THE UNIVERSITY OF MANITOBA

AN ELECTROGONIOMETRIC ANALYSIS OF KNEE JOINT MOVEMENT AND ELECTROMYOGRAPHIC STUDY OF THE PEAK ACTIVITY OF THE THIGH MUSCLES DURING THE STAIR CYCLE IN NORMALS AND SYME AMPUTEES

by

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A thesis submitted to the Faculty of Graduate Studies of the University of Manitoba in partial fulfillment of the requirements of the degree of

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To

ELIZABETH

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ABSTRACT

Ascending and descending stairs were studied in 18 normal and 10 Syme amputees. The normal group age range was 20-53 years with a mean of 27.7; and the amputee range was 19-56 years with a mean of 40.7 years.

The parameters studied were temporal (foot switch), knee electrogoniometry of flexion and extension, and peak EMG of the quadriceps and hamstrings.

The data of each variable for both limbs of both groups in both ascending and descending cycles was analysed and compared for differences within each group and between each group.

A reliability analysis was calculated for the three repetitions of each of the ascending and descending cycles, and sample size analysis calculated for all appropriate data.

The temporal data, established by foot switches were the cycle time, the cadence, and the swing and stance phases expressed as a percent of the cycle and in absolute time units.

The electrogoniometric data was established by an instrument capable of measuring single plane movement which occurs with a varying axis. The knee joint angle at various significant parts of the stair cycle, the maximum and minimum knee ranges, their time of occurrence, and finally, the total range of knee motion during the stair cycle were measured and analysed.

The peak electromyographic data established via surface electrodes and then measured and analysed were the position, amplitude and sequence of peaks in both muscle groups.

The electrogoniometer data were correlated with the peak EMG data, which related the peak EMG to the knee joint position.

The possible application of the information gained is discussed with respect to myoelectric control of prostheses, functional electrical stimulation and exercise therapy.

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CHAPTER I

INTRODUCTION

For the evaluation of clinical problems in rehabilitation medicine, three basic requirements must be met. First, a thorough understanding of the functional anatomy is essential. Second, a systematic and orderly examination of the problem is necessary. Finally, a comprehensive understanding of the pathology is required. This study affects primarily the first requirement.

This study was initiated in an attempt to increase the knowledge of the functioning of the quadriceps femoris. Although the quadriceps femoris are active during walking (horizontal gait) it is quite possible to walk without obvious gait abnormality with these muscles not contracting even though they may be capable of contracting. This has been exemplified clinically to the author at the Roehampton Amputee Centre in Britain and in his clinical experience and in that of others (Bowie, 1979), particularly in the case of reflex inhibited quadriceps femoris due to knee injury. For this reason an activity in which the quadriceps femoris were essential was chosen. Stair climbing (vertical gait) is an activity in which the quadriceps are essential for either single step or step over step gait.

Perhaps the first attempts at an objective understanding of stair climbing were undertaken by the photographer Eadweard Muybridge (1887) in the middle of the last century. However, in recent times little research seems to have been done in this area. A comprehensive literature review was undertaken and as well, Dr. J.V. Basmajian, an authority in the field of electromyography, was consulted and confirmed the lack of literature (Basmajian, 1979).

As can be seen by the literature review the knee joint and thigh muscles have been extensively researched in many other ways.

The first part of the study was undertaken to establish norms in stair climbing concurrently analysing temporal, electrogoniometric and peak electromyographic data which has not been done before.

The second part of the study was to compare the normal data with the data from Syme's amputees performing the same action. Unilateral Syme's amputees were chosen because of their unique end weight-bearing characteristics, allowing a comparison with the normal with only one foot missing. This limits the number of variables that can produce differences and allows a more sensitive comparison. Also Syme's amputees are generally of a younger age than the majority of higher below knee amputees.

It is expected that this information may be used clinically, in rehabilitative exercises, functional electrical stimulation, and in myoelectric control of prosthesis.

The sequence of the study was as follows: preliminary studies of appropriate electrode placement were made after a knee dissection study was done particularly with reference to the vastus medialis.

A foot switch insole was designed for the capture of the temporal data. After initial trials with various metallic electrodes a commercial design was chosen. These were attached to thin radiographic material and the unit was tested. Finally, the electronic circuitry associated with the switches was miniaturized within the output plug body.

An electrogoniometer that is capable of measuring single plane movement with a varying axis was conceived, designed and tested in conjunction with the Rehabilitation Centre for Children (formerly the Shriners Hospital in Winnipeg).

The electromyography system established previously in the Department of Anatomy was used for the capture of peak EMG activity.

CHAPTER II

REVIEW OF THE LITERATURE

EVOLUTION OF THE KNEE JOINT

It is not the purpose of this thesis to determine the detailed evolution of the knee joint, however, certain developmental stages are worthy of consideration.

The simplest form of knee joint existed in the now extinct tailed amphibia (Sutton, 1884) in which poorly developed femoral condyles articulated with both the tibia and fibula via a pliant fibrovesicular tissue with no joint cavity. The simplest present day vertebrate with a knee joint is the amphibian, for example, the salamander. In the salamander, the femur and single tibiofibular bone articulate via a thin interarticular membrane. No menisci, cruciates or patella are present (Herzmark, 1938). The frog, also in this class, has a similar structure, but the interarticular membrane has two crescent-shaped thickenings. Of the reptiles, the crocodile and alligator have separate tibiae and fibulae which articulate with the femur in a joint cavity. The cruciates, collateral ligaments and menisci are well developed, as is a femoro-fibular disc. There is no patella (Haines, 1942). The simpler mammals, e.g. the opossum, demonstrate further development with a thickening over the joint line to form a rudimentary patella, and the lack of an intra-articular disc between the fibula and femur, although they still articulate. The placentalia, e.g. the cat, demonstrate essentially all the features that exist in man.

Mammalian Knee Joint - Quadrupedal to Bipedal

Sutton (1884), in his theory of multilocular derivatives of the knee joint cavity, postulates that the primitive mammalian knee joint had three separate cavities, one between the femur and the patella, and two between the femur and tibia and fibula. However, it is difficult to be certain what the primitive mammalian state may have been.

As the basic locomotor pattern developed from quadrupedal to bipedal a variety of changes in the lower limb occurred:

1) The muscle masses of the lower limb of the biped became more central rather than posterior.
2) The leg of the biped became a propulsive strut exerting action parallel to its long axis, rather than a propulsive lever exerting action at right angles to the long axis.
3) The centre of gravity moved into line with the long axis of the leg from the central body position in the quadruped.
4) The biped has more selective hip, knee and ankle movement instead of the quadrupedal limitation of hip extension causing knee and ankle (hock) extension.
5) In bipedal locomotion the lower limb acts more as a propulsive strut with the ankle flexors as a retropulsive strut. In quadrupeds, forward thrust of the hind limbs is controlled by the forelimbs.

(Haxton, 1947; Krogman, 1962)

In the early Pleistocene age, approximately half a million years ago, the existence of the Australopithecinae from the fossil evidence in South Africa is well established (Broom, 1950; Le Gros Clark, 1955; Washburn and Howell, 1960). Assuming that the hip and knee muscle-bone relationships and actions to be the same in these man-apes as in modern man, there is no doubt that the Australopithescenae demonstrated the shift from semiquadrupedal to bipedal locomotion (Krogman, 1962).

The origin of bipedalism is obscure. It has not been established yet whether it occurred as a result of a selection of a variable structural form which previously existed, or whether it developed in response to some specific functional demand. Irrespective of its origin, once man established the bipedal gait the variety of changes mentioned occurred in the process of adaptation.

DEVELOPMENT OF THE KNEE JOINT

Prenatal Development

Introduction

The prenatal development of the knee has been studied more than that of any other joint (Gray and Gardner, 1950). Since the 1870's it has been thought that movement, or mechanical factors, rather than genetic factors, determined joint development (Rayher, 1874; Schulin, 1879). The early work of Bernays (1878) on the comparative development of the knee joint in man and other animals, appears to be the first published work arguing against this "Mechanical Theory of Development of Joints", although Bernays did not deny that activity in the later stages of development may well be important in altering the articular form. Most recently, Moore (1977) states that probably as a result of joint movement the mesenchymal cells subsequently disappear from the surface of the articular cartilage.

The prenatal development of the fertilized ovum may be divided into the following stages:

1) First three weeks

2) Embryonic period from the commencement of the fourth week to the end of the eighth week

3) Fetal period from commencement of ninth week to birth (Hamilton et al., 1972)

Development of the knee joint during these stages of development is as follows:

1) FIRST THREE WEEKS. During this period implantation into the uterine wall and the formation of the basic germ layers occurs. No development of the knee joint is present at this stage.

2) EMBRYONIC PERIOD, fourth to eighth week. The leg buds appear by the end of the fourth week between day 28 and day 31, Carnegie stage 13, when the embryo is approximately 4-6 mm in crown-rump length (Streeter, 1945). The base of the limb buds extends over six spinal segments, from the second lumbar to the second sacral, and consists of undifferentiated mesenchymal tissue derived from the somatic mesoderm, covered by a layer of ectoderm.

The inductive influence of the apical ectodermal ridge at the tip of the bud stimulates the growth and development of the limb. This tip becomes flattened into a paddle-shaped foot plate, with digits differentiating at the margin of the plate (Moore, 1977).

The critical period in the development of the knee corresponds to that of the lower limb, i.e., three to six weeks after fertilization, but it is likely to be somewhat shorter in duration (Gardner and O'Rahilly, 1968).

Development of the Skeleton. In the middle of the sixth week definite condensations of the mesenchymal cells are noted. These are the primordia of the skeletal structures. The primordium of the tibia and fibula is a single sheet (McDermott, 1943), but it later separates. Whether or not the fibula enters into the formation of the knee joint has been disputed. Evidence that for a brief period a femoro-fibular articulation exists has been indicated by Henke and Reyher (1874), Bardeen (1905), Keith (1923) and O'Rahilly (1951). The femur, tibia and fibula begin to chondrify at the end of the sixth week (O'Rahilly et al., 1957). The tibial and femoral condyles are most cartilagenous by 7½ weeks (Gardner and O'Rahilly, 1968).

At the end of the embryonic period the femur and tibia begin to ossify (O'Rahilly and Gardner, 1972), preceding fibular ossification (Bardeen, 1905; O'Rahilly and Gardner, 1972).

The mesenchymal condensations of the patella appear in the seventh week (Gardner and O'Rahilly, 1968) and by the end of the embryonic period chondrification is present (Walmsley, 1940).

Development of the Articular Cavities. The appearance of unchondrified mesenchyme between the developing cartilages of a joint has been named the "interzone" after Bernays (1878), and has been regarded as the first morphologic indication of a joint (Haines, 1947). However, although interzones usually occur in the chondrification process, they are not always an indication of joint differentiation. For example, the femoropatellar articulation develops without an interzonal stage (Gardner, 1950), and the femorofibular interzone becomes invaded by tibial tissue and does not become a joint (O'Rahilly, 1951). However, at the end of the embryonic period the interzones of the knee become less dense, and more loosely arranged fibres indicate the future site of cavities. Four small slit-like cavities may occur between the mesenchyme and the tibia and femur around the periphery of the joint. There is evidence of a cavity between the patella and femur and loosely arranged tissue is evident in the popliteal fossa and suprapatellar process (Bardeen, 1905; Gray and Gardner, 1950).

Development of Ligaments and Capsule. By 7½ weeks the mesenchymal condensations of the cruciates appear and become established with strands of collagen present within a week (Gray and Gardner, 1950). The fibular collateral ligament is visible during this time and precedes the development of the tibial collateral ligament (Gardner and O'Rahilly, 1968).

Development of the Menisci. These originate directly from the mesenchyme at the end of the embryonic period (Gardner and O'Rahilly, 1968). They appear as closely packed cells, the long axes of which are transverse to that of the limb.

Development of the Muscle Mass. The muscle mass of the limb is derived from the somatic layer of the lateral plate mesoderm and not from the mesenchyme of the myotome regions of the somites (O'Rahilly, 1967; Moore, 1977). By the seventh week all the muscles of the knee joint can be identified (Bardeen, 1905). The ligamentum patellae is definite by the end of the embryonic period (Gray and Gardner, 1950).

Development of Blood Supply and Nerve Supply. Blood vessels in the peripheral part of the knee may be present at the end of the embryonic period (Gardner et al., 1959; Gardner and O'Rahilly, 1968) but are usually characteristic of the fetal period (McDermott, 1943).

At the end of the embryonic period the lower limb has rotated medially through almost 90 degrees and the knees are directed ventrolaterally. The toes are separated having differentiated from the foot plate and the limb resembles the adult form (Moore, 1977).

3) FETAL PERIOD, ninth week to birth. At the start of this period the legs are short and the thighs relatively small. The soles of the feet face each other and the big toe is cranial in direction (O'Rahilly and Gardner, 1975). After 12 weeks the lower limbs are slightly shorter than their final relative proportions which are attained during weeks 17 to 20 (Moore, 1977). At the start of the fetal period the flexed knee joint clearly resembles the adult in form and arrangement (Langer, 1929; McDermott, 1943; Gardner and O'Rahilly, 1968).

Development of the Skeleton. In the early stages of the fetal period the cartilage model forming the knee joint has a definite limiting membrane, the perichondrium, and the cartilage becomes more mature (McDermott, 1943). As this cartilage model grows, diaphyseal ossification continues until at term both the femoral and tibial diaphyses are completely ossified, and the distal femoral epiphysis and proximal tibial epiphysis have begun ossification (O'Rahilly, 1962).

By the 10th week the patella is cartilagenous and remains so till term (Gray and Gardner, 1950). In the initial development the medial and lateral facets are equal but by the 23rd week the patella has acquired lateral facet predominence (Walmsley, 1940).

Development of Articular Cavities. During the ninth week definite and fairly extensive cavities are present, mainly femoropatellar and femoromeniscal. The tibiomeniscal cavity follows this development. The posterior cavity of the joint is established by the 12th week. By the 14th week the cavities mentioned above have coalesced to form a single cavity but the final extent of the cavity is only established during the remainder of the fetal period (Gray and Gardner, 1950). However, at the 14th week the space is similar in contour and relative size to the adult joint (Flint, 1904).

Development of Ligaments and Capsule. Excluding the quadriceps femoris muscle and the ligamentum patellae anteriorly, which develop in the embryonic period, the fibrous capsule first becomes evident in the ninth week by a midline concentration of collagenous fibres posteriorly. It is established by the 11th week (Gray and Gardner, 1950). The detailed development of the capsule is subject to variation.

The collagenous content of the cruciates increases significantly at the 12th week and three weeks later becomes densely collagenous as the penetration of vessels occurs during the 14th week. The cruciates become more sharply demarcated and increase in size till term (Gray and Gardner, 1950). At the beginning of the fetal period the fibular collateral and tibial collateral ligaments become more evident and between the 14th and 15th weeks resemble the adult structure, the fibular collateral ligament maintaining its earlier lead. The posterior meniscofemoral ligament is evident at 10 weeks and the transverse ligament by 12 weeks (McDermott, 1943).

Development of the Synovial Tissue. The primordium of the synovial tissue becomes established after cavitation occurs at the end of the embryonic period. It lines the articular cartilage and the menisci. However, the primordium lining the menisci does not have the vascularity associated with synovial tissue and this difference persists and becomes more marked during later fetal months. The synovial villi appear in the 29th-30th week (McDermott, 1943). The appearance of the synovial tissue suggests that it secretes before term (Gray and Gardner, 1950).

Development of Menisci. In the early fetal period the menisci are well defined, first containing numerous cells and then strands of collagen, but avascular. In the 14th week the menisci are of the same distribution and configuration as in the adult form (McDermott, 1943). As collagen deposition increases the menisci become more prominent and by the 15th week there is an increase in the penetration of blood vessels and nerves in the periphery. By the 23rd week there is a clear distinction between the vascularity of the synovial tissue and the avascular collagenous part of the menisci. A few vessels, however, do reach the inner borders of the menisci. From the 33rd week until term, there is a reduction in the number of vessels penetrating the menisci (Gray and Gardner, 1950).

Development of the Muscle Mass. The muscle mass being established in the embryonic period, with the rotation of the limbs the extensor muscles move from the dorsal to the ventral aspect (Moore, 1977).

Development of Blood Supply and Nerve Supply. Although blood vessels appear before the formation of articular cavities, and are present in the embryonic period, the general adult pattern is present after 11 weeks. Arteries, arterioles, capillaries, venules and veins which supply the joint are present (Gray and Gardner, 1950).

The nerves accompany the blood vessels and by the 12th week the articular innervation of the adult is present (Gardner, 1948).

Postnatal Development

From birth onward the changes in the knee joint are primarily those of growth and ossification. The menisci, however, become more avascular and increase their collagen content within the first two to three years. At about 2½ years of age the synovial membrane has completely disappeared from the weight-bearing areas of the menisci (McDermott, 1943).

The patella becomes ossified from several centres (McKenzie and Naylor, 1957) and this ossification commences at three years in females and a year later in males. Ossification is usually complete about the age of puberty (Meschan, 1975).

As mentioned previously, the femur is the first long bone to demonstrate traces of ossification in the late embryonic period with secondary centres of ossification occurring in the distal end of the femur in the last fetal month. The last femoral epiphysis to fuse is the distal epiphyseal plate which occurs at 16 years of age in females and two years later in males. Complete union may take until the 20th year of age (Gray, 1973). Tibial ossification also begins in the late embryonic period and the centre for the ossification of the proximal end of the tibia is present at birth (O'Rahilly, 1962).

At about 10 years of age the cartilagenous proximal end has another ossification centre in the region of the tibial tubercle, and by 12 years of age this has joined with the major proximal centre of ossification.

The last tibial epiphysis to fuse is the proximal epiphyseal plate and this follows the same pattern as the distal end of the femur (Gray, 1973).

KNEE JOINT ARTICULATIONS

Introduction

The knee joint consists of the articulation of three bones: the femur, tibia and patella. The combination of articulation is:

1) Patello-femoral

2) Tibio-femoral

The knee joint is classified as a synovial joint.

The knee possesses more than one pair of articulating surfaces and because of this is described as a compound joint. The intracapsular menisci allow the joint to be also termed complex (Gray, 1973).

Within the synovial classification the knee joint has been variously described as:

1) Two condylar joints (between the corresponding femoral and tibial condyles) and a sellar joint (between the patella and femur) (Gray, 1973).

2) A modified hinge joint (Steindler, 1955; Cunningham, 1972; Basmajian, 1976).

3) "Generally considered to be a strict hinge joint" (Lewin, 1952).

4) A condylar joint (Gardner et al., 1975).

5) "Mechanically undisputably of the hinge variety" (Kapandji, 1970).

6) A double condyloid joint (between femur and tibia) and a gliding joint (between patella and femur)(Lockhart et al., 1974).

7) Diarthrodial hinge (Grant, 1975; Langebartel, 1977; Jacob et al., 1978).

Articulating Surfaces

Femur

The distal end of the femur expands widely to accommodate two articular condyles which are continuous anteriorly and separated by an intercondylar fossa posteriorly. The condyles are covered by articular cartilage. Viewed anteriorly (Fig. 1), the articular surface seen is exclusively that of the patella, the lateral condyle extending higher than the medial. The higher lateral area of the lateral facet is called the supratrochear tubercle, the site of patella pressure with the knee in full extension (Ficat and Hungerford, 1977). The anterior surface is also called the "patellar groove", patellar facets, or trochlea. The inferior border of this articulating surface is marked by two grooves (Fig. 2, inferior view) the lateral being more distinct. In full knee extension, the anterior borders of both menisci are at the level of the ridges.

The thickness of the articular cartilage is 2-3 mm, thicker on the lateral surface (Ficat and Hungerford, 1977).

The tibial surfaces of the femur are convex in both mediolateral and anteroposterior dimensions. The anteroposterior curvature (Fig. 3, viewed laterally) is not uniform, being sharper posteriorly than anteriorly. The medial articular surface is longer anteriorly which is significant in relation to the rotation that occurs as the knee comes to full extension (Gray, 1973).

FIGURE 1.

The epiphysial line is shown by the dotted line.

The capsule attachment is shown by the interrupted line.

A) LEFT FEMUR--Anterior view of lower end.

B) LEFT FEMUR--Posterior view of lower end.

1) Articularis genu
2) Fibular collateral ligament
3) Tibial collateral ligament
4) Vastus medialis
5) Adductor magnus
6) Medial head--gastrocnemius
7) Posterior cruciate ligament
8) Anterior cruciate ligament
9) Lateral head--gastrocnemius
10) Plantaris
11) Short head of biceps
12) Vastus intermedius

C) LEFT TIBIA--Anterior view of upper end.

D) LEFT TIBIA--Posterior view of upper end.

1) Iliotibial tract
2) Tibialis anterior
3) Patellar ligament
4) Sartorius
5) Gracillis
6) Semitendinosus
7) Tibial collateral ligament
8) Semimembranosus
9) Vastus medialis
10) Popliteus
11) Soleus
12) Flexor digitorum longus
13) Tibialis posterior
14) Posterior cruciate ligament

(from McMinn and Hutchings, 1977)

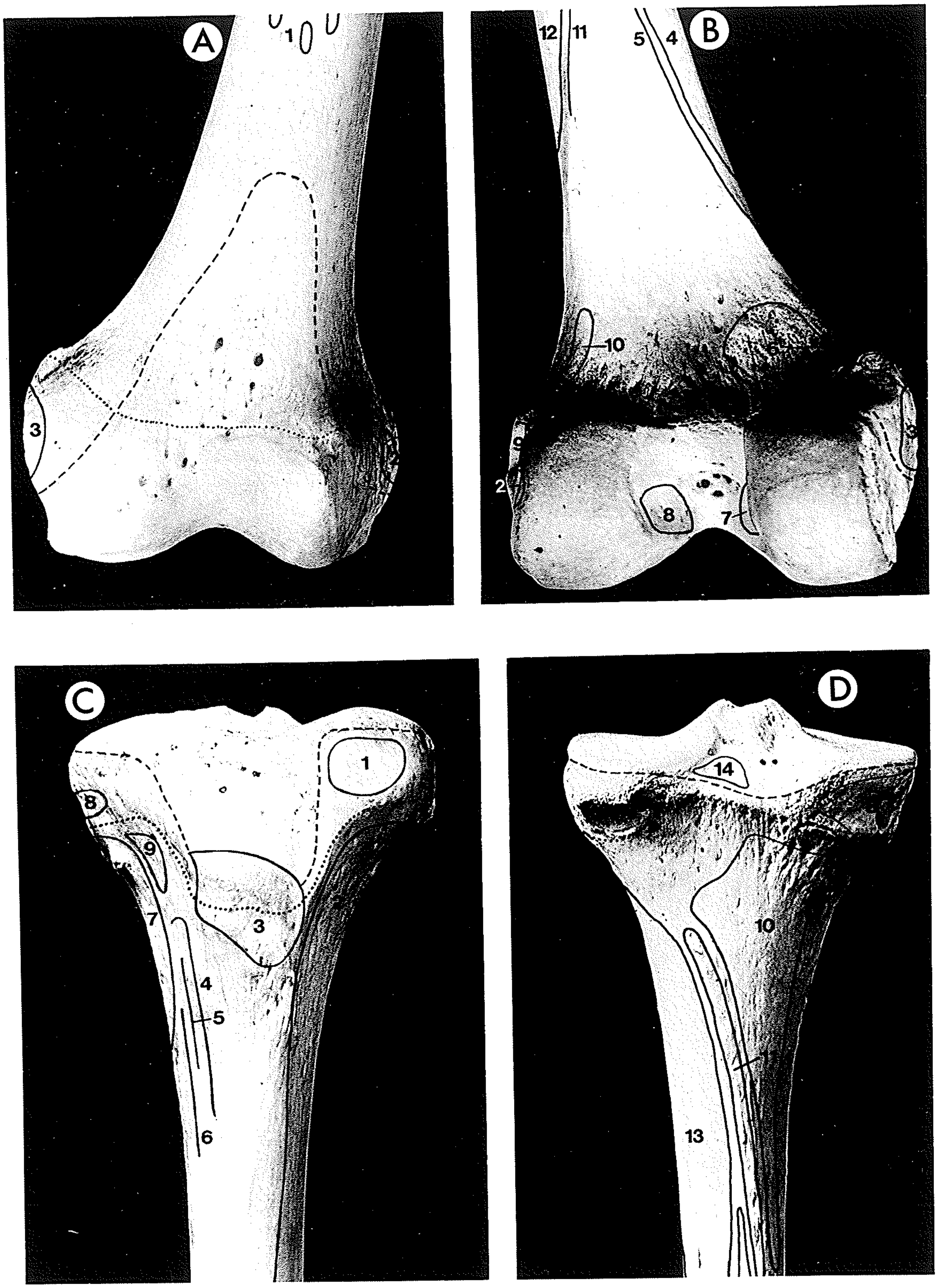

A
1
3
12
11
5
4
B
10
9
2
8
7
C
1
8
9
3
7
4
5
6
D
14
10
13

FIGURE 2.

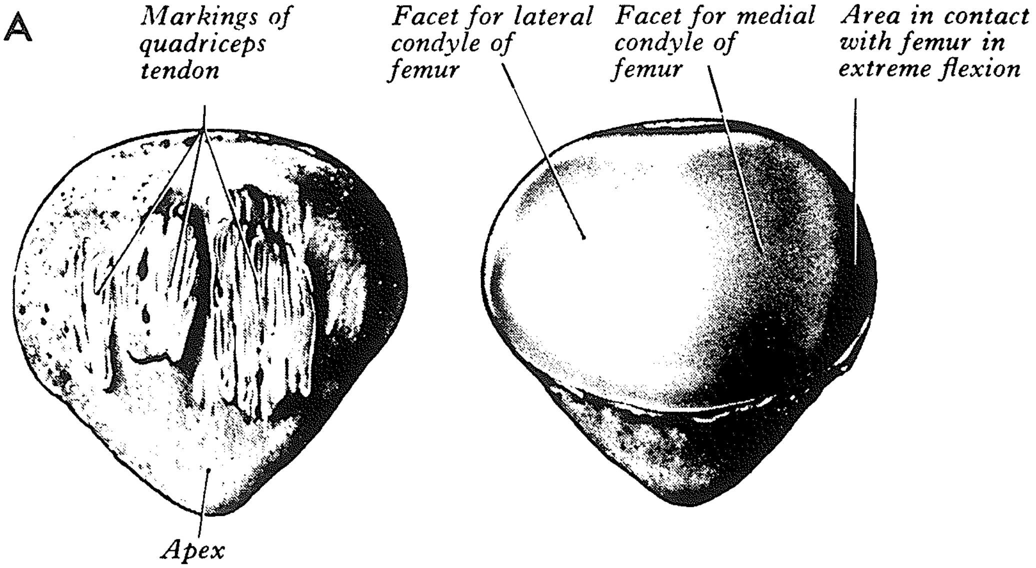
A) PATELLA--Anterior and posterior views. Odd facet is on the extreme right.

(from Gray, 1973)

B) LEFT FEMUR--Distal view.

(from McMinn and Hutchings, 1977)

A


Markings of quadriceps tendon
Facet for lateral condyle of femur
Facet for medial condyle of femur
Area in contact with femur in extreme flexion
Apex

B

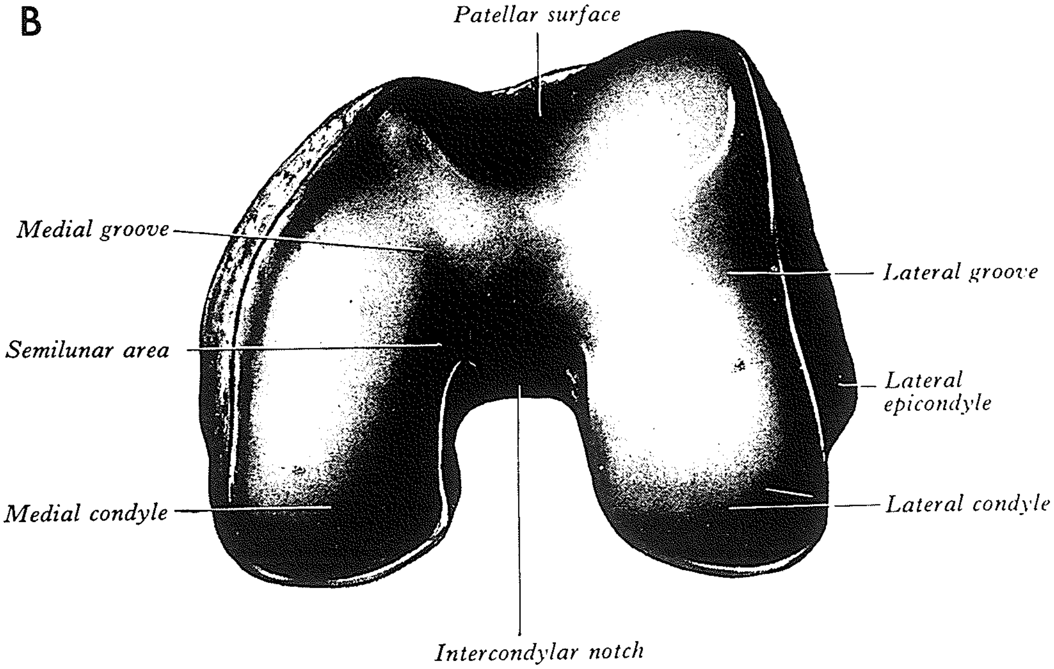

Patellar surface
Medial groove
Lateral groove
Semilunar area
Lateral epicondyle
Medial condyle
Lateral condyle
Intercondylar notch

FIGURE 3.

A) LEFT FEMUR--Medial view of lower end.

B) LEFT FEMUR--Lateral view of lower end.

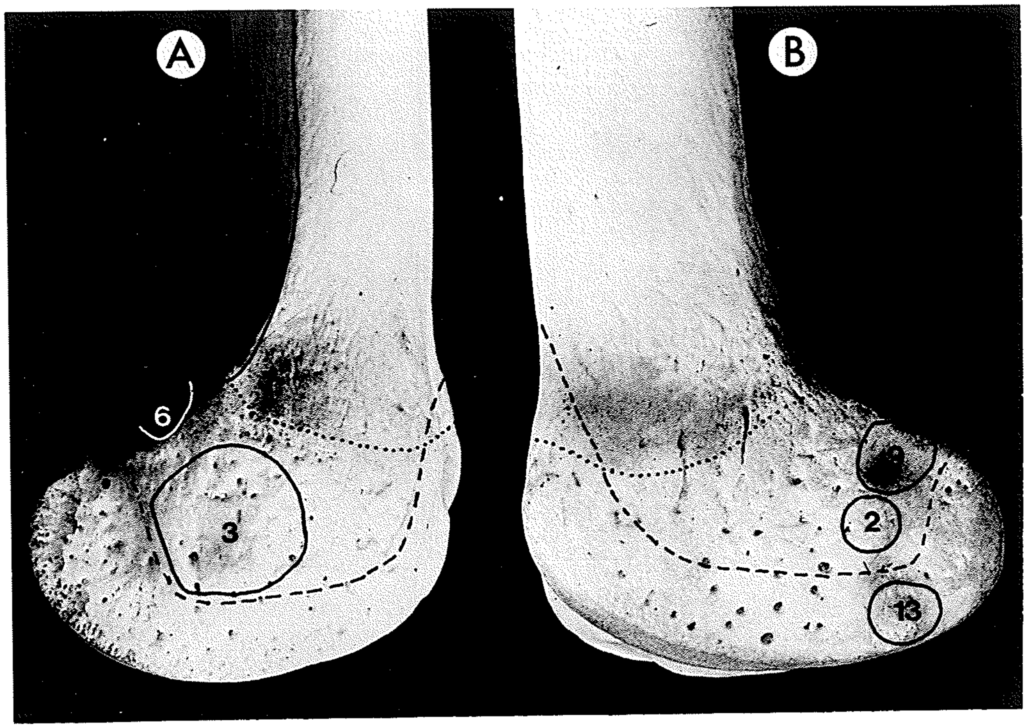
1) Articularis genu
2) Fibular collateral ligament
3) Tibial collateral ligament
4) Vastus medialis
5) Adductor magnus
6) Medial head--gastrocnemius
7) Posterior cruciate ligament
8) Anterior cruciate ligament
9) Lateral head--gastrocnemius
10) Plantaris
11) Short head of biceps
12) Vastus intermedius

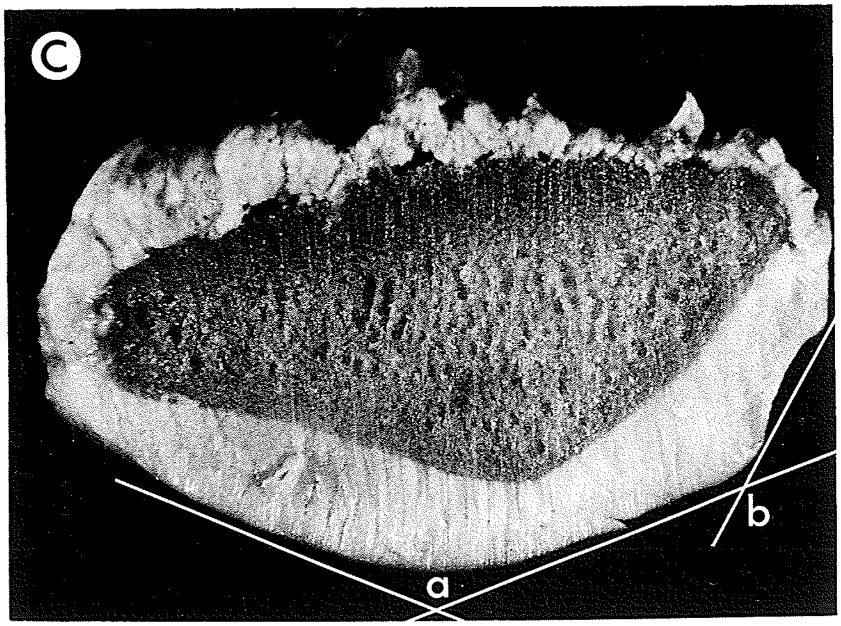
(from McMinn and Hutchings, 1977).

C) PATELLA --Cross section through the middle of the articular surface to demonstrate:

i) The variation in cartilage thickness and subchondral configuration.
ii) Angle (a) made between the medial and lateral facets.
iii) Angle (b) made between the medial and odd facets.

(from Ficat and Hungerford, 1977)


A
6
3
B
2
13


C
b
a

Tibia

The expanded proximal end of the tibia contains two flattened articular condyles for the corresponding femoral condyles, separated by a non-articular intercondylar area (Fig. 4). This superior articulating surface is tilted backwards in relationship to the long axis of the tibial shaft. This tilt is greatest at birth and decreases with age (Kate and Robert, 1965).

The oval medial articulating surface has its longer axis antero-posteriorly, corresponding to the longer medial femoral condyle. The area in contact with the medial meniscus is flattened and widest posteriorly, while the central area not related to the meniscus is concave. The lateral aspect of the medial intercondylar tubercle is covered with articular cartilage and the peripheral margins of the articular cartilage are sharp.

The circular lateral articular surface (Fig. 4, superior view) forms a corresponding similar pattern regarding the area of contact and the lateral intercondylar tubercle but the posterior edge of the articulating surface differs from the sharp margin of the medial surface as its smooth and rounded surface accommodates the tendon of popliteus which is its immediate relation (Gray, 1973).

The tibial articular cartilage ranges from 1.8 to 3.3 mm, being thicker in the contact area of the slight flexion range (0-30°) than in the full flexion range (60-120°) (Walker and Hajek, 1972).

FIGURE 4.

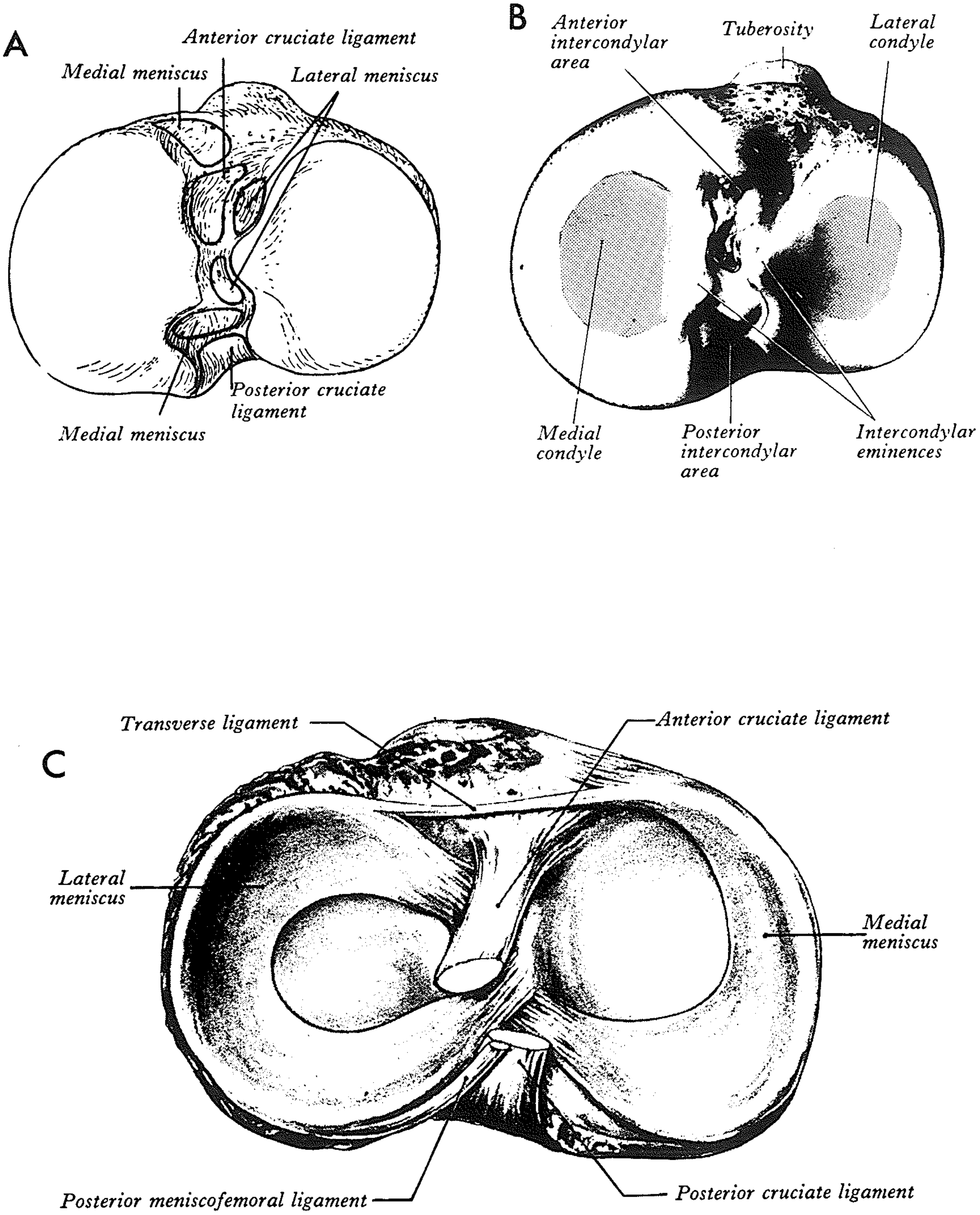
TIBIA--Superior views showing the:

A) Relationships of attachments of the non-articulated intercondylar region.

B) Direct contact areas of the medial and lateral condyles.

C) The menisci and associated ligaments.

(from Gray, 1973)


A
Anterior cruciate ligament
Medial meniscus
Lateral meniscus
Posterior cruciate ligament
Medial meniscus
B
Anterior intercondylar area
Tuberosity
Lateral condyle
Medial condyle
Posterior intercondylar area
Intercondylar eminences
C
Transverse ligament
Anterior cruciate ligament
Lateral meniscus
Medial meniscus
Posterior meniscofemoral ligament
Posterior cruciate ligament

Patella

This is the largest sesamoid bone. It has the thickest articular cartilage in the human body on its posterior articulating surface, being 4-5 mm thick (Ficat and Hungerford, 1977). The triangular bone, with its apex distal, is slightly wider than longer (De Vriese, 1913), and the variations in this are very slight (Vallois, 1914). The articular surface covers 75% of the posterior surface and is divided into a medial facet and a generally larger lateral facet by a vertical ridge (Fig.

The flat or slightly concave medial facet is divided by a vertical ridge at the upper medial margin and this extra facet is called the "odd" facet (Goodfellow et al., 1976A). This facet is highly variable, being nearly in the same plane as the medial facet or at as much as a 60° angle to it (Ficat and Hungerford, 1977)(Fig.

The ridge produced by the odd and medial facets may be purely cartilagenous and not indicated on the subchondral bone (Fig. 3). The odd facet is in contact with the femur in full flexion of the knee.

The lateral facet is larger in both dimensions and concave in both horizontal and vertical planes. It lies nearest the coronal plane of the body with the medial and its odd facet showing progressive obliquity.

The presence of one or two transverse ridges producing a corresponding two or three pairs of articulating areas is still not established. De Palma (1954), Cunningham (1972), Watanabe (1974), Lockhart et al. (1974), Grant (1975), Basmajian (1976) and Grant (1978) described two ridges and therefore three pairs of surfaces; while Emery and Meachim (1973), Ficat, C. (1974), Ficat, R.P. and Hungerford (1977), and McMinn and Hutchings (1977) are more inclined to the single ridge concept between the middle and lower thirds of the articulating surfaces. Gray (1973) states that there may be two faint horizontal ridges...(but) in many patellae only one horizontal ridge can be distinguished.

Generally there is only a single patella but there have been reported accessory patellae (Wright, 1903) and absence of patella (Rubin, 1915; Bell, 1955).

Function of the Patella. A variety of functions has been ascribed to the patella.

Steindler (1955) states that the patella increases the force of extension by as much as 50%.

The patella considerably decreases the friction of the extensor mechanism, because of the very low coefficient of friction of hyaline cartilage. Tendons are capable of withstanding great tensile loads but not high friction or compression, and the presence therefore of the patella in the extensor apparatus protects the quadriceps tendon from friction, and permits the extensor apparatus to tolerate high compression loads (Ficat and Hungerford, 1977).

It also corrects the lateral pull of the quadriceps during extension and centralizes the divergent contraction from the four muscles of the quadriceps (Steindler, 1955; Hungerford and Ficat, 1977).

In addition, it serves to protect the articular cartilage of the femur (Grant, 1975).

With regard to mechanical aid to extension, there appear to be significant differences of opinion. Some authors believe that the efficiency of the extensor action of the quadriceps is not improved with the patella, unimpaired without it, or that it acts only in an accessory or insignificant role (Brooke, 1937; Hey Groves, 1937; Watson-Jones, 1945;

Freehafer, 1962). Some long-term clinical studies of patellectomy support this view (Young and Regan, 1945; Boucher, 1958; West, 1962; Geckeler and Quaranta, 1962). On the other hand, some feel that the patella is an important functional component of the extensor mechanism (Steindler, 1955; McKeever, 1955; De Palma and Flynn, 1958; Smillie, 1970). This view is supported by clinical studies of Scott (1949), Todd (1950), and Insall et al. (1976A) and by experimental work (Haxton, 1945, Kaufer, 1971).

Sesamoid Bones of the Knee Joint

The patella and fabella are the sesamoid bones of the knee. The fabella (little bean) is in the lateral head of the gastrocnemius muscle and may be considered, like the patella, to be articular in classification as it articulates with the lateral femoral condyle (Gray, 1973). It occurs in 11-13% of the population and in over 50% of cases it is bilateral (Pancoast, 1909; Sutro et al., 1935; Falk, 1963).

Capsule

This is a complex, imprecisely defined fibrous structure (Warren and Marshall, 1979). It is deficient in parts and reinforced in others by expansion of various tendons. Its osseous attachments are demonstrated in Figures 1 and 3. Anteriorly the patella and the insertion of the quadriceps exclude the capsule formation. Laterally the fibres extend distally to the head of the fibula. Medially the capsule blends with the posterior part of the tibial collateral ligament and may be regarded as its deep component (Last, 1948). Posteriorly it blends with the head of gastrocnemius and is strengthened centrally by the oblique popliteal ligament (Gray, 1973).

Synovial Membrane

The synovium of the knee joint is the most extensive and complex of the body (Gray, 1973). Superiorly it forms a large suprapatellar bursa or pouch under cover of the quadriceps femoris muscle. It extends upward 4-5 cms from the proximal articular border of the femur but this is variable (Ficat and Hungerford, 1977).

Inferiorly it covers the infrapatellar fat pad and is continuous with the peripatellar folds of synovia on either side of the patella. Certain folds may be significant enough to simulate knee derangement (Pipkin, 1971). The synovium covering the cruciates is called the infrapatellar fold or ligamentum mucosum. This bell-shaped "ligament" has medial and lateral alar folds which almost divide the joint into medial and lateral halves (Steindler, 1955). Posteriorly the synovium is not a continuous sheet but indented by the structures of the intercondylar region which are extrasynovial but intracapsular.

Bursae of the Knee Joint

The bursae of the knee joint are numerous. They are not all consistent. Some communicate with the knee joint (Brantigan and Voshell, 1943; Gray, 1973).

They are: Anteriorly
1) Subcutaneous prepatellar bursa
2) Deep infrapatellar bursa
3) Subcutaneous infrapatellar bursa
4) Suprapatellar bursa (or pouch)

Laterally between
1) Lateral head of gastrocnemius and capsule
2) Fibular collateral ligament and biceps
3) Fibular collateral ligament and popliteus

4) Tendon of popliteus and lateral femoral condyle

Medially between 1) Medial head of gastrocnemius and capsule

2) Tibial collateral ligament and muscles of pes anserinus.

3) Tibial collateral ligament and femur, or capsule, or medial meniscus, or tibia, or tendon of semimembranosus. (variable in number)

4) Semimembranosus and capsule (semimembranous bursa)

5) Tendon of semimembranosus and semitendinosus

Ligaments

Introduction

Although there is no published classification of the ligaments of the knee into major and accessory groups, for descriptive purposes the ligaments are divided into the two groups listed.

GROUPS	MAJOR	ACCESSORY
	Anterior Cruciate Ligament	Arcuate Popliteal Ligament
	Posterior Cruciate Ligament	Transverse Ligament
	Tibial Collateral Ligament	Anterior and Posterior Meniscofemoral Ligaments
	Fibular Collateral Ligament	Tibiomensical Ligament
	Oblique Popliteal Ligament	Patellofemoral Ligament
	Ligamentum Patellae	Ligament of Barkow

An anatomical description is followed by functional analyses. It will be evident that unanimity of function in this area is often not established.

Since ligaments need stress to indicate their function, the measure of their tautness or tension during specific motions or static positions indicates their function in that motion or position. All the ligament studies are *in vitro* studies and the vast majority are qualitative, subjective and uncontrolled. The methodology has varied but generally severence of specific ligamentous structures and subsequent assessment has been the procedure. In excess of 300 studies since 1916 have used this methodology to study knee ligaments (Nicholas, 1977). The limitations of this method are evident but an understanding on this level in well controlled studies, will form the basis for the understanding of the complexities of the structures of the *in vivo* knee (Mains et al., 1977).

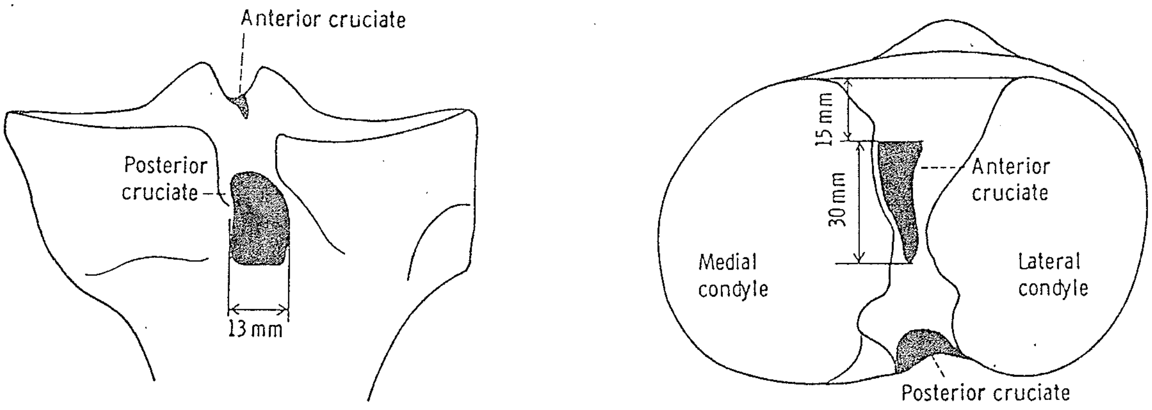
ANTERIOR CRUCIATE LIGAMENT

The anterior cruciate ligament is a complicated two-part ligament named because of its distal anterior attachment in the intercondylar area.

Its two parts are the anteromedial and posterolateral (Abbott et al., 1944; Girgis et al., 1975).

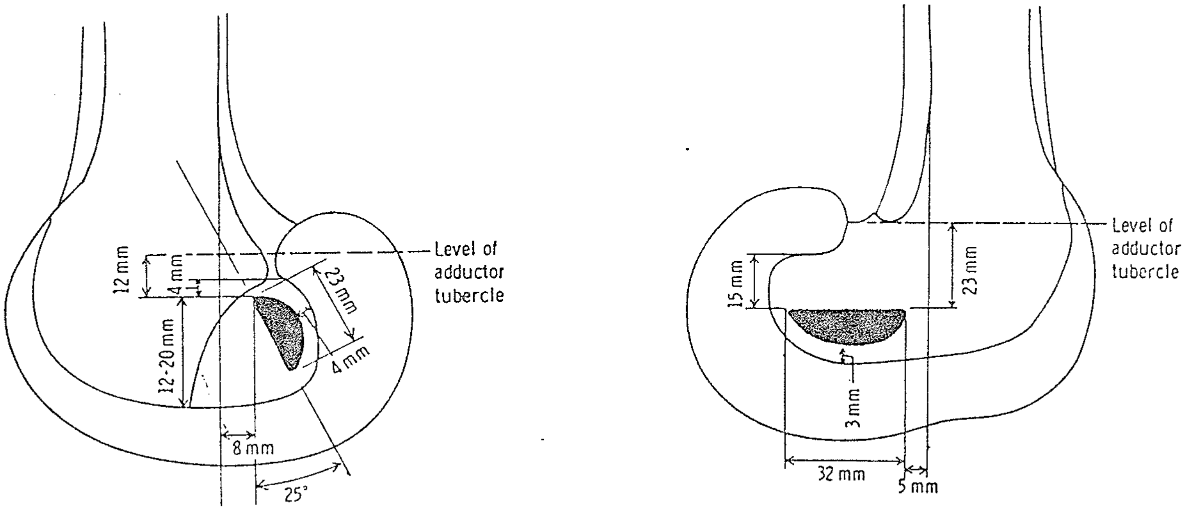
The proximal attachment of this fan-shaped ligament is to the posterior part of the medial surface of the lateral femoral condyle in the form of a segment of a circle (Last, 1951). The size and relationship is shown in Figure 5.

FIGURE 5. AVERAGE MEASUREMENTS AND RELATIONS OF THE ANTERIOR AND POSTERIOR CRUCIATE LIGAMENTS.



A. POSTERIOR SURFACE OF THE TIBIA (PROXIMAL)

B. SUPERIOR SURFACE OF THE TIBIA



C. MEDIAL SURFACE OF RIGHT LATERAL FEMORAL CONDYLE SHOWING ANTERIOR CRUCIATE ATTACHMENT

D. LATERAL SURFACE OF RIGHT MEDIAL CONDYLE SHOWING POSTERIOR CRUCIATE ATTACHMENT

(From Girgis et al. 1975)

The ligament is 38 mm long by 11 mm wide (Morris, 1933; Girgis et al., 1975).

The function of this ligament is described variously as being the keystone of stability by some and at the other extreme almost a vestige (Herzmark, 1938). There is no doubt that the functional role of this ligament has yet to be established and there appear to be many interpretations of function primarily by clinical observation of pathological cases (Hey Groves, 1920; Cowan, 1965; Lam, 1968; Liljedahl et al., 1966).

Flexion and Extension, Fig.5(1). In some studies the anterior cruciate is taut in extension (Bugnion, 1892; Poirer and Charpy, 1911; Brantigan and Voshell, 1941; Abbott et al., 1944; Kennedy et al., 1974; Girgis et al., 1975) and it is generally agreed that it controls hyperextension (Helfet, 1959; Slocum and Larson, 1968; Hughston et al., 1976A; Furman et al., 1976). Kapandji (1970), however, feels that the primary controller of extension is the posterior aspect of the capsule.

In flexion some feel that the anterior cruciate becomes lax (Abbott et al., 1944; Slocum and Larson, 1968; Hughston, 1969; Hughston et al., 1976) and some state part of it remains taut (Kennedy et al., 1974; Girgis et al., 1975). Others feel it is taut throughout flexion and extension (Brantigan and Voshell, 1941; Haines, 1941; Gray, 1973), and finally, taut in flexion and extension but slack in semiflexion (Corner, 1915; Kennedy, 1978).

Some of the current workers in this field are still in disagreement regarding the tautness and indeed the existence of the anteromedial band (Hughston et al., 1976A; Furman et al., 1976).

As the function of this ligament is not as yet defined, the significance of a major clinical test, the drawer sign and its variations, is still questioned and much literature has been published on this topic (Abbott et al., 1944; O'Donoghue, 1950; Hughston, 1962; Robichon and Romero, 1968; Slocum and Larson, 1968; Kennedy and Fowler, 1971; Wang and Walker, 1974; Shaw and Murray, 1974; McMaster et al., 1974; Marshal et al., 1975; Furman et al., 1976; Hughston et al., 1976). Much conflicting evidence exists from one extreme, that the drawer sign may be positive when the ligament is intact (Markolof et al., 1976) and negative when this ligament is severed (Reynolds, 1967; James, 1978). This is mentioned because of the major clinical implications.

The two-part ligament concept is shown in Figure 5(1) and although there may be a controversy regarding the anteromedial band it can be considered that the primary stability part of the ligament is the posterolateral component (Lembo et al., 1975).

Rotation. Because of the evident twisting of this ligament with the posterior cruciate ligament, its function as a rotatory stabilizer has been postulated both in flexion and extension, and again disagreement exists as to whether this occurs more in medial rotation of the tibia (Robichon and Romero, 1968; Kapandji, 1970; Kennedy et al., 1974; Girgis et al., 1975; Mains et al., 1977) or lateral rotation (Cailliet, 1973). Furman et al. (1976) indicated that both medial and lateral rotation were increased significantly on sectioning the anterocruciate; but Wang and Walker (1974) state that although the cruciates (anterior and posterior) come into tension, the lever arm about the centre of rotation is not sufficient to resist rotatory torque. However, Shaw and Murray (1974) state that the cruciates stabilize but do not control rotation.

Strength and Age Changes. The various strength parameters (linear load-first sign of failure, maximum load) of this ligament are dependent on the different testing procedures. For example, because of the different strain rates, knee angle position and method of gripping the specimen, the comparison between studies may be invalid (Noyes et al., 1974).

However, Kennedy et al. (1976) reported an ultimate tensile strength of 626 newtons and Trent et al. (1976) reported a range from 285 to 1718 newtons. Noyes and Grood (1976) noted that the strength parameters significantly reduced with increasing age up to 48 years of age. They also noted that the failure of the ligament in their younger human group occurred in the substance of the ligament (at 1730 newtons) while in the older human group the failure was due primarily to bony avulsion (at 734 newtons).

POSTERIOR CRUCIATE LIGAMENT

This two-part ligament has its distal attachment posterior to the anterior cruciate ligament, extending on to the posterior surface of the tibia (see Figure 5) and it is also attached to the posterior horn of the lateral meniscus.

The proximal attachment is to the lateral surface of the medial condyle of the femur in the form of a segment of circle. The size and relationship of the attachment is seen in Figure 5.

The proximal and distal attachments of this ligament are at right angles to each other [Fig. 5(1)].

FIGURE 5(1).

A) Anterior cruciate ligament in extension.

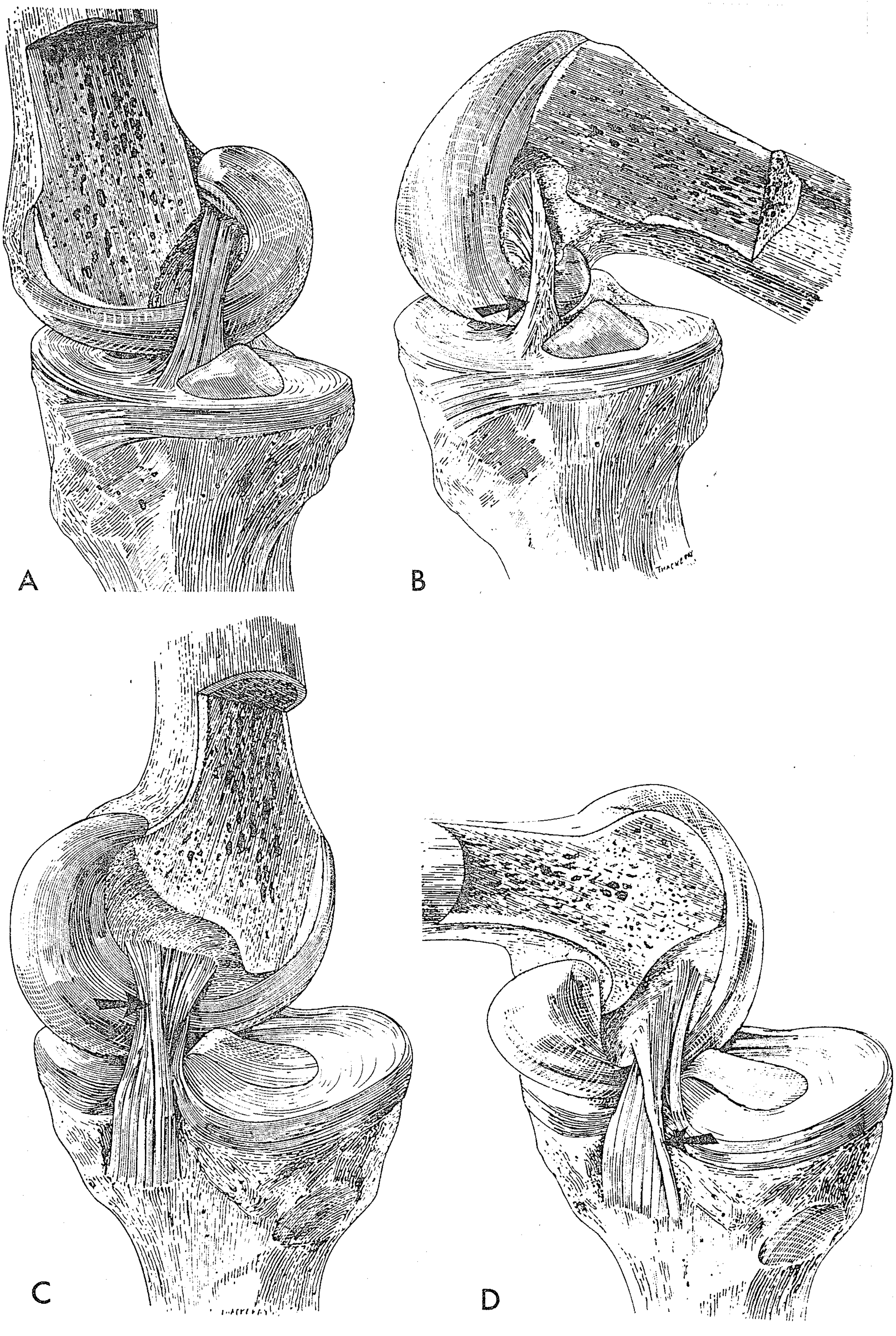
B) Anterior cruciate ligament in flexion. Arrow points out the anteromedial band.

C) Posterior cruciate ligament in extension.

D) Posterior cruciate ligament in flexion. Arrow points out the smaller posterior part, note the proximal and distal attachments are at right angles to each other.

(from Girgis et al., 1975)

Fig.5(1)



The ligament consists of a larger anterior and a smaller posterior part which are fused. The average length is 38 mm and the width 13 mm (Girgis et al., 1975).

Flexion and Extension, Fig.5(1). There is general agreement that in flexion the larger anterior part is taut and the posterior part is lax. The opposite is true in extension (Hey Groves, 1920; Brantigan and Voshell, 1941; Haines, 1941; Abbott et al., 1944; Gray, 1973; Girgis et al, 1975).

Hughston et al. (1976) offer an explanation of the constantly taut posterior cruciate ligament in that the ligament appears to be located in the centre of the joint and functions as the axis about which flexion/extension and rotation occurs. It is therefore considered by them as a fundamental stabilizer of the knee.

Rotation. The statements regarding the cruciates and rotation are made in the anterior cruciate ligament section. In addition, it controls medial rotation (Hughston, et al., 1976).

Strength. This ligament is considered to be the stronger and shorter of the cruciates (Gray, 1973).

There is both direct and indirect evidence for the strength contention. Kennedy et al. (1976) demonstrated that the yield point (when permanent deformation of the ligament occurs) of the posterior cruciate was nearly twice that of the anterior cruciate. Morrison (1968, 1969, 1970) also indicated that the forces on the posterior cruciate ligament are greater during level walking, ramp walking and stair climbing. Although there was a numerical variation Grood and Noyes (1976) were also in agreement.

The statement on ligament length, however, is not borne out by Girgis et al. (1975) though the average width was greater.

In summary, despite the complex architecture of the cruciate ligaments and the variation in the reports which have been described, there definitely appears to be some reciprocity between the anterior and posterior cruciate ligaments in flexion and extension so that some elements do remain taut in all positions and this provides stability (James, 1978).

TIBIAL COLLATERAL LIGAMENT

This ligament is also known as the medial collateral ligament. The ligament consists of a superficial and deep part. The superficial part consists of parallel or long fibres which run vertically, and an oblique part (Warren and Marshall, 1979). Numerous terms have been used for each part. The superficial part has been called variously the medial collateral ligament, superficial medial collateral ligament, the superficial medial ligament (Warren and Marshall, 1979), and the internal lateral ligament (Gray, 1973A). The superior oblique fibres are also called the posterior oblique ligament (Hughston and Eiliers, 1973; James, 1978).

The deep part has been called variously, the middle or medial capsular ligament (Slocum and Larson, 1968; Slocum et al., 1974), the short internal lateral ligament (Last, 1948) and the meniscofemoral and meniscotibial ligaments (Warren et al., 1974).

Superficial Fibres

The superficial part is triangular or delta-shaped and may be separated from the deeper part by one or more bursae (Brantigan and

Voshell, 1943). It extends from the medial epicondyle of the femur, distally to the medial tibial condyle and shaft. Its apex is located posteriorly on the joint line and its extent and even its existence is varied (Brantigan and Voshell, 1943; Warren and Marshall, 1979).

The superficial ligament is approximately 11 cm long by 1.5 cm wide (Brantigan and Voshell, 1943).

Superficial Fibres Function

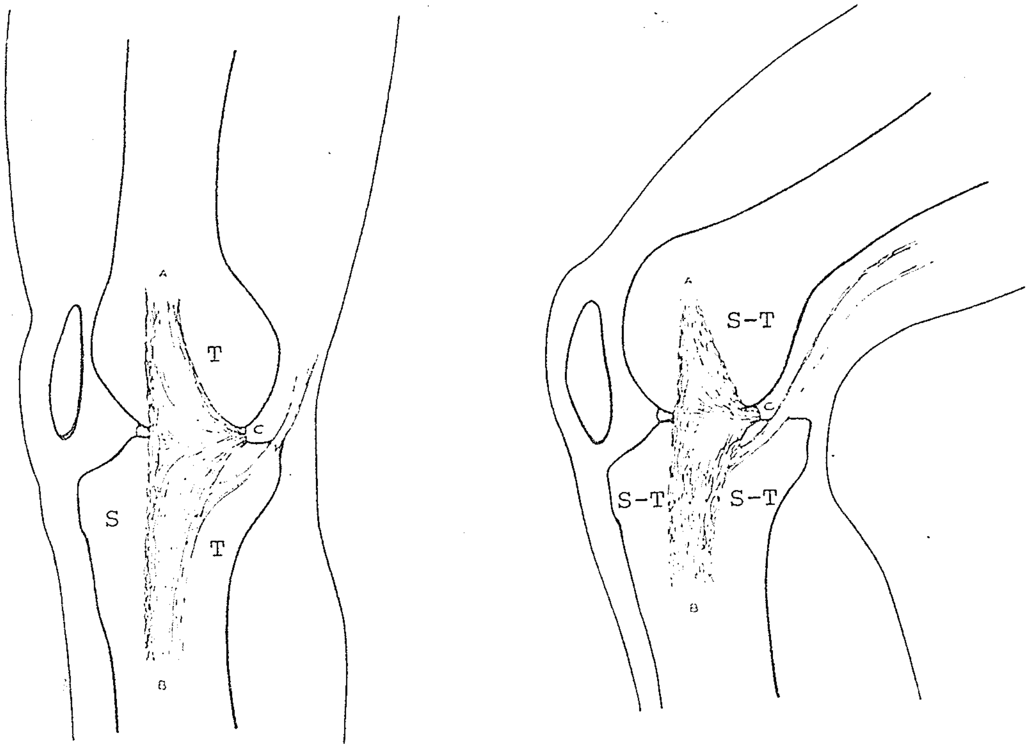
These fibres have different functions.

Flexion and Extension. Robichon and Romero (1968) state that the long anterior fibres are relaxed in extension and taut in full flexion and the posterior oblique fibres have the opposite function; in semi-flexion the ligament is relaxed (Fig. 6).

This general pattern of increased tension with flexion in the anterior component is agreed upon by Brantigan and Voshell (1941, 1943), Abbott et al. (1944), and Steindler (1955). However, Last (1948, 1972), Slocum and Larson (1968) and Smillie (1970) state that the long anterior fibres are relaxed in flexion. The long anterior fibres have been considered as a single homogenous unit (White and Raphael, 1972) but Warren et al. (1974) have distinguished anterior and posterior components which are tight in flexion and extension respectively because of their origin from a critical point on the medial femoral condyle that is relative to the instant centres of rotation (Fig. 7).

Rotation. The control of lateral rotation has been ascribed to the superficial part of this ligament (Brantigan and Voshell, 1941, 1943, Hallen and Lindahl, 1965) and its importance confirmed by Warren et al. (1974), Markoloff et al. (1976), and Hsieh and Walker (1976).

FIGURE 6. TENSION IN THE SUPERFICIAL TIBIAL COLLATERAL LIGAMENT FROM EXTENSION TO FLEXION OF THE KNEE.



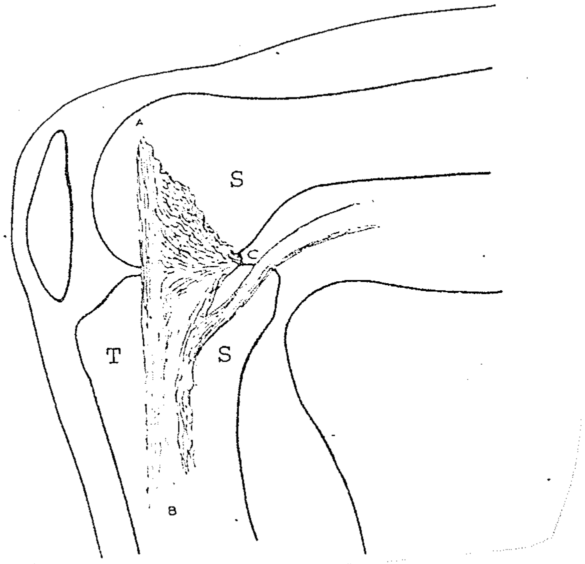
A. EXTENSION

Anterior Fibres Slack

Posterior Fibres Taut

B. SEMIFLEXION

All Same Tension (Semi-taut)



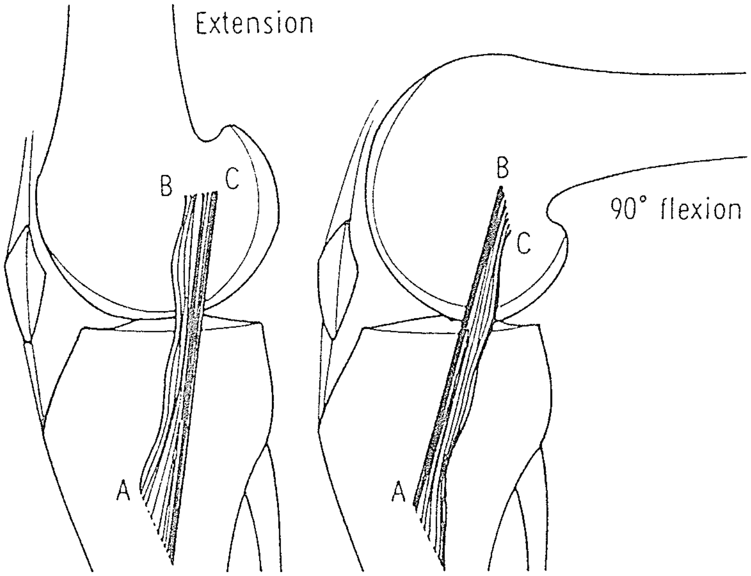
C. FULL FLEXION

Anterior Fibres Taut

Posterior Fibres Slack

(From Robichon and Romero 1968)

FIGURE 7. TENSION OF THE LONG FIBRES OF THE SUPERFICIAL TIBIAL COLLATERAL LIGAMENT IN EXTENSION AND FLEXION OF THE KNEE.



EXTENSION

Anterior Fibres Slack

Posterior Fibres Taut

FLEXION

Anterior Fibres Taut

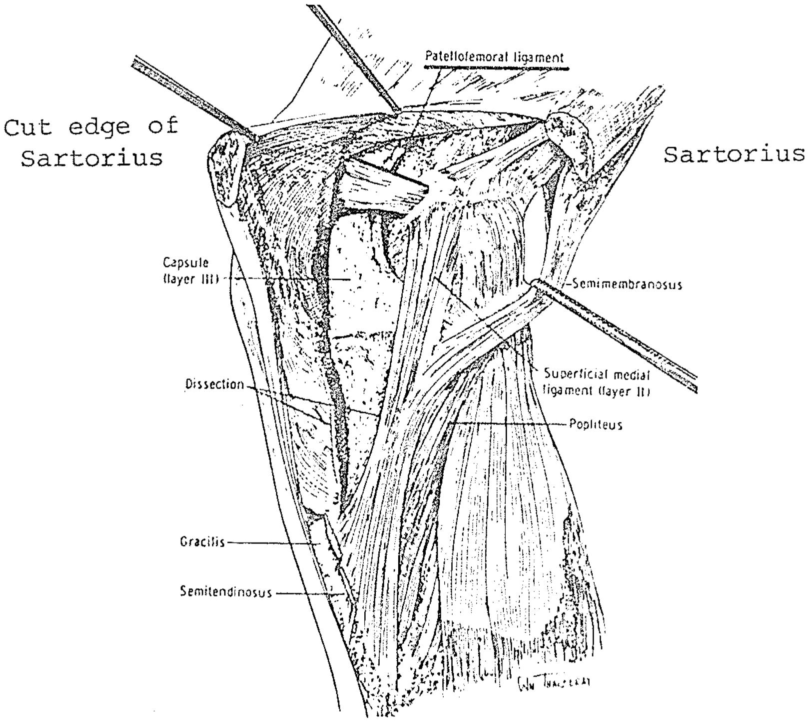
Posterior Fibres Slack

A AND B ARE THE ANTERIOR BORDER OF THE LONG FIBRES.

POINT C IS 5 mm POSTERIOR TO B.

(From Warren et al 1974)

FIGURE 8. PATELLO-FEMORAL LIGAMENT.



Medial Aspect of Right Knee

(From Warren and Marshall 1979)

Wang and Walker (1974) state that the collateral ligaments are twice as effective as the cruciate ligament in controlling joint laxity.

Valgus/Varus. The superficial portion of the ligament is vital to valgus/varus (medial/lateral) stability in flexion rather than extension (Markoloff et al., 1976). In full extension with intact cruciates the amount of valgus/varus movement is minimal (Hallen and Lindahl, 1965; Warren et al., 1974).

Deep Fibres

The deep fibres are described as being deep to the long fibres and along the lip of the tibial articular cartilage (Warren and Marshall, 1979) although Mains et al. (1977) describe a more anterior attachment on the tibia. The deep fibres are attached to the medial meniscus as are the posterior fibres of the superficial part (Warren and Marshall, 1979).

Deep Fibres Function. The deep portion's contribution to knee stability is considered to be of relatively minor importance partially due to its relatively weak anatomical structure (Mains et al., 1977). It was found to have a minimal effect on controlling rotation (Warren et al., 1974; Markoloff et al., 1976) or valgus/varus stability (Hallen and Lindahl, 1965; Warren et al., 1974) and is lax in extension (Main et al., 1977).

FIBULAR COLLATERAL LIGAMENT

This ligament is also known as the lateral collateral ligament. This rounded cord-shaped structure is attached proximally to the lateral epicondyle of the femur and extends to the anterior surface of the apex of the fibular head; it is unattached to the lateral meniscus (Gray, 1973).

Flexion and Extension. It is taut in extension and lax in flexion (Abbott et al., 1944; Last, 1972; Gray, 1973). In extension it is under greater tension than the medial structures of the joint (Hughston et al., 1976B).

Rotation. When the knee is flexed it controls medial and lateral rotation (Brantigan and Voshell, 1941).

Valgus/Varus. In extension, if severed the valgus/varus (medial/lateral) instability may not be detected clinically unless other structures are involved (Markoloff et al., 1976).

OBLIQUE POPLITEAL LIGAMENT

This ligament has also been called the oblique popliteal ligament of Winslow (Last, 1972). This is a posterior thickening of the capsule in the form of fasciculi, passing from the insertion of semimembranosus proximally and laterally to the lateral femoral condyle (Gray, 1973; Lewin, 1952). It is considered to be an extension of the semimembranosus tendon sheath (Warren and Marshall, 1979).

It controls hyperextension (Brantigan and Voshell, 1941) and in conjunction with the posterior capsule is important for stability in most positions (Markoloff et al., 1976) though O'Donoghue (1978) acknowledges that once flexion occurs it becomes lax.

LIGAMENTUM PATELLAE

This structure will be considered as part of the quadriceps femoris muscle.

ARCUATE POPLITEAL LIGAMENT

This ligament is also known as the arcuate ligament (James, 1978) and the posterior arcuate ligament (Kennedy, 1978; James, 1978).

This "Y" shaped ligament extends from the head of the fibula to its wider proximal component attached to the lateral femoral condyle. The anterior limb is also called the short lateral ligament (Gray, 1973). Though Last (1972) contends that there is a separate short external lateral ligament that is anterior to the "arcuate" ligament. It would appear to have the same functions as a component of the posterior capsule of the knee.

The term "arcuate complex" is used to indicate the arcuate ligament, tendon of popliteus and the lateral collateral ligament (James, 1978).

TRANSVERSE LIGAMENT

This variable ligament connects the anterior convex margins of the menisci and may be absent (Gray, 1973)(Fig. 4).

Its function is not described in the literature.

ANTERIOR AND POSTERIOR MENISCOFEMORAL LIGAMENTS

Humphry (1858) originally described both an anterior and posterior fibrous band accompanying the posterior cruciate ligament. Subsequently, however, the ligament of Humphry was described as the anterior meniscofemoral ligament and the posterior fibres were described as the ligament of Wrisberg (Grant, 1937; Heller and Langman, 1964). In French, German and Russian literature the posterior meniscofemoral ligament is also known as Robert's ligament (Kaplan, 1956). It is also known as the lateral meniscofemoral ligament.

The frequency of occurrence of these ligaments varies from 70% (Heller and Langman, 1964) to 100% (Brantigan and Voshell, 1946) and when present the posterior meniscofemoral ligament is more commonly found (Girgis et al., 1975) though this is disputed (Heller and Langman, 1964). A small minority have both (Heller and Langman, 1964).

The function of these two ligaments is to "control" the mobility of the lateral meniscus with popliteus (Last, 1948) and to increase the congruity of the lateral aspect of the joint during flexion (Heller and Langman, 1964). However, the lack of constancy of these accessory ligaments suggests a limited functional importance to Girgis et al. (1975).

TIBIOMENISCAL LIGAMENT

This ligament described by Girgis et al. (1975) states confirmation of an original description by Kaplan (1957) of a fibrous band from the posterior cruciate ligament to the posterior horn of the lateral meniscus. On reviewing Kaplan's (1957) paper no reference to the tibiomeniscal ligament was found.

PATELLOFEMORAL LIGAMENT

This ligament has also been called the epicondylopatellar ligament (Kaplan, 1962).

This ligament is deep to vastus medialis passing from the medial femoral epicondyle towards the patella (Warren and Marshall, 1979)(Fig.

The function of this ligament is not stated.

LIGAMENT OF BARKOW

This ligament, not described in current literature, passes from the posterior horn of the lateral meniscus to the anterior horn of the medial meniscus (Galeazzi, 1927).

Menisci of the Knee Joint (Medial and Lateral)

These internal articular structures are also known as the semilunar cartilages and are attached to the periphery of the articular surface of the tibia via the coronary ligament. The shape and attachment of each is seen in Figure 4. In cross section they are wedge shaped, the tapered part of the wedge facing the centre of the articular surface and the thicker part of the wedge peripheral. The inferior surface of the meniscus is flat, covering approximately 66% of each articulating surface (Gray, 1973). The superior surface is concave.

The shape of the medial meniscus is usually constant while the lateral meniscus has many variations (Levine and Blazine, 1966).

The menisci are fibrocartilagenous and avascular except for their peripheral attachment (Davies and Edwards, 1948). The extent of this vascularity has been described as the outer one-third (Scapinelli, 1968). The fibres of the menisci are aligned predominantly in the long axis of the meniscus with few radial, oblique and vertical fibres (Cameron and MacNab, 1972; Bullough et al., 1970).

Function

Though Gray (1973) states that the functional role of the interarticular fibrocartilage remains unproven because the views advanced are largely deductions from structure and phylogenetic data with the help of mechanical analysis, some recent data substantiate a few of the claims. The functions of the menisci have been described as:

1) Increasing weight distribution, weight bearing
2) Shock absorbing
3) Increasing congruency
4) Facilitating combined movements
5) Facilitating spread of lubricants
6) Protection (Steindler, 1955)

Weight Bearing

It is now well established that the menisci bear weight (Shrive, 1974; Seedhom et al., 1974; Waler and Erkman, 1975; Shrive et al., 1978; Goodfellow and O'Connor, 1978). Earlier, though some "cushioning" was acknowledged, weight bearing was described as not occurring in vertical compression because condylar articular contact between the femur and tibia always existed (Brantigan and Voshell, 1941). Contact with menisci and articular cartilage was thought to be pathologic (Last, 1972). Shrive et al. (1978) state that the menisci bear 45% of the total joint load from full extension to 30° of flexion, and Krause et al. (1976) state a range of 30-55% of the total joint load in extension. In comparing medial and lateral loading, the lateral meniscus takes a greater percentage of the load than the medial meniscus, which shares its load bearing with the articular cartilage equally (Walker and Erkmann, 1975).

The structural arrangement of the fibres of the meniscus permit tensile strength to develop to the same magnitude as that developed in articular cartilage (Bullough et al., 1970; Mather et al., 1949). By implication, because these similar adjacent surfaces have similar properties they both have similar functions of weight bearing.

Hoop stress (or circular type stress) within the meniscus which occurs during compressive loading has been demonstrated which is also indicative of weight bearing, as is the increased stress that occurs across the knee joint after a menisectomy (Krause et al., 1976). Maquet et al. (1975) cite the substantial reduction in contact area of the knee after menisectomy, which is indicative of weight bearing by the menisci.

Shock Absorbing

Brantigan and Voshell (1941) described the shock absorbing or cushioning effect of the menisci to occur only in hyperflexion or hyperextension (presumably the ends of range of motion), while others have considered vertical compression absorption as a function of the menisci (Gardner et al., 1975).

The importance of subchondral bone in sparing articular cartilage from impact has been reported (Radin et al., 1970; Radin and Paul, 1971; Simon et al., 1972) and the load bearing stated (see section on weight bearing). Krause et al. (1976) postulates that the articular cartilage covered by the menisci can be thought of as being sandwiched between two energy absorbing materials, the menisci and the subchondral bone. Uzaki et al. (1979) in a study of the viscoelastic properties of menisci states that the menisci do not play a role as a shock absorber because the properties demonstrated were mostly that of an elastic material.

Congruency

Increasing congruency by implication means increased stability, and it was considered "obvious" that the menisci were stabilizers of the knee (Corner, 1915). Shinno (1962) also stated the menisci acted as cushion which increased stability. However, after removal of both menisci, the valgus/varus laxity of the knee joint is relatively unaffected and only minor changes are recorded in torsional laxity (Markolf et al., 1976). However, Wang and Walker (1974) stated the menisci played a part in controlling joint laxity.

In addition, increasing congruency *per se* is not necessarily beneficial to function. Walker and Hsieh (1977) in their study of femoral and tibial prostheses indicated that both high and low congruency configurations produced increased abrasion and fatigue of components. They suggested that a partial conformity between the femoral and tibial component is the optimum configuration.

Combined Movements

The fact that the only mammal, the fruit bat, that has no menisci in its knee joint is the only mammal that lacks rotation at that joint, led to the belief that the most important function of the menisci is to allow rotation (Last, 1972).

Kaplan (1955) states that the similarity of the embryologic derivation of the menisci and the cruciates makes it seem likely that they function as an accessory part of the rotation at the knee.

Watson-Jones (1955) by a manual technique also attributes a role of the menisci in rotation of the knee joint.

The complexity of knee motion and its accommodation by the menisci is decribed thoroughly by Kapandji (1970) and Steindler (1955).

Lubrication

A plausible deduction from observing the structure of the menisci is that it facilitates the spread of synovial fluid. A qualitative study based on theories of lubrication states that a loss of menisci leads to about 20% increase in friction (MacConaill, 1932). Various theories of joint lubrication (Dintenfass, 1963; Dowson et al., 1969) do not quantify any increased joint lubrication by the menisci.

Protection

Some authors feel that the menisci keep the capsule and synovia from being pinched as they fill the space left by the incongruency of the femoral and tibial articulating surfaces (Brantigan and Voshell, 1941; Steindler, 1955).

The menisci may calcify (Cave, 1943) or be discoid, i.e., no contact existing between tibia and femur (Duncan, 1932; Brantigan and Voshell, 1941; Kaplan, 1957).

Contact Areas of the Knee Joint

Tibiofemoral

The early European radiographic studies in this field established varying tibiofemoral contact areas (Zuppinger, 1904; Fischer, 1907, Strasser, 1917). The more recent studies show widely differing figures which may be explained by the varying techniques used. Walker and Hajek (1972) stated that in knee extension a combined area (medial and lateral) of 3.2 sq cm (with a load of 150 kp) was found. Kettlekamp and Jacobs (1972) determined the average area to be in excess of twice that figure with a load of 3-8 kp. However, Maquet et al., (1975) using considerably greater loads (225 to 250 versus 3-8 kp) (Kettlekamp and Jacobs, 1972) obtained a contact area figure of 20.13 cm^2 which is over six times greater than that obtained by Walker and Hajek (1972).

The ratio between the smaller lateral and larger medial area was 1:1.6 (Kettlekamp and Jacobs, 1972), however, under "physiological loading" (i.e., heavy loading) there was only a slight difference in weight bearing between the medial and lateral plateau (Maquet et al., 1975). The contact area moved posteriorly and decreased as knee flexion increased (Walker and Hajek, 1972; Kettlekamp and Jacobs, 1972; Maquet et al., 1975) (Fig. 11).

Patellofemoral

The work of Wiberg (1941) was confirmed by Goodfellow et al. (1976A). Under light loads (2-10 kp) they established a patellar pattern which they felt was appropriate because the incongrous articular surfaces

FIGURE 9. PATELLA CONTACT AREAS IN VARYING DEGREES OF FLEXION.

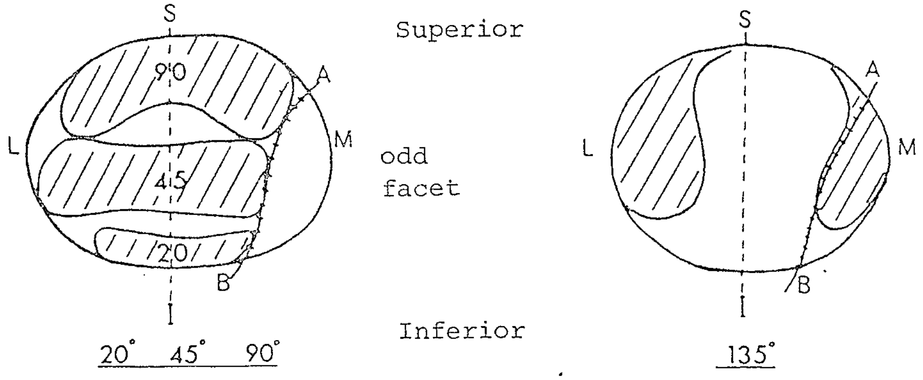
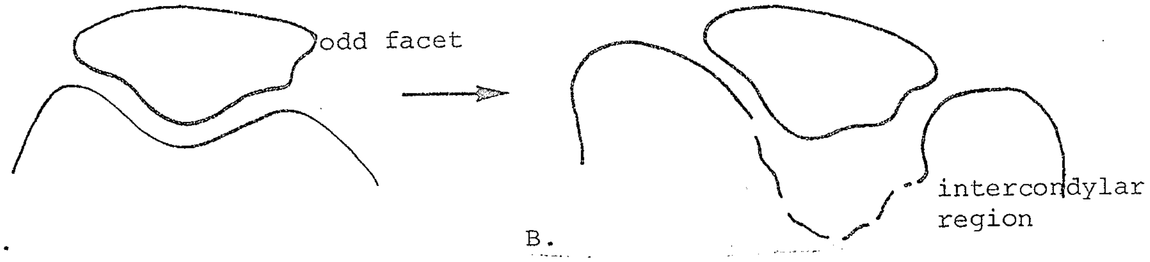


FIGURE 10. HORIZONTAL SECTION THROUGH PATELLA AND FEMORAL CONDYLES.

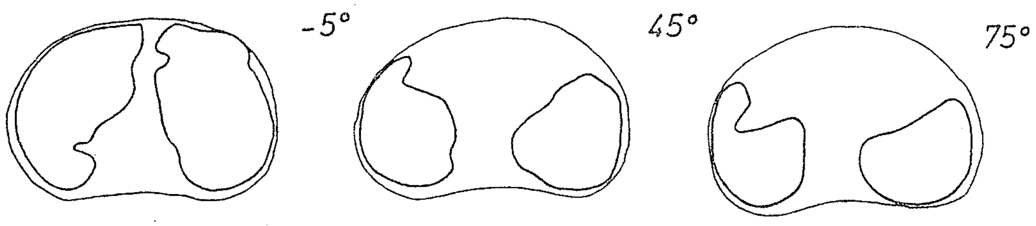
B. BEYOND 90° MEDIAL FEMORAL CONDYLE ARTICULATES WITH ODD FACET, WHILE THE WHOLE OF THE MEDIAL FACET LIES FREE IN THE INTERCONDYLAR REGION.

A. FROM FULL EXTENSION TO 90°



(From Goodfellow et al. 1976)

FIGURE 11. TIBIAL CONTACT AREAS IN VARIOUS DEGREES OF FLEXION.



(From Maquet et al. 1975)

were felt to be more position dependent rather than load dependent. The first contact of the patella and femur occurs at 10°-20° of flexion. The patellar surface area of contact moves from distal to proximal increasing in size (Fig. 9) in the first 90° of flexion (Aglietti et al., 1975). In full flexion the contact pattern is very different with the medial facet completely out of contact lying free in the intercondylar area with the entire odd facet engaging the lateral border of the medial femoral condyle. During this contact pattern the patella rotates (Goodfellow et al., 1976A)(Fig. 10). The patellofemoral contact areas are viewed together in Figure 12.

Trabecular Architecture of the Knee Joint

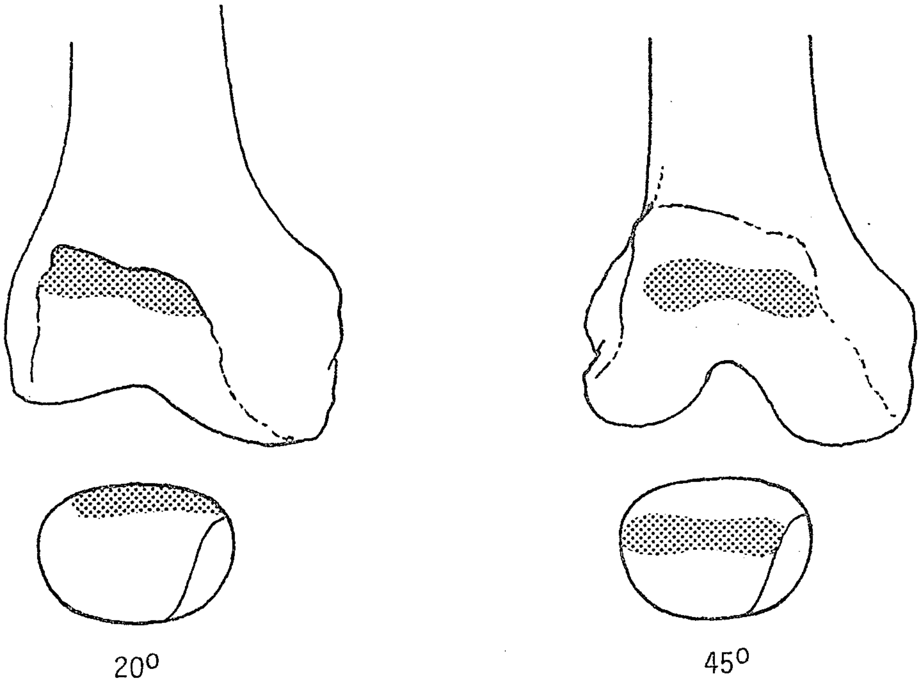
Femur

The trabeculae of the femoral condyles are arranged in five groups which allow the bone to adjust to mechanical compression and rotatory movement (Fig. 13).

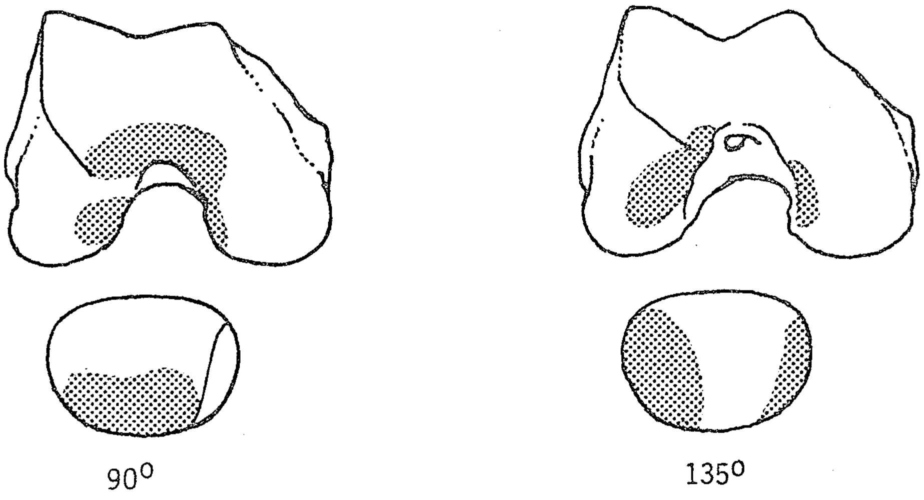
Tibia

Each tibial trabeculae arrangement has a different mechanical function. The epiphyseal line can be considered a laminated structure between the compressive forces from the articular surface above and the forces supporting the architecture of the metaphysis (Fig. 13).

FIGURE 12. PATELLO-FEMORAL JOINT CONTACT AREAS IN VARIOUS DEGREES OF KNEE FLEXION.



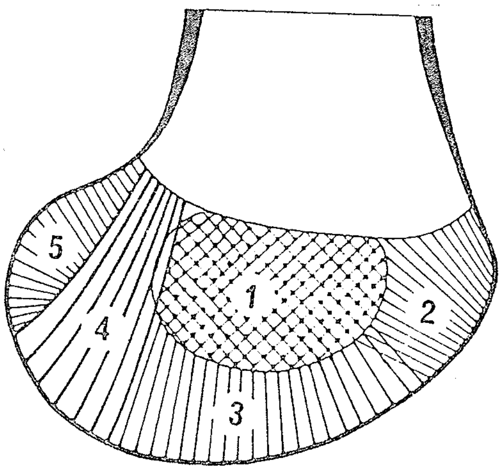
(THE EXTENSOR MECHANISM WITH THE PATELLA IS TURNED DOWN SO THAT THE FEMORAL AND PATELLAR CONTACT AREAS ARE REVEALED.)



From Goodfellow et al. (1976)

FIGURE 13. TRABECULAR ARCHITECTURE OF THE KNEE.

A. FEMORAL TRABECULAR ARRANGEMENT



FUNCTION

1 = Axis of wheel arrangement.

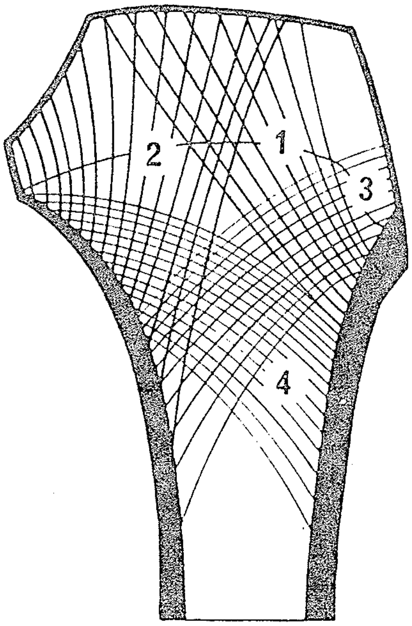
2 = To resist patella compression.

3 = Spokes of wheel.

4 = Supports weight-bearing force and outer contour of bone.

5 = Supports outer contour.

B. TIBIAL TRABECULAR ARRANGEMENT



FUNCTION

1 & 3 resist tension from quadriceps femoris muscle.

4 resist tension from hamstrings.

2 and posterior of 1 support compression from articular surface.

(From Takechi 1977)

Patella

Raux et al., (1975) determined that the predominant trabecular arrangements were in an anteroposterior direction between two layers of compacted bone. The lateral facet has a generally thicker compact subchrondral bone suggesting greater loading (Ficat and Hungerford, 1977).

In a study of the knee joint Behrens et al. (1974) showed the trabecular organization to be a significant factor in the strength of the bone while the bone material density did not.

Nerve Supply

Hilton's law (1891), that the muscles producing movement at a joint share their innervation with the articular structures of that joint, operates. The nerves are the femoral, obturator, tibial, common peroneal, and recurrent peroneal nerves (Gardner, 1948; Gray, 1973). Rarely the accessory obturator nerve supplies the joint. The posterior capsule has the greatest concentration of nerve fibres (Gardner, 1948).

The nerve supply is intimately associated with the blood supply. Both sympathetic and parasympathetic nerve fibres are present (Cooper et al, 1966).

Blood Supply

Arterial

The following arteries are the major blood supply to the knee joint: 1) descending genicular artery

2) medial and lateral superior genicular arteries

3) medial and lateral inferior genicular arteries

4) middle genicular artery

5) anterior and posterior tibial recurrent arteries

The arteries above form the geniculate anatomosis (Gray, 1973). There is an additional supply from:

1) descending branch of the lateral femoral circumflex

2) recurrent fibular arteries

(Scapinelli, 1968)

The prime supply of the patella enters through the middle third and inferior pole of the patella (Scapinelli, 1967).

Venous

The venous drainage of the knee is primarily through the deep vein system accompanying the arteries (Gray, 1973).

Lymphatic Drainage of the Knee

Most of the lymphatic drainage of the knee is into the popliteal lymph nodes situated in the popliteal fossa (Gray, 1973).

BIOMECHANICS OF THE KNEE JOINT

Introduction

Biomechanics is the larger field of study of mechanics (or motion) of living systems (Wartenweiler, 1973). Kinesiology, a branch of biomechanics, relates to human body movements (Dorland, 1974). Arthrokinematics, a part of kinesiology, is the study of movements that occur in joints and osteokinematics of movement of bones (MacConaill and Basmajian, 1977). An understanding of the arthrokinematics of the knee is essential for the design of an electrogoniometer. The development of an electrogoniometer for the knee is an integral part of this thesis.

Axes

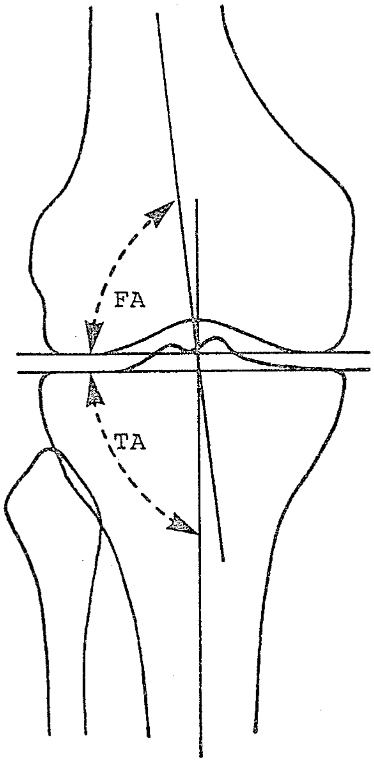
The static longitudinal axis of the tibia and femur derived radiologically are shown in Figure 14.

The determination of the dynamic axes of a joint allows analysis of the type of movement that occurs in the joint (Frankel and Burstein, 1970). It is apparent that the inequality of articular surfaces of the knee joint and the complex varying surface of the femoral component would vary the axis of movement (Steindler, 1955).

Flexion/Extension Axis

The largest functional movement of the knee has no single point which can be considered the axis. However, if the instant centres of motion, or centrodes, are linked together, a varying axis will be demonstrated. The classic work of Frankel and Burstein (1970), using a method developed a century earlier by Reuleaux (1876), stated that the instant centres were in a position shown in Figure 16 with the involute opening anteriorly. However, in this analysis the motion of the tibia and femur are considered planar and the existing tibial rotation, though it has been known for over a century, is ignored in the calculation (Weber and Weber, 1836; Meyer, 1853; Goodsir, 1868).

FIGURE 14. AXIAL RELATIONSHIPS OF THE KNEE.



FA = FEMORAL ANGLE

TA = TIBIAL ANGLE

FA = 80.2° 81.2°

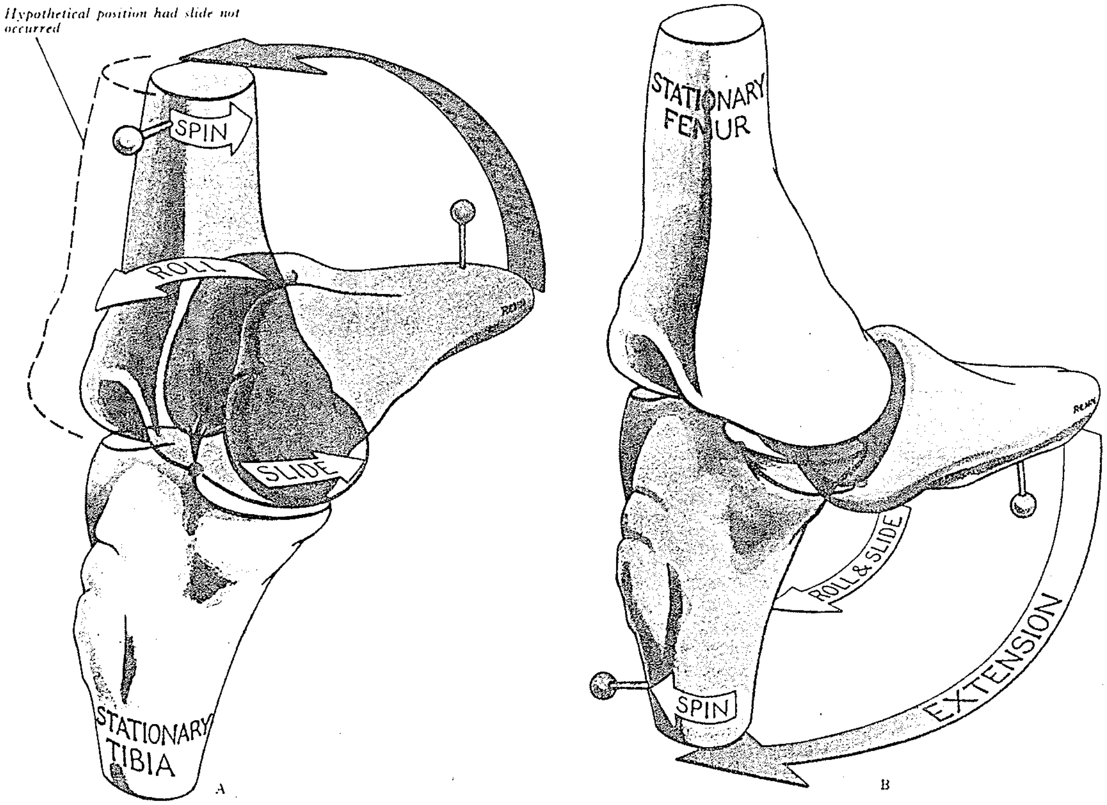
TA = 93.5° 92.4°

TOTAL: FA = 81°

TA = 93°

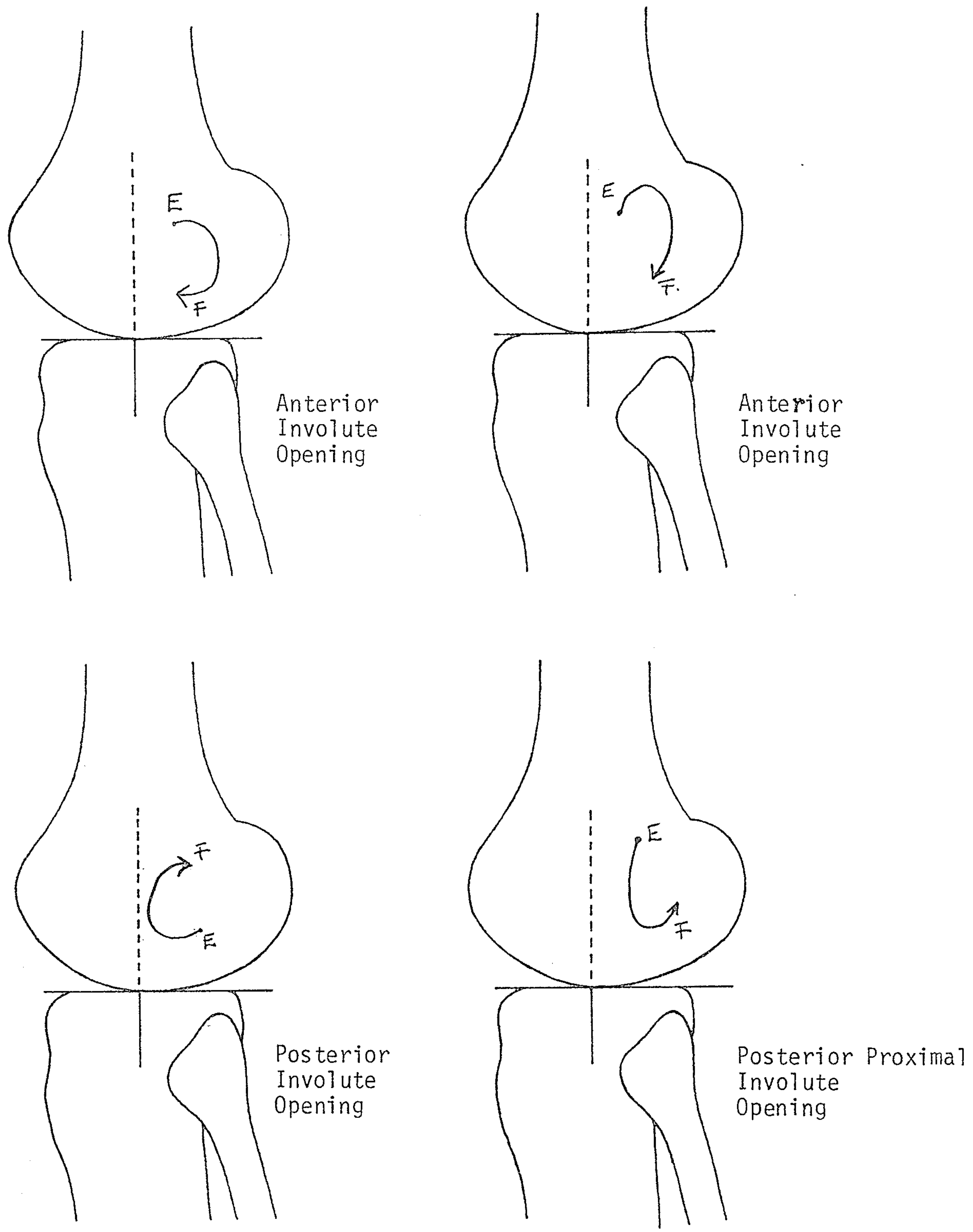
(From Keats et al., 1966)

FIGURE 15. KNEE MOTION.



(From Gray, 1973)

FIGURE 16. AXIS OF FLEXION AND EXTENSION OF THE KNEE



Frankel & Burstein (1970)

Smidt (1973)

Kettlekamp & Nasca (1973)

Kelly (1971)

Smidt (1973) also analysed the instant centres through a similar range of movement and similar technique and found a different pattern, with the involute opening anterodistally. The axis moves through 3.2 cm in 90° within a circle of 2.3 cm located at the level of the lateral epicondyle (Fig. 16).

In addition, Kelly (1971) described the axes of flexion and extension forming a curve with the opening superoposteriorly (Fig. 16).

Kettlekamp and Nasca (1973) reproducing the axis from Doctor V. H. Frankel (unreferenced) demonstrate the involute opening posteriorly and moving in a different direction (Fig. 16).

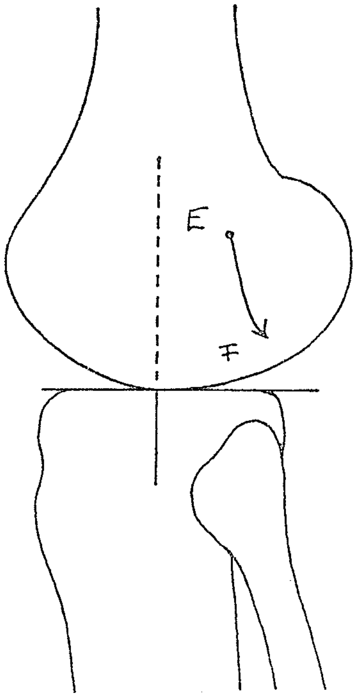
Walker et al. (1972) stated that a fixed axis of rotation is hardly applicable to the knee and that the loci (instant centres) of the rotational axis moved upwards, backwards, forwards and can form a loop. They also stated that the rotational position of the femur and tibia changes the instant centres.

Blacharski et al. (1975) and Harding et al. (1977) using a three-dimensional approach tend to discount the instant centres forming an involute curve and support the variable pattern distribution. They also acknowledge different patterns for each condyle which was elaborated by Strasser (1917)(Fig. 17).

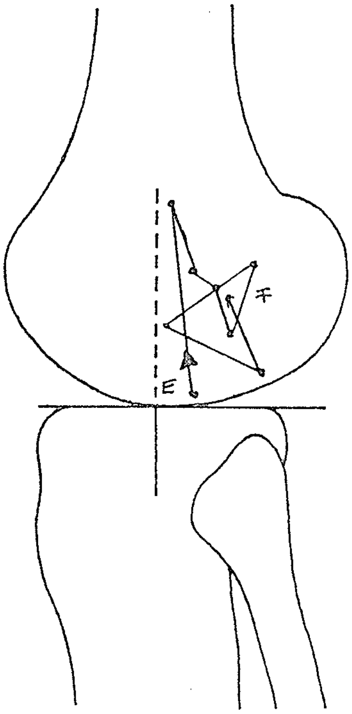
Recently Soudan et al. (1979) reviewing methodology, emphasized that the earlier instant centre concept is only valid for planar movement (which was recognized by the original authors), and that three-dimensional movements when presumed to be two dimensional can give misleading results. However, they calculate the instant centres to be more linear

FIGURE 17. AXIS OF FLEXION AND EXTENSION OF THE KNEE

Soudan et al. (1979)



Harding et al. (1977)



(Fig. 17) but stated that further examples were needed for conclusions to be made.

In summary, the instant centres exist somewhere at the level of the lateral femoral epicondyles, posterior to the long axis of the tibia.

Medial and Lateral Rotation Axis

Earlier the vertical axis which allows medial and lateral rotation to occur was difficult to find because of the limited movement in that plane (Haines, 1941). However, more recently there have been studies which do not demonstrate unanimity. The literature is summarized for three areas: 1) lateral region

2) intercondylar region

3) medial region

Lateral Region

Kelly (1971) states that the axis is in the centre of the lateral tibial condyle.

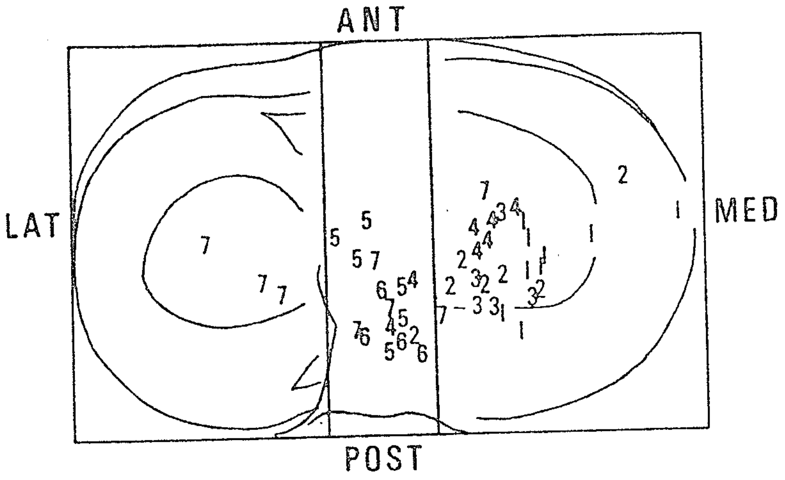
Barnett (1953) states that it is between the two horns of the lateral meniscus.

Intercondylar Region

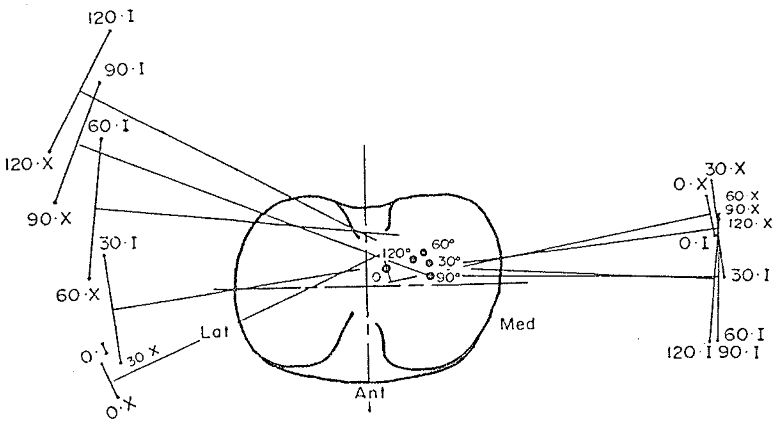
A second group of centres were in this region (Trent et al., 1976) (Fig. 18).

Shaw and Murry (1974) in their unloaded study stated that the axis passes through the medial intercondylar tubercle of the tibial pleateau. This concurs with Kaplan's earlier work (1962).

FIGURE 18. AXIS OF ROTATION OF THE KNEE.



A) SUPERIOR VIEW OF THE LEFT TIBIA. The numbers represent the centers of rotation in 15° increments for seven subjects numbered one to seven. (From Trent et al., 1976)



B) SUPERIOR VIEW OF THE RIGHT TIBIA. Using coordinates in a horizontal plane for neutral, internal and external rotation, the centre of rotation about an axis perpendicular to the horizontal plane is marked by the dark spots at various angles of flexion. (From Wang et al., 1973).

Medial Region

Brantigan and Voshell (1941) stated that this axis was near the medial femoral condyle with which Steindler (1955) agrees, stating that the lateral condyles rotate around the medial condyles.

Wang et al. (1973) calculated the centres of rotation from 0-120° of movement and they all occurred in the medial condyle and all but one of Trent's group concurs with this location (Trent et al., 1976)(Fig. 18).

Analysis of Knee Movement

The study of the movement of bones is called osteokinematics (MacConaill and Basmajian, 1977).

Viewed anteriorly, the longitudinal axes of the shafts of the femur and tibia are not vertical but subtend an angle of 171° on their lateral aspect (Engin and Korde, 1974). The knee joint surfaces are capable of slide, spin and roll movements (Fig. 15). As the femur and tibia move on each other a combination of rolling and sliding occurs simultaneously. At the end of extension a spin (or rotation) occurs. The direction of the spin, roll and slide depends on which component is moving on the other. For example, in walking, where the femur moves on the relatively fixed or stationary tibia, because a male surface is moving on a female surface, the roll and slide are in opposite directions and the spin is medial.

In kicking a ball, however, where the tibia is moving on a relatively fixed or stationary femur, the female component is moving on the male, here the roll and slide are in the same direction and the spin is lateral (Gray, 1973; MacConaill and Basmajian, 1977). The rotation in knee extension is the well known "screw home" (Lewin, 1952) and locking mechanism of the knee (Helfet, 1959; Hallen and Lindahl, 1966; Smillie, 1970).

Iseki and Tomatsu (1976) describe three phases of extension from flexion: 1) Early "hinge" phase from full flexion to 60° (a slide)

2) A "rolling" phase from 60° to 15° (with a slide)

3) Final "screw home" phase from 15° to 0°

The knee is considered "locked" in extension, the greater radius of curvature of the anterior part of the femoral condyles effectively tightening the ligaments in extension. However, a similar state of tension in the ligaments and resisted hyperextension occurs when the natural condyles are replaced with spherical ones. This work by Goodfellow and O'Connor (1978) postulates an anterior location of the contact areas and a posterior location of the tension elements (posterior part of the capsule and anterior cruciate ligament primarily). These are responsible for the development of a force which is capable of resisting hyperextension.

In "unlocking" of the knee, for example, in walking, which occurs on the relatively fixed tibia, the femur rotates laterally. The conformity of the joint surfaces and the position of the ligaments is important but the action of popliteus has been demonstrated to be of principal significance (Basmajian and Lovejoy, 1971).

The quantification of rotation phase has been mentioned earlier and variations in time and extent occur. Barnett (1953), for example, states that rotation occurs through the last 30° and is maximal in the last 10° in locking, but in "unlocking" the rotation is "confined" to the first few degrees of flexion.

Range of Knee Motion in Stair Climbing

The method for measuring and recording ranges of knee motion will be based on the Neutral Zero Method (Cave and Roberts, 1936) which has been accepted by the major orthopedic association of the world (Heck et al., 1965).

In this thesis, the largest functional range of the knee, i.e., flexion and extension, is studied. Studies on the movement of the knee in the sagittal plane have been many but primarily relate to a walking gait rather than to a climbing gait (Saunders et al., 1953; Elftman, 1955; Smith, 1956; Finley and Karpovich, 1964; Adrian and Karpovich, 1964; Gollnick and Karpovich, 1964; Gollnick et al., 1964; Murray et al., 1964; Tipton and Karpovich, 1965; Liberson, 1965; Inman, 1966; Murray et al., 1966; Murray, 1967; Morrison, 1968; Kettlekamp et al., 1970; Kettlekamp, 1973; Townsend et al., 1977).

Flexion/Extension

The following papers relate to stair climbing and have been summarized.

Laubenthal's et al. (1972) electrogoniometric study was of knee motion during activities of daily living in 30 subjects with a mean age of 25. The average maximum flexion was 83° for ascending and descending stairs, however the stair dimensions were not stated though a slope of 32° with only two steps was used. The cadence and temporal parameters were not defined.

In Berry's (1952) cinematographic technique five subjects of unstated age climbed six steps which measured 16.5 cm rise x 30.5 cm tread run and had a slope of 28°.

The average maximum and minimum flexion ranges which were unstated but calculated by this author from the graphs were, ascending maximum 89° and minimum 8°, descending maximum 92° and minimum 17°. The cadence was ascending 118, descending 121.

Shinno (1968), in a mathematic calculation, stated that the optimum slope for stair climbing was 30°, and that 85° of knee flexion was required. The optimum stair height or rise was 16.6 cm and tread run was 28.6 cm. A 20°-50° slope was considered "ordinary and convenient". At 35° slope, maximum flexion was calculated to be 89°, and with a slope of 40° maximum knee flexion was 91°. Because of the nature of the study, cadence and other variables were not considered.

McLeod et al., (1975) in their study of repetitive activities of the knee during 50 hours--with an electrogoniometer, did not state the range. However, from the graphs in the paper a maximum of 85° flexion was estimated in ascending and 88° in descending. The minimum ranges were -3° and -5° respectively.

The cadence for ascending was 86 and for descending 81.

It is stressed that these are approximate values calculated from the graphs in the paper.

Sturman's (1976) electrogoniometric study was of ascending stairs only, with 15 subjects who had a mean age of 22 years.

The 4 steps had a rise of 15.2 cm, tread run of 27.9 cm and a slope of 29°. The average maximum flexion was 85° and the average minimum was 10°. The average weight bearing occurred at 56° of knee flexion. The cadence was not stated.

Abduction/Adduction Rotation

Only one paper studies rotation and valgus-varus movement at the knee during stair climbing. Laubenthal et al. (1972) gives the following figures in ascending: an abduction and adduction movement of 17°, in rotation 16°, and in descending 14° and 15° respectively. However, on repeating the test the correlation was poor (i.e. <.45).

Compression Forces at the Knee Joint in Stair Climbing

Morrison (1969) states in his study of the tibiofemoral component of the knee, that although no definite trend can be deduced, it may be stated that the average values of maximum joint force are greater by 12 to 25 percent in stair climbing than in walking. Both in ascending and descending there are force peaks at the beginning and end of stance phase which are three to four times body weight (cadence 40-50).

Paul (1976) states that the patellofemoral force in ascending stairs is 1.7 times body weight, and in descending stairs is 2.7 times body weight.

Reilly and Martens (1972) describe similar results, stating a patellofemoral joint reaction force of 3.3 times body weight in descending stairs.

Seedhom and Terayama (1976), in their study of knee forces in getting out of a chair unaided, state that both the tibiofemoral and patellofemoral component forces were similar to that of walking. This study is mentioned here because of the flexed knee position.

As an indication of the relation of these units to more strenuous activity, Smith (1972), studying a drop landing from a height of approximately one meter, calculated the tibiofemoral force to be 24 times body weight and the patellofemoral to be 20 times body weight.

MUSCLES OF THE KNEE JOINT

The muscles studied in this thesis are the major muscles that provide knee movement in a sagittal plane. These are quadriceps femoris, biceps femoris, semitendinosus and semimembranosus. The latter three are studied as the medial and lateral hamstrings.

Quadriceps Femoris (Fig. 19)

This largest muscle of the body with its "four heads" or bellies is the prime extensor of the knee (Last, 1972). It covers the anterior, medial and lateral aspects of the thigh, and each of its four muscle components are given separate names and will be described separately.

Rectus Femoris. This muscle originates in two heads from the ilium above the acetabulum and is the only part of the quadriceps to span both the hip and the knee. This bipennate fusiform muscle attaches to the base of the patella via a thick, broad aponeurosis.

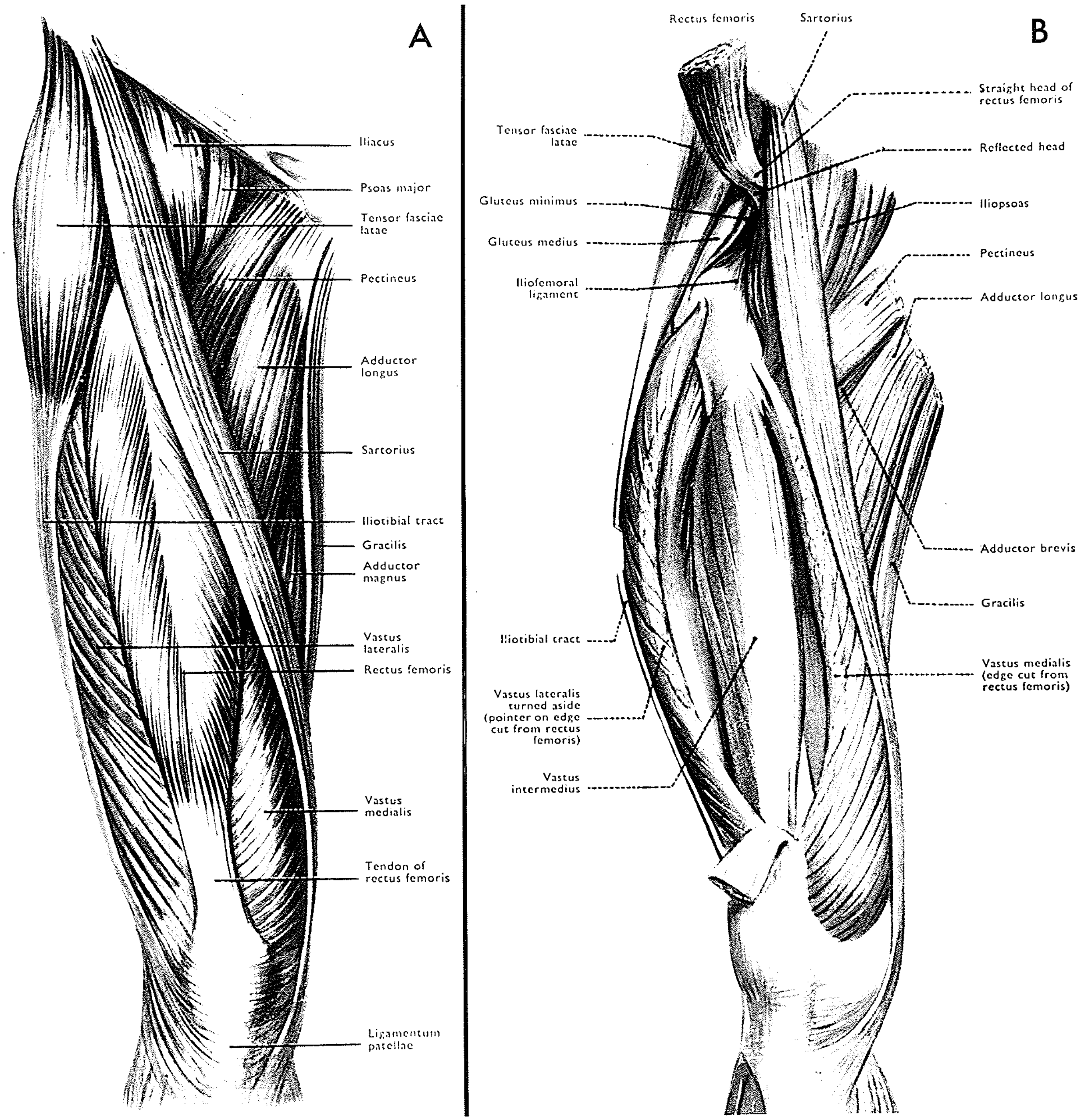
Vastus Medialis. This muscle is often fused with vastus intermedius (Gray, 1973), which in earlier texts resulted in the presently known quadriceps femoris being known as the femoral triceps (Duchenne, 1866). It originates from the lower part of the intertrochanteric line and the medial lip of the linea aspera to be inserted for the most part into the medial border of the patella. Though considered as a single entity by most standard texts (Cunningham, 1972; Gray, 1973; Last, 1972;

FIGURE 19.

A) RIGHT THIGH--Anterior view.

B) RIGHT THIGH--Anterior view with rectus femoris removed and vastus lateralis turned aside.

(from Cunningham, 1972)


A
Iliacus
Psoas major
Tensor fasciae latae
Pectineus
Adductor longus
Sartorius
Iliotibial tract
Gracilis
Adductor magnus
Vastus lateralis
Rectus femoris
Vastus medialis
Tendon of rectus femoris
Ligamentum patellae
B
Rectus femoris
Sartorius
Straight head of rectus femoris
Reflected head
Tensor fasciae latae
Gluteus minimus
Iliopsoas
Gluteus medius
Pectineus
Iliofemoral ligament
Adductor longus
Adductor brevis
Gracilis
Iliotibial tract
Vastus medialis (edge cut from rectus femoris)
Vastus lateralis turned aside (pointer on edge cut from rectus femoris)
Vastus intermedius

Grant, 1978), the lower fibres which insert "almost horizontally" (Gray, 1973; Last, 1972) have been differentiated from the upper fibres. Lieb and Perry (1968) called the upper fibres vastus medialis longus and the lower fibres vastus medialis oblique, and this terminology has been used subsequently by others (Cailliet, 1973; Speakman and Weisberg, 1977).

Vastus Lateralis. This muscle is the largest of the quadriceps femoris group. It arises from the lateral aspect of the greater trochanter and the lateral lip of the linea aspera. The extent of the origin varies from the upper half of this line (Gray, 1973) to the total length (Last, 1972; Grant, 1978). Its major insertion is into the lateral border of the patella.

Vastus Intermedius. This muscle is separated by a cleft from vastus medialis, or fused with it, and originates from the anterolateral shaft of the femur under the cover of vastus lateralis and rectus femoris. It fuses with the latter to insert into the patella via its common tendon (Gray, 1973). This muscle was also called crureus (Piersol, 1907; Davis, 1924; Buchanan, 1946).

Ligamentum Patellae. This is a continuation of the quadriceps tendon in which its sesamoid bone, the patella, is inserted. It extends from the patellar apex to the tubercle of the tibia and also receives the medial and lateral patellar retinaculae, the fibrous expansions of the quadriceps (Fig. 19).

Nerve Supply and Blood Supply of the Quadriceps Femoris

The quadriceps femoris is supplied by the femoral nerve from L2, L3 and L4, mainly from the latter two roots (Gray, 1973).

The main arterial supply is via the profunda femoris artery and its branches (Gray, 1973).

Function of the Quadriceps Femoris

Basmajian (1978) repeats the question asked in his previous editions, "why are there four heads for a muscle that has most of its insertion on the restricted edges of the patella? Do the various parts have individual functions?" Early EMG studies of this muscle concluded that the three vasti functioned in a similar manner (Pocock, 1963; Close, 1964). However, the apparatus and methodology were not the same as more recently have been used. For example, in Close's study the results were obtained by single channel recording and not by simultaneous multichannel recording.

Basmajian et al. (1972), more recently studied the interplay of the quadriceps with fine wire technique and demonstrated, amongst other things, a striking variation of activity between the different heads. A difference in function of the heads was also indicated by Wheatley and Jahnke (1951). Furthermore, Ravaglia (1957) demonstrated that the different heads acted in different ways in various phases of knee movement.

Rectus Femoris. As the only two-joint muscle of the quadriceps femoris, its function is undoubtly that of hip flexion and knee extension (Basmajian, 1978). (Two-joint function is expanded on in the Hamstrings section.) Fuijiwara and Basmajian (1975) state that the effect of contraction of rectus femoris is not limited to one joint and influences knee extension more than hip flexion. This is apparently confirmed by Houtz and Fischer (1959) who demonstrated that during cycle pedalling rectus femoris was more active during knee extension than hip flexion. In single joint action of knee extension or hip flexion Carlsöö and Fohlin (1969) demonstrated activity in rectus femoris.

Markee et al. (1955) stated the possibility of limited action to a single joint with which both Fuijiwara and Basmajian (1975), Basmajian (1957) disagree.

It is also a lateral rotator of the hip and not a medial rotator (Basmajian, 1978), and it aids in abduction of the hip (Wheatley and Jahnke, 1951).

It is usually supplied by two branches of the femoral nerve (Wheatley and Jahnke, 1951; Markee et al., 1955; Grant, 1975).

Vastus Medialis. A variety of functions have been attributed to this most controversial component of the quadriceps.

De Palma (1954) attributes the "screw home" movement to the vastus medialis, while a protective function against valgus joint stress is stated by Palmer (1938).

The most accepted function is that of patellar alignment due to the obliquity of the lower fibres (Lieb and Perry, 1968; Basmajian, 1970; Basmajian et al., 1972; Brunnstrom, 1972; Last, 1972). Perhaps the most controversial literature relates to the selective action of vastus medialis as a terminal extensor of the knee. Confusion arises as to the specificity of the term "selective" as this can mean "exclusively" a terminal extensor or "mostly or primarily active" in the final degrees of extension. Nevertheless both these views or variations on these views imply or state a selective action in terminal extension.

Lewin (1952) states that the vastus medialis is responsible for the last 10^{o}-15^{o}, while Nicoll (1943) states the last 15^{o} of extension even against minimal stress cannot be effected without the vastus medialis. Steindler (1955) emphasizes that vastus medialis especially is responsible for final forceful knee extension; and Smillie (1970) emphasizes the importance of vastus medialis as a terminal extensor, stating it to be the "key of the knee". He does, however, acknowledge that it may be used throughout the range of extension in overcoming marked resistance. Gardner et al. (1975) more cautiously state that, "it is usually stated that the vastus medialis contracts strongly only during the last phase of extension, but this has not been confirmed".

The original concept of vastus medialis as a terminal extensor appeared to originate from the clinical observation of apparent selective wasting of the vastus medialis and loss of function at the knee manifested by lack of knee extension (Lieb and Perry, 1968). This was well accepted (De Lorme and Watkins, 1948; Steindler, 1955; Brueckman, 1962; Mennel, 1964; D'Eshoughes and Waghemacker, 1970) and is repeated in the current Gray's Anatomy (1973).

However, Brewerton (1955) in an early EMG study refuted the current concept in a series of experiments. Subsequently, further publications by others concurred, refuting the selective action, and demonstrating activity throughout extension without a significant increase in vastus medialis compared with other heads (Pocok, 1963; Close, 1964;

Francis, 1967; Hallen and Lindahl, 1967; Basmajian, 1970; Lesage and Le Bar, 1970; Lieb and Perry, 1971; Perry, 1972; Knight et al., 1979). Lieb and Perry (1968) reporting an *in vitro* mechanical study, also stated their views opposing terminal extension as a function, and with their later paper (1971) state that the apparent marked wasting of vastus medialis is an indication of generalized quadriceps weakness manifested more in the vastus medialis because of its thin fascial covering and the obliquity of its fibres. In addition, the terminal 15° of extension requires a 60% increase in force.

It would appear then, that the matter had been resolved, i.e., that there is no selective terminal extension function, although as indicated previously it has not been acknowledged by the textbooks. Finally, however, Francis and Scott in 1974 limited the terminal extension of the knee in one group during exercise and compared the hypertrophy of the muscles of this group to a control group who were allowed to fully extend the knee. A significant difference was found in the hypertrophy of the vastus medialis in the control. Perry (1975), however, was critical of the technique used by Francis and Scott and their response to the criticism has not led to any final agreement on vastus medialis function as a terminal extensor.

This muscle is also supplied by two branches of the femoral nerve (Woodburne, 1957; Lieb and Perry, 1968).

Vastus Lateralis. This muscle is a powerful knee extensor and no specialized functions have been attributed to it (Basmajian, 1978). It is more active keeping the knee extended during hip flexion with the tibia medially rotated than laterally rotated (Wheatley and Jahnke, 1951). The efficiency of its pull is dependent on the corresponding contraction of vastus medialis (Lieb and Perry, 1968).

Vastus Intermedius. This muscle is a powerful knee extensor and no specialized function has been attributed to it (Basmajian, 1978).

Hamstrings (Fig. 20)

The muscles on the flexor aspect of the thigh are collectively known as the "hamstrings" and are the biceps femoris, semitendinosus and semimembranosus.

Biceps Femoris

The long head of this two-headed muscle arises as a common tendon with semitendinosus from the ischial tuberosity. The short head arises from the lateral lip of the linea aspera. The major part of the muscle inserts into the head of the fibula dividing into two slips around the fibular collateral ligament while other slips attach on to the lateral tibial condyle (Sneath, 1955). However, Marshall et al. (1972) do not agree with the precise details of the insertion.

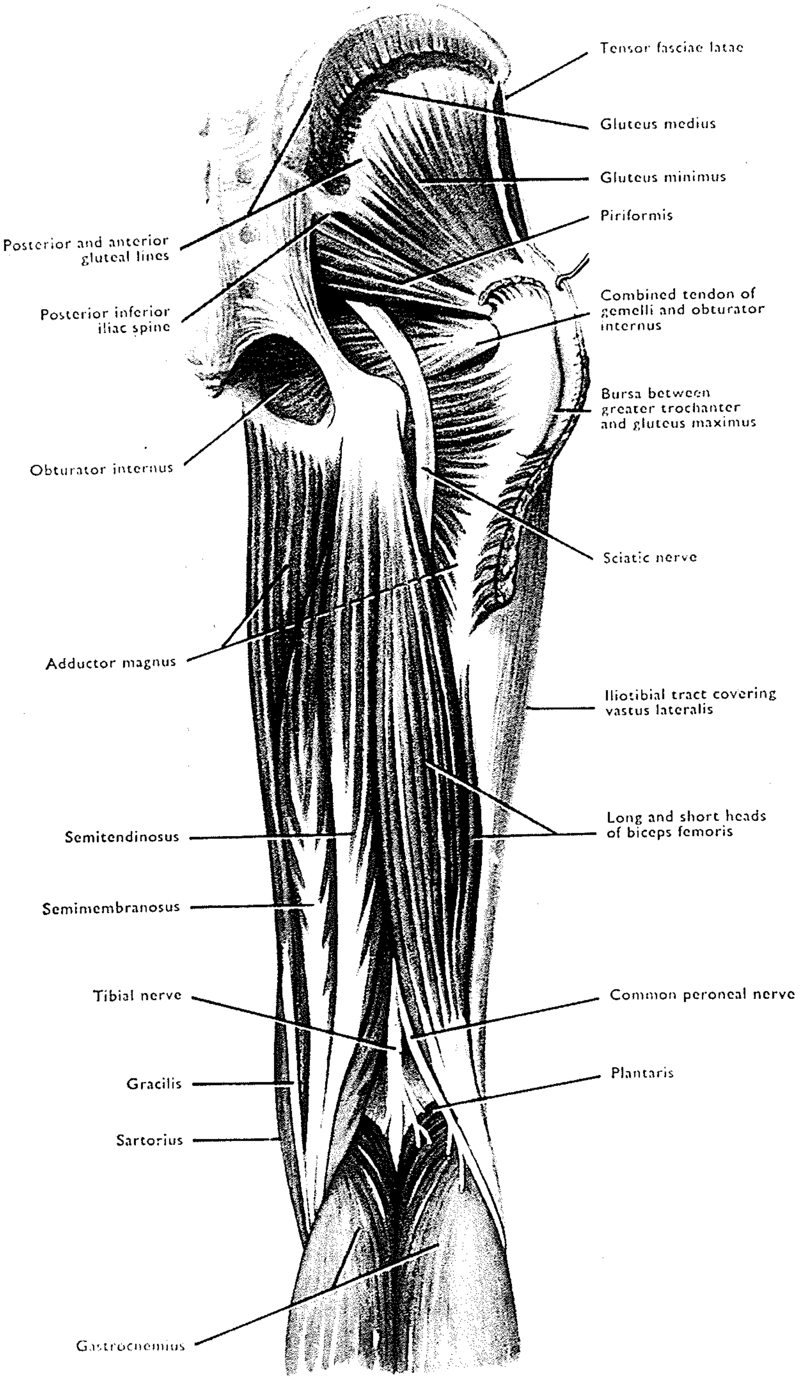
Nerve Supply and Blood Supply of the Biceps Femoris

The short head may be absent and the dual nerve supply from both the tibial and peroneal parts of the sciatic nerve (short head via the peroneal branch) (L_5, S_1, S_2) indicates the derivation from both ventral and dorsal musculature

FIGURE 20.

RIGHT THIGH--Posterior view showing hamstrings.

(from Cunningham, 1972)


Tensor fasciae latae
Gluteus medius
Gluteus minimus
Piriformis
Posterior and anterior gluteal lines
Posterior inferior iliac spine
Combined tendon of gemelli and obturator internus
Bursa between greater trochanter and gluteus maximus
Obturator internus
Sciatic nerve
Adductor magnus
Iliotibial tract covering vastus lateralis
Semitendinosus
Long and short heads of biceps femoris
Semimembranosus
Tibial nerve
Common peroneal nerve
Gracilis
Plantaris
Sartorius
Gastrocnemius

(Gray, 1973). The blood supply is via the perforating branches of the profunda femoris artery (Gray, 1973).

Semitendinosus

This muscle arises as a common tendon with biceps femoris from the ischial tuberosity, inserting via a long tendon into the tibia with the muscles of the pes anserinus. A tendinous interruption is usually present (Gray, 1973) which is often bridged by a number of fascicles (Markee et al., 1955).

Nerve Supply and Blood Supply of the Semitendinosus

This muscle is supplied by two branches from the tibial division of the sciatic nerve (L_5, S_1, S_2) (Markee et al., 1955). The blood supply is via the perforating branches of the profunda femoris artery (Gray, 1973).

Semimembranosus

This muscle originates from the ischial tuberosity and is deep to semitendinosus. Its main insertion with four minor slips (Cave and Porteous, 1958,1959) is into the tuberculum quadratus tendinitis, a tubercle on the posterior aspect of the medial condyle of the tibia.

Nerve Supply and Blood Supply of the Semitendinosus

There is usually a single branch from the tibial part of the sciatic nerve (L_5, S_1, S_2) (Markee et al., 1955; Gray, 1973). The blood supply is via the perforating branches of the profunda femoris artery (Gray, 1973).

Functions of the Hamstrings

The three hamstring muscles act both on the hip and knee joint. Basmajian (1957) demonstrated that two-joint muscles when moving a single joint do not have a selective muscle fibre contraction towards the joint moved. For example, proximal muscle fibres do not contract for movement of the proximal joint and vice versa as suggested in single joint movement of two-joint muscles by Markee et al. (1955). This is acknowledged not to apply to different or parallel heads of the same muscle (Basmajian and Latif, 1957, see quadriceps function).

As a group, the relative strength of the hamstrings as hip extensors is double that of the other hip extensors as the hip is flexed to 90° from extension. As a group they contribute between 31-48% of the total strength of the hip extensors depending on the hip position (Waters et al., 1974).

Biceps Femoris. In addition to hip extension and knee flexion, this muscle is active in lateral rotation of the extended hip and in adduction against resistance of the abducted hip (Wheatley and Jahnke, 1951). Furlani et al. (1977) demonstrated that the two heads of biceps had different functions in extension of the thigh. Also, only about half the subjects in their study demonstrated activity in the muscle during lateral rotation of the flexed knee, although this is commonly stated as a function (Gray, 1973). Skopintseva (1969) in an EMG study of the long head under dynamic conditions, differentiated both proximal and distal activity as being able to be both simultaneous and independent, and related this to its anatomical structure.

In an *in vitro* study Marshall et al. (1972) emphasized the functional importance of lateral stability of the knee by the expansion of biceps femoris tendon.

Semimembranosus and Semitendinosus. These muscles are active in hip extension and adduction against resistance of the abducted hip, and in flexion and medial rotation of the tibia at the knee (Basmajian, 1978).

Finally, there appears to be no specialized regional differences of muscles that act only selectively on one of the two joints (Basmajian, 1957). However, it is still mentioned as a possibility in Gray (1973) citing Markee et al. (1955), despite the fact that Markee's functional study was on dogs with only a morphological application to man.

Thigh Muscle Activity in Standing

It is well established that during relaxed standing both the quadriceps femoris and the hamstrings are minimally active or quiescent for varying periods of time (Weddell et al., 1944; Kelton and Wright, 1949; Wheatley and Jahnke, 1951; Joseph and Nightingale, 1954; Joseph et al., 1955; Portnoy and Morin, 1956; Joseph and Williams, 1957; Furlani et al., 1977; Basmajian, 1978).

EMG of the Thigh Muscles During Stair Climbing

There is little published data in this area (Basmajian, 1979). Joseph and Watson (1967) were the first to publish EMG data on stair climbing. They used a single channel telemetered system and cine with surface electrodes. Eight muscles were investigated separately. Six subjects were studied climbing three stairs with a rise of 16.5 cm and a tread run of 28 cm (slope 30.5°). The cadence was not stated. The electrode site for the hamstrings was over both the medial and lateral groups and the quadriceps was likewise generally represented by a pair of electrodes spanning vastus lateralis and vastus medialis (Battye and Joseph, 1966). The results were as follows:

Ascending

Quadriceps: Single phase of activity in early stance.

Hamstrings: Biphasic in swing and early stance.

Descending

Quadriceps: Biphasic at end of swing and start of stance.

Hamstrings: Single phase between end of stance to end of swing.

A second paper by Fujikawa (1968), using a cinematographic, analyzed ascending stairs only, however stair size, subject number, and technique to justify conclusions of "impellent" and "balance" muscles were not stated.

A third paper by Freeman et al. (1976) relates to a single step-down and a comparison with descending stairs would be inappropriate.

Finally, Townsend et al., recently published two papers (1978, 1978A). These two papers will be considered together. Nineteen subjects were studied, the majority of whom were highly trained athletes and seven of whom were not. Both groups sustained knee injuries. No cadence was stated though a velocity calculation ascertained the cadence to be 140 ascending and 153 descending. However, in a personal communication with one of the authors (R. Shiavi, 1979), he stated the velocity calculation to be related to progression in a horizontal plane and this converted to 87.6 ascending and 95.4 descending.

The rise was 19 cm, the tread run was 30.5 cm and the tread width 76 cm (stair slope 32°). The "on time" or synergy pattern was studied in three individual muscles and the medial hamstring group. The results were as follows:

Ascending (this showed the greatest number of variant patterns)

Quadriceps: Monophasic from early to middle stance.

Hamstrings: Monophasic from end of swing to middle of stance, but many biphasic variants.

Descending

Quadriceps: Biphasic: first phase start of stance, second phase end of stance. Half the subjects were monophasic through stance.

Hamstrings: Monophasic from end of stance through to swing ending in early stance. Biphasic activity was seen in a quarter of the subjects.

Although comparisons are made with the Joseph and Watson (1967) paper, it may be worth noting that the electrode position for the quadriceps and hamstrings is different in the two groups of investigations. For example, Joseph and Watson's paper (1967) had electrode positions for the quadriceps over a combined vastus medialis/vastus lateralis area as mentioned previously and Townsend et al. (1978,1978A) used rectus femoris. This may account for the "very different pattern" found in descending. No goniometric analysis was performed, and in the summary, the authors state "clearly, continuing investigation is required to relate efforts, muscle activity and motion".

SYME'S AMPUTATION

Introduction

James Syme (1799-1870), a Scot, was the last and greatest of the pre-Listerian surgeons. He was renowned in his day as the most eminent surgeon in the English-speaking world (Harris, 1961). Well-informed and well-travelled, he obtained improved results in the surgical treatment of disease at a time when anaesthesia and antisepsis were unknown. Joseph Lister came to work with Syme who was considered a leader in the movement towards conservative limb surgery, namely the removal of parts of bone that were damaged, rather than radical limb amputation which was the current vogue. Lister in fact married Syme's eldest daughter, Agnes (Davis and Christopher, 1972).

During Syme's occupancy of the Chair of Clinical Surgery at the University of Edinburgh, he developed and perfected many new surgical procedures. Time has outmoded all except his disarticulation amputation through the talocrural (ankle) joint, with the preservation of the heel flap to permit weight bearing on the end of the stump. This end-weight bearing amputation through the ankle joint still carries its originator's name. Following his earliest paper in 1843 (Syme, 1843A), Syme subsequently published further on this amputation (Syme, 1843B,1844,1845,1846,1857). However, a similar surgical technique was actually propounded earlier by Baudens (1842) in Paris but was discarded due to poor results (Shelswell, 1954). It is known that Syme developed his method 13 years after studying other amputation techniques in Europe in the late 1820's, and in fact he performed the first Chopart's amputation in Great Britain. This amputation was a disarticulation at the midtarsal joint with a long plantar skin flap and had greater success than the other techniques against the "hospital diseases", primarily infections, in the early 19th century. It is not known why Syme spent so many years developing the method of his ankle joint disarticulation (Harris, 1961).

The Syme's amputation was used successfully and widely in many parts of the world, but by 1920 the technique seems to have been abandoned by surgeons everywhere in the Western world except Canada where good results were being achieved (Dale, 1961). One reason why the below-knee amputation is preferred to the Syme is in the name of more radical treatment against the cost of repeated operations and long hospitalization which occurred if an attempt to preserve as much of the limb as possible was attempted (Lindquist and Riska, 1966). But the two major reasons against the use of Syme's procedure were, firstly the poor cosmetic results of the amputation with the bulbous stump, and secondly, the difficulty of providing an ankle mechanism in the limited space below the stump. For these reasons, therefore, the below-knee amputation became the operation of choice. These two major limitations were largely resolved by improved surgical techniques (Sarmiento et al., 1961; Mazet, 1968; Sarmiento, 1972) and the development of the Canadian plastic laminated Syme prosthesis with a rubber foot (Boccius, 1961). The rubber foot design was first introduced by Marks (1889)(Fig. 25).

The use by the Canadian group, at Sunnybrook Hospital, of an elastic material for the heel to provide a motion equivalent for the ankle, led to a review of the original concept of Marks (1889) and resulted in the solid ankle cushion heel (SACH) laminated foot introduced in 1956 by the Biomechanics Laboratory of the University of California. The SACH foot, now produced in a moulded version, accounts for the majority of artificial feet in Western Europe, North America and many other parts of the world (Wilson, 1972). The SACH foot is available from over nine commercial manufacturers and has been the subject of intensive testing (Daher, 1975).

Indications for Syme Surgery

The Syme's amputation is not indicated in patients with chronic occlusive vascular disease. An intact undamaged heel skin is essential. Severe crush injuries with contaminated open fractures of the metatarsus and tarsus, diabetics with infected or gangrenous process confined to the forefoot, foot deformities secondary to trauma (both physical and thermal), or congenital abnormalities and obvious intractable infection are all indications for the Syme procedure (Shea, 1972). It is also used in rare conditions, for example, focal scleroderma (Hinterbuchner et al., 1972) and is also appropriate when sensation loss occurs, e.g., leprosy (Srinivasan, 1973). The technique of the Syme's amputation is shown on Figure 21.

It is a successful amputation in children, having the advantage that the epiphyseal plate is left intact and the skin of the lower limb grows appropriately with the bone, the limb having minimal growth changes (Harris, 1956,1961; Thompson et al., 1957; Mazet, 1968; Davidson and Bohne, 1975).

Harding (1967) felt that the indications were few for adults, stating its lack of popularity particularly with women because the prosthesis is bulky and unsightly. Thus the below-knee amputation may be preferred. Mazet (1968) contends that this is no longer a problem with the surgical techniques available.

The Syme's amputation through the talocrural joint has, however, undergone few changes (Dale, 1961; Harris, 1961; Harding, 1967; Davis and Christopher, 1972). A smaller, less bulbous weight-bearing end due to bone trimming and early prosthetic fitting forming a more firmly fixed heel pad is a modification (Sarmiento et al., 1966). Despite some modifications

FIGURE 21. TECHNIQUE OF THE SYME AMPUTATION.

A) Skin incision on the medial side.

B) Skin incision on the lateral side.

C) Division of collateral ligaments from within the joint.

D) Dislocation of the talus downward from the mortise of the ankle joint.

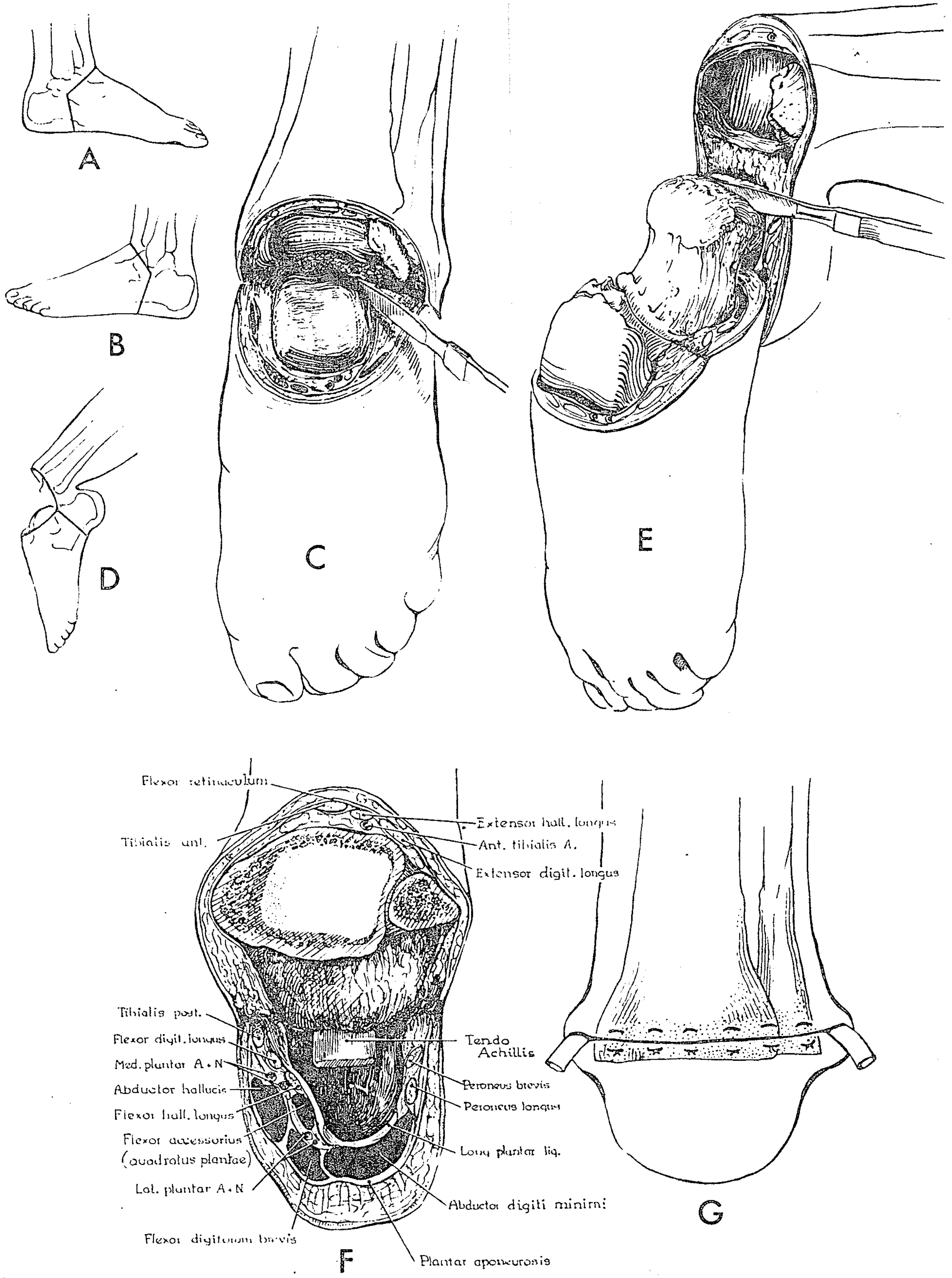
E) Talus dislocated. Calcaneus separated almost completely from the heel flap by subperiosteal dissection. The tendo Achilles is about to be divided at its insertion.

F) Anatomy of the field of operation after the talus has been removed from the heel flap.

G) Closure of the wound with drainage.

(from Harris, 1961)

Fig.21


A
B
D
C
E
Flexor retinaculum
Tibialis ant.
Extensor hall. longus
Ant. tibialis A.
Extensor digit. longus
Tibialis post.
Flexor digit. longus
Med. plantar A + N
Abductor hallucis
Flexor hall. longus
Flexor accessorius (quadratus plantae)
Lat. plantar A + N
Flexor digitorum brevis
Tendo Achillis
Peroneus brevis
Peroneus longus
Long plantar lig.
Abductor digiti minimi
Plantar aponeurosis
F
G

strict adherence to the basic principles and operative technique originally stated by Syme is essential for successful surgery (Harris, 1956,1961; Mazet, 1968; Shea, 1972)(Fig. 22).

Figure 23 shows a typical conventional stump.

The unique end-bearing characteristics of the Syme's amputation (Sarmiento, 1972) have been questioned by Hornby and Harris (1975) who state that the majority of patients require fitting with a prosthesis designed to relieve end-weight bearing.

The success of the Syme's procedure is well documented (Alldredge and Thompson, 1946; Warren et al., 1955) and long-term studies have established its effectiveness (Robertson, 1955; Dale, 1961; Gladstone and Iuliucci, 1961; Mazet, 1968). Specific longevity of 44 years (Ratliff, 1967), 64 years (Haig, 1972), 66 years (Catterall, 1967) and 74 years (Shelswell, 1954A) have been demonstrated. More recently with the greater awareness of energy expenditure of amputees, a comparative study of the energy costs of walking in lower limb amputees of different levels was undertaken (Waters et al., 1976). The results indicate that the amputee performance was significantly better the lower the amputation. Therefore, at the present time when preservation of function is the chief concern, they believe amputation should be performed at the lowest level.

Syme Prosthesis

In the introduction to this chapter, brief mention was made of the prosthesis used following Syme's amputation. Generally, early designs using wooden sockets and steel reinforced sockets (Winkley, 1910; Gaines-Erb, 1915) and even aluminum sockets (Marks, 1889) were available (Fig. 24 and 25) They had a variety of wooden and rubber feet and some were more complex

FIGURE 22. SYME AMPUTATION.

A) The proper level for transection of the tibia and fibula in the Syme amputation (from Harris, 1961).

B) Modified and conventional techniques. The conventional technique tends to produce a broad bulbous stump.
The modified technique necessitates:
  1) The metaphyseal flares of the tibia and fibula trimmed to create a narrow end.
  2) The heel flap fat locules are retained undisturbed by removing a thin wafer of the calcaneal plantar surface and fixing it to the distal tibia (from Shea, 1972).

Fig.22

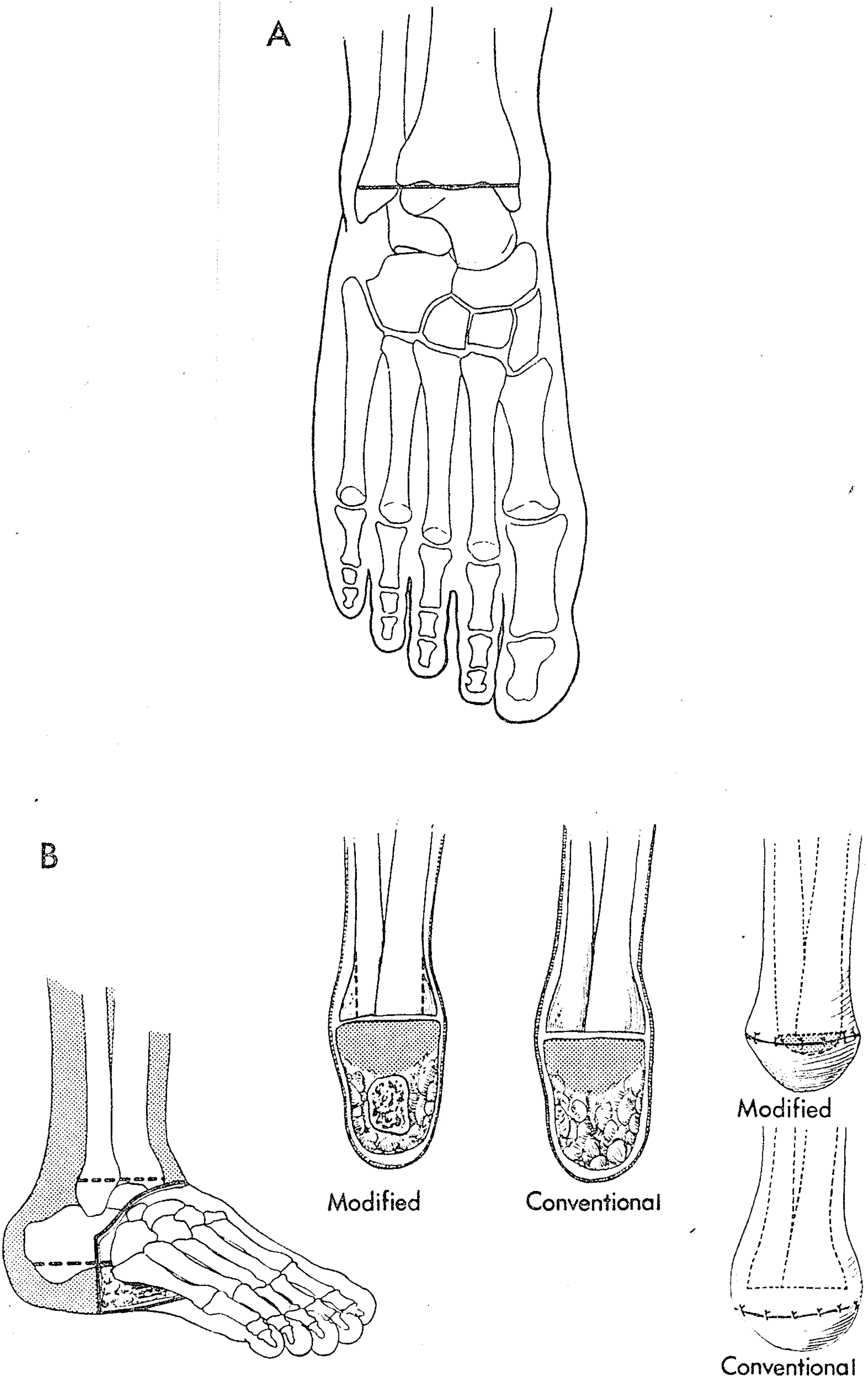


FIGURE 23. A CONVENTIONAL RIGHT SYME STUMP.

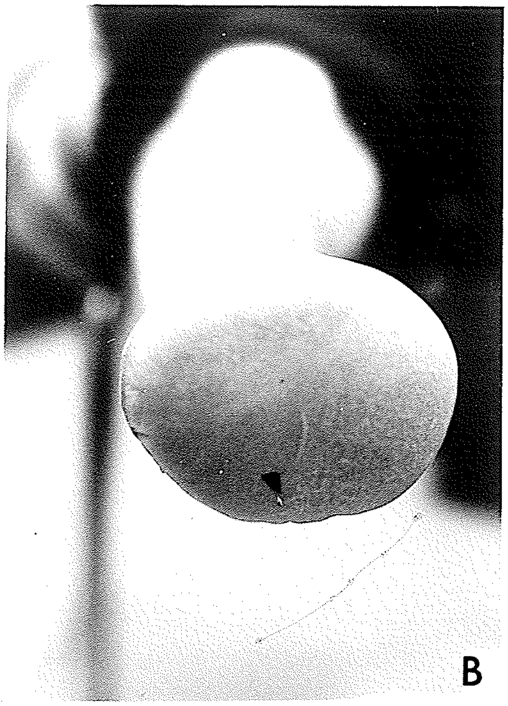
A) Anterior view- including knee.

B) Distal view.

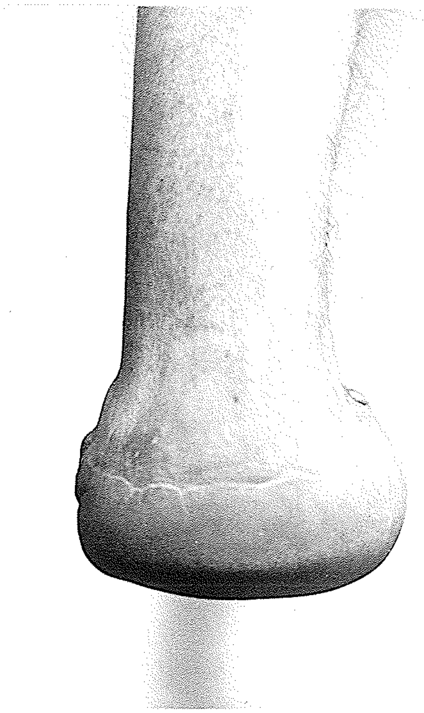
C) Anterior view- detail.

D) Medial view.

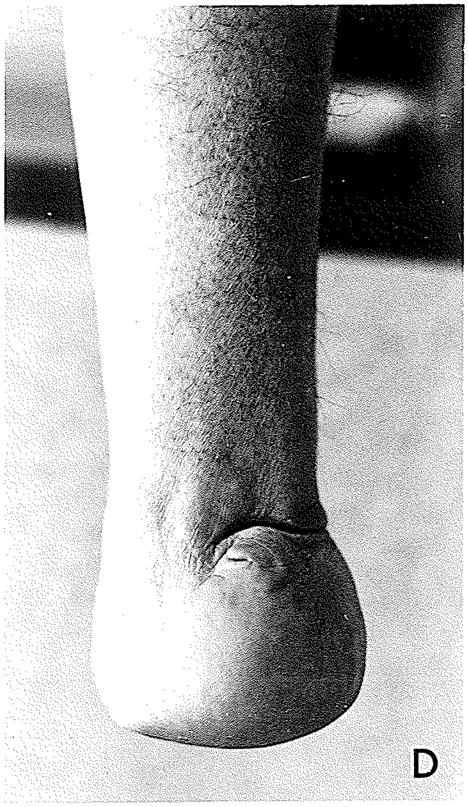
A



B

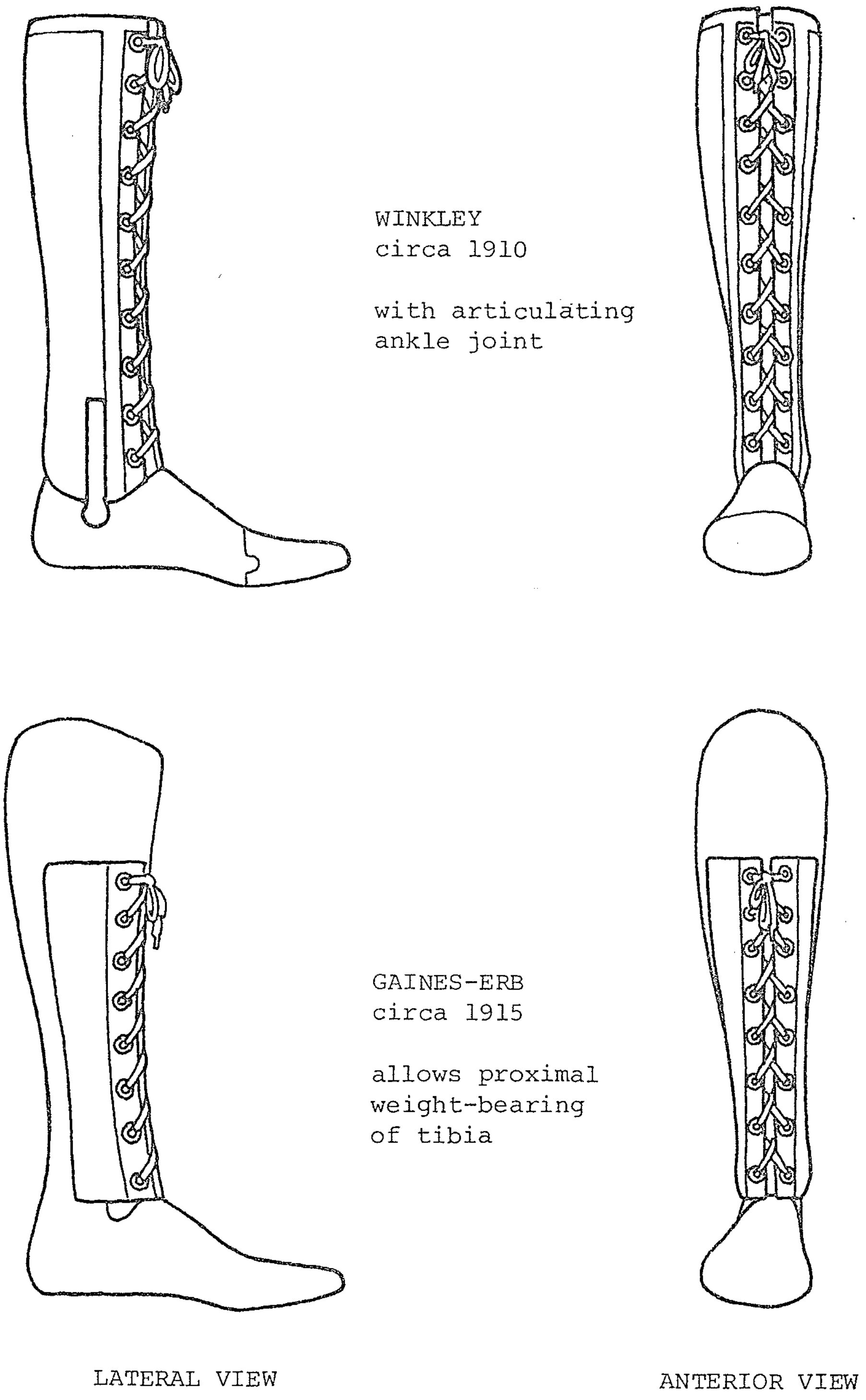


C



D

FIGURE 24. SYME'S PROSTHESES



WINKLEY
circa 1910

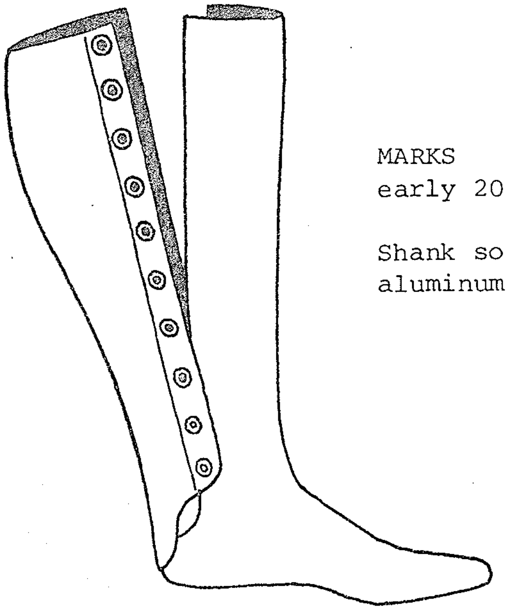
with articulating
ankle joint

GAINES-ERB
circa 1915

allows proximal
weight-bearing
of tibia

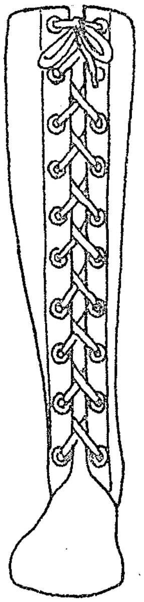
(From Wilson 1961)

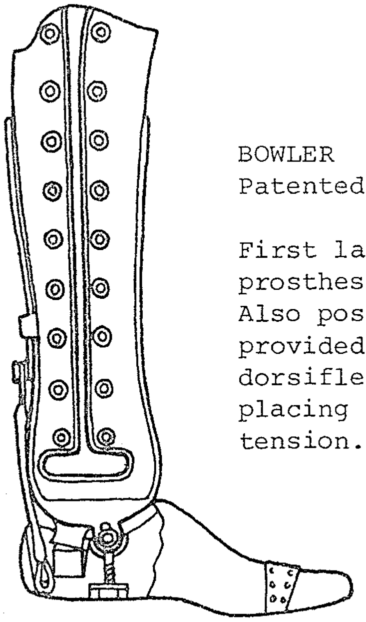
FIGURE 25. SYME'S PROSTHESES



MARKS
early 20th century

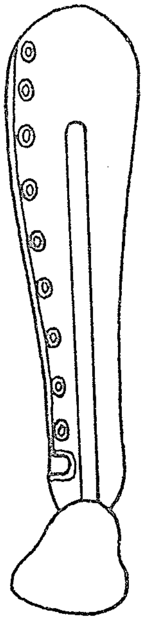
Shank socket in
aluminum with a rubber foot





BOWLER
Patented in 1919

First lateral opening prosthesis.
Also posterior cable provided resistance to dorsiflexion without placing ankle parts in tension.



LATERAL VIEW

ANTERIOR VIEW

(From Wilson 1961)

FIGURE 26. CANADIAN SYME PROSTHESIS.

A) Anterior view.

B) Medial view.

C) Posterior view.

D) Posterior view--showing interior.

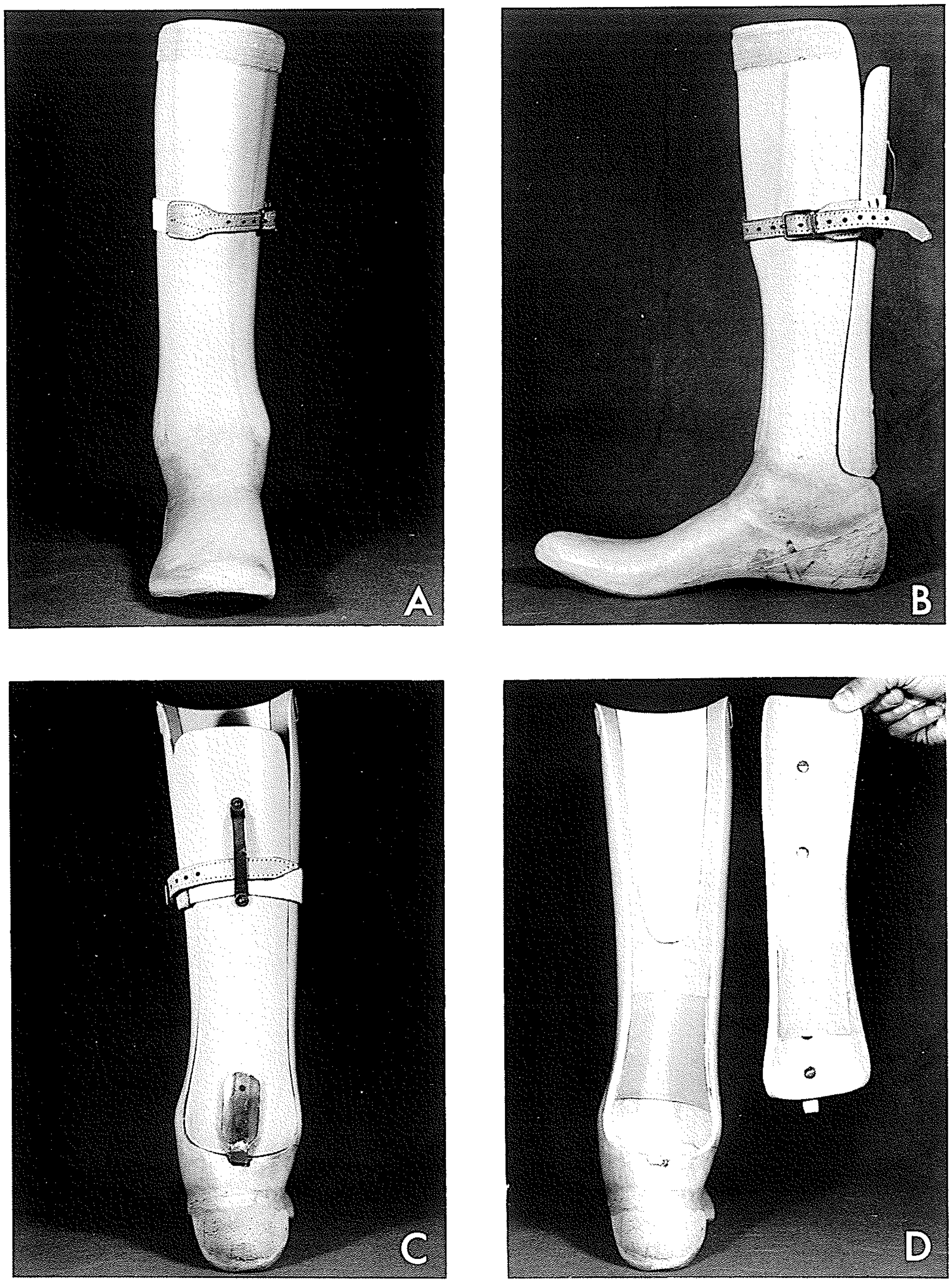

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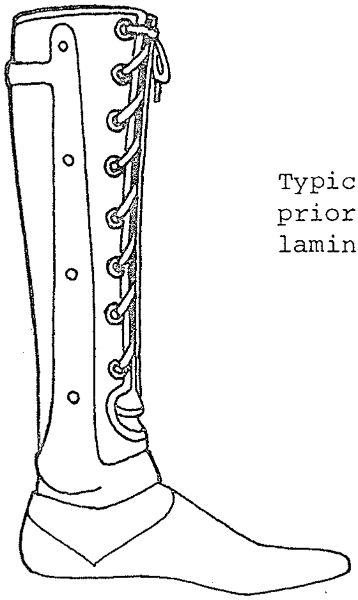
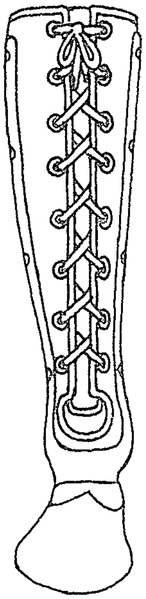
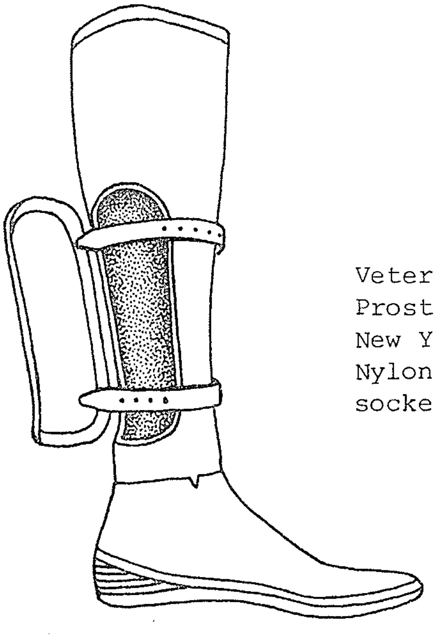
FIGURE 27. SYME'S PROSTHESES



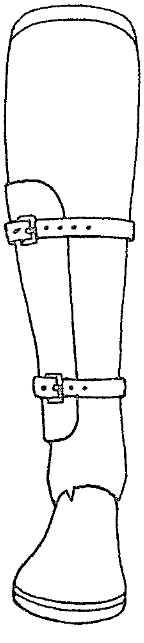
Typical prosthesis prior to the era of laminates.



LATERAL VIEW



Veterans Administration
Prosthetic Center
New York
Nylon-dacron-polyester socket.



MEDIAL VIEW

ANTERIOR VIEW

(From Wilson 1961)

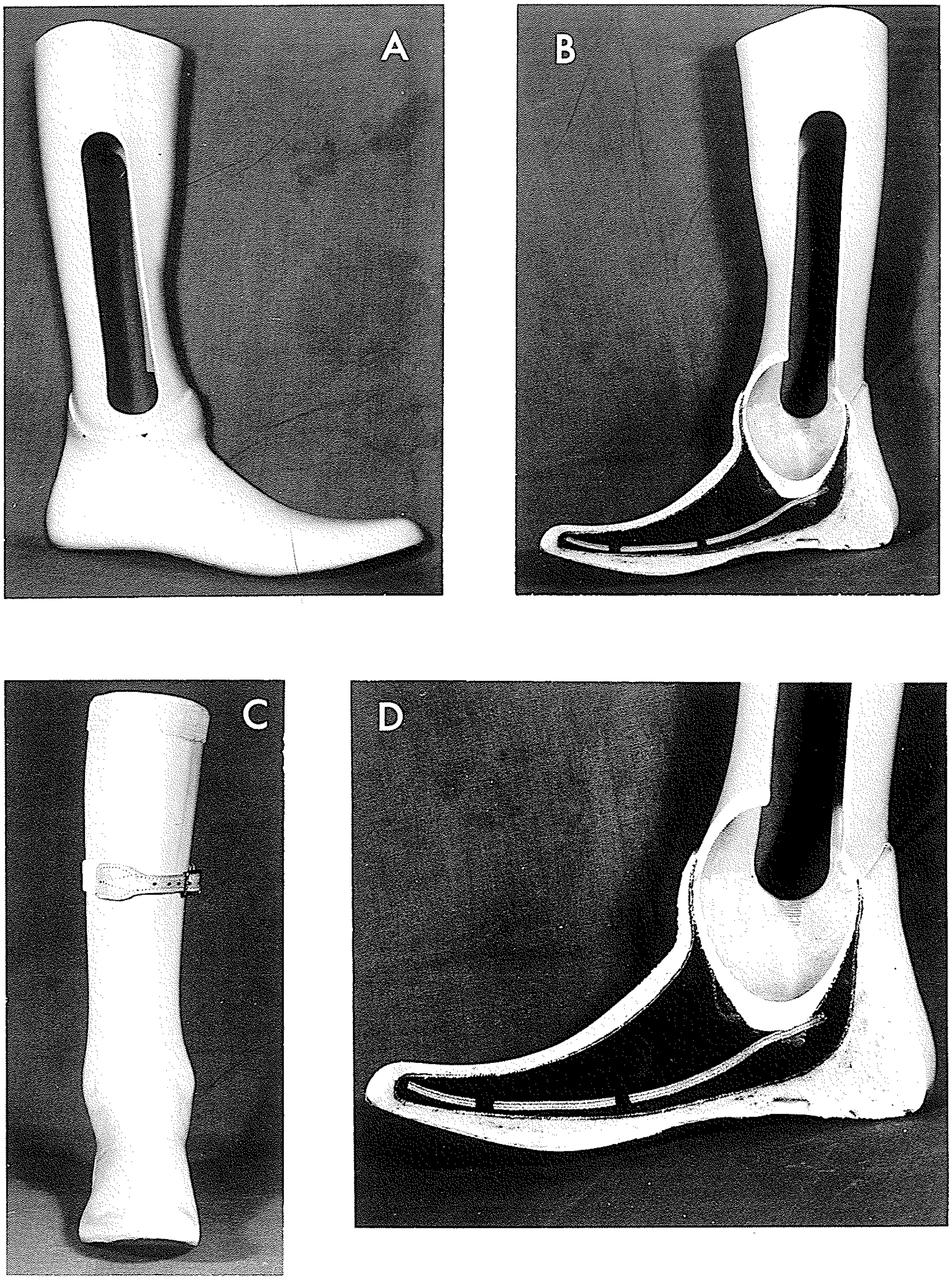
FIGURE 28. WINNIPEG TUCKER-SYME PROSTHESIS.

A) Lateral view.

B) Medial view with cut-away through SACH foot.

C) Anterior view.

D) Medial view of SACH foot.


A
B
C
D

in design accommodating for both cosmetic and functional use (Bowler, 1919)(Fig. 25). In the mid-fifties the Canadian Department of Veteran Affairs developed the firstplastic posterior opening Syme's prosthesis with a SACH foot which is still extensively used (Fig. 26). The Veterans Administration in New York designed a prosthesis with either a posterior or medial opening Syme prosthesis (Fig. 27). These were not without early failures (Wilson, 1961). More recently in 1965 the University of Miami Prosthetic and Orthotics Laboratory developed a windowless prosthesis with an expandable inner wall that is also patellar tendon weight-bearing and affords greater contact and stability (Sinclair, 1972). At approximately the same time other centres developed their modifications, e.g., the balloon prosthesis (Mazet, 1968).

Perhaps the latest in Syme prostheses is the Winnipeg Tucker-Syme prosthesis. This has an open frame design of a self-suspending socket with a rubber-bonded foot. This prosthesis is exclusive of metal and wood and has a continuous external skin with increased moisture resistance, added lightness and improved cosmesis when compared with previous designs (Cochrane, 1979; Cochrane and Tucker, 1979)(Fig. 28).

A REVIEW OF DATA ACQUISITION OF EMG SIGNALS

The EMG signal has to be received via electrodes, transmitted from the muscle site to the amplifiers, amplified, left in the unprocessed, natural or "raw" state, or processed and finally recorded for a method of evaluation.

Electrodes

Electrodes may be surface or indwelling. The indwelling electrodes are either inserted subcutaneously (Broughton, 1976), or intramuscularly (Basmajian, 1978).

Surface Electrodes

These are of varying size and type, constructed of metal (e.g., silver, platinum, or gold), or a combination of a metal and one of its salts, e.g., silver-silver chloride (Quanbury, 1972). The silver-silver chloride commercial electrodes, e.g., Beckman type are widely used and are excellent (Basmajian, 1978). The use of a metal and its salt minimizes the effect of electrode polarization. Furthermore, a bipolar arrangement, i.e., a pair of "active" electrodes with a separate ground electrode, greatly reduces any common noise signal picked up by both active electrodes (Quanbury, 1972).

Skin preparation was more critical prior to the advent of high input impedance amplifiers. It involved shaving, washing, alcohol rubbing and abrading the skin with abrasives ranging from coarse tissue paper to fine sandpaper. Scott et al. (1970) studied the characteristics of different applications of electrodes in their report. However, some present day amplifiers with high input impedance (> 7 megohms) allow the skin resistance not to be a critical factor (Quanbury, 1979). The development of high impedance amplifiers is recent and was dependent on the state of the art of semiconductor technology. These newer amplifiers allow a greater amount of the muscle signal to reach the amplifiers. In addition, and of even more importance, is their stabilization of the variations in pathway resistance that may occur during movement and time so that these variations do not affect the muscle signal.

Surface electrodes are usually applied with a saline contact paste or gel to increase conductivity and reduce electrode/skin resistance.

The electrodes may be attached to the skin by adhesive tape, elastic straps, double-sided adhesive collars or a vacuum. The latter two techniques are very suitable because they restrict any relative motion between the electrode and the skin and form a seal which prevents the saline contact jell from drying out (Quanbury, 1972). The needle electrode first assembled by Adrian and Bronk (1929) was also modified by Becker and Chamberlin (1960). This type of electrode is primarily used for diagnosis. It is extremely localizing and therefore is seldom used in kinesiology (Basmajian, 1978). Grieve (1976) is more emphatic, stating that this type of electrode is utterly unsuited for studies of movement and should be dismissed. In addition, with large muscle movements the artifact generation is increased and in general needle electrodes have a much higher electrical impedance than do surface electrodes (Quanbury, 1972). The variation in needle electrodes is carefully documented by Basmajian (1978).

The fine-wire electrodes introduced by Basmajian and Stecko (1962) are routinely used in kinesiological studies. They are inserted into a muscle, using a small (27) gauge hypodermic needle as a carrier which is withdrawn, leaving two fine wires of 25μ *in situ*. The wires are connected to small brass springs (Basmajian, 1978) or wound around a contact post (Brandell, 1979) at the insertion site for signal transmission.

APPLICATION AND LIMITATIONS OF SURFACE VERSES INDWELLING ELECTRODES

Fine wire electrodes have the following advantages:

1) They are extremely fine and therefore do not limit or inhibit measurement through pain.

2) They are easily implanted and withdrawn.

3) They are as broad in their pick-up from a specific muscle as are the best surface electrodes.

4) They produce sharp signals.

(Basmajian, 1978).

Surface electrodes have the following advantages:

1) They are non-invasive.

2) They are convenient to use and easy to apply.

3) They cause no discomfort to the patient.

4) They are less costly than fine wire electrodes and can be purchased commercially.

5) They are more acceptable to patient and doctor than fine wire electrodes.

The choice of electrodes is dependent on the objective of the study (Komi and Buskirk, 1970; Kelly, 1971; O'Connel and Gardner, 1963). It should not be made on the basis of convenience or lack of understanding of EMG technique. This means that the limitations of both techniques should be considered.

The major limitation of surface electrodes is that they can be used only with superficial muscles, and then pick up is generally widespread. This means that they should not be used in the study of fine movement or for deep muscles (Basmajian, 1978). Perhaps the chief usefulness of surface electrodes is when the simulated activity or interplay of activity is being studied in a fairly large group of muscles under conditions where palpation is impossible, e.g., in the muscles of the lower limbs during walking (Basmajian, 1978). The case is put more strongly by others who feel that where there is a large muscle mass the surface technique is more appropriate because it samples a greater muscle mass, while with the fine wire technique only a small volume of muscle is sampled (O'Connell and Gardner, 1963; Lippold, 1967; Jonsson and Komi, 1973; Grieve, 1976).

Gouisset and Maton (1972) in comparing both surface and indwelling electrodes suggested that the activity of the muscles fibres near the surface is representative of all the fibres involved in a given activity; and that because of the linearity of the relationship between both surface and indwelling electrodes the apparent difference was a constant coefficient.

Bottomley (1965) stated that surface electrodes get most of their signals from a fairly small group of muscle cells immediately beneath them. Kramer et al. (1972) showed that the electrode surface was not as important as pressure of application and interelectrode distance. The closer the electrodes the more localized is their pick-up.

Komi and Buskirk (1970) have studied the reliability of both suface and wire electrodes during a day, and from day to day. Within one day the surface electrode reliability was higher (0.9 to 0.6 reliability coefficient); and the day to day reliability of surface electrodes was .7, and for practical purposes nil for fine wire. In isotonic conditions the day to day reliability for surface electrodes was greater than .9.

Surface electrodes have been used extensively in the study of the quadriceps and hamstrings (Wheatley and Jahnke, 1951; Pocock, 1963; deAndrade, 1965; Battye and Joseph, 1966; Steiner, 1969; Tiselius, 1969; Gough and Ladley, 1971; Murphey et al., 1971; Merrifield and Kukulka, 1973; Jackson, 1973; Arsenault and Chapman, 1974; Brandell, 1976,1977; Viitasalo and Komi, 1978; Wahrenberg et al., 1978; Alon et al., 1979; Boccardi et al., 1979; Kumamoto, 1979; Shirai et al., 1979; Weiss and Hayes, 1979; Grantham and Peat, 1979; Hardt and Mann, 1979; Wolley et al., 1979; Vanden-Abeele et al., 1979; Shiavi et al., 1979).

Transmission of EMG to Amplifier

This may be done with cable ("solid wire"), or via a radio frequency signal from the subject to the receiver. The latter method is called "EMG telemetry". A number of excellent telemetry systems have been described (Winter and Quanbury, 1975) and good standard equipment is available from commercial sources (Basmajian, 1978).

The EMG signal from the electrode may be "buffered" or "pre-amplified" at the electrode site when using cables, to present a lower impedance to the long flexible cables (Milner, Basmajian and Quanbury, 1971). The cables must be of a special kind to reduce the noise due to the tribo electric phenomenon (Quanbury, 1972). Basmajian (1978) uses a device called a dual field-effect transistor source-follower, which, while giving no amplification at the source, completely removes interference noise from the cables.

Amplifiers

There are many commerical EMG amplifiers available. Many kinesiologists are confined to using discarded electroencephalographic equipment and if this equipment is not appropriately adjusted for the EMG spectrum it records a completely different phenomenon and should not be used (Basmajian, 1978).

Basically an electromyograph is a high gain amplifier with a frequency range of about 10 to several thousand hertz. Hayes (1960) suggests that the sharply peaked spectra of motor unit potentials derived from surface electrodes make the use of amplifiers with a limited frequency response of 20-200 Hz practical. This allows amplifier noise, general non-muscular "tissue-noise",and movement artifact to be eliminated without significant loss of motor unit potentials. Basmajian (1978) states that 1000 Hz as the upper limit of the band width is excellent but prefers the provision of high and low frequency cut-off switches.

EMG Signal Display

The EMG signal may be displayed in its unprocessed or natural "raw" form or be processed in some way.

There are a variety of processing methods of which the most common are summarized by Winter et al. (1979).

First, the natural or "raw" EMG signal may be full wave rectified, and then low pass filtered to form a linear envelope. The signal may also be integrated in a variety of ways. First, integration may occur over a period of time during an activity; or integrated over a small time unit (e.g., 50 ms) and then reset; or integrated to a preset voltage, then reset.

The full wave rectification and low pass filtering which produce a linear envelope are the most suitable of the methods for demonstrating peak activity. This is because the amplitude of the envelope closely follows the amplitude of the peaks of the raw EMG, as opposed to the integrated signal which is a given measure of the product of the EMG amplitude and time (Quanbury, 1979). It is useful to have the capability to display the raw and processed signal as a method of checking the display.

Another method of EMG display is to consider instead of the amplitude of the EMG signal, the number of turning points (NTP) of the signal across a baseline. This detects low levels of activity but does not emphasize the peaks (Grieve, 1976). This is a similar concept to the zero crossing (of the baseline) described by Fusefeld (1970). A comprehensive discussion of this area is given by Basmajian et al. (1975).

Recording EMG Signals

The most convenient and cheapest recorder is the pen-writing recorder which has the serious drawback that the inherent responsiveness of the mechanical system limits the frequency response to 100 Hz. In recent years, however, modified pen recorders have been produced that have proved to be efficient for many experiments (Basmajian, 1978). The drawback mentioned above by Basmajian would be more significant in apparatus that was measuring the raw signal as opposed to the processed signal which has a lower frequency band width (Quanbury, 1979).

Recorders may have up to 24 independent channels circuiting on a moving strip of paper. There are other types of recorders using both ultraviolet and infrared components of the electromagnetic spectrum relayed via fibre optic systems (Honeywell, 1969).

The most accurate technique is the multichannel FM tape recorder allowing great versatility of manipulation of the EMG data. However, at present this system is expensive (Basmajian, 1978).

Evaluation of the EMG Signal

There are a variety of ways the EMG signal can be evaluated. Perhaps the most comprehensive method at present is the use of an analog to digital conversion of the EMG signal using a computer which allows the most comprehensive qualitative and quantitative measurements to be made with the appropriate programming.

Basmajian, from the first edition (1962) to the fourth edition (1978) of his classic text "Muscles Alive" repeats the statement that the evaluation of recordings, whatever their type, is the most abused part of electromyography. At the 4th Congress of the International Society of Electrophysiological Kinesiology (ISEK) in Boston, Mass., Winter et al. (1979) expressed concern regarding the units terms and standards in the reporting of EMG research.

The computing method is not necessarily superior to other techniques and Basmajian (1978) states human judgement and tedium is still applicable in the writing of computer programs; and the problems and variations of analytical methods and the standard for quantification still exist when using a computer.

Should expensive and sophisticated computerization not be available other methods of EMG evaluation are possible.

Other methods of studying the EMG signal are used for phasic activity or "on time": a particular duration for which a muscle is active during a specific activity. In this type of analysis there has to be an establishment of a baseline signal which is used as a reference point. This can be a value usually stated as a unit above the basic relaxed muscle signal, which, since there is no signal from a relaxed muscle (Clemmesen, 1951; Basmajian, 1952; Ralston and Libet, 1953) is the noise level of the equipment. This can also be a percentage of the maximum activity level, for example, Battye and Joseph (1966) used a level of 15% of the maximum contraction as the indication for when the muscle was active. This type of analysis can be used when the understanding of muscle co-contraction is required (Joseph and Watson, 1967; Rau and Vredenbregt, 1973; Townsend et al., 1978,1978A). However, if this method is used exclusively the shape and intensity of the signal are not known, particularly during complex movements.

Some investigators have attempted to emphasize the general form of the envelope of the wave form by tracing around it (Sheffield et al., 1956). Others have studied peak EMG activity (Lopata et al., 1977; Johnson, 1978; Wahrenberg et al., 1978; Anderson et al., 1979; Hurbon et al, 1979).

The amplitude component can be expressed as a percentage of the maximum during a specified contraction; or the maximum that is seen during the activity, all other units being expressed as a percentage of this maximum. Hirose et al. (1974) showed that visual evaluation compared favourably with computer analysis.

Perhaps the last word should be from Dr. J.V. Basmajian (1978) "Experience has shown that the easiest and in most cases, most reliable evaluation is by the trained observer's visual evaluation of results colored by his knowledge of the technique involved."

CHAPTER III

MATERIALS AND METHODS

INTRODUCTION

Two groups were investigated, a normal group and a Syme's amputated group. The four major areas of interest in this investigation of stair climbing of the two groups are as follows:

1) Temporal data
2) Electrogoniometric data
3) Peak EMG sequence data
4) Relationship of peak EMG to electrogoniometric data

The descriptive data for each group is shown in Table 1.

The criteria for the normal group were that they had no previous history of injury, e.g., fractures, or surgery of any kind, e.g., meniscal removal or ligamentous repair of lower limb, and that they could climb up stairs, step over step.

The criteria for the Syme's amputee group was that they had a disarticulated amputation through the talocrural joint in one limb only and that they had no other history of knee injury or surgery to either knee or lower limb. They would also have to be good functional walkers and capable of climbing stairs step over step. This, of course, excluded most of the older Syme's group and those with additional problems which might influence stair climbing. There was one exception in the Syme's group to those criteria in that one subject had minor surgery at the age of two to one knee.

Although there are many ways of climbing stairs the step over step front approach was used for all subjects.

TABLE 1 - DESCRIPTIVE DATA

NORMAL GROUP

18 Subjects (12 male, 6 female)

AGE:	MEAN	27.7 ± 7.6 yrs.
	RANGE	20 - 53 yrs.
WEIGHT:	MEAN	66.5 ± 11.2 kg.
	RANGE	49.9 - 89.2 kg.
HEIGHT:	MEAN	171.3 ± 8.0 cm.
	RANGE	152.5 - 181.5 cm.

AMPUTEE GROUP

10 Subjects (9 male, 1 female - 5 right, 5 left)

AGE:	MEAN	40.7 ± 13.0 yrs.
	RANGE	19 - 56 yrs.
WEIGHT:	MEAN	76.2 ± 10 kg.
	RANGE	62.6 - 92.4 kg.
HEIGHT:	MEAN	176.3 ± 6.1 cm.
	RANGE	165.1 - 185.4 cm.

STUMP LENGTH: 39 ± 5.4 cm.

WINNIPEG TUCKER-SYME PROSTHESIS: 4 Subjects

CANADIAN SYME PROSTHESIS: 6 Subjects

The "handedness" of the subjects was not evaluated. Because of the difficulty in defining precise criteria for "footedness" this was also not evaluated. For example, to define that a person is right-footed because he kicks a ball with that foot may have limited application to an individual who is not a football player.

Subjects were selected on the basis of availability in both groups.

Because of the strict criteria required for the Syme's group, the minimum number accepted by the Thesis Advisory Committee was met but not exceeded.

The Syme's subjects were all at some time patients of the following prosthetic departments or clinics in the city: 1) Rehabilitation Centre, Health Sciences Centre, 2) Deer Lodge Hospital, 3) The Municipal Hospitals, 4) Anderson's House of Orthopaedic Appliances Ltd.

Statistics and data regarding availability of these patients were also obtained from the Manitoba Health Services Commission (Toll, 1977, 1979) some of which is summarized in Table 2.

All patients contacted for this study agreed to participate even though they were aware that there would not be any financial remuneration. However, all the studies were done at a time of the subjects' choosing and transportation was offered in all cases. Each amputee was sent a letter of thanks for participating in this study.

STAIRS

Five stairs were ascended and descended three times for each leg. The middle step was the only step to be analyzed to minimize acceleration and deceleration effects during climbing.

TABLE 2.

SYME'S AMPUTATION

INCIDENCE OF SURGERY IN MANITOBA:

1972 - 5
1973 - 7
1974 - 6
1975 - 7
1976 - 6
1977 - 3
1978 - 5

NUMBERS OF PROTHESES MADE IN MANITOBA

	CANADIAN TYPE	MODIFIED	TOTAL
1977	9 (a)	19 (b)	28
1978	4 (c)	14 (d)	18

(a) = 9 patients.

(b) = 15 patients, 4 received two devices, one of whom was a double amputee.

(c) = 3 patients, 1 received two devices.

(d) = 10 patients, 1 received two devices, one double amputee received four.

COSTS (1978/79)

Surgery and six weeks postsurgical care	$167.50
Anaesthesia basic value, plus time value fees	24.50
Prothesis: Canadian Type	695.00
Modified Type	717.00

The size of the stairs was as follows. Each step had a rise of 23 cm, a tread run of 28 cm, and a tread width of 99.7 cm. The resultant stair slope was 38° (Fig. 29). The stairs were placed against a corner of the laboratory so that the walls would have an added degree of security for the patients. No hand rails were constructed though subjects were allows to place a hand on the wall for balance if necessary. At the top of the stairs a platform 63.5 cm x 106.7 cm had a protective guard rail. The minimum height clearance for all stairs was 1.98 m and the maximum was 2.44 m. The clearance over the platform was 2.17 m.

The stairs were used in the University of Manitoba Medical College prior to this study and met the Provincial Building standards with respect to dimensions. The provincial standards allow a combination of rise and tread which affords a slope range of 20° to 41°. Shinno (1968) considers a slope of between 20° to 50° as normal. Fitch et al. (1974) in a study of stair dimensions state that stairs having a rise of 16.0-22.6 cm and a tread run of 19.6-36.1 cm result in stairs that are likely to be safer to climb. Their size of step for "fairly low rates of energy expenditure (at normal speeds) and low rate of missteps" is a rise of 10.2-17.8 cm and a tread run of 27.9-35.6 cm. Their stair slope affords a range of 16° to 33°. The significance of slope and stair size will be discussed in the results and discussion chapter.

The subjects were asked to climb at their own speed, and not to rush. The reason for this was predicated on the findings of Bard and Ralston (1959) where a correlation was made between differet cadences and energy expenditure. It was found that the free cadence had optimal energy consumption and both higher and lower speeds had greater energy expenditure.

FIGURE 29. STAIR DIMENSIONS.

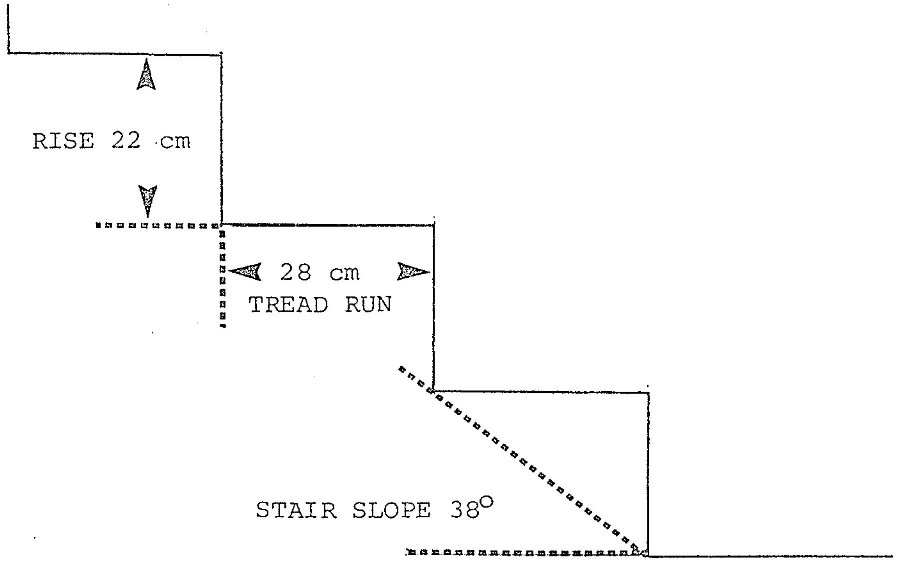


FIGURE 30. FOOT SWITCH CIRCUIT.

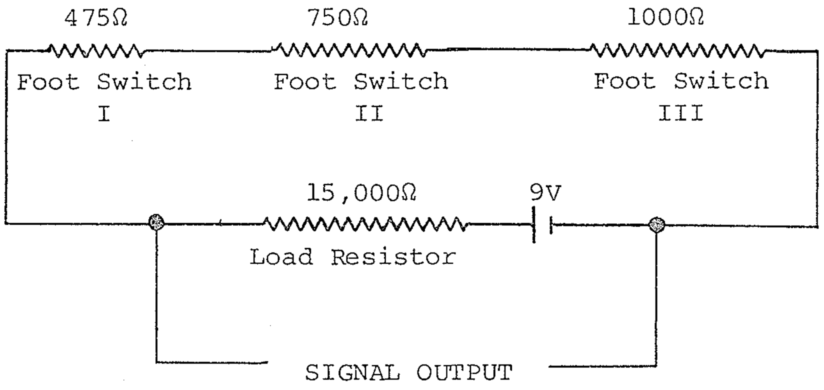
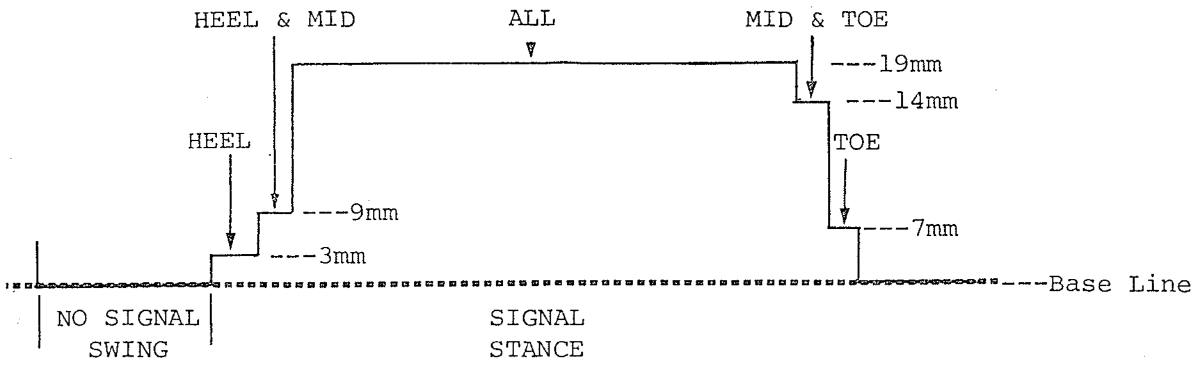


FIGURE 31. FOOT SWITCH SIGNAL.



TEMPORAL DATA ACQUISITION

The use of foot switches for temporal gait analysis is well established and widely used. The switches may be separate and applied individually to the shoe (Kettlekamp et al., 1970; Gyorg et al., 1976) or may be incorporated into the shoe (Winter et al., 1972). The following is the development of the customized insole with foot switches used in this study.

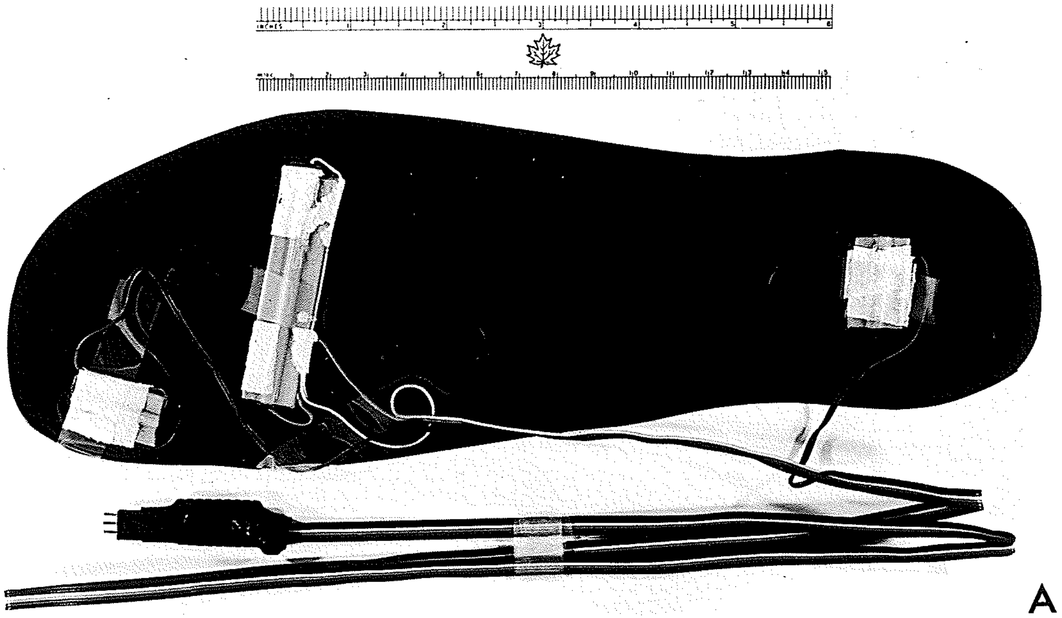
In a preliminary study ten thou (.010") brass shim stock, insulated by radiographic film, was used. However, this proved to be more cumbersome than the final design of a commercial foot switch which is produced by the Tapeswitch Corporation of America, New Jersey. The three foot switches used were connected in parallel with specified resistances so that when a switch or series of switches was closed a different signal output was obtained (Figs. 30 and 31). A similar system has been used in other foot studies (Letts et al., 1975). The output is displayed as a series of steps.

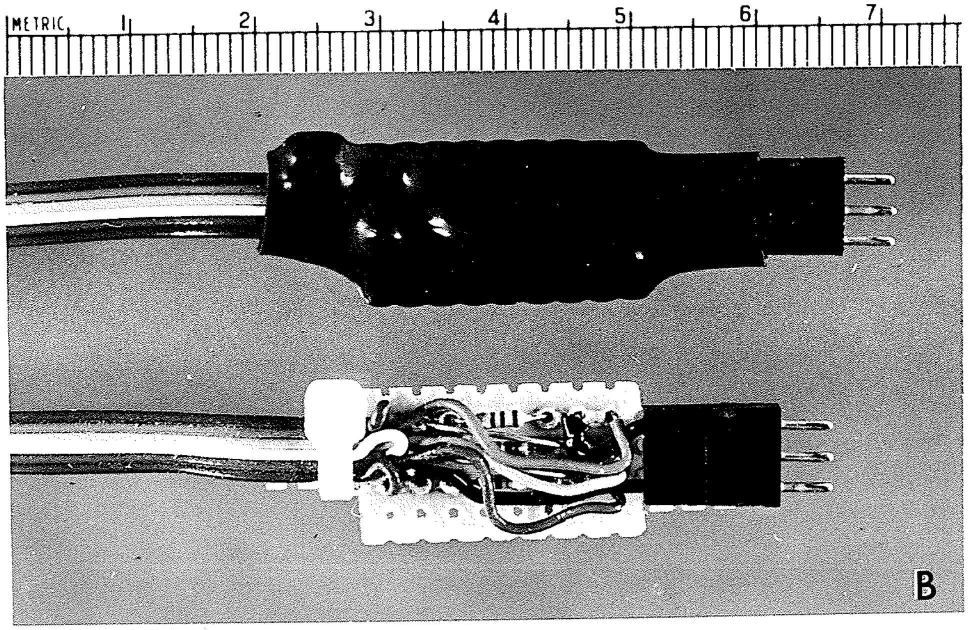
The foot switches are connected to the resistors via a fine multiplex cable and the resistors are miniaturized into the body of the foot switch outlet plug (Fig. 32). The foot switches were attached to radiographic film (.18 mm thick) which was cut out to the outline of the insole of the subject's shoe. A series of these insoles were made and the appropriate one chosen, prior to the attachment of the switches, by fitting it with the subject's shoe. The position of the switches stated below was determined by using the subject's foot as an "anatomical template" (Fig. 32).

FIGURE 32. FOOT SWITCHES.

A) Foot switches attached to customised insole of x-ray material with multiplex cable and output plug.

B) Details of minaturised circuitry of foot switch resistors in output plug.


INCHES
1
2
3
4
5
6
A


METRIC
1
2
3
4
5
6
7
B

1) Under the hallux-electrode size 2.5 cm x 1.5 cm
2) Under the metatarsophalangeal joints-electrode size 6.5 cm x 1.5 cm
3) Under the calcaneum-electrode size 2.5 cm x 1.5 cm

The triggering weight of these switches was 11.8-15.7 newtons force. When placed inside the shoe it was hardly detectable as the radiographic film was placed uppermost in the shoe, and only a smooth surface was presented to the foot which was covered with a sock. The subject was asked to climb the stairs at his normal pace, and not to rush. At the end of the run, the connectors of each of the foot switches was checked to ensure continuity. With some of the amputee group the foot switches were placed on the outside of the shoe as there was no anatomical reference point to use and the position was approximated to the non-amputated side.

This customized insole design provided a simple uncluttered foot switch unit which allowed accurate placement of the foot switches combined with minimal financial expenditure.

In this thesis only swing and stance phases are studied.

ELECTROGONIOMETRIC DATA ACQUISITION

Introduction

It was necessary to obtain accurate knee measurements through a large knee range, therefore knee angle measuring devices were reviewed.

A goniometer is an instrument for measuring angles (Dorland, 1974). The customary design used for clinical measurement of angles is that of two arms attached together at a single point near one end. The angular change between the two arms is assessed by a protractor attached to one arm at this axis. A simple electrogoniometer has a potentiometer at the axis, instead of the protractor, so that an electrical signal proportional to the angle between the arms is obtained. This variable signal is then transferred to display and recording equipment (Liberson, 1965). The gravity reference goniometer differs from the above design (Brown and Stevenson, 1953). Its subsequent improvement and electronic modification (Jones, 1970; Peat et al., 1976) would subserve the measurement of single plane movement but its application is limited to relatively slow movements without fast changes of direction, and excludes movements which occur around a vertical axis. The most accurate measurement of single-joint movements involves a consideration of the biomechanics of the joint studied, and a subsequent design to allow complete freedom of the joint motion while the goniometer is attached to both proximal and distal components of the joint. The section on the axis of flexion and extension in the review of the literature demonstrates that at present there is no accurate determination of the axis of movement except that it varies and occurs in the region of the lateral femoral condyle. Because of this, a single and fixed axis goniometer is unable to reproduce accurately the movement occurring in the saggital plane. In addition, the accurate determination of the centre of rotation in the living subject is almost impossible (Kettlekamp et al., 1970). To overcome this problem, various linkage systems have been devised to ascertain the relative positions of the thigh and the lower leg irrespective of the changing axis of the joint (Johnston and Smidt, 1969). The linked hinged parallelograms design of goniometer, originally a finger goniometer, (Thomas and Long, 1964; Long and Brown, 1964) has been enlarged to accommodate the knee joint, and is used in the Pathokinesiology Service of the Rancho Los Amigos Hospital in California. However, this electrogoniometer has an error of $\pm 6^{\circ}$ at 100° of knee flexion (Sturman, 1976A) Another design uses a variation of this parallelogram principle (Lamoureux, 1971), and this has been modified with a sliding bar (Banerjee et al., 1976). Double parallelograms and parallellogram chains in polyurethane elastomers have also been designed (Cousins, 1973).

More sophisticated systems of simultaneous three-dimensional analysis of knee range of movement have also been devised (Kettlekamp et al., 1970; Doyle et al., 1975; Townsend et al., 1977). Other electrogoniometers were reviewed that had limitations that precluded their use (Kinzel et al., 1972; Foort and Cousins, 1974).

The design of an electrogoniometer for this study required the accurate measurement of knee flexion-extension over a wide dynamic range but not the measurement in other dimensions of knee motion. Thus, although the lack of joint restriction offered by some of the above mentioned designs was appealing, their complexity of design for our requirement of only flexion-extension measurements was not. Consequently, an electrogoniometer was designed to permit all degrees of joint freedom without limiting the accuracy of the flexion and extension measurement. The instrument has a flexible junction bar connecting two main electrogoniometer arms, and the application of a trigonometric relationship for the determination of the actual joint angle. It permits easy alignment and calibration because its two bars can be used as a conventional goniometer which is aligned to bony points.

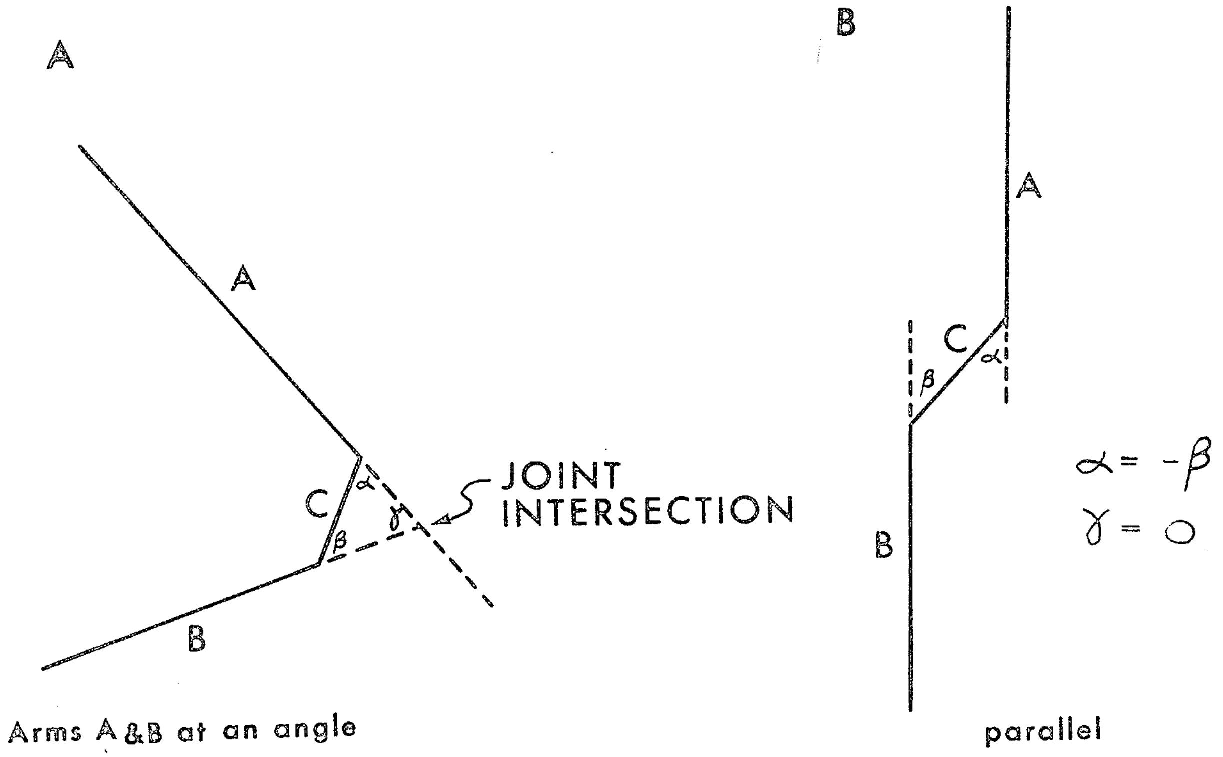
Some details of this goniometer have been published (Tata et al., 1979). However, a more comprehensive description is given as its design and development are an integral part of this study.

Principle

The electrogoniometer functions as a 3-bar, 2-joint mechanism that provides freedom of position for the undetermined instantaneous centre of rotation of the anatomical joint across which it is attached. The joint "intersection" (which subtends the desired joint angle) is free to take up any position as indicated by the virtual intersection of the two arms A and B (Fig. 33A). Since the double axis mechanism is a non-restraining one, the range over which this intersection can move is determined by the length of the intermediate junction bar C. In the unit constructed, this length was chosen to permit movement over the path traversed by the instant centre during flexion and extension of the knee joint, since this was the intended application of the electrogoniometer.

From Figure 33A, the desired joint angle, between arms A and B, is angle γ. A common trigonometric relationship states that $\gamma = 180 - (\alpha + \beta)$ where α and β are physically realizable and measurable angles. Thus, as long as arms A and B are parallel to the limb segments on either side of the joint the sum of the two angles α and β subtracted from a constant will yield the desired joint angle. When the two arms are parallel to each other, either angle β = angle $\alpha = 0$, or angle $\beta = -$ angle α as in Figure 33B and in either case the summation yields the correct joint angle at 0^{o}. Two potentiometers were used to obtain

Fig.33 ELECTROGONIOMETER PRINCIPLE



Arms A&B at an angle

parallel

$\alpha = -\beta$

$\gamma = 0$

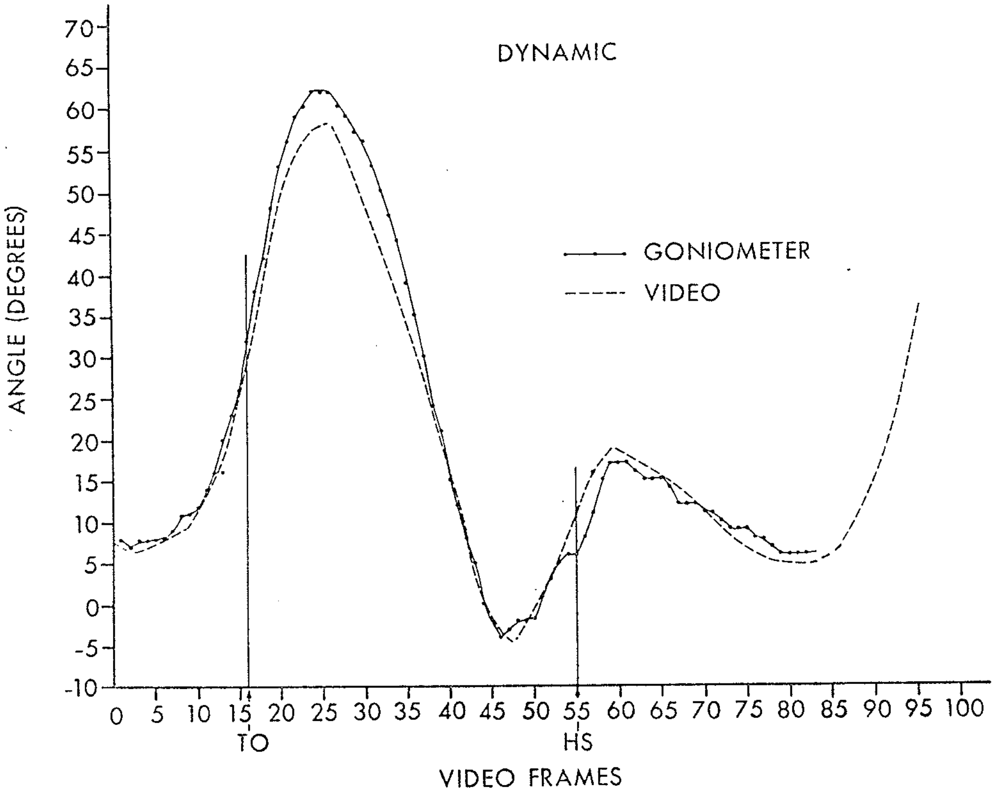


Fig.34 ELECTROGONIOMETER AND VIDEO SYSTEM OUTPUTS

electrical analogs of angles β and α and a summing circuit reproduced the trigonometric relationship.

Description

Mechanical Components

Since the study required flexion-extension measurements the design could be considerably simplified over a three degrees of freedom goniometer mentioned earlier that would respond to all joint motions. However, attention was given to the fact that the knee is not a simple hinge joint as well as to the fact that the instant centre of rotation varies with joint position and is presently undetermined.

The arms of the goniometer were fashioned from a lightweight aluminum alloy and lined with Velcro on their inner sides. They were contoured, as can be seen in Figure 35, to fit comfortably over the contours of the lateral aspect of the thigh and lower leg. In order to permit long axis rotation and abduction and adduction at the knee, the junction bar was made from thin flexible plastic that could be easily deformed in the above directions without affecting the accuracy of the flexion-extension measurements. The bodies of the potentiometers were attached to this thin plastic and their shafts to the aluminum bars (Figure 35). The linkage system offered no appreciable resistance to knee motion and the completed unit weighed 181 gm.

Electronic Components

The potentiometers were chosen for their full 360° rotation, small size and high degree of linearity (Bourns-Model 35305-1-502-linearity ± .5%). Each potentiometer was mounted so that the wiper

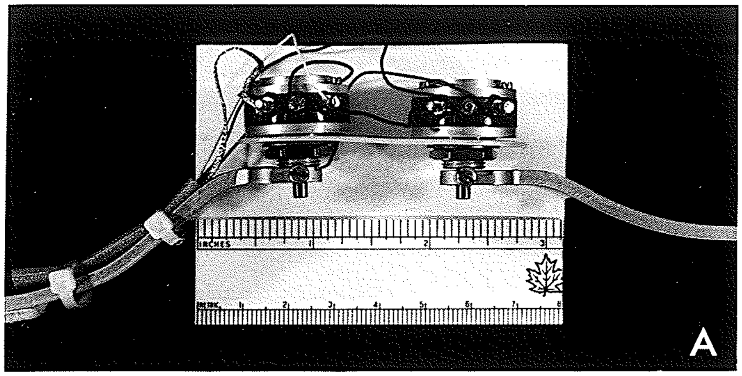
FIGURE 35. ELECTROGONIOMETER.

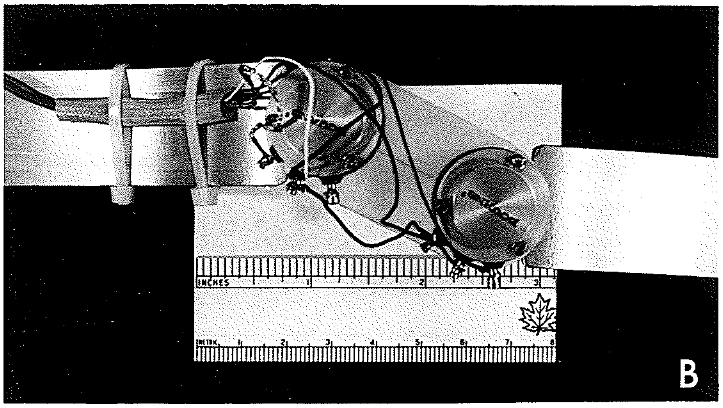
A) Anterior view showing potentiometers and selectively flexible intermediate junction bar.

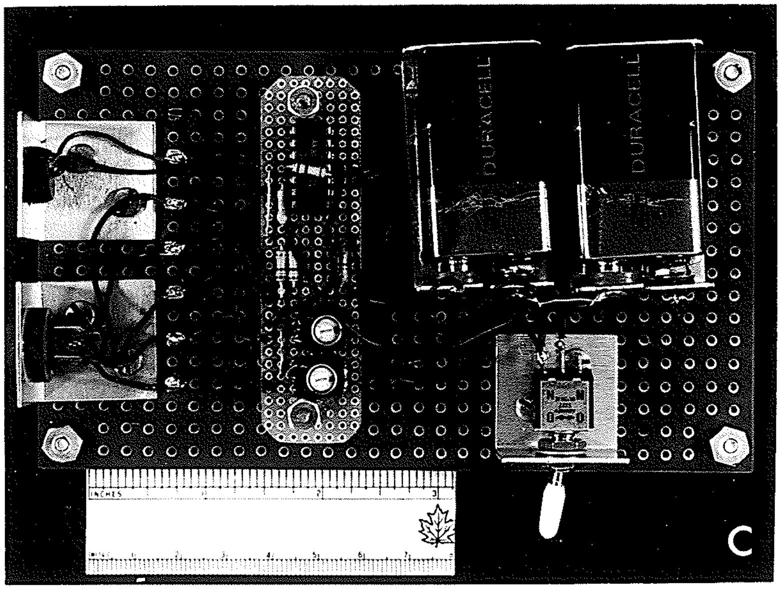
B) Lateral view of A.

C) Details of waist-belt pack.

D) Electrogoniometer attached.


INCHES
METRIC
A


INCHES
METRIC
B


DURACELL
DURACELL
INCHES
C


D

contact was approximately at mid-position, with angles β and α (Fig. 33B) equal to 0^{o} so that there was 180^{o} of movement available in either direction. The circuitry is shown in Figure 36. The 2 Kilohm trimpots were used to adjust each potentiometer output to exactly 0 volts when the angles equalled 0. The conventional method of recording knee range of movement is the accepted neutral zero method mentioned earlier (Heck et al., 1965). With the knee fully extended and the leg therefore straight, this position is considered to be zero degrees rather than 180^{o}. The 180^{o} constant term in the trigonometric relationship does not need to be included, and the desired output is then the sum of signals from the two potentiometers. The summing and balancing circuit shown in Figure 36 was mounted on a waist belt (Fig. 35) and flexible multiconductor cable connected it with the potentiometers. The output of the summing circuit was connected via an overhead cable to the chart recorder.

Evaluation

Three methods of evaluating the electrogoniometer were used, one dynamic and two static. These methods permitted an accurate assessment of performance, not only in fixed positions but also during a dynamic walking test.

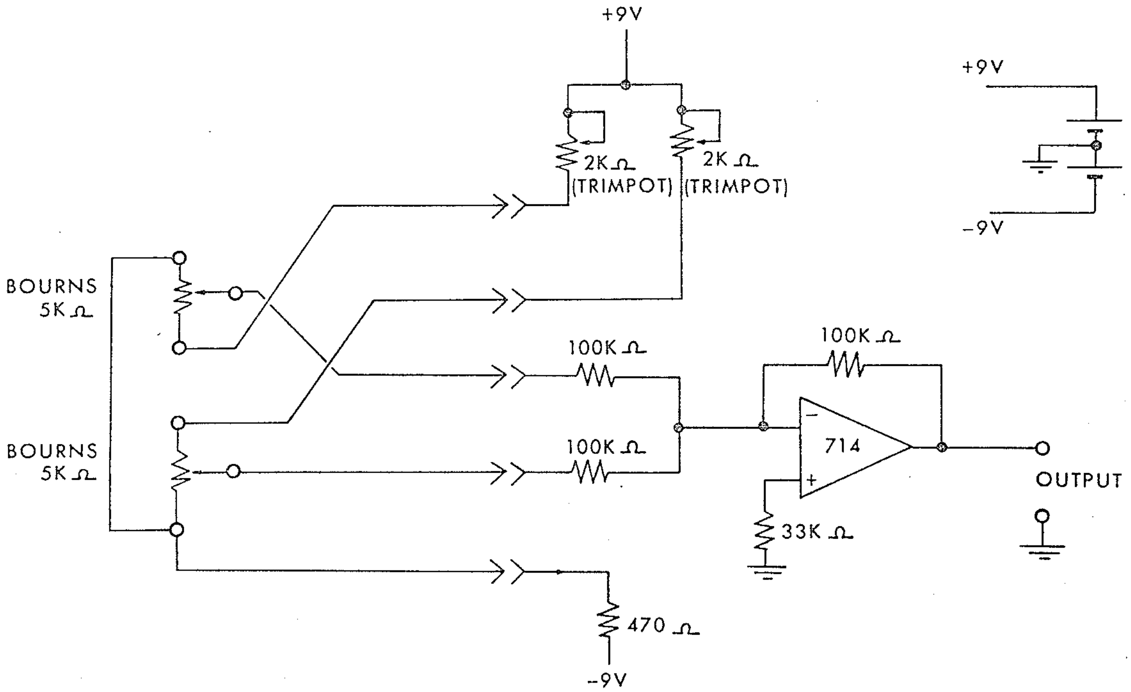
Dynamic Test

The electrogoniometer was attached with Velcro straps and reflective markers were placed on the leg over the greater trochanter, lateral epicondyle of femur, head of fibula and lateral malleolus. A video movement measuring system (Winter et al., 1972) was used to

Table 3 Electrogoniometer Test Data

SUBJECT LIMB POSITION	ANGLE BETWEEN FEMUR AND TIBIA MEASURED ON X-RAY (DEGREES)	ACTUAL DEGREES MOVED (DEGREES)	SIGNAL OUTPUT FROM ELECTRO GONIOMETER (VOLTS)	EQUIVALENT DEGREES/VOLT
POSITION 1 FULL EXTENSION	4°	0°	0.0	0.0
POSITION 2 SLIGHT FLEXION	49°	45°	1.4	32.1
POSITION 3 MORE FLEXION	76°	72°	2.3	31.3
POSITION 4 FULL FLEXION	122°	118°	3.7	31.9

Fig. 36 Electrogoniometer Circuit



measure the knee joint angle during walking. The electrical output of the goniometer was displayed on a chart recorder along with a pulse train locked to the TV field rate to synchronize this data with the video data. The curves from these two measurements were then compared as shown in Figure 34. As can be seen there is a close correlation with a maximum discrepancy between the two curves of 4° at the maximum flexion position. The overall accuracy of the video movement measuring system is in the order of $\pm 2^{\circ}$. Should the maximum error of the video movement measuring system coincide with the maximum discrepancy shown the error would be no more than 6° which occurs at maximum flexion.

Static Tests

POLAROID

In the first static test, reflective markers were again placed over the greater trochanter, lateral epicondyle of femur, head of fibula and lateral malleolus. The subject then held fixed angles of knee flexion while the electrical output of the goniometer was measured with a Hewlett-Packard digital multimeter and a Polaroid photograph was taken. Joint angle was measured with a protractor from the photographs, using the reflective markers as anatomical landmarks. The results were plotted and show a maximum deviation of 1.5° between the two measurement methods.

RADIOGRAPHIC

The second static test involved radiography to permit angle measurement directly between tibia and femur as revealed by roentgenograms (Fig. 37). The electrogoniometer was attached to the right leg and the signal relayed to a Hewlett-Packard digital multimeter. Four full-leg roentgenograms were taken using a General Electric 1200 m.a. generator at a focal film distance of 72 inches. The four positions were full extension, full flexion, and two arbitrary positions between these two extremes of range. In the four positions, the electrogoniometer signal was recorded. Bony points on the X-rays were then marked in a uniform manner and the angle between the femur and tibia measured with a protractor.

From Table 3 the measured angle on the X-ray in the four positions can be seen. The actual degrees moved for each position is obtained by subtracting four degrees from these measurements as full extension lacked four degrees when measured via the bony points on the X-ray. The signal output from the electrogoniometer was noted for each position and the equivalent degrees per volt derived by dividing the actual range of movement in degrees by the voltage signal. The data from Table 3 was plotted and a straight line with a slope of 32 degrees/volt demonstrated the overall sensitivity of the electrogoniometer.

Attachment

The Velcro straps are attached to the thigh and two to the lower leg. The leg is fully extended and the electrogoniometer is aligned using the greater trochanter and the lateral femoral condyle

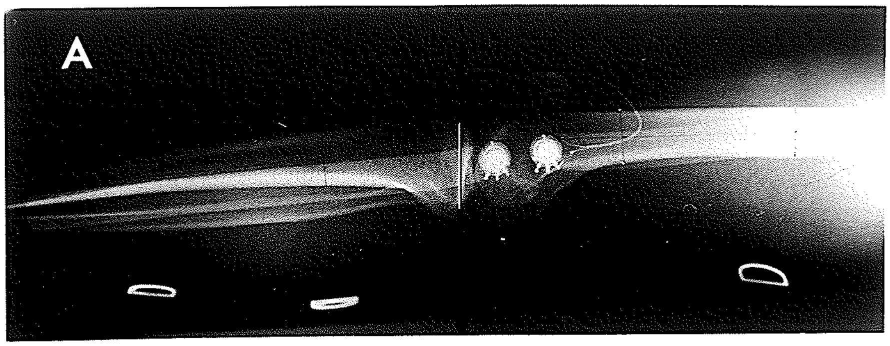
FIGURE 37. ROENTGENOGRAMS OF ELECTROGONIOMETER *IN SITU*.

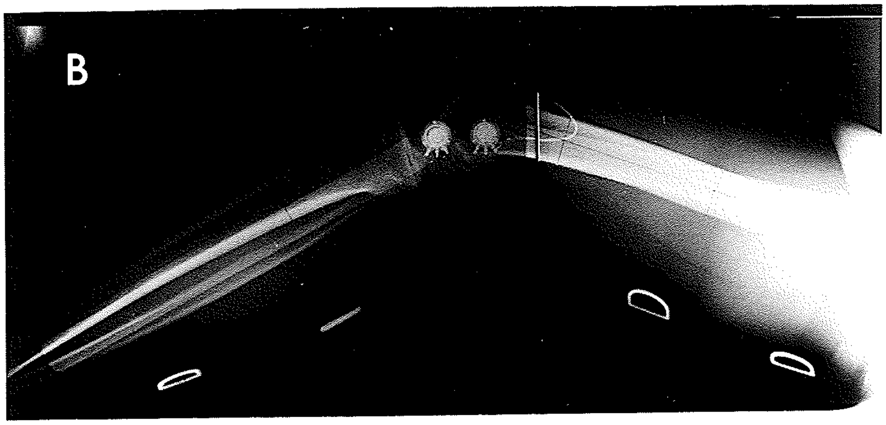
A) Knee extended.

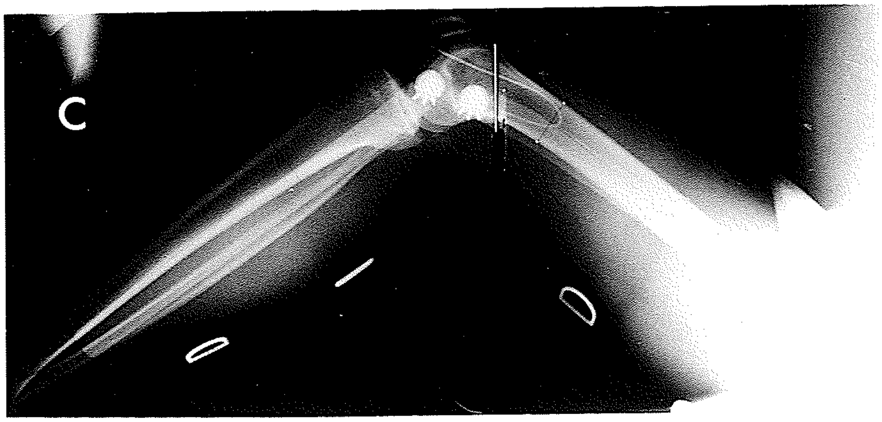
B) Slight flexion.

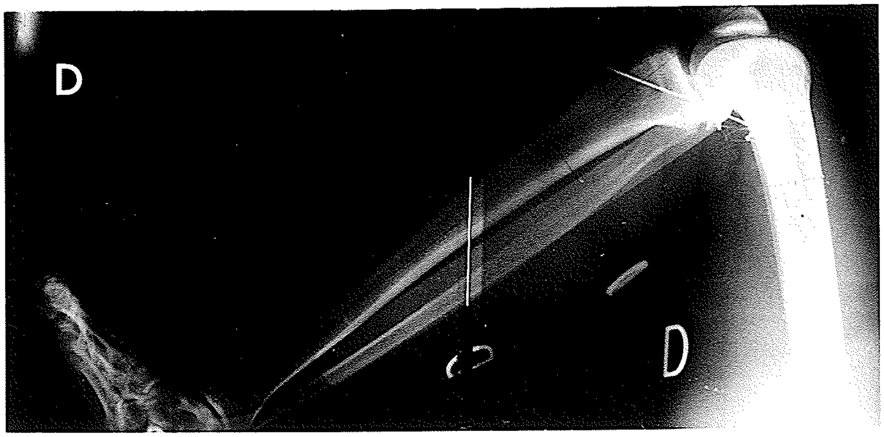
C) More flexion.

D) Full flexion.


A


B


C


D

as the proximal reference points, and the head of the fibula and the lateral malleolus as the distal reference points. The Velcro on the arms is firmly pressed onto the Velcro straps on the leg. Should the placement not be satisfactory, removal and re-application is quick and simple. The alignment is not critical as long as the junction bar is placed over the region of the lateral femoral condyle. Varying dimensions of the lower limbs as a result of quadriceps bulk are compensated for by the flexible junction bar.

Calibration

The goniometer was calibrated for the Honeywell Chart Recorder: An enlarged diagram with specified angles was used. The arms were aligned in these specific angles with varying junction bar positions. With the varying junction bar positions and constant arm positions the signal should not vary, and this was confirmed.

With varying arm positions the recorded deflection on the chart recorder was compared with the angle, and the degrees moved per millimeter calculated. The sensitivity of the instrument remained constant at one millimeter chart deflection representing a three degree angle change. The goniometer was checked for calibration regularly throughout the study.

Accuracy

The precision of the electrogoniometer is dependent on 1) the inherent inaccuracies in the mechanical and electronic components, and 2) error in attachment.

The potentiometers have a ±0.5% linearity over a 360^{o} range. Assuming a dynamic range of 120^{o} of motion, which is considerably higher than the range required for activities of daily living (Laubenthal et al., 1972), the total maximum error per potentiometer is 0.3° which is negligible.

In addition errors resulting from non-linearity in the electronic circuit are negligible (Quanbury, 1979). The major errors relate to the attachment of the electrogoniometer and even the maximum error in the video comparison would be acceptable. It should be noted that the video system's accuracy in measuring joint angle was not correlated with the actual bony position as was the electrogoniometer used in the study. The radiographic data is perhaps the most convincing in terms of accuracy as the actual positions of the tibia and femur were compared with signal output and the linearity is very high. The attached instrument accuracy on this basis is ±2°.

In summary, an accurate single plane electrogoniometer with a minimum of mechanical components that is light, unencumbering, easy to align and attach to the lower limb was designed, constructed and evaluated.

EMG DATA ACQUISITION

The purpose of this component of the study was to evaluate the sequence of peak EMG activity in the thigh during the climbing cycle. Though the phasic activity will be described this was not the objective of this part and the rigorous statistical analysis was applied only to the peak EMG sequence.

Electrodes

Beckman silver-silver chloride surface electrodes with an overall size of 16 mm diameter and an active electrode size of 9 mm were attached in pairs. The interelectrode distance was kept constant by the use of Beckman adhesive collars (kit 650429) which were used as a guide. The importance of this constancy was mentioned in the review of the literature. The interelectrode distance of the two closest points is 2.1 cm

Skin Preparation

As explained in the literature review, the high input impedance amplifier has certain advantages with respect to skin preparation. However, all subjects were shaved using a professional electric clipper from the Charlescraft Corporation Ltd., and the skin was rubbed firmly with 95% ethyl alcohol at each electrode site. The reasoning for this was that the electric clippers would assist in the slight debridement of the skin, the alcohol would remove the sebum and the friction would assist both the above objectives.

Skin Resistance

This was not recorded for each subject as it was not thought necessary because of the equipment design (see review and equipment data).

Electrode Site

Six muscle sites were chosen:

Quadriceps Femoris: 1) Vastus medialis oblique
2) Vastus medialis longus
3) Rectus femoris
4) Vastus lateralis

Because vastus intermedius has no superficial muscular component and only surface electrodes were used, it was felt that this muscle could not be adequately represented using this technique.

Hamstrings: 1) Lateral hamstrings
2) Medial hamstrings

Because of the surface technique no differentiation was made between the long and short heads of biceps femoris on the one hand and the semimembranosus and semitendinosus on the other.

Vastus Medialis

As has been noted in the review of the literature, the vastus medialis may be considered to have two functionally distinct parts, the vastus medialis oblique and vastus medialis longus.

To make the electrode positioning as accurate as possible a preliminary photographic study of the dissected vastus medialis was conducted. Twenty-four cadaver knees of 12 subjects from the University of Manitoba's Department of Anatomy Gross Laboratory were used. The subjects had a mean age of 67 years ± 11 years. Pins were placed at the following bony points:

1) apex of the patella

2) base of the patella in the midline

3) tibial tubercle

and also at the lowest point of the insertion of vastus medialis. A photograph was then taken with a scale ruler *in situ*, to record the position of knee flexion.

The 35 mm negatives were displayed on 8" x 10" prints and the following data collected after appropriate scaling using the ruler in the print:

Mean patellar length = 4.2 cm ± 0.6 cm (range 3.1-5.3 cm).

Angle of lowest fibres of vastus medialis oblique entering the patella = 60^{o} ± 6^{o} (range $48\text{-}73^{o}$).

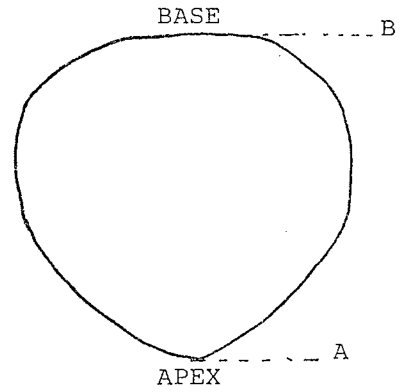
Point of lowest fibres on patella 3/5 the distance from apex to base from the apex = 40% ± 14% (range (25-91%).

Details of these findings are shown in Figure 38.

Lieb and Perry (1968) in their study of six cadavers had a range of $50\text{-}55^{o}$ from the vertical of the fibres of vastus medialis oblique. This corresponds closely with this study particularly when considering that we calculated the angle of the lowest fibres of vastus medialis oblique which have the greatest angle, and Lieb and Perry (1968) included all the fibres of vastus medialis oblique.

The limitations of a cadaver study of this kind are well known to the author and the major purpose of this study was to gain a better understanding of the fibre alignment and orientation to the area to obtain information about the lowest point at which the vastus medialis oblique electrode could be placed.

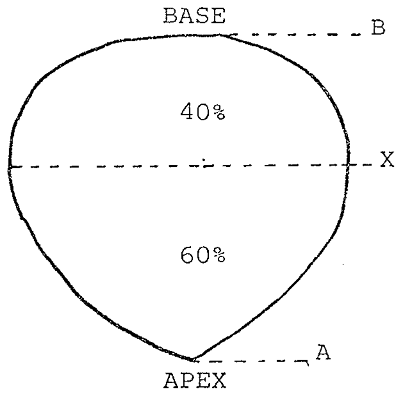
FIGURE 38. PRELIMINARY STUDY OF VASTUS MEDIALIS INSERTION.



PATELLA SIZE

BA = 4.2 ± 0.6 cm

Range = 3.1 cm to 5.3 cm



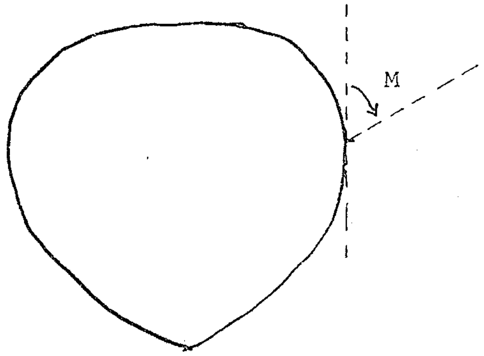
LOWEST POINT OF INSERTION

Expressed as a percentage of the total patella size.

X = 40 ± 14%

Range = 25% to 91%

$$\text{Formula} = \frac{BX}{BA} \times 100$$



ANGLE OF THE LOWEST FIBRES WITH RESPECT TO THE VERTICAL AXIS OF THE PATELLA

$M = 60 \pm 6^{\circ}$

Range = 48° to 73°

Vastus Medialis Oblique. The first pair of vastus medialis electrodes was placed over the oblique fibres, the lower electrode of the pair being placed near the base of the patella.

Vastus Medialis Longus. The second pair of vastus medialis electrodes was placed more laterally and as high as possible. Rectus femoris was palpated and avoided.

Rectus Femoris. The third pair of electrodes was placed on the middle of the belly of rectus femoris.

Vastus Lateralis. The fourth pair of electrodes was placed on vastus lateralis below the rectus femoris electrode, and anterior to the midline of the thigh when viewed from the lateral aspect.

Lateral Hamstrings. The fifth pair of electrodes was placed on the belly of the biceps femoris above the probable origin of the short head after palpating resisted flexion of the knee in standing. Care was taken to avoid vastus lateralis.

Medial Hamstrings. The sixth pair of electrodes was placed on the bellies of the semitendinosus and semimembranosus. These muscles could not be differentiated with surface electrodes. Care was taken to avoid the adductors.

The ground electrode was placed on the anterior aspect of the thigh for both limbs.

Figure 39 is an example of the electrode position for a normal and an amputee subject.

Electrode Cross Talk and Spatial Selectivity

In this study serious consideration was given to cross talk and the area from which the muscle signals were being received.Cross talk

FIGURE 39. SUBJECT PHOTOGRAPHS WITH ELECTRODES AND ELECTROGONIOMETER *IN SITU*.

A) Anterior view--normal subject 16.

B) Lateral view (right limb)--normal subject 16.

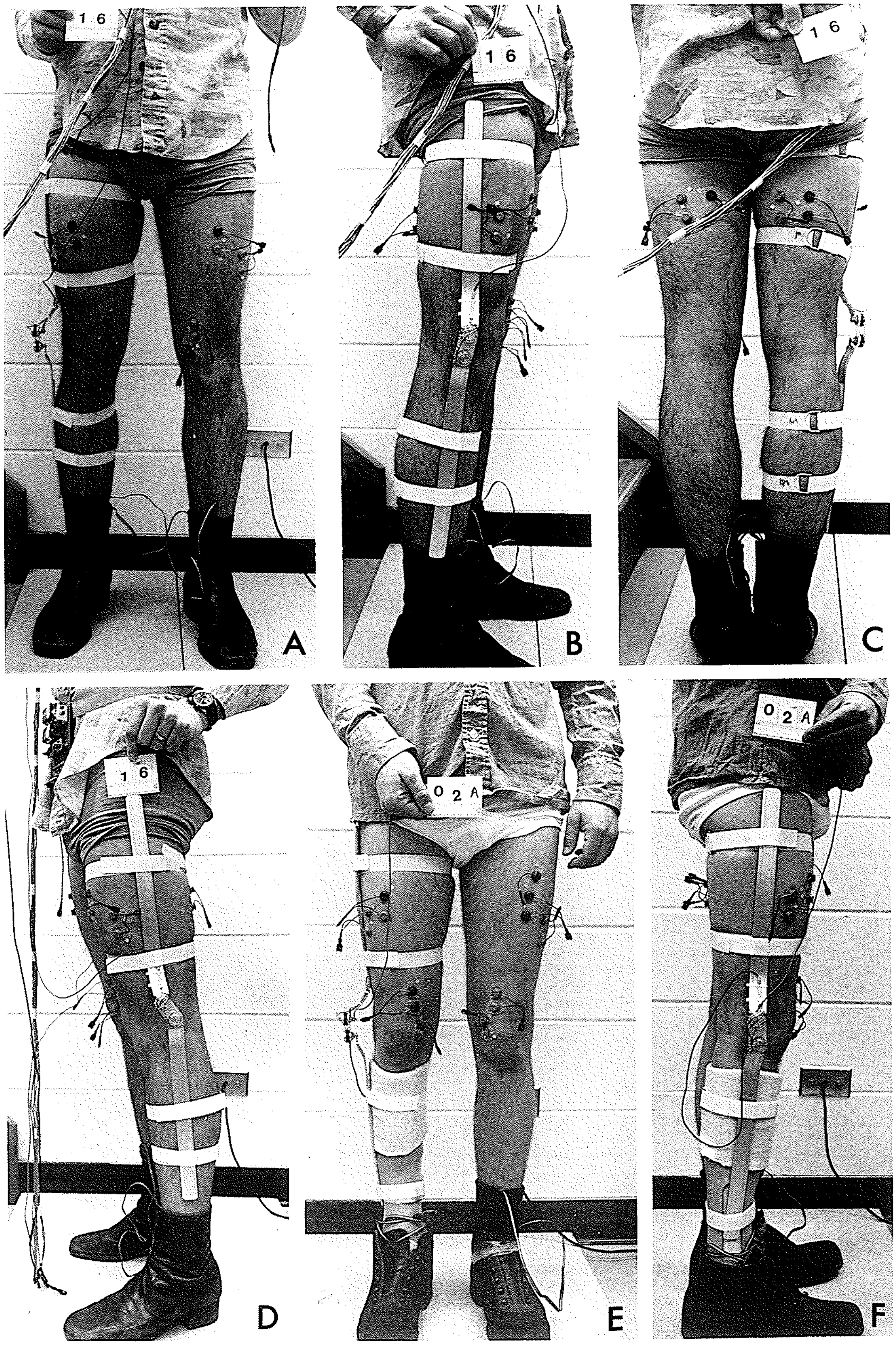
C) Posterior view--normal subject 16.

D) Lateral view (left limb)--normal subject 16.

These four photographs represent the full set for each subject for electrode and electrogoniometer positions.

E) Anterior view--Syme subject 2.

F) Lateral view (right limb)--Syme subject 2.


1 6
A
1 6
B
1 6
C
1 6
D
0 2 A
E
0 2 A
F

may be defined as the transfer of signals from one channel to another. This cross talk may occur at the signal site, i.e., at the electrodes or within the amplifier itself.

In quantitative EMG investigation by surface electrodes on large skeletal muscles the amplitude of the signal is dependent on the distance between the electrodes (Kramer and Kuchler, 1971). Kadefors et al. (1969) found that decreasing electrode distance decreased the RMS value of the signal though increased the upper frequency limit of it. When using a pair of surface electrodes, the closer the electrodes are to each other the greater will be the relative contribution of the superficial muscle fibres to the signal (Shwedyk, 1979).

If the electrode pair is aligned along the long axis of the muscle fibres a signal is almost certainly to be received from these muscle fibres rather than from a muscle at a different angle to it. The signal from another area of the muscle will be at its lowest when the electrode pair is at right angles to that area of muscle. This is because of the direction of depolarization of the muscle cell (Shwedyk, 1979).

The amplitude of the action potential is dependent on 1) the diameter of the muscle fibre, 2) the distance between the active muscle fibre and the recording site, and 3) the filtering properties of the electrode (De Luca, 1979). In human muscle tissue the amplitude increases as $V = Ka^{1.7}$, where a is the radius of the muscle fibre and K is a constant (Rosenfalck, 1969), and decreases approximately inversely proportional to the distance between the active fibre and the recording site (Buchthal et al., 1957). The filtering properties of a bipolar electrode are a function of the size of the recording contacts (Geddes, 1972), the distance between the contacts (Lindstrom et al., 1970), and the chemical properties of the metal electrolyte interface (De Luca and Forrest, 1972).

Since the recording electrodes, distance between them and electrode jets remained constant, the filtering properties of the electrodes were constant. As there are no studies comparing different muscle fibre radius in the quadriceps or hamstrings an assumption is made that they are not significantly different in any particular area. In addition, since the signal amplitude decreases inversely to the distance between the active fibre and recording site the signal pick-up of the surface electrode relates more strongly to the area under it. When the peak activity from two different groups of fibres occur at different times the peak detected by surface electrodes will almost surely represent the peak activity of the fibres directly beneath it, particularly when considering the signal attenuation that is discussed in the following paragraphs.

A fundamental consideration of the problem of cross talk relates to the actual capability of picking up an electrical signal from an action potential at a specified distance from the propagation of the signal. There is no doubt that theoretically there is a capability of picking up an electrical signal from an electrode from any electrically excitable tissue that is far away.

However, in the study of volume conduction of action potentials in human muscle, there was a 90% attenuation radius of the action potential recorded with a 25 micron electrode at approximately .2 mm from the electrode (Gath and Stalberg, 1978). Others also considering voltage drop all give significant losses within a millimetre (Krnjevic and Miledi, 1958; Rosenflack, 1969). Gath and Stalberg (1978) do, however, caution that the results cannot be generalized, since they depend on the filter function of the electrodes, the frequency spectrum of the signal and on the medium involved. However, they mention that their recent study (1978) agrees with their previous studies (1975, 1976) that the higher frequency content is attenuated more steeply with distance than is the low frequency content.

Finally, in a preliminary study, our complete equipment unit was studied for cross talk. Electrodes were placed over the vastus medialis and the muscles of the pes anserinus, especially sartorius. Resisted flexion and extension were performed and varying electrode distances and even the closest position (i.e., when the electrode cuffs were touching) were tested. There was no cross talk recorded between the electrodes. This was demonstrated by the fact that when flexion was resisted, no signal was received from the extensor (vastus medialis) electrodes and vice versa.

From all the above evidence, it would be reasonable to state that the signal received from the electrodes related to the area under them.

Photography

For each subject, all electrode and goniometer attachments and placements were photographed with a scale rule *in situ*.(Fig. 39). Initially, a Pentax ESII was used and latterly a Pentax ME camera using Kodak Tri-X pan film (400 ASA). Both cameras had 35 mm format and were fully automatic.

EMG Equipment

The EMG equipment has been used in previous studies (Peat and Grahame, 1977, 1977A). The following data details follow the guidelines on terminology suggested in the first interim report of the International Society of Electrophysiological Kinesiology's Committee on EMG Terminology (Winter et al., 1979).

The EMG signals from the electrodes passed through appropriately shielded cables to the amplifier. The signal amplifier for this study was developed, designed and constructed in the Electronics Shop of the Biomedical Engineering Research Department of the Rehabilitation Centre for Children, Winnipeg (formerly the Shriners Hospital). The circuit diagram for a single channel is shown in Figure 40.

The eight channel system provided two possible outputs, the raw or processed signal. The processed signal was used in the study.

Amplifier Details

The input impedance is 20 megohms and the common mode impedance 10 megohms.

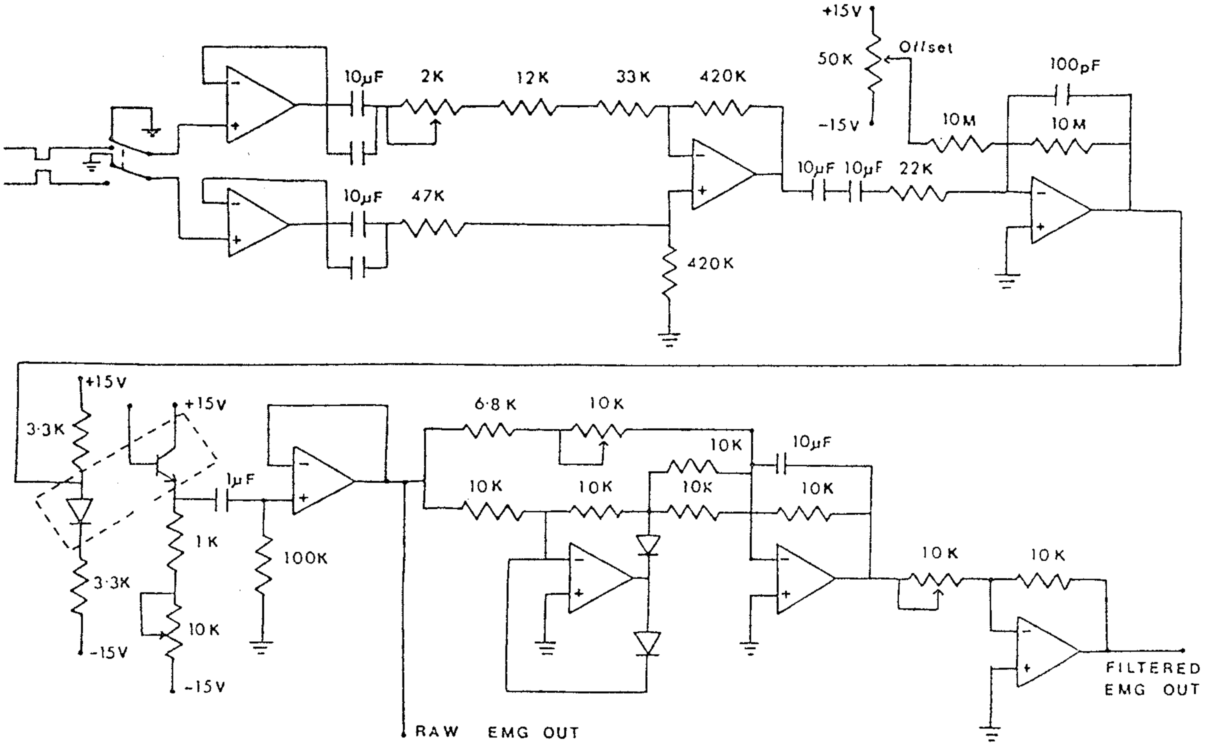
Band width is from 20 hertz (Hz) to in excess of 5 Kilohertz (KHz).

The common mode rejection ratio extends from 80 to 100 db at 60 Hz.

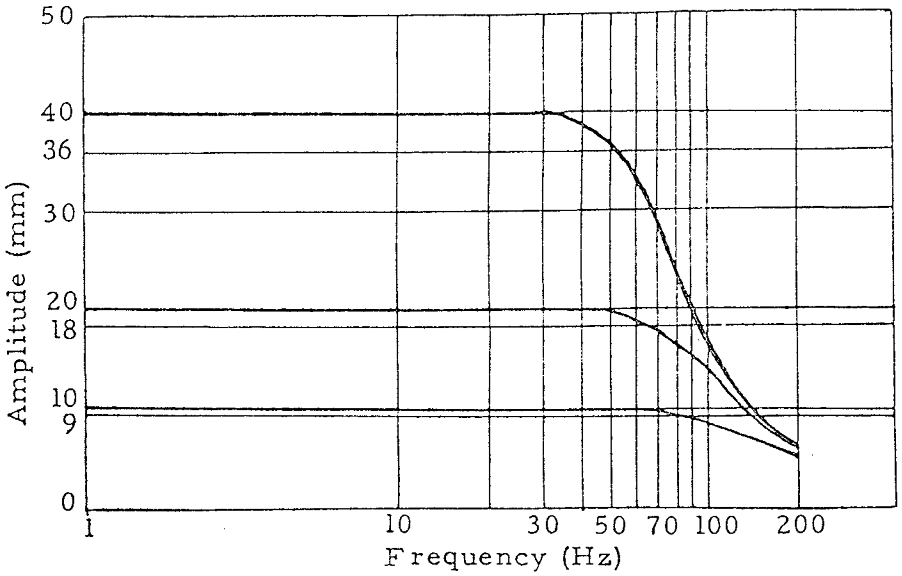
The overall voltage gain of the amplifier is 4700.

The signal was processed by means of full wave rectification and a first order low pass filtering with a 3 db cut off frequency of 10 Hz.

FIGURE 40. CIRCUIT OF A SINGLE EMG CHANNEL.



FREQUENCY RESPONSE OF INK RECORDER.



FLAT FREQUENCY RANGE

FREQUENCY RESPONSE AT 40 mm 30 Hz + 5-10%
AT 10 mm 80 Hz ± 5-10%

Ink Recorder Details

A Series 2500, eight channel Honeywell Pen Recorder (Model 1508) was used for all the recordings (Fig. 42).

This ink-writing oscillograph was bench-mounted and capable of simultaneously recording from a maximum of eight channels at frequencies varying from direct current to 50 Hz (with a 20 mm peak to peak deflection). The flat frequency range and frequency response are shown in Figure 40.

Using this recorder there was no overlap of channels during recording, each channel having a separate recording width of 40 mm.

The signals were attenuated by a Honeywell Accudata 125 attenuator which has nine ranges from 0.5-200 V/cm.

All subjects runs for the study were done at a paper speed of 50 mm/sec. The specified accuracy was within ±2%. The paper speed accuracy could be determined by measuring the distance on the paper of the timing marker. There was never any measurable variation.

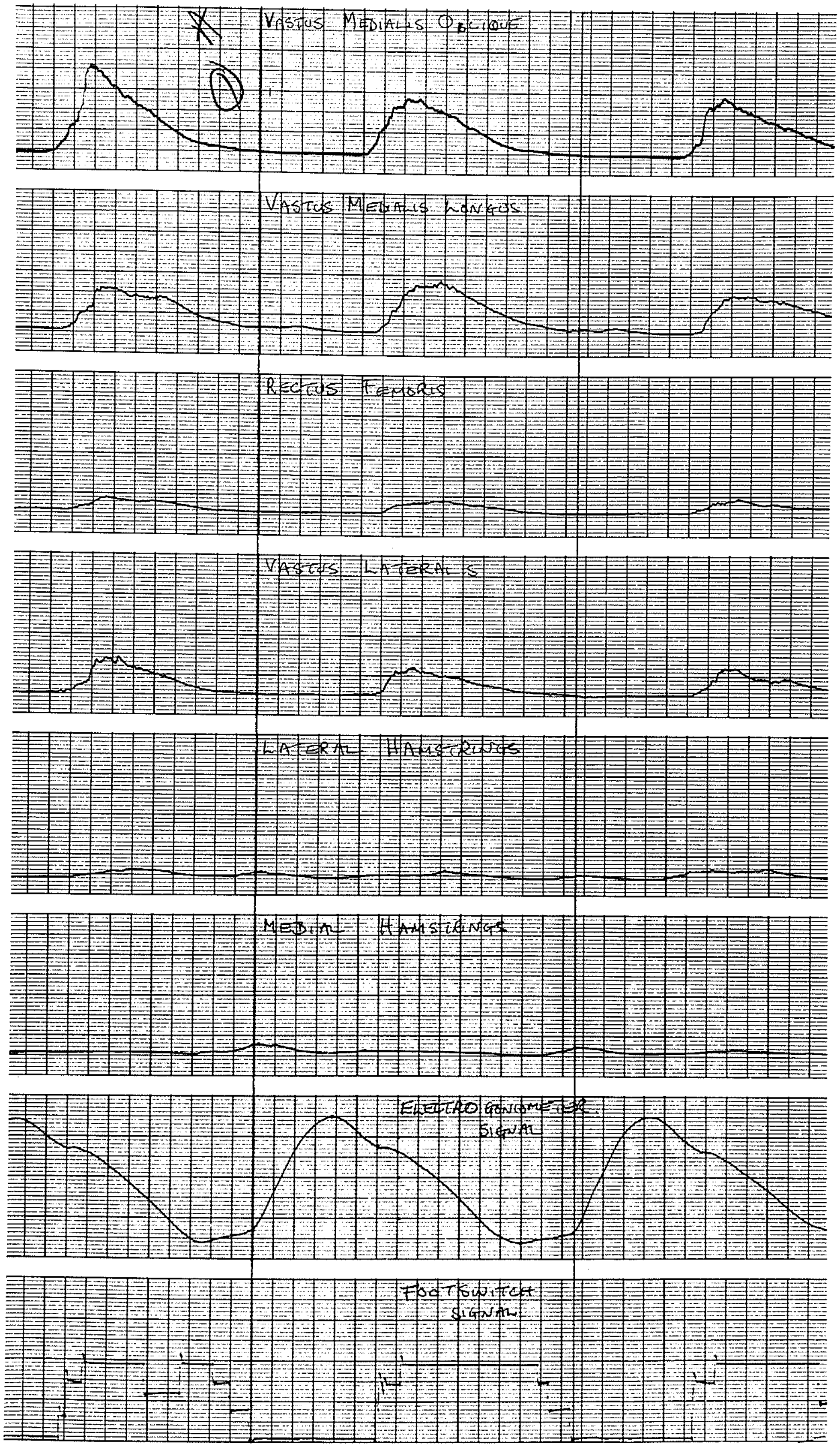
The recorder was operated within the specified environmental conditions. A sample of the ink recorder printout is shown opposite page 155.

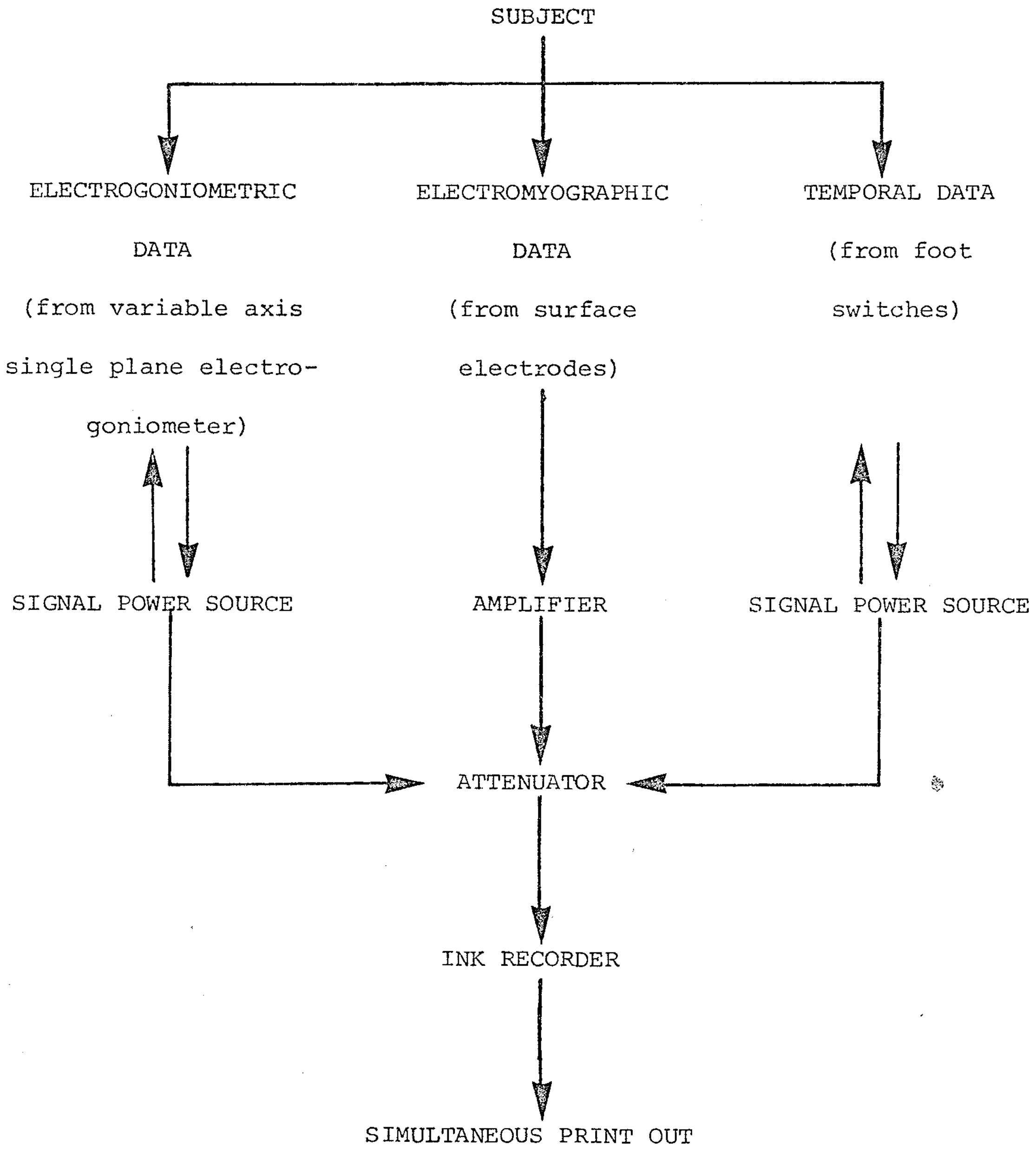
The scheme for data acquisition is shown on page 155A.

CONSENT FORM

This project was approved by the Faculty Committee on the Use of Human Subjects in Research. A copy of the consent form signed by all subjects prior to their partication is shown in Figure 43.

FIGURE 41. INK RECORDER PRINT OUT.


VASTUS MEDIALIS OBLIQUE
VASTUS MEDIALIS LONGUS
RECTUS FEMORIS
VASTUS LATERALIS
LATERAL HAMSTRINGS
MEDIAL HAMSTRINGS
ELECTROGONIOMETER SIGNAL
FOOTSWITCH SIGNAL



SCHEMATIC OF DATA ACQUISITION

See page 154

FIGURE 42.

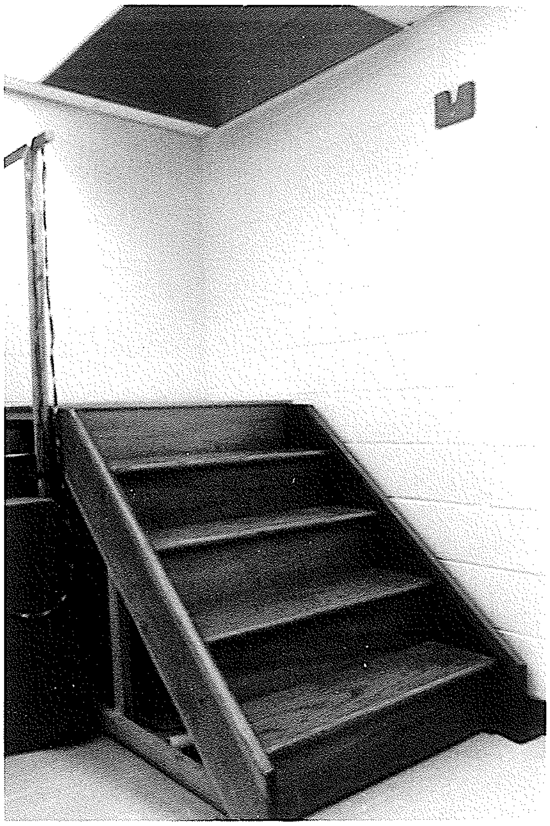
STAIRS USED IN STUDY.

EQUIPMENT DETAILS.

Ink Recorder.

EMG Amplifier.

Foot Switch Power Source.



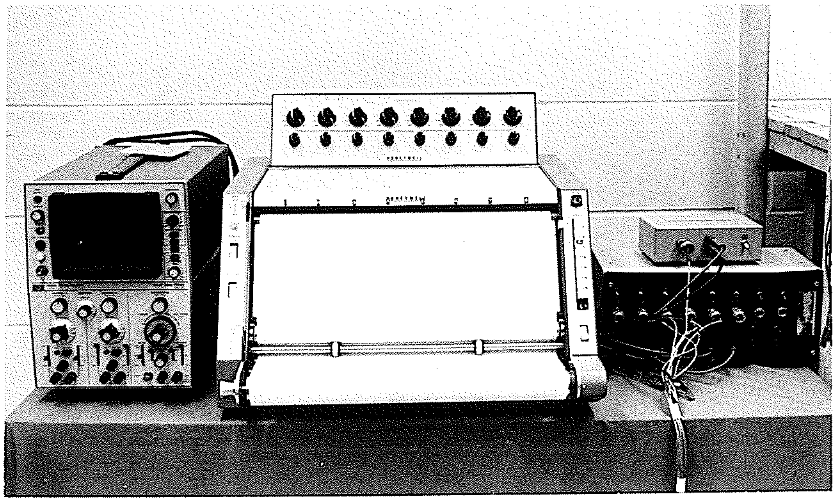


FIGURE 43.

Department of Anatomy

University of Manitoba

Electromyography Laboratory

I hereby consent to act as a subject in the Research Program on Muscle Activity in the Electromyography Laboratory of the Department of Anatomy. The procedure, which may include the insertion of fine wire electrodes into muscles, and photographic recording, has been explained to me fully, together with potential complications which may arise during the course of the study.

The photographic records will be used for analytical purposes and may also serve as documentation in research papers and medical lectures.

My participation is on a voluntary basis, and I reserve the right to withdraw immediately from the procedure whenever I wish.

SIGNATURE OF SUBJECT OR
RESPONSIBLE RELATIVE ______________________________

SIGNATURE OF MEDICAL
SUPERVISOR ______________________________

SIGNATURE OF WITNESS ______________________________

SUBJECT ______________________________

NUMBER ______________ DATE ______________

PATIENT HISTORY ______________________________

____

____

____

RUN OBJECTIVE ______________________________

____

____

SUBJECT ROUTINE

The following was established for the routine to be carried out for each subject:

1) Explain procedure. Recheck no injury or surgery.
2) Sign consent form.
3) Record: a) age, b) height, c) weight.
   In addition, for amputees: Date of amputation.
   Prosthesis type, Canadian or Tucker-Syme.
   Functional limitation.
   Associated injuries.
   Measure stump length.
4) Record subject number and code and place in holder.
5) Palpate muscles and shave appropriate areas of both legs.
6) Alcohol rub skin (minimum 10 times firmly).
7) Attach foot switches.
8) Attach electrodes (12 pairs).
9) Attach goniometer and belt (side chosen by coin toss: Right = heads
   Left = tails).
10) Photograph ensuring scale on number holder in view.
11) Check on position for five switches.
12) Check electrode positions: a) hip flex
    b) extend knee
    c) flex knee
13) Do maximum contractions 3 second hold, three times.
14) Record attenuator sensitivity for all channels.

15) Do first limb run. Ascend and descend three times.

16) Check run printout.

17) If okay change leads to opposite leg and reattach goniometer.

18) Photograph other leg.

19) Repeat 12 and 13.

20) Check run printout.

21) If okay subject run completed.

DATA COLLECTION

The data were recorded on capture sheets. The development and design for the capture sheets is part of the project. Four of the seven data sheets are appended (see Appendix A). The other three are similar to the appended sheets except that they are duplicated for the opposite limb. Since three steps in each cycle were analysed, all the sheets have three readings for each variable. Each of the three readings for each variable had to be coded and identified for the limb studied and for the cycle. Four hundred and fifty-six manual measurements were recorded on the capture sheet for each subject. Each of these measurements was remeasured for accuracy.

All measurements were recorded to the nearest half millimeter. In the temporal unit, which is measured horizontally, this is the equivalent in time to .01 second or 10 milliseconds. In the electrogoniometric unit, which is measured vertically, this is equivalent to 1.5 degrees.

The data from the capture sheets of each subject necessitated the use of thirty computer cards each capable of storing eighty units of information. The data was punched on the cards twice using the standard verification method.

A Statistical Package for the Social Sciences (SPSS) program established in the Computing Department for the Health Sciences was adapted to obtain the appropriate data. The precise program was considered to be too lengthy to be appended to this thesis, as in total it weighs in excess of eighteen kilograms. However, the calculations follow established mathematical and statistical procedure.

The statistical method established in this study is adaptable to other locomotion studies.

CHAPTER IV

RESULTS AND DISCUSSION

INTRODUCTION

The results are presented in the following order:

1) Temporal data
2) Electrogoniometric data
3) Peak electromyographic data
4) Electrogoniometric data correlated with peak electromyographic data

Within each of the above four major sections there are the following eight sub-groups.

1) Normal group during ascending cycle, right limb
2) Normal group during ascending cycle, left limb
3) Normal group during descending cycle, right limb
4) Normal group during descending cycle, left limb
5) Amputee group during ascending cycle, amputated limb
6) Amputee group during ascending cycle, non-amputated limb
7) Amputee group during descending cycle, amputated limb
8) Amputee group during descending cycle, non-amputated limb

RELIABILITY

The concept of reliability refers to how accurate, on the average, the estimate of the true score is in a population of measurements of a particular variable, and is a form of analysis of variance.

The mathematics of this are derived from Specht (1976) and is stated in the following way. If all the variation in the observed scores is due to errors of measurement the reliability coefficient will be zero. If there is no error of measurement the reliability coefficient will be one.

This calculation was made for each variable on the raw data for all eight combinations. Appendix A shows the data collected for which this calculation was applied and it is presented in appendix B.

SAMPLE SIZE

This statistical test gives an indication of the capability of detecting a true difference of a certain degree within a specified probability with the number of subjects used in the study (Snedecor and Cochran, 1967). This could not be used prior to the study as is the usual case as even rough estimates of the variables sought were not available in most cases.

For the number of subjects studied the probability of showing a statistically significant difference ($p = .05$) in the sample when the true population has a certain degree of difference is .8.

The size of this true population difference varies in the two groups studied as well as from variable to variable, and it is presented in appendix C.

DATA ACCURACY

The data are recorded to the significant digit that is possible on the measuring instruments.

Most sources of error were considered to be constant because all measurements were made by a single observer and repeated at later times.

Temporal Data and Electromyographic Peak Position

These are horizontal measurements. The accuracy was estimated to no less than half a millimeter as determined by the ruler marks, paper speed, paper marking and other variables.

Electrogoniometric Data

These are vertical measurements. Since it was not possible to determine fractions of a degree with the electrogoniometer, only whole numbers are presented (see section on electrogoniometer accuracy).

Electromyographic Peak Actively Expressed as Percentage of the Maximum Contraction

These are vertical measurements. It was not thought valid to express units in fractions of a percent. Only whole numbers corrected to the nearest whole number are presented.

DATA PRESENTATION

To clarify the data presentation all the data tables for each variable studied will follow the format shown in Table 4.

1) Normal Group Data. The data for both limbs during ascending and descending is presented. In the far right column a comparison is made between the right and left limbs in the appropriate cycle. In the lowest line a comparison is made between the ascending and descending cycle of each limb. When comparisons are made only the level of significant difference or lack of it is noted.

TABLE 4. VARIABLE

Unit.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND			DIFFERENCE
DESCEND			DIFFERENCE
ASCEND vs. DESCEND	DIFFERENCE	DIFFERENCE	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND			DIFFERENCE
DESCEND			DIFFERENCE
ASCEND vs. DESCEND	DIFFERENCE	DIFFERENCE	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	DIFFERENCE		DIFFERENCE	
DESCEND	DIFFERENCE		DIFFERENCE	

2) Amputee Group Data. This follows the format of the normal group except the limbs are designated amputated and non-amputated. In the far right column a comparison is made between the amputated and non-amputated limb in the appropriate cycle. In the lowest line a comparison is made between the ascending and descending cycle of each limb. When comparisons are made only the level of significant difference or lack of it is noted.

3) Normal Group Data Compared with Amputee Group Data. This section first compares the pooled normal data with the non-amputated limb of the amputee during ascending and then descending. Secondly, the pooled normal data is compared with the amputated limb of the amputee during ascending and then descending. In these four comparisons the raw data is excluded for brevity as the actual units can be seen in the first two groups; only the level of significant difference or lack of it is noted.

Exceptions. Should the normal limb data demonstrate a significant difference between either limb, pooling the data for the purpose of comparing it as a group of "normal limbs" may not be appropriate. In this case, when the final "between group" difference or comparison is made, the right and left limbs which are amputated or non-amputated are identified. This allows the evaluation of the difference occurring because it was from the right or left side of the body, and not necessarily because it was amputated or non-amputated. Fortunately the amputee group had five right and five left amputees.

When discussing the data, the word "difference" is used only where there is a statistically significant difference in the comparison being made. The words "statistically significant" will be omitted for brevity.

The following key words assist in identification:

1) Only two groups are described, normal and amputee.

2) Right limb and left limb refer only to the normal group.

3) Amputated and non-amputated refers only to the amputee group.

4) Normal limbs refers only to the pooled right and left limbs in the normal group, when the pooling is appropriate.

Should there be fewer than 18 subjects for the normal group or 10 for the amputee group a number will be recorded in parenthesis after the appropriate title. However, the precise case comparisons are presented in the EMG section.

The significance levels on all tables will record only one of the following levels of significance: .05, .01, .005 and .001. If the calculated significance is greater than .001, e.g., .0001, it is expressed by <.001.

TEMPORAL DATA

In this section the following data are examined:

1) The cycle time in seconds.

2) the cadence.

3) The swing phase as a percent of the total cycle.

4) The stance phase as a percent of the total cycle.

5) The swing time in seconds.

6) The stance time in seconds.

RELIABILITY - See Appendix B.

SAMPLE SIZE - See Appendix C.

TABLE 5. CYCLE TIME.

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	1.68 ± 0.24	1.70 ± 0.28	NS
DESCEND	1.61 ± 0.20	1.61 + 0.19	NS
ASCEND vs. DESCEND	.001	<.05	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	2.02 ± 0.62	1.89 ± 0.58	<.05
DESCEND	1.91 ± 0.69	1.86 ± 0.57	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*		NS	
DESCEND	NS*		NS	

*Dispersion Significance <.01. Amputee > Normal.

CYCLE TIME (TABLE 5)

Unit - Seconds.

Cycle time is the time it takes from the start of swing in one cycle to start of swing in the next cycle, climbing stairs step over step.

Normal

There was no difference between either limb during ascending or descending. However, comparing ascending to descending, the cycle time was less during descending for both limbs.

Amputee

The amputee did not follow the normal pattern. During ascending the cycle time was longer for the amputated limb but there was no difference during descending. However, comparing ascending to descending no difference was found between the cycles.

Normal versus Amputee

There was no difference between any of the comparisons between the groups. However, the dispersion of the data was greater for the amputee in each of the comparisons.

Summary

Normal group subjects have a longer cycle time when ascending than descending and amputees do not. However, during ascending the amputated limb has a longer cycle time than the non-amputated limb.

TABLE 6. CADENCE

Unit - Number of steps/minute.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	73 ± 10	72 ± 10	NS
DESCEND	76 ± 9	75 ± 9	NS
ASCEND vs. DESCEND	<.001	<.05	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	63 ± 14	67 ± 12	.05
DESCEND	68 ± 16	68 ± 13	NS
ASCEND vs. DESCEND	<.05	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.05		NS	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

CADENCE (TABLE 6)

Unit - Number of steps/minute.

Normal

There was no difference between limbs during ascending or descending. However, comparing ascending to descending the cadence was faster during descending for both limbs.

Amputee

The amputee did not follow the normal pattern. During ascending the non-amputated limb had a faster cadence, but during descending no difference was found between either limb. Comparing ascending to descending only the amputated limb had a lower cadence during ascending.

Normal versus Amputee

During ascending the amputated limb had a slower cadence than normal. No differences were found in any other comparisons. However, during descending the amputee cadence had a greater dispersion than the normal.

Summary

Normal subjects have a faster cadence during descending and amputees only demonstrate this in the amputated limb. During ascending the amputated limb has a slower cadence than the non-amputated limb, and also the normal limbs.

CYCLE TIME AND CADENCE DISCUSSION (FIGURE 44)

It might appear that the significant differences found when comparing the data of these two variables should be the same, since cadence is a function of cycle time. As can be seen this is not so. The reason for the difference is that although cadence is indirectly proportional to the cycle time, it is not a linear transformation. Cadence is the cycle time divided into sixty. This accounts for the differences and highlights the fact that these two variables are different entities.

The cadence of the subjects in this study is slower than those of Berry (1952) who in his five subjects had an ascending cadence of 118 and a descending cadence of 121. This greater descending cadence is similar to the present study. Townsend et al. (1978, 1978A) had an ascending cadence of 87.6 and a descending cadence of 95.4. This cadence is faster than those of our study, but similar in that the descending cadence is faster. The faster cadence may be due to the younger age group and the highly trained athletes used in the above studies. Our study most closely resembles the cadence in McLeod et al. (1975) which showed an ascending cadence of 86 and a descending cadence of 87. This cadence was calculated by this author from the graphics in the paper.

It is interesting to note that although there is a difference in the ages of our two groups there is no significant difference in the cycle time and cadence except for the amputated limb. This fact tends to make the comparison of the two groups more meaningful. Should there have been any significant differences in cycle time and cadence in both ascending group comparisons it might have been due to age differences. However, this was not the case.

Fig. 44

CYCLE TIME- NORMAL AND AMPUTEE GROUPS (seconds)

CADENCE- NORMAL AND AMPUTEE GROUPS (Steps/Minute)

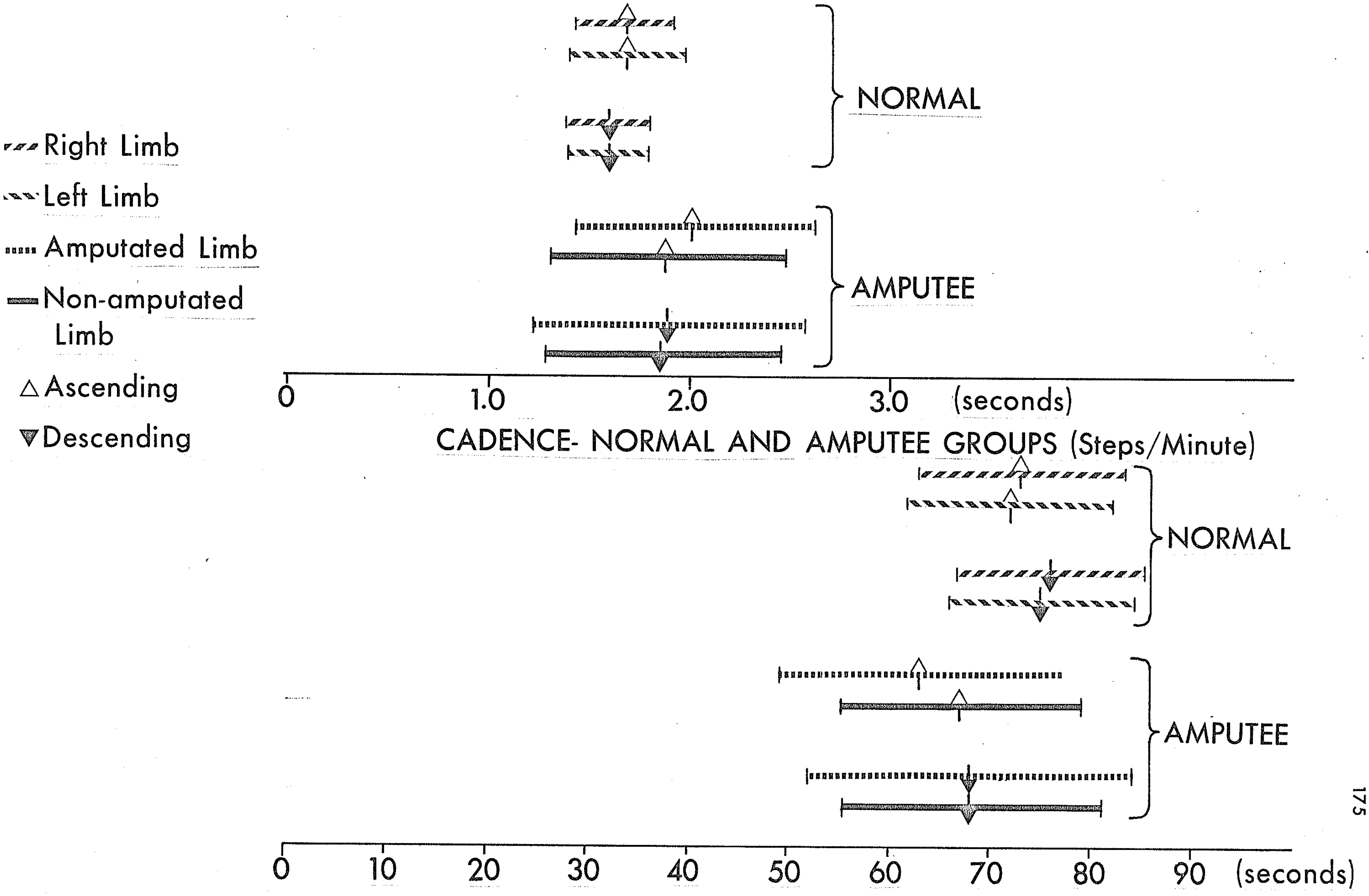


TABLE 7. SWING PHASE (The Non-Weight Bearing Phase of the Cycle)

Unit - Percentage of the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	39 ± 3	36 ± 4	<.05
DESCEND	41 ± 3	40 ± 3	NS
ASCEND vs. DESCEND	<.05	.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	42 ± 5	32 ± 6	<.001
DESCEND	50 ± 7	37 ± 6	<.01
ASCEND vs. DESCEND	<.05	<.01	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	.05	<.05	NS*	NS
DESCEND	<.001	<.05	NS*	NS

*Dispersion Significance <.01. Amputee > Normal.

SWING PHASE (The Non-Weight Bearing Phase of the Cycle) (TABLE 7)

Unit - Percentage of the cycle.

Normal

During ascending the right limb swing percent was greater than the left, however, there was no difference between either limb during descending. In comparing ascending to descending, a smaller percent of the cycle was spent in swing during ascending.

Amputee

The amputee did not follow the normal pattern. During both ascending and descending the amputated limb had a greater percentage of swing than the non-amputated limb. In comparing ascending to descending a smaller percent of the cycle was spent in swing during ascending.

Normal versus Amputee

As there were differences between normal right and left limbs the inter-group comparisons should be made between right and left limbs. As can be seen during ascending and descending the amputated limb had a greater swing percent than the normal. During ascending and descending the right non-amputated limbs had a greater dispersion than the right limbs.

Summary

During ascending the right limb had a larger swing percent than the left. Comparing ascending to descending both ascending right and left limbs had smaller swing percents.

During ascending and descending the amputated limb had a larger swing percent, and like the normal group both limbs during ascending had a smaller swing percent.

During ascending and descending the amputated limb had a greater swing percent.

TABLE 8. STANCE PHASE (The Weight Bearing Phase of the Cycle)

Unit - Percentage of the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	61 ± 3	64 ± 4	<.05
DESCEND	59 ± 3	60 ± 3	NS
ASCEND vs. DESCEND	<.05	.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	58 ± 5	68 ± 6	<.001
DESCEND	50 ± 7	63 ± 6	<.01
ASCEND vs. DESCEND	<.05	<.01	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	.05	<.05	NS*	NS
DESCEND	<.001	<.05*	NS*	NS

*Dispersion Significance <.01. Amputee > Normal.

STANCE PHASE (The Weight Bearing Phase of the Cycle) (TABLE 8)

Unit - Percentage of the cycle.

Here the significant differences that exist in the comparisons are the same as those of the swing percent because swing and stance are complementary namely swing and stance = 100% cycle.

SWING/STANCE DISCUSSION (FIGURE 45)

Normal

Perhaps the most striking observation that can be made in observing the swing/stance data is the high degree of similarity of normal stair climbing to normal walking, namely 40% swing, 60% stance (Murray et al., 1964; Murray et al., 1966; Murray, 1967). This may relate to some optimal biologically efficient ratio between swing and stance in horizontal and vertical locomotion. This was also noted in Townsend's studies (1977, 1977A).

During ascending a greater part of the cycle is spent in weight-bearing on the left leg than on the right leg. Although there is a significant difference the means are similar. The significance of this is not apparent and may be related to dominance of a limb. However this dominance is not demonstrated in descending and thus may be related to the more energy consuming activity of ascending. As explained earlier in the Materials and Methods section, detecting dominance in the lower limb is not feasible as in the upper limb.

During descending stairs less of the cycle is spent in weight-bearing than in ascending. This could be due to the gravity assistance of descending and perhaps more significantly, the longer time needed in weight-bearing during ascending to oppose the force of gravity.

Amputee

During ascending and descending the amputee spends longer on the non-amputated limb than the amputated limb. When comparing ascending to descending a pattern similar to the normal group is seen, namely that there is greater weight-bearing during ascending than descending. The most distinctive feature is perhaps the equal weight-bearing and non-weight-bearing of the amputated limb during descending which may indicate a reluctance to bearing weight through the prosthesis when going down stairs. Although there was no significant difference between the weight-bearing of the non-amputated limb and the normal in ascending there is a tendency for the amputee to spend longer on the non-amputated limb.

Normal versus Amputee

During both ascending and descending the amputated limb spent less of the cycle in weight-bearing than the normal. The non-amputated limb was not significantly different to the normal limbs but showed a tendency for a higher percent of the cycle to be spent in weight-bearing.

This tends to indicate that the amputee spends less time bearing weight through the prosthesis than the non-amputated limb and normal limbs.

Fig. 45

SWING AND STANCE PERCENT OF THE STAIR CYCLE

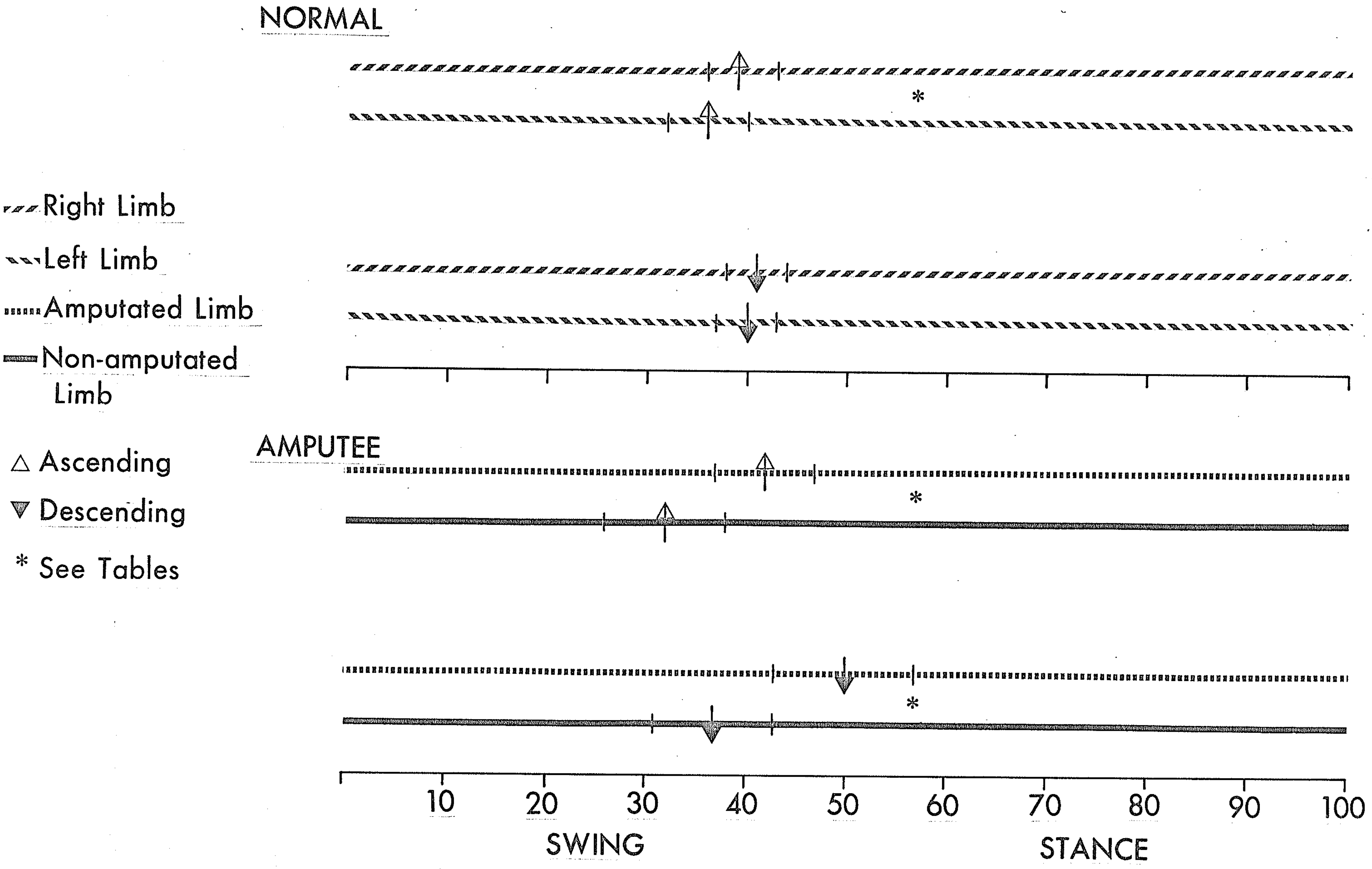


TABLE 9. SWING TIME (Duration of Non-Weight Bearing)

Unit - Seconds

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.66 ± 0.10	0.61 ± 0.84	<.05
DESCEND	0.66 ± 0.10	0.64 ± 0.08	NS
ASCEND vs. DESCEND	NS	<.05	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	0.83 ± 0.16	0.58 ± 0.07	.001
DESCEND	1.00 ± 0.55	0.67 ± 0.10	NS
ASCEND vs. DESCEND	NS	<.05	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.05	<.05*	NS	<.05*
DESCEND	<.01	NS*	NS	NS

*Dispersion Significance <.01. Amputee > Normal.

SWING TIME (Duration of Non-Weight Bearing) (TABLE 9)

Unit - Seconds

Normal

During ascending the right limb swing time was longer than the left, however, there was no difference between limbs in descending. In comparing ascending to descending no difference was found in the right leg but during descending the left leg had a longer swing time.

Amputee

The amputee followed a similar pattern to the normal in limb differences and cycle differences. During ascending the amputated limb had a longer swing time than the non-amputated, however in descending there was no difference between limbs. In comparing ascending to descending there was no difference in the amputated limbs, however the non-amputated limb had a longer swing time in descending.

Normal versus Amputee

During ascending the amputated limb had a longer swing time than the normal. During descending the amputated limb had a longer swing time than normal in the right but not left limbs.

During ascending the non-amputated limb had a shorter swing time than the normal in the left limb but not the right. During descending there was no difference between the normal and the non-amputated limbs swing time.

Summary

Both groups showed similar patterns. During ascending there was a difference between limbs in both groups but not in descending. In comparing groups the only instance where the swing time difference was evident in both limbs was in comparing the amputated limb to the normal limbs.

TABLE 10. STANCE TIME (Duration of Weight Bearing)

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	1.02 ± 0.17	1.09 ± 0.23	NS
DESCEND	0.95 ± 0.12	0.97 ± 0.12	NS
ASCEND vs. DESCEND	<.001	<.01	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	1.19 ± 0.48	1.31 ± 0.56	NS
DESCEND	0.91 ± 0.17	1.19 ± 0.53	NS
ASCEND vs. DESCEND	<.05	<.005	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*		NS*	
DESCEND	NS		NS*	

*Dispersion Significance <.01. Amputee > Normal.

STANCE TIME (Duration of Weight Bearing) (TABLE 10)

Unit - Seconds.

Normal

During ascending and descending there was no difference between limbs. In comparing ascending to descending the stance time was longer during ascending.

Amputee

The amputee followed a similar pattern to the normal in limb similarities and cycle differences. During ascending and descending there was no difference between limbs. In comparing ascending to descending the stance time was longer during ascending.

Normal versus Amputee

There was no difference between any of the group comparisons. There was a significantly greater dispersion in the amputee than the normal in the comparisons marked with an asterisk.

Summary

Both groups showed similar patterns, the ascending time being longer than the descending. There were no differences between groups.

STANCE AND SWING TIME DISCUSSION

The presentation of these two times is for the possible use in functional electrical stimulation.

Discussion of the significance is difficult as a frame of reference is not present as in comparing the percentages.

It is perhaps interesting to note that differences do exist between the groups in the swing time but not the stance time.

ELECTROGONIOMETRIC DATA

In this section the following data are examined:

1) The knee joint angle at the start of the swing phase.

2) The knee joint angle at the start of the stance phase.

3) (a) The maximum knee flexion during stair cycle.

(b) The position of this maximum knee flexion with respect to the cycle.

(c) The time of this maximum knee flexion with respect to the start of the cycle.

4) (a) The minimum knee flexion during the stair cycle.

(b) The position of this minimum knee flexion with respect to the cycle.

(c) The time of this minimum knee flexion with respect to the start of the cycle.

5) The total range of movement the knee moves through during the stair cycle.

ELECTROGONIOMETRY OF THE KNEE JOINT DURING THE STAIR CYCLE

The start of the stair cycle was chosen to be the start of the swing phase, which means that the sequence demonstrated is swing to stance. The sequence swing to stance as opposed to stance to swing was chosen because the electrogoniometric changes occurring from the swing to stance phase showed a more complex transition than from the stance to swing. In stance to swing in both ascending and descending cycles there is only a smooth transition from weight bearing to non-weight bearing while the knee is flexing. However, from swing to stance there is a significant change in knee motion.

The following is a description of the stair cycle.

Ascending

The ascending cycle demonstrates two dynamic angular changes per cycle (Figs. 46, 47). The knee is flexing from a nearly fully extended position as the foot leaves the step below being relieved of its weight. Swing phase commences with the knee flexing rapidly to clear the step above. Maximum flexion is reached three-fifths of the way through the swing phase. After the step is cleared the foot is lowered and the knee extended. As contact is made with the step, knee extension plateaus or extension increases more slowly, as weight is transferred to the step. Extension increases as the body is lifted towards the next step. Minimum flexion (or maximum extension) occurs during stance phase near the end of the cycle. Flexion now begins again as weight is mostly being born on the opposite leg, and swing phase starts again.

Descending

The descending cycle also demonstrates two dynamic angular changes per cycle (Fig. 48, 49). The knee is flexing to near its maximum flexed position as the foot leaves the step being relieved of its weight. Swing phase commences with the knee in a flexed position reaching maximum flexion almost immediately for the foot to clear the step it is leaving. The knee is rapidly extended as the foot seeks the step below reaching minimum flexion nine-tenths of the way through swing. The knee begins

Fig. 46

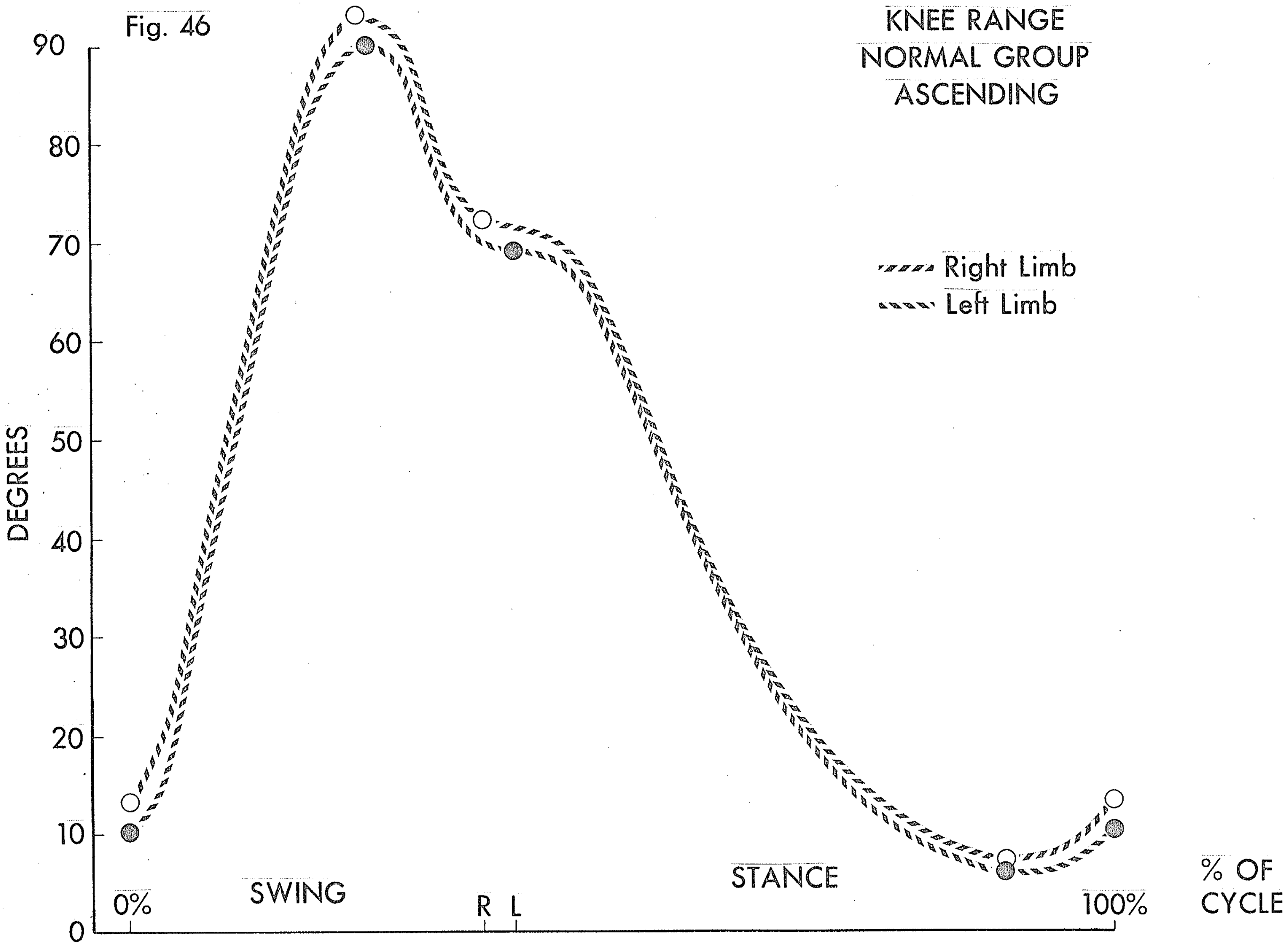
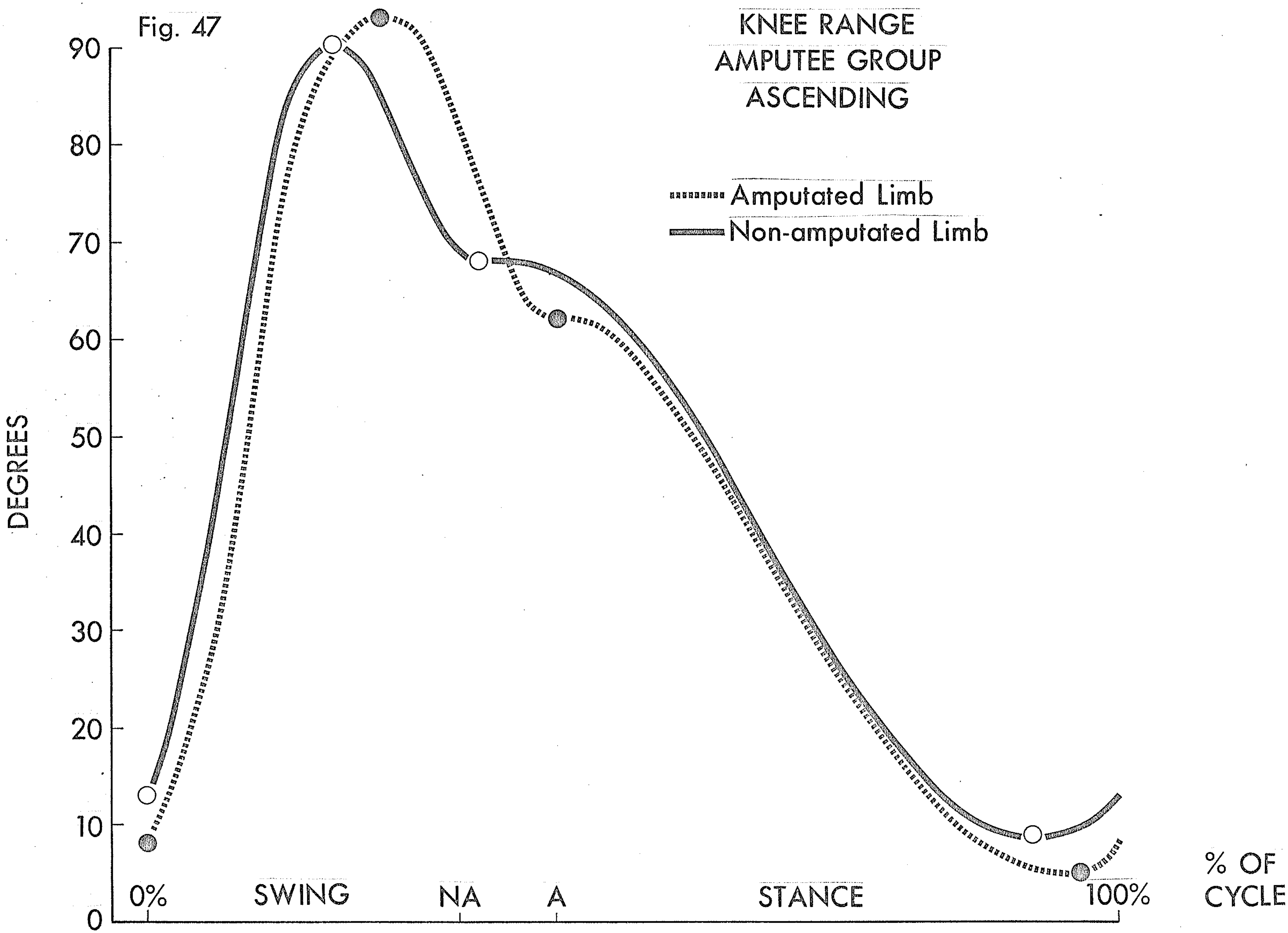

KNEE RANGE
NORMAL GROUP
ASCENDING
Right Limb
Left Limb
DEGREES
90
80
70
60
50
40
30
20
10
0
0%
SWING
R
L
STANCE
100%
% OF
CYCLE

Fig. 47


KNEE RANGE
AMPUTEE GROUP
ASCENDING
Amputated Limb
Non-amputated Limb
DEGREES
0
10
20
30
40
50
60
70
80
90
0%
SWING
NA
A
STANCE
100%
% OF
CYCLE

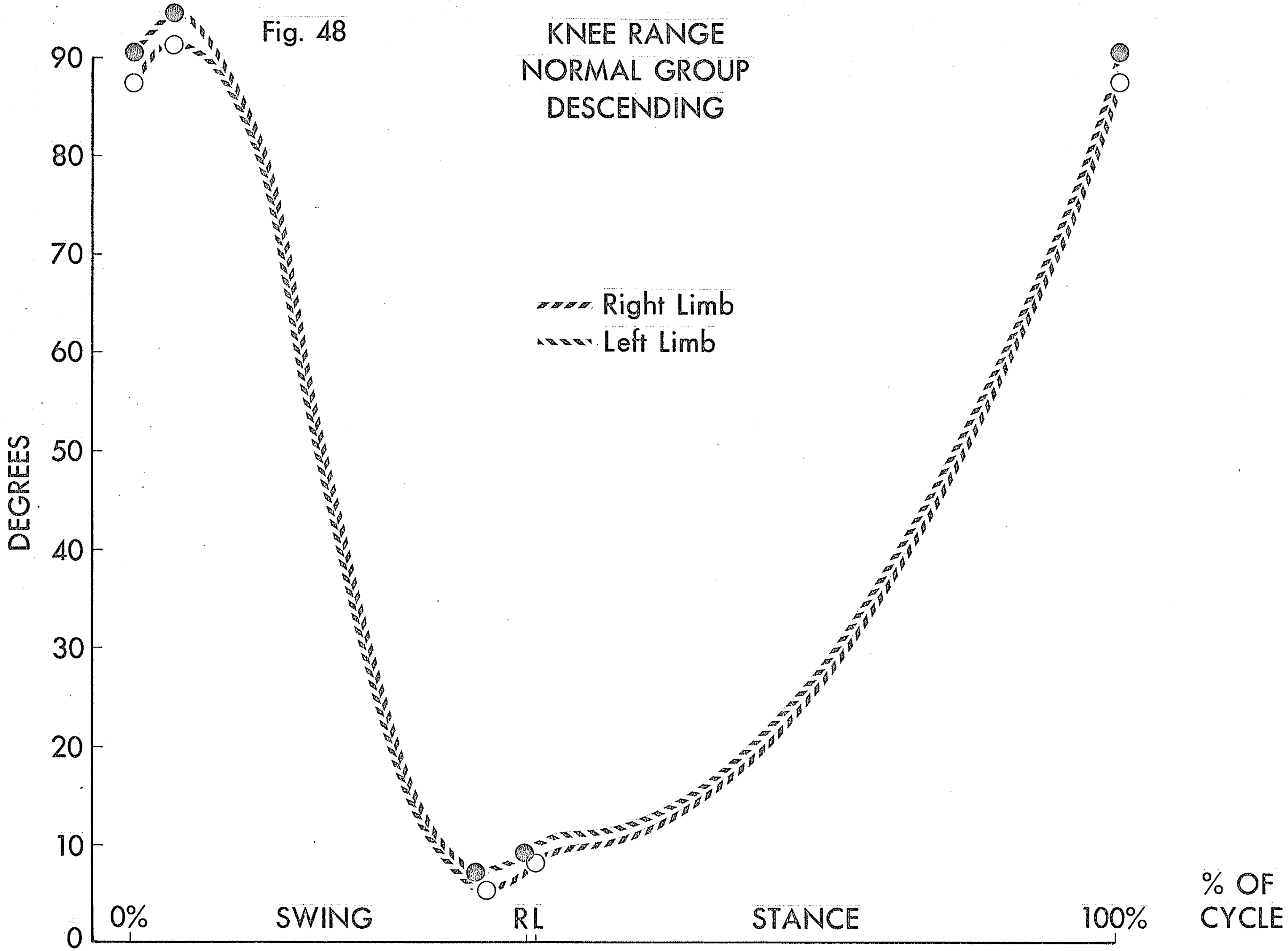


Fig. 48

KNEE RANGE
NORMAL GROUP
DESCENDING

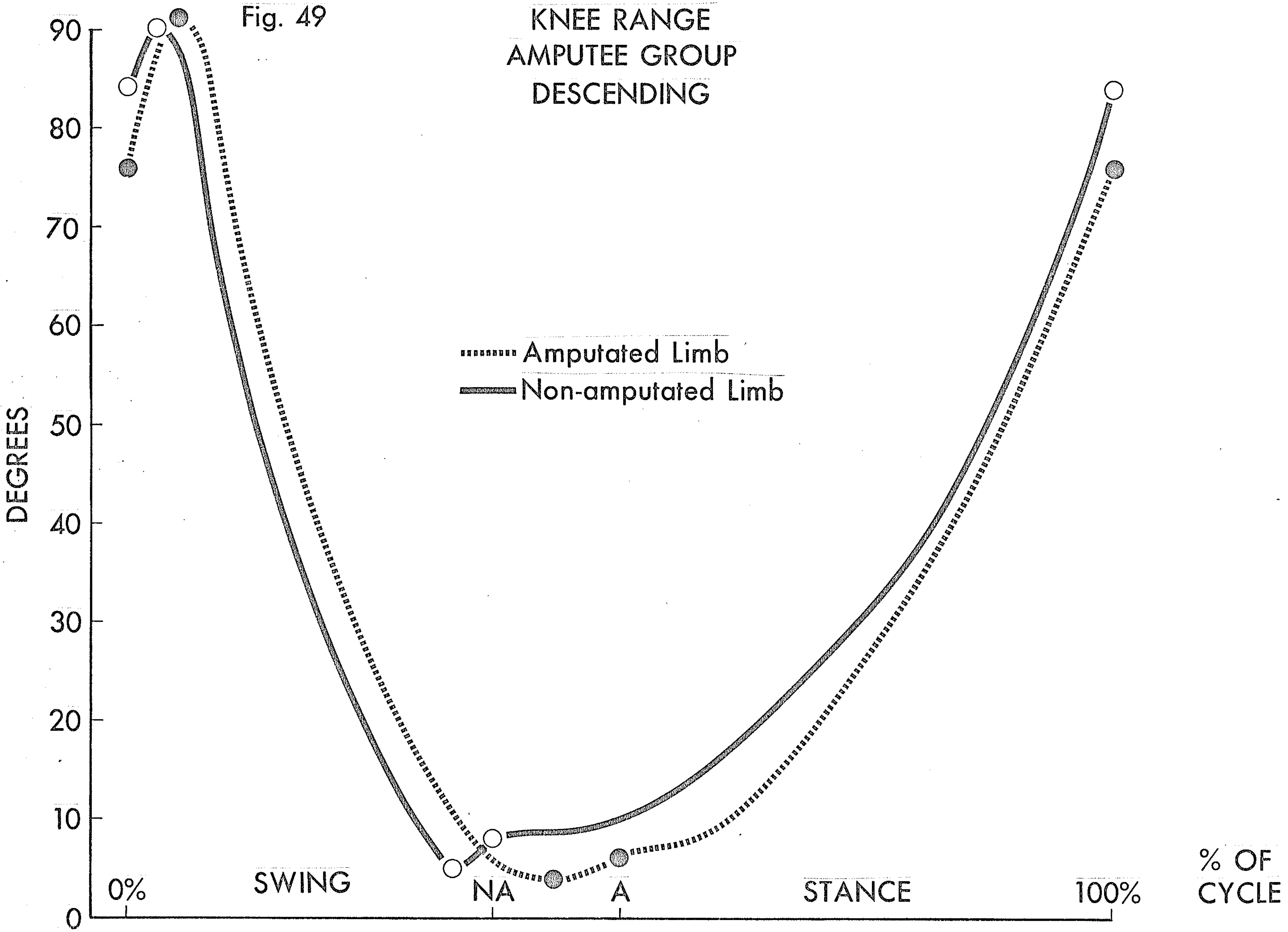


Fig. 49

KNEE RANGE
AMPUTEE GROUP
DESCENDING

flexion again in the swing phase. At the start of stance the knee is only slightly flexed and weight-bearing plateaus or slows flexion initially. Flexion continues as the body is lowered to the next step and concurrently weight is slowly relieved from the foot to commence the swing phase again.

It is interesting to note that one angular change occurs at the knee during ascending in swing phase, but both angular changes occur during descending in swing phase. In the stance phase one angular change occurs during ascending and never during descending.

Stair Size

There is no doubt that the size of the stair is important in the consideration of the goniometry. Unfortunately, the recording of the size of the steps and the number of stairs and the stair slope is not consistent in the literature. For example, in Laubenthal's et al. (1972) electrogoniometric study the dimensions of the two-step platform were not stated but the stair slope was. This alone does not allow a calculation of the stair dimension as can be seen below where two identical slopes have different stair size.

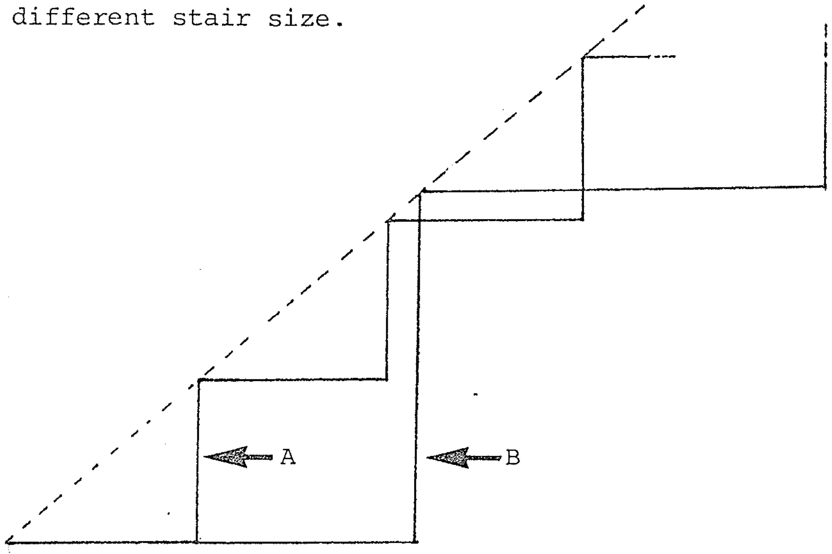


FIGURE 50. STAIR SLOPE.

Here the rise of the stair is over twice as great in (B) than in (A) but the slope is the same. It may be assumed that the higher the step the greater the flexion needed.

Other considerations should also be taken into account, such as the number of stairs. In short flights, e.g. two steps, the body may not need to incline the same way as when there are more than two steps. This inclination would most probably modify the knee angle.

The stair slope does give an idea of angle of inclination and the following are the slopes in all the stair studies.

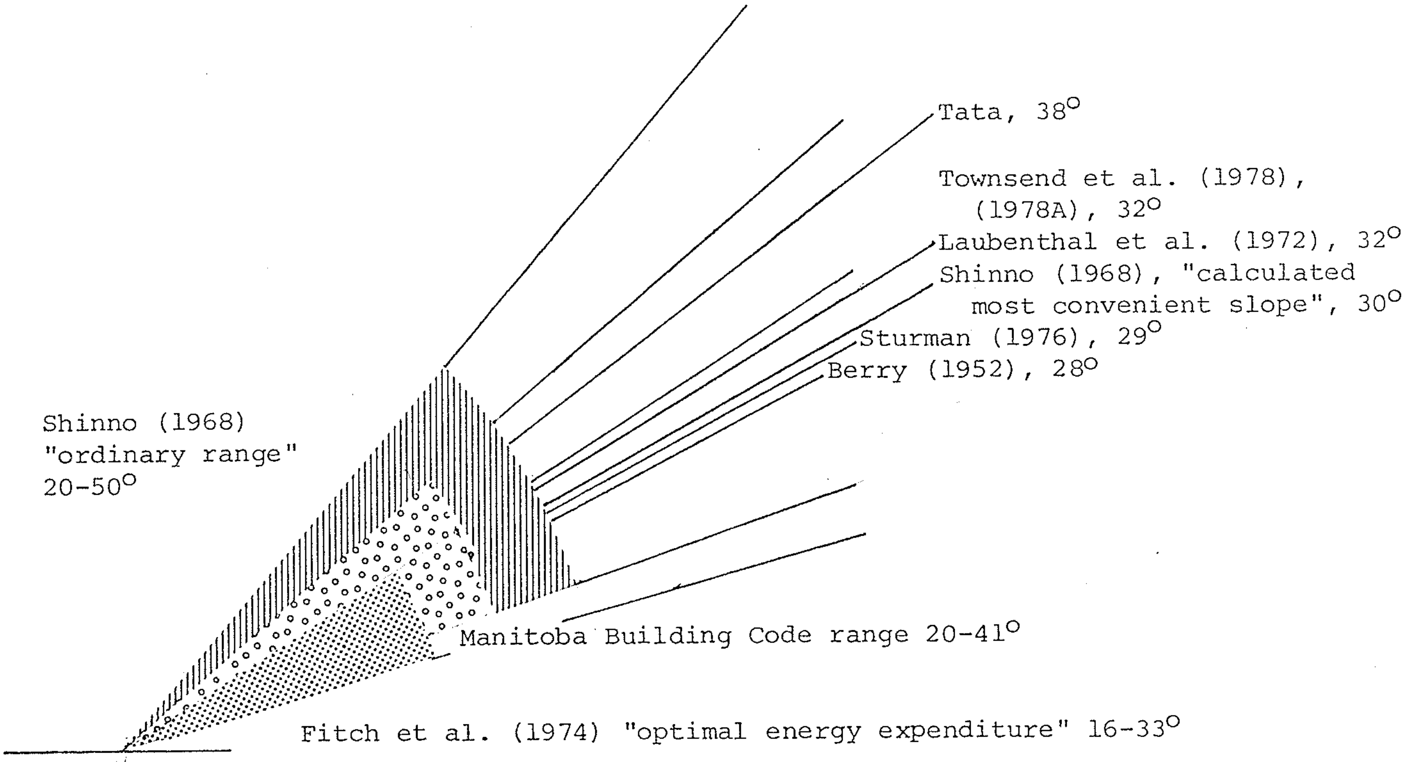


FIGURE 51. STAIR SLOPES IN STUDIES

TABLE 11. KNEE ANGLE AT THE START OF SWING PHASE

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	10 ± 7	13 ± 7	NS
DESCEND	87 ± 9	90 ± 7	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	8 ± 6	13 ± 7	NS
DESCEND	76 ± 16	84 ± 8	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	<.05*		NS	

*Dispersion Significance <.05. Amputee > Normal.

All the following electromyographic data is accurately transcribed to the appropriate electrogoniograms in the previous Figures 46 to 49.

KNEE ANGLE AT THE START OF SWING PHASE (TABLE 11)

Unit - Degrees.

Normal

There was no difference between either limb during ascending or descending. Comparing ascending to descending statistically is not meaningful because they are obviously different as can be seen from Figures 46 to 49.

Amputee

The amputee followed a similar pattern to the normal. There was no difference between either limb during ascending or descending.

Normal versus Amputee

The only difference was found during descending, in that the amputated limb flexed less than the normal and there was a greater dispersion of the amputee data.

Summary

During descending stairs the amputated limb did not flex as much as the normal. This may be due to the fact that the prosthesis does not allow dorsiflexion of the ankle which in turn limits the amount of flexion at the knee.

TABLE 12. KNEE ANGLE AT THE START OF STANCE PHASE

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	69 ± 9	71 ± 8	NS
DESCEND	8 ± 6	9 ± 6	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	62 ± 16	68 ± 8	NS
DESCEND	6 ± 9	8 ± 4	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*		NS	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

KNEE ANGLE AT THE START OF STANCE PHASE (TABLE 12)

Unit - Degrees.

Normal

There was no difference between limbs during ascending or descending. Comparing ascending to descending statistically is not meaningful because they are obviously different as can be seen from Figure 46 to Figure 49.

Amputee

The amputee followed a similar pattern to the normal. There was no difference between either limb during ascending or descending.

Normal versus Amputee

There was no difference in any of the group comparisons. However, in ascending and descending the amputated limb had a greater dispersion of the data than the normal.

Summary

There was no difference in the knee angle at the start of stance between the two groups.

TABLE 13. MAXIMUM KNEE FLEXION DURING THE STAIR CYCLE

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	90 ± 11	93 ± 10	NS
DESCEND	91 ± 10	94 ± 8	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	93 ± 13	90 ± 8	NS
DESCEND	91 ± 12	90 ± 7	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	NS		NS	

MAXIMUM KNEE FLEXION DURING THE STAIR CYCLE (TABLE 13)

Unit - Degrees.

Normal

There was no difference between limbs during ascending or descending. In comparing ascending to descending cycles there were no diferences found.

Amputee

The amputee followed a similar pattern to the normal in limb similarities and cycle similarities. There was no difference between limbs during ascending or descending. In comparing ascending to descending cycles there were no differences.

Normal versus Amputee

There was no difference between any of the group comparisons.

Summary

Amputees and normals flex their knees to the same extent during both stair cycles. This occurs during the swing phase for both cycles and it is interesting to note that the lack of flexion in the amputee at the start of swing is made up during the swing phase.

TABLE 14. POSITION OF MAXIMUM KNEE FLEXION WITH RESPECT TO THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	24 ± 1	23 ± 3	NS
DESCEND	4 ± 2	4 ± 2	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	23 ± 3	19 ± 4	<.001
DESCEND	5 ± 3	3 ± 2	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		<.005	
DESCEND	NS		NS	

POSITION OF MAXIMUM KNEE FLEXION WITH RESPECT TO THE CYCLE (TABLE 14)

Unit - Percentage point in the cycle.

Normal

There was no difference between limbs during ascending or descending. Comparing ascending to descending statistically is not meaningful because they are obviously different as can be seen from Figure 46 to Figure 49.

Amputee

The amputee did not follow a similar pattern to the normal. During ascending maximum flexion occurred earlier in the cycle in the non-amputated limb. However, there was no difference between limbs during descending.

Normal versus Amputee

During ascending the non-amputated limb reached maximum flexion earlier in the cycle than the normal limb. There was no difference in any of the other group comparisons.

Summary

During ascending the non-amputated limb reached maximum flexion earlier in the cycle than the amputated limb and the normal limb. The reason for this may be because during maximum flexion of the non-amputated limb, the amputee is bearing weight through the amputated limb. Since there is a preference for weight-bearing through the non-amputated limb, possibly because of the greater stability and sensory input, etc., maximum flexion occurs earlier in the cycle as an indication of the shorter swing phase. The knee is controlled well enough not to be flexed significantly more than the normal limb, but just earlier in the cycle.

TABLE 15. TIME OF MAXIMUM KNEE FLEXION WITH RESPECT TO THE START OF THE CYCLE

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.40 ± 0.06	0.39 ± 0.07	NS
DESCEND	0.06 ± 0.03	0.06 ± 0.03	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	0.48 ± 0.10	0.35 ± 0.04	<.005
DESCEND	0.11 ± 0.07	0.06 ± 0.03	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.005		<.05	
DESCEND	NS*		NS	

*Dispersion Significance <.001. Amputee > Normal.

TIME OF MAXIMUM KNEE FLEXION WITH RESPECT TO THE START OF THE CYCLE

(TABLE 15)

Unit - Seconds.

Normal

As for maximum flexion position.

Amputee

As for maximum flexion position.

Normal versus Amputee

During ascending the normal and non-amputated limbs follow the maximum flexion position. However, during ascending the amputated limb takes longer to reach maximum flexion than the normal.

Summary

The findings of a quicker maximum flexion tend to corroborate the statement in the maximum flexion summary of the amputee having a preference for weight-bearing through the non-amputated limb.

During ascending the delayed amputated limb peak compared with the normal may be due to the significantly longer swing time in the ascending cycle of the amputated limb.

TABLE 16. MINIMUM KNEE FLEXION DURING THE STAIR CYCLE

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	6 ± 6	7 ± 6	NS
DESCEND	5 ± 6	7 ± 6	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	5 ± 8	9 ± 4	NS
DESCEND	4 ± 10	5 ± 6	NS
ASCEND vs. DESCEND	NS	<.01	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

MINIMUM KNEE FLEXION DURING THE STAIR CYCLE (TABLE 16)

Unit - Degrees.

Normal

There was no difference between limbs during ascending or descending. In comparing ascending to descending cycles there was no difference.

Amputee

The amputee did not follow a similar pattern to the normal. There was no difference between limbs during ascending or descending. In comparing ascending to descending cycles there was no difference in the amputated limb. However, the non-amputated limb did not extend as much in the minimum flexion position during ascending.

Normal versus Amputee

There was no difference between any of the group comparisons. However, during descending the amputee data has a greater dispersion than the normal.

Summary

Amputees and normals extend their knees to the same extent during both stair cycles. However, in comparing ascending to descending the non-amputated limb extends less during ascending. This may be due to minimum flexion during ascending being in the non-weight-bearing period and during ascending in the weight-bearing one. The greater flexion (or lack of extension) may be for greater stability during weight-bearing.

(Stability refers not to the stability of the knee joint but the stability of the subject during the stair cycle. This is amplified in the electrogoniometry summary)

TABLE 17. POSITION OF MINIMUM KNEE FLEXION WITH RESPECT TO THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	89 ± 6	89 ± 7	NS
DESCEND	36 ± 5	35 ± 4	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	96 ± 3	91 ± 9	NS
DESCEND	43 ± 13	33 ± 6	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.001**		NS	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

**Dispersion Significance <.05. Normal > Amputee.

POSITION OF MINIMUM KNEE FLEXION WITH RESPECT TO THE CYCLE (TABLE 17)

Unit - Percentage point in the cycle.

Normal

There was no difference between limbs during ascending or descending. Comparing ascending to descending statistically is not meaningful because they are obviously different as can be seen from Figure 46 to Figure 49.

Amputee

The amputee followed a similar pattern to the normal. There was no difference between limbs during ascending or descending.

Normal versus Amputee

During ascending the amputated limb reached minimum flexion later in the cycle than the normal, and the normal data had a greater distribution than the amputated limb. Although there was no difference during descending between the amputated limb and the normal, the amputated limb data had a significantly greater dispersion than the normal.

Summary

Minimum flexion occurs later in the cycle in the amputated limb than in the normal. This delay or slower extension may be to allow the opposite non-amputated limb more time to reach the step above and thereby the amputee is more stable as minimum extension is reached. It is interesting to note that the normal data is more dispersed than the amputee. This is contrary to the majority of dispersion significance and highlights the greater uniformity of the amputees in this regard.

TABLE 18. TIME OF MINIMUM KNEE FLEXION WITH RESPECT TO THE START OF THE CYCLE

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	1.50 ± 0.23	1.50 ± 0.21	NS
DESCEND	0.60 ± 0.09	0.58 ± 0.08	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	1.94 ± 0.59	1.69 ± 0.40	<.05
DESCEND	0.91 ± 0.57	0.60 ± 0.06	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.05		NS*	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

TIME OF MINIMUM KNEE FLEXION WITH RESPECT TO THE START OF THE CYCLE

(TABLE 18)

Unit - Seconds.

Normal

As for minimum flexion position.

Amputee

During ascending the amputated limb took longer to reach the minimum flexion position than the non-amputated limb. However, there was no difference in descending.

Normal versus Amputee

The significant difference is the same as for the minimum flexion position. There was a significantly greater dispersion of the data for the amputee than the normal, in descending normal compared with amputated and ascending normal compared with non-amputated.

Summary

During ascending minimum flexion occurs later in the amputated limb than in the non-amputated limb and the normal.

TABLE 19. TOTAL RANGE OF KNEE MOVEMENT DURING THE STAIR CYCLE

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	85 ± 11	85 ± 9	NS
DESCEND	86 ± 11	87 ± 9	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	87 ± 8	81 ± 8	.001
DESCEND	86 ± 8	85 ± 7	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	NS		NS	

TOTAL RANGE OF KNEE MOVEMENT DURING THE STAIR CYCLE (TABLE 19)

Unit - Degrees.

Normal

There was no difference between limbs during ascending or descending. In comparing ascending to descending cycles there was no difference.

Amputee

The amputee did not follow the normal pattern. During ascending the non-amputated limb moved through a smaller range than the amputated limb. However, there was no difference in descending. In comparing ascending to descending cycles there was no difference.

Normal versus Amputee

There was no difference in total knee range between any of the group comparisons.

Summary

Amputees and normals move their knees through the same range during both stair cycle. However, during ascending stairs the non-amputated limb moves through a significantly smaller range than the amputated limb. It would appear that this difference occurs at the knee extension or minimum flexion range and not at the maximum flexion. (See previous data)

ELECTROGONIOMETRY SUMMARY

There appears to be an overall similarity between the electrogoniometric data of the two groups with the following exceptions.

1) At the start of swing phase during descending the amputated limb is not as flexed as the normal, possibily due to the lack of dorsiflexion of the prosthesis.

2) The position of maximum knee flexion with respect to the ascending cycle is earlier in the non-amputated limb than in the amputated, and also earlier than in the normal limb. This indicates faster flexion of the non-amputated limb, and is also indicative of the shorter swing and lack of reliance on the amputated limb.

3) The position of minimum knee flexion with respect to the ascending cycle occurs later in the amputated limb than the normal. This indicates slower extension of the amputated limb.

4) The total range ofknee motion is less during ascending in the non-amputated limb than in the amputated. The total range of movement in both groups, however, is not significantly different.

Stability refers not to the stability of the knee joint but to the stability of the subject during the stair cycle. With the knee flexed there is a greater capability for balance because the body can be moved in both directions by flexing and extending the knee to maintain balance of the body. However, with the knee more extended the ability to control movement is less as less range is available in one of the two directions.

ELECTROMYOGRAPHIC DATA

In this section the following data are examined:

1) The position of the EMG peak.

(a) The position of the EMG peaks expressed with respect to the cycle.

(b) The time and duration of specified peaks with respect to the start of the cycle.

2) The sequence of the EMG peak.

(a) The consideration of sequence or pattern of the quadriceps EMG peaks.

(b) The consideration of sequence or pattern of the hamstrings EMG peaks.

3) The amplitude of the EMG peaks.

(a) The amplitude of each of the quadriceps and hamstring peaks expressed as a percentage of a maximum voluntary contraction.

RELIABILITY - See Appendix B.

SAMPLE SIZE - See Appendix C.

INTRODUCTION

Prior to the consideration of the EMG peaks the following should clarify the data presentation.

1) *Vastus medialis oblique is designated M1.* This muscle demonstrates only one peak in each cycle.

2) *Vastus medialis longus is designated M2.* This muscle demonstrates only one peak in each cycle.

3) *Rectus femoris is designated M3.* This muscle demonstrated three possible peaks during the cycles:

(a) An early peak which occurs in swing phase, designated M31.

(b) The usual peak, which is the peak that occurs with the other quadriceps, designated M32.

(c) The late peak which occurs in late stance phase, designated M33.

No muscle demonstrated more than two peaks, i.e., all had an M32 and others had either an M31 or an M33 but not both.

Early Rectus Femoris Peak (M31)

In order to understand the incidence a table for the normal (Table 20) and amputee (Table 21) have been completed. As can be seen the patterns are complex.

During ascending 22% of the normals and 50% of the amputees had at least one early peak; however during descending 72% of the normals and 90% of the amputees had at least one peak.

In comparing both groups 72% of the normal and 90% of the amputees had at least one peak.

Table 22 shows the case numbers in the comparisons of M31 peaks.

Late Rectus Femoris Peak (M33)

The late rectus femoris peak was demonstrated only in the ascending cycle. In the normal group there were three left limbs demonstrating this peak. In the amputee group there were also three, all being in the non-amputated limb.

TABLE 20 - INCIDENCE OF EARLY RECTUS FEMORIS PEAK (M31) IN THE NORMAL GROUP.

(The number of subjects is indicated in parentheses)

CONSIDERING BOTH CYCLES

NO PEAK
28% (5)

PEAK
72% (13)

ASCENDING RIGHT	ASCENDING LEFT	DESCENDING RIGHT	DESCENDING LEFT	
X	X	X	X	6%(1)
	X	X	X	6%(1)
X			X	6%(1)
	X		X	6%(1)
		X	X	50%(9)
11%(2)	17%(3)	61%(11)	72%(13)	

CONSIDERING ASCENDING

NO PEAK
78% (14)

PEAK
22% (4)

UNILATERAL	BILATERAL
17% (3)	6% (1)

CONSIDERING DESCENDING

NO PEAK
28% (5)

PEAK
72% (13)

UNILATERAL	BILATERAL
11% (2)	61% (11)

TABLE 21 - INCIDENCE OF EARLY RECTUS FEMORIS PEAK (M31) IN THE AMPUTEE GROUP.

(The number of subjects is in parentheses)

CONSIDERING BOTH CYCLES

NO PEAK
10% (1)

PEAK
90% (9)

ASCENDING AMPUTATED	ASCENDING NON-AMPUTATED	DESCENDING AMPUTATED	DESCENDING NON-AMPUTATED	
X	X	X	X	20%(2)
X	X	X		10%(1)
X		X	X	10%(1)
X			X	10%(1)
		X	X	10%(1)
			X	20%(2)
		X		10%(1)
50%(3R + 2L)	30%(1R + 2L)	60%(3R + 3L)	70%(3R + 4L)	

CONSIDERING ASCENDING

NO PEAK
50% (5)

PEAK
50% (5)

UNILATERAL
20% (2)

BILATERAL
30% (3)

CONSIDERING DESCENDING

NO PEAK
10% (1)

PEAK
90% (9)

UNILATERAL
50% (5)

BILATERAL
40% (4)

TABLE 22. CASE NUMBER OF EARLY RECTUS FEMORIS PEAKS (M31)

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND			1 Case (NSD)
DESCEND			11 Cases
ASCEND vs. DESCEND	1 Case (NSD)	3 Cases	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND			3 Cases
DESCEND			4 Cases
ASCEND vs. DESCEND	4 Cases	2 Cases	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	2 N 3 A	3 N 2 A	2 N 1 A (NSD)	3 N 2 A
DESCEND	11 N 3 A	13 N 3 A	11 N 3 A	13 N 4 A

NSD = Not Sufficient Data to make a comparison.

4) *Vastus lateralis is designated M4.* This muscle demonstrates only one peak in each cycle.

5) *The lateral hamstring group is designated M5.* Some subjects demonstrated two peaks. M51 designates the only peak when a single peak is present and the first of two peaks when two are present. M51 is present in all subjects, normal and amputee, M52 in only some.

The second lateral hamstring peak (M52)incidence is shown in Tables 23 and 24. As in M31, the patterns are complex.

In comparing both groups 94% of the normal and 100% of the amputees showed at least one second peak.

During ascending 89% of the normal and 90% of the amputees had at least one peak; and during descending 78% of the normal and 80% of the amputees showed at least one second peak.

Table 25 shows the case numbers in the comparison of the M52 peaks.

6) *The medial hamstring group is designated M6.*

This muscle also demonstrated two possible peaks.

This first peak M61 includes the only peak when a single peak is present and the first of two peaks when two are present. M61 therefore has a full subject number in both groups (18 normal, 10 amputee).

The second medial hamstring peak incidence (M62) is shown in Tables 26 and 27.

As in the other two-peak muscles, the patterns are complex.

In comparing both groups 83% of the normal and 100% of the amputees showed at least one peak.

During ascending 78% of the normal and 100% of the amputees show at least one peak, and during descending 72% of the normal and 90% of the amputees showed at least one peak.

Table 29 shows the case numbers in the comparison of the M62 peaks.

TABLE 23 - INCIDENCE OF LATERAL HAMSTRINGS SECOND PEAK (M52) IN NORMAL GROUP.

(The number of subjects is in parentheses)

CONSIDERING BOTH CYCLES

NO PEAK
6%(1)

PEAK
94%(17)

ASCENDING RIGHT	ASCENDING LEFT	DESCENDING RIGHT	DESCENDING LEFT	
X	X	X	X	44%(8)
X	X	X		11%(2)
X	X		X	6%(1)
X		X	X	6%(1)
X	X			11%(2)
		X	X	6%(1)
	X	X		6%(1)
	X			6%(1)
78%(14)	83%(15)	72%(13)	61%(11)	

CONSIDERING ASCENDING

NO PEAK
11%(2)

PEAK
89%(16)

UNILATERAL
17%(3)

BILATERAL
72%(13)

CONSIDERING DESCENDING

NO PEAK
22%(4)

PEAK
78%(14)

UNILATERAL
22%(4)

BILATERAL
56%(10)

TABLE 24 - INCIDENCE OF LATERAL HAMSTRING SECOND PEAK (M52) IN AMPUTEE GROUP.

(The number of subjects is in parentheses)

CONSIDERING BOTH CYCLES

NO PEAK
0% (0)

PEAK
100% (10)

ASCENDING AMPUTATED	ASCENDING NON-AMPUTATED	DESCENDING AMPUTATED	DESCENDING NON-AMPUTATED	
X	X	X	X	10% (1)
X	X		X	30% (3)
	X	X	X	10% (1)
X	X			10% (1)
X			X	10% (1)
	X		X	10% (1)
			X	10% (1)
	X			10% (1)
60% (4R + 2L)	80% (4R + 4L)	20% (1R + 1L)	80% (4R + 4L)	

CONSIDERING ASCENDING

NO PEAK
10% (1)

PEAK
90% (9)

UNILATERAL
40% (4)

BILATERAL
50% (5)

CONSIDERING DESCENDING

NO PEAK
20% (2)

PEAK
80% (8)

UNILATERAL
60% (6)

BILATERAL
20% (2)

TABLE 25. CASE NUMBER IN LATERAL HAMSTRINGS SECOND PEAK (M52)

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND			13 Cases
DESCEND			10 Cases
ASCEND vs. DESCEND	11 Cases	9 Cases	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND			5 Cases
DESCEND			2 Cases
ASCEND vs. DESCEND	1 Case (NSD)	6 Cases	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	14 N 4 A	15 N 2 A	14 N 4 A	15 N 4 A
DESCEND	13 N 1 A (NSD)	11 N 1 A (NSD)	13 N 4 A	11 N 4 A

NSD = Not Sufficient Data to make a comparison.

TABLE 26 - INCIDENCE OF MEDIAL HAMSTRINGS SECOND PEAK (M62) IN NORMAL GROUP.

(The number of subjects is in parentheses)

CONSIDERING BOTH PEAKS

NO PEAK
17% (3)

PEAK
83% (15)

ASCENDING RIGHT	ASCENDING LEFT	DESCENDING RIGHT	DESCENDING LEFT	
X	X	X	X	33% (6)
X	X		X	11% (2)
X	X	X		6% (1)
X	X			11% (2)
X		X		11% (2)
	X	X		6% (1)
		X	X	6% (1)
72% (13)	67% (12)	61% (11)	50% (9)	

CONSIDERING ASCENDING

NO PEAK
22% (4)

PEAK
78% (14)

UNILATERAL	BILATERAL
17% (3)	61% (11)

CONSIDERING DESCENDING

NO PEAK
28% (5)

PEAK
72% (13)

UNILATERAL	BILATERAL
33% (6)	39% (7)

TABLE 27 - INCIDENCE OF MEDIAL HAMSTRINGS SECOND PEAK (M62) IN AMPUTEE GROUP.

(The number of subjects is in parenthesis)

CONSIDERING BOTH CYCLES

NO PEAK
0% (0)

PEAK
100% (10)

ASCENDING AMPUTATED	ASCENDING NON-AMPUTATED	DESCENDING AMPUTATED	DESCENDING NON-AMPUTATED	
X	X	X	X	10% (1)
X	X		X	40% (4)
	X	X	X	10% (1)
	X		X	20% (2)
X			X	10% (1)
X	X			10% (1)
70% (4R + 3L)	90% (5R + 4L)	20% (1R + 1L)	90% (5R + 4L)	

CONSIDERING ASCENDING

NO PEAK
0% (0)

PEAK
100% (10)

UNILATERAL
40% (4)

BILATERAL
60% (6)

CONSIDERING DESCENDING

NO PEAK
10% (1)

PEAK
90% (9)

UNILATERAL
70% (7)

BILATERAL
20% (2)

TABLE 28. CASE NUMBER IN MEDIAL HAMSTRINGS SECOND PEAK (M62)

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND			11 Cases
DESCEND			7 Cases
ASCEND vs. DESCEND	9 Cases	8 Cases	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND			6 Cases
DESCEND			2 Cases
ASCEND vs. DESCEND	1 Case (NDS)	8 Cases	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	13 N 4 A	12 N 3 A	13 N 5 A	12 N 4 A
DESCEND	11 N 1 A (NSD)	9 N 1 A (NSD)	11 N 5 A	9 N 4 A

NSD = Not Sufficient Data to make a comparison.

N = Normal.

A = Amputee.

TABLE 29. POSITION OF VASTUS MEDIALIS OBLIQUE PEAK (M1) IN THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	46 ± 2	44 ± 2	<.01
DESCEND	84 ± 3	84 ± 2	NS
ASCEND vs. DESCEND	<.001	<.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	55 ± 5	43 ± 5	.001
DESCEND	87 ± 11	85 ± 6	NS
ASCEND vs. DESCEND	<.001	<.001	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.005	.005*	NS*	NS*
DESCEND	NS*	NS*	NS*	NS

*Dispersion Significance <.05. Amputee > Normal.

POSITION OF VASTUS MEDIALIS OBLIQUE PEAK (M1) IN THE CYCLE (TABLE 29)

Unit - Percentage point in the cycle.

Normal

During ascending the right limb peaked later than the left. However, there was no difference between limbs during descending. In comparing ascending to descending cycles the peaks occurred in different parts of the cycle.

Amputee

The amputee followed a similar pattern to the normal in limb difference and cycle difference. During ascending the amputated limb peaked later than the non-amputated limb. However, there was no difference between limbs during descending. In comparing ascending to descending cycles the peaks occurred in different parts of the cycle.

Normal versus Amputee

As there were right and left leg differences in the normal group, the analysis had to be divided into eight instead of into four comparisons. However as can be seen during ascending the amputated limb peaks later than the normal limb. No difference was found in any of the other group comparisons. The asterisk indicates a significant greater dispersion of the data for the amputees.

Summary

During ascending there were limb differences in both groups. In comparing groups, during ascending, in the amputated limb vastus medialis oblique peaks later in the cycle than in the normal limb.

TABLE 30. POSITION OF VASTUS MEDIALIS LONGUS PEAK (M2) IN THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	50 ± 3	49 ± 3	<.05
DESCEND	87 ± 2	88 ± 3	NS
ASCEND vs. DESCEND	<.001	<.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	58 ± 5	46 ± 5	<.001
DESCEND	88 ± 11	87 ± 5	NS
ASCEND vs. DESCEND	<.001	<.001	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	.001	<.001	NS*	<.05
DESCEND	NS*	NS*	NS*	NS

*Dispersion Significance <.05. Amputee > Normal.

POSITION OF VASTUS MEDIALIS LONGUS PEAK (M2) IN THE CYCLE (TABLE 30).

Unit - Percentage point in the cycle.

Normal

The result of the data comparisons, with respect to differences detected are the same as vastus medialis oblique (M1).

Amputee

The results of the data comparisons, with respect to differences detected are the same as vastus medialis oblique (M1).

Normal versus Amputee

The results of the data comparisons, with respect to differences detected are the same as vastus medialis oblique (M1). In addition the ascending left non-amputated limb peaked earlier than the left normal limbs.

Summary

During ascending there were limb differences in both groups. In comparing groups during ascending, in the amputated limb vastus medialis longus peaked later in the cycle than in the normal limb. Also in ascending the left non-amputated limb peaked earlier than the left normal limb.

TABLE 31. POSITION OF RECTUS FEMORIS USUAL PEAK (M32) IN THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	52 ± 6	49 ± 5	<.05
DESCEND	83 ± 13	86 ± 9	NS
ASCEND vs. DESCEND	<.001	<.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	60 ± 8	50 ± 15	NS
DESCEND	84 ± 9	81 ± 10	NS
ASCEND vs. DESCEND	<.001	<.001	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.05	.001	<.05	NS*
DESCEND	NS	NS	NS	NS

*Dispersion Significance <.05. Amputee > Normal.

POSITION OF RECTUS FEMORIS USUAL PEAK (M32) IN THE CYCLE (TABLE 31)

Unit - Percentage point in the cycle.

This peak is present in all cases, therefore designated "usual" (see Introduction).

Normal

The results of the data comparisons, with respect to differences detected, are the same as vastus medialis oblique (M1).

Amputee

There was no difference between limbs during ascending or descending. However, comparing ascending to descending, the peak occurred at different times in the cycle.

Normal versus Amputee

The differences are as for vastus medialis oblique (M1). In addition, the ascending right non-amputated limb peaked earlier than the normal. However there was only one significantly greater distribution in the amputee group.

Summary

During ascending there was a limb difference in the normal group. In comparing groups during ascending the amputated limb peaked later in the cycle than the normal. Also in ascending the right non-amputated limb peaked earlier than the normal.

TABLE 32. POSITION OF RECTUS FEMORIS EARLY PEAK (M31) IN CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	9	13 (10 ± 2)	NSD(1)
DESCEND	46 ± 16	40 ± 17 (30 ± 19)	NS(11)
ASCEND vs. DESCEND	NSD(1)	NS(3)	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	15 ± 9 (20 ± 12)	28 ± 17 (31 ± 24)	NS(3)
DESCEND	40 ± 24 (34 ± 21)	38 ± 11 (42 ± 12)	NS(4)
ASCEND vs. DESCEND	NS(4)	NS(2)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	<.05	NSD	NS*
DESCEND	NS	NS	NS	NS

*Dispersion Significance <.05. Amputee > Normal.

POSITION OF RECTUS FEMORIS EARLY PEAK (M31) IN CYCLE (TABLE 32)

Unit - Percentage point in the cycle.

Normal and Amputee

There was no difference between limbs in ascending and descending, and no difference between ascending and descending cycles in both group comparisons.

In ascending the function of this early peak would appear to be that of hip flexion as it occurs early in the swing phase. This happens bilaterally in only 6% of normal subjects (one subject) and bilaterally in 30% of amputees (three subjects).

In descending it would appear that this early peak is an eccentric action in controlling hip extension as the foot reaches for the step below. This occurred in 61% of the normals (11 subjects) and 40% of the amputees (four subjects).

For a more exact distribution refer to Tables 20 and 21.

Normal versus Amputee

During ascending the left amputated limb peaked later than the normal. However since there were only three normals and two amputees in this comparison, it is difficult to make a definitive statement. All other comparisons were not significant and one comparison did not have sufficient data.

POSITION OF RECTUS FEMORIS LATE PEAK (M33) IN CYCLE

Unit - Percentage of cycle.

No table has been presented as 17% (three subjects) ascending (left limb) were at 96% ± 3% of the cycle and 30% (three subjects) ascending non-amputees were 91% ± 1% of the cycle. No comparisons are made because the peaks occur in one limb only and only in one cycle. It would appear that this late peak which coincides with knee flexion is functional for hip flexion in these cases.

In summary, of the other rectus femoris peaks (M31 and M33) it appears that this muscle is not of primary importance in flexing the hip during ascending stairs and it more often acts as a controller of the extending hip during descending stairs in an eccentric capacity.

TABLE 33. POSITION OF VASTUS LATERALIS (M4) PEAK IN THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	50 ± 3	50 ± 3	NS
DESCEND	87 ± 4	87 ± 3	NS
ASCEND vs. DESCEND	<.001	<.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	57 ± 5	46 ± 6	.005
DESCEND	89 ± 8	83 ± 10	NS
ASCEND vs. DESCEND	<.001	<.001	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.005*		<.05*	
DESCEND	NS*		NS*	

*Dispersion Significance <.05. Amputee > Normal.

POSITION OF VASTUS LATERALIS (M4) PEAK IN THE CYCLE (TABLE 33)

Unit - Percentage point in the cycle.

Normal

There was no difference between limbs during ascending or descending. However comparing ascending to descending, the peak occurred at different parts of the cycle.

Amputee

The results of the data comparisons, with respect to differences, are the same as vastus medialis oblique (M1).

Amputee versus Normal

The differences are as for vastus medialis oblique (M1). In addition, during ascending the non-amputated peak occurred earlier than the normal peak.

Summary

During ascending limb differences were found in the amputee group. In comparing groups during ascending the amputated limb peaked later than the normal, and the non-amputated limb peaked earlier.

POSITION OF QUADRICEPS PEAKS IN THE CYCLE--DISCUSSION (EXCLUDING M33)

All the peak positions for the two groups are summarized in Figures 52 and 53.

As can be seen from the groups (Fig. 54) during ascending all the usual peaks occur in early stance and during descending all the peaks occur in late stance. This would tend to substantiate the point that during ascending maximum contraction occurs early in the weight-bearing to elevate the body in a concentric action of the muscles. However, during descending the maximum contraction occurs mainly in a breaking effect to control knee flexion. Although this is evident, the precise positions of the peaks were not appreciated and it is surprising that they are as polarized within stance as they appear to be. Not only is the clear polarization evident but also the surprising localization of the peak position. In vastus medialis oblique the peak extends only over 4% of the cycle and has an ascending right and left limb difference with only a 1% variation in cycle mean.

Certain patterns emerge on observing the data:

Only during ascending were there <u>within</u> group differences; there was no peak difference in any of the descending cycles.

In all the above ascending normal within group differences (M1, M2 and M32), the left leg always peaked first although the mean difference was no more than 3%.

When differences existed within the amputee group (M1, M2 and M4) during ascending, the non-amputated limb always peaked first.

Only during ascending were there differences <u>between</u> the groups.

Between group differences were constant in the ascending cycle where the amputated limb always peaked after the normal (M1, M2, M32, M4).

Where the non-ampuated limb peak was different from the normal M2 left, M32 right, M4) the non-amputated limb always peaked before the normal.

It is difficult to give an explanation of the above but the high degree of consistency should lend itself to an explanation.

Limb dominance may account for the normal difference between limbs. It appears evident only during ascending possibly because of the greater work involved in ascending. When less work is done the dominance of the limb may not be evident because the load is more equally distributed between the limbs.

During ascending the amputated limb peaks may occur later because the less stable of the two limbs needs a longer weight-bearing period before performing the peak contraction. Likewise, the non-amputated limb peaks earlier perhaps to relieve weight from the opposite leg and prosthesis on the step below (Fig. 54).

It is not possible to compare the EMG study in this thesis to others as this work has not been done before.

However, there are some striking similarities with phasic activity studies.

Joseph and Watson (1967) using a single electrode spanning vastus medialis and vastus lateralis demonstrated during ascending a single phase of activity in early stance which is appropriate to the single peak found in this study, and during descending a biphasic pattern described at the end of swing and again at the start of stance is similar to that of rectus femoris early descending peak (Fig. 52).

Townsend et al. (1978, 1978A) who used rectus femoris as the representative muscle for the quadriceps, demonstrated a single phase of action from early to middle stance which is again appropriate to the single peak (M32) found in this study. During descending, half the subjects were monophasic through stance and half were biphasic at the start of stance and end of stance. This again correlates with the finding in this study (M32 and M31)(Fig. 52).

What is interesting to note is that the above papers studied phasic activity or on time, and the correlation between this activity and the peaks in our study is high, since the peaks also indicate the phasic activity to a certain extent.

Fig. 52

POSITION OF QUADRICEPS FEMORIS PEAK
NORMAL GROUP

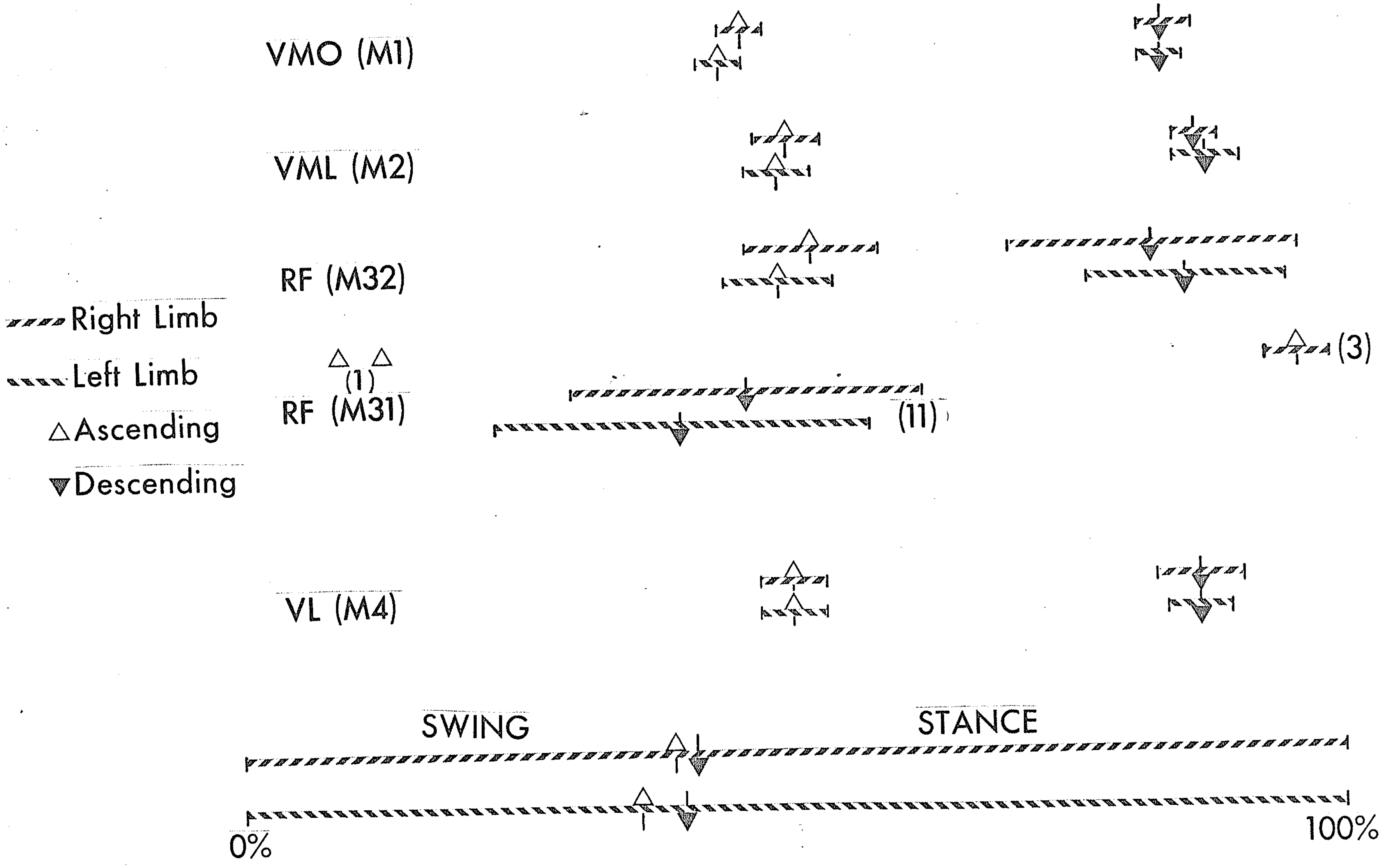


Fig. 53

POSITION OF QUADRICEPS FEMORIS PEAKS
AMPUTEE GROUP

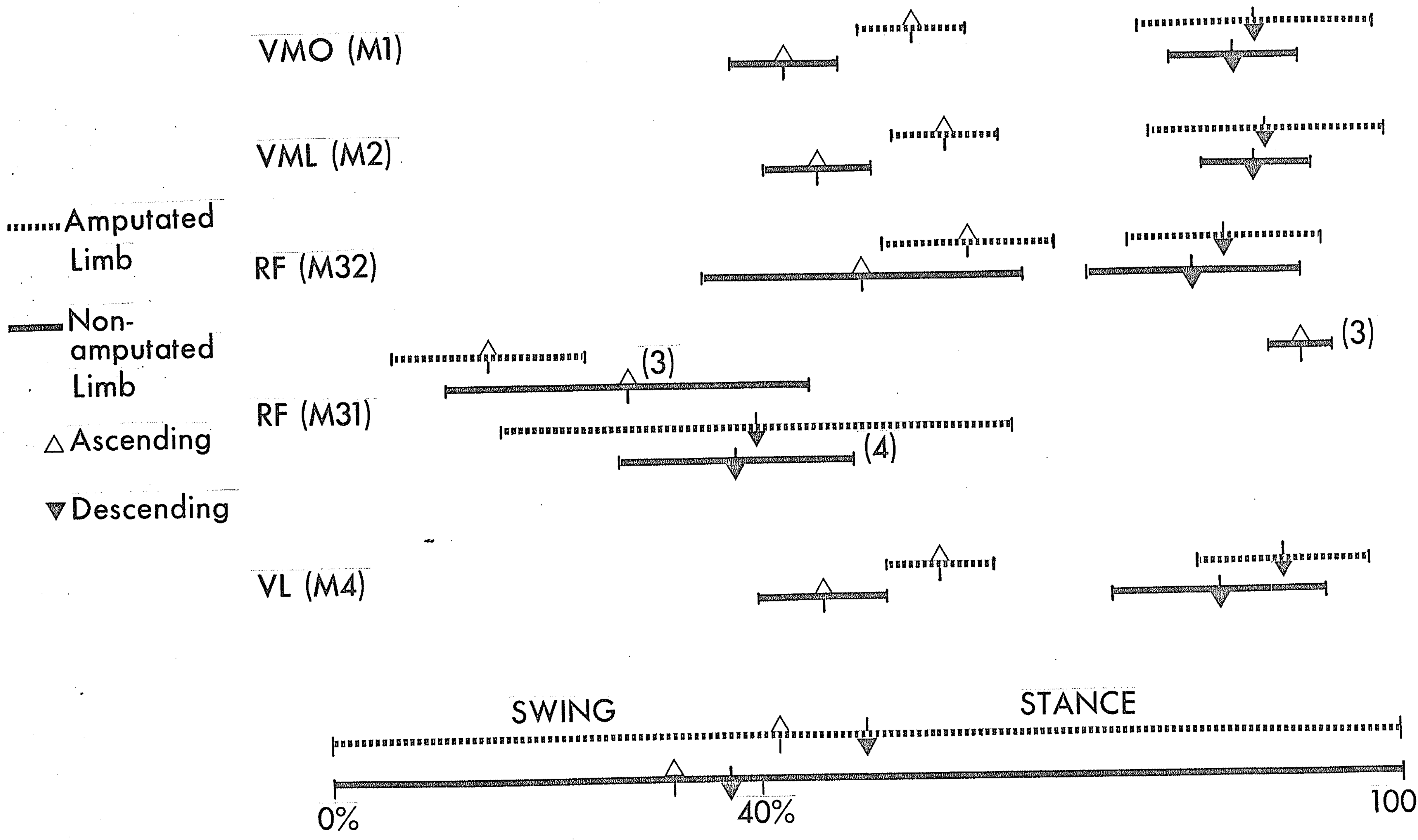


Fig. 54 SUMMARY OF MAJOR QUADRICEPS PEAKS AMPUTEES + NORMALS DURING BOTH CYCLES

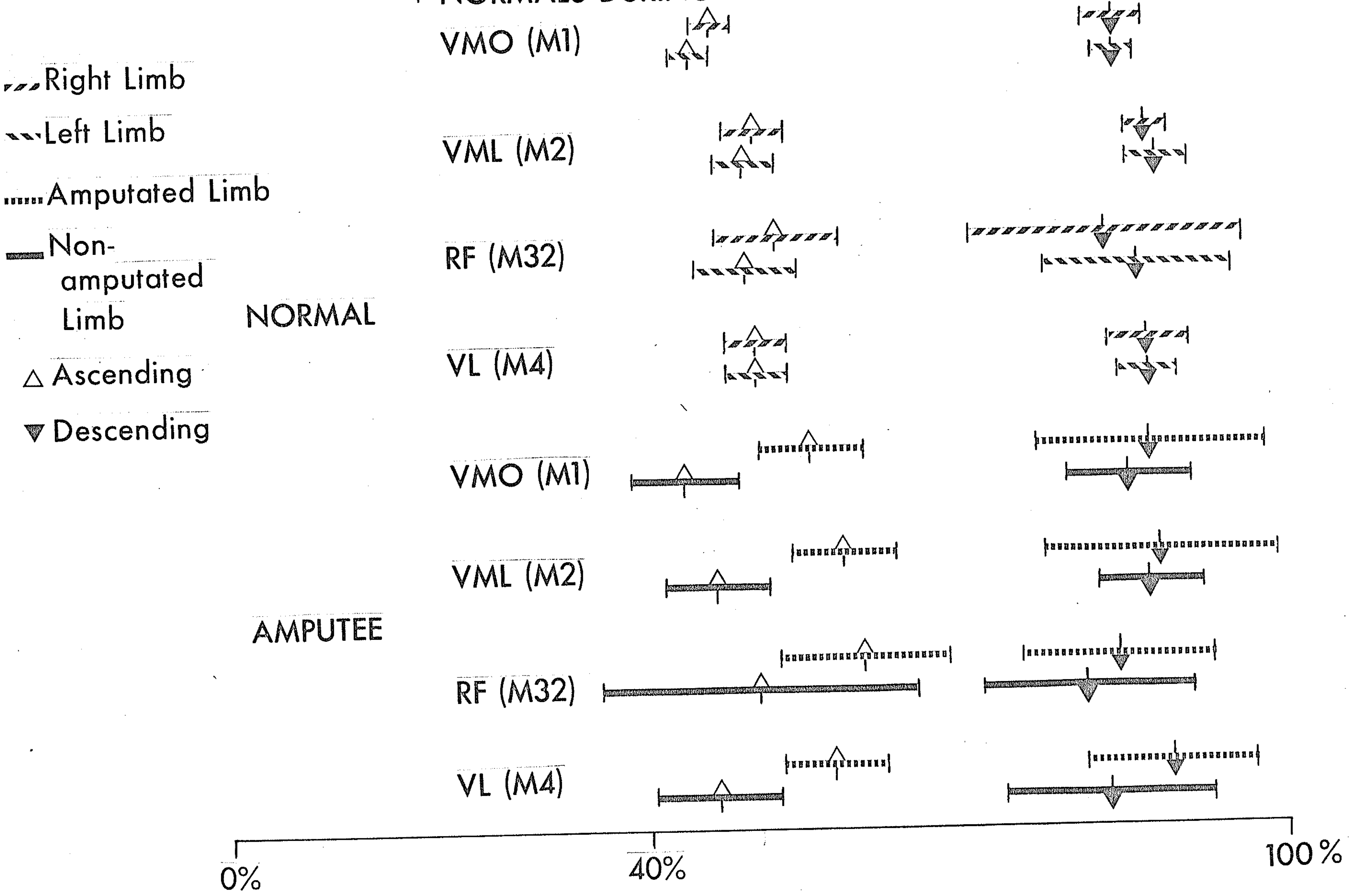


TABLE 34. POSITION OF THE LATERAL HAMSTRING FIRST PEAK (M51) IN THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	24 ± 33	33 ± 33	NS
DESCEND	44 ± 20	48 ± 28	NS
ASCEND vs. DESCEND	<.01	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	23 ± 28	21 ± 28	NS
DESCEND	68 ± 29	31 ± 28	<.05
ASCEND vs. DESCEND	<.05	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	<.05		NS	

POSITION OF THE LATERAL HAMSTRING FIRST PEAK (M51) IN THE CYCLE (TABLE 34)

Unit - Percentage point in the cycle.

Normal

There was no difference between limbs in ascending or descending. In comparing ascending to descending cycles the right limb showed an earlier ascending peak position but the left did not.

Amputee

The amputee did not follow the normal pattern. During descending the amputated limb peaked considerably later than the non-amputated limb. In comparing ascending to descending cycles the amputated limb showed a later peak in descending. However, no difference was found in the non-amputated limb.

Normal versus Amputee

During descending the amputated limb peaked later than the normal limb. There was no difference in any of the other group comparisons.

Summary

The normal group in comparing ascending to descending cycles showed an early right limb ascending peak.

The amputees showed in the descending cycle a later amputated limb peak than non-amputated limb. In comparing ascending to descending cycles, the descending amputated limb peak was later than the ascending peak.

In comparing groups, during descending the amputated limb peak was later than the normal peak.

TABLE 35. POSITION OF THE LATERAL HAMSTRING SECOND PEAK (M52) IN THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	59 ± 20 (58 ± 20)	68 ± 25 (69 ± 26)	NS(13)
DESCEND	83 ± 20 (84 ± 19	85 ± 17 (84 ± 18)	NS(10)
ASCEND vs. DESCEND	.01(11)	NS(9)	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	63 ± 10	61 ± 15 (59 ± 22)	NS(5)
DESCEND	69 ± 11	69 ± 40 (70 ± 28)	NS (2)
ASCEND vs. DESCEND	NSD(1)	NS(6)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NS	NS
DESCEND	NSD	NSD	NS	<.05

NSD = Not Sufficient Data.

POSITION OF THE LATERAL HAMSTRING SECOND PEAK (M52) IN THE CYCLE (TABLE 35)

Unit - Percentage point in the cycle.

Normal

72% of the normal group (13 subjects) showed a bilateral peak (peaks in both limbs) during ascending. 56% of the group (10 subjects) showed bilateral peaks during descending.

There was no difference between limbs during ascending or descending. In comparing ascending to descending cycles only the right limb showed an earlier ascending peak, there was no difference in the left limb.

Amputee

50% of the amputees (five subjects) showed bilateral peaks during ascending. 20% of the group (two subjects) showed bilateral peaks during descending.

There was no difference between either limb in ascending or descending. In comparing ascending to descending cycles, there was not sufficient data in the amputated limb and no difference was found in the non-amputated limb.

Normal versus Amputee

During ascending the left non-amputated limb peaked earlier than the left normal limbs. There was no difference in the other group comparisons, and the descending amputated limb comparisons could not be made for lack of sufficient data.

Summary

The normal group showed an ascending versus descending difference in the right leg.

In comparing groups during descending the left non-amputated limb peaked earlier than the normal left limb.

TABLE 36. POSITION OF THE MEDIAL HAMSTRINGS FIRST PEAK (M61) IN THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	18 ± 28	34 ± 35	NS
DESCEND	42 ± 31	46 ± 32	NS
ASCEND vs. DESCEND	.005	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	16 ± 23	17 ± 14	NS
DESCEND	69 ± 33	36 ± 27	<.05
ASCEND vs. DESCEND	<.005	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	<.05		NS	

POSITION OF THE MEDIAL HAMSTRINGS FIRST PEAK (M61) IN THE CYCLE (TABLE 36)

Unit - Percentage point in the cycle.

Normal

The results of the data comparisons with respect to difference detected are the same as the lateral hamstrings first peak (M51).

Amputee

The results of the data comparisons with respect to differences detected are the same as the lateral hamstrings first peak (M51).

Normal versus Amputee

The results of the data comparisons with respect to differences detected are the same as the lateral hamstrings first peak (M51).

Summary

The normal group, in comparing ascending to descending cycles, showed an earlier right limb ascending peak.

The amputees showed in the descending cycle an amputated limb peak later than in the non-ampuated limb. In comparing ascending to descending, the descending amputated limb peak was later than the ascending peak.

In comparing groups, during descending the amputated limb peak was later than the normal peak.

TABLE 37. POSITION OF THE MEDIAL HAMSTRING SECOND PEAK (M62) IN THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	70 ± 25 (63 ±26)	63 ± 27 (70 ± 28)	NS(11)
DESCEND	75 ± 24 (79 ± 23)	87 ±21 (82 ± 25)	NS(7)
ASCEND vs. DESCEND	NS(9)	NS(8)	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	58 ± 7	66 ± 27 (68 ± 25)	NS(6)
DESCEND	82 ± 80	68 ± 40 (76 ± 23)	NS(2)
ASCEND vs. DESCEND	NSD(1)	NS(8)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NS	NS
DESCEND	NS	NS	NS	NS

NSD = Not Sufficient Data.

POSTION OF THE MEDIAL HAMSTRING SECOND PEAK (M62) IN THE CYCLE (TABLE 37)

Unit - Percentage point in the cycle.

Normal

61% of the normal group (11 subjects) showed a bilateral peak (peaks in both limbs) during ascending. 39% of the group (seven subjects) showed bilateral peaks during descending.

There was no difference between limbs during ascending or descending. In comparing ascending to descending cycles, there were no differences.

Amputee

60% of the amputees (six subjects) showed a bilateral peak during ascending. 20% of the group (two subjects) showed bilateral peaks during descending.

The differences are as for the lateral hamstring second peak.

Normal versus Amputee

No differences were found in any of the group comparisons. However in descending normal versus amputated limbs there was not sufficient data.

Summary

There were no differences in any comparisons where there were sufficient data.

POSITION OF THE HAMSTRING PEAKS- DISCUSSION

In the normal, both the medial and lateral muscle groups showed similar patterns for first and second peaks (Fig. 55).

The first peaks of both groups have significant differences in the same comparison. One of the most striking observations is that the peak distribution is much more dispersed than the quadriceps and the descending amputated limb has a very late peak.

The second peaks have a similar pattern to one another except that the lateral hamstring second peak during descending in the left non-amputated limb peaks earlier than in the normal.

Ascending- Amputee and Normal (Figs. 55 and 56)

During ascending the first hamstring peak occurs almost exclusively during swing phase and would appear to be for knee flexion. The second peak in ascending appears almost exclusively in the stance phase, the means of which are all in the first half of stance. This would relate to the early weight-bearing phase and occurs just after the quadriceps peak. This may relate to the control of the powerful knee extension that is occurring as the body ascends the stairs.

Descending- Normal (Fig. 55)

During descending the first hamstring peak occurs in late swing and early stance, the means in the normal occurring in early stance. The possible functions are twofold. First, the eccentric control of the extending knee as the foot is reaching for the step below; and second, as the foot reaches the step the hamstrings work to stabilize the tibia to prevent it from flexing too quickly under the weight of the body. In effect the hamstrings are attempting to extend the knee which is the cause of the delayed flexion seen on initial contact. This appears to be contrary to the initial concept of the hamstrings as a knee flexor, which it undoubtly is. In this circumstance the foot is fixed and the body is moving around the foot which permits this apparent 'opposite' action.

This peak could also control hip flexion as weight is bearing on the limb, to prevent the trunk from falling forward.

A similar pattern is seen during level walking as a control mechanism (Milner et al., 1971; Battye and Joseph, 1966).

The second hamstring peak occurs at the same time as the peak quadriceps action in the descending cycle, and it would appear to also have a tibial stabilizing function.

Descending--Amputee (Fig. 56)

The descending pattern in that amputee is sufficiently different from the normal to warrant discussion. The first peak in the amputated limb is distinctly shifted more into stance phase. This would indicate that the amputated limb functions differently, in that tibial stabilization occurs much later in stance. This is most probably due to the lack of ankle dorsiflexion capability because of the prosthesis, and the unusually equal stance and swing phase discussed early in the Temporal Data section. The eccentric control in swing is not present.

The non-amputated limb peak functions more as the normal limb, with the peak mean occurring just a little earlier than the normal.

In comparing this study to others, Joseph and Watson (1967) stated that during ascending the activity was biphasic in swing and early stance. This corresponds to the peak means found in this study.

However, in descending they describe a single phase though in this study the majority demonstrated a biphasic pattern.

Townsend et al. (1978, 1978A) in their study describe the hamstrings in ascending acting as monophasic but with 'many' biphasic variants, in a similar part of the cycle to this study, i.e., end of swing to middle of stance.

In descending, the activity was described as monophasic with biphasic activity in 25%. In our study the percentage of biphasic was higher.

PEAK POSITIONS WITH RESPECT TO THE CYCLE

The peak positions in the cycle of the four individual quadricep muscles has been examined. The quadriceps will now be viewed as a group, not as individual muscles, and consideration given only to the peak positions in the cycle. The peaks are identified as first, second, third and last, and the early and late rectus femoris contractions are not included. The reason for the latter two exclusions is that it allows the extent of the usual peaks to be seen more clearly so that a greater understanding of the quadriceps muscles can be gained.

No comparison is made between the ascending and descending cycles in any of the peak positions because it has been established earlier that the peaks occur at distinctly different times (Fig. 54).

Fig. 55

POSITION OF HAMSTRING PEAKS NORMAL GROUP

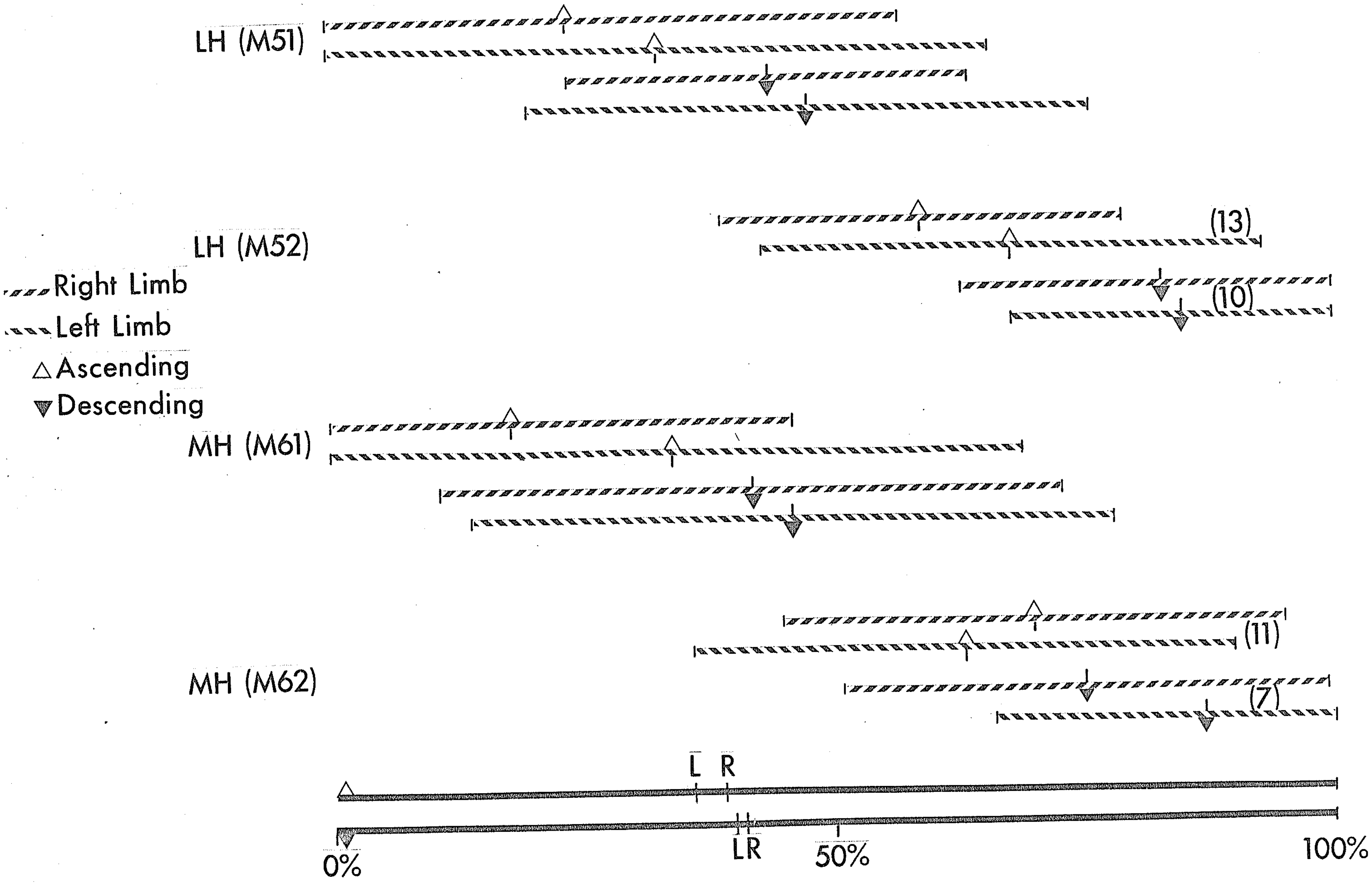


Fig. 56 POSITION OF HAMSTRING PEAKS AMPUTEE GROUP

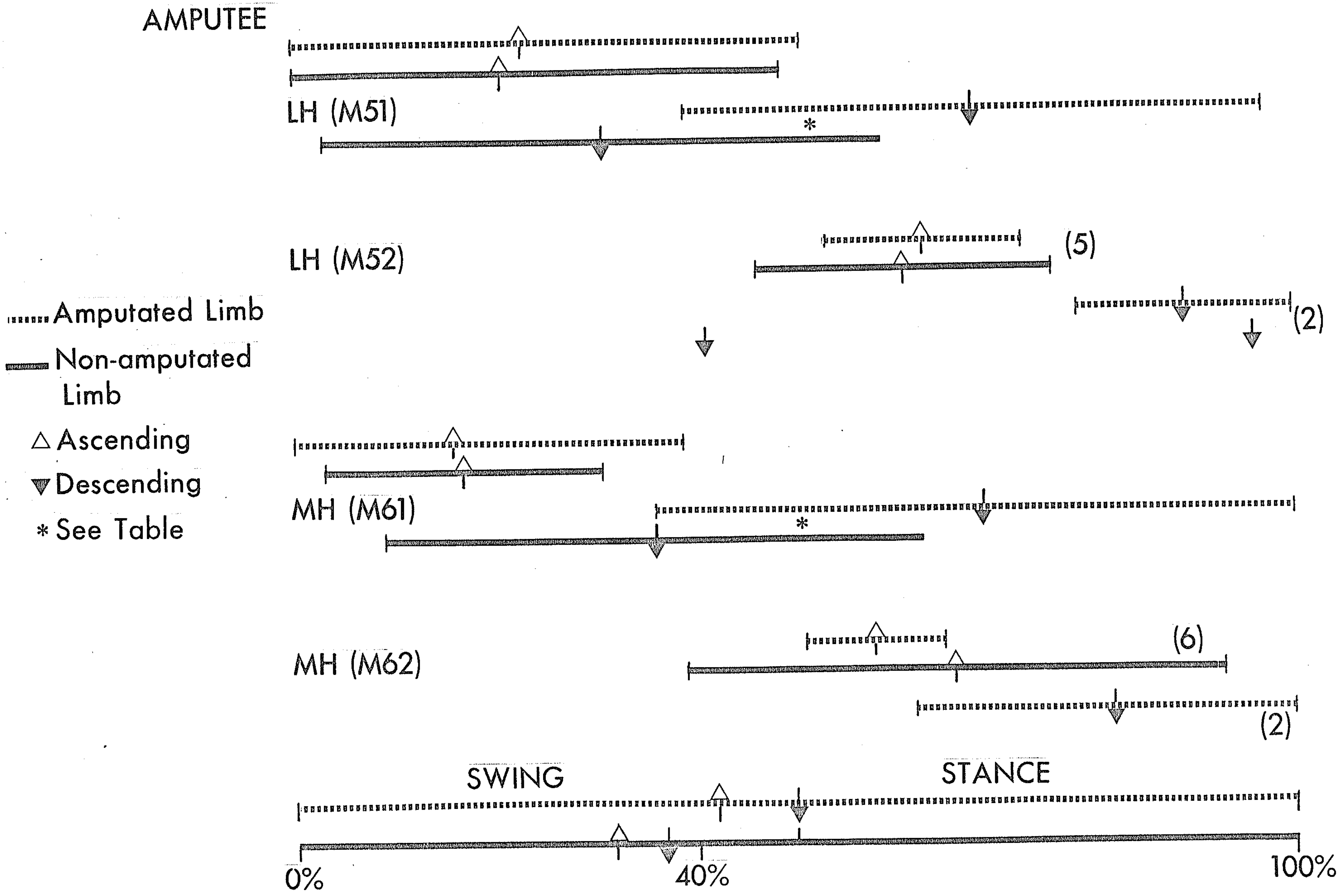


TABLE 38. FIRST QUADRICEPS PEAK WITH RESPECT TO THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	46 ± 2	44 ± 2	<.01
DESCEND	80 ± 11	82 ± 7	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	53 ± 3	42 ± 5	.001
DESCEND	81 ± 12	80 ± 10	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.001	<.001	NS*	NS
DESCEND	NS	NS	NS	NS

*Dispersion Significance <.05. Amputee > Normal.

FIRST QUADRICEPS PEAK WITH RESPECT TO THE CYCLE (TABLE 38)

Unit - Percentage point in the cycle.

Normal

During ascending the first peak occurred earlier in the right leg. There was no difference between limbs during descending.

Amputee

The amputee followed a similar pattern to the normal. During ascending the amputated limb peaked later than the non-amputated. There was no difference between limbs during descending.

Normal versus Amputee

During ascending the amputated limb peaked later than the normal. There were no differences in any of the other comparisons, but during ascending, the right non-amputated limb had a greater dispersion than the right normal limb.

Summary

During ascending in both groups, there were limb differences. In comparing groups the amputated limb first peak was later than in the normal limb.

TABLE 39. LAST QUADRICEPS PEAK WITH RESPECT TO THE CYCLE

Unit - Percentage point in the cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	54 ± 5	51 ± 4	NS (.055)
DESCEND	89 ± 3	90 ± 4	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	62 ± 6	47 ± 5	<.001
DESCEND	91 ± 6	87 ± 4	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.001		<.05	
DESCEND	NS*		NS	

*Dispersion Significance <.01. Amputee > Normal.

LAST QUADRICEPS PEAK WITH RESPECT TO THE CYCLE (TABLE 39)

Unit - Percentage point in the cycle.

Normal

No difference was found between either limb in ascending or descending. As can be seen the ascending probability was just too high.

Amputee

The amputee did not follow the normal pattern. During ascending the amputated limb peaked later than the non-amputated limb. There was no difference between limbs during descending.

Normal versus Amputee

During both ascending comparisons there were differences, the amputated limb peaking after the normal and the non-amputated limb before the normal. In descending the amputated limb had greater data dispersion than the normal limb.

Summary

During ascending the amputated limb peaked later than the non-amputated limb. In group comparison during ascending the amputated limb peaked later than the normal and the non-amputated limb peaked earlier.

TABLE 40. DURATION OF THE QUADRICEPS PEAKS WITH RESPECT TO THE CYCLE

Unit - Percentage of cycle.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	7 ± 5	8 ± 4	NS
DESCEND	9 ± 11	8 ± 9	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	9 ± 5	5 ± 2	NS (.06)
DESCEND	10 ± 10	7 ± 9	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS*	
DESCEND	NS**		NS	

*Dispersion Significance <.05. Normal > Amputee.

**Dispersion Significance <.05. Amputee > Normal.

DURATION OF THE QUADRICEPS PEAKS WITH RESPECT TO THE CYCLE (TABLE 40)

Unit - Percentage of cycle.

Having considered the differences in the position of the first and last peak, the information gained does not indicate whether the duration of the peaks expressed as a percentage of the cycle is different. For this reason the differences in cycle percent was analysed.

As can be seen from Table 40 there were no significant differences in any comparison. However it is worth noting that during ascending the non-amputated limb has the shortest duration of peaks and a significantly smaller dispersion than the normal limb. This means that the peak force from the muscle is applied over a shorter period of time.

A possible reason for this is that the amputee is more dependent on the non-amputated for greater propulsion in ascending than descending. This can also be described as a more impulsive action. An impulse is a force acting over a very short time (Duncan and Starling, 1927).

In Figure 57 the first and last peaks are shown. The two groups appear to be very similar particularly if one considers the apparent delay in ascending in the amputee group, a relationship to its longer swing phase. (see page 273)

THE TIME AND DURATION OF EMG PEAKS WITH RESPECT TO THE START OF THE CYCLE (TABLES 41 to 50)

The purpose of establishing the time and duration of certain peaks is twofold.

1) It is hoped that the clinical application of functional electrical stimulation will be made at some future date for appropriate patients in stair climbing. As the previous data regarding peaks is expressed as a percentage of or a percentage point in the cycle it cannot be easily converted into time units. However, with time units the maximum stimulation, which would appropriately coincide with the peaks found in this study, can be triggered from a foot switch unit on the same or opposite leg. This topic is discussed in the future clinical application section.

2) The verification of whether a peak sequence exists will be dependent upon the capability of the system to distinguish between peaks. As calculated earlier, should two peaks exist within .01 second of each other, the system used would not be able to detect the difference. If, however, the peaks are further apart than this then one can distinguish separate peaks. As can be seen in the following tables the peak separations are considerably longer than this and establishment of a sequence is therefore valid. The next section discusses this.

TABLE 41. TIME OF FIRST QUADRICEPS PEAK

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.78 ± 0.12	0.74 ± 0.11	NS
DESCEND	1.28 ± 0.22	1.36 ± 0.22	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	1.10 ± 0.42	0.79 ± 0.18	<.01
DESCEND	1.56 ± 0.66	1.47 ± 0.42	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.05*		NS	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

TABLE 42. TIME OF LAST QUADRICEPS PEAK

Unit - Second.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.90 ± 0.17	0.87 ± 0.14	NS
DESCEND	1.42 ± 0.18	1.45 ± 0.17	NS

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	1.27 ± 0.46	0.88 ± 0.18	<.005
DESCEND	1.76 ± 0.72	1.61 ± 0.40	NS

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.005*		NS	
DESCEND	NS*		NS*	

*Dispersion Significance <.05. Amputee > Normal.

TABLE 43. TIME FROM FIRST TO LAST QUADRICEPS PEAKS

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.13 ± 0.09	0.13 ± 0.07	NS
DESCEND	0.15 ± 0.21	0.11 ± 0.11	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	0.17 ± 0.12	0.09 ± 0.03	<.05
DESCEND	0.21 ± 0.21	0.13 ± 0.15	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*		NS**	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

**Dispersion Significance <.05. Normal > Amputee.

TABLE 44. TIME BETWEEN FIRST AND SECOND QUADRICEPS PEAKS

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.015 ± 0.04	0.06 ± 0.02	NS
DESCEND	0.10 ± 0.20	0.05 ± 0.08	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	0.07 ± 0.07	0.03 ± 0.02	NS
DESCEND	0.07 ± 0.06	0.05 ± 0.03	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*		<.005	
DESCEND	NS		NS*	

*Dispersion Significance <.05. Amputee > Normal.

TABLE 45. TIME BETWEEN SECOND AND THIRD QUADRICEPS PEAKS

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.02 ± 0.02	0.03 ± 0.03	NS
DESCEND	0.02 ± 0.03	0.02 ± 0.02	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	0.03 ± 0.02	0.04 ± 0.03	NS
DESCEND	0.07 ± 0.14	0.05 ± 0.11	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	NS*		NS*	

*Dispersion Significance <.05. Amputee > Normal.

TABLE 46. TIME BETWEEN THIRD AND FOURTH QUADRICEPS PEAKS

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.06 ± 0.07	0.04 ± 0.05	NS
DESCEND	0.02 ± 0.02	0.04 ± 0.05	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	0.08 ± 0.11	0.02 ± 0.02	NS
DESCEND	0.07 ± 0.11	0.03 ± 0.03	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*		NS**	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

**Dispersion Significance <.05. Normal > Amputee.

TABLE 47. TIME OF FIRST LATERAL HAMSTRING PEAK (M51)

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.37 ± 0.49	0.55 ± 0.55	NS
DESCEND	0.70 ± 0.31	0.76 ± 0.45	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	0.43 ± 0.49	0.38 ± 0.48	NS
DESCEND	1.35 ± 0.93	0.55 ± 0.48	<.05
ASCEND vs. DESCEND	<.05	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NS	NS
DESCEND	NS*	NS*	NS	NS

*Dispersion Significance <.05. Amputee > Normal.

TABLE 48. TIME OF SECOND LATERAL HAMSTRING PEAK (M52)

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	1.01 ± 0.35	1.16 ± 0.47	NS
DESCEND	1.35 ± 0.36	1.36 ± 0.33	NS
ASCEND vs. DESCEND	<.05	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	1.36 ± 0.74	1.24 ± 0.46	NS
DESCEND	1.56 ± 0.55	1.26 ± 0.89	NS
ASCEND vs. DESCEND	NSD	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NS	NS
DESCEND	NSD	NSD	NS*	NS

*Dispersion Significance <.05. Amputee > Normal.

NSD = Not Sufficient Data.

TABLE 49. TIME OF FIRST MEDIAL HAMSTRING PEAK (M61)

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	0.29 ± 0.42	0.58 ± 0.59	NS
DESCEND	0.68 ± 0.51	0.74 ± 0.51	NS
ASCEND vs. DESCEND	<.01	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	0.30 ± 0.43	0.33 ± 0.25	NS
DESCEND	1.38 ± 0.98	0.64 ±0.47	<.05
ASCEND vs. DESCEND	<.05	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NS	NS
DESCEND	<.05	NS*	NS	NS

*Dispersion Significance <.05. Amputee > Normal.

TABLE 50. TIME OF SECOND MEDIAL HAMSTRING PEAK (M62)

Unit - Seconds.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	1.22 ± 0.52	1.08 ± 0.54	NS
DESCEND	1.21 ± 0.45	1.36 ± 0.34	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	1.19 ± 0.52	1.28 ± 0.54	NS
DESCEND	1.48 ± 0.67	1.26 ± 0.90	NS
ASCEND vs. DESCEND	NSD	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NS	NS
DESCEND	NSD	NSD	NS*	NS

*Dispersion Significance <.05. Amputee > Normal.

NSD = Not Sufficient Data.

TIME AND DURATION OF EMG PEAK--DISCUSSION

As explained earlier in this section, the purpose of presenting temporal data is for the application of functional electrical stimulation and to quantify the systems capability to establish whether a sequence analysis should be pursued.

The sequence analysis of our data which follows has greater validity since it is now established that the system is capable of detecting all sequence differences should they exist.

Inter-peak time differences are not presented for the hamstrings because the hamstring peaks are more distinctly spaced then the quadricep peaks (Figs. 55 and 56).

EMG PEAK SEQUENCE

One of the objectives of this study was to ascertain whether or not there was a consistent sequence in the peak EMG activity of the quadriceps and hamstrings. The reason for this is that should there be a consistent sequence of contraction it would be more credible to ascribe specific function to each muscle. If there were no sequence or a random occurrence, it would be reasonable to say that there was no functional difference between the muscles. It is known that the three heads of triceps brachii work in different but recognized ways in elbow extension (Travill, 1962), however this is not the case in elbow flexion where there is rarely unanimity of action (Basmajian and Latif, 1957). In this study a sequence analysis was made to see if the knee extensors had a consistent EMG pattern, e.g., like the elbow extensors, or not, e.g., like the elbow flexors. The same analysis was done for the hamstrings.

Fig. 57

POSITION OF FIRST + LAST PEAK OF QUADRICEPS FEMORIS

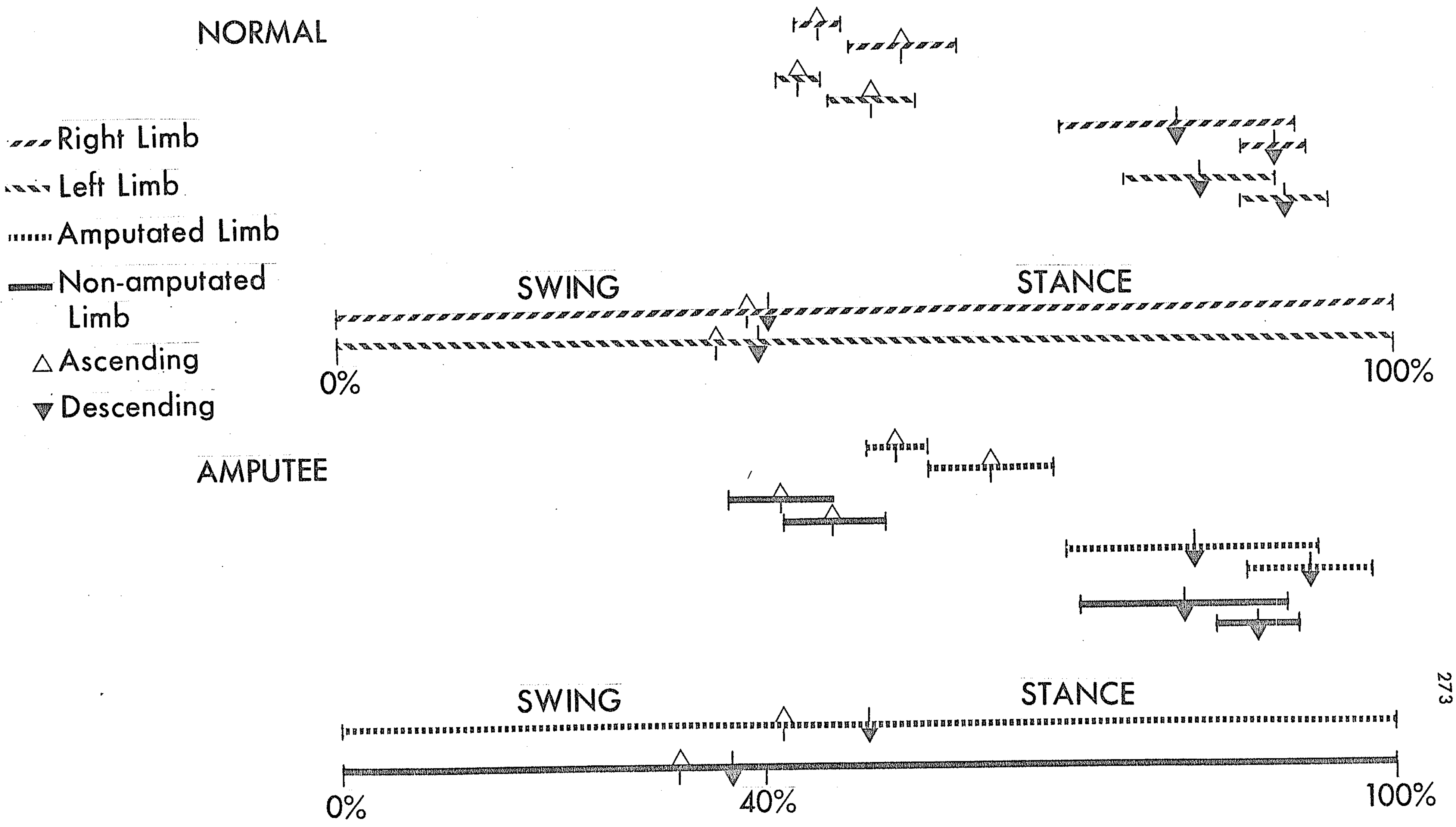


Fig. 58 PEAK TIMES OF QUADRICEPS FEMORIS FROM START OF CYCLE
NORMAL GROUP

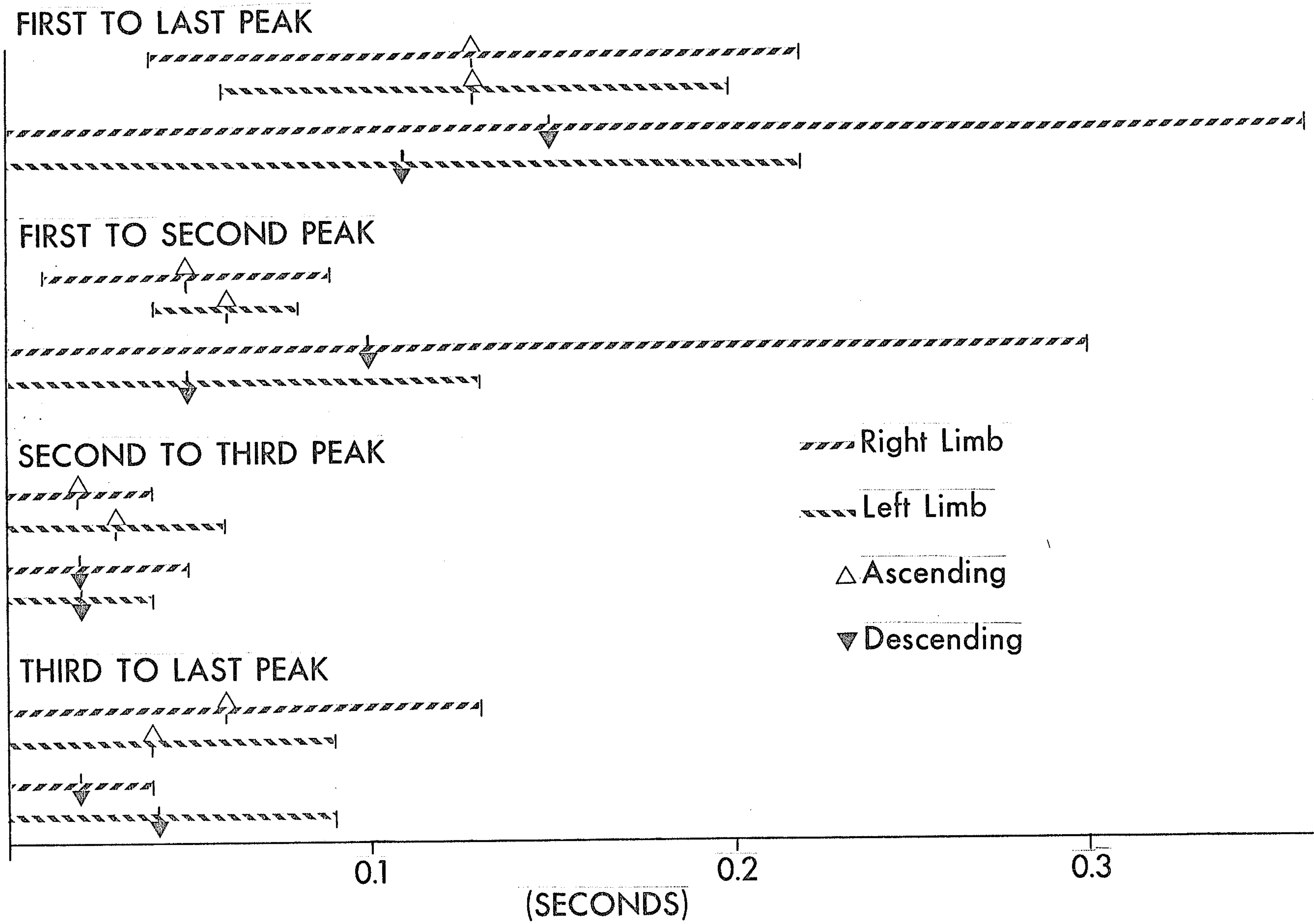
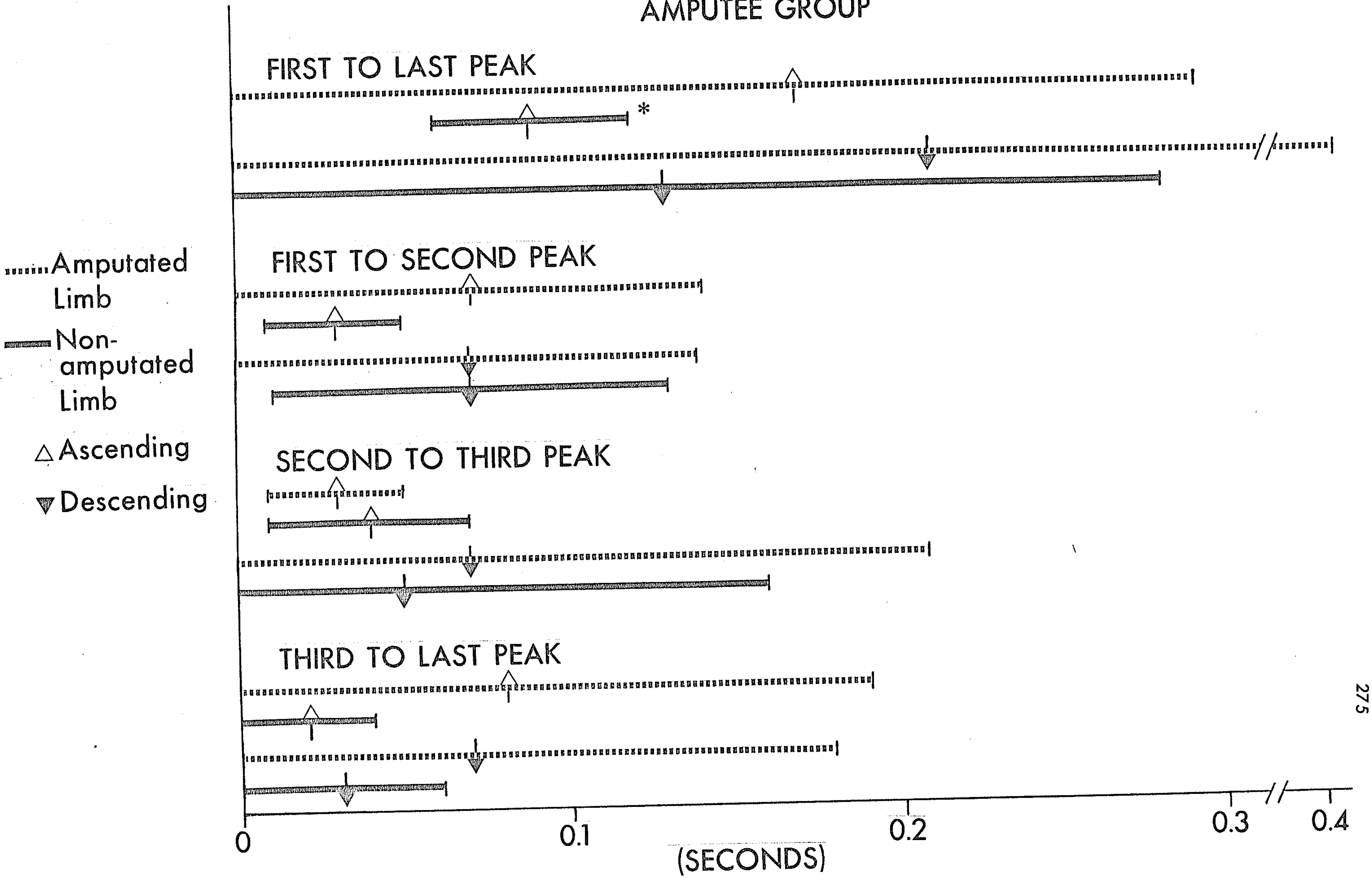


Fig. 59

PEAK TIMES OF QUADRICEPS FEMORIS FROM START OF CYCLE AMPUTEE GROUP



The method in which this was ascertained was by a statistical analysis called Kendall coefficient of concordance in which the hypothesis states that when there is no concordance, a low probability would reject this hypothesis.

To establish the sequence for each subject the peak distance from the start of the cycle to the appropriate muscle peak was measured in each of the three runs for both ascending and descending cycles. The measurements for each muscle in each cycle were added and the sequence for that subject established.

When considering the quadriceps sequence only, M32, the usual rectus femoris peak was used.

From the previous data presented one can see that each of the quadriceps and hamstring peaks occurs at separate and measurably different times. The following analysis now determines concordance or the establishment of a sequence.

TABLE 51. QUADRICEPS PEAK SEQUENCE IN NORMALS USING KENDALL'S COEFFICIENT OF CONCORDANCE

ASCEND RIGHT

MEAN RANK	VMO	VML	RF	VL
	1.056	2.889	3.278	2.778

CONCORDANCE SIGNIFICANCE <.00001 (1423)

SEQUENCE - VMO - VL - VML - RF

ASCEND LEFT

MEAN RANK	VMO	VML	RF	VL
	1.056	3.000	2.722	3.222

CONCORDANCE SIGNIFICANCE <.00001 (1324)

SEQUENCE - VMO - RF - VML - VL

DESCEND RIGHT

MEAN RANK	VMO	VML	RF	VL
	1.333	3.278	2.722	2.667

CONCORDANCE SIGNIFICANCE .0001 (1432)

SEQUENCE - VMO - VL - RF - VML

DESCEND LEFT

MEAN RANK	VMO	VML	RF	VL
	1.444	3.500	2.444	2.611

CONCORDANCE SIGNIFICANCE <.00001 (1342)

SEQUENCE - VMO - RF - VL - VML

QUADRICEPS PEAK SEQUENCE

Normal Group (Table 51)

Ascending. There was a highly significant concordance for the following sequence.

RIGHT LIMB - VMO - VL - VML - RF

LEFT LIMB - VMO - RF - VML - VL

The muscles that are consistently in the same sequence are the two parts of the vastus medialis, being first and third in sequence.

Descending. There was a highly significant concordance for the following sequence.

RIGHT LIMB - VMO - VL - RF - VML

LEFT LIMB - VMO - RF - VL - VML

The muscles that are consistently in sequence are the two parts of the vastus medialis, being first and last in sequence.

Normal Group Summary

Vastus medialis oblique was the only muscle to maintain a constant first position in all sequences, being the first in both limbs and both cycles.

Vastus medialis longus demonstrated the same position for both limbs in ascending and descending cycles. However in ascending it is the third peak for both limbs and in descending it is the fourth peak for both limbs.

Rectus femoris demonstrated in ascending a last and second peak and in descending a third and second peak. There appears to be a limb and not cycle consistency as it is the second peak in the left limb for both cycles.

Vastus lateralis demonstrated in ascending a second and fourth peak and in descending a second and third peak. There appears to be a limb and not cycle consistency as it is the second peak in the right limb for both cycles.

TABLE 52 - QUADRICEPS PEAK SEQUENCE IN AMPUTEES USING KENDALL'S COEFFICIENT OF CONCORDANCE

ASCEND AMPUTATED

MEAN RANK	VMO	VML	RF	VL
	1.300	3.400	3.000	2.300

CONCORDANCE SIGNIFICANCE .002 (1432)

SEQUENCE - VMO - VL - RF - VML

ASCEND NON-AMPUTATED

MEAN RANK	VMO	VML	RF	VL
	1.500	3.100	2.600	2.800

CONCORDANCE SIGNIFICANCE .03 (1342)

SEQUENCE - VMO - RF - VL - VML

DESCEND AMPUTATED

MEAN RANK	VMO	VML	RF	VL
	2.000	3.200	2.000	2.800

CONCORDANCE SIGNIFICANCE .09 $(\substack{1\\3}42)$

NO SIGNIFICANCE SEQUENCE

DESCEND NON-AMPUTATED

MEAN RANK	VMO	VML	RF	VL
	2.700	3.450	1.700	2.150

CONCORDANCE SIGNIFICANCE .02 (3412)

SEQUENCE - RF - VL - VMO - VML

Amputee Group (Table 52)

Ascending. There was a significant concordance for the following sequence.

AMPUTATED LIMB - VMO - VL - RF - VML

NON-AMPUTATED LIMB - VMO - RF - VL - VML

The muscles that are consistently in the same sequence are the two parts of vastus medialis being first and last in sequence.

Descending. As the concordance significance was .09 in the amputated limb no significant sequence should be attributed to the limb.

AMPUTATED LIMB - No sequence because concordance is not significant

NON-AMPUTATED LIMB - RF - VL - VMO - VML

Amputee Group Summary

Vastus medialis oblique maintained its first position in ascending but not in the single descending cycle available for comparison.

Vastus medialis longus was the only muscle to maintain a constant and specific last position in all the sequence analysed.

Rectus femoris demonstrated second and fourth peaks in ascending and the only first peak in descending, which was unique as vastus medialis oblique was first in ascending.

Vastus lateralis demonstrated second and third peaks in ascending and a second peak in descending.

The descending amputated limb had no sequence which tends to indicate a lack of specific function of the muscles during this action.

TABLE 53 - QUADRICEPS PEAK SEQUENCE IN NORMAL AND AMPUTATED LIMBS, USING KENDALL'S COEFFICIENT OF CONCORDANCE

ASCEND RIGHT

MEAN RANK	VMO	VML	RF	VL
	1.040	3.000	3.260	2.700

CONCORDANCE SIGNIFICANCE <.0001 (1423)

SEQUENCE - VMO - VL - VML - RF

ASCEND LEFT

MEAN RANK	VMO	VML	RF	VL
	1.170	3.090	2.740	3.000

CONCORDANCE SIGNIFICANCE <.0001 (1324)

SEQUENCE - VMO - RF - VML - VL

DESCEND RIGHT

MEAN RANK	VMO	VML	RF	VL
	1.570	3.170	2.520	2.740

CONCORDANCE SIGNIFICANCE <.0001 (1342)

SEQUENCE - VMO - RF - VL - VML

DESCEND LEFT

MEAN RANK	VMO	VML	RF	VL
	1.480	3.520	2.390	2.610

CONCORDANCE SIGNIFICANCE <.0001 (1342)

SEQUENCE - VMO - RF - VL - VML

Normal and Amputee Group Comparisons (Tables 53 and 54)

In comparing the two groups, the following must be born in mind: Since the amputees' limbs are not designated right and left, no comparison can be made unless the amputees' limbs are so designated. This means that in the comparison 18 normal limbs and five amputated limbs are compared; and because of the number variation, the possible sequence comparison may be affected.

What is perhaps most surprising is the constancy of sequence when the amputee group is added, and the following points are worth noting:

1) Vastus medialis oblique is consistently the first peak irrespective of group, limb or cycle. This is the only muscle of the four to remain in the same position for the 12 sequences studied.

2) Vastus medialis longus is always the third peak during the ascending cycle and always the last peak, during the descending cycle. This demonstrates a distinct and different ascending and descending function.

3) The most consistent sequence in eight of the 12, is: first, vastus medialis oblique; second, rectus femoris; third, vastus lateralis; fourth, vastus medialis longus.

Having established that a distinct sequence of contraction exists, particularly for the normal group, ascribing function is more speculative.

It would be reasonable to say that the function ascribed to the vastus medialis oblique, that of patellar alignment or correction of the lateral pull of the other quadriceps would be more appropriately subserved

TABLE 54 - QUADRICEPS PEAK SEQUENCE IN NORMAL AND NON-AMPUTATED LIMBS, USING KENDALL'S COEFFICIENT OF CONCORDANCE

ASCEND RIGHT

MEAN RANK	VMO	VML	RF	VL
	1.170	3.040	3.040	2.740

CONCORDANCE SIGNIFICANCE <.0001 (14^{2}_{3})

SEQUENCE - VMO - VL - VML
RF

ASCEND LEFT

MEAN RANK	VMO	VML	RF	VL
	1.130	2.910	2.780	3.170

CONCORDANCE SIGNIFICANCE <.0001 (1324)

SEQUENCE - VMO - RF - VML - VL

DESCEND RIGHT

MEAN RANK	VMO	VML	RF	VL
	1.700	3.220	2.650	2.430

CONCORDANCE SIGNIFICANCE .001 (1342)

SEQUENCE - VMO - RF - VL - VML

DESCEND LEFT

MEAN RANK	VMO	VML	RF	VL
	1.650	3.590	2.130	2.630

CONCORDANCE SIGNIFICANCE <.0001 (1342)

SEQUENCE - VMO - RF - VL - VML

if this muscle contracted before the other quadriceps so that the patella is appropriately aligned.

This explanation is appropriate for why the peaks should occur earlier and not later. However this does not account for a similar biomechanical result if the muscle contraction occurred at the same time. This may be because there may be a more efficient correction of the considerably higher lateral torque than medial torque on the patella when it occurs earlier. The higher lateral torque is produced by the rest of the quadriceps.

The evidence in this study is a quantified substantiation of this deduction which has been ascribed to this muscle by reason of its structure.

The vastus medialis longus ascending and descending pattern, and the most consistent sequence are an enigma and at this time the author cannot give an explanation for the distinct patterns seen in the other muscles.

TABLE 55 - HAMSTRING PEAK SEQUENCE IN NORMALS, USING KENDALL'S COEFFICIENT OF CONCORDANCE

ASCEND RIGHT (12 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.583	3.500	1.417	3.500
CONCORDANCE SIGNIFICANCE	<.00001	$(6_1 5_1 6_2$ $5_2)$		
SEQUENCE	MH1	LH1	MH2 LH2	

ASCEND LEFT (11 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.636	3.818	1.364	3.182
CONCORDANCE SIGNIFICANCE	<.00001	$(6_1 5_1 6_2 5_2)$		
SEQUENCE	MH1	LH1	MH2	LH2

DESCEND RIGHT (11 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.636	3.455	1.364	3.545
CONCORDANCE SIGNIFICANCE	<.00001	$(6_1 5_1 5_2 6_2)$		
SEQUENCE	MH1	LH1	LH2	MH2

DESCEND LEFT (9 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.667	3.556	1.444	3.333
CONCORDANCE SIGNIFICANCE	.0002	$(6_1 5_1 6_2 5_2)$		
SEQUENCE	MH1	LH1	MH2	LH2

HAMSTRINGS PEAK SEQUENCE

While the quadriceps have a common insertion, the hamstrings have a common origin. This means that the distal insertions, because they are at different places can more plausibly have different function and sequence. The hamstrings have all but one head of one muscle spanning two joints and the quadriceps have all but one muscle spanning a single joint.

However, as mentioned earlier, a sequence need not necessarily exist.

It should be noted that only subjects with all four peaks could be considered in the concordance so the case numbers are not constant.

Normal Groups (Table 55)

Ascending. There was a highly significant concordance for the following sequence.

RIGHT LIMB - MH1 - LH1 - MH2
LH2

LEFT LIMB - MH1 - LH1 - MH2 - LH2

The sequence of peaks is essentially the same for both right and left except the last two peaks are tied for the right limb.

Descending. There was a highly significant concordance for the following sequence.

RIGHT LIMB - MH1 - LH1 - LH2 - MH2

LEFT LIMB - MH1 - LH1 - MH2 - LH2

The first two peaks are the same for either limb and the third and fourth are reversed.

TABLE 56 - HAMSTRING PEAK SEQUENCE IN AMPUTEES, USING KENDALL'S COEFFIENCE OF CONCORDANCE

ASCEND AMPUTATED (6 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.667	3.833	1.333	3.167

CONCORDANCE SIGNIFICANCE .002 $(6_1 5_1 6_2 5_2)$

SEQUENCE - MH1 - LH1 - MH2 - LH2

ASCEND NON-AMPUTATED (8 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.500	3.378	1.500	3.625

CONCORDANCE SIGNIFICANCE .0002 $(5_1 5_2 6_2)$
6_1

SEQUENCE - MH1 - LH2 - MH2
LH1

DESCEND AMPUTATED (2 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.500	3.500	1.500	3.500

CONCORDANCE SIGNIFICANCE .1870 $(5_1 5_2)$
$(6_1 6_2)$

NO SIGNIFICANCE SEQUENCE

DESCEND NON-AMPUTATED (8 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.250	3.125	1.750	3.875

CONCORDANCE SIGNIFICANCE .0001 $(5_1 6_1 5_2 6_2)$

SEQUENCE - LH1 - MH1 - LH2 - MH2

Normal Group Summary

Lateral hamstring first peak (M51) maintained a consistently second placed peak in both limbs and both cycles.

Lateral hamstring second peak (M52) was either tied third, third or fourth peak.

Medial hamstrings first peak (M61) maintained a consistently first place peak in both limbs and both cycles.

Medial hamstrings second peak (M62) was either tied third, third or fourth peak.

There was a regular and consistent sequence for the first peaks of the two hamstring groups.

Amputee Group (Table 56)

Ascending. There was a significant concordance for the following sequence.

AMPUTATED LIMB - MH1 - LH1 - MH2 - LH2

NON-AMPUTATED LIMB - MH1 - LH2 - MH2
LH1

The first peaks of both the hamstring groups appear first.

Descending.

AMPUTATED LIMB - No sequence because concordance is not significant

NON-AMPUTATED LIMB - LH1 - MH1 - LH2 - MH2

Amputee Group Summary

Lateral hamstrings first peak (M51) was either first, tied first or second.

TABLE 57 - HAMSTRINGS PEAK SEQUENCE IN NORMAL AND AMPUTATED LIMBS, USING KENDALL'S COEFFICIENT OF CONCORDANCE

ASCEND RIGHT (16 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.630	3.560	1.380	3.440

CONCORDANCE SIGNIFICANCE <.0001 $(6_1 5_1 6_2 5_2)$

SEQUENCE - MH1 - LH1 - MH2 - LH2

ASCEND LEFT (13 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.620	3.850	1.380	3.150

CONCORDANCE SIGNIFICANCE <.0001 $(6_1 5_1 6_2 5_2)$

SEQUENCE - MH1 - LH1 - MH2 - LH2

DESCEND RIGHT (12 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.670	3.500	1.300	3.500

CONCORDANCE SIGNIFICANCE <.0001 $(6_1 5_1 \substack{6_2 \\ 5_2})$

SEQUENCE - MH1 - LH1 - MH2
LH2

DESCEND LEFT (10 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.600	3.500	1.500	3.400

CONCORDANCE SIGNIFICANCE <.0001 $(6_1 5_1 6_2 5_2)$

SEQUENCE - MH1 - LH1 - MH2 - LH2

Lateral hamstrings second peak (M52) was third or fourth.

Medial hamstrings first peak (M61) was either first, tied first or second peak.

Medial hamstrings second peak (M62) was either third or fourth.

The first two peaks were always the first peaks of the two hamstring groups, namely the second peak never occurred before the other hamstring group first peak. However, there was no consistent or regular pattern.

Normal and Amputee Group Comparisons (Tables 57 and 58)

As in the quadriceps sequence, the amputee sequence had to be divided into right and left limbs so that the appropriate groups could be compared.

As can be seen from the tables, the dominant sequence is first, medial hamstrings first peak, second, lateral hamstring first peak. In one instance they both occur first.

The third and fourth peaks show a more complex but significant sequence. In six cases the sequence is medial hamstrings second peak, the lateral hamstring second peak, in four cases they are tied, and in two cases the sequence is reversed.

Since these peaks occurred throughout the cycle, unlike the quadriceps peaks which are much more localized to specific areas (see previous Fig. 54), it is more appropriate to consider the first and second peaks separately.

TABLE 58 - HAMSTRINGS PEAK SEQUENCE IN NORMALS AND NON-AMPUTATED LIMBS, USING KENDALL'S COEFFICIENT OF CONCORDANCE

ASCEND RIGHT (16 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.560	3.500	1.440	3.500

CONCORDANCE SIGNIFICANCE <.0001 $(6_1 5_1 6_2)$
5_2

SEQUENCE - MH1 - LH1 - MH2
LH2

ASCEND LEFT (15 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.600	3.670	1.400	3.330

CONCORDANCE SIGNIFICANCE <.0001 $(6_1 5_1 6_2 5_2)$

SEQUENCE - MH1 - LH1 - MH2 - LH2

DESCEND RIGHT (15 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.530	3.330	1.470	3.670

CONCORDANCE SIGNIFICANCE <.0001 $(6_1 5_1 5_2 6_2)$

SEQUENCE - MH1 - LH1 - LH2 - MH2

DESCEND LEFT (13 Cases)

MEAN RANK	LH1	LH2	MH1	MH2
	1.540	3.460	1.540	3.460

CONCORDANCE SIGNIFICANCE <.0001 $(6_1 6_2)$
$(5_1 5_2)$

SEQUENCE - MH1 - MH2
LH1 - LH2

First Peaks

As the knee is flexed during all these peaks a rotatory force can be applied to the joint.

A possible explanation for the medial group being first to contract could be that this initially produces a more medial rotatory torque to the tibia and in this way moves the tibial tubercle medial with respect to the femur.

This alignment allows the quadriceps a more direct or straighter pull and they also happen to contract after the hamstrings. This could be incidental but is appropriate for both the ascending and descending cycles when considering the peak sequence of both these muscles. The amplitude of the peak emg tends to be, however, higher for the first lateral hamstrings group.

It is interesting to note that neither descending amputated limb cycles have a contraction sequence. Unfortunately in the amputee group the case number is too low to make any significant conclusion but this correlates with the short stance, and sudden burst.

TABLE 59. AMPLITUDE OF VASTUS MEDIALIS OBLIQUE PEAK (M1)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	144 ± 58	146 ± 9	NS
DESCEND	78 ± 49	87 ± 77	NS
ASCEND vs. DESCEND	<.001	.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	111 ± 82	220 ± 112	.001
DESCEND	66 ± 45	114 ± 78	<.05
ASCEND vs. DESCEND	<.05	<.01	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	NS		NS	

AMPLITUDE OF EMG PEAKS

The mean values for each run were expressed as a percentage of the maximum elicited contraction in the method described earlier.

AMPLITUDE OF VASTUS MEDIALIS OBLIQUE PEAK (M1) (TABLE 59)

Unit - Percentage of maximum elicited contraction.

Normal

There was no difference between either limb during ascending or descending. However, in comparing ascending to descending cycles the peak amplitude was less in descending.

Amputee

The amputee did not follow the normal pattern. During ascending and descending there was greater amplitude in the non-amputated limb. In comparing ascending to descending cycles the peak amplitude was less in descending.

Normal versus Amputee

There was no difference in any of the group comparisons.

Summary

Vastus medialis oblique showed less peak activity in descending than ascending stairs in both groups.

The amputee demonstrated in both cycles greater peak activity in the non-ampuated limb than the amputated limb.

TABLE 60. AMPLITUDE OF VASTUS MEDIALIS LONGUS PEAK (M2)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	116 ± 46	124 ± 74	NS
DESCEND	67 ± 31	81 ± 58	NS
ASCEND vs. DESCEND	<.001	<.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	127 ± 72	210 ± 85	<.005
DESCEND	92 ± 67	124 ± 55	NS
ASCEND vs. DESCEND	.001	.001	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		<.005	
DESCEND	NS		<.05	

AMPLITUDE OF VASTUS MEDIALIS LONGUS PEAK (M2) (TABLE 60)

Unit - Percentage of maximum elicited contraction.

Normal

The results of the data comparisons, with respect to differences, is the same as vastus medialis oblique (M1).

Amputee

The amputee did not follow the normal pattern. During ascending the non-amputated limb had greater peak activity than the amputated but not during descending.

In comparing ascending to descending cycles the peak amplitude was less in descending.

Normal versus Amputee

During ascending and descending the non-amputated limb had greater peak acting in both cycles than the normal. There was no difference in the amputated and normal limb comparisons in either cycle.

Summary

Vastus medialis longus showed less peak activity in descending than in ascending stairs in both groups.

The amputee demonstrated during ascending only, a greater peak activity in the non-amputated limb than in the amputated limb.

In comparing groups the non-amputated limb had greater activity in both cycles than the normal.

TABLE 61. AMPLITUDE OF RECTUS FEMORIS USUAL PEAK (M32)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	42 ± 31	48 ± 29	NS
DESCEND	31 ± 28	39 ± 26	NS
ASCEND vs. DESCEND	<.005	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	26 ± 19	65 ± 42	<.01
DESCEND	30 ± 23	43 ± 24	NS
ASCEND vs. DESCEND	NS	<.05	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	NS		NS	

AMPLITUDE OF RECTUS FEMORIS USUAL PEAK (M32) (TABLE 61)

Unit - Percentage of maximum elicited contraction.

Rectus Femoris

As stated earlier this muscle demonstrated two peaks in some subjects. The extra peak occurred early in the cycle or in the latter part.

The usual peak is first discussed.

Normal

There was no difference between either limb in ascending or descending. In comparing ascending to descending only the right leg showed a smaller peak amplitude.

Amputee

During ascending the non-amputated limb had a greater peak activity than the amputated but not during descending. In comparing the ascending with the descending cycle only the non-amputated limb showed a smaller peak amplitude in descending.

Normal versus Amputee

There was no difference in any group comparison.

Summary

In comparing ascending with descending cycle differences, the descending was significantly less in the normal right limb and the non-amputated limb.

TABLE 62. AMPLITUDE OF THE EARLY RECTUS FEMORIS PEAK (M31)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	31	31 (21 ± 11)	NSD(1)
DESCEND	21 ± 12	30 ± 19 (23 ± 9)	NS(11)
ASCEND vs. DESCEND	NSD(1)	NS(3)	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	13 ± 13 (11 ± 11)	15 ± 15 (18 ± 20)	NS(3)
DESCEND	21 ± 14 (21 ± 14)	76 ± 73 (37 ± 12)	NS(4)
ASCEND vs. DESCEND	<.05(4)	NS(2)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*	NS	NSD	NS
DESCEND	NS	NS	NS	NS

*Dispersion Significance <.05. Amputee > Normal.

NSD = Not Sufficient Data.

AMPLITUDE OF THE EARLY RECTUS FEMORIS PEAK (M31) (TABLE 62)

The incidence of this peak can be seen in Table 20 and 21.

In the normal group where comparisons were possible there was no difference in the data. In the amputee group there was a significantly greater activity during descending than ascending which may indicate this muscle's more important role in controlling the descending amputated limb rather than flexing the ascending amputated limb, however, relative amplitude of both are quite low.

In the group comparison no significant differences were found where there was sufficient data.

AMPLITUDE OF THE LATE RECTUS FEMORIS (M33)

In the normal group the three left ascending limbs the mean amplitude is 46 ± 29%.

In the amputee group the three non-amputated limbs the amplitude is 51 ± 8%.

TABLE 63. AMPLITUDE OF VASTUS LATERALIS PEAK (M4)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	113 ± 54	108 ± 46	NS
DESCEND	59 ± 37	56 ± 40	NS
ASCEND vs. DESCEND	<.001	<.001	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	92 ± 48	204 ± 101	<.005
DESCEND	61 ± 39	122 ± 65	<.01
ASCEND vs. DESCEND	<.01	<.001	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		<.005*	
DESCEND	NS		<.05*	

*Dispersion Significance <.05. Amputee > Normal.

AMPLITUDE OF VASTUS LATERALIS PEAK (M4) (TABLE 63)

Unit - Percentage of maximum elicited contraction.

Normal

The results of the data comparisons, with respect to differences, are the same as vastus medialis oblique (M1).

Amputee

The results of the data comparisons with respect to differences, are the same as vastus medialis oblique (M1).

Normal versus Amputee

During ascending and descending the non-amputated limb had greater peak activity in both cycles than did the normal. There was no difference in the amputated and normal limb comparisons in either cycle.

Summary

Vastus lateralis showed less peak activity in descending than ascending stairs in both groups.

The amputated limb showed greater activity in both cycles than the non-amputated limb and the normal limb.

AMPLITUDE OF QUADRICEPS EMG PEAKS--DISCUSSION

Certain distinct patterns emerge in studying the data. However, before discussing the data it should be noted that means of twice the maximum elicited contraction are obtained sometimes; the actual percentage as a numerical unit is not important. The reason for this occurrence is that in some individuals the dynamic and powerful movement of ascending stairs generates a greater EMG signal than an isometric maximum resisted movement elicited manually. Another reason for this is that the mean of three maximum manual resisted contractions was used to determine the maximum elicited contraction and this would also tend not to produce the maximum possible generated EMG activity, although it produces a more uniform reference point. Finally, the signal reference point could be affected by joint position and the duration of contraction, for example, if all the subjects were asked to jump from a three-foot height on to one leg the EMG generated would be greater than a manual resisted movement over a period of time.

However, the actual percentage as a numerical unit is not important. What is important is the relative positions of the peaks, not the absolute measurement, and the maximum elicited contraction is used only as a standard or reference point.

Normal Group (Figs. 60 and 61)

1) The vasti are more active than rectus femoris in both ascending and descending stairs.

2) There is a consistent pattern and sequence of activity levels in both limbs in ascending and descending stairs. The following is the order of decreasing activity: Vastus medialis oblique

Vastus medialis longus

Vastus lateralis

Rectus femoris.

3) There was no significant difference between the peak amplitudes of either the right or left limb in both ascending or descending cycles.

4) In all but one of the eight comparisons between ascending and descending cycles the descending peak was significantly less than the ascending peak.

The above demonstrates again that the separate heads of the quadriceps appear to have separate functions because of the regular and consistent pattern demonstrated.

Rectus femoris plays a minor role during stair climbing. This is not only evident from the distinctly lower peak activity compared with the others but also because rectus femoris tended to be in a shorter position when the maximum elicited contraction was performed because all the subjects were sitting during the resisted movement. This also would tend to make the EMG signal of its maximum elicited contraction lower.

Both limbs contributed to each ascending and descending cycle peak equally as there were no significant differences between the limbs.

The ascending versus descending peak difference is due most probably to the fact that during ascending the quadriceps are working in a concentric fashion, that is, the muscles' origin and insertion are approximate; and during descending the quadriceps are working eccentrically, namely the muscles' origin and insertion are lengthening. In addition, during ascending, work was done against gravity and during descending gravity assists the action. Ascending stairs may be considered to be positive work (concentric muscle contraction) and descending stairs, negative work (eccentric muscle contraction)(Knuttgen et al., 1971, Nielsen et al., 1972).

It has been frequently observed that the energy cost in humans is much less for eccentric work than concentric work (Abbott et al., 1952; Asmussen, 1952; Kamon, 1970; Monod and Scherrer, 1973).

Thys et al. (1972) in their study of the utilization of muscle elasticity in exercise, studied deep knee bending and the quadriceps. They note that electrical activity is appreciably less in the flexion phase during the negative work performance than in the phase of extension or positive work performance.

Pocock (1968) also showed similar results in the squatting action. Our study confirms the difference and acknowledges this to occur in all components except possibly the rectus femoris. A possible reason for this could be that as rectus femoris is a two-joint muscle the position of the hip plays an important part in considering the degree of eccentric and concentric movement. For example, during descending stairs when eccentric work is occurring in the quadriceps, if the hip is moving into flexion the resultant eccentric work is not as marked.

<u>Amputee Group</u> (Figs. 60 and 61)

1) The vasti are more active than rectus femoris in both ascending and descending stairs.

2) There is no consistent pattern and sequence of activity levels in both limbs during ascending and descending stairs.

During ascending and descending the amputated limb peak activity, in decreasing order is: Vastus medialis longus

Vastus medialis oblique

Vastus lateralis

Rectus femoris

3) There are significant differences between the amputated and non-amputated limb peaks in both ascending and descending. Specifically during ascending the non-amputated limb peak amplitude means are two to two and a half times greater than descending peak amplitudes.

During ascending and descending the pattern of mean differences in the amputated limb followed the normal more closely than did the non-amputated limb. In the non-amputated limb the mean peak activity of the vasti was at a similar level in ascending and similar levels in descending considering the cycles separately.

Considering that the non-amputated limb had considerably higher peaks the dominance of the limb is again demonstrated. However, one might expect that because the non-amputated limb had greater peaks the amputated limb might also have significantly less activity than the normal limb but this did not occur.

4) Vastus medialis oblique in the amputated limb demonstrates a low peak very close to the lowest of the vasti peaks and it is tempting to extrapolate this into some decrease in function that is related to the obviously smaller bulk of this muscle in the amputee. However that is very speculative.

5) In all but one of the eight comparisons between ascending and descending cycles the descending peak was significantly less than the ascending peak. The discussion in the normal group applies here as well. The exception is rectus femoris which shows a tendency in the opposite direction, namely a greater descending peak in the amputated limb. In the early peak of rectus femoris this is significantly different for the descending amputated limb.

Normal and Amputee Group (Fig. 62)

In comparing groups only the non-amputated limb vastus medialis longus and vastus lateralis showed significantly greater activity in both ascending and descending cycles than the normal. One could speculate that this is because they are the most important of the four extensors with respect to extension function. Vastus medialis oblique exerts a correcting oblique force and because of its limited size and function does not show variation; and rectus femoris is not sufficiently highly active in stair climbing to demonstrate a significant difference.

It is again worth noting the difference in the ascending and descending cycle in both groups. Of the 16 comparisons, all but two (both involving rectus femoris) show the descending cycle to have a lesser peak emg activity.

One thing that is interesting is that as the non-amputated limb shows more peak activity than the amputated limb, one would expect that the amputated limb would show correspondingly less activity. However this is not the case. This could be due to the increased requirements needed for vertical propulsion and descending control by the appropriate limb due to the lack of plantar flexion in the amputated limb. This is compensated for by the greater activity of the non-amputated limb.

Fig. 60

AMPLITUDE OF QUADRICEPS FEMORIS PEAK-ASCENDING CYCLE

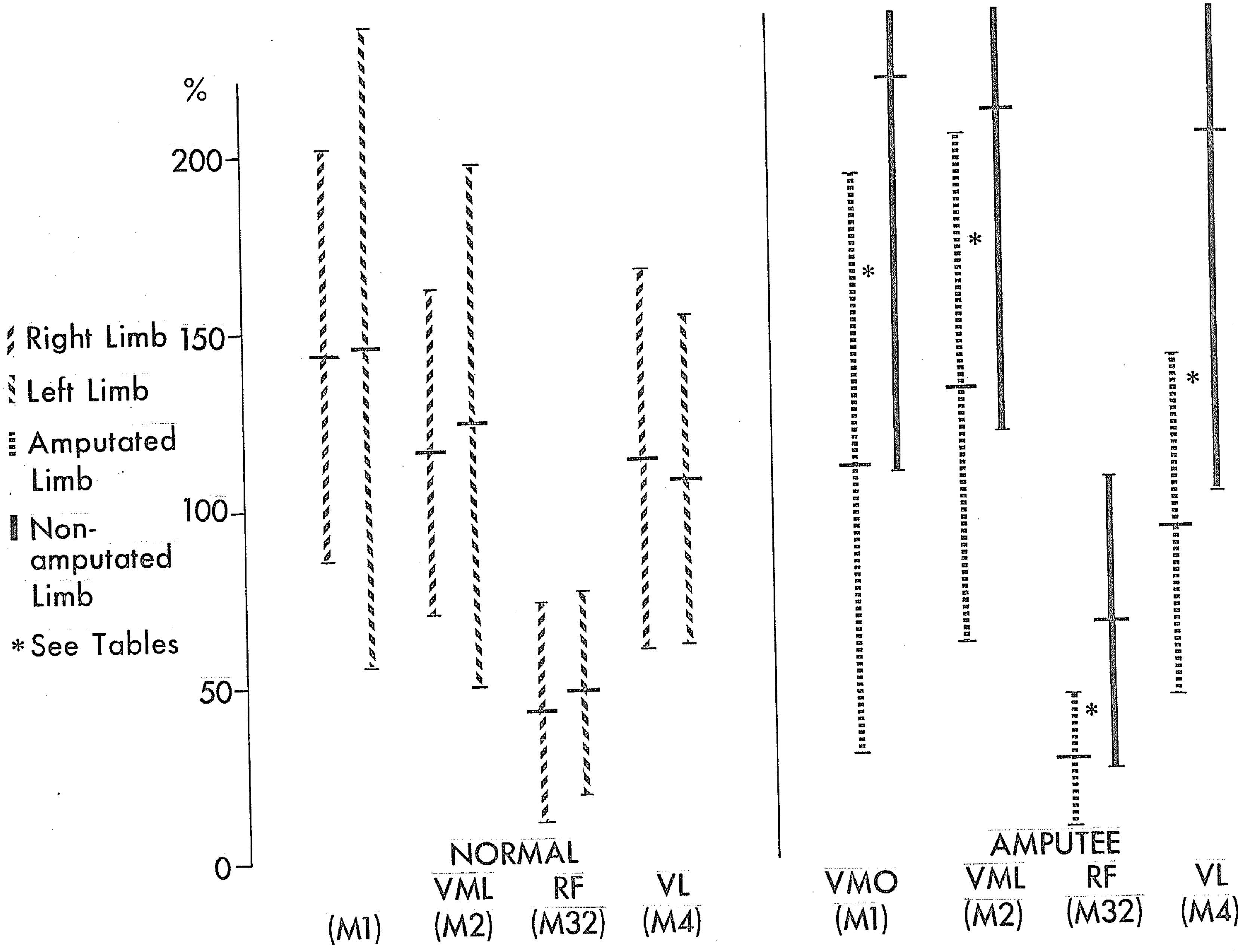


Fig. 61 AMPLITUDE OF QUADRICEPS FEMORIS PEAK-DESCENDING CYCLE

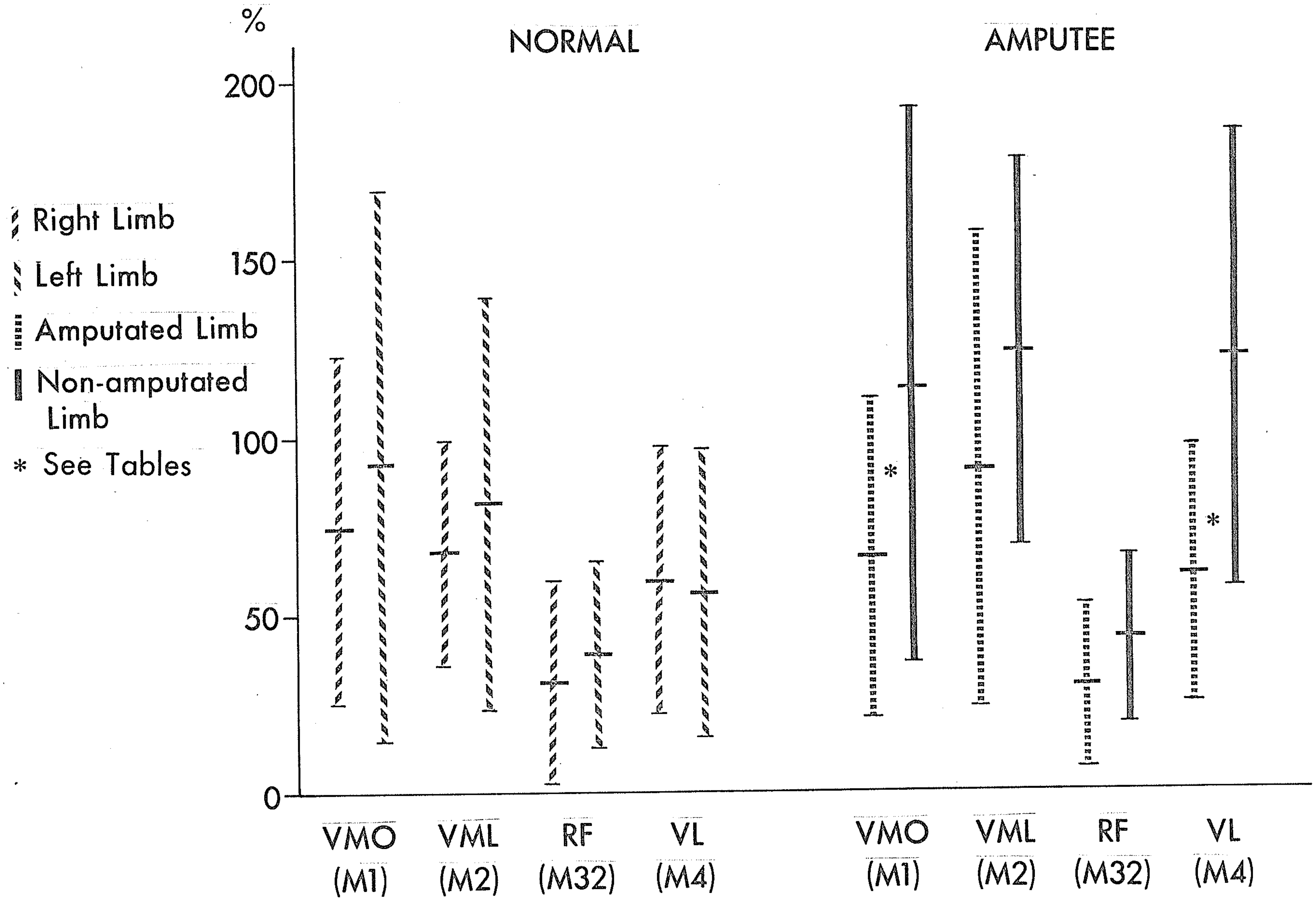


Fig. 62 AMPLITUDE OF QUADRICEPS FEMORIS PEAKS BOTH CYCLES

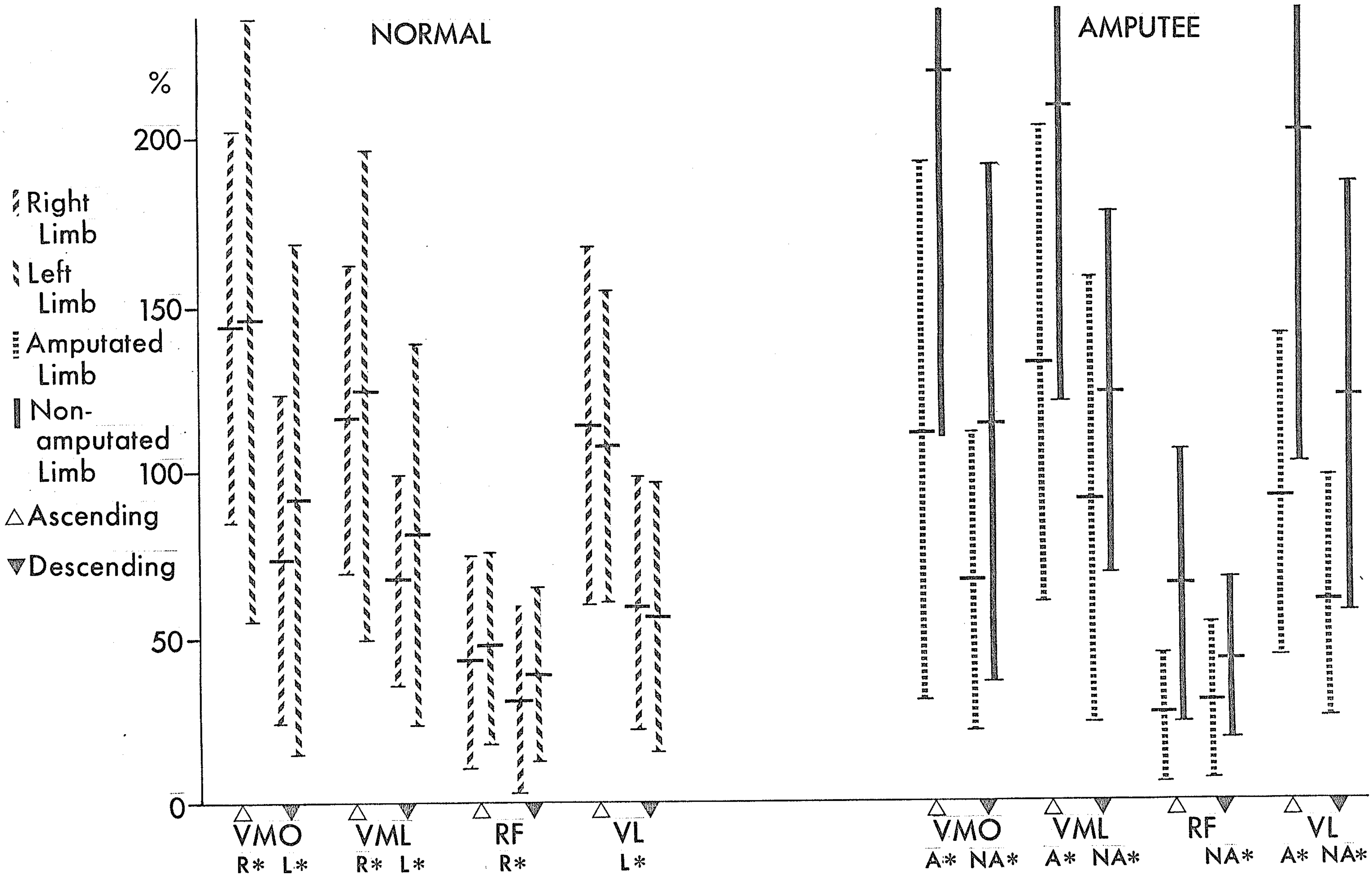


TABLE 64. AMPLITUDE OF THE LATERAL HAMSTRING FIRST PEAK (M51)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	27 ± 24	28 ± 23	NS
DESCEND	16 ± 11	21 ± 22	NS
ASCEND vs. DESCEND	<.05	<.05	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	39 ± 21	29 ± 14	NS
DESCEND	108 ± 114	27 ± 24	NS (.055)
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	<.05*		NS	

*Dispersion Significance <.05. Amputee > Normal.

AMPLITUDE OF THE LATERAL HAMSTRING FIRST PEAK (M51) (TABLE 64)

Unit - Percentage of maximum elicited contraction.

Normal

There was no difference between either limb during ascending or descending. However, in comparing ascending to descending cycles the peak amplitude was less in descending.

Amputee

The amputee did not follow the normal pattern. There was no difference between limbs during ascending and descending, nor in comparing ascending with descending cycles. However, it should be noted that the amputated limb had a greater peak activity than the non-amputated limb to a significance of .055.

Normal versus Amputee

During descending the amputated limb had a greater peak and a greater dispersion than the normal. No difference was found in any of the other group comparisons.

Summary

During descending the lateral hamstrings first peak was greater in the amputated limb than the normal.

TABLE 65. AMPLITUDE OF THE LATERAL HAMSTRING SECOND PEAK (M52)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	15 ± 8 (20 ± 18)	20 ± 14 (19 ± 14)	NS(13)
DESCEND	12 ± 9 (12 ± 8)	14 ± 12 (15 ± 13)	NS(10)
ASCEND vs. DESCEND	<.05(11)	NS(9)	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	95 ± 31	30 ± 10 (32 ± 10)	<.05(5)
DESCEND	34 ± 10	9 ± 6 (24 ± 16)	NS(2)
ASCEND vs. DESCEND	NSD(1)	NS(6)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*	<.001	NS	NS
DESCEND	NSD	NSD	NS	NS

*Dispersion Significance <.01. Amputee > Normal.

AMPLITUDE OF THE LATERAL HAMSTRING SECOND PEAK (M52) (TABLE 65)

Unit - Percentage of maximum elicited contraction.

Normal

There was no difference between limbs in ascending or descending. In comparing ascending with descending cycles the peak amplitude was less in descending for the right limb only.

Amputee

During ascending the amputated limb demonstrated greater activity than the non-amputated limb, but not in descending. In comparing ascending to descending cycles there was not sufficient data for the amputated limb and the non-amputated limb showed no difference.

Normal versus Amputee

During ascending the left amputated limb had a greater peak than the left normal and the right amputated limb had a greater dispersion than the normal. There were no differences in any of the other group comparisons.

Summary

During ascending the left amputated limb has greater activity than the left normal limb and the amputated limbs had greater activity than the non-amputated limb.

TABLE 66. AMPLITUDE OF MEDIAL HAMSTRINGS FIRST PEAK (M61)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	23 ± 13	17 ± 9	NS
DESCEND	10 ± 6	11 ± 5	NS
ASCEND vs. DESCEND	<.001	<.005	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	35 ± 24	34 ± 22	NS
DESCEND	44 ± 28	19 ± 14	<.01
ASCEND vs. DESCEND	NS	NS (.051)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*		NS*	
DESCEND	<.005		NS*	

*Dispersion Significance <.01. Amputee > Normal.

AMPLITUDE OF MEDIAL HAMSTRINGS FIRST PEAK (M61) (TABLE 66)

Unit - Percentage of maximum elicited contraction.

Normal

The results of the data comparison, with respect to differences is the same as for the lateral hamstrings first peak (M51).

Amputee

During descending the amputated limb had a greater peak than the non-amputated but not during descending. In comparing ascending to descending cycles there was no significant difference. However, it should be noted that during descending the non-amputated limb had a lesser peak than during ascending to a significance of .051.

Normal versus Amputee

During descending the amputated limb had greater activity than the normal. There were no differences in any of the other group comparisons. However all had a significantly greater dispersion in the amputee group data.

Summary

During descending the amputated limb shows greater activity than the normal limb and the non-amputated limb.

TABLE 67. AMPLITUDE OF THE MEDIAL HAMSTRINGS SECOND PEAK (M62)

Unit - Percentage of maximum elicited contraction.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	16 ± 7 (23 ± 23)	17 ± 9 (19 ± 7)	NS(11)
DESCEND	13 ± 13 (12 ± 12)	12 ± 4 (13 ± 4)	NS(7)
ASCEND vs. DESCEND	NS(9)	<.05(8)	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	33 ± 18	39 ± 36 (33 ± 33)	NS(6)
DESCEND	20 ± 16	7 ± 7 (13 ± 11)	NS(2)
ASCEND vs. DESCEND	NSD(1)	NS(8)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS*	NS*	NS
DESCEND	NSD	NSD	NS	NS

*Dispersion Significance <.05. Amputee > Normal.

AMPLITUDE OF THE MEDIAL HAMSTRINGS SECOND PEAK (M62) (TABLE 67)

Unit - Percentage of maximum elicited contraction.

Normal

There was no difference between limbs during ascending or descending. However, in comparing ascending to descending cycles the left limb showed less peak activity during descending but the right did not.

Amputee

There was no difference between limbs during ascending or descending. In comparing ascending to descending cycles there was not sufficient data for the amputated limb and the non-amputated limb showed no difference.

Normal versus Amputee

Where there was sufficient data no significant differences were found in the group comparisons. However, the left amputated and the right non-amputated limbs data had a greater dispersion than the normal.

Summary

No difference was found between limbs and between groups.

AMPLITUDE OF HAMSTRING PEAKS--DISCUSSION

Certain distinct patterns emerge in studying the data.

Normal Groups (Figs. 63 and 64)

1) During ascending and descending cycles there is no easily distinguishable sequence of amplitude, however, the lateral hamstrings first peak in both cycles has the overall greatest amplitude.

2) There was no significant difference between the peak amplitudes or either right or left limbs in both ascending or descending cycles.

3) In all but two of the eight comparisons between ascending and descending cycles the descending peak was singificantly less.

The mean activity of the hamstrings in ascending and descending stairs does not show a great variation in peak amplitude. This may indicate that they all contribute to stair climbing to approximately the same extent with the possible exception of the lateral hamstrings' first peak which works to a greater extent in the swing phase of both cycles.

The uniformity between limbs indicates that both limbs contribute equally to stair climbing.

The ascending and descending difference again relates to positive and negative work explained in the discussion of the normal quadriceps group.

Amputee Group (Figs. 63 and 64)

1) During ascending all the muscles demonstrated a tendency for the amputated limb peaks to be greater than the non-amputated limb peaks. In the lateral hamstring second peak this was significantly greater.

2) During descending this tendency for the amputated limbs to have greater peaks is more marked, the medial hamstrings first peak being significantly greater, andthe lateral hamstrings first peak having a significance of .055.

3) In comparing ascending to descending cycles there were no significant differences, although the medial hamstrings first peak had a greater ascending peak in the non-amputated limb of .051.

4) Again in comparing ascending to descending cycles, the unique pattern seen in rectus femoris is seen for both first hamstring peaks. In the amputated limb the means are greater for the descending cycle than for the ascending cycle. This is contrary to the pattern seen in every other comparison.

It is apparent that the amputated limb has almost consistently higher peaks than the non-amputated which indicates a limb preference or dominance.

There is no ascending versus descending difference through one must consider the unique pattern described, that of the descending amputated limb showing grater peak activity. A possible explanation of this is that during descending the hamstrings of the amputated limb must work harder to control the tibia. As mentioned earlier when discussing normal hamstrings peak positions in the cycle, the hamstrings control the tibia, extending it to control the descent and concurrently prevent excessive hip flexion. This hip flexion control allows the pelvis to stay in an appropriately balanced position instead of falling foward, which would be the natural tendency when going down stairs.

The tibial control is decreased in the amputated limb because of the loss of the plantar flexors. These muscles also moved the tibia backwards, or rather controlled the forward movement. Because of this the peak in the hamstrings in the amputated limb has to be much higher to compensate for the loss of plantar flexion power.

In addition, because of the loss of plantar flexion there is less balance control in stopping the body from moving forward and this is another possible reason for the amputated limb to have a higher peak.

In the non-amputated limb this is not necessary and the normal pattern of higher peaks for concentric contraction applies.

Normal versus Amputee (Fig. 63)

Certain distinct patterns emerge.

1) There is generally greater activity in the hamstrings of the amputated limb than in the normal.

2) Limb difference is not seen in the normal but is seen in the amputee.

3) In descending both first peaks of both medial and lateral hamstrings show significantly greater activity in the amputated limb than the normal.

The greater peak activity in the amputee indicates a greater reliance or importance of the hamstrings to that group.

The greater activity of the amputated limb over the normal in descending has been discussed with reference to the non-amputated limb and the same explanation applies. Firstly, the amputated limb lacks the power of plantar flexion. Because of this during descending stance phase the hamstrings have to work harder to control the tibia in attempting to move it backwards and also in balancing the pelvis preventing the trunk from falling forward. This explanation coincides most appropriately with the dramatic shift seen in the significantly shifted peak of the amputated limb in descending (Fig. 65).

EMG PEAK ACTIVITY HAMSTRINGS AND QUADRICEPS--DISCUSSION

The peak activity for both groups of muscles has been discussed within groups and between groups for both cycles.

A brief summary of the quadriceps and hamstrings peak activity is now given. Figures 62 and 64 summarize the 68 peaks discussed above and are drawn to the same scale to allow comparison. The following patterns are seen:

1) The quadriceps showed greater overall activity than the hamstrings during both stair cycles.

2) The non-amputated limb demonstrated greater activity than the normal in the quadriceps muscle, however the amputated limb demonstrates greater activity than the normal in the hamstring muscles.

3) Overall the two-joint muscles, hamstrings and rectus femoris, had lesser peak activity than the one-joint muscles, the vasti.

The fact that the quadriceps are the prime controllers of knee extension is obvious because they are the only functioning knee extensor.

A possible explanation for the non-amputated limb quadriceps greater peak activity and the amputated limb hamstrings greater peak activity has been given.

It is possible that the lesser activity in two-joint muscles relates to their greater flexibility of function, namely the control over two-joint actions at opposite ends. The vasti are knee extensors but the hip flexion function of rectus femoris can be performed by other muscles. With respect to the hamstrings there are other muscles that help control the tibia in weight-bearing, and gravity assists in knee flexion during ascending so that peak activity would not need to be as great.

Fig. 63

AMPLITUDE OF HAMSTRING PEAKS-ASCENDING CYCLE

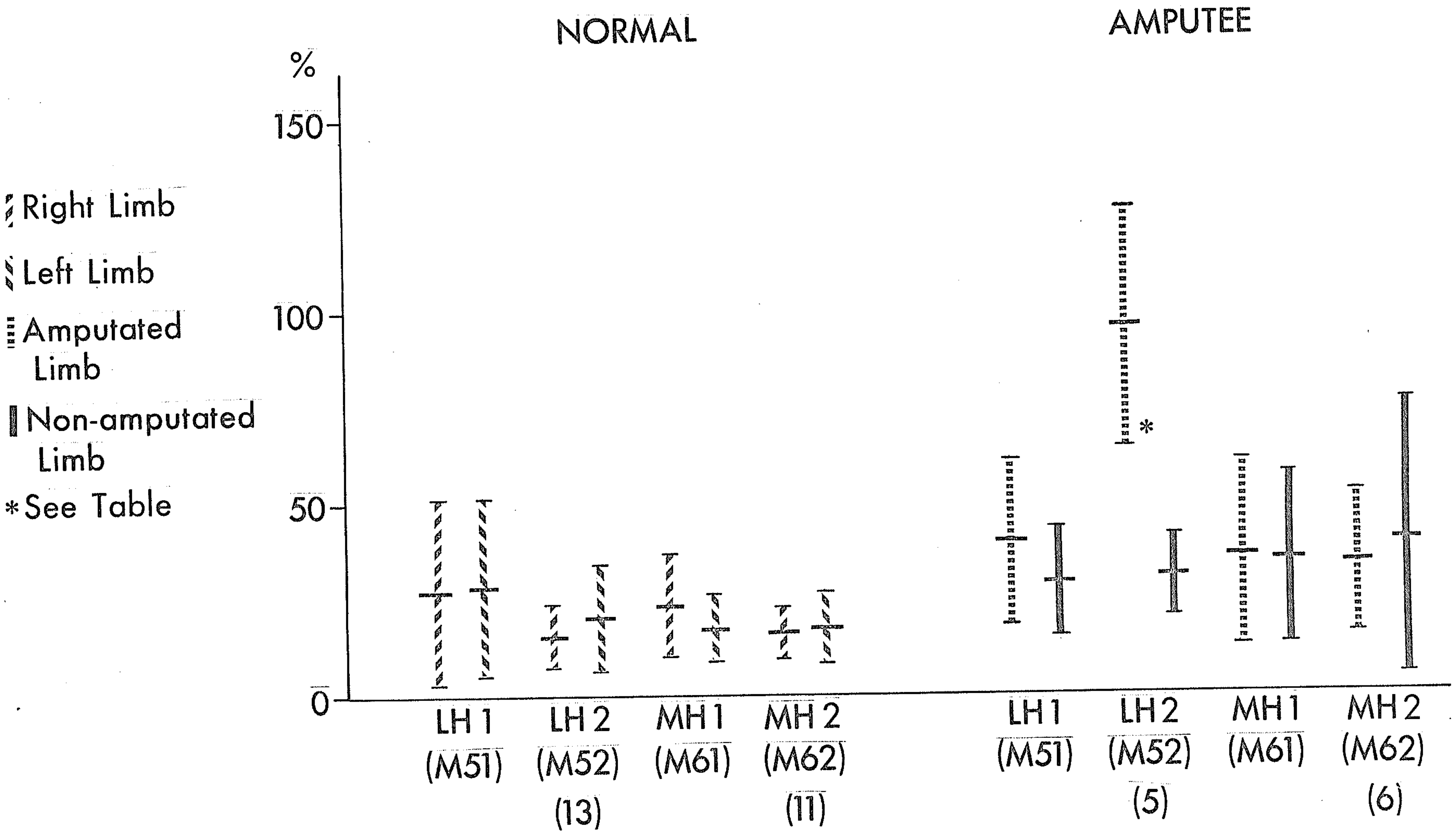
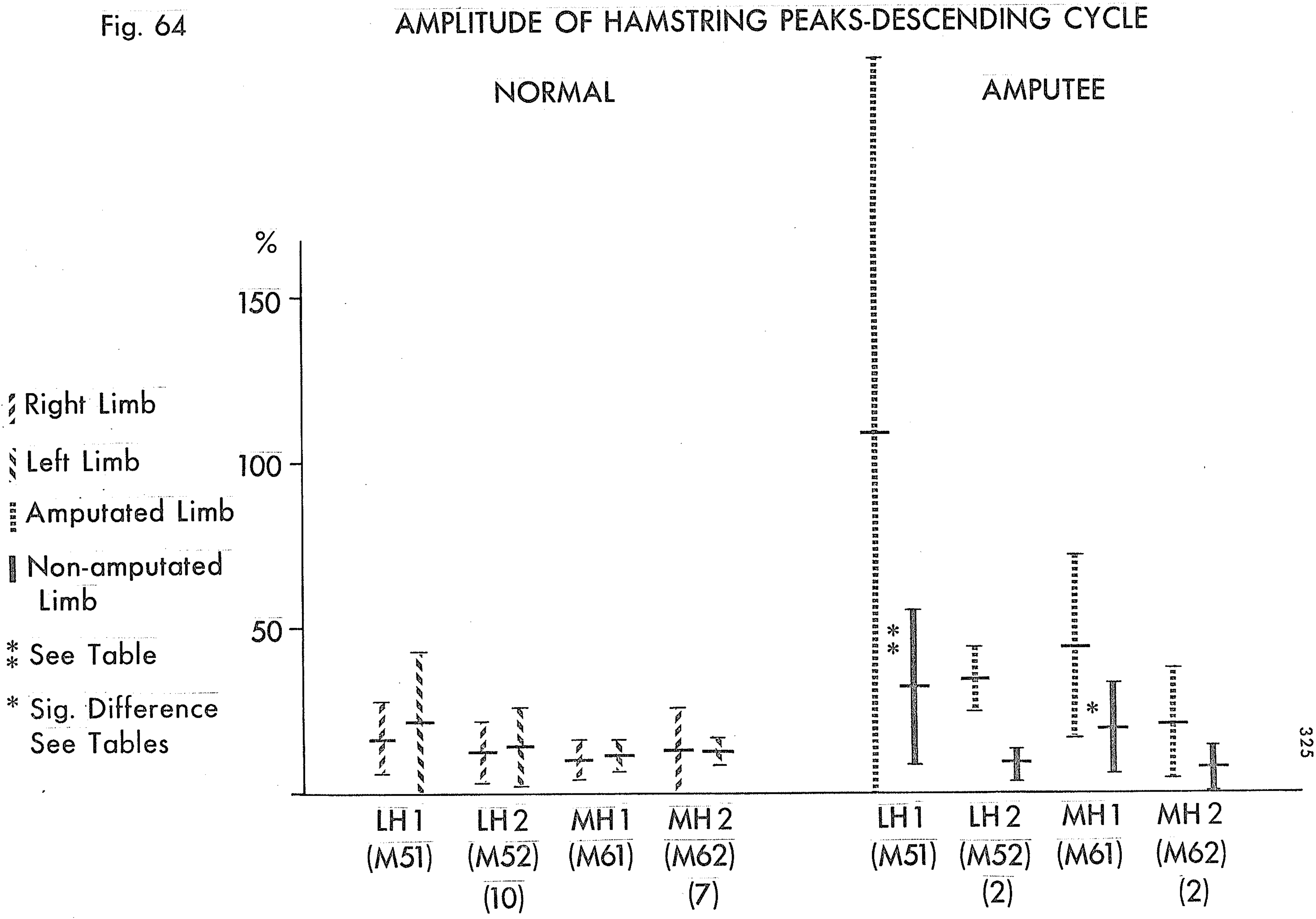
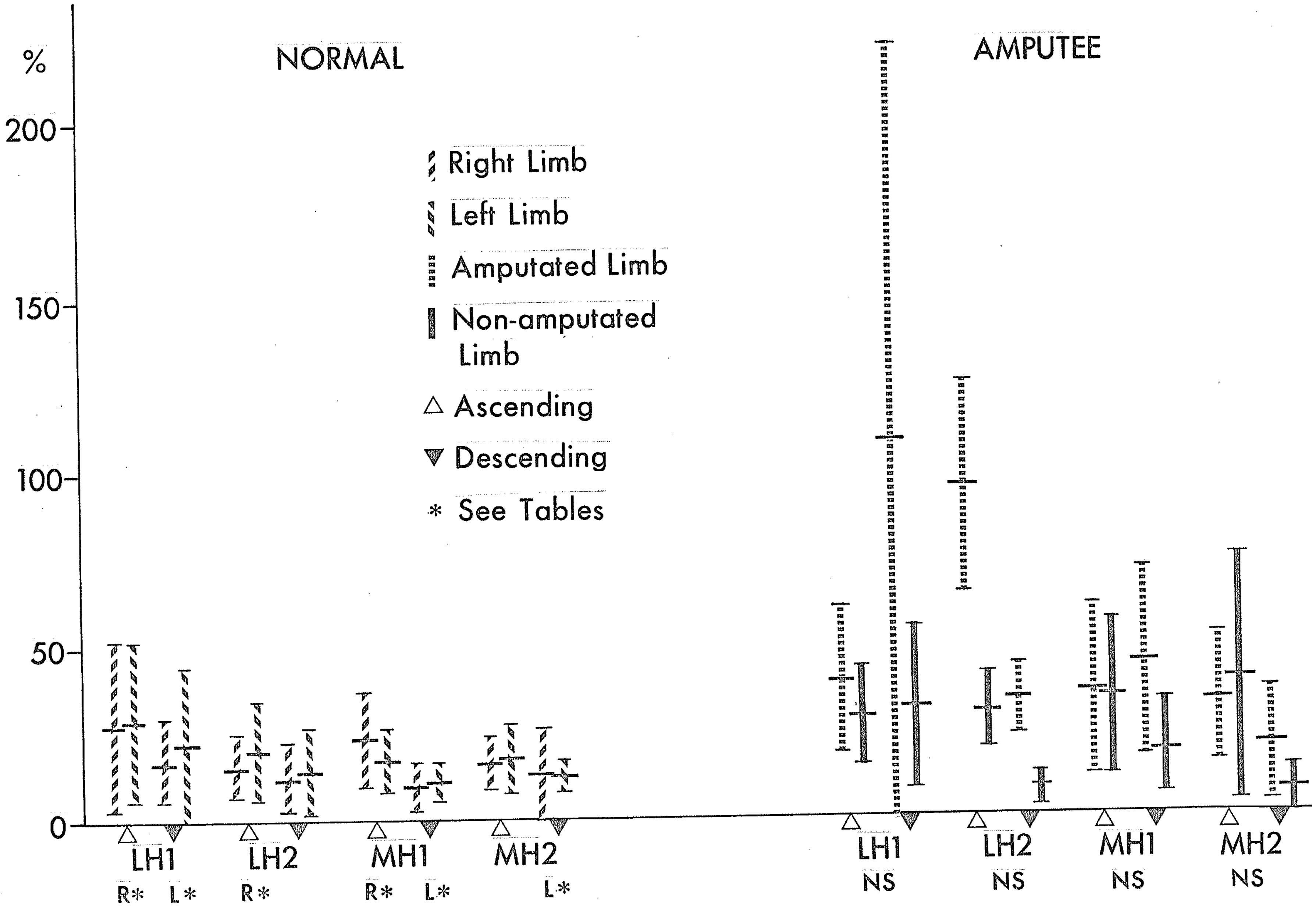


Fig. 64



AMPLITUDE OF HAMSTRING PEAKS-DESCENDING CYCLE
NORMAL
AMPUTEE
%
150
100
50
Right Limb
Left Limb
Amputated Limb
Non-amputated Limb
** See Table
* Sig. Difference See Tables
**
*
LH 1 (M51)
LH 2 (M52) (10)
MH 1 (M61)
MH 2 (M62) (7)
LH 1 (M51)
LH 2 (M52) (2)
MH 1 (M61)
MH 2 (M62) (2)

Fig. 65 AMPLITUDE OF HAMSTRING PEAKS-BOTH CYCLES



ELECTROGONIOMETRIC DATA CORRELATED WITH PEAK ELECTROMYOGRAPHIC DATA

In this section the following data are examined:

1) The knee angle at the quadriceps peaks.

2) The knee angle at the hamstrings peaks.

The purpose of this section was to see the correlation or lack of correlation between the knee joint angle and the peaks.

Although the peaks occur at different parts of the cycle during ascending and descending cycles, the ascending versus descending comparison was made to see if there was any correlation between knee angle and the muscle peak. In addition, the interplay of quadriceps and hamstring peaks with the knee angle are studied.

RELIABILITY - See Appendix B.

SAMPLE SIZE - See Appendix C.

TABLE 68. KNEE ANGLE AT VASTUS MEDIALIS OBLIQUE PEAK (M1)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	65 ± 10	69 ± 7	NS
DESCEND	46 ± 8	54 ± 17	NS
ASCEND vs. DESCEND	<.005	<.005	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	53 ± 15	63 ± 6	<.05
DESCEND	39 ± 22	60 ± 8	<.01
ASCEND vs. DESCEND	<.05	<.05	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	<.005*		NS	
DESCEND	NS*		<.01	

*Dispersion Significance <.05. Amputee > Normal.

KNEE ANGLE AT VASTUS MEDIALIS OBLIQUE PEAK (M1) (TABLE 68)

Unit - Degrees

Normal

There was no difference between either limbs during ascending or descending cycles. However, in comparing ascending to descending cycles the peak occurred with the knee at a more extended position during descending.

Amputee

During ascending and descending the non-amputated limb flexed more than the amputated. In comparing ascending to descending cycles, the peak occurred with the knee at a more extended position during descending.

Normal versus Amputee

During ascending the normal limb flexed more than the amputated limb. During descending the non-amputated limb flexed more than the normal. During descending, there was a significantly greater dispersion of the amputated limb data than in the normal.

Summary

Both groups have a greater knee flexion during the ascending peak than descending.

During ascending and descending stairs the non-ampuated limb is flexed more than the amputated during the peak.

In comparing groups during ascending, the amputated limb flexes less than the normal, and during descending the non-amputated limb flexes more than the normal during the vastus medialis oblique peak.

TABLE 69. KNEE ANGLE AT VASTUS MEDIALIS LONGUS PEAK (M2)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	58 ± 9	61 ± 7	NS
DESCEND	53 ± 7	58 ± 9	<.05
ASCEND vs. DESCEND	<.005	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	46 ± 18	61 ± 6	<.005
DESCEND	53 ± 19	63 ± 7	<.005
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	<.05*	NS	NS
DESCEND	NS	<.01*	<.05	NS

*Dispersion Significance <.05. Amputee > Normal.

KNEE ANGLE AT VASTUS MEDIALIS LONGUS PEAK (M2) (TABLE 69)

Unit - Degrees.

Normal

During descending the right peak occurred at a lesser degree of flexion than the left peak, however, there was no difference between either limbs during descending. Comparing ascending with descending cycles the descending right limb had a lesser degree of flexion than the ascending limb; at the time of the peak however there was no difference in the left limb.

Amputee

During ascending and descending the non-amputated limb was flexed more than the amputated limb during the peak. Comparing ascending to descending cycles there was no difference.

Normal versus Amputee

During ascending and descending the left amputated limb flexed less than the left normal limb during the peak. During descending the right non-amputated limbs flexed more than the right normal limbs during the peak. There were no differences in the other group comparisons.

Summary

In the normal group, during descending the vastus medialis longus peak of the right limb occurred at a lesser knee angle than the left limb. Comparing ascending to descending only the right leg had a less flexed limb position during descending.

The non-amputated limb peaks occurred at a greater angle of flexion than the amputated peaks. During both cycles the left amputated limb peak occurred at a lesser angle than did the normal.

TABLE 70. KNEE ANGLE AT RECTUS FEMORIS USUAL PEAK (M32)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	55 ± 13	60 ± 9	.05
DESCEND	50 ± 18	59 ± 12	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	41 ± 16	53 ± 13	NS
DESCEND	36 ± 21	56 ± 7	<.05
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	<.001	NS	NS
DESCEND	NS	NS*	NS	NS

*Dispersion Significance <.05. Amputee > Normal.

KNEE ANGLE AT RECTUS FEMORIS USUAL PEAK (M32) (TABLE 70)

Unit - Degrees.

Normal

During ascending the left limb peak occurred at a greater degree of flexion than the right, however there was no difference in descending. In comparing ascending to descending cycles there was no difference.

Amputee

During descending the non-amputated limb was more flexed during the peak than the amputated limb, however there was no difference in ascending. In comparing ascending to descending cycles there was no difference.

Normal versus Amputee

During ascending the left amputated limb peak occurred during less flexion than did the left normal limb peak. There were no differences in any of the other group comparisons.

Summary

During ascending the left limb was flexed more than the right during the rectus femoris peak.

During descending the non-amputated limb was flexed more than the amputated limb during the rectus femoris peak.

TABLE 71. KNEE ANGLE AT THE RECTUS FEMORIS EARLY PEAK (M31)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	44	63 (60 ± 3)	NSD(1)
DESCEND	25 ± 17	34 ± 28 (38 ± 45)	NS(11)
ASCEND vs. DESCEND	NSD(1)	NS(3)	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	64 ± 44 (60 ± 37)	76 ± 13 (70 ± 10)	NS(3)
DESCEND	24 ± 31 (24 ± 32)	27 ± 8 (25 ± 12)	NS(4)
ASCEND vs. DESCEND	NS(4)	NS(2)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NSD	NS
DESCEND	NS	NS	NS*	NS

*Dispersion Significance <.05.

NSD = Not Sufficient Data.

KNEE ANGLE AT THE RECTUS FEMORIS EARLY PEAK (M31) (TABLE 71)

Table 71 demonstrates the knee angle during this peak. Where there were sufficient data no significant difference was found.

KNEE ANGLE AT THE RECTUS FEMORIS LATE PEAK (M33)

The three ascending normal right limbs have a mean knee flexion of $7^{\circ} \pm 7^{\circ}$.

The three ascending non-amputated limbs have a mean knee flexion of $27^{\circ} \pm 7^{\circ}$. This is appropriate for their respective positions in the cycle. (See Position of Peak section)

TABLE 72. KNEE ANGLE AT VASTUS LATERALIS PEAK (M4)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	58 ± 8	58 ± 8	NS
DESCEND	54 ± 11	57 ± 7	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	49 ± 19	60 ± 7	<.05
DESCEND	42 ± 23	56 ± 9	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS*		NS	
DESCEND	NS*		NS	

*Dispersion Significance <.05. Amputee > Normal.

KNEE ANGLE AT VASTUS LATERALIS PEAK (M4) (TABLE 72)

Unit - Degrees.

Normal

There was no difference in limbs during ascending or descending. In comparing ascending to descending cycles there was also no difference.

Amputee

During ascending the non-amputated limb was flexed more at the peak than was the amputated, however there was no difference in descending. In comparing ascending to descending cycles there was no difference.

Normal versus Amputee

There were no differences in any of the group comparisons though the amputated limb data has a significantly greater dispersion than the normal.

Summary

During ascending the non-amputated limb peak of vastus lateralis occurred at greater flexion than did the amputated limb peak.

KNEE ANGLE AT THE QUADRICEPS PEAK EMG - DISCUSSION

Figures 66 and 67 summarizes the previous data.

Normal

During ascending all the peak means occurred within 14° of each other (from 55° to 69° flexion). During descending all the peak means occurred within 9° of each other (from 50° to 59° flexion).

Amputee

During ascending and descending cycles the amputated limb peaks occurred at a distinctly lower range of flexion than the non-amputated limb.

During ascending all the peak means of the amputated limbs occurred within 13° of each other (from 41° to 54°); during descending the amputated limb peak means occurred within 7° of each other (from 39° to 56°).

During ascending all the peak means of the non-amputated limb occurred within 9° of each other (from 54° to 63°); and during descending the non-ampuated limb peak means were within 7° of each other (from 56° to 63°).

Normal versus Amputee

During descending in all groups the peaks occurred through a smaller range than in ascending. The amputated limb clearly flexes less during the peak than the non-amputated and the normal.

It is difficult to explain the above findings. The peaks occurring in the amputated limb in greater degrees of extension in ascending may be due to the inability of the amputated limb to generate maximum peak activity as quickly as the normal, and therefore the limb extends further before the peak EMG is reached. An explanation for the peak occurring in more extension during descending could be due to the earlier weight-bearing on the non-ampuated limb since the amputated limb is not "allowed" to flex as much during descending stairs, therefore the peak occurs earlier. Another explanation is that during descending the stance phase in the amputated limb is the shortest and therefore, relatively, the peak occurs earlier.

Fig. 66

KNEE ANGLE AT QUADRICEPS PEAK - ASCENDING

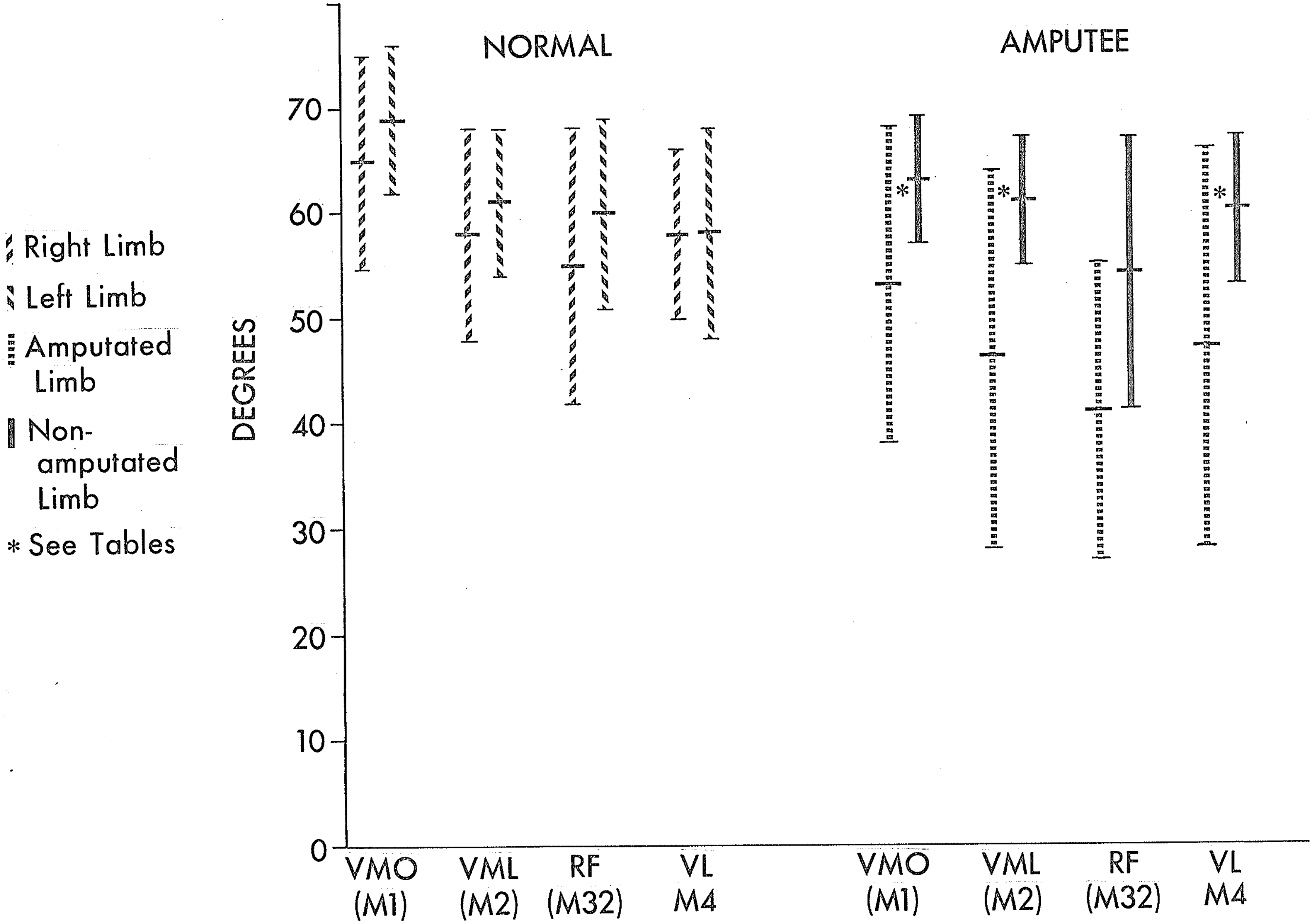
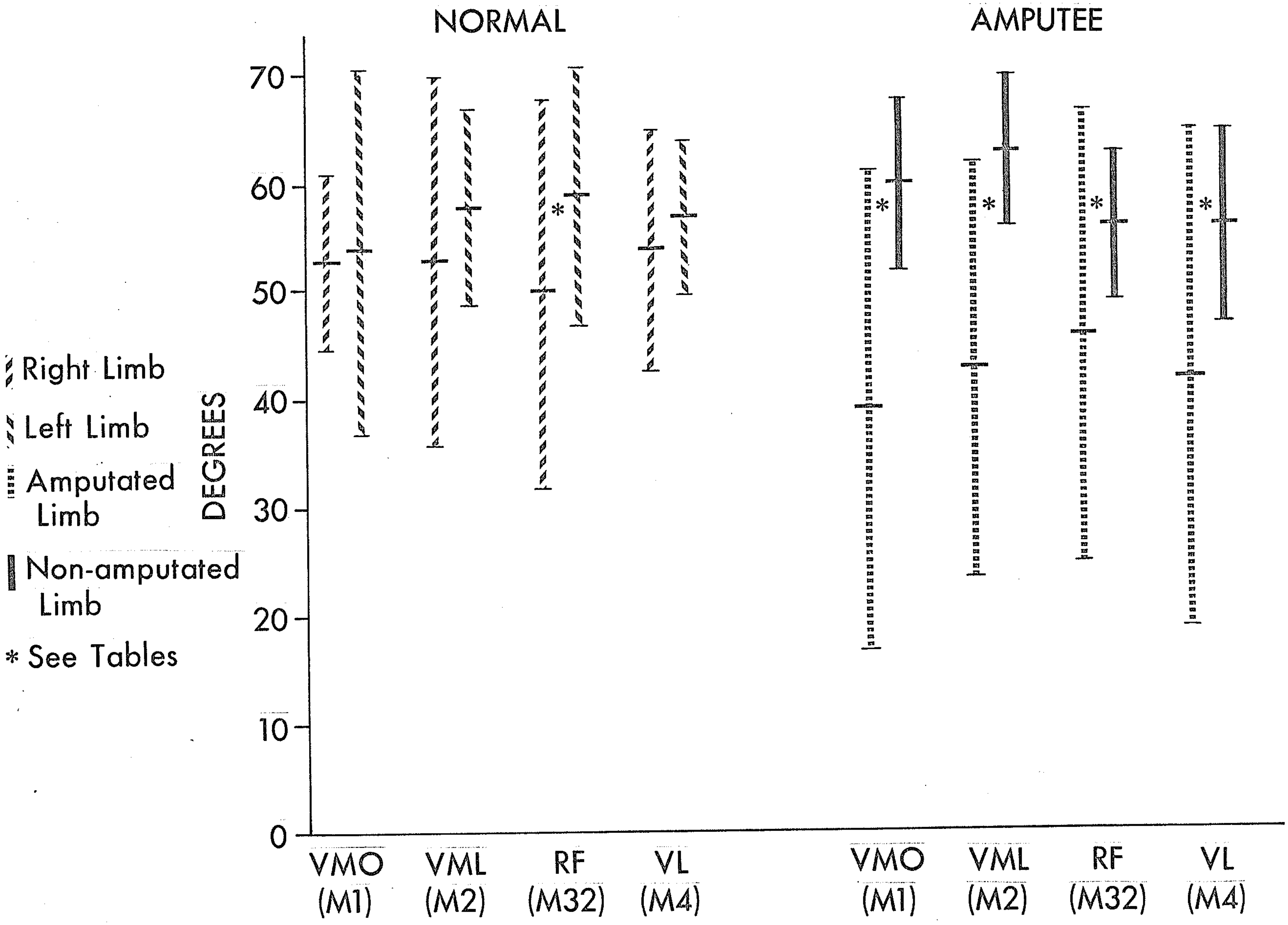


Fig. 67 KNEE ANGLE AT QUADRICEPS PEAK - DESCENDING



KNEE ANGLE AT THE HAMSTRING PEAKS (Figs. 68 and 69)

The first and second peaks of both hamstring groups (Tables 73 to 76) are considered together because of their almost uniform lack of significant differences in the comparisons, with the exception of the medial hamstring first peak when the right peak occurs at greater flexion in descending than in ascending. During descending the second peak data has not been placed on the graph because it represents only two cases and the standard deviation is large (Fig. 69).

The knee angles corroborate the previous statements of hamstring function. During ascending, the first peak is for flexing the knee, and during descending the first peak is most probably a tibial stabilizer since it occurs slightly earlier in knee flexion than the quadriceps peaks, probably an earlier peak to control or counter the quadriceps peak.

The second peaks are more difficult to evaluate and may tend to have a stability function as well. This is deduced from observing goniometric data and peak position. In descending, however, this peak may help lift the leg off the step.

TABLE 73. KNEE ANGLE AT LATERAL HAMSTRING FIRST PEAK (M51)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	26 ± 19	34 ± 19	NS
DESCEND	30 ± 24	43 ± 28	NS
ASCEND vs. DESCEND	NS	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	28 ± 16	37 ± 26	NS
DESCEND	35 ± 19	48 ± 30	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS*	
DESCEND	NS		NS*	

*Dispersion Significance <.05. Amputee > Normal.

TABLE 74. KNEE ANGLE AT LATERAL HAMSTRING SECOND PEAK (M52)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	41 ± 21 (41 ± 21)	34 ± 30 (38 ± 32)	NS(13)
DESCEND	58 ± 31 (58 ± 30	64 ± 26 (62 ± 27)	NS(10)
ASCEND vs. DESCEND	NS(11)	NS(9)	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	36 ± 18	37 ± 20 (41 ± 24)	NS(5)
DESCEND	34 ± 17	42 ± 48 (46 ± 34)	NS(2)
ASCEND vs. DESCEND	NSD(1)	NS(6)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NS	NS
DESCEND	NSD	NSD	NS	<.05

TABLE 75. KNEE ANGLE AT MEDIAL HAMSTRING FIRST PEAK (M61)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	26 ± 27	38 ± 26	NS
DESCEND	47 ± 27	52 ± 29	NS
ASCEND vs. DESCEND	<.05	NS	

AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	25 ± 19	51 ± 31	NS (.054)
DESCEND	43 ± 23	41 ± 28	NS
ASCEND vs. DESCEND	NS	NS	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS		NS	
DESCEND	NS		NS	

TABLE 76. KNEE ANGLE AT MEDIAL HAMSTRING SECOND PEAK (M62)

Unit - Degrees.

NORMAL GROUP

	RIGHT	LEFT	RIGHT vs. LEFT
ASCEND	33 ± 29 (40 ± 26)	44 ± 33 (36 ± 36)	NS(11)
DESCEND	48 ± 39 (55 ± 37)	69 ± 28 (63 ± 32)	NS(7)
ASCEND vs. DESCEND	NS(9)	NS(8)	

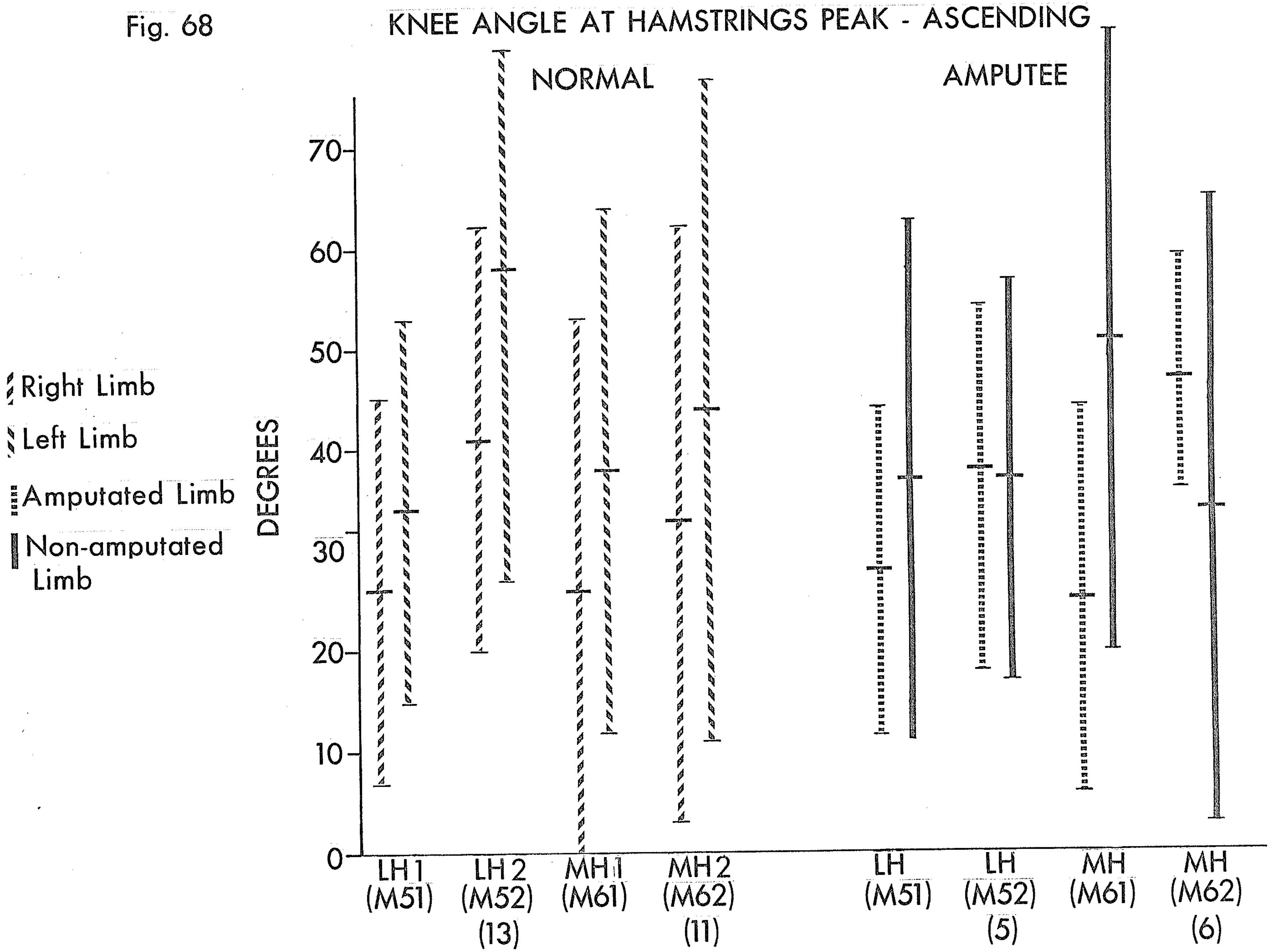
AMPUTEE GROUP

	AMPUTATED	NON-AMPUTATED	AMPUTATED vs. NON-AMPUTATED
ASCEND	47 ± 12	34 ± 31 (34 ± 27)	NS(6)
DESCEND	26 ± 31	42 ± 48 (57 ± 28)	NS(2)
ASCEND vs. DESCEND	NSD(1)	NS(8)	

NORMAL GROUP COMPARED WITH AMPUTEE GROUP

	NORMAL vs. AMPUTATED		NORMAL vs. NON-AMPUTATED	
	RIGHT	LEFT	RIGHT	LEFT
ASCEND	NS	NS	NS	NS
DESCEND	NSD	NSD	NS	NS

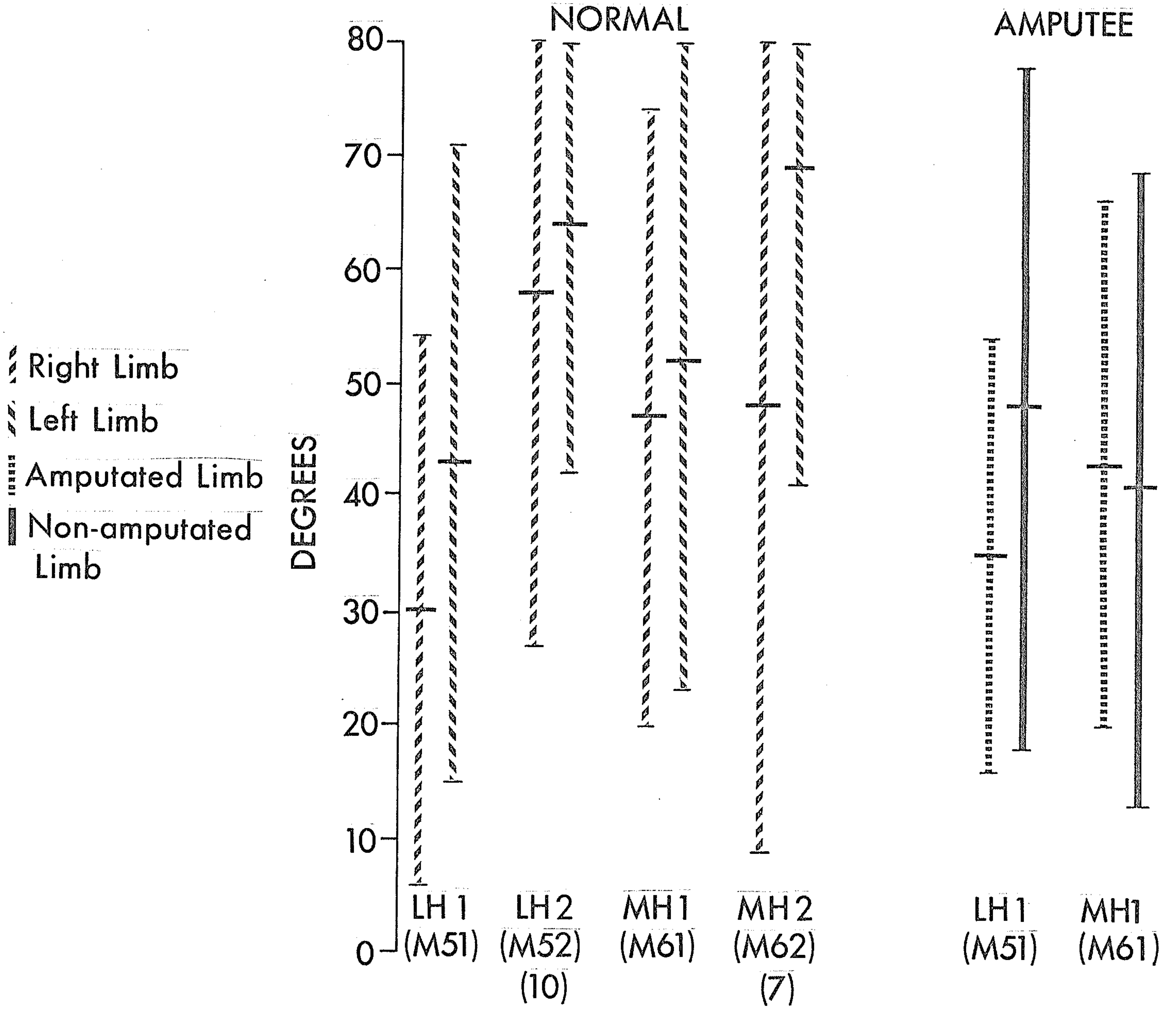
Fig. 68



KNEE ANGLE AT HAMSTRINGS PEAK - ASCENDING

Fig. 69

KNEE ANGLE AT HAMSTRINGS PEAK - DESCENDING



CHAPTER V

APPLICATION

The major purpose of this study was to derive an objective model of both stair climbing cycles in humans. The particular areas of interest were knee joint movements, and thigh muscle activity and the establishment of weight-bearing and non-weight-bearing of the limb during the cycles in normals and Syme amputees.

Future Clinical Application

The major reason for giving precise temporal data is for its possible application in two main clinical areas:

1) Myoelectric control
2) Functional electrical stimulation (FES)

Myoelectric Control

At the present time myoelectric devices for the control of upper limb orthoses and prostheses have been used for over two decades (Bottomley and Cowell, 1964), after the possibility was suggested by Battye et al. in 1955.

Many devices have been made and modified since then in North America and Europe (McKenzie, 1965; Knowles et al., 1965; Scott, 1967; Staros and Peizer, 1975), and in Russia (Popov, 1965).

In the lower limb, however, since the energy requirements of prostheses are very high and the problems of storage of the power requirements are not resolved, myoelectric devices are presently EMG controlled rather than powered (Horn, 1972; Dyck et al., 1975). With these prostheses above knee amputees can be trained in stair climbing "in a matter of hours" (Horn, 1972).

More efficient and longer lasting energy sources are being explored with greater intensity than ever before because of the current world awareness of energy shortage and the need for power conservation. Because of this the future development of a powered plantar flexing ankle controlled by a sequence of quadriceps contraction is possible.

It is acknowledged that the Syme's amputee has a generally better functional limb than amputees at a higher level and it is difficult to say how much extrapolation can be made from the data presented here to higher above knee amputees.

Functional Electrical Stimulation (FES)

Twenty years ago Liberson and his fellow workers (1961) selectively stimulated the dorsiflexion of the ankle joint in hemiplegic patients by means of a heel switch on the patient's leg triggered at the appropriate time in the gait cycle. This study initiated numerous research projects and was the forerunner of a new technology in rehabilitation establishing functional electrical stimulation as a field of research.

Interest in this field of rehabilitation increased both in research and in the development of commercial units. Multi-institutional and interdisciplinary work now occurs in the United States, Yugoslavia, Sweden and Switzerland (Kralj and Vodovnik, 1977, 1977a).

The initial single channel was developed into a double channel system in Rancho Los Amigos Hospital Rehabilitation Engineering Centre in California, over five years after multi-site FES was proposed (Vodovnik et al., 1965). This expanded into three channels and presently six channel systems are used (Stanic et al., 1978).

The applications of FES are wide and varied and, with the exception of stimulation after peripheral nerve lesion, an intact lower motor neurone is needed. The majority of work is done with hemiplegic patients (Acimovic et al., 1976; Savinelli et al., 1978) though areas of research include cerebral palsy children and idiopathic scoliosis (Kralj and Vodovnik, 1977).

FES is also used to suppress spasticity, to aid in incontinence, to prevent atrophy of muscle structures with peripheral nerve lesions, to induce bone growth and healing, and to assist in pain suppression (all of which are referenced in Kralj and Vodovnik, 1977).

A major area of study is with respect to gait. In gait studies there is need for control of onset, duration and amplitude of the electrical signal. These signals may be modified by electrogoniometers (Trnkoczy and Bajd, 1975) and by an on-time processor computer (Valencic et al., 1977).

At present the stimulation sequences are set for a given speed and largely determined intuitively, relying on normal muscle activity during gait. In addition, the amplitude changes that occur with time increase the complexity of the technique (Kralj and Vodovnik, 1977).

"Such stimulation may have many settings, timing parameters and amplitudes but, because the evaluation and analysis of gait are not yet sophisticated enough, the accurate presumption of a stimulation sequence for individual patients is not possible." (Kralj and Vodovnik, 1977).

In this study the sequence and peak amplitudes of the thigh muscles are stated in terms of absolute time and a relationship to maximal elicited contraction in normal subjects and Syme's amputees. Since the establishment of norms gives a baseline or frame of reference for further methodological study, it is hoped that this data can be used in FES and stair climbing.

Exercise Therapy

1) It is clear from our results that rectus femoris does not play a major part in either stair cycle though the other parts of the quadriceps studied do.

Integrating this information with that found by Deutsch and Lin (1978), who found that if strengthening of the vasti is of greater importance a hip angle of 90^{o} must be maintained; both amputees and normal subjects should have their hips at 90^{o} during some part of their exercise program. Stair climbing, step over step, is a function of the terminal stages of rehabilitation and at that time, more quadriceps exercise should be done in sitting rather than lying down. The latter position is a common method of quadriceps re-education in amputees at the present time.

2) It would also be appropriate to apply maximum resistance in exercise in a similar pattern to that associated with the peak activity found. Therefore, maximum concentric contraction should occur between 55^{o} and 65^{o} fo knee flexion and maximum eccentric contraction between 50^{o} to 60^{o} of flexion. The hip during these exercises should be between 30^{o} to 40^{o} of flexion. This is calculated from Berry (1952) and Joseph and Watson's (1967) paper.

3) During ascending stairs since the rectus femoris is not a constant hip flexor, this function must be subserved by the other hip flexors. Because of this, stair climbing, which is associated with the later stages of rehabilitation, should include re-education of all hip flexors with an emphasis on the muscles other than rectus femoris.

4) In stair climbing or the later stages of rehabilitation, the importance of the eccentric control of the knee should be stressed. It is often seen clinically in rehabilitating the injured knee that the descending stair activity is the last or often the more difficult of the two stair cycles to be performed normally.

5) Hamstring re-education should occur with the knee at 55° to 65° of flexion in both eccentric and concentric modes with the hip at 40° to 60° of flexion.

6) As the hamstring activity is greater in the amputee group than the normal, both concentric and eccentric peak activity in the above ranges with maximum resistance should be emphasized.

CHAPTER VI

SUMMARY

The purpose of this study was to have a comprehensive evaluation of stair climbing with respect to temporal data, knee electrogoniometry and thigh muscle peak electromyography, in two groups of subjects. Data from the normal group was compared with data from a Syme amputee group. Unilateral Syme amputees were chosen because of their unique end weight bearing characteristic and because as sensitive a comparison as possible was to be made to see how the functional changes in the thigh muscles and knee joint were affected by the loss of a single foot.

Eighteen normal subjects and 10 amputees were evaluated.

Both limbs of each group were evaluated in both ascending and descending cycles.

Five steps were ascended and descended three times and the middle step data for each of the cycles were recorded on specially designed capture sheets for computer analysis. The reliability of the data was high. The middle step was chosen to minimize the acceleration and deceleration effects.

The subjects wore a foot switch insole and a knee electrogoniometer capable of measuring single plane movement with a varying axis. These instruments were conceived, designed and tested in conjunction with the Rehabilitation Centre for Children Gait Laboratory (formerly the Shriners Hospital, Winnipeg).

The muscles studied were the prime flexors and extensors of the knee; specifically the oblique and long parts of vastus medialis, rectus femoris, and vastus lateralis, and the medial and lateral hamstring groups. Surface electrodes were used, and the signal was full wave rectified and low pass filtered through an established electromyographic amplifier system.

With respect to temporal data, ascending stairs takes longer than descending in normals but this difference does not occur in the amputee group. Although the amputees have a tendency to take longer in both cycles this was not found to be statistically significant. The amputee demonstrated a difference between the limbs in the ascending cycle only, spending longer on the amputated limb.

The swing/stance ratio in normal stair climbing is the same as in walking, namely 40% swing 60% stance. There was a statistically significant difference between the right and left limbs during ascending even though the difference in means was only 3% of the cycle. During descending stairs less time is spent in weight bearing than in ascending stairs.

The amputee spends less time on the amputated limb than the normal, and during descending the amputated limb has a 50%-50% weight bearing/ non-weight bearing split.

With respect to electrogoniometric data there was a great similarity between the limbs in the normal group. In the amputee group there were differences between the non-amputated and the amputated limb. During ascending, maximum flexion occurs earlier in the non-amputated limb than it does in the amputated or in the normal limb. This indicates faster flexion of the non-amputated limb.

In descending stairs the amputated limb is not flexed as much as the normal perhaps because of the lack of capability of dorsiflexion due to the prosthesis.

The amputee takes longer to extend the amputated limb during ascending and the non-amputated limb moves through a smaller range of movement than the amputated limb.

With respect to the electromyographic data, all the quadriceps in the normal group, with the exception of rectus femoris, were monophasic in both cycles. The position of the peaks of the quadriceps during ascending was in early stance and during descending in late stance. (Rectus femoris showed a complex pattern of peaks.)

The non-amputated limb quadriceps peaked before the amputated only during ascending.

The amputated limb quadriceps peaked after the normal in ascending only, and where there were differences between the normal and non-amputated limbs, the non-amputated limb always peaked earlier.

The normal hamstring peaks were generally biphasic and showed complex patterns. The first peak during ascending occurred in swing and the second in stance. During descending the first peak occurred in early stance and the second in late stance. The hamstring peaks occurred before the quadriceps peak, indicating a stabilizing function.

The heads of the quadriceps in the normal group contracted in a specific sequence, the most consistent of which was the oblique fibres of vastus medialis which always contracted first irrespective of limb, group or cycle. The hamstrings also had a peak sequence although they were more widely spread.

The amplitude of the peak EMG was compared with a maximum elicited isometric contract. In considering normal quadriceps peak amplitude, there was no difference between limbs. The vasti are more active than rectus femoris in both ascending and descending cycles. The peak amplitude was greater in ascending than in descending cycles.

In the amputee group there was a limb difference. The non-amputated limb had greater quadriceps peak amplitude than the amputated limb. As in the normal group the vasti were more active than rectus femoris in both ascending and descending cycles; and the peak amplitude was greater in ascending than descending.

The hamstrings showed considerably smaller peak amplitude than the quadriceps in both groups. In addition, in the amputee group there was a greater peak in the amputated limb rather than in the non-amputated limb, an opposite pattern to the quadriceps activity.

With respect to the knee position during the peak activity, there was no difference between limbs and all the normal quadriceps means occur within a small range from 50° to less than 70°.

The amputee group demonstrated a difference between limbs. The non-amputated limb had peaks at approximately the same knee angle as the normal group; however, the non-amputated limb had peaks at a lesser degree of flexion in both cycles than the normal.

The hamstring peaks occurred in a less flexed knee position than the quadriceps peaks but the standard deviation is greater.

The future clinical application of this study would be in the fields of myoelectric control, functional electrical stimulation and exercise therapy.

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APPENDIX A

DATA SHEETS

①

GENERAL DATA

I.D. ☐☐☐ 3 — FIRST = GROUP 1 = CONTROL
2 = AMPUTEE
SECOND + THIRD = NUMBER

1 = EGM
2 = EMG
3 = EMG v EGM — ☐☐☐ A 7

AGE ☐☐ 9 — YEARS

SEX ☐ 10 — MALE = 1
FEMALE = 2

WEIGHT ☐☐☐☐ 14 — KILOGRAMS

HEIGHT ☐☐☐☐ 18 — CENTIMETRES

AMPUTATION SIDE ☐ 19 — RIGHT = 1
LEFT = 2

PROSTHESIS TYPE ☐ 20 — OLD SYMES = 1
NEW SYMES = 2.

STUMP LENGTH ☐☐☐ 23 — CENTIMETERS.

MAXIMUM Ⓡ

MX	CNR	M11	26
MX	CNR	M12	29
MX	CNR	M13	32

M4

MX	CNR	M41	53
MX	CNR	M42	56
MX	CNR	M43	59

MX	CNR	M21	35
MX	CNR	M22	38
MX	CNR	M23	41

M5

MX	CNR	M51	62
MX	CNR	M52	65
MX	CNR	M53	68

3

MX	CNR	M31	44
MX	CNR	M32	47
MX	CNR	M33	50

M6

MX	CNR	M61	71
MX	CNR	M62	74
MX	CNR	M63	77

MAXIMUM Ⓛ

B 7

M1

MX	CNR	M11	10
MX	CNR	M12	13
MX	CNR	M13	16

M4

MX	CNR	M41	37
MX	CNR	M42	40
MX	CNR	M43	43

M2

MX	CNR	M21	19
MX	CNR	M22	22
MX	CNR	M23	25

M5

MX	CNR	M51	46
MX	CNR	M52	49
MX	CNR	M53	52

M3

MX	CNR	M31	28
MX	CNR	M32	31
MX	CNR	M33	34

M6

MX	CNR	M61	55
MX	CNR	M62	58
MX	CNR	M63	61

②

ELECTROGONIOMETER DATA.

I.D. ☐☐☐ 3   1 4   LEG 1 5

ASCEND

1 C 7

AS	WH	R1	10
AS	WH	R2	17
AS	WH	R3	24

ST.

A	T	H	R1	14
A	T	H	R2	21
A	T	H	R3	28

W

AS	TSW	R1	31
AS	TSW	R2	34
AS	TSW	R3	37

T

AS	TST	R1	40
AS	TST	R2	43
AS	TST	R3	46

L

V

AM	XFL	R1	49
AM	XFL	R2	55
AM	XFL	R3	61

H

AM	XFLH	R1	52
AM	XFLH	R2	58
AM	XFLH	R3	64

D 7

ow
ER

V

AM	IFL	R1	10
AM	IFL	R2	16
AM	IFL	R3	22

H

AM	IFLH	R1	13
AM	IFLH	R2	19
AM	IFLH	R3	25

DESCEND.

2 E 7

DS	WH	R1	10
DS	WH	R2	17
DS	WH	R3	24

D	T	H	R1	14
D	T	H	R2	21
D	T	H	R3	28

DS	TSW	R1	31
DS	TSW	R2	34
DS	TSW	R3	37

DS	TST	R1	40
DS	TST	R2	43
DS	TST	R3	46

V

DM	XFL	R1	49
DM	XFL	R2	55
DM	XFL	R3	61

H

DM	XFLH	R1	52
DM	XFLH	R2	58
DM	XFLH	R3	64

F 7

V

DM	IFL	R1	10
DM	IFL	R2	16
DM	IFL	R3	22

H

DM	IFLH	R1	13
DM	IFLH	R2	19
DM	IFLH	R3	25

I.D. ☐☐☐ 3 | 2 4 | LEG 1 5

409. ④

ASCEND

1 K 7

A	M1	MH	R1	11
A	M1	MH	R2	19
A	M1	MH	R3	27

A	M1	MV	R1	15
A	M1	MV	R2	23
A	M1	MV	R3	31

A	M2	MH	R1	35
A	M2	MH	R2	43
A	M2	MH	R3	51

A	M2	MV	R1	39
A	M2	MV	R2	47
A	M2	MV	R3	55

A	M31	MH	R1	59
A	M31	MH	R2	67
A	M31	MH	R3	75

A	M31	MV	R1	63
A	M31	MV	R2	71
A	M31	MV	R3	79

L 7

A	M32	MH	R1	11
A	M32	MH	R2	19
A	M32	MH	R3	27

A	M32	MV	R1	15
A	M32	MV	R2	23
A	M32	MV	R3	31

A	M4	MH	R1	35
A	M4	MH	R2	43
A	M4	MH	R3	51

A	M4	MV	R1	39
A	M4	MV	R2	47
A	M4	MV	R3	55

A	M51	MH	R1	59
A	M51	MH	R2	67
A	M51	MH	R3	75

A	M51	MV	R1	63
A	M51	MV	R2	71
A	M51	MV	R3	79

M 7

A	M52	MH	R1	11
A	M52	MH	R2	19
A	M52	MH	R3	27

A	M52	MV	R1	15
A	M52	MV	R2	23
A	M52	MV	R3	31

A	M61	MH	R1	35
A	M61	MH	R2	43
A	M61	MH	R3	51

A	M61	MV	R1	39
A	M61	MV	R2	47
A	M61	MV	R3	55

A	M62	MH	R1	59
A	M62	MH	R2	67
A	M62	MH	R3	75

A	M62	MV	R1	63
A	M62	MV	R2	71
A	M62	MV	R3	79

DESCEND

2 N 7

D	M1	MH	R1	11
D	M1	MH	R2	19
D	M1	MH	R3	27

D	M1	MV	R1	15
D	M1	MV	R2	23
D	M1	MV	R3	31

D	M2	MH	R1	35
D	M2	MH	R2	43
D	M2	MH	R3	51

D	M2	MV	R1	39
D	M2	MV	R2	47
D	M2	MV	R3	55

D	M31	MH	R1	59
D	M31	MH	R2	67
D	M31	MH	R3	75

D	M31	MV	R1	63
D	M31	MV	R2	71
D	M31	MV	R3	79

O 7

D	M32	MH	R1	11
D	M32	MH	R2	19
D	M32	MH	R3	27

D	M32	MV	R1	15
D	M32	MV	R2	23
D	M32	MV	R3	31

D	M4	MH	R1	35
D	M4	MH	R2	43
D	M4	MH	R3	51

D	M4	MV	R1	39
D	M4	MV	R2	47
D	M4	MV	R3	55

D	M51	MH	R1	59
D	M51	MH	R2	67
D	M51	MH	R3	75

D	M51	MV	R1	63
D	M51	MV	R2	71
D	M51	MV	R3	79

P 7

D	M52	MH	R1	11
D	M52	MH	R2	19
D	M52	MH	R3	27

D	M52	MV	R1	15
D	M52	MV	R2	23
D	M52	MV	R3	31

D	M61	MH	R1	35
D	M62	MH	R2	43
D	M61	MH	R3	51

D	M61	MV	R1	39
D	M61	MV	R2	47
D	M61	MV	R3	55

D	M62	MH	R1	59
D	M62	MH	R2	67
D	M62	MH	R3	75

D	M62	MV	R1	63
D	M62	MV	R2	71
D	M62	MV	R3	79

EMG PEAKS RELATED to GONIOMETER SIGNAL

⑥

I.D. ☐☐☐ 3 | 3 | 4 | LEG 1 | 5

410.

ASCEND

W 7

AM1	GV	R1	10
AM1	GV	R2	13
AM1	GV	R3	16
AM2	GV	R1	19
AM2	GV	R2	22
AM2	GV	R3	25
AM31	GV	R1	28
AM31	GV	R2	31
AM31	GV	R3	34
AM32	GV	R1	37
AM32	GV	R2	40
AM32	GV	R3	43
AM4	GV	R1	46
AM4	GV	R2	49
AM4	GV	R3	52
AM51	GV	R1	55
AM51	GV	R2	58
AM51	GV	R3	61
AM52	GV	R1	64
AM52	GV	R2	67
AM52	GV	R3	70
AM61	GV	R1	73
AM61	GV	R2	76
AM61	GV	R3	79

1

2

1

2

1

X 7

62.

AM62	GV	R1	10
AM62	GV	R2	13
AM62	GV	R3	16

DESCEND

2 Y

DM1	GV	R1	10
DM1	GV	R2	13
DM1	GV	R3	16
DM2	GV	R1	19
DM2	GV	R2	22
DM2	GV	R3	25
DM31	GV	R1	28
DM31	GV	R2	31
DM31	GV	R3	34
DM32	GV	R1	37
DM32	GV	R2	40
DM32	GV	R3	43
DM4	GV	R1	46
DM4	GV	R2	49
DM4	GV	R3	52
DM51	GV	R1	55
DM51	GV	R2	58
DM51	GV	R3	61
DM52	GV	R1	64
DM52	GV	R2	67
DM52	GV	R3	70
DM61	GV	R1	73
DM61	GV	R2	76
DM61	GV	R3	79

Z 7

DM62	GV	R1	10
DM62	GV	R2	13
DM62	GV	R3	16

SOME CODE EXPLANATIONS

MX = Maximum

CNR = Contraction

A = Ascending

D = Descending

SW = Swing

H = Horizontal (measurement, i.e., temporal)

T = Total

ST = Start and Stance

MXFL = Maximum flexion

MIFL = Minimum flexion

V = Vertical (measurement, i.e., amplitude)

Many permutations subsequently had to be formed to identify different variables, e.g., PCTM5IR = Peak expressed as a percentage, during descending, first peak of lateral hamstring, on the right

APPENDIX B

RELIABILITY

TEMPORAL DATA RELIABILITY

Four swing times and four total times (each limb for two cycles) had eight reliability coefficient calculations for each group (see Appendix A).

Normal Group - 2 reliability coefficient calculations were >.70
- 3 reliability coefficient calculations were >.80
- 3 reliability coefficient calculations were >.90

Amputee Group - 1 reliability coefficient calculation was >.80
- 7 reliability coefficient calculations were >.90

ELECTROGONIOMETRIC DATA RELIABILITY

Four measurements for each of the following variables was evaluated: start of swing, start of stance, maximum and minimum flexion and time. This means that there were 24 reliability coefficient calculations for each group (see Appendix A).

Normal Group - 6 reliability coefficient calculations were >.80
- 18 reliability coefficient calculations were >.90

Amputee Group - 2 reliability coefficient calculations were >.80
- 22 reliability coefficient calculations were >.90

ELECTROMYOGRAPHIC DATA RELIABILITY

In this section all but the temporal data is calculated from the measured data. This is exemplified by the peak position expressed as a percentage of the cycle or the peak amplitude being expressed as a percentage of the maximum elicited contraction. Because of this the following reliability calculations were made:

1) Maximum elicited contraction
2) Amplitude measurements
3) Temporal measurements

Maximum Elicited Contraction

These are vertical measurements. All repeated measurements for all muscles in both groups were between .96 and .99.

Amplitude Measurements

These are vertical measurements. Since nine muscle peaks were evaluated (see Appendix A) and both limbs in both cycles were considered, there were 36 reliability coefficient calculations made for each group:

Normal Group - 2 reliability coefficient calculations were >.74
- 2 reliability coefficient calculations were >.80
- 32 reliability coefficient calculations were >.90

Amputee Group - 3 reliability coefficient calculations were >.80
- 33 reliability coefficient calculations were >.90

Temporal Measurements

These are horizontal measurements. Following the above format:

Normal Group - 1 reliability coefficient calculations were >.79
- 7 reliability coefficient calculations were >.80
- 28 reliability coefficient calculations were >.90

Amputee Group - 1 reliability coefficient calculation was >.89
- 35 reliability coefficient calculations were >.90

EMG versus EGM RELIABILITY

Nine muscle peaks were evaluated for both limbs and both cycles (see Appendix A). Therefore 32 reliability coefficients were calculated for each group.

Normal Group - 4 reliability coefficient calculations were >.70
- 8 reliability coefficient calculations were >.80
- 24 reliability coefficient calculations were >.90

Amputee Group - 3 reliability coefficient calculations were >.80
- 33 reliability coefficient calculations were >.90

APPENDIX C

SAMPLE SIZE

It should be noted that the sample size has both ascending and descending values and the larger of the two rather than the smaller are recorded to give an overestimation rather than an underestimation of population difference detectable by the sample of 18 normals and 10 amputees.

TEMPORAL DATA SAMPLE SIZE

	Normal	Amputee
Cycle Time (Seconds)	.1	.08
Cadence (Steps/Minute)	3	4
Swing Percentage	3	1
Stance Percentage	2	3
Swing Time (Seconds)	.06	.46
Stance Time (Seconds)	.08	.37

ELECTROGONIOMETRIC DATA SAMPLE SIZE

	Normal	Amputee
Start of Swing (Degrees)	6	12
Start of Stance (Degrees)	7	9
Maximum Flexion (Degrees)	7	5
Position of Maximum Flexion %	2	2
Maximum Flexion Time (Seconds)	.03	.09
Minimum Flexion (Degrees)	5	7
Position of Minimum Flexion %	3	2
Minimum Flexion Time (Seconds)	.08	.49
Total Range of Movement (Degrees)	3	4

ELECTROMYOGRAPHIC DATA SAMPLE SIZE

Position of Peak as Percentage of Cycle	Normal	Amputee
M1	3	9
M2	3	11
M31	7	2
M32	1	2
M4	3	9
M51	22	35
M52	19	54
M61	23	37
M62	24	40

	Normal	Amputee
Difference Between First and Last	15	32
Last Peak Quadriceps	3	7
First Peak Quadriceps	14	49

Absolute Time (Seconds)

Time of Peaks	Normal	Amputee
Peak 1 (First Quadriceps)	.05	.35
Peak 4 (Last Quadriceps)	.08	.32
Peak 1-4	.28	.93
Peak 51	.35	.86
Peak 52	.34	.66
Peak 61	.43	.89
Peak 62	.47	.60

Amplitude Expressed as a Percentage of Maximum Contraction

	Normal	Amputee
M1	48	62
M2	34	61
M31	15	106
M32	19	31
M4	27	83
M51	16	104
M52	10	46
M61	10	21
M62	15	46

EMG versus EGM SAMPLE SIZE

The following are the population difference detable by the sample. All units in degrees.

	Normal	Amputee
M1	11	16
M2	6	13
M31	3	9
M32	2	1
M4	7	20
M51	27	26
M52	24	60
M61	28	26
M62	32	35

APPENDIX D

PUBLICATIONS AND PRESENTATIONS

Publications from thesis to date:

Tata, J.A., Quanbury, A.O., Steinke, T.G. and Grahame, R.E. (1978). A variable axis electrogoniometer for the measurement of single plane movement. J. Biomechanics 11: 421-425.

Tata, J.A. and Grahame, R.E. (1978). An electrogoniometric analysis of knee joint movement and an electromyographic analysis of the thigh muscles in ascending and descending stairs in normal subjects. Proc. Canad. Fed. Biol. Soc. 21: 90.

Tata, J.A. and Grahame, R.E. (1979). A temporal and electrogoniometric analysis of human knee joint movement during the stair cycle in normal subjects. Anat. Rec. 193: 701.

Paper Presentations:

1. Canadian Federation of Biological Societies Meeting, London, Ontario, 1978.
2. American Association of Anatomists Meeting, Miami, Florida, U.S.A., 1979