a two dimensional biomechanical model. Within the limitations of the present study it can be concluded that the muscular, compressive TF contact forces are reduced with increasing angular velocity of movement. These results also indicate that the forces developed during maximal isokinetic knee extension are significantly reduced relative to other dynamic activities and therefore isokinetic dynamometry is a safe and effective method for muscle function assessment, training and rehabilitation.

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SUMMARY

Isokinetic dynamometry is the assessment of dynamic muscle function during isolated joint movements performed at constant angular velocity. The optimal muscular loading (resistive moment equivalent to muscular moment) and control of the angular velocity during isokinetic movements resulted in widespread applications in the areas of muscle testing and rehabilitation. The assessment of muscle function from the resistive moment developed by the isokinetic dynamometer however, is affected by several mechanical and methodological problems. The present study considers the main problems of isokinetic dynamometry that influence muscular performance, measurement of isokinetic parameters, assessment of muscle function and development of dynamic joint forces.

The effects of visual feedback on isokinetic maximum torque and reciprocal muscle group ratio of the knee extensors and flexors at a slow (1.04 $rad \cdot s^{-1}$) and a fast (3.14 $rad \cdot s^{-1}$) angular velocity of movement were examined using the gravity corrected resistive moment of the dynamometer as the real-time visual feedback source. This elicited a significant increase in the maximum torque output of both muscle groups at the slow angular velocity. There was no significant improvement at the fast angular velocity. The knee flexor/extensor torque ratio was not effected by visual feedback or angular velocity of movement. It was concluded that visual feedback can improve muscular torque output under certain restrictions on velocity and range of movement. Visual feedback is therefore essential during maximum voluntary activation tests in isokinetic dynamometry.

The angular velocity development and maintenance during isokinetic knee extension was examined at preset angular velocities of 0.52, 1.57, 2.62

and 3.67 rad.s^{-1} using a computerised AKRON isokinetic dynamometer. Angular velocity was determined from differentiation of the angular position-time data after optimal smoothing using a low pass digital filter. Maximum torque was determined from the part of the movement with the angular velocity within $\pm$ 10% of the preset velocity. The mean maximum torque ranged from 264.7 ($\pm$43.8) Nm at 0.52 rad·s^{-1} to 198.8 ($\pm$27.9) Nm at 3.67 rad·s^{-1}. During the initial acceleration period the velocity of the limb exceeded the preset velocity by an average of 145%, 44%, 29% and 18% at the four preset velocities respectively. The constant velocity period ranged from 63.7% at 0.52 rad.s^{-1} to 40.3% of the total movement at 3.67 rad.s^{-1}. These results indicate that the angular velocity during isokinetic movements using isokinetic dynamometers, fluctuates even after the initial acceleration period and appropriate correction methods are required before the measurement of isokinetic parameters and the assessment of dynamic muscle function.

The measurement of dynamic joint forces during isokinetic knee extension requires the determination of a biomechanical model of the knee. The anatomical parameters required for this model are patellar tendon (PT) moment arm, tibial plateau-tibial axis angle and PT tendon-tibial axis angle. These parameters were determined *in vivo* during knee extension using videofluoroscopy. Image distortion in videofluoroscopy however requires appropriate non-linear correction methods, in order to obtain accurate biomechanical quantitative measurements. For this purpose an algorithm for two-dimensional coordinate reconstruction and non-linear distortion correction using a polynomial method was developed. The measurement error obtained using an image intensifier - video system was 0.246 $\pm$0.111 mm over a 180 mm x 180 mm field of view. Five males (mean age 20.8 $\pm$years, mass 79.2

$\pm$7.2 kg and height 179 $\pm$ 3.2 cm) without knee joint injury history participated in the study. The mean PT moment arm at full extension was 33.81 $\pm$3.44 mm, increased to a maximum of 39.87 $\pm$2.4 mm at 0.78 rad of knee flexion and decreased to 33.63 $\pm$4.01 mm at 1.57 rad. The PT-tibial plateau angle was 1.96 $\pm$0.12 rad at full extension and decreased linearly to 1.53 $\pm$0.05 rad at 1.57 rad of knee flexion. The mean angle between the tibial plateau and the tibial long axis was 1.48 $\pm$0.04 rad.

The muscular and tibiofemoral contact forces during isokinetic knee extension were examined at angular velocities ranging from 0.52 rad$\cdot$s^{-1} to 3.66 rad$\cdot$s^{-1}. The maximum moment (mean $\pm$SD) ranged from 226.20 $\pm$39.52 Nm at 0.52 rad$\cdot$s^{-1} to 166 $\pm$27.56 Nm at 3.66 rad$\cdot$s^{-1}. These differences were significant ($F_{3,12}=17.9$, $p<0.05$) and subsequent *post hoc* tests revealed that the significant differences were between the moments at 0.52 rad$\cdot$s^{-1} and 2.62-3.66 rad$\cdot$s^{-1}. The maximum muscular force ranged from 7.55 $\pm$0.49 times body weight (BW) at 0.52 rad$\cdot$s^{-1} to 5.72 $\pm$0.94 BW at 3.66 rad$\cdot$s^{-1}. The compressive tibiofemoral force ranged from 7.53 $\pm$0.49 BW at 0.52 rad$\cdot$s^{-1} to 5.68 $\pm$0.91 BW at 3.66 rad$\cdot$s^{-1} and the shear tibiofemoral force from 0.94 $\pm$0.48 BW to 0.83 $\pm$0.35 BW respectively. These differences were significant for both maximum muscular force ($F_{3,12}=13.7$, $p<0.05$) and compressive tibiofemoral force ($F_{3,12}=13.57$, $p<0.05$). Differences between the shear forces at the different angular velocities were not significant ($F_{3,12}=0.64$, $p>0.05$).

These results indicate that the forces developed during maximal isokinetic knee extension are significantly reduced relative to other dynamic activities and therefore isokinetic dynamometry is a safe and effective method for muscle function assessment, training and rehabilitation, provided that appropriate correction methods for the mechanical and methodological errors are implemented.

LIST OF FIGURES AND TABLES Page

APPENDIX I

COMPUTER PROGRAMS

INTRODUCTION

The development of specific software was required for data reduction and analysis in the studies presented in this thesis. Software was developed by the author using Borland's Turbo Pascal (version 5.5) on an OPUS PC-V (Intel 82386 based) microcomputer under MS-DOS (version 3.3) operating system. Turbo Pascal was used because of its modular programming nature and the integrated development environment allowing efficient editing, compilation and debugging of computer programs.

The following Turbo Pascal units were developed for use by the main programs:

Axis	A complete set of procedures for data display and presentation in graphical form. Arrays or single points are displayed on the monitor using different attributes. Hard copies of the display using a Hewlett-Packard plotter are also available.
Mouse	A complete set of procedures for handling the basic operations of the graphics cursor including initialisation, shape determination and input of video coordinates.
Gcr	A complete set of procedures for text input-output in a graphics environment. This unit was adapted from Weiskamp *et al.* (1989).
Bspline	A procedure for the computation of the coefficients of a B-Spline based on the algorithm presented by Rankin (1989).
Smspline	A procedure for the computation of the coefficients of a cubic spline based on the algorithm presented by Reinsch (1967).

Svd A singular value decomposition algorithm implemented for the solution of simultaneous linear equation systems. This procedure is based on the algorithm presented by Press *et al.* (1989).

Serial A complete set of procedures for handling the serial port of the computer including initialisation, input and output of data (Borland, 1989).

The following Turbo Pascal programs were developed for data collection, analysis and presentation:

Digital A program for data smoothing using a Butterworth digital filter and computation of kinematic parameters using numerical differentiation.

Video A program for coordinate reconstruction and correction of optical distortion of image intensifier-video systems using the method described in Chapter 5.

Kinematic A program for the measurements of the knee joint parameters described in Chapter 6 from video X-ray records.

Knee A program for smoothing and interpolation of the knee joint parameters.

Kinetic A program for the computation of the muscular and intersegmental forces using the method described in Chapter 7.

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```
unit axis;

{$E+}
{$N+}

interface

type
RealArray2=array[1..500,1..2] of real;
IntegerArray2=array[1..500,1..2] of integer;
axislabel=string[30];
WrkString=string [11];

var
Xdiv,Ydiv,Xmin,Xmax,Ymin,Ymax,Xrange,Yrange,mfX,mfY:real;
BaseX,BaseY,Dx,Dy:integer;
GrafBase: word;
com1:text;

procedure World(Xl,Xh,Yl,Yh:real);
procedure Findworld( var Data:RealArray2;  n1,n2:integer; sizeX,sizeY:real);
procedure PlotWorld;
procedure Drawarray( var Data:RealArray2; n1,n2,code,size:integer;
                        Display,Plot:boolean);
procedure DrawLine(X1,Y1,X2,Y2:real);
procedure DrawPoint(X1,Y1:real);

implementation
uses dos,printer,graph,Serial;

procedure World(Xl,Xh,Yl,Yh:real);
var
viewport:viewporttype;
begin
GetViewSettings(ViewPort);
with ViewPort do
begin
    BaseX:=x1+75;
    BaseY:=y2-35;
    Dx:=(x2-x1)-99;
    Dy:=(y2-y1)-40;
end;
Xrange:=Xh-Xl;
Yrange:=Yh-Yl;
Xmin:=Xl;
Xmax:=Xh;
Ymin:=Yl;
Ymax:=Yh;
Xdiv:=(Xh-Xl)/Dx;
Ydiv:=(Yh-Yl)/Dy;
end; {World}

procedure Findworld( var Data:RealArray2;  n1,n2:integer; sizeX,sizeY:real);
var
```

```
np,i:integer;
factorX,factorY:real;
begin
np:=n2-n1+1;
Ymax:=Data[n1,2];
for i:=n1+1 to n2 do
   if Data[i,2]>Ymax then Ymax:=Data[i,2];
Ymin:=Ymax;
for i:=n1 to n2 do
   if Data[i,2]<Ymin then Ymin:=Data[i,2];
if (Ymin>0) and ((Ymax-Ymin)*0.1>=Ymin) then Ymin:=0;

factorY:=((Ymax-Ymin)*(sizeY-1))/2;
Ymin:=Ymin-factorY;
Ymax:=Ymax+factorY;

Xmax:=Data[n1,1];
for i:=n1+1 to n2 do
   if Data[i,1]>Xmax then Xmax:=Data[i,1];
Xmin:=Xmax;
for i:=n1 to n2 do
   if Data[i,1]<Xmin then Xmin:=Data[i,1];
if (Xmin>0) and ((Xrange)*0.1>=Xmin) then Xmin:=0;

factorX:=((Xmax-Xmin)*(sizeX-1))/2;
Xmin:=Xmin-factorX;
Xmax:=Xmax+factorX;

World(Xmin,Xmax,Ymin,Ymax);
end; {FindWorld}

procedure PlotWorld;
begin
mfX:=1;
repeat
mfX:=mfX*10;
until Xrange*mfX/10>10;
mfX:=mfX/10;
mfY:=1;
repeat
mfY:=mfY*10;
until Yrange*mfY/10>10;
mfY:=mfY/10;
end;

procedure Drawarray( var Data:RealArray2; n1,n2,code,size:integer;
                     Display,Plot:boolean);
var
Pixel:IntegerArray2;
np,i:integer;
SymbolSize:Real;

procedure Cross(x,y,size:integer);
begin
   Line(x,y-(1+size),x,y+(1+size));
```

```
    Line(x-(1+size),y,x+(1+size),y);
end; {Cross}

begin
    np:=n2-n1+1;
    if Display then
    begin
        for i:=1 to np do
        begin
            Pixel[i,1]:=BaseX+round((Data[n1-1+i,1]-Xmin)/XDiv);
            Pixel[i,2]:=BaseY-round((Data[n1-1+i,2]-Ymin)/YDiv);
            case abs(code) of
            1:PutPixel(Pixel[i,1],Pixel[i,2],GetColor);
            2:cross(Pixel[i,1],Pixel[i,2],size);
            3:circle(Pixel[i,1],Pixel[i,2],1+size);
            4:rectangle(Pixel[i,1]-size,Pixel[i,2]-size,Pixel[i,1]+size,Pixel[i,2]+size);
            5:begin
                 Line(Pixel[i,1]-size,Pixel[i,2]-size,Pixel[i,1]+size,Pixel[i,2]+size);
                 Line(Pixel[i,1]+size,Pixel[i,2]-size,Pixel[i,1]-size,Pixel[i,2]+size);
              end;
            6:bar3d(Pixel[i,1]-size,BaseY,Pixel[i,1]+size,Pixel[i,2],0,true);
        end;
    end;
    if (code>0) or (code=0) then DrawPoly(np,Pixel);
end; {Display}

if Plot then
begin
    PlotWorld;
    for i:=n1 to n2 do
    begin
        Data[i,1]:=Data[i,1]*mfX;
        Data[i,2]:=Data[i,2]*mfY;
    end;
Xrange:=Xrange*mfX;
Yrange:=Yrange*mfY;
AssignSerial(Com1,0,231);
Rewrite(Com1);
Writeln(com1,'IN;');
Writeln(com1,'SC',Round(Xmin*mfX),',',Round(Xmax*mfX),',',
            Round(Ymin*mfY),',',Round(Ymax*mfY),';');
Writeln(com1,'IP2000,2000,8000,7000;');
Writeln(com1,'PT0.1;SP1;VS5;');
Writeln(com1,'SR',0.75*(Size+1):9:4,',',1.5*(Size+1):9:4,';');
SymbolSize:=0.002;
for i:=n1 to n2 do
begin
    case abs(code) of
    1:Writeln(com1,'PA',Data[i,1]:9:4,',',Data[i,2]:9:4,';PD;PU;');
    2:begin
            Writeln(com1,'PA',Data[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                 Data[i,2]:9:4,';PD;');
            Writeln(com1,'PA',Data[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                 Data[i,2]:9:4,';PU;');
            Writeln(com1,'PA',Data[i,1]:9:4,',',
```

```
                Data[i,2]-(Yrange)*SymbolSize*Size:9:4,';PD;');
            Writeln(com1,'PA',Data[i,1]:9:4,',',
                Data[i,2]+(Yrange)*SymbolSize*Size:9:4,';PU;');
        end;
    3:begin
            Writeln(com1,'PA',Data[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]:9:4,';PD;');
            Writeln(com1,'PA',Data[i,1]:9:4,',',
                Data[i,2]+(Yrange)*SymbolSize*Size:9:4,';');
            Writeln(com1,'PA',Data[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]:9:4,';');
            Writeln(com1,'PA',Data[i,1]:9:4,',',
                Data[i,2]-(Yrange)*SymbolSize*Size:9:4,';');
            Writeln(com1,'PA',Data[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]:9:4,';PU;');
        end;
    4:begin
            Writeln(com1,'PA',Data[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]+(Yrange)*SymbolSize*Size:9:4,';PD;');
            Writeln(com1,'PA',Data[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]+(Yrange)*SymbolSize*Size:9:4,';');
            Writeln(com1,'PA',Data[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]-(Yrange)*SymbolSize*Size:9:4,';');
            Writeln(com1,'PA',Data[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]-(Yrange)*SymbolSize*Size:9:4,';');
            Writeln(com1,'PA',Data[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]+(Yrange)*SymbolSize*Size:9:4,';PU;');
        end;
    5:begin
            Writeln(com1,'PA',Data[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
               Data[i,2]+(Yrange)*SymbolSize*Size:9:4,';PD;');
            Writeln(com1,'PA',Data[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]-(Yrange)*SymbolSize*Size:9:4,';PU;');
            Writeln(com1,'PA',Data[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]-(Yrange)*SymbolSize*Size:9:4,';PD;');
            Writeln(com1,'PA',Data[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                Data[i,2]+(Yrange)*SymbolSize*Size:9:4,';PU;');
        end;
    end;
    if ((code>0) or (code=0)) and (i<n2) then
    Writeln(com1,'PA',Data[i,1]:9:4,',',Data[i,2]:9:4,';PD',
               Data[i+1,1]:9:4,',',Data[i+1,2]:9:4,';');
end;
Writeln(com1,'PU;PA0,0;SP0;VS5;');

Close(Com1);
for i:=n1 to n2 do
begin
    Data[i,1]:=Data[i,1]/mfX;
    Data[i,2]:=Data[i,2]/mfY;
end;
Xrange:=Xrange/mfX;
Yrange:=Yrange/mfY;
end; {Plot}
```

```
end; {DrawArray}

Procedure DrawLine(X1,Y1,X2,Y2:real);
begin
Line(BaseX + round((X1-Xmin)/XDiv),
     BaseY-round((Y1-Ymin)/YDiv),
     BaseX + round((X2-Xmin)/XDiv),
     BaseY-round((Y2-Ymin)/YDiv));
end; {DrawLine}

Procedure DrawPoint(X1,Y1:real);
begin
PutPixel(BaseX + round((X1-Xmin)/XDiv),
        BaseY-round((Y1-Ymin)/YDiv),GetColor);
end; {DrawPoint}
end.
```

```
unit mouse;

interface

uses dos,graph;

type
st=string[16];
var
reg:registers;
mask:array [0..1,0..15] of word;
i,k,l,w,mul:integer;
preint:pointer;
s:st;
procedure MouseInit;
procedure ShowMouse;
procedure HideMouse;

implementation

function BinToDec(s:st):integer;
begin
w:=0;
for i:=1 to length(s) do
begin
if s[i]='1' then
begin
mul:=1;
for k:=1 to 16-i do
mul:=mul*2;
end
else
mul:=0;
w:=w+mul;

end;
bintodec:=w;
end;{procedure}

procedure MouseInit;
begin
mask[ 0,0]:=bintodec('1111111111111111');
mask[ 0,1]:=bintodec('1111111111111111');
mask[ 0,2]:=bintodec('1111111111111111');
mask[ 0,3]:=bintodec('1111111111111111');
mask[ 0,4]:=bintodec('1111111111111111');
mask[ 0,5]:=bintodec('1111111111111111');
mask[ 0,6]:=bintodec('1111111111111111');
mask[ 0,7]:=bintodec('1111111111111111');
mask[ 0,8]:=bintodec('1111111111111111');
mask[ 0,9]:=bintodec('1111111111111111');
mask[0,10]:=bintodec('1111111111111111');
mask[0,11]:=bintodec('1111111111111111');
mask[0,12]:=bintodec('1111111111111111');
mask[0,13]:=bintodec('1111111111111111');
```

```
mask[0,14]:=bintodec('1111111111111111');
mask[0,15]:=bintodec('1111111111111111');

mask[ 1,0]:=bintodec('0000000000000000');
mask[ 1,1]:=bintodec('0000000000000000');
mask[ 1,2]:=bintodec('0000000000000000');
mask[ 1,3]:=bintodec('0000000000000000');
mask[ 1,4]:=bintodec('0000000000000000');
mask[ 1,5]:=bintodec('0000000000000000');
mask[ 1,6]:=bintodec('0000001110000000');
mask[ 1,7]:=bintodec('0000001110000000');
mask[ 1,8]:=bintodec('0000001110000000');
mask[ 1,9]:=bintodec('0000000000000000');
mask[1,10]:=bintodec('0000000000000000');
mask[1,11]:=bintodec('0000000000000000');
mask[1,12]:=bintodec('0000000000000000');
mask[1,13]:=bintodec('0000000000000000');
mask[1,14]:=bintodec('0000000000000000');
mask[1,15]:=bintodec('0000000000000000');


with reg do
begin
AX:=9;
BX:=7;
CX:=7;
DX:=ofs(mask);
ES:=seg(mask);
end;
Intr(51,reg);

with reg do
begin
AX:=15;
CX:=40;
DX:=40;
end;
Intr(51,reg);

reg.ax:=1;
intr(51,reg);


reg.ax:=19;
reg.dx:=768;
intr(51,reg);

reg.ax:=7;
reg.cx:=0;
reg.dx:=639;
intr(51,reg);

reg.ax:=8;
reg.cx:=0;
reg.dx:=479;
```

```
intr(51,reg);

end; {procedure}

procedure ShowMouse;
begin
reg.ax:=1;
intr(51,reg);
end;

procedure HideMouse;
begin
reg.ax:=2;
intr(51,reg);
end;

end.
```

```
unit GCrt;

{The Turbo Pascal code for the following procedures was modified from Weiskamp et
al. (1989) (See References in page 128)}

GTextX,GTextY:integer;
function Is(Num: longint): string;
function Rs(n: real; width, decimals: integer): string;
function Xg(X: Integer): Integer;
function Yg(Y: Integer): Integer;
procedure GWrite(S: string);
procedure GWriteXY(x, y: integer; S: string);
procedure GDelLineXY(x, y: integer);
procedure GWriteCh(ch: char);
function GReal(var Num: real): boolean;
function GInt(var Num: Integer): boolean;
function GReadStr(var S: string): boolean;
function GReadCh(var ch: char): boolean;
```

```
unit Spline;

interface

uses axis;

type
XYArray=array[1..200,1..2] of Real;
var
i,k,n,gd,gm:integer;
t:Real;
BSplineCoef:record
          a,b,c,d:XYArray;
          end;

Procedure BSpline(var r,s:XYArray;n,d:integer);

implementation

Procedure BSpline(var r,s:XYArray;n,d:integer);
begin
  for i:=2 to n-4 do
  begin
  with BSplineCoef do
  begin
    a[i-1,1]:=-r[i-1,1]+3*r[i,1]-3*r[i+1,1]+r[i+2,1];
    b[i-1,1]:=3*r[i-1,1]-6*r[i,1]+3*r[i+1,1];
    c[i-1,1]:=-3*r[i-1,1]+3*r[i+1,1];
    d[i-1,1]:=r[i-1,1]+4*r[i,1]+r[i+1,1];
    a[i-1,2]:=-r[i-1,2]+3*r[i,2]-3*r[i+1,2]+r[i+2,2];
    b[i-1,2]:=3*r[i-1,2]-6*r[i,2]+3*r[i+1,2];
    c[i-1,2]:=-3*r[i-1,2]+3*r[i+1,2];
    d[i-1,2]:=r[i-1,2]+4*r[i,2]+r[i+1,2];
  end;
  end;
  for i:=1 to  (n-5)*d do
  begin
  t:=((i-1) mod d)*1/d;
  k:=(i-1) div d  + 1;
  with BSplineCoef do
  begin
    s[i,1]:=(a[k,1]*t*t*t+b[k,1]*t*t+c[k,1]*t+d[k,1])/6;

    s[i,2]:=(a[k,2]*t*t*t+b[k,2]*t*t+c[k,2]*t+d[k,2])/6;
  end;
  end;
end;

end.
```

```
unit smspl;

{$E+}
{$N+}

interface

uses dos,printer,graph,axis;

const
RealSize=6;
label draw,size,start,smsize,telos;
type
Array1=array[0..2] of real;
Array2=array[0..2,0..2] of real;
ArrayPointer1=^Array1;
ArrayPointer2=^Array2;
var
gdriver,gmode,nc:integer;
ch:char;
smooth:boolean;

procedure SplineFunction(var x,y,a,b,c,d:ArrayPointer1; n:integer; sf:Real);
procedure SplFit(var x,y,a,b,c,d:ArrayPointer1; var xp,yp:Real; n:integer; sf:Real);
Function splSF(var x,y,a,b,c,d:ArrayPointer1; Nmax:integer; Variance:Real):Real;

implementation

procedure SplineFunction;
label iteration,fin,finish;
var
xs,ys,r,r1,r2,t,t1,u,v:ArrayPointer1;
e,f,f2,g,h,p,sdy,di,de,temp,ws,sd,ad:real;
n1,n2,m1,m2,i,j,k,l,kmax,st:integer;
begin
GetMem(r,(n+2)*RealSize);
GetMem(r1,(n+2)*RealSize);
GetMem(r2,(n+2)*RealSize);
GetMem(t,(n+2)*RealSize);
GetMem(t1,(n+2)*RealSize);
GetMem(u,(n+2)*RealSize);
GetMem(v,(n+2)*RealSize);

n1:=1; n2:=n;
m1:=n1-1; m2:=n2+1;
r^[m1]:=0;r^[n1]:=0;r1^[n2]:=0;r2^[n2]:=0;r2^[m2]:=0;u^[m1]:=0;
u^[n1]:=0;u^[n2]:=0;u^[m2]:=0;p:=0;

m1:=n1+1; m2:=n2-1;
h:=x^[m1]-x^[n1];

f:=(y^[m1]-y^[n1])/h;

for i:=m1 to m2 do
begin
```

```
g:=h;
h:=x^[i+1]-x^[i];
e:=f;
f:=(y^[i+1]-y^[i])/h;
a^[i]:=f-e;
t^[i]:=2*(g+h)/3;
t1^[i]:=h/3;
r2^[i]:=1/g;
r^[i]:=1/h;
r1^[i]:=-1/g-1/h;
end;

for i:= m1 to m2 do
begin
b^[i]:=sqr(r^[i])+sqr(r1^[i])+sqr(r2^[i]);
c^[i]:=(r^[i]*r1^[i+1])+(r1^[i]*r2^[i+1]);
d^[i]:=r^[i]*r2^[i+2];
end;

f2:=-sf;

iteration:

for i:=m1 to m2 do
begin
r1^[i-1]:=f*r^[i-1];
r2^[i-2]:=g*r^[i-2];
r^[i]:=1/(p*b^[i]+t^[i]-f*r1^[i-1]-g*r2^[i-2]);
u^[i]:=a^[i]-r1^[i-1]*u^[i-1]-r2^[i-2]*u^[i-2];
f:=p*c^[i]+t1^[i]-h*r1^[i-1];
g:=h;
h:=d^[i]*p;
end;

for i:= m2 downto m1 do
begin
u^[i]:=r^[i]*u^[i]-r1^[i]*u^[i+1]-r2^[i]*u^[i+2];
e:=0;
h:=0;
end;

for i:=n1 to m2 do
begin
g:=h;
h:=(u^[i+1]-u^[i])/(x^[i+1]-x^[i]);
v^[i]:=(h-g);
e:=e+v^[i]*(h-g);
end;

g:=-h;
v^[n2]:=g;
e:=e-g*h;
g:=f2;
f2:=e*p*p;
```

```
if (f2>=sf) or (f2<=g) then
goto fin;

f:=0;
h:=(v^[m1]-v^[n1])/(x^[m1]-x^[n1]);

for i:=m1 to m2 do
begin
g:=h;
h:=(v^[i+1]-v^[i])/(x^[i+1]-x^[i]);
g:=h-g-r1^[i-1]*r^[i-1]-r2^[i-2]*r^[i-2];
f:=f+g*r^[i]*g;
r^[i]:=g;
end;

h:=e-p*f;
if h=0 then
goto fin
else
p:=p+(sf-f2)/((sqrt(sf/e)+p)*h);
goto  iteration;

fin:
for i:=n1 to n2 do
begin
a^[i]:=y^[i]-p*v^[i];
c^[i]:=u^[i];
end;

for i:=n1 to m2 do
begin
h:=x^[i+1]-x^[i];
d^[i]:=(c^[i+1]-c^[i])/(3*h);
b^[i]:=(a^[i+1]-a^[i])/h-(h*d^[i]+c^[i])*h;
end;

FreeMem(r,(n+2)*RealSize);
FreeMem(r1,(n+2)*RealSize);
FreeMem(r2,(n+2)*RealSize);
FreeMem(t,(n+2)*RealSize);
FreeMem(t1,(n+2)*RealSize);
FreeMem(u,(n+2)*RealSize);
FreeMem(v,(n+2)*RealSize);
end; {SplineFunction}

procedure SplFit;
var
klo,kup,k:integer;
h:real;
begin
klo:=1;
kup:=n;
while kup-klo>1 do
begin
k:=(kup+klo) div 2;
```

```
if x^[k] > xp then kup:=k else klo:=k
end;
h:=xp-x^[klo];
k:=klo;
yp:=a^[k]+(b^[k]*h)+(c^[k]*(sqr(h)))+
(d^[k]*(h*h*h));


end; {SplFit}

Function splSF;
var
ys:ArrayPointer1;
sumres:RealArray2;
i,k,opt:integer;
begin

GetMem(ys,(Nmax+2)*RealSize);
i:=0;
Repeat
i:=i+1;
SplineFunction(x,y,a,b,c,d,Nmax,i);

sumres[i,1]:=i;
sumres[i,2]:=0;
for k:=1 to Nmax-1 do
  begin
  SplFit(x,y,a,b,c,d,x^[k],ys^[k],Nmax,i);
  Sumres[i,2]:=sumres[i,2]+sqrt(sqr(y^[k]-ys^[k]));
  end;
  Sumres[i,2]:=sumres[i,2]/(Nmax-2);
writeln(lst,i:5,abs(SumRes[i,2]-Variance):10:2,abs(SumRes[i-1,2]-Variance):10:2);
until ((i>1) and (abs(SumRes[i,2]-Variance)>abs(SumRes[i-1,2]-Variance))) or
(i=Nmax*2);
SplSF:=i;

FreeMem(ys,(Nmax+2)*RealSize);

end; {splsf}
end.
```

```
unit svdext;

{The Turbo Pascal code for the following procedures was modified from Press et al.
(1989) (See References in page 128)}

procedure svdcmp(var a: ExtArrayNXM;
                   n,m: integer;
                 var w: ExtArrayNEq;
                 var v: ExtArrayNXM);
procedure svbksb(var u: ExtArrayNXM;
                 var w: ExtArrayNEq;
                 var v: ExtArrayNXM;
                   n,m: integer;
                 var b: ExtArrayNEq;
                 var x: ArrayNEq);
procedure svdfit(var a: ExtArrayNXM;
                 var x: ArrayNEq;
                 var b: ExtArrayNEq;
                   n,m: integer;
                 var u: ExtArrayNXM;
                 var w: ExtArrayNEq;
                 var v: ExtArrayNXM);
```

```
unit Serial;

{The Turbo Pascal code for the following procedures was modified from Borland (1989)
(See References in page 128)}

procedure AssignSerial(var F: Text; Port,Params: word);
procedure Readln(var F: Text; var Str: String);
```

```
program digital;

{$E+}
{$N+}

uses dos,printer,graph,axis,mouse;

type
ArrayPointer=^DataArray;
DataArray=array[1..2] of real;
label start,finish;
var
n,i,j,k,ii,m,iteration,gdriver,gmode,Dir,Dis,Beg,ExtFlx,
loop,movement,step,cn,fin,initbeg,initfin,DisplayStep,ExtPoints,sum:integer;
Data,FData,Temp,Trend:ArrayPointer;
Acc,Vel,Pos,tor:Realarray2;
c1,c2,c3,c4,c5,z,z1,z2,sv,a,b,wc,f,dt,cutoff,pi,sd,
AcMax,VeMax,VeConst,PresetVel,TorMax,TorOv,RomMin,RomMax,
TorVel,OvVel,TorTime,OvTime,TorPos,OvPos:real;
sumY,sumAS,sumA:array [1..100] of real;
sumcn,base,ssdep,ssbg,sswg,sum10A,sum10Y,sum10AS,CY,fdep,cutoffstep:real;
datai:file of integer;
datar:file of real;
name,filename:string[20];
violation:boolean;
Regs:Registers;
Ch:char;
ExportFile:Text;

procedure DigitalFilter(var Data:ArrayPointer; cf:extended);
label filter;
begin
cf:=cf/0.802;
violation:=false;
sv:=1.0/(cf*dt);
if sv<4 then
begin
violation:=true;
end;
pi:=4*arctan(1);
z:=pi*cf*dt;
z1:=sin(z);
z2:=cos(z);
wc:=z1/z2;
a:=2.0*wc*sqrt(0.5);
b:=0.5*a*a;

c1:=b/(1.0+a+b);
c2:=2*c1;
c3:=c1;
c4:=2.0*(1.0-b)/(1.0+a+b);
c5:=(a-b-1.0)/(1.0+a+b);

for i:=1 to n do
begin
```

```
Temp^[i]:=Data^[i];
end;
for j:=1 to 2 do
begin
FData^[1]:=Temp^[1];
FData^[2]:=Temp^[2];

for i:=3 to n do
begin
FData^[i]:=c1*Temp^[i]+c2*Temp^[i-1]+c3*Temp^[i-2]+c4*FData^[i-1]+c5*FData^[i-2];
end;
k:=n+1;
for i:=1 to n do
begin
Temp^[i]:=FData^[k-i];
end;
end;

for i:=1 to n do
begin
FData^[i]:=Temp^[i];


end;

cf:=0.802*cf;
end; {filter}


procedure MouseInput(var X,Y:integer);
begin
repeat
reg.ax:=3;
Intr(51,reg);
until reg.bx and 1=1 ;

x:=reg.cx;
y:=reg.dx;
repeat
reg.ax:=3;
Intr(51,reg);
until reg.bx and 1=0;
end;
procedure MoveXY(x,y:byte);
begin
with reg do
begin
ah:=2;dl:=x;dh:=y;bh:=0;intr(16,reg);
end;
end;
begin

write('DATA INPUT FROM FILE <1> OR KEYBOARD <2> :');readln(i);
if i=1 then
begin
```

```
write('FILE NAME > ');readln(name);
write('DATA FORMAT: 1=REAL  0=INTEGER >');readln(m);
case m of
1:begin
assign(datar,name);
reset(datar);
Getmem(Data,(filesize(datar)+1+30)*6);
Getmem(FData,(filesize(datar)+1+30)*6);
Getmem(Temp,(filesize(datar)+1+30)*6);

for i:=1 to filesize(datar) do
begin
  seek(datar,i-1);
  read(datar,Data^[i]);
  writeln(Data^[i]:10:5);
end;

n:=filesize(datar);
close(datar);

end;
0: begin
     assign(datai,name);
     reset(datai);


     gdriver:=detect;
     Initgraph(gdriver,gmode,'');
     mouseinit;
     beg:=0;
     fin:=filesize(datai) div 3-1;
     repeat
     ClearDevice;
     m:=0;
     for i:=Beg to Fin do
     begin
       seek(datai,i*3);
       read(datai,Dir);
       seek(datai,i*3+1);
       read(datai,Dis);
       DisplayStep:=(Fin-Beg) div GetMaxX;
       if DisplayStep<2 then DisplayStep:=2;
       PutPixel(m div DisplayStep,GetMaxY div 2-Dis,15);
       PutPixel(m div DisplayStep,GetMaxY div 2-Dir,15);
       m:=m+1;
     end;
     Fin:=Beg;
     MouseInput(j,k);beg:=Beg+j*DisplayStep;
     MouseInput(j,k);fin:=fin+j*DisplayStep;

     readln(ch);
     until Upcase(ch)='N';
     Closegraph;
     initbeg:=beg;
     initfin:=fin;
```

```
    ExtPoints:=30;

    Getmem(Data,500*6);
    Getmem(FData,500*6);
    Getmem(Temp,500*6);
    Getmem(Trend,500*6);
    step:=0;
    repeat
    step:=step+1;
    beg:=initbeg-ExtPoints*step;
    fin:=initfin+ExtPoints*step;
    i:=0;
    repeat
    seek(datai,(Beg+i*step)*3+1);
    read(datai,Dis);

    seek(datai,(Beg+i*step)*3);
    read(datai,Dir);
    i:=i+1;
    data^[i]:=Dis/2;
    Tor[i,2]:=Dir/1;
    until Beg+i*step>Fin;
    n:=i;
    ExtFlx:=i;
    until n=filesize(datai);
    close(datai);
    end;
end; {case}
end
else
begin
writeln('INPUT NUMBER OF POINTS  N    ');
write('N MUST BE AN EVEN NUMBER AND Y(1)=Y(N) :');readln(n);
Getmem(Data,(n+30)*6);
Getmem(FData,(n+30)*6);
Getmem(Temp,(n+30)*6);

for i:=1 to n do
begin
write('Y(',i,')=');readln(Data^[i]);
end;
write('SAVE DATA TO A FILE (1=YES 0=NO) :');readln(m);
if m=1 then
begin
write('NAME OF FILE :');readln(name);
assign(datar,name);
rewrite(datar);
for i:=1 to n do
write(datar,Data^[i]);
close(datar);
end;
end;
if n>500 then n:=500;
```

```
cutoffstep:=0.1;
cutoff:=1;
loop:=0;
n:=ExtFlx;
for i:=1 to n do
begin
Pos[i,2]:=data^[i];
end;

sd:=0.2221;

for i:=1 to n do
Temp^[i]:=Data^[i]-(Data^[1]+(Data^[n]-Data^[1])*(i-1)/(n-1));

for i:=1 to n do
begin
Trend^[i]:=Data^[i]-Temp^[i];
Data^[i]:=Temp^[i];
write(Data^[i]:10:2);
end;

start:

dt:=0.003125*step;
cutoff:=cutoff+cutoffstep;
loop:=loop+1;

DigitalFilter(Data,cutoff);


if violation then goto finish;

sumAS[loop]:=0;
for i:=ExtPoints to n-ExtPoints do
begin

sumAS[loop]:=sumAS[loop]+sqrt(sqr(Data^[i]-FData^[i]));
end;
sumA[loop]:=(sumAS[loop]/(n-2*ExtPoints));
writeln(lst,loop:2,cutoff:6:2,n:5,'RMS: ',sumA[loop]:5:3);
sumY[loop]:=abs(sumA[loop]-sd);

if loop>1 then
if cutoff<0 then
begin
cutoff:=cutoff-cutoffstep;
DigitalFilter(Data,cutoff);
loop:=loop-1;

goto finish;
end;

if violation=false then goto start;

finish:
```

```
for i:=1 to n do
begin
Pos[i,2]:=FData^[i]+trend^[i];
Data^[i]:=Data^[i]+trend^[i];
end;

{Acc[1,2]:=(FData^[3]-2*FData^[2]+FData^[1])/(dt*dt);

for i:=2 to n-1 do
Acc[i,2]:=(FData^[i+1]-2*FData^[i]+FData^[i-1])/(dt*dt);

Acc[n,2]:=(FData^[n]-2*FData^[n-1]+FData^[n-2])/(dt*dt);}

for i:=1 to 4 do
Vel[i,2]:=(-25*FData^[i]+48*FData^[i+1]-36*FData^[i+2]+16*FData^[i+3]-3*FDa
ta^[i+4])/(12*dt);
for i:=n downto n-4 do
Vel[i,2]:=(-25*FData^[i]+48*FData^[i-1]-36*FData^[i-2]+16*FData^[i-3]-3*FData^[i
-4])/(12*dt);

for i:=5 to n-5 do
Vel[i,2]:=(FData^[i-2]-8*FData^[i-1]+8*FData^[i+1]-FData^[i+2])/(12*dt);

ExtPoints:=30;

for i:= 1 to n-(ExtPoints) do
begin
FData^[i]:=FData^[i+ExtPoints];
Data^[i]:=Data^[i+ExtPoints];
Vel[i,2]:=Vel[i+ExtPoints,2];
Acc[i,2]:=Acc[i+ExtPoints,2];
Pos[i,2]:=Pos[i+ExtPoints,2];
Tor[i,2]:=Tor[i+ExtPoints,2];
end;

n:=n-2*ExtPoints;

i:=1;
repeat
i:=i+1;
until
abs(vel[i,2]-0)>abs(vel[i-1,2]-0);
beg:=i-1;
i:=n;
repeat
i:=i-1;
until

abs(vel[i,2]-0)>abs(vel[i+1,2]-0);

fin:=i+1;
i:=0;
repeat
FData^[i+1]:=FData^[beg+i];
```

```
Data^[i+1]:=Data^[beg+i];
Vel[i+1,2]:=Vel[beg+i,2];
Acc[i+1,2]:=Acc[beg+i,2];
Pos[i+1,2]:=Pos[beg+i,2];
Tor[i+1,2]:=Tor[beg+i,2];
i:=i+1;
until  i>fin;
n:=fin-beg+1;

sumY[1]:=0;
for i:=1 to n do
begin
Acc[i,1]:=(i-1)*0.003125*step;
Vel[i,1]:=(i-1)*0.003125*step;
Pos[i,1]:=(i-1)*0.003125*step;
Tor[i,1]:=(i-1)*0.003125*step;
Pos[i,2]:=Pos[i,2]*pi/180;
Data^[i]:=Data^[i]*pi/180;
Fdata^[i]:=Fdata^[i]*pi/180;
vel[i,2]:=vel[i,2]*pi/180;
sumY[1]:=sumY[1]+sqr(0-Vel[i,2]);
end;
sumY[1]:=(sumY[1]/n);

writeln(lst,dt:20:10,Fdata^[n div 2]:10:5,Fdata^[n div 2 + 1]:10:5,vel[n div 2, 2]:10:2);}

gdriver:=detect;
Initgraph(gdriver,gmode,'');
Setviewport(0,0,round(getmaxX/1),round(getmaxY/1),true);
setbkcolor(3);


FindWorld(Vel,1,n,1,1.05);
DrawArray(Vel,1,n,1,1,true,false);
Filename:='c:\hg\'+Copy(name,3,5)+'A.KIN';
Assign(ExportFile,Filename);
Rewrite(ExportFile);
writeln(ExportFile,'"","",""');
for i:= 1 to n do
writeln(ExportFile,vel[i,1]:10:2,',',vel[i,2]:10:2,',',Tor[i,2]:10:2,',',i:3);
Close(ExportFile);
readln(ch);if UpCase(ch)='Y' then Savescreen('DigVel.pic');

closegraph;
end.
```

```
program video;

{$E+}
{$N+}

uses dos,printer,graph,mouse,axis,SERIAL,svdext;

label InputPoints,Plot;
type
ExtArray=array[1..250] of extended;

var
reg:registers;
i,j,k,l,m,ncontrol,NProximal,np,gd,gm,xm,ym,com,size,Neq:integer;
sum,sumx,sumy,xp,yp,MaxDist,MinDist,TotalError,StDev,SymbolSize:extended;
cpx,cpy,dcpx,dcpy,CpDist:ExtArray;
ProxCpX,ProxCpY,ProxDcpX,ProxDcpY,XYError:ExtArray;
ProxOrder:array[1..250] of integer;

cfx,cfy:ArrayNEq;
cor,dig:Realarray2;
test:IntegerArray2;

PlotPoint:array[1..90,1..2] of integer;
error:boolean;
ExtFile:file of extended;
RealFile:File of Real;
IntFile:file of integer;
TextFile:Text;
fname:string[30];
ch:char;
s1,s2,XY:string[30];
Com1:Text;

procedure MouseInput(var X,Y:integer);
begin
repeat
reg.ax:=3;
Intr(51,reg);
until reg.bx and 1=1 ;

x:=reg.cx;
y:=reg.dx;
repeat
reg.ax:=3;
Intr(51,reg);
until reg.bx and 1=0;
end;

procedure MoveXY(x,y:byte);
begin
with reg do
begin
ah:=2;dl:=x;dh:=y;bh:=0;intr(16,reg);
end;
```

```
end;

procedure ClrLn;
begin
with reg do
begin
ah:=3;intr(16,reg);

for i:=1 to 80 do
begin
MoveXY(i,dh);
write(' ');
end;
end;
end;

function StrCon(i:integer): STRING;
var
s:string[6];
begin
  Str(i,s);
  StrCon:=s;
end;

procedure NormalEquations(var cpx,cpy,dcpx,dcpy:ExtArray;
                          var Ncontrol: integer;
                          var Cfx,Cfy:ArrayNEq);

var
i,j,NEq:integer;
Wmin,Wmax,condition:extended;
f:ExtArrayNEq;
a:ExtArrayNXM;
u:ExtArrayNXM;
v:ExtArrayNXM;
w:ExtArrayNEq;

begin
Neq:=6;
GetMem(a,(Neq*Ncontrol+1)*10);
GetMem(u,(Neq*Ncontrol+1)*10);
GetMem(v,(Neq*Ncontrol+1)*10);
GetMem(f,(NControl+1)*10);
GetMem(w,(Neq+1)*10);


for i:=1 to Ncontrol do
  begin
    a^[i,1]:=1;
    a^[i,2]:=dcpX[i];
    a^[i,3]:=dcpY[i];
    a^[i,4]:=dcpX[i]*dcpX[i];
    a^[i,5]:=dcpX[i]*dcpY[i];
    a^[i,6]:=dcpY[i]*dcpY[i];
```

```
  end;

for i:=1 to NControl do
  f^[i]:=cpX[i];

svdfit(a,cfX,f,NControl,NEq,u,w,v);

for i:=1 to Neq do
 {writeln(lst,w[i]:10:2);}
wmax:=0;
for i:=1 to Neq do
if w^[i]>wmax then wmax:=w^[i];
wmin:=wmax;
for i:=1 to Neq do
if w^[i]<wmin then wmin:=w^[i];
if wmin<>0 then condition:=wmax/wmin else condition:=0;

{for i:=1 to Neq do
writeln(lst,w[i]:50:10);}
writeln;
writeln('CONDITION :',condition:40:10);
{writeln(lst,'RECIPROCAL:',1/condition);
writeln(lst,'MACHINE  %:',(1/condition)/1.0e-14);}

for i:=1 to Ncontrol do
  begin
    a^[i,1]:=1;
    a^[i,2]:=dcpX[i];
    a^[i,3]:=dcpY[i];
    a^[i,4]:=dcpX[i]*dcpX[i];
    a^[i,5]:=dcpX[i]*dcpY[i];
    a^[i,6]:=dcpY[i]*dcpY[i];

  end;

for i:=1 to NControl do
  f^[i]:=cpY[i];

svdfit(a,cfY,f,NControl,Neq,u,w,v);

FreeMem(a,(Neq*Ncontrol+1)*10);
FreeMem(u,(Neq*Ncontrol+1)*10);
FreeMem(v,(Neq*Ncontrol+1)*10);
FreeMem(f,(NControl+1)*10);
FreeMem(w,(Neq+1)*10);


end; {NormalEquations}

begin

write('INPUT NUMBER OF CONTROL POINTS > ');readln(Ncontrol);
write('INPUT CONTROL POINTS FROM KEYBOARD (1) OR FILE (2) >');readln(k);
case k of
1:begin
```

```
  for i:= 1 to Ncontrol do
   begin
     write('cpX[',i,'] > ');readln(cpX[i]);
     write('cpY[',i,'] > ');readln(cpY[i]);
   end;
 write('SAVE CONTROL POINTS TO FILE (Y/N) >');readln(ch);
 if (ch='y') or (ch='Y') then
  begin
    write('INPUT FILENAME >');readln(fname);

    assign(ExtFile,fname);
    rewrite(ExtFile);
    for i:=1 to Ncontrol do
    write(ExtFile,cpX[i],cpY[i]);
    close(ExtFile);
  end;
 end;
2:begin
  {write('INPUT FILENAME >');readln(fname);}
  fname:='c:\tp\pas\xray240.dat';
  assign(ExtFile,fname);
  reset(ExtFile);
  for i:=1 to Ncontrol do
  read(ExtFile,cpX[i],cpY[i]);
  close(ExtFile);
 end;
end; {case}

for i:=1 to Ncontrol do
begin
cpX[i]:=(cpX[i]*10);
cpY[i]:=(cpY[i]*10);
writeln(i:5,cpX[i]:10:0,cpY[i]:10:0);
end;

fname:='c:\hg\cp240.asc';
  assign(TextFile,fname);
  rewrite(TextFile);
  for i:=1 to Ncontrol do
  writeln(TextFile,cpX[i]:5:0,',',cpY[i]:5:0);
  close(TextFile);

readln;
gd:=detect;
InitGraph(gd,gm,'');
mouseinit;
goto InputPoints;
for i:= 1 to Ncontrol do
begin

MoveXY(1,1);write('DIGITISE CONTROL POINT ',i:2,' > ');
MouseInput(xm,ym);
dig[i,1]:=xm*1;dig[i,2]:=ym*1;
reg.ax:=2;
Intr(51,reg);
```

```
Circle(Round(dig[i,1]),Round(dig[i,2]),2);
reg.ax:=1;
Intr(51,reg);
dcpX[i]:=(dig[i,1]-GetMaxX div 2)/100;
dcpY[i]:=((GetMaxY-dig[i,2])-GetMaxY div 2)/100;
{MoveXY(30,1);write('X =',dcpX[i]*100:8:4,'     ','Y =',dcpY[i]*100:8:4);}
end;
repeat
ClrLn;
MoveXY(1,1);write('RE-DIGITISE ANY CONTROL POINT (Y/N) >');read(ch);
if Upcase(ch)='Y' then
begin
MoveXY(1,1);write('RE-DIGITISE CONTROL POINT NUMBER :       ');read(i);
Setcolor(0);
reg.ax:=2;
Intr(51,reg);
Circle(Round(dig[i,1]),Round(dig[i,2]),2);
reg.ax:=1;
Intr(51,reg);
SetColor(15);
MouseInput(xm,ym);
dig[i,1]:=xm*1;dig[i,2]:=ym*1;
reg.ax:=2;
Intr(51,reg);
Circle(Round(dig[i,1]),Round(dig[i,2]),2);
reg.ax:=1;
Intr(51,reg);
dcpX[i]:=(dig[i,1]-GetMaxX div 2)/100;
dcpY[i]:=((GetMaxY-dig[i,2])-GetMaxY div 2)/100;
end;
until  UpCase(ch)='N';
MoveXY(1,1);ClrLn;


assign(ExtFile,'DcpXY.dat');
rewrite(ExtFile);
for i:=1 to Ncontrol do
begin
write(ExtFile,DcpX[i],DcpY[i]);
end;
close(ExtFile);

InputPoints:

assign(ExtFile,'DcpXY.dat');
reset(ExtFile);
for i:=1 to Ncontrol do
begin
read(ExtFile,DcpX[i],DcpY[i]);
dig[i,1]:=DcpX[i];dig[i,2]:=DcpY[i];
end;
close(ExtFile);

assign(IntFile,'c:\tp\pas\xrtest.dat');
reset(IntFile);
```

```
for i:=1 to 52 do
begin
read(IntFile,test[i,1],test[i,2]);
end;
close(IntFile);
readln;

ClearDevice;
MoveXY(1,1);write('NUMBER OF POINTS TO DIGITIZE > ');read(np);
MoveXY(1,1);write('                                        ');

for i:=1 to np do
begin
MoveXY(1,1);write('DIGITIZE POINT > ',i:5);
MouseInput(Xm,Ym);
HideMouse;
Circle(Xm,Ym,2);
ShowMouse;
dig[i,1]:=xm*1;dig[i,2]:=ym*1;
end;

for i:=1 to np do
begin
xp:=(dig[i,1]-GetMaxX div 2)/100;
yp:=((GetMaxY-dig[i,2])-GetMaxY div 2)/100;
end;

for j:=1 to Ncontrol do
  CpDist[j]:=sqrt(sqr(DcpX[j]-Xp)+sqr(DcpY[j]-Yp));
MaxDist:=0;
for j:=1 to Ncontrol do
if CpDist[j]>MaxDist then MaxDist:=CpDist[j];

Nproximal:=240;

for j:=1 to Nproximal do
begin
  MinDist:=MaxDist;
  for k:=1 to Ncontrol do
  begin
    if CpDist[k]<MinDist then
    begin
      MinDist:=CpDist[k];
      ProxOrder[j]:=k;
    end;

  end;
  CpDist[ProxOrder[j]]:=MaxDist;
end;

for j:= 1 to NProximal do
begin
  ProxCpX[j]:=CpX[ProxOrder[j]];
  ProxCpY[j]:=CpY[ProxOrder[j]];
  ProxDcpX[j]:=DcpX[ProxOrder[j]];
```

```
  ProxDcpY[j]:=DcpY[ProxOrder[j]];
end;
NormalEquations(ProxCpX,ProxCpY,ProxDcpX,ProxDcpY,Nproximal,cfx,cfy);
cor[i,1]:=cfx[1]+
     cfx[2]*xp+
     cfx[3]*yp+
     cfx[4]*xp*xp+
     cfx[5]*xp*yp+
     cfx[6]*yp*yp;
cor[i,2]:=cfy[1]+
     cfy[2]*xp+
     cfy[3]*yp+
     cfy[4]*xp*xp+
     cfy[5]*xp*yp+
     cfy[6]*yp*yp;

Xp:=cor[i,1];Yp:=cor[i,2];
MoveXY(25,2);write('X =',Xp:10:3,'    ','Y =',Yp:10:3);
XYError[i]:=sqrt(sqr(test[i,1]-xp)+sqr(test[i,2]-Yp));
writeln(lst,i:5,test[i,1]:5,',',test[i,2]:5,XYError[i]:10:3);
end;
TotalError:=0;
for i:=1 to np do
TotalError:=TotalError+XYError[i];
TotalError:=TotalError/np;
StDev:=0;
for i:=1 to Np do
StDev:=StDev+sqr(XYError[i]-TotalError);
StDev:=sqrt(StDev/(Np-1));

writeln(lst,'CONTROL POINTS: ',NProximal:5,
          ' TOTAL ERROR = ',TotalError:10:3,
          ' S.D. =',StDev:10:3);


{digXmax:=Dig[1,1];
for i:= 2 to np do
  if dig[i,1]>digXmax then DigXmax:=dig[i,1];
DigXmin:=DigXmax;
for i:= 1 to np do
  if dig[i,1]<digXmin then DigXmin:=dig[i,1];
digYmax:=Dig[1,2];
for i:= 2 to np do
  if dig[i,2]>digYmax then DigYmax:=dig[i,2];
DigYmin:=DigYmax;
for i:= 1 to np do
  if dig[i,2]<digYmin then DigYmin:=dig[i,2];

corXmax:=cor[1,1];
for i:= 2 to np do
  if cor[i,1]>corXmax then corXmax:=cor[i,1];
corXmin:=corXmax;
for i:= 1 to np do
  if cor[i,1]<corXmin then corXmin:=cor[i,1];
corYmax:=cor[1,2];
```

```
for i:= 2 to np do
  if cor[i,2]>corYmax then corYmax:=cor[i,2];
corYmin:=corYmax;
for i:= 1 to np do
  if cor[i,2]<corYmin then corYmin:=cor[i,2];}

{for i:=1 to np do
begin
cor[i,1]:=(cor[i,1]*(digXmax-digXmin))/(corXmax-corXmin);
cor[i,2]:=(cor[i,2]*(digYmax-digYmin))/(corYmax-corYmin);
end;
for i:=1 to np do
begin
cor[i,1]:=(cor[i,1]*10);
cor[i,2]:=(cor[i,2]*10);
end;}

Plot:

{ClearDevice;}
AssignSerial(Com1,0,231);
Rewrite(Com1);
Writeln(com1,'IN;');
Writeln(com1,'IP200,200,7000,7000;');
Writeln(com1,'SP1;VS10;');

PlotPoint[ 1,1]:= 1;PlotPoint[ 1,2]:= 2;
PlotPoint[ 2,1]:= 2;PlotPoint[ 2,2]:= 3;
PlotPoint[ 3,1]:= 3;PlotPoint[ 3,2]:= 4;
PlotPoint[ 4,1]:=20;PlotPoint[ 4,2]:=21;
PlotPoint[ 5,1]:=21;PlotPoint[ 5,2]:=22;
PlotPoint[ 6,1]:=22;PlotPoint[ 6,2]:=23;
PlotPoint[ 7,1]:=23;PlotPoint[ 7,2]:=24;
PlotPoint[ 8,1]:=24;PlotPoint[ 8,2]:= 5;
PlotPoint[ 9,1]:=19;PlotPoint[ 9,2]:=36;
PlotPoint[10,1]:=36;PlotPoint[10,2]:=37;
PlotPoint[11,1]:=37;PlotPoint[11,2]:=38;
PlotPoint[12,1]:=38;PlotPoint[12,2]:=39;
PlotPoint[13,1]:=39;PlotPoint[13,2]:=40;
PlotPoint[14,1]:=40;PlotPoint[14,2]:=25;
PlotPoint[15,1]:=25;PlotPoint[15,2]:= 6;
PlotPoint[16,1]:=18;PlotPoint[16,2]:=35;
PlotPoint[17,1]:=35;PlotPoint[17,2]:=48;
PlotPoint[18,1]:=48;PlotPoint[18,2]:=49;
PlotPoint[19,1]:=49;PlotPoint[19,2]:=50;
PlotPoint[20,1]:=50;PlotPoint[20,2]:=41;
PlotPoint[21,1]:=41;PlotPoint[21,2]:=26;
PlotPoint[22,1]:=26;PlotPoint[22,2]:= 7;
PlotPoint[23,1]:=17;PlotPoint[23,2]:=34;
PlotPoint[24,1]:=34;PlotPoint[24,2]:=47;
PlotPoint[25,1]:=47;PlotPoint[25,2]:=52;
PlotPoint[26,1]:=52;PlotPoint[26,2]:=51;
PlotPoint[27,1]:=51;PlotPoint[27,2]:=42;
PlotPoint[28,1]:=42;PlotPoint[28,2]:=27;
PlotPoint[29,1]:=27;PlotPoint[29,2]:= 8;
```

```
PlotPoint[30,1]:=16;PlotPoint[30,2]:=33;
PlotPoint[31,1]:=33;PlotPoint[31,2]:=46;
PlotPoint[32,1]:=46;PlotPoint[32,2]:=45;
PlotPoint[33,1]:=45;PlotPoint[33,2]:=44;
PlotPoint[34,1]:=44;PlotPoint[34,2]:=43;
PlotPoint[35,1]:=43;PlotPoint[35,2]:=28;
PlotPoint[36,1]:=28;PlotPoint[36,2]:= 9;
PlotPoint[37,1]:=15;PlotPoint[37,2]:=32;
PlotPoint[38,1]:=32;PlotPoint[38,2]:=31;
PlotPoint[39,1]:=31;PlotPoint[39,2]:=30;
PlotPoint[40,1]:=30;PlotPoint[40,2]:=29;
PlotPoint[41,1]:=29;PlotPoint[41,2]:=10;
PlotPoint[42,1]:=14;PlotPoint[42,2]:=13;
PlotPoint[43,1]:=13;PlotPoint[43,2]:=12;
PlotPoint[44,1]:=12;PlotPoint[44,2]:=11;
PlotPoint[45,1]:= 6;PlotPoint[45,2]:= 7;
PlotPoint[46,1]:= 7;PlotPoint[46,2]:= 8;
PlotPoint[47,1]:= 8;PlotPoint[47,2]:= 9;
PlotPoint[48,1]:= 5;PlotPoint[48,2]:=25;
PlotPoint[49,1]:=25;PlotPoint[49,2]:=26;
PlotPoint[50,1]:=26;PlotPoint[50,2]:=27;
PlotPoint[51,1]:=27;PlotPoint[51,2]:=28;
PlotPoint[52,1]:=28;PlotPoint[52,2]:=10;
PlotPoint[53,1]:= 4;PlotPoint[53,2]:=24;
PlotPoint[54,1]:=24;PlotPoint[54,2]:=40;
PlotPoint[55,1]:=40;PlotPoint[55,2]:=41;
PlotPoint[56,1]:=41;PlotPoint[56,2]:=42;
PlotPoint[57,1]:=42;PlotPoint[57,2]:=43;
PlotPoint[58,1]:=43;PlotPoint[58,2]:=29;
PlotPoint[59,1]:=29;PlotPoint[59,2]:=11;
PlotPoint[60,1]:= 3;PlotPoint[60,2]:=23;
PlotPoint[61,1]:=23;PlotPoint[61,2]:=39;
PlotPoint[62,1]:=39;PlotPoint[62,2]:=50;
PlotPoint[63,1]:=50;PlotPoint[63,2]:=51;
PlotPoint[64,1]:=51;PlotPoint[64,2]:=44;
PlotPoint[65,1]:=44;PlotPoint[65,2]:=30;
PlotPoint[66,1]:=30;PlotPoint[66,2]:=12;
PlotPoint[67,1]:= 2;PlotPoint[67,2]:=22;
PlotPoint[68,1]:=22;PlotPoint[68,2]:=38;
PlotPoint[69,1]:=38;PlotPoint[69,2]:=49;
PlotPoint[70,1]:=49;PlotPoint[70,2]:=52;
PlotPoint[71,1]:=52;PlotPoint[71,2]:=45;
PlotPoint[72,1]:=45;PlotPoint[72,2]:=31;
PlotPoint[73,1]:=31;PlotPoint[73,2]:=13;
PlotPoint[74,1]:= 1;PlotPoint[74,2]:=21;
PlotPoint[75,1]:=21;PlotPoint[75,2]:=37;
PlotPoint[76,1]:=37;PlotPoint[76,2]:=48;
PlotPoint[77,1]:=48;PlotPoint[77,2]:=47;
PlotPoint[78,1]:=47;PlotPoint[78,2]:=46;
PlotPoint[79,1]:=46;PlotPoint[79,2]:=32;
PlotPoint[80,1]:=32;PlotPoint[80,2]:=14;
PlotPoint[81,1]:=20;PlotPoint[81,2]:=36;
PlotPoint[82,1]:=36;PlotPoint[82,2]:=35;
PlotPoint[83,1]:=35;PlotPoint[83,2]:=34;
PlotPoint[84,1]:=34;PlotPoint[84,2]:=33;
```

```
PlotPoint[85,1]:=33;PlotPoint[85,2]:=15;
PlotPoint[86,1]:=19;PlotPoint[86,2]:=18;
PlotPoint[87,1]:=18;PlotPoint[87,2]:=17;
PlotPoint[88,1]:=17;PlotPoint[88,2]:=16;
Xrange:=240;Yrange:=240;size:=2;
Writeln(com1,'SC-120,120,-120,120;');
for i:=1 to 88 do
BEGIN
Writeln(com1,'PA',Cor[plotpoint[i,1],1]:10:1,Cor[plotpoint[i,1],2]:10:1,';');
Writeln(com1,'PD;');
Writeln(com1,'PA',cor[plotpoint[i,2],1]:10:1,cor[plotpoint[i,2],2]:10:1,';');
Writeln(com1,'PU;');
END;
Writeln(com1,'SR',0.75*(Size+1):9:4,',',1.5*(Size+1):9:4,';');
SymbolSize:=0.002;
for i:=1 to Np do
begin
  Writeln(com1,'PA',Cor[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                 Cor[i,2]:9:4,';PD;');
  Writeln(com1,'PA',Cor[i,1]:9:4,',',
                 cor[i,2]+(Yrange)*SymbolSize*Size:9:4,';');
  Writeln(com1,'PA',cor[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                 cor[i,2]:9:4,';');
  Writeln(com1,'PA',cor[i,1]:9:4,',',
                 cor[i,2]-(Yrange)*SymbolSize*Size:9:4,';');
  Writeln(com1,'PA',cor[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                 cor[i,2]:9:4,';PU;');
end;
Writeln(com1,'PA-100,-100;PD;');
Writeln(com1,'PA-100,0;PU;');
Writeln(com1,'PA-100,-100;PD;');
Writeln(com1,'PA0,-100;PU;');


readln(l);

Writeln(com1,'IP200,200,7000,7000;');
Writeln(com1,'SP1;VS10;');
Writeln(com1,'SC0,640,0,640;');
for i:=1 to 88 do
BEGIN
Writeln(com1,'PA',Dig[plotpoint[i,1],1]:10:3,640-Dig[plotpoint[i,1],2]:10:3,';');
Writeln(com1,'PD;');
Writeln(com1,'PA',Dig[plotpoint[i,2],1]:10:3,640-Dig[plotpoint[i,2],2]:10:3,';');
Writeln(com1,'PU;');
END;
Size:=4;
for i:=1 to Np do
begin
Writeln(com1,'PA',Dig[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
                 480-Dig[i,2]+(Yrange)*SymbolSize*Size:9:4,';PD;');
  Writeln(com1,'PA',Dig[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                 480-Dig[i,2]+(Yrange)*SymbolSize*Size:9:4,';');
  Writeln(com1,'PA',Dig[i,1]+(Xrange)*SymbolSize*Size:9:4,',',
                 480-Dig[i,2]-(Yrange)*SymbolSize*Size:9:4,';');
```

```
  Writeln(com1,'PA',Dig[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
               480-Dig[i,2]-(Yrange)*SymbolSize*Size:9:4,';');
  Writeln(com1,'PA',Dig[i,1]-(Xrange)*SymbolSize*Size:9:4,',',
               480-Dig[i,2]+(Yrange)*SymbolSize*Size:9:4,';PU;');
end;
Writeln(com1,'PA0,0;PD;');
Writeln(com1,'PA0,100;PU;');
Writeln(com1,'PA0,0;PD;');
Writeln(com1,'PA100,0;PU;');
Writeln(com1,'PA0,0;SP0;');
   readln;
CloseGraph;
end.
```

```
program Kinematic;

{$E+}
{$N+}

uses Dos,crt,printer,graph,axis,mouse,spline,svdext,Gcrt;

type

LineParameter=record
            a,b,c:Real;
            end;
PointCoordinates=record
                X,Y:Real;
                end;
FrameRecord=record
            TFAngle:Real;
            MomArm:Real;
            BAngle:Real;
            TFContact:PointCoordinates;
            AntPlat:PointCoordinates;
            PosPlat:PointCoordinates;
            Patella:PointCoordinates;
            TibTub:PointCoordinates;
            end;
ExtArray=array[1..250] of Extended;
label angle,here,retrieve;
var
i,j,k,n,FemCond,FramePointer,step,gd,gm,gerror,
Ncontrol,Nproximal,NFrame,xm,ym,x,y,x1,x2,y1,y2,error:integer;
FemCon,TibPlat,Tibia,Femur,DigPoints:XYArray;
TibiaAxis,FemurAxis,PatelTendon,TibPlatAxis:LineParameter;
KneeAngle,KneeAngleCos,FTDistance,
IntX,IntY,FemX,FemY,TibX,TibY,aa,bb,cc,TempX,TempY,det,Offset,FullExt:Real;
FemurContact,TibiaContact:array [1..2] of PointCoordinates;
MarkerBase,MarkerTip:PointCoordinates;
Cf:arrayNeq;
RealFile:File of Extended;
Frame:array [1..20] of FrameRecord;
cpx,cpy,dcpx,dcpy:ExtArray;
ExtFile:File of Extended;
IntFile:file of integer;
StoreFile:Text;
fname:string;
Ts:array [1..12] of String[8];
ch:char;
Xp,Yp,MaxDist,MinDist:extended;
CpDist: ExtArray;
ProxCpX,ProxCpY,ProxDcpX,ProxDcpY,XYError:ExtArray;
ProxOrder:array[1..100] of integer;
cfx,cfy:ArrayNEq;
a:ExtArrayNXM;
u:ExtArrayNXM;
v:ExtArrayNXM;
f:ExtArrayNEq;
```

```
w:ExtArrayNEq;

FileName:String;
TextFile:text;

function KeyPress: boolean;
var
Empty:char;
begin
while KeyPressed do Empty:=ReadKey;
Repeat Until KeyPressed;
KeyPress:=true;
end;

procedure Audio;
begin
Sound(1500);
Delay(50);
Nosound;
end;

procedure XYFilter(var x,y:Real);
begin
X:=(X-GetMaxX div 2)/100;
Y:=((GetMaxY-Y)-GetMaxY div 2)/100;
end;


procedure MouseInput(var X,Y:Integer);
begin
repeat
reg.ax:=3;
Intr(51,reg);
until reg.bx and 1=1 ;
x:=reg.cx;
y:=reg.dx;
repeat
reg.ax:=3;
Intr(51,reg);
until reg.bx and 1=0;
Audio;
end;



Procedure InputData(var XYPoint: XYArray; n:integer);
var
i,j,x,y:integer;
begin
  for i:= 1 to n do
  begin
    MouseInput(x,y);XYPoint[i,1]:=x*1;XYPoint[i,2]:=y*1;
    HideMouse;
    PutPixel(x,y,15);
    ShowMouse;
```

```
    XYFilter(XYPoint[i,1],XYPoint[i,2]);
    GwriteXY(Xg(25),Yg(1),'N :'+ls(i)+' X : '+ls(x)+' Y : '+ls(y));
  end;
end;

function MomentArm(var Con,Pat,Tib:PointCoordinates): real;
var
f,g:real;
begin
f:=Tib.X-Pat.X;
g:=Tib.Y-Pat.Y;
MomentArm:=sqrt(sqr(g*(f*(Con.Y-Tib.Y)-g*(Con.X-Tib.X)))+
sqr(f*(g*(Con.X-Tib.X)-f*(Con.Y-Tib.Y))))
 /(sqr(f)+sqr(g));
end;


procedure NormalEquations(var cpx,cpy,dcpx,dcpy:ExtArray;
                          var Ncontrol: integer;
                          var Cfx,Cfy:ArrayNEq);

var
i,j,NEq:integer;
Wmin,Wmax,condition:extended;


begin
Neq:=6;
GetMem(a,(Neq*Ncontrol+1)*10);
GetMem(u,(Neq*Ncontrol+1)*10);
GetMem(v,(Neq*Ncontrol+1)*10);
GetMem(f,(NControl+1)*10);
GetMem(w,(Neq+1)*10);

for i:=1 to Ncontrol do
  begin
    a^[i,1]:=1;
    a^[i,2]:=dcpX[i];
    a^[i,3]:=dcpY[i];
    a^[i,4]:=dcpX[i]*dcpX[i];
    a^[i,5]:=dcpX[i]*dcpY[i];
    a^[i,6]:=dcpY[i]*dcpY[i];

  end;

for i:=1 to NControl do
  f^[i]:=cpX[i];

svdfit(a,cfX,f,NControl,NEq,u,w,v);

for i:=1 to Neq do
 {writeln(lst,w[i]:10:2);}
wmax:=0;
for i:=1 to Neq do
if w^[i]>wmax then wmax:=w^[i];
```

```
wmin:=wmax;
for i:=1 to Neq do
if w^[i]<wmin then wmin:=w^[i];
if wmin<>0 then condition:=wmax/wmin else condition:=0;

{for i:=1 to Neq do
writeln(lst,w[i]:50:10);}
{writeln;
writeln('CONDITION :',condition:40:10);}
{writeln(lst,'RECIPROCAL:',1/condition);
writeln(lst,'MACHINE  %:',(1/condition)/1.0e-14);}

for i:=1 to Ncontrol do
  begin
    a^[i,1]:=1;
    a^[i,2]:=dcpX[i];
    a^[i,3]:=dcpY[i];
    a^[i,4]:=dcpX[i]*dcpX[i];
    a^[i,5]:=dcpX[i]*dcpY[i];
    a^[i,6]:=dcpY[i]*dcpY[i];

  end;

for i:=1 to NControl do
  f^[i]:=cpY[i];

svdfit(a,cfY,f,NControl,Neq,u,w,v);
FreeMem(a,(Neq*Ncontrol+1)*10);
FreeMem(u,(Neq*Ncontrol+1)*10);
FreeMem(v,(Neq*Ncontrol+1)*10);
FreeMem(f,(NControl+1)*10);
FreeMem(w,(Neq+1)*10);
end; {NormalEquations}

procedure DistortionTransformation(var cpx,cpy,Dcpx,dcpy:ExtArray; var X,Y:Real;
Nproximal:integer);
var
j,k:integer;
begin

xp:=X;
yp:=Y;
GDelLineXY(Xg(10),Yg(2));
GWriteXY(Xg(10),Yg(2),'X= '+Rs(Xp,10,3)+'    '+'Y= '+Rs(Yp,10,3));

for j:=1 to Ncontrol do
  CpDist[j]:=sqrt(sqr(DcpX[j]-Xp)+sqr(DcpY[j]-Yp));
MaxDist:=0;
for j:=1 to Ncontrol do
if CpDist[j]>MaxDist then MaxDist:=CpDist[j];

for j:=1 to Nproximal do
begin
  MinDist:=MaxDist;
  for k:=1 to Ncontrol do
```

```
  begin
   if CpDist[k]<MinDist then
   begin
     MinDist:=CpDist[k];
     ProxOrder[j]:=k;
   end;

  end;
  CpDist[ProxOrder[j]]:=MaxDist;
end;

for j:= 1 to NProximal do
begin
  ProxCpX[j]:=CpX[ProxOrder[j]];
  ProxCpY[j]:=CpY[ProxOrder[j]];
  ProxDcpX[j]:=DcpX[ProxOrder[j]];
  ProxDcpY[j]:=DcpY[ProxOrder[j]];
  {writeln(ProxCpX[j]:10:2,ProxCpY[j]:10:2,ProxDcpX[j]:10:2,ProxDcpY[j]:10:2);}
  {write(lst,ProxOrder[j]:5);}
end;
NormalEquations(ProxCpX,ProxCpY,ProxDcpX,ProxDcpY,Nproximal,cfx,cfy);
  x:=cfx[1]+
     cfx[2]*xp+
     cfx[3]*yp+
     cfx[4]*xp*xp+
     cfx[5]*xp*yp+
     cfx[6]*yp*yp;
  y:=cfy[1]+
     cfy[2]*xp+
     cfy[3]*yp+
     cfy[4]*xp*xp+
     cfy[5]*xp*yp+
     cfy[6]*yp*yp;
GDelLineXY(Xg(10),Yg(3));
GWriteXY(Xg(10),Yg(3),'X= '+Rs(X,10,3)+'    '+'Y= '+Rs(Y,10,3));
end; {DistortionTransformation}

procedure LineEquation(var XYarray:XYArray;
                           n:integer;
                           var cf:ArrayNEq);




begin
GetMem(a,(2*n+1)*10);
GetMem(u,(2*n+1)*10);
GetMem(v,(2*n+1)*10);
GetMem(f,(2*n+1)*10);
GetMem(w,(2*n+1)*10);



for i:=1 to N do
  begin
```

```
    a^[i,1]:=1;
    a^[i,2]:=XYArray[i,1];
  end;

for i:=1 to N do
  f^[i]:=XYArray[i,2];

svdfit(a,cf,f,n,n,u,w,v);

FreeMem(a,(2*n+1)*10);
FreeMem(u,(2*n+1)*10);
FreeMem(v,(2*n+1)*10);
FreeMem(f,(2*n+1)*10);
FreeMem(w,(2*n+1)*10);



end; {NormalEquations}

procedure InputCalibPoints;
begin
GDelLineXY(Xg(1),Yg(1));
GWriteXY(Xg(1),Yg(1),'INPUT NUMBER OF CONTROL POINTS >');
if not GInt(NControl) then Halt(1);

GDelLineXY(Xg(1),Yg(1));
GWriteXY(Xg(1),Yg(1),'INPUT CONTROL POINTS FROM KEYBOARD (1) OR FILE
(2)>');
if not GInt(k) then Halt(1);

case k of
1:begin
   for i:= 1 to Ncontrol do
     begin
       write('cpX[',i,'] > ');readln(cpX[i]);
       write('cpY[',i,'] > ');readln(cpY[i]);
     end;
  GDelLineXY(Xg(1),Yg(1));
GWriteXY(Xg(1),Yg(1),'SAVE CONTROL POINTS TO FILE  (Y/N) >');
if not GReadStr(fname) then Halt(1);
  if (fname='y') or (fname='Y') then
    begin
    GDelLineXY(Xg(1),Yg(1));
GWriteXY(Xg(1),Yg(1),'INPUT FILENAME >');
if not GReadStr(fname) then Halt(1);
      assign(ExtFile,fname);
      rewrite(ExtFile);
      for i:=1 to Ncontrol do
      write(ExtFile,cpX[i],cpY[i]);
      close(ExtFile);
    end;
  end;
2:begin
    GDelLineXY(Xg(1),Yg(1));
    GWriteXY(Xg(1),Yg(1),'INPUT FILENAME >');
```

```
    if not GReadStr(fname) then Halt(1);
    assign(ExtFile,fname);
    reset(ExtFile);
    for i:=1 to Ncontrol do
    read(ExtFile,cpX[i],cpY[i]);
    close(ExtFile);
   end;
end; {case}
for i:=1 to Ncontrol do
begin
cpX[i]:=(cpX[i]*10);
cpY[i]:=(cpY[i]*10);
{writeln(i:5,cpX[i]:10:0,cpY[i]:10:0);}
end;
end; {InputCalibPoints}


procedure DigitCalibPoints;
label Stored;
begin
goto Stored;
for i:= 1 to Ncontrol do
begin
GDelLineXY(Xg(1),Yg(1));
GWriteXY(Xg(1),Yg(1),'DIGITISE CONTROL POINT '+Is(i)+' > ');
MouseInput(xm,ym);
HideMouse;
Circle(Xm,Ym,2);
ShowMouse;
dcpX[i]:=Xm*1;
DcpY[i]:=Ym*1;
{MoveXY(30,1);write('X =',dcpX[i]*100:8:4,'    ','Y =',dcpY[i]*100:8:4);}
end;

repeat
GDelLineXY(Xg(1),Yg(1));
GWriteXY(Xg(1),Yg(1),'RE-DIGITISE ANY CONTROL POINT (Y/N) >');
if not GReadch(ch) then Halt(1);
if Upcase(ch)='Y' then
begin
GDelLineXY(Xg(1),Yg(1));
GWriteXY(Xg(1),Yg(1),'RE-DIGITISE CONTROL POINT NUMBER) >');
if not GInt(i) then Halt(1);
Xm:=round(dcpX[i]);
Ym:=round(DcpY[i]);
Setcolor(0);
HideMouse;
Circle(Xm,Ym,2);
ShowMouse;
SetColor(15);
MouseInput(xm,ym);
HideMouse;
Circle(Xm,Ym,2);
ShowMouse;
dcpX[i]:=Xm*1;
```

```
dcpY[i]:=Ym*1;
end;
until  UpCase(ch)='N';
Stored:
assign(ExtFile,'c:\tp\pas\DigCal.dat');
reset(ExtFile);
for i:=1 to Ncontrol do
read(ExtFile,dcpX[i],dcpY[i]);
close(ExtFile);

for i:=1 to Ncontrol do
begin
dcpX[i]:=(dcpX[i]-GetMaxX div 2)/100;
DcpY[i]:=((GetMaxY-dcpY[i])-GetMaxY div 2)/100;
end;

end; {DigitCalibPoints}

begin   {main}

gd:=Detect;
InitGraph(gd,gm,'c:\tp\pas');
Gerror:=GraphResult;
if gerror<0 then
begin
write('GraphicsError :');writeln(GraphErrorMsg(gerror));
end;

MouseInit;

Nproximal:=7;
InputCalibPoints;
DigitCalibPoints;
GDelLineXY(Xg(1),Yg(1));
GWriteXY(Xg(1),Yg(1),'INPUT NUMBER OF FRAMES >');
if not GInt(NFrame) then Halt(1);
ClearDevice;
for FramePointer:=1 to NFrame do
begin
for FemCond:= 1 to 2 do
begin

n:=15;
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'DIGITIZE CONDYLE '+Is(FemCond));Audio;

InputData(DigPoints,n);

for i:=1 to n do
  DistortionTransformation(cpx,cpy,Dcpx,dcpy,DigPoints[i,1],DigPoints[i,2],Nproximal);

Bspline(DigPoints,FemCon,n,10);
GDelLineXY(Xg(1),Yg(1));GWriteXY(Xg(1),Yg(1),'HIT A KEY TO CONTINUE');
Repeat until KeyPress;
GDelLineXY(Xg(1),Yg(1));
```

```
GwriteXY(Xg(1),Yg(1),'DIGITIZE TIBIA PLATEAU ' + Is(FemCond));Audio;
InputData(DigPoints,n);
for i: = 1 to n do
  DistortionTransformation(cpx,cpy,Dcpx,dcpy,DigPoints[i,1],DigPoints[i,2],Nproximal);
Bspline(DigPoints,TibPlat,n,10);
n: = (n-5)*10;

{for i: = 1 to n do
begin
PutPixel(round(FemCon[i,1]),round(FemCon[i,2]),15);
PutPixel(round(TibPlat[i,1]),round(TibPlat[i,2]),15);
DrawPoint(FemCon[i,1],FemCon[i,2]);
DrawPoint(TibPlat[i,1],TibPlat[i,2]);
end;}

FTDistance: = 0;
for i: = 1 to n do
  for j: = 1 to n do
  begin
    if i = 1 then
      begin
      FTDistance: = sqrt(sqr(FemCon[i,1]-TibPlat[j,1]) + sqr(FemCon[i,2]-TibPlat[j,2]));
      FemurContact[FemCond].X: = FemCon[i,1];
FemurContact[FemCond].Y: = FemCon[i,2];
      TibiaContact[FemCond].X: = TibPlat[j,1]; TibiaContact[FemCond].Y: = TibPlat[j,2];
      end
      else
      if sqrt(sqr(FemCon[i,1]-TibPlat[j,1]) + sqr(FemCon[i,2]-TibPlat[j,2])) < FTDistance
then
      begin
      FTDistance: = sqrt(sqr(FemCon[i,1]-TibPlat[j,1]) + sqr(FemCon[i,2]-TibPlat[j,2]));
      FemurContact[FemCond].X: = FemCon[i,1];
FemurContact[FemCond].Y: = FemCon[i,2];
      TibiaContact[FemCond].X: = TibPlat[j,1]; TibiaContact[FemCond].Y: = TibPlat[j,2];
      end;
   end;


GDelLineXY(Xg(1),Yg(1));
end; {i: = 1 to 2}

with Frame[FramePointer] do
begin
TFContact.X: = ((FemurContact[1].X + TibiaContact[1].X)/2 +
            (FemurContact[2].X + TibiaContact[2].X)/2)/2;
TFContact.Y: = ((FemurContact[1].Y + TibiaContact[1].Y)/2 +
            (FemurContact[2].Y + TibiaContact[2].Y)/2)/2;
DistortionTransformation(cpx,cpy,Dcpx,dcpy,TFContact.X,TFContact.Y,Nproximal);
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'DIGITIZE POSTERIOR TIBIA PLATEAU');Audio;
MouseInput(X,Y);
PosPlat.X: = X/1; PosPlat.Y: = Y/1;
  XYFilter(PosPlat.X,PosPlat.Y);
  DistortionTransformation(cpx,cpy,Dcpx,dcpy,PosPlat.X,PosPlat.Y,Nproximal);
GDelLineXY(Xg(1),Yg(1));
```

```
GwriteXY(Xg(1),Yg(1),'DIGITIZE ANTERIOR TIBIA PLATEAU');Audio;
MouseInput(X,Y);
AntPlat.X:=X/1; AntPlat.Y:=Y/1;
XYFilter(AntPlat.X,AntPlat.Y);
  DistortionTransformation(cpx,cpy,Dcpx,dcpy,AntPlat.X,AntPlat.Y,Nproximal);
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'DIGITIZE PATELLA');Audio;
MouseInput(X,Y);
Patella.X:=X/1; Patella.Y:=Y/1;
  XYFilter(Patella.X,Patella.Y);
  DistortionTransformation(cpx,cpy,Dcpx,dcpy,Patella.X,Patella.Y,Nproximal);
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'DIGITIZE TIBIA TUBEROCITY');Audio;
MouseInput(X,Y);
TibTub.X:=X/1; TibTub.Y:=Y/1;
  XYFilter(TibTub.X,TibTub.Y);
  DistortionTransformation(cpx,cpy,Dcpx,dcpy,TibTub.X,TibTub.Y,Nproximal);
MomArm:=MomentArm(TFContact,TibTub,Patella);
end;

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'HIT A KEY TO CONTINUE');Audio;

repeat until KeyPress;
GDelLineXY(Xg(1),Yg(1));
angle:
n:=2;
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'DIGITIZE TIBIA AXIS');
InputData(Tibia,n);
for i:=1 to n do
  DistortionTransformation(cpx,cpy,Dcpx,dcpy,Tibia[i,1],Tibia[i,2],Nproximal);

TibiaAxis.a:=-(Tibia[2,2]-Tibia[1,2]);
TibiaAxis.b:=Tibia[2,1]-Tibia[1,1];
TibiaAxis.c:=Tibia[1,1]*(-TibiaAxis.a)-Tibia[1,2]*TibiaAxis.b;


GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'DIGITIZE FEMUR AXIS');Audio;

InputData(Femur,n);
for i:=1 to n do
  DistortionTransformation(cpx,cpy,Dcpx,dcpy,Femur[i,1],Femur[i,2],Nproximal);

FemurAxis.a:=-(Femur[2,2]-Femur[1,2]);
FemurAxis.b:=Femur[2,1]-Femur[1,1];
FemurAxis.c:=Femur[1,1]*(-FemurAxis.a)-Femur[1,2]*FemurAxis.b;


Det:=TibiaAxis.a*FemurAxis.b-FemurAxis.a*TibiaAxis.b;
if Det=0 then halt(1)
 else
 begin
 IntX:=(TibiaAxis.b*FemurAxis.c-FemurAxis.b*TibiaAxis.c)*(1/Det);
```

```
 IntY:=(FemurAxis.a*TibiaAxis.c-TibiaAxis.a*FemurAxis.c)*(1/Det);
 end;

aa:=sqrt(sqr(Tibia[1,1]-Femur[1,1])+sqr(Tibia[1,2]-Femur[1,2]));
bb:=sqrt(sqr(IntX-Femur[1,1])+sqr(IntY-Femur[1,2]));
cc:=sqrt(sqr(Tibia[1,1]-IntX)+sqr(Tibia[1,2]-IntY));
KneeAngleCos:=((sqr(bb)+sqr(cc)-sqr(aa))/(2*bb*cc));
KneeAngle:=arctan(sqrt(1-sqr(KneeAngleCos))/KneeAngleCos);

if KneeAngle<0 then KneeAngle:=pi-abs(Kneeangle);
KneeAngle:=(KneeAngle*180)/pi;
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'ANGLE ='+Rs(KneeAngle,5,2)+
                  ' MOMENT ARM ='+Rs(Frame[FramePointer].MomArm,5,2));
Frame[FramePointer].TFAngle:=KneeAngle;
Audio;
Repeat until KeyPress;



with Frame[FramePointer] do
begin
PatelTendon.a:=-(TibTub.Y-Patella.Y);
PatelTendon.b:=TibTub.X-Patella.X;
PatelTendon.c:=Patella.X*(-PatelTendon.a)-Patella.Y*PatelTendon.b;

TibPlatAxis.a:=-(Posplat.Y-AntPlat.Y);
TibPlatAxis.b:=PosPlat.X-AntPlat.X;
TibPlatAxis.c:=AntPlat.X*(-TibPlatAxis.a)-AntPlat.Y*TibPlatAxis.b;

Det:=TibPlatAxis.a*PatelTendon.b-PatelTendon.a*TibPlatAxis.b;
if Det=0 then halt(1)
 else
 begin
 IntX:=(TibPlatAxis.b*PatelTendon.c-PatelTendon.b*TibPlatAxis.c)*(1/Det);
 IntY:=(PatelTendon.a*TibPlatAxis.c-TibPlatAxis.a*PatelTendon.c)*(1/Det);
 end;

aa:=sqrt(sqr(PosPlat.X-Patella.X)+sqr(PosPlat.Y-Patella.Y));
bb:=sqrt(sqr(IntX-Patella.X)+sqr(IntY-Patella.Y));
cc:=sqrt(sqr(PosPlat.X-IntX)+sqr(PosPlat.Y-IntY));
KneeAngleCos:=((sqr(bb)+sqr(cc)-sqr(aa))/(2*bb*cc));
KneeAngle:=arctan(sqrt(1-sqr(KneeAngleCos))/KneeAngleCos);

if KneeAngle<0 then KneeAngle:=pi-abs(Kneeangle);
KneeAngle:=(KneeAngle*180)/pi;
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'ANGLE ='+Rs(KneeAngle,5,2)+
                  ' MOMENT ARM ='+Rs(Frame[FramePointer].MomArm,5,2));
Frame[FramePointer].BAngle:=KneeAngle;
  end;


ClearDevice;
World(-100,100,-100,100);
```

```
with Frame[FramePointer] do
begin
  writeln(lst,FramePointer:3,TFAngle:10:2,MomArm:10:2,BAngle:10:2);
 {writeln(lst,TFContact.X:5:2,TfContact.Y:5:2);
  writeln(lst,TibTub.X:5:2,TibTub.Y:5:2);
  writeln(lst,Patella.X:5:2,Patella.Y:5:2);
  writeln(lst,AntPlat.X:5:2,AntPlat.Y:5:2);
  writeln(lst,PosPlat.X:5:2,PosPlat.Y:5:2);}
DrawPoint(TFContact.X,TfContact.Y);
DrawPoint(TibTub.X,TibTub.Y);
DrawPoint(Patella.X,Patella.Y);
DrawLine(AntPlat.X,AntPlat.Y,PosPlat.X,PosPlat.Y);
DrawLine(Tibia[1,1],Tibia[1,2],IntX,IntY);
DrawLine(Femur[1,1],Femur[1,2],IntX,IntY);
end;
Audio;
repeat until keypress;
ClearDevice;

end; {FramePointer}

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT FILE NAME :');
if not GReadStr(FileName) then halt(1);
if Length(FileName)>1 then
begin
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT OFFSET :');
if not GReal(Offset) then halt(1);
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'FULL EXTENSION :');
if not GReal(FullExt) then halt(1);


Assign(StoreFile,FileName);
Rewrite(StoreFile);
for i:= 1 to NFrame do
begin
Frame[FramePointer].TfAngle:=Frame[FramePointer].TFAngle+Offset;
Frame[FramePointer].TfAngle:=FullExt-Frame[FramePointer].TFAngle;


Writeln(lst,Frame[i].TFAngle:8:2,
          Frame[i].MomArm:8:2,
          Frame[i].BAngle:8:2,
          Frame[i].TFContact.X:8:2,Frame[i].TFContact.Y:8:2,
          Frame[i].AntPlat.X:8:2,Frame[i].AntPlat.Y:8:2,
          Frame[i].PosPlat.X:8:2,Frame[i].PosPlat.Y:8:2,
          Frame[i].Patella.X:8:2,Frame[i].Patella.Y:8:2,
          Frame[i].TibTub.X:8:2,Frame[i].TibTub.Y:8:2);

Writeln(StoreFile,
          Frame[i].TFAngle:8:2,
          Frame[i].MomArm:8:2,
          Frame[i].BAngle:8:2,
```

```
        Frame[i].TFContact.X:8:2,Frame[i].TFContact.Y:8:2,
        Frame[i].AntPlat.X:8:2,Frame[i].AntPlat.Y:8:2,
        Frame[i].PosPlat.X:8:2,Frame[i].PosPlat.Y:8:2,
        Frame[i].Patella.X:8:2,Frame[i].Patella.Y:8:2,
        Frame[i].TibTub.X:8:2,Frame[i].TibTub.Y:8:2);
end;
Close(StoreFile);
end;

closegraph;

end.
```

```
program Knee;

uses smspl,graph,gcrt,printer,axis;

const
RealSize=6;
label retrieve;
var
Data:Realarray2;
{x,y:ArrayPointer2;}
KneeAngle,MomArm,IntMomArm,Bangle,IntBAngle,Torque,GravAngle,AngAcc,a,b,c,d:A
rrayPointer1;
i,k,n,gdriver,gmode,NFrame,Npoint,Error:integer;
sf,Variance,FExt,Flex,IntAngle:Real;
FileName:String;
Ts8:Array [1..10] of String[8];
Ts10:Array [1..10] of String[10];
StoreFile:Text;
begin
n:=200;
GetMem(GravAngle,(n+2)*RealSize);
GetMem(AngAcc,(n+2)*RealSize);
GetMem(Torque,(n+2)*RealSize);
GetMem(KneeAngle,(n+2)*RealSize);
GetMem(MomArm,(n+2)*RealSize);
GetMem(BAngle,(n+2)*RealSize);
GetMem(IntMomArm,(n+2)*RealSize);
GetMem(IntBAngle,(n+2)*RealSize);
GetMem(a,(n+2)*RealSize);
GetMem(b,(n+2)*RealSize);
GetMem(c,(n+2)*RealSize);
GetMem(d,(n+2)*RealSize);
gdriver:=detect;
Initgraph(gdriver,gmode,'');


GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT FILE NAME :');
if not GReadStr(FileName) then halt(1);
{goto retrieve;}
if Length(FileName)>1 then
begin
Assign(StoreFile,FileName);
Reset(StoreFile);
i:=1;
while not eof(storefile) do
begin
for k:=1 to 3 do
Read(StoreFile,Ts8[k]);
        val(Ts8[1],KneeAngle^[i],Error);
        val(Ts8[2],MomArm^[i],Error);
        val(Ts8[3],Bangle^[i],Error);
Writeln(lst,KneeAngle^[i]:8:2,
        MomArm^[i]:8:2,
```

```
          Bangle^[i]:8:2);
Readln(storefile);
i:=i+1;
end;
NFrame:=i-1;
Close(StoreFile);
end;
Writeln(lst);

{GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT FILE NAME :');
if not GReadStr(FileName) then halt(1);
if Length(FileName)>1 then
begin
Assign(StoreFile,FileName);
Reset(StoreFile);
i:=1;
Readln(storefile);
while not eof(storefile) do
begin
for k:=1 to 3 do
begin
Read(StoreFile,Ts10[k]);
Read(StoreFile,ch);
end;
          val(Ts10[3],Torque^[i],Error);
Readln(storefile);
i:=i+1;
end;
NPoint:=i-1;
Close(StoreFile);
end;
Writeln(lst);

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT FILE NAME :');
if not GReadStr(FileName) then halt(1);
if Length(FileName)>1 then
begin
Assign(StoreFile,FileName);
Reset(StoreFile);
i:=1;
Readln(storefile);
while not eof(storefile) do
begin
for k:=1 to 2 do
begin
Read(StoreFile,Ts10[k]);
Read(StoreFile,ch);
end;
          val(Ts10[2],GravAngle^[i],Error);

Readln(storefile);
i:=i+1;
end;
```

```
NPoint:=i-1;
Close(StoreFile);
end;

writeln(lst);

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT FULL EXTENSION :');
if not GReal(FExt) then halt(1); }

{Extension}
KneeAngle^[NFrame+1]:=KneeAngle^[NFrame]+
                  (KneeAngle^[NFrame]-KneeAngle^[NFrame-1]);
MomArm^[NFrame+1]:=MomArm^[NFrame]+
                  (MomArm^[NFrame]-MomArm^[NFrame-1]);
BAngle^[NFrame+1]:=BAngle^[NFrame]+
                  (Bangle^[NFrame]-BAngle^[NFrame-1]);
sf:=0;
SplineFunction(KneeAngle,MomArm,a,b,c,d,NFrame+1,sf);
i:=0;
repeat
Flex:=i*5;
if Flex<KneeAngle^[NFrame] then
SplFit(KneeAngle,MomArm,a,b,c,d,Flex,IntMomArm^[i],NFrame+1,sf)

else
begin
IntAngle:=KneeAngle^[NFrame]-abs(KneeAngle^[NFrame]-Flex);
SplFit(KneeAngle,MomArm,a,b,c,d,
      IntAngle,
      IntMomArm^[i],NFrame+1,sf);
IntMomArm^[i]:=MomArm^[NFrame]+
          (MomArm^[NFrame]-IntMomArm^[i])
end;
{write(lst,IntMomArm^[i]:8:2);}
i:=i+1;
until Flex>Trunc(KneeAngle^[NFrame]);

writeln(lst);
sf:=0;
SplineFunction(KneeAngle,BAngle,a,b,c,d,NFrame+1,sf);
i:=0;
repeat
Flex:=i*5;
if Flex<KneeAngle^[NFrame] then
SplFit(KneeAngle,BAngle,a,b,c,d,Flex,IntBAngle^[i],NFrame+1,sf)

else
begin
IntAngle:=KneeAngle^[NFrame]-abs(KneeAngle^[NFrame]-Flex);
SplFit(KneeAngle,BAngle,a,b,c,d,
      IntAngle,
      IntBAngle^[i],NFrame+1,sf);
IntBAngle^[i]:=BAngle^[NFrame]+
          (BAngle^[NFrame]-IntBAngle^[i])
```

```
end;
{write(lst,IntBAngle^[i]:8:2);}
i:=i+1;
until Flex>Trunc(KneeAngle^[NFrame]);
NFrame:=i-1;
GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT FILE NAME :');
if not GReadStr(FileName) then halt(1);
{goto retrieve;}
if Length(FileName)>1 then
begin
Assign(StoreFile,FileName);
Rewrite(StoreFile);
for i:=1 to NFrame do
begin
Writeln(StoreFile,(i-1)*5:8,IntMomArm^[i]:8:2,IntBAngle^[i]:8:2);
Writeln(lst,(i-1)*5:8,IntMomArm^[i]:8:2,IntBAngle^[i]:8:2);
end;
Close(StoreFile);
end;
Writeln(lst,FileName);
Halt(1);

for i:=1 to Npoint do
writeln(lst,FExt-(GravAngle^[i]*180)/pi:8:2,Torque^[i]:8:2,IntMomArm^[i]:8:2,IntBAngl
e^[i]:8:2);

FreeMem(GravAngle,(n+2)*RealSize);
FreeMem(AngAcc,(n+2)*RealSize);
FreeMem(Torque,(n+2)*RealSize);
FreeMem(KneeAngle,(n+2)*RealSize);
FreeMem(MomArm,(n+2)*RealSize);
FreeMem(BAngle,(n+2)*RealSize);
FreeMem(IntMomArm,(n+2)*RealSize);
FreeMem(IntBAngle,(n+2)*RealSize);
FreeMem(a,(n+2)*RealSize);
FreeMem(b,(n+2)*RealSize);
FreeMem(c,(n+2)*RealSize);
FreeMem(d,(n+2)*RealSize);
end.
```

```
program Kinetic;

uses smspl,graph,gcrt,printer,axis;

const
RealSize=6;


label retrieve;
var
Data:Realarray2;
KneeAngle,MomArm,Bangle,
Torque,GravAngle,AngVel,AngAcc,Fm,Fs,Fc,a,b,c,d,Time:ArrayPointer1;
i,k,n,gdriver,gmode,NFrame,Npoint,Error,IArmPos:integer;
Tm,Tg,Tr,Tb,Tt,Fr,Fg,sf,Variance,FExt,Flex,Bmass,
PatMomArm,ISeg,IInpArm,ResMomArm,Dt,TanAcc,RadAcc,SegLen,SegRad:Real;
FileName:String;
Str8:Array [1..10] of String[8];
Str10:Array [1..10] of String[10];
StoreFile:Text;

function RadToDeg(var r: Real): Real;
begin
RadToDeg:=(r*180)/pi;
end;

function DegToRad(var r: Real): Real;
begin
DegToRad:=(r*pi)/180;
end;


begin
n:=200;
GetMem(GravAngle,(n+2)*RealSize);
GetMem(AngAcc,(n+2)*RealSize);
GetMem(AngVel,(n+2)*RealSize);
GetMem(Fm,(n+2)*RealSize);
GetMem(Fs,(n+2)*RealSize);
GetMem(Fc,(n+2)*RealSize);
GetMem(Time,(n+2)*RealSize);
GetMem(Torque,(n+2)*RealSize);
GetMem(KneeAngle,(n+2)*RealSize);
GetMem(MomArm,(n+2)*RealSize);
GetMem(BAngle,(n+2)*RealSize);
GetMem(a,(n+2)*RealSize);
GetMem(b,(n+2)*RealSize);
GetMem(c,(n+2)*RealSize);
GetMem(d,(n+2)*RealSize);
gdriver:=detect;
Initgraph(gdriver,gmode,'');


GDelLineXY(Xg(1),Yg(1));
```

```
GwriteXY(Xg(1),Yg(1),'INPUT X-RAY FILE NAME :');
if not GReadStr(FileName) then halt(1);
{goto retrieve;}
if Length(FileName)>1 then
begin
Assign(StoreFile,FileName);
Reset(StoreFile);
i:=1;
while not eof(storefile) do
begin
for k:=1 to 3 do
Read(StoreFile,Str8[k]);
        val(Str8[1],KneeAngle^[i],Error);
        val(Str8[2],MomArm^[i],Error);
        val(Str8[3],Bangle^[i],Error);
        KneeAngle^[i]:=DegToRad(KneeAngle^[i]);
        Bangle^[i]:=DegToRad(BAngle^[i]);;
{Writeln(lst,KneeAngle^[i]:8:2,
        MomArm^[i]:8:2,
        Bangle^[i]:8:2);}
Readln(storefile);
i:=i+1;
end;
NFrame:=i-1;
Close(StoreFile);
end;
Writeln(lst);

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT TORQUE FILE NAME :');
if not GReadStr(FileName) then halt(1);
if Length(FileName)>1 then
begin
Assign(StoreFile,FileName);
Reset(StoreFile);
i:=1;
Readln(storefile);
while not eof(storefile) do
begin
for k:=1 to 3 do
begin
Read(StoreFile,Str10[k]);
Read(StoreFile,ch);
end;
        val(Str10[2],AngVel^[i],Error);
        val(Str10[3],Torque^[i],Error);
Readln(storefile);
i:=i+1;
end;
NPoint:=i-1;
Close(StoreFile);
end;
Writeln(lst);

GDelLineXY(Xg(1),Yg(1));
```

```
GwriteXY(Xg(1),Yg(1),'INPUT BODY MASS (Kg) :');
if not GReal(BMass) then halt(1);

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT SEGMENT LENGTH (m) :');
if not GReal(SegLen) then halt(1);

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT ARM POSITION (5,6,7) :');
if not GInt(IArmPos) then halt(1);


GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT FULL EXTENSION (Degrees) :');
if not GReal(FExt) then halt(1);
FExt:=DegToRad(Fext);

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT TIBIAL PLATEAU-TIBIAL AXIS ANGLE (Degrees) :');
if not GReal(Tt) then halt(1);
Tt:=DegToRad(Tt);

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT ANG. POSITION FILE NAME :');
if not GReadStr(FileName) then halt(1);
if Length(FileName)>1 then
begin
Assign(StoreFile,FileName);
Reset(StoreFile);
i:=1;
Readln(storefile);
while not eof(storefile) do
begin
for k:=1 to 3 do
begin
Read(StoreFile,Str10[k]);
Read(StoreFile,ch);
end;
        Time^[i]:=(i-1)*0.01;
        val(Str10[2],GravAngle^[i],Error);
Readln(storefile);
i:=i+1;
end;
NPoint:=i-1;
Close(StoreFile);
end;
writeln(lst,Dt:10:5);

Dt:=0.01;
writeln(lst);


{Extension}
KneeAngle^[NFrame+1]:=KneeAngle^[NFrame]+
```

```
                    (KneeAngle^[NFrame]-KneeAngle^[NFrame-1]);
MomArm^[NFrame+1]:=MomArm^[NFrame]+
                    (MomArm^[NFrame]-MomArm^[NFrame-1]);
BAngle^[NFrame+1]:=BAngle^[NFrame]+
                    (Bangle^[NFrame]-BAngle^[NFrame-1]);

NPoint:=Npoint-1;
sf:=0;
SplineFunction(Time,AngVel,a,b,c,d,NPoint,sf);
for i:= 1 to NPoint-1 do
SplFit(Time,AngVel,a,b,c,d,Time^[i],AngAcc^[i],NPoint,sf,1);
NPoint:=Npoint-1;
for i:=1 to Npoint do begin
Data[i,1]:=Time^[i];
Data[i,2]:=AngAcc^[i];
end;

ClearDevice;
   FindWorld(data,1,NPoint,1,2);
   DrawArray(data,1,NPoint,2,2,true,false);

readln;
case IArmPos of
5:IInpArm:=0.451223;
6:IInpArm:=0.519155;
7:IInpArm:=0.593081;
end;

case IArmPos of
5:ResMomArm:=0.325;
6:ResMomArm:=0.355;
7:ResMomArm:=0.385;
end;

ISeg:=(0.061*BMass)*sqr(SegLen*0.735);

for i:=1 to Npoint do
begin
sf:=0;
Flex:=FExt-GravAngle^[i];
{write(lst,Flex:10:2);}
SplineFunction(KneeAngle,MomArm,a,b,c,d,NFrame+1,sf);
if Flex<KneeAngle^[NFrame] then
SplFit(KneeAngle,MomArm,a,b,c,d,Flex,PatMomArm,NFrame+1,sf,0)
else
SplFit(KneeAngle,MomArm,a,b,c,d,KneeAngle^[NFrame],PatMomArm,NFrame+1,sf,0);

SplineFunction(KneeAngle,BAngle,a,b,c,d,NFrame+1,sf);
if Flex<KneeAngle^[NFrame] then
SplFit(KneeAngle,BAngle,a,b,c,d,Flex,Tb,NFrame+1,sf,0)
else
SplFit(KneeAngle,BAngle,a,b,c,d,KneeAngle^[NFrame],Tb,NFrame+1,sf,0);


TanAcc:=SegRad*AngAcc^[i];
```

```
RadAcc: = SegRad*sqr(AngVel^[i]);
Fr: = Torque^[i]/ResMomArm;
Fg: = (BMass*0.061)/9.81;
Tg: = Tt-GravAngle^[i];
Tr: = (pi/2)-Tt;
Tm: = pi-Tb;
Fm^[i]: = (Torque^[i] + IInpArm*AngAcc^[i] + ISeg*AngAcc^[i])/(PatMomArm/1000);
Fc^[i]: = Fr*sin(Tr) + Fm^[i]*sin(Tm)-Fg*sin(Tg) + BMass*0.061*(RadAcc*sin(Tt)-TanAcc*
cos(Tt));

Fs^[i]: = -Fr*cos(Tr) + Fm^[i]*cos(Tm)-Fg*cos(Tg) + BMass*0.061*(RadAcc*cos(Tt) + TanA
cc*sin(Tt));

Fm^[i]: = Fm^[i]/(BMass*9.81);
Fc^[i]: = Fc^[i]/(BMass*9.81);
Fs^[i]: = Fs^[i]/(BMass*9.81);
Data[i,2]: = Fm^[i];

end;

ClearDevice;
   FindWorld(data,1,NPoint,1,2);
   DrawArray(data,1,NPoint,2,2,true,false);
   for i: = 1 to Npoint do
   Data[i,2]: = Torque^[i];
   FindWorld(data,1,NPoint,1,2);
   DrawArray(data,1,NPoint,2,2,true,false);
   readln;
   ClearDevice;
      for i: = 1 to Npoint do
   Data[i,2]: = Fs^[i];
   FindWorld(data,1,NPoint,1,2);
   DrawArray(data,1,NPoint,-2,2,true,false);
      for i: = 1 to Npoint do
   Data[i,2]: = Fc^[i];
   readln;
   ClearDevice;
   FindWorld(data,1,NPoint,1,2);
   DrawArray(data,1,NPoint,-2,2,true,false);
readln;

GDelLineXY(Xg(1),Yg(1));
GwriteXY(Xg(1),Yg(1),'INPUT ANG. POSITION FILE NAME :');
if not GReadStr(FileName) then halt(1);
if Length(FileName) > 1 then
begin
Assign(StoreFile,Filename);
Rewrite(StoreFile);
for i: = 1 to Npoint do
writeln(StoreFile,Fext-GravAngle^[i]:10:5,',',Fm^[i]:10:5,',',
                Fs^[i]:10:5,',',Fc^[i]:10:5,',',
                Fr:10:5,',',Torque^[i]:10:5);
Close(StoreFile);
end;
```

```
FreeMem(GravAngle,(n + 2)*RealSize);
FreeMem(AngAcc,(n + 2)*RealSize);
FreeMem(AngVel,(n + 2)*RealSize);
FreeMem(Fm,(n + 2)*RealSize);
FreeMem(Fs,(n + 2)*RealSize);
FreeMem(Fc,(n + 2)*RealSize);
FreeMem(Time,(n + 2)*RealSize);
FreeMem(Torque,(n + 2)*RealSize);
FreeMem(KneeAngle,(n + 2)*RealSize);
FreeMem(MomArm,(n + 2)*RealSize);
FreeMem(BAngle,(n + 2)*RealSize);
FreeMem(a,(n + 2)*RealSize);
FreeMem(b,(n + 2)*RealSize);
FreeMem(c,(n + 2)*RealSize);
FreeMem(d,(n + 2)*RealSize);
end.
```

APPENDIX II

TABLES OF RESULTS

Table II-1. Maximum resultant moment (Nm) during visual feedback (VF) and no-visual feedback (NVF) conditions during isokinetic knee extension and flexion at 1.06 and 3.14 rad·s^{-1} (Number of subjects N=10).

N	1.06 rad·s^{-1}				3.14 rad·s^{-1}			
	Extension		Flexion		Extension		Flexion	
	VF	NVF	VF	NVF	VF	NVF	VF	NVF
1	284	268	162	155	211	189	110	103
2	231	226	135	131	154	154	91	104
3	216	190	114	106	149	139	87	89
4	214	184	105	77	136	146	72	62
5	228	210	144	141	162	171	110	103
6	155	144	82	89	128	130	56	65
7	271	264	135	137	196	190	111	107
8	195	165	124	116	97	104	62	68
9	245	245	117	95	170	163	89	96
10	209	180	99	112	105	108	77	63

Table II-2. Analysis of variance summary table of maximum resultant moment for different visual feedback, muscle group and speed of movement conditions.

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Visual Feedback	1	775.00	775.00	13.54
Residual	9	515.12	57.236	
Muscle Group	1	130007.80	130007.80	148.08
Residual	9	7901.37	877.93	
Speed of Movement	1	48659.13	48659.13	135.541
Residual	9	3231.00	359.00	
Visual Feedback by Muscle Group	1	189.25	189.25	5.78
Residual	9	294.87	32.76	
Visual Feedback by Speed of Movement	1	556.50	556.5	10.02
Residual	9	499.62	55.51	
Muscle Group by Speed of Movement	1	5628.00	5628.00	28.97
Residual	9	1748.12	194.23	
Visual Feedback by Muscle Group by Speed of Movement	1	137.62	137.62	1.64
Residual	9	745.50	83.83	

Table II-3. Knee Flexion/Extension moment ratio during visual feedback (VF) and no-visual feedback (NVF) conditions at 1.06 and 3.14 rad·s^{-1}

N	1.06 rad·s^{-1}		3.14 rad·s^{-1}	
	VF	NVF	VF	NVF
1	0.57	0.58	0.52	0.54
2	0.58	0.58	0.59	0.68
3	0.53	0.56	0.58	0.64
4	0.49	0.42	0.53	0.42
5	0.63	0.67	0.68	0.60
6	0.53	0.62	0.44	0.50
7	0.50	0.52	0.57	0.56
8	0.63	0.70	0.64	0.65
9	0.48	0.39	0.52	0.59
10	0.47	0.62	0.73	0.58

Table II-4. Analysis of variance summary table for knee flexor/extensor moment ratio for different visual feedback and speed of movement conditions.

Source of Variation	d.f.	Sum of Squares	Mean Square	F-ratio
Visual Feedback	1	0.001	0.001	0.521
Residual	9	0.019	0.002	
Speed of Movement	1	0.006	0.006	1.208
Residual	9	0.045	0.005	
Visual Feedback by Speed of Movement	1	0.002	0.002	0.550
Residual	9	0.034	0.004	

Table II-5. Digital filter cut-off frequency for the smoothing of the angular position-time data. The cut-off frequency was determined using the method described in Chapter 4.

N	Digital Filter Cut-off Frequency (Hz)			
	0.52 $rad{\cdot}s^{-1}$	1.57 $rad{\cdot}s^{-1}$	2.62 $rad{\cdot}s^{-1}$	3.66 $rad{\cdot}s^{-1}$
1	2.8	3.8	4.4	5.1
2	3.0	4.0	4.4	4.5
3	2.7	4.1	5.3	5.7
4	2.7	3.8	4.0	4.4
5	2.3	4.0	4.8	4.8
6	2.9	4.3	4.9	5.2
7	3.0	4.6	4.0	4.9
8	2.6	3.8	4.2	4.5

Table II-6. Analysis of variance summary table for digital filter cut-off frequencies at different preset angular velocities of movement.

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Ang. Velocity	3	20.75	6.92	70.80
Residual	21	2.05	0.10	

Table II-7. Measurement error (mm) in control point (N=52) reconstruction using the Image intensifier-video system for joint kinematics described in Chapter 5. Different numbers of calibration points for the reconstruction were used, covering 20%, 40% and 60% of the field of view (FOV). Table continues on the next page.

Control Point No	% FOV		
	20%	40%	60%
1	0.171	0.208	0.281
2	0.176	0.155	0.270
3	0.106	0.067	0.111
4	0.263	0.186	0.123
5	0.342	0.286	0.264
6	0.285	0.315	0.336
7	0.171	0.098	0.029
8	0.078	0.129	0.092
9	0.397	0.401	0.398
10	0.221	0.221	0.309
11	0.482	0.389	0.381
12	0.373	0.166	0.201
13	0.465	0.256	0.309
14	0.582	0.506	0.605
15	0.323	0.240	0.192
16	0.074	0.134	0.174
17	0.191	0.219	0.314
18	0.247	0.229	0.228
19	0.107	0.131	0.190
20	0.194	0.183	0.170
21	0.389	0.410	0.477
22	0.055	0.117	0.254
23	0.357	0.367	0.288
24	0.150	0.224	0.329
25	0.473	0.382	0.252
26	0.231	0.246	0.247
27	0.316	0.279	0.152

28	0.270	0.213	0.273
29	0.159	0.175	0.288
30	0.334	0.353	0.304
31	0.481	0.495	0.481
32	0.350	0.267	0.175
33	0.176	0.121	0.060
34	0.246	0.250	0.219
35	0.148	0.250	0.317
36	0.144	0.254	0.305
37	0.380	0.443	0.444
38	0.094	0.156	0.237
39	0.329	0.342	0.370
40	0.175	0.145	0.132
41	0.421	0.410	0.388
42	0.196	0.135	0.148
43	0.252	0.340	0.514
44	0.154	0.312	0.341
45	0.212	0.225	0.340
46	0.340	0.345	0.447
47	0.087	0.037	0.079
48	0.343	0.320	0.389
49	0.167	0.122	0.199
50	0.210	0.183	0.204
51	0.193	0.255	0.297
52	0.053	0.118	0.199

Table II-8. Analysis of variance summary table for measurement error using different distortion uniformity (DU) constants.

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
DU Constant	2	17941.00	8970.50	2.43
Residual	102	377297.00	3698.99	

Table II-9. Digitising reliability (N = 10) of patellar tendon moment arm, knee flexion angle and patellar tendon-tibial plateau angle measurements using the method described in Chapter 6.

N	Patellar Tendon Moment Arm (mm)	Knee Flexion Angle (rad)	Patellar Tendon-Tibial Plateau Angle (rad)
1	35.95	1.60	1.55
2	36.69	1.58	1.56
3	36.90	1.59	1.55
4	36.22	1.62	1.56
5	36.05	1.59	1.56
6	36.47	1.62	1.57
7	35.69	1.63	1.55
8	35.75	1.62	1.56
9	36.48	1.63	1.56
10	36.91	1.63	1.55

Table II-10. Knee extensor moment arm (mm) at different knee positions measured using the method described in Chapter 6. Position 1 represents full extension and subsequent positions are at 0.0873 rad (5 degrees) increments of knee flexion.

Position	Subject No				
	1	2	3	4	5
1	39.06	33.44	34.91	30.17	31.49
2	39.12	35.44	35.40	30.92	32.52
3	40.08	37.31	35.99	31.70	33.62
4	41.03	38.68	36.43	32.50	34.90
5	41.59	39.52	36.64	33.32	36.32
6	41.97	39.95	36.74	34.23	37.88
7	42.18	40.11	36.83	35.30	39.72
8	42.00	40.15	36.91	36.60	41.58
9	41.61	40.16	36.98	37.76	42.68
10	41.24	40.11	37.02	38.01	42.98
11	40.80	39.96	37.02	37.60	42.74
12	40.06	39.65	37.00	37.20	42.08
13	38.87	39.15	36.95	36.79	41.06
14	37.56	38.36	36.89	36.13	39.72
15	36.53	37.08	36.81	35.28	38.08
16	35.75	35.12	36.73	34.42	36.22
17	35.09	32.95	36.64		34.29
18	34.43	31.65	36.56		
19		30.79	36.47		

Table II-11. Patellar tendon-tibial plateau angle (rad) at different knee positions measured using the method described in Chapter 6. Position 1 represents full extension and subsequent positions are at 0.0873 rad (5 degrees) increments of knee flexion.

Position	Subject No				
	1	2	3	4	5
1	1.84	1.90	2.06	1.87	2.10
2	1.83	1.88	2.05	1.87	2.09
3	1.81	1.88	2.04	1.86	2.07
4	1.80	1.87	2.01	1.84	2.04
5	1.80	1.86	1.98	1.83	2.02
6	1.79	1.86	1.95	1.81	2.00
7	1.77	1.85	1.93	1.77	1.98
8	1.76	1.84	1.91	1.74	1.94
9	1.75	1.84	1.88	1.72	1.92
10	1.74	1.83	1.84	1.70	1.90
11	1.73	1.81	1.79	1.68	1.87
12	1.72	1.79	1.73	1.66	1.84
13	1.71	1.77	1.69	1.63	1.80
14	1.69	1.74	1.65	1.59	1.76
15	1.67	1.70	1.63	1.54	1.73
16	1.63	1.64	1.60	1.49	1.69
17	1.60	1.59	1.57		1.64
18	1.56	1.57	1.54		
19		1.56	1.50		

Table II-12. Maximum resultant moment (Nm) at different angular velocities.

Subject No	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
1	272	240	225	215
2	258	203	167	157
3	179	173	162	153
4	196	182	166	156
5	226	189	156	149

Table II-13. Analysis of variance summary table for maximum resultant moment at different angular velocities

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Angular Velocity	3	10772.38	3590.79	17.90
Residual	12	2407.12	200.59	

Table II-14. Angular position (rad of knee flexion) of maximum resultant moment.

Subject No	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
1	1.1684	1.2684	1.0984	1.0684
2	0.8755	1.0433	1.1533	1.1284
3	1.2782	1.3182	1.2782	1.8582
4	1.0370	1.4530	1.4031	1.1530
5	1.2359	1.0959	1.2159	1.2359

Table II-15. Analysis of variance summary table for angular position of maximum resultant moment at different angular velocities.

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Angular Velocity	3	0.09	0.03	1.34
Residual	12	0.26	0.02	

Table II-16. Maximum muscular force (×BW) at different angular velocities.

Subject No	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
1	7.96	7.35	6.49	5.56
2	8.12	6.95	5.48	-
3	7.03	6.64	5.91	6.93
4	7.11	6.49	6.16	5.74
5	7.55	6.41	5.13	4.65

Table II-17. Analysis of variance summary table for maximum muscular force at different angular velocities

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Angular Velocity	3	11.15	3.72	13.70
Residual	12	3.26	0.27	

Table II-18. Angular position (rad of knee flexion) of maximum muscular force at different angular velocities.

Subject No	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
1	1.5084	1.2984	1.1284	1.0384
2	1.5655	1.3233	1.1533	1.1133
3	1.4868	1.7130	1.0530	1.1530
4	1.3182	1.3182	1.2782	0.5582
5	1.1759	1.3559	1.2559	1.1759

Table II-19. Analysis of variance summary table for maximum muscular force angular position at different angular velocities

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Angular Velocity	3	0.57	0.19	5.58
Residual	12	0.41	0.03	

Table II-20. Maximum compressive tibiofemoral force (xBW) at different angular velocities.

Subject No	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
1	8.00	7.36	6.47	5.54
2	7.92	6.93	5.40	-
3	7.08	6.63	5.95	6.80
4	7.14	6.52	6.22	5.80
5	7.53	6.4	5.09	4.59

Table II-21. Analysis of variance summary table for maximum compressive tibiofemoral force at different angular velocities

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Angular Velocity	3	11.28	3.76	13.57
Residual	12	3.33	0.28	

Table II-22. Angular position (rad of knee flexion) of maximum compressive tibiofemoral force at different angular velocities.

Subject No	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
1	1.5084	1.2984	1.1284	1.0388
2	1.5655	1.3233	1.1533	1.1133
3	1.4869	1.7130	1.2730	1.1530
4	1.3182	1.3182	1.2782	0.5582
5	1.1758	1.3558	1.2859	1.1759

Table II-23. Analysis of variance summary table of maximum compressive tibiofemoral force angular position at different angular velocities

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Angular Velocity	3	0.54	0.18	6.32
Residual	12	0.34	0.03	

Table II-24. Maximum shear tibiofemoral force (×BW) at different angular velocities.

Subject No	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
1	0.40	0.40	0.38	0.34
2	1.42	0.94	0.78	-
3	0.68	0.61	0.59	1.17
4	1.25	1.25	1.23	0.97
5	0.94	1.06	0.92	0.86

Table II-25. Analysis of variance summary table for maximum shear tibiofemoral force at different angular velocities

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Angular Velocity	3	0.06	0.02	0.64
Residual	12	0.40	0.03	

Table II-26. Angular position (rad of knee flexion) of maximum shear tibiofemoral force at different angular velocities.

Subject No	0.52 rad·s^{-1}	1.57 rad·s^{-1}	2.62 rad·s^{-1}	3.66 rad·s^{-1}
1	0.7184	0.6584	0.7884	0.7884
2	0.4655	0.7433	0.8233	0.6832
3	0.5069	0.6230	0.7830	0.7530
4	0.4182	0.4182	0.3682	0.2582
5	0.5959	0.8159	0.6559	0.5959

Table II-27. Analysis of variance summary table for maximum shear tibiofemoral force angular position at different angular velocities

Source of Variation	d.f.	Sum of Squares	Mean Square	F-Ratio
Angular Velocity	3	0.06	0.02	1.83
Residual	12	0.12	0.01	

APPENDIX III

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APPENDIX IV

RELEVANT PUBLICATIONS

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Isokinetic Dynamometry
Applications and Limitations

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Contents

Summary

Isokinetic contraction is the muscular contraction that accompanies constant velocity limb movements around a joint. The velocity of movement is maintained constant by a special dynamometer. The resistance of the dynamometer is equal to the muscular forces applied throughout the range of movement. This method allows the measurement of the muscular forces in dynamic conditions and provides optimal loading of the muscles.

However, during movements in the vertical plane, the torque registered by the dynamometer is the resultant torque produced by the muscular and gravitational forces. The error depends on the angular position and the torque potential of the tested muscle group. Several methods have been developed for the correction of gravitational errors in isokinetic data.

The torque output also contains artefacts that are associated with the inertial forces during acceleration and deceleration periods before the development of the constant preset angular velocity. For an accurate assessment of muscle function, only constant velocity data should be analysed.

The most frequently used isokinetic parameters are the maximum torque and the angular position where it was recorded, the torque output at different angular velocities of movement, the torque ratio of reciprocal muscle groups and the torque output during repeated contractions.

The unique features of isokinetic dynamometry are optimal loading of the muscles in dynamic conditions and constant preselected velocity of movement. These features provide safety in the rehabilitation of patients with muscular and ligamentous injuries. Isokinetic dynamometry has also been used for the training of various muscle groups in order to improve the muscular performance in dynamic conditions. The movement velocity of different activities can be simulated during training in order to improve the training effect.

Data acquisition and analysis have been improved by using computer systems interfaced to isokinetic dynamometers. Recently developed computer systems provide correction for gravitational and inertial errors, accurate computation of isokinetic parameters and real-time display of the torque output.

1. Definition of Isokinetics

The term 'isokinetics' is defined as the dynamic muscular contraction when the velocity of movement is controlled and maintained constant by a special device (Thistle et al. 1967). The resistance of the device is equal to the applied muscular torque over the range of movement. It is evident from the definition that isokinetic movements require the use of an electromechanical device capable of maintaining constant the velocity of movement. Thistle et al. (1967) presented the isokinetic contraction as a refinement of the controlled motion concept, where the velocity of movement is no longer an uncontrolled variable but may be preset according to the specific functional activity of the contracting muscle groups. The velocity control mechanism of the dynamometer is usually an electronic servomotor or a hydraulic valve. The velocity of movement is preset and the control mechanism is activated only when the preset velocity is attained by the moving limb. Any increase in muscular torque above this level results in the development of an equal-magnitude resistive force by the control mechanism of the dynamometer (Moffroid et al. 1969). The muscular force varies at different joint angles because of different biomechanical properties of the musculoskeletal system. With the isokinetic method, if maximum force is applied to the dynamometer over a range of movement, the resistance of the dynamometer is proportional to the muscular capacity at different joint angles, offering optimal loading of the muscles in dynamic conditions. Furthermore, isokinetic dynamometers, unlike gravity-loaded systems, do not store potential energy and therefore the return movement does not require eccentric contraction to control the return of the limb-lever arm system to the initial position (Thistle et al. 1967).

Hislop and Perrine (1967) compared muscle loading during isokinetic and isotonic (uncontrolled velocity) testing. The load applied to the contracting muscles during isotonic movements is maximal at points where the mechanical advantage of the muscles is minimal (e.g. at the limits of the range of movement in knee extension-flexion movements). On the other hand, during isokinetic movements the resistance is equal to the muscular capacity and therefore muscle loading is maximal at points where the mechanical advantage is maximal. With the isokinetic method the maximum muscular force that can be applied over a range of movement can be measured in dynamic conditions, provided that the preset velocity has been attained by the moving limb.

2. Gravitational Effect on Isokinetic Movements

During isokinetic tests involving movements in the vertical plane (e.g. knee extension-flexion), the forces acting on the limb-lever system are the muscular force (F_m) and the gravitational force (F_g) generated by the mass of the limb and the lever arm (fig. 1). The torque registered by the dynamometer is not the actual muscular torque but the torque generated by the resultant of the muscular and gravitational forces (Herzog 1988; Winter et al. 1981). Because the gravitational force remains constant for the same testing conditions, the percentage error in the recorded torque depends on the magnitude of the muscular force applied. In knee flexion movements the error is greater than the error in extension because the hamstrings are usually less powerful than the quadriceps, while the gravitational torque remains the same for both movements.

Winter et al. (1981) investigated the effect of gravitational forces on the recorded torque by the dynamometer during movements in the vertical plane. A correction factor was introduced to eliminate the gravitational error in the calculation of mechanical work generated by the muscular forces during knee extension-flexion movements. The correction factor was the work generated by the gravitational forces and it was determined using a piezoresistive accelerometer placed on the lever arm of the dynamometer. The magnitude of the gravitational error was demonstrated by comparing the mechanical work computed from the torque recorded by the dynamometer with the mechanical work corrected for the effect of gravitational forces. In the above study, 4 subjects performed 2 minutes of alternating knee extension and flexion on an isokinetic dynamometer at 20, 40 and 60 degrees per second. The error when the gravitational forces were not considered varied from 26 to 43% in extension and from 55 to 510% in flexion.

The effect of gravitational forces in the determination of the fatigue index was also investigated. Fatigue index was defined as the mean decline in maximum torque over 50 knee extensions at 180 degrees per second and was expressed as a percentage of the initial maximum torque (Thorstensson & Karlsson 1976). The error between corrected and uncorrected fatigue indices ranged from −6.5 to 26% and the correlation coefficient was r=0.80, indicating that the error is not a constant factor since the maximum torque is produced at different joint angles as muscular fatigue increases. It was suggested that the relationship between fatigue index and relative distribution of fast twitch fibres as reported by Thorstensson et al. (1976) could substantially change if the data were corrected for the effect of gravitational forces. The results of this study indicated the importance of gravity correction in the assessment of muscle function with isokinetic dynamometers. Nelson and Duncan (1983) presented a simplified method for the computation of the gravitational torque during knee extension-flexion movements. This method required only the recording of the gravitational torque generated by the weight of the limb-lever arm system at a specific angular position within the range of movement, while the limb-lever arm system is allowed to fall passively against the resistance of the dynamometer. The gravitational torque at every angular position is then calculated and this correction factor is added to the maximum torque produced by muscle groups opposed by gravity (quadriceps in the knee extension-flexion example) or subtracted from the recorded torque produced by muscle groups facilitated by gravity (e.g. hamstrings).

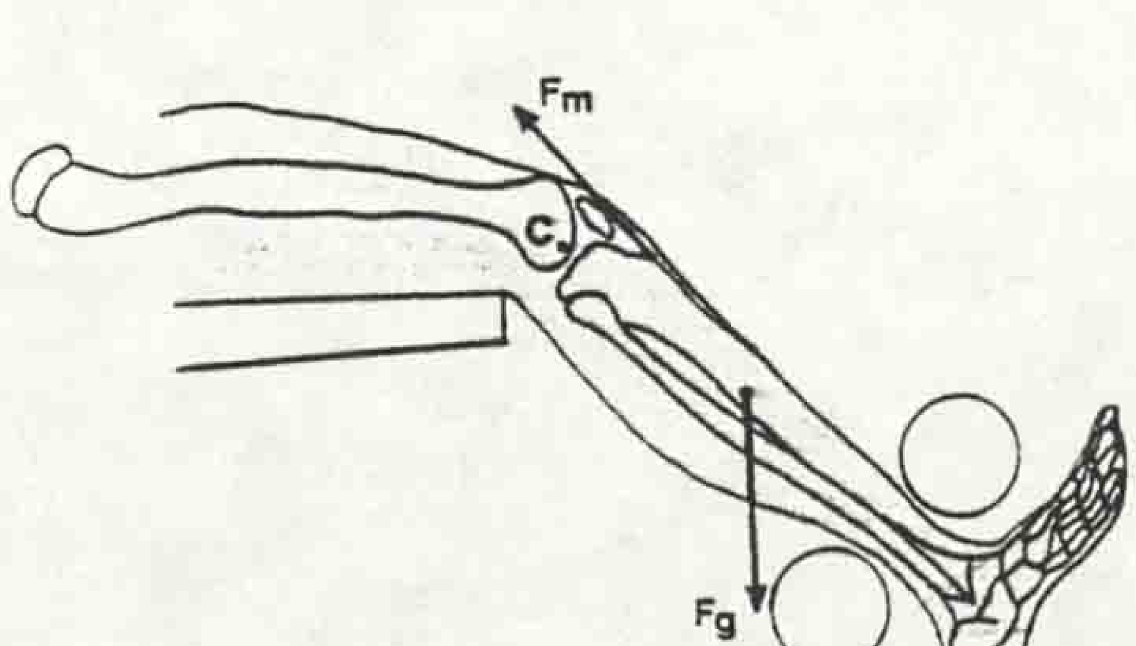

Fig. 1. Action of muscular (F_m) and gravitational (F_g) forces during isokinetic knee extension testing.

This method is accurate and simpler than the method proposed by Winter et al. (1981), requiring only the measurement of the gravitational torque in a specific position within the range of movement. However, in order to obtain valid results with this method the muscles must remain fully relaxed during the passive fall against the resistance offered by the dynamometer. In practice, several trials should be performed in order to obtain the actual gravitational torque, typically the minimum torque value recorded from the repeated trials.

3. Inertial Effect on Isokinetic Movements

The torque output during isokinetic movements frequently contains a prominent initial spike, which may be followed by torque oscillations of decreasing amplitude (Sapega et al. 1982). This phenomenon is usually referred to as the 'torque overshoot' and always appears in the initial part of the movement (fig. 2).

The feedback mechanism of the dynamometer is not activated if the velocity of movement is lower than the preset angular velocity. During this period the limb is free to accelerate as there is no resistive force exerted by the dynamometer (fig. 3). Subsequently the velocity of the limb is increased above the preset angular velocity. Sapega et al. (1982) filmed 2 tests on an isokinetic dynamometer using inert weights and a hip abduction movement. Analysis of the high-speed film revealed that during this free acceleration period, the angular velocity exceeded the preset velocity by 11% and 200% in the inert weights and hip abduction tests, respectively. When the feedback mechanism is activated, a resistive force is exerted by the dynamometer, in order to decelerate the limb to the level of the preset velocity (fig. 3). The overshoot in the torque output represents this 'reaction' of the dynamometer to the overspeeding limb-lever arm. Sapega et al. (1982) reported that in the hip abduction test the torque overshoot occurred during this deceleration period. It was calculated that the torque overshoot was the torque required by the dynamometer to produce the deceleration of the limb-lever system.

The torque of a rotating system is proportional to the angular acceleration and the moment of inertia of the system. During proximal joint testing, where a greater limb mass and a longer distance between the axis of rotation and the centre of mass are involved, the magnitude of the torque overshoot increases. Another factor affecting the magnitude of the torque overshoot is the mass of the dynamometer lever arm used for the test. The duration of the acceleration period is affected by the level of the present angular velocity and the power

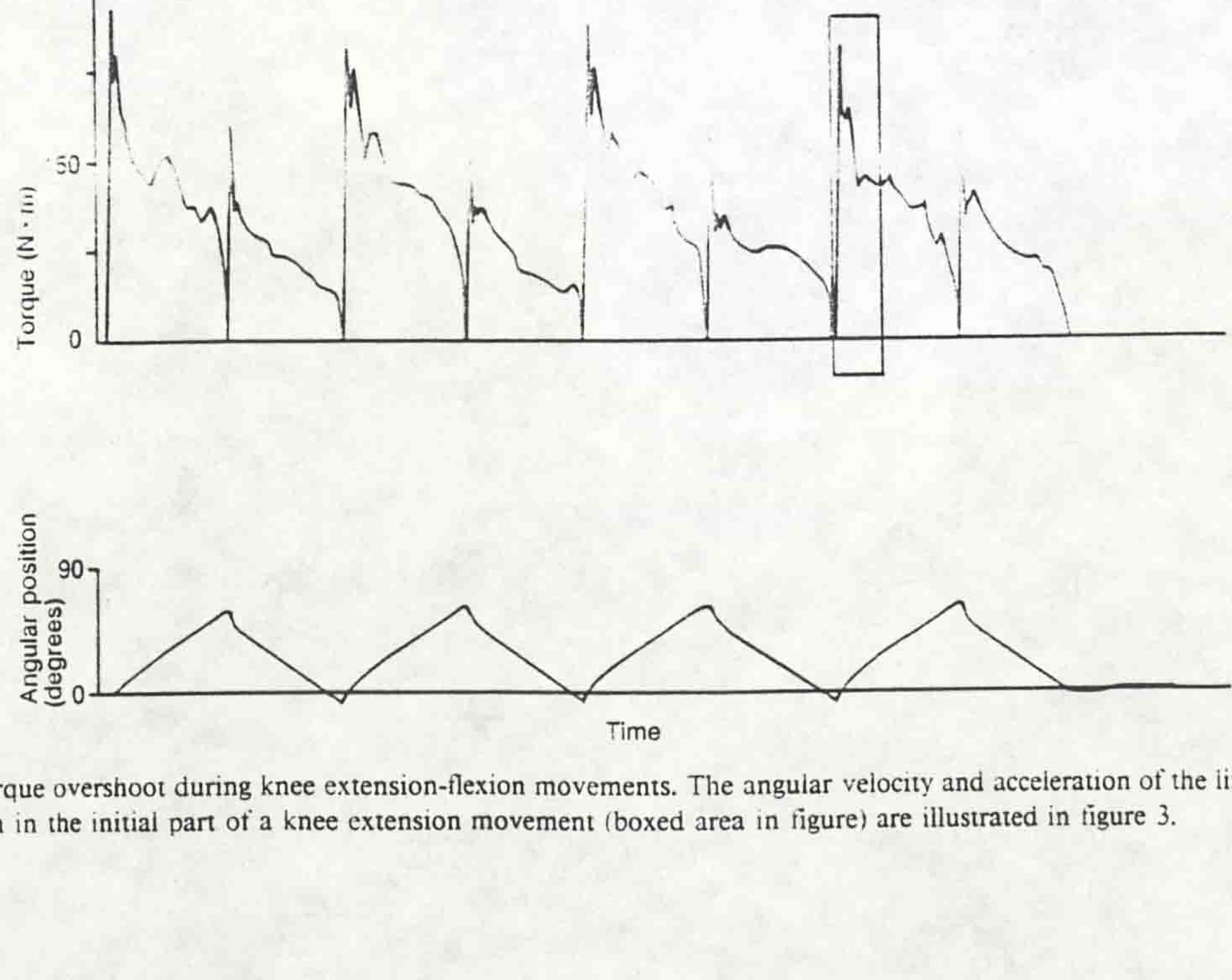

Fig. 2. Torque overshoot during knee extension-flexion movements. The angular velocity and acceleration of the limb-lever arm system in the initial part of a knee extension movement (boxed area in figure) are illustrated in figure 3.

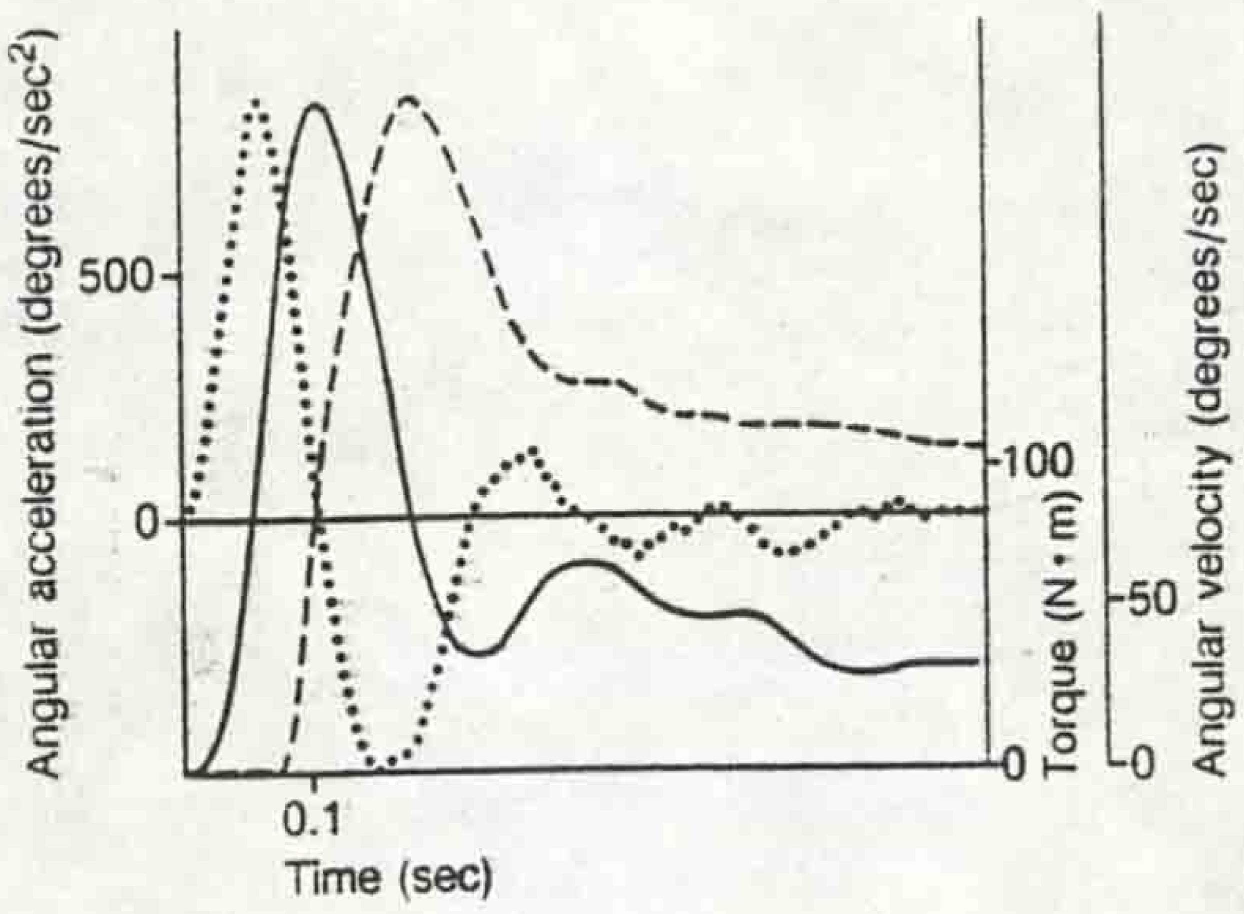

Fig. 3. Torque (– – –), angular velocity (——) and angular acceleration (· · · ·) of the limb-lever arm system during knee extension. The angular velocity of the dynamometer was preset at 30 °/sec. Notice that the preset velocity is exceeded by the velocity of movement during the free acceleration period and the torque overshoot is the torque required to decelerate the limb. The velocity of movement becomes constant and equal to the preset velocity after a series of acceleration and deceleration periods.

of the muscle group involved relative to the mass of the limb and the dynamometer lever arm.

During isokinetic testing, the overshoot is frequently the peak-point in the torque output. If this peak is interpreted as the subject's maximum torque, the muscular capability will be overestimated, influencing bilateral comparisons and reciprocal muscle group ratios.

The damp of the torque signal is a method that has been used to control the torque overshoot. Sinacore et al. (1983) investigated the effect of damp on isokinetic measurements and they reported that the damp resulted in a reduction of the torque signal amplitude throughout the range of movement and a displacement of the torque curve in the time axis. The effects of the damp method introduce errors in the maximum torque measurement and the torque-position relationship. Signal damp is therefore not an effective method for the elimination of the inertial artefact (Bemben et al. 1988; Murray 1986).

Gransberg and Knutsson (1983) connected a computer to the velocity control mechanism of a dynamometer in order to increase the acceleration period. The limb was resisted before the initiation of movement and during the acceleration period. The resisted acceleration method allowed a smooth transition from the acceleration to the constant velocity phase, with minimal torque oscillations. The acceleration period, however, was increased and the preset velocity was attained later in the range of movement.

Another method to overcome the inertial artefact is to use torque data only from constant velocity periods of the movement (Osternig et al. 1982; Perrine & Edgerton 1978). Since oscillations in the torque output represent alternating periods of acceleration and deceleration, artefact-free data can be obtained from the portion of the movement where the angular velocity remains constant and equal in magnitude with the preset velocity setting of the dynamometer.

4. *Isokinetic Parameters*

4.1 Maximum Torque

The maximum torque during isokinetic movements is a measure of the muscular force applied in dynamic conditions. Various testing protocols have been used for the assessment of maximum torque. The main difference between these protocols is the number of repetitions required in order to develop the maximum torque. Sawhill et al. (1982) investigated the number of repetitions required to achieve stable measurements during isokinetic testing at angular velocities ranging from 200 to 400 degrees per second. They suggested that 4 maximal repetitions are required in order to obtain stable isokinetic data. Johnson and Siegel (1978) reported that 3 submaximal followed by 3 maximal repetitions are essential for stable isokinetic data in knee extension movements. Appen and Duncan (1986) investigated the knee extensor and flexor muscles using 5 submaximal followed by 3 maximal repetitions. The testing protocol for the measurement of maximum torque of the knee extensors and flexors used by Jenkins et al. (1984) consisted of 5 maximal reciprocal (i.e. extension followed by flexion) repetitions, whereas Dibrezzo et al. (1985) used only 2 maximal repetitions. Baltzopoulos et al. (1988) used 6 reciprocal repetitions for the measurement of maximum torque in knee extension-flexion movements.

It is evident from the above studies that maximum torque is always evaluated from the first 2 to 6 maximal repetitions and is defined as the maximum single torque value measured during these repetitions.

However, Patton and Duggan (1987) defined maximum torque as the mean torque from 5 maximal repetitions and Morris et al. (1983) used the mean of 3 repetitions. The maximum torque depends on the angular position (i.e. the joint position) where it was recorded (Caiozzo et al. 1981: Osternig 1975: Osternig et al. 1983; Thorstensson et al. 1976). The mean torque calculated from torque values recorded at different angular positions is not a meaningful measure of muscle function, because there is no information about the angular position. This method is useful only when the torque value is recorded at a specific predetermined angular position in every repetition. In this case, however, the recorded torque at the predetermined specific angular position may not be the maximum torque in that repetition.

4.2 Angular Position

The angular position is important in the assessment of muscle function because it provides information about the mechanical properties of the contracting muscles. It can be used to evaluate the optimum joint angle for maximum muscular force. The maximum torque position is affected by the angular velocity of movement. Thorstensson et al. (1976) reported that during knee extensions the maximum torque occurred later in the range of movement as the preset angular velocity increased. Osternig et al. (1983) reported a transfer in the flexion maximum torque position from 32 to 61 degrees of knee flexion during an increase from 50 to 400 degrees per second, respectively. The transfer observed in the extension maximum torque position was from 87 to 63 degrees of knee flexion during an increase from 50 to 400 degrees per second. They also reported that with increasing velocity the maximum torque optimal position in flexion and extension tended to converge near the 60 degree position. Moffroid et al. (1969) also reported that the optimal position in extension was at 63 degrees.

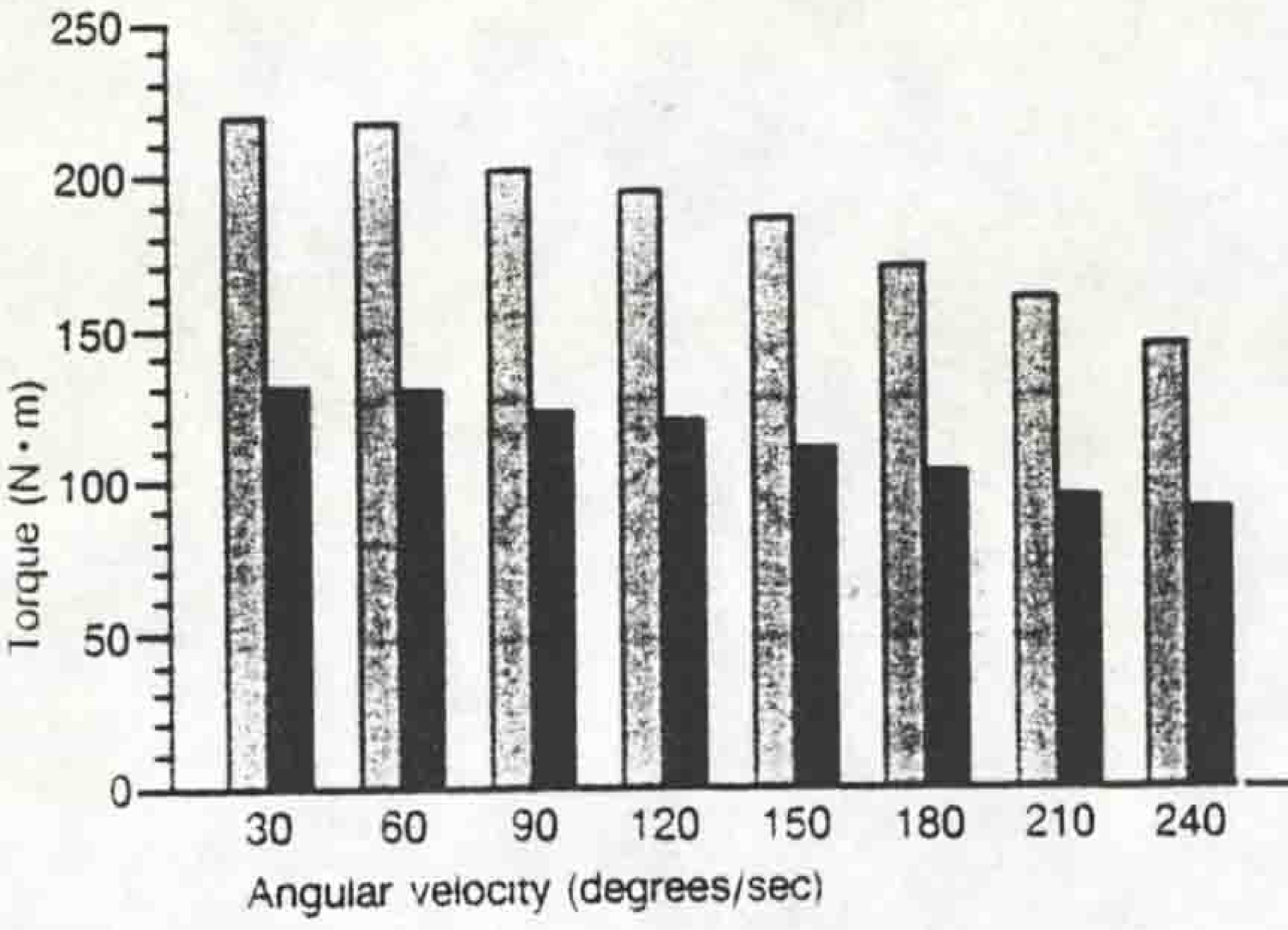

Fig. 4. Maximum torque at different angular velocities of knee extension (□) and flexion (■) [data from Baltzopoulos & Brodie 1987].

However, with increasing velocity, the acceleration period before the activation of the resistive mechanism of the dynamometer is longer and the limb may pass past the optimal position during this period. As a result the maximum torque tends to occur later in the range of movement with increasing velocity and not in the optimal joint position. Consequently, analysis of maximum torque data irrespective of angular position may lead to erroneous conclusions about muscle function.

4.3 Torque-Velocity Relationship

The muscular torque exerted during isokinetic testing decreases with increasing angular velocity of movement (Barnes 1980: Campbell 1979; Gregor et al. 1979; Moffroid et al. 1969; Osternig et al. 1983; Thorstensson et al. 1976: Yates & Kamon 1983) [fig. 4]. This decline in torque output has been attributed to different neurological activation patterns of motor units at different velocities (Barnes 1980: Milner-Brown et al. 1975). Moffroid et al. (1969) recorded the torque in knee extension movements at a specific position (65 degrees of knee flexion). With the velocity of movement increasing from 0 to 108 degrees per second, they reported a decrease in the torque output. However they observed an initial plateau in the torque output between 0 and 36 degrees per second. This plateau was attributed to possible human subject reluctance to exert more force at the slower velocities.

Perrine and Edgerton (1978) tested the torque of the knee extensors at angular velocities of movement ranging from 0 to 288 degrees per second. The torque was recorded at an angle of 70 degrees of knee flexion, in order for the muscle to develop maximum tension and attain the preset velocity. An intial plateau in the torque output was observed between 0 and 144 degrees per second and then the torque decreased with increasing velocity.

Lesmes et al. (1978) tested the maximum torque of the knee extensors and flexors irrespective of angular position at angular velocities ranging from 0 to 300 degrees per second. The maximum torque decreased with increasing velocity, but they also reported an initial plateau in the torque output between 0 and 60 degrees per second for both extension and flexion movements. In the above studies the obtained torque-velocity curve was compared to the classical *in vitro* force-velocity curve (Fenn & Marsh 1935: Hill 1938). The *in vivo* isokinetic torque-velocity curve was similar to the *in vitro* hyperbola at higher velocities of movement. In lower velocities, however, a plateau was observed in the torque output, whereas in the *in vitro* curve an increase in force occurs with decreasing velocity. This difference was attributed to a neural mechanism which limits the muscle tension development in lower velocities of movement during isokinetic evaluation of the torque-velocity relationship (Perrine & Edgerton 1978). However, Parker et al. (1983) tested knee extension at 54, 108, 162, 216, 270 and 300 degrees per second and concluded that the quadriceps torque-velocity relationship observed was in accordance with the Hill equation.

The Hill equation was derived from experiments with animal muscles free of the joint and therefore the force was acting in the same line as the actual tension development. This has very important implications in comparisons between the *in vitro* and *in vivo* force-velocity relationship. The velocity in the *in vitro* curve represents the actual velocity of the contraction, whereas the velocity in the *in vivo* curve represents the velocity of the moving limb under the influence of the contracting muscle. Hinson et al. (1979) reported that during elbow flexion and with the lower arm moving with constant angular velocity, the contraction velocity of the elbow flexors is not constant but contains only periods of acceleration and deceleration. They concluded that the term 'isokinetics' denotes the type of muscular contraction which accompanies constant angular velocity movements and not constant velocity of muscular contraction.

Another problem in the *in vivo* and *in vitro* force-velocity comparison is the angular position of the maximum torque during isokinetic testing. Theoretically the maximum torque in the *in vivo* testing is generated at a joint angle where the contracting muscle has an optimal mechanical advantage, provided that the muscle has developed maximum tension. Since it takes a finite amount of time for individual muscle fibres to develop maximum tension, the decrease in torque with increasing angular velocity could be a reflection of the muscle's inability to develop maximum tension at the optimal joint angle (Coyle et al. 1979). Increasing angular velocity would position the limb away from the optimal joint angle, when the muscle develops maximum tension.

Despite these problems, the torque-velocity relationship during isokinetic testing provides important information about muscle function at different movement velocities, especially when the muscle function is assessed in relation to the velocity of a particular activity.

4.4 Reciprocal Muscle Group Ratio

The reciprocal muscle group ratio is an indicator of muscular balance or imbalance around a joint. The hamstring to quadriceps ratio of the knee joint is one of the more important parameters in isokinetic assessment because the knee is one of the largest and most complex joints in the human body and its normal function is important for injury prevention. It has been suggested that the hamstring to quadriceps ratio is more important than the maximum torque in the assessment of muscle function (Campbell & Glenn 1982). Goslin and Charteris (1979) tested the knee extension-flexion movement of 60 untrained subjects at 30 degrees per second and reported a hamstring to quadriceps ratio of 0.44. Gilliam et al. (1979) tested high school football players at 30 and 180 degrees per second and found hamstring to quadriceps ratios of 0.60 and 0.77, respectively. Scudder (1980) tested the knee extensors and flexors of 10 normal untrained subjects and reported an increase in the hamstring to quadriceps ratio from 0.56 to 0.62 with an increase in the angular velocity from 0 to 72 degrees per second. A similar increase was reported by Davies et al. (1981) using professional football players. The ratio was increased from 0.61 at 45 degrees per second to 0.80 at 300 degrees per second. Wyatt and Edwards (1981) reported a similar increase with female subjects, from 0.71 at 60 degrees per second to 0.85 at 300 degrees per second. Housh et al. (1984) reported that the hamstring to quadriceps ratios in female throwers, jumpers, middle distance runners and sprinters were 0.70, 0.75, 0.81 and 0.71, respectively, at 180 degrees per second. Dibrezzo et al. (1985) reported that the mean ratio of 241 females between the age of 18 and 28 years was 0.54 at 60 degrees per second.

It is evident from the above studies that hamstring to quadriceps ratio is affected by age, sex and activity. It is also evident that the ratio is increased with an increase in the angular velocity of movement, indicating a possible decline in the relative quadriceps activity. However, it is important to note that the isokinetic data in the above studies were not corrected for the effect of gravity.

Schlinkman (1984) reported that the hamstring to quadriceps ratio of high school football players increased from 0.54 at 60 degrees per second to 0.67 at 300 degrees per second, but when the extension and flexion torque was corrected for the effect of gravity, the ratio was decreased by 8 to 12%. Appen and Duncan (1986) computed the corrected and uncorrected ratio in male track athletes

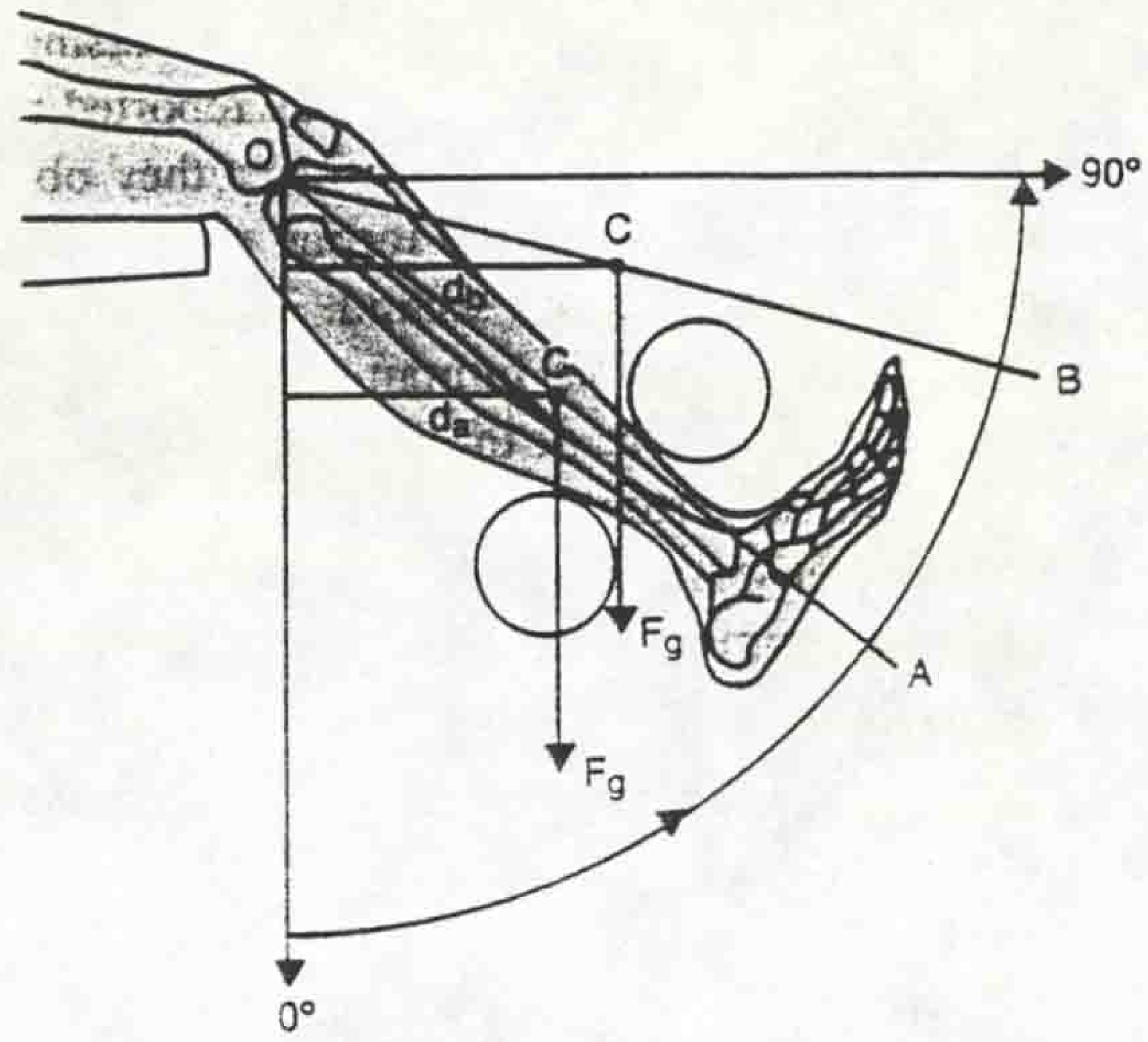

Fig. 5. The gravitational torque in position B. $T_b = F_g \cdot d_b$ and is greater than the torque in position A. $T_a = F_g \cdot d_a$ because $d_b > d_a$ while the gravitational force F_g remains the same throughout the range of movement.

at 60, 180, 240 and 300 degrees per second. The results of this study demonstrated that although the uncorrected ratios were similar to previous studies, indicating an increased hamstring to quadriceps ratio with increasing angular velocity, the gravity-corrected ratios remain constant with increasing angular velocity. The error in the computation of the ratio, with data not corrected for the effect of gravity, increased from 18.5% at 60 degrees per second to 37.7% at 300 degrees per second.

The error increase can be explained by the different angular position of the maximum torque with increasing angular velocity (Osternig et al. 1983; Thorstensson et al. 1976). The maximum torque is generated at increased knee joint angle with increasing velocity. The gravitational torque also increases with increasing knee joint angle because the horizontal distance between the centre of mass of the limb-lever arm system and the vertical axis of the dynamometer is increasing (fig. 5). In order to compute the gravity-corrected hamstring to quadriceps ratio the gravitational torque is added to the denominator (quadriceps) and subtracted from the numerator (hamstrings) resulting in a decrease of the ratio magnitude. At decreased knee joint angles the gravitational torque is minimal and the error is smaller. With increased knee joint angle, the gravitational torque increases, resulting in a further decrease of the hamstring to quadriceps ratio and a greater error.

It is evident from these studies that the interpretation of the reciprocal muscle group ratio without considering the gravity effect results in erroneous conclusions about muscle function (Fillyaw et al. 1986). Consequently, conclusions of previous studies with data uncorrected for the effect of gravity must be treated with caution, because the effect of the gravitational error in the validity of the results is unknown.

4.5 Muscular Endurance

Muscular endurance is the ability of the contracting muscles to perform repeated contractions against a load. The muscular endurance in dynamic conditions using isokinetic dynamometers is assessed by computing a fatigue index. However, different testing protocols and definitions have been used for the determination of the fatigue index. The testing protocol used by Thorstensson and Karlsson (1976) consisted of 50 maximal contractions of the knee extensors. Muscular endurance was assessed by expressing the mean torque from the last 3 contractions as a percentage of the mean torque from the initial 3 contractions. Patton et al. (1978) investigated the shape of fatigue curves using repeated contractions to exhaustion. Fatigue index was expressed as the time required for muscular exhaustion. Barnes (1981), in a similar study, used a testing protocol consisting of 10 maximal contractions and the fatigue index was computed by expressing the maximum torque in the last contraction as a percentage of the maximum torque during the 10 contractions. It is evident from the above studies that there is no standardised testing protocol and definition for the fatigue index and the assessment of muscular endurance. Patton and Duggan (1987) examined the relationship between the muscular endurance test introduced by Thorstensson and Karlsson (1976) and the 30-second Wingate test. No relationship was reported between fatigue indices measured by the 2 tests. However, the isokinetic data were not corrected for the effect of gravity. Baltzopoulos et al. (1988) defined fatigue index as the decline in maximum torque over time, using 30 seconds of repeated reciprocal contractions with gravity-corrected data. The results of this test were compared with the fatigue index from the 30-second Wingate test. A significant correlation ($r=0.86$, $p<0.001$) was found between the fatigue indices from the two tests.

The difference in angular position of the maximum torque and the reduction of the angular velocity with muscle fatigue may have an effect in the computation of fatigue index. The work performed is a more representative measure of muscle function because it takes into account the force output throughout the range of movement. However, Burdett and Swearingen (1987) computed the ratio of the work produced during the last 5 of 25 maximal contractions to the work during the first 5 and reported that the reliability of the work ratio was low and that the number of contractions to 50% of the initial torque level was a more reliable measurement of muscular endurance.

5. *Applications of Isokinetics*

5.1 Rehabilitation and Assessment

The advantages of isokinetic systems include variable resistance equal to the applied muscular force, and constant preselected velocity of movement. These unique features provide safety when used for rehabilitation of patients with muscular and ligamentous injuries and accuracy in the assessment of muscular performance at different functional velocities of movement.

The purpose of rehabilitation programmes following injury or surgery is to restore normal muscle function of the affected limb. However, the force-velocity relationship during isokinetic movements and the velocity specific training effects on muscular strength reported for normal subjects (Caiozzo et al. 1981; Coyle et al. 1981; Jenkins et al. 1984; Parker et al. 1983) had a considerable effect on the selection of training velocity in rehabilitation programmes. Parker (1982) proposed the use of an appropriate velocity according to the condition of the injured muscle. The velocity was calculated by substituting in Hill's equation for the force-velocity relationship the maximum isometric torque that a patient is able to exert. Sherman et al. (1982) recommended rehabilitation velocities ranging from 60 to 300 degrees per second in order to ensure that both muscle fibre types were recruited and trained. Grimby (1985) suggested that the training velocity should depend on the phase of rehabilitation, type and degree of muscular hypotrophy and individual reaction at different velocities.

Campbell and Glenn (1982) assessed the effect of rehabilitation programmes for patients with chondromalacia, ligamentous repairs and meniscectomies with isokinetic testing. An isokinetic dynamometer was used to evaluate the maximum torque and hamstring to quadriceps ratio at 30 and 180 degrees per second, before and after the rehabilitation programme of the affected and unaffected limb. Although the rehabilitation programme consisted of isometric contractions and functional activities of the affected limb, a significant increase in the isokinetic maximum torque was reported. The isokinetic test revealed that the extension maximum torque and the hamstring to quadriceps ratio were not rehabilitated to the levels of the unaffected limb but the opposite was found for the flexion maximum torque.

Armstrong et al. (1983) investigated the reliability and safety features of isokinetic dynamometry in patients with multiple sclerosis. The maximum torque and hamstring to quadriceps ratio of the right knee were evaluated for 10 patients and 20 healthy subjects at angular velocities ranging from 0 to 270 degrees per second. In order to assess the reliability of isokinetic dynamometry, the maximum torque of 3 patients was evaluated after 0, 6 and 11 weeks. The results demonstrated that the maximum torque of patients with multiple sclerosis was significantly lower than the maximum torque of healthy subjects, although the torque curves were similar in shape. The maximum torque output of 50% of the patients at 270 degrees per second was 0 N · m. Hamstring to quadriceps ratios at all angular velocities were not significantly different from the respective ratios of the healthy subjects. The test-retest reliability for patients with multiple sclerosis was 0.99 ($p<0.001$) with both tests performed in the same week. However, the maximum torque was variable after 6 and 11 weeks and it was suggested that when such patients are not familiar with isokinetic equipment, an increase in the maximum torque may not reflect an improvement in the functional condition, but a learning effect or familiarisation with the isokinetic apparatus. Watkins et al. (1984) examined 15 hemiparetic patients and 15 healthy subjects. They performed 5 bilateral consecutive repetitions of the knee extensors and flexors muscles at 30 degrees per second in order to evaluate the maximum torque and hamstring to quadriceps ratio. The maximum torque of the unaffected side of the patients was significantly lower than in healthy subjects and furthermore the maximum torque of the affected side was significantly lower than the unaffected side. The accuracy of isokinetic testing in detecting muscle function deficiencies was documented by evaluating the muscle function of the affected side of the patients with manual muscle testing. Although the maximum torque and hamstring to quadriceps ratio of the affected side using isokinetic dynamometry were significantly lower than healthy subjects, the recorded grades of manual testing were 'good' to 'normal', indicating the superiority of isokinetic dynamometry in detecting muscle function deficiencies.

Burnie and Brodie (1986) assessed the effectiveness of a rehabilitation programme for knee injury using isokinetic dynamometry. Muscle function of the knee extensors and flexors of a professional football player was assessed with an isokinetic dynamometer 12 weeks after an injury which involved the medial collateral ligament and both the anterior and posterior cruciate ligaments. Bilateral testing of the knee extensors and flexors was performed at 60 degrees per second 12, 20 and 27 weeks after surgery. The use of an isokinetic dynamometer was also included in the rehabilitation programme during this period. The results indicated a significant increase in extension and flexion maximum torque of the operated knee (304 in flexion and 344% in extension), reducing the bilateral deficit from 52 to 16% in flexion and from 70 to 26% in extension. The range of movement was increased from 40 to 106 degrees and the hamstring to quadriceps ratio was improved from 1 to 0.87 after the rehabilitation programme.

Similar improvements were reported by Thomee et al. (1987) after rehabilitation of patients with anterior cruciate ligament injury. The maximum torque of the knee extensors and flexors at 30, 60, 120, 180 and 300 degrees per second of 16 patients was evaluated before and after a rehabilitation programme of 8 weeks. The rehabilitation programme consisted of knee extension and flexion at 60 and 180 degrees per second using an isokinetic dynamometer. After the rehabilitation programme the operated knee extension maximum torque increased from 56 to 74% and the flexion maximum torque from 78 to 102% compared with the non-operated knee.

The results of the above studies indicate that isokinetics is an effective rehabilitation method and is also of value for rehabilitation assessment. It is also evident that the most frequently used isokinetic parameters in the assessment of muscle function are maximum torque and reciprocal muscle group ratio. However, the magnitude of errors in the evaluation of these parameters if the isokinetic data are not corrected for gravitational and inertial effects (Sapega et al. 1982; Winter et al. 1981) demonstrate the importance of appropriate filters in order to eliminate potential errors. Furthermore, the maximum torque of an injured or operated joint is very low, increasing further the magnitude of the percentage gravitational error. A typical example is the previously reported result by Armstrong et al. (1983) that many patients with multiple sclerosis were unable to produce extension and flexion maximum torque greater than 0 N · m at 275 degrees per second. Assuming that the limb was moving in extension for example with a constant velocity of 275 degrees per second, it is evident that the knee extensors were generating force and thus a finite amount of torque was applied to the dynamometer, but 0 N · m was recorded. In this case the muscular torque was either equal in magnitude with the gravitational torque and not 0 N · m as reported, or it was less than the torque signal resolution of the isokinetic system.

Isokinetic dynamometers have also been used to assess the effects of injuries on muscle function and the effect of various treatment and rehabilitation techniques. Among other applications, isokinetic dynamometry has been used to examine the synergetic action of the anterior cruciate ligament and the thigh muscles in maintaining joint stability (Solomonow et al. 1987), to assess muscle function and evaluate rehabilitation programmes for knee ligament injuries (Grimby et al. 1980; LoPresti et al. 1988; Murray et al. 1984; Noyes et al. 1987), to examine muscle function after bilateral femoral osteotomy (Olerud et al. 1984) and for arthroscopic meniscectomy with and without tourniquet control (Thorbland et al. 1985). It has also been used to evaluate the efficiency of a rehabilitation programme after arthroscopic meniscectomy (Shields et al. 1987), to assess the function of the knee extensors and flexors after diagnostic and operative arthroscopy and open meniscectomy (Hamberg et al. 1983), to examine the effect of patella brace on quadriceps torque (Lysholm et al. 1984), to examine the results of transcutaneous neural stimulation after arthroscopic knee surgery (Jensen et al. 1985) and to assess muscle function after lateral reconstruction for anterolateral rotary instability of the knee (Fleming et al. 1983). Mira et al. (1980) examined the shape of the isokinetic quadriceps torque in order to determine the type of femoral shaft fracture and the level of injury. Knutsson and Martensson (1985) used isokinetic measurements to examine the origin of hysterical paresis. Treatment methods for achilles tendon injuries have also been evaluated using isokinetic dynamometry (Beskin et al. 1987; Inglis et al. 1976; Nistor 1981; Pierre et al. 1984) and it has also been used for postoperative evaluation of shoulder dislocation (Miller et al. 1984) and assessment of trunk extensors and flexors in normal and low back dysfunction patients (Kishino et al. 1985; Mayer et al. 1985; Smidt et al. 1983).

5.2 Isokinetic Training

The constant preselected velocity during isokinetic movements allows the training and improvement of muscular performance in dynamic conditions. Isokinetic training at a specific angular velocity increases the maximum torque of the involved muscle groups at the training velocity (Lesmes et al. 1978). A transfer effect at other velocities (i.e. increased maximum torque at lower and higher velocities than the training velocity) has also been reported (Coyle et al. 1981; Lesmes et al. 1978). In these studies it was reported that maximum torque increased significantly at the training velocity and velocities below the training velocity. It was also reported that high velocity training has a better transfer effect to lower velocities than low velocity training to higher velocities of movement. Caiozzo et al. (1981) reported that high velocity training (240 degrees per second) produced increased maximum torque at lower velocities with an exception of 30 degrees per second. Jenkins et al. (1984) reported that training at 240 degrees per second produced improvements at 240 and 300 degrees per second, while training at 60 degrees per second produced improvements at 60 and 180 degrees per second. Garnica (1986) reported that improvements after low velocity training (60 degrees per second) occurred at a higher velocity (180 degrees per second) and that high velocity training increased the maximum torque at the training velocity only.

The improvement in muscular performance after isokinetic training has been explained by velocity-specific adaptation of motor units within the muscle (Milner-Brown et al. 1975; Sale et al. 1983) and velocity-specific adaptation within the nervous system (Barnes 1980; Sale et al. 1982). However, differences in the direction of the transfer effect can be explained by differences in sample size, muscle fibre distribution and training period and intensity.

5.3 Injury Prevention

In contrast to previous findings (Heiser et al. 1984; Mulder 1973; Slagle 1979), Grace et al. (1984) reported that an imbalance between right and left knee maximum torque or an imbalance in the hamstring to quadriceps ratio was not associated with increased incidence of knee joint injury. Preseason maximum torque and hamstring to quadriceps ratio of 206 male high school football players were evaluated with an isokinetic dynamometer at 60 and 240 degrees per second.

Maximum torque imbalance was defined as a difference between right and left knee of 10% or more. Hamstring to quadriceps ratio imbalance was defined as the difference between the mean and the actual ratio of 10% or more. Although an imbalance was detected for 33% of the tested subjects, no relationship was found between imbalance and joint injury susceptibility. However, further research is needed to examine the relationship between muscle imbalance assessed with the isokinetic method and injury (Grace 1985).

6. Computer Systems in Isokinetic Analysis

Manual analysis of isokinetic data involves the computation of the isokinetic parameters from the torque graph printed on a chart recorder. This method involves basic measurement techniques and can be time consuming and inaccurate. Furthermore the implementation of appropriate filters for the gravitational and inertial artefacts is restricted because of the amount and complexity of the mathematical computations involved (Watkins et al. 1984). The development of computer systems interfaced to isokinetic dynamometers provides a solution to the above problems and enhances the efficiency and accuracy of isokinetic dynamometry for training and rehabilitation.

Richards and Cooper (1982) described the interface of an Apple III microcomputer to a Cybex II isokinetic dynamometer. The isokinetic parameters computed from the isokinetic data include maximum torque, work, power, reciprocal muscle group ratio and range of movement. In order to avoid interpretation of torque overshoot as muscular torque, data sampled at the first 0.01 seconds of the movement were not included in the analysis. Data analysis time is approximately 10 seconds.

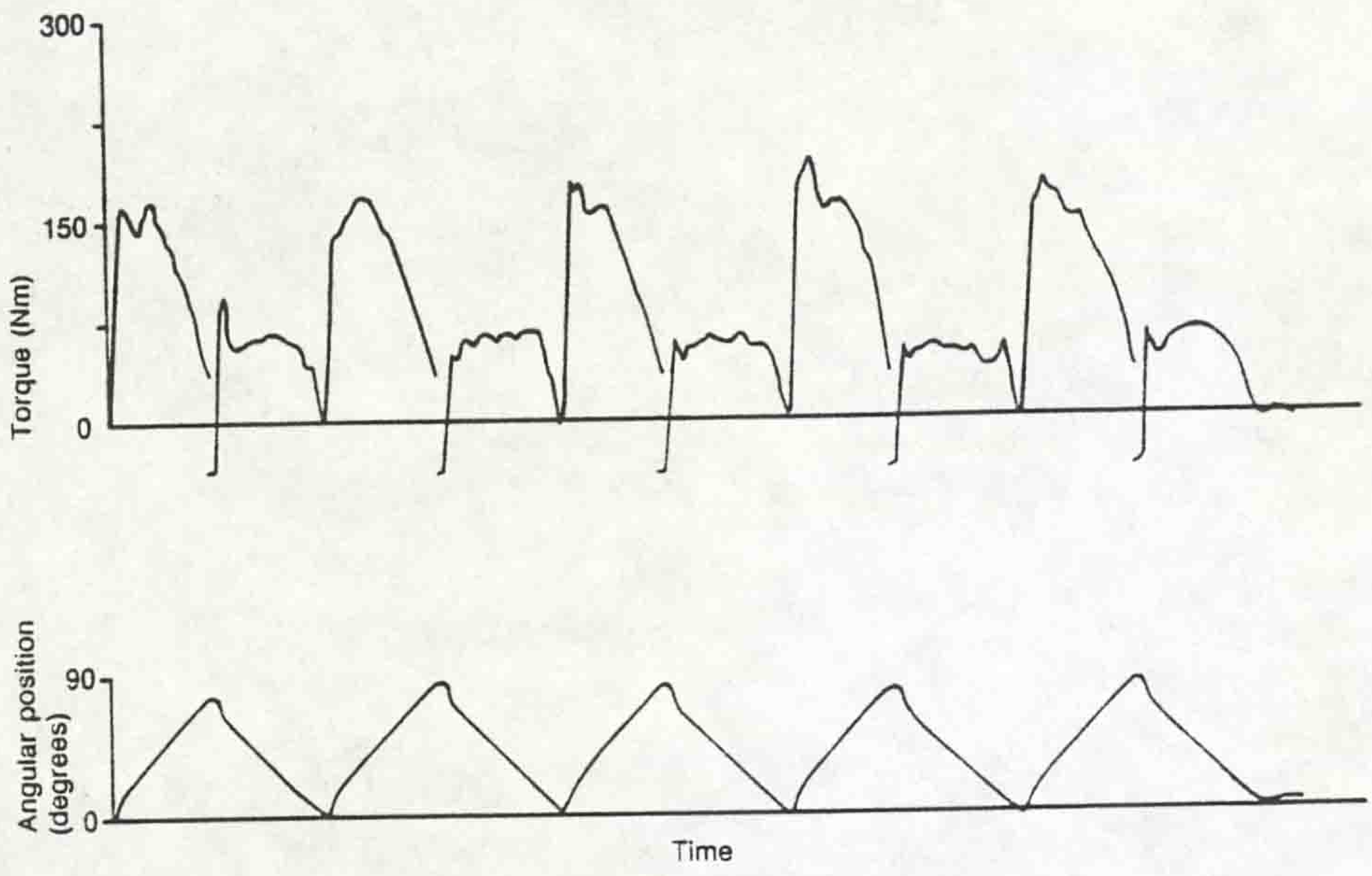

Fig. 6. Real-time display of the gravity-corrected torque and angular position during a knee extension-flexion test. Notice that at the end of extension movements a torque amount of about 30 N · m is registered by the system, representing the muscular torque required to maintain the limb-lever arm system in this upright position. The negative values at the start of flexion movements indicate that the muscular torque is applied in the opposite direction. Compare also with figure 2 where the torque output is not gravity-corrected.

The reliability of the system was determined by the intraclass correlation coefficient for the computation of torque, work and power. The reliability coefficients were greater than 0.99 ($p<0.001$), indicating reliable measurement of the isokinetic parameters.

Osternig et al. (1982) developed a computer system for data acquisition and analysis from a modified Orthotron isokinetic dynamometer. The angular velocity of movement is computed from the angular position data. With this method acceleration and deceleration phases can be identified, allowing the evaluation of maximum torque from constant velocity data.

Another computer system for the Cybex dynamometer was developed by Potash et al. (1983). An Apple II microcomputer was interfaced to the dynamometer. Two testing protocols for the evaluation of isokinetic parameters at 30 degrees per second from 6 repetitions or at 180 degrees per second from 20 seconds of continuous repetitions were implemented in the program. After data input completion the program evaluates maximum torque, power, reciprocal muscle group ratio and several timing parameters.

More recently Baltzopoulos (1988) has developed a computer system for the Akron isokinetic dynamometer which displays the gravity-corrected torque and the angular position in real time (fig. 6) and corrects the data for inertial errors before the computation of the isokinetic parameters described previously.

The replacement of manual data acquisition and analysis using computer systems, has reduced analysis time and computational error allowing the implementation of correction methods for any gravitational or inertial errors, and therefore enhancing the accuracy of isokinetic measurements.

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Brief Report

Development of a computer system for real-time display and analysis of isokinetic data

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Summary

An isokinetic analysis system was developed by interfacing an Akron isokinetic dynamometer to a microcomputer. The system was designed for real-time display of the muscular torque output in order to provide immediate visual feedback of the muscular performance. The muscular torque data were filtered for the effect of gravitational forces. The angular velocity of movement was monitored and only constant velocity data were used for analysis. Standard testing protocols for the assessment of muscle function in dynamic conditions have been implemented in the system.

Relevance

The system eliminates potential errors involved in isokinetic dynamometry and can be used for an efficient and accurate analysis of muscle function for rehabilitation purposes.

Key words: Isokinetic dynamometer, computer processing, real-time, rehabilitation

Introduction

Isokinetic dynamometers have been used for the assessment and improvement of muscle function for both rehabilitation and training[1–6]. With the isokinetic method the angular velocity of movement around a joint is maintained constant by an electromechanical dynamometer. The speed is preselected and the resistance of the dynamometer is equal to the muscular forces throughout the range of movement. In this way muscle loading is adapted to the muscular capacity.

However, the gravitational forces acting on the system affect the recording of the muscular forces by the dynamometer. The movement also contains periods of acceleration and deceleration before the development of the constant preselected velocity. Analysis of the isokinetic data without considering these problems results in erroneous conclusions about muscle function[9,10]. However, in a manual analysis of the data, the implementation of appropriate correction methods is restricted because of the amount and complexity of the mathematical computations involved.

The Akron dynamometer (Akron Therapy Products, Ipswich, UK) is a new isokinetic device for testing and training the major muscle groups. The operational features of the dynamometer include independent setting of the angular speed in the two directions of movement (e.g. extension–flexion), allowing simulation of the contraction of reciprocal muscle groups during different functional activities. This feature is critical in the rehabilitation of patients with musculoskeletal injuries or testing and training of athletes for different activities.

The purpose of this study was to develop a computer operated system for real-time display of the torque output and analysis of the isokinetic data by implementing correction methods for any gravitational and inertial errors. This will improve the efficiency of the dynamometer as a tool for the assessment of muscle function and the accuracy of the computed isokinetic parameters.

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Methods and Results

Instrumentation

The system consists of an Akron isokinetic dynamometer with a built-in A/D converter and an IBM compatible computer with a high resolution graphics card and a serial port. The dynamometer maintains the preselected speed of movement constant. The speed can be set independently for reciprocal muscle groups, ranging from 0 to 4·2 rad · s^{-1}. The torque applied to the dynamometer and the angular position of the moving limb are recorded by appropriate transducers. The analog signals of torque and angular position are converted to digital form by the A/D converter, giving a resolution of 0·40 N m and 0·5°. The output of the A/D converter is controlled by the computer.

System operation

The computer program was designed for real-time display of the torque output. The digital signals of torque and angular position are sampled by the computer at a rate of 290 samples per second. The computer, after the input of each sample, stores the values in memory for later processing and displays graphically the torque signal on the monitor, before the input of the next sample. With this method the user has immediate visual feedback of the muscular performance. After test completion the torque data are corrected for the effects of gravitational and inertial forces.

The torque registered by the dynamometer is not the actual muscular torque but the torque generated by the resultant of the muscular and gravitational forces. With the gravity correction method implemented in the present system, the limb-lever arm system is allowed to fall passively and the program samples the gravitational torque recorded by the dynamometer at a specific position within the range of movement (ROM). This information is used for the correction of the torque signal after data input completion[7]. The gravity corrected torque graph in extension (Figure 1, *upper left*), indicates that a finite amount of muscular torque is exerted at

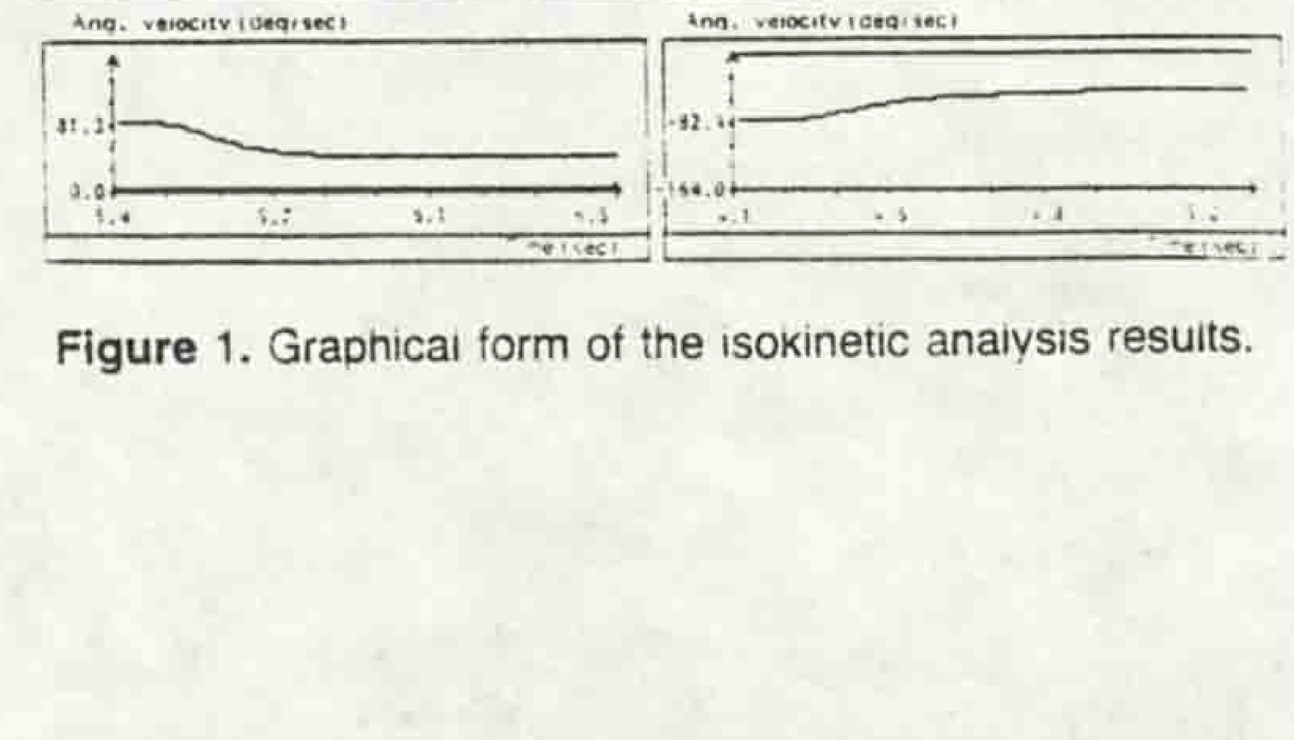

Figure 1. Graphical form of the isokinetic analysis results.

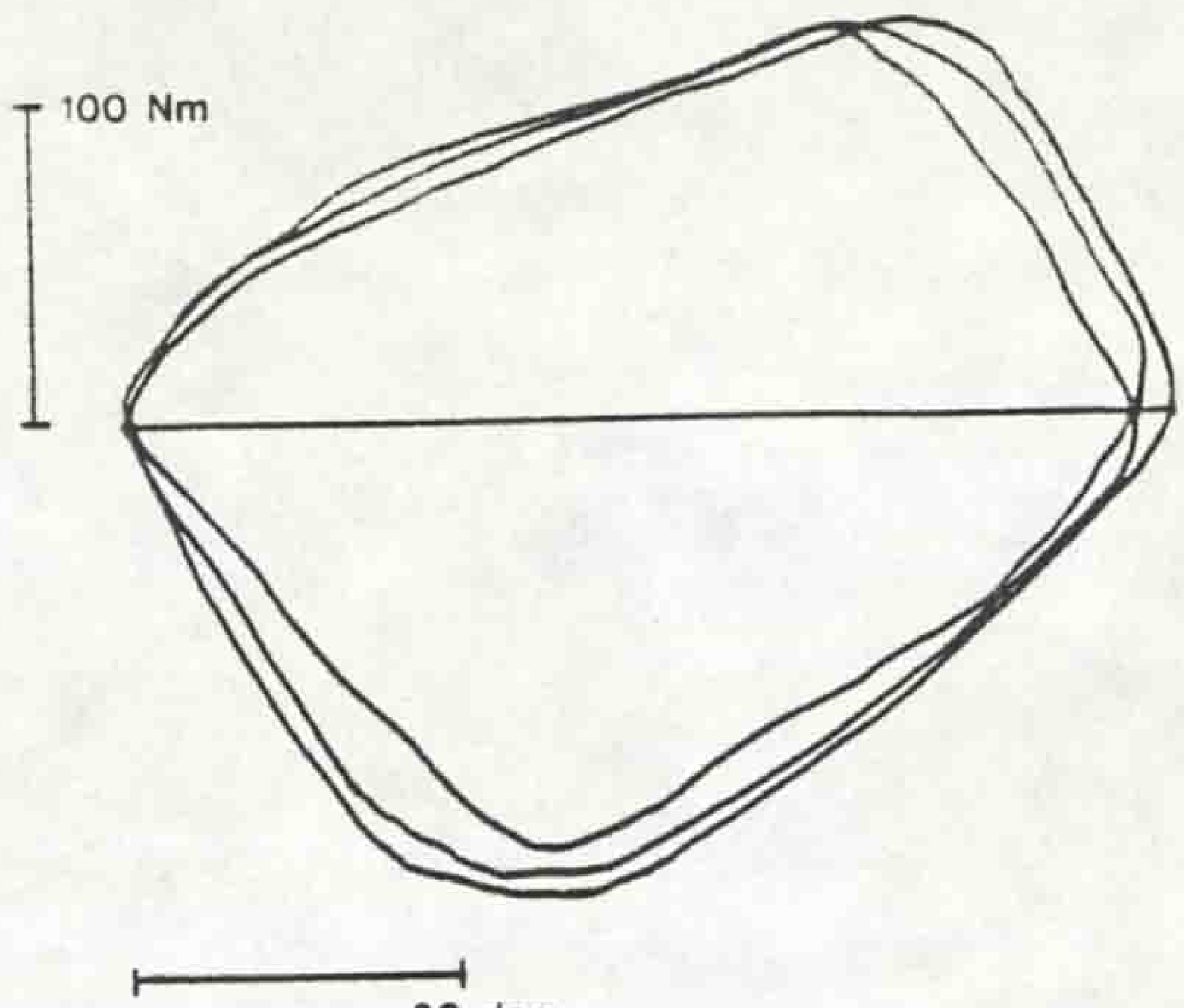

Figure 2. A typical analogue output using an X–Y plotter of the torque recorded during a knee extension (*lower part*)/ flexion (*upper part*) movement.

the end of the extension movement in order to hold the leg and lever arm upright, whereas the uncorrected analog output seen in Figure 2 shows an apparent zero torque.

The torque overshoot which is frequently observed in the torque output of isokinetic dynamometers represents the forces associated with the initial velocity fluctuations of the limb-lever arm system, prior to the development of the constant preselected velocity[3].

During isokinetic testing, the overshoot is frequently the peak point in the torque output. If this peak is interpreted as the subject's maximum torque, the muscular capability will be overestimated, also affecting bilateral comparisons and reciprocal muscle group ratios.

One method used for the elimination of the torque overshoot is the damping of the torque signal. However, this method affects the amplitude of the torque signal and the torque-time relationship throughout the ROM[11]. The method implemented in the present system is the monitoring of the angular speed of movement. Since oscillations in the torque output represent alternating periods of acceleration and deceleration, artifact-free data can be obtained from the part of the movement where the angular velocity remains constant. The angular velocity is computed from the angular position-time raw data.

Several isokinetic parameters are computed from the filtered data, including maximum torque, fatigue index, reciprocal muscle group ratio and work. Data analysis time is approximately 20 seconds. The results may be presented in numerical (Figure 3) or graphical form (Figure 1) and stored in record files for later comparisons.

In an attempt to standardize the testing protocols and allow valid comparisons between subjects, two testing protocols for the measurement of muscular strength and muscular endurance have been implemented in the

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********* ISOKINETIC DATA ANALYSIS *********

NAME : BB
WEIGHT :80.00 Kg
JOINT : RIGHT KNEE
MOVEMENT : EXTENSION/FLEXION
EXTENSION ANG. SPEED :240.00 Deg/Sec
FLEXION   ANG. SPEED :240.00 Deg/Sec
DATE : 28/ 7/1988
*******************************************

EXTENSION MAXIMUM TORQUE :129.00 Nm
ANGULAR POSITION :41.00 DEG
FLEXION   MAXIMUM TORQUE :68.00 Nm
ANGULAR POSITION :25.50 DEG
EXTENSION MAXIMUM TORQUE/WEIGHT : 1.61 Nm/Kg
FLEXION   MAXIMUM TORQUE/WEIGHT : 0.85 Nm/Kg
EXTENSION WORK :30.54 J
FLEXION   WORK :16.05 J
KNEE FLEXION   / EXTENSION RATIO :   0.53
EXTENSION FATIGUE INDEX : 0.00 Nm/Sec
FLEXION   FATIGUE INDEX : 0.00 Nm/Sec
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Figure 3. Numerical form of the isokinetic analysis results.

program. The strength test consists of six maximal reciprocal contractions and is used to determine the maximum torque capability of the subject. The endurance test consists of repeated contractions for 30 seconds and is used to estimate the fatigue of the contracting muscles. A training mode allows use of the dynamometer for training. The user specifies the training time and intensity and the computer displays the torque output during the training period.

Reliability and Validity

The reliability of the torque measurements was examined by repeated loading of the dynamometer with three different inert weights. A one-way related ANOVA test was performed on the recorded torque values and the intraclass correlation coefficient computed from the ANOVA results[8] was $r = 0{\cdot}99$ ($P<0{\cdot}001$).

The validity of the torque measurements was assessed using loads with known inertial properties. Linear regression analysis of the recorded torque on the mathematically computed applied torque was conducted and the validity coefficient[8] was $r = 0{\cdot}99$ ($P<0{\cdot}001$). The mean difference between recorded and applied torque was 2·41% of the mean applied torque. The results indicate that the measurement of torque with the present system is both valid and reliable.

Conclusion

The newly developed computer system has replaced manual analysis of the isokinetic data recorded in analogue form. Figure 2 illustrates a typical analogue output recorded on an X–Y plotter prior to the availability of the A/D converter and the computer system described in this report. Any isokinetic parameters to be measured had to be extracted from this graph using simple techniques involving rulers and planimetry. This is time consuming, inaccurate, tedious and prone to operator error. The computer system has replaced these manual analysis procedures. The system also provides real-time display of the muscular torque, while data analysis time has been reduced and the accuracy of the computed isokinetic parameters increased using correction methods for any gravitational or inertial errors. The combination of immediate data presentation, the available storage capabilities and the relatively low cost of the system (£12 000), permits the conclusion that it can be used for an efficient and accurate assessment of isokinetic muscle function.

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[For further enquiries, copies of the computer program or details concerning the operation of the system, please contact the authors.]

Sources of Error in Isokinetic Dynamometry: Effects of Visual Feedback on Maximum Torque Measurements

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The purpose of this study was to examine the effects of visual feedback on isokinetic maximum torque and reciprocal muscle group ratio of the knee extensors and flexors at a slow (60°/sec) and a fast 180°/sec) speed of movement. The real-time gravity-corrected torque output, used as the visual feedback source, elicited a significant increase in the maximum torque output of both muscle groups at the slow speed. There was no improvement at the fast speed of movement. The knee flexor/extensor ratio was not effected by visual feedback or speed of movement. It was concluded that visual feedback of the torque output can improve maximum voluntary contraction in isokinetic dynamometry under certain restrictions on speed and range of movement.

Accurate and objective assessment of muscle function is essential in both injury rehabilitation and the development of strength for specific purposes. Isokinetic dynamometry is widely used in such assessment because it offers the capability of providing variable resistance that is equal to the muscular forces and constant preselected speed of movement (2, 7, 9, 10).

The main isokinetic parameters used in the assessment of muscle function are maximum torque and reciprocal muscle group ratio. Maximum torque is defined as the highest muscular torque value from a number of repetitions. Reciprocal muscle group ratio is the quotient of maximum torque of two reciprocal muscle groups [e.g., hamstrings/quadriceps (h/q) ratio]. This ratio is an indication of joint balance and stability, and its accurate measurement is important for injury prevention and rehabilitation (4). The measurement of these parameters, however, is affected by gravitational and inertial forces during the test. Accurate assessment of muscle function requires appropriate correction methods.

The effect of gravitational forces on the measurement of several isokinetic parameters has previously been investigated, and significant measurement errors have been reported (1, 16). The importance of gravity correction for intrasubject comparisons, however, has been questioned because the effect of the gravitational forces is uniform over the same experimental procedures (8). This approach is valid only if the torque is recorded at a constant, predetermined angular position (i.e., similar gravitational force), using the same experimental procedure in all tests.

The development of the constant preset speed is another potential methodological problem in isokinetic data analysis. A finite period of time is necessary for the development of the preset speed. This acceleration period increases with increasing preset speed. The initially overspeeding limb is decelerated to the level of the preset speed by the resistive mechanism of the dynamometer. The torque overshoot that is frequently observed in the beginning of movement represents this resistive torque and must not be interpreted as muscular torque (13).

Other sources of variability in isokinetic testing include the positioning and stabilization of the subject on the dynamometer, rest periods be-

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tween tests at different angular velocities, test instructions, and motivation during the test. Accurate assessment of muscle function and valid comparisons of isokinetic data require standardized testing and measurement protocols implementing correction methods and maximizing voluntary muscular output.

Visual feedback (VF) of the muscular torque output during isokinetic testing is a source of variability in the measurement of isokinetic parameters and, consequently, in the assessment of muscle function (6, 8, 12). Riggsbee (12) suggested that using the dynamometer's analogue recorder as a visual feedback source can improve the patient's response during isokinetic testing, although no experimental data to support this hypothesis were reported at that time. Subsequently, however, Figoni and Morris (6) reported that VF improved the maximum torque output of both quadriceps and hamstrings by 12 percent at a slow speed of movement (15°/sec), but there was no improvement at a fast speed (300 °/sec). Hald and Bottjen (8) reported that VF improved the maximum torque output of both muscle groups by 6 percent at 60°/sec and approximately 3 percent at 180°/sec. The analog torque recorder of the isokinetic dynamometer was used as the VF source in the above studies.

It is evident from these results that VF has a significant effect on the torque output. The magnitude of this effect depends on the angular speed of movement. The isokinetic parameters, however, were not corrected for the effect of gravitational and inertial forces, although Figoni and Morris (6) measured the maximum torque after the first torque peak in order to avoid interpretation of the torque overshoot artifact as muscular torque. Therefore, it is not clear, whether VF affects torque output or this effect is a methodological artifact.

The purpose of this study was to examine the effect of VF on the maximum torque and reciprocal muscle group ratio. The gravity-corrected, real-time display of the torque output was used as the VF source. Torque data were corrected for the effects of inertial forces before measuring the above isokinetic parameters.

METHODS

Instrumentation

An Akron isokinetic dynamometer (Akron Therapy Products, Norfolk, England) was used to measure muscular torque. This system permits isolated joint testing at a constant, preset speed of movement that can be set independently for reciprocal muscle groups. The dynamometer was interfaced with an Intel 8086 based microcomputer for data collection and analysis. Torque and angular position data were sampled at 290 Hz. The gravitational torque throughout the ROM was registered before the test. The torque data during the test were corrected for gravity and displayed on the monitor in real-time (3) (Figure 1). Data from the constant angular velocity periods only were used to determine the isokinetic parameters. The above computer system is described in detail by Baltzopoulos and Brodie (3).

Subjects

Ten healthy males without any history of joint injury gave informed consent and volunteered to participate in this study. The subjects had a mean age of 25.8 (±1.7) yrs, a mass of 69.9 (±4.1) kg, and averaged 177.7 (±5.9) cm in height. Right dominance was determined using an isometric contraction of both legs at approximately 60 degrees of knee flexion on the dynamometer, simulating a simple kicking action.

Procedures

A pilot study indicated that the effect of VF was similar for male and female subjects supporting previous findings that the effect of VF on maximum torque was not sex related (11). Therefore, only male subjects participated in the present study.

The testing protocol consisted of five maximal reciprocal repetitions of the knee extensors and flexors, since development of the maximum torque requires three to five repetitions (2). The test was performed at a slow (60°/sec) and a fast (180°/sec) speed of movement with and without visual feedback. The range of movement (ROM) for all tests was from 90 to 30 degrees of knee flexion.

The tests were completely randomized, and rest periods of five minutes were given between the tests. A familiarization and warm-up period was given five minutes before the test. The tests were performed with the subjects seated and the

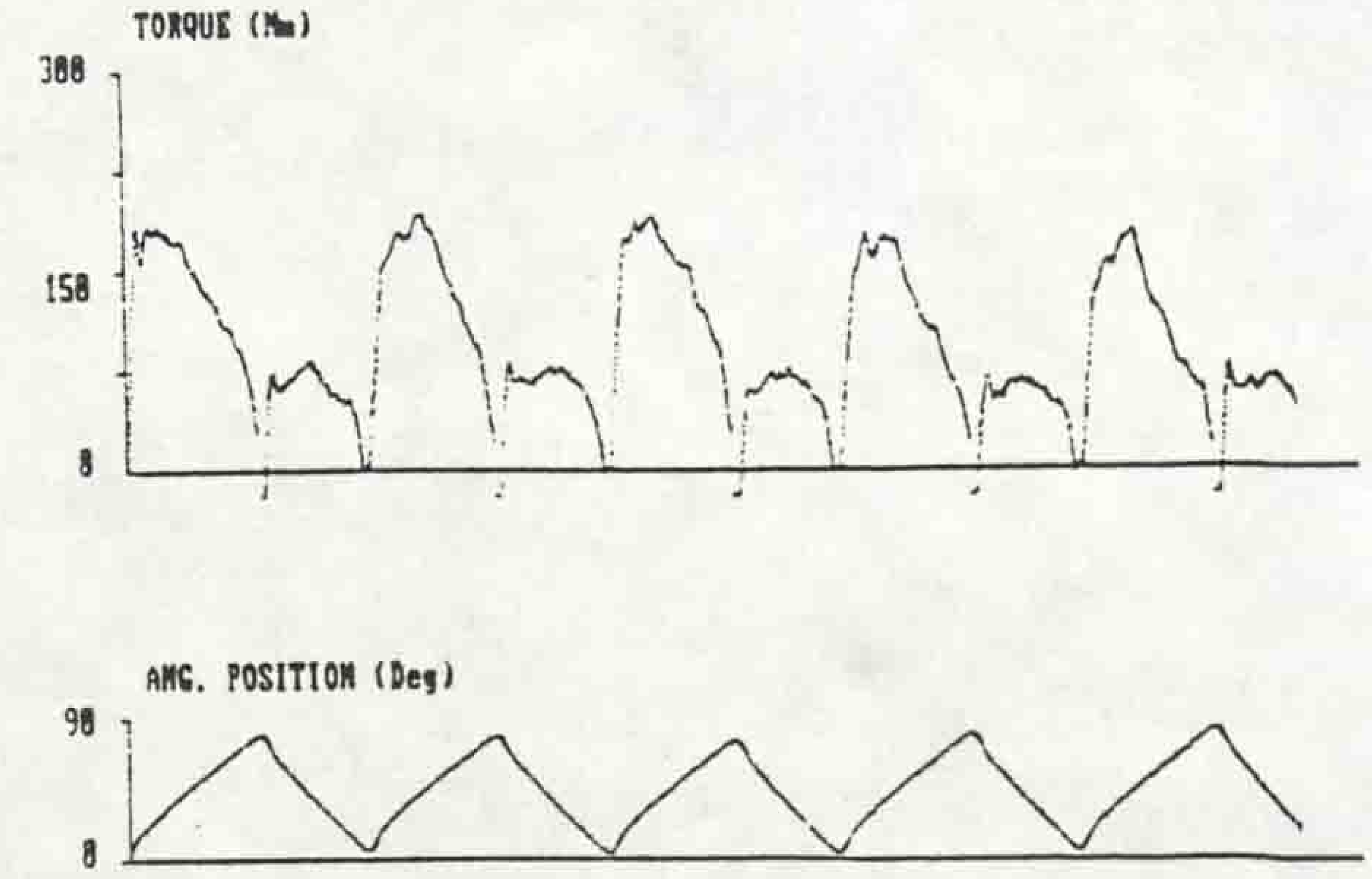

Figure 1. Sample of real time torque and joint position output.

thigh, pelvis, and noninvolved foot stabilized with appropriate belts. The axis of rotation of the dynamometer was aligned with the most prominent point of the lateral femoral condyle.

During all tests, the computer monitor was positioned approximately one meter from the subject at eye level. All subjects were instructed to carefully observe the monitor. During the VF tests, real-time display of the gravity-corrected muscular torque was provided. During the no-VF tests, the monitor was blank. All subjects were given written, standardized instructions to work as hard and fast as possible against the resistance of the dynamometer during the tests and to try to overcome the torque curves from the previous repetitions displayed on the computer monitor.

Gravity-corrected torque data from constant speed periods only were used to calculate the following isokinetic parameters: *1*) maximum torque in extension (quadriceps) and flexion (hamstrings), 2) knee angle of maximum torque, and *3*) hamstrings/quadriceps ratio.

Muscular torque was not normalized to body mass because the correlations obtained between maximum torque and body mass were weak—coefficients of determination were less than 53 percent.

Data Analysis

Differences between the VF and no-VF conditions, at different speeds of movement (slow-fast), and different muscle groups (quadriceps-hamstrings) were analyzed using a three-factor (2 × 2 × 2) repeated measures ANOVA test, with muscular torque measurements as the performance variable. Significance was accepted at the 0.05 probability level.

RESULTS

There was a main effect for visual feedback (F1,9 = 13.6, p < 0.05), speed of movement [$F(1,9) = 148.1$, $p < 0.05$], and muscle group [$F(1,9) = 135.5$, $p < 0.05$]. The mean (SD) of the torque measurements are presented in Table 1 and graphically represented in Figure 2. At the slow speed, visual feedback improved the torque output by 8 percent and 6 percent in extension and flexion, respectively. There was no improvement at the fast speed.

TABLE 1

Mean (SD) of maximum torque (Nm) measurements (No VF, no visual feedback; VF, visual feedback)

	Slow (60 °/sec)		Fast (180 °/sec)	
	No VF	VF	No VF	VF
Extension	207.6 (42.2)	224.8 (36.9)	149.4 (30.1)	150.8 (36.3)
Flexion	115.9 (24.9)	121.7 (23.4)	86.0 (19.2)	86.5 (19.8)

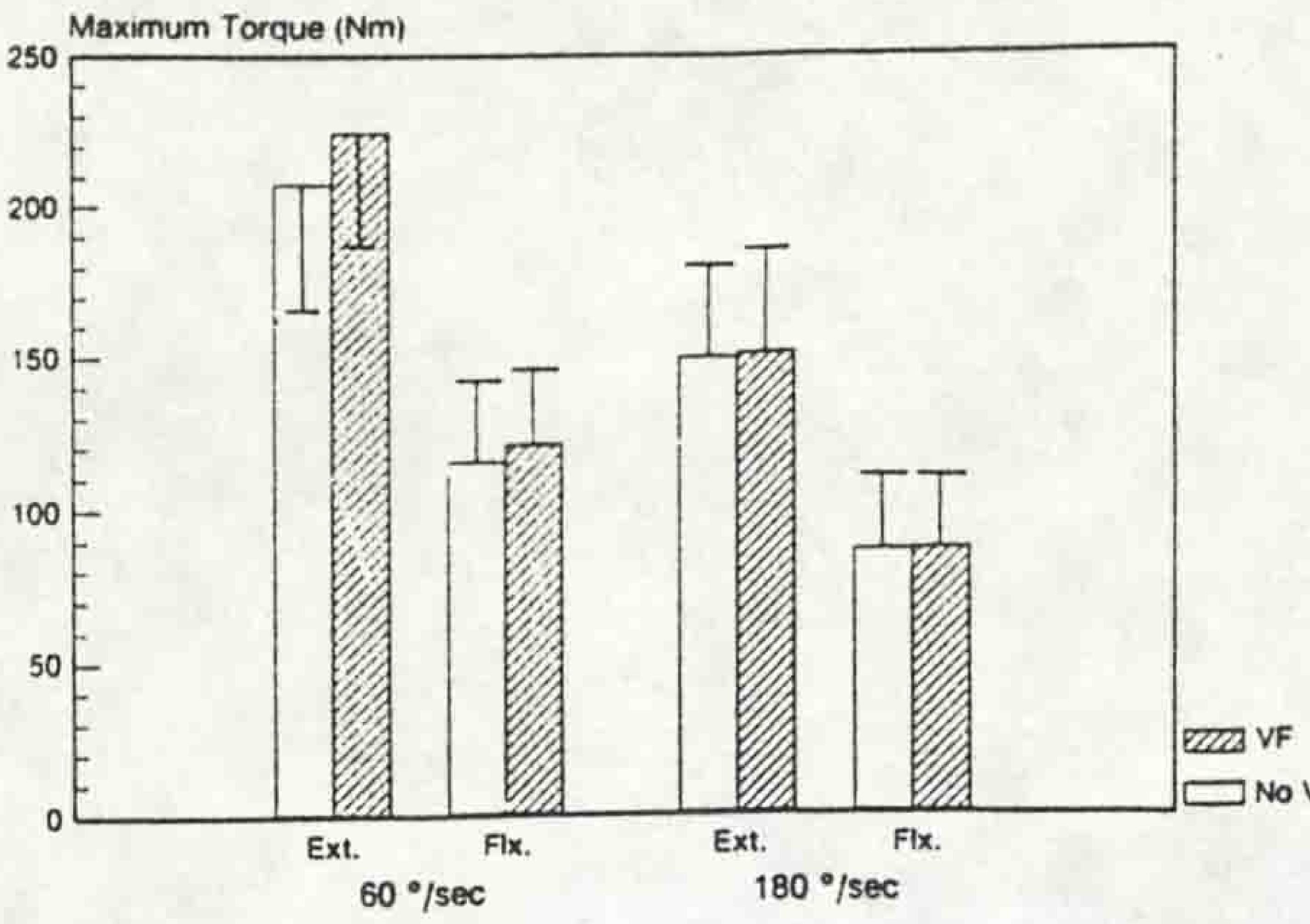

Figure 2. Maximum torque in extension and flexion under different visual feedback and angular velocity conditions.

TABLE 2

Mean (SD) of hamstring/quadriceps (h/q) ratio measurements (No VF, no visual feedback; VF, visual feedback).

Slow (60 °/sec)		Fast (180 °/sec)	
No VF	VF	No VF	VF
0.57 (0.100)	0.54 (0.059)	0.58 (0.077)	0.58 (0.085)

TABLE 3

Mean (SD) of maximum torque angular position (degrees) (No VF, no visual feedback; VF, visual feedback)

	Slow (60 °/sec)		Fast (180 °/sec)	
	No VF	VF	No VF	VF
Extension	21 (3.8)	25 (6.6)	30 (5.3)	30 (8.3)
Flexion	38 (4.5)	33 (7.1)	30 (9.3)	37 (8.9)

A two-factor (VF-speed) repeated measures ANOVA test was used to analyze the hamstrings/quadriceps ratio. There were no significant effects. The mean (SD) of the hamstrings/quadriceps ratio measurements are presented in Table 2. This ratio was approximately 0.57 under all conditions of VF and speed of movement.

DISCUSSION

The mean extension torque obtained in this study under the VF condition is approximately 4 percent higher than respective values in previous VF studies (6, 8). The flexion torque, however, is lower, with differences ranging from 13 to 23 percent. These differences may be attributed to individual differences and the fact that no gravity correction procedures were used in the above studies.

The results of the present study indicate that the position of the maximum torque under the VF condition was shifted later in the ROM during the slow speed test (Table 3). A shift in the opposite direction was observed during the fast speed test. These results underline the importance of gravity correction in isokinetic dynamometry, especially

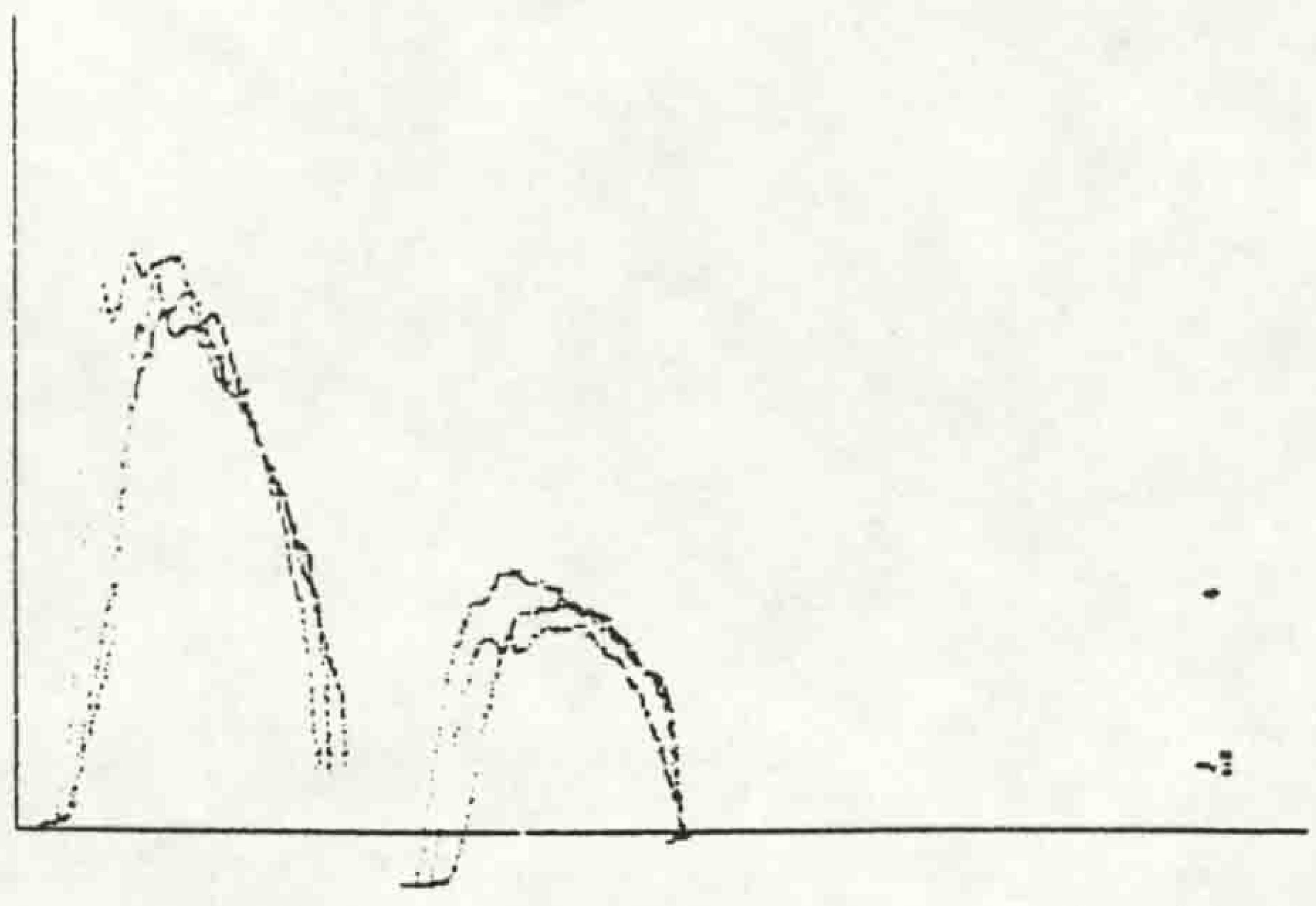

Figure 3. Sample of torque output used for visual feedback.

when the experimental variable is likely to affect the position of the maximum torque.

The constant velocity periods of the movement were determined by measuring the angular velocity from the torque-position data (2). Torque data from the isokinetic part of the movement only were used for subsequent analysis. This method does not affect the amplitude and phase of the torque signal (14) and ensures that the torque overshoot during the initial acceleration period is not interpreted as muscular torque.

Visual feedback had a positive effect on the maximum torque of both muscle groups (quadriceps-hamstrings) at the slow speed only. Similar findings have been reported by Figoni and Morris (6), although they found an increase of 12 percent for both muscle groups, compared to 8 and 6 percent for extensors and flexors in the present study. Hald and Bottjen (8) reported a significant increase of 3 percent at the fast speed and 6 percent at the slow speed. Statistical significance, however, was determined by using a series of *t*-tests and not an ANOVA design.

It is evident that muscular torque output presented as extrinsic, concurrent, visual feedback (13) has a positive, motivational effect on muscular response. The subjects were required to process the visual feedback and adjust their muscular response during the ongoing repetition. The temporal constraints of the movement may, therefore, explain the absence of a VF effect at the fast speed test (6).

During the slow speed test (60°/sec), the 60 degree motion arc was completed in approximately one sec. During the fast speed test, however, the movement time was approximately 300 msec. The reaction time to visual information was approximately 160 to 180 msec (15). This suggests that during the slow speed test, approximately 80 percent of the movement time was available for information processing and response adjustment. On the contrary, during the fast speed test, any response occurred during the last third of the movement when the muscles were beyond their optimal anatomical position for maximum torque production.

The effectiveness of VF in isokinetic dynamometry appears to be related to the movement time of a particular testing procedure as a function of ROM and speed of movement. A positive effect of VF on maximum torque during a fast speed test is possible, provided that the ROM is appropriately set to allow the postfeedback muscular response to occur before the optimal anatomical position of the activated muscle group (e.g., shoulder extension-flexion) is reached. In isokinetic dynamometry, however, anatomical and apparatus constraints may limit the ROM and the effect of visual feedback.

The effectiveness of visual feedback is also influenced by its precision and accuracy (15). With the present isokinetic system, the real-time display of torque and angular position could be presented in different forms. To enhance precision, the display was modified to present the torque output only over the entire computer monitor (Figure 3), without information that could distract the attention of the subjects (e.g., axes legends, angular position, etc). Furthermore, the torque output from the different repetitions was superimposed allowing easier comparisons with the previous repetitions during the test.

The hamstrings/quadriceps torque ratio was approximately 0.57 at both speeds. An increase in this ratio with increasing speed has been reported previously (5, 17). The result of this study, however, support recent findings (1) that this increase is a gravitational artifact and that the gravity-corrected ratio remains relatively constant with increasing speed. Despite a significant increase in the maximum torque under the VF condition, there was no significant difference in the hamstrings/quadriceps ratio, since the maximum torque increase was similar for both muscle groups.

CONCLUSIONS

Within the limitations of the present study, visual feedback yielded: *1*) a significant increase in the maximum torque output which was similar for both muscle groups tested; therefore, no effect on the reciprocal muscle group ratio was observed, and *2*) an effect that depends on the movement time of a particular testing procedure as a function of ROM and speed of movement. ❑

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