ELECTROGONIOMETRY
OF THE KNEE

BY

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A Thesis
Submitted to the Faculty of Graduate Studies
in Partial Fulfillment of the Requirements
for the Degree of

MASTER OF SCIENCE

Department of Physical Education and Recreation Studies
University of Manitoba
Winnipeg, Manitoba

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TABLE OF CONTENTS

Abstract....i
Acknowledgements....ii
List of tables....iii
CHAPTER 1 Introduction....1
Statement of the problem....3
Hypothesis....3
Operational definitions....3
Assumptions....3
Limitations....3
Delimitations....4
Significance....4
CHAPTER 2 Review of Literature....7
Gait....7
Electrogoniometers....10
Chondromalacia and Gait....16
Foot-Knee Interactions....17
Foot Orthotics....23
Musculature....27
CHAPTER 3 Methods and Procedures....33
Subjects....33
Instruments....33
Procedures....36
Statistical Analysis....38
Pilot Study Results....38
Main Study Procedures....41
Statistical Analysis....41
CHAPTER 4 Results....43
Main Study....43
CHAPTER 5 Discussion....46
Pilot Study....46
Main Study....47
Comparison to Literature Values....49
CHAPTER 6 Summary and Conclusion....54
Conclusions....56

Improvements/Recommendations 56
REFERENCES 58

APPENDIX 1....................64
Eligibility questionnaire
APPENDIX 2....................66
Diagram showing potentiometer configuration for determination of internal/external rotation
APPENDIX 3 68
Diagram explaining equation 1
APPENDIX 4....................70
Scattergrams and regression equations showing the linearity of the potentiometers
APPENDIX 5....................74
Diagram of electrogoniometer
APPENDIX 6....................76
Dimensions of the parts of the electrogoniometer
APPENDIX 7....................78
Wiring diagram
APPENDIX 8....................80
Excel™ spreadsheet for voltage conversions and degree value determination
APPENDIX 9....................82
Consent form
APPENDIX 10....................85
ANOVA tables comparing the 4 trials of the pilot study. All parameters are analyzed for both subjects.
APPENDIX 11....................92
Power analysis
APPENDIX 12....................94
Scattergrams and regression equations for pre and post-test linearity checks on the potentiometers
APPENDIX 13....................100
Stride (H2H) ANOVA tables
APPENDIX 14....................104
Stance (H2T) ANOVA tables

APPENDIX 15 108
Stride and stance time ANOVA tables
APPENDIX 16 111
Literature values of the three rotational parameters used in this study

ABSTRACT

The reliability of a newly designed electrogoniometer, capable of measuring the three rotational degrees of freedom of the knee, was tested in this study. The electrogoniometer was tested on 20 normal subjects while treadmill walking at a speed of 2 MPH. Each subject was tested three times with removal and re-fitting of the electrogoniometer occurring between trials 2 and 3. An analysis of variance (ANOVA) between trials determined the device to possess both mechanical and placement reliability for all three rotational parameters studied. The results also indicated that the device had a higher degree of reliability in the stance phase of gait compared to the whole stride. Finally, the results generated by the newly constructed electrogoniometer were consistent with accepted values in the literature.

ACKNOWLEDGEMENTS

I would like to thank my committee members for their speedy work in correcting and returning of the various copies of this thesis. I would like to give special thanks to Dr. Dean Kriellaars for his technical assistance, Robert Graham, P.Eng. for assistance with the data acquisition system and finally Rick Hall for his help with the computer program.

LIST OF TABLES

Table 1.....12
Control group's rotational values of the knee collected over an entire gait cycle. (Marans, Jackson, Glossop, & Young (1989))
Table 2.....12
Experimental group's rotational values of the knee collected over an entire gait cycle ACL unaffected knee (Marans, Jackson, Glossop, & Young (1989))
Table 3.....12
Epidemiological data for control and ACL unaffected groups. (Marans, Jackson, Glossop, & Young (1989))
Table 4.....13
Internal/external rotational values for normal knees collected during the stance phase only. (Czerniecki, Lippert, & Olerund (1988))
Table 5.....14
Control group's rotational values for the knee collected over an entire gait cycle. (Isacson & Brostrom (1988))
Table 6.....15
Various parameters of knee motion for normal subjects. (Chao, Laugman, Schneider, & Stauffer (1983))
Table 7.....39
Means and standard deviations of the three rotational parameters. Average of the 4 trials in the pilot study (n=2)
Table 8.....40
ANOVA results for subject KCARL
Table 9.....40
ANOVA results for subject DALSTE
Table 10.....44
The mean values of the three rotational parameters for both stride and stance phase in the main study (n=20)
Table 11.....45
The ANOVA p-values comparing the three trials of each rotational parameter in the main study. Both stride and stance phase shown (n=20)

ELECTROGONIOMETRY OF THE KNEE

CHAPTER 1

INTRODUCTION

A very common complaint of runners is knee pain (James, Bates, & Ostering, 1978), specifically in the area of the patella. This condition is often referred to as runner's knee and diagnosed as chondromalacia patellae. According to Ficat and Hungerford (1977) this diagnosis should only be made in cases of gross destruction of the articular cartilage on the undersurface of the patella. It is thought that less than 20% of people with pain about the patello-femoral joint actually have gross destruction (Percy, & Strother, 1985). A more accurate description of the runners' pain would be patello-femoral arthralgia (PFA). This term encompasses pain in or about the patello-femoral joint. Many feel this is the initial stage in the development of chondromalacia patellae, and if allowed to progress will proceed to degenerative lesions.

The exact cause of the pain associated with this condition is unknown since the cartilage itself is devoid of nerve fibers (Grana, & Kriegshauser, 1985). Three adjacent tissues though are highly innervated and could possibly be responsible for the pain. The first is the synovium, which could be irritated by by-products of cartilage degeneration. The second tissue implicated is the subchondral bone. As cartilage degeneration occurs, the energy absorbing ability of this tissue decreases. Subsequently, loads are passed on to the subchondral bone, resulting in increased intra-osseus pressure and pain (Grana, & Kriegshauser, 1985). Fulkerson (1989) also notes the pain can arise from the retinacula. In cases of abnormal tracking the medial or lateral retinaculum can be stretched and be the source of pain.

Regardless of from where the pain is coming, the degeneration of the articular cartilage is a result of excessive pressure (Ficat, & Hungerford, 1977). This excessive pressure is a result of abnormalities which alter the contact areas of the patella (pressure = force /area). The structures responsible for normal patellar tracking are numerous and consist of both static and dynamic elements. Some of the more common factors that predispose individuals to PFA are; increased Q-angle (Fox, 1975), quadriceps imbalances (Fox, 1975), and pronated feet (Jernick, & Heifitz, 1979).

James, Bates, and Ostering (1978) found that 58% of runners with lower extremity problems were (over) pronated: of these, 18% had knee problems. In Jernick's 1979 study of 19 female runners with PFA, they found a high correlation between foot pronation and PFA. This study showed that there was a link between the two, but failed to explain it. Others have attempted to explain the kinematics of a pronated lower limb and how this pronation would affect the knee (Beckman, 1980 ; Buchbinder, Napora, & Biggs, 1979 ; D'Amico, & Rubin, 1986 ; Tiberio, 1987).

Even though the mechanics are poorly understood a foot orthotic is often prescribed to runners with over-pronation. The orthotic's effectiveness in reducing the symptoms of PFA has been reported (James, et al., 1978) (Eggold, 1981). Still the effect of the orthotic on altering the mechanics of the knee so as to alter patello-femoral (P-F) tracking has not been demonstrated empirically. Speculation has been that the connection between foot pronation and PFA has to do with the transverse rotations of the tibia which may be altered during the abnormal pronation.

Statement of the problem

To design and construct an original tri-planer electrogoniometer that was foremost sensitive to the rotations of the tibia through its long axis, and to test the reliability of this device in a normal population .

Hypothesis

The design of the electrogoniometer would allow it to reliably measure the transverse rotations of the tibia, relative to the femur, in subjects from a normal population.

Sub-hypothesis

The constructed electrogoniometer would reliably output values for knee flexion/extension and abduction/adduction in a normal population.

Operational definitions

1) Internal/External rotation - rotation of bones through their long axes in relation to the weight bearing foot or relative to adjacent bones.

2) Normal population - a group of individuals who do not present with PFA and who have no history of leg or hip problems.

Assumptions

1) the changes in the rotations of the femur and tibia are measurable

2) subjects are able to adopt a consistent gait during testing (ie. effects of the treadmill, goniometer and foot switch application are negligible)

Limitations

1) most rotational values in the literature are from goniometers of various designs and may not express true values or be comparable with one another

2) there is inherent invalidity in an exoskeletal measuring device and therefore obtained values may not be comparable to those obtained from more invasive measures

3) the concept of a moving quadrilateral is only accurate if the length of the sides remain constant. (complete movement cannot be totally eliminated)

Delimitations

1) this study uses only one pre-determined speed of walking for all subjects

2) due to the design of the electrogoniometer, only the right legs of subjects were tested

Significance

The lack of information on the effect of an orthotic on the mechanics of the knee might be due to the lack of agreement on how over-pronation alone affects the knee. It is agreed that during over-pronation the tibia internally rotates more and for a longer period of time than normal (Carson, 1985 ; Lutter, 1978 ; Rothbart, & Estabrook, 1988). Disagreement occurs in regard to what the femur is doing at this time. Normal gait mechanics dictate that from heel strike (HS) to flat foot (FF) the foot pronates, the tibia internally rotates as does the femur and pelvis. From FF to toe off (TO) body weight shifts over the foot, the foot supinates and the segments externally rotate. It is generally accepted that an over-pronator has prolonged internal tibial rotation: internally rotating when it should be externally rotating with the other body segments. Some authors feel that the normal external rotations of the pelvis and femur will prevail and continue to externally rotate. This will cause incongruent rotations at the knee (tibial-femoral joint) (Beckman, 1980 ; D'Amico, & Rubin, 1986 ; James, Bates, & Ostering, 1978 ;

Rothbart, et al., 1988). The end result is abnormal patellar tracking. Other authors feel that compensatory internal femoral rotation will occur (Buchbinder, et al., 1979 ; Tiberio, 1987). This is thought to also result in abnormal tracking.

During normal kinematics the knee flexes with internal rotation of the tibia and extends with external rotation (Tiberio, 1987). Herzog-Franco (1987) felt that, as a compensatory movement, knee flexion will increase. Increased knee flexion will cause higher compression forces in the P-F joint which over time could cause PFA.

A few authors have noted that the quadriceps will be affected by the over-pronation (Beckman, 1980 ; D'Amico, et al., 1986 ; Herzog-Franco, 1987 ; Tiberio, 1987). This could be significant since the musculature is the major determinant of patellar tracking. Tiberio (1987) and Buchbinder (1979) speculated that the pull of these muscles as a group would be directed more laterally when accompanied by over-pronation. There exists an antagonistic relationship between the vastus medialis (VM) and vastus lateralis (VL) which controls the patella. Ineffectiveness of either one will cause the patella to track to the opposite side. Beckman (1980) theorized that the incongruent rotation occurring at the knee (between the tibia and femur) would effectively shorten the VM and decrease its effectiveness. The resultant imbalance between the two heads of the quadriceps would allow lateral tracking of the patella. If Herzog-Franco (1987) is correct, and the pronation causes an increase in flexion at the knee, greater quadriceps activity will be required to offset the increased flexion moment.

The first step in determining what is occurring at the knee is to have the ability to accurately measure the leg rotations of two populations, normals and PFA subjects. A data collection system and an electrogoniometer are needed which would allow researchers the ability to collect the required data. The design of the electrogoniometer should be such that it is lightweight and easy to use. Once this equipment is obtained, studies could be conducted which would help explain the relationship between over-pronation, rotations of the leg, and PFA.

CHAPTER 2
REVIEW OF LITERATURE

Gait

In order to understand pathologic gait one must first understand what is normal. In most gait studies the authors have focused on walking. The stance phase, which is of major concern in this paper, consists of 3 phases; contact, midstance and propulsion (Tiberio, 1987). The contact phase occurs from HS to FF, while FF to heel raise (HR) is considered midstance. Propulsion is the final phase of stance and occurs from HR to TO.

As the foot strikes the ground, it is in a supinated position (or a less pronated position) and then starts to pronate. Normal pronation (abduction, eversion and dorsi-flexion) occurs up to 25% of the stance phase (Santopietro, 1988) or approximately throughout the contact phase to FF. At this point (≈ 25% stance) the body weight passes over the foot and the foot starts to re-supinate. Supination continues throughout midstance and propulsion to TO.

As pronation occurs, the boney configuration of the talus in the ankle mortise acts like a torque converter and causes an obligatory internal rotation of the leg (tibia and fibula) (James, et al., 1978). Likewise external rotation of the leg accompanies supination. Also during the gait cycle there are coordinated movements between leg rotation and the flexion and extension occurring at the knee. According to Tiberio (1987) the knee is close to full extension at heel strike, then flexes 15°-20° during the contact phase. During midstance the knee then extends, and finally from HR to TO the knee again flexes in order to prepare for the swing through. When comparing the motion at the knee (sagittal plane) with the rotations in the leg (transverse plane), we see that, in normal gait during initial foot contact, the tibia internally rotates as the knee flexes and externally rotates as the knee extends through midstance (Tiberio, 1987). Tiberio (1987) states that this coupling is obligatory and is often called automatic rotation. This is different from the rotations that can occur independent of flexion and extension when the knee is flexed more than 20°.

When a forefoot varus is present the normal actions of pronation and supination are altered. A forefoot varus is said to be present when the rearfoot (calcaneus) is in neutral and the forefoot is elevated on the medial side in the non-weight bearing position. The foot contacts the ground in the normal fashion but then over pronates due to the deformity in the forefoot. Over-pronation means that pronation is occurring for more than 25% of the stance phase (Santopietro, 1988). The foot will pronate normally to FF, but then continue to pronate as the forefoot attempts to make contact with the ground in order to push off.

If a rearfoot deformity is present (such as calcaneal varus), the over-pronation will consist of a greater amount of pronation throughout the contact phase. A rigid flatfoot is a deformity that causes the foot to contact the ground in a pronated position and remain pronated until the foot leaves the ground. These abnormal pronatory movements also bring with them abnormal tibial rotations.

Levens, Berkeley, Inman, and Blosser (1948) studied the transverse rotations of the lower limb in normal individuals. Using high speed cameras and skeletal pins, they measured the rotational ranges and relative magnitudes of the pelvis, femur and tibia. They found that in normal gait the three segments were synchronized, all internally and externally rotating together, although they all did not rotate equivalent amounts. From the time the foot left the ground (swing phase) until it reached FF, the entire lower limb was rotating internally, with the distal segments rotating more than the proximal ones. From FF on, the segments then externally rotated, again with the distal segments rotating more than the proximal ones.

The fact that some segments were rotating more than others meant that there were relative differences between the segments. Levens et al. (1948) found that from approximately HC to FF the tibia rotated internally 3.5° relative to the femur. During midstance there was first a relative outward rotation of the tibia on the femur of 1.5°, then a small internal rotation of 0.5 degrees. From HR to TO there was another relative external rotation of the tibia on the femur of 3.5 degrees.

When Levens et al. (1948) considered the rotations between the femur and the pelvis, they found there was a 7° internal rotation of the femur on the pelvis from approximately HS to FF. Then from FF to TO there was a relative external rotation of 6.5 degrees. These results are based on averages for 12 apparently normal individuals.

Kettelkamp et al. (1970) also measured the transverse rotations of the lower limb in normal gait. These researchers used an electrogoniometer which measured the tibial rotations relative to the femur. In their 22 subjects (44 knees), they found a large degree of variation in the measured movements during the stance phase. With subjects walking at a self-selected speed, most subjects displayed a sequence of motions throughout contact phase and midstance which supported that described by Levens et al. (1948). However, during the final phase from HR to TO the majority of people showed relative internal rotation of the femur on the tibia or at most minimal external rotation. This contradicts other descriptions of normal joint mechanics. These researchers did not state if they controlled for abnormal biomechanics of the foot during the selection of their subjects.

In a more precise test of normal knee joint motion during walking, Lafortune and Cavanagh (1989) used an invasive technique utilizing intracortical pins inserted into the femur, tibia, and patella. Target clusters were then attached to the pins and mathematical relationships were used to determine relative spatial arrangements of the pins at various points in the gait cycle. They found that with subjects walking at a speed of 1.5 m/sec, the tibia internally rotated approximately 7° immediately after heel strike and remained in this position until the knee reached full extension prior to toe-off. Around the beginning of knee flexion, prior to toe-off, the tibia internally rotated an additional 7°. External rotation then followed throughout the swing phase to just before heel strike. This study found no external rotation occurring during the stance phase. Furthermore, no abduction or adduction about the knee occurred in the stance phase which had been found in other studies.

This study was also able to determine the motion of the patella. At heel strike the patella moved laterally across the femur (~ 6mm), and it continued this motion for approximately 50% of the stance phase, or just prior to full extension. After its lateral shift the patella moved medially slightly, then continued its lateral motion throughout toe-off into the swing phase. This meant that as the tibia was internally rotating for the second time, the patella was shifting laterally.

These researchers concluded that patellar motion was less dependent on tibial motion, and more dependent on femoral and patellar restraints.

Electrogoniometers

A number of studies have used electrogoniometers of various designs to study abnormal knee motion secondary to a mechanical deficiency (most often a cruciate deficiency). Most of these studies compare their experimental population to a population designated as normal. These studies serve as a source of data from which to determine the "normal" pattern of knee motion obtained with an exoskeletal measuring device.

In a study to determine the motion of anterior cruciate ligament (ACL) deficient knees, Marans, Jackson, Glossop, and Young (1989) designed a spatial linkage device (electrogoniometer) which enabled the evaluation of the three rotational and three translational movements occurring at the knee joint during level walking. This study used a group of 30 normal individuals as a control group. This control group contained both males and females and these subjects were tested while walking at a self-selected speed on the floor. The rotational values are listed in tables 1 and 2. Since it was important for the control group to be comparable to the experimental group, the two groups were matched on the basis of age, sex, height, thigh circumference, calf circumference, and cadence (see table 3).

Marans et al. (1989) found no statistically significant difference between their control group and the ACL deficient group with respect to rotational values. Furthermore, no significance was found between the knees of the same experimental individual (ie. good knee and bad knee). There was however, a significant difference found in one of the translational parameters, namely anterior/posterior translation. In this case the ACL deficient knees possessed greater ranges of this movement.

Table 1
Control group's rotational values of the knee collected over an entire gait cycle (N=30) (Marans, et al., 1989)

ROTATION	mean (standard deviation)
flexion/extension	48.6° (6.0)
axial rotation	9.2° (3.7)
angulation	3.9° (1.5)

Table 2
Experimental group's rotational values of the knee collected over an entire gait cycle (N=20 ACL unaffected knees) (Marans, et al., 1989)

ROTATION	mean (standard deviation)
flexion/extension	48.2° (6.6)
axial rotation	8.9° (4.1)
angulation	3.3° (1.3)

Table 3
Epidemiological data of control and ACL unaffected groups (Marans, et al., 1989)

	Control Group mean (standard deviation)	ACL Group mean (standard deviation)
Age (yrs)	23.4 (4.2)	27.9 (6.7)
Sex (M/F)	2/1	3/1
Height (cm)	173.7 (8.3)	176.0 (10.5)
Thigh cicum (cm)	44.6 (2.9)	46.4 (3.7)
Calf circum (cm)	32.9 (2.9)	33.9 (3.0)
Cadence (steps/ sec)	1.10 (0.08)	1.13 (0.07)

Czerniecki, Lippert, and Olerund (1988) studied the rotational parameters of ACL deficient knees compared to normal knees. The researchers used a control group which consisted of 9 normal individuals. For this study the researchers used the MERU tri-axial electrogoniometer. Data was collected for both right and left legs at 3 different speeds of treadmill ambulation. A significant increase in the internal/external rotation was seen with increasing speeds in the entire population. There was however, no significant difference between the normal and experimental groups. The data for normal individuals is shown in table 4.

Table 4
Internal/external rotational values for normal knees collected during the stance phase only (Czerniecki, et al., 1988)

speed (m/min)	normal right knee rotation (degrees)	normal left knee rotation (degrees)
84	11.3° (4.0)	9.7° (3.0)
132	12.7° (4.8)	12.9° (4.2)
156	14.8° (6.4)	13.3° (4.3)

mean (standard deviation)

In the same year Isacson and Brostrom (1988) performed an electrogoniometeric experiment to examine gait abnormalities in patients with rheumatoid arthritis. Here again control groups were used for comparison. The first control group consisted of 11 females walking at a self-selected speed. The second control group consisted of 6 females walking at a pre-determined slow speed (0.6 m/sec), both groups walked on a treadmill. On average the normal subjects in the first control group ambulated at 1.2

(0.1) m/sec., which was significantly faster than the test population. The goniometer used was a modified C.A.R.S.-U.B.C. goniometer which simultaneously measured 3-D motion about the hip, ankle, and knee. The values obtained from the normals, for the knee joint, are listed in table 5. The results indicated an altered amount of abduction/adduction occurring about the knee in the rheumatoid gait. Other aberrations were noted at the other two joints studied.

Table 5
Control group's rotational values for the knee collected over an entire gait cycle (Isacson, & Brostrom, 1988)

	normal subjects (N=11) self-selected speed	normal subjects (N=6) pre-determined speed
ROTATION	9° (2)	7° (4)
AB/ADDUCTION	8° (2)	7° (3)
FLEXION/EXTENSION	58° (3)	55° (5)

mean (standard deviation)

Chao, Laugman, Schneider, and Stauffer (1983) also used goniometers to collect some of their data on knee joint motion during level walkway walking. Their study was designed to collect data from a large number of subjects, both male and female, with respect to the various parameters of walking, ground reaction forces, and knee joint motion. Subjects were allowed to walk at their own speed and were tested in their own shoes. The speed of walking (m/min.) for the men as a group was 74.4 (15.1) and for the women it was 67.5 (11.2). The type of electrogoniometer used to collect this data was not stated, but normative values for 110 individuals was given. This data was subdivided into total motion during a cycle and total motion in the stance phase. Furthermore, subjects were separated into male and female groups. Both the sex groupings were again divided into two age categories. Group 1 contained individuals in the age range 32-85, while group 2 spanned 19-32.

Table 6
Various parameters of knee motion for normal subjects
(Chao, et al., 1983)

	MEN			WOMEN		
Tibio-femoral move-ments	Group 1 n=32	Group 2 n=21	Total	Group 1 n=37	Group 2 n=20	Total
flexion at heel strike	1°(4)	7°(4)	2°(6)	0°(5)	4°(6)	1°(5)
total sagittal motion	72°(6)	68°(8)	71°(7)	66°(9)	70°(8)	68°(8)
total stance ab/adduction	7°(2)	6°(2)	7°(2)	7°(2)	6°(2)	7°(2)
total frontal motion	12°(4)	11°(3)	12°(3)	10°(4)	9°(2)	10°(3)
total stance motion	9°(3)	11°(3)	10°(3)	10°(3)	9°(2)	9°(3)
total transverse motion	14°(4)	14°(3)	14°(4)	14°(4)	13°(3)	13°(4)

mean (standard deviation)

It is interesting to note that there was very little variation across age and sex with respect to the transverse rotations.

Chondromalacia and Gait

It should be noted again that, unless evidence of gross destruction of the patellar cartilage is present (and identified), the term PFA is more appropriate than chondromalacia patellae (CMP), although many studies still use this term.

It is understood that in cases of CMP due to patellar tracking problems, the relationship between the patella and the femur is distorted. In 1979, Sikorski attempted to study the rotations of the femur in patients suffering from CMP symptoms. Realizing that the relationship between the patella and the femur had previously been studied only in the non-weight bearing position, he developed a method by which the weight bearing condition was simulated and the position of the femur inferred from radiographs. He found that, in the simulated weight bearing position, the femur of control subjects rotated internally (medially). Although flexion angles used in this study were greater than in Levens et al. (1948), both report similar findings. Levens et al. (1948) found that the femur and other segments rotated internally from minimal weight bearing to full weight bearing (ie. from HS to FF). This though, did not happen in patients with CMP symptoms. With the onset of muscular activity the femur in these subjects externally rotated. This fact supports the idea that persons with CMP have altered femoral rotations and possibly abnormal knee mechanics.

More recently Dillon, Updyke, and Allen (1983) compared the gait patterns of patients with CMP symptoms with those of controls. The results showed significant differences between the two groups. Not only did the CMP group have less flexion in the stance phase, but more interestingly, femoral rotations were different. The only statistically significant difference in femoral rotations occurred during the swing phase. Although not statistically significant, they also noted marked differences in the stance phase. Again these studies did not control for over-pronation in their subjects. These two studies show that people with PFA have altered rotations in their lower segments. It is possible that the altered rotations are a protective response to patello-femoral pain, but it is equally possible that the PFA is a result of the altered rotations.

Foot-Knee Interactions

As stated, over-pronation is pronation that occurs at an inappropriate time or to a greater degree than normal, and causes larger and prolonged internal tibial rotations. The question is now, how does this alter the mechanics of the knee so as to disrupt the P-F joint? Few studies to date have attempted to answer this question, although many authors have speculated as to the events occurring at the knee in an over-pronating individual and how this would affect the position of the patella. In general, two schools of thought exist: those that believe congruent rotation will occur between the tibia and femur (Buchbinder, et al., 1979 ; Ramig, Shadle, Watkins, Cavolo, & Kreutzberg, 1977 ; Tiberio, 1987 ; Williams, 1977), and those that believe incongruent rotation will be the result of over-pronation (Beckman, 1980 ; D'Amico, et al., 1986 ; James, et al., 1978 ; Rothbart, & Estabrook, 1988).

James (1978) recognized the increased and prolonged internal tibial rotation that accompanies over-pronation and felt that this abnormal transverse rotation would have to be absorbed in the knee. This in turn, he felt, would disrupt the normal tibial-femoral relationship and was probably the cause of the high incidence of knee injuries in runners.

Beckman's (1980) ideas agreed with those of James' (1978), in that they both believed the femur would externally rotate because of the force of body weight. He added that the resultant torque could be very harmful to knees with decreased joint integrity. Beckman (1980) further speculated that this motion at the knee would affect the surrounding musculature. He theorized, " As the tibia rotates internally it pulls the patella medially altering the force vectors of the muscles involved"(p.52). The VM was thought to be functionally weakened, especially its medial pull vector, which controls the patella.

In an article on foot orthoses and the Q-angle, D'Amico (1986) stated that with over-pronation, " the femur rotates with a greater excursion than the tibia, causing the patella to move plantarly and medially. Therefore there is a concomitant increase in the quadriceps angle accompanying pronation" (p.339). Even though the authors attempted to relate the rotations to the actions of the patella, there was a discrepancy between their mechanics and those defined by Levens, et al. (1948). They previously stated that the distal segments rotate more than the proximal ones, with the tibia rotating internally 3.5° relative to the femur. Therefore, for the femur to internally rotate more than the tibia, it would have to rotate more than this amount. Furthermore, in the over-pronator, the tibia is internally rotating more than usual. At the same time, the femur rotates internally 7° relative to the pelvis. If the femur was to rotate more than tibia, large amounts of rotation (internal) would have to be occurring at the hip when the mechanics dictated by Levens et al. (1948) say external rotation should be occurring. It is possible that the mechanics discussed by D'Amico (1986), if present during running, would create greater problems at the hip that would overshadow those at the knee.

Rothbart and Estabrook (1988) speculated that the patello-femoral problems caused by over-pronation are occurring because the tibia is rotating faster than the femur. This would probably be occurring between HC and FF. In previous models the incongruent rotation was thought to occur later in the stance phase. Rothbart and Estabrook (1988) felt that asynchronous motion of more than 4°- 6° was responsible for an obliquely tracking patella. The patella was thought to track obliquely across the femoral condyles toward the tibial tubercle and back again, eventually eroding the undersurface. Presumably it is this excessive normal oblique tracking that is responsible for the cartilage destruction.

Since a pronator with a forefoot varus is thought to pronate normally in the initial contact phase and then deviate in midstance, this model may be addressing a calcaneal varus deformity.

Buchbinder's (1970) work supported the theory of <u>congruent</u> rotation occurring at the knee during the over-pronated gait cycle. He theorized that over-pronation caused both the tibia and the femur to internally rotate when they should be externally rotating. This in turn would cause the quadriceps to exert an abnormal pull on the patella. "Since with prolonged pronation both the origin and insertion of the quadriceps are located lateral to the patella, contraction of the quadriceps tend to pull the patella in a lateral direction" (p.160). By having both the tibia and the femur internally rotated, mechanics of the hip would be upset, as there should be a relative external rotation of the femur to the pelvis. Also, as Tiberio (1987) pointed out earlier, relative internal rotation of the tibia on the femur <u>must</u> be present for flexion to occur at the knee during the first 15°-20° (automatic rotation).

Williams (1977) in an earlier article wrote," ...there is no way in which internal rotation of the leg as a whole can influence the patellofemoral congruity in flexion unless the knee joint is itself deranged, for example in medial rotary instability. The argument therefore that excessive pronation of the foot produces prolonged internal rotation of the leg and forces the patella laterally out of the patellofemoral groove of the femur is not tenable in the face of objective biomechanical analysis"(p.11).

Ramig et al. (1980) also stated that over-pronation will cause prolonged internal rotation of the entire extremity. They reported that this condition would force the patella laterally out of the groove. As Tiberio (1987) pointed out, if the femur is also remaining internally rotated the quadriceps alignment is straightened out and would seem not to cause lateral patellar tracking.

Tiberio (1987) put forth a biomechanical model to explain abnormal patellar tracking during over-pronation. He speculated that; at the beginning of midstance knee flexion should be over and the knee should begin to extend. For this extension to occur, there must be relative external rotation of the tibia on the femur (automatic rotation). Since with over-pronation the tibia is internally rotating at this point there is a biomechanical "dilemma" occurring at the tibio-femoral joint. Tiberio (1987) felt that the body would compensate by internally rotating the femur so as to develop the relative external tibial rotation and allow extension to occur. This would alter the normal patello-femoral mechanics. With internal rotation of the femur, the patella, relatively speaking, laterally tracks in the femoral groove and is compressed against the lateral femoral facet during extension. This model adds to the congruent rotation theory by considering differential motion of the segments which are both internally rotated. This model still has the same problem as that of D'Amico and Rubin's (1986) model, that is, motion between the femur and pelvis. With the internal femoral rotation there would have to be very large abnormal femoral rotations relative to the pelvis. During midstance, the opposite leg is swinging through, with the associated external pelvic rotation. The lateral patellar tracking part of this model is supported by Olerud and Berg (1984), who found that with internal rotation of the leg, the Q-angle increases because the pelvis (origin of the rectus femoris) is excluded from the rotations. It is still questionable whether running could occur with these mechanics occurring at the hip.

The aspect of pelvic rotation is what Santopietro (1988) addresses most specifically in his model of incongruent rotation. "As long as the body is moving forward and continues to do so, the internal and external pelvic rotations will prevail whether or not there is pronation or supination"(p.568). Thus, when pronation is prolonged, torsional stress is created because the pelvis and femoral segments are externally rotating and the tibia is locked in an internally rotated position.

The patella, because of its attachments, may ride on the lateral ridge of the femoral condyle (Santopietro, 1988). This statement is confusing because, with the tibia internally rotating the tibial tubercle would be moved medially and with the femur externally rotating, the medial condyle would be presented to the patella, not the lateral one. Although, with muscle contraction of the quadriceps, the patella would be pulled laterally, if the idea of an ineffective vastus medialis is accepted in this situation.

The idea of torsional stress developing in the knee is mentioned by many authors (Bates, Ostering, Manson, & James, 1979 ; Eggold, 1981 ; Santopietro, 1988 ; Williams, 1977). The occurrence of this is supported by the work of Coplan (1989) who studied the rotational laxity of knees in persons with mild over-pronation (calcaneal angle of >2°). Coplan (1989) measured the total range of passive transverse rotary movement of the knee of 15 subjects using the Cybex II™ isokinetic dynamometer. The knees were tested at three angles; 90°, 15°, and 5° of flexion. Even though these subjects were only mild pronators, she found significant differences at 5° of knee flexion. The pronating subjects had greater rotational laxity than did the control subjects. This study would seem to indicate that torsional stress is being generated by over-pronation and is being absorbed in the soft tissue about the knee. This could be the "derangement" that Williams (1977) made reference to earlier, which he felt was necessary for altered P-F mechanics.

The relationship of over-pronation and knee mechanics is referred to in a review paper by Herzog-Franco (1987). He felt that during the initial phase of gait (contact phase) the over-pronation would cause the knee to flex sooner than normal. This would in turn abnormally increase the stress in the quadriceps. This complies with normal mechanics (automatic rotation) of flexion requiring internal tibial rotation in the first 15°-20°. If over-pronation causes flexion to be prolonged, then there would be an increase in the patello-femoral joint reaction force (PFJRF) applied to the joint and could lead to PFA. The type of abnormal pronation referred to here seems to be either a rigid flat foot or calcaneal varus because, as stated earlier, a forefoot varus deformity produces a deviation during midstance. Not a lot of deviation occurs in the contact phase. Although, Santopietro (1988) felt that both rigid flat foot and forefoot varus would cause the knee to flex to a greater degree. This too would cause an increase in the PFJRF because with increased flexion there is increased quadriceps activity (Elliott, & Blanksby, 1979).

While describing how orthotics work, Williams (1977) and Beckman (1980) may have inferred support for the hypothesis of increased knee flexion in the genesis of PFA. Williams (1977) speculated that the foot orthotic works by producing a more complete extension of the knee through a sensory bio-feedback mechanism. Beckman (1980) felt that by keeping the foot in a more supinated position the tibia is "kicked back" and the knee consequently extends which decreases the PF compression force.

As one can see, most of the literature in this area is both speculative and contradictory. The fact that most runners with PFA developed this condition gradually may suggest that mechanical deviations are slight. This same fact may also support the idea of normal pelvic and femoral rotations with abnormal tibial rotations, and the resultant incongruent rotation at the knee. Congruent rotation, on the other hand, would seem to produce large deviations at the hip and would probably not allow normal running to occur.

Foot Orthotics

The general feeling regarding foot orthotics is that they are effective in treating runners with knee problems (Bates, et al., 1979 ; Donatelli, 1987 ; Subotnick, 1980), but the question of why they work still remains. According to Subotnick (1980) an orthosis works through two mechanisms a) biomechanical balancing of the foot encouraging re-supination and neutral position at the middle of midstance and b) through a bio-feedback mechanism. Presumably the former mechanism will synchronize the rotations of the segments as in a normal individual. The bio-feedback was unexplained by Subotnick (1980) but Williams (1977) earlier explained that, "Insoles invariably alter the sensory feedback and as a result may lead to significant changes of gait (as anyone who has walked any distance with a pebble in his shoe will be ready to attest !). Such a gait change can be all that is necessary to overcome a biomechanical problem proximally in the leg"(p.51).

Most articles dealing with foot orthotics have dealt with their effect on altering the various parameters of pronation in or about the foot. For example Bates et al. (1979) used a small sample size (N=6) of runners who he had successfully treated with orthotics for one year. The reasons why the orthotics were prescribed was not mentioned. When considering differences between a shoe and an orthotic application, the only significant differences found were in the time to maximum ankle dorsi-flexion and the angle of the posterior shank at maximum pronation. In both cases the orthotic increased the values.

Rodgers and Leveau (1982) studied the effect of orthotics in a more "real world setting". Unlike the study by Bates et al.(1979) who used a treadmill and a standardized test shoe, Rodgers and Leveau (1982) filmed their subjects running on a track in their own shoes. Again these researchers used subjects who had previously been wearing orthotics for some period of time. Also in this experiment, " No attempt was made to select subjects who were excessive pronators, although the conditions which necessitated the runners' use of FOD (foot orthotic devices) were related to excessive pronation"(p.89). This, they felt, also added to the external validity of this experiment. The reasons given for using the orthotics ranged from knee pain (41.1%) to shin splints (10.3%). The results showed non-significant differences in the effectiveness of orthotics between right and left legs, and significant differences with the use of orthotics in maximal angle of pronation and percentage of support time in pronation (in left foot only). The orthotics decreased both parameters. Due to the variability of the data, these researchers concluded that the effectiveness of the orthotics was questionable.

The effectiveness of orthotics in relieving symptoms of lower leg problems was explored by Eggold's 1981 survey. Of the 146 respondents, ~74% obtained at least 80% relief with the use of orthotics, ~40% reported 100% relief. In this group of runners ~40% reported knee pain as the major complaint and this accounted for the largest single complaint. Eggold (1981) felt the orthotic would allow synchronized external rotation of the tibia and the femur, and decrease the torque that over-pronation caused at the knee. He also felt that the orthotic would help bring the knee out of its adducted position. Santopietro (1988) also believed that over-pronation would cause the knee to be medially displaced to a greater degree in the frontal plane . He felt this adduction would accompany the incongruent rotations of the limb segments.

Taunton, Clement, Smart, Wiley, and McNicol (1987) measured a number of parameters of both knee and ankle/foot motions in runners with compensatory over-pronation in an attempt to study the effect of a foot orthotic. A treadmill and a C.A.R.S.-U.B.C. electrogoniometer positioned at the knee and ankle were used. They found that, with the orthotic, there was no difference in the valgus displacement of the knee. It must be kept in mind that the reasons for these runners having the orthotics prescribed was not given. The runners may not have had any knee problems at all. The stress accompanying over-pronation in these runners may have been concentrated elsewhere. Also, these runners had been wearing their orthotics for some time before testing was done (ie. were now asymptomatic). It is not known if an orthotic has a lasting effect on the limb, and if it does, it is not known how long it lasts before the previous symptom-producing biomechanics manifest themselves again.

Taunton et al. (1987) found no significant changes in support phase knee flexion or knee internal rotation with the orthotic application as compared with a regular running shoe. Also no significant alterations were found between maximal knee internal rotation (indicative of tibial rotation) and any of the 3 components of pronation (abduction, eversion or dorsi-flexion) or maximal knee flexion and the 3 components of pronation.

"Implying that the temporal relationships which exist between knee and ankle parameters are not significantly altered by CROD's (corrective running orthotic devices)"(p.114). The orthotic did significantly decrease the period of knee internal rotation ($p>0.08$) in these subjects.

The finding of no change in support phase knee flexion would tend to detract from the belief that the PFA is occurring due to over-pronation causing increased knee flexion (Herzog-Franco, 1987 ; Santopietro, 1988). If this were true, relief of the symptoms should be accompanied by a reduction in knee flexion. Again however, the reasons for prescription of the orthotics were not given. This is important to know this because not everyone that has over-pronation develops symptoms. Likewise people with similar pronation may develop different symptoms in different areas of the lower limb. The stress of over-pronation will manifest itself in different areas depending on a number of individual factors. For example, over-pronators who develop knee problems may be predisposed to this particular condition due to laxity in their knees.

Adduction of the knee, mentioned earlier, would tend to increase the Q-angle by increasing the physiologic valgus of the knee. D'Amico and Rubin (1986), realizing the importance of the Q-angle on patellar tracking, wished to explore the effect of a foot orthotic on the Q-angle. They used a static test and found that standing on the orthotic decreased the Q-angle on average 6° (N=21). No mention was made as to the reason for the orthotic being prescribed. These authors believed incongruent rotation was occurring at the knee as a result of over-pronation and that the orthotic would alter this relationship. More specifically, as mentioned before, they believed the femur rotated internally more than the tibia, as was mentioned before. These results could be explained by Olerund and Berg (1984). They stated that with pronation the entire limb internally rotates along with the patella and the tibial tubercle, but excluding the pelvis. The origin of the rectus femoris is relatively lateralized and this increases the Q-angle. Assuming the subjects in the previous experiment were over-pronators, the application of the orthotic reduced the tibial and femoral rotations, thus effectively decreasing the Q-angle.

D'Amico and Rubin (1986) also theorized that an increase in the Q-angle would affect patellar tracking, "The line of pull of vastus lateralis is shorter with an increased Q-angle, and as a result, the muscle develops in a contracted state. Correspondingly, the vastus medialis is more prone to fatigue in overuse situations" (p.338). This idea of over-pronation affecting muscle function is not a new one. Beckman (1980) felt an imbalance would occur in this situation due to a functional weakening of the medial pull of the vastus medialis. He states, "As the tibia rotates internally, it pulls the patella medially, altering the force vectors of the muscles involved. This has an effect of functionally weakening the medial pull of the vastus medialis muscle (and increasing its vertical pull)" (p.52).

Musculature

The musculature controlling the tracking of the patella, extension of the knee and controlled flexion of the knee is the quadriceps femoris muscle group. This group consists of 4 muscles; vastus lateralis (VL), vastus medialis (VM), vastus intermedius and rectus femoris. Of most concern in this paper are the first two components, the VL and VM.

The VL is the largest component of the quadriceps. It originates from the inter-trochanteric line, greater tuberosity, proximal half of the lateral linea aspera and is inserted into the lateral border of the patella, quadriceps tendon and the tendinous expansion to the capsule (Clemente, 1985). Fibers on average run 30°- 40° to the long axis of the femur (Jacobson, & Flandry, 1989). The VM on the other hand is much smaller. It arises from the lower half of the inter-trochanteric line, medial lip of the linea aspera, medial supra-condylar line and tendons of the adductors longus and magnus and medial intermuscular septum (Clemente, 1985). The VM is inserted into the extensor aponeurosis and the medial patella (Solcum, & Larson, 1968). According to Slocum and Larson (1968) this extensor aponeurosis inserts into the antero-medial aspect of the upper end of the proximal tibia through its capsular and deep fascial attachments, and constitutes the anatomical reason for the VM's ability to resist external rotation of the tibia through the first 60° of flexion.

The VM is said to have two components characterized by an abrupt change in the direction of its fibers (Brunet, & Steward, 1989). The fibers of the vastus medialis longus (VML) are at ~50° to the long axis of the femur while the fibers of the vastus medialis oblique (VMO) are more transversely oriented and are at ~65° (Jacobson, & Flandry, 1989).

The most often described function of the VL and VM is knee extension. Individually both these muscles are able to extend the knee: in fact all the components of the quadriceps are able to produce knee extension except the VMO (Lieb, & Perry, 1968). The function of this latter muscle is patellar alignment in the last 30° of extension (Bose, Kanagasunther, & Osman, 1980). The pull of this muscle resists the tendency of the patella to track laterally due to the pull of the VL and the physiologic valgus of the leg. In the last 30° its function becomes increasingly important because the effect of static (bony and ligamentous) constraints against lateral tracking are reduced (Bose, et al., 1980).

By HS the quadriceps are already active and continue to be active until just after peak knee flexion (around FF) (McClay, Lake, & Cavanagh, 1990). Mann and Hagy (1980) have found that the duration of the quadriceps activity as a percentage of stance phase increases with increasing velocity. They stated that in walking the quadriceps are active for ~15% of the stance phase, in running ~50% and in sprinting ~80%. The principle function of these muscles in the stance phase is to control the descent of the body's center of gravity while the knee is flexing (McClay, et al., 1990). This means that during this time the muscles are eccentrically contracting.

Most electromyographic (EMG) studies involving the VM use the VMO muscle because of its prominence in the thigh. These studies generally have found that in normal subjects there exists an antagonistic relationship between the VM and the VL that allows for normal patellar tracking (Elliott, & Blanksby, 1979 ; MacIntyre, & Roberson, 1987 ; Mariani, & Caruso, 1979). Elliott and Blanksby (1979) recorded the EMG activity of 10 female runners on a treadmill. The electrical activity, recorded in average IEMG units (integrated EMG), of the VM and VL were similar (similar waveforms), with the VM having higher readings for both test velocities. In all cases there was a significant increase in both muscles as the speed of running increased. Elliott & Blanksby (1979) felt this was related to the increased flexion occurring at the knee with the higher speeds.

McIntyre and Robinson (1987) found similar results, again using normal runners on a treadmill. Over the stance phase, both muscles showed similar curves (linear envelope EMG) with the VL having a slightly larger amplitude during the contact phase. These differences though were not significant. This was the "grand ensemble average" of 10 samples of 11 subjects. The curves of these two muscles were the most similar of the knee muscles tested.

Reynolds et al. (1983) compared the EMG activity in the VL to that in the VMO during the last 30° of extension in a weight bearing position. These authors reported the activity as a percentage of the activity during a maximal isometric contraction. The results of the mean differences in the normalized EMG activity of the VMO and the VL were compared. They found no significant differences in the activity levels of these muscles in this range of extension. In this experiment the VMO had a slightly higher mean (% max), 5.76 ± 3.84 compared to 4.72 ± 3.44.

EMG and patellar subluxation was studied by Mariani and Caruso (1979). The EMG activity of patients suffering from patellar subluxation were compared before and after an operation to realign the tibial tuberosity (ie. the insertion of the quadriceps) medially. This operation decreases the Q-angle. These researchers found that before the operation there was an electromyographic imbalance between the VM and VL. This difference occurred throughout the entire range of 0°- 90°, but was most obvious in the last 30°. After the operation the activity level of the VM recovered to levels which were more similar to the VL. This study used raw EMG signals which can not be quantified. The differences, though obvious, are still subjective. The fact that moving the insertion of the quadriceps affected the EMG activity in the VM lends credence to Beckman's (1980) theory of segment rotations affecting the VM's effectiveness. Incongruent rotation would have the tibia moving medially while the femur is moving laterally, giving the tibial tuberosity a more medial displacement.

As with any muscle, the ability of the VM and VL to generate force varies through a range of lengths. This is frequently referred to as the length-tension relationship. During level walking the vasti as a group lengthen between 0.2 - 3.4 inches from a standing length (Morrison, 1970). Morrison found that activity was not constant over this length change, greatest activity usually occurred between the mean and the maximum values. Similarly, the greatest force values exerted by a muscle occurred close to maximum length. This would seem to suggest that the quadriceps operate most efficiently not in a shortened position, but in a lengthened one. A muscle is generally shortened when its origin is brought closer to its insertion, and this usually occurs upon concentric contraction of the muscle.

Therapists have long theorized that strengthening the VM of a patient would help to pull the patella medially and reduce patellar problems. But generally strengthening the quadriceps would also increase the pull of the VL. They therefore have looked for exercises that would selectively strengthen the VM. Slocum and Larson (1968) previously stated that due to the anatomical attachments of the VM, it possessed the ability to prevent external rotation of the tibia during the first 60° of flexion. Conversely it should be able to internally rotate the tibia. Hanten and Schulthies (1990) tested this theory and found that there was no significant difference between the normalized EMG readings of the VMO and VL during resisted internal rotation. A significant difference was found though using adduction exercises. This could possibly be explained because the VMO originates from the tendons of the adductors longus and magnus (Bose, et al., 1980).

A similar result was found earlier by Wheatley and Jahnke (1951). They found elevated EMG signals from the VM when adduction of the thigh was performed with the leg in extension, while the VL showed increased activity during abduction. These researchers also noted, "There is more activity of the vastus medialis in keeping the leg in extension during thigh flexion with the leg laterally rotated and by the vastus lateralis in keeping the leg in extension during thigh flexion with the leg medially rotated" (p.513).

These experiments show that there are certain movements that will cause a functional imbalance in the activity of the VM and VL. The incongruent rotations or the valgus displacement of the knee, discussed earlier, may alter the origins and insertions of these muscles and in turn alter the length-tension relationships and hence affect their patellar tracking abilities.

CHAPTER 3
METHODS AND PROCEDURES

Subjects

This study used one group of normal subjects which was tested three times. Twenty male and female subjects (Marans, Jackson, Glossop, & Young, 1989) (Chao, Laughman, Schneider, & Stauffer, 1983) were used which were enlisted from the campus population. The subjects were approached by the tester and asked to be involved in the study. The subjects were screened to ensure that they had no PFA symptoms and no history of major knee, hip, or ankle problems that would alter their knee mechanics. A questionnaire was used for this purpose (see appendix 1).

Instruments

In order to measure the rotational movement of the segments of the lower limb a tri-planer electrogoniometer was designed and constructed. This apparatus allows 3 degrees of rotational freedom designed to measure; knee flexion/extension, knee adduction/abduction, and internal/external tibial movements. The devise is light weight and non-restrictive. Similar devices have previously been found not to hinder running style and to give valid (Chao, 1980) and reliable (Laugman, Askew, Bleimeyer, & Chao, 1984) measurements.

The main purpose of the goniometer design was to obtain increased sensitivity to changes in internal/external rotations of the tibia relative to the femur, hence it is here that this design differs from others in the literature. Previous designs (C.A.R.S.-U.B.C. , etc.) have used a single potentiometer orientated so that its axis of action was parallel and lateral to that of the lower shank. This was found to give a valid measure of tibial rotation by Chao

(1980), although the validation process included a non-anatomically correct knee joint simulation. According to Kapandji (1983) and Daniel, Akeson, and O'Connor (1990), the long axis of the lower shank passes through the medial side of the tibial plateau, close to, if not through the medial tibial spine. This would seem to indicate that the rotation of the lower shank is not symmetrical (as was the simulation by Chao, 1980) , but rather that the lateral portion of the shank is passing through a larger arc than the medial portion during rotation.

The design of the goniometer used in this study utilized a tibial plate and wrap around arm for direct measure of the tibial motion. Three potentiometers were arranged into a quadrilateral-type configuration, with the tibial long axis of rotation being the fourth point in the figure (see appendix 2). As a result, it was not necessary to know exactly where the fourth point was located. With the four points forming a quadrilateral, it was known that the angles contained within would equal 360°. It was also known that the changes that occurred to the angles of the quadrilateral would always equal 0° (assuming the lengths of the sides remained constant). Therefore in order to determine the amount of tibial rotation, the angular changes occurring at the 3 potentiometers were added together and set equal to -X.

(eq. 1)

$$\Delta pot1 + \Delta pot2 + \Delta pot3 + X = 0$$

$$\Delta pot1 + \Delta pot2 + \Delta pot3 = -X$$

$$-(\Delta pot1 + \Delta pot2 + \Delta pot3) = X \text{ ; } X = \text{change in the rotation of the tibia}$$

(see appendix 3 for further explanation)

The placement of the other two potentiometers which determine abduction/adduction and flexion/extension are similar to previous designs and are both positioned at the level of the lateral epicondyle. The flexion /extension potentiometer is positioned in the sagittal plane, while the one for measuring abduction/adduction is in the coronal plane. A slider bar joining the upper and lower segments of the goniometer was designed to prevent stress on the lower apparatus as a result of anterior/posterior femoral translation during flexion and extension. This slider bar was designed so as not to allow rotation through its long axis.

All components of the goniometer are constructed of either aluminum or plastic, except for the inner component of the sliding rod and the positioning prongs, which are made of steel. The potentiometers are linear, single turn, 10 kΩ units. The linearity of each potentiometer was tested and a regression equation determined prior to use (see appendix 4). The device is powered by a 6 volt DC power supply. The voltage regulator was plugged into a GFI (ground fault interupter) receptacle on a isolation unit so as to protect the subject from possible electrical shock.

Attachment of the electrogoniometer to the thigh and shank were made using elastic straps and velcro. A rubber condyle cup was used to position the upper apparatus against the lateral femoral epicondyle (see appendix 5). This area of the femur contains the instant center of rotation for the sagittal plane as determined by the Reuleaux method (Nordin, & Frankel, 1989).

The dimensions of the electrogoniometer are given in appendix 6 and a wiring diagram is presented in appendix 7.

Voltage signals from the 3 potentiometers used to determine transverse rotations and the potentiometers for abduction/adduction and flexion/extension was captured with an IBM 286 computer via a 12 bit analogue to digital converter. The Lab-Tech™ software program operated at a rate of 40 samples per second and measured voltage to two decimal places. The data was then transferred to the Quattro Pro™ software program, where the voltage changes that occurred at each potentiometer were displayed. It also displayed heel strike and toe off with similar voltage changes generated from foot switches at the heel and toe. The voltage changes from the potentiometers are linearly related to the changes in degrees occurring in the limb. The maximum and minimum voltage values for each cycle were recorded and arranged into a separate table in the Quattro Pro™ software program and then transferred to an EXCEL™ spreadsheet on a Macintosh SE™. The Macintosh spreadsheet converted the voltage values to degree values using the specific regression equations for each potentiometer (see appendix 8). The cycle ranges were then averaged to determine a value for each trial. These calculations determined average degree changes for potentiometers 1-5 for each trial (average Δ pot 1-5). The three values which determine internal/external rotation (ie. Δ pot 3-5) were then entered into the equation previously described (eq. 1) to determine tibial rotation for each trial.

The foot switches were made of two flat pieces of copper, covered by a rubber coating. A wire was attached to each of the copper plates and then to opposite ends of a 1.5 volt battery. This apparatus was then wired into the data acquisition system with the other potentiometers. Both the foot switches were placed 2 cm. from the ends of the shoe.

Procedures

pilot study

Two male subjects were used for the pilot study. They were instructed to bring their regular footwear as per Chao, et al., 1983, and a pair of shorts to the testing session. Before testing began, the procedure was explained, the consent form signed (see appendix 9) and questionnaire filled out. At this point the subject's right leg was fitted with the goniometer and the heel and toe switches taped to the right foot. Fitting of the goniometer occurred with the subject standing at ease as per Isacson, Gransberg, and Knutsson, 1986. The rubber condyle pad was placed over the right lateral femoral epicondyle and the femoral bar was directed to the femoral greater trochanter with the thigh pad positioned so that the slider bar was vertical. The tibial portion of the goniometer was then attached by sliding the steel portion of the slider bar into the upper portion and then the tibial plate was positioned on the medial surface of the tibia. The subject then began to walk on the Quinton Q 65 treadmill at the test speed of 2 MPH and zero elevation. When the subject indicated their normal walking style had been obtained, recording began. The recording consisted of approximately 15 strides. The treadmill was then stopped and the subject was allowed to get off. The brace was not be removed while the subject had a 3 minute break. Once the subject was back on the treadmill and comfortable walking was again indicated, another 15 strides were recorded. This allowed for determination of mechanical reliability.

This procedure was repeated after removal and re-fitting of the goniometer. A 5 minute break was taken after removal of the goniometer. No marks on the skin were purposely made to indicate placement. This allowed for determination of placement reliability (Isacson, et al., 1986).

A time interval of approximately 24 hours was used to separate this testing session from the fourth test. This part of the study was used to determine if validity was maintained over time. This fourth test used the same treadmill speed and goniometer attachment procedure. The subject came in, was fitted with the goniometer and switches as in the previous tests, walked until comfort was indicated and again approximately 15 strides collected.

Although the data was collected for approximately 15 strides, only 10 strides were analyzed. The 10 strides to be analyzed were chosen after the maximum and minimum values were displayed in the Quattro spreadsheet. The first 2-3 strides were not used, the next 10 were analyzed provided heel and toe switch information were reliable. This was determined by the consistency of the time span of the stride. Strides in which foot switches either did not engage or remained shut were not used.

Statistical Analysis

Only 2 subjects were used in the pilot study, therefore in order to determine the reliability between trials in such a small sample, the measurements in each trial were compared for each subject independently. The raw data (voltage readings) from each of the four tests were divided into strides using the heel strikes as markers, then maximum and minimum values determined for each potentiometer within each stride. The voltage values (maximum and minimum) for the 10 strides to be analyzed in each trial were converted to degree values using the EXCEL™ spreadsheet. This generated 10 degree values for each of: flexion/extension, abduction /adduction and tibial rotation. These values were then compared to the corresponding re-test degree values using a repeated measures ANOVA.

Follow up tests were used to determine if significant differences existed between the trials.

Pilot Study Results

As was mentioned earlier the pilot study used only two subjects, each tested four times. The descriptive statistics are shown in table 7. An analysis of variance (ANOVA) was run on the 10 stride values for each of the three rotational parameters to determine if significant differences existed between the trials. This was done for each subject. Tables 8 and 9 show the determined probability values for each of the ANOVA tables and they also show which trials were determined to be significantly different. For both subjects the tibial rotation parameter was the only one to show reliability over the four trials ($p > .05$). For complete ANOVA tables of the pilot subjects see appendix 10.

Table 7
Means and standard deviations of the three rotational parameters. Average of the 4 trials in the pilot study (n=2)

	MEAN (STANDARD DEVIATION)
TIBIAL ROTATION	4.78° (.6)
AB/ADDUCTION	10.84° (2.12)
FLEXION/EXTENSION	45.33° (3.91)

Table 8
ANOVA results for subject KCARL

	p value	significant differences
TIBIAL ROTATION	p = .0555	none
AB/ADDUCTION	p = .024*	trial 1 and trial 3 trial 2 and trial 3
FLEXION/EXTENSION	p = .0101*	trial 2 and trial 4 (24hrs) trial 3 and trial 4 (24hrs)

*significant at .05 level of confidence

Table 9
ANOVA results for subject DALSTE

	p values	significant differences
TIBIAL ROTATIONS	p = .802	none
AB/ADDUCTION	p = .0007*	trial 1 and trial 2 trial 1 and trial 3
FLEXION/EXTENSION	p = .0001*	all except trial 1 and trial 4

*significant at .05 level of confidence

Since the tibial rotation data was of the greatest interest in this study, it was this data that was used to determine the number of subjects that would be required in the main study. The power analysis can be seen in appendix 11. Twenty subjects was determined to be sufficient to expose differences between 3 trials if in fact real differences were occurring.

A pre- and post-test check of the linearity of the potentiometers was carried out at the beginning and end of all testing to ensure that potentiometer wear would not be a factor. Approximately 10 voltage readings were taken for each potentiometer and a scattergram and regression equation generated. (see appendix 12)

Main Study Procedures

Once it was determined that there was no difference in reliability between running a re-test with a 24 hour time interval and re-testing with only a 5 minute interval, the 5-minute interval method was adopted for the re-test situation in the main study. This reduced the number of trials required to three. Also, a 2 minute accustomization trial was added to the beginning of the testing session so that the subjects could get used to the goniometer and treadmill before a trial was run in which data was collected. All trials used the same treadmill speed of 2 MPH.

With the above changes, testing could be completed in one session. The same procedure was employed as in the pilot study in terms of pre-test preparation and goniometer application. After the 2 minute accustomization trial, the subject remained on the treadmill for a break of approximately 1 minute. The first trial was then run, with data being collected after comfortable walking was indicated. The subject again remained on the treadmill for a 3 minute break before walking for the second test. After the second test the subject was allowed off the treadmill and the goniometer was removed for a 5 minute break. The goniometer was re-fitted as previously stated and the third test was administered. The testing was complete at this point.

Statistical Analysis

With 20 subjects being used, the average degree value of 10 strides was used as the value of that trial. Therefore, each trial yielded an average value for each of : flexion/extension, abduction/adduction and tibial rotation. These values were derived in the same manner as in the pilot study (ie. using maximum and minimum values and the EXCEL spreadsheet) and were used to determine if significant differences existed between the three trials. Again, the three trials were compared using an ANOVA and follow up tests. This was done for each of the three rotational parameters.

For the main study, not only were maximal ranges for the entire stride compared, but also ranges for only the stance phase. The IBM computer accomplished this by using the heel strike and toe off as markers. This allowed comparison of the three rotational parameters in the stance phase separately.

CHAPTER 4
RESULTS

Main Study

There were 20 subjects tested in this experiment. The subject pool consisted of 8 females and 12 males. Unfortunately, the initial data from 2 subjects, both female, was unusable. In both instances the positioning prongs broke from their attachment to the slider bar which gave the slider bar too much play in the sagittal plane. These two subjects were subsequently re-tested after the positioning prongs were re-fastened, even though their previous experience could have biased their results (ie. increased consistency). Furthermore, 3 subjects gave unreliable toe switch information due to the switch sticking during testing. The data from these same subjects was still used as it contained valuable stride information. This resulted in a reduced N in the stance (heel strike to toe off (H2T)) data (N=17). The subjects had a mean age and S.D. of 25.2 (3.1) years.

Three trials for each of the 20 subjects generated a total of 60 values for each of the three rotational parameters (flexion/extension, abduction/adduction and tibial rotation) for the whole stride (H2H) and 51 values for each parameter in the stance phase (H2T). These values are presented in table 10.

Table 10
The mean degree values of the three rotational parameters for both stride and stance phase

	H2H (stride) n=20	H2T (stance) n=17
TIBIAL ROTATION	9.23° (2.5)	6.92° (2.8)
AB/ADDUCTION	12.0° (3.1)	7.51° (2.4)
FLEXION/EXTENSION	47.41° (6.2)	24.21° (7.1)

mean (standard deviation)

The data for the whole stride (H2H) will first be considered. The first parameter, tibial rotation, had its three trials analyzed with an ANOVA which determined $p = .2749$. Therefore, at the alpha level of .05 the follow up tests indicated no significant difference between any of the trials. The ANOVA for abduction/adduction (H2H) showed $p = .0622$. No significant differences existed between any of the trials for this parameter either. Similarly no significant differences between any of the trials were indicated by the flexion/extension ANOVA which had $p = .192$. For the complete H2H ANOVA tables see appendix 13.

With regard to the stance phase (H2T) data, the first ANOVA for the tibial rotation parameter determined $p = .5142$. No differences were shown to occur between any of the trials. Again, as in the H2H data, no significant differences were found between any of the trials for either stance phase (H2T) abduction/adduction or flexion/extension. The probability values were . 6019 and .9305 respectively. For complete H2T ANOVA tables see appendix 14.

Table 11
The ANOVA p-values comparing the three trials of each rotational parameter. Both stride and stance phase shown

	H2H (stride) n=20	H2T (stance) n=17
TIBIAL ROTATION	p = .2749	p = .5142
AB/ADDUCTION	p = .0622	p = .6019
FLEXION/EXTENSION	p = .192	p = .9305

CHAPTER 5
DISCUSSION

Pilot study

The results from the pilot study showed that the electrogoniometer was reliable with respect to tibial rotations for all aspects tested, that is, for mechanical reliability (trials 1 and 2), placement reliability (trials 1 and 3) and placement reliability over time (trials 1 and 4). This was reflected in the probability values found in both subjects' analysis of variance. Furthermore, the tibial rotation measures for both subjects combined (2x 4 trials = 8) had a very low standard deviation of 0.6°. This may have been biased however, due to the fact that during initial testing of the electrogoniometer design the pilot subjects may have become accustomed to walking on the treadmill. This could also explain why the standard deviation (a measure of variability) in the main study was so much higher (2.5° H2H) than that of the pilot study. The differences in sample size alone could also be responsible for this fact. It was suspected that the standard deviation would be higher in the main study due to the larger N and was estimated at between 1° and 1.5°. It was on these values that the power analysis was based. The standard deviation of the differences between the trials (the square root of the means square value corresponding to the residual source of variance in the ANOVA table) for the main study was 1.46° for the tibial rotation H2H data. For the H2T data it was 1.24° which means that even with the subject total reduced to 17 for this section, the study was still powerful enough to determine differences between trials if in fact differences existed (see appendix 11).

The ANOVA tables for both flexion/extension and abduction /adduction showed a number of significant differences between trials. It was noticed though, that the differences in the means were fairly small, and it was assumed that with a larger sample size in the main study, the increased variance would eliminate some of these differences. This turned out to be the case.

Of final note, the means of all three rotational parameters in the pilot study fell within the ranges of the H2H parameters collected in the main study.

Main Study

The main purpose of this study was to determine if the constructed electrogoniometer was a reliable measuring tool in a normal population. The reliability was broken down into two parts, mechanical and placement. Testing the subjects and then re-testing them without removing the electrogoniometer tested mechanical reliability, while testing, removing and re-fitting the electrogoniometer tested the ability of the electrogoniometer to be used at some other time and still achieve the same results.

The design of the study was to test the research hypothesis that the means of the three trials would not differ significantly from each other. The null hypothesis then became; the means of the three trials would not equal each other. At the alpha level of .05, the null hypothesis had to be rejected and the research hypothesis was accepted. This was true for each parameter studied, for both the whole stride (H2H) and the stance phase alone (H2T). This suggests that the electrogoniometer design is mechanically reliable and it also has the ability to maintain reliability after removal and re-fitting.

With respect to the stride (H2H) data, the tibial rotation means were the most reliable of the three parameters. This is consistent with the pilot data which only considered the whole stride. Looking more closely at the means of each trial, there was a higher degree of mechanical reliability (trial 1- trial 2) than placement reliability (trial 1- trial 3) as might be expected. Both mean differences, however, were very small, .32° and .433° respectively.

The abduction/adduction parameter had the highest variance in its results. This is consistent with the study by Kettlekamp et al. (1970) which showed test/re-test reliability scores for abduction/adduction to be the lowest of the three rotational parameters studied. It should be noted, however, that the data from the present study still exhibited a high degree of mechanical reliability. Most of the variance in the ANOVA occurred between trials 1 and 3 (and trials 2 and 3). This was due mainly to trial 3's large variability (largest S.D.).

The flexion/extension ANOVA showed less consistency in the stride (H2H) data than in the stance (H2T) data. In fact, this was true for all parameters studied. Consistently, the H2T ANOVA tables determined higher probability values than the corresponding H2H ANOVA tables. A possible explanation for this is the consistency of stance time compared to stride time. Data was collected for both the time of each stride and each stance phase. The stride time ANOVA table showed a $p = .0189$ with significant differences between trials 1 and 2 (borderline) and trials 1 and 3, whereas, the stance time ANOVA determined $p = .971$, that is, no differences between the trials (see appendix 15). This greater variation in subjects strides during the swing phase of gait may account for some of the variability in the H2H data. Furthermore, the fact that the greatest variability in the stride time occurred in trial 3 may account for the greater variance seen in this trial's results over the three parameters studied.

One reason that may account for the greater variability seen in the stride times of the third trial relates to experimental design. Before trial 1 the subject had the chance to walk on the treadmill in the accustomization trial, and before trial 2 the subject had both the accustomization trial and trial 1, in which to find their comfortable gait pattern. Before trial 3, however, the subjects had a 5 minute break, after which they were asked to immediately do a trial. A second accustomization trial, before trial 3, may have been beneficial in increasing consistency, not only in terms of time, but also in achieving the same gait patterns as in previous trials. This would be the case especially with subjects who were walking on the treadmill for the first time. It was noted that some subjects initially had quite a difficult time walking on the treadmill.

Comparison to Literature Values

Even though this study was not designed to validate the results generated from the electrogoniometer, it is of interest to see how these values compare to the values in the literature. It has to be remembered that the values in the literature are from goniometers of various designs, and as stated in the limitations, these values may not be comparable. Also the values from one study (Lafortune & Cavanaugh, 1985) are from an entirely different method (intracortical pins and video analysis) which has its own set of limitations. Finally, the fact that two different populations are being compared should be kept in mind.

When looking at the data for each parameter in both the H2T and H2H situations, one can see that, consistently, the values of each parameter are higher in the H2H data. (This was not only true for the means but also for each individual stride as well.) This would suggest that the largest range of movement was occurring during the swing phase of the stride, rather than in the stance phase. Therefore, this allows the stride data (H2H) to be viewed as swing data.

When comparing the values of tibial rotations from this study to those in the literature, the factor of speed needs to be addressed. Czerniecki et al. (1988) have shown that tibial rotation values increase with increasing speed. The subjects in this present study were walking at a speed of 2 MPH, which is at the lower end of the velocity range used in the literature. When comparing the tibial rotation value found in this study to those in the literature which found their maximal value in the swing phase, it can be seen that, again, the value from this study is at the lower end of the range found in the literature. Isacson and Brostrom (1988) used a speed of 1.34 MPH in their study and found a value for tibial rotation of 7°(4), which was lower than that found in this study (9.23°(2.5)). This would be expected due to the difference in speeds used. The next closest speed reported was 2.51 MPH by Chao et al. (1983), which produced tibial rotation values of 14°(4) from 20 females. There are three studies which allowed the subjects to walk at their normal walking speed and reported the speed as self-selected. The values cited in the study by Marans et al. (1989) are very comparable at 9.2°(3.7) for males and 8.9°(4.1) for females. The second study using an unreported self-selected speed (Kettlekamp, et al., (1970)) found values of 12.9°(4.41) for the test situation and 13.1°(4.39) for the re-test situation. (Note: the standard deviations in these experiments were higher than those found in the present study.) Finally, Isacson and Brostrom, (1988) found their self-selected speed to average 2.69 MPH. This produced a value of 9°(2). As can be seen, the values for tibial rotations obtained in this study are very comparable to those reported in the literature when speed and differing electrogoniometer designs are taken into consideration.

No literature was found to indicate that the speed of walking had an effect on the amount of abduction/adduction occurring. Therefore, this was not considered as a factor in this section. Still, the value of 12.0°(3.1) determined for the swing phase in this experiment exceeded all other values cited in the literature. The literature values ranged from 3.6°(2.1) to 10.5°(4.41). Realistically, the value of abduction/adduction determined in this study was higher but not dramatically different from the range cited in the literature, in fact, it was not significantly different from the higher end of the range (alpha = .05, $n_1 = 60$, $n_2 = 32$, t = 1.71).

Judging from the literature values for flexion/extension, there is a trend toward increasing flexion with increasing speeds. Even considering speed though, the value of 47.4°(6.2) from this experiment is lower than the value of 55°(5) from the study by Isacson and Brostrom, (1988) which used the lowest speed of 1.34 MPH. The value of 47.4°(6.2) found in this experiment is comparable with the values in the study by Marans et al. (1989), which used self-selected speeds. They obtained values of 46.8°(6) and 48.2°(6.6) for the control group and unaffected leg group respectively.

With regard to the stance phase data, this study obtained a value of 6.9°(2.8) for the range of tibial rotations. Three studies in the literature determined values for tibial rotations in the stance phase, however all used higher speeds than the 2 MPH used here. The closest speeds were 2.51 MPH and 2.77 MPH in the study by Marans et al. (1989). Their respective values were 9°(3) for males and 10°(3) for females. The other studies which used higher speeds generated higher values, the highest being 14.8°(6.4) at 5.8 MPH (Czerniecki, et al., (1988). Chao et al. (1983) was the only study which cited values for both the stance and swing phases. In that study larger values were found in the swing phase, which is consistent with the data in this study.

Only one study looked specifically at stance phase abduction/adduction and two looked at stance phase flexion/extension. Chao et al. (1983) found his subjects, both male and female, each displayed 7°(2) of abduction/adduction in the stance phase. This is comparable to the value of 7.5°(2.4) determined from this experiment.

Kettlekamp et al. (1970), using an unreported self-selected speed, found his subjects displayed 20.6°(4.4) of flexion in the stance phase. The 24.2°(7.1) found in this study was slightly higher and also had a higher degree of variability associated with it. In the second study, Chao et al. (1983), determined that their male subjects, walking at a self-selected speed of 2.77 MPH, exhibited 32°(6) of flexion in the stance phase. The females in the same experiment, walking at 2.51 MPH, exhibited 30°(6) of flexion. For complete summaries of literature values see appendix 16.

As can be seen, the values obtained from the electrogoniometer designed for this experiment are very comparable to those in the literature. Again, this comparison was not intended to validate the instrument but rather to simply determine if the new design, especially the way it measures the range of tibial rotations, generated values in the range of those from instruments previously validated.

There is a final note regarding the design of the electrogoniometer. In all ANOVA tables the first F - test, which indicates the ability of the device to determine differences between subjects, was very large. The corresponding p - values were all .0001, indicating that the goniometer had the ability to determine differences between each subject with a very high degree of certainty, and at the same time, was able to determine that the trails of each subject did not differ significantly. Furthermore, scattergrams were plotted to find out if there were any interactions between the various parameters (ie.cross-talk). No significant correlations were found. The highest r^2 value was .153 with the others below .085. Therefore, it can be concluded that the goniometer measured the three parameters independently.

CHAPTER 6
SUMMARY AND CONCLUSIONS

The lack of information regarding the rotations occurring in the leg of a person with patello-femoral arthralgia initiated this introductory study. The electrogoniometer designed for this study, with its ability to measure motion in three planes, will enable some of the questions in the literature to be answered. This device could be used to compare a normal population to a pronating population with patello-femoral symptoms. This would hopefully resolve the question of which type of motion is occurring at the tibio-femoral joint in this population, congruent (Buchbinder, et al., 1979 ; Ramig, Shadle, Watkins, Cavolo, & Kreutzberg, 1977 ; Tiberio, 1987 ; Williams, 1977) or incongruent motion (Beckman, 1980 ; D'Amico, & Rubin, 1986 ; Rothbart, et al., 1988 ; James, Bates, & Ostering, 1978). That is, is there a larger amount of transverse tibial rotation occurring in the pronating patello-femoral population. Furthermore, this electrogoniometer should be able to determine if the test population, as compared to normals, has greater knee flexion during gait as theorized by Herzog-Franco (1987). This greater knee flexion would increase the patello-femoral joint reaction force and could lead to patello-femoral symptoms. Finally, this device could help determine if there is any difference in the amount of abduction or adduction between the two populations. If in fact the pronating patello-femoral population demonstrates greater abduction, as D'Amico and Rubin, (1986) and Santopietro (1988) have speculated, this could have an effect on the musculature of the knee, especially the effectiveness of the VMO.

The electrogoniometer designed for this study used five potentiometers to measure motion in three planes. The device measured:

tibial rotations relative to the femur, knee abduction/adduction, and knee flexion/extension. In order to measure transverse tibial rotations this goniometer used a "moving quadrilateral", which is unique to this design.

The reliability of the device was initially tested in a small pilot study which yielded encouraging results. Two subjects were each tested four times to determine if the device could reliably measure the three rotational parameters. The goniometer was attached to the subjects in a prescribed manner and the subjects walked on a treadmill at 2 MPH.

Mechanical reliability (trials 1-2), placement reliability (trials 1-3), and placement reliability over time (trials 1-4) were all tested. Reliability between trials was determined using an analysis of variance and follow up tests. The results indicated that the most reliable measurements were those of tibial rotations. The results also indicated that there was no difference in the reliability of trials run on the same day as those run with a 24 hour interval.

The reliability of the electrogoniometer was then tested in a larger study using twenty normal subjects. The subjects were each tested three times with all three trials occurring on the same day. The ability of the device to measure the three rotational parameters was tested for both mechanical (trials 1-2) and placement (trials 1-3) reliability. In this main study, the data was broken down so that the reliability of three rotational parameters being studied could be analyzed for both a full stride and for the stance phase alone. Again, as in the pilot study, the reliability between trials was determined using an analysis of variance and follow up tests. Furthermore, the results of the three rotational parameters from the constructed electrogoniometer in this study were compared to the results obtained in other studies that also used normal subjects.

Conclusions

The various ANOVA tables indicate that the electrogoniometer that was designed and constructed for this study was a reliable measuring tool. The device displayed both mechanical and placement reliability for all three of the rotational parameters. Furthermore, the results determined that the goniometer had a higher degree of reliability in the stance phase (H2T) than for the whole stride. The ANOVA tables also indicated that the goniometer was sensitive enough to determine differences between subjects, and regression equations determined that device was measuring the three parameters independently.

The comparison of the three rotational values generated by this electrogoniometer with those of the same parameters in the literature, indicated that this device produced values consistent with accepted values.

Improvements/recommendations

If this electrogoniometer was to be validated there are a few improvements that should be made to decrease error in measurement.

1. Straps should be made out of a material that would be more adhesive to the subjects skin.

2. The strap that is in place at the condyle pad could be replaced with a sleeve (adjustable) that would be more comfortable and would assure the condyle pad remained in its place.

3. The slider bar should have a "flat" machined into the inside so as to reduce the piston's ability to rotate. As it is, using a nylon screw, a few degrees of rotation are permitted. Furthermore, with prolonged use the nylon screw will wear out increasing the amount of rotation permitted.

4. Precision potentiometers should be used, although modifications to inexpensive ones seemed to work.

5. The wrap around bar could be made smaller so as to reduce momentum that may distort results. This would become more important as the speed of testing increased.

6. The wrap around bar could be made with a break and rivet in the middle so that it could become reversible and left legs could also be tested. This though would require using a greater than one turn potentiometer at the flexion/extension position.

7. A new device on the lower end of the slider bar could be designed which would give greater certainty that the fourth side of the moving quadrilateral is remaining constant.

8. Higher gauge wire could be used in order to reduce its bulkiness.

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APPENDIX 1

Eligibility questionnaire

Eligibility Questionnaire
for electrogoniometry of the knee study

DATE______________________________________ AGE________
NAME______________________________________ SEX________
ADDRESS___________________________________

1) Are you experiencing any pain or discomfort in your hips, knees, ankles, feet or any other part of your lower limbs? YES__ NO___

2) Have you ever experienced any major trauma or disease process in your hips, knees, ankles, or feet? (ie. congenital hip problems, arthritis of joints, fractures, surgery, cartilage problems, etc.) YES__ NO__

if YES explain: ______________________________________

____

____

3) Do you, or have you ever, worn orthotics?

YES__ NO___

4) Are you experiencing any problems with your knee caps? (ie. any pain after prolonged sitting, clicking under the knee cap associated with pain, locking, catching, or giving out of the knee, pain going up or down stairs)

YES__ NO___

5) Can you think of anything that is hindering either your ability to walk, or your walking style, at this time? (ie. blisters, calluses, corns, strains, cramps, etc.)

YES__ NO___

if YES explain:______________________________________

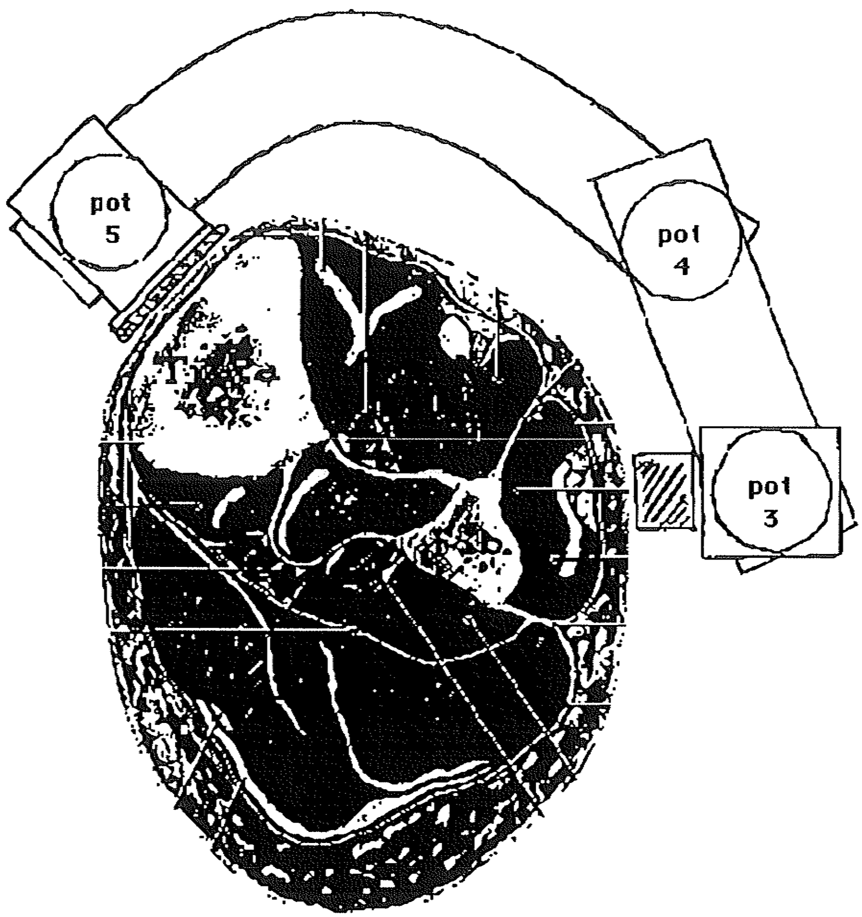
____

____

APPENDIX 2

Diagram showing potentiometer configuration for determination of internal/external rotation

Cross section of the tibia showing potentiometer
configuration for determination of internal/external rotation

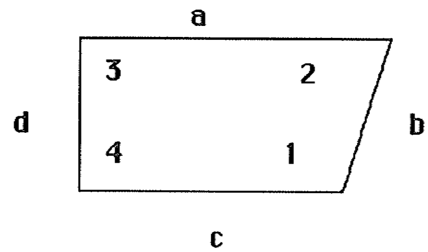


APPENDIX 3

Diagram explaining equation 1

Explanation of equation 1

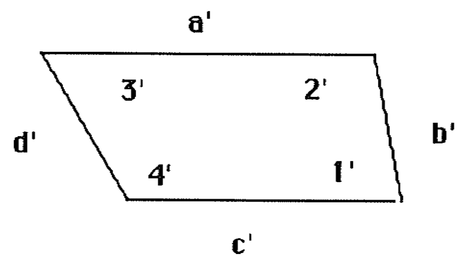
starting position



the apex of angles 1, 2, and 3 correspond to potentiometer positions

the apex of angle 4 represents the long axis of the tibia

internally rotated position



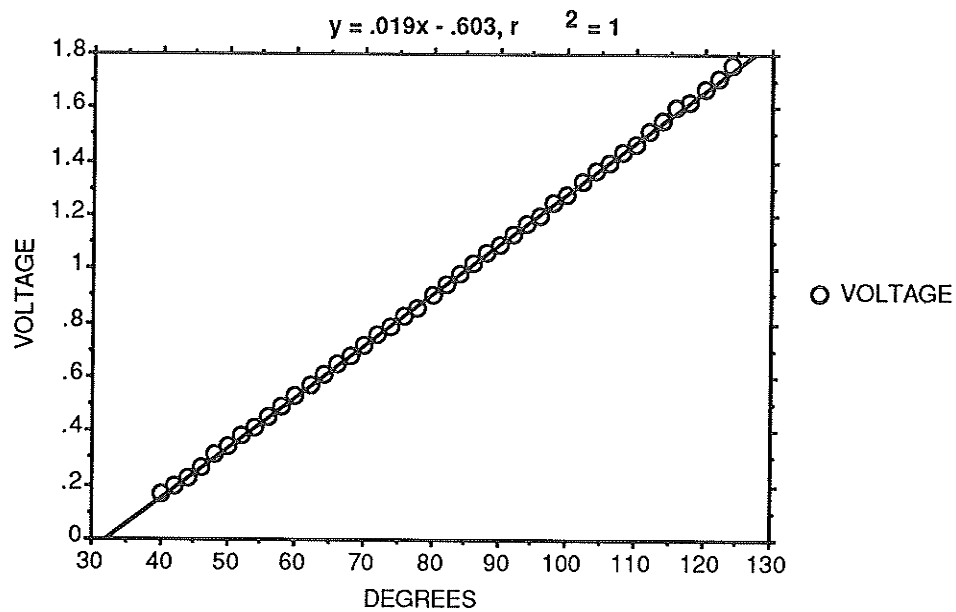
$\angle 1 + \angle 2 + \angle 3 + \angle 4 = 360°$
$\angle 1' + \angle 2' + \angle 3' + \angle 4' = 360°$
subtract
$\Delta\angle 1 + \Delta\angle 2 + \Delta\angle 3 + \Delta\angle 4 = 0$
$\Delta\angle 1 + \Delta\angle 2 + \Delta\angle 3 = -\Delta\angle 4$
$-(\Delta\angle 1 + \Delta\angle 2 + \Delta\angle 3) = \Delta\angle X$; X = tibial rotation

assuming : $a = a'$
$b = b'$
$c = c'$
$d = d'$

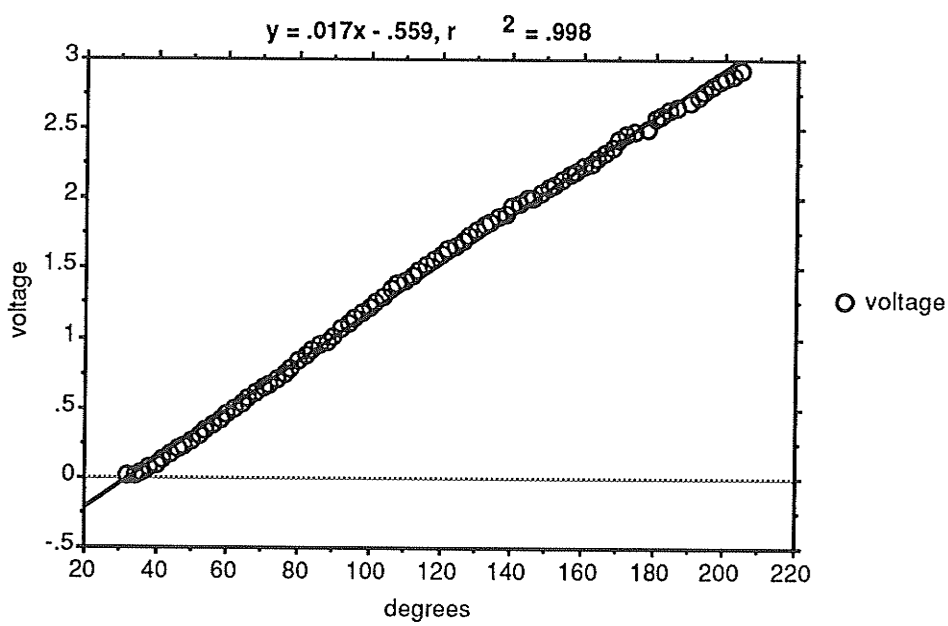
APPENDIX 4

Scattergrams showing the linearity of the potentiometers

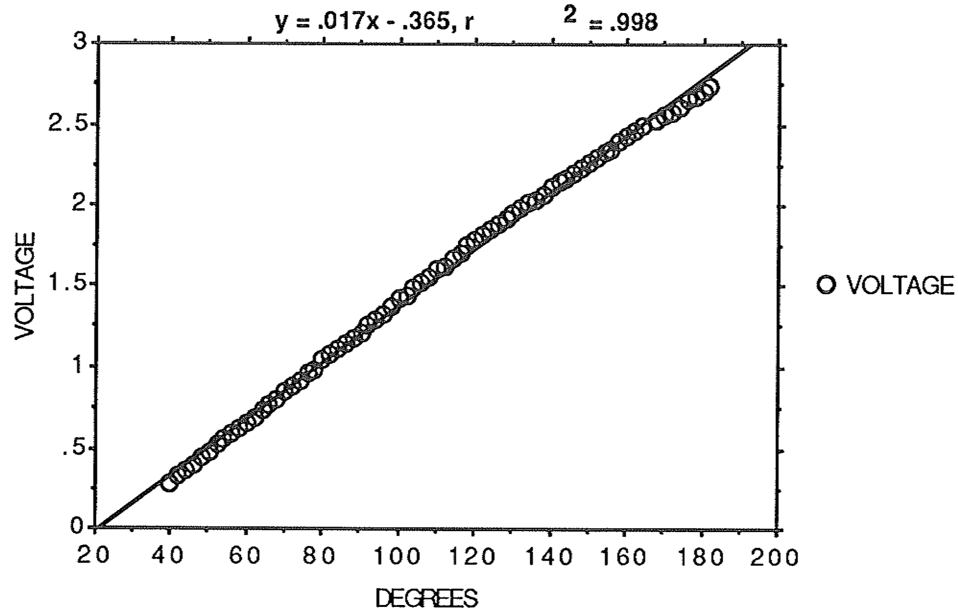
VIEW OF POTENTIOMETER 1



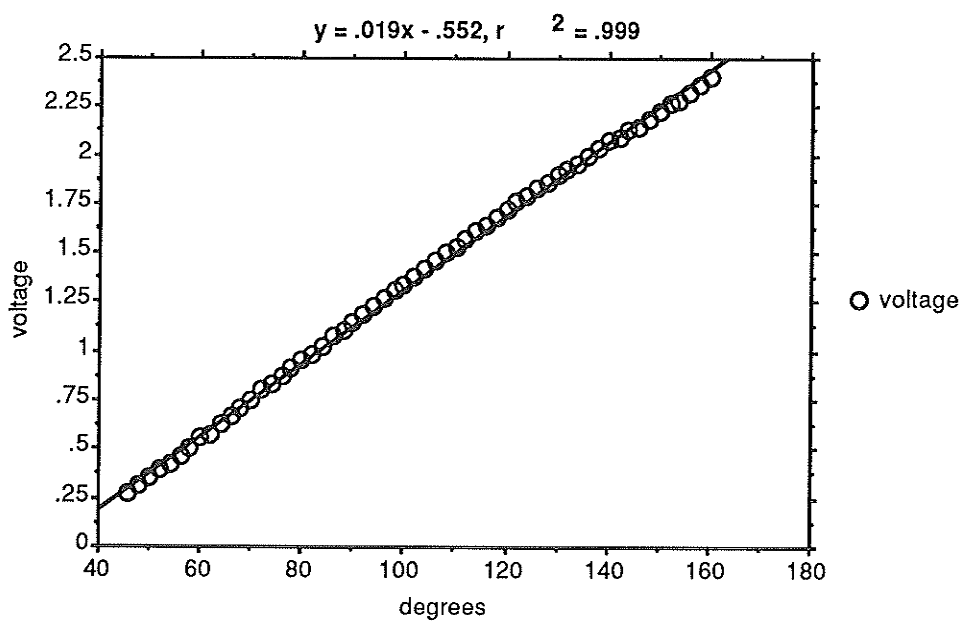
VIEW OF POTENTIOMETER 2



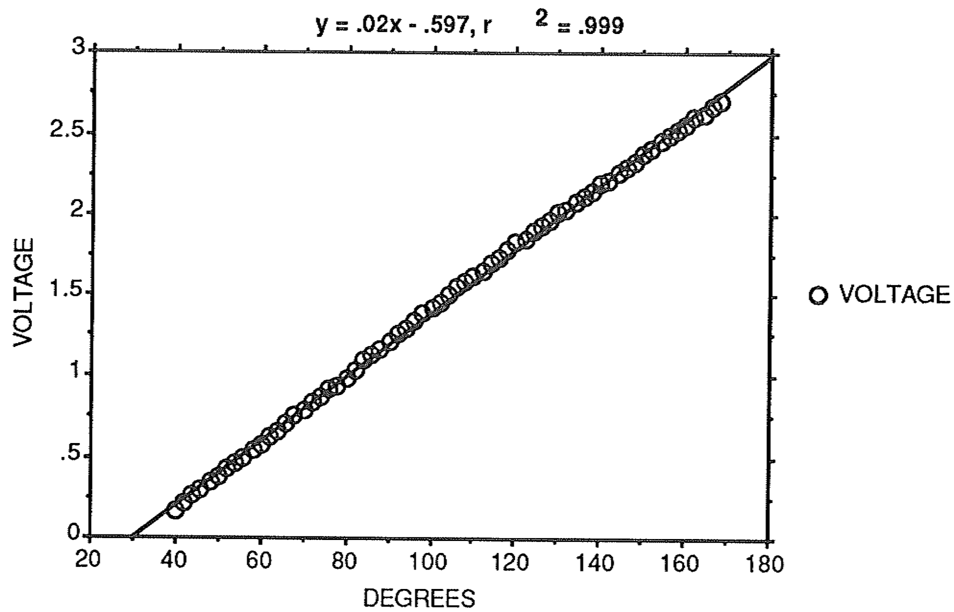
VIEW OF POTENTIOMETER 3



VIEW OF POTENTIOMETER 4



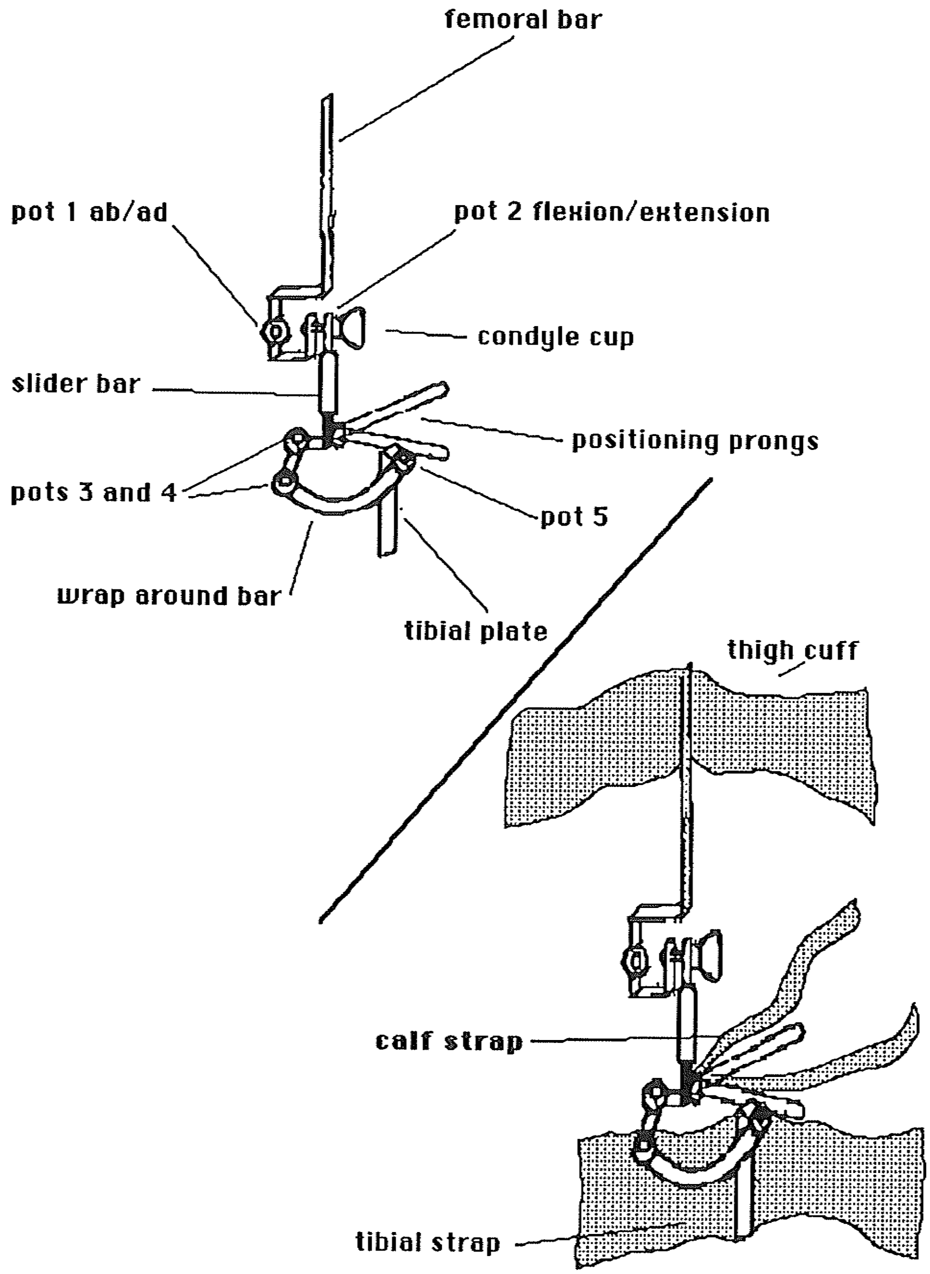
VIEW OF POTENTIOMETER 5



APPENDIX 5

Diagram of electrogoniometer

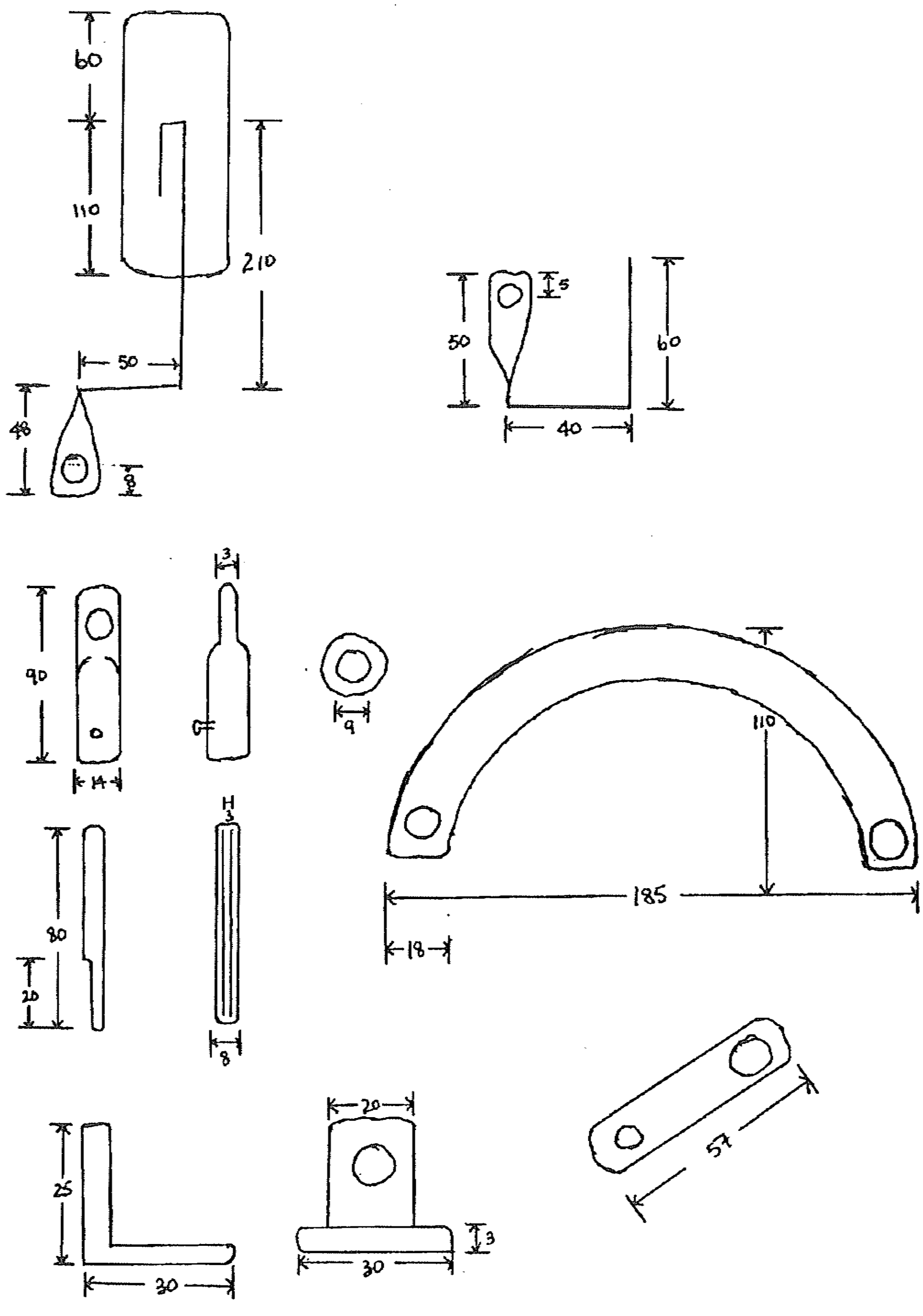
Diagram of electrogoniometer



APPENDIX 6

Dimensions of the parts of the electrogoniometer

Dimensions of the parts of the electrogoniometer

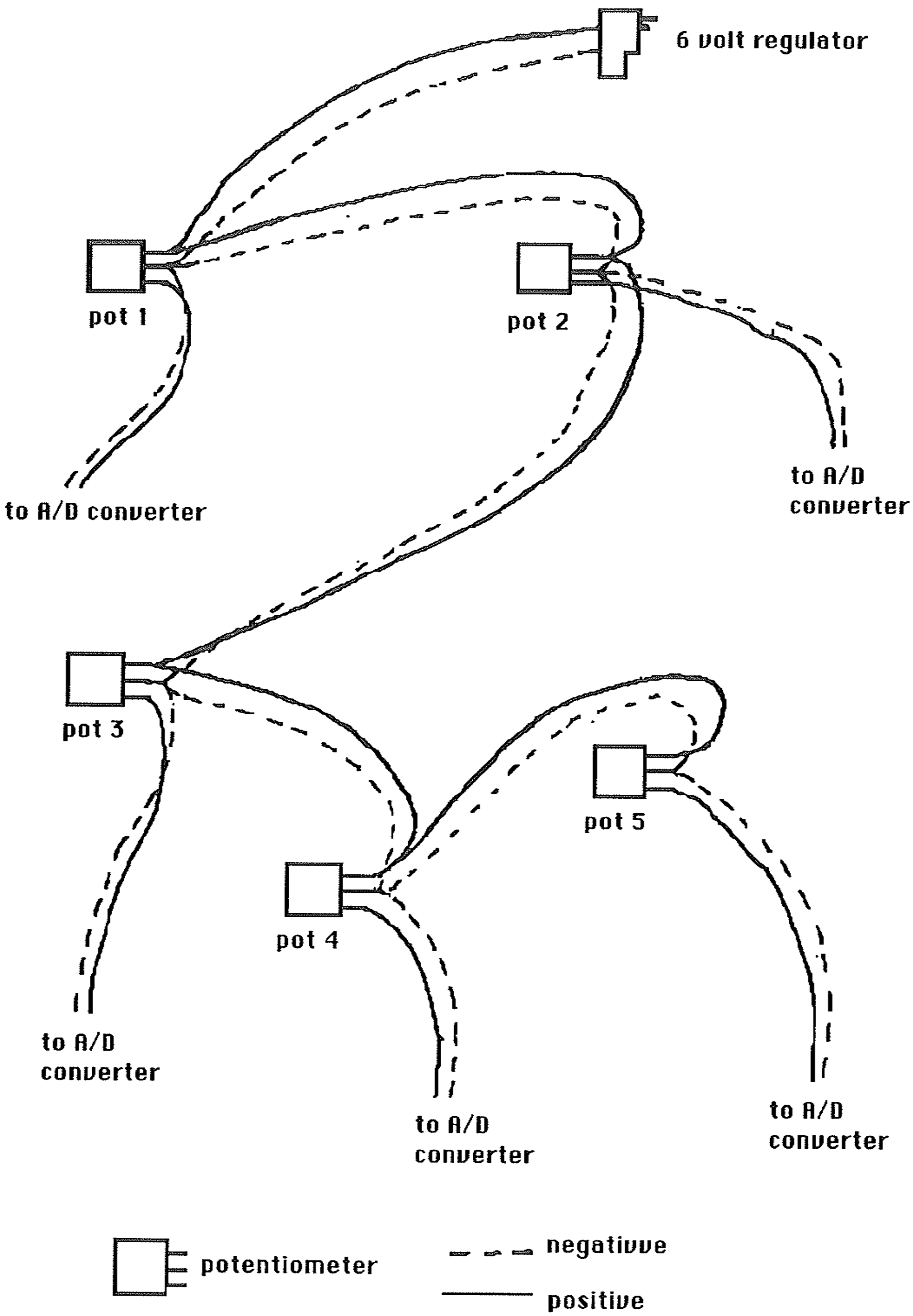


(all measurements in millimeters)

APPENDIX 7

Wiring diagram

Wiring diagram



APPENDIX 8

Excel™ spreadsheet for voltage conversions and degree value determination

Excel™ spreadsheet for voltage conversions and degree value determination

SUBJECT			CYCLE VALUES						
POT3	MAX 1	MIN 1	MAX 2	MIN 2	MAX 3	MIN 3	MAX 4	MIN 4	MAX 5
VOLTAGE									
DEGREES	27.65	27.65	27.65	27.65	27.65	27.65	27.65	27.65	27.65
RANGES		0		0		0		0	
AVERAGE	0								
SUBJECT			CYCLE VALUES						
POT 4	MAX 1	MIN 1	MAX 2	MIN 2	MAX 3	MIN 3	MAX 4	MIN 4	MAX 5
VOLTAGE									
DEGREES	29.0526316	29.0526316	29.0526316	29.0526316	29.0526316	29.0526316	29.0526316	29.0526316	29.0526316
RANGES		0		0		0		0	
AVERAGE	0								
SUBJECT			CYCLE VALUES						
POT 5	MAX 1	MIN 1	MAX 2	MIN 2	MAX 3	MIN 3	MAX 4	MIN 4	MAX 5
VOLTAGE									
DEGREES	29.85	29.85	29.85	29.85	29.85	29.85	29.85	29.85	29.85
RANGES		0		0		0		0	
AVERAGE	0								
			AB/ADD						
SUBJECT			CYCLE VALUES						
POT1	MAX 1	MIN 1	MAX 2	MIN 2	MAX 3	MIN 3	MAX 4	MIN 4	MAX 5
VOLTAGE									
DEGREES	31.7368421	31.7368421	31.7368421	31.7368421	31.7368421	31.7368421	31.7368421	31.7368421	31.7368421
RANGES		0		0		0		0	
AVERAGE	0								
SUBJECT			TIBIAL ROTATIONS						
CYCLE	1	2	3	4	5	6	7	8	9
RANGE/CYCLE	0	0	0	0	0	0	0	0	0

APPENDIX 9

Consent form

The University of Manitoba

Faculty of Physical Education and Recreation Studies

INFORMED CONSENT FOR ELECTROGONIOMETRY OF THE KNEE STUDY

EXPLANATION OF THE STUDY

The purpose of this study is to test a newly designed piece of equipment which measures the rotations of the leg. This device is called an electrogoniometer and is similar to a leg brace that is connected to a computer. On the brace are a number of sensing devices which will indicate the movements of your knee joint. This information will be gathered while you walk on a treadmill. You will be asked to walk on the treadmill four times. That is 4 back-to back trials will be required, with the removal and re-fitting of the electrogoniometer occurring between trials 3 and 4. Each trial will consist of approximately 10 strides each. Also two small electrical switches will be taped to the heel and toe of your shoe to indicate each time your heel strikes, and your toe leaves the floor.

DISCOMFORTS

It is very unlikely that the electrogoniometer will cause you any discomfort or restrict your walking style.

INQUIRIES

If you have any questions throughout the study feel free to ask. An attempt will be made to explain the procedures and the results as clearly as possible. Should this be insufficient, ask questions.

CONSENT

I understand that my participation in this study is voluntary. I also understand that I may withdraw at any point in the study and that subject confidentiality will be respected.

I have read this form, understand the procedures involved in the study and willingly agree to participate.

________________________ ______________________________

date signature of participant

____

signature of witness

APPENDIX 10

Pilot study ANOVA tables

KCARL TIBIAL ROTATION

One Factor ANOVA X 1 : TRIAL Y 1 : RANGE

Analysis of Variance Table

Source:	DF:	Sum Squares:	Mean Square:	F-test:
Between groups	3	5.219	1.74	2.772
Within groups	36	22.595	.628	p = .0555
Total	39	27.814		

Model II estimate of between component variance = .111

One Factor ANOVA X 1 : TRIAL Y 1 : RANGE

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
KCARL1	10	5.591	.673	.213
KCARL2	10	4.904	1.155	.365
KCARL3	10	5.182	.757	.24
KCAR (24HRS)	10	5.84	.388	.123

One Factor ANOVA X 1 : TRIAL Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
KCARL1 vs. KCARL2	.687	.719	1.253	1.939
KCARL1 vs. KCARL3	.409	.719	.444	1.154
KCARL1 vs. KCAR (24HRS)	-.249	.719	.165	.703
KCARL2 vs. KCARL3	-.278	.719	.205	.785
KCARL2 vs. KCAR (24HRS)	-.936	.719*	2.326	2.642

* Significant at 95%

One Factor ANOVA X 1 : TRIAL Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
KCARL3 vs. KCAR (24HRS)	-.658	.719	1.15	1.857

KCARL ABDUCTION/ADDUCTION

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Analysis of Variance Table

Source:	DF:	Sum Squares:	Mean Square:	F-test:
Between groups	3	16.689	5.563	3.544
Within groups	36	56.512	1.57	p = .024
Total	39	73.201		

Model II estimate of between component variance = .399

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
KCARL1	10	9.682	.274	.087
KCARL2	10	9.789	2.278	.72
KCARL3	10	8.157	.622	.197
KCAR (24HRS)	10	9.262	.792	.25

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
KCARL1 vs. KCARL2	-.107	1.136	.012	.191
KCARL1 vs. KCARL3	1.525	1.136*	2.469	2.722
KCARL1 vs. KCAR (24HRS)	.42	1.136	.187	.75
KCARL2 vs. KCARL3	1.632	1.136*	2.828	2.913
KCARL2 vs. KCAR (24HRS)	.527	1.136	.295	.941

* Significant at 95%

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
KCARL3 vs. KCAR (24HRS)	-1.105	1.136	1.296	1.972

KCARL FLEXION/EXTENSION

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Analysis of Variance Table

Source:	DF:	Sum Squares:	Mean Square:	F-test:
Between groups	3	67.362	22.454	4.371
Within groups	36	184.927	5.137	p = .0101
Total	39	252.289		

Model II estimate of between component variance = 1.732

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
KCARL1	10	49.206	1.073	.339
KCARL2	10	47.647	3.506	1.109
KCARL3	10	47.413	2.115	.669
KCAR (24HRS)	10	50.628	1.622	.513

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
KCARL1 vs. KCARL2	1.559	2.056	.789	1.538
KCARL1 vs. KCARL3	1.793	2.056	1.043	1.769
KCARL1 vs. KCAR (24HRS)	-1.422	2.056	.656	1.403
KCARL2 vs. KCARL3	.234	2.056	.018	.231
KCARL2 vs. KCAR (24HRS)	-2.981	2.056*	2.883*	2.941

* Significant at 95%

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
KCARL3 vs. KCAR (24HRS)	-3.215	2.056*	3.354*	3.172

* Significant at 95%

DALSTE TIBIAL ROTATION

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Analysis of Variance Table

Source:	DF:	Sum Squares:	Mean Square:	F-test:
Between groups	3	.623	.208	.332
Within groups	36	22.504	.625	p = .802
Total	39	23.128		

Model II estimate of between component variance = -.042

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
DALSTE1	10	4.25	.837	.265
DALSTE2	10	4.098	.858	.271
DALSTE3	10	4.447	.677	.214
DALST (24HRS)	10	4.227	.778	.246

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
DALSTE1 vs. DALSTE2	.152	.717	.062	.43
DALSTE1 vs. DALSTE3	-.197	.717	.103	.557
DALSTE1 vs. DALST (24H...	.023	.717	.001	.065
DALSTE2 vs. DALSTE3	-.349	.717	.325	.987
DALSTE2 vs. DALST (24H...	-.129	.717	.044	.365

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
DALSTE3 vs. DALST (24H...	.22	.717	.129	.622

DALSTE ABDUCTION/ADDUCTION

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Analysis of Variance Table

Source:	DF:	Sum Squares:	Mean Square:	F-test:
Between groups	2	8.118	4.059	9.701
Within groups	27	11.297	.418	p = .0007
Total	29	19.415		

Model II estimate of between component variance = .364

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
DALSTE1	10	13.735	.461	.146
DALSTE2	10	12.632	.783	.247
DALSTE3	10	12.631	.656	.207

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
DALSTE1 vs. DALSTE2	1.103	.594*	7.269*	3.813
DALSTE1 vs. DALSTE3	1.104	.594*	7.283*	3.816
DALSTE2 vs. DALSTE3	.001	.594	5.975E-6	.003

* Significant at 95%

DALSTE FLEXION/EXTENSION

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Analysis of Variance Table

Source:	DF:	Sum Squares:	Mean Square:	F-test:
Between groups	3	78.119	26.04	14.87
Within groups	36	63.042	1.751	p = .0001
Total	39	141.161		

Model II estimate of between component variance = 2.429

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
DALSTE1	10	41.763	1.143	.361
DALSTE3	10	40.233	2.059	.651
DALSTE2	10	44.118	1.037	.328
DALST (24HRS)	10	41.588	.62	.196

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
DALSTE1 vs. DALSTE3	1.53	1.2*	2.228	2.585
DALSTE1 vs. DALSTE2	-2.355	1.2*	5.278*	3.979
DALSTE1 vs. DALST (24H...	.175	1.2	.029	.296
DALSTE3 vs. DALSTE2	-3.885	1.2*	14.365*	6.565
DALSTE3 vs. DALST (24H...	-1.355	1.2*	1.747	2.29

* Significant at 95%

One Factor ANOVA X 1 : TRIALS Y 1 : RANGE

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
DALSTE2 vs. DALST (24H...	2.53	1.2*	6.092*	4.275

* Significant at 95%

APPENDIX 11

Power analysis

POWER ANALYSIS

the alpha level is .05

the test statistic to be used is $\Delta/Ó$

where; Δ = the smallest difference to be detected with high probability

and Ó = the standard deviation

- for this study the smallest difference to detect was set at 1.5°
- the standard deviation of the tibial rotation data in the pilot was .6°. Since it was determined that this would probably increase with a larger sample size, both 1° and 1.5° were used.
- at this point tables were consulted to determine the number of subjects needed

$\Delta/Ó = 1.5/1 = 1.5$

POWER	number of subjects needed
.70	8
.80	10
.90.	13
.95	15

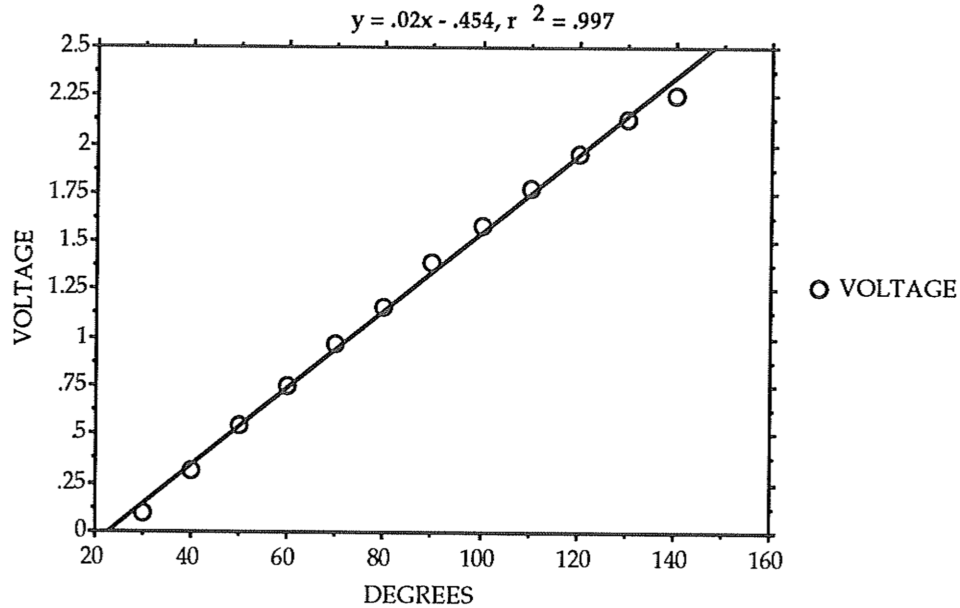
$\Delta/Ó = 1.5/1.5 = 1$

POWER	number of subjects needed
.70	17
.80	21

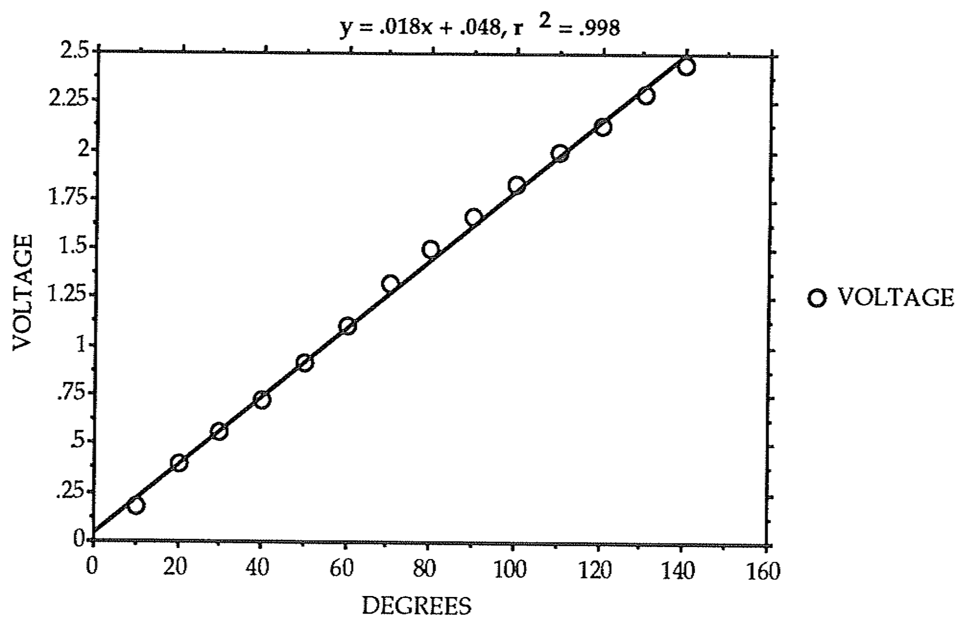
APPENDIX 12

Pre- and post-test potentiometer linearity check

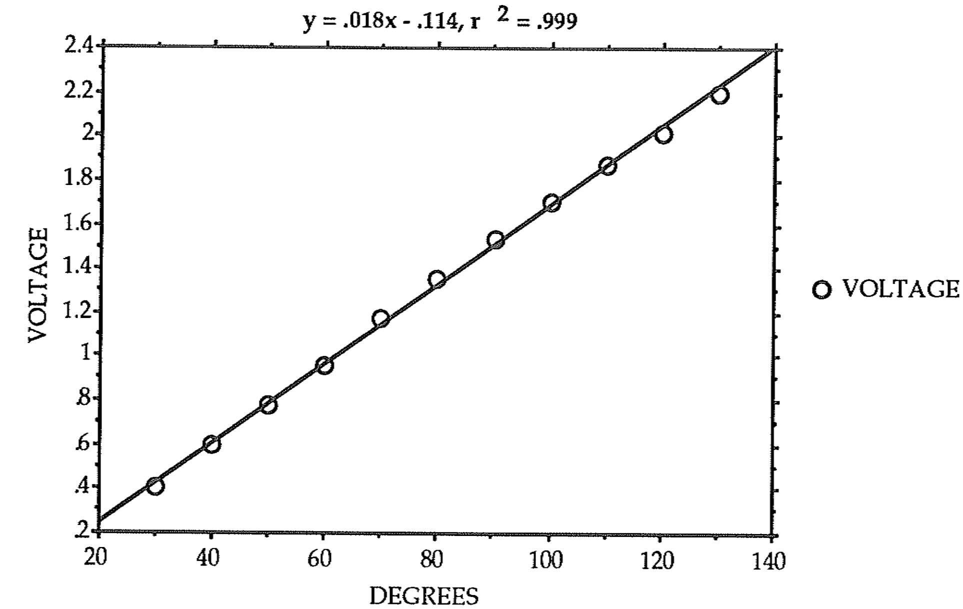
POTENTIOMETER 1
PRE-TEST



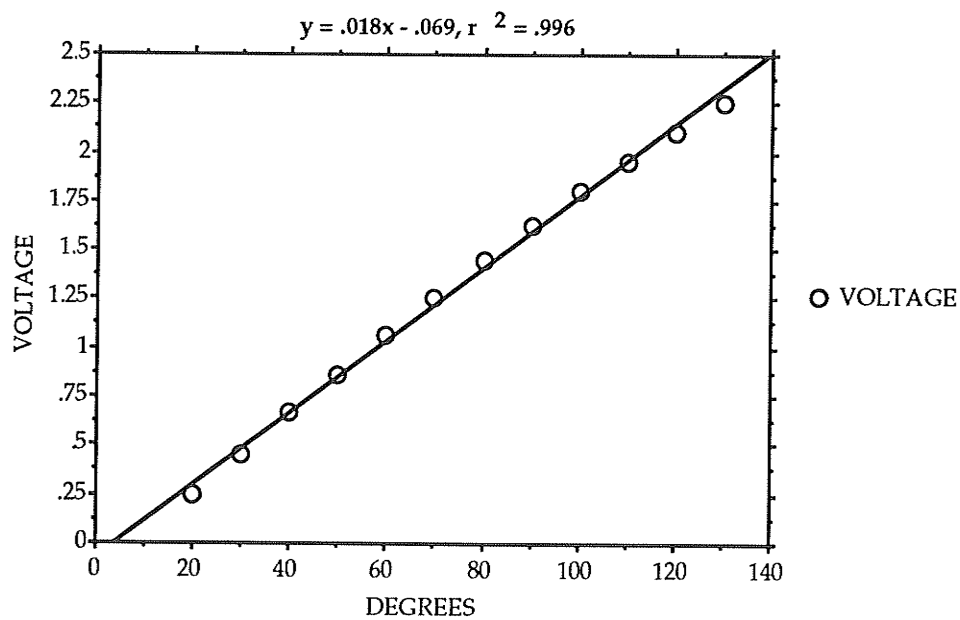
POTENTIOMETER 2
PRE-TEST



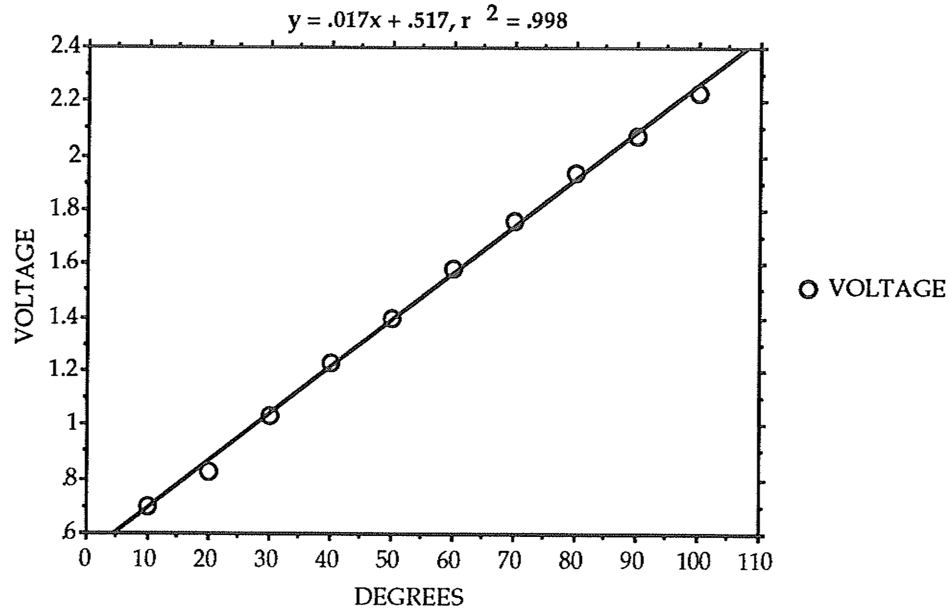
POTENTIOMETER 3
PRE-TEST



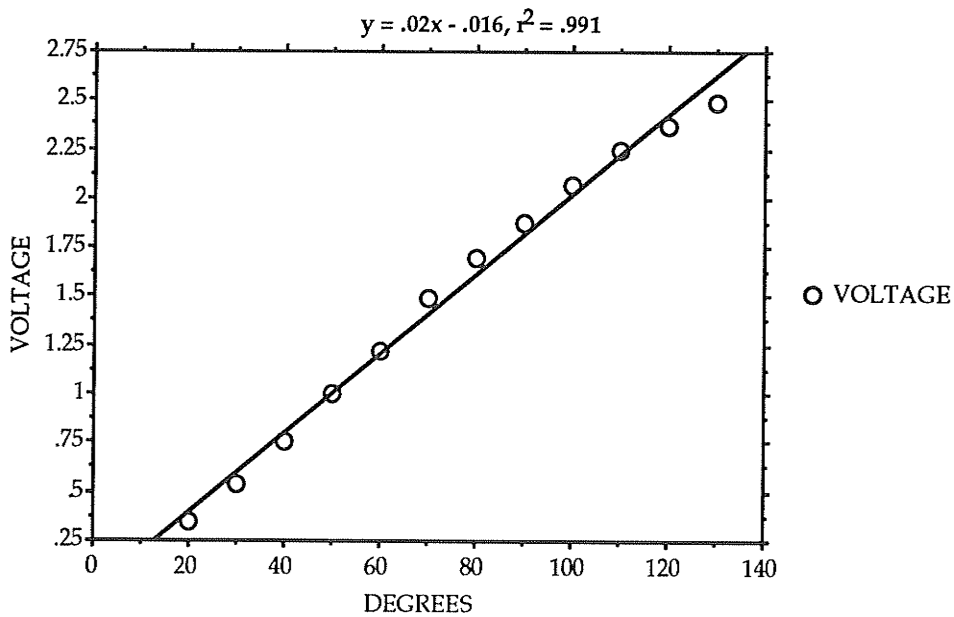
POTENTIOMETER 4
PRE-TEST



POTENTIOMETER 5
PRE-TEST

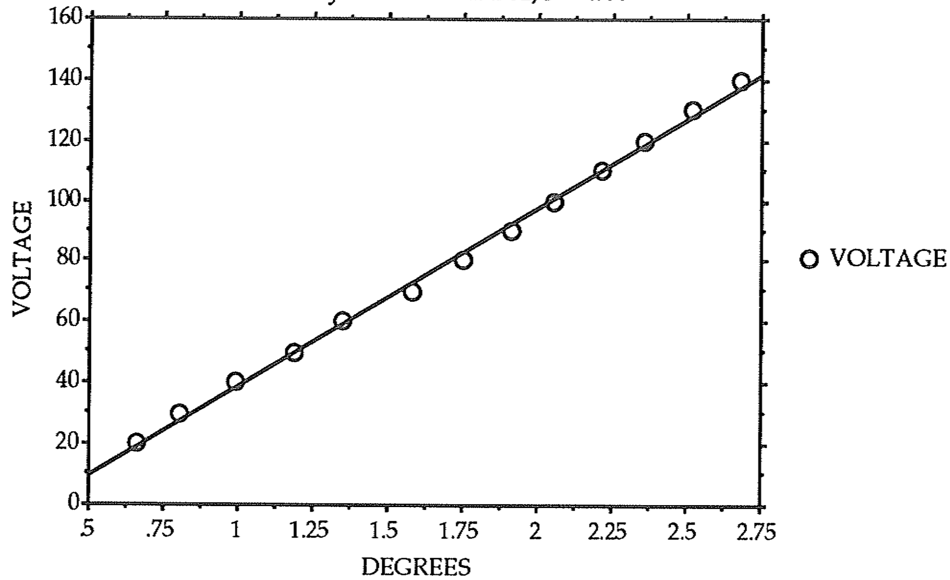


POTENTIOMETER 1
POST-TEST



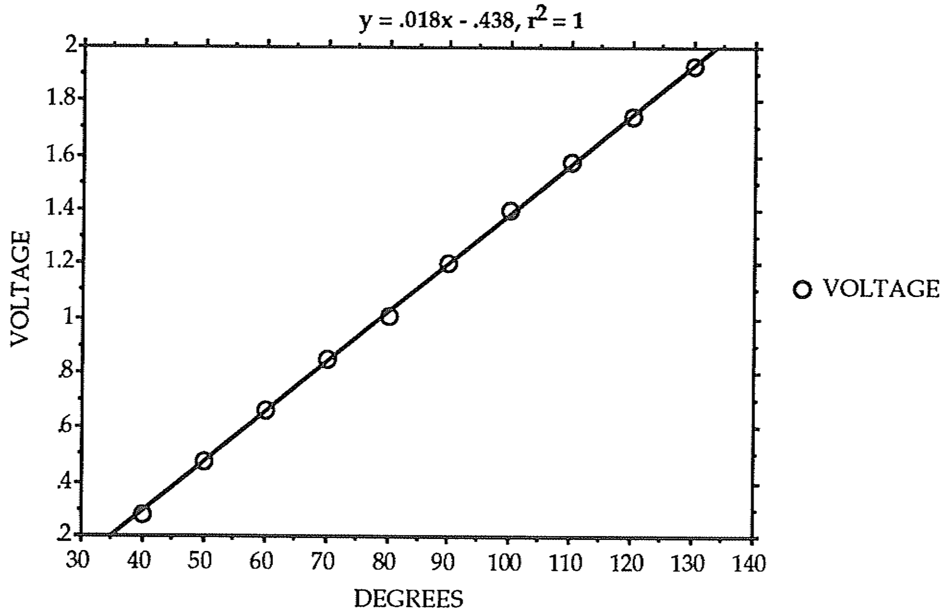
POTENTIOMETER 2
POST-TEST

$y = 58.581x - 19.362, r^2 = .997$



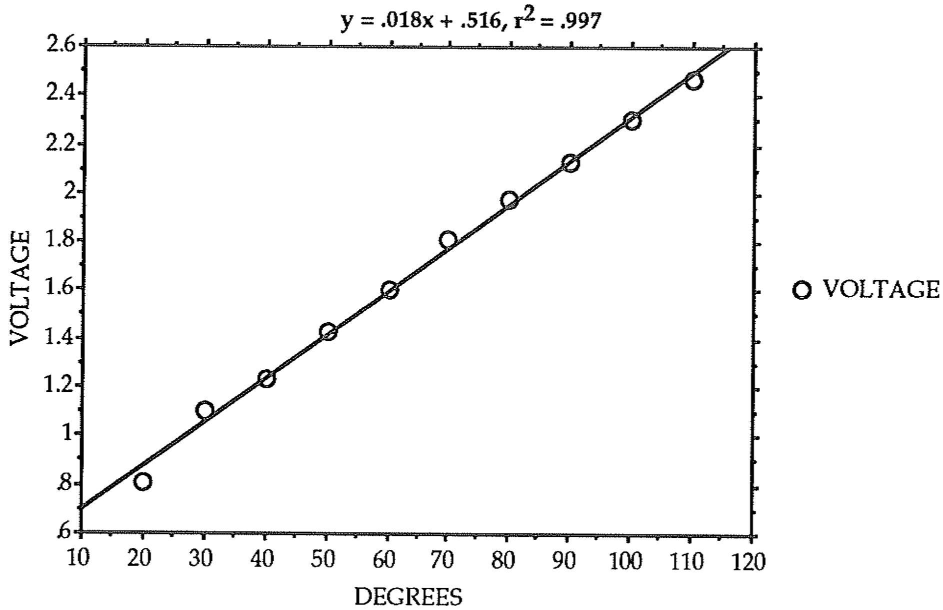
POTENTIOMETER 3
POST-TEST

$y = .018x - .438, r^2 = 1$



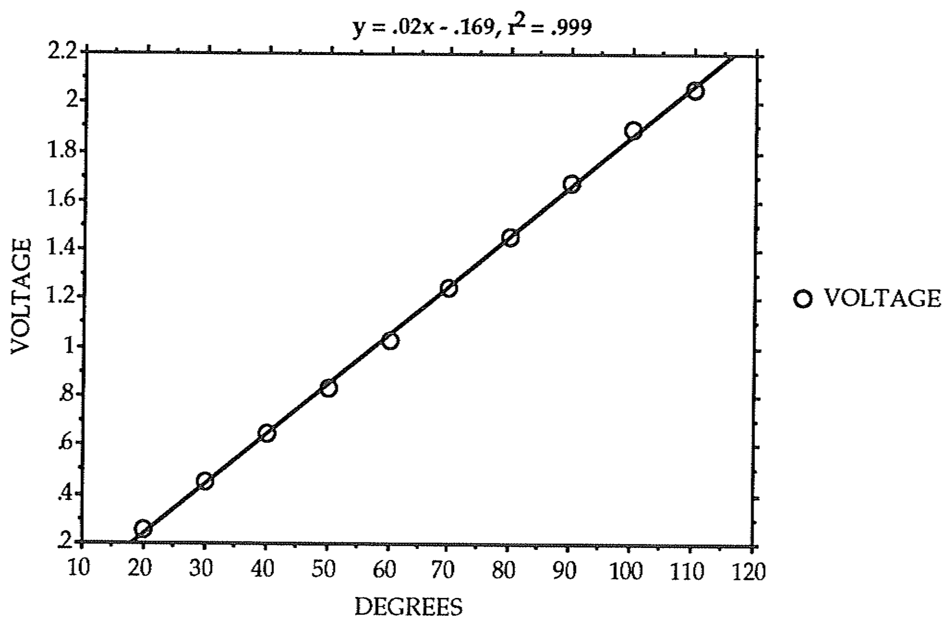
POTENTIOMETER 4

POST-TEST



POTENTIOMETER 5

POST-TEST



APPENDIX 13

H2H ANOVA tables

TIBIAL ROTATION H2H

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between subjects	19	294.36	15.493	7.119	.0001
Within subjects	40	87.047	2.176		
treatments	2	5.72	2.86	1.336	.2749
residual	38	81.327	2.14		
Total	59	381.407			

Reliability Estimates for- All treatments: .86 Single Treatment: .671

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
TRIAL1 - TIB ROT	20	9.268	2.787	.623
TRAIL2 - TIB ROT	20	9.589	2.828	.632
TRIAL3 - TIB ROT	20	8.835	2.002	.448

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
TRIAL1 - ... vs. TRAIL2 -...	-.32	.937	.24	.693
TRIAL1 - ... vs. TRIAL3 -...	.433	.937	.438	.936
TRAIL2 - ... vs. TRIAL3 -...	.753	.937	1.326	1.629

ABDUCTION/ADDUCTION H2H

One Factor ANOVA-Repeated Measures for $X_1 \dots X_3$

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between subjects	19	419.215	22.064	5.673	.0001
Within subjects	40	155.562	3.889		
treatments	2	21.16	10.58	2.991	.0622
residual	38	134.402	3.537		
Total	59	574.776			

Reliability Estimates for- All treatments: .824 Single Treatment: .609

One Factor ANOVA-Repeated Measures for $X_1 \dots X_3$

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
TRIAL1 - AB/AD	20	11.623	2.57	.575
TRIAL2 - AB/AD	20	11.537	2.457	.549
TRIAL3 - AB/AD	20	12.838	4.062	.908

One Factor ANOVA-Repeated Measures for $X_1 \dots X_3$

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
TRIAL1 - ... vs. TRIAL2 -...	.087	1.204	.011	.146
TRIAL1 - ... vs. TRIAL3 -...	-1.214	1.204*	2.083	2.041
TRIAL2 - ... vs. TRIAL3 -...	-1.301	1.204*	2.393	2.188

* Significant at 95%

FLEXION/EXTENSION H2H

One Factor ANOVA-Repeated Measures for $X_1 \dots X_3$

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between subjects	19	1924.359	101.282	11.991	.0001
Within subjects	40	337.86	8.446		
treatments	2	28.106	14.053	1.724	.192
residual	38	309.754	8.151		
Total	59	2262.218			

Reliability Estimates for- All treatments: .917 Single Treatment: .786

One Factor ANOVA-Repeated Measures for $X_1 \dots X_3$

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
TRIAL1 - F/E	20	46.511	6.011	1.344
TRIAL2 - F/E	20	47.536	6.723	1.503
TRIAL3 - F/E	20	48.173	6.021	1.346

One Factor ANOVA-Repeated Measures for $X_1 \dots X_3$

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
TRIAL1 - F... vs. TRIAL2 ...	-1.025	1.828	.644	1.135
TRIAL1 - F... vs. TRIAL3 ...	-1.661	1.828	1.693	1.84
TRIAL2 - F... vs. TRIAL3 ...	-.637	1.828	.249	.706

APPENDIX 14

H2T ANOVA tables

TIBIAL ROTATION H2T

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between subjects	16	342.405	21.4	14.146	.0001
Within subjects	34	51.437	1.513		
treatments	2	2.094	1.047	.679	.5142
residual	32	49.343	1.542		
Total	50	393.842			

Reliability Estimates for- All treatments: .929 Single Treatment: .814

Note: 3 cases deleted with missing values.

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
TRIAL1 - TIB ROT	17	6.959	2.855	.693
TRAIL2 - TIB ROT	17	7.143	3.091	.75
TRIAL3 - TIB ROT	17	6.652	2.604	.631

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
TRIAL1 - ... vs. TRAIL2 -...	-.184	.868	.093	.431
TRIAL1 - ... vs. TRIAL3 -...	.308	.868	.261	.722
TRAIL2 - ... vs. TRIAL3 -...	.491	.868	.665	1.153

ABDUCTION/ADDUCTION H2T

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between subjects	16	197.016	12.314	4.669	.0001
Within subjects	34	89.659	2.637		
treatments	2	2.8	1.4	.516	.6019
residual	32	86.858	2.714		
Total	50	286.675			

Reliability Estimates for- All treatments: .786 Single Treatment: .55

Note: 3 cases deleted with missing values.

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
TRIAL1 - AB/AD	17	7.359	2.446	.593
TRAIL2 - AB/AD	17	7.319	1.767	.429
TRIAL3 - AB/AD	17	7.835	2.939	.713

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
TRIAL1 - ... vs. TRAIL2 -...	.04	1.151	.003	.071
TRIAL1 - ... vs. TRIAL3 -...	-.476	1.151	.355	.842
TRAIL2 - ... vs. TRIAL3 -...	-.516	1.151	.417	.913

FLEXION/EXTENSION H2T

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between subjects	16	1873.229	117.077	5.958	.0001
Within subjects	34	668.061	19.649		
treatments	2	3.003	1.501	.072	.9305
residual	32	665.059	20.783		
Total	50	2541.29			

Reliability Estimates for- All treatments: .832 Single Treatment: .623

Note: 3 cases deleted with missing values.

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
TRIAL1 - F/E	17	24.543	8.308	2.015
TRIAL2 - F/E	17	23.969	7.102	1.722
TRIAL3 - F/E	17	24.121	6.259	1.518

One Factor ANOVA-Repeated Measures for X_1 ... X_3

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
TRIAL1 - F... vs. TRIAL2 ...	.574	3.185	.067	.367
TRIAL1 - F... vs. TRIAL3 ...	.422	3.185	.036	.27
TRIAL2 - F... vs. TRIAL3 ...	-.152	3.185	.005	.097

APPENDIX 15

Stride and stance time ANOVA tables

STRIDE TIME

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between subjects	19	.481	.025	25.817	.0001
Within subjects	40	.039	.001		
treatments	2	.007	.004	4.413	.0189
residual	38	.032	.001		
Total	59	.52			

Reliability Estimates for- All treatments: .961 Single Treatment: .892

Note: 42 cases deleted with missing values.

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
TRIAL 1	20	1.186	.086	.019
TRIAL 2	20	1.204	.094	.021
TRIAL3	20	1.212	.104	.023

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
TRIAL 1 vs. TRIAL 2	-.019	.019*	2.086	2.043
TRIAL 1 vs. TRIAL3	-.026	.019*	4.174*	2.889
TRIAL 2 vs. TRIAL3	-.008	.019	.358	.847

* Significant at 95%

STANCE TIME

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between subjects	16	.188	.012	12.081	.0001
Within subjects	34	.033	.001		
treatments	2	6.075E-5	3.037E-5	.03	.971
residual	32	.033	.001		
Total	50	.221			

Reliability Estimates for- All treatments: .917 Single Treatment: .787

Note: 5 cases deleted with missing values.

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Group:	Count:	Mean:	Std. Dev.:	Std. Error:
TRIAL1 - stance ...	17	.744	.068	.017
TRIAL2 - stance ...	17	.745	.059	.014
TRIAL3 - stance ...	17	.742	.075	.018

One Factor ANOVA-Repeated Measures for $X_1 \ldots X_3$

Comparison:	Mean Diff.:	Fisher PLSD:	Scheffe F-test:	Dunnett t:
TRIAL1 - s... vs. TRIAL2 ...	-.001	.022	.004	.091
TRIAL1 - s... vs. TRIAL3 ...	.002	.022	.011	.15
TRIAL2 - s... vs. TRIAL3 ...	.003	.022	.029	.241

APPENDIX 16

Literature values of the three rotational parameters

LITERATURE INTERNAL/EXTERNAL ROTATION VALUES

AUTHOR	SUBJECTS	VELOCITY	MAX VALUE	OCCURRED
LaFortune et al.	n=3 ? males or females	3.36 MPH	14°	stance
Marans et al.	n=50 males and females	self selected	9.2°(3.7) males 8.9°(4.1) females	swing
Kettlekamp et al.	n=32 males and females	self selected	12.9°(4.41) test 13.1°(4.39) re-test	swing
Chao et al.	n=21 males n=20 females	self selected 2.77 MPH males 2.51 MPH females	9°(3) males 10°(3) females in stance	14°(4) males 14°(4) females in swing
Isacson et al.	n=17 females	s/s 2.69MPH imposed 1.34 MPH	9°(2) s/s 7°(4) imposed	swing
Czerniecki et al.	n=20 males and females	3.13 MPH 4.9 MPH 5.8 MPH	11.3°(4) 12.7°(4.8) 14.8°(6.4)	stance

s/s - self selected speed

LITERATURE ABDUCTION/ADDUCTION VALUES

AUTHOR	SUBJECTS	VELOCITY	MAX VALUE	OCCURRED
LaFortune et al.	n=3 ? males or females	3.36 MPH	6.5°	swing 0° in stance
Marans et al.	n=50 males and females	self selected	3.9°(1.5) control 3.6°(2.1) unaffected	swing
Kettlekamp et al.	n=32 males and females	self selected	9.7(3.56) test 10.5(4.41) re-test	swing
Chao et al.	n=21 males n=20 females	self selected 2.77 MPH males 2.51 MPH females	7°(2) males 7°(2) females in stance	12°(4) males 10°(4) females in swing
Isacson et al.	n=17 females	s/s 2.69MPH imposed 1.34 MPH	s/s 8°(2) 7°(3) imposed	over whole stride

s/s - self selected speed

LITERATURE FLEXION/EXTENSION VALUES

AUTHOR	SUBJECTS	VELOCITY	MAX VALUE	OCCURRED
LaFortune et al.	n=3 ? males or females	3.36 MPH	67°	swing phase
Marans et al.	n=50 males and females	self selected	46.8° (6) control 48.2° (6.6) unaffected leg	swing phase
Kettlekamp et al.	n=32 males and females	self selected	68.1° (6.47) swing	20.6°(4.4) stance
Chao et al.	n=21 males n=20 females	self selected 2.77 MPH males 2.51 MPH females	72°(6) males 66°(9) females swing	32°(6) males 30°(6) females stance
Isacson et al.	n=17 females	s/s 2.69MPH imposed 1.34 MPH	58°(3) s/s 55(5) imposed	swing

s/s - self selected speed