ANALYSIS OF THE KINETIC, KINEMATIC AND TEMPOROSPATIAL PARAMETERS OF GAIT, AND THEIR RELATIONSHIP TO FUNCTIONAL AMBULATION FOLLOWING TOTAL KNEE REPLACEMENT USING TWO DIFFERENT PROSTHETIC DESIGNS

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ABSTRACT

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The aim of this research is to assess the efficacy of two different prosthetic designs used in total knee replacement operations in osteoarthritic patients. The study involved the analysis of the kinetic, kinematic and temporospatial parameters of gait, as well as the assessment of pain and the physiological cost of walking and functional activities.

This study involves examining the joint range of motion, moment, power and lower muscle activity by using an optoelectronic motion analysis system, a Kistler force plate and telemetered electromyography. First, 54 osteoarthritic subjects were studied one week before total knee replacement (TKR) to compare them with 25 healthy control subjects in order to establish the baseline data. The second concern of this study was to follow up these patients 3, 6 and 9 months following TKR using the same protocol. Fifteen patients received a posterior cruciate ligament (PCL) sacrificing prosthesis and thirty-nine patients had a PCL-retaining prosthesis.

The temporospatial gait parameters in both groups showed no significant differences between the two types of prosthesis during the study period. The range of knee motion during the stance and swing phase of gait was similar for both knee replacement groups, but increased progressively post-operatively in both groups, the range of motion being greatest at 9 months after operation. Knee moments in mid-stance for the patients with implants with PCL-retaining prosthesis was not statistically different from the PCL-sacrificing prosthesis until 9 months after surgery. The knee power at mid-stance in both groups revealed no significant differences at any stage. Abnormal muscle activity was noted during level walking in four lower limb muscles at 3 and 6 months after TKR, but phasic activity was noted at 9 months after TKR. All the gait parameters and electromyography improved 9 months after surgery but did not reach the values of control subjects. The clinical scores improved at 3, 6 and 9 months after surgery compared with the preoperative value. The pain scores decreased significantly at 3, 6 and 9 months after surgery compared with the preoperative value.

The main conclusion from this study is that prosthetic knee replacement produces a significant improvement in gait and symptoms. Posterior cruciate ligament sacrificing and retaining prostheses resulted in similar improvements in the temporospatial and kinematic parameters of gait over entire period of study. The knee moments in mid-stance were lower post-operatively in subjects with a PCL-sacrificing prosthesis. This was statistically significant at 3 and 6 months but not at 9 months, thus conferring potential advantage on the subjects.

AC	Acceleration
ACL	Anterior Cruciate Ligament
Ant. Fem	Anterior Femoral
Ant. Tib	Anterior Tibial
AROM	Active Range of Motion
ASE	Active Surface Electrode
ASIS	Anterior Superior Iliac Spine
CKRS	Cincinnati Knee Rating Scale
CM	Centimetre
CODA	Cartesian Optoelectronic Dynamic Anthropometer
COG	Centre of Gravity
CPM	Continuous Passive Motion
CVA	Cerebral Vascular Accident
DB	Decibels =20 log10 (value)
DE	Deceleration
DJD	Degenerative Joint Disease
DSP	Digital Signal Processors
EMG	Electromyography
FF	Foot-flat
GRF	Ground Reaction Force
HO	Heel-Off
HS	Heel-Strike
HSS	Hospital for Special Surgery
IC	Initial Contact
IQR	Inter-Quartile Range
ISW	Initial Swing
KSS	Knee Society scale
L	Left
LEDs	Light Emitting Diodes
LH	Lateral Hamstring
LR	Loading-Response

MBK	Mobile Bearing Knee
MH	Medial Hamstring
MADFSW	Maximum Ankle Dorsiflexion in Swing
MAMMS	Maximum Ankle Moment in Stance
MAPMS	Maximum Ankle Power in Stance
MHEST	Maximum Hip Extension in Stance
MKFLR	Maximum Knee Flexion in Loading Response
MKFSW	Maximum Knee Flexion in Swing
MKMMST	Maximum Knee Moment in Stance
MKPMST	Maximum Knee Power in Stance
MST	Mid-Stance
MSW	Mid-Swing
N	Newtons (unit of force)
OA	Osteoarthritis
OGA	Observational Gait Analysis
PC	Personal Computer
PCI	Physiological Cost Index
PCL	Posterior Cruciate Ligament
PCLS	Posterior Cruciate Ligament Sacrificing
Pos. Fem	Posterior Femoral
Pos. Tib	Posterior Tibial
PCLR	Posterior Cruciate Ligament Retaining
PFC	Press-Fit Condylar
PSIS	Posterior Superior Iliac Spine
PSW	Pre-Swing
PSE	Passive Surface Electrode
R	Right
RA	Rheumatoid Arthritis
RF	Rectus Femoris Muscle
SD	Standard Deviation
ROM	Range of Motion
ST	Stance

SW	Swing
TKR	Total Knee Replacement
TO	Toe-Off
TST	Terminal Stance
TSW	Terminal Swing
UHMWPE	Ultra-High Molecular Weight Polyethylene
VAS	Visual Analogue Scale
VL	Vastus Lateralis Muscle
VM	Vastus Medialis Muscle
W	Watts (Unit of power)

Preface

The initial baseline data of this work has been presented in the International Symposium on Gait Disorders, Prague Czech Republic, 4-6 September 1999.

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CHAPTER ONE

INTRODUCTION

Prosthetic joint replacement is becoming increasingly important in the surgical treatment of the arthritic knee. Many types of prosthesis are now available. Total knee replacement (TKR) has become an established method of treatment of severe degenerative disease of the knee joint to reduce pain and to restore function and mobility *(Miller and Carr 1993)*.

Constant updating and innovation in the design of prostheses for TKR surgery have given rise to arguments about the use of prostheses as replacements for normal ligaments, and the impact that these have had on human gait. This is probably due to the difficulty in measuring the effect of TKR on gait in routine clinical practice.

The procedures used currently for TKR fall into two categories: posterior cruciate ligament (PCL)-retaining or PCL-sacrificing operations.

Those who argue in favour of PCL-retention take the view that the natural ligament following the operation will continue to provide the patient with knee stability and a reasonable quality of locomotion. Those who disagree do so on the grounds that modern prostheses have been developed which can successfully replace the function of the PCL with a posterior stabilizing articular surface (plastic cam), without decreasing the quality of the patient's locomotion. It is still a controversial question, however, as to whether the PCL should be retained or substituted during knee replacement surgery.

Because of this controversy we propose to look at both arguments objectively by using a three-dimensional motion analysis system, a force plate and electromyography to measure functional knee motion, knee moment and power and lower limb muscle activity during gait in patients with retention of the PCL and in those with a PCL-sacrificing TKR design.

By analysing the gait of subjects at regular intervals following their operation, and using a control group of healthy subjects who had no operation, we hope to provide new information for surgeons to aid in the decision on whether to retain or sacrifice the PCL.

1.1. Organisation of the thesis

The thesis is presented in five chapters. The first chapter presents a description of the human gait cycle and its various events. The six determinants of gait, and the characteristics of the adult gait are summarised. The remaining part of this chapter gives a brief summary of the methods of gait analysis, section 1.7 reviewing methods of gait analysis and section 1.8 reviewing the historical background of gait analysis and the advantages and limitations of different measurement systems.

Chapter two gives a brief summary of the structure and function of the knee joint, and the osteoarthritic knee. This is followed by a historical review of knee replacement and outlines some of the controversies still current over choice of prosthesis and the types of prostheses currently used for knee replacement, with particular emphasis on the advantages and disadvantages of PCL-retaining and PCL-sacrificing. In addition, the literature on gait analysis in patients with osteoarthritis of the knee, and gait analysis before and after TKR will be reviewed. This literature review leads to the formulation of specific hypotheses, which are tested in the experimental work reported in the subsequent chapters. Summary is given at the end of this chapter.

Chapter three describes a pilot study carried out to examine the intra-investigator reliability of the CODA mpx30 scanner system and the best test conditions.

Chapter four sets out the plan of the experimental study, and describes the materials and methods of data collection. This includes a clinical description of the patients and a description of the inclusion and exclusion criteria employed to select the study sample. The range of measurements used in the study is described, including the visual analogue scale and the Cincinnati knee rating scale. The chapter also describes the use of gait analysis equipment, such as the CODA mpx30 motion analysis system, the Kistler force plate, the electromyograph (EMG), and the recording procedures. The types of prostheses used, the surgical technique for sacrificing and retaining PCL and post-operative management of the patient following TKR are outlined. The methods of data analysis are described.

Chapter five presents the complete results of the main study, including the kinematic of the hip and kinematic and kinetic data for the ankle and knee in control subjects and osteoarthritic patients. A pre-operative comparison between the three different types of prostheses is reported. The effect of TKR on gait using the three different types of prosthesis, at 3, 6 and 9 months following TKR is discussed. The two groups of PCL-sacrificing and PCL-retaining TKR are also compared, as well as two groups operated on by the same surgeon. The chapter also reports the self assessment of pain and function and the physiological cost index result of all patients at 3, 6, and 9 months following total TKR and interprets the results.

Chapter six discusses the findings and presents the conclusions and suggestions for future research.

1.2. Walking and gait

Walking can be defined as a 'method of locomotion involving the use of the two legs, alternately, to provide both support and propulsion, with at least one foot being in contact with the ground at all times' *(Whittle 1996a).*

Gait can be defined as, 'the manner of moving the body from one place to another by alternately and repetitively changing the location of the feet, with the condition that at least one foot is in contact with the walking surface' *(Smidt 1990)*.

1.3. Description of the gait cycle

This section contains a description of the main features of the gait cycle. The gait cycle is defined as, 'the time interval between two successive occurrences of one of the repetitive events of walking', e.g. heel strikes of the same foot *(Whittle 1996a)*. The period from the heel strike of one leg to the next heel strike of the same leg equals 100 per cent of the gait cycle. The normal gait cycle is generally divided into the stance (ST) phase (weight bearing period) during which the foot is on the ground, and the swing (SW) phase during which the foot is in the air. The ST occupies approximately 60% and the SW phase 40% of the gait cycle in normal walking.

1.3.1. Phases of the gait cycle

The phases of the gait cycle have been described by *Inman et al (1981)* and *Perry (1992)*. The two descriptions are similar in many respects. The new nomenclature tends to describe the phases of the cycle in terms of function (for example loading response) whereas the former sees the cycle as a series of actions (one of which is heel strike for example). The major advantage of the new nomenclature is that it applies equally well to healthy subjects and pathological cases (for example patients whose gait pattern does not include heel strike on one side or even bilaterally). Table 1.1 illustrate the old nomenclature description and its relationship to new nomenclature terminology.

Table 1.1 Old nomenclature and new nomenclature gait component terminology.

Old nomenclature	New nomenclature
Heel-strike	Initial contact
Foot-flat	Loading response
Mid-stance	Mid-stance
Heel-off	Terminal stance
Toe-off	Pre-swing
Acceleration	Initial swing
Mid-swing	Mid-swing
Deceleration	Terminal-swing

Old nomenclature:

Heel-strike (HS) (0 %) is measured at the moment when the heel makes initial contact with the ground. Foot-flat (FF) (0-10 %) starts when the whole of the foot makes contact with the ground. Mid-stance (MST) (10-30 %) occurs when the centre of gravity passes directly over the supporting limb. Heel-off (HO) (30-50 %) occurs as the body's centre of gravity moves directly over the supporting limb and continues to move forward contributing to the lifting of the heel. Toe-off (TO) (60 %) occurs at the instant when the toe loses contact with the ground. Acceleration (AC) (60-70 %) is that part of the phase in which the limb is propelled forwards beyond the body's centre of gravity. Mid-swing (MSW) (70-80 %) occurs when the leg is directly beneath the body. Deceleration (DE) (80-100 %) starts when the forward motion of the leg is restrained to control the position of the foot before the next heel strike (Figure 1.1).

Spatial and temporal parameters are used to describe normal gait, including double support, single support, stride length, step length, velocity, and cadence. Double support is the instance when both feet are touching the ground at the same time.

The single support phase begins when the toe of the opposite limb leaves the ground. In normal walking, in one complete gait cycle, the double support phase and single support occur as follows: double support → right single support → double support → left single support → double support.

Stride length is the linear distance between two successive initial contacts of the same leg. The distance from the initial contact of one lower extremity to the initial contact of the opposite extremity is the step length. In each gait cycle during normal walking, there are two step lengths for each stride length. The velocity of walking is the distance covered by the whole body in a given time. Cadence is the number of steps taken in one minute.

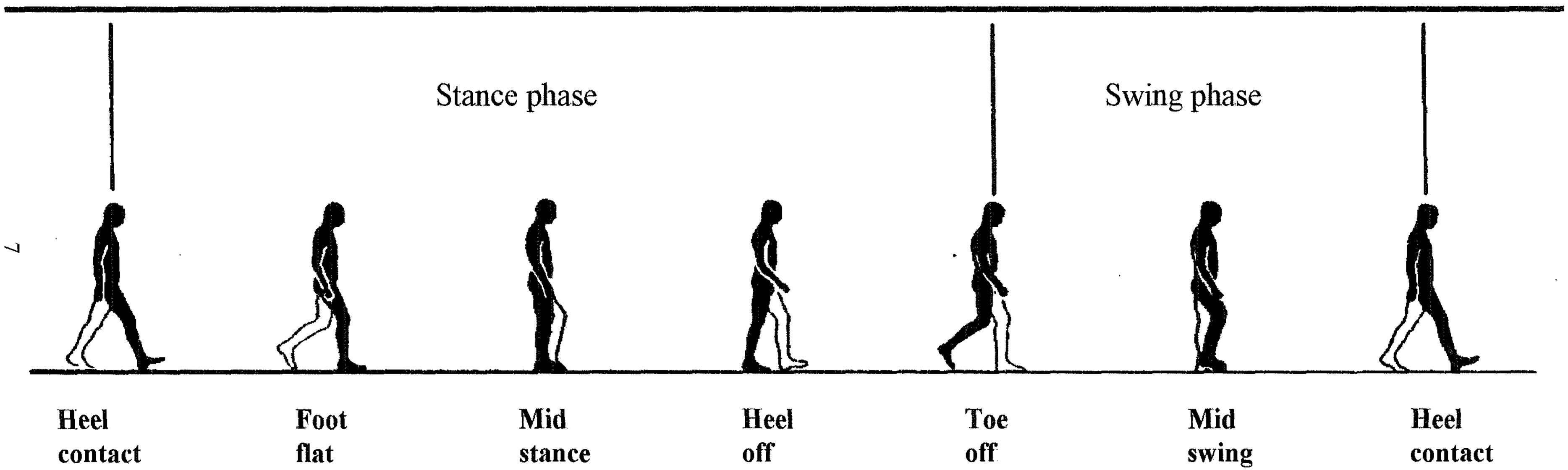


Figure 1-1 Phases of the gait cycle. Old nomenclature (Adapted from Whittle 1991).

New nomenclature:

Initial contact (IC) (0-2%) is measured at the moment when the foot makes contact with the ground (Figure 1.2). Loading response (LR) (0%-10%) (double support period): The phase begins with initial floor contact and continues until the other foot is lifted for swing (Figure 1.3). Mid-stance (MST) (10%-30%): It begins as the other foot extends to neutral (Figure 1.4). Terminal stance (TST) (30%-50%): It begins with heel rise and continues until the other foot makes contact with the ground (Figure 1.5). Pre-swing (PSW) (50%-60%) (double stance period): It begins with initial contact of the opposite limb and ends with ipsilateral toe-off (Figure 1.6). Initial swing (ISW) (60%-73%): It begins with a lift of the foot from the floor and ends when the swinging foot is opposite the stance foot (Figure 1.7). Mid-swing (MSW) (73%-87%): During this period the foot is in the air (Figure 1.8). Terminal swing (TSW) (87%-100%): This completes the gait cycle. The limb decelerates in preparation for the start of the next cycle (Figure 1.9).

Figures 1.2-1.9 show the force vector during the initial contact transient, at loading response, at mid-stance (when the swing phase leg passes the stance phase leg), at terminal stance and at pre-swing.

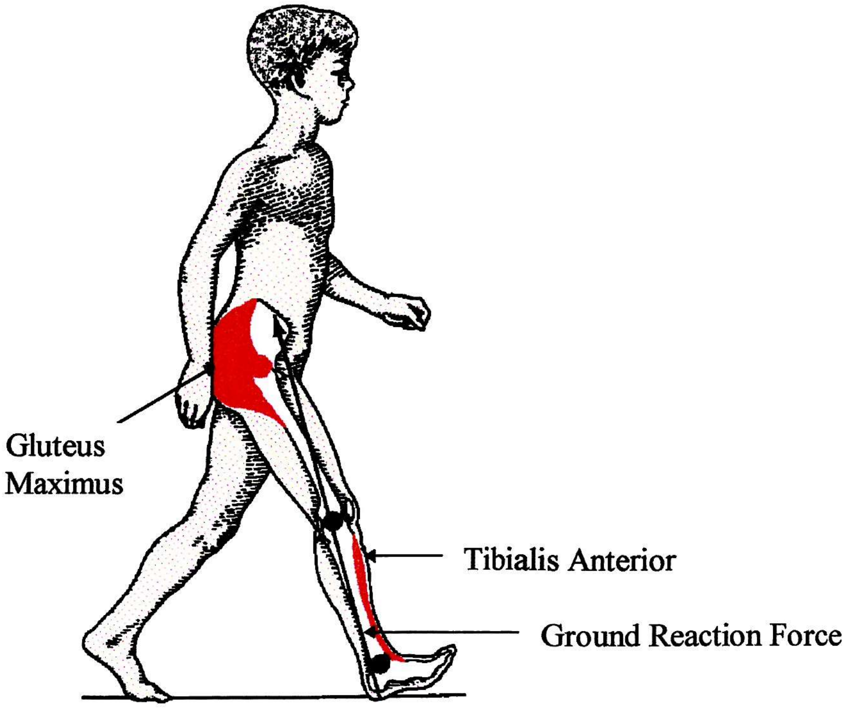


Figure 1.2 Initial contact. (Adapted from Gage JR 1990).

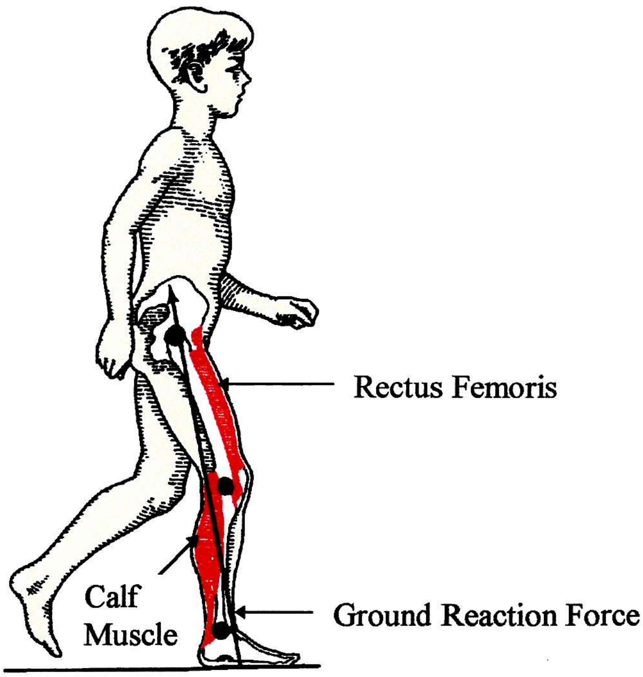


Figure 1.3 Loading response. (Adapted from Gage JR 1990).

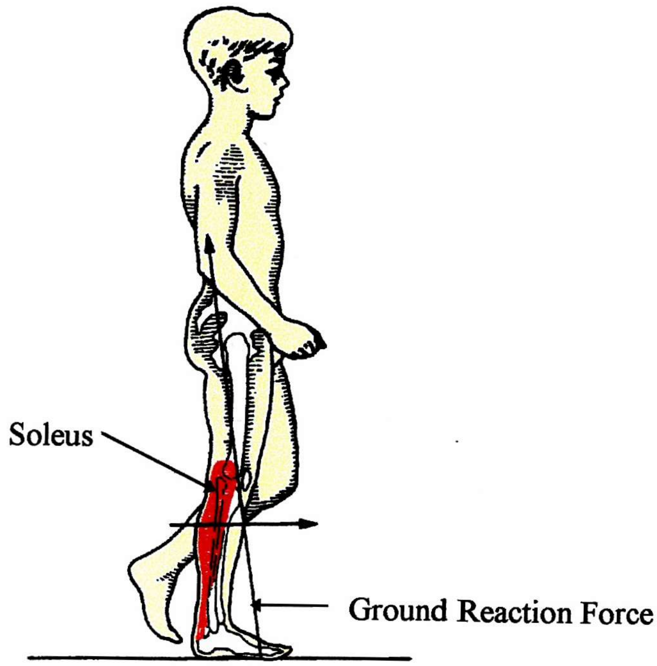


Figure 1.4 Mid-stance. (Adapted from Gage JR 1990).

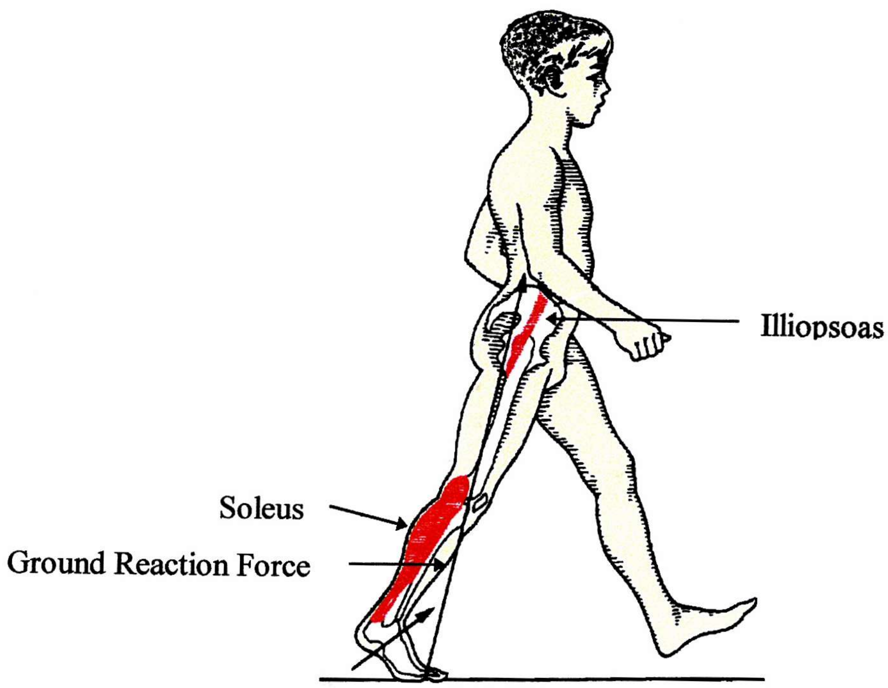


Figure 1.5 Terminal stance. (Adapted from Gage JR 1990).

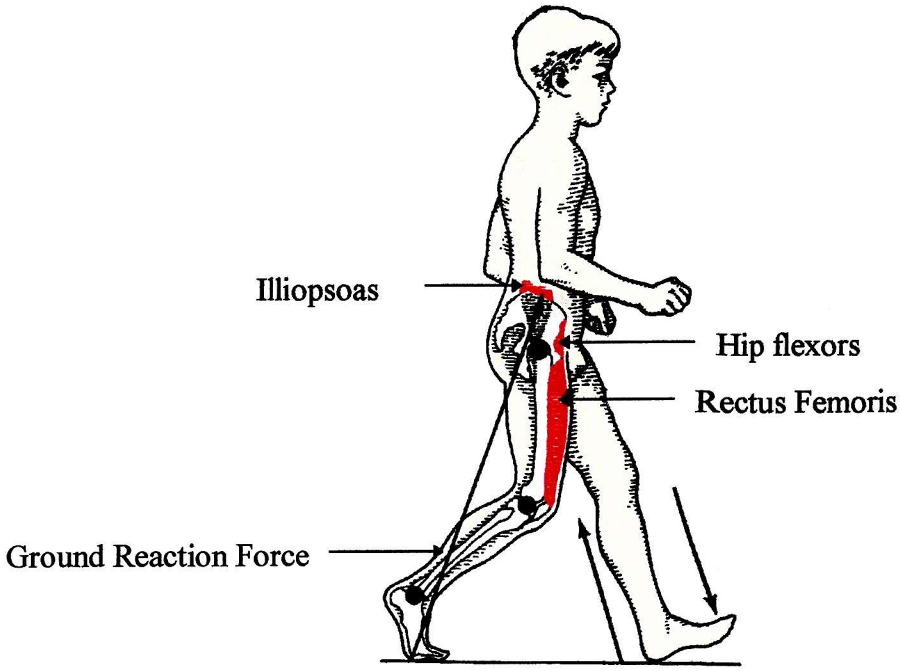


Figure 1.6 Preswing. (Adapted from Gage JR 1990).

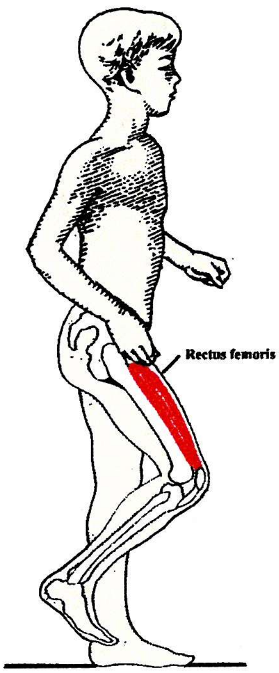


Figure 1.7 Initial swing. (Adapted from Gage JR 1990).

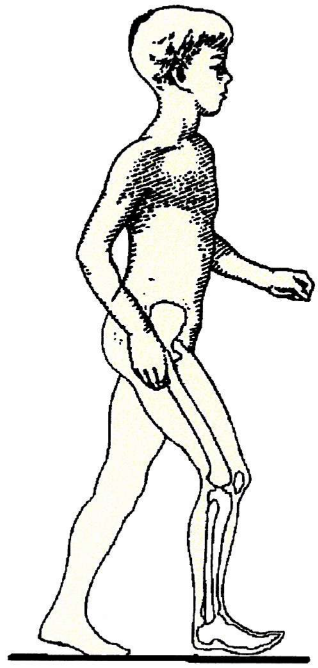


Figure 1.8 Mid-swing. (Adapted from Gage JR 1990).

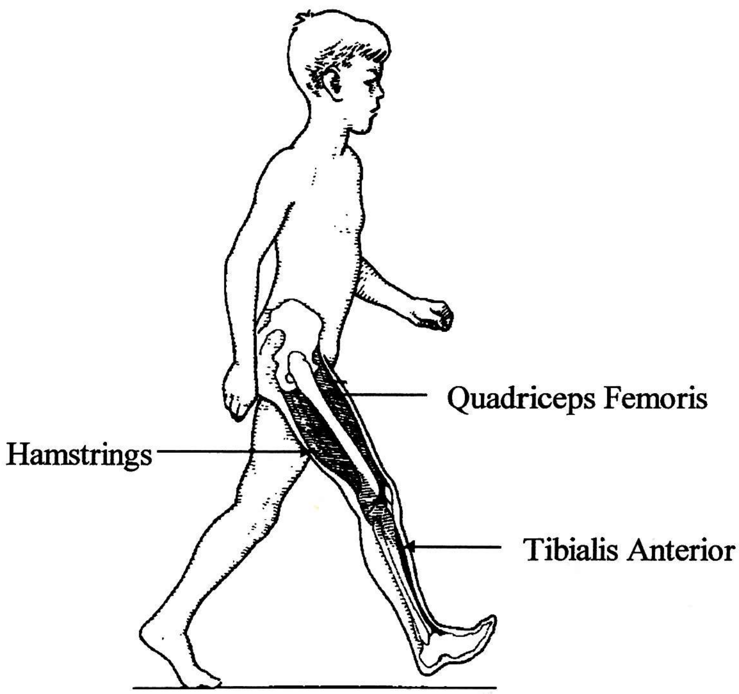


Figure 1.9 Terminal swing. (Adapted from Gage JR 1990).

1.4. Determinants of gait

Saunders et al (1953) described the major determinants of normal gait. The determinants of gait are: pelvic rotation, pelvic tilt, knee flexion in stance, knee and ankle mechanisms and lateral displacement of the pelvis. The centre of gravity of the body is displaced over a smooth pathway over the weight-bearing limb twice during the complete gait cycle. Using a hypothetical model of ''compass gait'', with the legs articulated only at the hips and fixed in extension, the pathway of the centre of gravity would be located at the points of intersecting arcs as shown in Figure 1.10.

Determinant one: Pelvic rotation

The pelvis in normal walking rotates alternately in a clockwise and anti-clockwise direction on the swing side. This prevents an excessive drop in the body's centre of gravity during periods of double limb support. The magnitude of pelvic rotation is about 4 degrees in either direction, a total of about 8 degrees. Pelvic rotation provides more flattening of the arc of the path of the centre of gravity which leads to reduction of the energy cost of locomotion (Figure 1.11).

Determinant two: Pelvic tilt

In normal walking the pelvis drops downwards relative to the horizontal plane on the side of the swinging leg. The amount of this downward drop is about 5 degrees during mid-stance and prevents an excessive rise by flattening the arc of the path in the body's centre of gravity (Figure 1.11).

Determinant three: Knee flexion in stance

The knee is fully extended during heel strike. It then flexes to about 15 degrees in the mid-stance phase. This reduces the impact of body weight and inertia and acts as a shock absorber. Knee flexion appears to be the most important determinant in terms of energy expenditure. Without knee flexion in the stance phase, the centre of gravity would rise more during mid-stance and the total vertical amplitude of centre of gravity movement would be greater (Figure 1.11).

Determinants four and five: Foot and knee mechanisms

The vertical displacement of the centre of body mass is smoothed out by the coordinated movements of the knee and ankle joints (Figure 1.12).

Determinant six: Lateral displacement of the pelvis

The pelvis shifts from side-to-side in the horizontal plane, resulting in minimizing the lateral displacement of the centre of gravity over the weight bearing leg. This occurs twice in each gait cycle.

It has been claimed that these six factors are important in controlling the excursion of the centre of gravity, and in energy conservation.

Abnormalities of two or more of these determinants, as a result of injury or disease, lead to an increase in the level of energy expenditure and make effective compensation difficult *(Saunders et al 1953)*.

Gage (1991) stated that there are five prerequisites for normal gait which are frequently lost in pathological gait and lead to decreased efficiency of gait and increased energy expenditure:

(1) Stability in stance.

(2) Adequate foot clearance in swing.

(3) Pre-position of the foot in terminal swing.

(4) An adequate step length.

(5) Energy conservation.

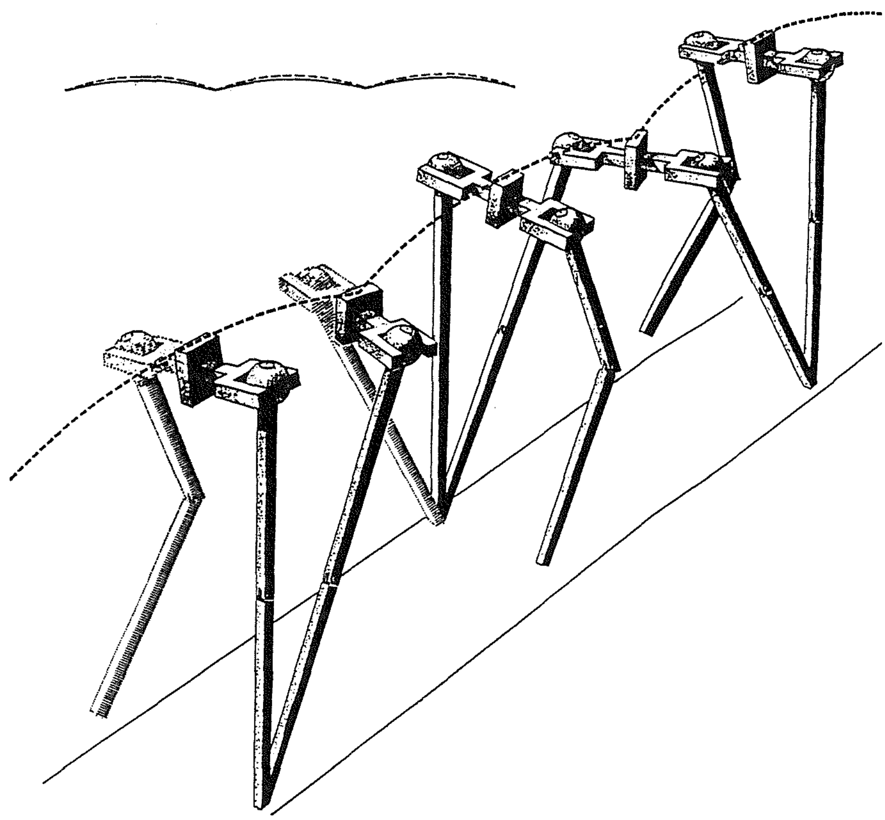


Figure 1-10 Hypothetical compass gait. (Copied with permission from Saunders et al 1953).

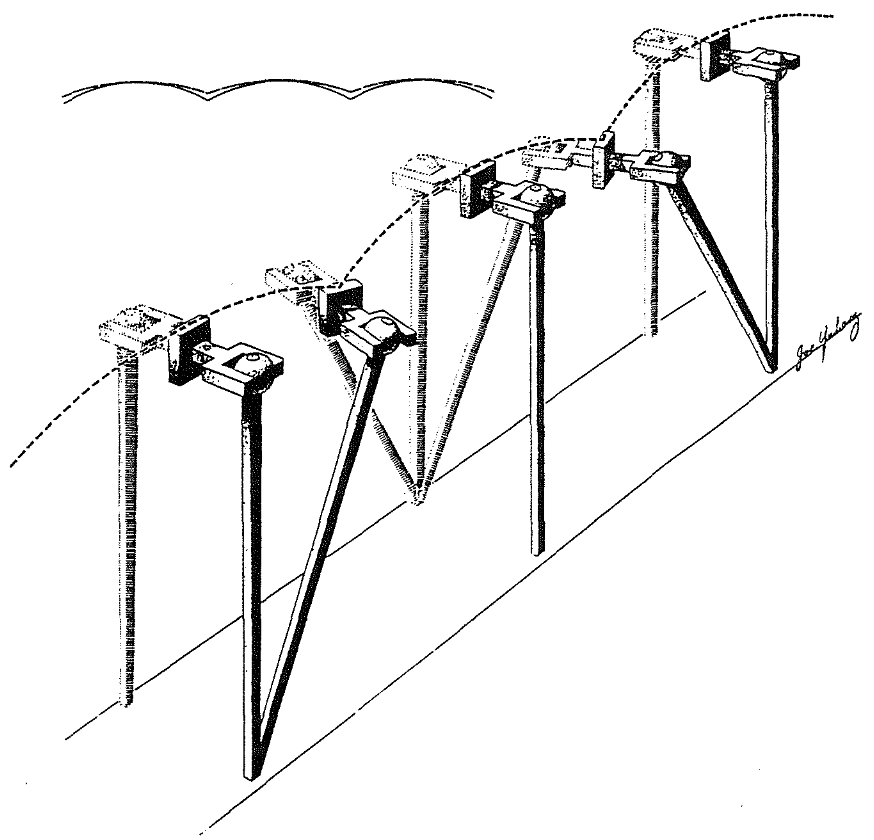


Figure 1-11 Knee flexion coupled with pelvic rotation and pelvic tilt in order to achieve the least vertical displacement of the centre of gravity. (Copied with permission from Saunders et al 1953).

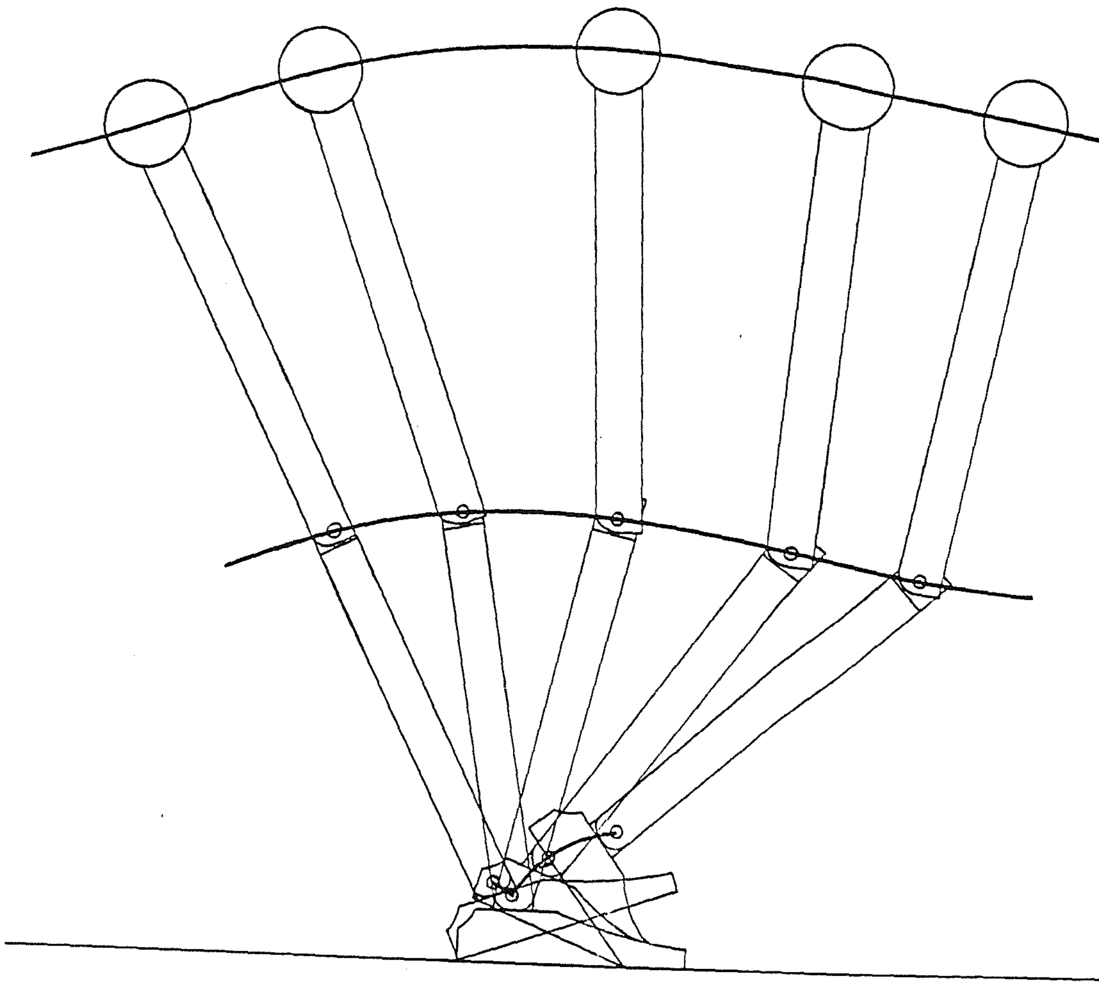


Figure 1-12 Interaction of ankle and knee rotation serving to smooth out the pathway at the intersections of the arcs of the centre of gravity. (Copied with permission from Saunders et al 1953).

1.5. Characteristics of gait in elderly subjects

In healthy elderly subjects gait may be viewed as 'a slowed-down version of the gait of younger adults' *(Whittle 1996a)*. Changes associated with age may be said to take place from the age of 60 onwards. They affect stride length, step length and walking speed *(Jones and Barker 1996, Ferrandez, et al 1990)*. Decreased range of motion in the lower limbs also occurs in elderly subjects. The changes help the subjects to maintain balance and stability *(Whittle 1996a)*. Several studies have attempted to describe the normal gait pattern of adults in an age range of 20 to 87 years, and the effects of old age on gait.

Studies of the kinematic gait patterns of normal men in different age groups (64 healthy subjects aged from 20 to 87) confirmed a reduction in cadence and stride length, and an increase in double support time in subjects over 65 years of age. They also showed a decreased range of hip flexion and extension, decreased flexion and extension of the knee, decreased plantar flexion and dorsiflexion of the ankle *(Murray et al 1964, Murray et al 1969)*.

Similarly, a further study by *Finley et al (1969)*, compared the kinematic gait pattern of 12 young and 23 elderly women in the age group; 19-38 years of age and 64-68 years of age, respectively. They found that the older women walked with a shorter step length, slower velocity, and smaller swing to stance ratio.

Murray et al (1970) reported the walking pattern of healthy women of different age groups. They selected 30 healthy women aged between 20 and 70 years and classified them into 5 groups. They found that elderly women walked with shorter step and stride lengths, and a smaller swing to stance ratio. Their walking velocity was reduced. *Gabell and Nayak (1984)* studied the variability in gait in 32 young and 32 elderly men aged 21 to 47 years, and 66 to 84 years respectively. They did not observe a significant variability in gait as measured by coefficient of variation of step length, stride time, stride width and double-support time.

Hageman and Blanke (1986) compared the gait pattern of young and elderly women in the 20-60 age group. They found that elderly women have a wider base of support than younger women. The step and stride length values of elderly women were reduced. These findings are similar to those of *Murray et al (1970) and Chao et al (1983).*

Blanke and Hageman (1989), in a later report, also studied the gait pattern in 12 young and 12 elderly men aged 20 to 32 years, and 60 to 74 years respectively. They did not observe significant differences between young and elderly men in step and stride length, and in gait velocity, but there were significant differences between the groups in stride length. The lack of statistically significant difference in that study may have been associated with the small sample size in both test groups.

1.6. Methods of gait analysis

Gait analysis was defined by *Davis (1988a)* as 'systematic measurement, description and assessment of those quantities thought to characterise human locomotion'. Gait studies have been performed in subjects with normal and abnormal gait for many years using various measurement techniques. The most commonly used methods of gait analysis are the following:

1.6.1. Observational analysis of gait

The conventional method for identifying or describing a subject's gait is by visual observation. Observational gait analysis (OGA) depends on the observer's judgment without the aid of any electronic devices *(Malouin 1995).* Simple aids can be used, for example, a stop-watch or a measured walkway. OGA assesses cadence, step and stride length, in addition to the range of motion at the pelvis, hip, knee and ankle during the gait cycle *(Saunders et al 1953).*

Saleh and Murdoch (1985) found visual observational gait analysis to be an unreliable technique. They claimed that detection of gait abnormalities was low at high gait velocity. Gait parameters (e.g. step length and step time), which are considered to be of particular value in the assessment of gait, are difficult to assess visually. It has also been noted that the accuracy of observational gait analysis differs from one observer to another and that intra-observer reliability is better than inter-observer reliability *(Kerbs et al 1985).*

Gage and Ounpuu (1989) reported several limitations which could arise when evaluating gait deviations with the naked eye. They state that the human eye cannot recognise certain movements occurring at higher frequencies, and it cannot simultaneously observe details of more than one position. If high speed events cannot be observed, some abnormalities may not be detected. Furthermore, many patients partially overcome gait problems by compensatory mechanisms. This may make identification of the primary defect difficult.

1.6.2. Kinematics

Kinematics is the study of body motion but without consideration of the forces that cause the motion. Kinematic data have been used widely in the analysis of gait pathology and can be presented on three principle planes; sagittal, frontal and transverse at the pelvis, hip, knee and ankle joints *(Harris et al 1994).* Kinematic studies of gait are often combined with measurements of stride length, step length, velocity, and cadence *(Gill et al 1997).*

1.6.3. Kinetics

The study of the relationship between body motion and the forces which act on the body is called kinetic *(Winter 1990)*. Kinetic data can measure hip, knee, and ankle joint moments and powers *(Harris et al 1994, Nordin and Frankel 1989)*.

A moment is defined as a force which acts at a distance to produce angular displacement about the (joint) centre of rotation. Power is the product of the moment generated by muscles and ligaments crossing the joint and the angular velocity of the joint *(Chao and Cahalan 1990)*.

In normal walking there are two forces (internal and external) acting on the joints or body segments. The internal forces are mainly generated by muscles and ligaments, whereas the external forces are produced by gravity and ground reaction forces (GRF) in response to the body weight transmitted through the feet *(Norkin and Levangie 1992)*.

The ground reaction force is the most important force acting during the stance phase of gait, acting between the foot and ground and consisting of three forces (vertical, anteriorposterior and mediolateral). These forces are measured with force plates embedded into a walkway *(Cook et al 1997)*.

1.6.4. Electromyography (EMG)

Dynamic EMG is the process of recording the myoelectrical activity of skeletal muscles during movement. The application of EMG in the assessment of lower extremity muscle function has been extensively studied by several researchers for many years. It has been applied to normal and pathological gait in clinical practice, to determine phasic patterns for individual muscles or muscle groups *(Drillis 1958, Sutherland and Hagy 1972, Murray et al 1984)*.

Surface and intramuscular electrodes have been used for EMG studies of gait. The selection of electrode type depends on the muscle under study. Large superficial muscles, for example the knee extensors and flexors, require surface electrodes, whereas deep muscles require intramuscular electrodes. Surface electrodes are useful for the study of large muscles or groups of muscles *(Robinson and Kellogg 1995)*.

Intramuscular electrodes, for example needle and fine-wire-based electrodes, are used to identify the activity of small or deep muscles. Needle electrodes are invasive and are generally used in physiological research work rather than gait analysis, while fine-wire electrodes are more frequently used in clinical practice. *Kadaba et al (1985)* reported that data recorded from surface electrodes represent a more consistent measure of muscle activity than intramuscular electrodes and are more reliable in comparing data from different test days.

The following advantages of the surface EMG electrodes *(Basmajjian 1967, Aminoff 1992)*:

(1) They are non-invasive

(2) They are easy to apply to the skin and can be re-applied frequently without discomfort to the patient.

(3) There is no risk of infection.

(4) They are more acceptable on ethical grounds.

1.6.5. Energy expenditure

Measurements of the energy cost of walking are an important part of gait analysis. Any deviation from the normal walking pattern, for example caused by pain, muscle weakness, or any flexion, valgus or varus deformity, will lead to an increase in the energy spent by the body *(Doderlein 1997)*. Measurement of energy expenditure therefore provides the clinician with valuable information about a person's functional capability and gait performance, and may provide a guideline for further treatment *(Waters et al 1987, Waters and Mulroy 1999, Doderlein 1997)*. The common methods used to measure the energy cost of walking, and the advantages and disadvantages of each method are described below.

Oxygen consumption measurement

Oxygen consumption measures the amount of oxygen consumed per kilogram body weight per unit of time (ml/kg). The ratio of carbon dioxide produced to oxygen consumed provides information on the type of metabolism taking place. The classic method that is used to measure oxygen consumption is called the Douglas bag system (in which expired air is collected for later analysis). Accurate results can be obtained from the oxygen consumption measurement *(Perry 1992)*. The disadvantages of oxygen consumption measurement are the following:

- They are very expensive.
- The patient finds it uncomfortable.
- The measuring equipment is heavy (800-2500 g).
- The equipment requires regular maintenance.

Heart rate measurements and 'physiogical cost'

Heart rate measurements have demonstrated that there is a linear relationship between heart rate and oxygen uptake. Energy expenditure is considered to be proportional to the heart rate and changes in heart rate can be used to calculate the physiological Cost Index (PCI). This method of estimating energy expenditure was chosen for this study because of its simplicity.

In recent years PCI has been used in research and as a clinical tool to study pathological gait in children and adults *(MacGregor 1979, Butler et al 1984, Rose et al 1990)*. It was found to be reliable in test-retest experiments on healthy subjects walking at self-selected speed on a treadmill *(Bailey and Ratcliffe 1995)*. The advantages of heart rate measurement are:

- It is simple and easy to measure and does not require special instruments.
- It is inexpensive
- It does not inhibit normal walking patterns.

The disadvantages of heart rate measurements and 'physiogical cost'are the following:

- The heart rate can be affected by changes in room temperature and by certain drugs.
- It does not measure energy expenditure directly, and it is not possible therfore to correct readings to take account of the presence of anxiety or the influence of drugs.

1.7. Historical review of methods of gait analysis

The purpose of this section is to review the history and advances in the technology of gait analysis especially during the past two decades. The study of human gait has grown in recent years, to become one of the most interesting and exciting branches of medicine. As can be seen from this review, several devices for the measurement of gait have been invented.

Medical science has long been concerned with the diagnosis of human gait, but has only recently extensively used gait analysis in clinical practice for the diagnosis and treatment of musculo-skeletal and neurological conditions.

Further advances were made with the use of computing technology, video cameras and improved electromyographic equipment. Instrumental gait analysis has contributed to significant advances in understanding the impact of orthopaedic and neurological conditions on human gait *(Masdue et al 1997, Polak 1998)*.

It is hoped that this review will give readers and researchers a general understanding of the concept of gait and the variety of measurement systems currently available for the analysis of human motion.

Historically, the analysis of human gait did not advance very far between classical times and the mid-nineteenth century. This was due to a lack of effective scientific methods of measurement *(Murray et al 1964)*. Consequently, the application of instrumental gait analysis in clinical practice was slow to develop.

In the seventeenth century, *Borelli (1608-1679)* was a pioneer in the science of locomotion and 'the first to apply Newtonian principles of mechanics to the movement of the human body' *(Cavanagh and Henley 1993)*. His work was published in the *De Motu Animalium (1682)* after his death and can be regarded as one of the most important advances in the development of the understanding of locomotion and muscle action, to which modern analysis is indebted. He wrote about biomechanics and gave detailed descriptions of walking events. Borelli was the first to measure the body's centre of gravity, and described how the balance of the body is maintained in walking through the 'forward displacement of the centre of gravity beyond the supporting area' *(Steindler 1953)*. During the first half of the nineteenth century most researchers were influenced by Borelli's ideas.

A new era of technology began in the study of the mechanisms of walking with the contribution of the Weber brothers, whose work was based on a sound concept of the mechanical principles of the human body. In *1836, Wilhelm and Eduard Weber* in Germany developed the first methods of quantitative measurement of locomotion using observations of the stance and swing phases of gait. They also postulated the pendulum theory of locomotion, which considered the swing phase of the gait as a purely passive movement. Although their research was conducted using primitive methods, it has been argued that it produced reasonably accurate results *(Paul 1998).* As a result of their measurements, they were able to draw up images of the body structure of a man in the process of walking.

Most of the credit for the early descriptions of the kinematics of gait using photographic methods belongs to *Marey* in Paris, and to *Muybridge* in California.

Marey (1885) was the first to use photographic methods to analyse patterns of body movements during gait. He introduced a system called chronophotography to analyse human gait. This yielded figures which are known today as stick-diagrams *(Kleissen et al 1998).*

Edward Muybridge (1902) was a famous photographer of the late 19th century and should probably be considered the father of human motion analysis. He used a number of stationary cameras in an innovative method to study the motion of a horse running at normal speed. Photography was an effective method of studying human gait for that time. Its main disadvantage was the length of time required to collect and analyse the data *(Banta 1999).*

During the 1890s in Germany, *Braune* and *Fischer* (an anatomist and mathematician respectively) introduced a method of three-dimensional analysis of human movement. Light-emitting markers were used in conjunction with trigonometric measurement to produce pictures with a frequency of 26 images per second. They were able to study the angular displacements of the lower limb joints using this technique. Their work, published *in Der Gang Des Menschen (1895),* was a major contribution to the understanding of human locomotion *(Steindler 1953).*

Motion analysis developed further in the following two decades after the advent of photography and the study of the mechanics of gait (i.e. kinetics) began. Kinetics is the study of the forces which act on the joints or body segments and the changes in motion which they produce. Kinetic analysis allows the measurement of hip, knee, and ankle joint moments and powers *(Nordin and Frankel 1989, Harris et al 1994).*

Elftman (1939 a&b) studied the energy and resultant forces acting at the hip, knee, and ankle joints provided by the muscles during walking, but he did not calculate the actual muscle and ligament loads. Elftman's work was limited due to the small sample of subjects that he studied, and also the fact that he used a system of two-dimensional analysis of human gait.

Research into human gait improved significantly during the 1940s and 1950s. Pioneering work by *Verne Inman* and his team at the College of Engineering and the Medical School of the *University of California* made a major contribution to human gait kinematics and mechanics *(Whittle 1996b). Inman* drew upon an extensive background in engineering, orthopaedics and anatomy to bring together a team specialising in gait analysis. This work covered the displacements and rotations of limbs in space, velocities and accelerations, external forces on the limbs, energy expenditure and dynamic electromyography (EMG - the process of recording the myoelectrical activity of muscles during movement). Inman's work was published by members of the University of California in *Human Walking (1981)* shortly after his death. This work greatly advanced our understanding of gait kinematics and kinetics, thereby contributing to improved design of artificial limbs.

Bresler and Frankel (1950) calculated the external resultant forces and moments acting at the hip, knee and ankle during level walking in four subjects. The techniques used in obtaining the data included cine-photographs, a force plate and skin markers to define joint centres.

The use of accelerometers to measure limb accelerations was introduced into biomechanics by several investigators, such as Gage *(1964),* Eberhart et al *(1954),* Liberson *(1965),* and Morris *(1973).*

Saunders et al (1953) described the excursions of the lower extremities with pelvic movement during gait. They analysed the main angular displacements of the lower limbs during normal human gait by using high-speed motion picture photography and also described the six fundamental features of the movement pattern which minimise excursion of the centre of gravity. They suggested that these are the most important features which determine whether a movement pattern is normal or pathological. These principles still guide researchers and clinicians studying human gait and are described in section (2.4).

Close (1959), one of Professor Inman's students, attempted to increase the clinical value of gait analysis by putting electromyography (single channel) on film. In this way, muscle activation could be studied in the context of observed movements. *(Sutherland 1997).*

In the subsequent phase of research, involving poliomyelitis patients, three-channel EMG was superimposed on film, in the same way, by Sutherland and his co-workers.

In the 1960s, *Murray (1964)* and her team in Milwaukee used interrupted-light photography to measure the angular displacement of hip, knee, ankle, and pelvic tilt. The tests were carried out using white stick diagrams, a mirror positioned overhead to measure locomotion in the sagittal plane, and photographs taken of the subjects walking. Her main interest was in studying the gait patterns of normal men and women, amputees, and Parkinson's disease patients.

Furnee (1967) in Delft, reported the development of the first automated kinematic measurements with video camera and dedicated computer interface with direct extraction of coordinates of imaged markers from video signal. He was developing a system able to quantify arm movement, because his field of work was in myoelectric prosthesis *(Kleissen, 1998 personal communication).*

In *1974, Perry* used a foot switch and a four-bar linkage goniometer. Perry's stride analyser was further developed by Bontrager, and this and her other achievements in the field, such as work on the specificity of surface and fine wire EMG recordings, are presented in her book *Gait Analysis: Normal and Pathological Functions (1992).*

Goniometry was introduced in clinical practice in the 1970s. For example, the sagittal, coronal and transverse rotations about the hip and knee joints were measured by *Johnston and Smidt (1969)* in thirty-three healthy men using a three plane, exoskeletal, electro-goniometric method. Subsequently, *Kettelcamp et al (1970)* used an electrogoniometer to measure the range of motion in normal and pathological knee joints. They recommended this method for routine clinical practice. However, the limitations of this method, which include errors due to the geometrical offset of the goniometer with respect to the joint centre, and due to the mobility of the soft tissues to which the goniometer is strapped, reduced its utility in everyday practice.

Lamoureux (1971) designed a large exoskeletal electro-goniometric device to measure the angular displacement occurring at the hip, knee and ankle joint. The apparatus was made of two light metal arms which were connected to the two segments of the limb to be measured.

Around the same time, *Sutherland and Hagy (1972)* described the use of a Vanguard Motion analyser to analyse lower limb motion. The cameras operated at 50 frames/sec. in order to freeze the motion of the subject for the analysis portion of the study. The advantages of this method were that no apparatus was attached to the subject, and multiple measurements could made in the same session. EMG may be superimposed on the motion picture film for simultaneous recording, and the recording of both legs can be made at the same time. The main disadvantage of this system was the time required to collect and analyse the data.

In *1972, Baumann* acquired a six-channel EMG with telemetry and a high speed cine camera that recorded at the same speed as the moving patient, providing more reliable measurements. He collected a vast amount of data on both cerebral palsy and normal children *(Sutherland 1997).*

About the same time, *Winter et al (1972)* at Waterloo University in Canada used a television camera 2-D system to track the movements of markers attached to the limb of a subject and a video tape recorder to record motion data. A computer was used to calculate the values of the joint angles to produce a simple description of the locomotion system during walking. The authors used this system to analyse the gait of 12 subjects at three walking speeds. The main advantage of this system is that the patient is not required either to wear power packs, or to be hard-wired, thus allowing the free movement of the subject. However, it is not without disadvantages. It presents only two-dimensional motion data, and in addition, the data collection requires a large computer memory. Winter published influential articles, and a book entitled *Biomechanics and Motor Control of Human Movement (Winter 1990).*

Simon's laboratory in Boston in the mid seventies used three dimensional techniques. Prior to the work of Simon, all gait analysis had been two-dimensional. He had spent a year with Sutherland and Perry before returning to the Boston Children's Hospital. His contribution was to add a sonic digitiser to three high speed movie cameras with a Vanguard Motion Analyser to make a three-dimensional coordinate system. This was an improvement on the previous two-dimensional system used by the Sutherland gait laboratory. However, the data still had to be digitised manually *(Gage 1998, personal communication).*

By *1970* more advanced systems of motion analysis had been developed using computers. These produced a computerized visual image of body movement which could be displayed graphically on the PC screen in less than a minute.

The three dimensional data proved to be useful in research work. This was facilitated by the introduction of a set of small, lightweight markers, either active small infra-red light emitting diodes (LEDs) or passive infra-red reflective spheres which were placed over selected body landmarks.

Jarret 1976 developed the first automatic motion measurement systems (**SELSPOT**) based on the use of a television video camera and computer. The system utilises from 1 to 16 cameras, a set of segment markers, LEDs placed over anatomical landmarks, and a data transfer system to a controlling computer.

The particular advantage of this system is that since the active markers are pulsed sequentially, the identity (anatomical location) of each marker is known by the controlling computer. Its main disadvantage is that the patient has either to wear power packs or be hard-wired to a computer in order to receive the necessary power for marker illumination. A second disadvantage is the considerable amount of time required to process the data. Somewhat later, the **VICON** and **CODA** systems were introduced.

VICON is an opto-electronic motion measurement system which uses five television cameras placed around a 10m walkway. The markers used are passive retro-reflective infra-red on each camera and each marker is initially identified by the operator. The system is capable of tracking up to 30 markers simultaneously. A major advantage of this system are that the patient does not need to wear a battery pack or be hardwired to a computer *(Kadaba et al 1987)* and the patient can be recorded on video. Despite these advantages, **VICON** has the following limitation:

(1) The system uses passive markers, and that affects the resolutions.

(2) It needs to be recalibrated for every test, because the camera sensors are sensitive to the movement of the camera, temperature and humidity, and so this is time consuming.

The first design studies for the **CODA** system were carried out at the Department of Ergonomics and Cybernetics at Loughborough University in 1970. The acronym '**CODA**' stands for Cartesian Optoelectronic Dynamic Anthropometer *(Mitchelson 1990)*. A primary device, completed in *1973* and developed by *Mitchelson (1974)* was commercially marketed as CODA 3. This instrument provided much better performance using 3 mirror scanners, rather than television, to capture the three-dimensional coordinates of limb position *(Mitchelson 1990)*.

The detection system consisted of compound cylindrical lenses in three electronic cameras. This was an advanced system, much better than others used at that time. The main disadvantage of the system was that it was limited to 8 markers. *Miller (1987)* described the CODA-3 as a 'pre-calibrated, optical measurement system at a pre-production stage. It bears much promise as a turnkey motion analysis system and we have found it very easy to use'.

Further development of the CODA system continued between *1985* and *1987* and an improved version of CODA-3, called CODA mpx30 was produced, meeting all of the initial design goals. The CODA mpx30 differs from the old CODA-3 in several important ways:

(1) It is all solid state with no moving parts, so it is more reliable.

(2) It is much smaller and lighter, and therefore portable.

(3) The markers are active LEDs, not passive corner cube prisms. This allows the use of many more markers with completely secure automatic identification.

(4) Markers can be placed as close together as desired (This is not possible with passive markers such as those used in video based systems).

(5) The system uses a scanner unit, not rotating mirrors.

(6) Sampling rates up to 800Hz are possible.

(7) The software is much more extensive and is Windows based.

The CODA mpx30 has the following limitation:

(1) Inability to record both limbs at the same time. To overcome this, one scanner is needed for each side.

The CODA mpx30 system is not a well known system due to the problems experienced in the early design of CODA-3 and this causes limitations as regards information exchange between researchers.

The CODA mpx30 differs from other motion analysis systems in two main ways:

(1) It has a resolution which is at least five times better than other systems which are based on video cameras such as VICON.

(2) It also has a greater dynamic range which means that fine details can be seen, even in large scale movements, with much greater resolution than with other motion analysis systems.

There are other several systems which are commercially available. These systems use either active infra-red LEDs or passive markers. Markers using active infra-red LEDs are used with the Selspot *(Selcon System, Ltd., Southfield, MI),* and Optotrack systems *(National Digital, Inc., Ontario, Canada).* Passive reflective markers are used with MacReflex *(Sweden),* Kinemetrix *(Medical Research Ltd., Wortley Moor Road, Leeds, United Kingdom)* Expert Vision *(Motion analysis corporation, Santa Rosa, CA),* Peak Performance *(Peak Performance Technologies Inc., Englewood, CO),* Elite *(Bioengineering Technology & systems, Milano, Italy)* and Ariel *(Life Systems Inc. La Jolla, CA).* Force plates, EMG or other analogue devices can be integrated by a synchronised analogue input accessory *(Davis 1998b, Davis and DeLuca 1996).*

Gage (1978) introduced gait analysis for routine clinical practice at Newington Children's Hospital in 1981. The Newington laboratory was unique in that it was the first to use automatic digitisation. The numbers of the markers did not need to be entered manually. Gage then moved to the Gillette Children's Hospital in Minnesota as a Medical Director where he set up gait analysis services along with his team, including Drs Koop, Novacheck, and Stout *(Gage 1998, Davis 1998a, personal communication).*

The most recent innovation in tracking method has been electromagnetic tracking using fixed sensors attached to the subject. A preliminary study determining the feasibility of electromagnetic tracking for collecting kinematic data at the ankle joint complex was carried out by Woodburn et al 1999.

There are different types of electromagnetic trackers on the market such as those developed by Polhemus and Ascension (Kirtley 2000, personal communication).

These are motion analysis systems based on magnetic field signals. They have the advantages of high level of accuracy (resolution) and the data is obtained in real-time. The analysis appears quick and easy to perform with no need for post-processing such as is often needed with video-based systems. They do however have some limitations. The sensors must be within range of the source and away from stray magnetic fields and metal objects. Most systems use cables which can inhibit normal walking patterns. The currently available magnetic field systems are not recommended for gait research, but they are useful for applications, such as spine motion analysis, posture work in which the subject does not move very much and virtual reality where the real time feature is useful (Kirtley 2000, personal communication).

In the last three years, Polhemus and Ascension have developed products together with Skill's 6D RESEARCH motion capture and kinematic analysis system software. They claim that this electromagnetic tracking system is a superior product and truly useful for gait analysis. Skill Technologies, Inc. has confirmed the points mentioned above regarding the limitations of this product and described their latest features and capabilities. They also recommend ways in which to overcome the stated problems, such as using non-metallic force plates and various methods for orgnising the cables. There is an eight sensor wire-less version which is half price of a video-based motion analysis system The currently available magnetic field systems are highly recommended for gait research (Cheetham 2000, personal communication).

CHAPTER TWO

LITERATURE REVIEW

2.1. Structure and function of the knee joint

The knee is a complex hinge joint situated in the middle of the lower limb. It is composed of three bones: the femur, tibia and patella. The joint has three articulating surfaces, two occurring between the condyles of the femur and tibia (tibiofemoral), and a third between the femur and the patella (patellofemoral) (Figure 2.1) *(Lockhart et al 1974, Moore and Agur 1996).* The ligaments provide stability to the knee joint during ambulatory activities (Figures 2.2-2.4). The main functions of the muscles are to provide control and coordinate movement required for ambulation. In addition, they support the knee joint during weight transfer and shock absorption (Figures 2.5-2.6) *(Lockwood 1998).*

The distal end of the femur is enlarged into two rounded condyles, forming a U shape, separated posteriorly by a deep intercondylar notch, but fusing anteriorly into a trochlear groove for articulation with the patella. The lateral femoral condyle is more prominent, and is slightly shorter than the medial. The most prominent lateral parts of the condyles are called the epicondyles.

The upper surface of the tibia has two rounded condyles. The expanded proximal end of the tibia, which widens transversely, acts as a bearing surface for body weight transmission through the femur. The tibial condyles articulate with the two femoral condyles and with the fibula at the tibiofibular joints. Irregularities of the upper surface between the two tibial condyles form anterior and posterior intercondylar areas. The intercondylar area, between the condylar articular surfaces, is separated by the intercondylar notch *(Warwick and Williams 1989).*

The patella is superficial and easily located on the anterior surface of the knee. The patella or kneecap is the largest sesamoid bone in the body. It is roughly triangular with rounded angles. The superior border is the base of the triangle, and the inferior border is the apex. The inner surface of the patella is covered with cartilage and glides on the cartilage of the femoral condyle notch.

The anterior surface of the patella is separated inferiorly from superficial tissues and skin by the subcutaneous infrapatellar bursa. The posterior surface is primarily articular and smooth *(Hungerford and Barry 1979, Warwick and Williams 1989, Lockhart et al 1974).*

The main biomechanical function of the patella is to transfer quadriceps muscle force into patellar ligament force and, thus, increase the power of extension and produce an extension moment at the knee joint *(Kauffer and Michigan 1971, Draganich et al 1987).* Also, the patella acts as a leverage system to facilitate the movement during walking.

Normal knee flexion is from 0 degrees to 135 degrees. From a position of nearly full extension at initial contact, the rolling movements of femoral condyles over the tibial plateau start with about 15 degrees of flexion at loading response and continue until 20 degrees flexion is reached in the mid-stance phase. After 20 degrees of flexion the ligaments become relaxed and permit both gliding and axial rotation. The patella glides over the end of the femoral trochlear groove as the knee bends. The knee joint starts flexing once again during the middle of the swing phase and reaches its maximum 50-75) degrees).

2.1.1. Ligamentous and muscular structures

Anterior structure

This includes the patellae, ligamentum patellae, the quadriceps muscle group and the expansions of the medial and lateral retinacula. The ligamentum patellae is a strong, flattened band about 8 cm in length extending from the inferior aspect and apex of the patella to the tibial tuberosity. The quadriceps femoris group of muscles are the main extensor acting on the knee. It consist of the rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius.

All fibers of the quadriceps muscle are inserted by a common tendon into the anterior tuberosity of the tibia *(Warwick and Williams 1989)*.

Posterior structure

This includes the oblique popliteal ligament, the gastrocnemius and the hamstring muscles. The hamstring muscles are the main flexors acting on the knee. These are the semitendinosus, semimembranosus and biceps femoris. They all originate from the ischial tuberosity. The semitendinosus and semimembranosus are inserted into the medial surface of the upper part of the tibia and the posterior part of the medial condyle of the tibia, respectively, whereas the biceps femoris is inserted into the lateral side of the head of the fibula, sending a slip to the lateral condyle of the tibia *(Warwick and Williams 1989)*.

Medial structure

This includes the tibial medial collateral ligament, pes anserinus complex and medial retinaculum. The tibiocollateral ligament is a strong flat band, 8 to 9 cm long, which extends from the medial epicondyle of the femur passing downwards and slightly forwards to attach the medial condyle of the tibia and the medial side of its shaft. The tibiocollateral ligament consists of both a superficial and a deep part. The superficial part is attached proximally into the medial femoral condyle and distally to the medial tibial condyle. The deep part of the medial ligament is divided into a meniscofemoral and meniscotibial portion. The medial collateral ligament is the primary stabiliser of the medial side of the knee against excessive valgus forces and external rotation. The medial pes anserinus complex is made up of the sartorius, gracilis and semitendinosus *(Kennedy and Fowler 1971).*

Lateral structure

This includes the fibular collateral ligaments, the arcuate ligament, the cruciate ligament, the lateral meniscus and tensor fasciae latae. The fibular lateral ligament is a rounded cord about 5 cm long and is attached from the lateral epicondyle of the femur to the lateral surface of the head of the fibula. It lies free from the capsule and lateral meniscus. The fibular collateral ligament helps prevent lateral displacement of the knee *(Warwick and Williams 1989)* (Figures 2.2-2.4).

2.1.2. Intra-articular structures of the knee

In addition to the ligaments that control and strengthen the joint capsule, there are two sets of intra-capsular structures which play an important part in the functioning of the knee from both anatomical and mechanical viewpoints; these are:

Cruciate ligaments

These strong intracapsular bands of fibrous tissue stretch upward between the tibia and the femur, crossing each other on the way. The PCL is thicker than the ACL and is considered to be one of the strongest ligaments in the knee joint *(Kennedy and Graiger 1967, Bayley et al 1988).*

The anterior and posterior cruciate ligaments (ACL, PCL) are named according to their tibial origins, and pass upwards to attach to the intercondylar notch of the femur *(Moore 1980, Ellis 1983).* In normal knee function, the ACL and PCL in the stance are instrumental in controlling the backward and forward displacement of the tibia on the femur, and in the swing phase to prevent excessive external and internal rotation of the tibia under the femoral condyle *(Kapanji 1970, Vandomelen and Fowler 1989, Laskin 1995).*

Menisci

The menisci are two C - shaped wedges of fibrocartilaginous intra-articular structures. The medial and lateral menisci rest between the condyles of the femur and tibia, and are connected to each other by the anterior transverse ligament. The medial meniscus is semicircular and about 3.5 cm long; it has a wider curve than the lateral mensicus. The lateral mensicus is more circular and covers a larger portion of its tibial surfaces than the medial. Both menisci glide posteriorly in knee flexion and anteriorly in extension *(Warwick and Williams 1989).* The menisci facilitate rotary movements, act as shock absorbers and provide some degree of stability to the knee joint *(Grana 1993).*

2.1.3. Blood vessels

The blood supply to the knee is extensive and is derived from both the femoral and popliteal arteries. The superior genicular artery rises from the femoral artery and supplies the vastus medialis and various portions of the knee joint and surrounding musculature. The medial and lateral superior genicular arteries are branches of the popliteal arteries.

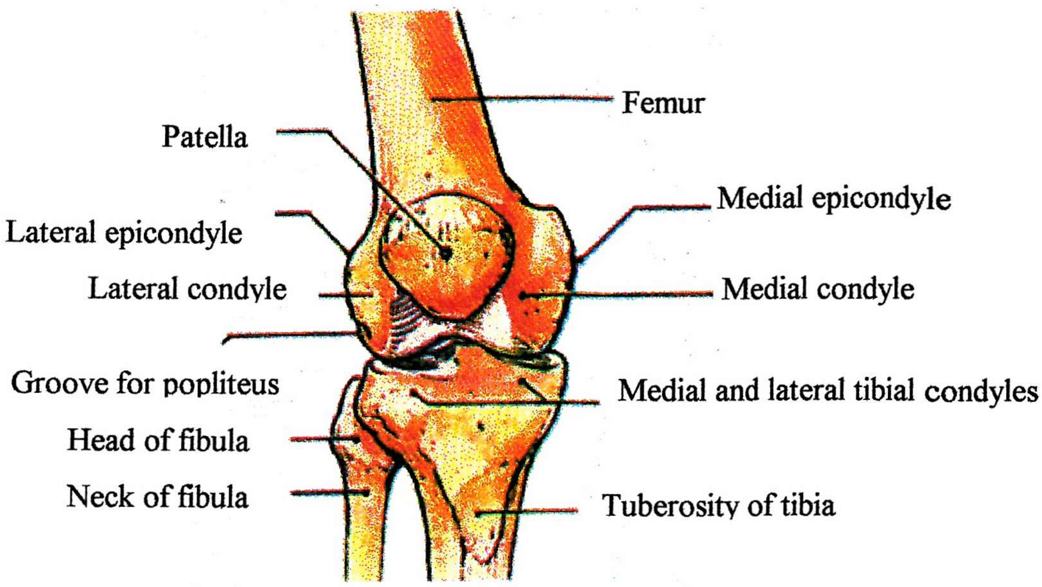
The middle and lateral inferior genicular arteries arise from the popliteal vessel and run towards the front of the knee, deep to the cruciate and collateral ligaments. The anterior and posterior tibial recurrent arteries supply the anterior aspect of the knee, the superior tibiofibular joint, and the lateral condyle of the tibia. The middle genicular artery arises from the popliteal artery and penetrates the posterior capsule in the intercondylar notch *(Warwick and Williams 1989).*

2.1.4. Innervation of the knee joint

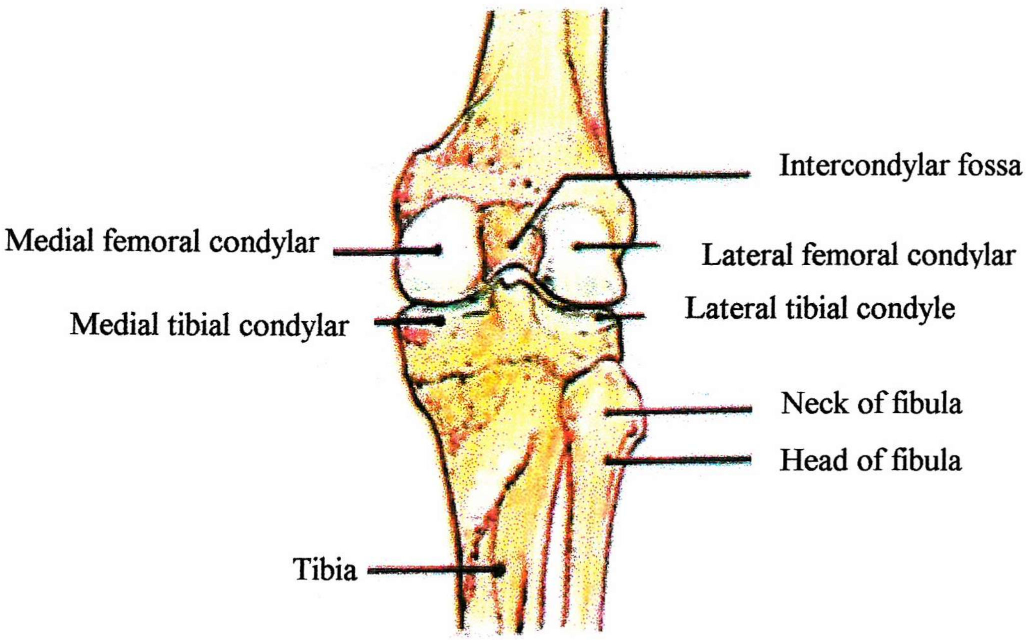
The knee joint is innervated by several nerves that branch out from the obturator, the femoral, the tibial and the common peroneal nerves. The most consistent and largest nerve supplying the knee is the posterior articular nerve. The posterior group is composed of the prominent posterior articular tibial nerve and a terminal branch of the obturator nerve. This prominent branch of the posterior tibial nerve rises from the tibial nerve at a variable level above the knee or within the popliteal fossa. The terminal branch of the obturator nerve reaches the knee joint via the popliteal fossa.

The anterior group consists of branches of the femoral, common peroneal, and saphenous nerves. The femoral nerve branches to the vastus lateralis, vastus medialis, and the vastus intermedius, contributing articular branches to the lateral, central and medial portion of the capsule, respectively. The common peroneal nerve innervates the posterolateral corner of the knee, whereas the recurrent peroneal branches off to the anterolateral capsule.

The final anterior articular afferent nerve is the infrapatellar branch of the saphenous nerve. This branch has particular clinical significance because it also provides sensory branches to the patellar tendon *(Kennedy et al 1981).*



A- Anterior view



B - Posterior view

Figure 2-1 Bones of knee joint. Anterior view (A) and posterior view (B). (Copied with permission from Moore and Agure, Williams and Wilkins 1996)

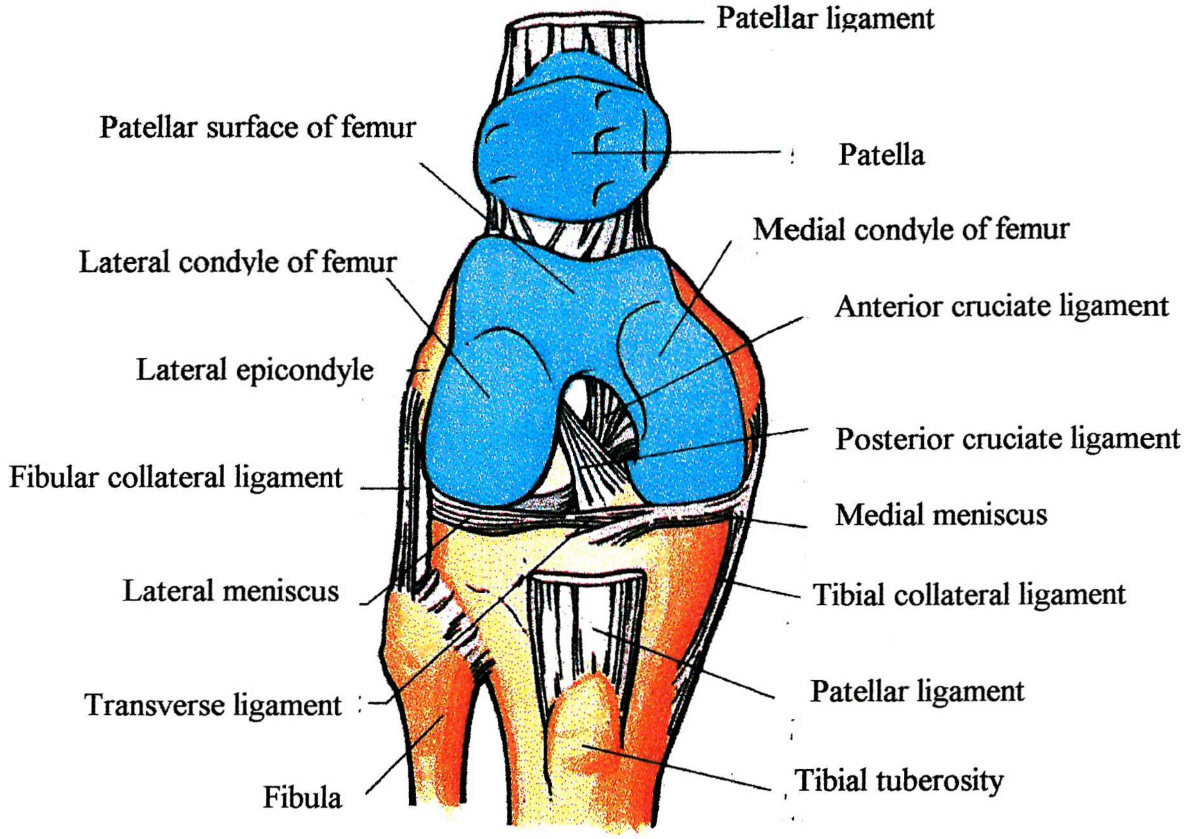


Figure 2-2 Anterior view of knee joint. (Copied with permission from Moore and Agure, Williams and Wilkins 1996)

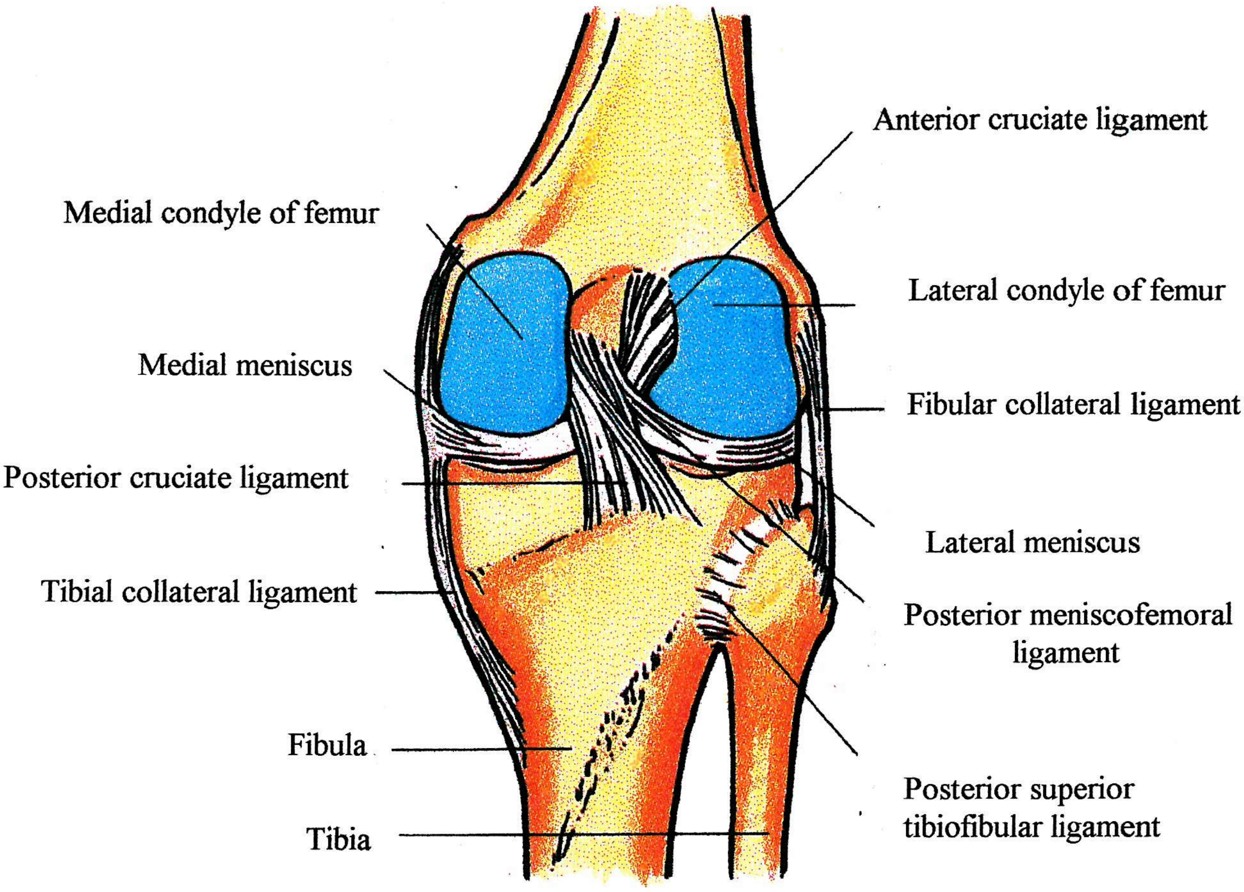


Figure 2.3 Posterior view of knee joint. (Copied with permission from Moore and Agure, Williams and Wilkins 1996)

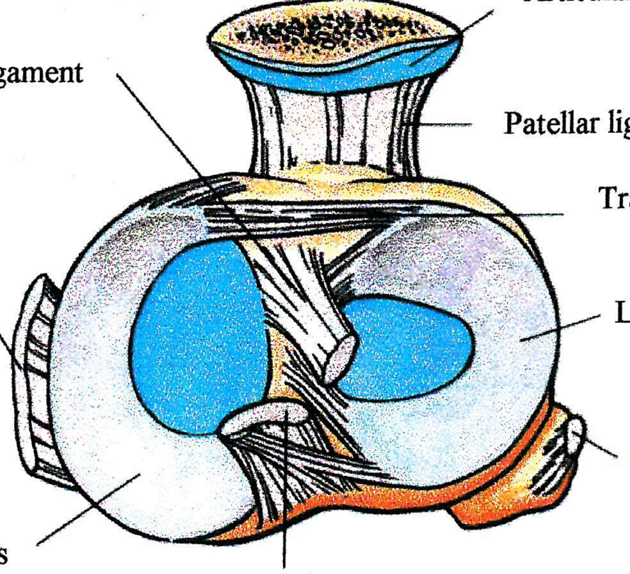


Figure 2.4 Superior view of knee joint. (Copied with permission from Moore and Agure, Williams and Wilkins 1996)

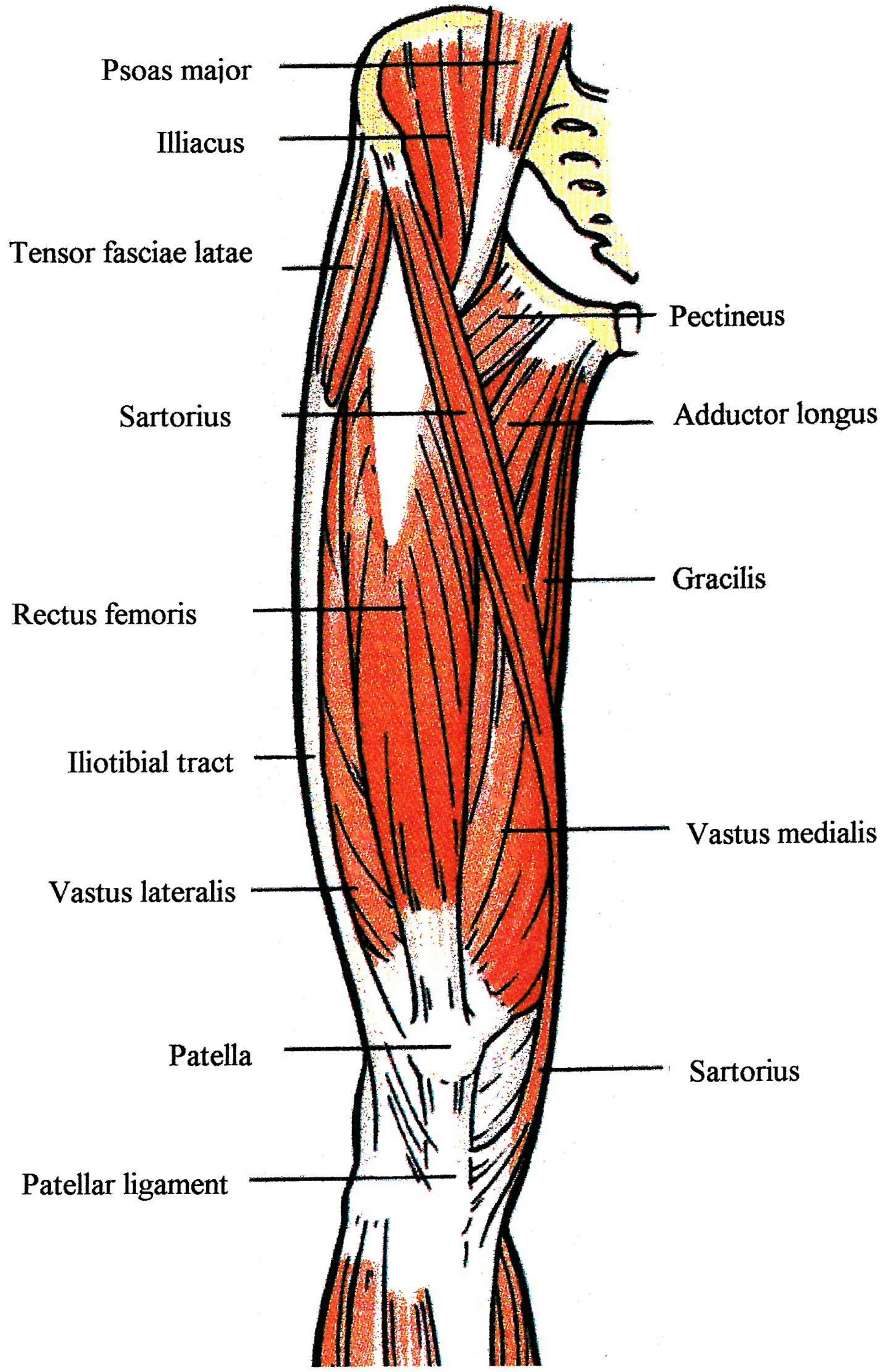


Figure 2.5 Front of the thigh - Superficial dissection. (Copied with permission from Moore and Agure, Williams and Wilkins 1996)

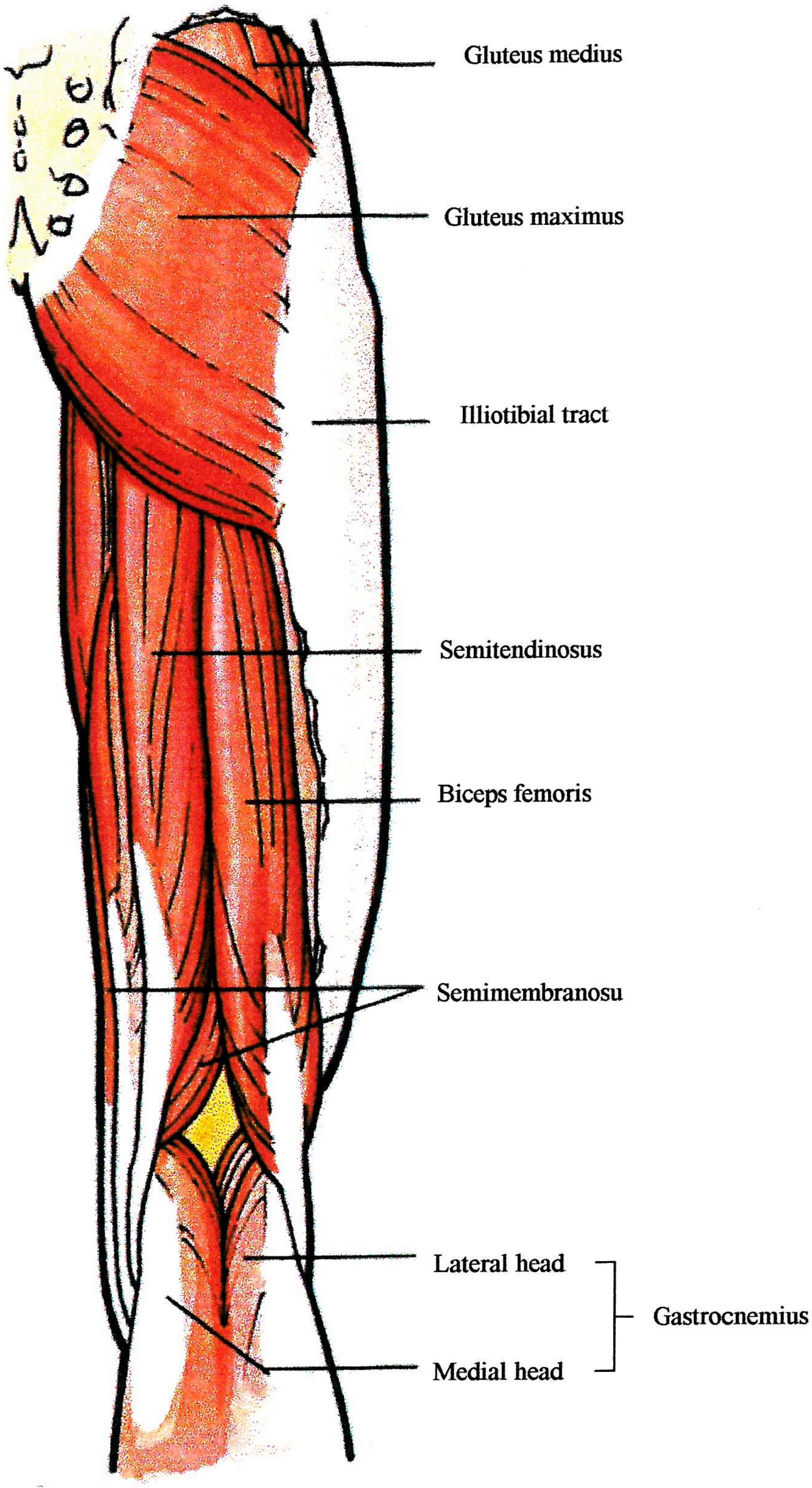


Figure 2.6 Back of the thigh - Superficial dissection. (Copied with permission from Moore and Agure. Williams and Wilkins 1996)

2.2. The knee joint in osteoarthritis

Osteoarthitis is a non inflammatory disorder of joints characterized by defects in articular cartilage and related changes in subchondral bone, joint margins, synovium, and paraarticular structures *(Altman 1986)*. It is the most frequent joint disorder in the World today and the knee is the most commonly involved joint *(Spector et al 1992)*. The clinical changes are pain, swelling, stiffness, deformity, and muscle weakness *(Altman 1990)*.

The primary pathological change in OA is cartilage degeneration. Osteophyte formation and inflammation are secondary changes as a result of damage to the cartilage and joint incongruity.

The main clinical changes are pain after prolonged rest, stiffness on initiation of gait in the morning or after sitting which improves when the knee warms up, and giving way due mainly to quadriceps wasting or to laxity in the ligament *(Jackson and Waugh 1984)*.

Gait pattern is affected with regard to reduction in walking speed, stride length and ROM of the knee in the stance and swing phases of the gait cycle.

The clinical methods of assessment of the OA knee are by assessing pain, function, ROM, quadriceps thickness and X-rays. The radiological findings of osteoarthitis remain the best clinical examination for diagnosis of OA. Indicators include joint space narrowing, osteophyte cortical sclerosis and subchondral sclerosis.

2.3. Historical review of total knee replacements

The surgical procedure of total knee replacement was introduced in the 1950s. A large variety of prostheses have been developed and modified over the past thirty years. TKR is widely used for pain relief, restoration of function, and improved quality of life *(Martin et al 1998).*

In *1860 Verneuil* was the first to interpose soft tissues between bone ends to reconstruct the articular surface of a joint. *(Murray 1991).* Later on *Campbell* in 1940 reported two patients in which he interposed Vitalium plates in the entire contact surface of the knee joint. The concept of knee replacement has remained since then, but the problem was to find a suitable material to interpose between the two resected joint surfaces *(Riley et al 1984, Goldberg et al 1992).*

Over the years a variety of materials have been implanted between joint surfaces in the knee, such as fat, prepatellar bursa tissue, fascia lata, skin, cellophane and nylon *(Riley 1976).* The history of the replacement of the knee joint by modifying the articular surfaces continued up to the latter part of the 20*th* century. A large number of investigations have been published describing the variety of prostheses that have been designed. However, no ideal type of prosthesis for all cases has been developed *(Scott et al 1979, Riley and Healy 1984).*

Walldius in 1950 designed his hinged linked implant, which is considered the earliest successful tibiofemoral replacement. The prosthesis consisted of long-stemmed devices inserted into the medullary canal of the femur and tibia and articulated together through a hinge mechanism. It constrained the joint to allow only one simple axis of movement without rotation. In addition, it did not take into consideration the importance of the ligamentous structure in supporting the prosthesis *(Crenshaw 1987, Lotke 1981, Laskin 1984).*

Within a few years, several researchers tried different modifications and developments of the hinged knee prosthesis *(MacIntosh 1966, Shiers 1974, McKee 1974, Wilson et al 1974).*

McKeever (1960) introduced a new version of the MacIntosh prosthesis. *MacIntosh* in 1966 reported a review of 58 rheumatoid knees with metal tibial plateau.

The modern era of total knee replacement started in the early 1970s with the work of *Gunston (1971, 1973)*. He introduced the first non-hinged prosthesis with cemented prosthetic components. He used a prosthesis with a metallic femoral part that articulated against polyethylene troughs fitted to the tibial side. The support between femoral and tibial parts derived from the surrounding ligaments and soft tissues.

The next group of prostheses was called nonconstrained total knee prostheses (total condylar). They consist of a cylindrical component positioned between the femur and tibia, allowing more contact surface between these bones. The function of this prosthesis depends on the integrity of the surrounding ligaments and soft tissues. This type was developed to avoid the early complications that were experienced with a hinged prosthesis *(Ranawat and Sculcue 1985, Walker 1985)*. The Freeman Swanson knee *(Freeman et al 1973)* and the Kinemax knee are examples from this group of prostheses.

The main indications for total knee replacement in patients with OA are intolerable pain, mechanical deformity, stiffness, and joint instability. Contraindications to total knee replacement include gross quadriceps weakness, genu recurvatum associated with muscular weakness, paralysis, poor general health, and severe osteoporosis *(Crenshow 1987, Windsor and Insall 1991)*. The major causes of TKR failure are early and late infection, malalignment, vascular complications and wound necrosis *(Insall 1993)*

The optimal prosthetic knee should be resistant to wear and tear, resemble the normal knee joint, have the ability to maintain the normal anatomical and biomechanical interaction between the shapes of the replaced articular surfaces and the soft tissues (ligaments and muscle) and be free from loosening *(Andriacchi 1993).*

There are two types of TKR design in current use. These are as follows.

(1) Posterior Cruciate-retaining

This type of prosthesis allows preservation of the PCL (examples: Leeds, Polycentric, Geomedic and Duo-condylar) *(Longton and Seedhom 1981).* It has been shown that the PCL assists in femoral roll-back and gives more flexion and provides greater posterior stability to the knee *(Insall 1984, Andriacchi et al 1982, Dorr et al 1985, Andriacchi et al 1986, Andriacchi and Galante 1988),* and maintains a natural flexibility of the knee joint *(Insall et al 1977, Goodfellow and O'Connor 1978, Sledge and Walker 1984, Scott and Volatile 1986).* Other authors have shown that retaining the PCL alows it to transmit and absorb horizontal forces arising during movement. These forces are not then transmitted to the prosthesis and cement bone interface, so the force on the patellofemoral joint is less *(Soudry et al 1984, Bayley et al 1988).* Preservation of the PCL also improves proprioception of movements of the knee joint *(Barrett et al 1991). Cash et al 1996* reported no difference in the threshold to perception of motion between those who received PCL-retaining and those with PCL-substitution.

(2) Posterior cruciate-substituting

This type of prosthesis sacrifices the PCL, anterioposterior stability being provided by the presence of a solid cam device to allow femoral rollback (examples: Insall-Berstein, Posterior Stabilized, Kinematic stabiliser and Press-Fit Condylar stabilized). This type of prosthesis replaces the function of the PCL with a posterior stabilizing articular surface.

It has been shown that PCL sacrificing prostheses provide greater stability than PCL-retaining prostheses. This additional stability is accomplished by increasing conformity between the femoral and tibial surfaces *(Insall 1984)*. It has been argued that retention of the PCL does not increase knee flexion due to femoral rollback. because the contact point, instead of going backwards, actually slides forwards.

Table 2.1 summarises the advantages and disadvantages of the PCL-retaining and PCL-sacrificing prostheses.

Table 2.1 Summary of the advantages and disadvantages and disadvantages of the PCL-retaining and PCL-sacrificing prostheses.

	PCL-retaining	PCL-sacrificing
Example	Miller-Galante 2,Genesis, and Anatomic Modular Knee	Insall-Berstein, Press-Fit Condylar stabilized
Advantages	1. Preserves PCL 2. Absorption of the horizontal forces by the ligaments 3. Preserve knee proprioception. 4. Provides greater posterior stability if PCL functioning normally	1. Gives better result in correction of deformity 2. Optimal fixation 3. Gives a bigger range of motion than the PCL-retaining prosthesis 4. Technically an easier operation.
Disadvantages	1. Provides less posterior stability if PCL not functioning normally 2. It is a technically difficult operation	1. Reduces knee proprioception

2.4. Review of gait analysis in patients with knee osteoarthritis

Osteoarthritis of the knee joint is a common cause of abnormal gait in elderly subjects. Understanding of the gait abnormalities in patients with OA of the knee is essential for the optimal management of these patients and for the evaluation of treatment outcomes. However, to date, no comprehensive studies of the gait characteristics of these patients have been carried out. The above studies examined the temporospatial parameters (walking velocity, cadence and stride length) and generally demonstrated that patients with arthritis of the knee walked more slowly, had a reduced stride length and a longer stance phase of the gait cycle than healthy control subjects. In some cases the study of the temporospatial parameters was combined with measurements of the range of motion at the lower limb joints during walking (kinematics). However, the study of the moments and powers generated during ambulation (kinetics) and the pattern of activation of lower limb muscles received little attention.

Györy and co-workers (1976) reported the results of gait analysis in 65 patients with degenerative arthritis and 30 patients with rheumatiod arthritis (RA) and compared them with 29 healthy subjects. The patients in this study exhibited marked reductions in temporal and distance parameters (velocity, cadence and stride length). In addition, stance phase flexion and extension, and peak values of vertical ground reaction forces, were reduced in subjects with degenerative joint disease.

In a similar study *Stauffer et al (1975)* described the changes found in normal and diseased knees in patients with rheumatoid and degenerative arthritis during gait analysis. A knee electrogoniometer with three orthogonally oriented potentiometers was used for measuring motion in three planes during the gait cycle. The study demonstrated differences in most functional parameters compared to age-matched controls. These included a reduced velocity of gait, a shorter single-limb stance phase, increase in medial-lateral shear forces and prolonged double-limb support. Sagittal plane knee motion was significantly reduced in patients with diseased knees compared with the range of motion in the knees of healthy subjects. Patients with OA achieved 82% of the range of motion of the control group. Patients exhibited limited internal and external rotation at the knee, which could have been due to reduction in knee flexion during gait. The authors suggsted that abnormal gait patterns in patients with arthritic knees could be due to diminished knee joint proprioception.

By contrast, *Suzuke and Takahama (1979)* reported that the average range of sagittal plane knee motion was less in patients with OA than in those with RA. *Stauffer et al (1975)* observed that degenerative arthritis of the knee causes a limitation of internal and external rotation as a result of decreased flexion during gait. A positive correlation was observed between knee flexion and extension range and maximal isometric quadriceps and hamstring muscle force. The stronger the muscles, the greater was the range of motion of the knee in the sagittal plane. The authors claimed that flexion and extension range of motion during gait could be determined by the strength of muscles acting on the knee. The objective data obtained in the gait laboratory correlated with a clinical and radiographic scoring system. A significant correlation was also found between knee flexion in stance and pain during weight bearing.

Messier et al (1992) examined the changes in gait mechanics, and isokinetic knee strength, in 15 OA patients. Fifteen age-matched subjects without neuromuscular disease were recruited as a control group. The study revealed significant differences in gait parameters between the OA patients and the normal control group.

2.5. Review of gait analysis following total knee replacement

Several authors have studied gait characteristics following different types of TKR using various techniques. However, to date, there is still controversy regarding whether or not retention or sacrifice of the PCL in different types of prosthesis will improve gait and function after TKR.

All the above studies have utilised some form of gait analysis to assess patients' gait pattern after TKR. Some of these studies demonstrated considerable improvement in some gait parameters when comparing pre- and post-operative gait data. On the other hand, other studies revealed gait abnormalities following surgery. Increase in walking speed, stride length and knee range of motion during mid-stance and mid-swing are the most positive findings. Persistent abnormalities of the knee flexion-extension moment, and residual weakness of the quadriceps muscles have also been reported. The current evidence of the effects of PCL retaining and sacrificing TKR procedures on gait is inconclusive.

TKR is now becoming a standard method of treatment of OA. The problem is that the two major types of prostheses (retaining-PCL and sacrificing-PCL) are mechanically and structurally different, but it is not clear which of the two prostheses improve gait more, and it is exactly this point that my research is addressing.

Retention of the Posterior cruciate ligament is a controversial subject in TKR. Theoretically, retention of the PCL should be superior because PCL stabilises the knee in stance, which is essential for normal energy efficient walking. Reducing stability would result in poor gait. Sacrificing the PCL and replacing it with a posterior stabilizing articular surface provides greater stability. This implant attempts to mimic as closely as possible the natural movement of the knee joint. This additional stability is accomplished by a compromise surface geometry between the femoral and tibial surfaces. The difficulty in achieving the correct balance in the PCL could be a main argument against its usage.
The goal of this study was to address this issue objectively in order to see whether there is a difference between the patterns of movement associated with the two designs by means of a comprehensive gait analysis of patients with both types.

In 1974 *Chao and Stauffer* performed a quantitative gait evaluation of 29 healthy subjects and 95 patients with OA of the knee. The temporal parameters of gait, and the three components of the ground reaction force were recorded

using knee electrogoniometers, foot switches and a force plate. The team made an important advance in methodology, creating a program for the evaluation of the biomechanics of knee replacement. However, much of the data generated could not be clearly interpreted. No attempt was made to calculate moments and powers across the joints.

Later on, *Collopy et al (1977)* studied the functional performance in 20 patients with OA and RA of the knee before and one year following TKR with geometric prosthesis (PCL-retaining). They recorded velocity, cadence, stride length and duration of weight-bearing. Forward, lateral and vertical motion of the head was measured, as well as the range of knee flexion and extension. Walking speed, knee extension and isometric knee flexor muscle strength, and weight bearing on the operated limb during standing, improved more in patients with RA than in those with OA. Generally, however, patients with RA did not reach the functional level attained by the patients with OA. The authors also found an improvement in knee extension and flexion torques in patients with RA which continued for up to 12 months after surgery. By contrast, these decreased in patients with OA at three months postoperatively. At 6 and 12 months after surgery, these indices returned to pre-operative levels.

Murray et al (1983) studied 21 patients who had undergone TKR with the Marmor prosthesis. The gait measurements were carried out before surgery, and again 6, 12, and 24 months after the surgery. The authors documented improvements in knee function occurring in the first six months, and slow improvement up to the second year post-surgery. Murray and co-workers also found that the overall pre-operative to postoperative change was greater in patients diagnosed as having RA, than in patients with degenerative joint disease. Normal gait had not been restored. These results are in agreement with the study by *Collopy et al (1977)* of 20 patients with OA or RA who received geometric TKR, and that by *Kroll et al (1989)* of 18 patients with OA who underwent total condylar TKR.

Berman et al (1987) performed quantitative gait analysis after unilateral or bilateral PCL-sacrificing TKR. They subdivided their OA patients into three groups. Sixteen had unilateral TKR with a normal contralateral knee, 12 had unilateral TKR with radiographic evidence of moderately severe or severe arthritis of the contralateral knee, and 7 had bilateral TKR. The mean follow-up times for the first, second, and third groups were 18, 24, and 15 months respectively. The authors reported that the first and third groups demonstrated the greatest improvement in stride characteristics while the second group demonstrated the least improvement. However, gait parameters post-operatively were still comparable with those of age-matched normal controls.

Chen et al (1991) evaluated knee function following TKR in 13 OA patients with unilateral disease during level walking and stair climbing, using quantitative biomechanical kinematics and force plate measurements. An Insall-Burstein (I/B) prosthesis which sacrifices the PCL was inserted into 9 knees (6 patients), and a Miller-Galante (M/G) prosthesis which retains the PCL was inserted into 9 knees (8 patients). Patients after both types of TKR tended to walk slowly with shorter step length. The authors found no significant differences between the two types of prostheses in the kinetic and kinematic parameters. However, this could be due to the small sample size and the poor sensitivity of their methods of assessment.

A study by *Simon et al (1983)* compared the gait patterns of 12 patients with unilateral knee OA and fifteen age-matched controls without neuromuscular disease using a high-speed motion camera, two force plates, and EMG. At two years following TKR with a PCL retaining prosthesis, no differences were found between the patients and the control group for any of the gait parameters studied. EMG showed normal phasic muscle activity during level walking two years following TKR.

Andersson and associates (1981) compared the gait pattern in 26 patients with three different prostheses: 10 polycentric (PCL-retaining), 10 Gunston (PCL-retaining), and 6 geomedic. Gait measurements were made in all subjects just before surgery, and again at three to six months for 16 patients, at seven to twelve months for 15 patients, and at 13 months or more for ten patients post surgery. The instrumentation used included photocells, foot switches, and a multicomponent force-plate. Gait analysis included time-distance measurements, cadence, duration of single and double-limb support, and timing of swing phase, and the three components of ground reaction force, medio-lateral, antero-posterior, and vertical. The study revealed improvements in gait velocity, stride length, and double limb and single limb-support time and ground reaction forces after TKR. The authors claimed that improvements were significantly correlated with a reduction in pain.

The various types of prostheses used for TKR have different effects on gait. In one study by *Andriacchi et al (1982)* five prostheses of different designs (geometric (PCL-retaining), Gunston (PCL-retaining), total condylar, duoptellar (PCL-retaining), and Cloutier) were compared during level walking and stair climbing in 26 patients. All the above prostheses preserve PCL except the total condylar implant. An age-matched group of 14 elderly normal healthy subjects was also studied. Motion of the lower limbs was monitored by observing the spatial position of six light-emitting diodes. A piezoelectric force platform provided the three components of ground-reaction force, vertical twisting moments, and location of the resultant forces acting at the foot. The study confirmed that the least constrained cruciate-retaining prostheses resulted in a better gait regarding knee range of motion, while the cruciate sacrificing prostheses provided a smaller functional range of motion during stair climbing. However, this was associated with a slightly shorter stride length, a reduced mid-stance knee flexion, and an abnormal pattern in the external flexion-extension moment.

Normal gait patterns were not restored in any of the patients after TKR, in spite of excellent clinical results. The study indicated that after surgery, even asymptomatic patients with a successful clinical outcome demonstrated decreased walking speed, stride length and knee flexion during mid-stance. An abnormal pattern of flexion-extension moments at the knee joints during level walking was seen with all five prosthetic designs. It was suggested by the authors that the abnormal flexion-extension knee joint moments in patients with PCL-sacrificing prosthesis are due to an abnormal tibial/femoral contact area during knee flexion, diminished knee joint proprioception, and reduced mechanical efficiency of the extensor mechanism. EMG activity was not recorded during level walking and stair climbing in this study.

In another study, *Andriacchi et al (1985)* compared level walking and stair climbing between 14 healthy adult subjects and 48 patients (56 knees) fitted with 3 different designs of TKR. Sixteen patients had total condylar designs (PCL sacrificed), 18 had Cloutier total knee designs, and 14 had posterior stabilized total condylar designs. The authors found that patients with PCL sacrificing designs demonstrated a dynamic gait adaptation characterised by forward lean of the body during stair climbing, and speculated that this adaptation was secondary to a weakness in quadriceps activity or due to poor knee stability during stair climbing. The more normal gait in the PCL-retaining knees could be due to more normal femoral rollback, adequate proprioceptive feedback and good quadriceps muscle function. The findings of this study were similar to those reported by *Andriacchi et al (1982), Chao et al (1980), Simon et al (1983), Rittmen et al (1981), Stauffer et al (1981).*

Kelman et al (1989) studied a small number of elderly patients with unilateral PCL-retaining prostheses. Advanced gait analysis technology (high-speed cinephotograghy, EMG, and Kistler force-plates) was used. The team reported that in patients with PCL-retaining TKR, the motion in the sagittal plane of the replaced knee was very similar to that of the same motion of the contralateral knee, and nearly equal to that of a control group when ascending and descending stairs. *Andriacchi et al (1982)* reported similar findings in patients with PCL-retaining prostheses. EMG data revealed significantly greater muscle activity about the replaced knee in all muscle groups tested.

A functional comparison of PCL-retaining and PCL- sacrificing TKR was carried out by *Leffers et al (1983).* They assessed 12 patients who had undergone replacement in one knee with a PCL-retaining prosthesis and in the contralateral knee with a total condylar PCL-sacrificing prosthesis. Differences were noted between the two designs on level walking and stair climbing. Patients with a PCL-sacrificing prosthesis exhibited an inferior performance compared to patients with a PCL-retaining prosthesis. In neither group of patients did gait approximate normal patterns, despite an excellent clinical result with complete pain relief and increased physical endurance.

In a later study from the same centre, *Dorr et al (1988)* performed gait analysis and clinical evaluation in 11 patients with bilateral TKR. Each patient had a PCL-retaining procedure on one knee and a PCL-sacrificing operation on the other. Instrumental gait analysis was carried out pre-operatively and at six months and two years after surgery. The authors found differences between the two prosthetic designs on both level walking and stair ascending. They noted that PCL-sacrificing procedures were associated with greater loading and higher interface forces in stance and speculated that these effects may reduce the durability of the prosthesis. In contrast to the findings of *Andriacchi et al (1982)* no difference in angle of knee flexion was noted between the two designs during stair descent *(Rosenberg et al 1994).* The study demonstrated maximum quadriceps activity during stair climbing in patients with PCL-retaining and -stabilizing designs. Patients with PCL-sacrificing designs required increased use of the soleus muscle while stair ascending, suggesting that a forward lean was used by patients with the PCL-sacrificing designs, similar to that described by *Andriacchi (1988), Andriacchi and Galante (1988).*

The study indicated that maximum quadriceps activity was noted during stair climbing with PCL-retaining TKR. The PCL-sacrificing prostheses required more muscle activity for stability during stance. In particular, soleus muscle activity was increased during stair descent, suggesting that this muscle attempts to substitute for the PCL. Furthermore, varus and flexion moments were higher following PCL-sacrificing knee prosthesis. The authors claimed that this may lead to decreased longevity, durability and effectiveness of the prosthesis

However, the above findings have been challenged by other studies. *Rittman et al (1981)* compared the range of knee joint motion in three planes in patients with four types of total knee implants. Nine subjects had variable axis knees, 9 had geometric knee implants (PCL-retaining), 8 had Herbert knee, and 6 had Shiers knee implants. There were no differences in the gait parameters between the groups, and flexion during both stance and swing phases were reduced in all cases. The authors concluded that these alterations in the knee motion pattern during level walking were not implant-specific.

Similar results were reported by *Chao et al (1980),* who used a three dimensional electrogoniometer, two Kistler force plates and foot switches to evaluate the gait patterns of 154 patients before surgery, and 89 patients one year after surgery. Sixty-nine healthy subjects served as a control group. Three different groups of prosthesis (constrained, semiconstrained, and unconstrained) were studied. The constrained prostheses showed improved results in patients who had poor function pre-operatively. The unconstrained and semiconstrained prostheses improved gait in patients whose pre-operative function was not severely impaired. Patients with RA achieved a better functional score after surgery compared to patients with OA, in spite of the fact that patients with RA had worse joint function before surgery. Eight significant gait variables were found to be important in discriminating between those with normal and abnormal gait. The study demonstrated increased stride length and knee range of motion after total knee replacement.

These findings are in agreement with those of *Wilson et al (1996)* who demonstrated that there were no significant differences between the gait of patients who had a PCL sacrificing total knee arthroplasty and a control group. *Wilson et al* studied sixteen patients implanted with the posterior-stabilised prosthesis (Insall-Burstein Posterior Stabilised) and thirty two age-matched healthy subjects using a comprehensive gait analysis approach, measurement of isokinetic muscle function and EMG data analysis of the entire gait cycle following total knee arthroplasty. There was a significant decrease in functional range of motion during level walking and stair descent in patients with PCL-sacrificing prosthesis. They found that patients following TKR had normal isokinetic strength and normal overall phasic motor firing on EMG. The authors demonstrated that there were no significant differences between patients with a PCL-sacrificing prosthesis and a control group during level walking and stair climbing. but 25% reduction in the peak flexion moment during stair ascending. The lack of statistically significant differences in this study could be a result of the small sample size *(Andriacccchi and Hurwitz 1997).*

Steiner et al (1989) studied a group of 11 patients in whom 14 TKR prostheses were implanted. Patients were studied pre-operatively, then at 3, 6, and 12 months post-operatively. While the study demonstrated that most of the improvements in gait parameters had occurred by three months after TKR, at six months, knee moment had returned to the pre-operative levels. The researchers reported that only 50% of patients demonstrated phasic quadriceps activity and improved phasic hamstrings and anterior tibialis patterns at six months after surgery.

Kramers-de Quervain et al (1997) assessed 5 elderly OA patients 2 to 5 years following bilateral total knee arthroplasty. The mean age of the subjects was 75 years. Two different designs of prostheses (semiconstrained loose hinged prosthesis and unconstrained low-contact-stress mobile) were used in each subject. A three-dimensional motion analysis system (Vicon), a Kistler force plate for ground reaction force measurements, and dynamic EMG were used. In contrast to the findings of *Simon et al (1983)* the velocity and stride length of the patients were reduced. EMG data also demonstrated abnormal muscle activity of the quadriceps and hamstrings. The study did not demonstrate design specific functional changes of gait during level walking. The authors concluded that asymmetric dynamic knee flexion/extension motion patterns were explained not only by the differences in implant design but by other important variables which affect motion patterns. These include interference by the pattern of the contralateral side, the effect of the prosthesis on the patella/extensor mechanism on the passive range of motion, ligament balancing, impaired proprioception, and pre-operative gait habit.

Bolanos et al (1998) performed an isokinetic strength testing and gait analysis in 14 patients with bilateral TKR. Each patient had a PCL-retaining prosthesis on one knee and a PCL-stabilised prosthesis in the contralateral knee. Measurement of isokinetic muscle function and instrumental gait analysis were carried out pre-operatively. The mean follow-up was ninety-eight months after surgery. The authors found no differences in the isokinetic muscle testing parameters for both quadriceps and hamstrings. Also, no significant differences were found in gait parameters and electromyographic waveforms between the two prosthetic designs on both level walking and stair climbing.

In the same year, a Japanese study by *Ishii et al (1998)* performed a gait analysis in 11 patients with PCL-retaining prosthesis and in 9 patients with PCL-stabilised prosthesis. Knee motion was measured using an electrogoniometer to measure the three-dimensional motion of the joint. The average time after surgery was 42.6 months for patients with retention of the PCL, and 25.9 for patients with the posterior stabilizing articular surface. The authors found no significant differences regarding peak knee flexion in the stance and swing phase between the group with retention of the PCL and that with the posterior stabilizing articular surface. The study did not examine the differences in the kinetic parameters and muscle activity during level walking.

In a recent quantitative gait analysis study, *Otsuke et al (1999)* studied gait patterns in 38 women with two types of semiconstrained implants. Gait analysis was conducted using a pressure measuring system before surgery, and again every three months for a year. Gait parameters included the timing of the phases of the gait cycle, measurement of the vertical ground reaction forces and location of centre of pressure on the footprint.

These studies demonstrate the importance of instrumental gait analysis in the functional assessment of gait in patients with TKR. Numerous other studies have also addressed this subject (*Chao 1975, Stauffer et al 1975, Andriacchi et al 1977, Kimura et al 1978, Andriacchi et al 1980, Ferkul et al 1981, Rittman et al 1981, Skinner et al 1983, Peat et al 1984, Smidt et al 1984, Gore et al 1986, Sweeney et al 1986, Horwood et al 1988, Steiner et al 1989, Mattsson et al 1990ab, Berger et al 1990, Berman et al 1991, Chao et al 1991, Giannini et al 1992, Mikosz et al 1993, Weidenhielm et al 1993, Ramakrishnan et al 1994).*

Several authors have studied gait characteristics following different types of TKR using various techniques. However, to date, there is still controversy regarding whether or not retention or sacrifice of the PCL in different types of prosthesis will improve gait and function after TKR.

All the above studies have utilised some form of gait analysis to assess patients' gait pattern after TKR. Some of these studies demonstrated considerable improvement in some gait parameters when comparing pre- and post-operative gait data. On the other hand, other studies revealed gait abnormalities following surgery. Increase in walking speed, stride length and knee range of motion during mid-stance and mid-swing are the most consistently reported positive findings. Persistent abnormalities of the knee flexion-extension moment, and residual weakness of the quadriceps muscles have also been reported. The current evidence of the effects of PCL retaining and sacrificing TKR procedures on gait is inconclusive.

2.6 Hypothesis

The hypothesis to be tested is that a posterior stabilized knee prosthesis sacrificing the PCL will result in an advantage over a retaining prosthesis in respect of kinetic (knee moment and power), kinematic (knee flexion in mid-stance and mid-swing) and temporospatial (walking speed, stride length and percentage of stance phase) gait parameters and that this improvement would be reflected in better functional ambulation.

<u>Summary:</u>

Osteoarthritis of the knee joint is a common cause of abnormal gait in elderly subjects. Understanding of the gait abnormalities in patients with OA of the knee is essential for the optimal management of these patients and for the evaluation of treatment outcomes.

Joint replacement is a well recognised way of managing such problems if other methods have failed. The knee joint is unique in having intra-articular ligaments to provide stability. TKR is now becoming a standard method of treatment of OA. The problem is that it is not clear which of the two major types of prostheses (retaining-PCL and sacrificing-PCL) which are mechanically and structurally different improves gait more, and it is exactly this point that my research is addressing as the current evidence is inconclusive.

Theoretically, retention of the PCL should give better results because the PCL stabilises the knee in stance, which is essential for normal energy efficient walking. Reducing stability would result in poor gait. On the other hand it could be argued that sacrificing the PCL and replacing it with a posterior stabilizing articular surface provides more stability.

The cam mechanism helps to substitute for the biological stability provided by the PCL and aids in femoral rollback, thereby attempting to mimic as closely as possible the natural movement of the knee joint. Also, stability is provided by the similar surface geometry of the femoral and tibial surfaces of the prosthesis.

The present study aims to determine objectively whether there is a difference between the patterns of movement associated with the two designs by means of a comprehensive gait analysis of patients fitted with each type.

CHAPTER THREE
PILOT STUDY

3.1. Introduction

Before undertaking the main project, the investigator practised using the CODA motion analysis system and its software, the Kistler force plate and electromyography (EMG) in order to familiarise himself with the equipment and to reduce the possibility of operator error occurring during recording. Due to the fact that the apparatus was new to the investigator, formal instruction and guidance from the Research Unit experimental officer (Mr. Burnett) was obtained over a period of 3 weeks, thereafter assistance and advice continued to be available on request. At the beginning of the study the investigator wished to ascertain the repeatability of data collection. Therefore, before undertaking the main project, the investigator carried out intra-rater reliability tests of gait analysis using the CODA motion analysis system.

3.2. Aims

The aims of the pilot study were :

1- to familiarise the researcher with the measurement techniques and to solve any unexpected problems with data collection before the main study;
2- to check the intra-rater reliability of gait analysis with the Coda mpx30 system and the repeatability of the EMG pattern of normal walking;
3- to test the acceptability of walking during gait with one arm flexed, both arms flexed and both arms swinging and determine whether the gait pattern is affected by this or not;

4- to amend the data collection procedures in order to avoid possible sources of error.

3.3. Equipment

The equipment used in the study was located in the Gait Laboratory at Southampton General Hospital (SGH). For a detailed description of the equipment used, please refer to sections 4.8.1 to 4.8.3.

3.4. Intra-investigator reliability

3.4.1. CODA mpx30 system

The purpose of this test was to assess the intra-rater reliability of recording using the Coda mpx30 motion analysis system.

A group of 8 healthy subjects (4 male, 4 female) ranging in age from 20 to 45 years were instructed to walk at self-selected walking speed.

Kinetic data were recorded using a Kistler force plate embedded in the walkway. The force plate was used in conjunction with the Coda mpx30 system. Measurements were taken on three separate occasions: two on the same day (morning and afternoon) and one a week later. Joint range of motion was determined for the hip, knee and ankle joints.

3.4.2. Repeatability of the EMG pattern of normal walking

In order to observe whether the pattern of the rectified EMG of a healthy subject was consistent when measured on different occasions, the gait of the above 8 subjects in section 3.4.1 was recorded using contact surface electrodes on the following four muscles: rectus femoris, medial hamstring, tibialis anterior and gastrocnemius. The first occasion was in the morning and the second was in the afternoon of the same day. The third occasion was a week later.

3.5. Recording with one arm flexed, both arms flexed and both arms swinging

The signals from markers 1 and 2 i.e. the anterior superior iliac spine and posterior superior iliac spine (ASIS and PSIS) are usually obscured by the arm swinging between the CODA and the marker. Frequently the recording is made with one arm flexed and held on the side. This would give an uninterrupted view of the CODA markers but it is not a normal strategy during walking.

The aim of this pilot study was therefore to investigate whether recording gait with one arm flexed, both arms flexed or both arms swinging affected temporospatial, kinematic and kinetic gait parameters in the trunk and lower limbs.

Eight healthy subjects (4 male, 4 female) ranging in age from 50 to 80 years were instructed to walk at their preferred walking speed. Three tests were conducted for each subject. In one test, the arm on the same side as the tested knee was flexed. In a second test, both were flexed across the chest, and in the third, both arms were swinging.

3.6. Data analysis:

Mean and standard deviation were calculated by using Statistical Package for Social Sciences (SPSS Version 6.1.3, 1995) to analyse the data.

The analysis of variance for repeated measurement (ANOVA) was used to determine the statistical significance of differences in the gait parameters between the three test conditions and the T-test for differences between recording of gait with one aım flexed and both arms swinging. A p-value of 0.05 was regarded as the threshold for significance. The tests were recommended by Dr. Pickering in the Medical Statistics Department (personal communication).

3.7. Results

3.7.1. Intra-rater reliability test of the Coda mpx30 system

Summary statistics of the kinematic parameters of 8 normal subjects comparing the record on the three occasions are given in Table 3.1. There was no statistically significant difference between the 1st, 2nd, and 3rd recordings with respect to the range of motion of the hip, knee and ankle joints or EMG records. This is illustrated graphically in Figures 3.1-3.6.

One way analysis of variance of the kinematic parameters over the three occasions revealed no statistically significant difference by the same observer ($p > 0.05$). This suggests that the measurements of lower limb range of motion using the Coda mpx 30 can reliably be obtained by the same observer on separate occasions.

Table 3-1 Kinematic data from 8 healthy subjects whose gait was recorded on three different occasions at free walking speed.

Gait parameters (degrees)	1st Occasion Mean (SD)	2nd Occasion Mean (SD)	3rd Occasion Mean (SD)	P[1] value
Max hip extension in (ST)	42.66 (6.35)	43.34 (8.55)	44.96 (7.07)	0.88
Max knee flexion in (LR)	18.81 (6.26)	21.50 (6.29)	20.98 (6.17)	0.74
Max knee flexion in (SW)	59.93 (9.18	60.22 (6.26)	62.45 (7.79)	0.83
Max ankle dorsiflexion in (SW)	7.10 (3.48)	7.01 (2.71)	7.36 (2.16)	0.97

Key : Max=maximum; ST= stance phase; SW swing phase; LR=loading response; DF=dorsiflexion. [1]Analysis of variance

3.7.2. Repeatability of the EMG pattern of normal walking

Repeatability of the EMG pattern during walking at self selected speed was also checked in the present study by recording the gait of 8 healthy subjects on three different occasions. Inter-individual variability could be due to factors such as skin resistance, thickness of subcutaneous fat, and variability in placement of the surface electrodes.The results indicated that EMG patterns were reasonably consistent for a given subject.

3.7.3. Recording with one arm flexed and both arms flexed

The data recorded with both arms swinging were incomplete and have not been included in the statistical analysis, due to interrupted vision of the markers to the coda. Analysis of the results for the two remaining test conditions including the temporospatial, kinematic, and kinetic parameters are shown in Tables 3.2-3.4. There were no statistically significant differences when the test was performed with one, or both arms flexed during walking at the preferred speed except for the hip ROM ($p = 0.05$). It was possible to record from all subjects with one arm flexed, both arms flexed and from some subjects with both arms swinging, and this is shown graphically in Figures 3.7-3.8 for comparison.

Table 3-2 The mean, standard deviation (SD) and level of statistical significance of temporospatial parameters for 8 healthy subjects with one arm swinging and both arms flexed at free walking speed.

Gait parameters	One arm swinging Mean (SD)	Both arms flexed Mean (SD)	P*- value.
Walking speed (m/s)	1.28 (0.20)	1.25 (0.14)	0.69
Stride length (m)	1.35 (0.19)	1.28 (0.13)	0.78
Mid stance (% cycle)[1]	30.05 (0.97)	29.99 (0.65)	0.14
Mid swing (% cycle)[1]	80.05 (0.97)	79.99 (0.65)	0.14

[1]Timing of mid stance and mid swing expressed as percentage of the gait cycle.
**T-test*

Table 3-3 The mean, standard deviation (SD) and level of statistical significance of kinematic parameters for 8 healthy subjects with one arm swing and both arms flexed at free walking speed.

Gait parameters (ROM) in degrees	One arm swinging Mean (SD)	Both arms flexed Mean (SD)	P*- value.
Max hip extension in stance	42.66 (5.58)	40.84 (3.14)	0.05
Max knee flexion in loading phase	16.92 (6.17)	17.65 (6.05)	0.99
Max knee flexion in swing	59.65 (7.90)	60.63 (7.23)	0.78
Max ankle dorsiflexion in swing	7.86 (2.25)	7.28 (2.26)	0.84

Key: Range of motion (ROM); Maximum (Max).
**T-test*

Table 3-4 The mean, standard deviation (SD) and level of statistical significance of kinetic parameters for 8 healthy subjects with one arm swinging and both arms flexed at free walking speed.

Parameters	One arm swinging Mean (SD)	Both arms flexed Mean (SD)	P*- value.
Max Knee moment in mid-stance (Nm/kg)	0.14 (0.07)	0.14 (0.08)	0.81
Max Knee power in mid-stance (W/kg)	0.10 (0.08)	0.17 (0.09)	0.75
Max ankle moment in pre-swing(Nm/kg)	0.71 (0.28)	0.76 (0.33)	0.23
Max ankle power in pre-swing(W/kg)	4.90 (1.17)	4.29 (0.99)	0.35

Key : *Maximum (Max); mid-stance (MS); pre-swing (PS). Moment in Nm/Kg and power in W/Kg.*
**T-test*

Summary:

A pilot study was carried out to familiarise the researcher with the measurement techniques to be adopted in the main study. To test the intra-rater reliability of gait analysis with the Coda mpx30 system, to test the repeatability of the EMG pattern of normal walking, and to test the applicability of recording walking during gait with one arm flexed, both arms flexed and both arms swinging. No statistically significant difference was found in the range of motion of the hip, knee and ankle joints on three occasions.

There was no statistically significant difference between recordings with one arm flexed and both arms flexed. Recording subjects with both arms swinging was found to be impractical. The pilot study therefore established the intra-investigator reliability of the test procedure and indicated that the best way to record the data is with either one or both arms folded. Based on these findings, the researcher was able to proceed to the main study with reasonable confidence in the protocol to be used. The methods of the main experimental phase are described in the following chapter.

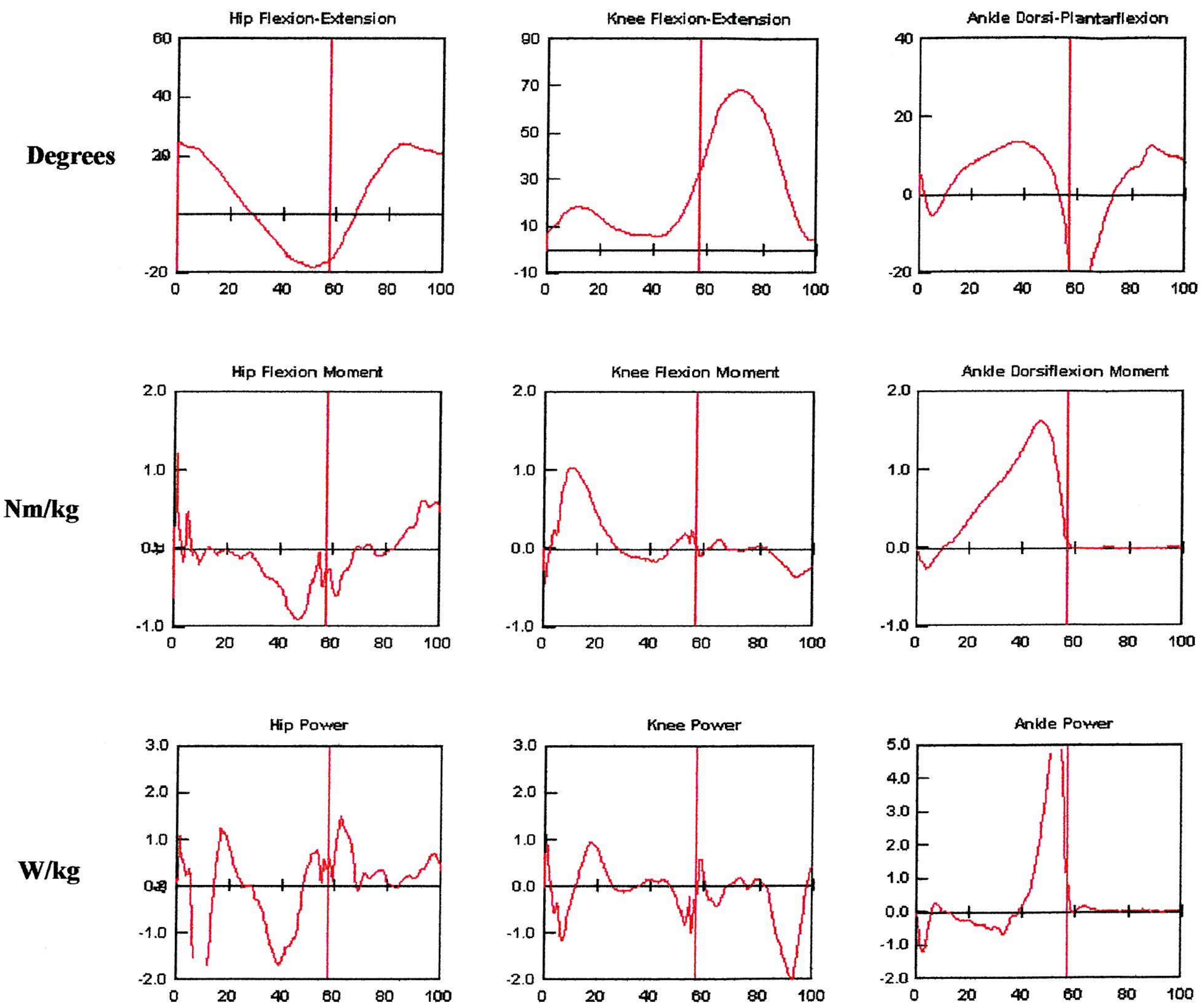


Figure 3-1 Sagittal plane joint kinematics and kinetics for the hip, knee and ankle of a healthy subject (DS1R4) in order to check the reproducibility of the gait pattern. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in watts/kg (first record).

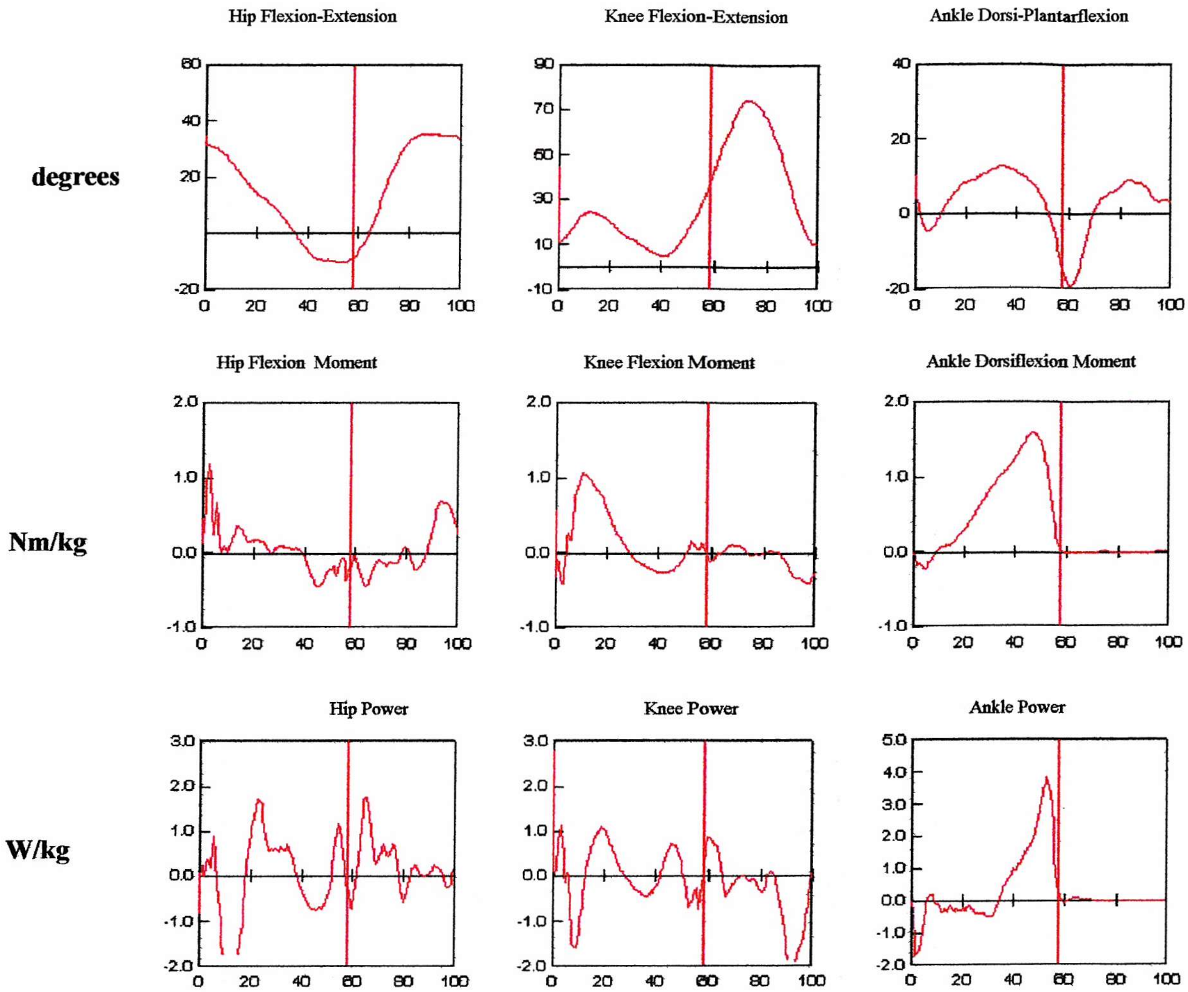


Figure 3-2 Sagittal plane joint kinematics and kinetics for the hip, knee and ankle of a healthy subject (DS2R4) in order to check the reproducibility of the gait pattern. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in watts/kg (second record).

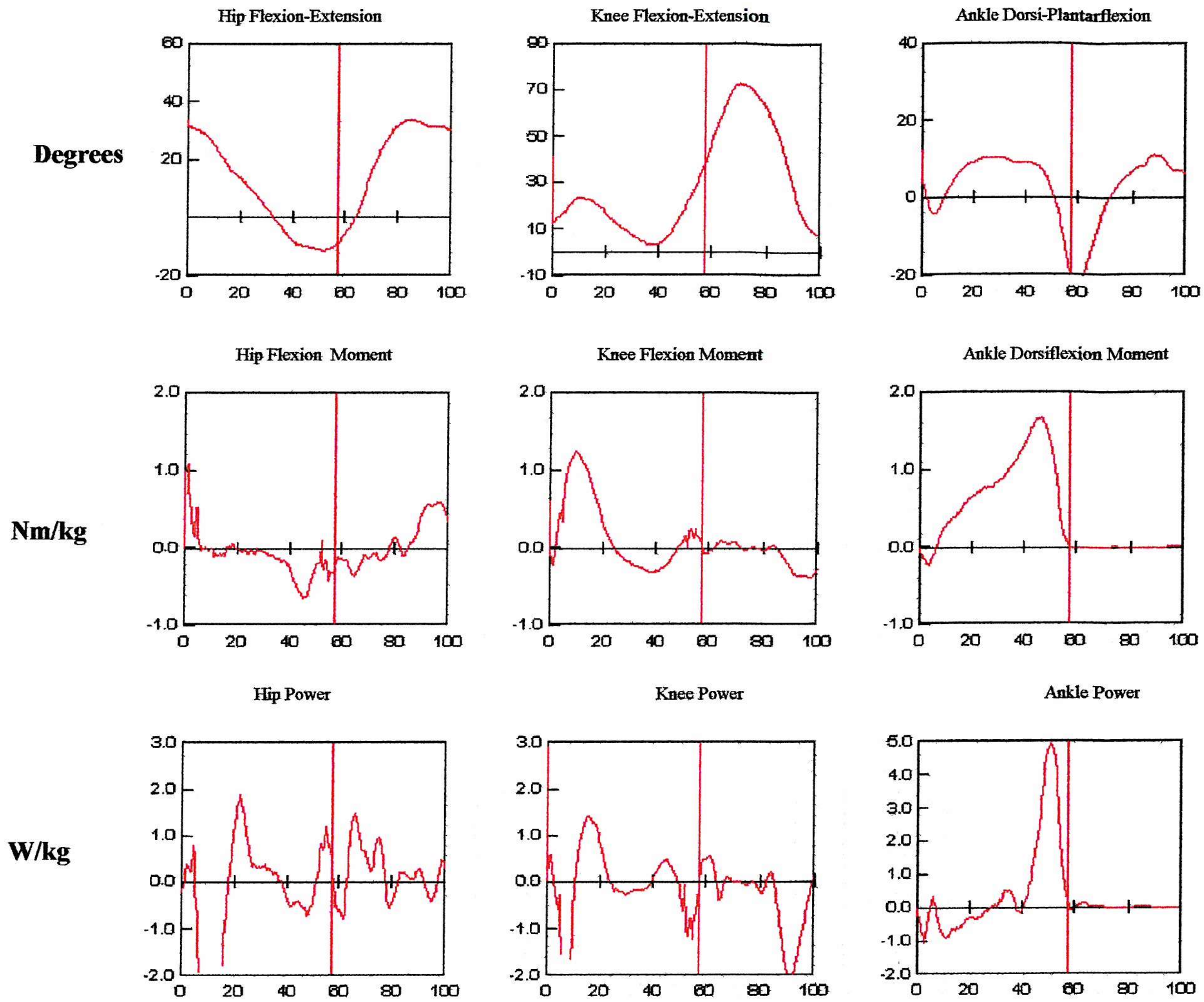


Figure 3-3 Sagittal plane joint kinematics and kinetics for the hip, knee and ankle of a healthy subject (DS3R4) in order to check the reproducibility of the gait pattern. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in watts/kg (third record).

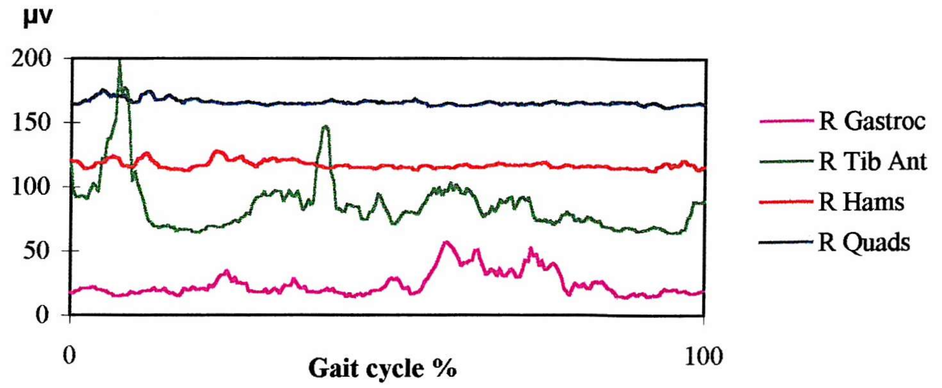


Figure 3-4 Recorded EMG of rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of healthy subject (OM1R5) in order to check the reproducibility of the sequences of the EMG activation (first record).

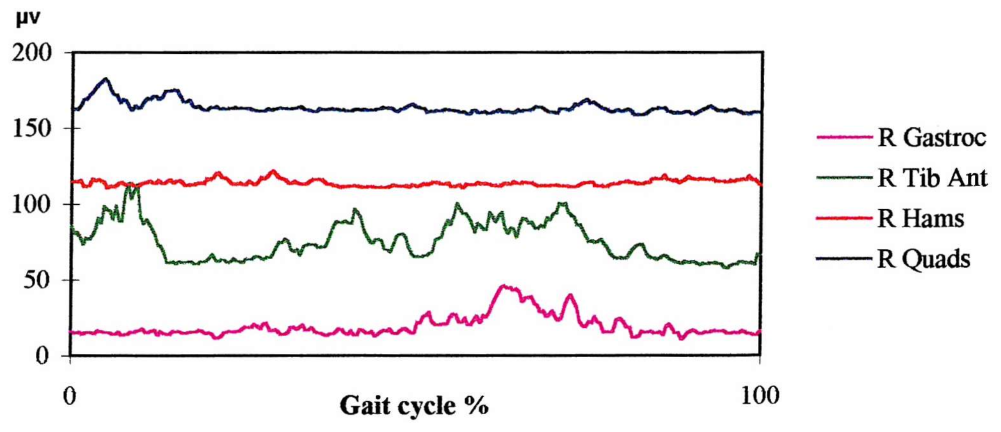


Figure 3-5 Recorded EMG of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of healthy subject (OM2R2) in order to check the reproducibility of the sequences of the EMG activation (second record).

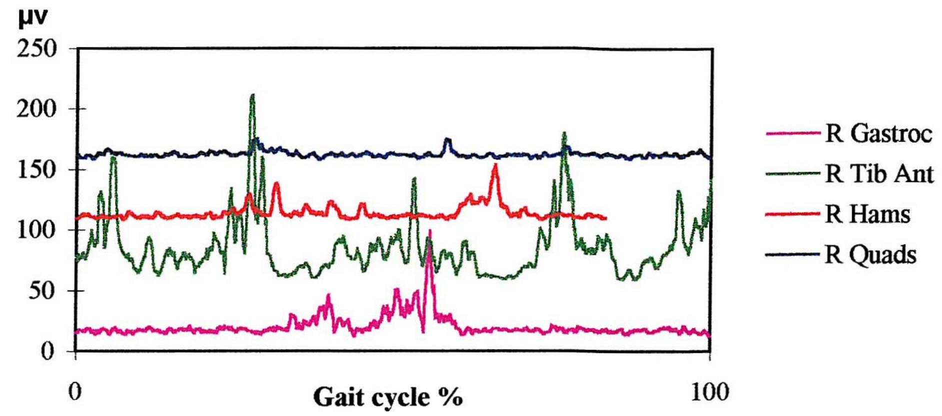


Figure 3-6 Recorded EMG of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of healthy subject (OM3R3) in order to check the reproducibility of the sequences of the EMG activation (third record).

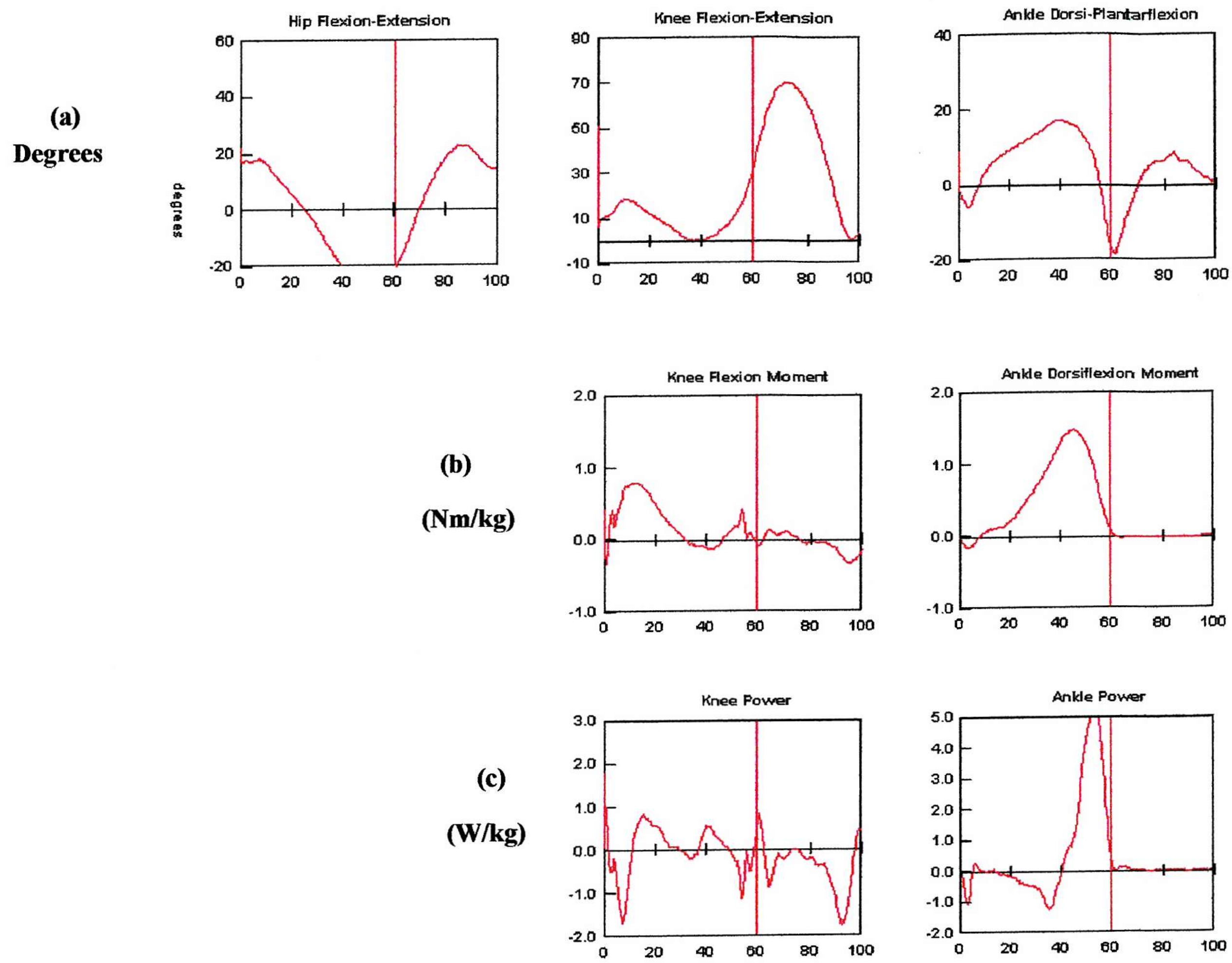


Figure 3-7 Sagittal plane joint kinematics for the hip, knee and ankle (a) and joint kinetics for the knee and ankle (b and c) of a healthy subject (XY) with one arm swinging.

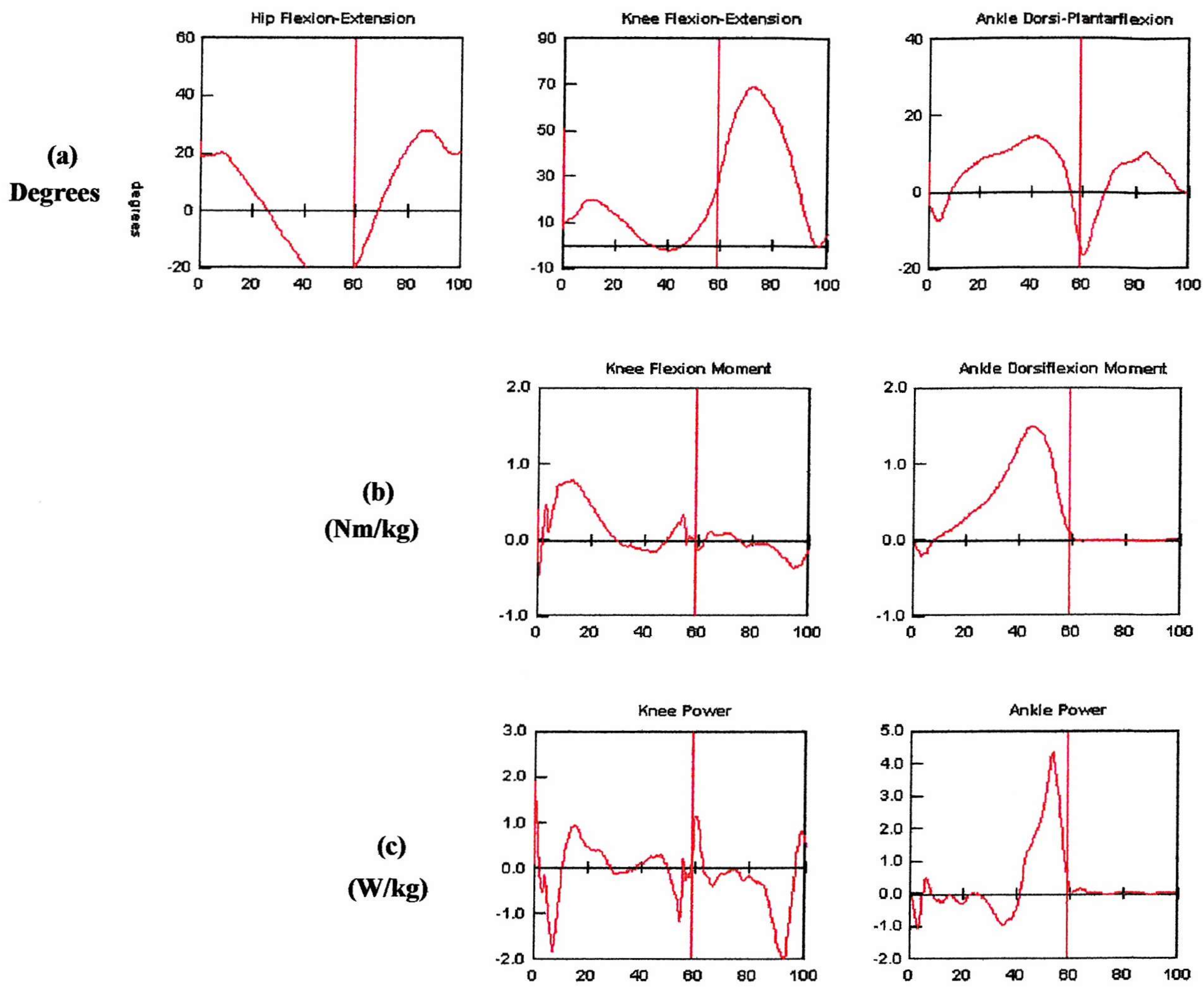


Figure 3-8 Sagittal plane joint kinematics for the hip, knee and ankle (a) and joint kinetics for the knee and ankle (b and c) of a healthy subject (XY) with both arms flexed.

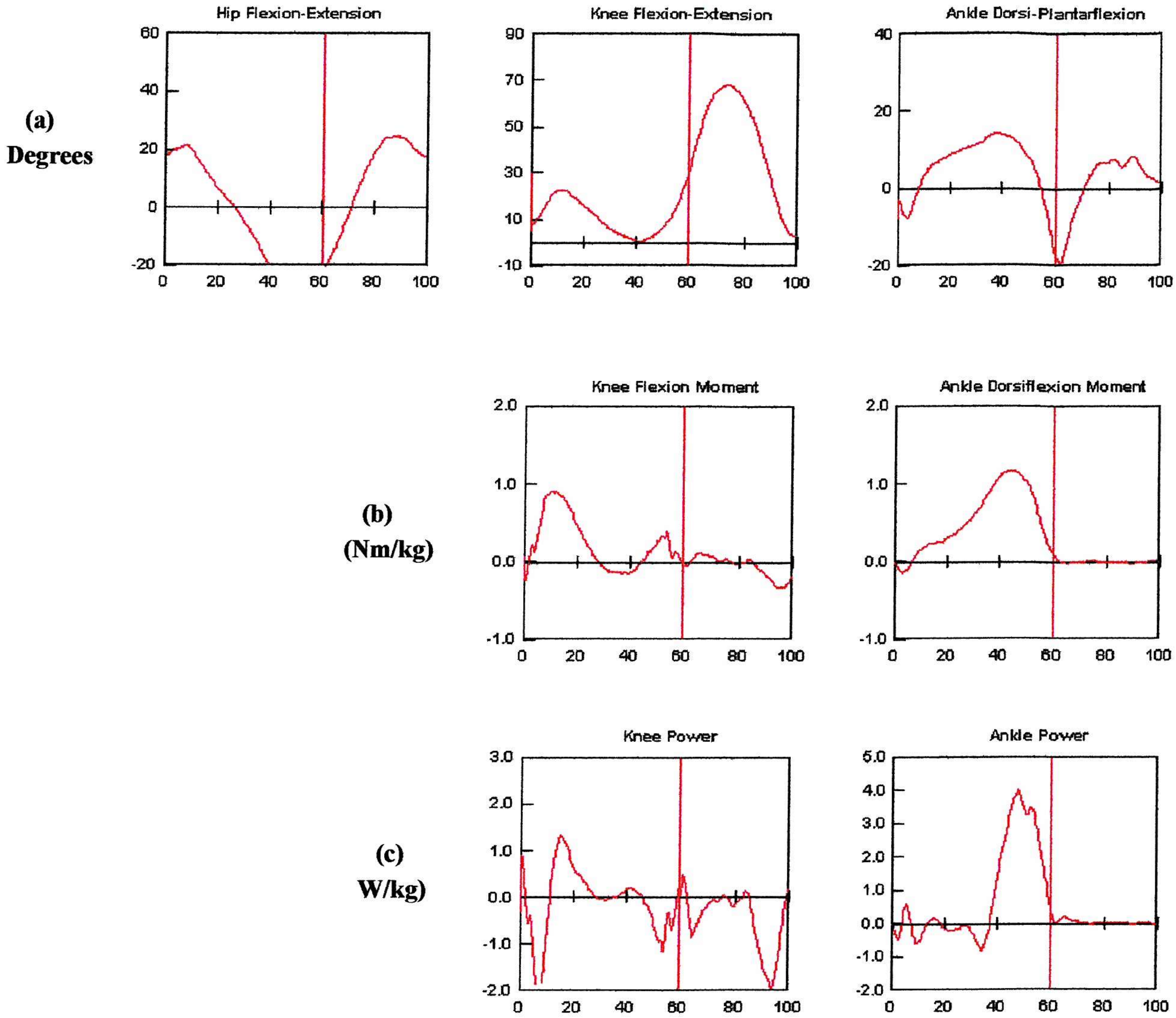


Figure 3-9 Sagittal plane joint kinematics for the hip, knee and ankle (a) and joint kinetics for the knee and ankle (b and c) of a healthy subject (XY) with both arms swinging.

CHAPTER FOUR

MATERIALS AND METHODS

4.1. Introduction

This chapter describes the study design, the population studied, ethical issues that were considered, the equipment used in the research and the procedures for collecting the data.

4.2. Aims of the study

The aims of this study were:

(1) To compare gait patterns of healthy people with patients with OA of the knee who are undergoing total knee replacement.

(2) To evaluate the impact on gait of two contrasting surgical procedures of TKR on patients with knee OA. One approach focuses on the use of PCL-sacrificing prostheses, while the other focuses on PCL-retaining prostheses.

(3) To evaluate the functional outcome of TKR on the patients' symptoms, e.g. pain and functional ability.

4.3. Recruitment of control subjects and patients

4.3.1. Control subjects

Fifty-five subjects were initially recruited and assessed. As all surgically treated subjects were aged 50 or above, only the results of control subjects of that age group will be presented in this thesis. There were 25 volunteers who were hospital employees, members of the hospital League of Friends, patients' relatives or personal contacts. All subjects were interviewed by the investigator to confirm the following criteria:

(1) No previous injury or history of knee surgery;
(2) No functional limitations of the hip, knee or ankle on walking;
(4) No leg length discrepancy;
(5) No medical history of musculoskeletal or neurological problems.

4.3.2. Patient group

Fifty-eight patients with OA who were admitted to the Department of Orthopaedic Surgery at Southampton General Hospital for TKR during the period between July 1997 and June 1998 were recruited. Thirty-four cases had involvement of one knee joint, and the rest had bilateral joint involvement.

4.3.3 Clinical description of the patients

The diagnosis of osteoarthritis of the knee was made by the orthopaedic surgeons on the basis of the clinical findings (e.g. pain, decreased range of movement, tenderness, swelling, crepitus on movement and walking distance affecting quality of life). In addition, radiological changes (e.g. narrowing of the joint space and bone formation) were considered.

In order to be recruited in to the study subjects had to meet the following inclusion and exclusion criteria:

Inclusion criteria were as follows :

(1) Age greater than 50 years;
(2) Subjects with radiological evidence of severe OA diagnosed by a consultant radiologist;
(3) All subjects due to have unilateral primary TKR surgery;

(4) The ability to walk 15m independently without an assistive device;

(5) Subjects without real leg length discrepancy (up to 0.5cm was permitted);

(6) Subjects with no symptoms in their hip and ankle joints in either limb;

(7) Subjects were not to have taken any analgesics for a minimum of three hours before the gait recording was conducted.

Exclusion criteria were as follows :

(1) Subjects having undergone previous major surgery on their other knee (e.g. replacement, osteotomy, fracture), their lower limbs, or their spine;

(2) Neuromuscular conditions (e.g. stroke, Parkinson's disease) or severe pulmonary or cardiac conditions that cause difficulties with ambulation;

(3) The presence of severe low back pain and / or lower limb disability affecting the gait;

(4) Patients undergoing any surgical procedure during the follow-up period which might affect gait;

(5) Subjects with leg length discrepancy of more than 0.5 cm. Limb length was measured with a tape measure from the anterior superior iliac spine to the medial malleolus from a supine position with the limbs parallel *(Hoppenfeld, 1976)*. This is necessary because leg length discrepancy affects the symmetry of hip, knee and ankle motion during walking *(Kaufman, et al 1996)*.

4.4. Ethical approval

The study was approved by the Southampton and South West Hampshire Joint Ethics Committee (Appendix 1). The Accident and Orthopaedic Directorate at Southampton General Hospital also agreed to allow the research project (Appendix 2). The purpose of the study was explained to the control subjects and patients and written information about the research and a consent form were provided (Appendix 3).

4.5. Assessment intervals

Control subjects were seen on one occasion. Patients were assessed one week before surgery, and 3, 6 and 9 months following TKR in the Gait Laboratory at Southampton General Hospital.

The patients were required to record the severity of pain one week before the operation and 3, 6, and again 9 months after their operation. The post-operative Cincinnati Knee Rating Scale and physiological cost index were completed one week before the operation and at the end of the third, sixth and ninth months after their operation. Gait analysis was performed one week before surgery and 3, 6 and 9 months following the knee replacement.

4.6. Primary outcome measures

Instrumental gait analysis allows the study of different aspects of gait. The parameters chosen for measurements in the gait analysis of patients with knee OA and following TKR were:

4.6.1. Temporospatial measures of gait

These are: walking speed, timing of mid-stance and mid-swing and stride length. These temporospatial parameters were chosen because of their importance for safe and energy efficient ambulation.

4.6.2. Kinematic analysis

The CODA motion analysis system was used to measure the maximum joint range of motion at the hip, knee, and ankle. These parameters were selected to characterize the dynamic range of motion of the joints and to facilitate comparison between different sessions in the follow-up period, not only graphically but also numerically and statistically. The following parameters were selected:

(a) Maximum hip extension in stance phase
(b) Maximum knee flexion during loading response (weight acceptance)
(c) Maximum knee flexion during swing phase
(d) Maximum ankle dorsiflexion in swing

Also, these kinematic parameters were chosen because of their importance for safe and energy efficient ambulation.

4.6.3. Kinetic analysis

Analysis of moments and powers, used together with temporospatial measures and kinematic analysis, give a more complete description of gait. The maximum knee joint moments and powers were directly measured in mid-stance, i.e. when the ground reaction force moves across the hip joint, anterior to the knee and ankle. At this point the ground reaction force reaches its maximum vertical height. The maximum ankle joint moments and powers were measured in the pre-swing phase when the ground reaction force passes posterior to the hip and knee joints and anterior to the ankle joint. (Moments give an indication of the tension of ligaments and muscles across the joint, i.e. dynamic stability. Powers relate to movement that is actually produced).

4.7. Secondary outcome measures

4.7.1. EMG data

Myoelectric signals were recorded from four muscles: rectus femoris, medial hamstring, tibialis anterior and gastrocnemius using surface electrodes.

4.7.2. Visual analogue scale (VAS)

Pain severity at rest was measured using the VAS (Appendix 4). Assessment of the degree of pain was made by patients as follows:

The patient determines the level of pain by marking a horizontal line 100mm in length, one end of which indicates 'No pain' and the other end, 'Pain as bad as it could be' *(Scott and Huskisson, 1976).* The patient is asked to record the level of pain that is being experienced at the time of the test. The distance from the left end of the line, 'No pain', to the patient's mark was then measured in millimetres and called the pain score. This scale has the advantage of being simple and can be completed by the patients so that any observer bias is excluded.

4.7.3. Cincinnati knee rating scale (CKRS)

Patients' functional activities were assessed using the CKRS. The CKRS is a functional questionnaire which is completed by the subject. It was originally designed to assess the motor functional ability of patients with knee ligament injuries following knee reconstructive surgery *(Noyse et al 1984).*

The scale has been found to be reliable and sensitive in the assessment of patients with knee problems *(Noyse et al 1984 & 1987).* It is a questionnaire which is completed by the patient and has 100 points as a total score. It measures severity of pain (20 points) (20 ='no pain' to 0 ='pain present all the time'), severity of swelling (10 points) (10 ='no swelling' to 0 ='severe problems'), degree of giving way (20 points) (20 ='no giving way' to 0 ='severe problems'), overall physical activity (20 points) (20 =' normal' to 0 =' severe problems'), walking activity (10 points) (10 ='normal' to 2 ='severe problems'), stair activity (10 points) (10 ='normal' to 2 ='severe problems'), running (5 points) (5 ='normal' to 1 ='severe problems') and jumping (5 points) (to 5 ='normal' to 1 ='severe problems').

Other items recorded for clinical purposes but not included as part of the score are location and type of pain, knee stiffness, kneecap grinding, and knee locking (Appendix 5). The method was chosen because it is simple and of proven validity and reliability in patients who had knee surgery.

4.7.4. Physiological cost index (PCI)

PCI was used to evaluate energy expenditure during walking. In recent years PCI has been used in research and as a clinical tool to study children and adults with pathological gait. It was found to be easy, reliable and valid *(MacGregor 1979 and 1981, Butler et al 1984, Rose et al 1990)*. Measurements of the energy cost of walking are an important part of gait analysis. Any deviation from the normal walking pattern, for example caused by pain, muscle weakness, or any flexion, valgus or varus deformity, will lead to an increase in the energy spent by the body *(Doderlein 1997)*. Assessment of energy expenditure therefore provides the clinician with valuable information about a person's functional capability and gait performance, and may influence further treatment *(Waters 1992, Waters and Mulroy 1999, Doderlein 1997)*.

The PCI was calculated by subtracting the pulse rate at rest (baseline) from pulse rate at the end of 25m. walk (beats per minute) and dividing that by walking velocity (metres per minute).

The patient was asked to sit down without speaking for five minutes before the resting pulse rate was recorded. Pulse rate was calculated for one minute. A red adhesive tape was placed on the floor at the starting point of the walk and another red line was placed at the end of the 25m walkway. The walkway was flat and smooth.

The investigator asked each patient to walk along the measured distance of walkway at a comfortable speed wearing their own shoes. At the starting point, each patient received a 'go' command. The stop watch was started as soon as the patient was given the 'go' command. When the patient reached the second red line, the time was recorded. The pulse rate at rest was counted again immediately, for one minute, by the investigator. The PCI was chosen for this study because of its simplicity.

Any medication the patients were taking which might affect heart rate (e.g. B-blocker) or analgesics were documented.

In the present study, the data derived from the PCI were uncertain value because most patients were receiving medication which could potentially have affected the heart rate. Therefore, no PCL data have been included in the statistical analysis.

4.8. Gait analysis equipment

Instrumental gait analysis was performed to provide objective measurements of gait. Kinetics, kinematics and EMG data were collected. The floor area of the walkway was 8 metres long and 4 metres wide and contained one force plate embedded in the floor. The gait analysis equipment used in the study consisted of:

- a CODA mpx 30 - a motion analysis system (Figure 4.1).
- a Telemetered Dynamic Electromyography (EMG) System (Figure 4.2).
- a Kistler force plate (Figure 4.3).

4.8.1. CODA mpx30 scanner unit

The CODA mpx30 was used in this study to record the temporospatial parameters and functional range of motion of the hip, knee and ankle joints. CODA mpx30 is a computerised three dimensional movement analysis system. CODA has a scanner consisting of three optical sensors mounted on a rigid frame. The scanner captures infra red light signals pulsed sequentially by markers placed on anatomical landmarks, a pelvic frame and thigh and shin wands.

The middle sensor tracks vertical movements, whilst the outer ones track the horizontal movements of the markers. The kinematic and force plate data were computed using a method of link-segmental analysis and inverse dynamics *(Charnwood Dynamics 1995)*. This mathematical model predicts the joint's centre from data recorded in two co-ordinates and anthropometric measurements, namely the patient's weight, height, width of the ankle and knee joints and the width and depth of the pelvis.

The CODA mpx30 system resolution is 0.1mm along the axis parallel to the machine (in X and Z), and 0.6mm along the perpendicular axis (Y) at a distance of three metres from the CODA scanner unit *(Charnwood Dynamics 1998)*. It is capable of locating targets with 0.1- 0.2 mm accuracy in the x, y, and z axes.

CODA mpx30 uses small infra-red light emitting diodes (LEDs) (Figure 4.4) which are placed on the subject to be examined. Each of the markers is number coded for automatic identification. These markers are powered by small rechargeable battery packs which are placed on the subject close to the markers (Figure 4.5). Each battery pack has two small sockets into which the individual LED markers are plugged. Each socket has a number which indicates the identity of the marker to be plugged into that socket and the movements are tracked by the CODA mpx30 sensors.

4.8.2. Force plate

Kinetic data were recorded using a Kistler 9281B force plate (Kistler Instruments Limited, Switzerland), which is integrated with the CODA mpx30 system. When the foot touches the force plate the reaction forces are converted into electrical signals relating to their direction and magnitude. The force vector and intersection point with the floor are used to calculate moments and powers that have been generated during walking. The force plate is an aluminium plate, 40cm by 60cm, placed in the middle of the walkway, with its surface level to the floor surface. In this study the moments and powers at the knee joint were recorded. All data were obtained from subjects walking barefoot to eliminate variations caused by shoe design.

4.8.3. Telemetered dynamic electromyography system

The integrated telemetered Charnwood Dynamic electromyography (EMG) system consists of pre-prepared adhesive conductive electrodes (silver-silver chloride, 6mm in diameter) that are attached over the muscle to be tested. The electrodes connect to pre-amplifiers via press-studs. The amplifiers connect to the transmitting telemetry unit via light wires.

The pre-amplifier modules are wired to a belt pack in which analogue data is converted to digital data for transmission via infra-red telemetry to the host computer. The transmitting telemetry unit contains circuitry which further amplifies and rectifies the EMG signals from each of the amplifier modules. The transmitter unit encodes the data from each of the EMG channels and transmits it via infra-red optical signals to the receiver unit situated near the host computer.

The receiver unit is connected via a digital link into one of the three digital signal processor (DSP) cards in the host PC used by the CODA system. The EMG signals are acquired synchronously with the CODA and force plate data. The processor then transmits the raw rectified EMG data to a receiver module in the host computer using infra-red telemetry.

At the same time, the processor on the belt pack produces low frequency envelope signals corresponding to the EMG activity sensed by each pre-amplifier. These low frequency signals return to the pre-amplifier modules via the same leads as the original signals, where they illuminate red LEDs. Thus, indication of the muscle activity can be visualised. The Motion Analysis application software processes and displays the EMG data. The EMG signals are full wave rectified and sampled at a constant rate of 1600 Hz, giving effective frequency bandwidth of 35-800Hz. Successive samples of a given channel are then averaged and the result transmitted at 400 Hz. Transmission of the EMG signals occurs during time intervals when CODA markers are not emitting light. The advantage of the dynamic EMG telemetry system is that during recording there is no need for wires to trail from the subject, which might alter gait, particularly in patients who have locomotion impairment.

4.8.4. Other measurements

Anthropometric measurements were made. These are necessary for the calculations of the kinetic parameters of gait. The height of the patient when barefoot was measured using a standard scale mounted on a wall. Weight was also measured barefoot using an electronic weighing scale. The width of the ankle and knee joints were measured with dial calipers (Figure 4.6). A metal ruler was used to measure the width of the front pelvic frame, and the depth of the side pelvic frame. An alignment frame was used to define the axis of rotation of the ankle joint (Figure 4.7).

Figure 4.1 CODA mpx30 scanner unit

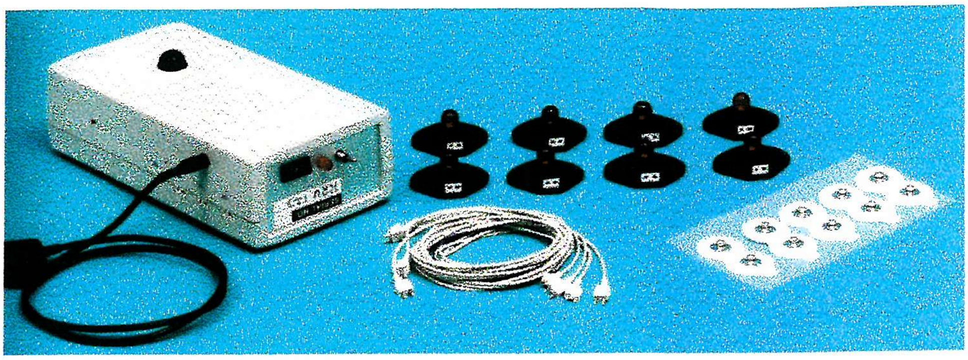


Figure 4.2 Electromyography surface electrodes and transmitter pack.

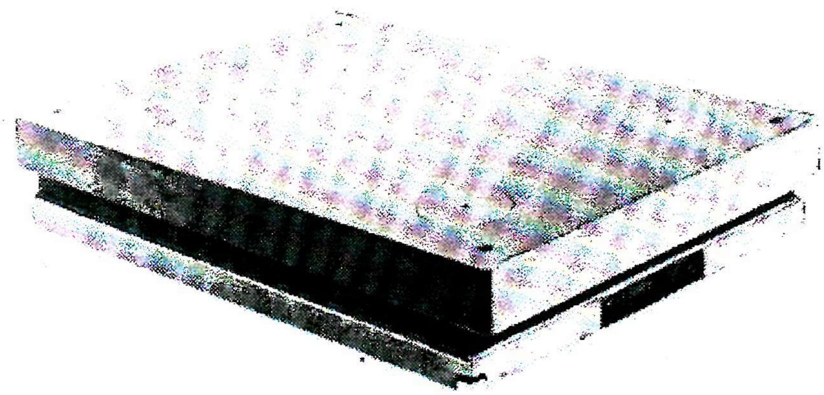


Figure 4.3 Kistler force plate.

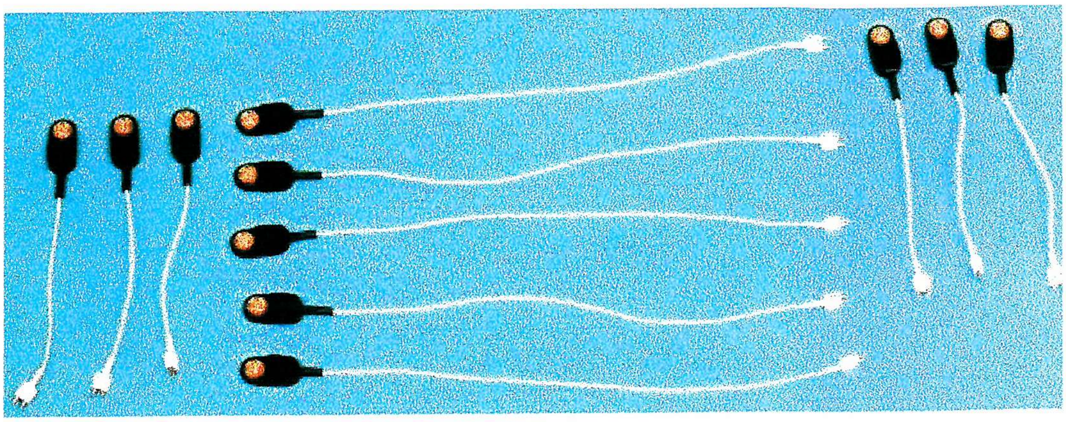


Figure 4.4 Coda markers.

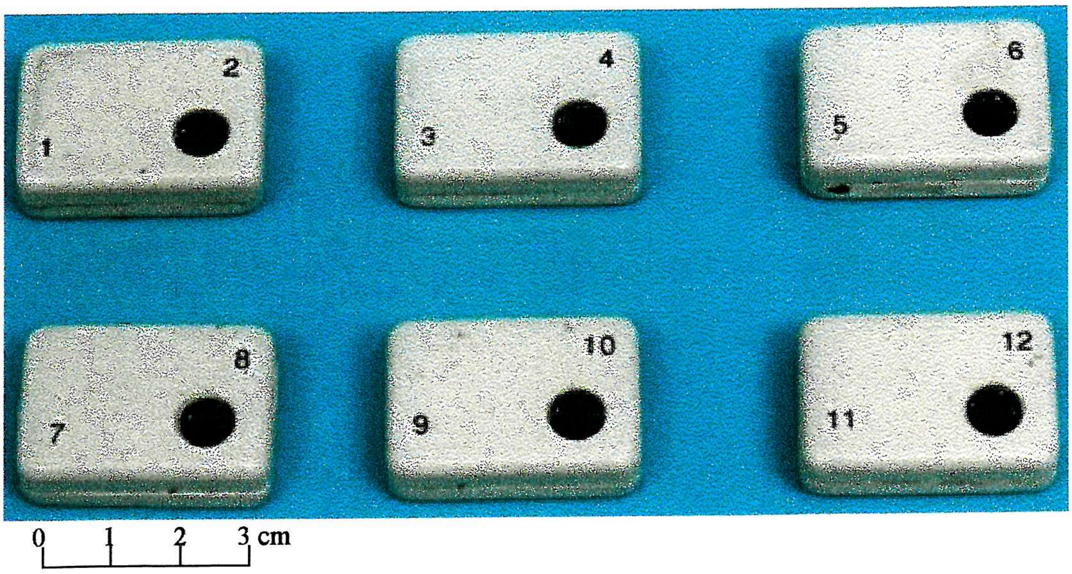


Figure 4.5 Battery packs.

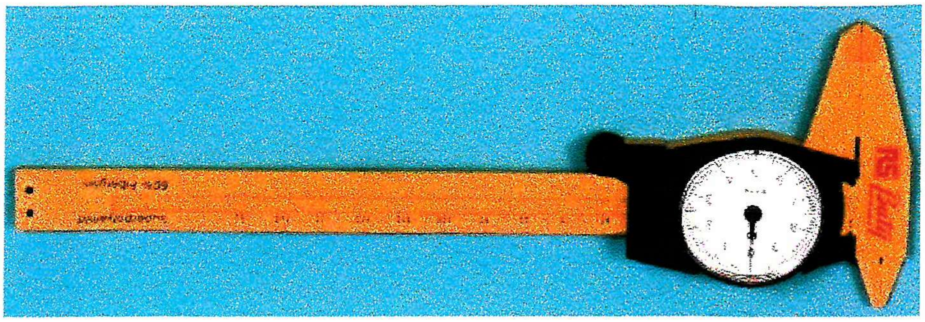


Figure 4.6 Dial calipers

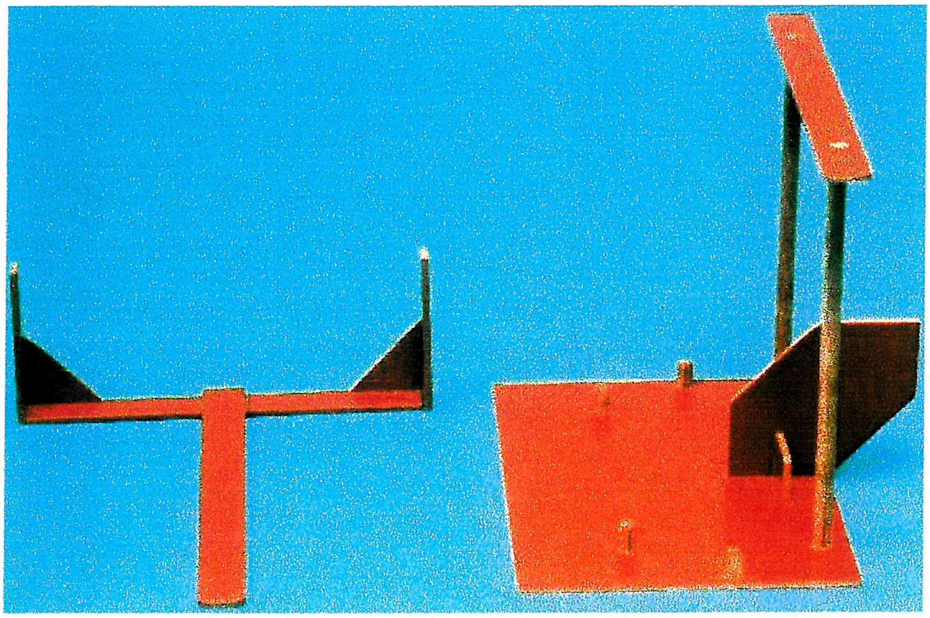


Figure 4.7 Alignment frame

4.9. Procedure

4.91. Position of the subject

A verbal explanation was provided before the subjects signed a consent form. Before recording gait, each subject was seated on a comfortable stool and the whole procedure was explained in detail.

4.9.2. Placement of markers

The placement of the markers was carried out in accordance with the guidelines in the CODA mpx30 user manual. Eleven markers were positioned as shown in (Figure 4.8).

Number 1 was positioned near the end of the sacral wand.

Numbers 2 and 3 were attached to the pelvic frame, positioned over the posterior and anterior superior iliac spine (PSIS and ASIS).

The thigh and shank wands were fastened and adjusted using velcro straps. Thigh markers were fixed on thigh wands just above the knee and consisted of two markers for the anterior and posterior femoral (Ant.Fem and Pos.Fem) positions.

Number 5 'Ant. Fem' was positioned near the anterior end of the thigh wand.

Number 6 'Post. Fem' was positioned near the posterior end of the thigh wand.

Two tibial markers, anterior and posterior (Ant.Tib and Post.Tib) were attached to the shank wand just below the knee.

Number 7 'Ant. Tib' was positioned near the anterior end of the shank wand.

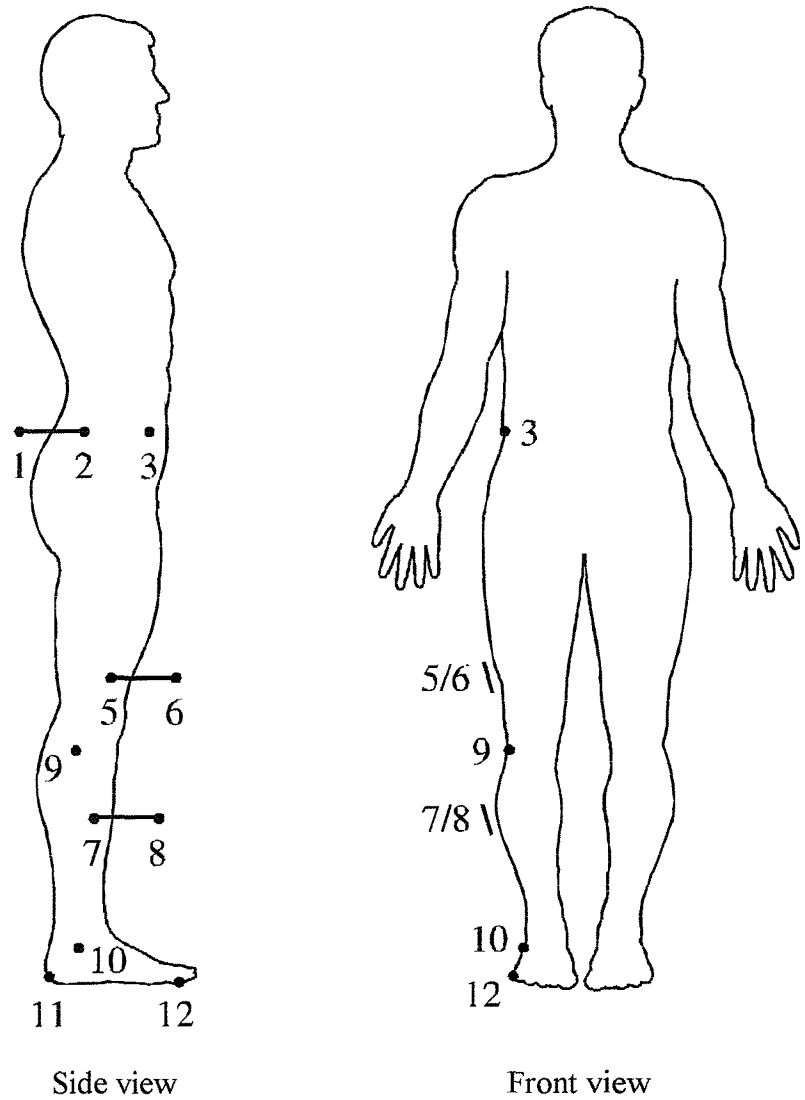
Number 8 'Post. Tib' was positioned near the posterior end of the shank wand.

Number 9 was positioned over the lateral epicondyle of the femur.

Number 10 was positioned over the lateral malleolus.

Number 11 was positioned at the lateral base of the heel.

Number 12 was positioned at the base of the lateral 5th metatarsal.



1	Sacral wand
2	Posterior superior illiac spine
3	Anterior superior illiac spine
5	Posterior end of the thigh wand
6	Anterior end of the thigh wand
7	Posterior end of the shank wand
8	Anterior end of the shank wand
9	Lateral epicondyle of the femur
10	Lateral malleolus
11	End of the heel
12	5th metatarsal base

Figure 4.8 Markers numbers and their locations

Each subject was dressed in shorts so that the markers on the tested leg would be visible. All subjects were asked to sit on a bench (height 50 cm and seat depth 30 cm without back or head support) with their arms by their side and knees flexed at 90 degrees during marker placement. The feet were fully supported on the floor or on the adjustable foot support. A parallel T-bar was placed between the knees in order to adjust the thigh wand (Figure 4.9). Alignment of the shank wand was made with subject standing erect in order to define the axis of rotation of the ankle joint (Figure 4.10).

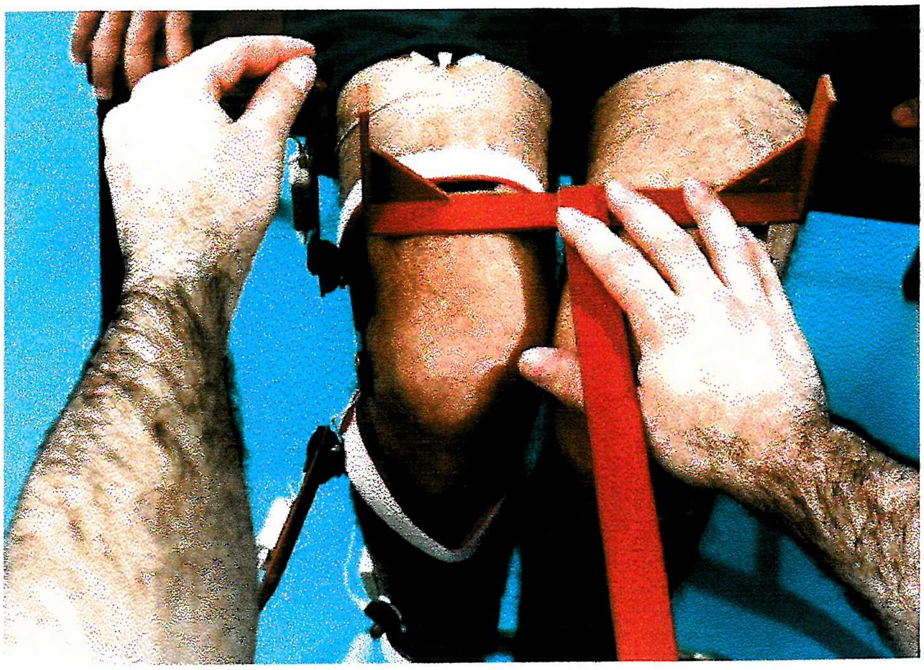


Figure 4.9 Thigh wand alignment using a T-bar from sitting position.

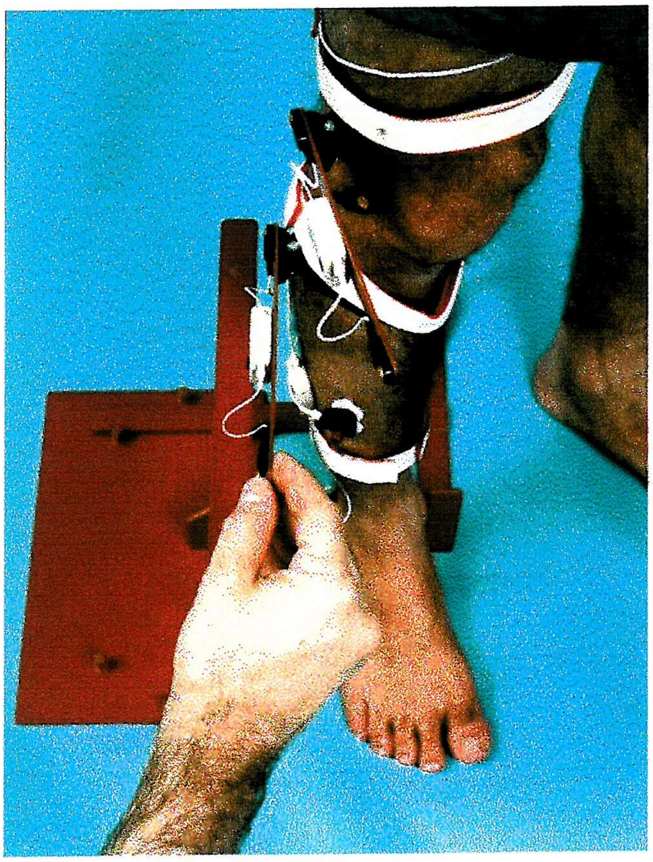


Figure 4.10 Shank wand alignment frame from standing position.

4.9.3. Recording of EMG during walking

The telemetry transmitter unit was attached with a velcro-strap to the centre of the subject's back and was secured around the subject's waist with velcro. A strap was fitted over the shoulder which was fully adjustable and also held in position by velcro.

4.9.4. Location of the surface EMG electrodes

(1) Skin preparation

Before placing the EMG electrodes, the skin was shaved, if necessary, over the site of application, and gently cleaned with methylated spirit in order to reduce the electrical impedance between the skin and the electrodes. It was then left to dry.

(2) Muscles tested :

A surface EMG was performed from the following four muscles simultaneously with CODA mpx30 and force plate recording:

(a) Rectus femoris;
(b) Medial hamstring;
(c) Tibialis anterior;
(d) Gastrocnemius;

(3) Positions of electrodes

The EMG electrodes were placed along the muscle fibre orientation. The EMG reference electrode was placed over the base of the neck. The placement of the EMG electrodes took place with the subjects sitting on a bench (without back or head support) for anterior leg muscles, and from a standing position for the posterior leg muscles.

The electrode position for the rectus femoris muscle was located on the point halfway between the ASIS and the superior border of the patella. The electrode for the medial hamstring muscle was placed on the lateral border of the muscle about 13 cm above the insertion on the fibula. For the tibialis anterior muscle, the electrode was located over the muscle mass in the middle third of the tibia. The electrode for the gastrocnemius muscle was placed over the visible muscle bulk between the lateral and medial head of the muscle. These are standard sites for EMG electrode placement used in routine clinical practice *(Goodgold 1974)*.

4.9.5 Final procedure for CODA mpx 30 recording

Figure 4.11 shows the lateral view of a subject wearing all of the equipment required for the recording of data. Before actual recording, each subject made three trial runs in order to become familiar with both the environment and the instruments attached to the lower limb. The patient was required to start at the same end of the walkway each time, returning to the start position for the next run.

The subjects were not aware that they were activating a force plate. The investigator made sure that the subject approached the force plate so that the leg being tested would step onto the force plate, whilst the other leg walked as normal on the floor.

To ensure that the subjects did not change their gait pattern, each walk was closely observed and all subjects were instructed to look straight ahead and to walk towards a large red cross positioned on the facing wall directly in front of them. Subjects were asked to walk at a comfortable walking speed on hearing the command 'off you go', and were constantly observed in order to make sure that they walked with their usual gait pattern, especially when stepping on the force plate.

It was necessary for the recording procedure that the subject was barefoot. The gait of each patient was recorded for one complete gait cycle; from initial contact to the point where the same foot made contact with the floor again. A successful recording occurred when the complete gait cycle was captured by the CODA mpx30 and when the subject's foot had struck the force plate accurately. All subjects were allowed to take a rest if needed.

No walking aids were used by the subjects during the recording. Between each run, the attachment of the markers and surface EMG electrodes was checked to make sure that each was still in the correct place.

To prevent fatigue and activity-induced knee pain, a maximum of three runs were recorded for each patient for each leg in the first session, and a maximum of five runs on the following sessions, 3,6 and 9 months post-operatively. An acceptable run met the following requirements:

(1) The leg to be tested struck the force plate.

(2) The other leg was outside the force plate.

(3) No visible alteration in the gait pattern were noted during walking.

During recording, the investigator made sure that patient's foot struck the force plate. Following the recording, the data was computed and kinematic, kinetic, and EMG data and graphs were available immediately.

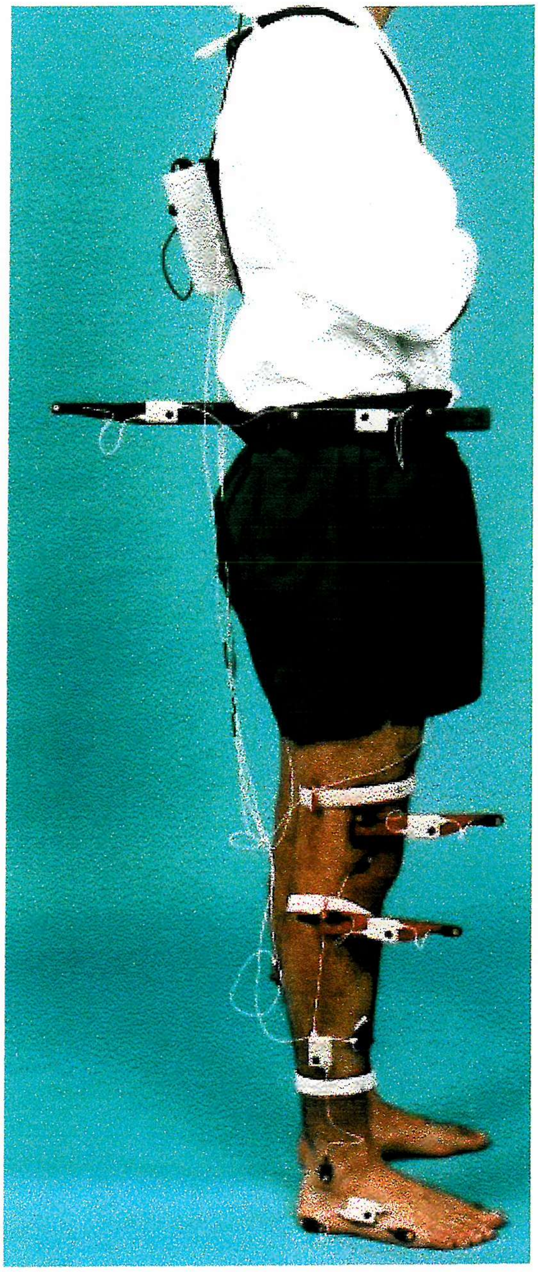


Figure 4.11 Lateral view showing the position of the equipment in the standing setup

4.10. Types of prostheses used :

1- Press-fit condylar PCL-sacrificing prosthesis *(Johnson & Johnson Raynham, MA 1989).*

The femoral condyles have an inward and upward sweep in the coronal and the matching insert condylar geometry insures maximum contact area and appropriate constraint (Figure 4.12). 'The sagittal plane has a tangent radius of curvature to simulate the centre of rotation of the normal knee'. The radii of the tibial and femoral components are almost equivalent. This reduces the contact between the two plastic surfaces and reduces wear and tear. The posterior stabilized prosthesis has a box feature between the femoral condyle. The posterior aspect of the box has a cam feature which interacts with the tibial spine. The cam provides true mechanical substitution for the function of the PCL and helps to aid in femoral rollback on the tibia during knee flexion and prevent posterior subluxation of the tibia *(Ranawat et al 1997, Vince et al 1989)*

2- Press-fit condylar PCL-retaining prosthesis

'The femoral condyles have a gentle radius of curvature in the coronal plane to allow for normal varus/valgus moments during gait. The sagittal plane view has a flat tibial component and a tangent radius of curvature to simulate the centre of rotation of the normal knee. This condylar design allows the component to articulate with both cruciate-retaining inserts of two different levels of constraint' *(Johnson & Johnson Orthopaedics Research & Development 1989).* The tibial titanium trays are shaped anatomically to maximally cap the available cut surface of the tibial plateau while minimizing overhang (Figures 4.13) (*Scott and Thornhill 1989).*

2- Mobile Bearing Knee (MBK)

The MBK prosthesis consists of three parts (Figure 4.14). 'There is a flat tibial plate with three protrusions on its upper surface. The first of these is a curved back stop on the medial side, the second is a T-stud, and the third is an anterior rail. The second component is a plastic surface that is located on top of the metal base plate. The lower surface of the plastic is flat and has a T-slot that engages the T-stud, allowing rotation on the metal base plate. Anterior displacement of the plastic is limited by the anterior rail. Posterior displacement is limited to 3mm by the medial back stop, but is unlimited on the lateral side. The external rotation is limited to 10 degrees, while the internal rotation is not restrained. The upper surface of the plastic has a bearing surface designed to be in conformity with the femoral component. The femoral component resembles a standard condylar replacement, but with a constant sagittal radius from distal to posterior. In addition, there are two notches on the lateral and medial side, which are anterior extensions of the posteriordistal bearing surface, to provide a complete contact area throughout the whole range of motion. The prosthesis is designed to retain the PCL' *(Menchetti and Walker 1997).*

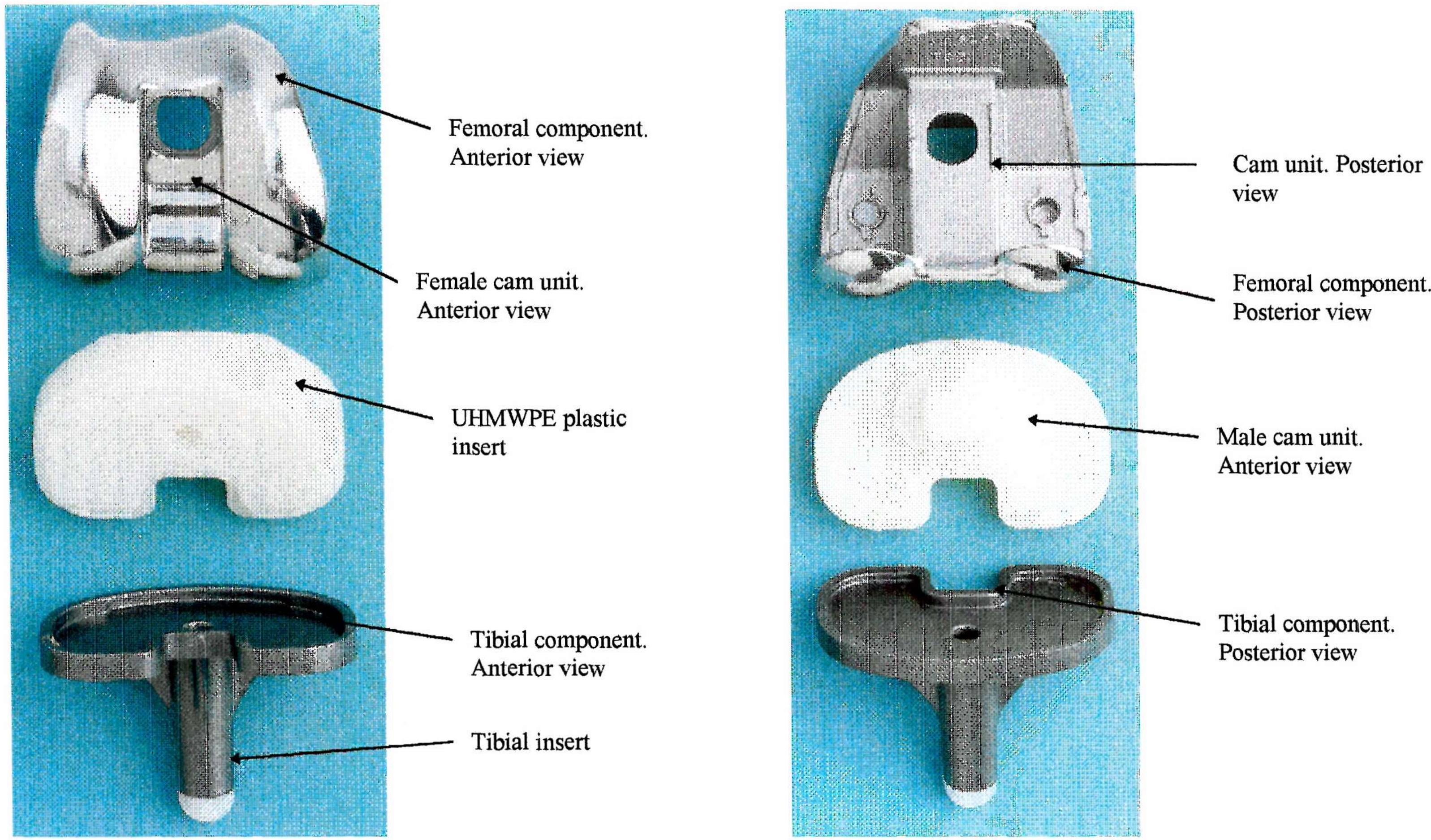


Figure 4.12 Press-fit condylar cruciate ligament sacrificing modular knee system.

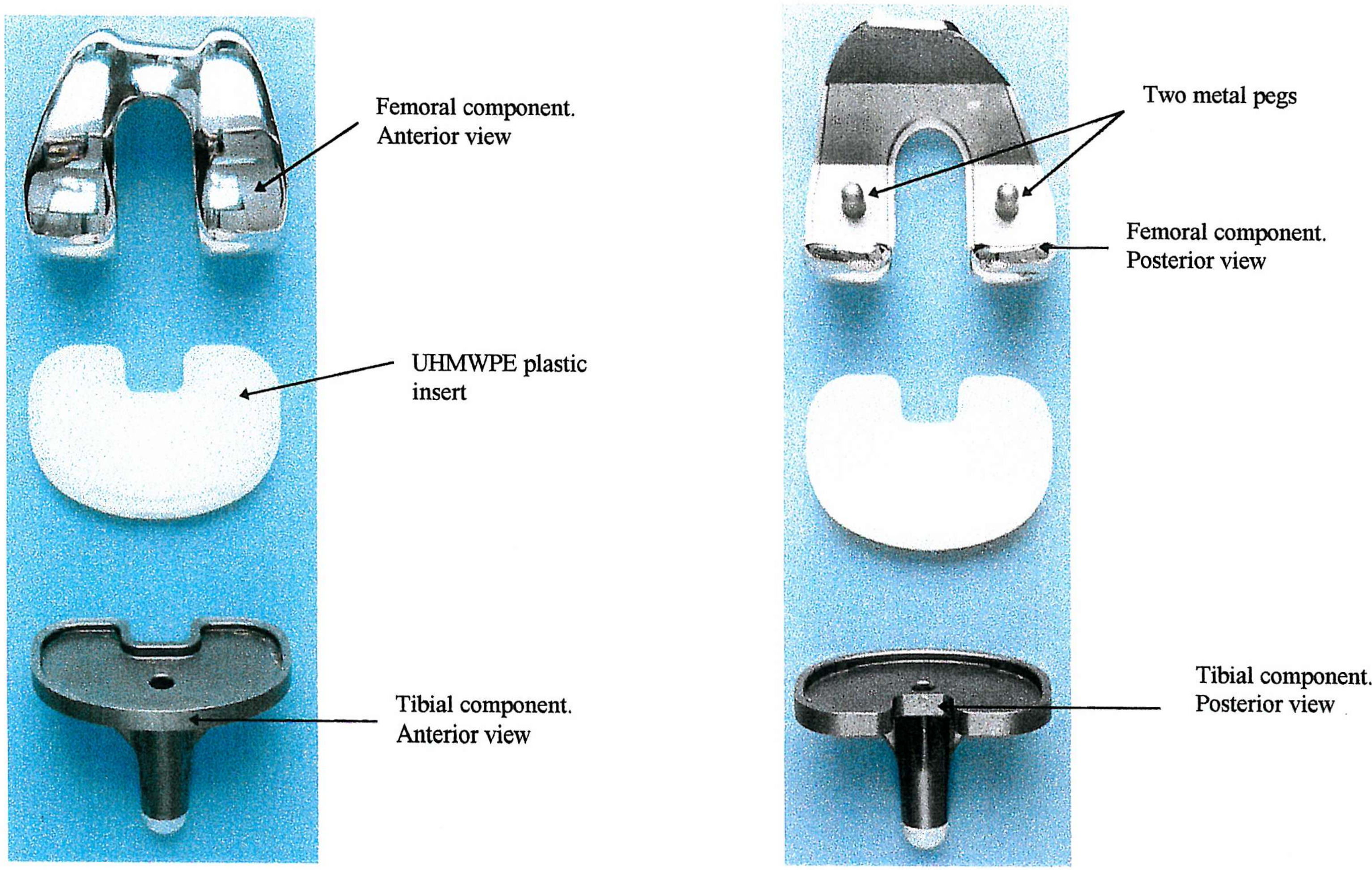


Figure 4.13 Press-fit condylar cruciate ligament retaining modular knee system.

Figure 4.14 Mobile Bearing Knee (From Zimmer, Inc, 1997).

4.11. Surgical technique for sacrificing and retaining PCL.

The information in this section was obtained from a recording of an interview I made with Mr. *Barrett (1997)* (Orthopaedic Consultant Surgeon) at Southampton General Hospital. The patient is anaesthetised and the leg being operated on is appropriately prepared and draped. A tourniquet is usually applied and remains in place for the duration of the surgery. An incision is made in the midline, and a medial parapatellar capsular approach is made. The patella is dislocated laterally to expose the knee joint and the knee is placed in full flexion. The ACL and the menisci are excised. Attention is then turned to the femur, and appropriate bone cuts are made using the jigging guides, with an intramedullary rod to ensure alignment. Alignment is normally 5 degrees for both the PFC and MBK. The femur is then cut so that it can be fitted accurately using the jigging guides. Attention is then turned to the tibia, and with external tibial guides, appropriate bone cuts are made. The alignment of the knee and any soft tissue releases are checked again to ensure that the knee is stable. A high density polyethylene spacer is inserted into the knee joint to achieve maximum flexion and full extension in order that the knee becomes fully straight and bends at about 110 degrees. The patella is then resurfaced using a plastic button. All the prostheses are implanted using acrylic cement. The wound is closed in layers using absorbable material (Vicryl) and the skin is then sewn together.

4.12 Decisions concerning the use of a given type of prosthesis

There are two schools of thought based on preservation of the posterior cruciate ligament or sacrificing the ligament and substituting a plastic cam. There is no hard evidence on which is best, and thus opinion is divided, based on experience and bias. However there is a more clear indication regarding types of prosthesis for specific conditions and I would say that for knees that are worn out in an established normal fashion, a condylar type unconstrained prosthesis is employed, be it with or without the posterior cruciate. For unstable knees with ligament laxity I think most surgeons would use a more constrained or linked prosthesis involving a higher and more stable posterior cam or indeed a physical hinge which would give better stability of the knee *(Mr. Barrett 1999 personal communication).*

4.13. Post-operative management

The routine procedures for the care of patients following TKR surgery were as follows. On day one the patient had the drain taken out. The knee was placed in a continuous passive-motion (CPM) under supervision of the physiotherapist. The settings on the machine were started with a 30 degree arc of flexion over a two minute cycle. The arc of flexion was increased within tolerable limits until the knee reached 90 degrees of flexion. The main physiotherapeutic goals were twofold: to achieve maximum range of movement and muscular control, and to achieve maximal functional independence.

On the second day the physiotherapist initiated isometric quadriceps-strengthening exercises, active range of motion exercises and gait training. Partial and full weight bearing attempts were started on the second or third post-operative day. Before discharge from the hospital the patients were given an appropriate walking aid, and instructed on how to sit and/or stand, how to turn, and how to use the stairs. All patients were discharged within two weeks post-operatively and the active range of motion was recorded for all patients at the last treatment session prior to discharge. All patients were instructed to continue training using a physiotherapy programme. All patients were given a follow-up appointment six weeks after the operation to see the surgeon.

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Patients with knee flexion of less than 70 degrees, poor quadriceps strength, or those experiencing stiffness were referred for outpatient physiotherapy.

4.14. Data analysis of the main study

To compare the data of the patients with the data of the healthy control subjects when walking at their preferred speed, non-parametric statistics were used, due to the fact that the baseline data did not have a normal distribution.

A comparison of the differences between groups was made using the Mann-Whitney U test and functional changes over time were analysed with the Kruskal Wallis test (See appendix 6). All data analysis was performed on PC using the Statistical Package for Social Sciences (SPSS Version 6.1.3, 1995).

Summary

After obtaining ethical committee approval, the researcher studied a group of OA patients, before and after total knee replacement. These patients were selected according to specific selection criteria. In addition, a group of healthy control subjects were recruited for comparison. All subjects were assessed using a motion analysis system that was integrated with telemetered electromyography and a force plate. In addition, pain, functional activities and energy expenditure were assessed using the Visual Analogue Scale, the Cincinnati Knee Rating Scale and a Physiological Cost Index. All the patients were assessed at baseline (just before surgery) and 3, 6 and 9 months after total knee replacement. All patients had total knee replacement using the midline para-patellar capsular approach. Post-operative management was carried out for all patients, involving inpatient treatment and a home programme. The data collected from the tests were analysed using non-parametric statistics on a PC, using the Statistical Package for Social Sciences (SPSS Version 6.1.3, 1995).

CHAPTER FIVE

RESULTS

5.1. Characteristics of the study sample

5.1.1. Control subjects

Fifty-five healthy potential control subjects were recruited for the study. However, 30 of these could not be matched to the age range of the patients and therefore the control data were derived from 25 subjects, 10 men and 15 women.

5.1.2. Patients

Patients were recruited for the study from outpatient clinics and the orthopaedic ward. Fifty eight patients with OA of the knee joint were studied. All OA patients fulfilled the study entry criteria and consented to take part in the study. However, four patients were later excluded from the final analysis because one left the Southampton area, one had severe oedema and pain of the operated knee, one was unable to attend the required assessment sessions, and another patient withdrew because she required further surgery following the initial TKR. The final study sample consisted of 54 patients.

The demographic characteristics of patients and control subjects are summarised in Table 5.1.

Table 5-1 The mean and range of age, height, and weight of the group of patients with osteoarthritis and control subjects.

	Patients		Control subjects	
Variable	Mean	Range	Mean	Range
Age (years)	71	50-84	69.00	56-82
Height (cm)	165	152-184	167.40	156-184
Weight (kg)	77	57-105	75.10	58-100

Thirty four patients had involvement of one knee joint, and the rest had bilateral joint involvement.

Scores from the Visual Analogue Scale, Cincinnati Knee Rating Scale and Physiological Cost Index data before and 3, 6 and 9 months following surgery were compared and the statistical significance was tested using the Kruskal-Wallis test.

Surgeon and prosthesis

The TKR operations were carried out by five consultant surgeons as shown in Table 5.2. Of the 54 patients, 41 were treated by surgeons 1 and 2 who had a special interest in knee surgery.

Table 5-2 Analysis of the types of prosthesis used.

	Number of prostheses		
Surgeon	PFC stabilised	PFC retaining	MBK retaining
1	15		16
2		10	
3		6	
4		4	
5		3	

5.2. Distribution of data

Plots were made of all the temporospatial, kinematic and kinetic gait parameters in all patients with OA of the knee at baseline and for the control group. The distribution of data was examined in order to decide on the appropriate methods of statistical analysis. Since the data were not normally distributed, the Mann-Whitney U test and Kruskal Wallis test were chosen for data analysis.

5.3. Comparison between control subjects and the study group at baseline.

5.3.1. Temporospatial gait parameters

Tables 5.3-5.5 summarise the temporospatial parameters and kinematic and kinetic data of the patients and control subjects.

Summary statistics of the walking speed, stride length, the timing of mid-stance, and the timing of mid-swing expressed as a percentage of the gait cycle of the patients and control groups are given in Table 5.3. The differences between control and patient groups for walking speed, stride length, timing of mid-stance and mid-swing in the gait cycle are also shown in Figures 5.1-5.2. The left-hand box represents the OA group and the right-hand box the control group.

As shown in Table 5.3 patients with OA of the knee walked more slowly and had a shorter stride length than the control subjects. Their median walking speed was 0.58 m/s compared to 1.18 m/s in the control group. The timings of mid-stance and mid-swing in the gait cycle in the OA patients were significantly delayed by comparison to the control group. The differences between the two groups were statistically significant for all of these variables (P< 0.0005).

Table 5-3 Median, IQR and range of temporospatial gait parameters for control and patient groups.

Gait parameters	Control group (n=25)	Patient group (n=54)	P-value
	Median (IQR) (Range)	Median (IQR) (Range)	
Walking speed (m/s)	1.18 (1.06-1.26) (0.93-1.50)	0.58 (0.40-0.71) (0.10-1.18)	0.0005
Stride length (m)	1.31 (1.18-1.38) (1.07-1.61)	0.76 (0.63-0.93) (0.26-1.37)	0.0005
Timing of mid-stance (% cycle)	30.22 (29.98-30.67) (28.93-31.43)	32.15 (31.14-34.56) (28.90-46.39)	0.0005
Timing of mid-swing (% cycle)	80.22 (79.98-80.67) (78.93-81.43)	82.15 (81.14-84.56) (78.90-96.39)	0.0005

IQR, Inter-quartile range.

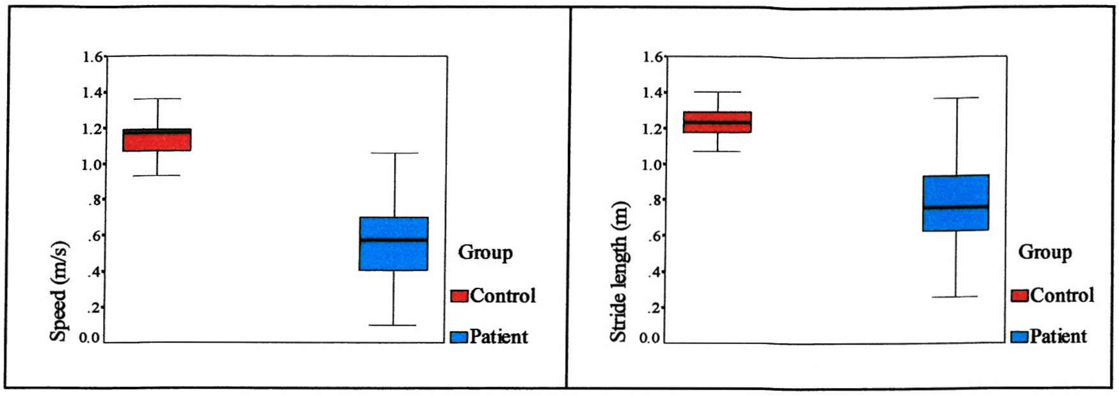

Figure 5-1 Boxplot showing the walking speed and stride length of the control and patient groups. The median, IQR and range are shown.

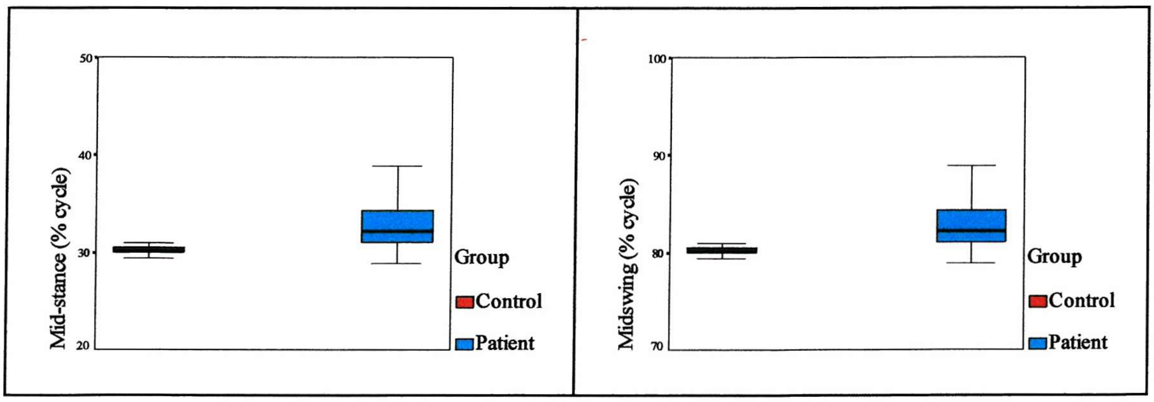

Figure 5-2 Boxplot showing the timing of mid-stance and mid-swing in the gait cycle of the control and patient groups. The median, IQR and range are shown.

5.3.2. Kinematic Parameters

The kinematic gait parameters of the control and patient groups were assessed by recording the hip, knee and ankle joint ranges of motion in the sagittal plane.

Analysis of the kinematic data showed that patients had a smaller range of motion at the hip, knee and ankle joints than the controls. The differences reached statistical significance for all of these parameters ($P< 0.05$) (Table 5.4). These results are also graphically represented in the boxplot diagrams below (Figures 5.3 and 5.4).

Table 5-4 Median, IQR and range of kinematic gait parameters for control and patient groups.

Gait parameters (degrees)	Control group (n=25)	Patient group (n=54)	P-value
	Median (IQR) (Range)	Median (IQR) (Range)	
Max hip extension (ST)	36.80 (33.80-43.35) (29.50-53.50)	26.55 (21.00-33.65) (11.40-42.70)	0.0005
Max knee flexion (LR)	13.20 (10.40-18.50) (4.10-22.10)	4.70 (2.40-6.40) (0.00-9.20)	0.0005
Max knee flexion (SW)	60.90 (58.60-66.50) (42.90-76.10)	34.60 (27.88-45.15) (4.60-54.00)	0.0005
Max ankle dorsiflexion (SW)	27.20 (26.30-28.35) (25.30-30.10)	22.10 (20.43-23.73) (14.70-26.00)	0.0005

IQR, Inter-quartile range.

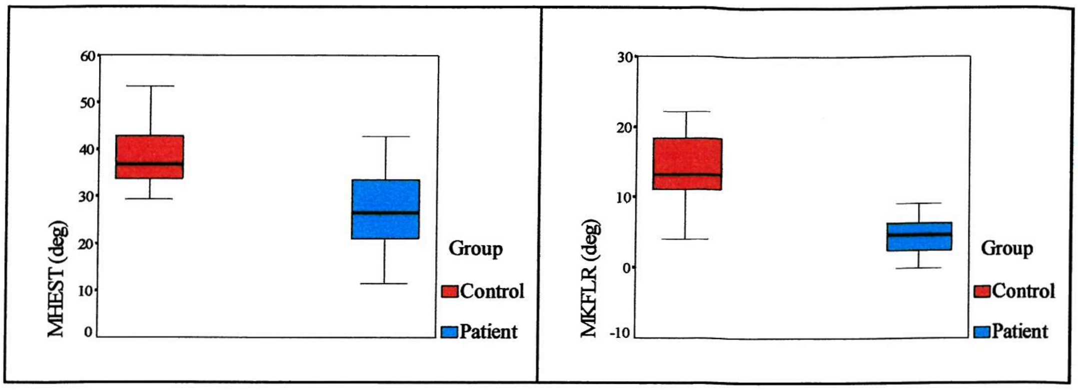

Figure 5-3 Boxplot showing the comparison of the maximum hip extension in stance (MHEST) and maximum knee flexion in loading response (MKFLR) of the control and patient groups. The median, IQR and range are shown.

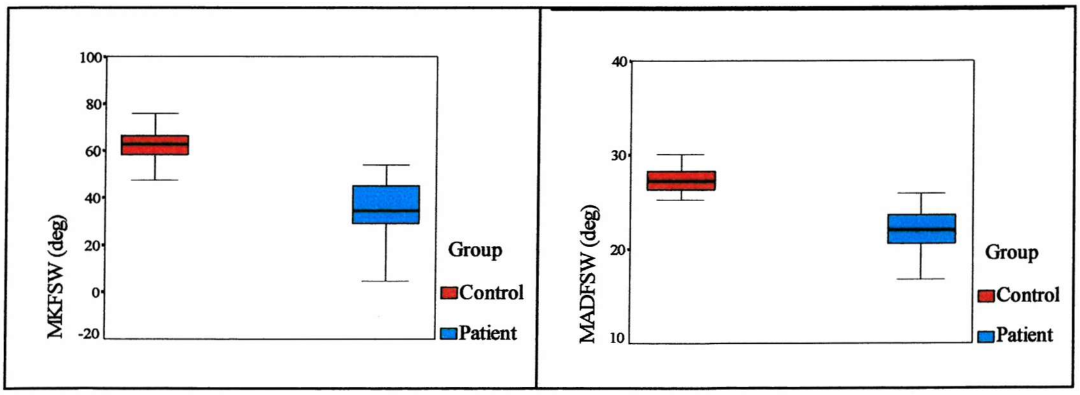

Figure 5-4 Boxplot showing the comparison of the maximum knee flexion in swing (MKFSW), maximum ankle dorsiflexion in swing (MADFSW) of the control and patient groups. The median, IQR and range are shown.

5.3.3. Kinetic Parameters

Summary statistics of the moments and powers generated at the knee and ankle of the study and control groups are given in Table 5.5 and graphically illustrated in the boxplot diagrams (Figures 5.5-5.6).

The moments generated at the knee were higher in the patients than in the control subjects. The median was 0.32 Nm/kg. This contrasts with a value of 0.12 Nm/kg in control subjects. These differences were statistically significant ($P < 0.0005$). Patients showed increased moments and reduced powers at the knee joint compared to the control subjects. The moments and powers generated at the ankle were significantly less in OA patients than in the control subjects.

Table 5-5 Median, IQR and range of kinetic gait parameters for control and patient groups.

Gait parameters	Control group (n=25)	Patient group (n=54)	P-value
	Median (IQR) (Range)	Median (IQR) (Range)	
Max knee moment in mid-stance (Nm/kg)	0.12 (-0.35-0.21) (-0.13-0.33	0.32 (0.20-0.48) (-0.02-1.09)	0.0005
Max knee power in mid-stance (W/kg)	0.12 (0.03-0.29) (-0.20-0.46)	-0.0 (-0.05-0.05) (-0.38-0.27)	0.0005
Max ankle moment in pre-swing (Nm/kg)	0.80 (0.66-1.02) (0.28-2.10)	0.47 (0.35-0.70) (0.10-1.36)	0.0005
Max ankle power in pre-swing (W/kg)	3.64 (3.14-4.71) (2.33-7.02)	1.25 (0.45-2.41) (-0.02-4.39)	0.002

IQR, Inter-quartile range.

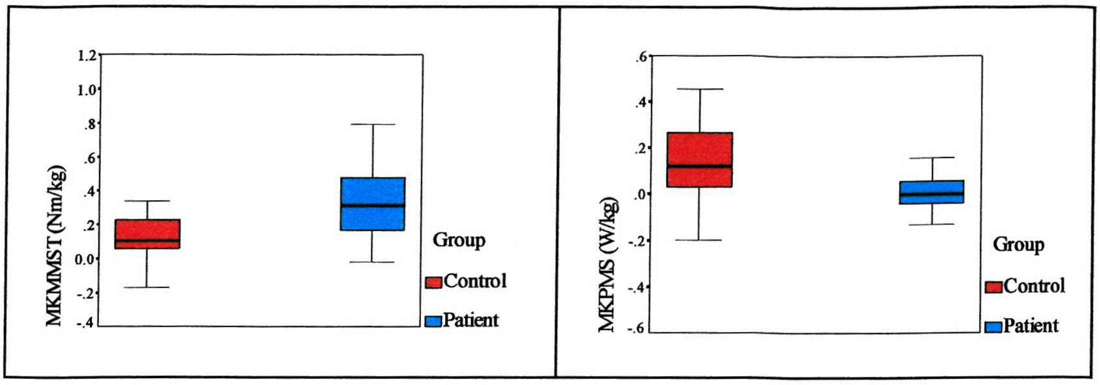

Figure 5-5 Boxplot showing the differences in maximum knee moment in mid-stance (MKMMST) and maximum knee power in mid-stance (MKPMST) of the control and patient groups. The median, IQR and range are shown.

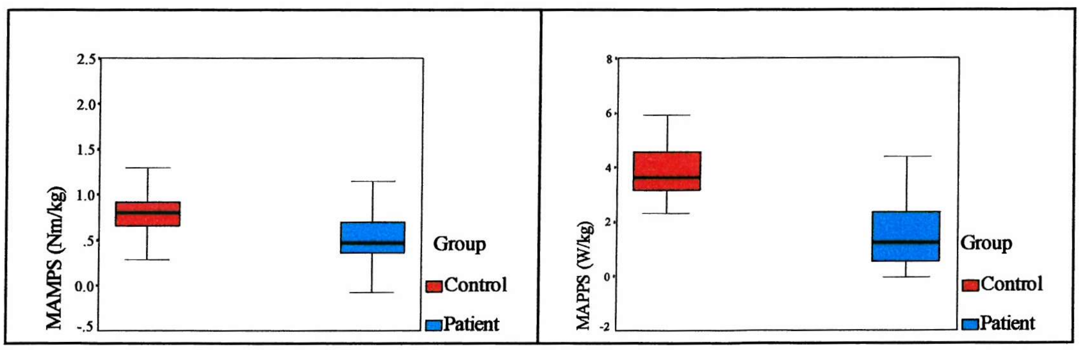

Figure 5-6 Boxplot showing the differences in maximum ankle moment in pre-swing (MAMPSW) and maximum ankle power in pre-swing (MAPPSW) of the control and patient groups. The median, IQR and range are shown.

5.4. Example of the kinetic and kinematic data obtained from one subject in the control group

This section will analyse the hip, knee and ankle data for one healthy control subject, whose measurements were a typical example of those found in the control group.

Figures 5.7- 5.9 are typical records of the hip, knee and ankle joint range of motion (ROM) in the sagittal plane of one complete gait cycle obtained from a control subject during walking at a preferred speed. The figures include the simultaneous hip, knee, and ankle joint moments and powers measured in Nm/kg and W/kg, respectively.

5.4.1. Kinematics of the hip

It will be noticed from Figure 5.7 that the maximum hip flexion is about 38 degrees at initial contact. The hip then extends to a maximum of 0.0 degrees of extension to lie in line with the trunk at terminal stance. The hip is again flexed in mid-swing to the same degree as at initial contact. Minor variations between individuals were noted, which may be due to the alignment of the pelvic frame.

5.4.2. Kinematics and kinetics of the knee

Figure 5.8 illustrates the knee kinematics and kinetics of a healthy subject. It will be noticed from Figure 5.8 that the knee flexion in stance is about 23 degrees during loading response phase. From its initial flexion wave the knee reaches its closest position to extension (10 degrees of flexion) during the pre-swing. The knee moves further into flexion again in the swing phase, reaching a maximum of about 62 degrees of flexion in mid-swing.

It can also be seen that the initial knee flexion moment reverses into an extension moment (reaching a peak of 1.1 Nm/kg at about 10 % of the gait cycle).

5.4.3. Kinematics and kinetics of the ankle.

Figure 5.9 illustrates the ankle kinematics and kinetics of a healthy subject. The graph shows the ROM at the ankle in stance and swing phase. A minimal plantarflexion starts at initial contact and is followed by dorsiflexion of about 16 degrees. The ankle is again dorsiflexed in mid-swing in order to allow foot clearance from the ground.

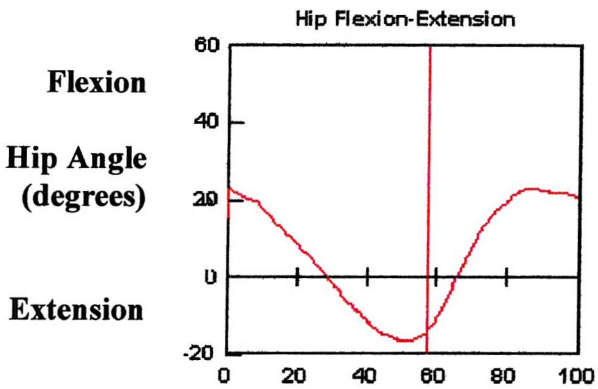

Figure 5-7 Computer generated plots of the hip kinematic parameters of a healthy subject (CF1R4). Joint range of motion is shown in degrees.

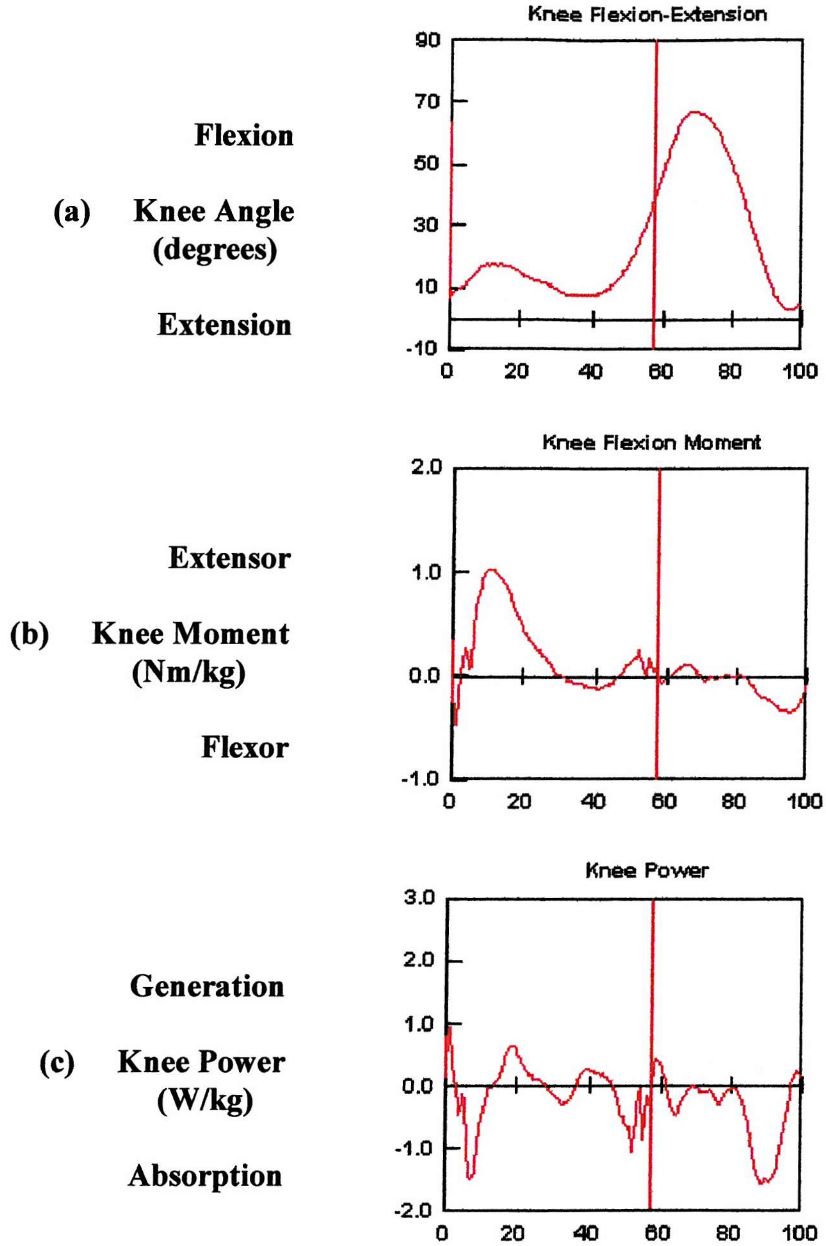

Figure 5-8 Computer generated plots of the knee kinematic and kinetic parameters of a healthy subject (CF1R4). Joint range of motion is shown in degrees (a), joint moment in Nm/kg (b), and joint power in W/kg (c).

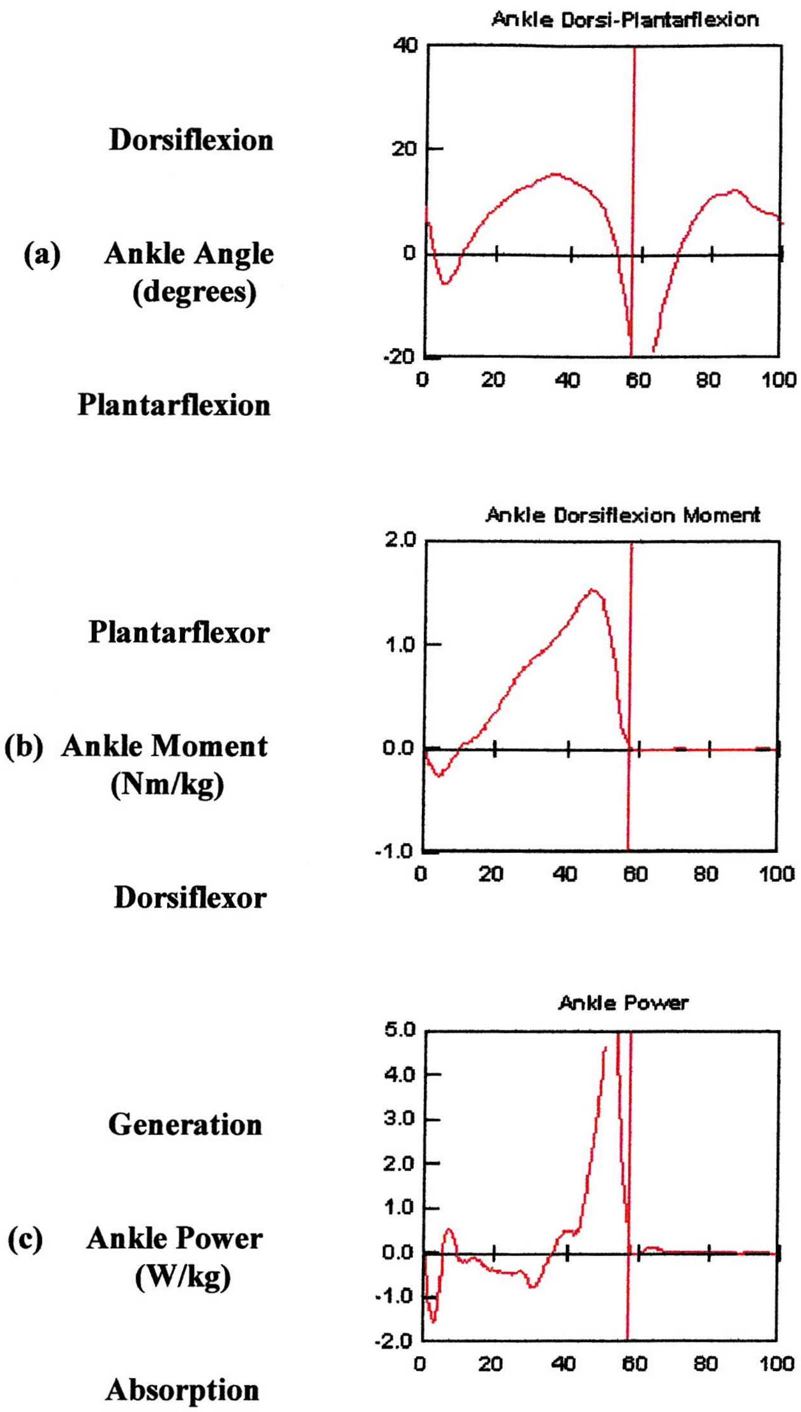

Figure 5.9 Computer generated plots of the ankle kinematic and kinetic parameters of a healthy subject (CF1R4). Joint range of motion is shown in degrees (a), joint moment in Nm/kg (b), and joint power in W/kg (c).

5.5. Electromyography during gait

5.5.1. Pattern of lower limb muscle activity during walking in normal subjects

EMG activity was recorded from four muscles by surface contact electrode pre-amplifiers. The four muscles studied were the rectus femoris, medial hamstring, tibialis anterior and gastrocnemius.

The patterns of EMG of the four muscles during one complete gait cycle for two healthy subjects are illustrated in figures 5.10 and 5.11. The vertical scale represents the amplitude of EMG in μv for each of the four muscles recorded. The horizontal scale represents the percentage of the gait cycles.

Activity of the rectus femoris muscle at free walking speed:

The EMG pattern of the rectus femoris activity consisted of an initial burst at the initial contact of the stance phase. Maximal activation was noted during initial contact and loading response. In mid-stance the rectus femoris did not show activity.

Activity of the medial hamstring muscle:

The pattern of EMG activity of the medial hamstring muscles at self selected walking speed showed a peak of activity twice during the gait cycle. The first burst of activity occurred at the beginning of the stance phase (on initial contact), and the second at the end of the swing phase.

Activity of the tibialis anterior muscle:

There were two peaks of muscle activity which occurred at initial contact and the end of stance.

Activity of the gastrocnemius muscle:

The activity of the gastrocnemius reaches its maximum after mid-stance.

5.5.2. Pattern of lower limb muscle activity during walking in the OA patients

Activity of the rectus femoris muscle at free walking speed:

The rectus femoris muscle of the OA patients with knee arthritis demonstrated an abnormal pattern of activation during the gait cycle. This was characterised by prolonged activity throughout the stance phase. Prolonged activity was noted, beginning before the end of the swing phase in 12 patients during the transition period from stance to swing phase when the knee began to flex. In 10 OA patients the rectus femoris was almost silent throughout the gait cycle. In 4 OA patients the activity of the rectus femoris was similar to that of healthy controls. The remaining OA patients demonstrated slight to moderate activity during the stance phase.

Activity of the medial hamstring muscle:

In 6 OA patients the medial hamstring muscle showed an almost normal muscle pattern. The medial hamstring muscle was silent throughout the stance phase in 9 OA patients and in the swing phase in 16 OA patients. The remaining OA patients with knee arthritis demonstrated slight to moderate activity between mid-stance and toe-off.

Activity of the tibialis anterior muscle:

The tibialis anterior muscle of the OA patients throughout the gait cycle showed abnormal activity. There was continuous anterior tibial muscle activity in 47 out of 54 OA patients throughout the gait cycle. There was no patient in whom the anterior tibial muscle was silent during the gait cycle.

The remaining OA patients demonstrated a variety of activity during stance phase and early swing phase.

Activity of the gastrocnemius muscle:

Activation of the gastrocnemius was delayed or absent in the pre-swing phase in 41 out of 54 OA patients. Seven OA patients showed prolonged activation during the stance phase. In three OA patients, there was prolonged activation during the stance and swing phases. In four OA patients, the calf muscles showed an almost normal pattern of activity. The gastrocnemius muscle was almost silent throughout the gait cycle in 11 OA patients. Examples of the EMG records are shown in figures 5.12-5.15.

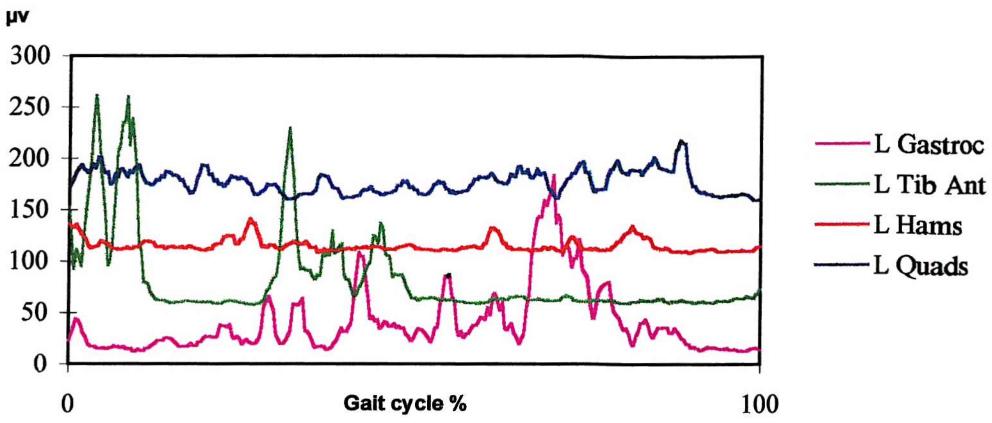

Figure 5.10 Recorded EMG of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of a healthy subject (ST3L3). The record represents one gait cycle. Each line is offset by 50μv for clarity.

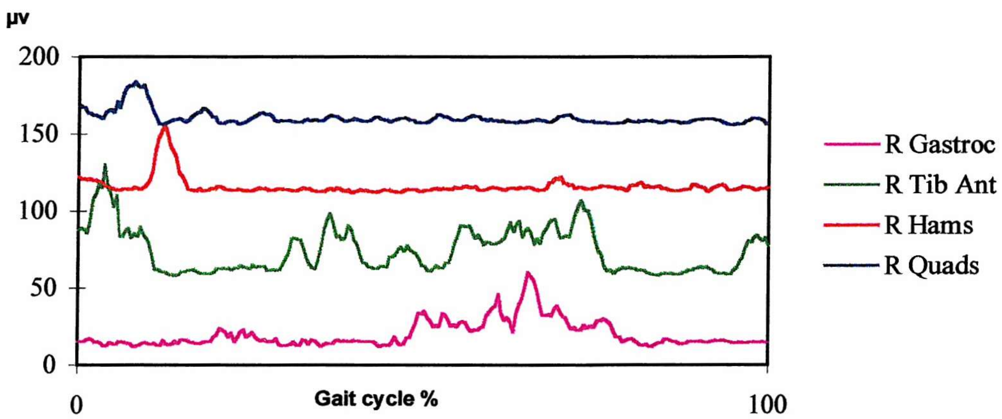

Figure 5.11 Recorded EMG of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of a healthy subject (GS1R5). The record represents one gait cycle. Each line is offset by 50μv for clarity.

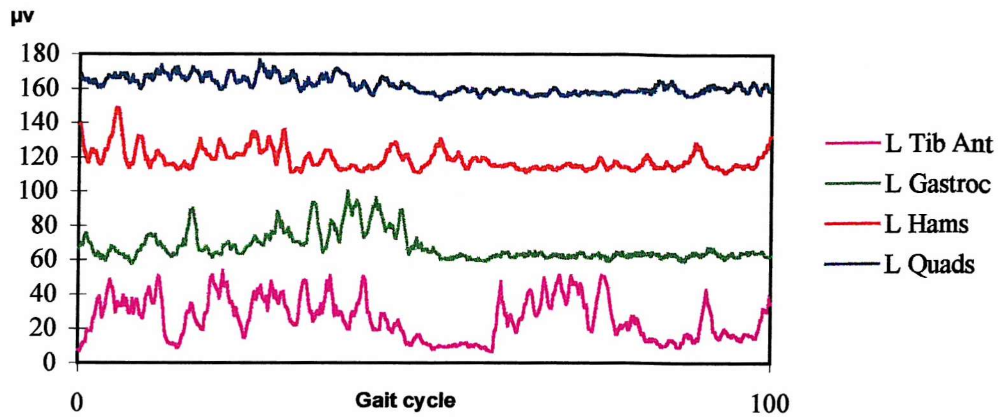

Figure 5.12 Recorded EMG of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of OA subject (LE1L5). The record represents one gait cycle. Each line is offset by 50µv for clarity.

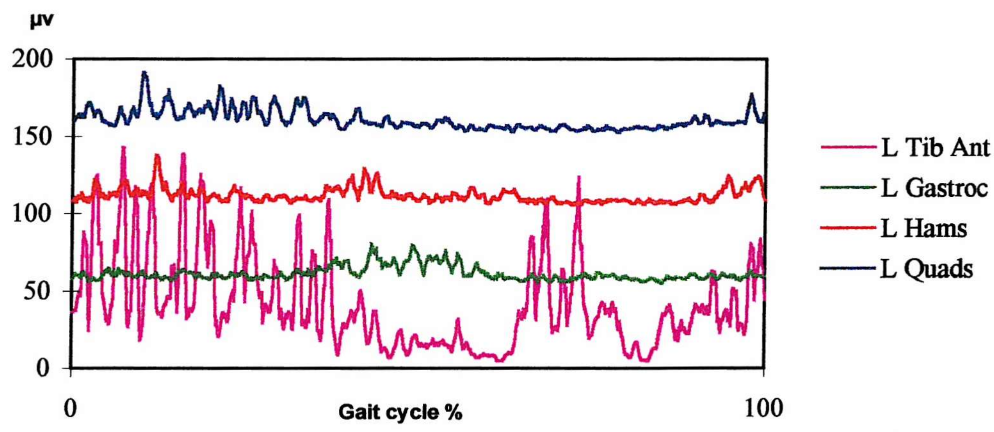

Figure 5.13 Recorded EMG of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of OA subject (MA1L5). The record represents one gait cycle. Each line is offset by 50µv for clarity.

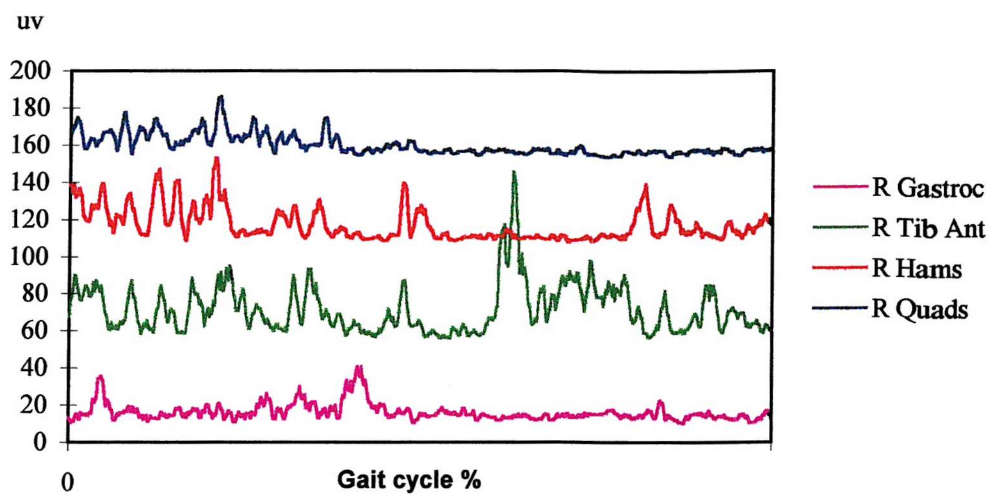

Figure 5.14 Recorded EMG of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of OA subject (GM1R1). The record represents one gait cycle. Each line is offset by 50μv for clarity.

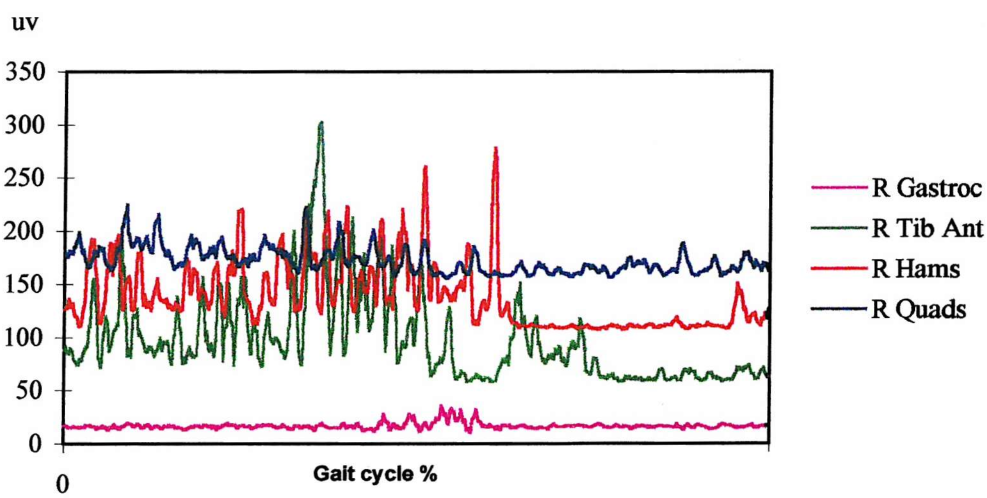

Figure 5.15 Recorded EMG of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of OA subject (OM2R2). The record represents one gait cycle. Each line is offset by 50μv for clarity.

Summary:

Gait analysis was carried out of 54 OA patients awaiting TKR. Twenty-five healthy subjects were selected for gait evaluation. The age, height and body weight of these subjects were comparable to the patient population used. The results obtained in this section have demonstrated the following:

(a) The temporospatial data for the 54 patients with knee osteoarthritis showed that these patients walked with the following characteristics:

(1) Significantly shorter stride length and consequently a very slow walking speed;

(2) Significantly delayed mid-stance and mid-swing;

(3) A longer gait cycle than in control subjects.

(b) The pattern of kinematic and kinetic data showed that the patients walked with the following characteristics:

(1) Significantly smaller hip extension in the stance phase, knee flexion at mid-stance and mid-swing, and ankle dorsiflexion in the swing phase;

(2) Significantly smaller knee moments and ankle moments and powers. The powers generated at the knee were less than those of the control subjects. However, the differences were not statistically significant.

(c) The activity of the rectus femoris, medial hamstring, tibialis anterior and gastrocnemius muscles were monitored simultaneously using telemetered electromyography (EMG) and showed abnormal muscle activation in the OA patients.

(d) The recording from the rectus femoris, medial hamstring, tibialis anterior and gastrocnemius muscles of the healthy subjects demonstrated some interindividual variability in the pattern of EMG activity during the stance phase. However, the EMG pattern for a given individual remained the same on repeated testing.

5.6. A pre-operative comparison of gait in the three groups of patients who received the three different types of prosthesis used in the study

Fifteen patients received press-fit condylar prostheses (PCL-sacrificing) (group A), 23 received press-fit condylar prostheses (PCL-retaining) (group B) and 16 had mobile bearing knee prostheses (PCL-retaining) (group C).

5.6.1. Temporospatial gait parameters

Tables 5.6-5.8 summarise the pre-operative temporospatial, kinematic and kinetic parameters of the OA groups with press-fit cruciate ligament sacrificing condylar prostheses (group A), press-fit cruciate ligament retaining condylar prostheses (group B) and mobile bearing knee prostheses (group C). Summary statistics of the walking speed, stride length, the timing of mid-stance, and mid-swing of the three pre-operative groups are given in Table 5.6. The Kruskal Wallis test showed that there were no statistically significant differences in the temporospatial gait parameters between the three OA groups pre-operatively. The boxplots illustrate the results for the group of control subjects together with the three OA groups for comparison (Figures 5.16-5.17).

Table 5-6 Comparison of the temporospatial gait parameters in patients with OA of the knee before TKR. Patients were grouped according to the type of prostheses implanted.

Gait parameters	Group A[1] (n=15)	Group B[2] (n=23)	Group C[3] (n=16)	P-value
	Median (Range)	Median (Range)	Median (Range)	
Walking speed (m/s)	0.60 (0.50-0.81)	0.53 (0.38-0.66)	0.59 (0.39-0.69)	0.53
Stride length (m)	0.83 (0.68-0.96)	0.73 (0.63-0.93)	0.76 (0.62-1.00)	0.62
Mid-stance (% cycle)	32.00 (30.52-34.52)	33.36 (31.50-35.09)	33.02 (31.09-35.00)	0.49
Mid-swing (% cycle)	82.00 (80.52-84.52)	83.71 (81.50-85.09)	82.02 (81.09-85.31)	0.49

[1]Underwent PFC PCL-sacrificing, [2]Underwent PFC PCL-retaining, [3]Underwent MBK PCL-retaining.

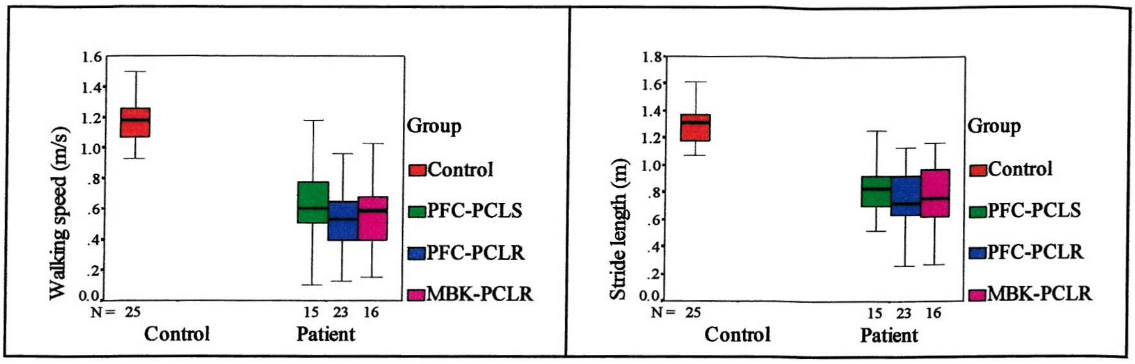

Figure 5-16 Boxplots showing the pre-operative differences in walking speed and stride length between the three groups of TKR patients and the control group. Pressfit condylar posterior cruciate ligament sacrificing (PFC-PCLS) (group A), pressfit condylar posterior cruciate ligament retaining (PFC-PCLR) (group B) and mobile bearing knee posterior cruciate ligament retaining (MBK-PCLR) (group C). The median, IQR and range are shown.

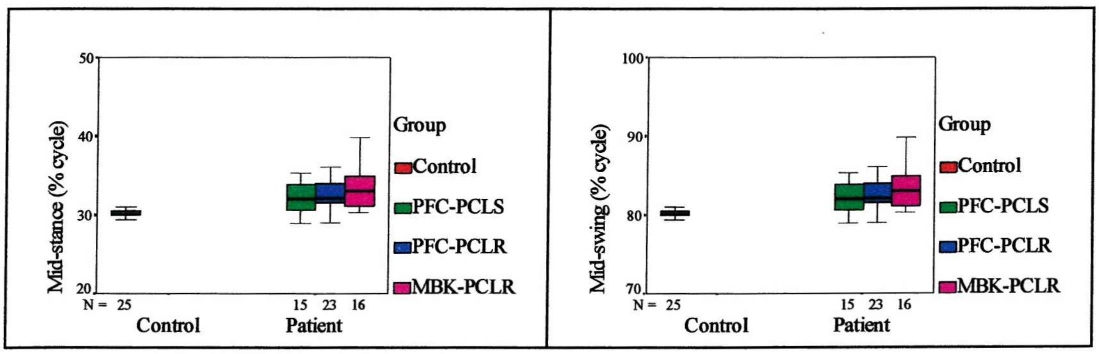

Figure 5-17 Boxplots showing the pre-operative differences in the timing of mid-stance and mid-swing in the gait cycle between the three groups of TKR patients and the control group. Pressfit condylar posterior cruciate ligament sacrificing (PFC-PCLS) (group A), pressfit condylar posterior cruciate ligament retaining (PFC-PCLR) (group B) and mobile bearing knee posterior cruciate ligament retaining (MBK-PCLR) (group C). The median, IQR and range are shown.

5.6.2. Kinematic Parameters

Analysis of the maximum hip extension in stance, maximum knee flexion in the loading response phase, maximum knee flexion in swing and maximum ankle dorsiflexion in swing data using the Kruskal Wallis test shows that there were no statistically significant differences between the three pre-operative patient groups (A, B and C) in the kinematic gait parameters (Table 5.7). The plots show the results for the control subjects and the three OA groups (Figures 5.18-5.19).

Table 5-7 Comparison of the kinematic gait parameters in patients with OA of the knee before TKR. Patients were grouped according to the type of prostheses implanted.

Gait parameters (degrees)	Group A[1] (n=15) Median (Range)	Group B[2] (n=23) Median (Range)	Group C[3] (n=16) Median (Range)	P-value
Max hip extension (ST)	27.00 (20.30-34.00)	26.50 (21.00-33.50)	25.70 (20.63-33.80)	0.97
Max knee flexion (LR)	4.19 (2.10-6.40)	4.80 (2.50-6.40)	4.10 (1.80-6.43)	0.81
Max knee flexion (SW)	33.60 (26.30-40.10)	34.90 (29.00-45.30)	34.25 (26.77-44.35)	0.83
Max ankle dorsiflexion (SW)	21.50 (19.80-23.40)	22.40 (20.60-23.80)	22.60 (19.15-23.76)	0.80

[1]Underwent PFC PCL-sacrificing, [2]Underwent PFC PFC-retaining, [3]Underwent MBK PCL-retaining.

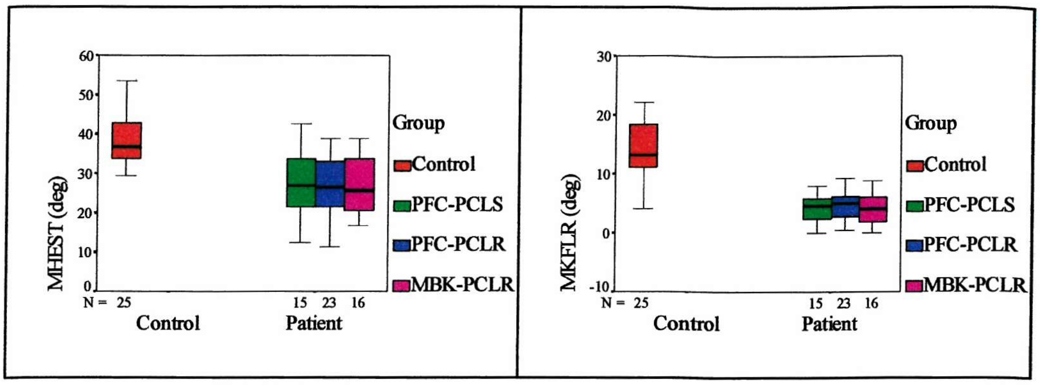

Figure 5-18 Boxplots showing the pre-operative differences in the maximum hip extension in stance (MHEST) and maximum knee flexion in loading response (MKFLR) between the three groups of TKR patient and the control group. Pressfit condylar posterior cruciate ligament sacrificing (PFC-PCLS) (group A), pressfit condylar posterior cruciate ligament retaining (PFC-PCLR) (group B) and mobile bearing knee posterior cruciate ligament retaining (MBK-PCLR) (group C). The median, IQR and range are shown.

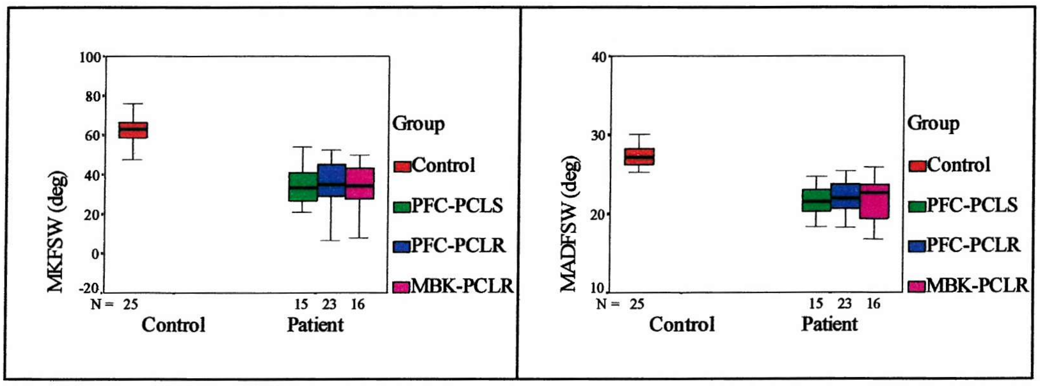

Figure 5-19 Boxplots showing the pre-operative differences in the maximum knee flexion in swing (MKFSW) and maximum ankle dorsiflexion in swing (MADFSW) between the three groups of TKR patients and the control group. Pressfit condylar posterior cruciate ligament sacrificing (PFC-PCLS) (group A), pressfit condylar posterior cruciate ligament retaining (PFC-PCLR) (group B) and mobile bearing knee posterior cruciate ligament retaining (MBK-PCLR) (group C). The median, IQR and range are shown.

5.6.3. Kinetic Parameters

No significant differences in the moments and powers generated at the knee and ankle between the three pre-operative OA groups (A, B and C) could be detected using the Kruskal Wallis test (Table 5.8). The plots compare the results for the control subjects and the three patient groups (Figures 5.20-5.21).

Table 5-8 Comparison of the kinetic gait parameters in patients with OA of the knee before TKR, using the Kruskal Wallis test. Patients were grouped according to the type of prostheses implanted.

Gait parameters	Group A[1] (n=15)	Group B[2] (n=23)	Group C[3] (n=16)	P-value
	Median (Range)	Median (Range)	Median (Range)	
Max Knee moment in mid-stance (Nm/kg)	0.32 (0.14-0.51)	0.31 (0.20-0.48)	0.37 (0.21-0.51)	0.53
Max knee power in pre-swing (W/kg)	0.01 (0.04-0.06)	0.03 (0.06-0.04)	0.05 (0.12-0.03)	0.12
Max Ankle moment in mid-stance (Nm/kg)	0.44 (0.43-0.65)	0.50 (0.36-0.70)	0.58 (0.38-0.77)	0.42
Max ankle power in pre-swing (W/kg)	1.06 (0.43-2.83)	1.25 (0.55-2.07)	1.28 (0.54-2.09)	0.90

[1]Underwent PFC PCL-sacrificing, [2]Underwent PFC PCL-retaining, [3]Underwent MBK PCL-retaining.

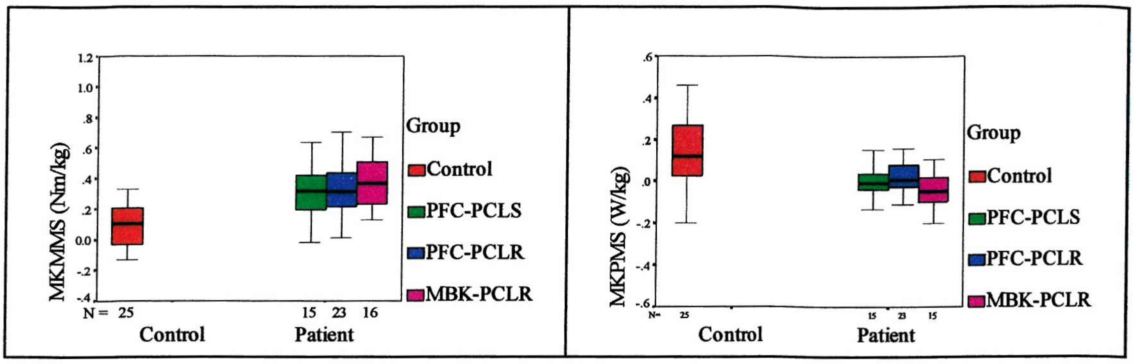

Figure 5-20 Boxplots showing the pre-operative differences in maximum knee moment in mid-stance (MKMMST) and maximum knee power in mid-stance (MKPMST) between the three groups of TKR patients and the control group. Pressfit condylar posterior cruciate ligament sacrificing (PFC-PCLS) (group A), pressfit condylar posterior cruciate ligament retaining (PFC-PCLR) (group B) and mobile bearing knee posterior cruciate ligament retaining (MBK-PCLR) (group C). The median, IQR and range are shown.

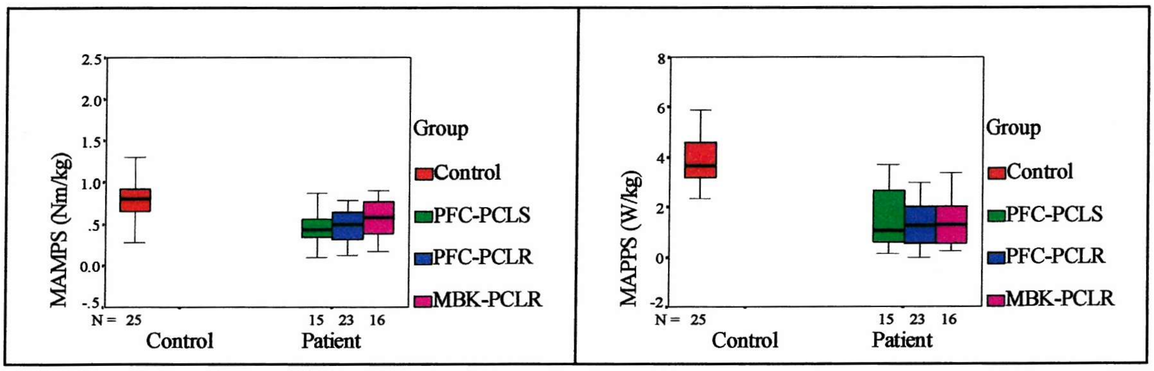

Figure 5-21 Boxplots showing the pre-operative differences in maximum ankle moment in pre-swing (MAMPSW) and maximum ankle power in pre-swing (MAPPSW) between the three groups of TKR patients and the control group. Pressfit condylar posterior cruciate ligament sacrificing (PFC-PCLS) (group A), pressfit condylar posterior cruciate ligament retaining (PFC-PCLR) (group B) and mobile bearing knee posterior cruciate ligament retaining (MBK-PCLR) (group C). The median, IQR and range are shown.

5.6.4. A pre-operative comparison of the Cincinnati Knee Rating Scale (CKRS) and the Visual Analogue Scale (VAS) scores.

A pre-operative comparison of CKRS and VAS scores was carried out. No significant differences in the CKRS or VAS were found between the three groups of subjects pre-operatively using the Kruskal Wallis tests (Table 5.9). These results are graphically illustrated in the boxplot diagrams below (Figure 5.22).

Table 5-9 Cincinnati knee rating scale (CKRS) and visual analogue scale (VAS) scores at baseline in patients with OA of the knee before TKR.

	Group A[1] (n=15)		Group B[2] (n=23)		Group C[3] (n=16)		
Parameters	Median	Range	Median	Range	Median	Range	P[1]-value
CKRS	19.00	7.50-29.00	24.00	14.00-41.00	20.50	16.00-27.50	0.41
VAS (cm)	7.05	5.48-7.88	7.10	5.90-7.80	7.30	4.95-8.20	0.83

[1]Underwent PFC PCL-sacrificing, [2]Underwent PFC PCL-retaining, [3]Underwent MBK PCL-retaining .

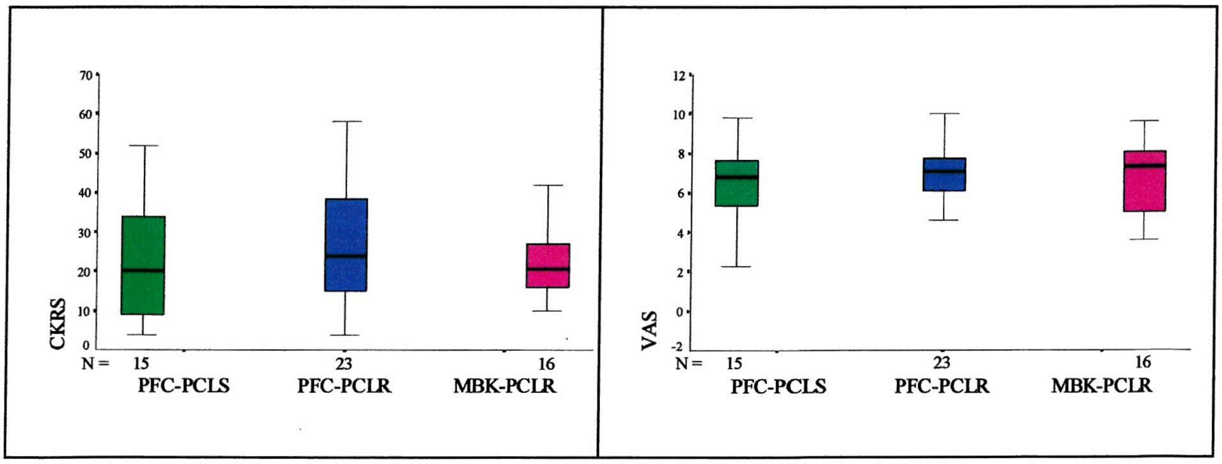

Figure 5-22 Boxplots comparing the pre-operative scores of the Cincinnati knee rating scale (CKRS) and visual analogue scale (VAS) between the three groups of TKR patients. PFC-PCLS (group A), PFC-PCLR (group B) and MBK-PCLR (group C). The median, IQR and range are shown.

Summary:

The temporospatial, kinematic and kinetic parameters of the pre-operative OA groups with press-fit cruciate ligament sacrificing condylar prostheses (group A), press-fit cruciate ligament retaining condylar prostheses (group B) and mobile bearing knee retaining prostheses (group C) were compared. Also, a pre-operative functional comparison using the Cincinnati knee rating scale (CKRS) and visual analogue scale (VAS) scores was carried out.

The temporospatial, kinematic and kinetic data of the three groups of OA patients showed that they did not significantly differ from each other. Generally, all three groups of patients walked with shorter stride length and consequently slow walking speed. The timing of mid-stance and mid swing were delayed. The pattern of kinematic and kinetic data showed that patients walked with smaller hip extension in the stance phase, knee flexion at mid stance and mid swing, and ankle dorsiflexion in the swing phase. They also had smaller knee moments and ankle moments and powers. The powers generated at the knee were less than those of the control subjects. A pre-operative comparison of CKRS and VAS scores among the three groups showed no significant differences on either measure.

5.7. Effects of TKR on gait

5.7.1. Press-Fit Condylar Posterior Cruciate Ligament Sacrificing (PFC-PCLS) knee replacement (group A).

This section will describe the changes over time in this group of 15 patients.

5.7.1.1. Temporospatial gait parameters

Pre- and post-operative changes for each temporospatial gait variable are shown in Table 5.10. Statistically significant increases in walking speed and stride length from baseline were observed at all follow-up assessments. There was a progressive improvement towards normal values in these parameters with time, but at 9 months they still had not reached the level of the control subjects. There were no significant changes in the timing of mid-stance and mid-swing in the gait cycle during the observation period at 3 and 6 months. However, significant changes occurred at 9 months.

The plots show the results for the control subjects and the baseline and 3, 6, and 9 months follow-up after PFC-PCLS TKR (Figures 5.23-5.24).

Table 5-10 Median, IQR and range of temporospatial parameters while walking at preferred speed. Figures are shown for the control group and for four test periods before and after PFC-PCLS implantation.

Gait parameters	Control (n = 25)	PCLS			
		Baseline (n = 15)	3 months (n= 14)	6 months (n = 11)	9 months (n = 13)
	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Walking speed (m/s)	1.18 (1.06-1.26) (0.93-1.50)	0.60 (0.50-0.81) (0.10-1.18)	0.78 (0.59-0.97) (0.08-1.21)*	0.94 (0.81-1.07) (0.06-1.18)*	0.99 (0.87-1.09) (0.17-1.23)*
Stride length (m)	1.31 (1.18-1.38) (1.07-1.61)	0.83 (0.68-(0.96) (0.31-1.37)	0.97 (0.77-1.12) (0.19-1.37)*	1.06 (0.77-1.15) (0.14-1.39)*	1.12 (0.98-1.18) (0.39-1.36)*
Mid-stance (% cycle)	30.22 (29.98-30.67) (28.93-31.43)	32.00 (30.52-34.52) (28.90-42.60)	30.76 (30.38-31.41) (29.45-44.26)	30.39 (29.79-30.76 (27.69-45.82)	30.50 (29.65-30.89) (29.41-37.59)*
Mid-swing (% cycle)	80.22 79.98-80.67 (78.93-81.43)	82.00 (80.52-84.52) (78.90-92.60)	80.76 (80.38-81.41) (79.45-94.26)	80.39 (79.79-80.76) (77.69-95.82)	80.50 (79.65-80.89) (79.41-87.59)*

**Denotes statisticaly significant differences (P < 0.05) (Mann-Whitney U test) from the baseline(group A).*

IQR, Inter-quartile range.

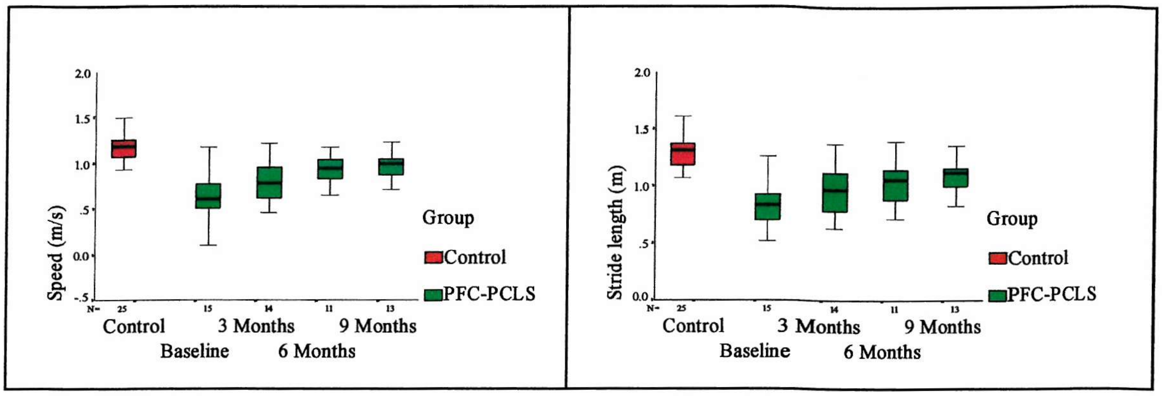

Figure 5-23 Boxplots comparing the differences in walking speed and stride length between baseline and 3, 6, and 9 months follow-up after PFC-PCLS TKR. The red boxplot shows the control subject for comparison. The median, IQR and range are shown.

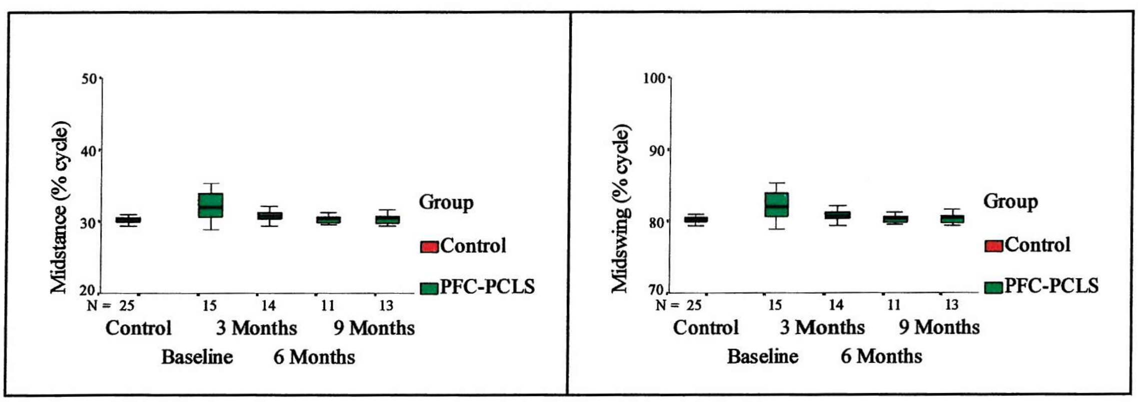

Figure 5-24 Boxplots comparing the differences in the timing of mid-stance and mid-swing in the gait cycle between baseline and 3, 6, and 9 months follow-up after PFC-PCLS TKR. The red boxplot shows the control subjects for comparison. The median, IQR and range are shown.

5.7.1.2.Kinematic parameters during gait

The median (range) of hip, knee and ankle range of motion during level walking at the baseline and at 3, 6 and 9 months after surgery are shown in Table 5.11. The maximum hip extension during stance (MHEST) improved significantly at 3, 6 and 9 months after surgery. There was a statistically significant difference in MHEST between the baseline values and the data at 6 and 9 months after TKR.

Some gradual improvement occurred in maximum knee flexion during the loading response phase from 6 to 9 months follow-up period. At 9 months the maximum knee flexion during the loading response phase in the study group (6.90°) was significantly lower ($P < 0.05$) than the maximum knee flexion during the loading response phase in the control group (13.22°).

The maximum knee flexion during the mid-swing phase in the study group at 9 months (50.30°) was also significantly lower ($P < 0.05$) than the maximum knee flexion during the mid-swing in the control group (60.90°). The maximum ankle dorsiflexion during swing increased at 3, 6 and 9 months after surgery. No statistically significant differences were observed between controls and patients in the maximum ankle dorsiflexion during swing phase 9 months after surgery.

These results are plotted for the control subjects and patients at baseline and 3, 6, and 9 months follow-up after PFC-PCLS TKR (Figures 5.25-5.26).

Table 5-11 Median, IQR and range of kinematic parameters while walking at preferred speed. Figures are shown for the control group and for four test periods before and after PFC-PCLS implantation.

Parameter (degrees)	Control (n = 25)	Group A			
		Baseline (n = 15)	3 months (n= 14)	6 months (n = 11)	9 months (n = 13)
	Median (IQR) Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Max hip extension (ST)	36.80 (33.80-43.35) (29.50-53.50)	27.00 (20.30-34.00) (12.50-42.70)	28.95 (23.60-35.40) (9.80-40.60)	35.90 (27.70-38.40) (11.80-44.60)*	32.80 (25.60-38.85) (9.90-45.60)*
Max knee flexion (LR)	13.20 (10.40-18.50) (4.10-22.10)	4.60 (2.10-6.40) (0.00-8.00)	3.05 (2.15-8.18) (0.10-10.50)	5.50 (2.90-7.20 (2.10-11.20)	6.90 (4.65-10.20) (3.70-15.90)*
Max knee flexion (SW)	60.90 (58.60-66.50) (42.90-76.10)	33.60 (26.30-42.10) (4.60-54.00)	43.35 (32.85-48.85) (21.10-56.40)*	47.70 (36.70-52.60) (11.00-60.40)*	50.30 (45.15-55.40) (26.50-59.70)*
Max ankle dorsiflexion (SW)	27.20 (26.30-28.35) (25.30-30.10)	21.50 (19.80-23.40) (14.70-24.70)	23.85 (21.65-24.23) (21.30-25.60)*	25.60 (22.30.26.20) (17.90-26.90)*	25.90 (24.75-27.20) (23.40-28.90)*

**Denotes statisticaly significant differences (P < 0.05) (Mann-Whitney U test) from the baseline(group A).*

IQR, Inter-quartile range.

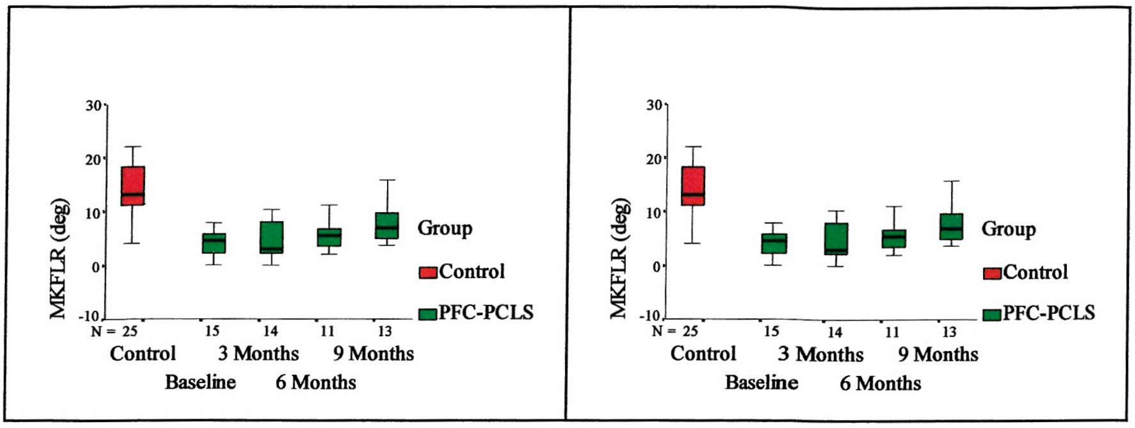

Figure 5-25 Boxplots comparing the differences in maximum hip extension in stance (MHEST) and maximum knee flexion in loading response (MKFLR) between baseline and 3, 6, and 9 months follow-up after PFC-PCLS TKR. The red boxplot shows the control subjects for comparison. The median, IQR and range are shown.

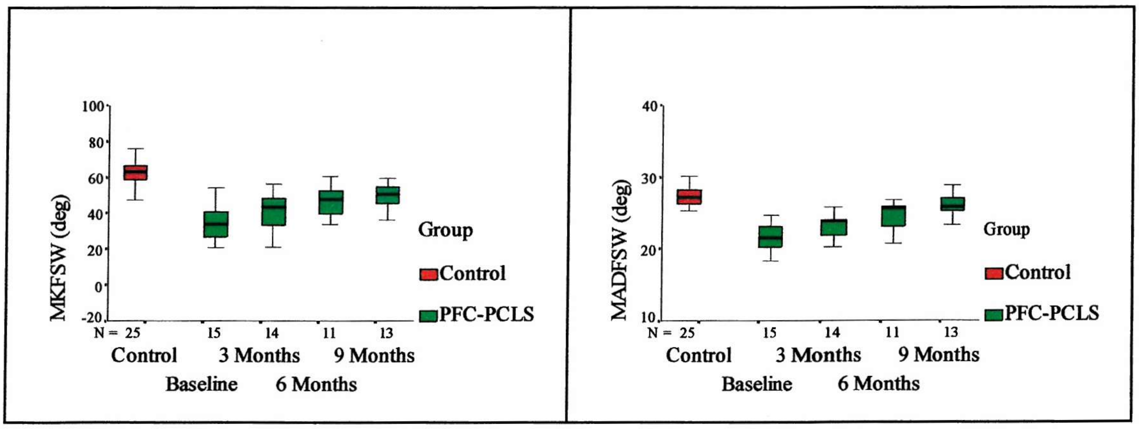

Figure 5-26 Boxplots comparing maximum knee flexion in swing (MKFSW), maximum ankle dorsiflexion in swing (MADFSW) between baseline and 3, 6, and 9 months after PFC-PCLS TKR. The red boxplot shows the control subjects for comparison. The median, IQR and range are shown.

5.7.1.3. Kinetic parameters during gait

Table 5.12 lists The median (range) values obtained for the moments and powers generated at the knee and ankle joints in control subjects and patients in the pre-operative and post-operative periods over 9 months of follow-up. As can be seen from the results, the moments generated at the knee decreased progressively in the patient group but did not reach the level of the control group.

Knee powers was less after surgery than at baseline. They improved to a positive value after 9 months but did not reach the level of the control groups and the difference was statistically significant.

Table 5.12 also illustrates the results of the maximum ankle moment and power in pre-swing. The maximum ankle dorsiflexion moment in pre-swing improved at 6 months and 9 months, reaching normal values. The patients generated more muscle plantarflexion power at this joint during pre-swing on each of the three test periods after surgery. However, the change in plantarflexion power after surgery was significantly less than that of the control subjects ($P < 0.05$).

The plots show the results for the control subjects and patients at baseline and 3, 6, and 9 months post-operatively for comparison (Figures 5.27-5.28).

Table 5-12 Median, IQR and range of the kinetic parameters while walking at preferred speed. Figures are shown for the control group and for four test periods before and after PFC-PCLS implantation.

Gait parameters	Control (n = 25)	Group A			
		Baseline (n = 15)	3 months (n= 14)	6 months (n = 11)	9 months (n = 13)
	Median (IQR) Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Max knee moment in mid-stance (Nm/kg)	0.12 (-0.35-0.21) (-0.13-0.33	0.32 (0.14-0.51) (0.02-0.64)	0.20 (0.13-0.32) (0.04-0.67)*	0.13 (0.09-0.26) (-0.15-0.89)*	0.19 (0.09-0.31) (0.09-0.36)*
Max knee power in mid-stance (W/kg)	0.12 0.03-0.29 (-0.20-0.46)	0.01 (-0.04-0.06)(- 0.38-0.22)	0.03 (-0.07-0.05) (-0.53-0.23)*	0.03 (-0.07-0.03) (-0.31-0.20)	0.04 (-0.03-0.17) (-0.36-0.44)*
Max ankle moment in pre-swing (Nm/kg)	0.80 (0.66-1.02) (0.28-2.10)	0.44 (0.33-0.65) (0.10-0.87)	0.64 (0.32-0.77) (0.17-1.08)*	0.69 (0.59-0.87) (0.45-1.24)*	0.77 (0.66-0.91) (0.49-1.21)*
Max ankle power in pre-swing (W/kg)	3.64 (3.14-4.71) (2.33-7.02)	1.06 (0.43-2.83) (0.14-3.70)	20.06 (1.11-3.20) (0.06-4.96)*	2.40 (1.66-3.45) (0.06-40.00)*	2.34 (1.35-3.70) (0.18-4.78)*

**Denotes statisticaly significant differences (P < 0.05) (Mann-Whitney U test) from the baseline(group A).*

IQR, Inter-quartile range.

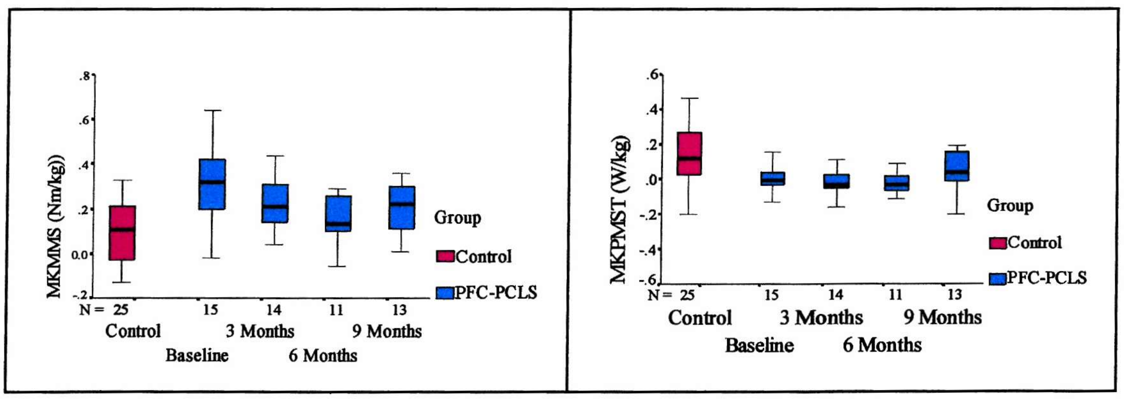

Figure 5-27 Boxplots comparing the differences in maximum knee moment in mid-stance (MKMMST) and maximum knee power in mid-stance (MKPMST) at baseline and 3, 6, and 9 months follow-up after PFC-PCLS TKR. The red boxplot shows the control subjects for comparison. The median, IQR and range are shown.

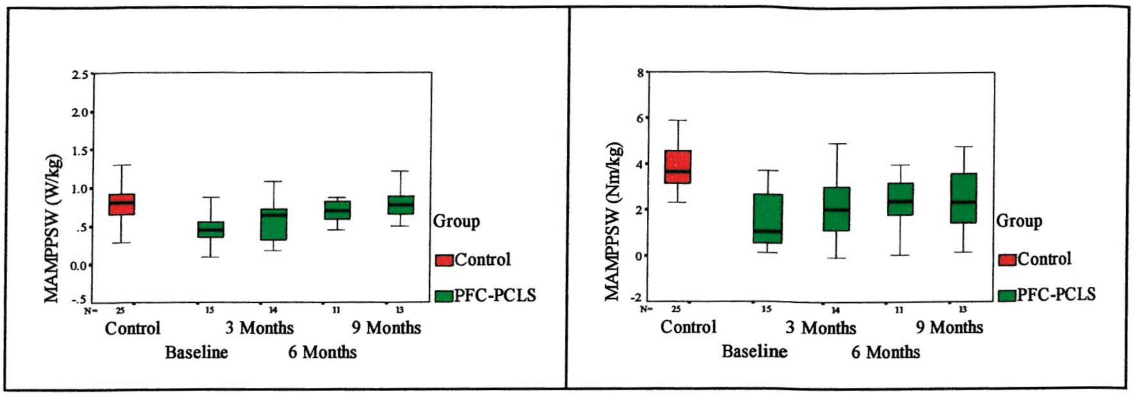

Figure 5-28 Boxplots comparing the differences in maximum ankle moment in pre-swing (MAMPSW) and maximum ankle power in pre-swing (MAPPSW) at baseline and 3, 6, and 9 months follow-up after PFC-PCLS TKR. The red boxplot shows the control subjects for comparison. ***The median, IQR and range are shown.***

5.7.1.4. Example of the kinetic and kinematic data obtained from one patient in this group at baseline and 3, 6, and 9 months after PFC-PCLS TKR.

Figures 5.29- 5.31 are examples recorded in the sagittal plane for joint ROM in degrees for the hip, knee and ankle, with the simultaneous hip, knee, and ankle joint moments and powers measured in Nm/kg and W/kg respectively. The measurements were taken at baseline and 3, 6, and 9 months after operation.

Kinematics of the hip

It will be noticed from Figure 5.29 that the maximum hip extension in the stance phase changed slightly from baseline to 9 months after operation.

Kinematics and kinetics of the knee

Figure 5.30 shows the knee kinematics and kinetics of the same patient at baseline and 3, 6, and 9 months after surgery. The analysis of kinematics demonstrates an abnormal knee flexion during the loading response and in the mid-stance phase after surgery. An improvement in the knee flexion during swing can be observed from the baseline data to the 9 months after operation.

It can be seen from the figure that the knee moment decreased at 9 months. There were no changes in the knee power over the follow up period.

Kinematics and kinetics of the ankle.

Figure 5.31 illustrates the ankle kinematics and kinetics of the same patient. The analysis of ankle kinematics demonstrates excessive dorsiflexion at mid-stance and limited plantarflexion at the push-off 3, 6 and 9 months after surgery. There was hardly any change in maximum ankle dorsiflexion in the swing phase from baseline to 9 months after operation. The ankle moment and power increased progressively during the follow up period.

Healthy subject	PFC-PCLS TKR			
	Baseline	3 months post-operative	6 months post-operative	9 months post-operative

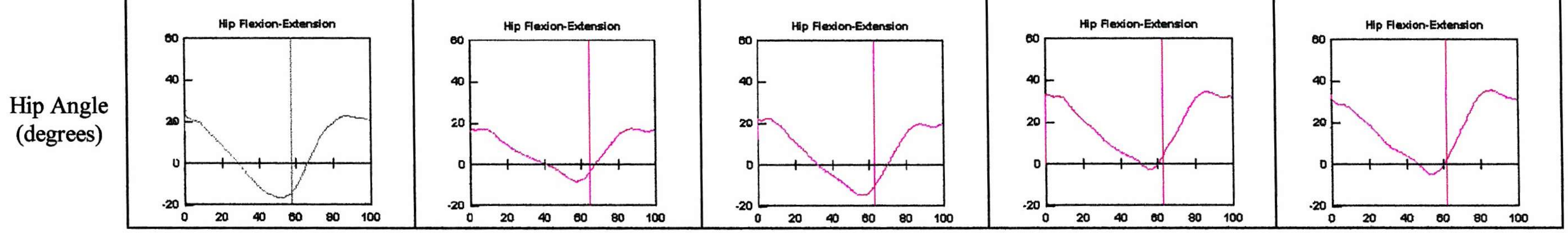

Figure 5.29 Sagittal plane kinematics of the hip joint of a patient (FP) at baseline and 3, 6 and 9 months after PFC-PCLS TKR and a healthy subject. Joint range of motion is shown in degrees.

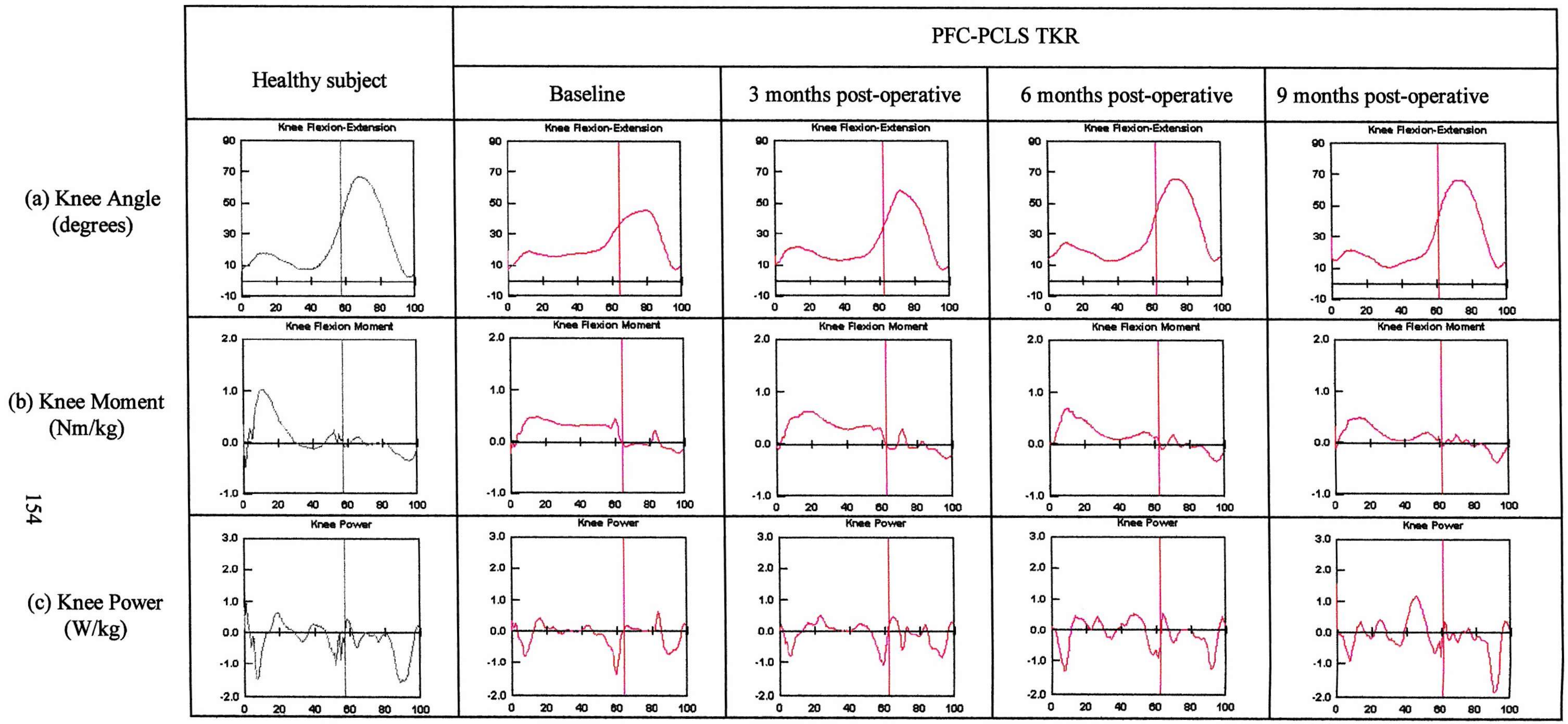

Figure 5.30 Sagittal plane kinematics of the knee joint (a) and kinetics data (b and c) of a patient (FP) at baseline, and 3, 6 and 9 months after PFC-PCLS TKR and a healthy subject. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in W/kg.

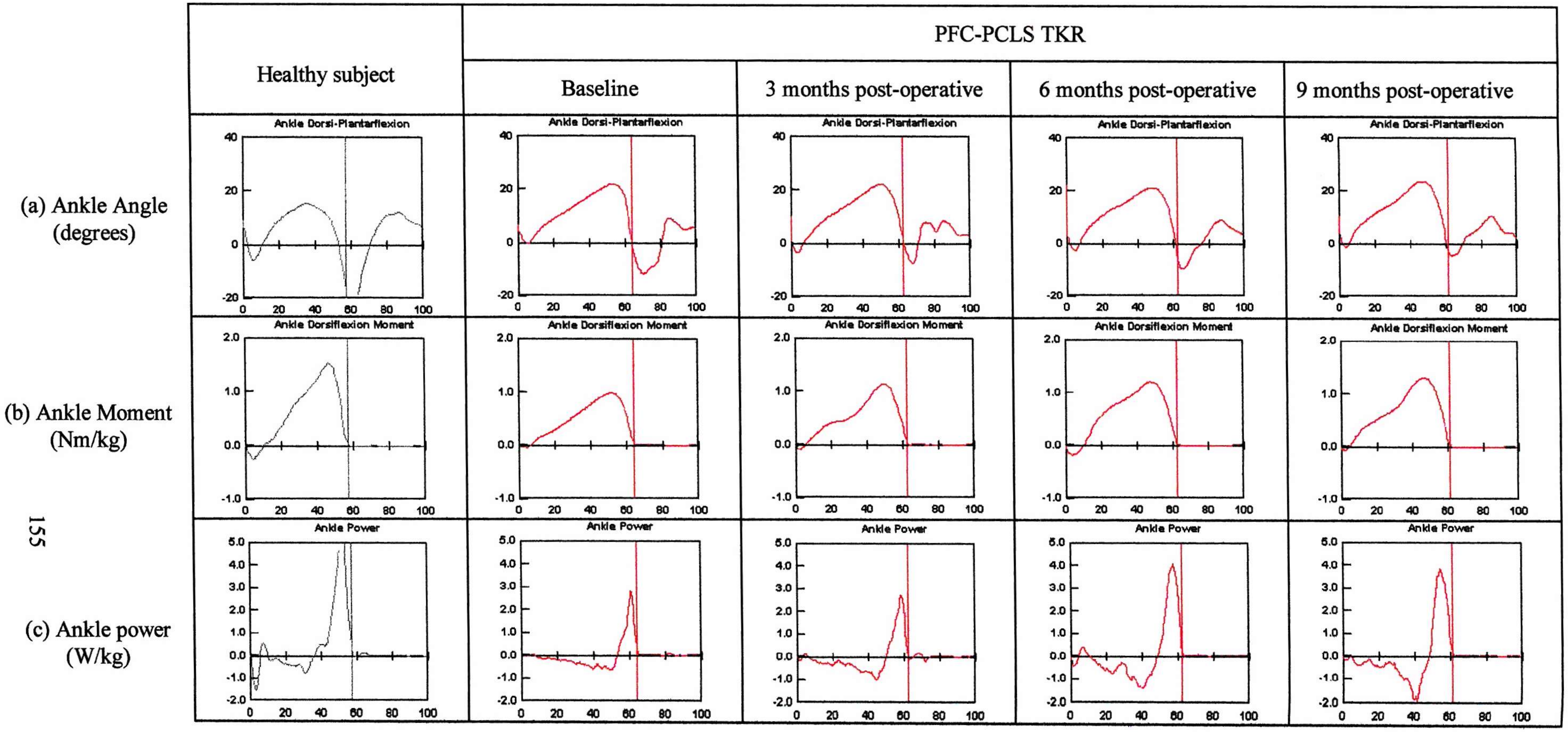

Figure 5.31 Sagittal plane kinematics of the ankle joint (a) and kinetics data (b and c) of a patient (FP) at baseline, and 3, 6 and 9 months after PFC-PCLS TKR and a healthy subject. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in W/kg.

5.7.1.5. Pattern of EMG activity in patients following PFC-PCLS TKR.

The rectus femoris muscle in this group demonstrated an abnormal pattern of activation during the gait cycle. This was characterised by prolonged activity or slight activity throughout the stance phase before the knee replacement. Prolonged activity started before the end of the swing phase in 3 patients. In 6 patients the rectus femoris was almost silent throughout the gait cycle before the operation and the activity increased at 6 and 9 months after surgery. In 10 patients, rectus femoris activation was generally normal at 9 months after operation. The remaining patients demonstrated slight to moderate activity during the stance phase.

The medial hamstring muscle demonstrated an abnormal pattern of activation between mid-stance and toe-off before the operation in most of the patients in this group. In 3 patients, the medial hamstring muscle showed an almost normal pattern at 6 and 9 months after the operation. The medial hamstring muscle was silent throughout the stance phase and swing phase in most of the patients throughout the study period.

The tibialis anterior muscle of the 15 patients was abnormally active throughout the gait cycle and showed prolonged activity during the stance and swing phases at 3, 6, and 9 months follow-up. There was continuous anterior tibial muscle activity in 10 out of 15 patients throughout the gait cycle. In the remaining patients, there was a variety of activity during the stance phase and early swing phase.

The gastrocnemius muscle in the majority of the patients was active in the pre-swing phase. Two patients showed prolonged activation during the stance phase. In three patients there was a prolonged activation during the stance and swing phase. In four patients the calf muscles showed an almost normal pattern of activity. The gastrocnemius muscle was almost silent throughout the gait cycle in one patient.

Figures 5.32-5.33 are records obtained from one patient at 3, and 9 months after the operation, and are representative of the abnormal patterns of activation seen in 15 patients.

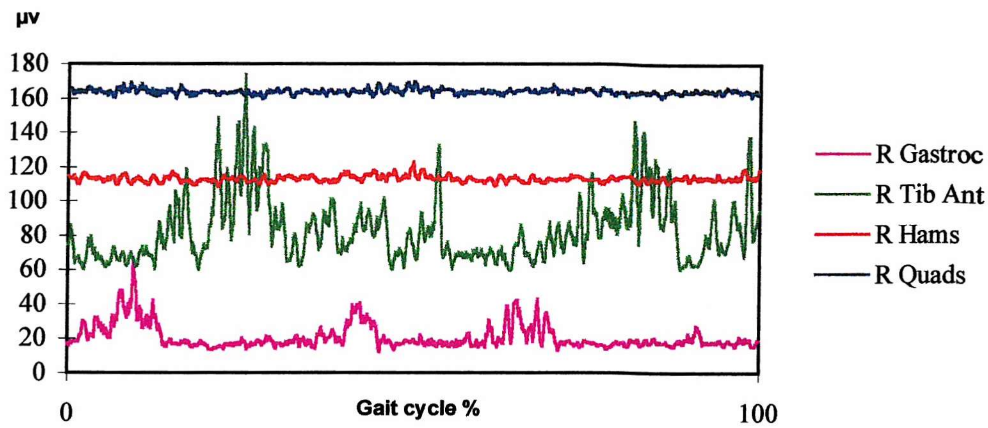

Figure 5-32 Rectified EMG sampled at 1600 Hz of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of one patient (FP2R3) 3 months after PFC-PCLS TKR during walking at free speed. The record represents one gait cycle.

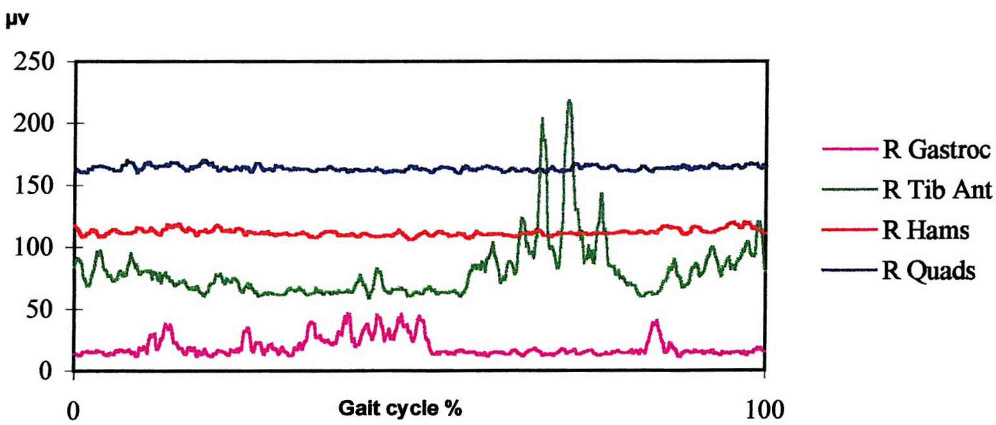

Figure 5-33 Rectified EMG sampled at 1600 Hz of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of one patient (FP4R3) 9 months after PFC-PCLS TKR during walking at free speed. The record represents one gait cycle.

Summary:

The temporospatial, kinematic and kinetic parameters of the patients implanted with press-fit cruciate ligament sacrificing condylar prostheses (group A) were analysed. All patients in this group were followed up 3, 6 and 9 months after surgery. All the fifteen patients were treated by a single surgeon using the posterior cruciate ligament sacrificing prosthesis.

The number of patients reduced to 14, 11 and 13 at 3, 6 and 9 months after operation respectively.

A significant increase, compared to baseline, in walking speed and stride length was observed at 3, 6 and 9 months after surgery.

A significant increase in the maximum knee flexion at loading response was observed at 9 months after surgery.

A significant increase in the maximum knee flexion at mid-swing was observed at 3, 6 and 9 months after surgery.

A significant decrease in the maximum knee moments at mid-stance was observed at 3, 6 and 9 months after surgery.

A significant increase in the maximum knee powers at mid-stance occurred only 9 months after surgery.

Abnormal muscle activity was noted during level walking in four lower limb muscles at 3 and 6 months after TKR, but most of the patients showed phasic activity at 9 months after TKR.

Overall, the knee moments generated at the knee joint were decreased significantly at 3, 6 and 9 months after surgery. Pain was either reduced or disappeared completely at 9 months after surgery.

5.8.2. Press-Fit Condylar PCL-Retaining (PFC-PCLR) knee replacement (group B).

This section will describe the changes over time in this group. There were 23 patients in this group.

5.8.2.1. Temporospatial gait parameters

The temporospatial gait parameters measured during gait analysis for the both PFC-PCLR TKR and the control group are listed in Table 5.13. The patients in this group walked slower at 3, 6 and 9 months after operation than the control group and had a shorter stride length.

The values are given as median, IQR and the range. Statistically significant improvements were observed at 3, 6 and 9 months post-operatively for walking speed and stride length ($P < 0.05$). The timing of mid-stance and mid-swing of the gait cycle were significantly delayed in the patient group during the study period.

The plots show the results for the control subjects and the patients at baseline and 3, 6, 9 months follow-up after PFC-PCLR TKR for comparison (Figures 5.34-5.35).

Table 5-13 Median, IQR and range of temporospatial parameters at baseline and the three test periods following PFC-PCR TKR at free walking speed

Gait parameters	Control (n = 25)	Group B			
		Baseline (n = 23)	3 months (n = 23)	6 months (n = 20)	9 months (n = 18)
	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Walking speed (m/s)	1.18 (1.06-1.26) (0.93-1.50)	0.53 (0.37-0.66) (0.13-0.96)	0.86 (0.59-0.92) (0.42-1.16)*	0.91 (0.69-1.07) (0.51-1.38)*	0.90 (0.71-1.02) (0.39-1.36)*
Stride length (m)	1.31 (1.18-1.38) (1.07-1.61)	0.72 (0.63-0.93) (0.26-1.13)	0.90 0.79-1.08) (0.67-1.17)*	1.02 (0.89-1.13) (0.70-1.23)*	1.07 (0.83-1.16) (0.67-1.32)*
Mid-stance (% cycle)	30.22 (29.98-30.67) (28.93-31.43)	32.11 (31.56-35.09) (29.01-42.70)	31.15 (20.08-32.89) (28.30-34.15)*	31.88 (30.73-32.68) (29.75-34.57)*	31.58 (30.87-32.76) (29.13-46.23)*
Mid-swing (% cycle)	80.22 79.98-80.67 (78.93-81.43)	82.11 (81.56-85.09) (79.01-92.70)	81.15 (80.08-82.89) (78.30-84.15)*	81.88 (80.73-82.68) (79.75-84.57)*	81.58 (80.87-82.76) (79.13-96.23)*

**Denotes statisticaly significant differences (P < 0.05) (Mann-Whitney U test) from the baseline(groupB).*

IQR, Inter-quartile range.

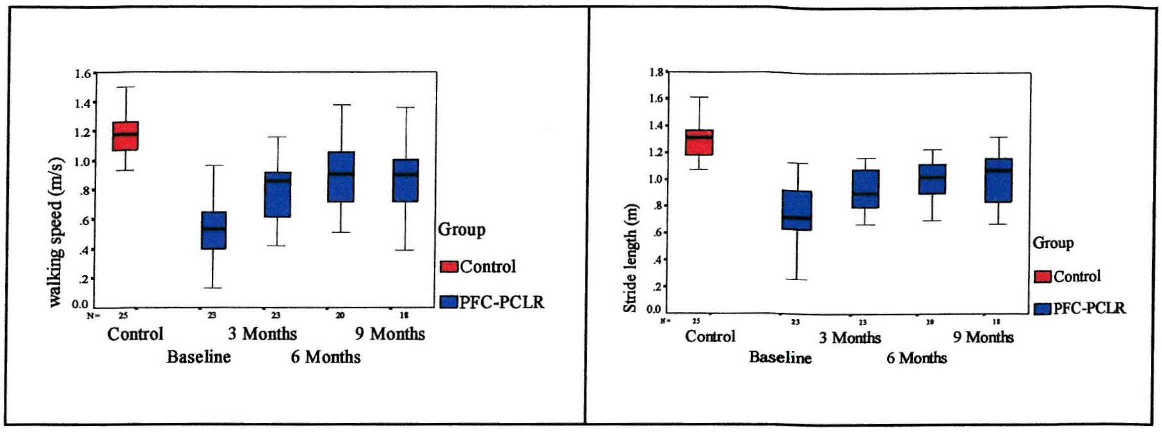

Figure 5-34 Boxplots comparing the differences in walking speed and stride length between control subjects and patients at baseline and 3, 6, and 9 months follow-up after PFC-PCLR TKR. The median, IQR and range are shown.

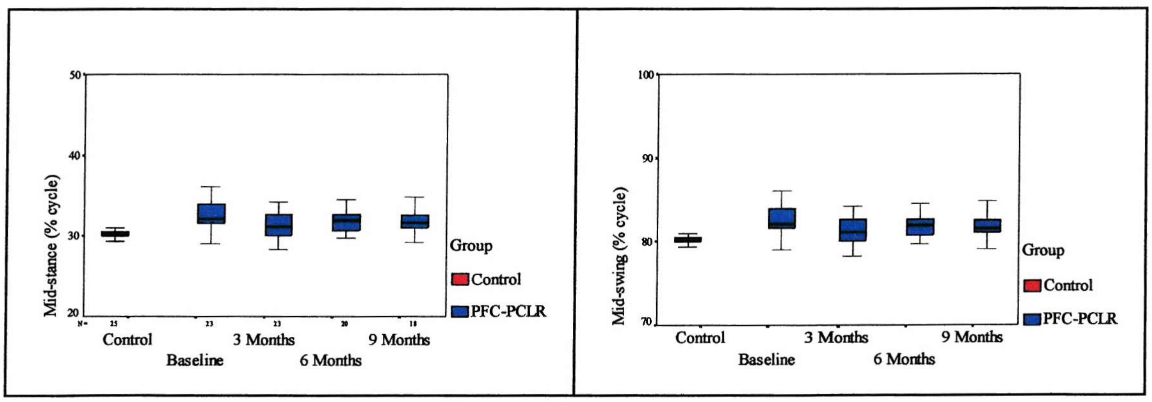

Figure 5-35 Boxplots comparing the differences in the timing of mid-stance and mid-swing in the gait cycle between control subjects and patients at baseline and 3, 6, and 9 months follow-up after PFC-PCLR TKR. The median, IQR and range are shown.

5.8.2.2. Kinematic parameters during gait

The kinematic gait parameters of patients who received PFC-PCLR TKR are given in Table 5.14. Maximum hip extension in stance increased at the 3 and 6 months post-operative evaluation. Although it reduced slightly at a 9 months, it was still more than the baseline value. The maximum knee flexion during the stance phase was slightly less at 3 and 6 months follow-up compared to the pre-operative value. The largest increase in the maximum knee flexion during swing phase occurred at 6and 9 months after surgery. All the above values were statistically significant at the three follow-up periods after surgery, compared to the control group. The maximum ankle dorsiflexion in swing increased significantly 9 months after surgery compared to baseline but was still less than that in the control group. The plots show the results for the control subjects and patients at the baseline and 3, 6, and 9 months follow-up after PFC-PCLR TKR for comparison (Figures 5.36-5.37).

Table 5-14 Median, IQR and range of the kinematic parameters at baseline and the three test periods following PFC-PCR TKR at free walking speed.

Gait parameters (degrees)	Control (n = 25)	Group B			
		Baseline (n = 23)	3 months (n = 23)	6 months (n = 20)	9 months (n = 18)
	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Max hip extension (ST)	36.80 (33.80-43.35) (29.50-53.50)	26.60 (21.40-33.40 (11.40-38.90)	29.20 (26.30-37.00) (23 21-78.00)	34.90 (27.75-38.30) (26.30-43.60)*	31.35 (26.58-36.60 (22.30-44.51)*
Max knee flexion (LR)	13.20 (10.40-18.50) (4.10-22.10)	5.00 (2.50-6.40) (0.50-90.20)	5.71 (3.60-6.70) (0.20-14.70)	6.15 (4.73-9.75) (2.30-16.60)*	6.45 (3.90-9.88) (2.20-14.00)*
Max knee flexion (SW)	60.90 (58.60-66.50) (42.90-76.10)	35.10 (29.00-46.00) (6.50-52.40)	43.40 (35.60-49.20) (22.57-57.90)*	47.60 (45.35-51.83) (33.50-57.60)*	48.65 (39.73-51.10) (33.80-56.80)*
Max ankle dorsiflexion (SW)	27.20 (26.30-28.35) (25.30-30.10)	21.90 (20.70-23.90) (18.30-25.50)	23.40 22.30-25.50) (19.70-26.80)	24.50 (23.33-26.50) (20.30-27.90)*	25.55 (24.65-26.43) (24.10-27.50)*

**Denotes statisticaly significant differences (P < 0.05) (Mann-Whitney U test) from the baseline(group B).*

IQR, Inter-quartile range.

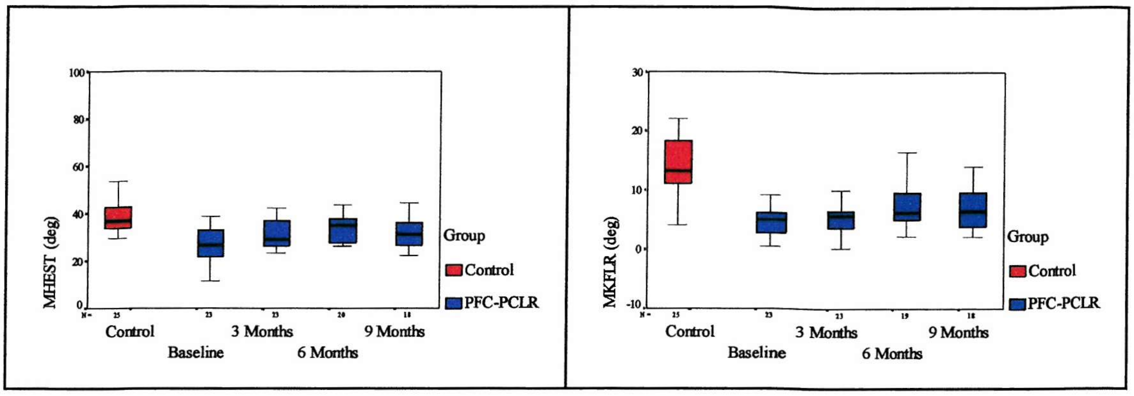

Figure 5-36 Boxplots comparing the differences in maximum hip extension in stance (MHEST) and maximum knee flexion in loading response (MKFLR) between control subjects and patients at baseline and 3, 6, and 9 months follow-up after PFC-PCLR TKR. The median, IQR and range are shown.

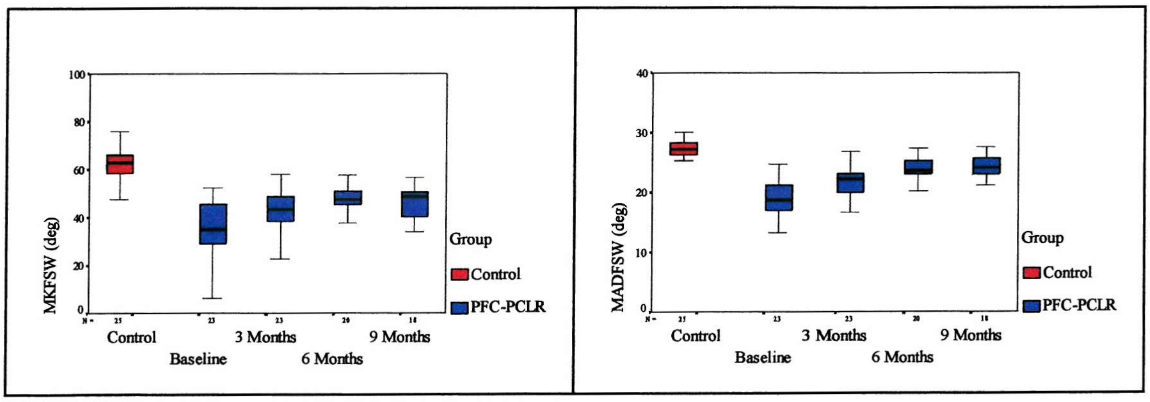

Figure 5-37 Boxplots comparing the differences in maximum knee flexion in swing (MKFSW), and maximum ankle dorsiflexion in swing (MADFSW) between control subjects and patients at baseline and 3, 6, and 9 months follow-up after PFC-PCLR TKR. The median, IQR and range are shown.

5.8.2.3. Kinetic parameters during gait

Results of the moments and powers generated at the knee and ankle of the study and control groups are given in Table 5.15. The maximum knee moment in mid-stance was reduced at the three months follow-up period after operation.

This was slightly less at 6 months but it was still higher than that of control group (P < 0.05). Throughout the post-operative follow-up period the maximum power generated in mid-stance was less compared to the control group and the differences were statistically significant. The maximum ankle moment increased at 3 months and 6 months after the operation.

The maximum ankle power showed significant improvement at 3, 6 and 9 months after the operation, but again there were no further significant changes at 9 months. The plots show the results for the control subjects and the patients at baseline and 3, 6, and 9 months follow-up after PFC-PCLR TKR for comparison (Figures 5.38-5.39).

Table 5-15 Median, IQR and range of kinetic parameters at baseline and for the three test periods following PFC-PCLR TKR at free walking speed.

Gait parameters	Control (n = 25)	Group B			
		Baseline (n = 23)	3 months (n = 23)	6 months (n = 20)	9 months (n = 18)
	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Max Knee moment in mid-stance (Nm/kg)	0.12 (-0.35-0.21) (-0.13-0.33)	0.31 0.19-0.45) (0.01-0.79)	0.30 (0.21-0.47) (0.00-0.54)	0.29 (0.18-0.36) (0.11-0.56)*	0.35 (0.21-0.46) (0.06-0.56)*
Max Knee power in mid-stance (W/kg)	0.12 (0.03-0.29) (-0.20-0.46)	0.01 (-0.04-0.09) (-0.11-0.27)	0.03 (-0.09-0.08) (-0.29-0.58)	0.04 (-0.07-0.18) (-0.29-0.50)	0.02 (-0.16-0.14) (-0.65-0.56)*
Max ankle moment in pre-swing (Nm/kg)	0.80 (0.66-1.02) (0.28-2.10)	0.50 (0.28-0.67) (0.12-1.36)	0.64 (0.39-0.78) (0.17-1.27)	0.76 (0.65-1.00) (0.11-1.67)*	0.75 (0.67-0.89) (0.37-1.26)*
Max ankle power in pre-swing (W/kg)	3.64 (3.14-4.71) (2.33-7.02)	1.25 (0.55-2.07) (0.02-4.39)	1.81 (1.05-2.59) (0.73-3.26)*	2.11 (1.56-3.13) (0.54-4.92)*	2.43 (1.30-3.01) (0.78-6.55)*

**Denotes statisticaly significant differences (P < 0.05) (Mann-Whitney U test) from the baseline(group B).*

IQR, Inter-quartile range.

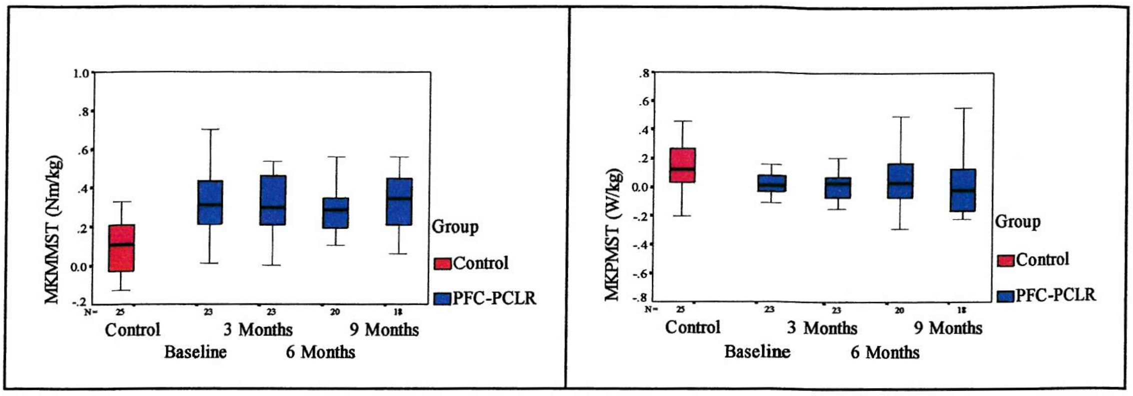

Figure 5-38 Boxplots comparing the differences in maximum knee moment in mid-stance (MKMMST) and maximum knee power in mid-stance (MKPMST) between control subjects and patients at baseline and 3, 6, and 9 months follow-up after PFC-PCLR TKR. The median, IQR and range are shown.

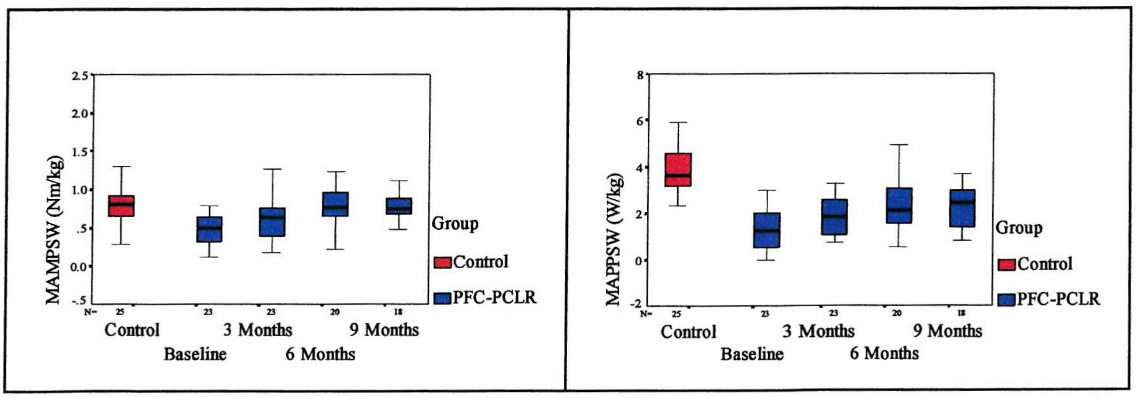

Figure 5-39 Boxplots comparing the differences in maximum ankle moment in pre-swing (MAMPSW) and maximum ankle power in mid-stance (MAPPSW) between control subjects and patients at baseline and 3, 6, and 9 months follow-up after PFC-PCLR TKR. The median, IQR and range are shown.

5.8.2.4. Example of the kinetic and kinematic data obtained from one patient in this group at baseline and 3, 6, and 9 months after PFC-PCLR TKR.

Figures 5.40-5.42 are examples recorded in the sagittal plane of the joint ROM in degrees for the hip, knee and ankle with the simultaneous hip, knee, and ankle joint moments and powers measured in Nm/kg and W/kg respectively. The measurements were taken at baseline and 3, 6, and 9 months after operation.

Kinematics of the hip

It can be noticed from Figure 5.40 that the maximum hip extension in the stance phase increased slightly from baseline at 3 and 6 months after the operation. However, it was similar to the baseline value at 9 months.

Kinematics and kinetics of the knee

Figure 5.41 shows the knee kinematics and kinetics of the same patient at baseline and 3, 6, and 9 months after operation.

The maximum knee flexion in the loading response phase and in mid-stance and the maximum knee flexion in mid-swing increased significantly at 9 months after operation compared with baseline values

The knee moment and power in mid-stance changed slightly from the baseline during the study period.

Kinematics and kinetics of the ankle.

Figures 5.42 shows the ankle kinematics and kinetics of the same patient. The maximum ankle dorsiflexion in the swing phase remained unchanged from baseline at 9 months of the operation. The maximum ankle dorsiflexion moment in pre-swing also remained unchanged form baseline after the operation. However, the maximum ankle power in pre-swing increased at 3 months after operation and remained at approximately the same level at the second and third follow-up visits.

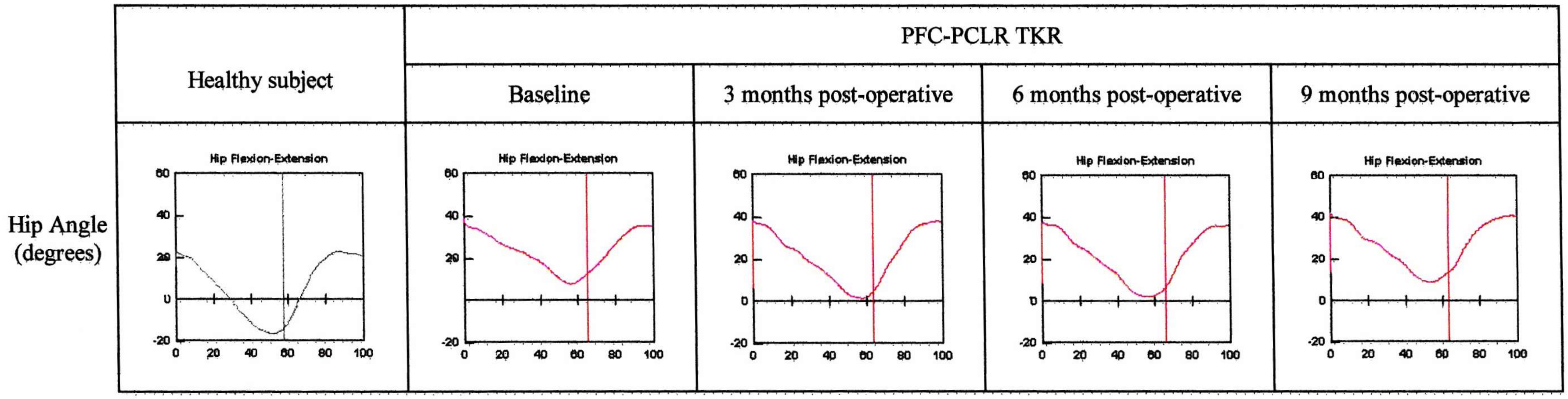

Figure 5.40 Sagittal plane kinematics of the hip joint of a patient (AB) at baseline and 3, 6 and 9 months after PFC-PCLR TKR and a healthy subject. Joint range of motion is shown in degrees.

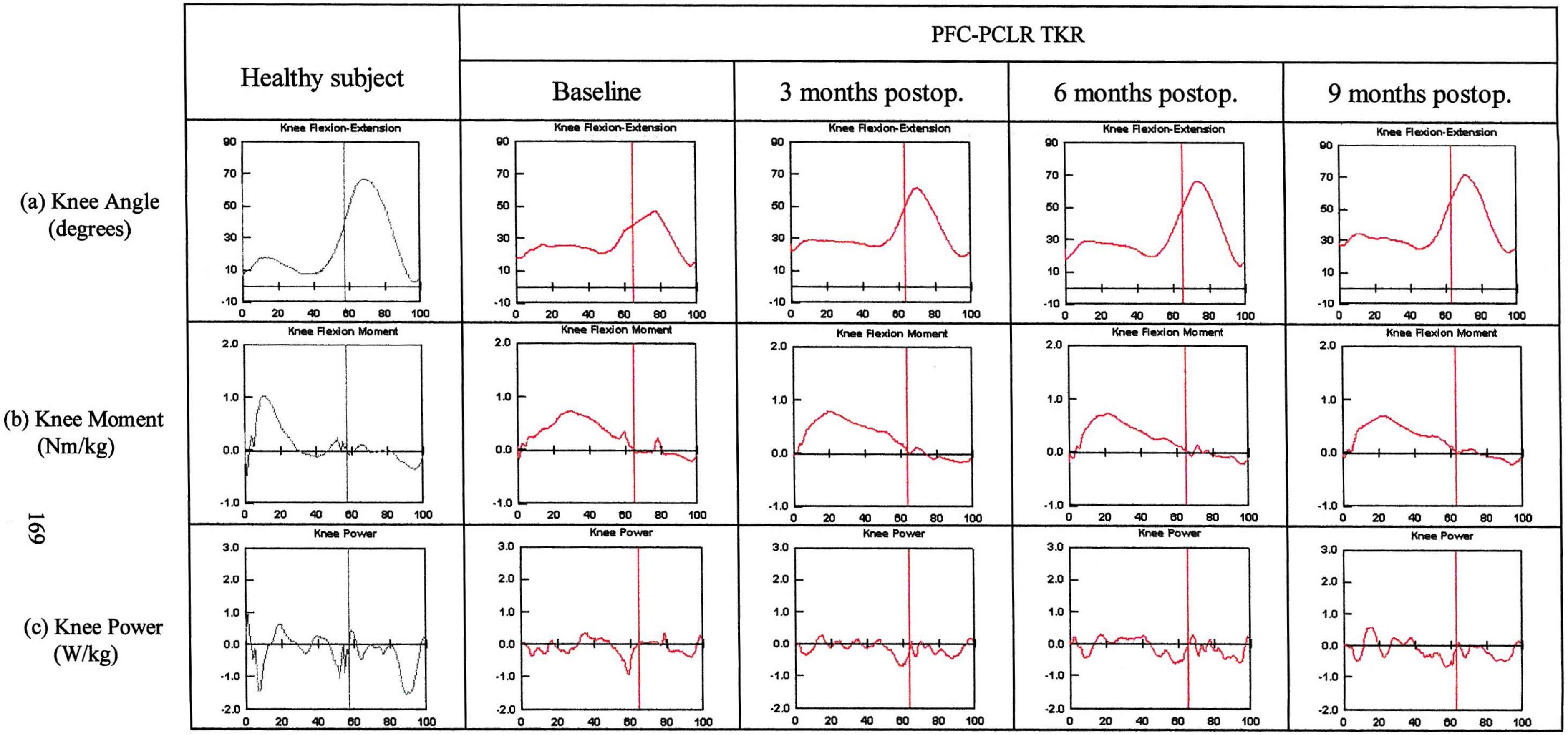

Figure 5.41 Sagittal plane kinematics of the knee joint (a) and kinetics data (b and c) of a patient (AB) at baseline, and 3, 6 and 9 months after PFC-PCLR TKR and a healthy subject. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in W/kg.

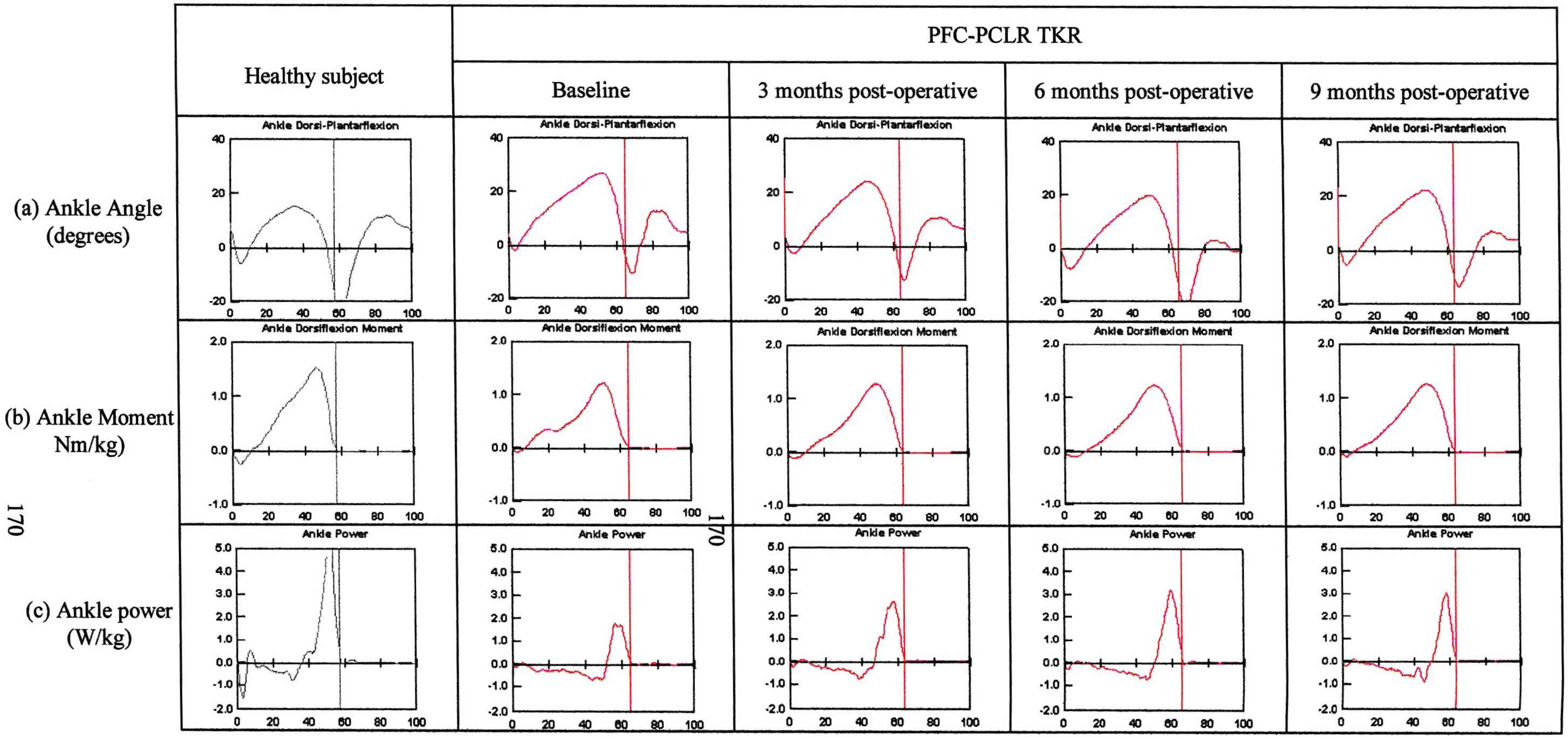

Figure 5.42 Sagittal plane kinematics of the ankle joint (a) and kinetics data (b and c) of a patient (AB) at baseline, and 3, 6 and 9 months after PFC-PCLR TKR and a healthy subject. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in W/kg.

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5.8.2.5. EMG activity in patients following PFC-PCLR TKR.

The rectus femoris muscle in this group demonstrated an abnormal pattern of activation characterised by prolonged activity throughout the stance phase and swing phase before the operation in most of the patients. In 2 patients, the rectus femoris was almost silent or only slightly active throughout the gait cycle throughout the study period.

The medial hamstring muscle also demonstrated prolonged activation between mid-stance and toe-off.

The tibialis anterior muscle was active throughout the gait cycle before surgery and after the operation.

Activation of the gastrocnemius was minimal in the pre-swing phase in most of the patients at baseline. This did not change after the TKR.

Figures 5.43-5.44 are records obtained from one patient at 3, and 9 months after the operation and are representative of the abnormal patterns of activation seen in 23 patients.

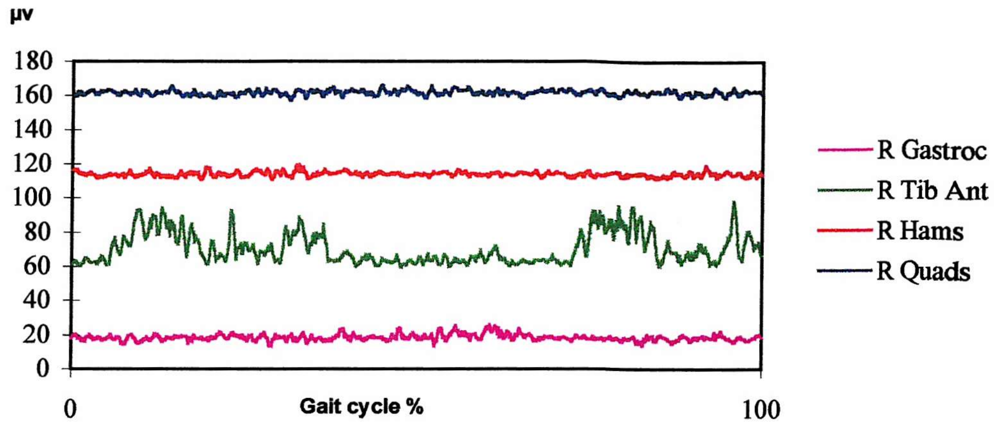

Figure 5 .43 Rectified EMG sampled at 1600 Hz of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of one patient 3 (AB2R2) months after PFC-PCLR TKR during walking at free speed. The record represents one gait cycle.

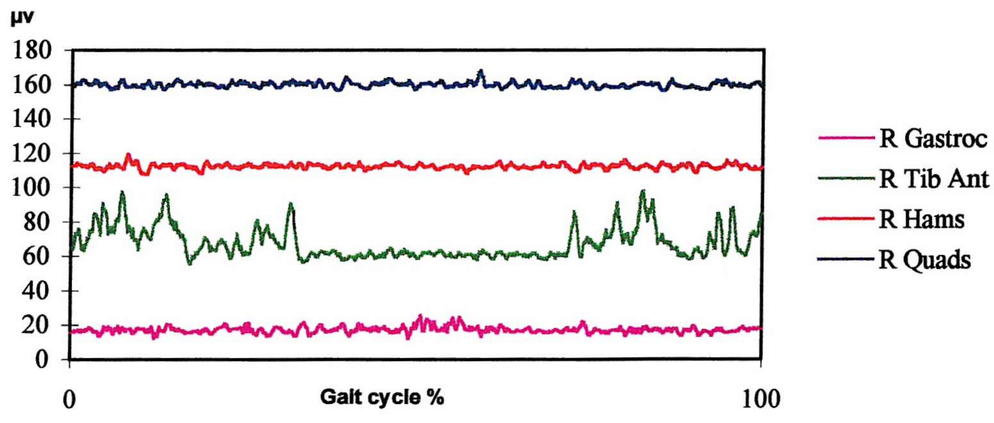

Figure 5.44 Rectified EMG sampled at 1600 Hz of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of one patient (AB4R3) 9 months after PFC-PCLR TKR during walking at free speed. The record represents one gait cycle.

Summary:

The temporospatial, kinematic and kinetic parameters of the patients implanted with press-fit cruciate ligament retaining condylar prostheses (group B) were collected. All patients in this group were followed up 3, 6 and 9 months after surgery. The number of patients reduced to 20 and 18 at 6 and 9 months after operation respectively.

Twenty three patients were treated using PFC-PCLR prostheses by four different surgeons. Ten cases were performed by a single surgeon and the remainder by the other three consultant surgeons.

A significant increase in walking speed and stride length was observed at 3, 6 and 9 months after surgery.

A significant increase in the maximum knee flexion at loading response was observed at 9 months after surgery.

A significant increase in the maximum knee flexion at mid-swing was observed at 3, 6 and 9 months after surgery.

A significant decrease in the maximum knee moments at mid-stance was observed only at 9 months after surgery.

A significant increase in the maximum knee powers at mid-stance occurred only 3 and 6 months after surgery.

Abnormal muscle activity was noted during level walking in four lower limb muscles at 3 and 6 months after TKR, but most of the patients showed phasic activity at 9 months after TKR.

Overall, knee moments generated at the knee joint were decreased significantly at 3, 6 and 9 months after surgery. Pain was either reduced or disappeared completely at 9 months after surgery.

5.8.3. Mobile bearing posterior cruciate ligament retaining (MBK-PCLR) knee replacement (group C).

This section will describe the changes over time in this group. There were 16 patients in this group.

5.8.3.1. Temporospatial gait parameters

Analysis of the temporospatial gait parameters (Table 5.16) shows that the patients who underwent MBK-TKR had significant improvements in the medianvalues of their walking speed and stride length at follow-up 3, 6 and 9 months after surgery compared with values before surgery.

The walking speed averaged 0.59m/s before surgery and increased to 1.00 m/s nine months later. Although there was some improvement in the walking speed and stride length, the values did not reach those of control subjects. The stride length averaged 0.76m before surgery, compared to 1.08m nine months after surgery. Mid-stance and mid-swing timings decreased at 3 months but remained unchanged thereafter.

The plots show the results for the control subjects and patients at the baseline and 3, 6, and 9 months follow-up after MBK-PCLR TKR for comparison (Figures 5.45-5.46).

Table 5-16 Median, IQR and range of temporospatial parameters at baseline and the three test periods following MBK-PCLR TKR at free walking speed.

Gait parameters	Control (n = 25)	Group C			
		Baseline (n = 16)	3 months (n = 16)	6 months (n = 10)	9 months (n = 14)
	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Walking speed (m/s)	1.18 (1.06-1.26) (0.93-1.50)	0.59 (0.39-0.69) (0.15-1.03)	0.78 (0.58-0.96) (0.35-1.24)*	0.97 (0.66-1.14) (0.26-1.31)*	1.00 (0.79-1.14) (0.18-1.44)*
Stride length (m)	1.31 (1.18-1.38) (1.07-1.61)	0.76 (0.62-1.00) (0.27-1.17)	1.02 (0.75-1.09) (0.61-1.33)*	1.08 (0.81-1.17) (0.44-1.20)*	1.08 (0.84-1.21) (0.34-1.46)*
Mid-stance (% cycle)	30.22 (29.98-30.67) (28.93-31.43)	33.02 (31.09-35.31) (30.32-46.39)	30.92 (29.99-33.58) (29.39-34.77)*	30.66 (30.06-31.96) (29.33-35.85)*	30.89 (29.98-32.36) (28.57-39.81)*
Mid-swing (% cycle)	80.22 79.98-80.67 (78.93-81.43)	83.02 (81.09-85.31) (80.32-96.39)	80.92 79.99-83.58) (79.39-84.77)*	80.66 (80.06-81.96) (79.33-85.85)*	80.89 (79.98-82.36) (78.57-89.81)*

**Denotes statistically significant differences (P < 0.05) (Mann-Whitney U test) from the baseline(group C).*
IQR, Inter-quartile range.

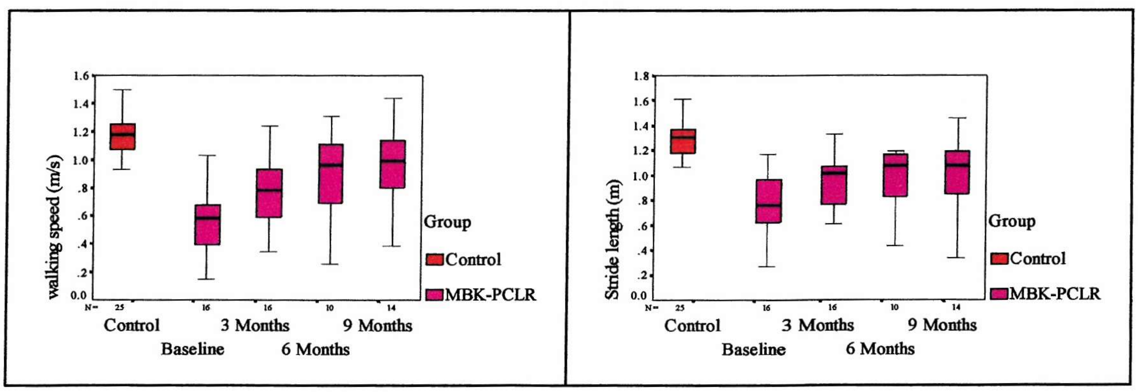

Figure 5-45 Boxplots comparing the differences in walking speed and stride length in the gait cycle between control subjects and patients at baseline and 3, 6, and 9 months follow-up after MBK-PCLR TKR. The median, IQR and range are shown.

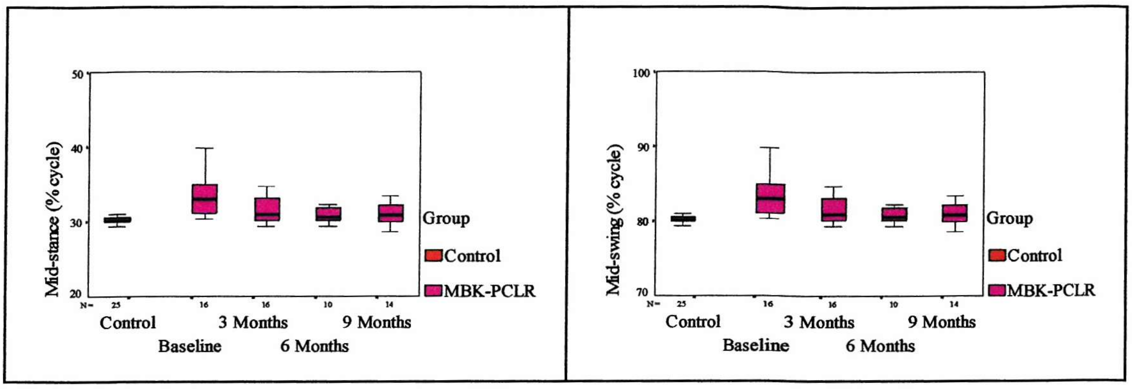

Figure 5-46 Boxplots comparing the differences in the timing of mid-stance and mid-swing in the gait cycle between control subjects and patients at baseline and 3, 6, and 9 months follow-up after MBK-PCLR TKR. The median, IQR and range are shown.

5.8.3.2. Kinematic parameters during gait

The median hip, knee and ankle range of motion during level walking of patients treated with MBK-PCLR prostheses at the baseline and 3, 6, and 9 months after surgery as well as the values for control subjects are shown in Table 5.17. Maximum hip extension in stance changed after surgery but was still significantly lower in the study group compared to the control group (P <0.05) at 9 months follow-up.

Little change in the maximum knee flexion during stance occurred from the pre-operative to the post-operative period and remained significantly lower than in the control group (P < 0.05). Maximum knee flexion during swing increased at each post-operative evaluation but was significantly lower in the study group compared to the control group (P <0.05) at all follow-up visits.

Progressive increase was observed in maximum ankle dorsiflexion in swing at 3, 6 and 9 months after surgery but did not reach the values of those in the control group. The plots show the results for the control subjects and the patients at baseline and 3, 6 and 9 months follow-up after PFC-PCLR TKR for comparison (Figures 5.47-5.48).

Table 5-17 Median, IQR and range of kinematic parameters at baseline and the three test periods following MBK-PCLR TKR at free walking speed.

Gait parameters (degrees)	Control (n = 25)	Group C			
		Baseline (n = 16)	3 months (n = 16)	6 months (n = 10)	9 months (n = 14)
	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Max hip extension (ST)	36.80 (33.80-43.35) (29.50-53.50)	25.70 (20.63-33.80) (16.80-38.90)	30.30 (26.03-34.90) (15.20-40.10)*	32.00 (28.78-34.75) (24.80-41.00)*	33.75. (30.23-36.42) (16.20-42.40)*
Max knee flexion (LR)	13.20 (10.40-18.50) (4.10-22.10)	4.10 (1.80-6.43) (0.01-8.90)	3.05 (1.28-5.15) (0.01-11.00)	2.95 (1.45-4.88) (1.30-9.30)	4.65 (2.30-6.63) (0.90-15.60)*
Max knee flexion (SW)	60.90 (58.60-66.50) (42.90-76.10)	34.25 (26.78-44.35) (7.70-50.00)	45.45 (37.23-55.25) (28.90-63.70)*	47.50 (42.05-53.93) (21.10-57.70)*	52.10 (40.98-57.73) (26.90-61.70)*
Max ankle dorsiflexion (SW)	27.20 (26.30-28.35) (25.30-30.10)	22.60 (19.15-23.78) (16.80-26.00)	23.25 (22.43-24.38) (20.50-25.90)*	24.15 (23.15-25.25) (21.90-26.50)*	25.85 (24.63-26.35) (21.30-27.10)*

**Denotes statistically significant differences ($P < 0.05$) (Mann-Whitney U test) from the baseline(group C).*

IQR, Inter-quartile range.

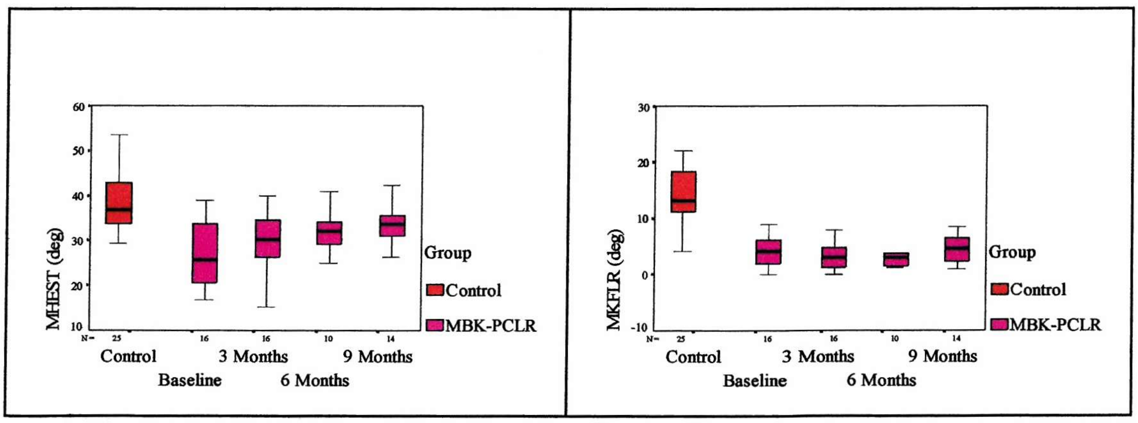

Figure 5-47 Boxplots comparing the differences in maximum hip extension in stance (MHEST) and maximum knee flexion in loading response (MKFLR) between control subjects and patients at baseline and 3, 6, and 9 months follow-up after MBK-PCLR TKR. The median, IQR and range are shown.

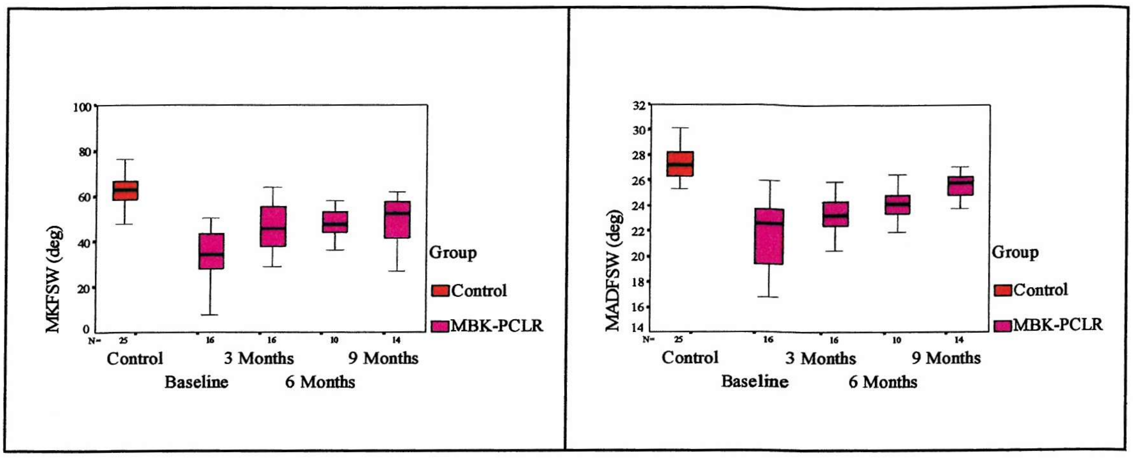

Figure 5-48 Boxplots comparing the differences in maximum knee flexion in swing (MKFSW) and maximum ankle dorsiflexion in swing (MADFSW) between control subjects and patients at baseline and 3, 6, and 9 months follow-up after MBK-PCLR TKR. The median, IQR and range are shown.

5.8.3.3. Kinetic parameters during gait

Results of the moments and powers generated at the knee and ankle joints of the study and control groups are given in Table 5.18. The maximum knee moment in mid-stance became progressively less over the follow-up period. There were statistically significant differences between the study group and the control group at all follow-up visits. There were a significant differences in the maximum knee power in mid-stance between the study group and the control group before and after surgery. Table 5.18 also illustrates the results of the maximum ankle moment and power in pre-swing.

The maximum ankle dorsiflexion moment in pre-swing improved at 6 months and 9 months, and was close to the normal value. In contrast, patients generated significantly less muscle plantarflexion power at this joint during pre-swing in each of the three test periods after surgery, compared to the control subjects. The plots show the results for the control subjects and the patients at baseline and 3, 6, 9 months follow-up after MBK-PCLR TKR for comparison (Figures 5.49-5.50).

Table 5-18 Median, IQR and range of kinetic parameters at baseline and the three test periods following MBK-PCLR TKR at free walking speed.

Gait parameters	Control (n = 25)	Group C			
		Baseline (n = 16)	3 months (n = 16)	6 months (n = 10)	9 months (n = 14)
	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)	Median (IQR) (Range)
Max Knee moment in mid-stance (Nm/kg)	0.12 (-0.35-0.21) (-0.13-0.33	0.37 (0.22-0.51) (0.13-1.09)	0.32 (0.24-0.43) (0.03-0.69)*	0.27 (0.14-0.37) (0.12-0.69)*	0.22 (0.18-0.32) (0.04-0.38)*
Max Knee power in mid-stance (W/kg)	0.12 (0.03-0.29) (-0.20-0.46)	-0.05 (-0.12-0.03) (-0.20-0.11)	-0.03 (-0.12-0.08) (-0.17-0.17)	-0.02 (-0.06-0.05) (-0.13-0.10)	0.00 (-0.15-0.09) (-0.29-0.25)*
Max ankle moment in pre-swing (Nm/kg)	0.80 (0.66-1.02) (0.28-2.10)	0.58 (0.38-0.77) (0.17-0.90)	0.55 (0.42-0.80) (0.32-1.11)*	0.66 (0.63-0.95) (0.54-1.00)*	0.68 (0.51-0.92) (0.04-1.04)*
Max ankle power in pre-swing (W/kg)	3.64 (3.14-4.71) (2.33-7.02)	1.28 (0.54-2.09) (0.23-3.35)	2.21 (1.49-2.98) (0.51-4.42)*	2.89 (1.74-3.34) (0.66-3.70)*	2.41 (1.69-3.16) (0.18-4.50)*

**Denotes statistically significant differences ($P < 0.05$) (Mann-Whitney U test) from the baseline (group C).*

IQR, Inter-quartile range.

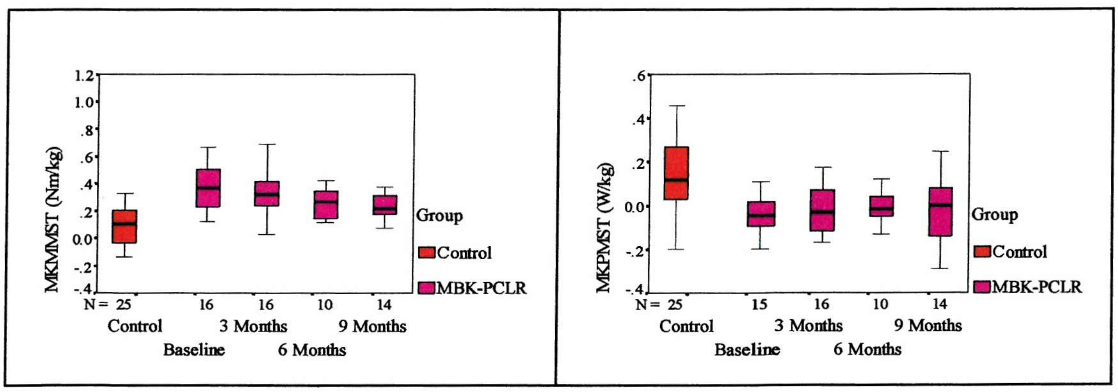

Figure 5-49 Boxplots comparing the differences in maximum knee moment in mid-stance (MKMMST) and maximum knee power in mid-stance (MKPMST) between control subjects and patients at baseline and 3, 6, and 9 months follow-up after MBK-PCLR TKR. The median, IQR and range are shown.

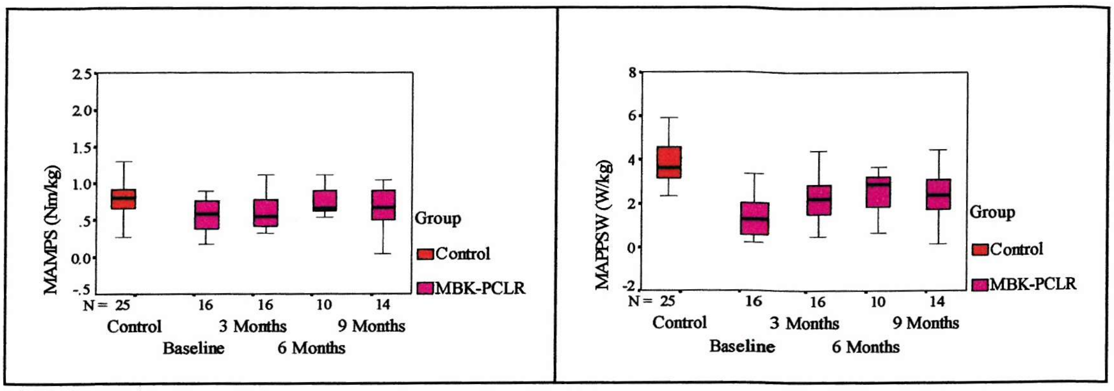

Figure 5-50 Boxplots comparing the differences in maximum ankle moment in pre-swing (MAMPSW) and maximum ankle power in pre-swing (MAPPSW) between control subjects and patients at baseline and 3, 6, and 9 months follow-up after MBK-PCLR TKR. The median, IQR and range are shown.

5.8.3.4. Example of the kinetic and kinematic data obtained from one patient in this group at baseline and 3, 6, and 9 months after MBK-PCLR TKR.

Figures 5.51- 5.53 are examples recorded in the sagittal plane of joint ROM in degrees for the hip, knee and ankle with the simultaneous hip, knee, and ankle joint moments and powers measured in Nm/kg and W/kg respectively. The measurements were taken at baseline and 3, 6, and 9 months after operation.

Kinematics of the hip

It can be noticed from Figure 5.51 that the maximum hip extension in the stance phase progressively increased from baseline to 9 months after operation.

Kinematics and kinetics of the Knee

Figure 5.52 shows the knee kinematics and kinetics of the same patient at baseline and 3, 6, and 9 months after operation.

It can be seen that the maximum knee flexion at loading response and the maximum knee flexion in mid-swing progressively increased from baseline to 6 and 9 months after operation.

The maximum knee moments in mid-stance decreased slightly at 3, 6 and 9 months after the operation. The maximum knee power in mid-stance was unchanged during the follow-up period.

Kinematics and kinetics of the ankle.

Figure 5.53 shows the ankle kinematics and kinetics of the same patient. The maximum ankle dorsiflexion in swing phase progressively increased from baseline to 9 months after operation. The maximum ankle plantarflexion moment and the power in pre-swing increased from baseline after surgery.

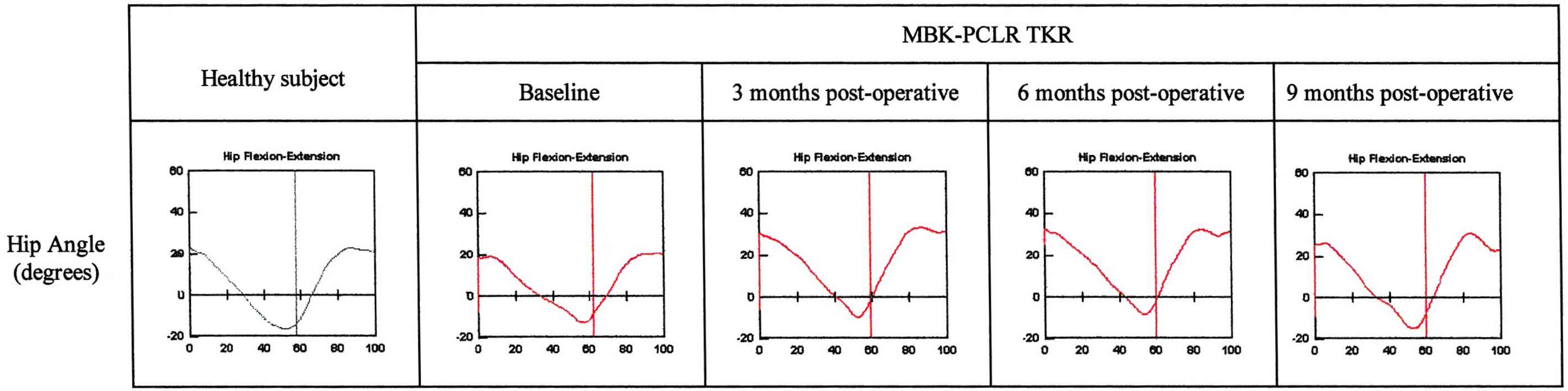

Figure 5.51 Sagittal plane kinematics of thehip joint of a patient (CH) at baseline and 3, 6 and 9 months after MBK-PCLR TKR and a healthy subject. Joint range of motion is shown in degrees.

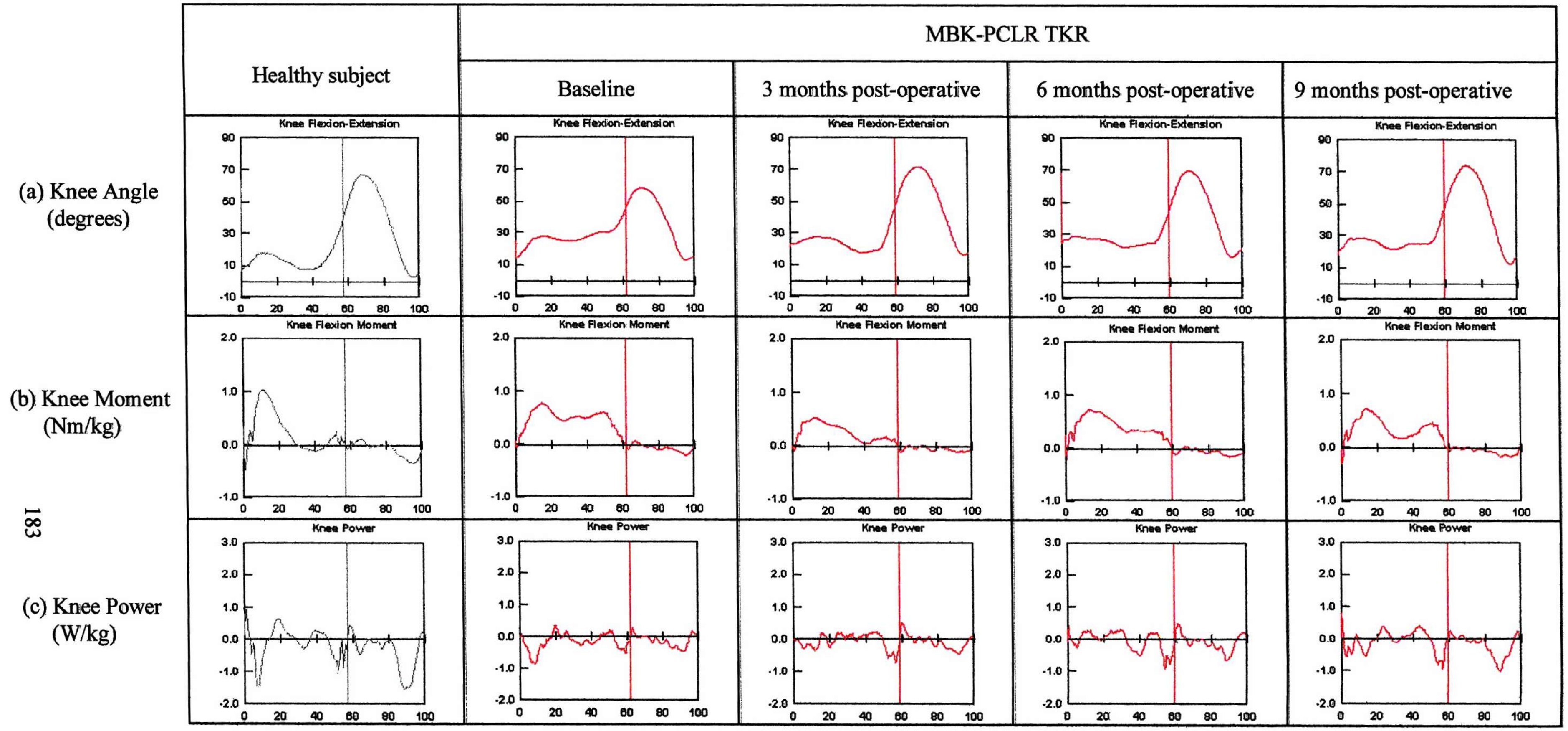

Figure 5.52 Sagittal plane kinematics of the knee joint (a) and kinetics data (b and c) of a patient (CH) at baseline, and 3, 6 and 9 months after MBK-PCLR TKR and a healthy subject. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in W/kg.

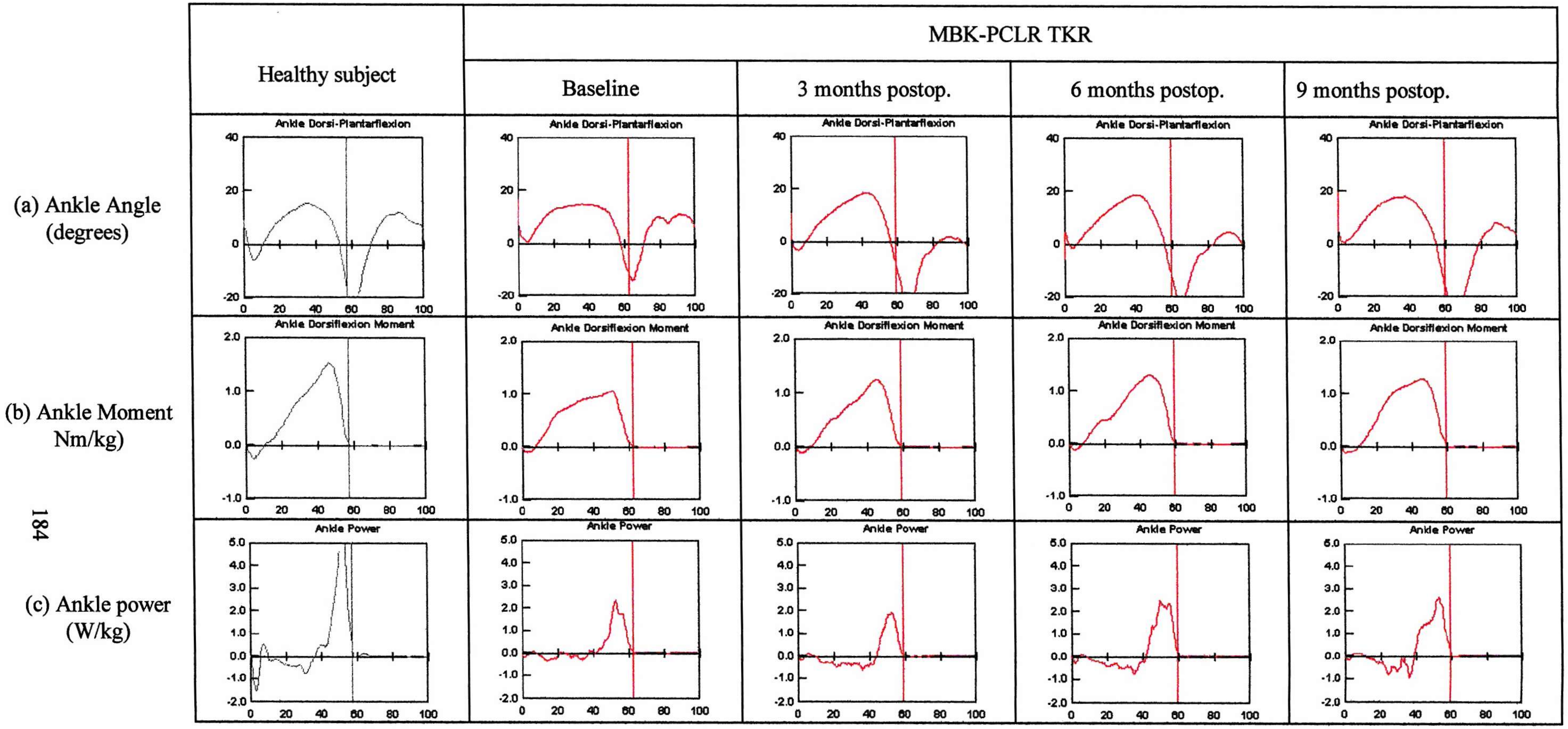

Figure 5.53 Sagittal plane kinematics of the ankle joint (a) and kinetics data (b and c) of a patient (CH) at baseline, and 3, 6 and 9 months after MBK-PCLR TKR and a healthy subject. Joint range of motion is shown in degrees, joint moment in Nm/kg, and joint power in W/kg.

5.8.3.5. EMG in MBK-PCLR TKR group

The rectus femoris muscle in this group was active throughout the stance phase and swing phase after the operation in most of the patients. In 6 patients the rectus femoris was almost silent throughout the gait cycle before surgery but activity increased slightly at 6 and 9 months after the operation.

The medial hamstring muscle was active at the time of initial contact and again between mid-stance and toe-off in most of the patients at 3, 6 and 9 months after the operation.

Tibialis anterior muscle activity was present during the stance and swing phase in most of the patients at 3, 6, 9 months follow-up. In 11 patients the phasic pattern of gastrocnemius muscle absent.

Figures 5.54-5.55 are examples of EMG records of one patient at 3, and 9 months after surgery.

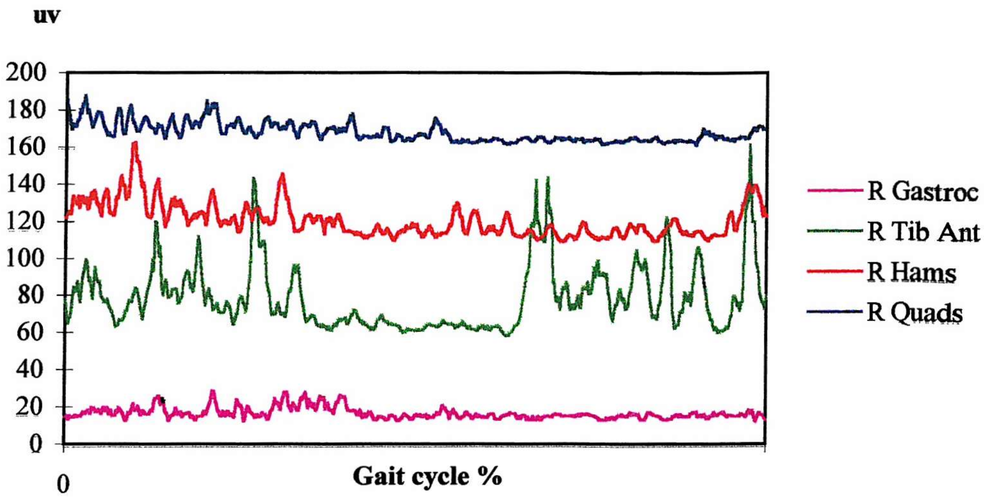

Figure 5-54 Rectified EMG sampled at 1600 Hz of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of one patient (CH2R3) 3 months after MBK-PCLR TKR during walking at free speed. The record represents one gait cycle.

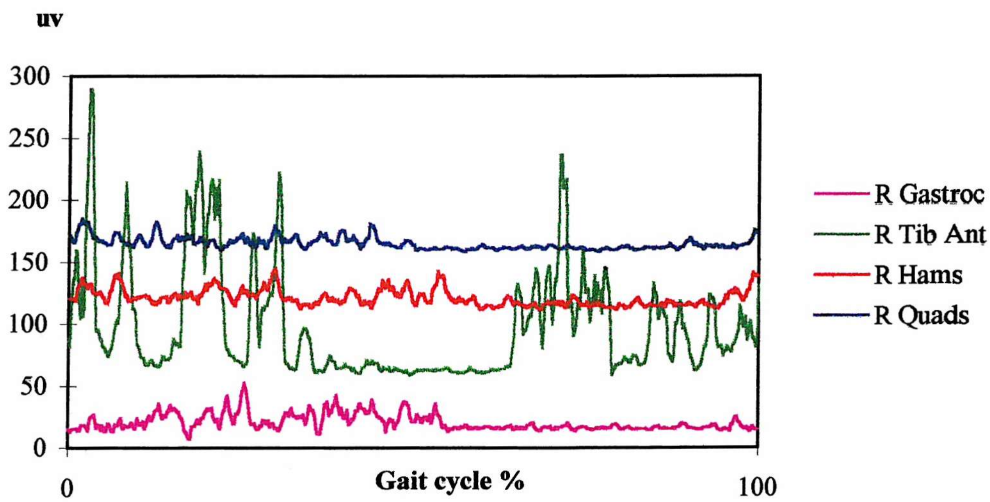

Figure 5-55 Rectified EMG sampled at 1600 Hz of the rectus femoris (Quads), medial hamstring (Hams), tibialis anterior (Tib Ant), and gastrocnemius (Gastroc) muscle groups of one patient (CH4R3) 9 months after MBK-PCLR TKR during walking at free speed. The record represents one gait cycle.

Summary:

The temporospatial, kinematic and kinetic parameters of the patients implanted with mobile bearing knee retaining prostheses prostheses (group C) were collected. All patients in this group were followed up 3, 6 and 9 months after surgery. All the sixteen cases were treated by a single surgeon using the posterior cruciate ligament retaining mobile bearing knee prosthesis.

A significant increase in walking speed and stride length was observed at 3, 6 and 9 months after surgery.

A significant increase in the maximum knee flexion at loading response was observed at 9 months after surgery.

A significant increase in the maximum knee flexion at mid-swing was observed at 3, 6 and 9 months after surgery.

A significant decrease in the maximum knee moments at mid-stance was observed at 3, 6 and 9 months after surgery.

A significant increase in the maximum knee powers at mid-stance occurred only 9 months after surgery.

Abnormal muscle activity was noted during level walking in four lower limb muscles at 3 and 6 months after TKR, but most of the patients showed phasic activity at 9 months after TKR.

Overall, knee moments generated at the knee joint were decreased significantly at 3, 6 and 9 months after surgery. Pain was either reduced or disappeared completely at 9 months after surgery.

5.9. Comparison between the three groups of TKR over time

5.9.1. Temporospatial gait parameters

Summary statistics of the walking speed, stride length, the timing of mid-stance, and mid-swing of the three groups at 3, 6 and 9 months follow-up are given in Table 5.19. The Kruskal Wallis test showed that there were no statistically significant differences in the temporospatial gait parameters in the follow-up period among group A (PFC-PCLS), group B (PFC-PCLR) and group C (MBK-PCLR).

5.9.2. Kinematic gait parameters

Summary statistics of the maximum hip extension in stance, maximum knee flexion in loading response, maximum knee flexion in swing and maximum ankle dorsiflexion in swing, of the three groups at 3, 6 and 9 months follow-up are given in Table 5.20.

The Kruskal Wallis test showed that there were no statistically significant differences among the three groups in any of the kinematic gait parameters in the follow-up period.

5.9.3. Kinetic gait parameters

Summary statistics of the maximum knee moments in stance, the maximum knee power in stance, maximum ankle moment in stance and maximum ankle power in stance of the three groups at 3, 6 and 9 months follow-up are given in Table 5.21.

The Kruskal Wallis test showed that there were no statistically significant differences in all kinetic gait parameters in the follow-up period among the three groups, except for the maximum knee moment in stance at 9 months follow-up. These were highest for group B and lowest for group A.

Table 5-19 Comparison of the temporospatial gait parameters between groups over time 3, 6 and 9 months after TKR. Patients were grouped according to the type of prostheses implanted.

Gait parameter	Period months)	Group A[1] Median (IQR)	Group B[2] Median (IQR)	Group C[3] Median (IQR)	P*-Value
Walking speed (m/s)	3	0.78 (0.59-0.97)	0.86 (0.59-0.92)	0.78 (0.58-0.96)	0.91
	6	0.94 (0.81-1.07)	0.91 (0.69-1.06)	0.97 (0.66-1.13)	0.92
	9	0.99 (0.86-1.08)	0.90 (0.71-1.02)	1.00 (0.79-1.14)	0.49
Stride length (m)	3	0.97 (0.76-1.12)	0.90 (0.79-1.08)	1.02 (0.75-1.88)	0.83
	6	1.06 (0.77-1.15)	1.02 (0.89-1.13)	1.08 (0.80-1.17)	0.92
	9	1.12 (0.97-1.17)	1.07 (0.83-1.16)	1.08 (0.84-1.21)	0.79
Mid-stance (% cycle)	3	30.76 (30.38-31.41)	31.15 (30.08-32.89)	30.92 (29.99-33.58)	0.65
	6	30.39 (29.78-30.75)	31.88 (30.72-32.68)	30.66 (30.06-31.95)	0.06
	9	30.50 (29.65-30.89)	31.58 (30.87- 32.76)	30.89 (29.98- 32.36)	0.07
Mid-swing (% cycle)	3	80.76 (80.38-81.41)	81.14 (80.08-82.89)	80.92 (79.99- 83.58)	0.65
	6	80.39 (79.78-80.75)	81.88 (80.73-82.68)	80.66 (80.06-81.95)	0.06
	9	80.50 (79.65-80.89)	81.58 (80.87- 82.76	80.89 (79.98- 82.36)	0.07

[1]Underwent PFC PCL-sacrificing, [2]Underwent PFC PCL-retaining, [3]Underwent MBK PCL-retaining.
* Kruskal Wallis test among the 3, 6 and 9 months in the three groups A, B and C.

Table 5-20 Comparison of the kinematic gait parameters between groups over time 3, 6 and 9 months after TKR. Patients were grouped according to the type of prostheses implanted.

Gait parameter (degrees)	Period months)	Group A[1] Median (IQR)	Group B[2] Median (IQR)	Group C[3] Median (IQR)	P*-Value
Max hip extension (ST)	3	28.95 (23.60-35.40)	29.20 (26.30-37.00)	30.30 (26.03-34.90)	0.76
	6	35.90 (27.70-38.40)	34.90 (27.75-38.30)	32.00 (28.77-34.75)	0.56
	9	32.80 (25.60-38.85)	31.35 (26.57-35.60)	33.75 (30.22-36.42)	0.75
Max knee flexion (LR)	3	3.05 (2.15-8.17)	5.70 (3.60-6.70)	3.05 (1.28-5.15)	0.92
	6	47.70 (36.70-52.60)	6.15 (4.73-9.75)	2.95 (1.45-4.87)	0.11
	9	6.90 (4.65-10.20)	6.45 (3.90-9.86)	4.65 (2.30-6.63)	0.13
Max knee flexion (SW)	3	43.35 (32.85-48.85)	43.40 (35.60-49.20)	45.45 (37.23-55.45)	0.64
	6	25.60 (22.30-26.20)	47.60 (45.35-51.83)	47.50 (42.05-53.93)	0.99
	9	50.30 (45.15-55.40)	48.65 (39.73-51.10)	52.10 (40.98-57.73)	0.30
Max ankle dorsiflexion (SW)	3	23.85 (21.65-24.23)	23.40 (22.30-25.50)	23.25 (22.43-24.38)	0.81
	6	25.60 (22.30-26.20)	24.65 (23.33-26.50)	24.15 (23.15-25.25)	0.67
	9	25.90 (24.75-27.20)	25.55 (24.65-26.43)	25.85 (24.62-26.35)	0.61

[1]Underwent PFC PCL-sacrificing, [2]Underwent PFC PCL-retaining, [3]Underwent MBK PCL-retaining. * Kruskal Wallis test among the 3, 6 and 9 months in the three groups A, B and C.

Table 5-21 Comparison of the kinetic gait parameters between groups over time 3, 6 and 9 months after TKR. Patients were grouped according to the type of prostheses implanted.

Gait parameter	Period months)	Group A[1] Median (IQR)	Group B[2] Median (IQR)	Group C[3] Median (IQR)	P*-Value
Max Knee moment in mid-stance (N/kg)	3 6 9	0.20 (0.12-0.34) 0.13 (0.09-0.25) 0.19 (0.08-0.29)	0.30 (0.21-0.47) 0.28 (0.18-75.00) 0.35 (0.21-0.46)	0.32 (0.24-0.43) 0.29 (0.14-0.37) 0.22 (0.18-0.32)	0.11 0.06 0.03
Max Knee power in mid-stance (W/kg)	3 6 9	0.03 (0.07-0.05) 0.03 (0.07-0.03) .040 (0.03-0.16)	0.02 (-.09-.08) 0.04 (0.07-0.18) 0.02 (0.16-0.14)	0.03 (0.12-0.08) 0.02 (0.06-0.05) 0.00 (-0.15-0.09)	0.49 0.45 0.61
Max ankle moment in pre-swing (N/kg)	3 6 9	0.64 (0.32-0.77) 0.69 (0.59-0.870) 0.77 (0.66-0.91)	0.64 (0.39-0.78) 0.76 (0.64- 1.00) 0.75- (0.67-0.89)	0.55 (0.42-0.80) 0.66 (0.67-0.95) 0.68 (0.51-0.92)	0.93 0.54 0.65
Max ankle power in pre-swing (W/kg)	3 6 9	2.06 (1.11-3.20) 2.40 (1.66-3.45) 2.34 (1.34-3.70)	1.81 (1.05-2.59) 2.11 (1.55- 3.12) 2.43 (1.30-3.01)	2.21 (1.49-2.98) 2.89 (1.74-3.33) 2.41 (1.69-3.16)	0.43 0.68 0.97

[1]Underwent PFC PCL-sacrificing, [2]Underwent PFC PCL-retaining, [3]Underwent MBK PCL-retaining.
* Kruskal Wallis test among the 3, 6 and 9 months in the three groups A, B and C.

The data analysed in section 5.9 covered all the patients who took part in tests, but did not cover those specific groups of patients who completed all the tests. For example, the data for PFC-PCLS in figure 5.56 cover 15 patients who participated in the baseline tests; only 14 of those patients completed the postoperative (3-months) test, and only 11 and 13 did the 6-months and 9-months tests, respectively. There were instances where, for example, a patient completed the 3-months and 9 month tests, but not the 6-months.

It was therefore decided to plot these data against data for specific groups who had completed all tests. For the checks, four of the most important parameters (walking speed, maximum knee flexion in loading response, maximum knee flexion in mid-swing and maximum knee moment in mid-stance) were plotted in order to determine whether significantly different results might be produced in this way. The data are graphically illustrated in Figures 5.56-5.63. It may be observed that for each pair of boxplots, the parameters measured show similar results between the two sets of data compared.

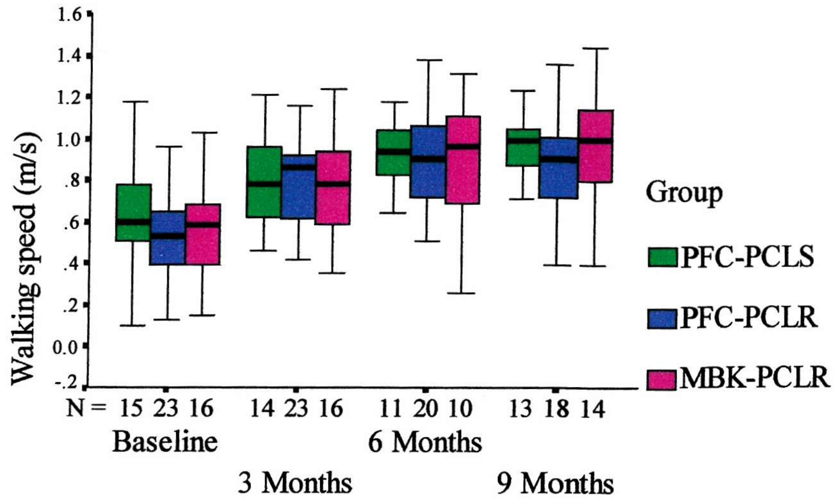

Figure 5-56 Boxplots comparing the differences in walking speed in all operated patients at baseline and 3, 6, and 9 months after PFC-PCLS, PFC-PCLR and MBK-PCLR TKR. The median, IQR and range are shown.

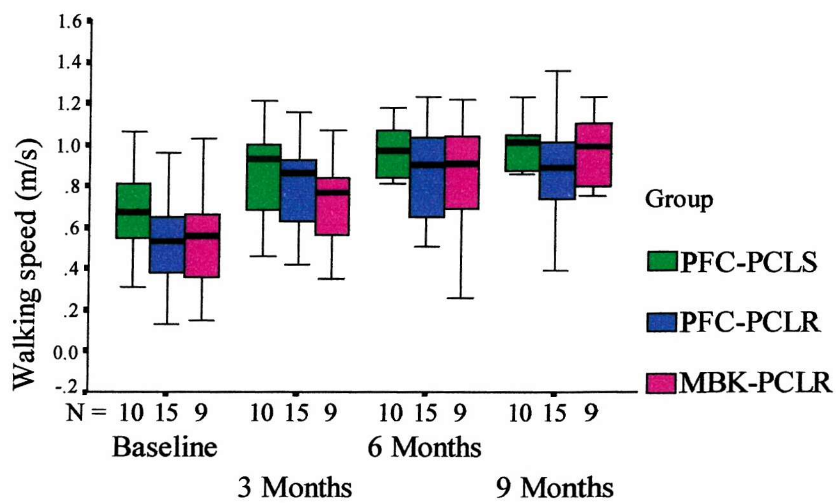

Figure 5-57 Boxplots comparing the differences in walking speed in those operated patients who attended all assessments at baseline and 3, 6, and 9 months after PFC-PCLS, PFC-PCLR and MBK-PCLR TKR. The median, IQR and range are shown.

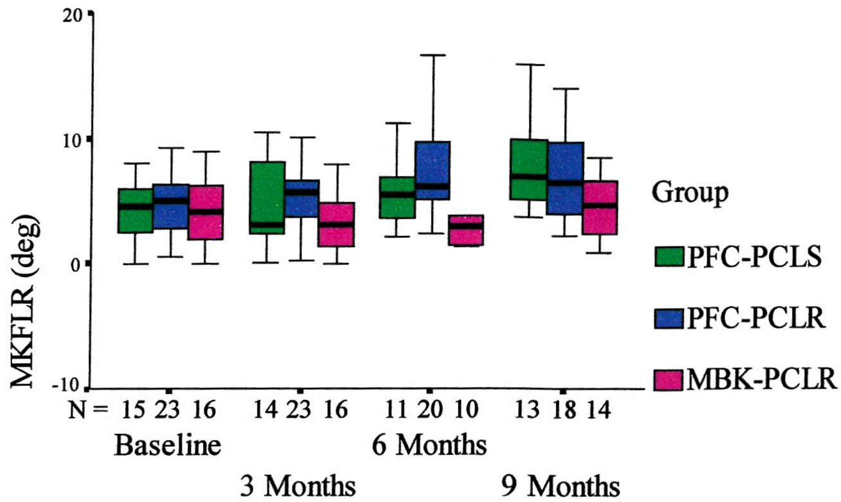

Figure 5-58 Boxplots comparing the differences in the maximum knee flexion in loading response phase (MKFLR) in all operated patients at baseline and 3, 6, and 9 months after PFC-PCLS, PFC-PCLR and MBK-PCLR TKR. The median, IQR and range are shown.

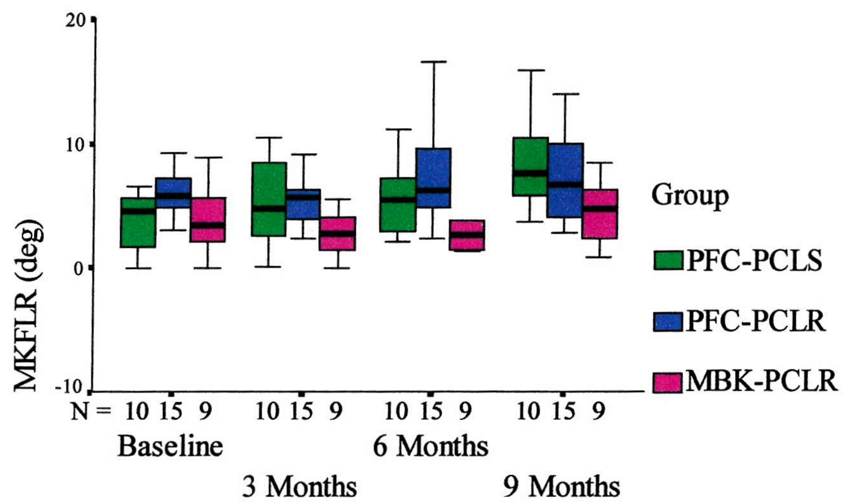

Figure 5-59 Boxplots comparing the differences in the maximum knee flexion in loading response phase (MKFLR) in those operated patients who attended all assessments at baseline and 3, 6, and 9 months after PFC-PCLS, PFC-PCLR and MBK-PCLR TKR. The median, IQR and range are shown.

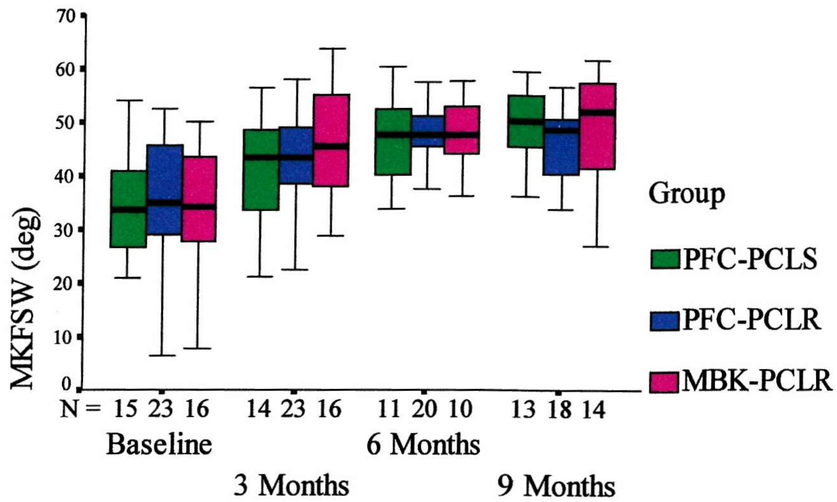

Figure 5-60 Boxplots comparing the differences in the maximum knee flexion in swing (MKFSW) in all operated patients at baseline and 3, 6, and 9 months after PFC-PCLS, PFC-PCLR and MBK-PCLR TKR. The median, IQR and range are shown.

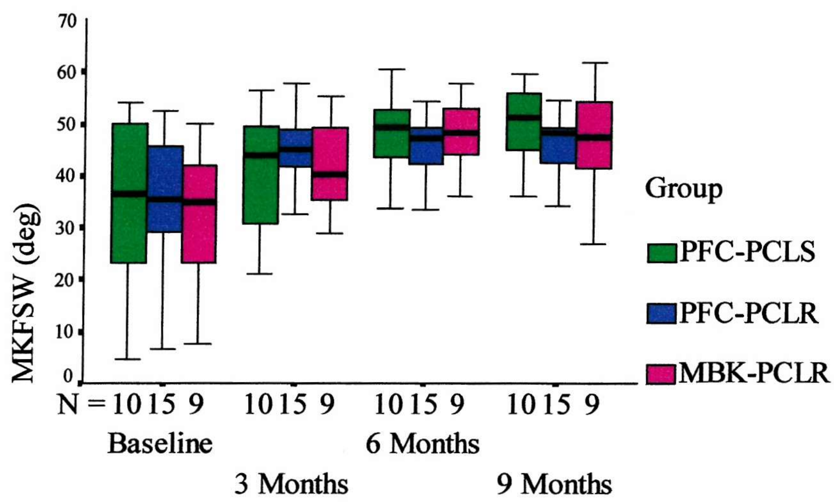

Figure 5-61 Boxplots comparing the differences in the maximum knee flexion in swing (MKFSW) in those operated patients who attended all assessments at baseline and 3, 6, and 9 months after PFC-PCLS, PFC-PCLR and MBK-PCLR TKR. The median, IQR and range are shown.

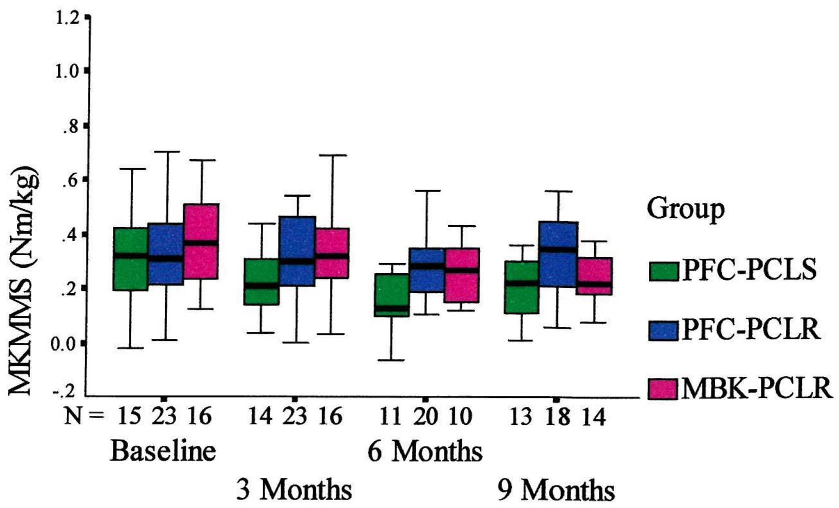

Figure 5-62 Boxplots comparing the differences in the maximum knee moment in mid-stance (MKMMST) in all operated patients at baseline and 3, 6, and 9 months after PFC-PCLS, PFC-PCLR and MBK-PCLR TKR. The median, IQR and range are shown.

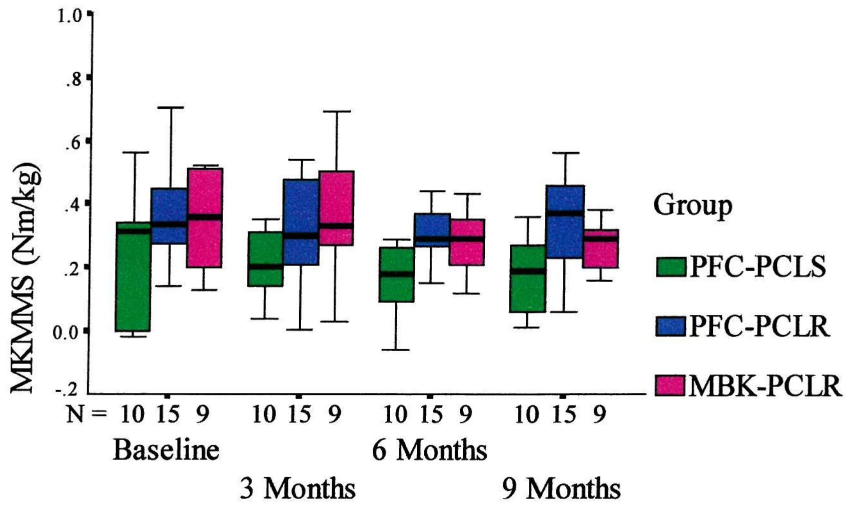

Figure 5-63 Boxplots comparing the differences in the maximum knee moment in mid-stance (MKMMST) in those operated patients who attended all assessments at baseline and 3, 6, and 9 months after PFC-PCLS, PFC-PCLR and MBK-PCLR TKR. The median, IQR and range are shown.

Summary:

The temporospatial, kinematic and kinetic parameters of the three groups with press-fit cruciate ligament sacrificing condylar prostheses (group A), press-fit cruciate ligament retaining condylar prostheses (group B) and mobile bearing knee retaining prostheses (group C) 3, 6 and 9 months after total knee replacement were compared.

A significant improvement, compared to baseline, in walking speed and stride length was observed in all the three groups.

Between group comparisons of the temporospatial, kinetic and kinematic parameters of gait were carried out. There was no statistically significant difference between the groups at 3, 6 and 9 months in the temporospatial, kinetic and kinematic parameters of gait, apart from significant differences in knee moments at mid-stance observed among the three groups at 9 months after surgery.

5.10. A comparison of gait analysis in patients with PCLS (group A) and PCLR (groups B and C) TKR.

To study the effect of retention or removal of the PCL on gait, the results of the two procedures were compared. Groups B and C were combined and compared to group A.

5.10.1. Temporospatial gait parameters.

The temporospatial gait parameters measured during gait analysis for both types of TKR at baseline and 3, 6 and 9 months after surgery are listed in Table 5.22. It can be seen that there were no significant differences in either walking speed or stride length between PCLS and PCLR .

5.10.2. Knee angles, moments and powers during gait

The median knee angle, moment and power values during level walking for both the PCL-sacrificing prosthesis and the PCL-retaining prosthesis are shown in Table 5.23. There were no significant difference between the two groups regarding the maximal knee flexion during the loading response phase or knee flexion during swing ($P > 0.05$) at 3, 6 and 9 months after surgery.

The maximum knee moments at 3, 6 and 9 months were greater in the PCLR group but differences did not reach the level of statistical significance with the PCLS group at 9 months after surgery.

The knee powers during mid-stance were greater for the PCLS group compared to the PCLR group but the differences did not reach the level of statistical significance.

Table 5-22 Median and IQR and level of statistical significance of temporospatial parameters of PCLS and PCLR patients at different follow-up times.

Gait parameters	Period (months)	PCLS (Preop. n = 15) (3 months n = 14) (6 months n = 11) (9 months n = 13)		PCLR (Preop. n = 39) (3 months n = 39) (6 months n = 30) (9 months n = 32)		P[1] value
		Median	IQR	Median	IQR	
Walking speed (m/s)	Preop.	0.60	0.50-0.81	0.56	0.38-0.66	0.32
	3	0.78	0.59-0.97	0.81	0.59-0.92	0.98
	6	0.94	0.81-1.07	0.91	0.68-1.08	0.80
	9	0.99	0.87-1.09	0.92	0.76-1.09	0.46
Stride length (m)	Preop.	0.83	0.68-0.96	0.75	0.63-0.93	0.33
	3	0.97	0.76-1.12	0.95	0.79-1.08	0.79
	6	1.06	0.77-1.15	1.03	0.88-1.15	0.71
	9	1.12	0.98-1.18	1.07	0.84-1.19	0.68

PCLS, Posterior cruciate ligament sacrificing; PCLR, Posterior cruciate ligament retaining. [1]Mann-Whitney U test.

Table 5-23 Median and IQR and level of statistical significance of knee angles, moment and power parameters of PCLS and PCLR patients at different follow-up times.

Gait parameters	Period (months)	PCLS (Preop. n = 15) (3 months n = 14) (6 months n = 11) (9 months n = 13) Median	IQR	PCLR (Preop. n = 39) (3 months n = 39) (6 months n = 30) (9 months n = 32) Median	IQR	P[1] value
Max knee flexion (LR) (degrees)	Preop.	4.60	2.10-6.40	4.80	2.50-6.40	0.81
	3	3.05	2.15-8.17	4.10	2.70-6.50	0.69
	6	5.50	2.90-7.20	5.80	3.23-9.33	0.94
	9	6.90	4.65-10.20	5.30	3.68-7.50	0.14
Max knee flexion (SW) (degrees)	Preop.	33.60	26.33-42.10	34.90	29.00-45.30	0.72
	3	43.35	32.85-48.85	43.40	36.50-50.90	0.60
	6	47.70	36.70-52.60	47.60	44.98-52.58	0.96
	9	50.30	45.15-55.41	48.95	40.65-54.48	0.51
Max knee moment in mid-stance (Nm/kg)	Preop.	0.32	0.14-0.51	0.32	0.20-0.46	0.60
	3	0.20	0.13-0.32	0.32	0.21-0.46	0.04
	6	0.13	0.09-0.26	0.29	0.17-0.35	0.02
	9	0.19	0.09-0.31	0.29	0.19-0.38	0.11
Max knee power in mid-stance (W/kg)	Preop.	0.01	-0.04-0.06	0.03	-0.06-0.04	0.96
	3	0.03	-0.07-0.05	0.01	-0.09-0.08	0.51
	6	0.03	-0.07-0.03	0.01	-0.06-0.12	0.30
	9	0.04	-0.03-0.17	-0.01	-0.16-0.11	0.34

PCLS, Posterior cruciate ligament sacrificing; PCLR, Posterior cruciate ligament retaining. *[1]Mann-Whitney U test.*

5.10.3. Cincinnati Knee Rating Scale (CKRS) and Visual Analogue Scale (VAS) scores in patients with PCLS (group A) and PCLR (groups B and C) TKR

A comparison between the CKRS and Visual Analogue Scale (VAS) scores in patients with PCLS and PCLR groups was carried out.

The results for the PCLS and PCLR groups are shown in Table 5.24. It can be seen that there was no statistically significant difference in CKRS and VAS at

3, 6 and 9 months after surgery between the groups. PCLR group had greater improvement in CKRS at 3 months while PCLS had more at 9 months post surgery. However, the differences were not statistically significant. PCLR patients had better pain relief than PCLS at the 9 months follow-up visit but this was not statistically significant.

The plots show the results for the patients at baseline and at 3, 6 and 9 months follow-up after TKR (A, B, and C) for comparison (Figure 5.64).

Table 5-24 Median and range of CKRS and VAS scores at baseline and the three test periods for the PCLS and PCLR groups with TKR.

Variable	Period (months)	PCLS (3 months n = 14) (6 months n = 11) (9 months n = 13)		PCLR (3 months n = 39) (6 months n = 30) (9 months n = 32)		P - value
		Median	Range	Median	Range	
CKRS	3	49.00	14.00-86.00	59.00	14.00-98.00	0.56
	6	68.00	30.00-84.00	59.00	89.34-0.66	0.98
	9	64.00	27.00-90.00	66.00	6.00-90.00	0.59
VAS (cm)	3	1.60	0.30-5.30	1.60	0.00-7.20	0.85
	6	1.60	0.00-5.60	0.95	0.00-6.80	0.71
	9	0.90	0.00-7.60	0.90	0.00-6.60	0.75

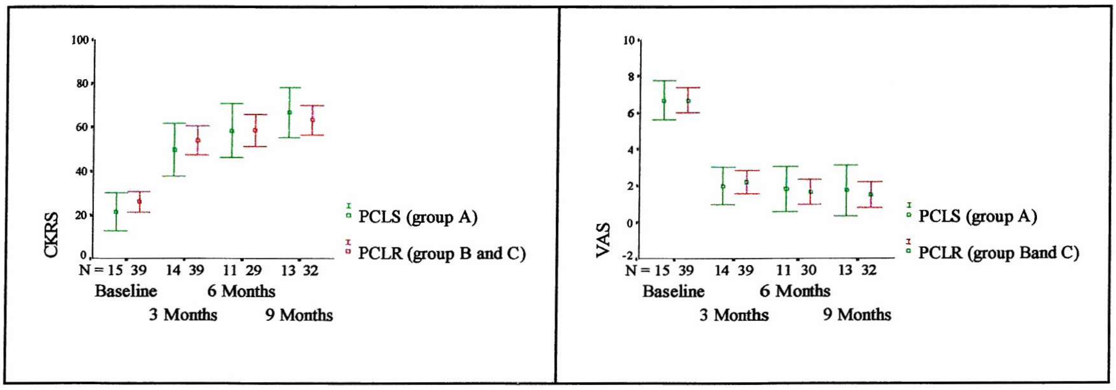

Figure 5-64 Error bar comparing the pre-operative values with those at 3, 6 and 9 months follow-up on the Cincinnati Knee Rating Scale (CKRS) and visual analogue scale (VAS) between the PCLS and PCLR. The median and range are shown.

Summary:

comparisons of the temporospatial, kinetic and kinematic parameters of gait were made between the posterior cruciate ligament sacrificing prostheses (group A) and posterior cruciate ligament retaining prostheses (groups B and C).

There were no significant differences between the PCLS and PCLR TKRs in the walking speed and stride length at 3, 6 and 9 months after surgery.

There were no significant differences found regarding the maximum knee flexion in mid-stance and mid-swing between the two groups. The knee moments in mid-stance were less in those receiving PCL-sacrificing prostheses.

The functional assessment and pain scales demonstrated no significant differences in patients with PCLS (group A) and PCLR (groups B and C) TKR.

5.11. A comparison of gait analysis in patients with PFC-PCLS (group A) and MBK-PCLR (group C) TKR

The previous section compared the results between group A (PCL-sacrificing prostheses) and groups B and C (PCL-retaining prostheses). This section compares the results between two groups of patients (A and C) who were operated on by the same surgeon.

5.11.1. Temporospatial gait parameters.

The temporospatial gait parameters measured during gait analysis for both types of TKR at baseline and 3, 6 and 9 months after surgery are listed in Table 5.25. It can be seen that there were no significant differences in either walking speed or stride length between both PFC-PCLS and MBK-PCLR either at baseline or at 3, 6 and 9 months after surgery.

Similarly, there were no statistically significant differences with respect to the timing of mid-stance in the gait cycle at baseline and 3, 6 and 9 months after surgery between the groups with the PFC-PCLS and MBK-PCLR.

5.11.2. Knee angles, moments and powers during gait

The median knee angle, moment and power values during level walking for both the PCL-sacrificing prosthesis and the PCL-retaining prosthesis are shown in Table 5.26. There were no significant differences between the two groups regarding the maximal knee flexion during the loading response phase at baseline and 3, and 6 months after surgery, but a difference occurred at 9 months. There were no significant differences between the two groups regarding the maximal knee flexion during the swing phase at baseline and 3 and 9 months after surgery.

The maximum knee moments at 3, 6 and 9 months were greater in the MBK-PCLR group and statistically significant differences were noted only at 3 months after surgery. There were no statistically significant differences in the powers during mid-stance between the two groups.

Table 5-25 Median and IQR and level of statistical significance of temporospatial parameters of PFC-PCLS and MBK-PCLR patients at different follow-up times.

		PCLS (Preop. n = 15) (3 months n = 14) (6 months n = 11) (9 months n = 13)		PCLR (Preop. n = 16) (3 months n = 16) (6 months n = 10) (9 months n = 14)		
Gait parameters	Period (months)	Median	IQR	Median	IQR	P[1] value
Walking speed (m/s)	Preop.	0.60	0.50-0.81	0.59	0.39-0.69	0.55
	3	0.78	0.59-0.97	0.78	0.58-0.96	0.95
	6	0.94	0.81-1.07	0.97	0.66-1.14	0.83
	9	0.99	0.87-1.09	0.99	0.79-1.14	0.98
Stride length (m)	Preop.	0.83	0.68-0.96	0.76	0.62-1.00	0.45
	3	0.97	0.76-1.12	1.02	0.75-1.09	0.88
	6	1.06	0.77-1.15	1.08	0.81-1.17	1.00
	9	1.12	0.98-1.18	1.08	0.84-1.21	0.96
Mid-stance (% cycle)	Preop.	32.00	30.52-34.52	33.02	31.09-35.31	0.24
	3	30.76	30.38-31.41	30.92	29.99-33.58	0.62
	6	30.39	29.79-30.76	30.66	30.06-31.96	0.44
	9	30.50	29.65-30.89	30.89	29.98-32.36	0.33

PFC-PCLS, Press-fit condylar posterior cruciate ligament sacrificing; MBK-PCLR, mobile bearing knee posterior cruciate ligament retaining.
[1]Mann-Whitney U test.

Table 5-26 Median and IQR and level of statistical significance of knee angles, moment and power parameters of PFC-PCLS and MBK-PCLR patients at different follow-up times.

Gait parameters	Period (months)	PCLS (Preop. n = 15) (3 months n = 14) (6 months n = 11) (9 months n = 13)		MBK-PCLR (Preop. n = 16) (3 months n = 16) (6 months n = 10) (9 months n = 14)		P[1] value
		Median	IQR	Median	IQR	
Max knee flexion (LR) (degrees)	Preop.	4.60	2.10-6.40	4.10	1.80-6.43	0.98
	3	3.05	2.15-8.17	3.05	1.28-5.15	0.48
	6	5.50	2.90-7.20	2.95	1.45-4.88	0.07
	9	6.90	4.65-10.20	4.65	2.30-6.63	0.05
Max knee flexion (SW) (degrees)	Preop.	33.60	26.33-42.10	34.25	26.78-44.35	0.97
	3	43.35	32.85-48.85	45.45	37.23-55.25	0.43
	6	47.70	36.70-52.60	47.50	42.05-53.93	0.88
	9	50.30	45.15-55.41	52.10	40.98-57.73	0.77
Max knee moment in mid-stance (Nm/kg)	Preop.	0.32	0.14-0.51	0.37	0.22-0.51	0.35
	3	0.20	0.13-0.32	0.32	0.24-0.43	0.05
	6	0.13	0.09-0.26	0.27	0.14-0.37	0.09
	9	0.19	0.09-0.31	0.22	0.18-0.32	0.63
Max knee power in mid-stance (W/kg)	Preop.	0.01	-0.04-0.06	-0.05	-0.12-0.03	0.28
	3	0.03	-0.07-0.05	-0.03	-0.12-0.08	0.98
	6	0.03	-0.07-0.03	-0.02	-0.06-0.05	0.59
	9	0.04	-0.03-0.17	0.00	-0.15-0.09	0.33

PFC-PCLS, Press-fit condylar posterior cruciate ligament sacrificing; MBK-PCLR, mobile bearing knee posterior cruciate ligament retaining. [1]Mann-Whitney U test

5.11.3. Cincinnati Knee Rating Scale (CKRS) and Visual Analogue Scale (VAS) scores in patients with PFC-PCLS (group A) and MBK-PCLR (group C) TKR

A comparison between the CKRS and Visual Analogue Scale (VAS) scores in patients with PFC-PCLS and MBK-PCLR groups was carried out.

The results for the PFC-PCLS and MBK-PCLR groups are shown in Table 5.27. It can be seen that there was no statistically significant difference in CKRS and VAS at 3, 6 and 9 months after surgery between the groups. MBK-

PCLR group had greater improvement in CKRS at 9 months while PFC-PCLS had more at 3 and 6 months post surgery. However, the differences were not statistically significant. MBK-PCLR patients had better pain relief than PFC-PCLS at the 3 and 6 months follow-up visit but this was not statistically significant.

Table 5-27 Median and range of CKRS and VAS scores at baseline and the three test periods for the PFC-PCLS (group A) and MBK-PCLR (group C) with TKR.

Variable	Period (months)	PFC-PCLS (3 months n = 14) (6 months n = 11) (9 months n = 13) Median (range)	MBK-PCLR (3 months n = 16) (6 months n = 10) (9 months n = 14) Median (range)	P - value
CKRS	3	49.00 (14.00-86.00)	62.00 (24.00-86.00)	0.09
	6	68.00 (30.00-84.00)	74.00 (0.66-90.00	0.26
	9	64.00 (27.00-90.00)	73.00 (6.00-90.00)	0.92
VAS (cm)	3	1.60 (0.30-5.30)	1.30 (0.00-6.30)	0.62
	6	1.60 (0.00-5.60)	0.65 (0.00-4.50)	0.09
	9	0.90 (0.00-7.60)	0.45 (0.00-5.00)	0.27

CHAPTER 6

DISCUSSION

The hypothesis to be tested is that posterior stabilized knee prosthesis sacrificing the PCL will result in an advantage over a retaining prosthesis in respect of kinetic (knee moment and power), kinematic (knee flexion in mid-stance and mid-swing) and temporospatial (walking speed, stride length and percentage of stance phase) gait parameters and that this improvement would be reflected in better functional ambulation.

First, the historical development of gait analysis and functional anatomy of the knee were reviewed, to establish the theoretical context of the study.

Pilot studies were carried in order to test and refine the experimental protocol in preparation for the main study. At this stage, the first aim was to determine the repeatability of the CODA mpx30 system. The second aim was to determine the repeatability of the EMG pattern of normal walking. The third aim was to determine whether gait pattern is affected by walking with one arm flexed, or with both arms flexed.

For the main study, the gait patterns of OA patients were analysed one week before TKR, to establish baseline data. A control group of healthy subjects, matched for age, height and body weight was also studied. This allowed the data obtained from normal subjects to be compared with that of OA patients. Patients were re-assessed 3, 6 and 9 months after TKR and gait parameters compared between the three types of prosthesis.

The number of patients who completed the study was sufficient in spite of the difficulties in getting everyone to attend. However, there some patients who completed the 3-month and 9-month tests, but not the 6-months tests.

For example, of the PFC-PCLS TKR (group A) patients, fifteen participated in the baseline tests, but only 14 of these patients completed the postoperative (3-months) tests, and only 11 and 13 did the 6-month and 9-month tests, respectively. The drop-out rate from groups B and C at 6 and 9 months after surgery were similar. These difficulties may have been reduced if all the patients had been paid, both for attendance and the costs of their journey during follow-up.

Great attention was taken in the data collection. During recording, the investigator checked that patients struck the force plate. The investigator excluded any run when he was not sure whether the whole foot had touched the plate successfully. The investigator suggested that data capture would be improved by fitting two or more cameras on different sides to check on the foot position and have simultaneous recordings from both legs.

Different sizes of the pelvic frame were used depending on the patients size and the shank wand was padded when necessary to ensure comfort during walking.

Overall, I believe that the above difficulties did not affect the results of the study or the overall findings.

6.1. Historical review of gait analysis

The study of human gait has grown in recent years, to become one of the most interesting and exciting branches of medicine. As can be seen from the review in section 1.7, several devices for the measurement of gait have been invented. Analysis of human gait has been improved with the introduction of new technologies. The main milestones in the history of gait analysis include the use of an electrogoniometer, motion analysis systems and force plate measurement.

Further advances were made with the use of computing technology, video cameras and improved electromyographic equipment. Instrumental gait analysis has contributed to significant advances in understanding the impact of orthopaedic and neurological conditions on human gait *(Masdeu et al 1997, Polak 1998)*.

These developments have also opened up new arguments in the treatment of osteoarthritis of the knee. Constant updating and innovation in the use of prosthetics in total knee replacement surgery has raised arguments about the use of prostheses as replacements for normal ligaments, and the impact that these have had on human gait. The present study investigated the gait pattern of patients with knee osteoarthritis using a comprehensive three dimensional motion analysis system with integrated force plate and telemetered EMG.

6.2. Complexity of the knee joint and the importance of knee stability during walking

Many studies have confirmed that knee motion is not a result of a simple hinge articulation between the condyles of the femur and those of the tibia with the patella. Knee movements are complex and occur in three separate planes during the complete gait cycle. Knee flexion and extension are accomplished by gliding, rolling and rotation between the femoral and tibial condyles

(Crenshaw 1987). The purpose of this complex biomechanical articulation is to allow efficient transmission of power from the hip to the ankle in a variety of ambulating modes.

The main function of the knee joint is to provide mobility and stability to the lower extremity during level walking. Functional and positional stability of the knee joint are provided by mensici and ligaments respectively. The stability at the knee joint is dependent upon the complex interaction of ligaments and other soft tissues which provide the dynamic forces required for activity around the joint *(Kaplan 1962, Robichon and Romero 1968, Kapandji 1970, Sledge and Walker 1984, Schipplein and Andriacchi 1991)*. The knee joint has maximum stability when it is locked in full extension.

Locking is brought about and maintained by the action of the collateral and cruciate ligaments, the posterior capsule, quadriceps muscle expansion, and the tendons about the joint. Complex gliding, rolling and rotation are also important for normal knee function. Ligaments have two major roles in maintaining normal knee joint kinematics. They act as a dynamic guide to knee motion, and also act as a passive mechanical restraint that prevents the occurrence of abnormal translations when stresses are applied. Ligament tears disrupt these functions and often lead to symptomatic instability *(Warwick and Williams 1989).*

6.3. Pilot study.

At the beginning of the research the investigator recruited a small number of subjects in order to familiarise himself with data collection before the main study. The other aims of the pilot project were to examine the test-retest reliability of the CODA motion analysis system and to assess the effects of walking with one arm flexed, both arms flexed and both arms swinging on gait analysis data collection.

Intra-rater reliability refers to the repeatability of the same measurement on different occasions by one observer *(Rothstein 1985)*. The test was repeated three times in order to ascertain whether the same readings for the functional range of motion at the hip, knee and ankle could be obtained when the test was repeated 3 times.

The ANOVA statistical test was used to determine the variation between three tests carried out on the same day. The results showed a slight variation in the gait parameters between different sessions. This could be due to performance variation between different times. Overall, the intra-rater reliability was quite high and the study confirmed that no significant differences in any gait parameters were measured in normal walking on the three different occasions.

It was difficult to record gait with both arms swinging, due to covering of the pelvic markers during arm swing. Alteration in the gait could occur during walking with one arm swinging. However, the gait pattern of the individual is unlikely to change during the follow-up period if the same conditions are applied during all assessments. The pilot study confirmed that no significant differences occurred in gait when the subject walked with one arm flexed or both arms flexed.

6.4. The baseline data

6.4.1. Subject recruitment

Fifty-five healthy subjects were recruited and assessed. However, as all surgically treated subjects were aged 50 or above, only control subjects of that age group were selected. The control group consisted of 25 subjects and was larger than the cases in other published studies (e.g. *Dorr et al 1988, Bolanos et al 1998)* that had aimed to produce normative data for gait analysis after TKR.

The mean age of the normal subjects was the same as the patients in this study. Although all the healthy subjects were asymptomatic and without physical disorders in either lower limb, mild osteoarthritis could not be completely excluded without further radiographic assessment. However, this was not carried out because it could not be justified on ethical grounds.

6.4.2. Patient group

All the osteoarthritic subjects recruited for the study were attending the out-patient orthopaedic clinic and were aged between 50-85 years. They were seen one week before operation and 3, 6 and 9 months after surgery. The inclusion and exclusion criteria for patient selection restricted the participation of many patients in this study, and difficulties were encountered in finding a sufficient number of patients matching the specific criteria of selection.

The biomechanical gait analysis results are discussed under the following two headings:

- Temporospatial, kinematic and kinetic gait parameters.
- EMG.

6.5. Temporospatial, kinematic and kinetic gait parameters

The findings of this study confirm that patients with osteorthritis of the knee walked with shorter stride length, reduced walking velocity and had a longer duration of the stance phase of the gait cycle when compared to those of control subjects.

The reduction in the average stride length of the patients seemed to correlate with a significant reduction in walking speed. This could have reflected knee pain during walking. The findings of this study confirm previous reports. *Győry et al (1976)* found walking speed and stride length during gait to be lower in arthritic patients than in a control group. Similar observations were made by *Stauffer et al (1977)*. An interesting finding of this study is that mid stance occurred later in the gait cycle in patients compared to control subjects. Consequently, mid swing was also delayed.

The delay in mid-stance suggests that the patients tended to have a longer loading response time in stance. This pattern of prolonged stance and delayed mid swing could be a result of reduced weight transfer during walking, possibly because of knee joint instability.

During walking, the patient group demonstrated a significantly reduced range of motion at the hip and knee. Knee flexion in mid stance and flexion in mid swing are very important from the clinical point of view because they affect weight transfer on the stance limb and foot clearance. In mid stance, the knee is in full extension, so that the loading response is complete (weight bearing), while in mid swing, the leg should be flexed at the knee sufficiently to allow clearance from the ground.

It is interesting to note that the patients had less knee flexion in the stance phase, which interferes with loading on the stance limb and transference of body weight. This may due to an attempt to avoid discomfort and pain caused by movement at the knee.

Kettelkamp et al (1970), Laubenthal et al (1972), Györy et al (1976), Brinkman and Perry (1985) reported that normal walking requires 67°of knee flexion during the swing phase of gait. This was slightly more than the range of motion in our control group. In this study, the means of maximum knee flexion in the mid-stance and mid-swing phase in healthy control subjects were 15 and 61 degrees respectively which is similar to the means of 14 and 60 degrees reported by *Suzuki and Takahama (1979)*.

Knee motion is decreased, as reported by *Kettelkamp et al (1970), Stauffer et al (1977), Brinkman and Perry (1985), Steiner et al (1989), Messier et al (1992), Skinner (1993),* in the presence of arthritic changes. The most significant alteration in knee motion during walking was a decrease in knee ROM in the stance-phase. The findings of these studies are confirmed by the findings of the present study, which show reduction in the knee range of motion during level walking.

A possible explanation for this would be that structural changes in the joint reduced the functional range of motion. This is in agreement with the observation of *Stauffer et al (1977)* who found a correlation between knee flexion in stance and severity of osteoarthritis. The range of motion at the ankle in swing was significantly reduced in patients compared to the control group. A possible explanation for this would be that patients were inhibited by pain from pushing off for propulsion at the beginning of the swing phase. This could also be due to be reduced knee flexion in swing.

The moments and powers generated at the knee joint were measured exactly at mid stance. Patients showed increased moments and reduced powers at the knee joint compared to healthy control subjects. An explanation for the increase in moments particulary in patients suffering from osteoarthritis of the knee is that larger torques are required to achieve transfer of body weight more quickly.

Radin et al (1991) have shown that patients with knee pain have a faster transfer rate of force after initial contact with the ground, presumably to avoid pain. This was not borne out by the findings of this study. Our patients generated excessively high moments at the knee joint without a corresponding generation of force, probably to stabilise the osteoarthritic knee for weight transfer in stance. The increase in moments would require additional muscle contraction in stance by the continuous activation of the rectus femoris observed in this study.

The impairment of proprioception at the knee would not fully explain the observed gait abnormalities in these patients. Although the knee joint position sense was not objectively assessed in this study, none of the patients studied had a clinically detectable neurological deficit. Evidence from a study by *Marks et al (1993)* shows that there is no correlation between the walking ability of patients with severe osteoarthritis of the knee and impairment of the proprioceptive function.

6.6. A pre-operative comparison between the three study groups who received the different types of prosthesis

There were no statistically significant differences between the three OA study groups regarding temporospatial, kinematic and kinetic parameters during level walking. In addition, no significant differences were present in CKRS and VAS in the three groups. These results suggest that any changes in the parameters after surgery were due to the effects of TKR. Before surgery the patients with OA differed from the control subjects in all the gait parameters measured.

6.7. Gait analysis after total knee replacement.

The hypothesis to be tested is that posterior stabilized knee prosthesis sacrificing the PCL will result in an advantage over a retaining prosthesis in respect of kinetic (knee moment and power), kinematic (knee flexion in mid-stance and mid-swing) and temporospatial (walking speed, stride length and percentage of stance phase) gait parameters and that this improvement would be reflected in better functional ambulation.

The findings indicated significant differences between the study group after TKR, and control subjects, with respect to walking speed, stride length and hip, knee and ankle range of motion and knee and ankle moments and powers. The percent times of mid-stance and mid-swing in the gait cycle were similar to the normal value of the control group. Although the gait parameters of patients improved after TKR over the 9 months follow-up, they did not reach the values of healthy control subjects.

The results show that knee moments in patients with TKR did not fall to the normal value, indicated by the lack of full extension during the mid-stance of the gait cycle in any of three TKR groups.

The results also demonstrated no statistically significant differences between the three TKR groups regarding temporospatial, kinematics and kinetics parameters at 3 and 6 months follow-up. At nine months there were significant differences between the treated groups in the maximum knee moment at mid-stance. The knee power in the three groups showed lower than normal levels of power generated at the knee joint, consistent with reduced stride length and lower walking speed.

The results reveal no significant differences between patients who had PCLS (group A) or PCLR (groups B and C) prostheses with regard to walking speed, stride length either pre-operatively or at any points during the post-operative stage. Both walking speed and stride length tended to increase 3, 6 and 9 months after surgery, probably because of the relief of pain. Up to 9 months after surgery, the gait pattern of the patients was characterised by reduced walking velocity, associated with a reduction in the stride length. In our study the knee range of movement during stance and swing phases did not differ between the two designs and this has previously been reported by *Ishii et al (1998)*. The knee kinematics showed reduced knee flexion during loading response and mid-swing.

There are two possible explanations as to why the abnormalities mentioned above persist following TKR. One explanation is that patients develop a pattern of walking during a long pre-operative period, and that this pattern is maintained post-operatively *(Dorr et al 1988, Ishii et al 1998)*. The stiffness that occurs in the knee due to arthritis before TKR tends to consolidate this gait habit *(Chao et al 1980, Andriacchi et al 1982)*.

A second possible explanation may be the decline in joint position sense due to osteoarthritis *(Skinner et al 1984)*. However, this concept was not supported by *Marks et al (1993)* who failed to find a significant relationship between gait abnormalities and the degree of osteoarthritis. The current study did not specifically address itself to this effect of osteoarthritis on joint position sense.

PCL-sacrificing prostheses resulted in lower knee moments than devices in which the PCL is retained for the first 6 months. This could potentially confer an advantage in terms of loosening of the implant from the bones in which it is embedded. Comparison of the power generated in the two groups revealed lower levels of work produced at the knee joint with PCL-retaining prostheses than with PCL-sacrificing prosthesis. The presence of a smaller knee moment in the operated knee at mid-stance, associated with more phasic activity of the rectus femoris at 9 months after surgery, would tend to reduce the load and thus provide more stability for the operated knee.

It is interesting to analyse the work of one surgeon who uses two different prosthetic designs for different patients, as differences in results between PCLR and PCLS prostheses are more unlikely to be due to surgical experience. Generally, each surgeon will keep strictly to the procedure with which he feels most comfortable. However the surgeon who used both types of prosthesis had indicated that knee stability was an important factor in the choice made for individual patients. Despite this no pre-operative differences in knee moment were found between the two groups of patients. This suggests either that in practice, other factors influence the choice of prosthesis, or alternatively, that knee moments do not give an accurate reflection of knee instability as assessed clinically. This matter was not specifically addressed in this thesis and requires further investigation and future studies.

The subjects studied in this study were a relatively active part of the population of patients having knee joint replacement in that they could walk without major difficulties, an entry criterion being the ability to walk along the walkway several times.

Results from the PCLS (group A) and PCLR (group C) implanted by the same surgeon were compared. No significant differences were demonstrated between patients who had PCLS (group A) or PCLR (group C) prostheses with regard to temporospatial gait parameters either pre-operatively or at any stage post-operatively. The knee kinematics showed reduced knee flexion during loading response and mid-swing and increased knee flexion during swing phase in patients implanted with MBK-PCLR.

Although, knee moments in mid-stance were initially less for the patients with PCLS implants, the differences were not statistically significant at 6 and 9 months after surgery.

To date, only two studies by *Dorr et al (1988)* and *Bolanos et al (1998)* have compared the effects of cruciate ligament retaining and sacrificing TKR procedures on gait.

Dorr et al (1988) performed gait analysis and clinical evaluation on 11 patients with bilateral TKR. Each patient had a PCL-retaining procedure on one knee and a PCL-sacrificing operation on the other. Instrumental gait analysis was carried out pre-operatively and at six months and two years after surgery. The authors found differences between the two prosthetic designs on both level walking and stair ascending.

They noted that PCL-sacrificing procedures were associated with greater loading and higher interface forces in stance and speculated that these effects may reduce the durability of the prosthesis. The study demonstrated maximum quadriceps activity during stair climbing in patients with PCL-retaining and -sacrificing designs. Patients with PCL-sacrificing designs required increased use of the soleus muscle while stair ascending, suggesting that patients who had TKR with the PCL-sacrificing designs leaned forward excessively during this activity. Similar findings were reported by *Andriacchi (1988).*

Andriacchi's study indicated that maximum quadriceps activity was noted during stair climbing with PCL-retaining TKR. The PCL-sacrificing prostheses required more muscle activity for stability during stance. In particular, soleus muscle activity was increased during stair descent, suggesting that this muscle attempts to substitute for the PCL. Furthermore, varus and flexion moments were higher when a PCL-sacrificing knee prosthesis was implanted. The author claimed that this may lead to decreased longevity, durability and effectiveness of the prosthesis. The potential importance of including activities such as stair-climbing in such comparison was demonstrated in this study.

In the third study of patients who underwent bilateral TKR, *Bolanos et al (1998)* performed isokinetic strength testing and gait analysis on 14 patients. Patients had a PCL-retaining prosthesis on one side and PCL-sacrificing on the other. Assessments were carried out pre-operatively and on follow-up on average 98 months after surgery. The authors reported no differences in the isokinetic muscle testing parameters for both quadriceps and hamstrings. Furthermore, no significant differences were found in gait parameters and electromyographic muscle activity between the two groups on both level walking and stair climbing.

To the investigator's knowledge, the only study that compared knee motion during free walking speed between PCL-retaining and PCL-sacrificing is the study by *Ishii et al (1998)*. However, this study was limited to examination of the kinematic gait parameters only.

Ishii et al (1998) performed gait analysis on 11 patients with PCL-retaining prostheses and on 9 patients with PCL-stabilised prostheses. Knee motion was measured using an electrogoniometer. The average time of assessment after surgery was 42.6 months for patients with retention of the PCL, and 25.9 for patients with the posterior sacrificing stabilizing articular surface. The authors reported no significant differences regarding peak knee flexion in the stance and swing phase between the two groups. No attempts were made to study the differences in the kinetic parameters and muscle activity during walking.

The present study is the first to use comprehensive gait analysis including study of temporospatial, kinematic and kinetic parameters and electromyographic changes in the patients.

The present study shows that knee range of motion in stance and swing was not affected by the type of prosthesis used. This finding agrees with those in two previous studies (*Dorr et al 1988, Bolanos et al 1998*). By contrast, *Ishii et al (1998)* found that implantation of a PCL-retaining prosthesis increased the range of knee flexion in stance to normal values, probably by increasing femoral rollback.

In the present study patients were not evaluated during stair ascending and descending between PCL-retaining and PCL-sacrificing designs. However, *Becker et al (1991)* showed no statistically significant differences between the PCL-sacrificing and PCL-retaining during stair climbing. Also, *Shoji et al (1991)* showed no statistically significant differences between the PCL-substituting and PCL-retaining during stair climbing.

One strength of the present study is the large number of patients studied compared with other reports in the literature. In addition, because of the fact that most TKR operations were performed by only two surgeons, potential variation in clinical outcomes were reduced. Another strength of this study is that it was possible to measure the outcomes of two different prostheses implanted by the same surgeon, as analysed in section 5.11.

6.8. EMG

Dynamic EMG allows us to examine the patterns of muscle activity during movement. The application of EMG in the assessment of lower extremity muscle function has been studied by several researchers for many years. It has been applied to normal and pathological gait in clinical practice, to determine phasic patterns of activation for individual muscles or muscle groups *(Drillis 1958, Sutherland and Hagy 1972, Murray et al 1984).*

Surface EMG electrodes were used in this study to identify the activity of four lower limb muscles. They were preferred to needle electrodes because of the advantages proposed by *Basmajjian (1967) and Aminoff (1992)* (See Literature Review, Chapter one, page 21).

6.8.1. Repeatability of the EMG recording

Repeatability of the EMG pattern during walking at self selected speed was also checked in the present study by recording the gait of 8 healthy subjects on three different occasions. The results indicated that EMG patterns were fairly consistent for a given subject. Intrer-individual variability could have been to factors such as skin resistance, thickness of subcutaneous fat, and variability in placement of the surface electrodes.

6.8.2. Pattern of lower limb muscle activity during walking in normal subjects

Many researchers have reported intra-individual variability in EMG patterns regarding the timing of peak activity and the magnitude of the EMG signals. *Shiavi et al (1981)* reported intra-individual variability in EMG patterns regarding the timing of peak avtivity and in the magnitude at self selected speed of eight muscles in 25 normal persons between the ages of 20 and 40 years.

The pattern of activity of the rectus femoris, medial hamstring, tibialis anterior and gastrocnemius muscles recorded by surface EMG in 25 healthy subjects walking at their preferred speed demonstrated intra-individual variability. It is likely that the sequence and pattern of muscle activation during walking, particularly during the stance phase, related to intra-individual variability in walking speed in normal subjects. In some subjects the rectus femoris muscle showed gradual increase in amplitude from initial contact to the loading response phase.

The medial hamstring muscle in a few subjects showed prolonged activity from initial contact to mid stance. Variability in the tibialis anterior muscle was greatest at initial contact. The variability in the gastrocnemius muscle was seen mostly at push off.

These findings suggest that the methods of analysis used in this study were comparable in their results to these reported in the literature.

6.8.3. Pattern of muscular activity during walking in the pre-operative OA patients group

The present study confirmed that the sequence of muscle activation in patients suffering from knee osteoarthritis was grossly abnormal. It was characterised by continuous activation of the rectus femoris muscle which probably maintained knee extension in mid stance. The highest level of activity in the medial hamstrings muscle occurred in mid stance when the knee was extending.

The tibialis anterior muscle in the most of the patients was continuously active during the stance and swing phases. The activation of the gastrocnemius was delayed. This may have reflected the fact that the stance phase of gait was longer because subjects walked more slowly than normal. The increase in knee moments would require additional muscle contraction in stance as shown by the continuous activation of the tibialis anterior and the rectus femoris muscles observed in this study.

All these abnormal patterns of muscle activity were present most of the patients studied. It could be deduced that the prolonged activity of the rectus femoris and medial hamstring will restrain the flexion of the knee in the mid stance and consequently will reduce pain due to the increase in generated moments.

6.8.4. Pattern of muscular activity during walking in patients after TKR.

Most of the patients studied had EMG abnormalities before TKR. Nine months after surgery the stiff-knee flexed gait pattern at mid-stance had improved. Increased knee moment and low power generated at mid-stance were still a crucial point in the biomechanics of the replaced knee and altered the normal knee function. That may have led to decrease in the activity of the rectus femoris and medial hamstring at 3 and 6 months after surgery. In this study there were no consistent differences in EMG findings between patients with PCL-retaining and sacrificing prostheses.

Similarly, *Dorr et al (1988)* reported no significant difference in the EMG activity of quadriceps, hamstring and soleus muscles during level walking and stair climbing between PCL-retaining and PCL-sacrificing knees. However, patients with PCL-sacrificing designs required increased use of the soleus muscle while stair ascending, suggesting that a forward lean was used in the PCL-sacrificing designs. This finding is similar to that described by *Andriacchi (1988)*.

Kelman et al (1989) studied a group of patients with unilateral PCL-retaining prostheses. EMG data recorded from the vastus medialis of the quadriceps muscle, medial hamstring, lateral hamstring and gastrocnemius-soleus revealed significantly greater duration of muscle activity on the side of the TKR in all muscle groups tested. The authors suggested that loss of joint proprioception might be a cause, as described by *Skinner et al (1984)*.

Bolanos et al (1998) found no differences in the electromyographic waveforms of the rectus femoris, vastus lateralis, vastus medialis, medial hamstring, lateral hamstring and gastrocnemius muscles between the two prosthetic designs on both level walking and stair climbing.

6.9. Functional assessment

Functional improvement following TKR was assessed in this study with CKRS. The questionnaire measures severity of pain, severity of swelling, degree of giving way, overall physical activity, walking activity, stair activity and jumping or twisting activities. It is simple to use in practice and easy for patients to understand.

Significant improvement in CKRS occurred at 3, 6, and 9 months after surgery in the three different types of prostheses. There were no significant differences in the CRS between posterior cruciate ligament-sacrificed and posterior cruciate ligament-retaining total knee replacement in our study.

Previous investigation by *Shoji et al (1987), Shoji et al (1991) and Becker et al (1991)* reported no significant differences between posterior cruciate ligament excision and posterior cruciate ligament retention knee when evaluated by the Hospital for Special Surgery Knee Score Scale. Similarly, *Dorr et al (1988)* reported no difference between PCL-retaining and PCL-sacrificing knees in functional results when evaluated by the Hospital for Special Surgery Knee Score Scale.

We are aware of other study by *Hirsch et al (1994)* that found no significant differences among the three groups of total knee replacements in which PCL was sacrificed, retained or substituted, in functional results based on the Hospital for Special Surgery Knee Score Scale.

6.10. Effects of TKR on pain and function in the two different designs

The functional assessment and pain scales demonstrated no significant differences in between patients with PCLS and PCLR TKR.

The effect of TKR on pain should be interpreted with caution. This is because some patients continued to take analgesic drugs after surgery. It would have been unethical to discontinue the patients medication. To reduce the effect of analgesic drugs the investigator instructed all patients not to take any analgesics for a minimum of three hours before the gait recording. Most of the gait pre and post-operative assessment were carried out in the morning, so the time between the last dose of analgesic and the recording was in many cases much greater than 3 hours. Nevertheless, the investigator was aware that some of analgesic drugs may act for more than twelve hours.

6.11. CONCLUSIONS

The hypothesis to be tested is that a posterior stabilized knee prosthesis sacrificing the PCL will result in an advantage over a retaining prosthesis in respect of kinetic (knee moment and power), kinematic (knee flexion in mid-stance and mid-swing) and temporospatial (walking speed, stride length and percentage of stance phase) gait parameters and that this improvement would be reflected in better functional ambulation.

The knee moment was considered an important measure because reduction in the rotational force acting in the knee joint would be expected to reduce symptoms and wear and tear on the prosthesis.

The following are important points arising from the study:

1. Patients with osteoarthritis of the knee adopted a gait pattern which was significantly different from that of the control group. The temporospatial and kinematic gait abnormalities observed in patients with osteoarthritis of the knee were consistent with instability of the knee joint.

2. Increased joint moments (which are mainly maintained by the continuous contraction of the rectus femoris muscle throughout the stance phase of the gait cycle) may have helped to stabilise the knee joint.

3. Instrumental gait analysis, including the study of kinetic, kinematic and dynamic EMG, provided the clinician with detailed information that helped to illuminate the underlying causes of gait abnormalities in patients with osteoarthritis of the knee joint and changes after TKR.

4. PCL- sacrificing prostheses are similar in their effects to PCL- retaining designs regarding temporospatial and kinematic gait parameters at 9 months.

5. On the level of walking, the mechanical posterior cruciate ligament seems an adequate substitute for the biological stability provided by the natural posterior cruciate ligament.

6. PCL-sacrificing prostheses resulted in lower knee moments than devices in which the PCL was retained. This could potentially be an advantage because it may prolong the life span of the prosthesis.

It is clear that TKR is a good procedure. Patients experience considerable relief from their symptoms, pain being either reduced or eliminated. There is definite improvement in ROM which may take up to 9 months to reach its maximum. Gait improvement is reported by the patients and can clearly be demonstrated by objective gait analysis.

6. 12. Suggestions for future research

Although the equipment used in this research was satisfactory, there ways in which researchers undertaking similar work in future could benefit from additional equipment if available.

1. In my opinion the use of one CODA system and one force plate proved adequate for the purposes of the current study. However, it would have been helpful to record the gait parameters bilaterally, using two CODA and two force plates.

2. The use of a portable oxygen consumption measurement system (e.g. Cosmed K2, Italy) for measuring energy consumption directly may have avoided the problem caused by the medication being taken by many of the patients.

Apart from considerations about the equipment and methods used, there are other ways in which research of this kind can be improved. This study involved patients walking on a flat surface over a short distance in an indoor sheltered environment. This kind of study could be extended in various ways to cover a wider range of real-life situations and biomechanical stresses. Future studies could show how post-operative OA patients perform on a less even surface, how they perform in an outdoor environment, and whether performance is different under the burden of stair-climbing or when carrying heavy objects (such as a shopping bag). Further comparisons could then be made between the various prostheses.

Future research might ideally employ a prospective randomised controlled trial using different types of prosthesis in collaborative centres in the U.K. A successful trial could be difficult and expensive, requiring the cooperation and collaboration of many surgeons. It might be continued for several years in order to assess whether reduction in knee moments are indeed associated with better control of symptoms and better survival of the prosthesis.

Finally, further work is needed to assess the effects of the level of the expertise of the surgeon in selecting patients for surgery and in performing the joint replacement. The value of physiotherapy for improving outcomes of prosthetic surgery also requires controlled prospective evaluation.

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Appendix 1

Our Ref : TEW/11/WGP/MN

9 April 1997

Mr Al-Zahrani
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Please take note of paragraph 4

Dear Mr Al-Zahrani

Submission Number (66/97) - Kinetic and kinamtic analysis of lower limb function and gait parameters and their relationship to functional ambulation following total knee replacement

The Joint Ethics Committee considered the above application at its recent meeting. I am pleased to inform you that ethical approval has been given to this study subject to the consent form and patient information sheet being put on to locally headed paper. The Committee also requested that the wording relating your research project to degree requirements be deleted from the consent form and patient information sheet.

Would you please ensure that a record is made in the Medical Records of patients who agree to participate in a research project, to the effect that they have given their consent to involvement in this research study. The title of the research project should be clearly indicated. Please note that this only applies to patients who are willing to be involved in the study and does not apply to those who do not wish to participate.

Should any unforeseen problem of either an ethical or procedural nature arise during the course of this research where you feel that the Joint Ethics Committee may be of assistance, please do not hesitate to contact me.

Since the introduction of the NHS R&D initiative, and in particular the Culyer report, it is now necessary for the R&D Directorate at Southampton General Hospital to be notified of all research projects before commencement of the research including all ethically approved projects.

Yours sincerely

Dr. T. E. Woodcock
Honorary Secretary
Joint Ethics Committee

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Appendix 2

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Our Ref: JMH/KAH

3 November 1997

Dr K AL-Zahrani
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Victoria House
SGH

Dear Dr AL-Zahrani

Thank you for your letter regarding lower limb function and gait parameters following total knee replacement.

This was discussed at a recent Directorate Meeting and it was agreed that the Directorate would have no objection to this taking place.

I would only add that no patient in the Unit should be expected to be involved in more than one research project.

Yours sincerely

Mr J M Harley
Clinical Services Director

Appendix 3

PATIENTS' INFORMATION SHEET AND CONSENT FORM

I am a student at the University of Southampton Rehabilitation Research Unit. I am writing to ask you to take part in this research study. The aim of the study is to see how total knee replacement improves walking. If you take part in the study you will be required to have a simple test before the operation and 3, 6, and 9 months after the knee replacement. During the test, which will be carried out in the Gait Laboratory at Southampton General Hospital, a machine will record your walking speed and how your leg muscles are working. The test takes about one hour to complete, but you can take rest if needed. I will also complete some further clinical outcome tests with you. If you wish to withdraw from the study at any stage you will be free to do so, without giving a reason and this will not affect your treatment in any way. You are free to ask any questions to which answers will be fully provided.

Thank you for your help

Yours sincerely,

Khaled Al-Zahrani

I...have read and understood the above and I agree to take part in the above study.

Signature...   Date..

Home address...

....

Appendix 4

Pain Assessment

Name Date

Age Sex: F / M

1- Pain is in:-

Left knee ( ) Right knee ( )

2- Duration of pain is:- days month years

3- Intensity of pain:-

(Please mark on the line below for your pain intensity level)

No pain Pain as bad it could be

Appendix 5

Cincinnati Knee Rating System

1- Pain

20.....	No pain, normal knee, performs 100%
16.....	Occasional pain with strenuous sports or heavy work; knee not entirely normal; some limitation, but minor and tolerable.
12.....	Occasional pain with light recreational sports or moderate work activities; frequently brought on by vigorous activities, running, heavy labor, strenuous sports.
8.....	Pain, usually brought on by sports, light recreational activities or moderate work. Occasionally occurs with walking, standing, or light work.
4.....	Pain is a significant problem with activities as simple as walking. Relieved by rest. Unable to do sports.
0.....	Pain present all time; occurs with walking, standing, and at nighttime. Not relieved with rest.

Intensity of pain: -----Mild. -----Moderate. -----Severe.

Frequency of pain: -----Intermittent. -----Constant.

Location of pain:

Medial (inner side).
Lateral (outer side).
Anterior (front).
Posterior (back).
Diffuse (all over).

Type of pain:

......Sharp.
......Aching.
......Kneeling.
......Throbbing.
......Burning.

2- Swelling.

10.....	No swelling, normal knee, 100% activity.
8.....	Occasional swelling with strenuous sports or heavy work. Some limitation but minor and tolerable.
6.....	Occasional swelling with light recreational sports or moderate work activities; frequently brought on by vigorous activities, running, heavy labor, strenuous sports.
4.....	Swelling limits sports and moderate work. Occurs infrequently with simple walking activities or light work (about three times/year).
2.....	Swelling brought on by simple walking activities and light work. Relieved with rest.
0.....	Severe problem all of the time, with simple walking activities.

Appendix 5 (cont.)

3- Giving-way

20..... No giving-way, normal knee, performs 100%
16..... Occasional giving-way with strenuous sports or heavy work. Can participate in all sports, but some guarding or limitation are still present.
12..... Occasional giving-way with light recreational activities or moderate work. Able to compensate; limits vigorous activities, sport or heavy work; not able to twist suddenly.
8..... Giving-way limits sports and moderate work; occurs infrequently with walking or light work (about three times/years).
4..... Giving-way with simple walking activities and light work. Occurs once a month. Requires guarding.
0..... Severe problem with simple walking activities; cannot turn or twist while walking without giving way.

4- Other symptoms (unrecorded)

Knee stiffness	Kneecap grinding	Knee locking
None	None	None
Occasional	Mild	Occasional
Frequent	Moderate	Frequent
	Severe	

4-Overall activity level:

20..... No limitation; normal knee; able to do everything including strenuous sports or heavy labour.
16..... Perform sports, including vigorous activities, but at a lower performance level, involves guarding or some limits to heavy labour.
12..... Light recreational activities possible with rare symptoms; more strenuous activities cause problems. Active but in different sports; limited to moderate work.
8..... No sports or recreational activities possible. Walking activities possible with rare symptoms; limited to light work.
4..... Walking, activities of daily living cause moderate symptoms; frequent limitation.
0..... Walking, activities of daily living cause severe problems; persistent symptoms.

5- Walking:
10..... Normal unlimited.
8..... Slight/mild problem.
6..... Moderate problem: smooth surface possible up to 800 meters.
4..... Severe problem: only two to three blocks possible.
2..... Severe problem: requires cane, crutches.

5- Stairs:
10..... Normal unlimited.
8..... Slight/mild problem.
6..... Moderate problem: only 10-15 steps possible.
4..... Severe problem: requires banister, support.
2..... Severe problem: only one to five steps possible.

Appendix 5 (cont.)

5- Running

5.....	Normal unlimited: fully competitive, strenuous.
4.....	Slight/mild problem: run half-speed.
3.....	Moderate problem: only two to four km possible.
2.....	Severe problem: only one to two blocks possible.
1.....	Severe problem: only one to five steps.

5- Jumping or twisting activities

5.....	Normal unlimited: fully competitive, strenuous.
4.....	Slight/mild problem: some guarding, but sports possible.
3.....	Moderate problem: give up strenuous sports; recreational sports possible.
2.....	Severe problem: affects all sports, must constantly guard.
1.....	Severe problem: only light activity possible (golf, swimming).

Appendix 6

Statistical tests used in this study

1. Student t-test
2. Analysis of variance (ANOVA)
3. Mann-Whitney U test
4. Kruskal Wallis test

Student t-test (independent t-test)

This test compares the mean of two independent variables. In this thesis, this test was used to see if there is a significant difference in spatiotemporal, kinematic and kinetic parameters when the recording was performed with one, or both arms flexed during walking at the preferred speed. The data used for this type of test is required to have a normal distribution curve.

Analysis of variance (ANOVA)

Parametric test where data are required to have a normal distribution curve. One way analysis of variance is a method for comparing more than two groups of observations. In this thesis, the ANOVA was used to see if there is a significant variation between the three occasions of recordings, using CODA with respect to the kinematic (range of motion of the hip, knee and ankle joints) parameters.

Mann-Whitney U test

Non-parametric test where data are not required to have a normal distribution curve. It is a method for comparing the median of two independent variables. In this thesis, this test was used to see if there is a significant difference in certain variables in patients with PCLS and PCLR groups at 3, 6 and 9 months after surgery.

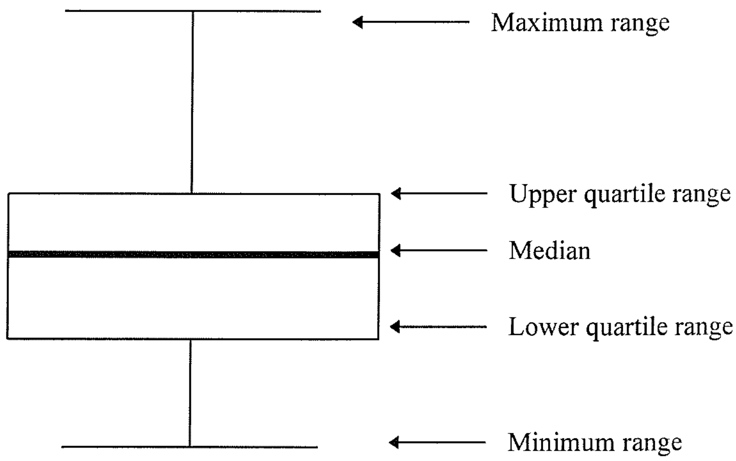
Appendix 6 (cont.)

Kruskal Wallis test

A method for comparing more than two groups. It is non-parametric test where data are not required to have a normal distribution curve. In this thesis, this test was used to see if there is a significant variation between the three groups of patients with OA of the knee before TKR. It was also used to see if there is a significant variation between the three groups of TKR at 3, 6 and 9 months after surgery with regard to spatiotemporal, knee angles, moments and powers gait parameters and Cincinnati Knee Rating Scale and Visual Analogue Scale in patients with PCLS and PCLR groups. In this thesis, means and standard deviations were used to enable comparison with other studies.

Box- and whisker plot

A method of describing the distribution of a quantitative variables graphically. The central line is the median, the inter-quartile range is shown by the box and the actual range displayed by the whiskers.



Ambulation	Walking
Anthropometry	Refers to physical measurements of the human body, i.e. age , height, weight and joint width measurement.
Kinematic	Refers to the study of body motion but without consideration to the forces that cause the motion.
Kinetics	Refers to the study of the relationship between body motion and the forces which act on the body.
Centre of Gravity (COG)	'COG of the body refers to a point in the body at which the entire weight of the body is more concentrated' (Smidt, 1990).
Coronal plane	The imaginary plane which divides the body into front and back portions.
Sagittal plane	The imaginary plane which divides the body into symmetrical right and left halves.
Spatial	Distance
Temporal	Timing
Transverse plane	The imaginary plane in which the body is divided into upper and lower sections.

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