DYNAMIC JOINT MECHANICS AS AN OBJECTIVE CLINICAL MEASURE OF ANKLE FUNCTION

Rita Morier
School of Physical and Occupational Therapy
McGill University, Montreal
July, 1988

A thesis submitted to the Faculty of Graduate Studies and Research in partial fulfillment of the requirements for the degree of
M.Sc. in Rehabilitation Science

UNIVERSITÉ McGILL

FACULTÉ DES ÉTUDES AVANCÉES ET DE LA RECHERCHE

Date ______________

NOM DE L'AUTEUR: ____

DÉPARTEMENT: ______________________ GRADE: ______________________

TITRE DE LA THÈSE: ____

____

____

1. Par la présente, l'auteur accorde à l'université McGill l'autorisation de mettre cette thèse à la disposition des lecteurs dans une bibliothèque de McGill ou une autre bibliothèque, soit sous sa forme actuelle, soit sous forme d'une réproduction. L'auteur détient cependant les autres droits de publications. Il est entendu, par ailleurs, que ni la thèse, ni les longs extraits de cette thèse ne pourront être imprimés ou reproduits par d'autres moyens sans l'autorisation écrite de l'auteur.

2. La présente autorisation entre en vigueur à la date indiquée ci-dessus à moins que le Comité exécutif du conseil n'ait voté de différer cette date. Dans ce cas, la date différée sera le ____

____

Signature de l'auteur

Adresse permanente:

____

____

____

____

Signature du doyen si une date figure à l'alinéa 2.

(English on reverse)

ABSTRACT

The purpose of the work contained in this thesis was to investigate whether experimental paradigms based on the system identification approach are an effective clinical assessment tool. This was accomplished by conducting two companion studies: a reliability study on a group of fifteen control subjects and a case study of an individual who had sustained a unilateral undisplaced ankle fracture. The data collected in both studies included ankle angular position, torque, and Tibialis Anterior and Triceps Surae electromyograms. From these data, measures of both static (e.g. range of motion) and dynamic (e.g. estimated elastic stiffness) joint function, were obtained.

A number of clinically relevant variables (plantarflexion MVC, dorsiflexion MVC, range of motion, passive torque, K offset, low K region, and the intercept of the K-absolute torque relation) were shown to be reliable. In addition, the results of the case study demonstrated that it would be feasible to use these experimental procedures and analytic methods on individuals who have sustained orthopedic trauma. Finally, certain variables (the K offset, the slope and intercept of the K-torque relation, and a low stiffness region) appeared to be sensitive to the clinical changes associated with orthopedic pathology.

RÉSUMÉ

Le but de cet ouvrage était de réaliser une étude de fiabilité de mesures de dynamiques de la cheville en tant qu'outils clinimétriques objectifs, fiables, et quantitatifs dans l'évaluation et le suivi clinique de la cheville suite à un traumatisme. Le paradigme et le montage expérimental ont été évalués à l'aide d'un groupe témoin et d'un cas pathologique. Des techniques d'identification de systèmes ont servi à déterminer les paramètres qui ont le mieux décrit le comportement de la cheville. La collecte des données normatives fut suivi par une évaluation de la fiabilité de certaines mesures tel que l'amplitude articulaire, la contraction maximal volontaire, le moment passif, et certaines mesures dynamiques dont le terme élastique (K). Nous avons procédé à l'étude du cas pathologique suivant le même plan.

Ce travail démontre que l'approche d'identification de systèmes, le paradigme expérimental, et le traitement des données conviennent très bien à une utilisation clinique. Les mesures mécaniques passives statiques et dynamiques sont très fiables et certaines d'entre elles, tel que l'aire de la boucle de moment-position, l'ordonnée à l'origine et la pente de la relation K-moment, et une région articulaire à faible rigidité font possiblement preuve d'une sensibilité aux changements due à une pathologie et sa guérison.

ACKNOWLEDGMENTS

I wish to express my deepest gratitude to my research director Dr. P. L. Weiss for her continuous support and her confidence in my abilities. Our many hours of discussion have contributed enormously to my professional formation. Her support and supervision has led me to a greater appreciation of scientific research and their authors.

I am also grateful to Dr. R.E. Kearney for his expertise in a matter originally so unfamiliar to me. He has taught me to persevere in the quest for answers.

I extend my appreciation to S. Lafontaine for his generosity and benevolence. His wealth of knowledge and willingness to share it were a great help in the completion of this work.

I thank L. Belzak and B. McNeil for their invaluable assistance and pleasurable camaraderie and extend this same gratitude to all my lab mates who shared with me space, computer time, lunches, and ideas.

I sincerely thank C. Cardinal-Garland for her help, dependable smile and moral which added to the pleasant work atmosphere.

À Maurice

Table of Contents

GLOSSARY OF ABBREVIATIONS

A/D	Analog-to-Digital
B	Viscous parameter
D/A	Digital-to-Analog
DF	Dorsiflexion
EMG	Electromyograms
I	Inertial parameter
IRF	Impulse response function
K	Elastic parameter
MVC	Maximal Voluntary Contraction
NP	Neutral-position
PF	Plantarflexion
PRBS	Pseudo-random binary sequence
ROM	Range of motion
SD	Standard deviation
VAF	Variance accounted for

CHAPTER I

INTRODUCTION

The ankle is a weight bearing joint that plays a crucial role in bipedal locomotion and is important in the maintenance of upright stance and equilibrium. It is subjected to a multitude of internal and external forces that result in a distinctive pattern of behavior. For example, joint compression forces of up to five times body weight occur during the push off phase in the normal gait cycle and change in the presence of pathology (Stauffer et al., 1977). Since the ankle is the focal point of total body weight transmission and poorly tolerates even small changes in its anatomical configuration (Frankel and Nordin, 1980), high demands are placed on the treatment and the return to normal function of ankle injuries. Injury, immobilization and therapeutic intervention will all induce structural and physiological changes at the cellular level which can manifest themselves clinically as a decrease in range of motion (ROM) or a decrease in muscle strength. Other clinically observed phenomena such as the joint's increased resistance to passive movement and the subjective impression of joint stiffness are poorly measured (Wright, 1973). The purpose of this study is to investigate the feasibility of using the system identification approach in the clinical assessment of joint function. The thesis is divided into the following chapters:

Chapter II briefly reviews the background material and the previous work in this field required to understand the problem formulation, significance and interpretation of the thesis results.

Chapter III refers to the general methods involved in the study including the apparatus, experimental paradigms, description of stimuli and general treatment of the data. The justification for the choice of parameters and methodology is given.

Chapter IV consists of a paper describing the construction of a custom made fibreglass boot used in the fixation of the foot to the experimental apparatus.

Chapter V outlines the reliability study performed on various measures of joint behavior in particular the dynamic joint mechanics and details the case study which was undertaken.

Chapter VI includes a summary of the major findings, limitations and recommendations for future work.

Chapter II

REVIEW OF THE LITERATURE

The first section of this review will deal with the notation and terminology that will be used throughout this thesis. A brief description of ankle joint anatomy and kinematics pertinent to this study will be presented in the next part of the review followed by an overview of the effect of injury on these structures at the cellular and functional level. The classical clinical approaches of evaluation of joint function for example, goniometry and the manual muscle test and their limitations will be discussed. A new assessment tool, the measurement of active and passive ankle mechanics will be introduced and the potential of this tool as a quantitative objective measure of ankle function will be addressed. A discussion leading to the problem formulation and objective of this thesis will conclude the review.

Definitions and notation

The movement of the ankle about the tibiotalar transverse axis is one of rotation. Rotation of the joint towards the lower leg is called dorsiflexion and rotation of the joint away from the lower leg is plantarflexion. In this study the motion was defined with respect to the mid position, a point which divided the subject's ROM in two. Any movement away from the mid position toward the leg was labeled dorsiflexion while movement away from the midposition in the opposite direction was labeled plantarflexion. The subject's available ROM was defined subjectively, based on the subject's report of pain or extreme discomfort and/or the occurrence of compensatory movements at the more proximal joints.

Dorsiflexing torques were denoted by positive values and tended to move the ankle into a more dorsiflexed position. In contrast, plantarflexing torques which were denoted by negative values tended to move the ankle into a more plantarflexed position.

Ankle Joint Anatomy

The purpose of this section is to offer a brief overview of ankle joint anatomy. Trauma, surgical intervention, or repeated minor trauma can induce changes in ankle joint structures as well as in the muscles that function about the joint. Changes in bone structure as a result of a fracture can alter the physical orientation and relationships of articular surfaces.

Bony configuration

The ankle joint consists of the articulation between the tibia, fibula and the talus and is also known as the talocrural joint. The tibia and the fibula are the two long bones of the lower leg. The tibia lies medially, is thicker and ends distally as the medial malleolus. The fibula lies laterally, is more slender but has an estimated transfer of as much as one sixth of the forces (Lambert, 1971). It ends distally as the lateral malleolus approximately 10 to 20 mm lower than the medial malleolus. The talus has a squarish body, a short neck and a head with an oval articular surface. The body lies between the malleoli of the tibia and fibula which form a socket and its upper surface articulates with the cartilage covered inferior surface of the tibia. The ankle joint has an efficient mechanism of load transmission which is thought to be a factor in its relative resistance to primary osteoarthritis and cartilage fibrillation (Wynarsky and Greenwald, 1983).

Joint Capsule

The ankle is a synovial joint possessing a potential cavity, lubricated articular cartilage, a capsule of fibrous tissue and a synovial membrane (Basmajian, 1979). The fibrous capsule surrounds the joint and is attached to the margins of the tibia and fibula articular surfaces, passing below to the talus close to the articular areas on the upper, medial and lateral surfaces of its body. Synovial membrane, also known as synovium, is a condensation of connective tissue that lines the inner surface of the fibrous capsule. It does not cover the weight bearing surface of articular cartilage (Hettinga, 1979a).

Ligamentous structures

The ankle joint capsule is weak anteriorly and posteriorly but is reinforced laterally by strong collateral ligaments. The anterior and posterior ligaments consist of localized thickenings of the capsule. On the medial aspect lies the stout medial (deltoid) ligament which is triangular in shape and is attached by its apex to the medial malleolus and by its base to the navicular bone in front and to the posterior part of the body of the talus behind. The lateral ligaments consist of three separate bands originating from the lateral malleolus. The calcaneofibular ligament inserts onto the lateral surface of the calcaneus restricting inversion and dorsiflexion. The anterior talofibular courses forward and medially to the neck of the talus limiting inversion and plantarflexion and the posterior talofibular ligament passes medially and slightly backwards to the lateral tubercle of the talus limiting inversion and dorsiflexion.

Musculature

The muscles acting about the ankle joint are divided into three compartments, the anterior compartment containing the dorsiflexor muscles, the peroneal compartment containing the peroneal muscles and the posterior compartment containing the plantarflexor muscles. For the purposes of this study attention will be drawn to two major muscle groups, Tibialis Anterior (TA), and the Gastrocnemius-Soleus (GS) complex which are located in the anterior and posterior compartments respectively.

TA was chosen in this study as the representative of the dorsiflexing contractile element of the ankle joint. This muscle is subcutaneous and arises from the lateral condyle and the upper two thirds of the lateral aspect of the tibia, gaining additional origin from the interosseus membrane and from deep fascia. The tendon becomes free in the lower third of the leg and passes medially to insert onto the medial aspect of the medial cuneiform and the base of the first metatarsal bone.

The GS Complex is the main plantarflexor of the ankle joint (Dersheid and Brown, 1985). Gastrocnemius has two bellies, the lateral head originating from the upper, posterior part of the lateral surface of the lateral femoral condyle and the medial head, originating from the popliteal surface of the femur above its medial condyle. The two fleshy bellies insert into the posterior surface of a broad membranous tendon of Soleus muscle to form the calcaneal tendon which inserts onto the posterior surface of the calcaneum.

The Soleus muscle originates from the posterior surfaces of the head and the upper third body of the fibula, from the soleal line of the tibia and from a tendinous arch between tibia and fibula. The plantarflexors are much more powerful than the dorsiflexors. Thi may be related to the functional requirements of this muscle group. For example, the body's center of gravity falls anterior to the ankle joint (Steindler, 1955) requiring a sporadic plantarflexing torque to maintain an erect posture. In addition, the plantarflexors provide the propulsive force in ambulation at push-off and absorb energy during early and mid stance as the leg rotates over the foot (Winter, 1987).

Ankle joint kinematics

Axis of rotation

Movement at the ankle joint takes place about an oblique transverse axis which can be approximated by a line joining a point 50 mm distal to the medial malleolus to a point 30 mm distal and 80 mm anterior to the lateral malleolus (Inman, 1976).

Range of Motion

There is a lack of consistency in the literature regarding the magnitude and the method of measuring ankle ROM. Radiograghic methods (Sammarco et al., 1973), measurements during weight bearing (Sammarco et al., 1973), passive as opposed to active ROM measures and ROM measurements under conditions of knee flexion (Ekstrand et al., 1982) are some of the factors affecting ROM measurement. There is also a tendency for ROM to decrease with increasing age (Sammarco et al., 1973; Boone et al., 1979). In general, the reported ROM ranged from 0.87 rad to about 1.31 rad (Caillet, 1968; American Academy of Orthopedic Surgeons, 1965). There is little mention of the criteria used to define the endpoints of such measurements and it would appear that most ROM measurements and methods have a subjective component.

Functional considerations

The range of ankle motion required for functional tasks such as normal walking is in the order of 0.17 rad of dorsiflexion and 0.35 of plantarflexion (Kotwick, 1982; Lindsjo, 1985). There do not seem to be any differences on the basis of sex although heel height can considerably alter joint angles (Murray et al., 1964; Murray et al., 1970).

Intra-subject joint angles vary little from stride to stride (less than 2% variability at all lower extremity joints (Winter, 1987). Winter (1987) compared joint angles for slow, natural and fast walking in 19 normal individuals and noted that for the ankle and with respect to the stride period, the joint angle profiles are essentially the same.

This author also studied angular velocities during gait and their variation with the speed of walking in again 19 subjects (Winter, 1987). A natural cadence (105 $\pm$ 6 steps/min) can develop maximal DF velocities of about 2.5 rad/s and maximal PF velocities of approximately 3.5 rad/s. At a fast cadence (natural + 20 steps/min) maximal DF velocities of approximately 3.0 rad/s and maximal PF velocities of approximately 4.5 rad/s were recorded (Winter, 1987).

Injury and Immobilization effects

The choice of variables that are measured in clinical assessments of joint function and dysfunction is based on the understanding of the changes that occur as a result of injury and immobilization. This forms the rationale for the selection of parameters that best characterize these changes. The following section summarizes some of the cellular and structural changes associated with injury.

Joint Capsule

Post traumatic synovitis can be considered as the synovium's simplest response to minor nonhemorrhagic trauma (Hettinga, 1979a). It consists essentially of an inflammatory response. Minor trauma, not requiring medical attention and with few symptoms, can set up this severe vasomotor reaction with significant symptoms apparent only months later (Soren et al., 1976). This phenomena is seen as fibrosis and thickening of the subsynovial connective tissue layer. If the trauma occurs only once, the inflammatory response subsides quickly. If mechanical irritation continues due to secondary laxity of ligaments, free joint bodies or recurrent trauma, a proliferative synovitis involving a progressive sclerosis of the synovial membrane four to six months post trauma with additional thickening of the connective tissue layer (Hettinga, 1979a) can occur. There are also distinct changes in the composition of synovial fluid within the joint that can further irritate and perpetuate the synovitis reaction. In spite of this, synovial tissue has a remarkable regenerative capacity due to its rich vascularization and, if not further traumatized, can completely regenerate within a few months (Hettinga, 1979a). Synovial fluid functions to nourish the articular cartilage and lubricate the joint and periarticular soft tissues (Dowson et al., 1969). It is extremely viscous at low velocities corresponding to slow movements and becomes less so and more elastic at high velocities (fast movements) (Dowson et al., 1969; Dowson, 1973). Synovial fluid forms a fluid film that protects the articular cartilage from erosion (Hettinga, 1979b).

A nonhemorrhagic reaction of the synovium can result in changes in synovial white blood cell, protein, and hyaluronate concentration (Bozdech, 1976). This can lead to a decrease in viscosity with possible alteration of joint congruency and pressure distribution across the articular surface (Wynarsky and Greenwald, 1983). Traumatic effusions that involve bleeding into the joint can occur with fractures and tearing of articular blood vessels. The iron that is liberated as a result of degradation of blood is a strong irritant to the synovial membrane (Hettinga, 1979b).

The fibrous capsule reacts to trauma by way of an inflammatory response as well. There is increased vascularity and proliferation of fibrous tissue (Hettinga, 1979b). A decrease in flexibility of periarticular structures is due to adhesion formation or intermolecular crosslinking of collagen (Woo et al., 1975). This is also known as contracture formation. The thickening of the synovium connective tissue layer and the formation of adhesions decrease the joint's compliance and can hinder its range of free movement (Woo et al., 1975).

Ligaments and Tendons

The collagenous tissues which support the skeletal system are ligaments (including joint capsules) and tendons. These structures are passive in that they are mainly loaded by the tension produced by joint motion and muscle contraction. Ligaments and tendons will remodel in response to the mechanical demands placed upon them. When subjected to increase stress ligaments become stronger and stiffer (Kotwick, 1982).

Noyes (1977) found a decrease in stiffness which he defined in terms of the slope of the tension to ligament failure curve, in primate anterior cruciate ligaments following eight weeks of knee immobilization; the ligaments were less capable of resisting load and elongation and failed at much lower tensile loads. Even after five months of resumed normal animal activity there was only partial recovery of ligament strength although the stiffness parameters had returned to normal. Investigations of canine ligament properties demonstrated an increase in medial collateral knee ligament stiffness as well as an increase in the diameter of collagen fibre bundles when animals were extensively trained on the treadmill (Tipton et al., 1970). Immobilized ligaments were weaker in terms of withstanding tensile loads than the normal knee ligaments and the normal knee ligaments were weaker than the trained knee ligaments. Similarly, studies on swine tendon also showed an increase in stiffness and strength with intensive exercise training, suggesting that tendons remodel much the same way ligaments do (Woo et al., 1975).

Muscle

Muscles can also be subjected to trauma, either directly as from a blow or fall or indirectly as from stretching a muscle beyond its normal range. Moreover, joint immobilization and soft tissue trauma can result in structural and physiological changes in muscle.

Contractile protein and connective tissue are synthesized by myoblasts and fibroblasts respectively in response to muscle tissue destruction. The physiological conditions at the site of injury are usually more favorable to fibroblast activity filling the breech with collagenous scar tissue (Allbrook, 1973). Rat studies involving laceration and surgical repair have shown that the areas of muscle devoid of innervation will have an increase in connective tissue laid down between muscle fibres (Garett et al., 1984). The extent of trauma may be much less and still elicit structural changes.

Muscle injury may be subtle and even asymptomatic. Gastrocnemius muscle biopsies from marathon runners showed as much as 10% of cells injured. Chronic overuse (veteran runners) caused an increase in intercellular collagen deposition suggestive of a fibrotic response to low level injury (Warhol et al., 1985). However chronic overuse was not defined quantitatively and the time frame for these changes was not clarified.. Thus the evidence seems to indicate that muscle repair and response to immobilization involves the increased deposition of collagen connective tissue. Collagen itself has an extremely low compliance so that small increases in the quantity of collagen in muscle increases tissue stiffness considerably (Alnaqeeb et al., 1984).

Joint immobilization can lead to similar structural changes in muscle. Immobilization of the cat ankle joint resulted in an alteration in Soleus muscle extensibility and an increase or decrease in the number of sarcomeres arranged in series depending on the immobilization position (Goldspink et al., 1974). Muscle shortening occurred when the limb was immobilized in a muscle shortened position. These changes occurred in as little as four weeks and were reversible. Tabary et al. (1972) demonstrated up to a 40% decrease in the number of sarcomeres of cat Soleus muscle fixed in a shortened position. Muscles of immobilized mouse ankles showed reduced extensibility which was attributed to deposition of collagenous connective tissue (Williams and Goldspink, 1981). Cardenas (1977) has shown a decrease in rat soleus cross-sectional area as a result of immobilization although muscle fibre counts remained relatively unchanged.

The loss of sarcomeres is one method of evaluating the extent of muscle atrophy. Atrophy can also be characterized by a loss of muscle cross sectional area or by a loss of muscle mass. Cross sectional area may be affected by other factors such as connective and fatty tissue content. These tissue proportions may vary rendering a muscle's cross sectional area less representative of the contractile element content. Moreover muscle mass can be affected by the water content of muscle and hydration can be difficult to control in experimental muscle specimens. Such factors should be taken into consideration and, if possible, corrected before meaningful interpretation of the experimental results.

There is evidence in the animal literature that certain slow postural muscles such as Soleus may atrophy to a greater extent than others

(Edgerton et al., 1975). Studies as to the preferential atrophy of certain muscle fibre types have generated controversial results. Preferential atrophy of type I (slow, low oxidative fibres) (Maier et al., 1976), type II (fast, high oxidative fibres) (Pachter and Eberstein, 1984), and atrophy of both types I and II (MacDougall et al., 1980) have been observed in fast contracting animal muscles which were immobilized in a neutral position i.e. the muscle at resting length. The cross sectional area of Type I fibres in human Vastus Medialis muscle was shown to preferentially decrease in subjects with chronic knee dysfunction (Edstrom, 1970). The comparison between different quadruped species and man should be approached cautiously.

It is difficult to assess muscle atrophy in humans due to ethical considerations and technical limitations as it is an invasive procedure. The small samples of muscle that are biopsied are usually the surface fibres which tend to be predominantly of type II (Halkjaer-Kristensen and Ingeman-Hansen, 1981), therefore the depth of biopsy is an important determinant of fibre type concentration. The type and extent of muscle atrophy observed may also be dependent on the intensity of the noxious stimulus and type of underlying pathology. Animal studies have shown that a nociceptive stimulus is associated with more atrophy than either denervation or immobilization (Hnik et al., 1977). When comparing atrophy studies the influences of the underlying pathology/intervention should be kept in mind.

Anatomical Configuration

The ankle joint has a relatively large weight bearing surface and consequently the load per unit surface area is less than that of the hip and knee joint (Frankle and Nordin, 1980). A 1 mm lateral displacement of the talus can reduce the weight bearing surface by as much as 42% (Ramsey and Hamilton, 1976). Such shifts could occur following ligamentous injury and fractures and could possibly precipitate early degenerative changes in the articular surface.

Summary

In summary, joint injury and immobilization can result in adaptive changes at the cellular and structural level. The literature demonstrates an overall increase in the concentration of connective tissue in muscle and adhesion formation in the joint capsule. Chronic inflammation and overuse also involve an increased deposition of collagen tissue in capsule, tendon, and ligament. Immobilized ligaments are more compliant and have less tensile strength. Ligament laxity and ankle joint mortise disruption can alter joint congruency which may possibly induce degenerative changes in cartilage (Frankel and Nordin, 1980). An orthopedic injury will have sequelae that can be characterized by such clinically observable entities as altered joint ROM, increased resistance to passive movement and a decrease in the contractile ability of the muscles acting on the joint (Derscheid and Brown, 1985).

Signs and symptoms of joint trauma

Pain

These cellular changes can result in pain, swelling, decreased ROM, decreased muscle strength, joint instability and possible loss of proprioception. Pain and tenderness over the site of ligamentous injury is often the last abnormal physical sign to disappear in ankle injuries (Freeman, 1965). These tissues, richly innervated with the free nerve endings that form part of the articular pain receptor system, are extremely sensitive to the mechanical deformation (Wyke, 1967) which can result from excessive scar tissue, defective apposition of surgically sutured ligaments or unattended ligament ends (Prins, 1978).

Swelling

Periarticular swelling can result from chronic joint inflammation or the persistent joint irritation that occurs with ankle instability or excessive scar tissue formation (Prins, 1978). Sondenaa et al. (1986) reported swelling in 95% of ankle fractures three months post immobilization and cast removal. Prins (1978) noted that in his study of ankle ligament injuries 30% of the tested ankles still demonstrated swelling six months following the completion of treatment. The swelling generally occurred after a period of prolonged weight bearing. Excessive and persistent swelling of the soft tissues can lead to periarticular adhesion formation (Salter, 1970) contributing to the clinically stiff joint.

ROM

Injury and immobilization can result in a decrease in joint range of motion. This may be due to limitations in joint capsule mobility as well as shortening of the soft tissues. Muscle shortening occurs when a limb is immobilized in a muscle shortened position (Goldspink et al., 1974; Williams and Goldspink, 1981). The ankle joint restriction in the posterior glide of talus during dorsiflexion may be due to scarring and decreased elasticity of the anterior capsule. Similarly, the anterior glide of the talus during plantarflexion is restricted by posterior capsule tightning (Derscheid and Brown, 1985). Most ankle injuries have more involvement of the anterior capsule than posterior, and dorsiflexion is more limited and more difficult to regain than plantarflexion (Derscheid and Brown, 1985; Sondenaa et al., 1986). Periarticular and intraarticular adhesions resulting from chronic swelling, excessive scar tissue formation and hemarthrosis will also contribute to the limitation of joint range (Salter, 1970). Sondenaa et al. (1986) have demonstrated a mean difference initially of 0.26 rad. with unaffected ankle in dorsiflexion and a mean of 0.31 rad. difference in plantarflexion. This ROM is quickly regained especially in the first six weeks post immobilization of an ankle fracture; the maximal gain occurs by 12 weeks in subjects receiving physiotherapy (Sondenaa et al., 1986).

Muscle strength

Muscles of immobilized animal limbs show a decrease in time to peak tension as well as a decrease in maximum twitch tension (Maier et al., 1976). This decrease in tension or force production presumably reflects a smaller contractile protein mass since some investigators have paradoxically shown that tension per gram of wet muscle weight had increased (Mann and Salafsky, 1970). Ankle injury is associated with muscle weakness presumably due to disuse atrophy. Pain and swelling will alter the level of muscle activity (Derscheid and Brown, 1985) and promote atrophy.

Neurological function

Mechano-receptors and/or muscle receptors can be lost or damaged due to injury. Ligamentous and capsular trauma may lead to partial joint de-afferentiation since nerve fibres in the ligaments and capsule have a lower tensile strength than collagenous fibres (Freeman et al., 1965). The ability to replicate joint angles with the unaffected extremity has been shown to be deficient in post injury ankles as long as eight months post trauma (Glencross and Thornton, 1981). However the magnitude of this deficiency was less than the error limits of goniometric measurements (Boone et al., 1978). There is as yet insufficient evidence that fractured joints and their surrounding tissues are neurologically impaired.

Increased resistance to movement

The resistance to passive joint movement is often encountered in a clinical orthopedic and neurological conditions (Wright, 1973) and has a varied nomenclature including: spasticity, rigidity, hypoextensibility, stiffness, compliance, hyper/hypo tonicity, flexibility, hyperreflexia, and muscle tightness. These terms have been used interchangeably and much confusion reigns due to imprecise definition, the subjective nature of this complaint and the inadequate tool with which we measure such phenomena (Weigner and Watts, 1986).

Summary

Joint injury and immobilization have been shown to alter the morphology of both passive (tendon, ligament, muscle, bone and capsule) and active structures (contractile proteins) as well as the contractile ability of the muscle. Clinically, these changes manifest themselves as a decreased ROM, an increased resistance to passive joint movement, and a decrease in muscle strength. These macroscopic manifestations of the injury induced changes are the focus of a clinician's assessment to determine the extent of injury, effect of rehabilitative interventions, and progression/ regression of dysfunction. Clinical measurement tools and techniques are essential in objectively monitoring these variables.

Clinical assessment of joint function

The post immobilization course of a fractured ankle joint with soft tissue injury often includes joint swelling, pain, tenderness, possible anatomic instability, a decrease in ROM, and a decrease in muscle strength. These entities are likely to affect the functional ability of the joint and, as such, form the basis and focus of therapeutic intervention and assessment. According to Feinstein (1983) the collection, measurement, and analysis of clinical data should satisfy certain criteria. The measurement should be preservable or recordable, the observations should be objective and the expressions should be dimensional. Qualitative and quantitative accuracy should be demonstrated in terms of validity and reliability.

The current, clinically accepted assessment tools are limited as they do not satisfy all of the above. For example, the manual muscle test is the most commonly used clinical method for assessing muscle strength (Nicholas et al., 1978; Pedneault, 1985). In this test, weaker muscles are graded by their ability to overcome the resistance of gravity throughout their available ROM (Daniels and Williams, 1946; Janda, 1983; Kendall and Kendall, 1949; Lovett and Martin, 1916). This is a highly subjective, and qualitative method of muscle grading that uses an ordinal scale (Michels, 1983). It lacks the ability to discriminate small but clinically important changes in muscle strength. Moreover, the performance of 40% of the extremity muscles such as the foot, hand and rotator muscles are not significantly affected by gravity and hence, these methods, which are based on gravity, lose validity (Wakim et al., 1950).

Limb girth or circumference has been used to assess the amount of muscle atrophy and/or the amount of swelling in a limb (Hoppenfeld, 1978). Such measurements are taken at standardized locations such as the greatest circumference of the limb. This technique does not differentiate between the relative contributions of intracellular fluid accumulation, structural changes in the muscle fibres or subcutaneous fat, all of which could alter limb girth. Limb girth measurements do not differentiate between the relative amounts of atrophy of muscles agonists or antagonists. Muscle cross sectional area poorly accounts for the differences in ability to produce tension. Muscle areas calculated from limb circumference correlate to a fair degree with muscle strength in elderly people (Pearson et al., 1985a) but there is a large variability in the strength per unit area. Maughan et al. (1983) have suggested that some of the variability in strength per area could be attributed to the proportion of fibre types. Pearson et al. (1985b) have suggested that contractile elements can hypertrophy and atrophy at the expense of such non-contractile elements as fat. Simple measurement of limb girth is a poor predictor of strength since fibre type and content can vary and produce changes in contractile ability of the muscle without apparent changes in limb girth.

Goniometry is defined as the measurement of joint range of motion with a protractor like instrument. Although this technique is quantitative, it is subject to measurement error (Stratford et al., 1984) with the examiner, the instrumentation and the subject all contributing to this error. The examiner must strictly adhere to a standardized technique based on the placement of the instrument over anatomical landmarks (Trombly, 1983). These are obvious in lean subjects but obesity, scars and bony deformities can lead to inaccuracies in landmark location. Moreover goniometry involves a static as opposed to a dynamic measurement and may not be representative of joint range during movement. The choice of goniometers is important since small goniometers intended for finger measurements are inadequate for larger lower extremity measurements. The scaling is not uniform on all goniometers and some can be read only to the nearest .09 rad or .04 rad. Reading of the scale also provides potential for error. Low (1976) demonstrated an end digit preference for numerical values ending in zero. The expectation of a value based on a previous measurement will introduce measurement error (Stratford et al., 1984). The reliability of goniometry has been the subject of much controversy and Boone et al. (1978) concluded that when more than one observer is responsible for ROM evaluation, changes of 0.09 rad for the upper limb and 0.1 rad for the lower limb should be taken into account before one can declare improvement or significant change. When it is considered that 0.17 rad dorsiflexion is the required functional range for sedentary people (Lindsjo et al., 1985) it is obvious that clinically important changes could occur and not be detected by this assessment technique.

In addition to the techniques outlined above, the patient's subjective account of functional ability, pain and stability are often critical to the clinician's assessment of joint function. Language and communication barriers can make this information inaccessible or it can lead to misinterpretation of both questions and answers. The credibility of such statements has to be taken into account and little can be done to quantitatively verify these claims. Other methods of assessment including dynamometry, isokinetic testing, gait analysis, roentogenic (X-ray) examination and electromyographic studies are used less frequently, since they are less readily available and also have their limitations. There is varied consensus as to how the results of these measurements correlate with ankle function. This can be exemplified by the fact that some authors stress the radiographic evidence of anatomical reduction of the fracture site as the important determinant of normal ankle function (Mitchell et al., 1979; Tunturi et al., 1983) while others do not support this notion (Bauer et al., 1985).

There are few available clinical assessment tools that satisfy the criteria of being quantitative, objective, reliable, and valid measures of joint function. Moreover there is a problem legitimizing and quantifying some of the subjective complaints which are so often an essential component in the evaluation of joint behavior. It is clear that improved methods and tools are needed.

Quantitative assessment of joint function

Resistance to passive movement

To date the quantitative assessment of joint resistance to passive movement has been studied primarily in patients with neurological conditions displaying alterations in muscle 'tone'. This increase in resistance to passive motion has been largely attributed to hyperexcitable reflexes with its central and peripheral influences (Burke, 1983). There is, however, increasing evidence that the resistance to joint displacement may be, in part, attributed to non-neuronal physical changes such as possible inactivity-related muscle shortening (Gossman et al., 1982) or altered mechanical properties of muscle fibres or the connective tissues within these muscles (Dietz et al. 1981; Dietz and Berger, 1983). Changes in resting muscle length would also affect the torque-angle relation at the joint acted upon by the shortened muscle.

Watts et al. (1986) measured resting 'tone' or stiffness at the elbow joint to study Parkinsonian rigidity. Compliance values, defined by these authors as the slope of the linear relation between torque and position were found to be moderately correlated with upper arm volume in the control subjects but less so in Parkinsonian subjects. The authors conclude that the observed changes in joint stiffness result from alterations in the passive mechanical properties of the upper limb, however, the authors did not correct for differences in subcutaneous fat and bone volume when making their volume calculations.

Tardieu et al. (1982) termed the resistance to passive movements as 'hypoextensibility'. In the investigation of ankle plantarflexor and dorsiflexor torques in cerebral palsy (CP) children, the authors stressed the importance of controlling for muscle activity in order to isolate the torque produced by the elastic properties of the passive tissues. In their study the ankle joint was passively moved to various joint positions 0.09

rad apart. One minute between successive test positions was allowed for the visco-elastic events to stabilize and then the torque was measured at that position. Where complete electrical silence was not obtained upon voluntary relaxation, an ischemia technique was used to reduce the ankle reflexes and suppress voluntary contraction. The results of this study must be viewed with caution since there were no normal controls for the imposed ischemia condition. Moreover, the ROM tested was limited by a predetermined ceiling torque of 5.25 7.0 Nm which corresponded, in some cases, to extremely limited values.

Dynamic passive torques

Dynamic passive torques were investigated in clinically spastic and Parkinsonian subjects at varying angular velocities (Bromberg and Grimby, 1983). The slope of the passive torque-position curves demonstrated velocity dependence in the spastic subjects but not in those with Parkinsonian rigidity. These slopes also changed with medication and therapy. This study demonstrated that passive torque recordings could be used as an objective evaluation of hypertonia and as an evaluation of treatment effects. However, EMG were not recorded and therefore the influence of contraction and reflex activity, and learned voluntary responses on their results could not be determined.

Dynamic Joint Mechanics

Ankle dynamic stiffness has been studied in normal subjects by several investigators. Agarwal and Gottlieb (1977) used sinusoidal perturbations of torque to characterize stiffness in relaxed and tonically active ankles. Stochastic band limited Gaussian torques were later used by the same authors to eliminate evoked adaptive changes in the system dynamics due to periodic inputs (Gottlieb and Agarwal, 1978). Perturbations of torque about mean joint positions demonstrated the dependence of joint compliance on joint angle. Hunter and Kearney (1982) applied stochastic ankle perturbations of position instead of torque and demonstrated the ankle stiffness' variation with mean ankle torque. They demonstrated that the dynamic behaviour of a joint was independent of the use of either torque or position perturbations. Joint stiffness has been modeled as a second order linear system, having inertial (I), viscous (B), and elastic (K) terms (Gottlieb and Agarwal, 1978; Hunter and Kearney, 1982).

Dynamic stiffness studies using perturbations of position have investigated the dependence of joint mechanics on various physiologic and kinesiologic variables. Dynamic stiffness varies systematically as a function of mean displacement amplitude (Kearney and Hunter, 1982). The elastic and viscous terms decrease as the displacement amplitude increases. The magnitude of tonic muscle contraction also has considerable effect on ankle stiffness (Hunter and Kearney, 1982; Weiss et al., 1988). The elastic parameter (K) increased in proportion to the level of contraction (torque) while the viscous parameter changed in such a way as to keep the damping factor constant (Hunter et al., 1983).

Muscle fatigue, provided that torque levels are maintained, does not seem to affect ankle mechanics (Hunter and Kearney, 1983) whereas the mean joint position at which the perturbations are applied does, even in the absence of muscle contraction (Weiss et al. 1986a). The mean torque and stiffness are small at mid-range but thereafter increase greatly as the joint reaches its extremes of ROM. Under active conditions of muscle contraction joint stiffness does not demonstrate this position sensitivity.

Joint stiffness has been shown to rise very steeply without an increase in net joint torque under conditions of muscle co-contraction (Hunter et al., 1983), therefore torque recordings with EMG recordings are essential to determine the contribution of active components to the final results.

A few investigators have also used these techniques to investigate abnormal joint mechanics primarily in neurologically impaired individuals. Gottlieb et al. (1978) applied sinusoidal perturbations of torque in clinically spastic patients and measured the effective joint compliance and the evoked EMG activity. Tonically active muscles even with complete subject relaxation precluded a purely mechanical explanation of stiffness. Work is currently under way in this laboratory to characterize ankle mechanics in a Parkinsonian popluation under conditions of varying activation tasks (Dannenbaum et al., 1988).

The investigations of dynamic stiffness (or its inverse compliance) are insufficient in number and have been limited to neurologically impaired individuals where reflex as well as mechanical changes have occurred and possible confound the measurement results. Moreover these studies have severe limitations that preclude generalizations to a larger population.

Problem formulation

The current clinical literature attests to the lack of objective, quantitative, and reliable approaches to the evaluation of joint function/dysfunction and the effects of therapeutic or remedial intervention. Previous studies of joint resistance to passive movement, a frequent characteristic in abnormal joint function, have focused on the difference between normal and neurologically impaired individuals having possible physiological and structural changes in muscle as well as alterations in motor and reflex responses. The results of these studies strongly support the occurrence of physiological, non-neuronal changes in muscle and joint structures. The orthopedic literature also provides evidence that orthopedic events induce these changes, e.g. an increase in tissue collagen content following immobilization or injury, yet quantitative assessment of these possible mechanical changes has received little attention.

The past investigations of stiffness in pathological conditions have had methodological weaknesses such as the use of limited joint ranges, limited contraction levels, and static conditions. There has been little quantitative measurement of the overall joint mechanics in any subject population.

Fast quantitative methods are available but to date, have been used to study normal joint mechanics. There have been no reliability studies to assess the stability to these measures over time in a normal population nor which of the dependent variables are most sensitive to changes in joint function. The investigation of stiffness in a case study of a functionally deafferented patient (Weiss et al., 1984b) has indicated that the procedure would be tolerated by persons with physical dysfunction. These methods can provide valuable information on the contribution of passive and active structures to joint behaviour and may possibly monitor changes that to date are subjectively evaluated. It is therefore appropriate to investigate next joint stiffness in individuals with orthopedic pathology in order to evaluate its potential as a clinically significant measure of joint function.

Objective

The objective of this study was to measure dynamic ankle mechanics (stiffness) in a group of normal subjects to investigate the feasibility of using this measure to monitor joint dysfunction and recovery to normal function in a subject who had sustained a uni-lateral orthopedic ankle injury. This was accomplished by:

1. establishing the feasibility of the systems identification approach and experimental protocol for clinical evaluation;
2. determining the reliability of the various measures of ankle joint behavior;
3. examining measures of joint function that may be sensitive to changes in joint function in an orthopedic case study.

CHAPTER III

GENERAL METHODS

This chapter will outline some of the general experimental methods, data processing and analysis used in the study. The justification for the present experimental paradigm will be presented. Some of the methods are brief or omitted since these sections are addressed in more detail in chapters four and five.

All experiments tock take place in Dr. Robert Kearney's laboratory located in the McGill Biomedical Engineering Unit. The study plan was approved by the combined McGill and Royal Victoria Hospital ethics committee.

Subject profile

Fifteen volunteer control subjects (mean age of 27.5 years), referred to by number and initials throughout the thesis, with no known orthopedic or neurological dysfunction, participated in the reliability study. One 26 year old male individual in the post immobilization period following an undisplaced malleolar fracture took part in the case study. None of the subjects were highly trained athletes or participated in competitive sports on a regular basis.

Experimental equipment and variables

Experimental instrumentation

Fig. 3.1 shows the experimental apparatus which consisted of a table upon which the subject lay supine. Straps over the pelvis and thighs were used to stabilize the trunk, restrict movement and maintain body position relative to the foot. The subject's right foot was supported on a cushioned aluminum plate and the left foot was attached to the electrohydraulic actuator pedal by means of an individually constructed fibreglass cast. A torque transducer and a rotary position transducer were mounted coaxially with the actuator. A dual channel oscilloscope providing visual ankle torque feedback was positioned above the subject.

Four mechanisms, including a mechanical stop, a hydraulic stop, servo control parameters, and a panic button within the subject's reach, ensured the subject's safety (Kearney et al., 1983).

Ankle Angular Position and Torque

Ankle angular position was recorded with a Beckman potentiometer (Model 6263-R5K-1.50) having a conductive plastic resistance element with a total resistance range of (0-5k). Ankle angles were measured using a reference position determined by a line parallel to the first metatarsal of the foot oriented perpendicularly to a line joining the medial tibial plateau to the medial malleolus. Ankle torques were recorded with a Lebow torque transducer (Model 2110-5k). This transducer consisted of a 4 arm bonded strain gage bridge with a rated capacity of 5,000 in-lb.

Fig. 3.1. General experimental apparatus. The ankle actuator is mounted on the experimental table which is equipped with straps to stabilize thighs and hips. The fibreglass cast limb fixation device is attached to the actuator pedal.

Actuator input

A 1024 point pseudo-random binary sequence (PRBS) repeatedly displayed at 500 Hz was generated by the computer (MicroVax II) and was used to drive the ankle actuator. The perturbation sequence had a peak-to peak amplitude of 0.05 rad and contained power adequate for system identification purposes up to 50 Hz (Kearney et al., 1983). A sample of torque and ankle position records and the impulse response function (see below) relating the two are shown in Fig. 3.2.

Experimental paradigms

Orientation and training session

Prior to the collection of experimental data each subject was requested to attend an orientation session to become familiar with the instrumentation and experimental procedures. Subjects were asked to sign an informed consent statement. A fibreglass cast was individually constructed during the initial orientation session, a technique which will be described in chapter four.

Three experimental paradigms were carried out during each experimental session. These included:

1 - Maximum Voluntary Contraction (MVC)

2 - Position ramp without perturbation (Passive static paradigm)

3 - Position Ramp with perturbation (Passive dynamic paradigm)

MVC Paradigm

Subjects were required to respond to a step change in the tracking stimulus as displayed on the oscilloscope. A maximal dorsiflexor torque was generated and subjects were instructed to maintain this maximal muscle contraction for 5 s after which they were allowed to rest. The trial was accepted provided the coefficient of variation of the last three seconds did not exceed .10. If a coefficient of variation of greater than this value occurred the attainment of an MVC was questionable (Kroemer and Marras, 1980) and the subject was requested to repeat the procedure. The 5 s maximal contraction was then repeated for the plantarflexor muscles to obtain the maximal plantarflexor torque. Maximal EMG amplitudes were also recorded. Typical recordings of dorsiflexing and plantarflexing MVC's are shown in Fig. 3.3. The arrows denote the portion of the record used in the maximal torque and EMG calculations. As expected, the MVC torque was quite stable throughout the 3 s interval.

Passive static paradigm

The joint was rotated twice through its available ROM at 0.05 rad/sec. This angular velocity was chosen to avoid evoking reflex responses to the muscle stretch or shortening which could alter the passive joint torque profile. The subject was instructed to relax during the trial and the absence of reflex activity was verified by monitoring the EMG signals.

Passive dynamic Stiffness

This paradigm was essentially the same as the position ramp paradigm except that a random perturbation (0.05 rad peak-to-peak) was superimposed on the ramp displacement.

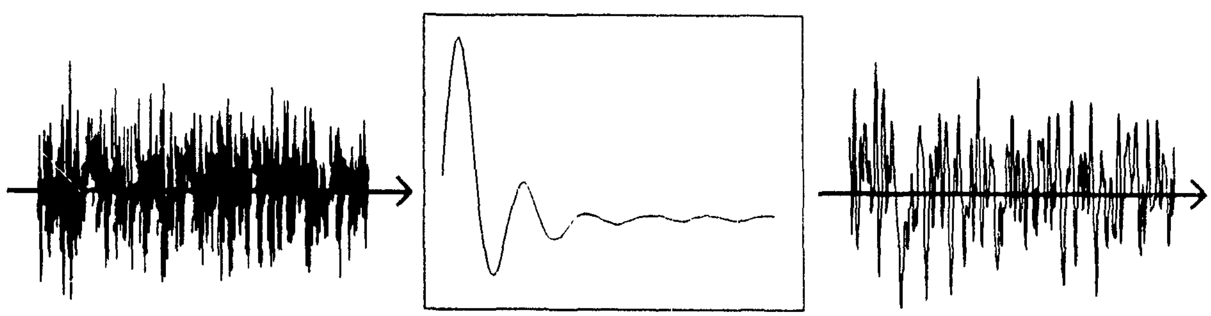

Fig. 3.2. Sample of torque and position records and the impulse response function (IRF) relating the two.

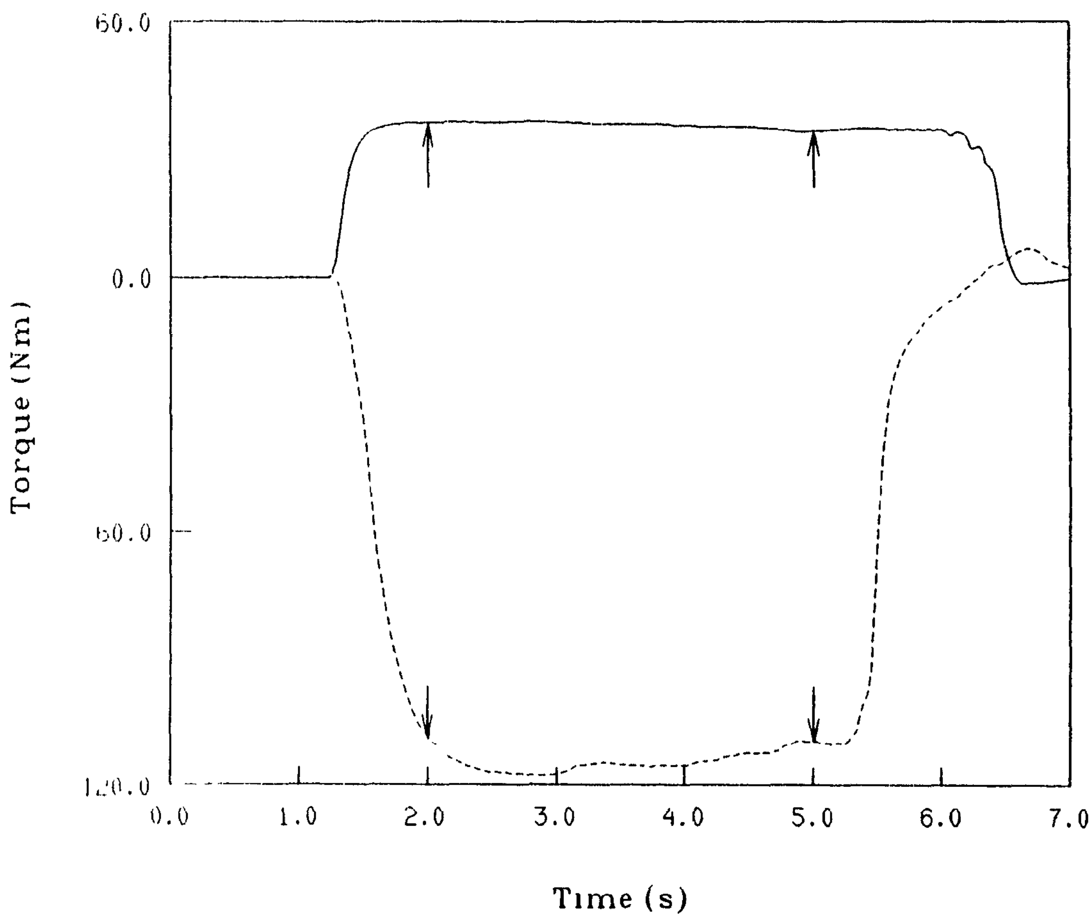

Fig. 3.3. Dorsiflexor (upper signal) and plantarflexor (lower signal) torque records for one individual (RM) sampled during the MVC paradigm. The limits of the record length used in the MVC calculations are indicated by the arrows.

Experimental Schedule

The control subjects were assessed twice on separate days (mean interval 16 days). The patient was assessed six times over a period of 5 months throughout the post immobilization recovery period.

Clinical evaluation

Anthropometric measures were collected and a clinical evaluation was performed on each individual to determine whether any unreported disorders of the foot and ankle were present. In the case study these measures served to monitor the patient's status using standard clinical tests.
These procedures are outlined in Appendix I.

General data analysis and statistical treatment

This section will deal in a more expanded fashion with some of the anlytical techniques briefly mentioned in chapters four and five. Those topics not covered here are to be dealt with in the above mentioned chapters. All data analysis modules are listed in Appendix II.

Actuator and Boot Inertial Corrections

Part of the sampled torque signal was due to the dynamics of the ankle actuator and the fibreglass cast and had to be removed. The impulse response function (IRF) between boot angular displacement and torque was determined and convolved with each combined ankle/boot experimental record to generate an estimate of the boot's contribution to the sampled torque. This estimate was then subtracted from the sampled torque to yield the portion of the torque due to the subject. This inertial correction was performed on every set of dynamic data.

Dynamic Measurements

System identification techniques were used to analyze data from the passive dynamic paradigm. These techniques produce optimum results when the data signals are stationary or free from linear trends. Linear trends were removed from each 2.05 s section of the position and torque records corresponding to each PRBS sequence. This improved the variance accounted for by the compliance IRFs relating torque and angular position. Ankle stiffness was then modeled parametrically using non-linear minimization techniques on the non-parametric IRF to estimate values of the inertial (I), viscous (B), and elastic (K) parameters of a second-order, differential equation of the form:

$$T(t) = K\theta(t) + B\dot{\theta}(t) + I\ddot{\theta}(t),$$

where: t = time (s),

T = ankle torque (Nm),

I = inertial term (Kgm^2),

B = viscous term (Nms/rad),

K = elastic term (Nm/rad),

Reliability

Repeated measures of the MVC, passive static mechanics and passive dynamic mechanics were obtained. The reliability of the various measures was estimated using Pearson's product moment correlation. For purposes of this study a variable with a coefficient greater than 0.80 was considered reliable. These results are discussed further in chapter five.

Choice of subject parameters

Since the purpose of this study was to investigate the feasibility of using a system identification approach in the clinical assessment of joint function, a joint, clinical population or pathology, and a subject group must be selected. The rationale for the various choices is outlined in the following sections.

Selection of joint

The ankle joint was investigated because of its high incidence of fractures relative to other joints (Buhr and Cooke, 1959; Fife and Barancik, 1985), its importance to locomotion, and its role in many occupational tasks.

Selection of pathology

Previous investigations of joint stiffness (or its inverse compliance) under pathological conditions have focused mainly on subjects with neurological disorders. There is evidence that physiological and structural changes contribute to the stiffness seen in these individuals. The literature also suggests the occurrence of these same changes in orthopedic injuries with limb immobilization and yet there are few quantitative studies addressing joint stiffness measures in this group. The investigation of joint mechanics in a situation of mechanical orthopedic dysfunction is therefore appropriate.

Selection of subjects

The results are to be inferred on a population consisting of adults ages 23-40 years. This age group was selected in the light of epidemiological and practical considerations. Ankle fractures are more common in this age group which corresponds to the second increase in incidence in the bimodal distribution of ankle fractures (Landin and Danielsson, 1983; Nilsson, 1969), and is associated with the working age population. The availability of subjects in a university setting for both experimental and control groups also favors this age group.

An approximate 15 year age span was used in the study to limit the variability due to age since ROM, muscle strength, and contractile properties are known to change with age (Davies and White, 1983; Pearson et al., 1985b; Sammarco et al., 1973). We also excluded elite athletes or individuals who may have incurred unusual adptations of musculoskeletal structure by virtue of their occupation or recreational activities. An example of this would be a ballet dancer where the joint is subjected to large forces at the extremes of the ROM and muscles and ligaments are trained for joint flexibility and muscle extensibility (Teitz, 1982).

Justification of experimental paradigms

The purpose of the first two paradigms (MVC and passive static paradigms) is to collect data comparable to the accepted clinically significant measures of ankle function; ROM and muscle strength as characterized by the MVC, and passive torque which corresponds to the resistance to passive joint displacement. These paradigms serve as a link between the clinically accepted measures of ankle function and the more novel parameters of passive static and passive dynamic ankle mechanics for example, the passive torque-position curve loop area and the slope and intercept of the relation between elastic stiffness and torque that will be explained further in chapter five.

CHAPTER IV

LOW INERTIA, RIGID LIMB FIXATION USING FIBREGLASS CASTING BANDAGE

Introduction

The development of clinical and experimental measurement devices to assess joint function, the effects of pathology and remedial intervention constitute an important goal in research. An accurate and objective method of assessing joint behavior using stochastic analysis, large frequencies, and large torques is being investigated in this laboratory. This required the development of an improved method of limb fixation to the experimental apparatus.

Traditionally, limbs have been attached to clinical and laboratory measurement devices using straps, cuffs and slings (Muwanga and Dove, 1985; Holmes and Alderink, 1984; Hof and Van Den Berg, 1977). There are a number of problems with these methods which may affect the reliability and validity of measurements. First, the attachment link must be tight to transmit forces accurately. As a result large forces will be concentrated over a small surface area, which may result in subject discomfort, circulatory disturbances, and soft tissue injury. Second, simple straps and cuffs, even when tightly secured, cannot totally eliminate movement between the limb and actuator. Finally, it is difficult to replicate accurately limb position (e.g. the joint axis of rotation) relative to the apparatus when reproducible measures of force and displacement are needed.

Previously we developed a technique using expanded polyurethane foam block casts to obtain rigid, low inertia, and comfortable fixation of the foot to an electrohydraulic actuator (Weiss et al., 1984a). Recently we have improved this method significantly through the use of the Dynacast XR Fiberglas bandage. This material is used clinically for the immobilization of fractures; its high strength with early load bearing, moisture permeability and low mass make it a substantial improvement over plaster of Paris bandages for limb immobilization. These characteristics are also ideal for the construction of a low inertia limb fixation device. In this paper we describe the new fixation device, the method of construction, and discuss its advantages in a clinical and experimental context.

Description of fixation

Fig. 4.1 shows a fibreglass cast mounted in our actuator. The cast is fixed to the actuator by two small aluminum sleeves that are attached to the sides of the cast with epoxy and secured to the actuator pedal with two bolts.

Comparison to previous technique

This new method of limb fixation has several advantages over our previous technique. First, the fibreglass material is rigid enough to provide adequate structural rigidity. The cast is attached to the actuator with small aluminum sleeves rather than with an aluminum support plate as before. As a result the inertia of the device is greatly reduced. Secondly, the method of construction is more efficient and the final product more comfortable than the previous polyurethane foam cast method. Finally the fibreglass material is easy to work with and the cast can be built in a single stage.

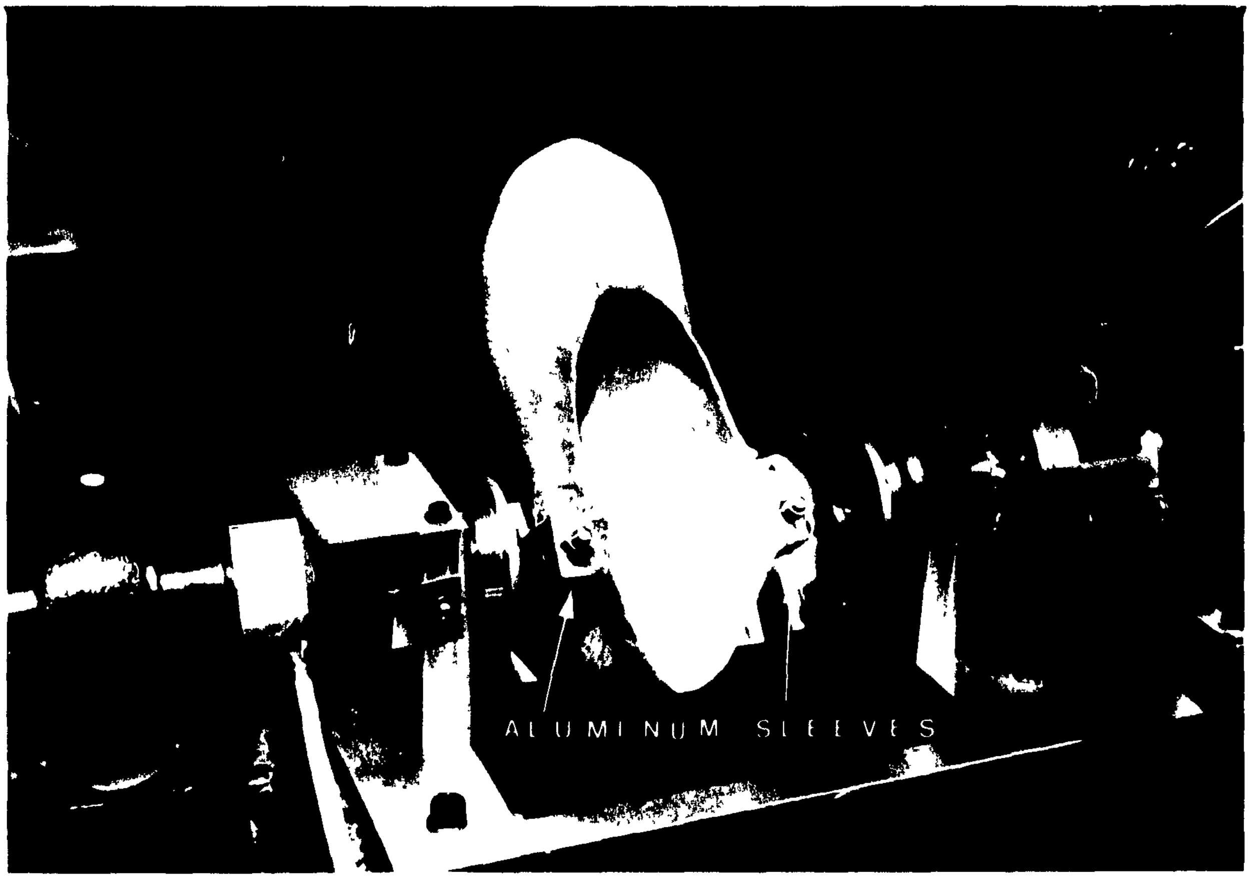

Fig. 4.1: Finished fibreglass cast attached to the actuator pedal by means of aluminum sleeves bolted to the pedal.

Cast Construction

Fibreglass casting material

Casting bandage consists of a knitted fibreglass substrate impregnated with a water activated polyurethane resin. It produces a light, rigid cast that sets within several minutes and is load bearing within 30 minutes. The actual setting time may be increased by storage at lower than recommended temperatures (i.e. 17°C instead of 22°C) or can be decreased by immersing the bandage roll in tepid water and squeezing prior to wrapping.

The material can be gently stretched in all directions allowing ease of wrapping and molding to the contours of the foot. Once cured the cast can be cut and trimmed using shears or a plaster saw and rough edges can be filed easily or sanded smooth. The finished cast is aesthetically pleasing and non-toxic[1] although care must be taken to avoid direct skin contact with the polyurethane resin during wrapping. In such case, small amounts of the resin can be easily removed from the skin with acetone.

Preparation

Fragile skin or open sores, if present, were lightly dressed to avoid unnecessary pressure. The subjects sat comfortably and wore a nylon knee-high (to facilitate cast removal) and a cotton stockinette over the foot and lower leg (to protect the skin). The stockinette then adhered to the fibreglass forming the lining of the finished cast. Rubber gloves were worn to prevent the resin from adhering to the technician's skin.

---

1. The manufacturer reports that approximately 5% of the population appear to be allergic to direct contact with the cured product when exposed for prolonged periods.

Wrapping Technique

One or two rolls of 7.5 cm by 3.6 metre of Smith & Nephew Dynacast XR Orthopaedic Casting Bandage were used. It has been our experience that any foot larger than a woman's size 9 or men's 7 requires from one and a half to two rolls. Two reinforcement strips of the casting bandage were applied before wrapping. The first strip was placed along the plantar surface from the tendino achilles-calcaneal junction to the metatarsal phalangeal joint. The second strip covered the calcaneal fat pad spanning the distance between the superior tips of the malleoli. The malleoli must be covered to ensure that the ankle axis of rotation could be identified on the cast.

The foot and reinforcement strips were then wrapped under slight tension using a modified figure of eight pattern, similar to the standard plaster of Paris wrapping technique (Lewis, 1977). The limb was wrapped so as to provide the most material in areas of highest stress. Wrinkles in the material were avoided since these can cause painful areas of localized pressure. This is particularly important over the malleoli and calcaneal tendon.

The fibreglass rolls are not immersed in water prior to wrapping as in the standard cast room procedure. This slowed the setting time and provided the additional time required to mold the cast. Once wrapping was completed the cast foot was rubbed briskly for 4-5 minutes using a lubricant (not oil or petroleum jelly). Our experience was that this improved contouring, accelerated setting, and promoted a good lamination which was desirable for overall cast strength. This strength is dependent in part on the speed of wrapping, the amount of contact between the layers and alignment of the fibreglass threads. Care must be taken to ensure proper foot and toe positioning as toes tend to adopt a position of flexion and the foot one of inversion-plantarflexion.

The cast instep was cut and widened (with bandage shears) just before setting was complete to facilitate its removal. The malleoli were then palpated and marked on the cast. The cast was removed only when set in order to preserve the molded shape and was then washed to remove any lubricant residue. It was left to dry overnight.

Finishing the fibreglass cast

Once set, the cast must be attached to the actuator in such a way as to align the ankle axis of rotation with that of the ankle actuator. This was done using a jig, shown in Fig. 4.2, which was an exact replica of the actuator pedal. Two 75 mm aluminum sleeves were suspended in the mold by means of a 2.5 mm diameter stainless steel pin driven through holes previously drilled in the cast at the ankle axis. The location of the ankle axis of rotation was estimated from the malleolar markings as described by Inman (1976). The new method reduced considerably the error in ankle axis estimation which allowed an anatomically truer and more comfortable range of ankle movement.

An industrial strength steel-filled epoxy putty (Devcon Plastic Steel Putty) was used to fasten the aluminum sleeves to the cast such that the axis of rotation was coaxial to the actuator and the foot horizontal to the foot pedal. The time and amount of adhesive required to glue the sleeves to the cast depended on the obliquity of the ankle axis of rotation since the

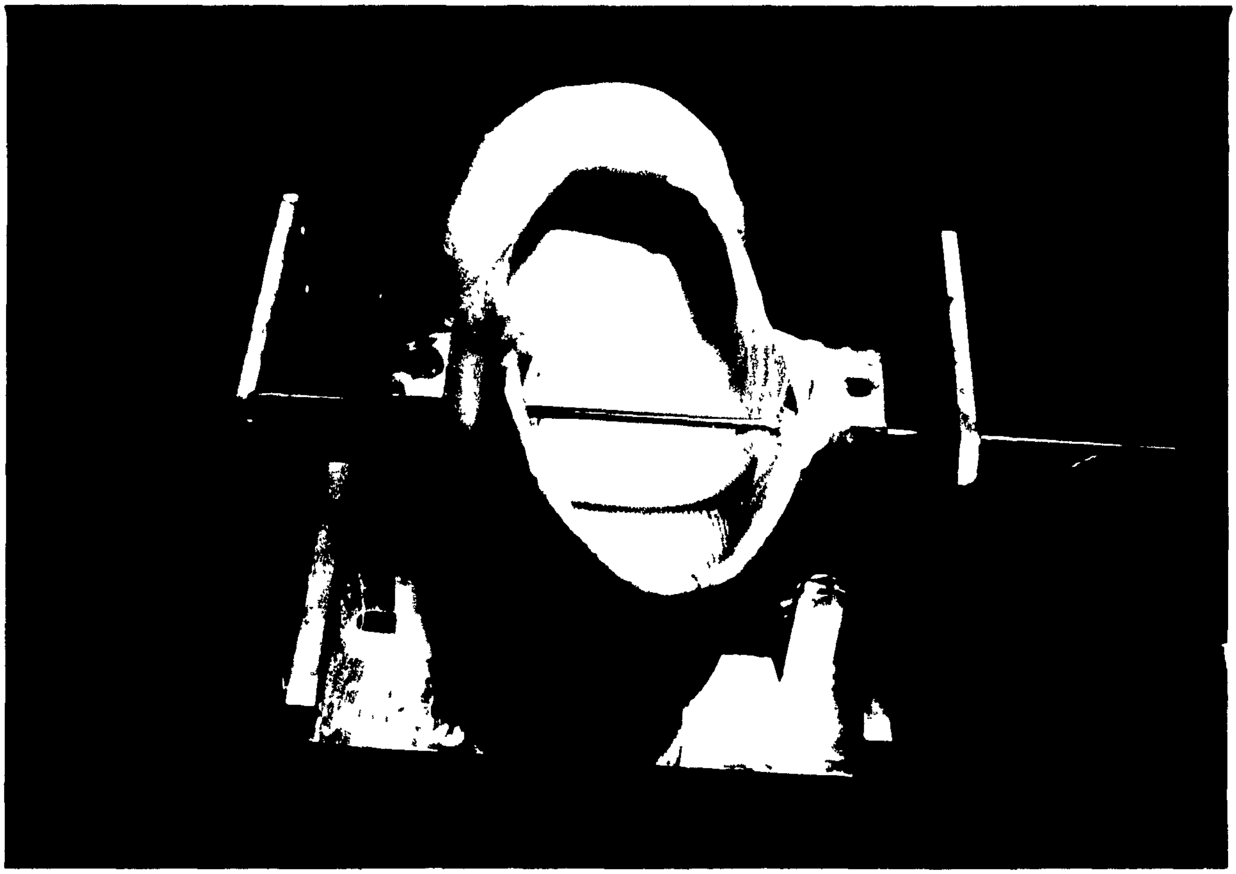

Fig. 4.2: Fibreglass cast suspended in jig by means of stainless steel pin inserted at approximated joint axis of rotation.

space between cast and sleeves had to be filled. The subject population and experimental objective must also be considered when gluing the sleeves since a highly trained subject will generate maximal torques that are much larger than an untrained individual. To withstand these larger torques the adhesive bond must be strengthened with further applications of glue.

Once cured and dried the cast was trimmed of excess material particularly over the tuberosity of the cuboid on the lateral aspect of the foot to facilitate foot insertion. Note, however, that it was important to maintain the "cup" shape of the heel and the malleoli in order to keep the cast on the foot. The cut edges of the cast were sanded and the stockinette lining secured to the exterior using porous skin tape. The cast fit like a tight, rigid shoe and subjects were advised to gently "ease" the foot into the cast; in some cases a shoe horn had to be used. Fig. 4.2 shows a completed cast bolted to the actuator-driven foot plate by means of the aluminum sleeves. A velcro band was fastened over the foot instep to compensate for the weakness induced by the removal of fibreglass material over the dorsum of the foot.

Discussion

Little subject preparation was required and the technique was so simple that we were able to construct these casts in the subject's home. The finished product was light (mass ranging from 300 to 400 gm) and had a small moment of inertia. The moment of inertia of the combined fibreglass cast and foot pedal was in the order of 0.007 kgm^2. This compares favorably with previous fixation devices which resulted in inertias as large as 0.025 kgm^2 (Weiss et al., 1984a). Moreover it was durable, rigid, and tolerated well by both our control and clinical subjects. Casts constructed according to this method can tolerate large forces without breakdown and have withstood ankle torques of over 130 Nm.

The reduction of inertia is important since large fixation device inertias will slow the dynamic response of the actuator. They will also produce large forces during rapid limb displacement which may mask the smaller elastic and viscous forces that are of central interest. This is particularly important when experimental inputs spanning a broad range of frequencies are used (as in our own studies of dynamic ankle joint mechanics) since at higher frequencies inertial forces dominate.

Indeed, reducing the cast inertia can effectively increase the resolution of the torque measurement. Thus, for a n bit A/D converter, the smallest torque that can be resolved will be,

$$\frac{T_{max}}{2^n - 1}$$

where T_{max} is the maximum torque signal to be recorded. Inertial effects dominate joint dynamics so that at high frequencies and to a good approximation,

$$T_{max} \propto I_{cast} + I_{foot}$$

Thus the resolution of the torque measurement will be proportional to,

$$\frac{I_t}{2^n - 1}$$

where I_t is equal to the total inertia of the cast and foot. Using typical values it can be seen that by reducing the moment of inertia of the fixation device we have effectively doubled the resolution of our torque measurements.

Even the small inertia associated with the light casts is of some concern and has to be removed by analytic methods. The reduction of cast/footplate inertia is achieved using analytical techniques based on system identification and is similar to that used with the polyurethane foam cast blocks (Weiss et al., 1986a).

We believe that these fibreglass casts provide excellent, rigid fixation of the foot and that they minimize relative movement between foot and the actuator. Quantitative measures of how well this is achieved are difficult to obtain. Indirect evidence is, however, available by examining our estimate of joint compliance obtained with the cast. Linear impulse response functions accounted for almost all of the observed torque variance, and these IRFs were, in turn, described very well by a second order model having parameters corresponding to those expected for the ankle. Bending of the cast or relative movement of the foot relative to the actuator would be expected to give rise to more complex high frequency dynamics (e.g. Ma and Zahalik, 1985) and/or to reduce the variance accounted for. Consequently the high VAFs of our linear IRFs and the excellent second order fits to them indicate that the cast provided excellent fixation over the range of frequencies studied.

CONCLUSIONS

We have developed a technique for the construction of a fibreglass cast used in the fixation of the foot and ankle to an experimental apparatus. The technique is simple and requires very little subject cooperation and time. The cast is light, comfortable and conforms well to the subject's foot. Motor control studies and the more classical joint functional assessments have been performed with this device and the results to date have been excellent.

Chapter V

DYNAMIC ANKLE MECHANICS AS A RELIABLE MEASURE OF ANKLE JOINT FUNCTION

INTRODUCTION

Both neurological disorders and orthopedic injury can alter the hard and soft tissues surrounding a joint and change its mechanical behaviour. For example, the examination of EMG and joint torque profiles in spastic individuals by Dietz and Berger (1983) demonstrated an increase in active joint torque despite reduced EMG activity. The authors suggested that the increase in joint torque was due to altered mechanical properties of the muscle. Passive ankle joint torques were also increased in subjects with chronic cerebral lesions following joint immobilization (Tardieu et al., 1982), possibly as a result of muscle shortening. There is evidence in the orthopedic literature that limb immobilization and orthopedic injury, either alone or in combination, lead to an increase in tissue collagen content (Noyes, 1977; Warhol et al., 1985). These events can also lead to an increased resistance to passive joint movement in excess of ten times that of the unaffected contralateral limb (Heerkens et al., 1987).

It is evident from the above that in injury and disease one can expect changes in joint mechanical behavior which can be observed clinically (Salter, 1970) by means of evaluations based largely on signs and symptoms such as pain, swelling, weakness, and loss of range of motion (ROM). The dilemma is that despite a plethora of assessment tools, few quantitative and objective techniques are available (Dersheid and Brown, 1985).

Previous investigations of pathological joint mechanics have had methodological weaknesses including the use of limited joint ranges (Tardieu et al., 1982; Weigner and Watts, 1986) and qualitative observation of joint behaviour (Bromberg and Grimby, 1983). Quantitative methods based on system identification techniques are available but to date, have been applied to a small numbers of normal subjects in basic research paradigms (Agarwal and Gottlieb, 1977; Gottlieb and Agarwal, 1978; Hunter and Kearney, 1982; Kearney and Hunter, 1982; Weiss et al., 1986a, 1986b). These methods have the potential to provide an objective measure of behaviour that to date has been only subjectively evaluated.

The objective of this study was to determine whether experimental paradigms based on the systems identification approach are an appropriate clinical assessment tool. This was accomplished by conducting two companion studies: a reliability study on a group of fifteen control subjects and a case study of an individual who had sustained a unilateral undisplaced ankle fracture. The objective of the first study was to establish whether selected measures of static and dynamic ankle joint function were reliable. The objective of the case study was to determine whether these procedures could be tolerated by clinical subjects, (2) their data could be analyzed using similar algorithms, and (3) the experimental variables were sensitive to changes in clinical signs and symptoms.

METHODS

Subjects

Fifteen volunteer subjects between the ages of 23 and 39 years (mean 27.5 SD = 4.5), with no symptoms of neurological or orthopedic dysfunction were examined.

Experimental instrumentation

The apparatus has been described in detail elsewhere (Kearney et al., 1983). Briefly, the subject lay supine on a table with straps stabilizing the pelvis and the thighs. The left foot was attached to the pedal of an electrohydraulic actuator by means of an individually constructed fibreglass cast (Morier et al., 1988) and the right foot was supported on an aluminum plate.

A computer generated 1024 point pseudo-random binary sequence (PRBS) displayed at 500 Hz was used to drive the ankle actuator. The resulting position perturbation had a peak-to-peak amplitude of 0.05 rad and contained power adequate for system identification purposes up to 50 Hz (Kearney et al., 1983).

Ankle Angular Position and Torque

Ankle angular position was recorded with a Beckman potentiometer (Model 6263-R5K-L.50) while ankle torque was measured with a Lebow torque transducer (Model 2110-5k). By convention dorsiflexor angles and torques were denoted as positive whereas plantarflexor angles and torques were denoted as negative. Angles were measured with respect to a reference position determined by a line parallel to the plantar surface of the foot oriented perpendicularly to a line joining the head of the fibula to the lateral malleolus.

The range of motion (ROM) of each subject's ankle was determined prior to the first experimental trial. The foot was attached to the actuator and the ankle rotated passively from a neutral position toward maximum dorsiflexion and then toward maximum plantarflexion. Displacement in each direction was continued until the subject could tolerate no further movement (i.e. the onset of pain) or until motion of the knee or lower leg was observed. This procedure was repeated three times and the mean value used to define the end points of the ROM.

All ankle displacements induced by the actuator were initiated at the mid-postion which was defined as the position equidistant from the extremes of the ROM. The potentiometer output was then scaled with respect to this position (as measured with a goniometer) so that all reported ankle positions corresponded to actual anatomical joint angles.

Electromyograms

Surface electromyograms (EMG) were recorded from Triceps Surae (TS) and Tibialis Anterior (TA). The subject's skin was prepared by shaving and briskly rubbing with ethanol. Disposable Ag/AgCl surface electrodes (Hewlett-Packard, 10 mm diameter, Model 14445A) were placed in a bipolar configuration parallel to the muscle fibre direction. The TA electrodes were placed lateral to the tibial crest about 50 mm distal to the tibial tuberosity. The TS electrodes were placed on the lateral head of the

Gastrocnemius just superior to its inferior border. A reference electrode was placed over the patella. EMG signals were high pass filtered at 10 Hz (second order Butterworth), full-wave rectified and then low pass filtered at 1000 Hz (fourth order Bessel filter). EMGs as well as the torque and position signals were then anti-alias filtered (100 Hz) with eight pole Bessel filters and sampled at 500 Hz by a 16 bit A/D converter.

Clinical assessment

Standard goniometric ROM measures (American Association of Orthopedic Surgeons, 1965) were made, manual muscle tests (Daniels and Worthingham, 1972) performed, and anthropometric measures collected on all subjects after each experiment. A sample of the assessment form is included in Appendix I.

Experimental paradigm

Subjects were examined twice at a mean interval of 16 days (SD 32 days). Three experimental trials were carried out during each experimental session. These included:

MVC Paradigm

Subjects were required to generate a maximum voluntary contraction (MVC) in response to a step change in a tracking stimulus displayed on an oscilloscope. Five second records were taken for both dorsiflexing and plantarflexing MVCs.

Passive Static Paradigm

The ankle joint was rotated at 0.05 rad/s from the mid-position to maximum dorsiflexion, back to maximum plantarflexion, and finally to the initial mid-position. The subject was instructed to relax and to neither help nor hinder the movement of the ankle. EMG signals were monitored to confirm that the subject remained relaxed.

Passive Dynamic Paradigm

This paradigm was similar to the passive static paradigm except that a stochastic perturbation of position (0.05 rad peak-to-peak) was superimposed on the ramp displacement. The subject was again instructed to relax and the EMG of both TA and TS was monitored to detect the presence of muscular activity.

Data analysis

Data analysis was done using NEXUS, a computer language for signal and system analysis (Hunter and Kearney, 1984).

MVC Torques

The mean torque generated during the last three seconds of the five second contraction as well as the coefficient of variation (torque SD/mean torque) was calculated. This mean torque was designated to be the MVC provided the coefficient of variation was less than 0.10. If the coefficient of variation exceeded this value the attainment of a true MVC was questionable (Kroemer and Marras, 1980) and the procedure was repeated.

Static Analysis

Passive joint torques were recorded as the ankle was rotated through its ROM and the work required to displace the joint was calculated by integrating along the path.

Inertial Correction

Part of the sampled torque signal was due to the dynamics of the ankle actuator and the fibreglass boot. This was removed prior to further analyses by (1) calculating the impulse response function (IRF) between the actuator/cast position and torque records collected during a calibration trial; (2) convolving the actuator/cast IRF with the position record recorded during the experimental trial and (3) subtracting the resulting predicted torque from the sampled torque. The residual torque representing the contribution from the ankle was thereafter referred to as the torque.

Dynamic Analysis

Linear trends in the data segments resulting from the position ramp paradigm were removed from each position and torque segment. Stiffness impulse response functions relating position and torque were then computed for successive 2.05 s intervals using the algorithm described in Hunter and Kearney (1983). Non-linear minimization techniques were used to determine the values of the inertial (I), viscous (B), and elastic (K)parameters which provided the best-fit between the IRFs and that of a second order, differential equation of the form:

$$T(t) = K\theta(t) + B\dot{\theta}(t) + I\ddot{\theta}(t),$$

where: t = time (s),

T = ankle torque (Nm),

θ = ankle angular displacement (rad),

I = inertial term (Kgm^2),

B = viscous term (Nms/rad),

K = elastic term (Nm/rad),

Estimate of measurement reliability

The reliability of the repeated measures from all paradigms was estimated using Pearson's coefficient of correlation (Carmines, 1979). For the purposes of this study, a variable with a coefficient (r) of greater than 0.80 was considered reliable. From the MVC and passive static paradigm measures of the dorsiflexing MVC, plantarflexing MVC, and ROM (analogous to clinically accepted measures of joint function) and two measures of joint behavior, the passive torque associated with the ramp displacement and the position-passive torque loop area (work) were examined. The variables assessed for reliability in the passive dynamic paradigm included: a predicted minimal K value (K offset), a low stiffness region (defined as the ROM for which $K \leq 30$ Nm/rad) and the slope and intercept of linear relation between K and torque.

Results

The maximum torque generated during both PF and DF MVCs are summarized in Table I. Note that the PF MVC values were consistently much larger than

Subject No.	Code	MVC Paradigm: DF MVC (Nm) Day 1	DF MVC (Nm) Day 2	PF MVC (Nm) Day 1	PF MVC (Nm) Day 2	Passive Static Paradigm: ROM (rad) Day 1	ROM (rad) Day 2	Passive TQ (Nm) Day 1	Passive TQ (Nm) Day 2	Work (J) Day 1	Work (J) Day 2
1	AS	54.5	42.7	89.1	91.5	1.28	1.37	36.7	44.5	2.81	4.20
2	BA	40.0	38.1	126.0	146.0	1.59	1.69	47.0	58.1	5.18	6.10
3	HM	25.9	22.2	28.4	34.7	1.24	1.29	23.3	28.9	3.95	4.71
4	JF	15.6	17.5	31.6	33.3	1.29	1.36	23.9	25.7	1.87	2.47
5	LB	28.8	26.6	54.6	65.4	1.42	1.27	51.7	41.7	3.02	3.32
6	LD	32.6	33.8	109.3	112.6	1.32	1.14	51.5	33.9	4.17	2.66
7	MB	30.8	31.7	50.8	48.4	1.44	1.47	38.4	40.1	4.03	5.41
8	MP	46.8	43.1	85.1	83.8	1.56	1.56	65.1	71.8	6.09	6.62
9	MR	28.0	32.3	87.5	91.8	1.52	1.26	40.6	26.4	4.45	3.01
10	MV	24.2	26.4	31.5	41.7	1.23	1.15	28.3	22.6	1.91	1.56
11	PW	27.7	29.6	56.9	63.1	1.10	1.07	22.4	17.1	2.01	1.39
12	RK	55.8	49.9	125.0	117.5	0.87	0.85	37.1	34.4	2.95	2.54
13	RM	34.4	35.9	112.9	113.9	1.23	1.23	40.9	38.6	4.40	3.40
14	SB	23.3	24.5	48.5	51.2	1.67	1.59	71.7	61.1	9.54	5.15
15	TG	31.9	32.3	56.9	55.6	1.29	1.39	38.7	42.7	3.17	3.47
Mean		32.8		74.8		1.32		40.2		3.85	
SD		10.0		34.2		0.20		14.2		1.62	
r		0.96		0.98		0.84		.87		.68	

Table 1: Group means, standard deviations (SD), and Pearson's coefficient (r) are provided for the experimental parameters for the 15 control subjects. The parameters for the maximal voluntary contraction (MVC) paradigm include the repeated measures (Day 1 and Day 2) for both dorsiflexion MVC (DF MVC, and plantarflexion MVC (PF MVC). The passive static paradigm parameters include the repeated measures of the range of motion (ROM), the passive torque and the loop area of the passive torque-position relation.

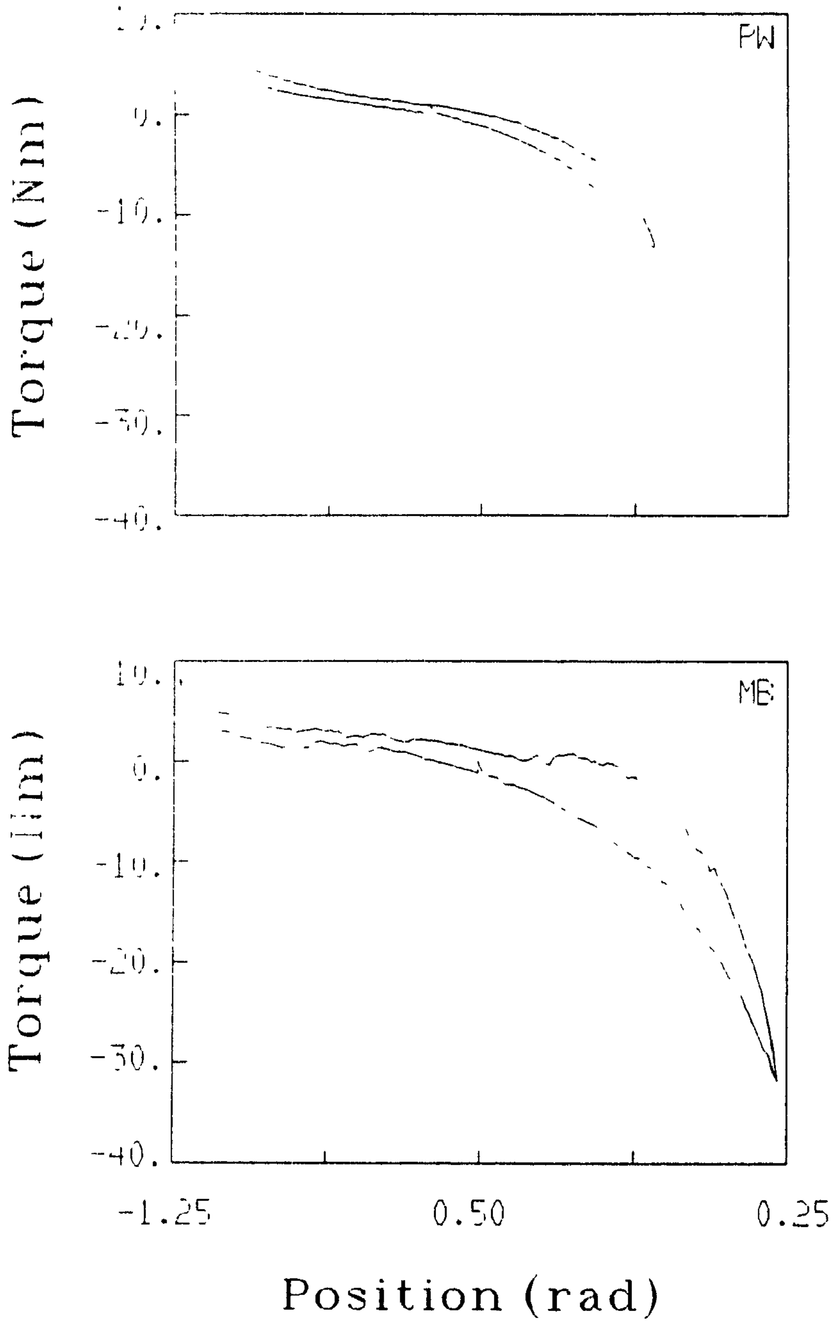

Fig. 5.1: Passive torque-position curves for two individuals, one, PW (upper panel) with a smaller than average ROM and another MB (lower panel) with a larger than average ROM.

the DF MVCs (means = 74.8 and 32.8 respectively). Furthermore, both MVCs provide a reliable measure of muscle function since their r values are high (0.96 and 0.98).

Static Paradigm

The relations between passive joint torque and position for two subjects (PW and MB) are shown in Fig. 5.1. The curves are sigmoidal in shape; passive torques were small and relatively insensitive to change at joint angles near the mid-range but increased sharply at the extremes of ankle position. Torques at extreme dorsiflexion are always greater than at extreme plantarflexion. Subject PW, shown in the upper panel, had a smaller than average ROM (1.10 rad) and a small range of passive torque (22.4 Nm). In contrast, subject MB, shown in the lower panel had a a relatively large ROM (1.47 rad) and a larger range of passive torque (40.1 Nm). Also shown in Table I are the ROM and the passive torque values as well as their reliability estimates. Both ROM (r = 0.87) and passive torque (r = 0.84) reliability coefficients were less than the MVC r values but nevertheless very high. These three measures were very consistent indices of ankle behaviour.

It can also be seen from Fig. 5.1 (both panels) that the passive torque curve followed different pathways when the ankle was rotated from plantarflexion to dorsiflexion (lower curves) then from dorsiflexion to plantarflexion (upper curves). The area between these two curves is a measure of the work required to displace the joint. The passive work values for subject PW and MB are 2.01 J and 4.03 J respectively. This measure has been used to assess joint mechanics quantitatively (Bromberg and

Grimby, 1982) and quantatively (Mortimer and Webster, 1978). As shown in Table I, work for the fifteen control subjects varied from 1.56 to 6.62 J (mean = 3.85 J, SD = 1.62 J).

Passive dynamic paradigm

The dynamic relation between torque and position can be represented nonparametrically by the compliance IRF. Separate IRFs, calculated for each repetition of the perturbation sequence accounted for more than 90% of the observed variance. A set of compliance IRF's obtained during a typical dynamic displacement (Paradigm 3) is shown in fig. 5.2. IRFs were determined over 0.1 rad intervals as the ankle was rotated from maximum dorsiflexion to maximum plantarflexion. Although the shapes of these IRFs were very similar, the gain and natural frequency varied systematically with joint position. Thus the ankle became more compliant (less stiff) as it was rotated from maximum dorsiflexion to mid-position and less compliant (more stiff) as it was rotated from mid-position to maximum plantarflexion. As demonstrated in Fig. 5.2, the compliance IRF gain was highest at mid-position.

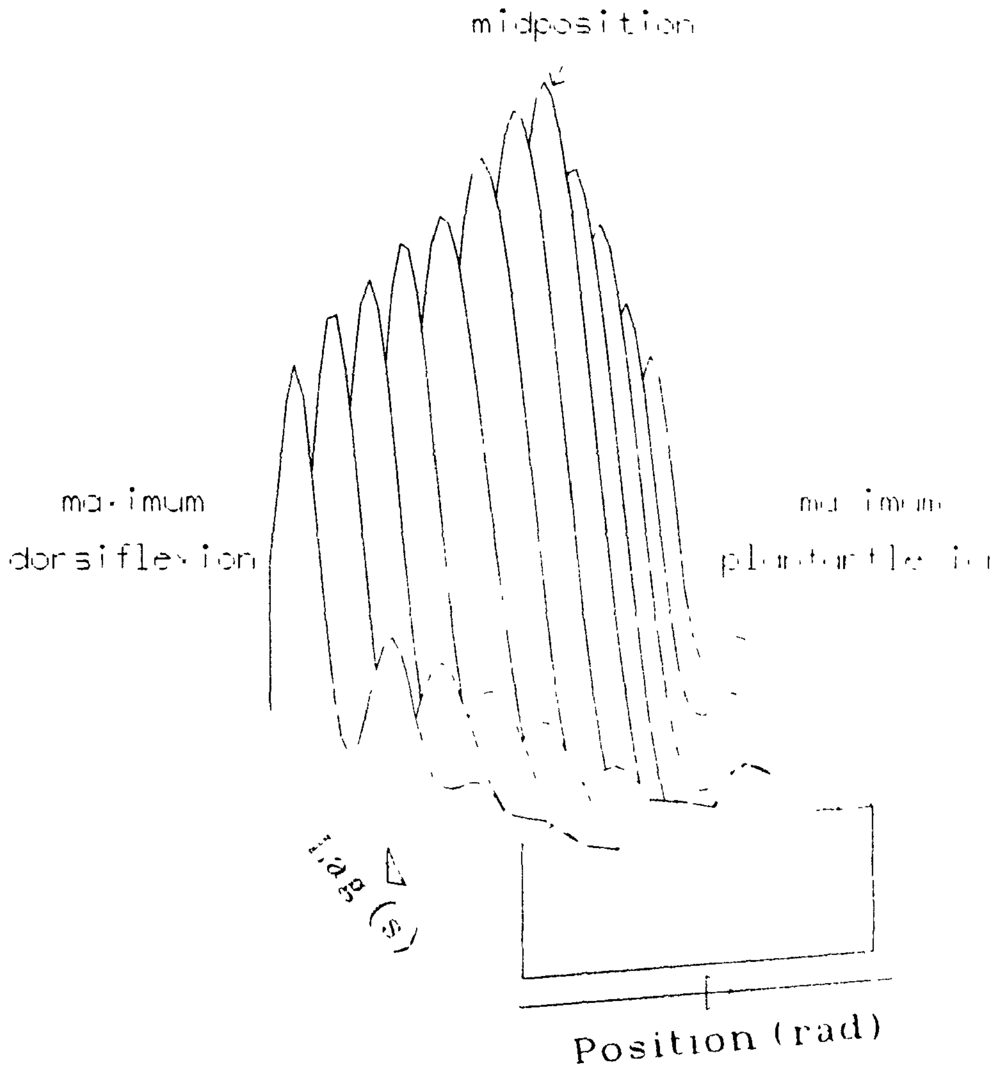

Fig 5.2: Plot of an ensemble of 12 ankle joint compliance IRF curves (subject MR) as the ankle joint was rotated from a position of maximum dorsiflexion through the mid-position to a position of maximum plantarflexion.

No.	Code	INERTIA ($kg\text{-}m^2 * 10^{-3}$) mean Day 1	mean Day 2	SD Day 1	SD Day 2	DAMPING FACTOR mean Day 1	mean Day 2	SD Day 1	SD Day 2	STIFFNESS RANGE (Nm/rad) Day 1	Day 2
1	AS	9.4	8.6	1.9	1.8	0.40	0.42	0.10	0.10	287	291
2	BA	8.9	9.9	2.0	1.9	0.35	0.33	0.09	0.10	204	301
3	HM	4.8	5.0	0.8	0.7	0.32	0.33	0.06	0.06	97	111
4	JF	3.5	3.5	0.4	0.3	0.34	0.38	0.09	0.10	112	97
5	LB	6.4	4.9	2.0	0.5	0.36	0.38	0.07	0.08	267	238
6	LD	5.4	5.2	0.6	0.4	0.44	0.51	0.12	0.14	282	176
7	MB	5.8	5.7	0.7	0.5	0.35	0.37	0.10	0.12	225	196
8	MP	9.3	8.6	1.6	1.2	0.33	0.37	0.08	0.10	410	112
9	MR	5.5	4.9	0.7	0.3	0.31	0.35	0.08	0.09	128	102
10	MV	5.4	5.3	0.6	0.8	0.34	0.34	0.08	0.08	181	103
11	RW	4.1	4.5	0.5	0.4	0.61	0.36	0.11	0.09	84	61
12	RK	8.8	8.5	0.5	0.6	0.25	0.28	0.05	0.05	184	191
13	RM	6.3	6.2	0.9	0.9	0.29	0.30	0.06	0.07	293	205
14	SB	5.9	5.5	0.9	0.9	0.29	0.31	0.07	0.05	265	359
15	TG	12.3	14.6	4.0	6.5	0.31	0.31	0.07	0.08	203	208

Table IIa: Values are from the passive dynamic paradigm. The repeat (Day 1 and Day 2) control means and standard deviations (SD) for the moment of inertia and damping factor, and stiffness range are given.

	K OFFSET (Nm/rad)		LOW K REGION (rad)		K vs TQ SLOPE (Nm/rad/rad)		K vs TQ INTERCEPT (Nm/rad)	
	Day 1	Day 2	Day 1	Day 2	Day 1	Day 2	Day 1	Day 2
	30	31	0.66	0.79	7.5	8.5	23	15
	18	20	0.95	0.98	6.7	7.1	13	14
	16	12	0.84	0.82	7.3	7.3	18	12
	14	12	0.83	0.85	7.1	7.4	13	11
	21	17	0.81	0.82	15.0	10.3	10	6
	20	18	0.75	0.71	8.3	9.5	16	10
	16	14	0.77	0.73	6.7	6.7	15	14
	31	25	0.88	0.84	8.9	9.3	24	6
	14	16	0.84	0.82	5.8	6.2	20	15
	11	14	0.80	0.82	10.7	8.3	7	7
	21	18	0.75	0.85	8.9	7.5	21	15
	54	46	0.51	0.45	8.3	8.3	53	46
	16	16	0.66	0.66	13.7	8.6	6	15
	23	21	0.78	0.82	9.2	9.8	25	22
	27	33	0.83	0.93	6.9	6.3	19	25
mean	21.5		0.78		8.4		17	
SD	9.8		0.11		1.8		10	
r	0.94		0.89		0.70		0.83	

Table IIb: Measures for the passive dynamic paradigm. The K offset, a position range where K $\leq$ 30 Nm/rad (low K region), the slope of the K absolute torque relation (K vs TQ SLOPE), the intercept for the K absolute torque relation (K vs TQ INTERCEPT) are given for the 15 controls. Means, standard deviations (SD) and Pearson's coefficient (r) are given.

A second order model having inertial (I), viscous (B), and elastic (K) terms was fitted to each compliance IRF. This model accounted for over 95% of the variance of the original IRF. The means and SD of I and the damping factor (ζ) and the range of K are presented in Table IIa. The damping parameter ζ which is related to the I, B, and K (Weiss et al., 1988) by the relation

$$\zeta = \frac{B}{2\sqrt{I\ K}}$$

was also determined.

These variables are also plotted against joint position and are shown in Fig. 5.3. It is apparent on examination of both Fig. 5.3 and Table IIa that the sensitivity of K to changes in joint position is greater than that of the other parameters. This parameter was therefore selected for further analysis.

The K-position curves obtained on days 1 and 2 for subject BA are shown Fig. 5.4 (upper panel). These curves are cup shaped with a flattened central region. It can be seen that K is sensitive to changes in joint position at both extremes of the available range and relatively insensitive at mid-range as has been previously reported (Weiss et al., 1986a). The BA1 (obtained on day 1) curve shows as much as a 15 fold increase of K over the minimal value. While the shape of the curves are quite similar, there does appear to be a position offset of approximately 0.05 rad.

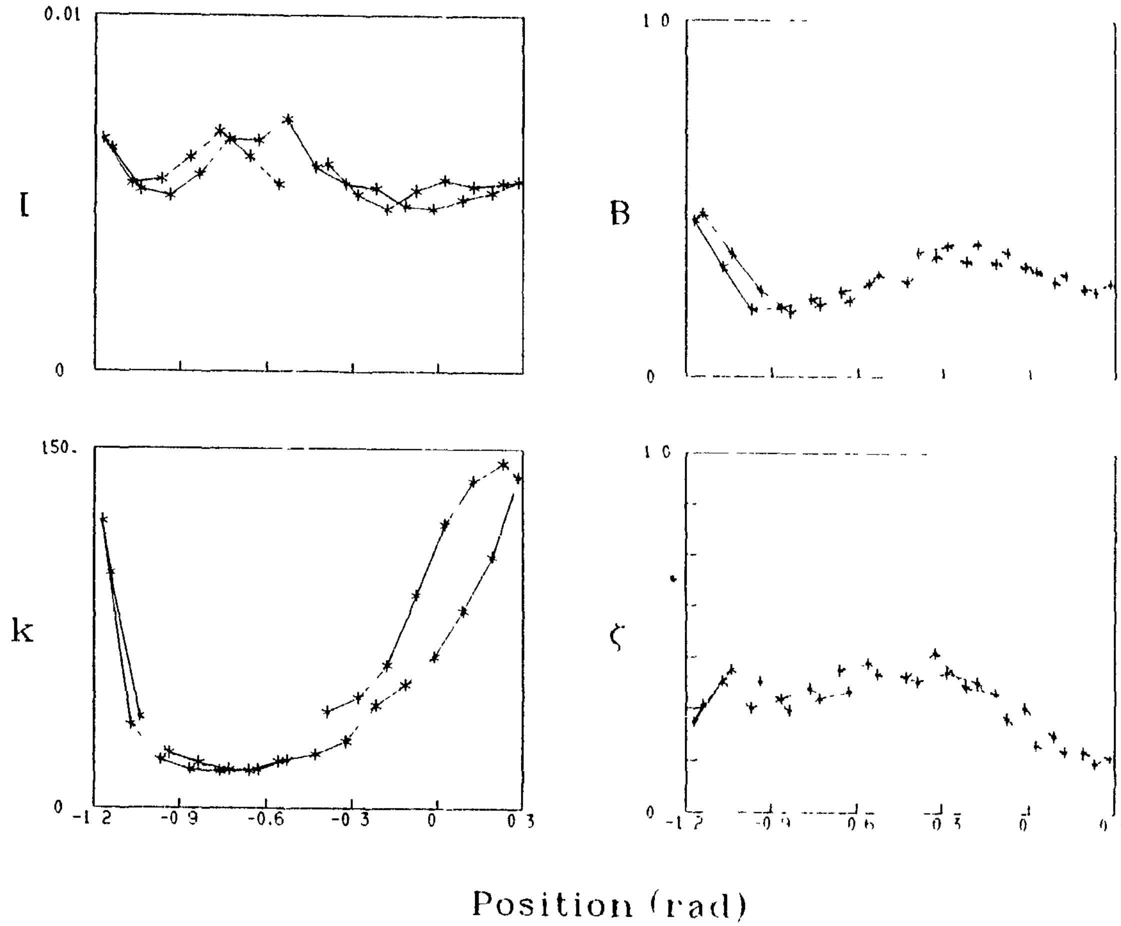

Fig. 5.3: The parameters of the second orde: model; I (kgm^2), B (Nms/rad), and K (Nm/raC) and a damping parameter ζ plotted as functions of ankle position for one subject (MR).

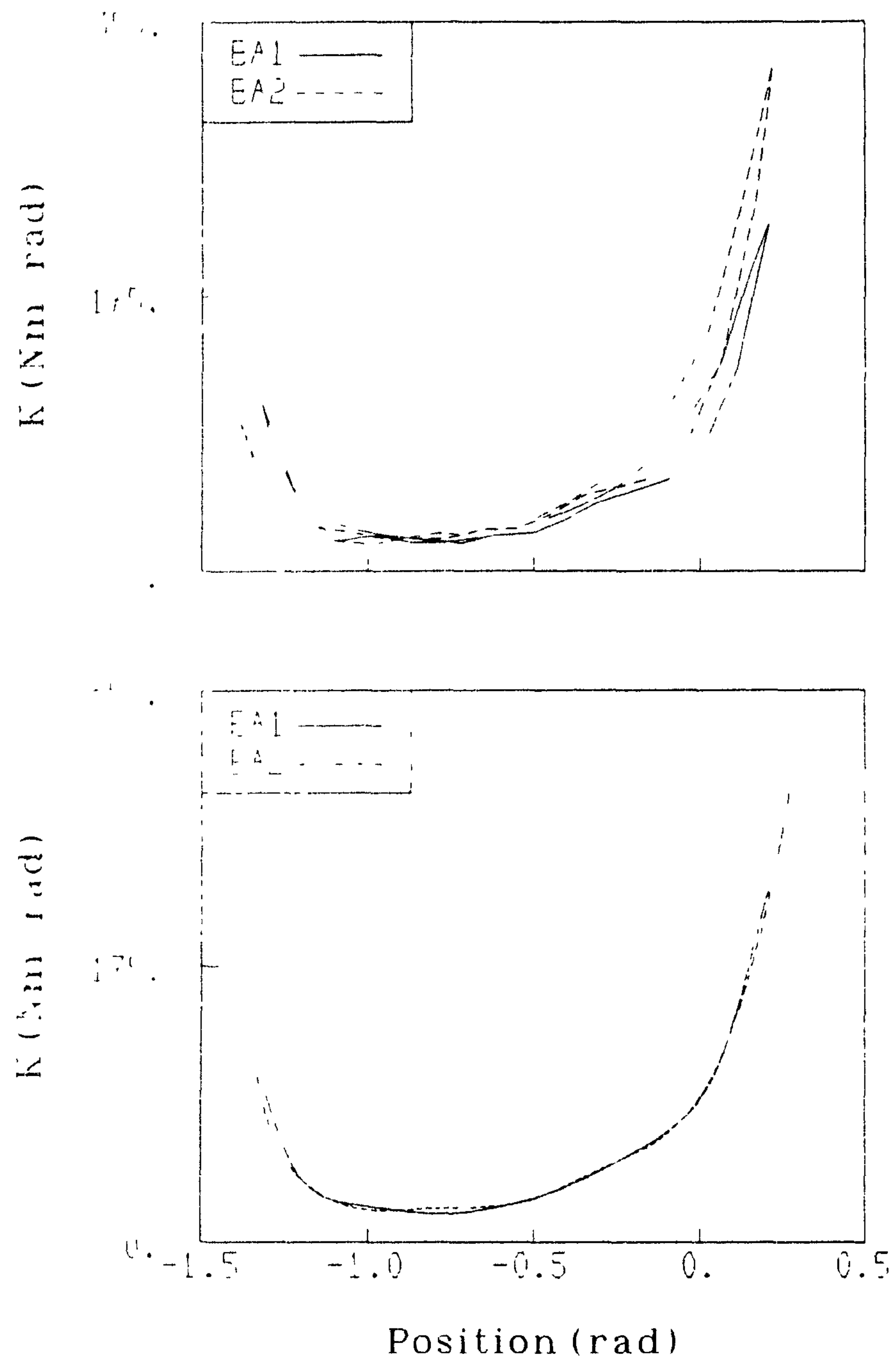

Fig. 5.4: Upper panel: The repeated measures (BA1,BA2) of the elastic parameter (K) plotted as a function of ankle position for one individual (BA).

Lower panel: The predicted Chebychev polynomial fitted to the original data and corrected for position offset for the same set of repeated measures (BA1,BA2).

Although the joint was perturbed continuously as it was rotated through the ROM, each estimate of K was obtained over a small range of joint angles (approximately 0.1 rad). The calculated mid-range and hence these small ranges were not the same on the two test days since their locations depended upon the values of maximum plantarflexion and dorsiflexion determined each day. This presented a problem for the reliability study since K was sensitive to variations in joint positions, particularly at the extremes of the ROM. An alternate representation of the K-position relation was required to enable a comparison. Chebyshev 10th order polynomials were therefore fitted to the ankle K-position curves which accounted for, in most instances, over 98% of the variance in the original curves. The position offset between these two (Day 1 and Day 2) predicted K-position was then determined and used to correct difference in the initial mid-position. One curve was shifted with respect to the other by the determined offset which permitted us to assess the reliability of the K-position data without altering its basic characteristics.

Fig. 5.4 (lower panel) shows the fitted and shifted curves corresponding to the original data presented in Fig. 5.4 (upper panel). Subsequent analysis of K was carried out using the corrected curves. Visual inspection of the two polynomial curves demonstrates the remarkable similarity and the repeatability of the two trials.

Two variables were used to assess the reliability of the stiffness data, the minimal K associated with each K-position curve (K offset) and the ROM over which K was less than 30 Nm/rad (low stiffness region). The values of the K offsets, listed in Table IIb, varied from 11 Nm/rad to 54 Nm/rad and had a reliability coefficient equal to 0.94. The low stiffness region, used to delineate the flattened, plateau-like portion of the curve is shown in Fig. 5.5 as the region located between the two arrows. This region ranged from 0.45 rad to 0.98 rad. The low stiffness region results listed in Table IIb also demonstrate it is consistent within the individual ($r = 0.89$).

In order to characterize further the properties of the elastic stiffness, the linear relation between K and the absolute value of the passive torque was investigated using linear regression techniques. The values for the slopes and intercepts are shown in Table IIb. A linear relation between K and the absolute value of torque described more than 90% of the variance. This is illustrated in Fig. 5.6 by the straight line approximating the absolute torque data points for one individual.

The reliability of the slope of the relation between K and the absolute torque is lower ($r = 0.70$) although a paired T-test did not demonstrate a significant difference between the repeated measures ($p > .90$). The two sets of measures could therefore not be considered significantly different. In contrast the reliability of the estimated K value when torque is equal to zero (intercept value) is quite acceptable at $r = 0.83$).

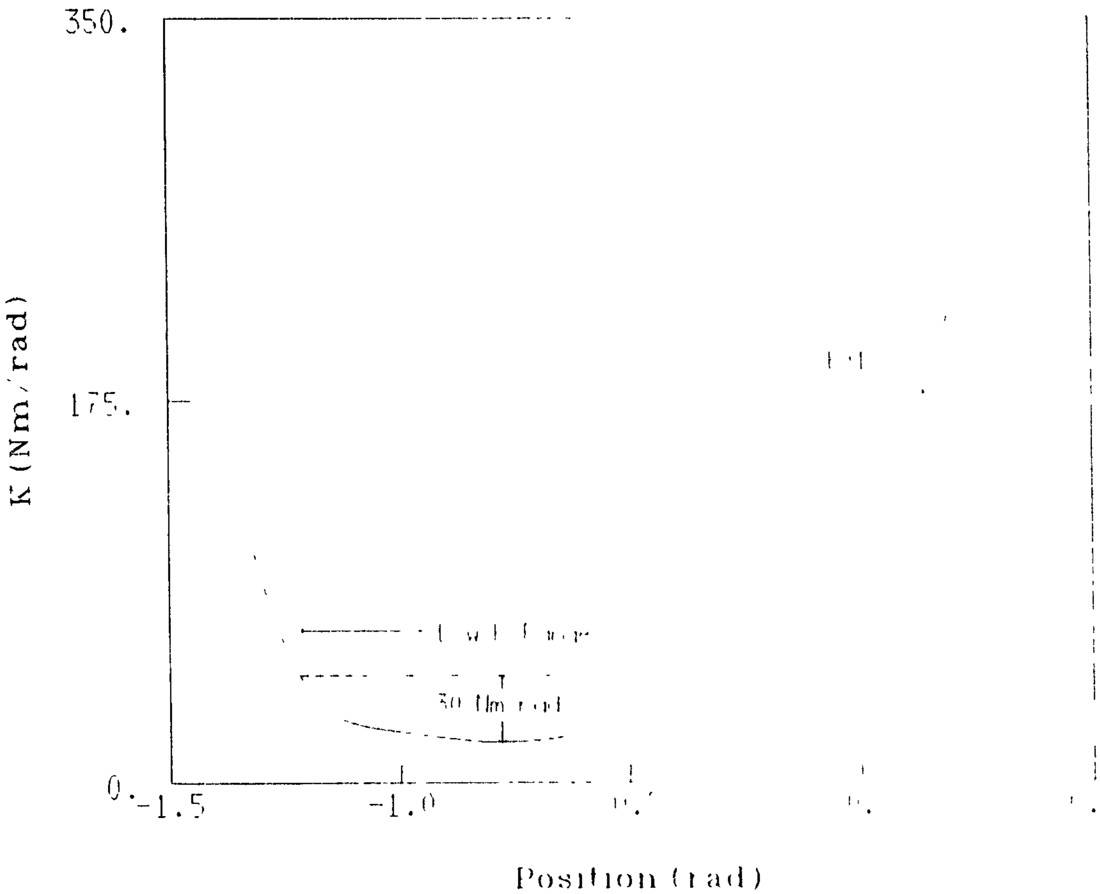

Fig. 5.5: A Chebychev polynomial with the predicted K plotted as a function of ankle position and a low stiffness region where K < 30 Nm/rad above the minimal predicted K (region indicated by the dashed line). (Data are from subject BA1).

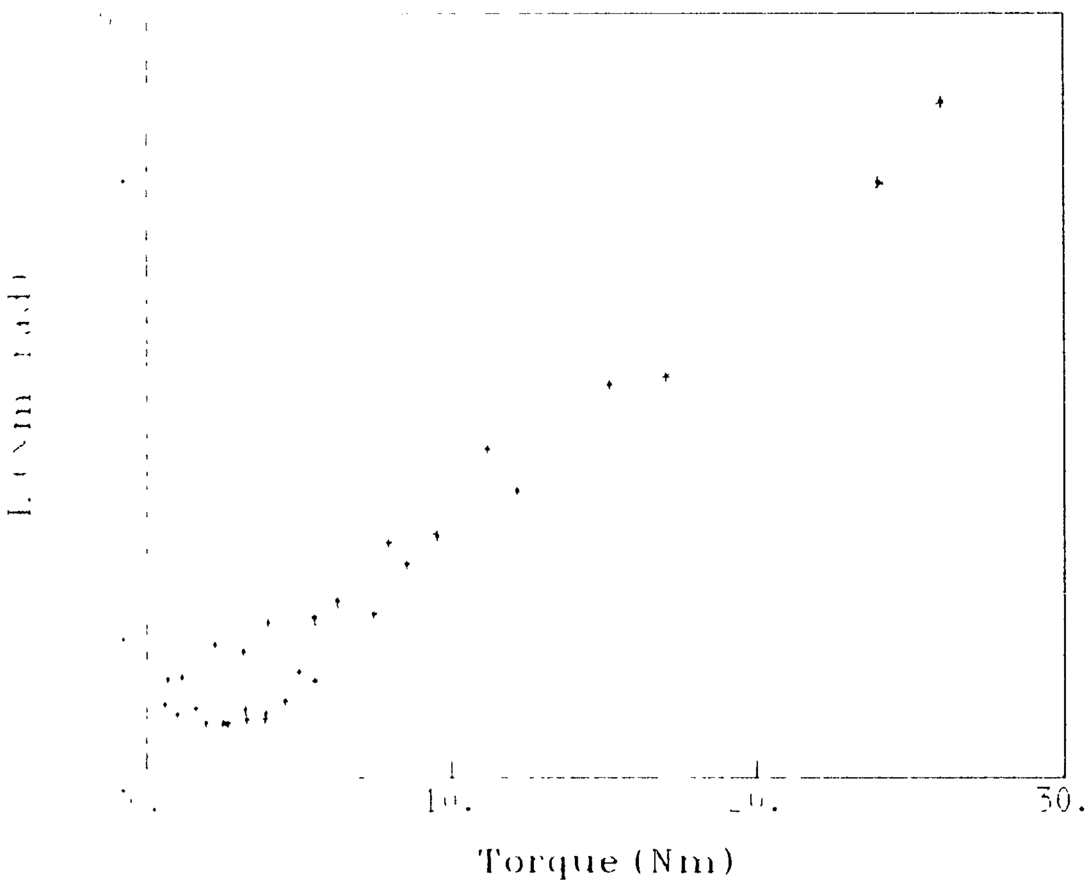

Fig. 5.6: The elastic parameter (K) plotted as a function of the absolute torque for one individual (BA) and the regression line (K = 6.7 * Tq + 13) fitted to the relation between K and the absolute value of torque.

Case study

The system identification approach to the evaluation of joint function was used to assess an individual recovering from orthopedic dysfunction to gain some insight as to the feasibility of these variables for examining the effects of pathology.

Methods

The case study involved a 26 year old male individual who had sustained a left undisplaced fracture that was treated by six week immobilization in a plaster cast. There was no history of orthopedic nor neurological dysfunction prior to the recent fracture, no use of medication for the condition, and the subject was not following any regular rehabilitation program.

The patient was initially assessed 8 days post cast removal and re-evaluated at approximately 2 week intervals with the same paradigms as in the reliability study. The time course of this investigation was determined from the expected course of post immobilization recovery of function.

Since immobilization edema is a common occurrence among ankle fractures (Sondenaa et al., 1986), an attempt was made to control it by supplying the patient with elastic support stockings. Despite this, foot volume did change over the 19 week period requiring recasting of the foot. Three such Fiberglass boots were made (at 8, 16 and 34 days post immobilization). Foam and rubber inserts were not used to correct for decreased foot volume since these might have introduced unwanted viscous and elastic forces into our dynamic measurements. The same analysis procedures were used as for the reliability study.

Case Study Results

The variance accounted for by the compliance IRFs was consistently greater than 92%. The parametric representation of joint behaviour in the form of the second order equation with inertial (I), Viscous (B), and elastic (K) parameters accounted for over 80% of the variance in the original IRF. These values are similar to those obtained from the control subject

As shown in Table III, the plantarflexing MVC, dorsiflexing MVC, joint ROM and passive torque values increased over the first five weeks (up to day 34) of the post-immobilization recovery period and then tended to stabilize. For example, there was an approximate 2 fold increase in passive torque over the first 5 weeks post-immobilization then an increase of about 30% over the last 16 weeks post-immobilization.

When the case study measures of both passive static and passive dynamic paradigms are compared to the control group means and SD, it is evident that there were some large differences. Fig. 5.7 shows the predicted polynomials corrected for position offset for three experimental evaluations (days 8, 24, and 135). The shapes of the curves are similar to the control curves (Fig 5.4) with the higher K values in extreme dorsiflexion as compared to extreme plantarflexion. The K values were also sensitive to extremes of position but were in general, much larger than the K values for our controls.

The case study K offsets (minimal K values) were greater than the control mean value (mean = 21.5, SD = 9.8 Nm/rad) plus two standard deviations for every session. When the low K region was investigated it was apparent that this region changed, increasing at first then stabilizing by about day 34. In contrast with the control values (mean = 0.78 rad, SD = 0.11 rad), the case study had a low K region value of 0.34 on the initial assessment which is considerably smaller (at least four standard deviations) than the control values. However, the most remarkable feature of the case study data was the large difference in the pathways of the three curves shown in Fig. 5.7 are compared to the two nearly superimposed repeated measures shown in the lower panel of Fig 5.4 While conclusions based on the results of a case study must be cautious, we were encouraged to note that our measures appear to reflect the recovery from dysfunction.

	MVC Paradigm		Passive Static Paradigm			Passive Dynamic Paradigm				
Variable Unit	PF MVC (Nm)	DF MVC (Nm)	ROM (rad)	PTQ (Nm)	W (J)	KO	Low K R	K vs TQ m	K vs TQ b	%VAF
Day 8	73.4	39.0	0.70	43.4	2.54	96	0.34	14.6	84	63
Day 16	79.5	40.0	0.87	49.9	3.94	65	0.29	19.3	77	92
Day 23	85.0	48.6	0.86	47.0	3.52	58	0.36	22.3	40	94
Day 34	116.1	51.0	1.17	83.6	8.05	62	0.43	12.8	87	91
Day 73	100.6	48.5	1.21	82.8	9.90	88	0.43	12.0	113	84
Day 135	92.0	51.2	1.33	110.1	14.31	92	0.45	8.9	152	66

Table III: Case study measures of joint function for the six experimental sessions. The MVC paradigm includes plantarflexion MVC (PF MVC) and dorsiflexion MVC (DF MVC). The passive static paradigm includes values for range of motion (ROM), passive torque (PTQ), and work (W). The passive dynamic paradigm includes the K offset (KO), low K range (Low K R), the slope of the K absolute torque relation (m), the intercept of the K-absolute torque relation (b) and the variance accounted for by the linear fit (%VAF).

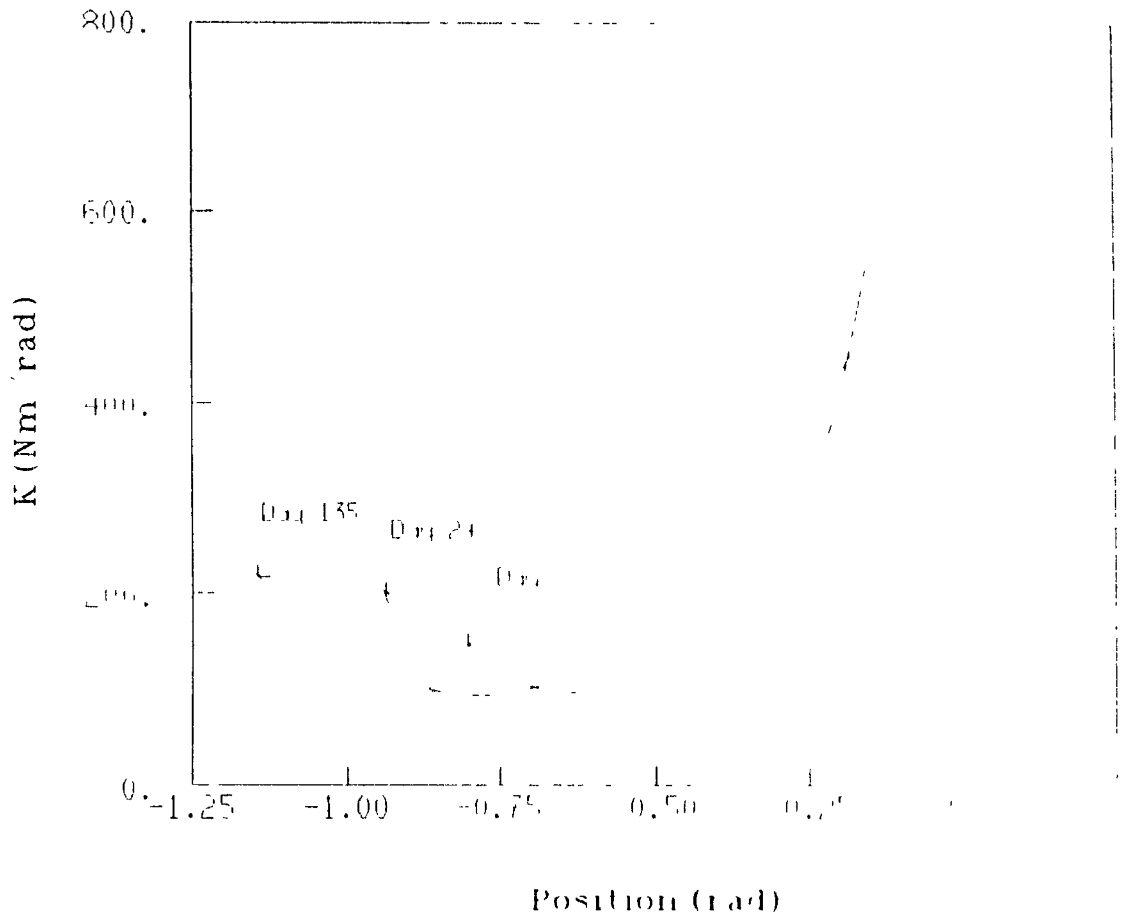

Fig. 5.7: Predicted polynomials for the K-position relation for the case study on three different days.

DISCUSSION

Valid, reliable, and sensitive outcome measures are essential for the development of good clinical assessment tools (Feinstein, 1983). Such tools would help the physician determine the effectiveness of the pharmaceutical or surgical intervention and help the therapist assess the benefits of different treatment modalities. In most clinical settings, however, disorders of motor performance are diagnosed and evaluated by methods which do not satisfy these criteria. For example, hypertonicity is assessed by observation of a patient's response to tendon taps and the sensation of a limb as it is passively moved through its range of motion. In the case of trauma, x-rays. arthroscopy, and positive contrast arthrograms are used to identify mechanical obstructions. Such methods generally provide information regarding static anatomical and physiological function rather than a quantitative and dynamic assessment of the problem.

The objective of the present study was to assess the ability of techniques based on system identification to evaluate abnormal joint function and the recovery from dysfunction. This is a natural extension of our previous work which focused on studies of the neuromuscular system controlling the intact human ankle. In this approach, descriptions of system behavior are determined from the analysis of the relation between inputs and outputs. These techniques have been used successfully in engineering but only recently have they been applied to biological problems. The suitability of this approach in terms of its validity, reliability, and sensitivity is discussed.

Validity

We have established experimental and analytic methods for determining quasi-linear models of joint mechanics and stretch reflex dynamics under stationary or slowly varying conditions. Joint mechanics, characterized in terms of dynamic stiffness (the dynamic relation between position and torque) were found to be well described by the parameters of a second order model having inertial (I), viscous (B), and elastic (K) terms (e.g. Hunter and Kearney, 1982). These models represent only a part of joint behavior since they change with the experimental conditions as a result of underlying nonlinearities. Nevertheless, they provide an excellent description of behavior for a given set of conditions. It is clear from the extremely good fits obtained by our models under a variety of different operating conditions that this method results in a representation of normal joint behavior that is more accurate than those obtained by purely static studies (Weigner and Watts, 1986).

Previously we have used these quantitative methods to document changes in joint stiffness with level of motor activity (Hunter and Kearney, 1982), perturbation amplitude (Kearney and Hunter, 1982), and fatigue (Hunter and Kearney, 1983). We have also characterized the effects of ankle position on joint stiffness (Weiss et al., 1986a). The present study represents our first attempt at developing paradigms for the application of system identification techniques to the study of the changes in joint mechanics resulting from neuromuscular dysfunction and trauma. Although a formal test of validity was not performed we were encouraged to note that the same algorithms used to analyze our control data worked well on the results of the patient's dynamic perturbation paradigm. The variance accounted for by the linear dynamic relation between joint torque and position (the compliance impulse response functions) was consistently greater than 90%. Moreover, the three parameter second order model accounted for more than 85% of the variance in these functions. Such values are comparable to those obtained from our control subjects.

Test-Retest Reliability

The reliability of clinical measures have been investigated in many ways (Stratford et al., 1984). Factors which appear to affect the outcome of such tests include the time interval over which the test-retest data are sampled (Gajdosik and Bohnon, 1987), whether the data are subjective or objective (Peat and Campbell, 1979), and the index used to evaluate reliability (Stratford et al., 1984).

It is rarely possible to design a test-retest study which conforms to ideal conditions since the selection of experimental variables and time course are usually constrained by factors related to the research question. However we felt it important to address the most major concerns in the design of this reliability study.

In some cases, previous investigations of clinical measures have focused their attention on the reliability of the instrumentation as opposed to the reliability of the variable itself. For example, the reliability of torque measurements on a Cybex dynamometer is often quoted as very high with r values greater than 0.9 (Armstrong et al., 1983; Mawdsley and Knapic, 1982). This result is misleading, however, since the test-retest variables were sampled with the use of calibration weights. Although such measures have the same units as those obtained when the experimental subject generates a voluntary contraction, the sources of measurement error are quite distinct. In contrast our results were obtained while the subjects performed experimental tasks which corresponded to clinically relevant assessment procedures.

Another approach has been to assess reliability on the basis of paired t-tests (Mawdsley and Knapic, 1982) where the absence of significant difference between the test-retest measures is taken to be an indication of reliability. This can also be misleading since repeated measures that have relatively low r values (e.g. the test-retest work values for this study, r = 0.68) may show no significant difference when a paired t-test is used to compare the two trials. For this reason we recommend that only measures which had high coefficients of variability (PF and DF MVC, ROM, passive torque, K offset, low K region, and the intercept of the K-absolute torque relation) be used to assess clinical change.

Although it is recommended that the test-retest interval be kept small, the interval used in this study was intentionally kept large (mean interval = 16 days) in order to abide by the time course over which such measures are normally obtained in a clinical setting. This may be less than ideal statistically but important clinically for assessing the stability of the measure.

Finally, it is important to note that some of our measures in particular, the ROM determination, had a subjective component which was beyond our experimental control. The reliability of this measure (r = 0.87), similar to that found by other investigators (Boone et al., 1978; Miller, 1985) is due, in part, to the method by which it was obtained (i.e. the subject's report of discomfort or pain. A more consistent method of determining true available ROM using objective criteria is needed reduce the variability in this parameter and in those sensitive to joint position, for example K and its derived measures. In addition, the Pearson's product moment coefficient, r, may not have sufficient power to demonstrate reliability if the intrasubject and intersubject variability are of the same magnitude (i.e. the slope of the K-torque relation). A larger sample size or a reduced intrasubject variability would alleviate this problem.

Sensitivity

The case study values for the low K region and K offset were considerably different than the mean values obtained for our controls. For example, the K offset range (58-92 Nm/rad) was more than 2 SD greater than the control mean (mean = 21.5, SD = 9.8 Nm/rad). The low K range (0.29 - 0.45 rad) was at least 3 SD less than the control mean (mean = 0.78, SD = 0.11 rads). Although these differences are important and the results encouraging, caution must be exercised with respect to conclusions obtained from a single case study.

Feasibility of the System Identification Approach

In theory, the use of these techniques for clinical assessment is attractive because this approach has the following valuable attributes:

1) The characterization of the mechanics and reflex dynamics can be done very rapidly; the mechanics can be determined from as little as 2.5 s of experimental data while the reflex dynamics require 10 s of data.

2) It permits the use of random inputs which prevent subjects from anticipating the stimulus waveform and generating "preprogrammed" or "voluntary" responses. This eliminates the confounding effects of voluntary activity making it possible to focus on more automatic mechanisms (e.g. stretch reflexes, muscle contractile mechanics).

3) It can correct for artifacts arising from the properties of the stimulus waveform and experimental apparatus. The responses observed under any experimental condition will depend upon the underlying physiological mechanisms, the characteristics of the stimulus waveform, and the experimental apparatus. The quantitative input/output analysis carried out in system identification generates models of the overall system which are independent of the stimulus waveform. Furthermore, the effects of the apparatus can be determined independently and then removed from the overall response to reveal the physiological system of interest.

4) The quantitative models resulting from system identification studies

have predictive capabilities; they can be used to estimate the response to inputs other than those used in the original experiments. Consequently, they can be used as components of more global models.

5) The systems descriptions of neuromuscular behavior generated in our experiments are completely objective; the experimental protocol is predetermined and under computer control, data qualification is done using rigorous quantitative tests, and the analysis procedures are standardized.

6) The analytic models resulting from our work can be remarkably parsimonious. Thus, for example, for a given set of conditions, dynamic ankle stiffness can be described by just three parameters.

The outcome of this study has demonstrated that the system identification approach can indeed be used to assess the changes in joint mechanics resulting from trauma of orthopedic origin. In particular, the results of the case study have enabled us to establish that the use of this approach is feasible for the study of joint pathology. The patient tolerated all experimental procedures and was able to perform all required tasks. The subject reported only minor discomfort when the foot was placed near the extremes of the range of motion and did not appear to be disturbed by the equipment. Moreover, the time required to execute all paradigms including attachment of the boot to the actuator and fixation of the body to the experimental table was well within the time often allotted for clinical tests.

Conclusion

In summary, the system identification approach provided measures of joint mechanics which are reliable and well tolerated. The case study results are encouraging and, although preliminary, demonstrate potential as valid clinical measures.

Chapter VI

SUMMARY AND RECOMMENDATIONS

Summary

The objective of this thesis was to investigate the feasibility of using measures of dynamic ankle mechanics (stiffness) as an objective and quantitative tool in the assessment of joint function. This entailed the development of a low inertia, rigid fixation technique for the attachment of the foot to the assessment device, the testing of 15 control subjects using paradigms designed to identify static and dynamic ankle mechanics, and serial examinations of an individual who had sustained a unilateral ankle fracture during the post immobilization period.

Major Findings

This work has made four significant contributions to the development of a reliable and clinically feasible assessment tool. These include:

1) The development of an improved method of limb fixation to the experimental apparatus. The use of fibreglass casting bandage material provides an extremely rigid fixation that can be simply and quickly constructed. Its low inertia resulted in a significant improvement in the resolution of the system dynamics.
2) The collection of normative data for static and dynamic ankle mechanics. The numbers of subjects examined in previous studies has been limited. As a result of this work, clinical data can be compared to age-matched controls.

3) The reliability of measures of joint function has been established. This represents an important first step in the development of a clinical assessment tool.

4) The clinical feasibility of both experimental protocol and analysis procedures has been established. The procedures were fast, required minimal cooperation and were well tolerated by all subjects.

Significance of study

There is ample evidence in the clinical literature suggesting that currently available, clinically accepted measures of ankle function such as goniometry, the manual muscle test and the assessment of muscle tone by passive movement lack objectivity, reliability, validity, and dimension. Furthermore they do not reflect the technical advancements made in the past several years in biomedical engineering. Clinically the development of an improved, objective, quantitative and reliable assessment tool is needed in:

1) the development of safer and more efficient work surfaces and supports;

2) the assessment of degree of joint dysfunction (to facilitate clinical management), the monitoring of disease progression or regression and the detection of dysfunction or disease in its initial stage where early intervention has a tremendous impact on outcome results;

3) the validation of treatment modalities and the development of new ones;

4) the appraisal of residual ability for adjudicating compensation

claims;

5) the design of limb prostheses and joint implants that simulate more normal joint behaviour;

6) the development of adaptive equipment to lessen disability or improve performance.

Recommendations

The present study had certain limitations in methodology, equipment, and scope. The purpose of the following section is to expand upon these and suggest improvements in each areas.

Experimental methodology

The results of this study are limited by the subjective nature of the ROM assessment, the method used to record absolute ankle position, and the numbers and types of subjects examined. First, it is necessary to develop an objective test of the true passive ROM. Although the ROM measures were shown to be reliable it is not known whether these represented the actual extremes of dorsiflexor and plantarflexor range. It may be that the subject's reports of ROM limits (based on pain and/or discomfort) are reliable but, nevertheless, inaccurate. It is recommended that the use of an objective method, possibly using the slope of the ankle position/torque curve, be explored.

Second, better methods of recording lower limb and absolute ankle position are required. Since the goniometric measurement of the absolute ankle angle was difficult to achieve (due to the location of the foot in the apparatus) changes in the subjects' initial ankle position could not be reliably recorded. Moreover, minor changes in body fixation (e.g. the extent of hip abduction) may have resulted in alterations of ankle position and foot fixation comfort. This in turn may have reduced the reliability of the subsequent results.

Third, it is possible that the manipulation of the joint during the experimental preparation (checking boot fit, subject positioning and ROM determination) may have had a conditioning effect on the soft tissues that could erroneously inflate or deflate our ROM values on repeated trials (Moller et al., 1985).

Finally further studies are needed to increase the size of the data base. Although differences due to limb dominance were not anticipated the possible effect cannot be totally dismissed. Subjects representing other age groups and activity or occupational levels should also be examined.

Experimental apparatus

The fibreglass casts used in this study were of sufficient stiffness and strength. Other study designs, particularly where parameters such as different population age, or occupation might result in the generation of much larger torques, may necessitate a stiffer cast. The stiffness of the casts used in limb fixation could be increased easily by incorporating ridges of fibreglass material or aluminum implants along the sole of the foot. It is also recommended that other types of the adhesive compound used to attach the aluminum sleeves be tested for the strength of bond.

This feasibility study is the first of many steps in the development of an assessment tool. In the future the ankle actuator and table will have to be modified before a similar system is used in a clinical setting. It is apparent that the experimental apparatus should allow for examination of either ankle and should be relatively mobile. The design should include an improvement in accessibility, i.e. a lower table and an improved method for subject stabilization which would facilitate duplication of test position. An ankle actuator with a higher performance level (adequate power to 100 Hz) would improve the estimate of the second order model and resolution of the I, B, and K parameters.

Scope of study

The results from the case study demonstrated the potential of the assessment procedure but one should take care not to draw premature conclusions from these results. Indeed many more subjects with orthopedic joint pathology as well as other joint conditions (e.g. rheumatoid arthritis, joint implants) should be investigated following the prescribed paradigm before any decisions are made regarding the parameters that best characterize joint function/dysfunction and those joint conditions most suited to the assessment technique.

APPENDIX I CLINICAL EVALUATION FORM

ASSESSMENT OF DYNAMIC JOINT MECHANICS IN AN ORTHOPEDIC PATIENT POPULATION
CLINICAL ASSESSMENT

NAME ______ DATE: ______
ADDRESS ______ REFERRING PHYSICIAN ______
PHONE ______ TREATING THERAPIST ______
SESSION ______ STATUS ______
BIRTHDATE ______ HEIGHT ______ WEIGHT ______ SEX ______
Clinical Diagnosis ______
Date of problem onset ______
Method of fixation ______ Date of cast removal ______

1. Level of activity ______

   walking tolerance ______

   weight bearing status (aids) ______

2. # of formal Rx sessions since last experiment/cast removal

   ______

3. Present level of pain or discomfort on walking

   ______

4. Medication

   ______

5. Any unusual symptoms (swelling, excess pain)

   ______

		R	L
6. Ankle ROM	-active dorsi		
	-active plantar		
	-passive dorsi		
	-passive plantar		
7. Manual muscle test grade	dorsiflexors		
	plantarflexors		
8. Calf circumference (largest)			
9. Leg length (Med Tib Plat to Med Mall)			

Comments ______

APPENDIX II

NEXUS MODULES FOR DATA ANALYSIS

Module Aninit.	To initiate analysis parameters.
Module Recall.	To recall data and initiate analysis according to paradigm.
Module Loopcode.	To calculate work value for Passive Static Paradigm.
Molule Sweepirfb.	To calculate IRFs during Passive Dynamic Paradigm.
Module Sweepfitb.	To estimate the values of I, B and K for Passive Dynamic Paradigm.
Module Lag.	To fit a polynomial to Passive Dynamic Paradigm data prior to determing position offset.
Module Compare.	To determine the position offset of the K position curve
Module LKrange.	To determine the position range where K < 30 Nm/rad.

```
# RM Aninit   INITIALIZE ANALYSIS PARAMETERS


ANSWER = 'Y'
INPUT 'INITIALIZE ANALYSIS PARAMETERS ["Y"]? ',ANSWER
IF ANSWER <> 'Y' THEN GOTO [START]
INPUT 'FILE NAME, IN QUOTES? ',FNAME
INPUT 'FILE STORAGE NAME, IN QUOTES? [FILE.DAN]', DANFILE
#
#RECALL SCALE FACTORS, ZERO LEVELS (OFFSETS), AND
#CALIBRATION IRF
#
CASE =1
INPUT 'CASE NUMBER FOR SCALE FACTORS ['_STR(CASE)_']?',CASE
SCL=RCL(;FNAME,CASE,,1,!)
CASE=CASE+1
INPUT 'CASE NUMBER FOR ZERO LEVELS ['_STR(CASE)_']?',CASE
ZER=RCL(;FNAME,CASE,,1,!)

return

CASE=CASE+1
INPUT 'CASE NUMBER FOR CALIBRATION IRF ['_STR(CASE)_']?',CASE
CALIRF=RCL(;FNAME,CASE,,1,!)
PLGKS(CALIRF;'N','Y','N','Y',?)
CASE=CASE+1

# NEXT YOU SHOULD RUN RECALL
```

```
# RM
# RECALL V01-01 April 27

# Recall data from FILELB (fname) file
#
# INPUTS: CASE, FNAME, SCL, ZER, CALIRF


.PURGE
SET QUIET ON
SET LOG OFF
INPUT 'Filename ?', FNAME
INPUT 'Case ?', CASE
#
EXECUTE "POS"_STR(CASE)_" = RCL (;FNAME,CASE,,1,1,,SCL[1],ZER[1])"
TQR = RCL (;FNAME,CASE,,2,1,,SCL[2],ZER[2])
EXECUTE "TA"_STR(CASE)_" = RCL (;FNAME,CASE,,3,1,,SCL[3],ZER[3])"
EXECUTE "GS"_STR(CASE)_" = RCL (;F ^ME,CASE,,4,1,,SCL[4],ZER[4])"
#
PARADIGM = 'SWP'
INPUT 'WHAT PARADIGM ,MVC,SWP,PRB,TOR ?', PARADIGM
IF (UPCAS(PARADIGM) == 'MVC') | (UPCAS(PARADIGM) == 'SWP');
         THEN GO TO [TQCASE]
PRINT "Correcting ineitial effect."
EXECUTE "TQ"_STR(CASE)_" = TQR - FLT (CALIRF,;
        POS"_STR(CASE)_";'Y')"
GOTO [CHOICE]

[TQCASE]
EXECUTE "TQ"_STR(CASE)_" = TQR"
#
# This is a good time to look at extracts of raw data to
# make sure it looks O.K.
#
ANSWER= 'Y'
INPUT 'DO YOU WISH TO LOOK AT DATA [Y] ?', ANSWER
IF (ANSWER <> 'Y') & (ANSWER <> 'y') THEN GOTO [CLEAN UP]

[CHOICE]
C= STR(CASE)
IF (UPCAS(PARADIGM) == 'MVC') THEN GO TO [MVC]
IF (UPCAS(PARADIGM) == 'SWP') THEN GO TO [SWEEP]
IF (UPCAS(PARADIGM) == 'PRB') THEN GO TO [SWEEPPRBS]
IF (UPCAS(PARADIGM) == 'TOR')  THEN GO TO [TORSWEEP]
```

```
[MVC]
EXECUTE "MVCTQ=CDET(ALT("_'TQ'_C_";'D',20,1);'C',FNAME_';
        MVC DEC TQ CASE='_C,)"
EXECUTE "MVCTA=CDET(ALT("_'TA'_C_";'D',20,1);'C',FNAME_';
        MVC DEC TA CASE='_C,)"
EXECUTE "MVCGS=CDET(ALT("_'GS'_C_";'D',20,1);'C',FNAME_';
        MVC DEC GS CASE='_ ,)"
PLGKS(MVCTQ,MVCTA,MVCGS;'N',?,'N','Y',)
PAUSE
EXECUTE "TA=EXT(TA"_C_";1500,1000)"
EXECUTE "GS=EXT(GS"_C_";1500,1000)"
EXECUTE "TQ=EXT(TQ"_C_";1500,1000)"
MTQ=MEAN(TQ)
MTA=MEAN(TA)
MGS=MEAN(GS)
CROSS= (MTA/-MGS)
PRINT "MTQ = ",  MTQ, "        MTA = ", MTA
PRINT " MGS = ", MGS, "        CROSS = ", CROSS

GOTO [CLEAN UP]

[SWEEP]
EXECUTE "SWPTQ=CDET(ALT("_'TQ'_C_";'D',20,1);'C',FNAME_';
         SWP DEC TQ CASE='_C,)"
EXECUTE "SWPPOS=CDET(ALT("_'POS'_C_";'D',20,1);'C',FNAME_';
         SWP DEC POS CASE='_C,)"
EXECUTE "SWPTA=CDET(ALT("_'TA'_C_";'D',20,1);'C',FNAME_';
         SWP DEC TA CASE='_C,)"
EXECUTE "SWPGS=CDET(ALT("_'GS'_C_";'D',20,1);'C',FNAME_';
         SWP DEC GS CASE='_C,)"
PLGKS(SWPPOS,SWPTQ,SWPTA,SWPGS;"N",?,"N","Y",)
PAUSE
PRINT "Evaluating EMG means ..."
EXECUTE "MTA=MEAN(TA"_C_")"
EXECUTE "MGS=MEAN(GS"_C_")"
PRINT
PRINT "MTA = ", MTA, "          MGS = ", MGS
PAUSE
GOTO [LOOP]
EXECUTE 'LENGTH = LEN(TA'_C_')'
PRINT " Computing means of position extracts."
PRINT "This is going to go for ",nint((length/1024) + 0.5),;
         " loops"
SWEEP = 0
SWEEP2 = -1023
MEANTA = ''
MEANGS = ''
SW = ''
```

```
[START]
SWEEP = SWEEP + 1
PRINT SWEEP
SWEEP2 = SWEEP2 + 1024
IF SWEEP2 => LENGTH THEN GOTO [STORE]
EXECUTE 'TA =EXT(TA'_C_';SWEEP2,1024)'
MEANTA = MEANTA_MEAN(TA)
EXECUTE 'GS = EXT(GS'_C_';SWEEP2,1024)'
MEANGS = MEANGS_MEAN(GS)
GO TO [START]

[STORE]
MEANTA = CDET(MEANTA;'C','MEANTA CASE='_C,)
MEANGS = CDET(MEANGS;'C','MEANGS CASE='_C,)
STO(MEANTA,MEANGS;DANFILE,'C')

EXECUTE 'ST1(EXT(ABS(TQ'_C_');2500,,);"Y")'
EXECUTE 'TNP=POS'_C_'[DMIN]'
EXECUTE "ST1(TQ"_C_";'Y')"
# MIN TQ, MAX TQ, TOTAL PASSIVE TQ, TRUE NEURAL POSITION
SET PRECISION 2
PRINT "MAX PLANTAR TQ= ", MIN, "           MAX DORSI TQ = "; MAX
PRINT "TOTAL PASSIVE TORQUE = ", RANGE, "          TNP = ", TNP
SET PRECISION 5
PAUSE

EXECUTE "ST1(POS"_C_";'Y')"
# MIN POS, MAX POS, TESTED ROM
PRINT "MAX PLANTAR POS = ", MIN, " MAX DORSI POS = ", MAX
PRINT "TESTED RANGE OF MOTION = ", RANGE
PAUSE

PLGKS(SWPTQ,SWPPOS;'X','N',?,'N','Y',)

run loopcode.
```

```
[SWEEPPRBS]
EXECUTE "PLGKS(ALT(POS"_C_";'D',10,1),ALT(TQ"_C_";'D',10,1);
        'N',?,'N','Y',)"
EXECUTE "LENGTH = LEN(POS"_C_")"
PRINT "Watch out! Expecting ", nint(length/1024 + 0.5),;
        " loops"
RUN [RITA.MOD]SWEEPIRF.
RUN [RITA.MOD]SWEEPIRFB.
MEANP = CDET (MEANP;'C',fname_' MEANP PRB CASE='_C,)
MEANT = CDET (MEANT;'C',fname_' MEANT PRB CASE='_C,)
VAFTOT = CDET (VAFTOT;'C',FNAME_' VAFTOT PRB CASE='_C,)
MTATOT = CDET (MTATOT;'C',FNAME_' MTATOT PRB CASE='_C,)
MGSTOT = CDET (MGSTOT;'C'.FNAME_' MGSTOT PRB CASE='_C,)
set quiet off
STO(MEANP,MEANT,VAFTOT,MTATOT,MGSTOT;DANFILE,'A')
set quiet on
PLGKS(MEANP,MEANT,VAFTOT;'M','Y','N','y','n','y',)
.plt plot.tek
PLGKS(MTATOT,MGSTOT;'M','Y','N','y','N','Y',)
.plt plot.tek
sw=dom(meanp)
set error off
.DEL TITLE.DAT;*
.DEL CHART.DAT;*
set error on

print @chart 'Filename:  ', FNAME, '  Case =  ', CASE, '  PRB'
PRINT @CHART ' '
PRINT @TITLE '  SWEEP   MEANP     MEANT     VAFTOT     MTATOT;
           MGSTOT'
PRT(SW,MEANP,MEANT,VAFTOT,MTATOT,MGSTOT;'dat.txt','6g10.3')
.APPEND TITLE.DAT,DAT.TXT CHART.DAT
.PRINT CHART.DAT
PAUSE
.DEL TITLE.DAT;*
.DEL DAT.TXT;*
.DEL CHART.DAT;*

PRINT " You about to run SWEEPFIT, with about, ",nint
        (length/1024 +0.5)," loops"
RUN [RITA.MOD]SWEEPFIT.
RUN [RITA.MOD]SWEEPFITB.
KTOT=CDET(KTOT;'C',FNAME_' KTOT PRB',)
SET QUIET OFF
PLGKS(KTOT;'A','Y','M','Y','N')
PAUSE
PLGKS(KTOT,MEANP;'M','Y','X','A','Y','N')
PAUSE
PLGKS(KTOT,MEANT;'M','Y','X','A','Y','N')
PAUSE
```

```
SET QUIET ON
SET ERROR OFF
.DEL TITLE.DAT;*
.DEL CHART.DAT;*
SET ERROR ON
PRINT @CHART 'FILENAME: ', FNAME, '  CASE = ', CASE, ' PRB'
PRINT @CHART ' '
DUMMY = 'SWEEP       GAIN      FREQ    DAMPTOT'
DUMMY = DUMMY_'     KTOT        ITOT      BTOT      VAHTOT'
PRINT @TITLE DUMMY
sw=dom(meant)
PRT (SW,GAINTOT,FREQTOT,DAMPTOT,KTOT,ITOT,BTOT,VAHTOT;'DAT;
        .TXT',!)
.APPEND TITLE.DAT,DAT.TXT CHART.DAT
.PRINT CHART.DAT
PAUSE
.DEL TITLE.DAT;*
.DEL DAT.TXT;*
.DEL CHART.DAT;*
.DEL DUMMY.NXC;*
STO(GAINTOT,FREQTOT,DAMPTOT,KTOT,ITOT,BTOT,VAHTOT;DANFILE,'A')
GOTO [CLEAN UP]

[TORSWEEP]
EXECUTE "TORTQ=CDET(ALT("_'TQ'_C_";'D',10,1);'C',FNAME ' ;
DEC TQ TOR CASE='_C,)"
SET QUIET OFF
PLGKS(TORTQ;'A','Y','N')
PAUSE
SET QUIET ON
EXECUTE "TORTA=CDET(ALT("_'TA'_C_";'D',10,1);'C',FNAME ' DEC;
         TA TOR CASE='_C,)"
EXECUTE "TORGS=CDET(ALT("_'GS'_C_";'D',10,1);'C',FNAME_' DEC;
         GS TOR CASE='_C,)"
PLGKS(TORTA,TORGS;'N',?,'N','Y',)
PAUSE
execute 'length = len(pos'_c_')'
print "You are about to run SWEEPIRF, expect ",nint;
        (length/1024 + 0.5)," loops"
RUN [RITA.MOD]SWEEPIRF.
MEANT = CDET (MEANT;'C',fname_' MEANT TOR CASE='_C,)
VAFTOT = CDET (VAFTOT;'C',FNAME_' VAFTOT TOR  CASE='_C,)
PLGKS(MEANT,VAFTOT;'M','Y','N',?,'N','Y',)
PAUSE
MTATOT = CDET (MTATOT;'C',FNAME_' MTATOT TOR CASE='_C,)
MGSTOT = CDET (MGSTOT;'C',FNAME_' MGSTOT TOR CASE='_C,)
PLGKS(MTATOT,MGSTOT;'M','Y','N',?,'N','Y',)
PAUSE
set error off
.DEL TITLE.DAT;*
.DEL CHART.DAT;*
set error on
pause
```

```
print @chart 'Filename:  ', FNAME, '  Case =  ', CASE, '  TOR'
PRINT @CHART ' '
PRINT @TITLE '  SWEEP   MEANP     MEANT     VAFTOT    ;
        MTATOT   MGSTOT'
PRT (SW,MEANP,MEANT,VAFTOT,MTATOT,MGSTOT;'dat.txt','6g10.3')
.APPEND TITLE.DAT,DAT.TXT CHART.DAT
.PRINT CHART.DAT
PAUSE
.DEL TITLE.DAT;*
.DEL DAT.TXT;*
.DEL CHART.DAT;*
STO(MEANP,MEANT,VAFTOT,MTATOT,MGSTOT;DANFILE,'A')
PRINT " ABOUT TO RUN SWEEPFIT, 50 LOOPS"
RUN  [RITA.MOD]SWEEPFIT.
KTOT=CDET(KTOT;'C',FNAME_' KTOT TOR CASE='_C,)
SET QUIET OFF
PLGKS(KTOT,VAHTOT;'M','Y','A','Y','N')
PAUSE
PLGKS(KTOT,MEANT;'X','M','Y','A','Y','N')
PAUSE
SET QUIET ON
SET ERROR OFF
.DEL TITLE.DAT;*
.DEL CHART.DAT;*
SET ERROR ON
PRINT @CHART 'FILENAME: ', FNAME, '  CASE = ', CASE, ' TOR'
PRINT @CHART ' '
DUMMY = 'SWEEP      GAIN     FREQ    DAMPTOT'
DUMMY = DUMMY_'     KTOT       ITOT      BTOT     VAHTOT'
PRINT @TITLE DUMMY
PRT (SW,GAINTOT,FREQTOT,DAMPTOT,KTOT,ITOT,BTOT,VAHTOT;'DAT;
        .TXT',!)
.APPEND TITLE.DAT,DAT.TXT CHART.DAT
.PRINT CHART.DAT
PAUSE
.DEL TITLE.DAT;*
.DEL DAT.TXT;*
.DEL CHART.DAT;*
.DEL DUMMY.NXC;*
STO(GAINTOT,FREQTOT,DAMPTOT,KTOT,ITOT,BTOT,VAHTOT;DANFILE,'A')

[CLEAN UP]
PRINT "Cleaning up..."
set error off
set error on
.purge
set log on
set quiet off
```

```
# Rita M    Loopcode a module to calculate loop area
# of the passive Tq-pos curve
# Index to find out what case numbers I need for swp

run [rita.mod]index.

# Aninit for initializing analysis by rcl scaling and
# zero levels


run [rita.mod]aninit.

INPUT 'Case ?', CASE
EXECUTE "POS"_STR(CASE)_" = RCL (;FNAME,CASE,,1,1,,SCL[1];
        ,ZER[1])"
TQR = RCL (;FNAME,CASE,,2,1,,SCL[2],ZER[2])
EXECUTE "TA"_STR(CASE)_" = RCL (;FNAME,CASE,,3,1,,SCL[3];
        ,ZER[3])"
EXECUTE "GS"_STR(CASE)_" = RCL (;FNAME,CASE,,4,1,,SCL[4];
        ,ZER[4])"

execute "pos=pos"_str(case)

vel=dif (pos)
pow=vel*tqr
work=int(pow)
start(work)
loop= work[end(work)]-work[start(work)]
print " The loop area for ",fname," is ",loop
.purge
```

```
# RM Module SWEEPIRFB:   To calculate IRFs during ankle sweep.
# Input details

VAFTOT = ''
CONSET (;'SEGLEN',1024)
# INPUT "Case? ", CASE
C = STR(CASE)
EXECUTE 'LENGTH = LEN(POS'_C_')'
CONSET (;'SWEEP',0)
CONSET(;'SWEEP2',-1023)
MEANP = ''
MEANT = ''
MTATOT= ''
MGSTOT= ''
TPIRFB = ''
SW = ''
[START]
CONSET(;'SWEEP', SWEEP + 1)
PRINT SWEEP
S = STR(SWEEP)
CONSET(;'SWEEP2',SWEEP2 + SEGLEN)
IF SWEEP2 => LENGTH THEN GOTO [STORE]
EXECUTE 'POS = EXT(POS'_C_';SWEEP2,SEGLEN)'
MEANP = MEANP_MEAN(POS)
POS = POS - SLOPE(POS)*DOM(POS)
EXECUTE 'TQ = EXT(TQ'_C_';SWEEP2,SEGLEN)'
MEANT = MEANT_MEAN(TQ)
TQ = TQ - SLOPE(TQ)*DOM(TQ)
EXECUTE 'TA=EXT(TA'_C_';SWEEP2,SEGLEN)'
EXECUTE 'GS=EXT(GS'_C_';SWEEP2,SEGLEN)'
MTA=MEAN(TA)
MGS=MEAN(GS)
MTATOT=MTATOT_MTA
MGSTOT=MGSTOT_MGS
IRF = FIL(TQ,POS;201,"Y")
VAFTOT=VAFTOT_VAF
TPIRF= -EXT(SMO(IRF;2);100,100)
TPIRFB=TPIRFB_TPIRF
SW=SW_SWEEP

GO TO [START]
[STORE]
#
# This would be a good time to plgks MEANP,MEANT,VAFTOT to look
#                     problems
MEANP=CDET(MEANP;'C','MEANP CASE='_C,)
MEANT=CDET(MEANT;'C','MEANT CASE='_C,)
VAFTOT=CDET(VAFTOT;'C','VAFTOT CASE='_C,)
LENSWEEP = LEN(MEANP)

# Return to recall module
```

```
# RM  March 88
# Module SWEEPFITb: To calculate IRFs fits during ankle sweep.

# Input details

# INPUT "Case? ", CASE
C = STR(CASE)
CONSET(;'SWEEP', 0 )
CONSET(;'IRFLEN',100)
CONASK(;'IRFLEN','Length of IRF',1,1000,?)
VAHTOT = ''
GAINTOT = ''
FREQTOT = ''
DAMPTOT = ''
KTOT = ''
BTOT = ''
ITOT = ''
SW = ''
P1 = .01
P2 = 100
P3 = .5

TPFIT = FIRF2(TPIRFB;IRFLEN,P1,P2,P3)
KTOT = ABS(1/P1)
ITOT = (ABS(1/P1)/(P2^2))
BTOT = (2*P3*ABS(1/P1)/P2)
VAHTOT = VAF
GAINTOT=p1
FREQTOT = p2
DAMPTOT=p3


GAINTOT=CDET(GAINTOT;'C', FNAME_'GAINTOT CASE='_C,)
FREQTOT=CDET(FREQTOT;'C', FNAME_'FREQTOT CASE='_C,)
DAMPTOT=CDET(DAMPTOT;'C', FNAME_'DAMPTOT CASE='_C,)
KTOT=CDET(KTOT;'C', FNAME_'KTOT CASE='_C,)
ITOT=CDET(ITOT;'C', FNAME_'ITOT CASE='_C,)
BTOT=CDET(BTOT;'C', FNAME_'BTOT CASE='_C,)
VAHTOT=CDET(VAHTOT;'C', FNAME_'VAHTOT CASE='_C,)
```

```
# RM module lag - To determine offset between two K-position
# curves

INPUT 'SUBJECT code',CODE
set quiet on

fname1=code '1.dan'
fname2=code_'2.dan'
execute 'p'_code_'1 =rcl(;fname1,!)'
execute 'p' code '1=ext(p'_code_'1;2,len(p'_code_'1)-2)'

execute 'k' code_'1 =rcl(;fname1,9,!)'
execute 'k' code_'1=ext(k'_code_'1;2,len(k'_code_'1)-2)'

execute 'p'_code_'2 =rcl(;fname2,!)'
execute 'p' code_'2=ext(p'_code_'2;2,len(p'_code_'2)-2)'

execute 'k'_code_'2 =rcl(;fname2,9,!)'
execute 'k'_code_'2=ext(k'_code_'2;2,len(k'_code_'2)-2)'

execute 'k1=k'_code_'1'
execute 'k2=k' code_'2'

execute 'pos1=p'_code_'1'
execute 'pos2=p'_code_'2'

K = K1_K2
[restart]

offmin=-.4
offmax=1.0
offincr=.01

# CONASK(;'OFFMIN','Minimum offset',-1.,1.,,)
# CONASK(;'OFFMAX','Maximum offset',OFFMIN,1.,,)
# CONASK(;'OFFINCR','Offsset increment',-1.,1.,?)
CONSET(;'OFFSET',OFᶠMIN)
XOFF=
XVAF=
[LOOP]
P=POS1 (POS2-OFFSET)
F=POLYFIT(P,K;10,10)
XOFF=XOFF OFFSET
XVAF=XVAF VAF
CONSET(;'OFFSET',OFFSET+OFFINCR)
IF (OFFSET < OFFMAX) THEN GOTO [LOOP]
ST1(XVAF)
CONSET(;'OFFSET',XOFF[DMAX])
run [rıta.mod]compaıe.
```

```
# RM COMPARE   A module that compares two K-Position curves
# by fitting a polynomial then determining the position offset
# Inputs POS1,K1
#        POS2,K2
#        OFFSET



CONASK(;'OFFSET','Offset ',-1.,1.,,)
P2 = POS2-OFFSET
F1= polyfit (pos1,k1;10,10)
f2=polyfit (p2,k2;10,10)
#
CONSET(;'XMIN',MAX(MIN(POS1)_MIN(P2)))
CONSET(;'XMAX',MIN(MAX(POS1)_MAX(P2)))
#CONASK(;'XMIN','Minimum X ',-2.,2.,,)
#CONASK(;'XMAX','MaximumX',XMIN,2,,)
X = (ramp(;101)-1)*(XMAX-XMIN)/100
X = X + XMIN
KP1=polyval(f1,x;MIN(POS1),MAX(POS1))
KP2=polyval(f2,x;MIN(P2),MAX(P2))
PLGKS (kp1,x,kp2,x;'x','s',)
set quiet off
F=MREG(kp1,kp2;)

# INPUT 'Store ?',ANSWER
# IF (UPCAS(ANSWER)=='Y') THEN GOTO [STORE]

EXECUTE 'pp'_CODE_'1=kp1'
EXECUTE 'pp'_CODE_'2=KP2'
EXECUTE 'LAG'_CODE_'=OFFSET'
print " the code is    ",code
print " lag =  ",offset
```

```
#  RM Lkrange   A module that determines a low stiffnes
#  position range

# Extracting the position range that is <= 30 Nm after the
# offset is removed.
# The predicted polynomial with actual ROM corrected for

INPUT 'What is the four letter subject code (in quotes)',CODE
INPUT 'What is tne two letter code, initials (in quotes)',CODEB
INPUT 'WHAT WAS LAG?', LAG

FNAME1=CODE '1.DAN'
FNAME2=CODE '2.DAN'
# RECALLING POS CHANNELS, CORRECTING THEM AND EXPANDING
# THEM TO 1001 PNTS
EXECUTE 'POS' CODEB '1=RCL(;FNAME1,!)'
EXECUTE 'POS' CODEB '1=EXT(POS'_CODEB_'1;2,LEN(POS'_CODEB_'1)
                                                          -2)'

EXECUTE 'P' CODEB_'2=RCL(;FNAME2,!)'
EXECUTE 'P' CODEB_'2=EXT(P'_CODEB_'2;2,LEN(P'_CODEB_'2)-2)'
EXECUTE 'POS' CODEB_'2=P'_CODEB_'2-LAG'

EXECUTE 'ST1(POS'_CODEB_'1;"Y")'
P1=(RAMP(;1001)-1)/1000
P1=P1*RANGE+MIN

EXECUTE 'ST1(POS'_CODEB_'2;"Y")'
P2=(RAMP(;1001)-1)/1000
P2=P2*RANGE+MIN
EXECUTE 'PPOS' CODEB_'1=P1
EXECUTE 'PPOS'_CODEB_'2=P2

# DETERMINING THE K OFFSET AS THE MINIMUM OF THE PREDICTED K
EXECUTE 'ST1(APP'_CODE_'1)'
MIN1=MIN
EXECUTE 'OFF' CODE_'1=APP'_CODE_'1-MIN'

EXECUTE 'ST1(APP'_CODE_'2)'
MIN2=MIN
EXECUTE 'PRINT "THE MINIMUM OF APP'_CODE_'1 IS ", MIN1'
EXECUTE 'PRINT "THE MINIMUM OF APP'_CODE_'2 IS ",MIN2'
EXECUTE 'OFF' CODE '2=APP' CODE '2-MIN'
PAUSE

EXECUTE 'CX' CODE_'1=CEXT((OFF' CODE_'1 <= 30),P1)'
EXECUTE 'CX' CODE_'2=CEXT((OFF' CODE_'2 <= 30),P2)'
EXECUTE 'ST1(CX' CODE '1)
EXECUTE 'LSb' CODE '1=RANGE
pause

EXECUTE 'ST1(CX'_CODE_'2)
EXECUTE 'LSb' CODE_'2=RANGE
EXECUTE 'PRINT "LSb'_CODE_'1 IS = ", LSb'_CODE_'1'
EXECUTE 'PRINT "LSb'_CODE_'2 IS = ", LSb'_CODE_'2'
```

BIBLIOGRAPHY

Agarwal, C.G. and Gottlieb, G.L. Oscillation of the human ankle joint in response to applied sinusoidal torque on the foot. J Physiol. 268: 151-176, 1977.

Allbrook, D. Muscle Regeneration. Physiother Can. 59: 240-247, 1973.

Alnaqeeb, M.A., AlZaid, N.S., and Goldspink, G. Connective tissue changes and physical properties of developing and ageing muscle. J Anat. 139: 677-689, 1984.

American Academy of Orthopaedic Surgeons. Joint Motion Methods of Measuring and Recording. published by American Academy of Orthopaedic Surgeons. Chicago, 1965.

Armstrong, L., Winant, D., Swasey, P., Seidle, M., Carter, A. and Gehlsen, G. Using Isokinetic Dynamometry to Test Ambulatory Patients with Multiple Sclerosis. Phys Ther. 63: 1274-1279, 1983.

Bauer, M., Jonsson, K. and Nillsson, B. Thirty year follow up of ankle fractures. Acta Orthop Scand. 56: 103-106, 1985.

Basmajian, J.V. Primary Anatomy 7 th ed. Baltimore. Williams and Wilkins Co. 1979.

Boone, D.C., Azen, S.P., Lin, C.M., Spence, C., Baron, C., and Lee, L. Reliability of goniometric measurement. Phys Ther. 58: 1355-1360, 1978.

Boone, D.C. and Azen, S.P. Normal range of motion of joints in male subjects. J Bone Joint Surg. 61: 756-759, 1979.

Bozdech, Z. Postraumaticka sinovitis. Acta Clin Orthop Traumatol Cech 43: 244-247, 1976.

Broberg, C. and Grimby, G. Measurement of torque during passive and active ankle movements in patients with hypertonia, a methodological study. Scand J Rehabil Med. (Suppl) 9: 108-117, 1983.

Buhr, A.J. and Cooke, A.M. Fracture patterns. The Lancet 1 :531-536,March 14, 1959.

Burke, D. Critical examination of the case for or against fusimotor involvement in disorders of muscle tone. In Motor Control Mechanisms in Health and Disease, edited by J.E. Desmedt, New York, Raven Press, 1983.

Caillet, R. Foot and Ankle Pain. Philadelphia: F.A. Davis Company, 1968.

Cardenas, D.D., Stolov, W.C., and Hardy, R. Muscle fibre number in immobilization atrophy. Arch Phys Med Rehabil. 58: 423-426, 1977.

Carmines, E.G. Reliability and validity assessment. Beverly Hills, Sage Publications, 1st ed., 1979.

Daniels, L. and Worthingham, C. Muscle Testing, Techniques of Manual Examination. 3rd ed. Philadelphia: W.B. Saunders Company , 1972.

Daniels, L. and Williams, M. Muscle Testing, Techniques of Manual Examination. 1st ed. Philadelphia: W.B. Saunders Company, 1946.

Dannenbaum, E.S., Weiss, P.L., Kearney, R.E., Gauthier, S. and Gregor, R.W. The effect of activation on Parkinsonian rigidity. Society for Neurosciences Abstracts, 1988, in press.

Davies, C.T.M. and White, M.J. Contractile properties of elderly Triceps Surae. Gerontology. 29: 19-25, 1983.

Derscheid, G.L. and Brown, W.C. Rehabilitation of the ankle. Clin Sports Med. 4: 527-544, 1985.

Dietz, V. and Berger, W. Normal and impaired regulation of muscle stiffness in gait: A new hypothesis about muscle hypertonia. Exp Neurol. 79: 680-687, 1983.

Dietz, V., Quintern, J. and Berger, W. Electrophysiological studies of gait in spasticity and rigidity. Brain. 104: 431-449, 1981.

Dowson, D., Wright, V., and Longfield, M.D. Human joint lubrication. Bio Med Eng. 4: 160-165, 1969.

Dowson, D. Lubrication and wear of joints. Physiother. 59: 104-106, 1973.

Edgerton, V.R., Barnard, R.J., Peter, J.B., Maier, A. and Simpson, D.R. Properties of immobilized hind-limb muscles of the Galago Senegalensis. Exp Neurol. 46: 115-131, 1975.

Edstrom, L. Selective atrophy of the red muscle fibres in the quadriceps in long standing knee joint dysfunction injuries to the anterior cruciate ligament. J Neurol Sci. 11: 551-558, 1970.

Ekstrand, J., Wiktorsson, M., Oberg, B., and Gillquist, J. Lower extremity goniometric measurements: A study to determine thier reliability. Arch Phys Med Rehabil. 63: 171-175, 1982.

Feinstein, A.R. An additional science for clinical medicine: IV The development of clinimetrics. An Int Med. 99: 843-848, 1983.

Fife, D. and Barancik, J. Northeastern Ohio trauma study III: Incidence of Fractures. Ann Emerg Med. 14: 244-248, 1985.

Frankel, V.H. and Nordin, M. Biomechanics of the Ankle. In: Basic Biomechanics of the Skeletal System. edited by Frankel,V.H.and Nordin,M. Philadelphia,Lea and Febiger, 1980.

Freeman, M., Dean, M., and Hanham, I. The etiology and prevention of functional instability of the foot. J Bone Joint Surg. 47B: 678-685, 1965.

Gajdosik, R. and Bohnon. Clinical measurement of ROM: Review of goniometry emphasizing reliability and validity. Phys Ther. 67: 1867-1872, 1987.

Garett, W.E., Seaver, A.V., Boswick, J.S., Urbaniak, J.R., and Goldner, L. Recovery of skeletal muscle after laceration and repair. J Hand Surg. 9: 683-692, 1984.

Glencross, D. and Thorton, E. Position sense following joint injury. J Sports Med. 21: 23-27, 1981.

Goldspink, G., Tabary, C.,Tabary, J.C. and Tardieu, C. Effect of denervation on the adaptation of sarcomere number and muscle extensibility to the functional length of the muscle. J Physiol. 236: 733-742, 1974.

Gossman, M.R., Sahrmann, S.A., and Rose, S.J. Review of length-assiciated changes in muscle. Experimental evidence and clinical implications. Phys Ther. 62: 1799-1808, 1982.

Gottlieb, G.L. and Agarwal, G.C. Dependence of human ankle compliance on joint angle. J Biomech. 11: 177-181, 1978.

Gottlieb, G.,Agarwal, G.,and Penn, R. Sinusoidal oscillation of the ankle as a means of evaluating the spastic patient. J Neurol Neurosurg Psychiatry. 41: 32-39, 1978.

Halkjaer-Kristensen, J. and Ingemann-Hansen, T. Variations in single fibre areas and fibre composition in needle biopsies from the human quadriceps muscle. Scand J Clin Lab Invest. 41: 391-395, 1981.

Heerkens, Y., Woittiez, R., Huijing, P., Huson, A., vanIngen Schenou, G. and Rozendal, R. Passive resistance of the human knee: the effect of remobilization. J Biomed Eng. 9: 69-76, 1987.

Hettinga, D.L. I. Normal joint structures and their reaction to injury. J Orthop Sports Phys Ther. 1: 16-22, 1979a.

Hettinga, D.L. II. Normal joint strutures and their reaction to injury. J Orthop Sports Phys Ther. 1: 83-88, 1979b.

Hnik, P., Holas, M., Payne, R. Reflex muscle atrophy induced by chronic peripheral nociceptive stimulation. J Physiol. (Paris) 73: 241-250, 1977.

Hof, A.L., and Van Den Berg, JW. Linearity between the weighted sum of the EMGs of the hyman Triceps Surae and the total torque. J Biomechanics. 10: 529-539, 1977.

Holmes, J.R. and Aldernik, G.L. Isokinetic strength characteristics of the quadriceps femoris and hamstring muscles in high school students. Phys Ther. 64: 914-918, 1984.

Hoppenfeld, S. Physical Examination of the Spine and Extremities. 1st ed. Norwalk: Appleton-Century-Crofts, 1978.

Hunter, I.W. and Kearney, R.E. Dynamics of Human Ankle Stiffness: Variation with Mean Ankle Torque. J Biomech. 15: 747-752, 1982.

Hunter, I.W. and Kearney, R.E. Invariance of ankle dynamic stiffness during fatiguing muscle contractions. J Biomech. 16: 985-991, 1983.

Hunter, I.W. and Kearney, R.E. Nexus: A computer language for physiological systems and signal analysis. Comp Biol and Med. 14: 386-401, 1984.

Hunter, I.W., Kearney, R.E. and Weiss, P.L. Muscle stiffness measurements in the presence of reflexes. Proceedings of 10th Canadian Medical and Biological engineering Conference, pp. 15-16. Canadian Medical and Biological Engineering Society, 1983.

Inman, V.T. The Joints of the Ankle. Baltimore: Williams and Wilkins, 1976.

Janda, V. Muscle Function Testing. 1st English ed. London: Butterworths, 1983.

Kearney, R.E. and Hunter, I.W. Dynamics of human ankle stiffness: Variation with displacement amplitude. J Biomech. 15: 753-756, 1982.

Kearney, R.E.,Hunter, I.W., Weiss, P.L. and Spring, K. Tilt-table/ankle-actuator system for the study of vestibulo spinal reflexes. Med Biol Eng Comput. 21:301-305, 1983.

Kendall, H.O. and Kendall, F.P. Muscles, Testing and Function, Baltimore, Williams and Wilkins Co., 1st ed., 1949.

Kotwick, J.E. Biomechanics of the Foot and Ankle. Clin Sports Med. 1: 19-34, 1982.

Kroemer, K.H.E. and Marras, W.S. Towards an objective assessment of the 'maximal voluntary contraction' component in routine muscle strength measurements. Eur J Appl Physiol. 45: 1-9, 1980.

Lambert, K. The weight-bearing function of the fibula. A strain gauge study. J Bone Joint Surg. 53a: 507, 1971.

Landin, L. and Danielsson, L. Children's ankle fractures. Acta Orhtop Scand. 54: 634-640, 1983.

Lewis, R.C. Handbook of Traction, Casting and Splinting Techniques, Philadelphia, J.B.Lippincott Co., 1st ed., 1977.

Lindsjo, U., Danckwardt, G., and Sahlstedt, B. Measurement of the motion range in the loaded ankle. Clin Orthop. 199: 68-71, 1985.

Lovett, R.W. and Martin, E.B. Certain aspects of infantile paralysis with a description of muscle testing. JAMA. 66: 729-733, 1916.

Low, J.L. The reliability of joint measurement. Physiotherapy. 62: 227-229, 1976.

Ma, S.P., and Zahalak, G. The mechanical response of the active human triceps brachii muscle to very rapid stretch and shortening. J Biomech. 18: 585-598, 1985.

MacDougall, J.D., Elder, G.C.B., Sale, D.G., Moroz, J.R. and Sutton, J.R. Effects of strength training and immobilization on human muscle fibres. Eur J Appl Physiol Occup Physiol. 43: 24-34, 1980.

Maier, A., Crockett, J.L., Simpson, D.R., Saubert, I.V. and Edgerton, V.R. Properties of immobilized guinea pig hindlimb muscles. Am J Physiol. 231: 1520-1526, 1976.

Mann, W. and Salafsky, B. Enzymic and physiological studies on normal and disused developing fasc and slow cat muscles. J Physiol. 208: 33-47, 1970.

Maughan, R.J., Watson, J.S. and Weir, J. Strength and cross-sectional area of human skeletal muscle. J Physiol. 338: 37-49, 1983.

Mawdsley, R.H. and Knapik. J.J. Comparison of isokinetic measurements with tests repetions. Phys Ther. 62: 169-172, 1982.

Michels, E. Measurement in physical therapy. Phys Ther. 63: 209-215, 1983.

Miller, P. Assessment of joint motion. In Rothstein,JM.(editor): Measurement in Physical Therapy: Clinics in Physical Therapy. New York, Churchill Linvingstone Inc. 1985.

Mitchell, W.C., Shaftan, G.W. and Sclafani, S.J.A. Mandatory open reduction: it's role in displaced ankle fractures. J Trauma. 19: 602-615, 1979.

Moller, M., Ekstrand, J., Oberg, B., and Gillquist, J. Duration of stretching effect on range of motion in lower extremities. Arch Phys Med Rehabil. 66: 171-173, 1985.

Morier, R.L., Weiss, P.L. and Kearney, R.E. Low inertia rigid limb fixation using fibreglass casting material. Canadian Medical and Biological Engineering Society Conference Abstract. 1988, in press.

Mortimer, J. and Webster, D. Relationship between quantitative meeasures of rigidity and tremor and the electromyographic responses to load perturbations in unselected normal subjects and Parkinson patients. In J.E. Desmedt, Cerebral Motor Control in Man: Long Loop Mechanisms. Prog Clin Neurophysiol vol 4, Karger, Basel, 1978.

Murray, M.P., Kory, R.C. and Sepic, S.B. Walking patterns of normal women. Arch Phys Med Rehabil. 51:637, 1970.

Murray, M.P., Drought, A.B., and Kory, R.C. Walking patterns of normal men. J Bone Joint Surg. 46A:335, 1964.

Muwanga, C.L. and Dove, A.F. The measurement of ankle movements - a new method. Injury. 16: 312-314, 1985.

Nicholas, J.A., Sapaga, A., Kraus, H. and Webb, J.N. Factors Influencing Manual Muscle Tests in Physical Therapy. J Bone Jnt Surg. 60a: 186-190, 1978.

Nilsson, B.E.R. Age and sex incidence of ankle fractures. Acta Orthop Scand. 40: 122-129, 1969.

Noyes, F.R. Functional properties of knee ligaments and alterations induces by immobilization. Clin Orthop. 123: 210-242, 1977.

Pachter, B.R. and Eberstein, A. Neuromuscular plasticity following limb immobilization. J. Neurocytol. 13: 1013-1025, 1984.

Pearson, M.B., Bassey, E.J. and Bendall, M.J. Muscle strength and anthropometric indices in elderly men and women. Age and Ageing 14: 49-54, 1985a.

Pearson, M.B., Bassey, E.J. and Bendall, M.J. The effects of age on muscle strength and anthropometric indices within a group of elderly men and women. Age and Ageing. 14: 230-234, 1985b.

Peat, M. and Campbell, G. Measurement systems in physical therapy. Physiother Can. 31: 132-136, 1979.

Pedneault, M. Mesure de la force musculaire: methodes privilegiees par les physiotherapeutes d'etablissements Montrealais. Physiother Can. 37: 370-375, 1985.

Prins, J.G. Diagnosis and treatment of injury to the lateral ligament of the ankle: A comparative clinical study. Acta Chirug Scand. Supp 486: 1-149, 1978.

Ramsey, P.L. and Hamilton, W. Changes in tibiotalar area of contact caused by lateral talar shift. J Bone Jnt Surg. 58: 356-357, 1976.

Salter, R.S. Textbook of Disorders and Injuries of the Musculoskeletal System. Baltimore: Williams and Wilkins Co., 1st ed., 1970.

Sammarco, G.J., Burstein, A.H., and Frankel, V.H. Biomechanics of the ankle: A kinematic study. Orthop Clin North Am. 4: 75-96, 1973.

Sondenaa, K., Hoiguard, U., Smith, U., and Alho, A. Immobilization of operated ankle fractures. Acta Orthop Scand. 57: 59-61, 1986.

Soren, A., Klein, W. and Huth, F. Microscopic comparison of the synovial changes in post traumatic synovitis and osteoarthritis. Clin Orthop Rel Res. 121: 191-195, 1976.

Statford, P., Agostino, V., Brazeau, C., and Gowitzke, B.A. Reliability of joint measurement: a discussion of methodology issues. Physiother 36: 5-9, 1984.

Steindler, A. Kinesioloygy of the human body 1st ed. Charles C.Thomas, Fort Lauderdale, 1955.

Tabary, J.C., Tabary, C., Tardieu, C., Tardieu, G., and Goldspink, G. Physiological and structural changes in the cat's soleus muscle due to immobilization at different lengths by plaster casts. J Physiol. 224: 231-244, 1972.

Tardieu, C., Huet de la Tour, E., Bret, M.D., and Tardieu, G. Muscle hypoextensibility in children with cerebral palsy.I Clinical and experimental observations. Arch Phys Med Rehabil. 63: 97-102, 1982.

Teitz, C.C. Sports medicine concerns in dance gymnastics. Ped Clin North Am. 29: 1399-1421, 1982.

Tipton, C.M.,James, S.L., Mergner, W. and Tcheng, T. Influence of exercise on strength of medial collateral knee ligaments of dogs. Am J Physiol. 21: 894-902, 1970.

Trombly, C.A. editor, Occupational Therapy for Physical Dysfunction. 2nd ed. Baltimore: Williams and Wilkins Co., 1983.

Tunturi, T.,Kemppainen, K., Patiala, H., Suakas, M.,Tamminen, O., and Rokkanen, P. Importance of anatomical reduction for subjective recovery after ankle fracture. Acta Orthop Scand. 54: 641-647, 1983.

Warhol, M.J.,Siegel, A.I.,Evans, W.I., and Silverman, L.M. Skeletal muscle injury and repair in marathon runners after competition. Am J Pathol. 118: 331-339, 1985.

Wakim, K.G., Gersten, J.W., Elkins, E.C., and Marin, G.M. Objective recording of muscle strength Arch Phys Med 31: 90-101, 1950.

Watts, R.L.,Wiegner, A.W. and Young, R.R. Elastic properties of muslces measured at the elbow in man: II Patients with parkinsonian rigidity. J. Neurol Neurosurg Psych., 49: 1177-1181, 1986.

Weiss, P.L., Hunter, I.W., and Kearney, R.E. Human ankle joint stiffness over the full range of muscle activation levels. J Biomech., 1988, in press.

Weiss, P.L., Kearney, R.E. and Hunter, I.W. Position dependence of ankle joint dynamics Part I Passive Mechanics. J Biomech. 19: 727-735, 1986a.

Weiss, P.L., Kearney, R.E. and Hunter, I.W. Position dependence of ankle joint dynamics Part II Active Mechanics. J Biomech. 19: 737-751, 1986b.

Weiss, P.L., Hunter, I.W. and Kearney, R.E. Rigid polyurethane foam casts for the fixation of human limbs. Med & Biol Eng & Comput. 22: 603-604, 1984a.

Weiss, P.L., Forget, R., Kearney, R.E., Hunter, I.W. and Lamarre, Y. Ankle stiffness in a functionally deafferented human subject. Soc Neurosci Abst. 10: 334, 1984b.

Wiegner, A.W. and Watts, R.L. Elastic properties of muscles measured at the elbow in man. I. normal controls. J Neurol Neurosurg Psych. 49: 1171-1176, 1986.

Williams, P.E. and Goldspink, G. Changes in the connective tissue component of muscle during periods of decreased or increased activity. J Anat. 133: 133, 1981.

Winter, D.A. The Biomechanics and Motor Control of Human Gait. University of Waterloo Press, Waterloo, 1987.

Woo, S.L.,Matthews, J.V., Akeson, W.H., Amiel, D.and Convery, F.R. Connective tissue response to immobility: correlative study of biomechanical and biochemical measurements of normal and immobilized rabbit knees. Arth Rheum. 18: 257-264, 1975.

Wright, V. Stiffness: A review of its measurement and physiological importance. Physiother. 59: 107-111, 1973.

Wyke, B. The neurology of joints. Ann R Coll Surg Eng. 41: 25-50, 1967.

Wynarsky, G.T. and Greenwald, A.S. Mathematical model of the human ankle joint. J Biomech 16: 241-251, 1983.