THE UNIVERSITY OF CALGARY

Gait Changes Following Total Hip Replacement

by

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A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES

IN PARTIAL FULFILLMENT OF THE REQUIREMENTS FOR THE

DEGREE OF MASTER OF SCIENCE

DEPARTMENT OF MEDICAL SCIENCE

CALGARY, ALBERTA

JUNE, 1997

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0-612-24641-8

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Abstract

To investigate the gait dynamics and hip muscle recruitment patterns associated with pre- and post-operative total hip replacement (THR), a gait analysis was performed on THR patients prior to surgery, and at four and eight months after surgery. Asymmetries between the affected and contralateral limbs were significant for ground reaction forces and hip kinematics at all time points. Patients used less hip extension moment, not hip abduction moment, possibly to reduce hip compressive loads. Cane use decreased affected hip abduction moment and ground reaction forces, but had no significant effect on lateral torso sway or hip kinematics. Cane use may have increased the abduction moment in the good hip, potentially accelerating disease. At four months post-operative, tensor fascia lata activity was longer, potentially as compensation for a surgically damaged gluteus medius. Results of this study provide quantitative data about the post-operative progress of THR patients with regard to gait dysfunction.

Acknowledgments

There are many people who have made a difference to my masters program, and I would like to specifically name a few. First, I would like to acknowledge Dr. Ron Zernicke, my supervisor, for training me in effective research and experimentation. Whenever I had a problem or question regarding my research, Dr. Zernicke was available and eager to help in any way possible. I wish to thank him for entrusting me with the set up and management of the gait lab. This has given me a unique and valuable opportunity for training.

The team of physiotherapists, Deb Thon, Peter Clare and Lauri Kaul, have been instrumental throughout the project from its very beginnings. I would like to acknowledge each of them for their expertise and efforts to ensure the relevancy of this research to the field of medicine.

This project was computer intensive, and would not have been possible without the support of Byron Tory, who was generous with his time and expertise, and was accessible at all times.

I would like to thank Dr. Tak Fung for his contribution to the statistical aspects of this project. His enthusiasm for statistics and his dedication to sharing this is inspiring.

I would like to especially acknowledge my wife, Grace, for her tireless, loving encouragement and support. The hours she spent editing the numerous drafts and working with me for the successful completion of this thesis have been invaluable.

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Chapter 1

Background

1.1 Introduction

Total Hip Replacement (THR) is usually an elective surgical procedure with the primary indication being pain relief [Olsson et al., 1985]. Secondary to pain relief, but still very important, is a patient's desire to improve his or her physical function and quality of life. In particular, patients may elect for THR to allow continued participation in activities such as walking and golfing. The patient's long term expectations for participation in these activities are linked to the outcome of their rehabilitation.

Total hip replacement is performed in patients with significant pain or functional disability of the hip for whom less invasive procedures have not been successful or are not medically feasible. More than half (54%) of all THRs are to treat osteoarthritis, with the remainder of diagnoses including treatment of fractures (18.3%), implants requiring revision (13.7%), aseptic necrosis and other bone disorders (6.9%), and rheumatoid arthritis (2.2%) [Praemer et al., 1992]. Total hip replacement is generally done in the over 65 year old population (Fig. 1). Over 200,000 hip replacements are performed annually in the U.S., with 123,000 being THRs. Allowed charges for THR totaled over $190 million to the U.S. Medicare in 1989.

In this study, we were interested specifically in which factors contribute to an abnormal gait pattern frequently observed in post-operative THR patients. Improved understanding of this abnormal gait pattern may have substantial impact on the post-operative management of THR patients with regard to physiotherapy treatments targeted at correcting gait dysfunction.

Long-term improvement in symptoms and function are expected in most patients, but mechanical or biological failure leading to revision surgery is still common. Conventional THRs are not likely to have a favorable outcome after 15 years [Yamamuro et al., 1990]. By 12 years post-operative, 20% of cemented and non-cemented THRs need revision [Cheal et al., 1992; Schurman et al., 1989]. Age and obesity are the most significant factors leading to poor long term outcome. Patients over 80 years old reported no loosening in follow-ups for up to five years [Levy, 1995], but the revision rate for patients less than 45 years old has been reported to be greater than 35% for cemented prostheses and 12% for non-cemented, press-fit prostheses after five to ten years following surgery [Mont et al., 1993]. These failure rates are substantial considering the duration that a young patient will expect to use the prosthesis. Revision surgery is possible, but the rate of success diminishes with each revision [Booth et al., 1988; Dee et al., 1989]. Overloading of the prosthesis through impact activity and obesity has been proposed as the cause of failure in young and obese patients [Dubs et al., 1983; Kilgus et at., 1991, Mont et al., 1993; Ritter and Meding, 1987; White, 1992]. Mont et al.[33] refuted these findings. They reported that revision surgery, obesity, or level of activity was not significantly correlated with failure rates or functional grading of the hips. Low numbers of subjects may have accounted for their finding [Mont et al., 1993].

In the weeks following THR surgery there is generally a marked decrease in pain following the initial soft tissue healing [Olsson et al., 1985]. As pain decreases, patients may become increasingly aware of physical limitations, such as decreased cardiovascular endurance, and the effect this has on overall function. Also, at this point (generally eight to twelve weeks post-operative), many patients and family members express dissatisfaction with the aesthetic quality of gait and the inability to ambulate long distances. After surgery, patients often continue to demonstrate a "lurching" gait pattern, or limp, characterized by excessive lateral trunk movement over the affected hip. The abnormal pattern is often associated with hip abductor muscle weakness, but many patients demonstrate this pattern despite adequate "strength" of these muscles. Many studies have examined the survivorship of THR, but few have examined the quality of life after surgery [Edworthy et al., 1993]. Literature describing the mechanics of the hip joint of the THR patient is rare, and no one has examined the precise pattern of gait involved with the lurch.

1.2 Gait

1.2.1 Mechanics of the limp

The stance phase of human gait places large compressive forces on the hip joint. About 5/6 of body weight lies above the level of the hip during the stance phase of gait, the other 1/6 representing the stance limb [Delp et al., 1996], but compressive forces at the hip have been estimated to be greater than 2.5 times body weight (BW) during gait [Bergmann et al., 1993; Brand and Crowninshield, 1980; Crowninshield et al., 1978;

Davy et al., 1988]. The activity of muscles crossing the hip to maintain posture during single limb support accounts for the majority of the compressive forces at the hip [MacKinnon and Winter, 1993; Neumann and Cook, 1985; Neuman et al., 1992; Neumann and Hase, 1994]. In the static model of the hip during single limb stance (Fig. 2), the force of gravity acting on the center of mass of the body creates an axial load at the hip joint and a moment about the hip. This moment is opposed by the force of the hip abductor muscles that also create a compressive force at the hip. The abductor force is much larger than the weight of the body since the moment arm of the muscles is much smaller than the moment arm of the body weight vector. Hip contact force may be even larger due to co-contraction of the hip adductor muscles and the hip abductors, or due to the role of the muscles in correcting medial and lateral imbalances during gait [MacKinnon and Winter, 1993].

Patients with painful hips attempt to avoid the high compressive forces seen at the hip during gait. To reduce the compressive force of the abductor muscles, many patients lurch over their painful hip during gait. This places the center of mass more over the hip to reduce the moment arm of the body weight about the hip. This in turn reduces the force requirement of the abductor muscles, and therefore the compressive load on the hip joint. Pelvic stability is maintained mechanically, and abductor muscle force can be reduced [Blout, 1956]. Lurching gait requires little activation of the hip abductor muscles, and these muscles subsequently grow weak. Conversely, patients with weak hip abductor muscles or an improper muscle recruitment may be required to walk with a limp or other compensatory gait to maintain stability during each single limb support on the affected side. The Trendelenburg test measures the weakness in hip abductor muscles that may result from a lurching gait. During this test, the patient is asked to stand on one leg, and weakness is present if the pelvis on the contralateral side drops [Echternach, 1990]. It is estimated that a decrease of 42% in hip abductor strength in a normal hip will result in an inability to generate the moment about the hip to maintain a level hip [Delp et al., 1996]. Olsson et al. [1985] reported that 38% to 44% of pre-operative THR patients in his study displayed a positive Trendelenburg test, and 92% of the patients displayed a moderate or severe limp. Mont et al. [1993] reported a moderate or severe limp in 73% of pre-operative THR patients. Both studies show that at least a slight limp is present in more than 95% of patients [Mont et al., 1993; Olsson et al., 1985].

After THR surgery, many patients continue to walk with a limp and often display residual weakness in the abductor muscles. Olsson et al. [1985] reported a moderate or severe limp in 35% and 30% of patients at six months and twelve months post-operatively. By five years post-surgery, Mont et al. [1993] reported that 16% of patients had a mild limp and 9% had a moderate limp. Murray et al. [1981] presented data on the degree of limp as a function of the lateral head motion. Significant decreases in lateral head motion were noticed after THR, especially within the first six months, but by two years the average lateral head motion was greater than for the normal population. Significant abductor muscle weakness was present in this population, however, it is unclear as to whether this weakness resulted in a limp, or if a persistent limp allowed for the abductor muscles to grow weak.

1.2.2 Asymmetry of gait

In pre-operative patients with unilateral disability, the degree of asymmetry, often related to the degree of lurch, is the most notable abnormality of gait. Pre-operatively, there is a 5% to 17% difference in single limb support times between affected and unaffected limbs [Long et al., 1993; Murray et al., 1981; Olsson et al., 1985]. This difference continues to decrease after surgery, and by one year after surgery, there is only a 1% to 8% difference. The asymmetry in step length decreases from 5% to 10% before surgery to nearly symmetrical by two years post-operative [Murray et al., 1981].

Gait asymmetry is reflected in the amount of loading on the affected and unaffected limbs. Pre-operatively, the affected limb experiences a 6% to 8% smaller maximum vertical ground reaction force (GRF) during gait than the unaffected limb [Long et al., 1993; Olsson et al., 1985]. Olsson et al. [1985] and Long et al. [1993] reported pre-operative vertical GRFs of 0.96 BW and 1.07 BW on the affected limb. Although patients place higher loads on their affected limb after THR surgery, GRF loading asymmetry does not return to normal after surgery. The difference in weightbearing between the affected and unaffected limbs during quiet standing remains to be 4% to 8% at one year after surgery [Long et al., 1993; Olsson et al, 1985], and 8% to 12% at two years after surgery [Long et al., 1993; Murray et al., 1981].

Before surgery, weight acceptance time, defined as the time to the first peak of the vertical component of the ground reaction force, was much longer for the affected limb than the unaffected limb. Weight acceptance time for the involved limb of the pre-operative THR patients was 21% of the gait cycle, decreasing to 17% by one year post-operative [Olsson et al., 1985], whereas normal subjects have full weight acceptance within the first 10% of the gait cycle [Perry, 1982]. Therefore, THR patients guard their hip by loading their limb more slowly. During quiet standing during there is a difference in vertical ground reaction force of 28% between affected and unaffected limbs [Murray et al., 1981].

1.2.3 Joint loading

Weight bearing pain in THR patients has been shown to correlate with the vertical GRF [Olsson et al., 1985], but no study has presented correlations between pain at the hip and hip joint loading. Hip joint loading during normal gait has been estimated using optimization methods and measured using instrumented prostheses. Using minimum total muscle stress as the criterion, Crowninshield et al. [1978] calculated peak hip joint contact forces that ranged from 3.3 BW to 5 BW during normal gait. Brand and Crowninshield [1980] later used this technique to determine hip joint loading in THR patients. Estimated hip contact forces for free walking speed were 3.4 BW pre-operatively and 3.7 BW at one year post-operatively [Brand and Crowninshield, 1980].

Direct measurement of hip joint loading requires the use of invasive techniques, therefore human studies are limited [Bergmann et al., 1993; Davy et al., 1988]. Davy et al. [1988] and Bergmann et al. [1993] studied hip joint forces after THR using an instrumented prosthesis. At one month post-surgery, Davy et al. [1988] reported a hip contact force of 2.1 BW during static single limb stance, and between 2.6 BW and 2.8

BW during gait. Bergmann et al. [1993] followed two patients with instrumented prostheses for 18 and 30 months respectively, and reported resultant hip joint forces during gait to range from 2.8 BW at a speed of 0.28 m/s to 4.8 BW at a speed of 1.39 m/s.

Joint loading is related, in part, to the activity of muscles crossing the hip joint. Electromyography (EMG) is a tool used to evaluate muscle activity by amplifying the electrical impulses generated during muscular contraction [Winter, 1979]. Timing and relative intensity of muscle activity can be measured, but this activity cannot be equated to muscle force [Bogey et al., 1992; Perry, 1992; Winter, 1979]. Recruitment of muscles about the hip is normally phasic, with abductor muscles being active for the loading response and most of the midstance of gait [Bogey, 1992; Perry, 1992]. In a qualitative report by Long et al. [1993], abnormal EMG patterns during gait were reported pre-operatively in 8 out of 18 patients. These patients included five patients with continuous firing of several of the muscles that cross the hip, and three patients who lacked gluteus medius and upper gluteus maximus activity. The three patients who lacked abductor activity had an associated Trendelenburg gait and decreased hip extension. By two years post-surgery, all abnormal EMG patterns had returned to normal, however, four patients with normal pre-operative EMG developed abnormal patterns post-operatively.

1.2.4 Time and distance parameters of gait

Walking speed is significantly slower in THR patients than in normal subjects [Andriacchi et al., 1977]. Pre-operative hip patients walk at a free speed of 0.45 m/s to 0.60 m/s [Brand and Crowninshield, 1980; McBeath et al., 1980; Murray et al., 1981;

Olsson et al., 1985] and a maximum walking speed of 0.93 m/s to 1.03 m/s [Mattson et al., 1990; Olsson et al., 1985]. Free walking speed increases by 32% to 55% from before THR surgery to six months after surgery [McBeath et al., 1980; Murray et al., 1981; Olsson et al., 1985], and it increases a further 7% to 10% between six and twelve months [Mont et al., 1993; Perry, 1992]. One year after surgery, free walking speed (0.77 m/s to 0.98 m/s) [Brand and Crowninshield, 1980; McBeath et al., 1980; Olsson et al., 1985] remains much lower than the normal adult walking speed of 1.33 m/s. Long et al. [1993] reported that walking speed returned to 94% of normal after one year and to 100% by two years, but normal walking speed was not defined in their study. Maximum walking speed for THR patients one year post-operatively reaches 1.17 m/s to 1.33 m/s [Mattson et al., 1990; Olsson et al., 1985]. Walking speed equals the product of step length and cadence. Studies have shown that both step length and cadence increase with an increase in walking speed [Andriacchi et al., 1977; Crowninshield et al., 1978]. Reports indicate that cadence increases with time after surgery from 95% of normal pre-operatively to 100% of normal by two years post-operatively [Long et al., 1993; Murray et al., 1981]. Step length increases from 41 cm to 52 cm by one year post-operatively [Olsson et al., 1985].

Step length can be influenced by the range of motion in the hip. Total hip replacement patients experience large increases in the range of motion of the hip following surgery [Mont et al., 1993; Murray et al., 1981]. Mont et al. [33] reported an increase in Harris hip score for range of motion from 2.8 out of 5 points pre-operatively to 4.3 points five years post-operatively. Long et al. [28] described the functional range of motion during gait to increase from 27° pre-operatively to 37° by one year post-operatively, which is considered to be normal by the authors [Long et al., 1993; Perry, 1992].

1.2.5 Energy of gait

Oxygen consumption is used as a measure of the energy expended during walking in THR patients. Despite large increases in walking speed, McBeath et al. [1980] reported no changes in oxygen consumption per minute in patients from before surgery to four years after surgery. This resulted in a decrease in oxygen consumption per meter, or an increased walking efficiency, from 0.500 ml/kg/m pre-operatively to 0.174 ml/kg/m at four years post-operatively. Increases in energy efficiency were confirmed by Mattson et al. [1990] using a fixed walking speed for up to a year after surgery. It is believed that the increased efficiency may be the result of changes in gait patterns which reduce the energy cost.

1.2.6 Summary

Total hip replacement patients walk distinctly differently from normal elderly patients with normal gait patterns. Their abnormal gait patterns are likely to be, in part, mechanisms to reduce pain during gait. Their walk is typically characterized by a limp, which is thought to decrease pain by reducing the load on the hip. They walk at a slower walking speed which has also been shown to decrease limb loading. Their asymmetrical gait further reduces limb loading on the affected side. However, their antalgic gait is also associated with a higher energy cost and a limited walking distance. Post-operatively, patients report a reduced amount of pain in the affected hip, and although their gait is closer to normal, they often continue to walk with significant gait abnormalities.

1.3 Cane use

1.3.1 Cane use in rehabilitation

The use of a cane can improve gait when there is weakness or pain at the hip [Brand and Crowninshield, 1980; Ely and Smidt, 1977; Joyce and Kirby, 1991]. Pre-operatively, walking aids are necessary for many patients. Mont et al. [1993] reported that 50% of the pre-operative patients in their study used one or two walking aids most of the time, and 18% used a walker or could not walk [Mont et al., 1993]. Other research consistently reports at least one half of THR patients use a cane during gait pre-operatively [Mattson et al., 1990; Murray et al., 1981; Olsson et al., 1985]. An additional 5% to 15% used two canes or crutches [Mattson et al., 1990; Murray et al., 1981; Olsson et al., 1985], and Murray reported 3% of the patients receiving THR with a trochanteric osteotomy used a walker during gait pre-operatively [Murray et al., 1981]. McBeath et al. [1980] reported that only 29% of the patients in their study could ambulate pre-operatively without the use of an aid.

Historically, patients and health-care workers have been reluctant to prescribe or use a cane, in part because of the stigma attached to using an assistive device. Studies using assistive devices as an outcome measure often neglect to address the quality of gait with and without the use of these devices, and a simple decrease in use of these devices is considered a better outcome. However, the use of a cane can reduce joint loading and possibly compensate for muscle weakness [Brand and Crowninshield, 1980; Ely and Smidt, 1977]. Cane use may improve the quality of gait post-operatively and could be considered as a tool for the rehabilitation of THR patients.

1.3.2 Ipsilateral versus contralateral cane use

Patients with hip or knee instability often prefer to use the cane on the ipsilateral side to the affected limb, and, thus, to use the cane as a splint and for weightbearing [Edwards, 1986]. Total hip replacement patients, generally, use the cane on the contralateral side to the affected hip [Dean and Ross, 1993; Edwards, 1986; Ely and Smidt, 1977; Joyce and Kirby, 1991; McConnel, 1991]. This is recommended since a cane in the contralateral hand encourages a normal reciprocal gait pattern, and it reduces the hip joint forces more effectively than an ipsilateral cane [Edwards, 1986; Joyce and Kirby, 1991; McConnel, 1991]. A cane reduces the force at the hip joint by supporting the weight of the torso, and more importantly, by reducing the force requirement of the abductor muscles by providing a stabilizing moment about the hip [Edwards, 1986; Ely and Smidt, 1977; Joyce and Kirby, 1991]. During single limb stance, the upward force of the cane on the contralateral arm causes a moment that opposes the moment caused by the body weight (Fig. 2). It is recommended that THR patients exert a maximum of 0.25 BW on the cane. These small forces significantly reduce the hip joint loading due to the large moment arm of the contralateral cane about the hip [Edwards, 1986; Ely and Smidt, 1977].

1.3.3 Limb loading

Many researchers have reported decreased loading on the limb during cane-assisted gait. Ely and Smidt [1977] reported that for patients who limp due to a hip disability, 0.15 BW applied to a cane in the contralateral hand reduces the maximal vertical GRF from 1.00 BW to 0.89 BW. While speed remained the same for gait with or without a cane, cane-assisted gait had a longer stride length and a lower cadence. Opila et al. [1987] presented impulse data from GRFs of THR patients 11 to 15 days after surgery. Although these patients had a unilateral disability, they walked using two points of aid. The aids supported weight at all times during gait, thereby decreasing the vertical GRF on both limbs and assisting in restraint and propulsion of the body. Edwards [1977] compared GRFs for the affected limb during ipsilateral and contralateral cane use. Free walking speed increased from 0.47 m/s for ipsilateral cane use to 0.55 m/s for contralateral cane use. At these speeds, cane force of 0.31 BW to 0.32 BW corresponded to maximum vertical GRFs on the affected limb of 1.03 BW and 0.99 BW for ipsilateral and contralateral cane use respectively. Edwards' study [Edwards, 1986] supports the use of contralateral rather than ipsilateral cane use except when reduced hip motion is desired.

The effect of a contralateral cane on limb loading is most noticeable at the hip, due to the reduction in hip abductor muscle activity. Brand and Crowninshield [1980] reported a 56% decrease in hip contact force during gait with the use of a cane in the contralateral hand as compared to without a cane in pre-operative THR patients.

However, patients in their study walked slower with a cane than without a cane (0.43 m/s with, and 0.57 m/s without), which may account for some of the difference. Kleissen et al. [1989] demonstrated the decrease in hip abductor muscle activity with the use of a cane. He reported a 25% decrease in gluteus medius muscle activity with the use of a cane as compared to without the use of a cane.

Neumann and colleagues [Neumann and Cook, 1985; Neumann et al., 1992; Neumann and Hase, 1994] have presented data concerning the EMG activity of hip abductors during gait while carrying loads in various positions. During single limb stance, a hand-held load has the opposite effect at the hip as a cane. A load held contralaterally would be added to the body weight and would shift the body weight vector away from the affected hip, thereby increasing the abductor muscle activity required to stabilize the pelvis. A load of 30% BW held in the contralateral hand causes a 243% increase in abductor muscle EMG magnitude as compared to walking without a load [Neumann and Hase, 1994]. An ipsilateral load, on the other hand, would cause a moment about the hip that opposes the body weight and results in less activation of the abductor muscles. Loads of 30% BW held ipsilaterally do not cause any increase in the abductor muscle activity as compared to walking without a load [Neumann and Hase, 1994].

1.3.4 Summary

Canes are used in post-operative rehabilitation to reduce the force on the prosthetic hip during healing by reducing the force requirements of abductor muscles crossing the hip, thereby compensating for weak abductor muscles. Canes used in the contralateral hand may prevent lurching gait by reducing weightbearing pain and assisting weak abductor muscles. Canes have been shown to reduce limb loading, hip muscle requirements, and hip contact forces. Post-operatively, if patients are encouraged to walk without a cane before adequate strength has been regained by the hip muscles, they may return to the pre-operative lurching pattern that had developed as a means of stabilizing the pelvis.

1.4 Objective

The objective of this study was to examine the relation between a lurching gait pattern, the use of a cane, and the recruitment of lower extremity muscles. The primary goal was to identify the gait dynamics and hip muscle recruitment patterns associated with pre- and post-operative THR.

1.5 Hypothesis

Patients will have an abnormal gait pattern prior to THR characterized by a significant asymmetry in their gait parameters (i.e., joint kinematics, joint moments, ground reaction forces, and muscle recruitment patterns). These asymmetries will persist for at least eight months post-operatively, and use of a cane will decrease the degree of asymmetry.

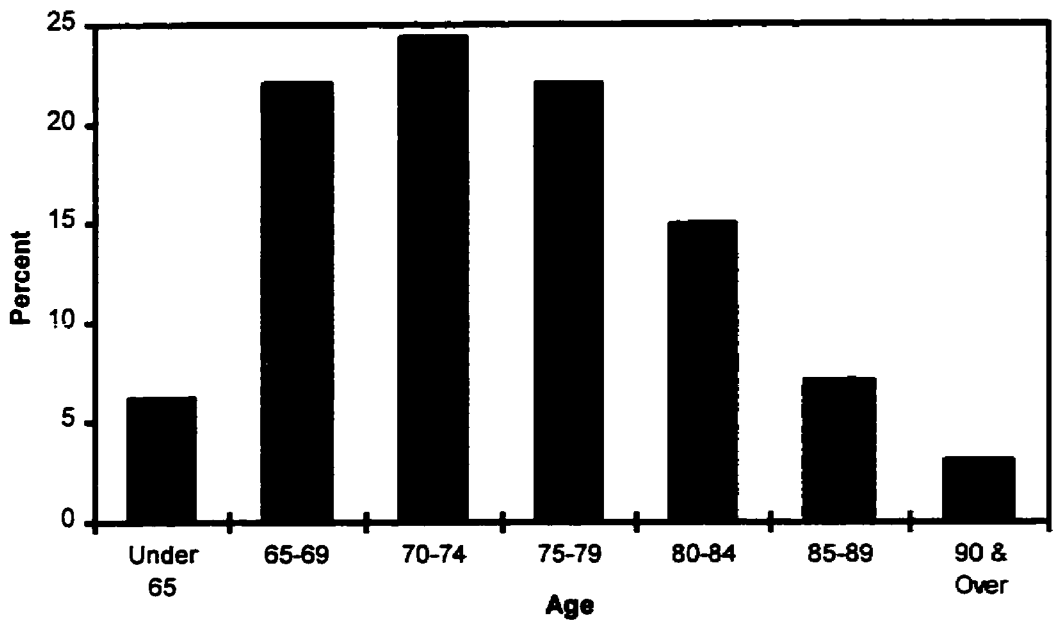


Fig. 1: Percent distribution of total hip replacement by age (U.S. data) [Praemer, 1992].

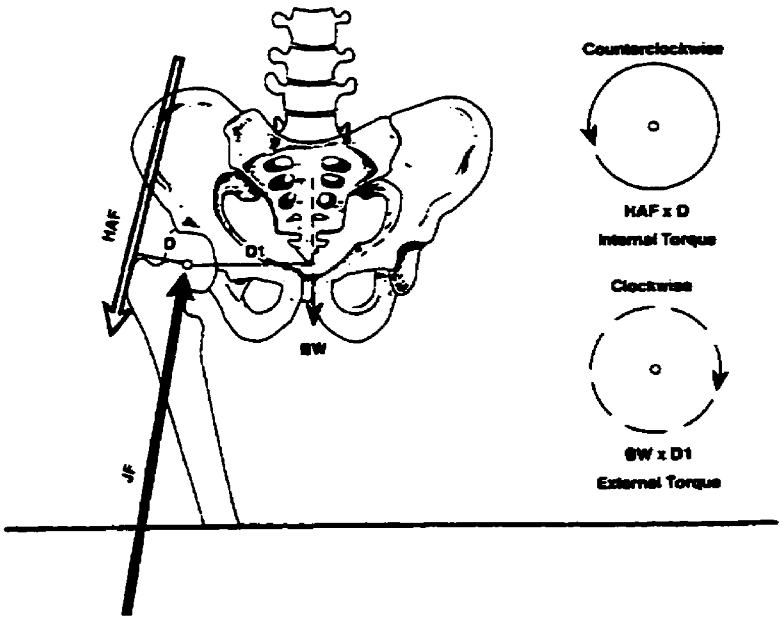


Fig. 2. Right hip abductor force (HAF) stabilizing the pelvis and trunk about the hip during single limb support. The internal torque produced by the HA muscle about moment arm D must be equal but opposite to the external torque produced by the body weight (BW) about moment arm D1. The resultant joint force (JF) equals the sum of the vectors HAF and BW [Neumann and Hase, 1994].

Chapter 2

Methods

2.1 Patients

Fourteen patients scheduled to undergo total hip replacement between November, 1994 and March, 1996, were recruited through the orthopaedic services at three adult medical centers in Calgary: the Foothills Hospital, the Calgary General Hospital, and the Rockyview Hospital. Patients were recruited by physiotherapists or nurses at the pre-operative clinics at these hospitals. Patients were introduced to the goals of the study by the health care professional and given a written description of their responsibility in the study. When the patients had the opportunity to review the written material, they were asked to sign an informed consent form (Appendix), which was approved by the University of Calgary Conjoint Medical Ethics Committee. Patients' participation was voluntary with reimbursement for mileage and parking for each testing session. At no time was the patient's pre-operative or post-operative care influenced in any way by the decision to participate in or to be excluded from the study.

The patients recruited for this study included only those scheduled for unilateral primary THR as a treatment for osteoarthritis of the hip. The osteoarthritis was diagnosed as uniarticular, and patients lived in or near Calgary. The following exclusion criteria were used for this project: previous surgery to either hip; recent trauma (e.g., fall, accident) precipitating the THR; pain or gait abnormality involving the non-operative lower extremity, the knee, or ankle of the operative limb; surgical complication; pulmonary or cardiac disease that presented potential risk to the patient during data collection; pre-existing neuromuscular disease that might manifest itself as lack of balance, absent or diminished strength in the trunk or extremities, absent or diminished sensation in the lower extremities; or an inability to ambulate prior to THR.

2.2 Study design

In this longitudinal study, THR patients were examined on three occasions and each patient served as their control. The first testing session occurred within three weeks prior to surgery, the second and third testing sessions occurred at four months and eight months after surgery respectively. During each testing session, a clinical assessment of the hip and a gait analysis was performed.

2.3 Clinical assessments

At each testing session, all participants had a standardized clinical assessment carried out by a physiotherapist. The assessment included range of motion and muscle testing, as well as an interview. Static range of motion of the hip was measured with a goniometer accurate to within 1° (Fred Sammons Inc., Brookfield, IL) using standard measuring procedures [Heck et al., 1965]. Strength of the hip joint musculature was measured using a conventional manual muscle testing scale of 0 to 5 (Table 1) [Clarkson and Gilewich, 1989; Daniels and Worthingham, 1986; Magee, 1992]. This method was chosen for its speed and ease of collection. Patients were interviewed using the Calgary Total Hip Assessment questionnaire [Thon et al., 1992], which specifies degree of pain and disability of the hip in daily living. Each patient's height and weight were measured while the patients wore their shoes, all clothing, and the belt pouch used for the EMG data collection during the gait analysis. Kinetic data were expressed as per unit body weight.

2.4 Gait analysis

Gait characteristics were recorded while patients walked the length of a 6 m walkway. They were asked to walk with their personal walking shoes on and at a self-selected speed. Three types of data were collected synchronously during all walking trials: high-speed video of subject movement, ground reaction forces, and EMG. To collect data for the patients' affected and contralateral limbs, patients walked with their right limb towards the video system for the first six trials during each testing session, and then walked for six more trials with their left limb facing the cameras. For each direction, patients walked without a cane and with a cane for three trials each. A standardized cane (Guardian Sunrise Medical, Arleta, CA) was supplied and adjusted so that the hand grip was level with the wrist joint when the patient was standing with his or her arms fully extended. Patients were required to use the cane in the contralateral hand to the affected hip, and to walk while moving the cane and the affected limb at the same time. Patients wore a T-shirt and shorts which exposed the greater trochanter during walking trials.

2.4.1 Kinematics

Three dimensional motions of the body segments were recorded using a two camera high-speed video digitizing system (Motion Analysis Corp., Santa Rosa, California) which captured the images of reflective markers placed on the patient. Cameras were placed in the anterior lateral direction to the patient when the patient was standing on the walkway, with a camera separation angle of approximately 45°. Camera positions were chosen to achieve the highest separation angle while maintaining adequate marker views from both cameras. Images from both cameras were synchronized and digitized at a frame rate of 200 Hz. The laboratory coordinate system was defined as a right handed system having the x axis in the direction of walking, the z axis pointing vertically, and the y axis orthogonal to the x and z axes, using a calibration frame with eight isotropically distributed markers.

A right handed coordinate system for each of the thigh, shank, and foot segments was defined having the origin at the proximal joint center, the x axis pointing inferiorly, the y axis pointing anteriorly, and the z axis pointing to the right. For the foot, inferior was defined as perpendicular to the sole of the foot, and anterior was defined as along the long axis of the foot. On each segment, including the torso, three non-collinear markers were taped in specific anatomical locations (Table 2) for recording the three-dimensional movement. Six markers were placed on either the front or lateral side of the segment at a known distance distally from the joint center. A trial was collected with the subject standing in the anatomical position in line with the walkway, and a computer algorithm (Matlab, The Mathworks Inc., Natwick, Mass.) calculated the lower limb marker locations with respect to the joint coordinate system (JCS) axes which is a pre-defined axis system for each segment [Grood and Suntay, 1983]. Markers on the limb were taped to the skin over bony landmarks, when possible, to avoid excessive marker movement.

Markers were only placed on the limb closest to the cameras and were transferred when the second limb was recorded. Joint centers were estimated using the location of bony landmarks. The ankle and knee joint centers were estimated to be the two maleoli and femoral epicondyles respectively. The hip joint center was estimated to be the midpoint between the anterior superior iliac spine and the symphysis pubis, deep to the greater trochanter.

The two video files for each trial were tracked using the Expert Vision software (Motion Analysis Corp.). This process used a direct linear transformation (DLT) to determine the three dimensional positions of each of the markers in each frame of data. Gaps in tracked marker data were estimated by linearly interpolating between measured points. No mathematical smoothing was performed during the tracking process.

2.4.2 Ground reaction forces

Ground reaction forces (GRF) were measured using a piezoelectric force plate (Kistler Instruments Corp., Amherst, N.Y.) embedded in the center of the walkway with the major axis of the force plate lining up with the direction of walking. The raw force plate signal was amplified, then digitized (EGAA, RC Electronics Inc., Santa Barbara, CA) at 1 kHz and stored on a personal computer (Model 6387, IBM, Armonk, NY). Trials were accepted only if the patient landed with only one foot entirely on the force plate, and the cane did not come in contact with the plate.

2.4.3 Electromyography

Surface electromyography (EMG) recorded the electrical activity from the following eight muscles: bilateral gluteus medii, bilateral tensor fascia latae, bilateral erector spinae, and unilateral rectus femoris and lateral hamstrings on the operative limb. Indwelling electrodes are required to measure EMG in small, deep muscles (Basmajian and De Luca, 1985), but the muscles we examined were superficial and large. Therefore, surface electrodes were used for this study since they reliably acquired signals from the muscles studied, and caused less discomfort and risk for patients. Surface electrodes have also been shown to produce more repeatable results (Kabada et al., 1985) and do not affect gait as much as indwelling electrodes (Young et al., 1989).

Prior to electrode placement, the skin was prepared by shaving and then cleaning with isopropyl alcohol. The EMG signal was pre-amplified 100 times at a distance of 4 cm from the electrodes prior to being sent to the signal multiplexer on a belt pouch worn by the patient (Bortec Electronics Inc., Calgary, AB). The multiplexed signal was transmitted over a 11 m cable to the amplifier where it was amplified to a 5 volt range. The amplified signal was digitized at a frequency of 1 kHz (EGAA, RC Electronics) and stored on a personal computer (IBM). Prior to further data analysis, EMG data high-pass filtered at a cutoff frequency of 25 Hz to remove movement artifacts.

2.5 Data analysis

2.5.1 Kinematics

Tracked video data, raw GRF data, and raw EMG data were analyzed using the Kintrak biomechanical analysis software (Motion Analysis Corp.) on a Sparc station (IPC, Sun Microsystems, Mountain View, CA).

Lower limb joint angles were calculated as JCS angles [Grood and Suntay, 1983], where joint angles described an ordered rotation about a selected Cartesian coordinate system described in terms of three independent angles. For the hip joint, the proximal segment was represented by the room coordinate system and the distal segment was represented by the thigh segment. Therefore, hip joint angles represented movement of the thigh segment relative to the horizontal and vertical axes of the room rather than with respect to the torso. With this method, resultant angles were similar to the clinical description of joint angles. All joint angles were filtered using a zero-lag four-pass low pass Butterworth filter at a cutoff frequency of 10 Hz. Walking speed was determined by calculating the average velocity of the greater trochanter marker in the direction of walking, over the full duration of the trial (Kintrak).

Lateral torso sway was calculated using a computer algorithm (Matlab). Positions of the torso markers were smoothed by a zero-lag four-pass low pass Butterworth filter at a cutoff frequency of 10 Hz using the Kintrak software. A line was calculated between the mid-sternum marker and the midpoint between the right and left pelvis markers. This line was projected on the y-z plane. A vertical line was considered to be a torso sway angle of 0°.

2.5.2 Ground reaction forces

Vertical, anterior-posterior, and medial-lateral force vectors were calculated from the raw force channels and filtered using a zero-lag four-pass low pass Butterworth filter at a cutoff frequency of 200 Hz. All forces reported in this study were normalized to body weight (BW).

2.5.3 Joint moments

Joint moments were calculated using an inverse dynamics model of the lower limb [Bresler and Frankel, 1950]. This model assumed that the lower limb was a set of three rigid ellipsoids linked by freely rotating joints. Input into the model included the magnitude and direction of the GRF, positions, velocities, and accelerations of the body segments, locations of the joint centers of the body segments, and estimates of the mass and inertial properties of the segments. Estimates of the locations of the centers of masses and the relative masses of the segments were based on data from Clauser et al. [1969], and moments of inertia were based on data from Dempster [1955]. Force data were filtered at 25 Hz, and video data were filtered at 5 Hz prior to joint moment calculations. Joint moments were normalized to body weight and reported as internal joint moments.

2.5.4 Muscle activity

Although we collected EMG from eight muscles, we only analyzed the gluteus medius and tensor fascia lata muscle activities. Erector spinae, quadriceps and hamstring muscles were not directly related to the hip and therefore their analysis goes beyond the scope of this paper.

Electromyography (EMG) was filtered using a zero-lag four-pass low pass Butterworth filter at a cutoff frequency of 200 Hz. Data were rectified and then smoothed using a zero-lag four-pass low pass Butterworth filter at a cutoff frequency of 25 Hz. Time normalized, filtered EMG data were averaged across the three trials for each condition using the Matlab software (Matlab, The Mathworks Inc., Natwick, MA). EMG onset and duration were calculated using Matlab. EMG was considered "on" when its amplitude was greater than 15% of the maximal filtered EMG measured during the testing session. To eliminate isolated spikes of EMG activity, the time to cross threshold was recorded only if the activity remained on or off for more than 10% of stance [Bogey et al., 1992]. The total duration for which the muscle was considered "on" during the stance phase of gait was calculated. To determine differences in the timing of the muscles, the stance phase was divided into time intervals (bins) equaling 10% of stance. If the muscle was "on" during any portion of each bin, the bin was assigned a value indicating "on." In this fashion, each of the ten bins during the stance phase as well as the bin immediately prior to foot-strike, would be considered either "on" or "off."

2.6 Statistical analyses

Clinical ranges of motion at the three testing sessions and for both limbs were compared using a two-way repeated measures analysis of variance (ANOVA), with session and limb being the two main effects (BMDP Statistical Software Inc., Los

Angeles, CA). Simple effects were tested for variables with a significant interaction effect, and contrasts determined differences between individual time points where there was a session effect.

Kinematic, GRF, and joint moment data were averaged for the three trials within each condition. The average of these three trials for each subject was used in all statistical tests.

Comparisons of trials without a cane were made using a two-way repeated measures analysis of variance for each gait variable (BMDP Statistical Software). The two main effects in this model were the time before or after surgery (session), and the affected or contralateral limb. Significant session effects were subsequently analyzed using contrasts to determine where the differences were between time points. Significant interaction effects were further analyzed using simple effects. Contrasts were again performed if the session effect was significant within the simple effects model.

To determine the effect of speed on the gait variables in this study, ANOVA tests were repeated with speed as a covariate (ANCOVA) (BMDP Statistical Software). This two-way repeated measures design was similar to the ANOVA design, with session and limb as main effects. Simple effects and contrasts were used to determine interaction effects and differences between each time point. Means for each variable at each condition were adjusted for the speed of walking and reported as adjusted means.

To assess the effect of the use of a cane on the gait variables, a three-way repeated measures ANOVA was performed with session, limb, and cane as main effects (BMDP Statistical Software). Contrasts were performed to determine where differences lay within the session main effect, and simple effects were performed for significant interaction effects.

All differences for the analysis of variance tests were significant if the α level was greater than 0.05. All error bars presented in the figures and errors reported in the text were standard errors of the mean.

Table 1: Manual muscle test grading scale*

Grade	Movement
5	Complete range of motion against gravity with maximal resistance
5-	Complete range of motion against gravity with near maximal resistance
4+	Complete range of motion against gravity with strong resistance
4	Complete range of motion against gravity with moderate resistance
4-	Greater than one-half the available range of motion against gravity with moderate resistance
3+	Less than one-half the available range of motion against gravity with moderate resistance
3	Complete range of motion against gravity
3-	Greater than one-half the available range of motion against gravity
2+	Less than one-half the available range of motion against gravity
2	Complete range of motion with gravity eliminated
2-	Greater than one-half the available range of motion with gravity eliminated
1+	Less than one-half the available range of motion with gravity eliminated
1	Palpable or observable muscle contraction but no joint motion
0	No palpable or observable muscle contraction

* Clarkson and Gilewich, 1989; Magee, 1992.

Table 2: Anatomical positions for reflective markers.

Name	Anatomical Position
5th Metatarsal	Superior to 5th metatarsal head
Navicular	Mid-sagittal at height of lateral maleolus
Calcaneus	Lateral to calcaneus inferior to lateral maleolus
Distal Shank	Lateral to distal fibula
Proximal Shank	Lateral to proximal fibula
Tibial Tuberosity	Mid-sagittal on tibial tuberosity
Lateral Epicondyle	On primary axis of rotation of knee joint
Anterior Thigh	Anterior thigh at half limb length
Greater Trochanter	50 mm inferior to greater trochanter
Right Pelvis	Right anterior superior iliac spine
Left Pelvis	Left anterior superior iliac spine
Mid-Sternum	Sternum at level of 2nd rib

Chapter 3

Clinical examination results

3.1 Patients

Fourteen patients volunteered for this study and were tested pre-operatively. By four months post-operatively, two patients had withdrawn from the study, and by eight months post-operatively, a third patient had withdrawn. Therefore, 11 patients completed the eight month protocol, of which 9 were male and 2 were female (Table 3). The mean age of all 14 patients was 63.7 years (46 years to 74 years).

3.2 Range of motion

Clinical assessment revealed a significant increase in the affected hip range of motion after surgery. The average clinical hip flexion range of motion at the three testing sessions is illustrated in Fig. 3. At all time points, the affected hip had a significantly smaller flexion range than the contralateral hip. The affected hip flexion range of motion increased after surgery and showed a significant increase by eight months after surgery. With time, the clinical hip extension range of motion increased for both limbs (Fig. 4). However, the simple effects tests indicated that only the affected limb increased significantly between each testing session. Pre-operatively and at four months post-operatively, the affected hip had a significantly smaller extension range, but by eight months post-operatively, symmetry was restored. For the abduction range of motion, the results of the ANOVA were a significant main effect for both session and limb, where abduction angle increased after surgery, and the contralateral limb had a larger abduction range than the affected limb. The interaction effect ($p = 0.07$) was tested using a simple effects model (Fig. 5). Within the simple effects model, the contralateral hip remained at a constant abduction range, while the affected hip increased significantly by four months after surgery. Pre-operatively and at four months post-operatively, the affected hip had a significantly smaller abduction range of motion than the contralateral hip, and at eight months, this effect was nearly significant ($p = 0.06$).

3.3 Manual muscle tests

Manual muscle tests revealed an incremental change in the strength of the affected hip after surgery. The frequencies of patients at each grade of muscle strengths for hip flexion, extension, and abduction are illustrated in Fig. 6, Fig. 7, and Fig. 8. Pre-operatively, all patients were graded a 4 or better for flexion and extension strength, with the exception of patient 12 who scored a 2 in both extension and abduction strength of the affected hip. All patients, except for patient 6, were graded a 4 or lower for the pre-operative hip abduction strength. By eight months post-operative, 75% of patients were graded a 5 for the flexion and extension strengths, and 67% of patients were graded a 5 or 5- for the abduction strength of the affected hip. The flexion, extension, and abduction strengths of the contralateral hip were graded a 5 for at least 75% of patients at all testing sessions (Fig. 9, Fig. 10, Fig. 11).

Table 3: Testing session date relative to surgery date for THR patients.

Patient	Gender†	Age at time of surgery (years)	Pre-Operative (days prior to surgery)	4 Months Post-Operative (months after surgery)	8 Month Post-Operative (months after surgery
1	M	58	6	4.4	8.6
2	M	58	3	4.1	9.0
3	M	67	8	4.2	9.0
4	M	60	10	4.4	7.6
5	M	69	9	4.4	8.4
6	M	72	16	4.0	8.2
7	M	59	5	4.4	9.1
8	F	71	9	4.0	8.3
9	F	63	7	n/a‡	n/a‡
10	M	74	22	5.7	8.0
11	M	71	6	4.1	n/a‡
12	M	55	12	5.0	7.6
13	F	46	5	4.0	8.0
14	M	69	4	n/a‡	n/a‡
Mean ± Std. Dev.		63.7 ± 8.0	9.8 ± 5.0	4.4 ± 0.5	8.3 ± 0.5

† M is male, F is female.

‡ Patient did not complete this testing session.

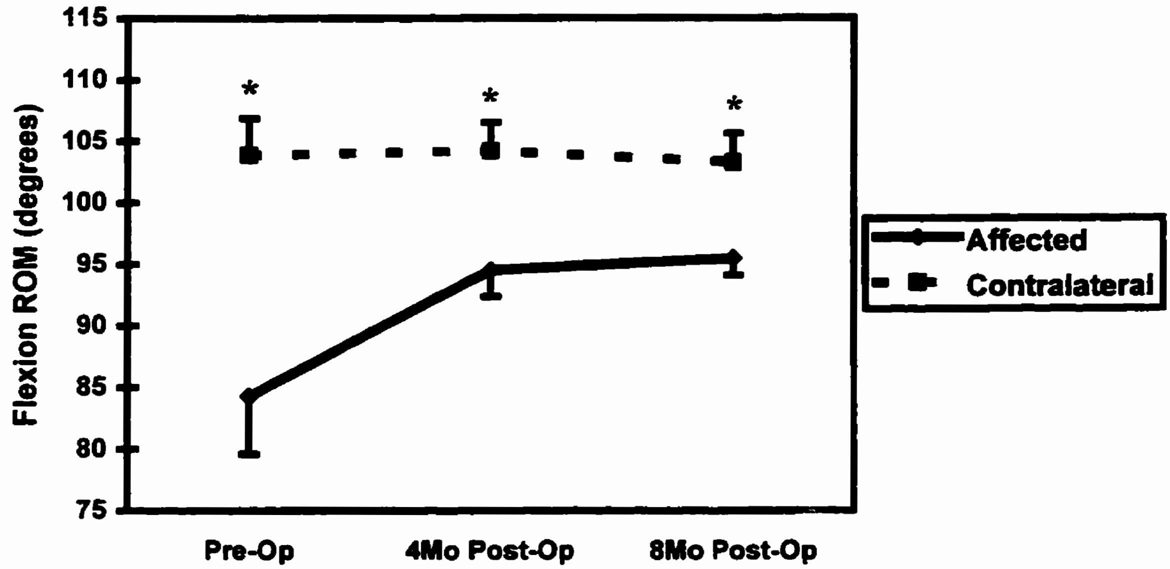


Fig. 3. Clinically measured hip flexion range of motion. Asterisks (*) represent differences between affected and contralateral limbs ($p < 0.05$). The affected hip flexion was significantly greater at eight months post-operatively than pre-operatively.

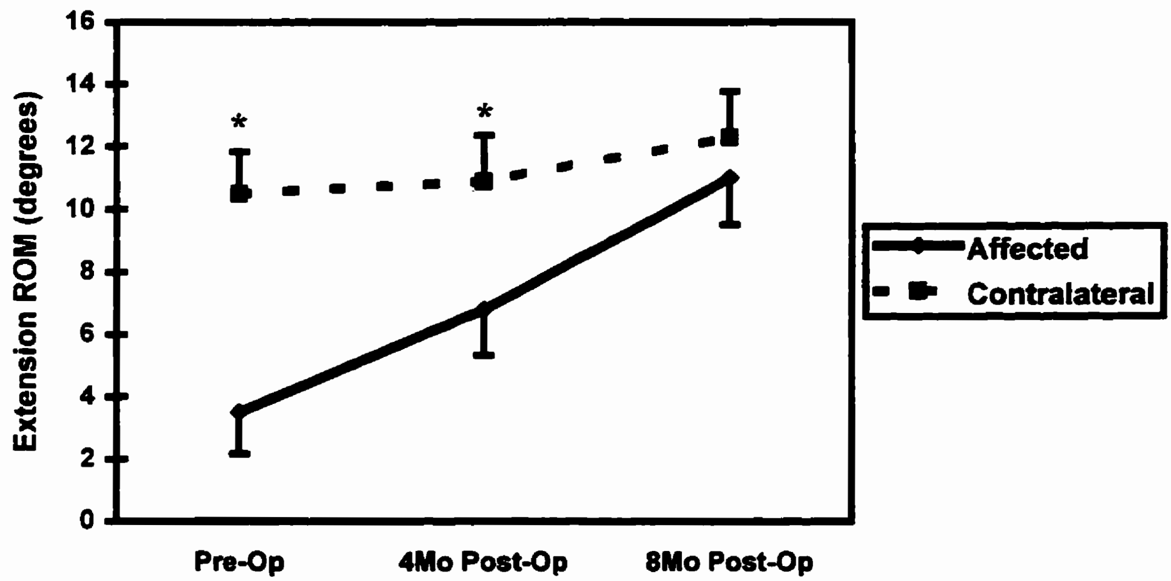


Fig. 4. Clinically measured hip extension range of motion. Asterisks (*) represent differences between affected and contralateral limbs ($p < 0.05$). The affected hip extension range increased significantly between pre-operative and four months post-operative, and between four and eight months post-operative.

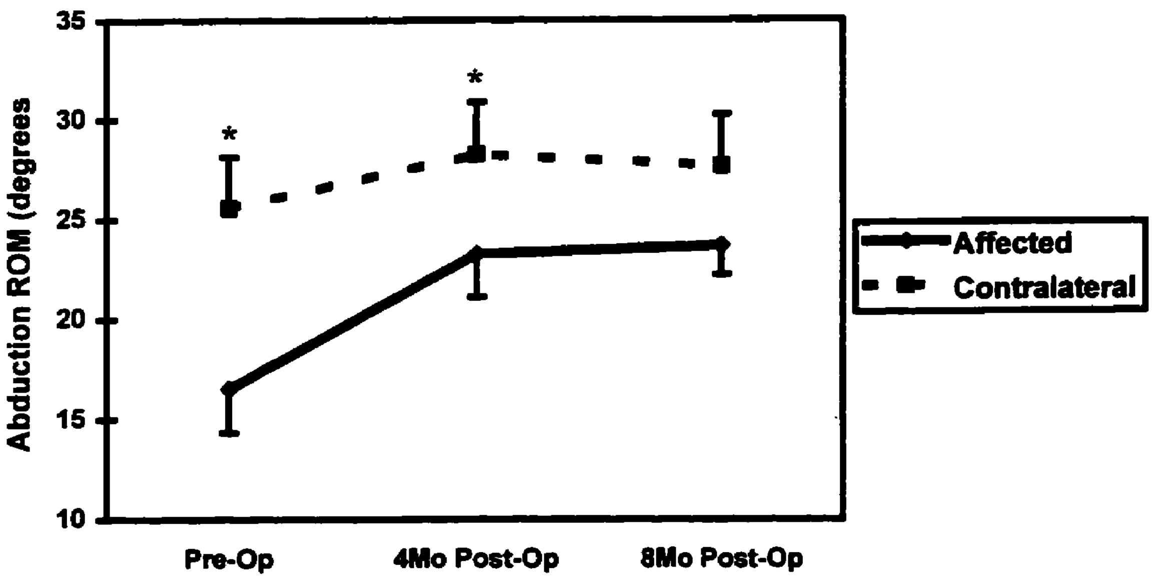


Fig. 5. Clinically measured hip abduction range of motion. Asterisks (*) represent differences between affected and contralateral limbs ($p < 0.05$). The four and eight month post-operative hip abduction ranges were significantly greater than the pre-operative abduction on the affected side.

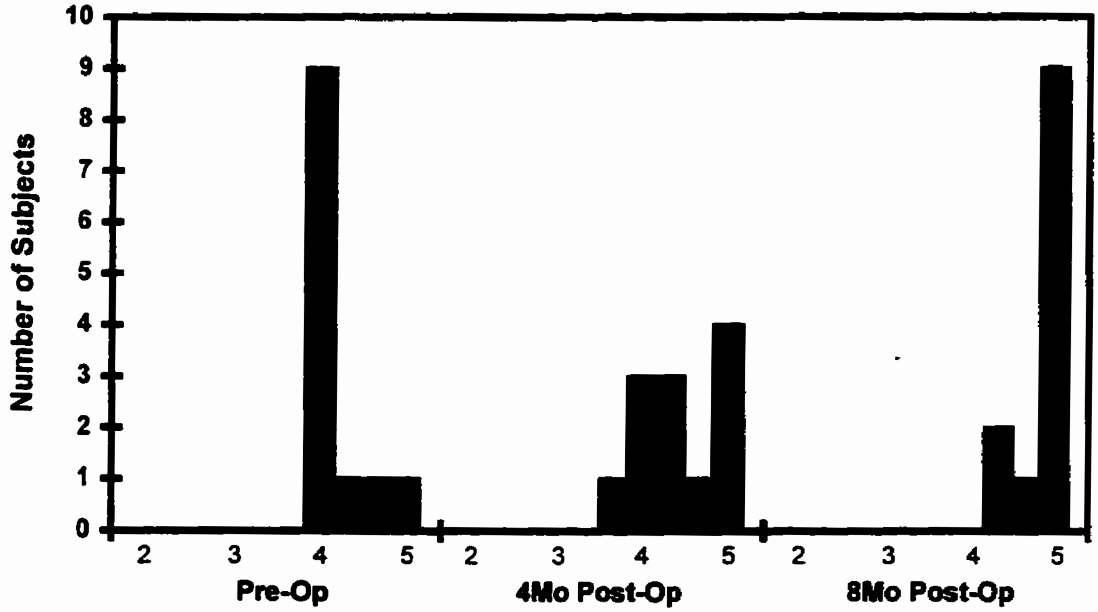


Fig. 6. Frequency histogram for the affected hip flexion strength of THR patients. Strength was graded on a 0 to 5 scale, indicated by shades of grey. Only integer grades are labeled, but each grade was assigned plus and minus.

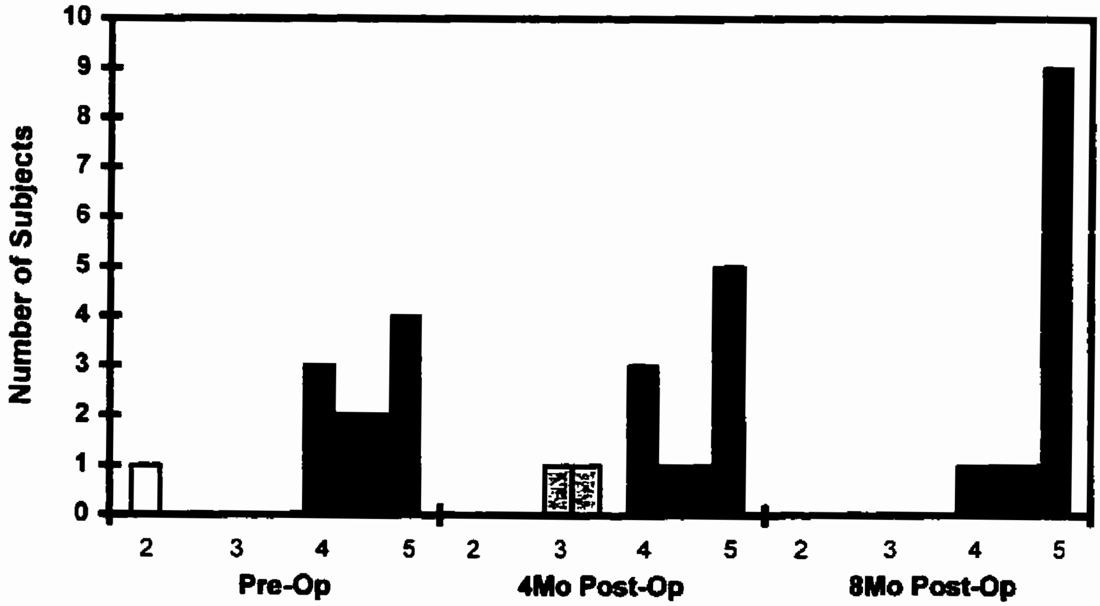


Fig. 7. Frequency histogram for the affected hip extension strength of THR patients. Strength was graded on a 0 to 5 scale, indicated by shades of grey. Only integer grades are labeled, but each grade was assigned plus and minus.

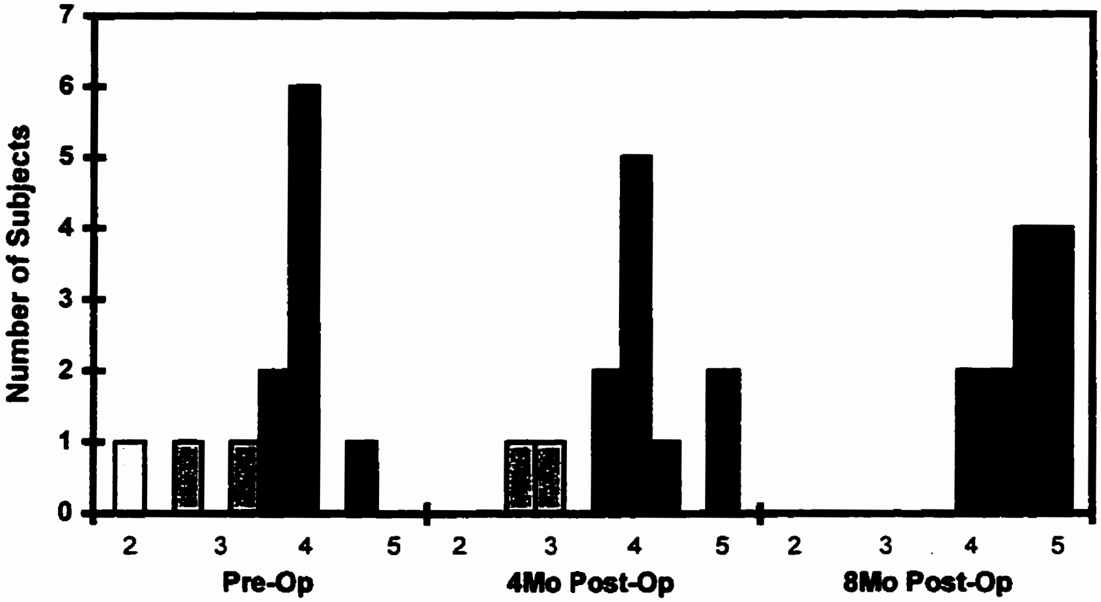


Fig. 8. Frequency histogram for the affected hip abduction strength of THR patients. Strength was graded on a 0 to 5 scale, indicated by shades of grey. Only integer grades are labeled, but each grade was assigned plus and minus.

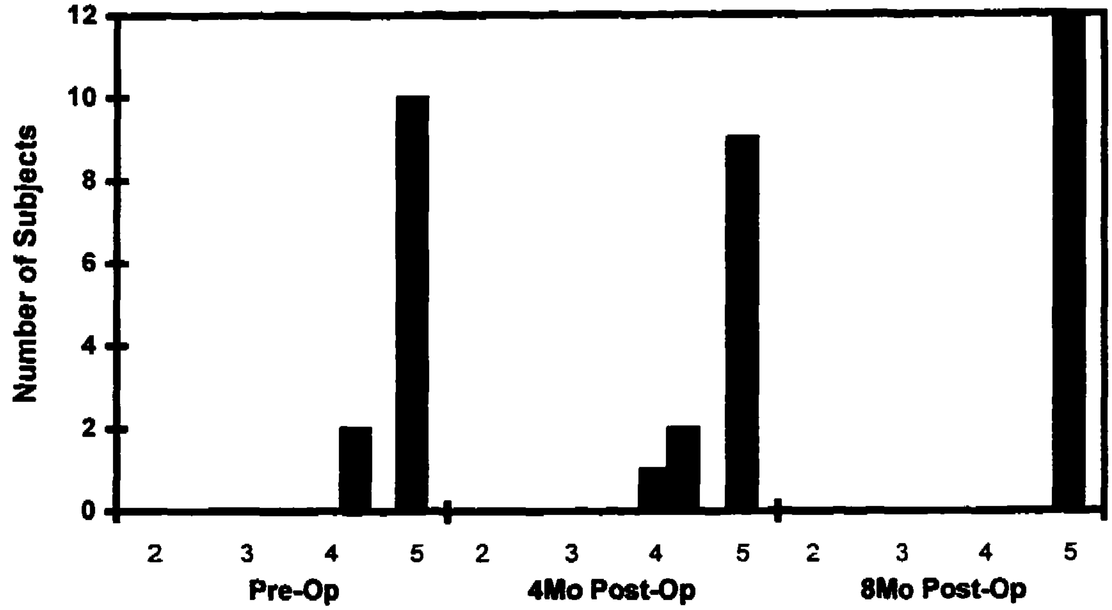


Fig. 9. Frequency histogram for the contralateral hip flexion strength of THR patients. Strength was graded on a 0 to 5 scale, indicated by shades of grey. Only integer grades are labeled, but each grade was assigned plus and minus.

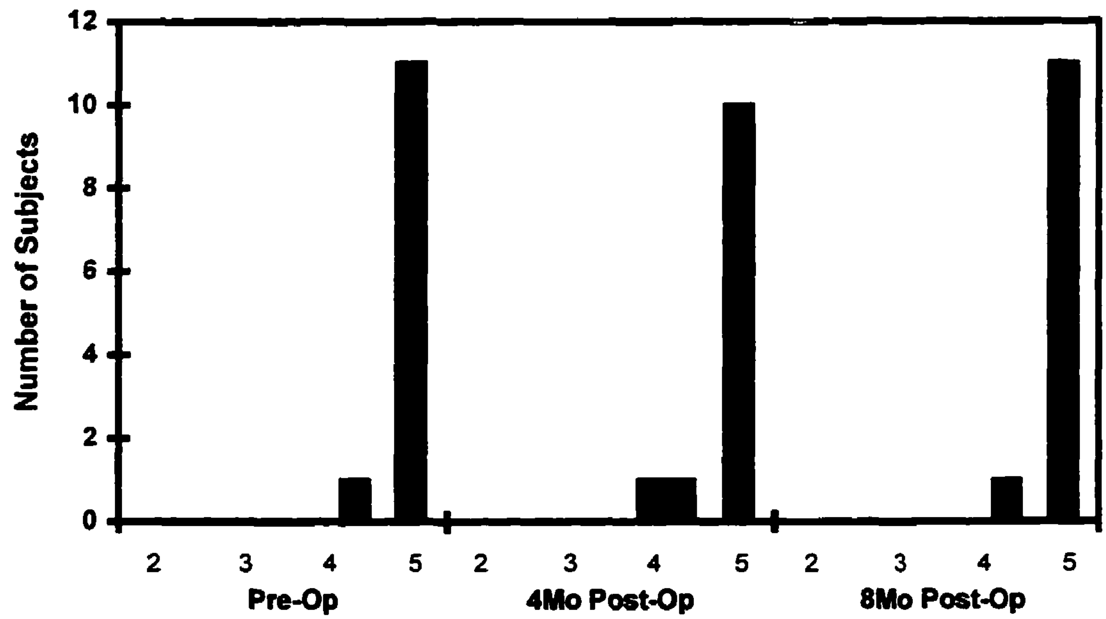


Fig. 10. Frequency histogram for the contralateral hip extension strength of THR patients. Strength was graded on a 0 to 5 scale, indicated by shades of grey. Only integer grades are labeled, but each grade was assigned plus and minus.

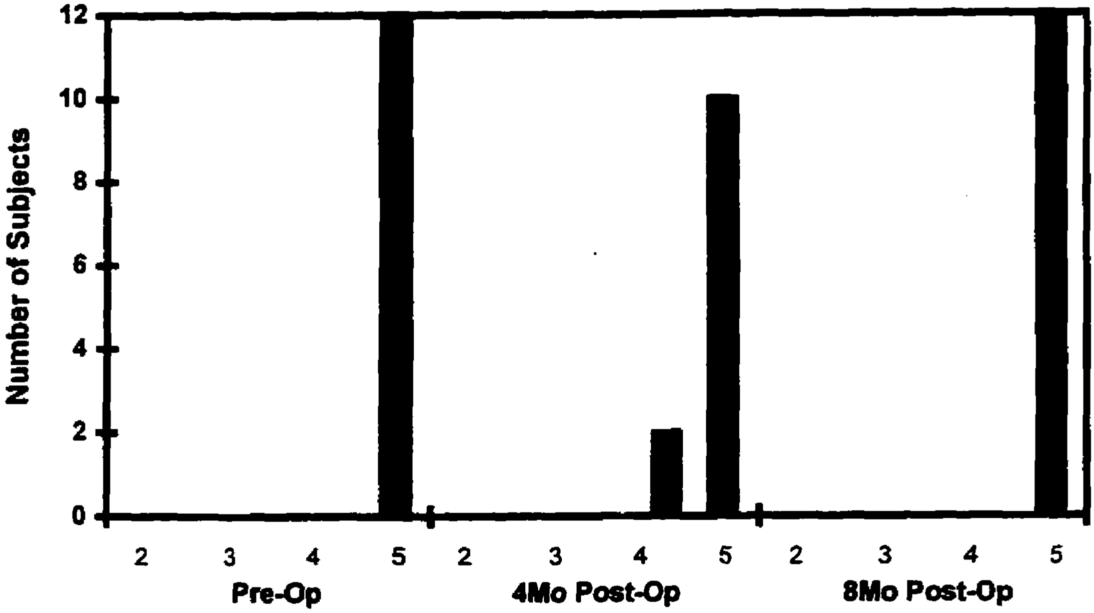


Fig. 11. Frequency histogram for the contralateral hip abduction strength of THR patients. Strength was graded on a 0 to 5 scale. Only integer grades are labeled, but each grade was assigned plus and minus.

Chapter 4

Gait analysis results

4.1 Analysis of variance

The analysis of variance (ANOVA) was performed for the trials in which no cane was used. The two main effects in the ANOVA were session (pre-operative, four months post-operative, and eight months post-operative) and limb (affected and contralateral).

4.1.1 Walking speed

The freely selected speed of walking for patients is shown in Fig. 12. There was no difference in speed during the trials collected for the affected limb as compared to the contralateral limb for all testing sessions. Speed did increase significantly after surgery, with both the four month and eight month post-operative speeds significantly greater than the pre-operative speed.

4.1.2 Lateral torso sway

The lateral sway of the torso averaged $4.4° \pm 0.2°$ toward either side (Fig. 13). There were no significant differences between testing sessions, nor were there any differences between affected and contralateral limbs.

4.1.3 Ground reaction force

The vertical component of the GRF curve (Fz) had two distinct peaks for patients walking in this study (Fig. 14). The first peak, or landing peak (Fz1), represented the acceptance of weight onto the limb. The second peak, or pushoff peak (Fz2), represented the vertical component of the pushoff onto the opposite limb. Both of these peaks were significantly reduced for the affected limb as compared to the contralateral limb during all three time points, but there were no significant differences from pre- to post-operatively. There was no difference for the minimum point between the two curves (Fzmin) between the affected and contralateral limbs, but there was a significant decrease in the four and eight month values as compared to pre-operatively. When examined as a range between the peaks and Fzmin, there were significant differences both across time points and between limbs. The difference between Fz1 and Fzmin, or yield range (FzR1), tended to increase after surgery and was significantly less on the affected side. The graph for the difference between Fz2 and Fzmin, or pushoff range (FzR2), showed the same trend as the yield range, except the session effect was significant between pre-operative and eight months post-operative sessions. The pushoff range for the affected limb was significantly reduced compared to the contralateral limb.

The anterior-posterior component of the GRF (Fy) had two peaks: an anterior peak during the loading response, and a posterior peak during pushoff. There were no significant main effects for the anterior peak, but there was a significant interaction between the session and limb effects (Fig. 15). The simple effects test revealed a significant increase in the affected limb from pre- to post-operative sessions, whereas the contralateral limb remained stable. Pre-operatively, significant asymmetry existed between the affected and contralateral limbs that was not present by four months post-operatively. The posterior peak on the affected limb was significantly smaller than the contralateral limb, but there were no differences across time points (Fig. 16).

The medial-lateral shear component of the GRF (Fx) typically had a medial peak shortly after foot-strike and a lateral peak which lasted throughout the remainder of stance (Fig. 17). Simple effects revealed that the medial peak increased for the contralateral limb after surgery, but that the affected limb did not change (Fig. 18). Pre-operatively, the affected and contralateral limbs had equal medial peaks, but by four and eight months after surgery, the contralateral limb increased and was significantly greater than the affected limb. There were no significant effects for the lateral peak, despite a trend towards a decreasing shear over time (Fig. 19).

4.1.4 Hip kinematics

During gait, the hip reaches its maximum flexion angle at around foot strike and its maximum extension angle at pre-swing [Perry, 1992]. For the patients in this study, the maximum flexion angle did not significantly increase post-operatively, nor was the difference between the affected limb and the contralateral limb significant (Fig. 20). The maximum hip extension angle was significantly smaller for the affected limb than the contralateral limb (Fig. 21). The extension angle differences between testing sessions were not statistically significant.

4.1.5 Joint moments

Joint moments were different among hip, knee, and ankle joints. The maximal ankle plantar flexion moment was significantly smaller on the affected side than the contralateral side, but there were no differences across time points (Fig. 22). Knee flexion and extension moments showed no significant differences (Fig. 23 and Fig. 24).

Differences were not significant for the maximum hip flexion moment between testing session or limbs (Fig. 25). The maximum hip extension moment tended to be smaller for the affected limb than the contralateral limb ($p = 0.08$) (Fig. 26). Neither the limb or session effects were significant for the hip abduction moment (Fig. 27).

4.1.6 Hip abductor muscle duration

The duration of activity during the stance phase for gluteus medius and tensor fascia lata are shown in Fig. 28 and Fig. 29. There were no statistically significant differences between sessions or limbs for either muscle. The gluteus medius muscle duration of activation on the contralateral limb increased by 11% of the stance phase by four months post-operative, while the duration on the affected limb remained stable. The tensor fascia lata duration on the contralateral limb decreased by 7% of the stance phase by four months post-operative, while the affected limb increased by 6% of the stance phase. By eight months post-operative, the tensor fascia lata duration was 2% smaller on the affected side, and the gluteus medius duration remained reduced on the affected side by 11% of the stance phase as compared to the contralateral limb.

4.1.7 Hip abductor muscle timing

The timing of the affected gluteus medius muscle did not differ significantly between testing sessions. Pre-operatively, 79% of patients had some gluteus medius activity from pre-stance to 60% of stance and only 14% had activity by 80% of stance (Fig. 30). At four months post-operative, 71% of patients had some activity from pre-stance to 60% of stance and only 7% of patients had activity by 80% of stance (Fig. 31).

At eight months post-operatively, 82% of patients had some gluteus activity from pre-stance to 60% of stance and only 9% of patients had some activity by 80% of stance (Fig. 32).

The timing of the affected tensor fascia lata was similar to that of gluteus medius for most patients. At all three testing sessions, more than 86% of patients had some tensor fascia lata activity from foot strike to 60% of stance (Fig. 33, Fig. 34 and Fig. 35). From 80% of stance to toe-off, at most, 27% of patients had activity for any of the testing sessions. During pre-stance, fewer patients had tensor fascia lata activity than after foot strike.

4.2 Analysis of covariance

The variables were analysed using an analysis of covariance (ANCOVA). Walking speed was used as a covariate to control for the potentially different walking speeds of the trials. Speed was entered as a covariate on a condition by condition basis, where the mean speed for the three trials of each condition was associated with the mean of each variable for each condition. The ANCOVA had two main effects, session and limb, as did the ANOVA. The means for each variable at each condition were adjusted for speed to plot results of the ANCOVA.

4.2.1 Lateral torso sway

The ANCOVA results for the lateral torso sway, with speed as a covariate, showed no significant differences between testing sessions or between limbs.

4.2.2 Ground reaction force

The landing peak of the vertical component of the GRF curve decreased significantly after surgery when it was adjusted for speed (Fig. 36). Both limbs had a significantly smaller adjusted landing peak by four months post-operative, with the affected limb being significantly smaller than the contralateral limb. The adjusted pushoff peak differences were not significant (Fig. 37). The minimum between the two peaks (Fzmin) showed no significant difference between sessions or limbs when adjusted for speed. The adjusted yield (FzR1) and pushoff (FzR2) ranges were smaller on the affected side than the contralateral side with the pushoff range being significant.

The anterior component of the GRF, when adjusted for speed, was significantly smaller by four months after surgery for the contralateral limb, but the affected limb remained stable (Fig. 38). The adjusted posterior component was significantly different between each of the sessions (Fig. 39). The adjusted posterior peak was significantly smaller on the affected side than the contralateral side.

The medial component of the GRF was significantly smaller on the affected side than the unaffected side when controlling for speed (Fig. 40). The medial peak for the contralateral limb did not differ over time, when controlling for speed, but the adjusted medial peak on the affected side decreased significantly by eight months post-operative. The adjusted lateral peak tended to decrease after surgery, but the effect was not significant (Fig. 41).

4.2.3 Hip kinematics

The sagittal hip range of motion during gait (when adjusted for speed) was smaller on the affected side than the unaffected side throughout the eight month protocol. This effect was significant for the maximum extension angle (Fig. 43) and total range of motion (Fig. 44), but not for the maximum flexion angle (Fig. 42). The adjusted sagittal hip range of motion during gait did not increase significantly in the affected limb, but decreased significantly in the contralateral limb by eight months post-operative.

4.2.4 Joint moments

Joint moments with speed as a covariate were similar to the joint moments without speed. The adjusted ankle plantar flexion moment was significantly smaller on the affected side than the contralateral side (Fig. 45). Adjusted knee flexion and extension moments were not significantly different between sessions or limbs (Fig. 46, and Fig. 47). The adjusted hip flexion moment was significantly larger on the affected side than the contralateral side (Fig. 48). The adjusted hip extension moment was smaller for the affected limb than the contralateral limb ($p = 0.07$) (Fig. 49). The adjusted hip abduction moment did not show any statistically significant differences between testing sessions or limbs (Fig. 50).

4.2.5 Hip abductor muscle duration

The adjusted means for the duration of activity of gluteus medius and tensor fascia lata are illustrated in Fig. 51 and Fig. 52. There were no statistically significant differences between testing sessions or limbs for either muscle.

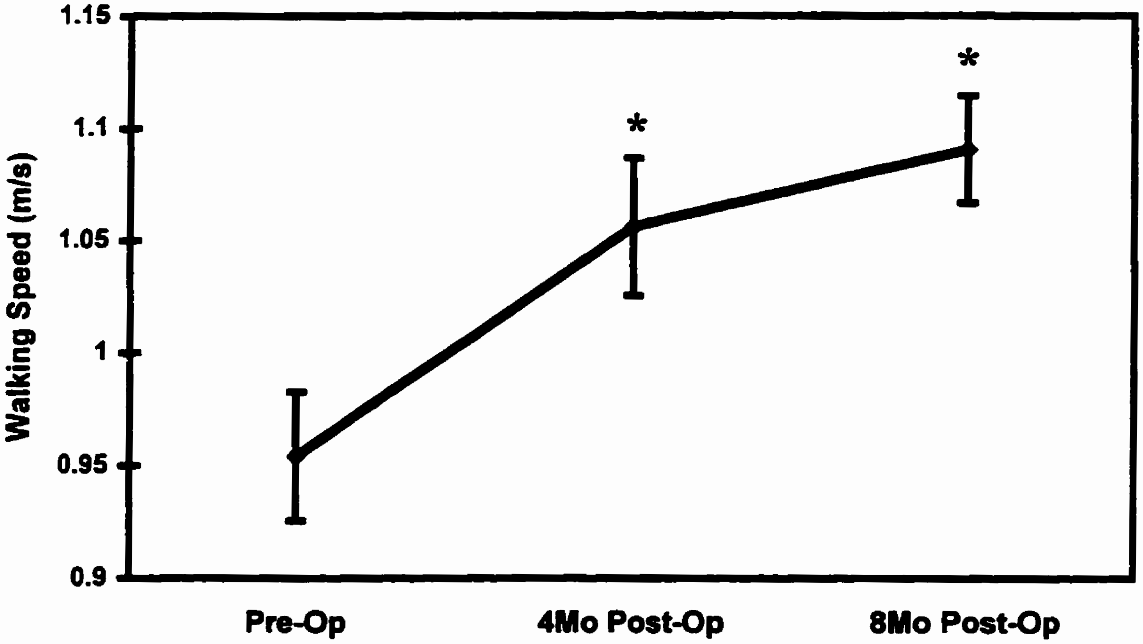


Fig. 12. Free walking speed of pre- and post-operative THR patients. Four and eight month post-operative speeds were significantly larger than pre-operative speed (*, $p < 0.05$).

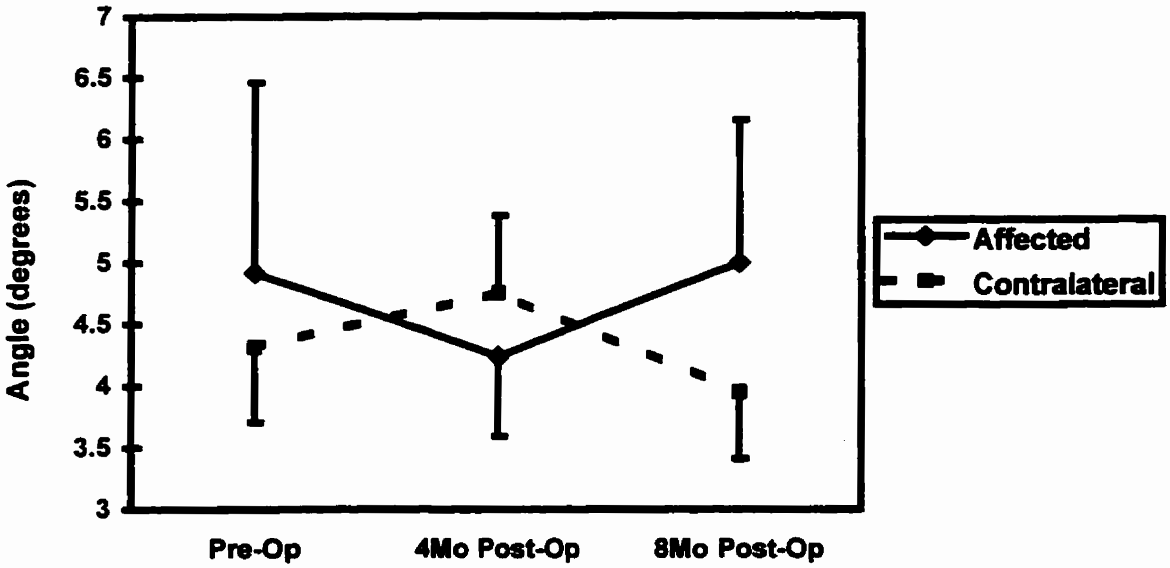


Fig. 13. Lateral torso sway for THR patients.

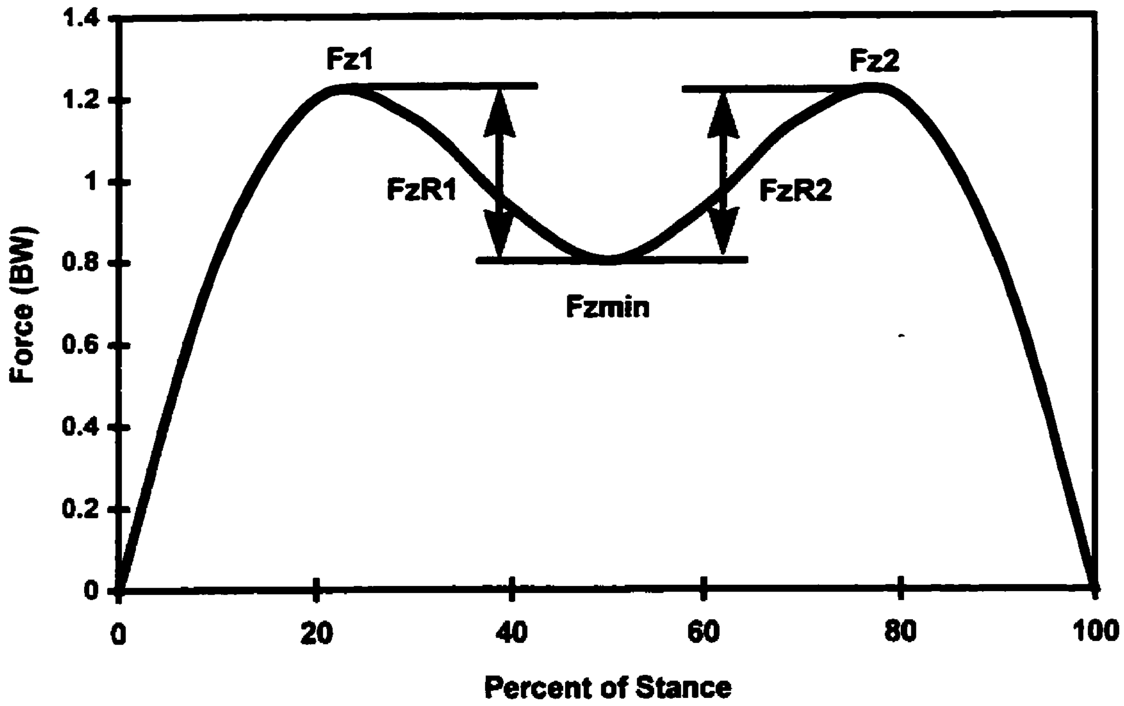


Fig. 14. Typical vertical GRF curve. Labels represent landing peak (Fz1), Pushoff peak (Fz2), minimum during mid stance (Fzmin), yield range (FzR1), and pushoff range (FzR2).

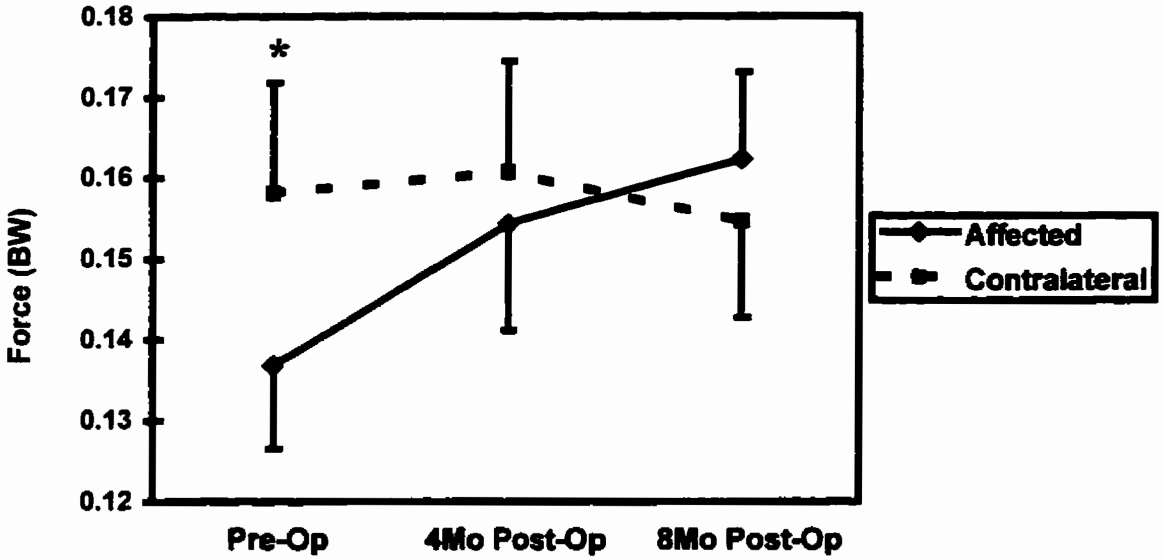


Fig. 15. Anterior shear peak of the GRF for THR patients. The asterisk (*) represents a difference between affected and contralateral limbs ($p < 0.05$). The shear on the affected side increased significantly across testing sessions.

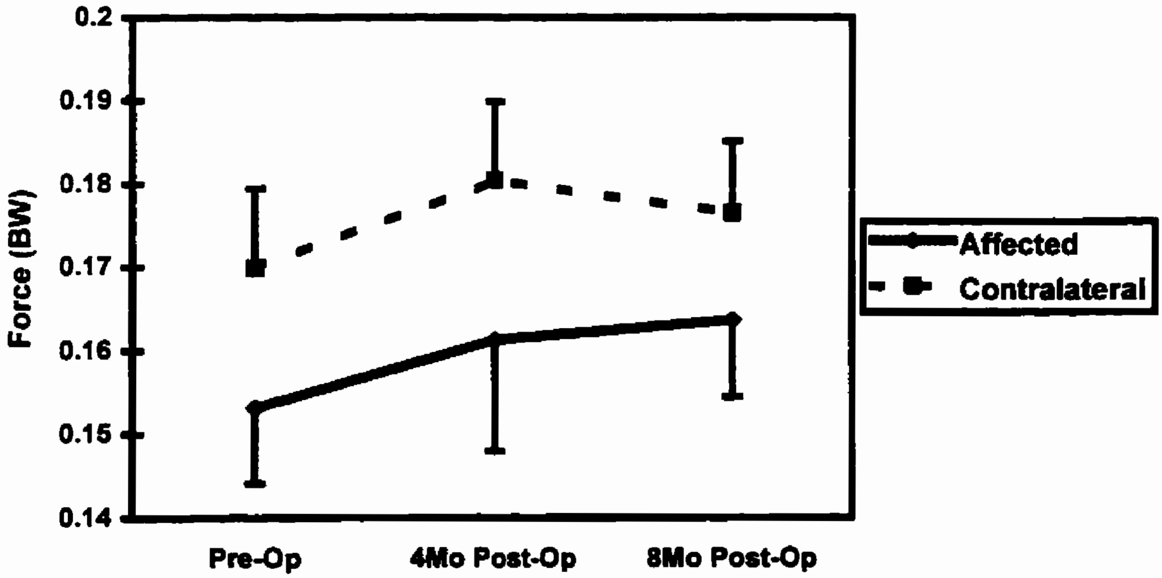


Fig. 16. Posterior shear peak of the GRF for THR patients. The affected leg had a significantly smaller shear peak than the contralateral leg.

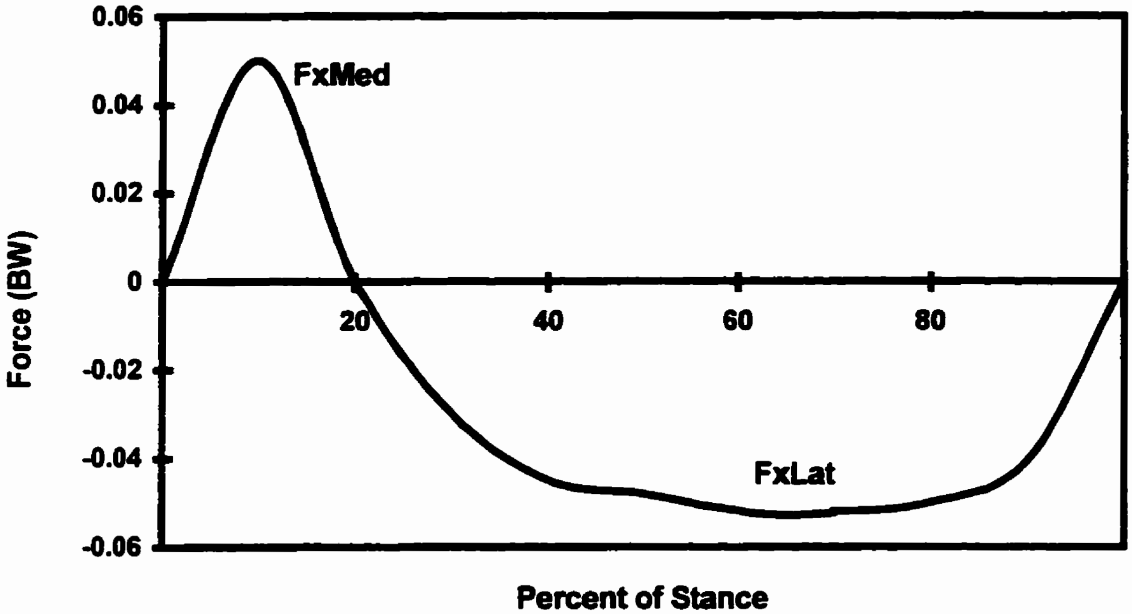


Fig. 17. A typical curve for the medial-lateral component of the GRF. The medial peak (FxMed) typically occurred during the loading response, and the lateral peak (FxLat) lasted for much of the stance phase.

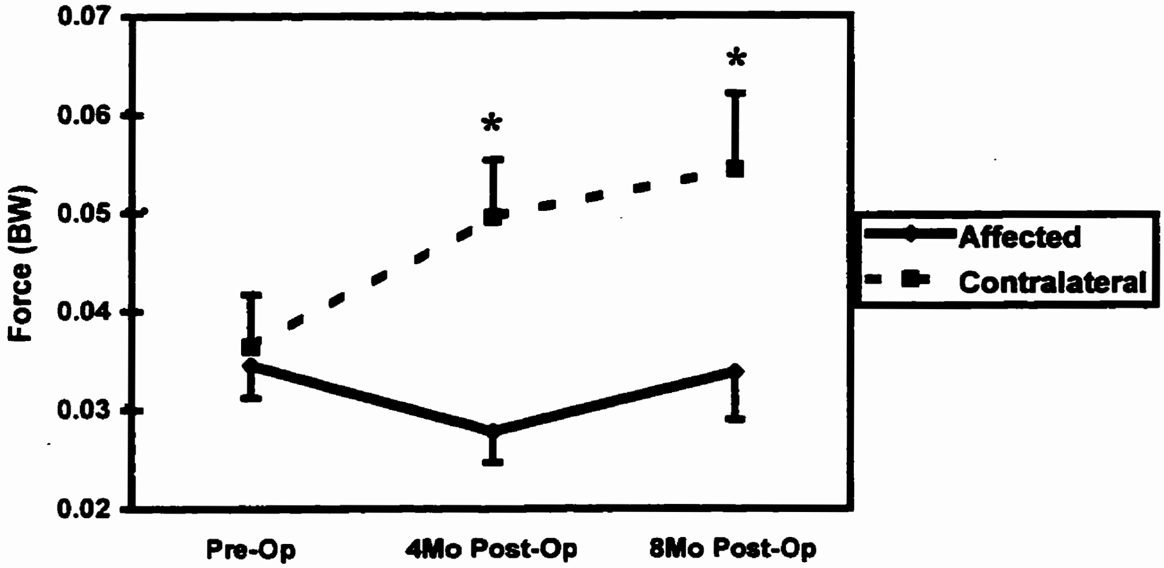


Fig. 18. Medial shear peak of the GRF for THR patients. Asterisks (*) represent differences between the affected and contralateral limbs. The contralateral limb increased significantly across testing sessions.

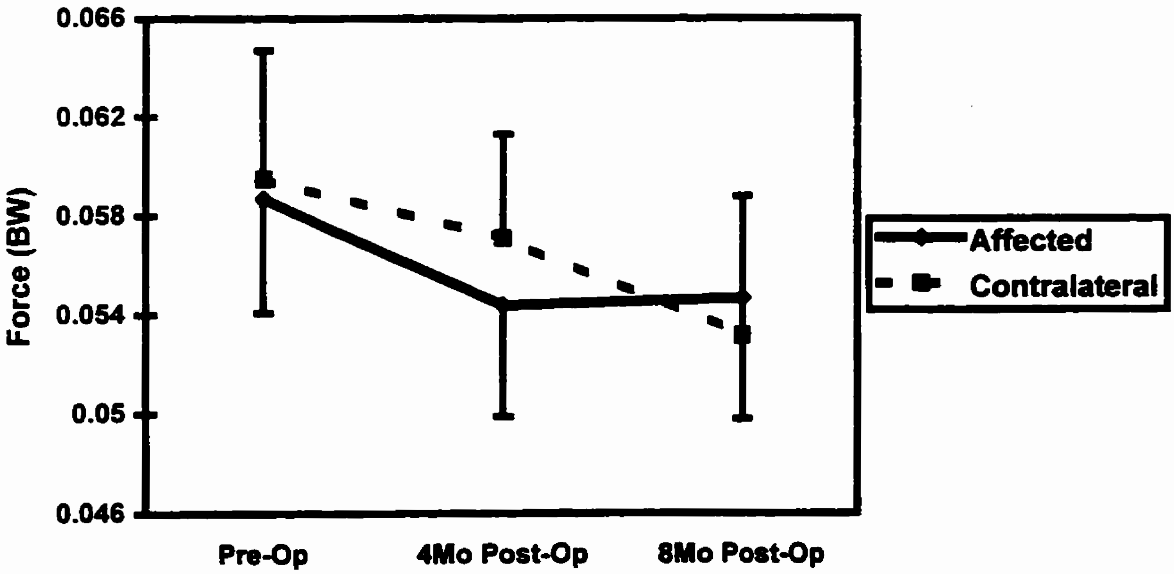


Fig. 19. Lateral shear peak of the GRF for THR patients.

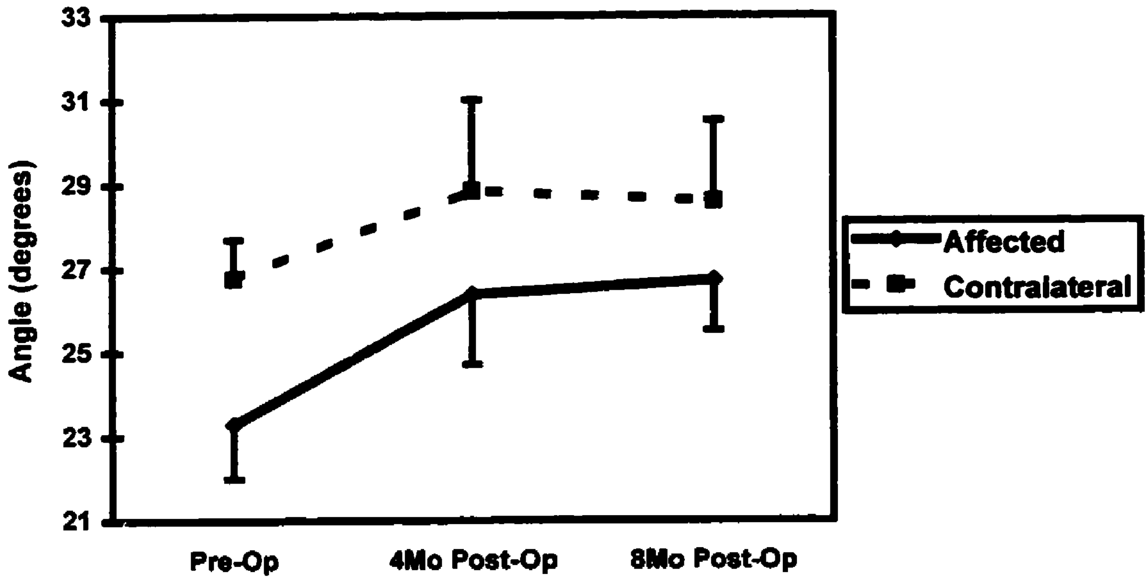


Fig. 20. Maximum hip flexion during gait for THR patients.

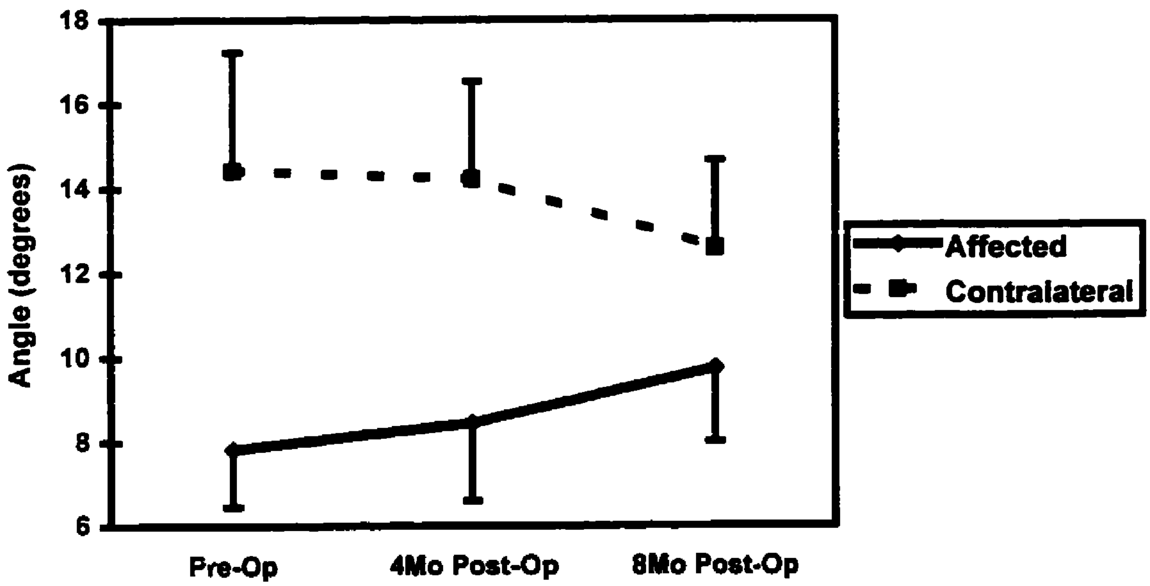


Fig. 21. Maximum hip extension during gait for THR patients. The hip extension angle was significantly smaller for the affected limb than the contralateral limb.

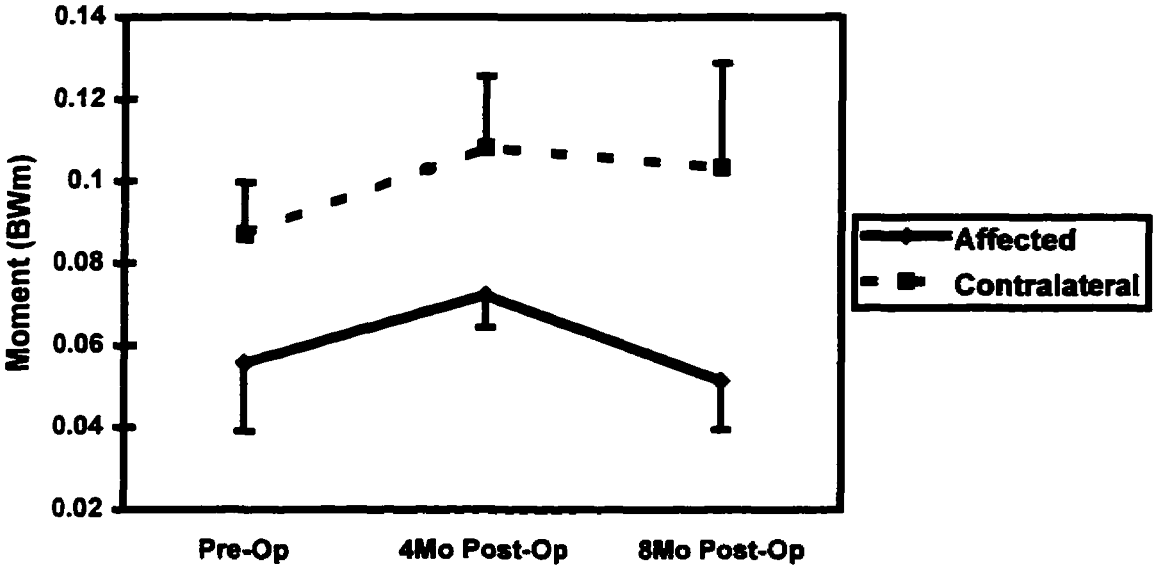


Fig. 22. Maximum ankle plantar flexion moment for THR patients. The ankle plantar flexion moment was significantly smaller on the affected limb than the contralateral side.

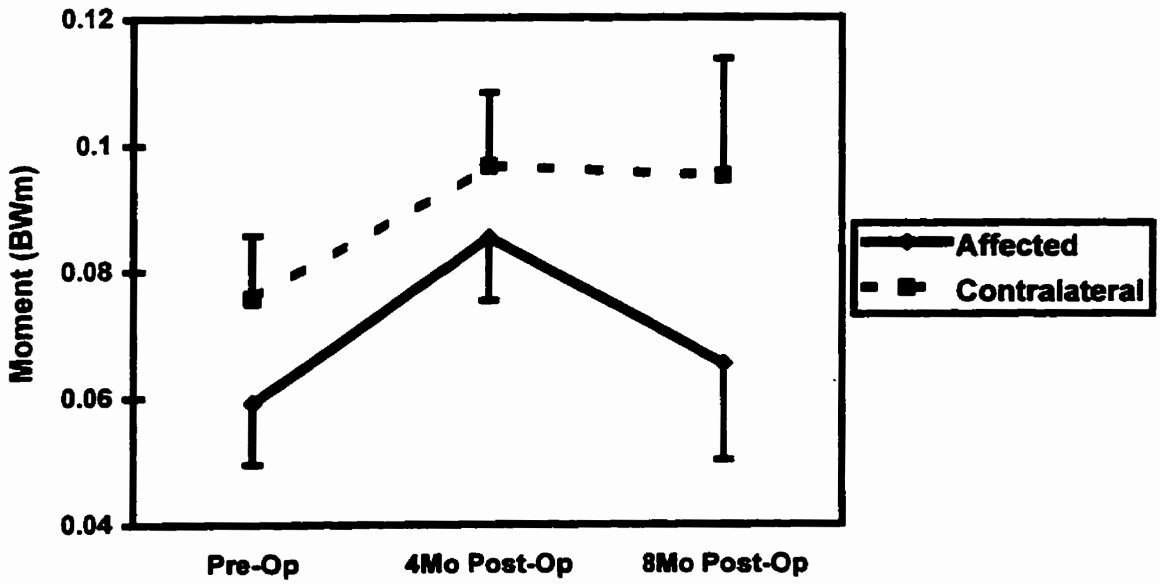


Fig. 23. Maximum knee flexion moment for THR patients.

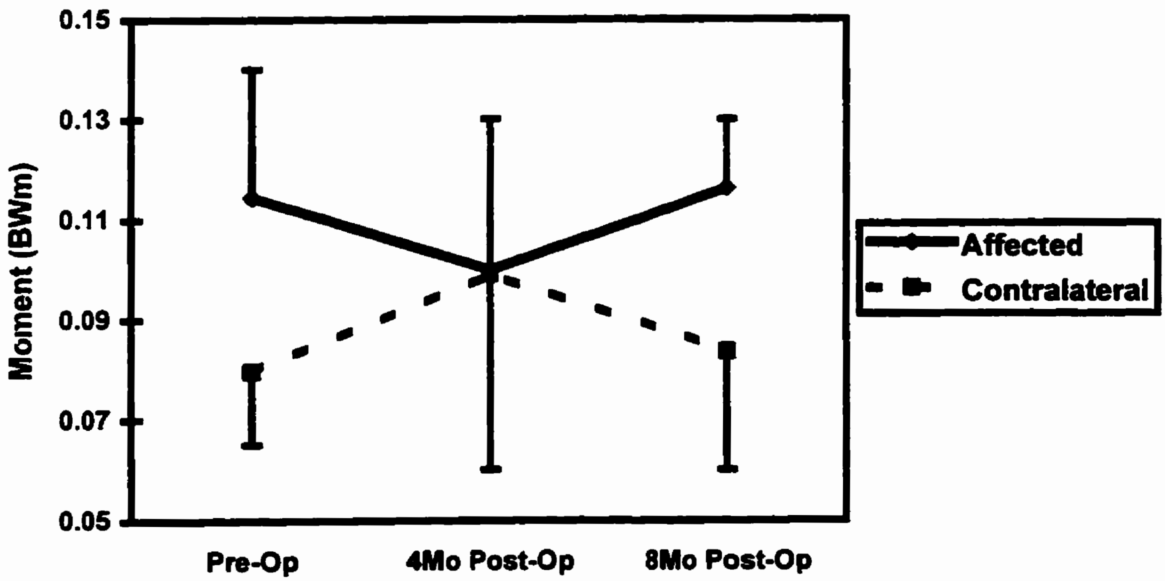


Fig. 24. Maximum knee extension moment for THR patients.

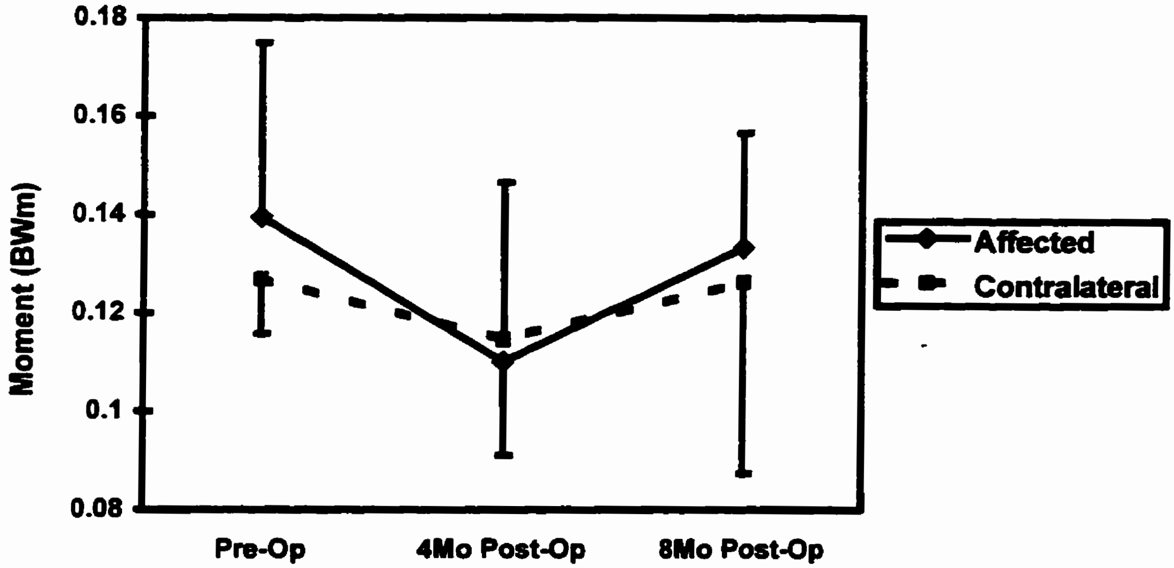


Fig. 25. Maximum hip flexion moment for THR patients.

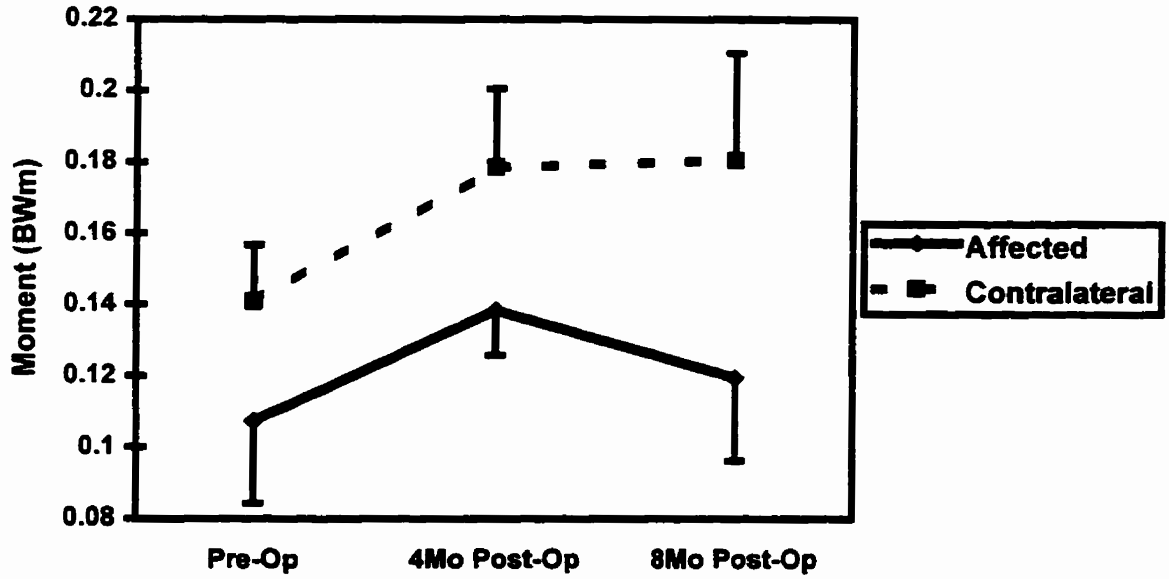


Fig. 26. Maximum hip extension moment for THR patients. The hip extension moment on the affected side was significantly smaller than the contralateral side.

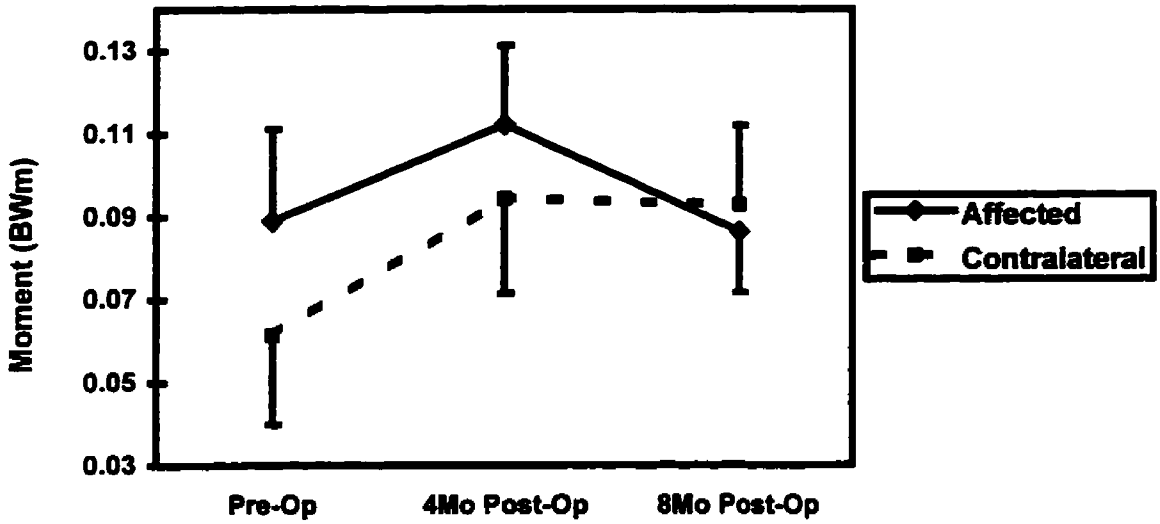


Fig. 27. Maximum hip abduction moment for THR patients.

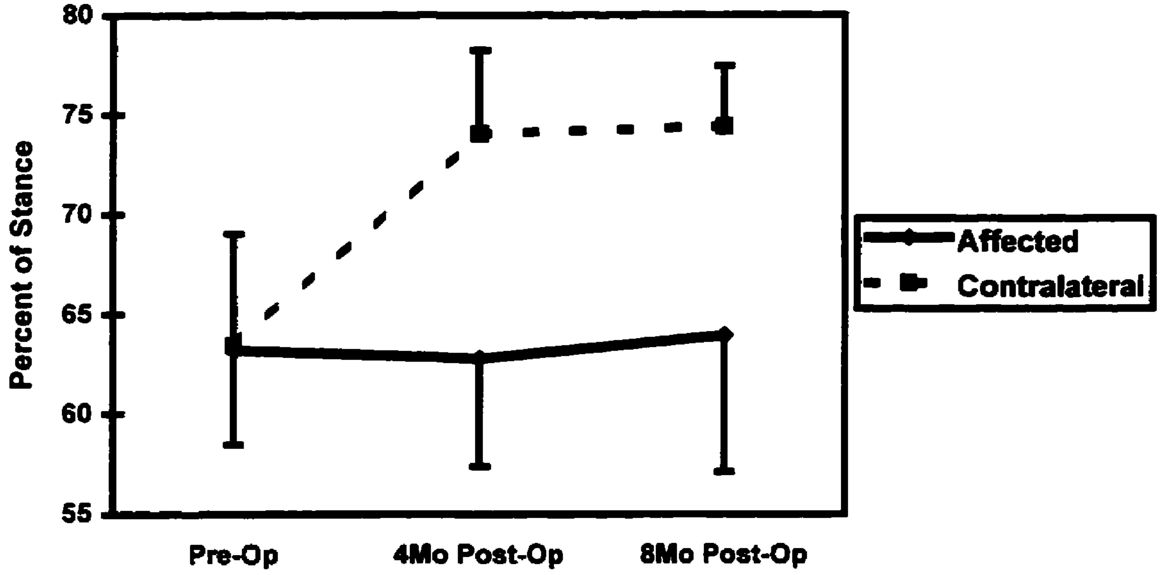


Fig. 28. Duration of activity of the gluteus medius during the stance phase for THR patients.

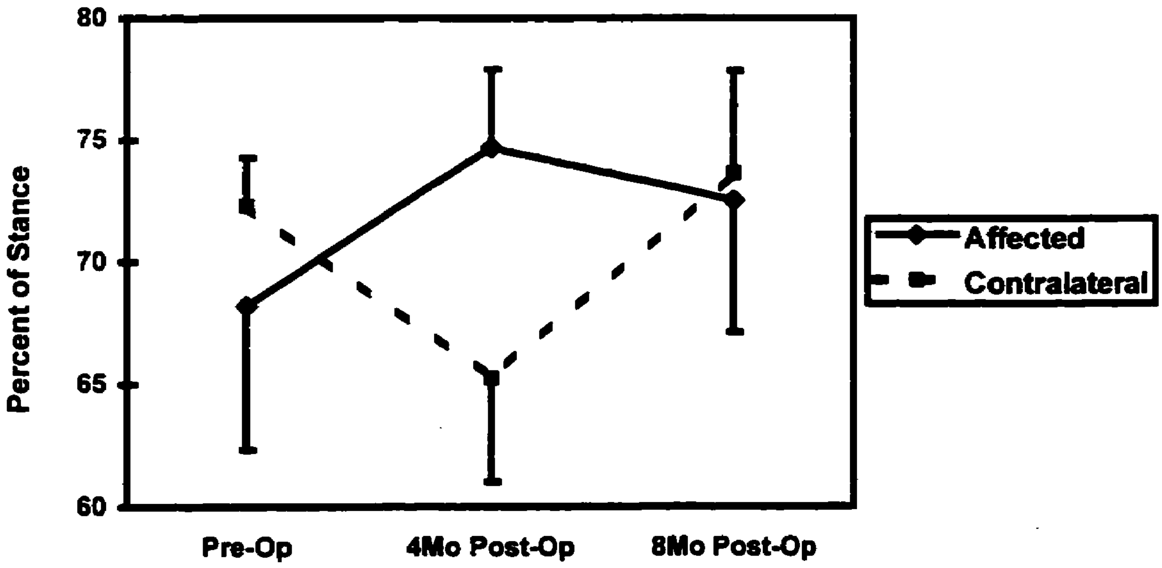


Fig. 29. Duration of activity of the tensor fascia lata during the stance phase for THR patients.

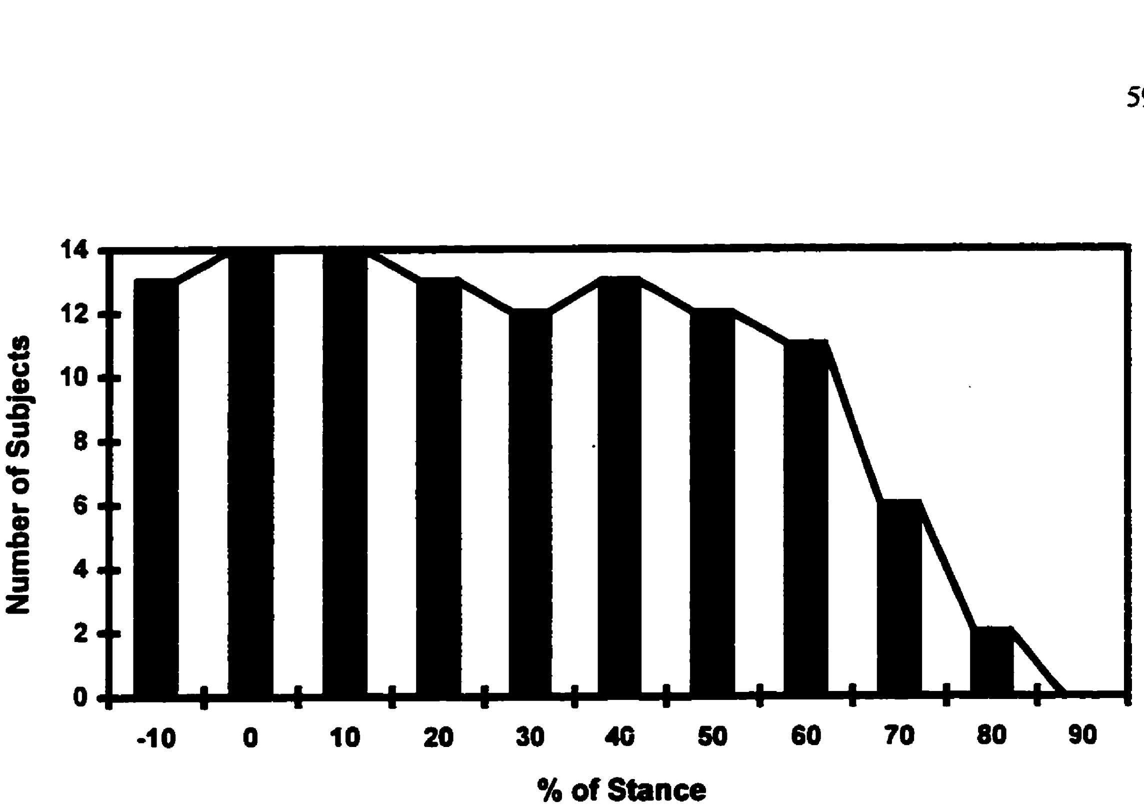


Fig. 30. Frequency histogram of the pre-operative affected gluteus medius activity during each of the 10% of stance time intervals. Total number of subjects in this plot was 14.

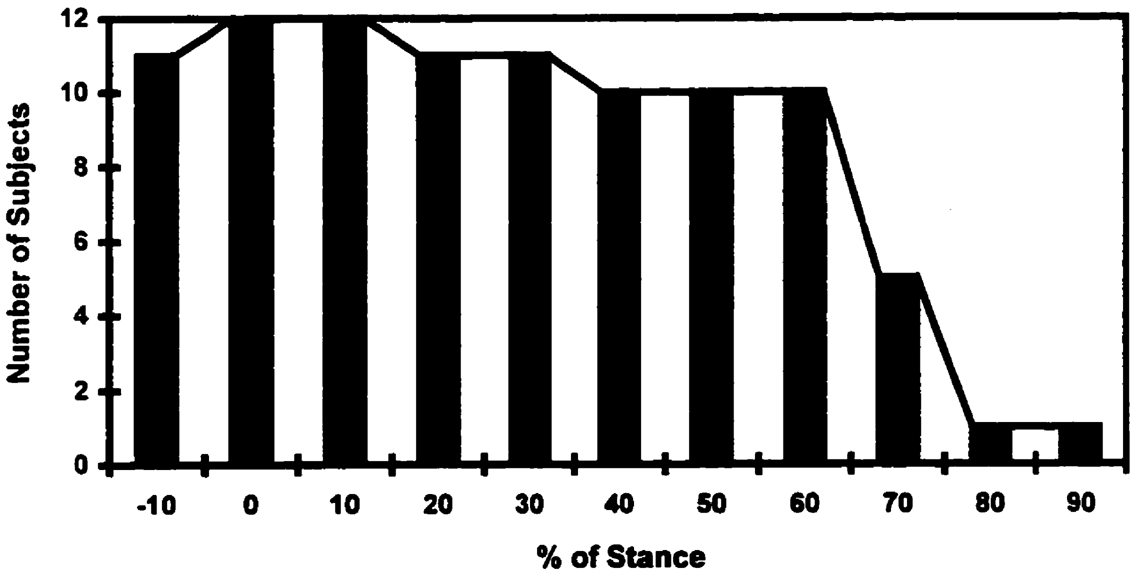


Fig. 31. Frequency histogram of the 4 month post-operative affected gluteus medius activity during each of the 10% of stance time intervals. Total number of subjects in this plot was 12.

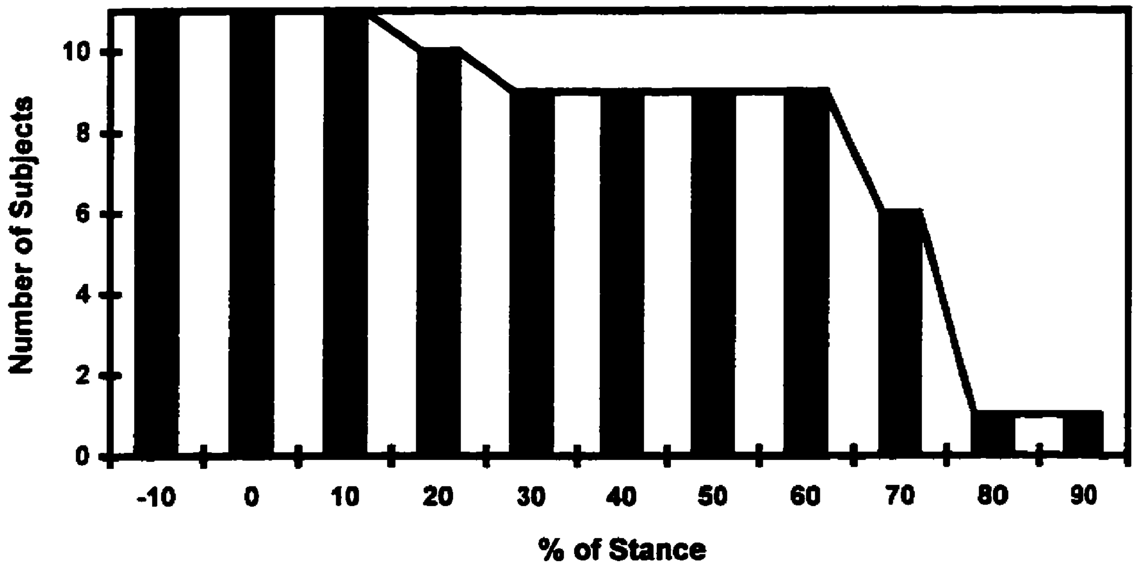


Fig. 32. Frequency histogram of the 8 month post-operative affected gluteus medius activity during each of the 10% of stance time intervals. Total number of subjects in this plot was 11.

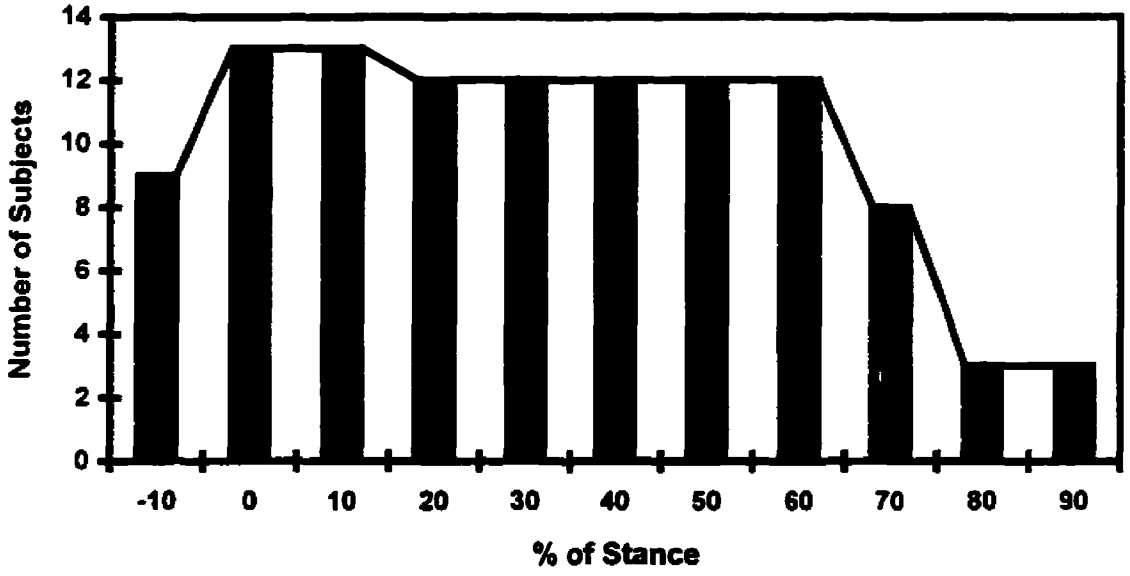


Fig. 33. Frequency histogram of the pre-operative affected tensor fascia lata activity during each of the 10% of stance time intervals. Total number of subjects in this plot was 14.

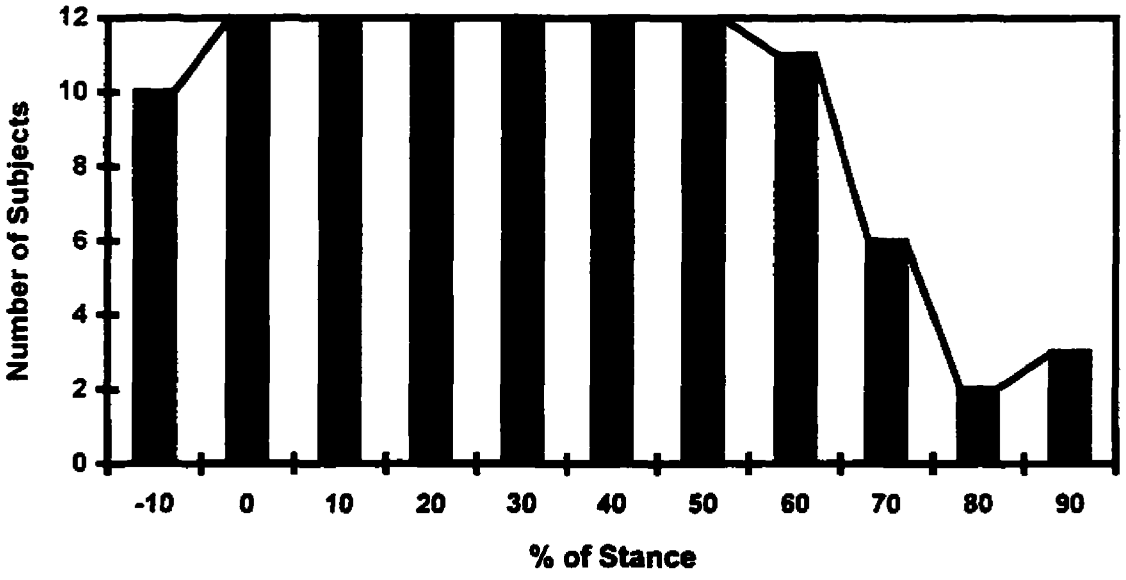


Fig. 34. Frequency histogram of the 4 month post-operative affected tensor fascia lata activity during each of the 10% of stance time intervals. Total number of subjects in this plot was 12.

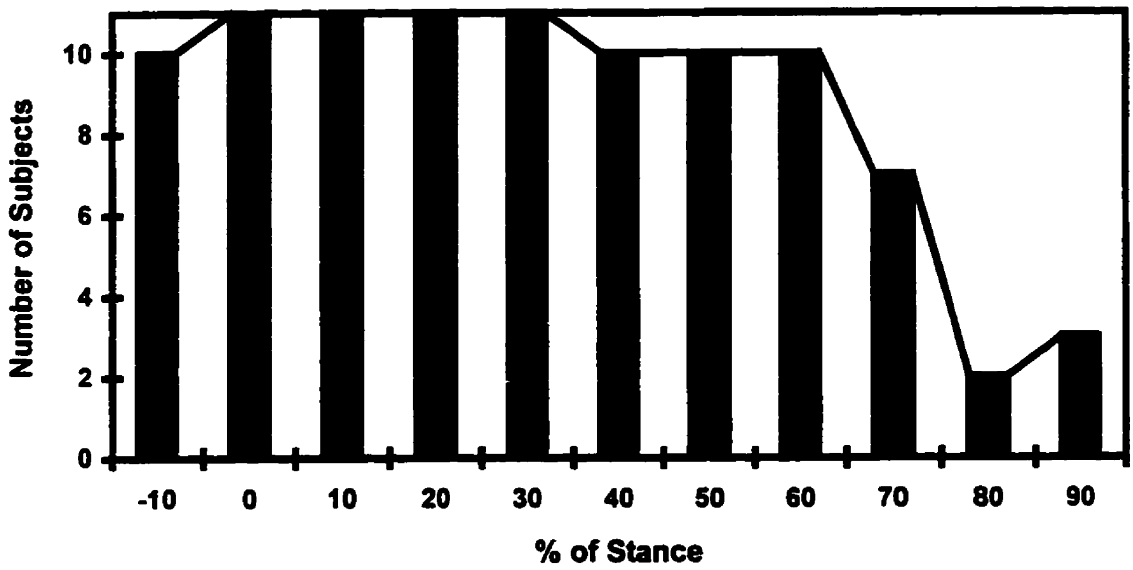


Fig. 35. Frequency histogram of the 8 month post-operative affected tensor fascia lata activity during each of the 10% of stance time intervals. Total number of subjects in this plot was 11.

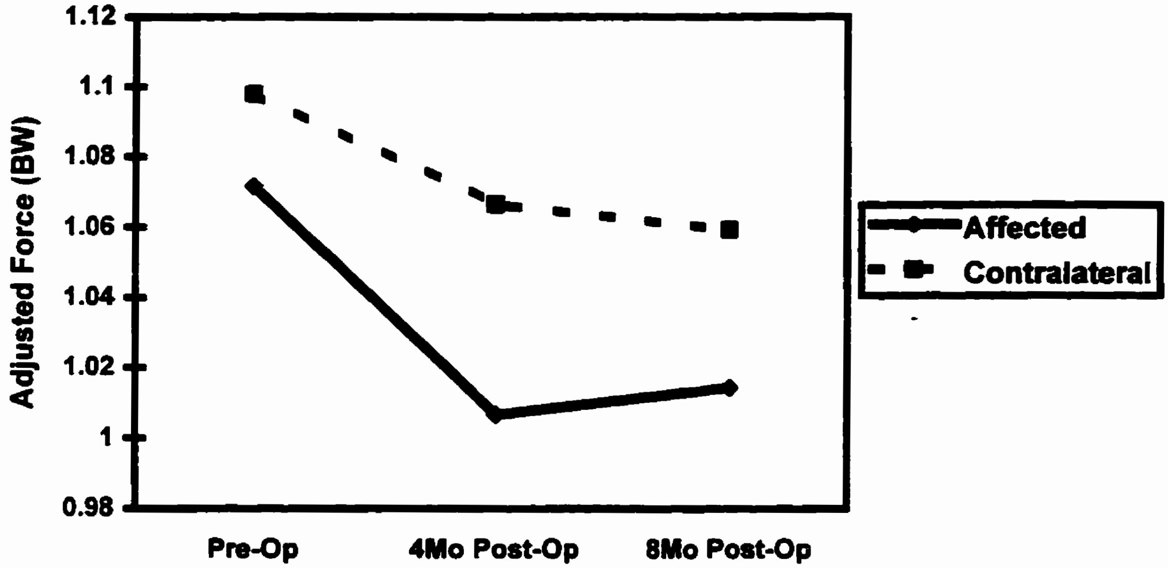


Fig. 36. Landing peak of the vertical GRF curve, adjusted for speed of walking. By four months post-operative, the adjusted landing peak was significantly smaller than the pre-operative peak. The affected limb had a smaller landing peak than the contralateral limb ($p < 0.05$).

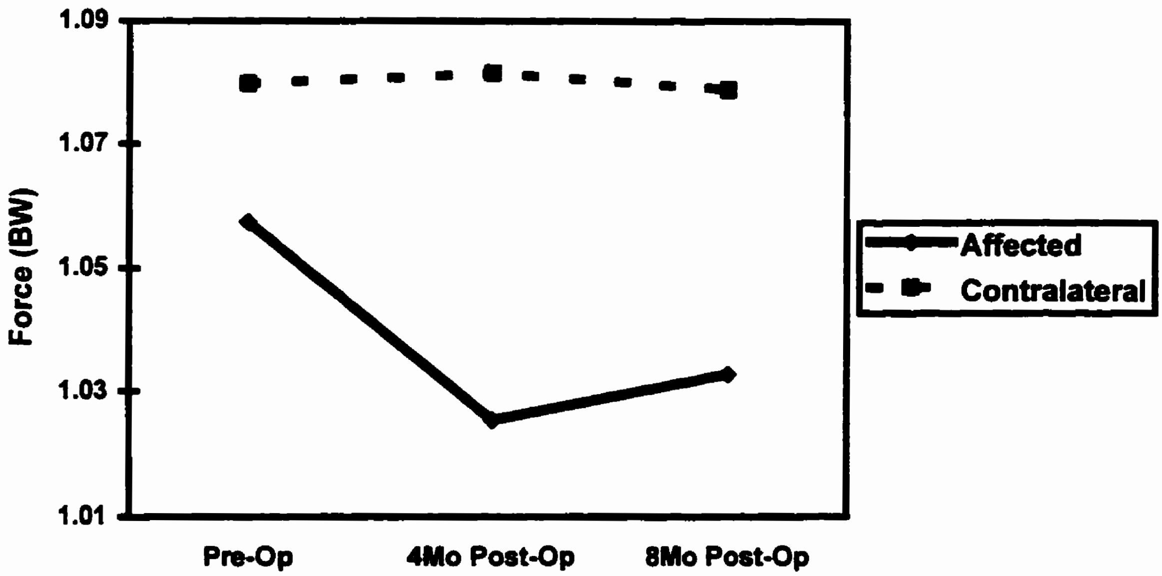


Fig. 37. Pushoff peak of the vertical GRF curve, adjusted for speed of walking.

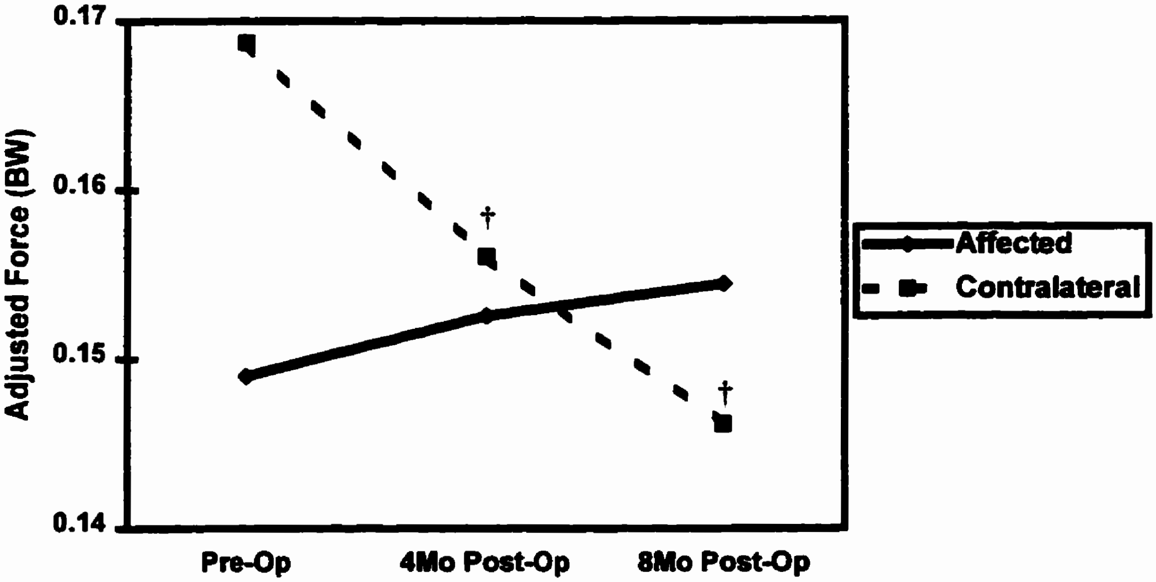


Fig. 38. Peak anterior shear of the GRF, adjusted for speed of walking. † represent differences from pre-operative to four and eight month post-operative forces ($p < 0.05$).

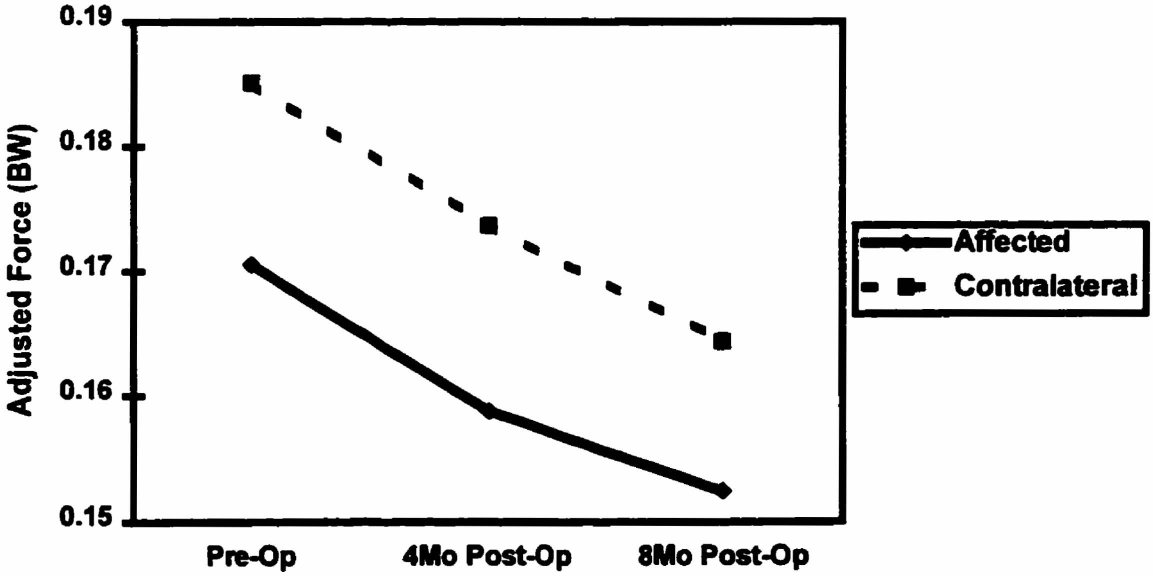


Fig. 39. Peak posterior shear of the GRF, adjusted for speed of walking. The affected limb had a significantly smaller adjusted posterior shear than the contralateral limb.

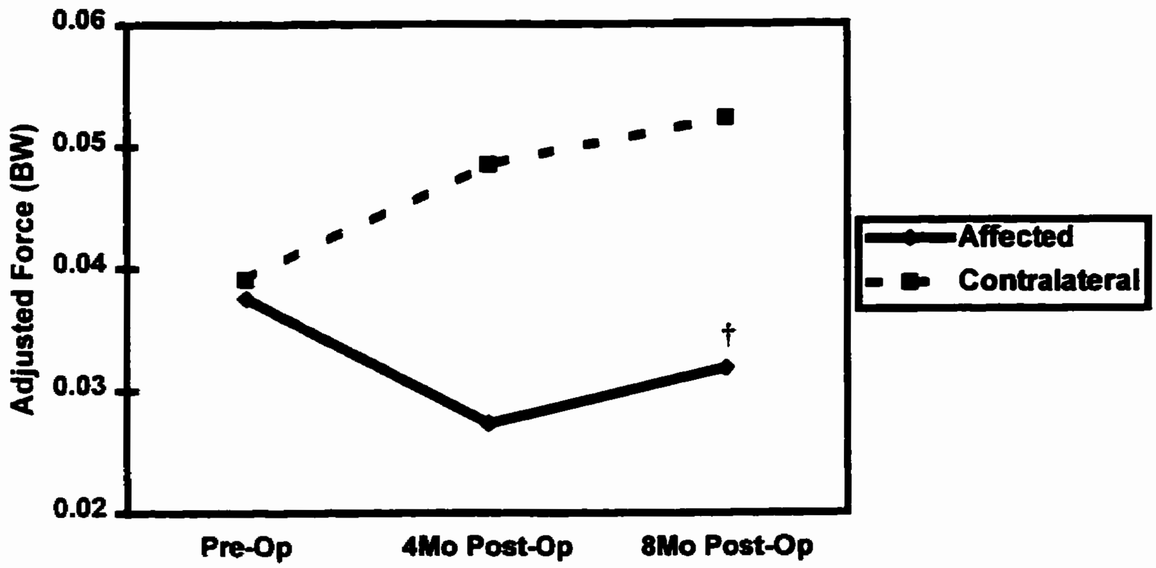


Fig. 40. Peak medial shear of the GRF, adjusted for speed of walking. † represents a difference from pre-operative to eight month post-operative ($p < 0.05$).

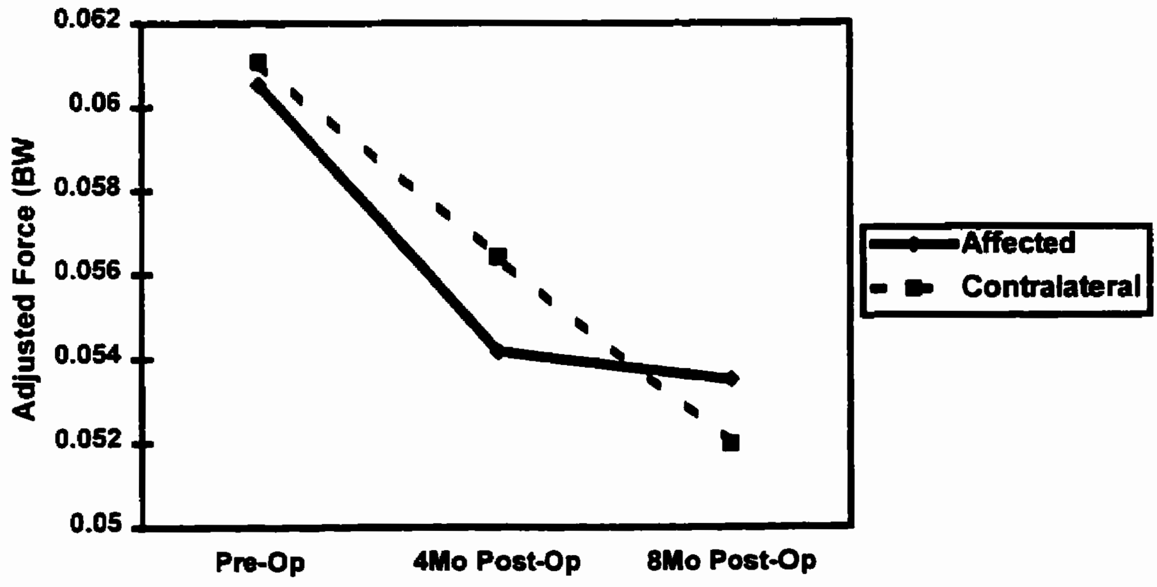


Fig. 41. Peak lateral shear of the GRF, adjusted for speed of walking.

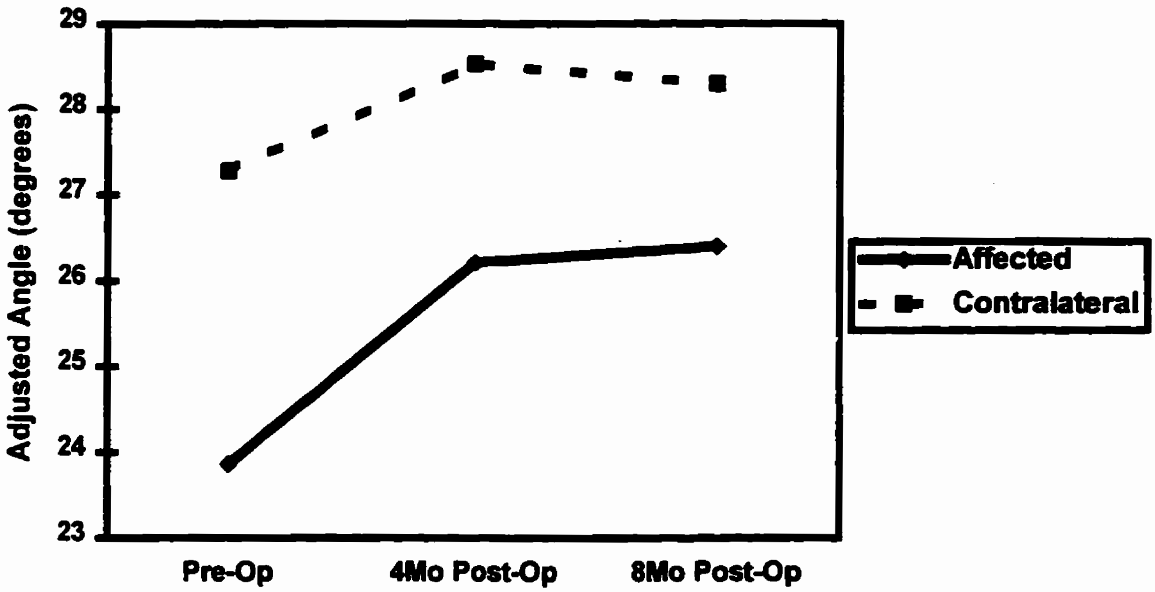


Fig. 42. Peak hip flexion angle during gait, adjusted for speed of walking.

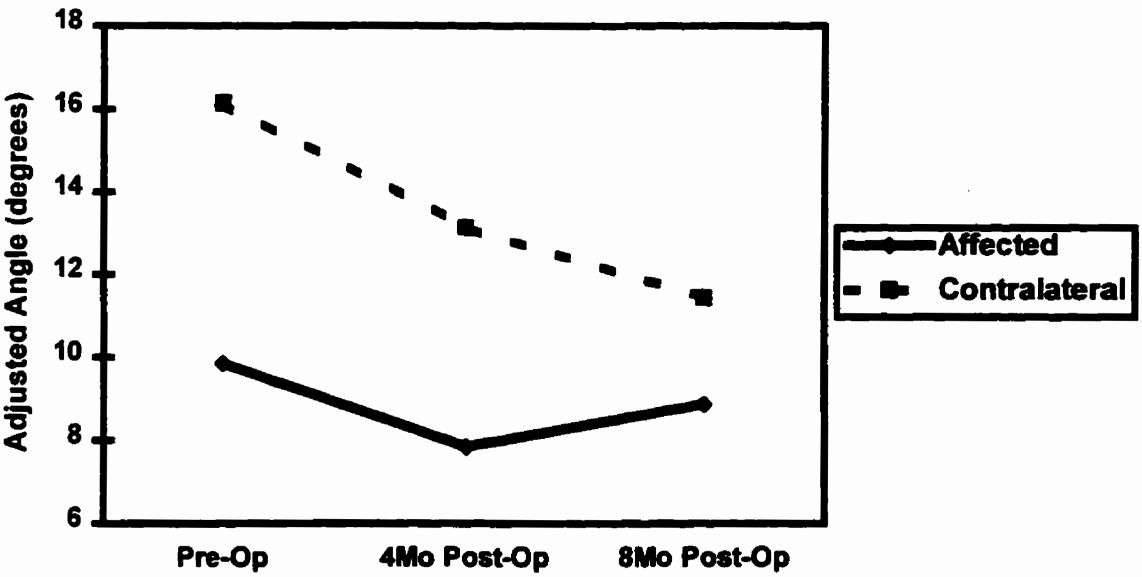
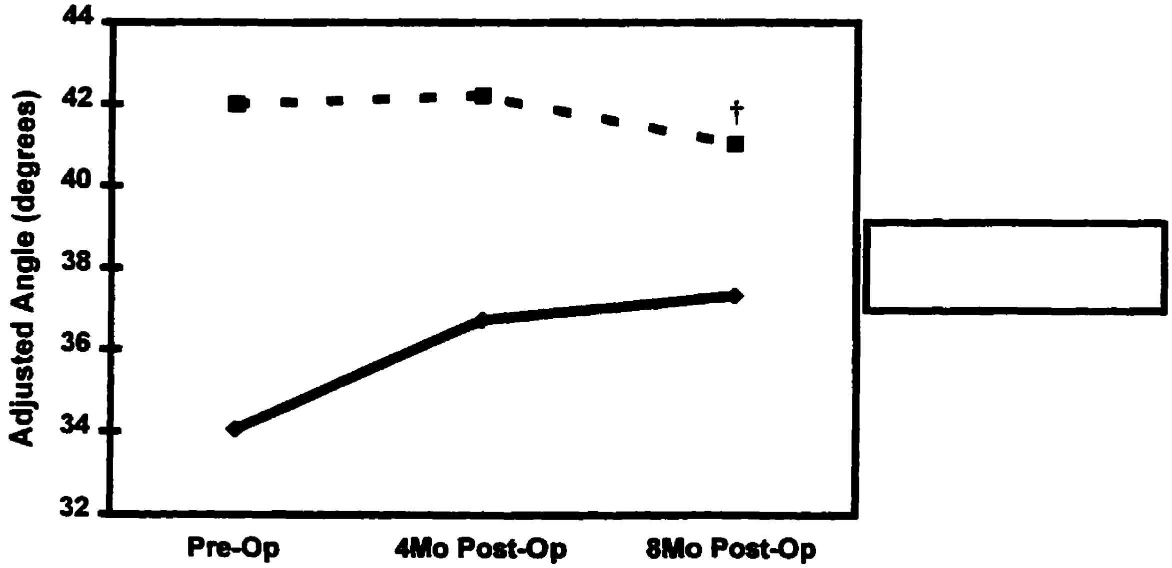


Fig. 43. Peak hip extension angle during gait, adjusted for speed of walking. The affected hip extension angle was significantly smaller than the contralateral extension angle.



Adjusted Angle (degrees)
44
42
40
38
36
34
32
†
Pre-Op
4Mo Post-Op
8Mo Post-Op

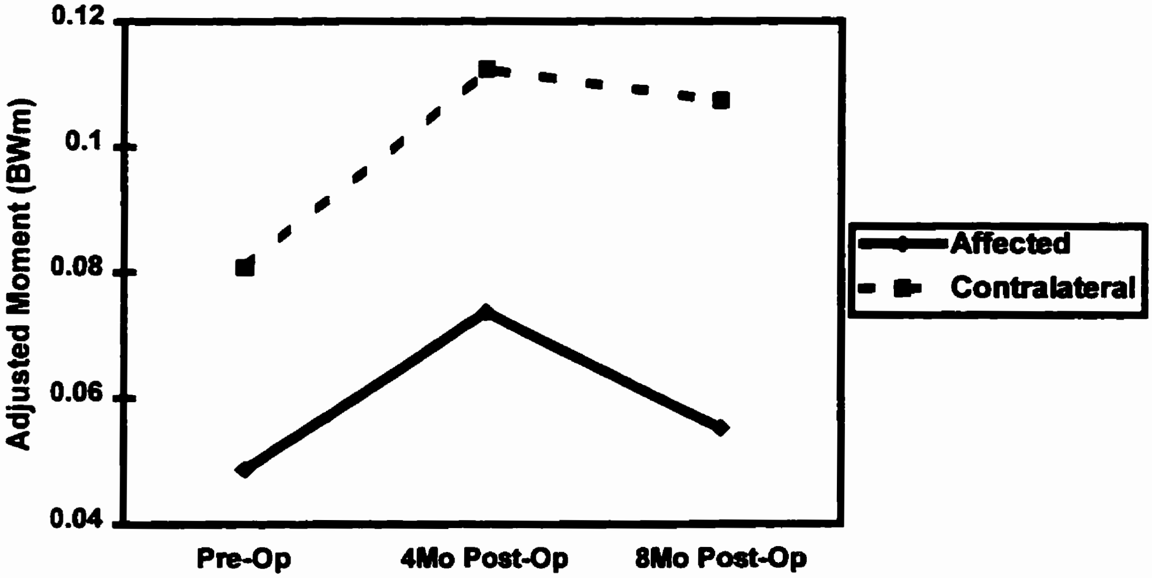


Fig. 45. Peak ankle plantar flexion moment, adjusted for speed of walking. The moment on the affected side was significantly smaller than the contralateral side.

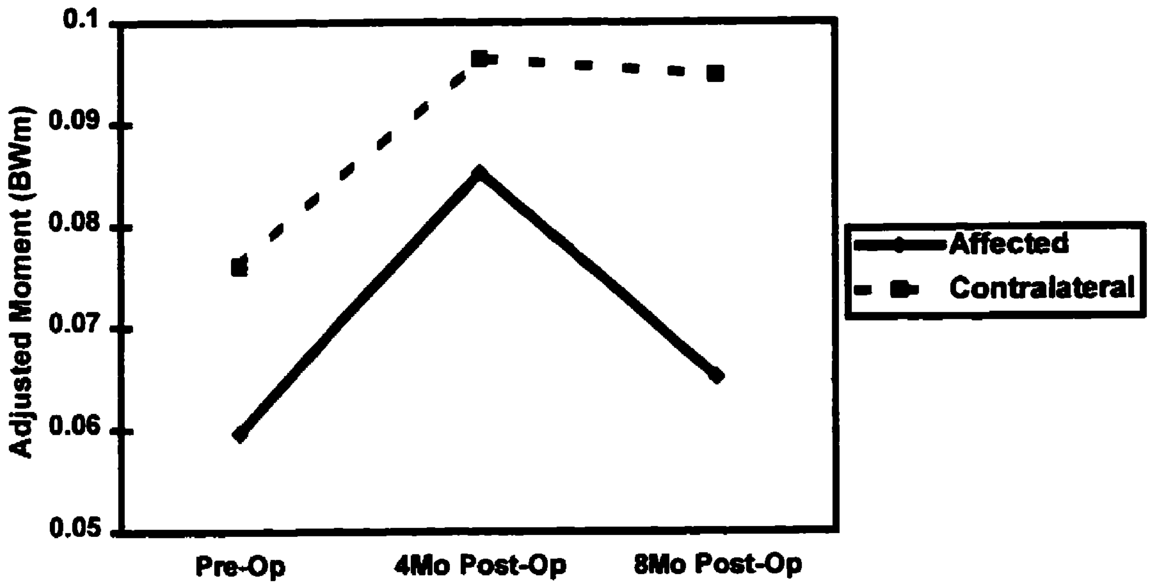


Fig. 46. Peak knee flexion moment, adjusted for speed of walking.

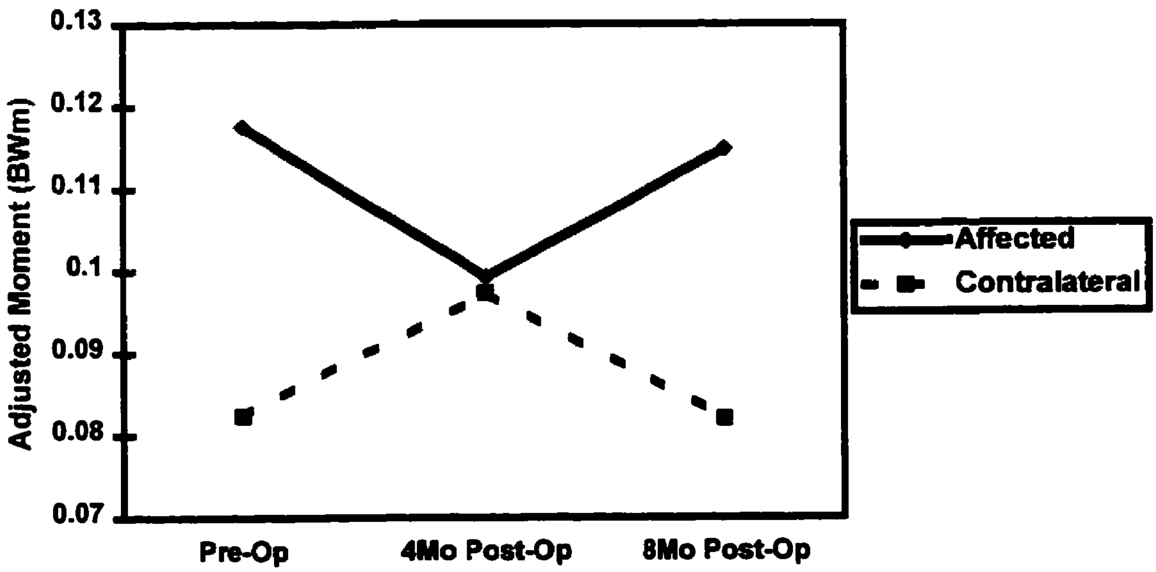


Fig. 47. Peak knee extension moment, adjusted for speed of walking.

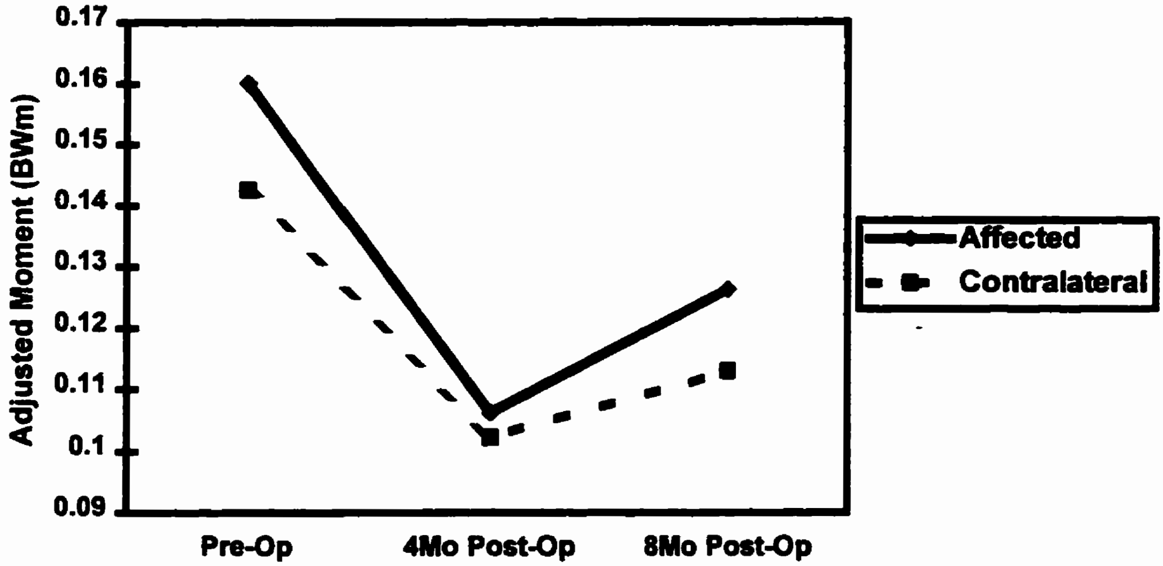


Fig. 48. Peak hip flexion moment during gait, adjusted for speed of walking. The adjusted hip flexion moment was significantly larger on the affected side than the contralateral side ($p < 0.05$).

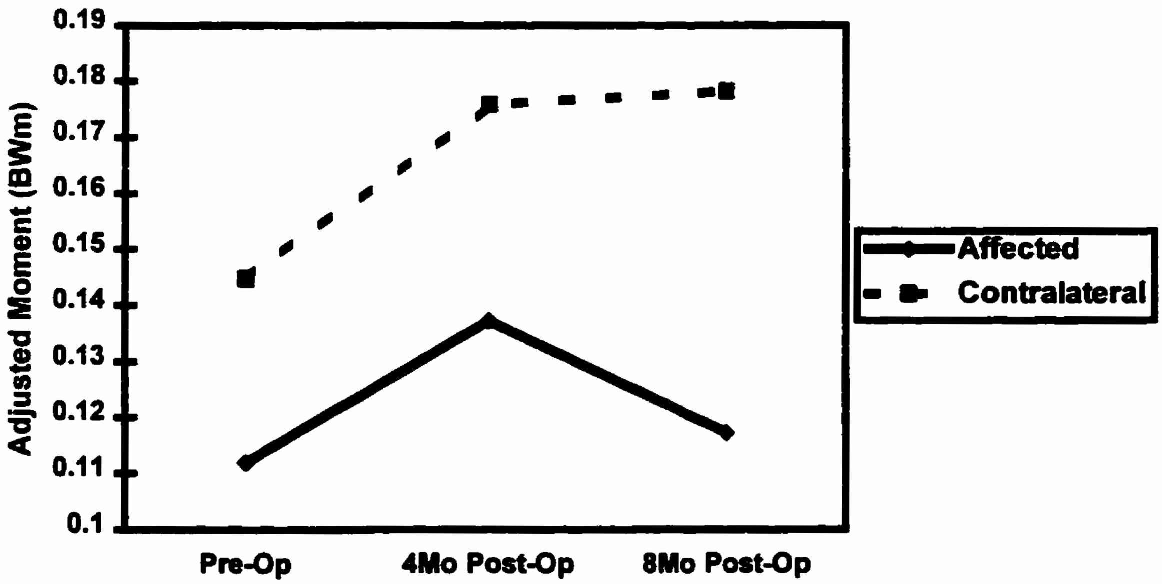


Fig. 49. Peak hip extension moment during gait, adjusted for speed of walking. The adjusted affected moment was smaller than the contralateral adjusted moment ($p = 0.07$).

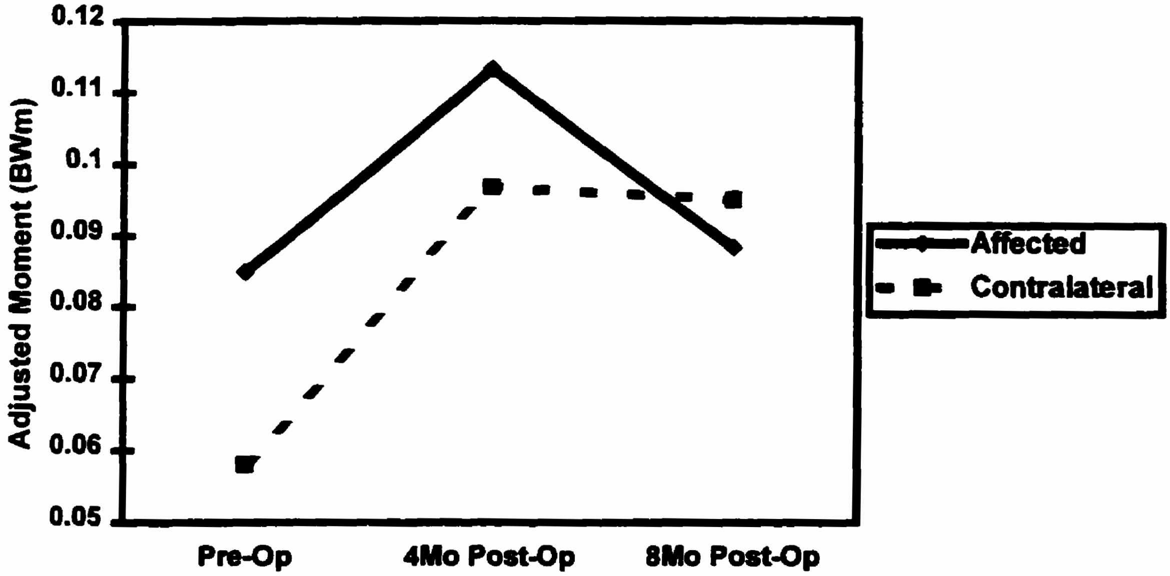


Fig. 50. Peak hip abduction moment during gait, adjusted for speed of walking.

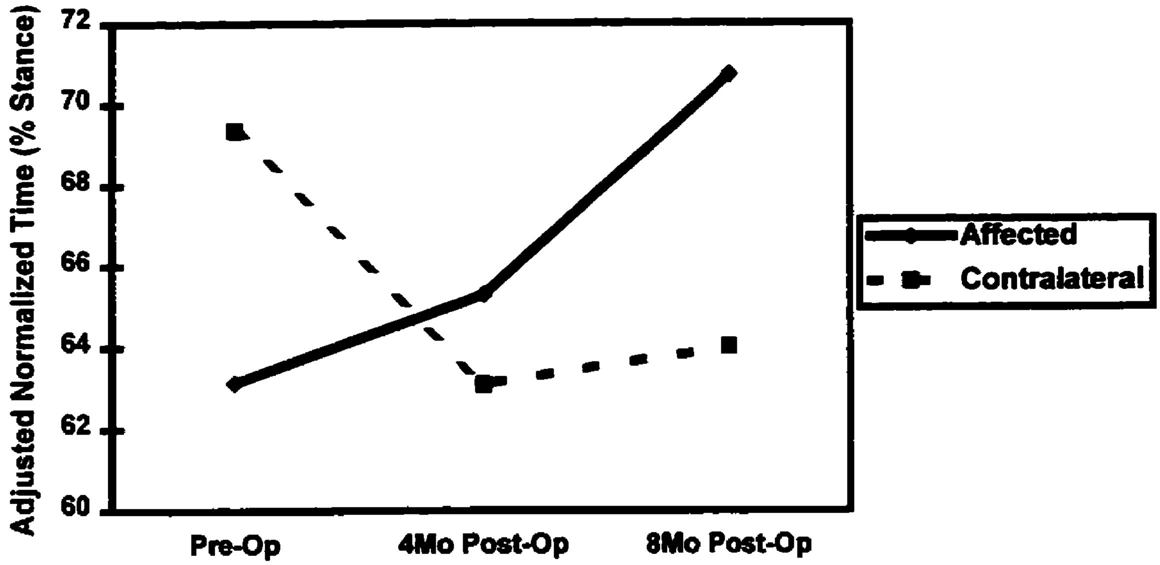


Fig. 51. Gluteus medius duration of activity during stance, adjusted for speed of walking.

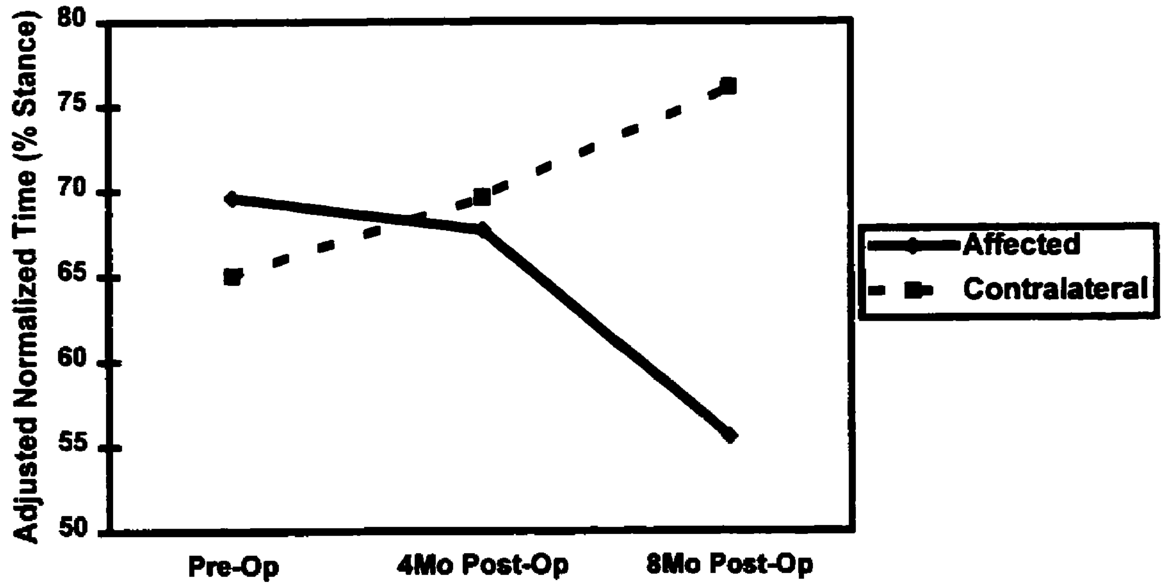


Fig. 52. Tensor fascia lata duration of activity during stance, adjusted for speed of walking.

Chapter 5

Cane results

5.1 Analysis of variance

The effect of a cane on THR gait was tested using a three way ANOVA, with the main effects being session (pre-operative, four months post-operative, and eight months post-operative), limb (affected and contralateral), and cane (cane and no cane).

5.1.1 Walking speed

The free speed of walking increased over time, with the increase being significant between pre-operative and eight months post-operative (Fig. 53). Patients using a cane walked an average of 3% slower than the same patients without a cane ($p = 0.07$). Our patients walked 1% slower when they walked with their affected limbs facing the cameras, than with their contralateral limbs facing the cameras ($p = 0.02$).

5.1.2 Lateral torso sway

The lateral torso sway showed no significant differences between sessions or limb main effects. The use of a cane did not significantly influence the degree of torso sway (Fig. 54).

5.1.3 Ground reaction force

Neither the landing peak (Fz1) or the pushoff peak (Fz2) of the vertical GRF curve differed between testing sessions, but there were main effects for both the limb and cane effects. The results of the simple effects analysis for these two variables are illustrated in Fig. 55 and Fig. 56. The landing peak (Fz1) and pushoff peak (Fz2) for the affected limb were significantly smaller than for the contralateral limb with and without the use of a cane. The use of a cane significantly reduced both peaks for the affected limb but not for the contralateral limb.

The minimum point between the two peaks of the vertical GRF curve (Fzmin) was significantly smaller at four and eight months post-operatively than pre-operatively. The use of a cane significantly decreased the Fzmin, but this effect was only present on the affected side. Without a cane, the affected and contralateral limbs had a similar Fzmin, but the affected side was significantly decreased with a cane.

There was a significantly smaller yield range (FzR1) and pushoff range (FzR2) for the affected side than the contralateral side. Both ranges tended to increase after surgery, but these differences were not significant. For the yield range, there was a significant interaction effect between session and cane, but the simple main effects test showed no differences. The pushoff range had an interaction effect between limb and cane. Only the affected limb pushoff range decreased, moderately, with the use of a cane ($p = 0.07$) and was significantly smaller than the contralateral limb with or without a cane.

The anterior and posterior shear components of the GRF were significantly smaller on the affected limb than the contralateral limb. There was a significant interaction for the anterior shear between the limb and session and the limb and cane effects. Simple effects showed a significant increase in anterior shear for the affected limb over the eight month protocol, and no change in the anterior shear on the contralateral side (Fig. 57). The affected limb had a significantly smaller anterior shear than the contralateral limb pre-operatively, but by four months post-operatively, the difference was no longer significant. The simple effects test for the limb versus cane interaction showed a significant reduction in the anterior shear on the affected side with the use of a cane, and no effect of a cane on the contralateral side (Fig. 58). The posterior shear component was 3% smaller with the use of a cane than without ($p = 0.03$).

The ANOVA results for the medial shear peak of the GRF are illustrated in Fig. 59. The medial shear peak had a significant session effect in which the peak was significantly greater by eight months post-operatively than pre-operatively, but the simple effects model revealed that the only increase was in the contralateral limb and not the affected limb. There was a significant main effect of the contralateral limb having a greater medial peak than the affected peak. In the simple effects model, both four and eight month post-operative time points had a significantly larger medial peak on the contralateral side. The lateral shear peak was decreased significantly by 15% with the use of a cane, but there was no significant difference across time points or between limbs.

5.1.4 Hip kinematics

The maximal hip flexion-extension angles did not show a significant effect of cane. However, the hip extension angle tended to be larger with the use of a cane ($p = 0.06$). The range of motion between the maximum hip flexion and extension had both a significant limb effect (affected was less than contralateral) and session effect (pre-operative was less than either four or eight months post-operative). Fig. 60 illustrates the interaction effect between limb and session. The affected limb had a significantly smaller hip range of motion at all time points, but was largest pre-operatively. The range of motion for the affected hip increased significantly after surgery, but the hip range of motion for the contralateral hip remained stable.

5.1.5 Joint moments

Sagittal plane joint moment results were different for the hip, knee, and ankle joints. The ankle plantar flexion moment and the hip extension moment were both significantly smaller on the affected side than the contralateral side. At the knee, the extension moment was significantly larger on the affected side than the contralateral side. The peak knee flexion moment was smaller on the affected side than the contralateral side, but the difference was not significant ($p = 0.08$). The use of a cane decreased the knee extension moment ($p = 0.02$) and the hip flexion moment ($p = 0.06$).

The maximal hip abduction moment showed no differences over time, but there was a significant interaction effect between the limb and cane effects. The use of a cane (Fig. 61) decreased the hip abduction moment on the affected limb ($p = 0.02$), but did not significantly increase the hip abduction moment on the contralateral limb ($p = 0.14$). Without a cane, the hip abduction moment on the affected limb was larger than the contralateral limb ($p = 0.07$), but with the use of a cane, the hip abduction moment on the affected side was smaller than the contralateral side ($p = 0.07$).

5.1.6 Hip abductor muscle duration

The use of a cane significantly influenced the activity of hip abductor muscles. The duration of activity during the stance phase for gluteus medius is illustrated in Fig.

62. There was a significant interaction between the cane and session effects. At eight months post-operative, the use of a cane significantly decreased the duration of gluteus medius, but pre-operatively and at four months post-operatively there were no differences between trials with and without the use of a cane. The use of a cane significantly reduced the duration of tensor fascia lata (Fig. 63).

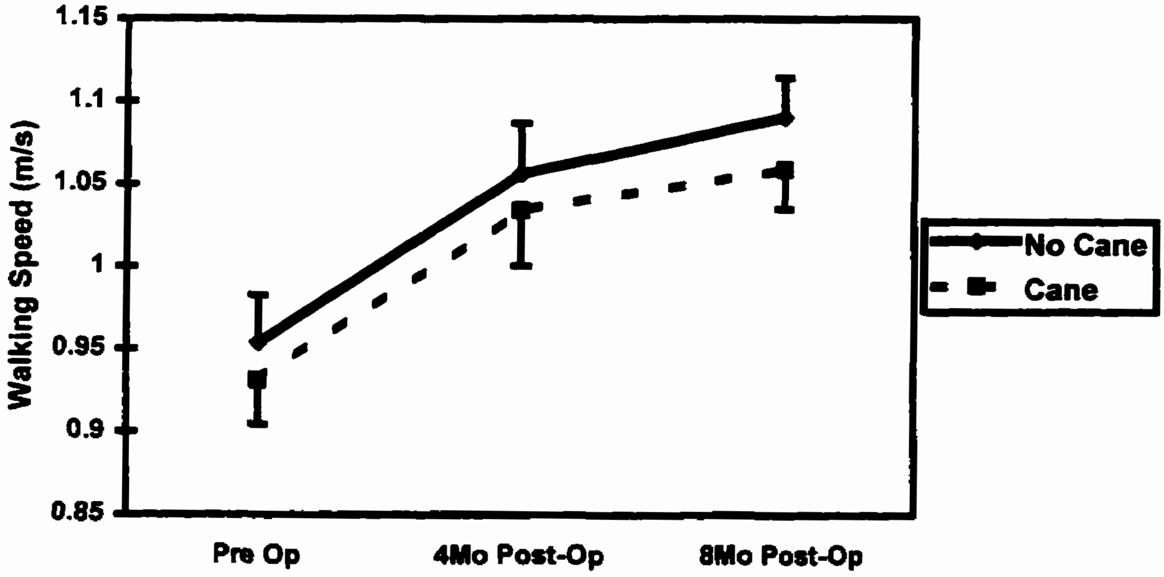


Fig. 53. Free walking speed for THR patients with and without the use of a cane. Eight month post-operative walking speed was significantly larger than pre-operative speed. A cane reduced the speed by 3% ($p = 0.07$).

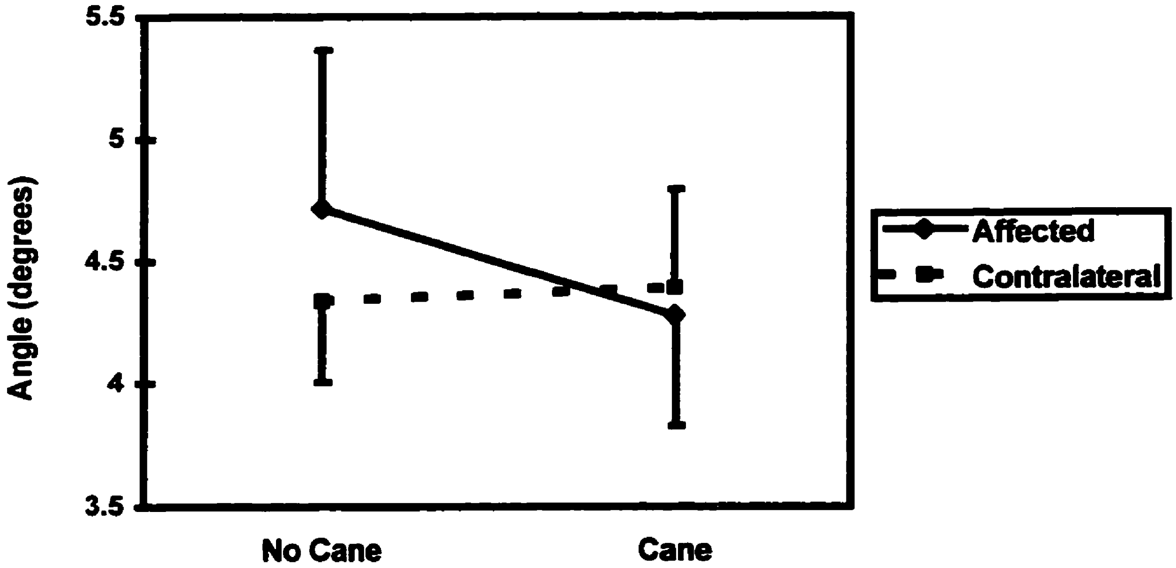


Fig. 54. Lateral torso sway for THR patients for the combined effect of the three testing sessions.

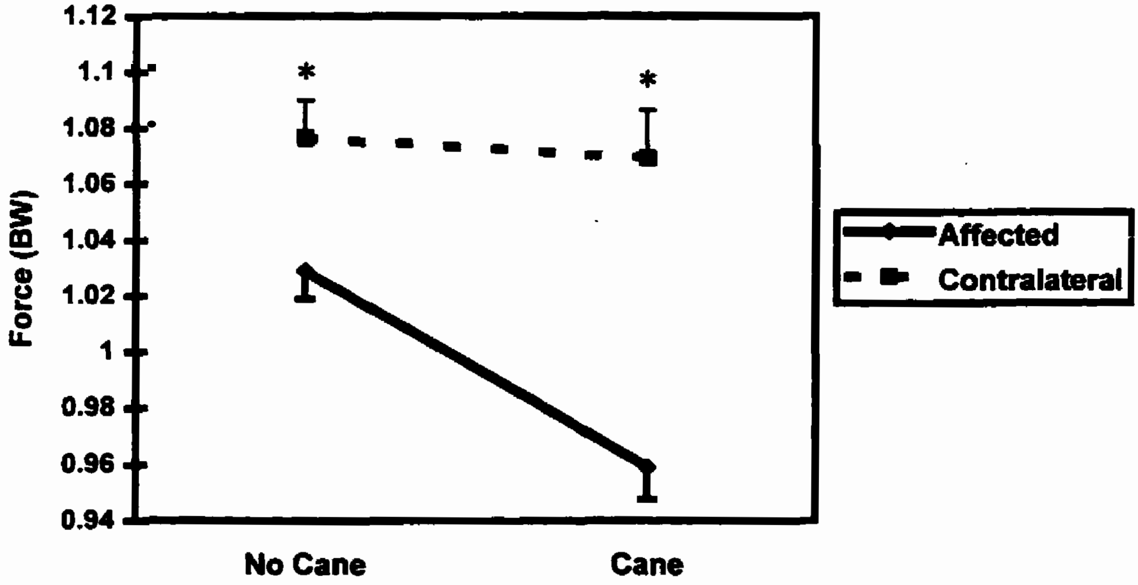


Fig. 55. Landing peak of the vertical GRF curve for THR patients. Asterisks (*) represent differences between affected and contralateral limbs ($p < 0.05$). A cane significantly reduced the landing peak of the affected limb.

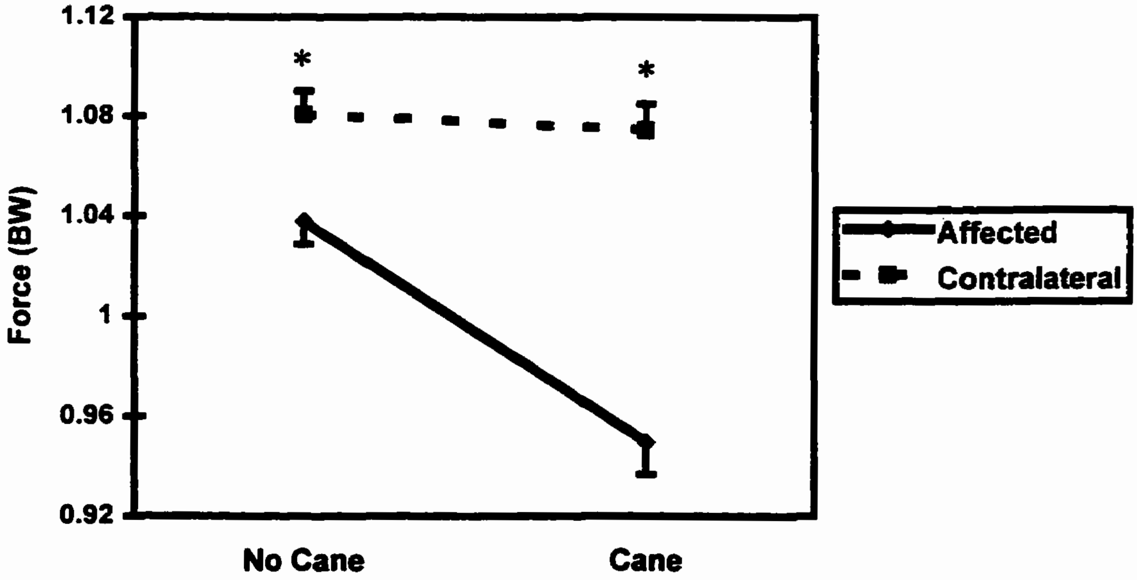


Fig. 56. Pushoff peak of the vertical GRF curve for THR patients. Asterisks (*) represent differences between affected and contralateral limbs ($p < 0.05$). A cane significantly reduced the pushoff peak of the affected limb.

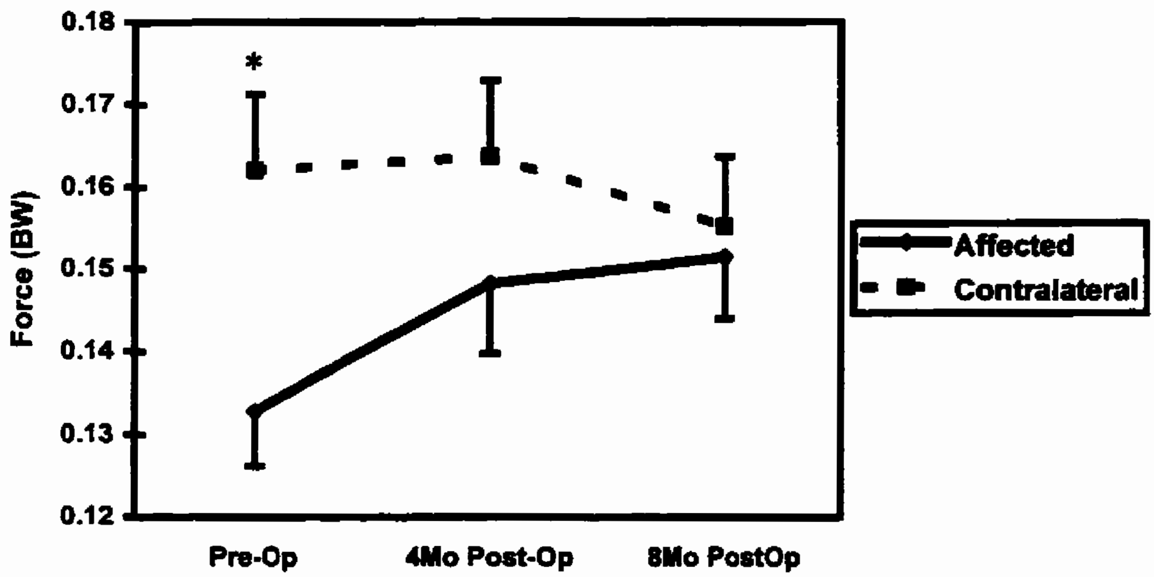


Fig. 57. Anterior shear peak of the GRF for THR patients for the combined effect of trials with and without the use of a cane. The asterisk (*) represents a difference between affected and contralateral limbs ($p < 0.05$). The anterior shear increased significantly for the affected limb across testing sessions.

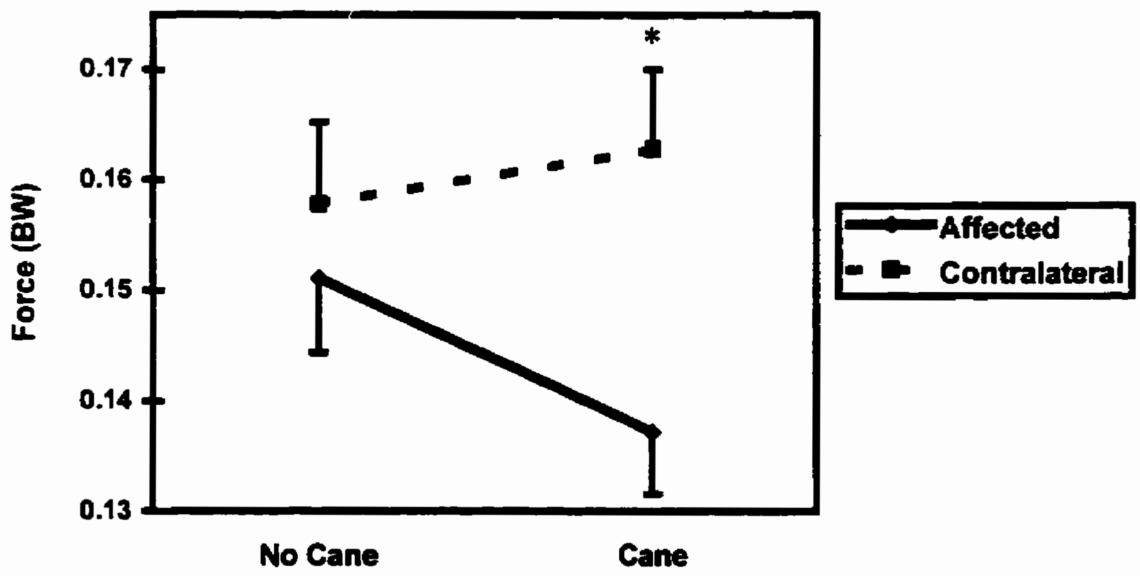


Fig. 58. Anterior shear peak of the GRF for THR patients for the combined effect of the three testing sessions. The asterisk (*) represents a difference between the affected and contralateral limb ($p < 0.05$). A cane decreased the anterior shear significantly for the affected limb.

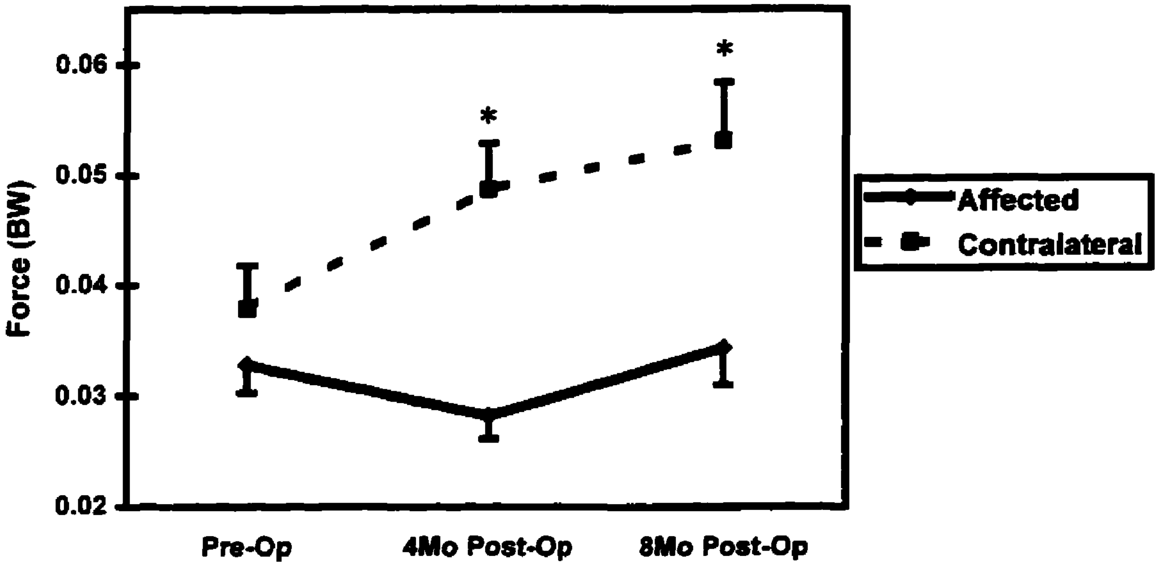


Fig. 59. Medial shear peak of the GRF for THR patients for the combined effect of trials with and without the use of a cane. Asterisks (*) represent differences between the affected and contralateral limbs. The contralateral limb increased significantly across testing sessions.

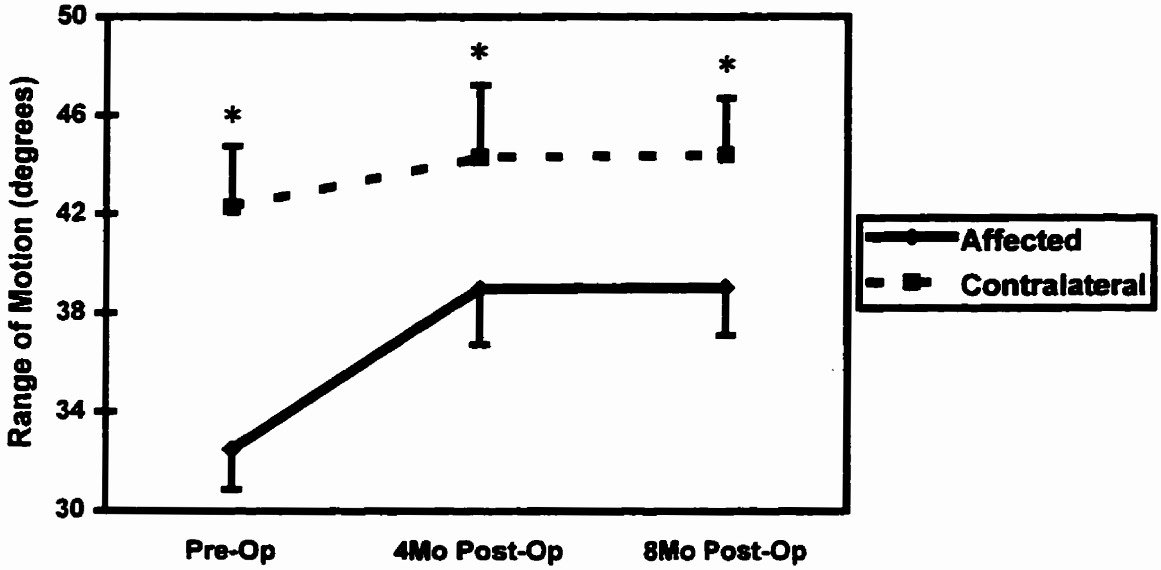


Fig. 60. Flexion-extension range of motion of the hip during gait for THR patients for the combined effect of trials with and without the use of a cane. Asterisks (*) represent differences between the affected and contralateral limbs ($p < 0.05$). The affected hip range of motion increased significantly across testing sessions.

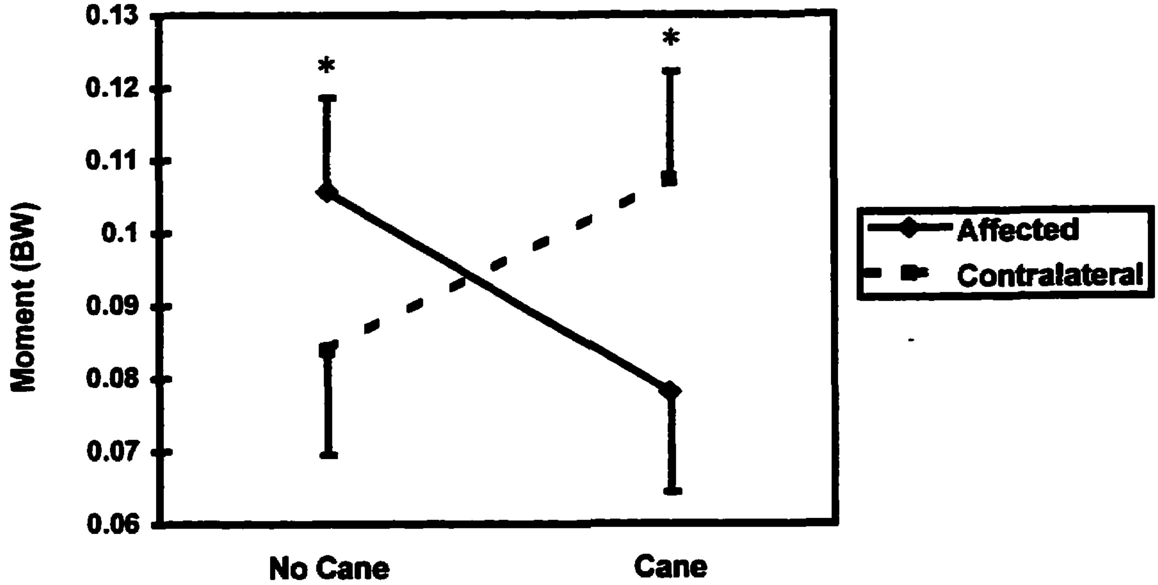


Fig. 61. Maximum hip abduction moment for THR patients. Asterisks (*) represent differences between affected and contralateral limbs ($p < 0.07$). A cane significantly decreased the affected hip abduction moment.

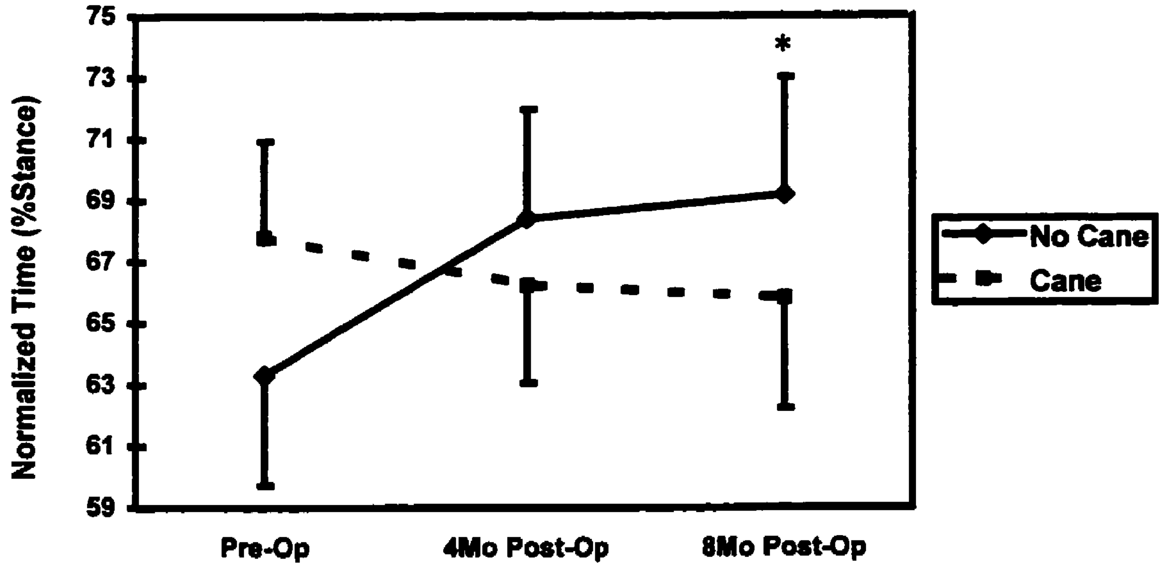


Fig. 62. Duration of activity of the gluteus medius during the stance phase for THR patients for the combined effect of affected and contralateral limbs. The asterisk (*) represents a difference between affected and contralateral limbs ($p < 0.05$).

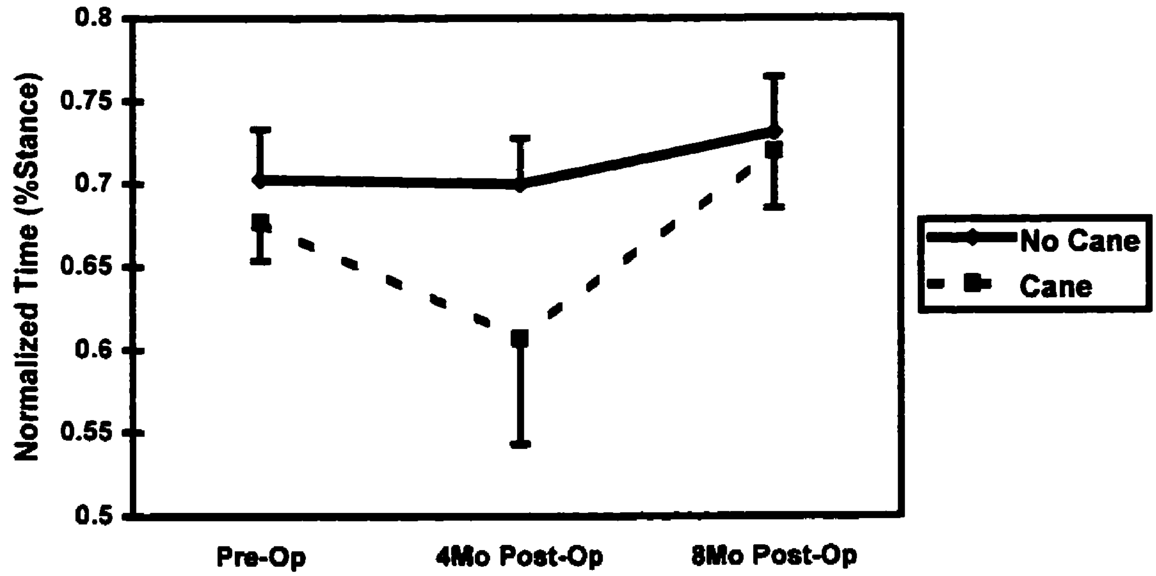


Fig. 63. Duration of activity of the tensor fascia lata during the stance phase for THR patients for the combined effect of affected and contralateral limbs. The duration of activity of tensor fascia lata was significantly reduced with the use of a cane.

Chapter 6

Discussion

6.1 Clinical

The patients involved in our study were selected for having unilateral osteoarthritis of the hip, with no other joint disorder that would affect their gait. Thus, our patients did not have multiple joint involvement or other neuromuscular disorders commonly found in pre-operative THR patients. Furthermore, we selected patients prior to their first hip surgery, eliminating all patients who were returning for revision surgery. All patients in this study complained of pain and limited function pre-operatively, but may have been healthier than the average THR candidate. We chose these inclusion and exclusion criteria to ensure the gait changes seen were due to the THR surgery and not the influence of a separate pathology. These criteria, as well as logistical issues in recruiting from multiple sites, put severe limitations on the number of patients eligible for our study, and in the 17 months of recruitment from three hospitals in Calgary, only fourteen volunteer patients entered into this study.

The clinically measured ranges of motion for the affected hip increased after surgery. The significant asymmetry found pre-operatively was no longer significant by eight months after surgery for both the abduction and extension angles, and was greatly diminished for the flexion angle. Mont et al. [1993] reported an increase in flexion angle from 71° pre-operatively to 99° by three to seven years post-operatively. Our patients increased in flexion from 84° pre-operatively to 96° at eight months post-operatively. The range of flexion for the contralateral limb was constant over time at 104°. Our patients may not have reached the plateau in flexion by eight months post-operative.

Hip muscle strength increased for the affected leg after surgery for most THR patients in our study. The flexion and extension strength was graded normal (grade 5) by eight months post-operative in 75% of patients. The abduction strength was graded somewhat lower at eight months post-operative, but there was an improvement in strength from the pre-operative strength for most patients. Pre-operative flexion and extension strength was graded a 4 or better in all but one patient, indicating a relatively strong sample of THR patients pre-operatively. However, there was an asymmetry in most patients, as the contralateral hip strength was graded a 5 in most patients pre- and post-operatively.

Pre-operatively, nearly half of the patients could not abduct their hip through full range of motion under moderate resistance. We did not measure strength quantitatively, but it is questionable as to whether these patients had adequate strength in their hip for normal daily activities. Delp et al. [1996] estimated that the moment required to stabilize the hip during one-legged standing in a 75 kg male was 51 Nm, and that the maximum abduction moment is normally 88 Nm. The grading scale used considers a grade of 5 to be normal. Thus, grades of less than 5 in abduction strength might correlate to strengths less than or near the threshold required for one-legged stance. Murray et al. [1981] measured an increase in maximum hip abduction moment from 49 Nm to 74 Nm for men and from 29 Nm to 50 Nm for women without a trochanteric osteotomy by 2 years after

THR surgery. During gait, THR patients may need to employ dynamic gait strategies to reduce the demand on the abductors to maintain balance during the stance phase of the affected limb.

6.2 No Cane

6.2.1 Walking speed

Freely selected speed of walking has been shown to decrease with age [Himann et al., 1988]. For normal subjects aged 61 years to 68 years, the free speed of walking is 1.22 m/s to 1.33 m/s [Cunningham et al., 1982; Waters et al., 1988]. Patients for this study, with an average age of 63.7 ± 8 years, had a pre-operative walking speed of 0.95 m/s. Although this speed was slower than age-matched normal subjects, it was considerably faster than the speeds reported for typical pre-operative patients (0.45 m/s to 0.60 m/s) [Brand and Crowninshield, 1980; McBeath et al., 1980; Murray et al., 1981; Olsson et al., 1985]. The difference between our results and the reported pre-operative results may be due to the patients selected for our study. Patients in similar gait studies were not selected for uniarticular osteoarthritis. By four months post-operative, patients in our study walked 1.05 m/s, and they walked 1.09 m/s by eight months post-operative. These post-operative speeds are also greater than the one year post-operative results of other studies (0.77 m/s to 0.98 m/s) [Brand and Crowninshield, 1980; McBeath et al., 1980; Olsson et al., 1985]. Although the speed increase after surgery was statistically significant for our patients, the magnitude of the increase was only 15%. Reports on the effect of walking speed on gait variables suggest that speed affects gait kinematics and kinetics, but the magnitude of speed increase is not likely to affect gait variables significantly [Andriacchi et al., 1977; Crowninshield et al., 1978]. It is possible that the patients in our study did not reach their free walking speed within the 6 m walkway, but since all tests were performed on the same walkway, the between test comparisons are likely to be accurate, and their self-selected walking speeds were markedly faster than patients in other studies.

Our patients showed a 1% difference in walking speeds between trials in which the affected limb was facing the cameras versus contralateral limb trials. Patients walked the length of the walkway first with their right limb facing the cameras, then their left. Of the 14 patients involved in our study, 6 had right hip involvement, and 8 had left hip involvement. Thus, the affected limb trials were collected first for some patients and second for others. Every assurance was made to encourage patients equally during all trials to ensure reproducible gait patterns. Therefore, the difference in speeds of walking between affected and contralateral limb trials is likely to be a random effect, not representative of THR gait.

6.2.2 Lateral torso sway

Patients in our study did not demonstrate a significant asymmetry in the degree of torso sway over the affected and contralateral limbs. Torso sway is a measure of limp, thus our patients did not demonstrate a limp over the affected limb, as seen in most pre-operative patients [Mont et al., 1993; Olsson et al., 1985]. Researchers generally measure the degree of limp clinically. Murray et al. [1981] measured limp as the lateral motion of the head, however, this variable may be measuring head tilt or other motions. Our method of measuring the angular motion of the trunk is sensitive in measuring quantitatively the degree of limp in the torso. Murray et al. [1981] reported a significant decrease in the lateral head motion of THR patients by 6 and 24 months post-operative, which suggests that the limp decreased in their patients. However, torso sway did not change by eight months post-operative in our patients.

6.2.3 Ground reaction force

The pre-operative landing peak of the vertical GRF on the affected side (1.04 BW) was 3% smaller than on the contralateral limb, and the pushoff peak on the affected limb (1.03 BW) was 2% smaller than the contralateral limb. These asymmetries were smaller than the 6% to 8% differences reported in the literature [Long et al., 1993; Olsson et al., 1985], possibly as a result of the patient sample in our study having no lurch. Long et al. [1993] reported an 11% increase in the magnitude of the landing peak and a 9% increase in the pushoff peak by one year post-operative, and Olsson et al. [1985] reported a 5% increase in the peak vertical force by one year post-operative. The magnitude of the vertical GRF peaks did not differ pre-operatively to eight months post-operative. When the speed of walking was controlled by using speed as a covariate in the analysis, the peaks of the vertical GRF curve decreased after surgery. The increase in vertical GRF reported by Long et al. and Olsson et al. may have been related to an increase in walking speed in their patients.

Pre-operatively, both the anterior and posterior shear peaks of the GRF were 13% and 10% smaller, respectively, on the affected limb than the contralateral limb. The posterior peak remained at 7% asymmetry up to eight months post-operatively. The anterior peak, however, increased significantly for the affected limb, to restore symmetry by four months post-operatively. Long et al. [1993] reported that the anterior peak had a 21% difference between limbs pre-operatively and a 14% difference at one year post-operative, and that the posterior peak was 10% different pre-operatively and 8% different at one year. The anterior peak, when adjusted for speed, remained stable for the affected limb after surgery, and the posterior peak decreased significantly. This suggested that the increase in the anterior shear was a function of an increase in the speed of walking, and if patients walked at a constant speed, the posterior peak would actually decrease after surgery in contrast to what is reported in the literature.

The maximal medial shear peak of the GRF, which typically occurred during the weight acceptance phase of gait, increased significantly after surgery for the contralateral limb but not the affected limb. When controlling for speed, the contralateral medial peak remained stable, and the affected medial peak decreased after surgery. The maximal lateral shear force tended to decrease after surgery with or without speed as a covariate, suggesting a decrease in the lateral motion of the center of mass of the subject.

Walking speed has been correlated with GRF over a wide range of speeds [Andriacchi et al., 1977; Crowninshield et al., 1978]. Our results suggest that even subtle increases in walking speed corresponded to significant changes in the GRF. If restricted to a constant pre-operative speed, our patients may have shown a decrease in the peak GRF after surgery.

6.2.4 Hip kinematics

The hip range of motion during gait increased by 5° for the affected limb by eight months post-operatively, but the angle at eight months remained significantly less than the contralateral limb range of motion. Long et al. [1993] reported a 10° increase in hip arc of motion after one year post-operatively, and the post-operative range was at a normal level of 37°. Murray et al. [1981] also presented a 10° to 15° increase in functional range of motion by six months post-operatively, but the post-operative range (30°) did not reach normal (46° for men and 40° for women). Our patients had a contralateral hip range of 42° and a post-operative affected hip range of 38°. Long et al. [1993] reported that 70% of the increase in range of motion was attributed to an increase in hip extension, whereas, 64% of the range for our patients was attributed to an increase in flexion angle.

The maximum extension angle during gait for our patients increased after surgery on the affected side, but decreased on the contralateral side. To maintain a more nearly normal walking speed, the contralateral limb may have been extending further to increase the step length on the contralateral side to compensate for the decreased hip extension on the affected side. As the affected hip extended further post-operatively, the contralateral limb no longer had to compensate, and bilateral hip extension became more symmetrical.

When we controlled for walking speed (ANCOVA analysis), the hip extension angle on the affected side did not increase after surgery.

6.2.5 Joint moments

Hip extension moments and ankle plantar flexion moments were smaller for the affected limb than the contralateral limb, but the knee extension moment was larger for the affected limb than the contralateral limb. The affected limb used greater knee and less hip and ankle moments for propulsion. With smaller hip extension moments on the affected side, patients may have reduced the sagittal plane moments to reduce the muscle activity in the large flexion-extension muscles of the hip to decrease hip compressive loads.

Typically, researchers have attributed the reduction in hip compressive loads to a reduction in hip abduction moment on the affected side [Delp et al., 1996]. This interpretation may be the result of studies showing a larger magnitude of moment in the frontal plane than the sagittal plane [Crowninshield et al., 1978]. The patients in our study did not have a significant reduction in hip abduction moment on the affected side. This difference from other studies may have been related to the type of patient we studied since the patients in our study had larger hip sagittal plane moments than frontal plane moments. Therefore, the patients in our study may have decreased their hip compressive loads by only decreasing their sagittal plane moments and not their hip abduction moment.

Typically, the hip joint moments are the least reliable of the variables studied in gait analysis. The source of error rests primarily with the location of joint centers. Small errors in hip joint center location could lead to errors in moment arm and therefore moment calculation errors. In order to determine the precise location of the hip joint center, we would be required to use invasive methods, such as x-rays. This was not feasible for our study, and we were forced to use anatomical landmarks to estimate the hip center. Between subject comparisons, as made in this paper, could still be made since the same method was used for all subjects.

The results of the ANCOVA on joint moment variables, with speed as a covariate were similar to the results of the ANOVA, with the exception of the hip flexion moment. This suggests that the increase in walking speed seen post-operatively was not of a sufficient magnitude to affect joint moments.

6.2.6 Hip abductor muscle activity

The duration of activity during the stance phase of gluteus medius and tensor fascia lata tended to be different with respect to testing session, however, our data were not powerful enough to show any significant differences in muscle activity. The gluteus medius duration increased by 11% of the stance phase by four months post-operative on the contralateral side but remained stable on the affected side. Tensor fascia lata duration decreased by 7% of the stance phase by four months on the contralateral side and increased by 6% of the stance phase on the affected side. This study did not control for surgical approach, but the surgical approach most commonly used for these patients was an anterolateral approach [Hoppenfeld and deBoer, 1994], which accesses the hip through the intermuscular plane between gluteus medius and tensor fascia lata. The anterior part or the entire gluteus medius is cut for adduction of the hip during surgery. These two muscles play a synergistic role for hip abduction. The increased duration of the affected tensor fascia lata at four months may have been a compensation for the surgically damaged gluteus medius. With the speed as a covariate in the analysis, the affected gluteus medius duration increased after surgery while the tensor fascia lata duration decreased.

The timing of hip abductor muscle activity has been presented by many researchers for the normal population. The gluteus medius muscle has been reported to be active from pre-stance to approximately 50% of stance [Perry, 1992; Jaegers et al., 1996]. Cessation time for the gluteus medius has been presented as later [Kleissen, 1990] and earlier [Lyons et al., 1983] than mid-stance. Our data suggest that for most THR patients, the gluteus medius was active from at least 10% prior to foot strike to between 60% and 80% of stance. Tensor fascia lata in normal populations is active from foot strike to the end of mid-stance [Perry, 1992; Jaegers et al., 1996]. In a young normal population, the tensor fascia lata is reported to be active for only a short burst during mid-stance [Perry, 1992; Lyons et al., 1983]. Our patients had tensor fascia lata activity from pre-stance to between 60% and 80% of stance in most patients. Long et al. [1993] reported that 3 out of 18 pre-operative THR patients had no gluteus medius activity, and five patients had continuous firing of the tensor fascia lata. By two years post-operative, all pre-operatively abnormal patients returned to a normal firing pattern, but four hips became abnormal.

None of our patients had either of these two gait abnormalities pre-operatively, but two patients had a continuous firing of tensor fascia lata by eight months post-operative.

6.3 Cane

All of the patients in our study were able to walk without the use of a cane or other walking aid at all testing sessions. Our patients walked 3% slower with the use of a cane than without a cane pre- and post-operatively. Brand and Crowninshield [1980] reported a 23% lower speed of walking in pre-operative THR patients. Their patients walked without a cane at 0.57 m/s in contrast to our patients who walked at 0.95 m/s. Ely and Smidt [1977] reported no difference in walking speed with the use of a cane in 15 patients with hip disease or hip replacement. Kleissen et al. [1989] reported a case study of a THR patient in which the patient walked 4% slower with the use of a cane. The use of a cane may diminish walking speed, but the magnitude of the decrement appears to be negligible for this sample.

The use of a cane did not significantly affect the torso sway in our patients, although Fig. 40 shows a slight decrease in torso sway over the affected limb. A cane may reduce the limp seen in patients with a moderate to severe limp, but our patients with no limp did not demonstrate this effect significantly.

The use of a cane reduced vertical and shear GRF for the affected limb. The landing and pushoff vertical GRF peaks were reduced 7% and 9% respectively with the use of a cane. These decreases were expected since the cane was in contact with the ground and transmitting load during the stance phase of the affected limb, thereby reducing the load on the limb [Ely and Smidt, 1977]. The cane had no effect on the contralateral limb since it was not in contact with the ground during contralateral stance. The cane significantly reduced the braking shear component and propulsive shear component of the GRFs for the affected limb, possibly by transmitting some of these forces as well. The lateral shear was 15% smaller for both limbs with the use of a cane. The cane may have decreased the lateral oscillation of the center of mass, which could have reduced the lateral shear force component of the GRF.

The effect of a cane on sagittal plane moments was only noticeable in knee extension and hip flexion. The use of a cane reduced these moments for both the affected and contralateral limbs, and may have been a result of the slower speed of walking with the use of a cane.

The hip abduction moment in the affected hip was significantly reduced by 26% with the use of a cane in the contralateral hand. We found, however, that the contralateral hip abduction moment increased by 28% with the use of a cane. An important role of a cane in THR rehabilitation is to reduce the load on the hip and, thereby, allow for healing of the bone and soft tissues. Our results suggest that this reduction was successful on the affected side, but the contralateral side may have experienced an increase in the hip abduction moment. This increase in the hip abduction moment may have a potential deleterious effect on the contralateral limb, possibly accelerating hip disease. Further research is necessary to determine if this effect is ubiquitous for all patients with unilateral hip disease.

The post-operative use of a cane decreased the duration of activity of the hip abductor muscles tested. Pre-operatively, the gluteus medius duration activity was greater with the use of a cane, but this effect was not statistically significant. There was no differential effect between the affected and contralateral hips for either muscle. To produce a larger abduction moment on the contralateral limb than the affected limb with the use of a cane, the hip abductor muscles likely stronger, but we did not estimate the magnitude of force of the hip abductor muscles or the amplitude of EMG signal of these muscles. Our EMG results reflect that the duration of activity of these muscles was not different between sides. The amplitude of the EMG signal may have been different between limbs, but this comparison was not feasible using EMG electrodes over separate muscles and on separate testing sessions. The same muscle contraction could produce different EMG signals depending on the location of the electrodes and the preparation of the skin [Basmajian and DeLuca, 1985]. Long et al. [1989] reported that the amplitude of EMG signal from the gluteus medius activity decreased by 25% with the use of a cane. The EMG electrodes were left in the same locations on the muscle for both tests, and thus this comparison was legitimate.

6.4 Conclusion

This study identified the gait dynamics and hip muscle recruitment patterns associated with pre-and post-operative THR. Many THR patients walk with a lurching gait pattern, which is often associated with weak hip abductor muscles. We examined the effects of the use of a cane in post-operative rehabilitation of THR patients.

The patients selected for our study were selected for having uniarticular osteoarthritis of the hip prior to primary THR. This criteria was chosen to study the effects of THR on gait with no other joint degeneration which may affect gait. This selection criteria led to a study population which did not demonstrate a lurching gait pattern and represented a select portion of the THR population.

Patients involved in this study demonstrated an increase in hip strength, an increase in clinically assessed hip range of motion, and an increase in speed of walking by eight months post-operative. Ground reaction forces were asymmetrical between the affected and contralateral limbs pre-operatively and did not return to symmetry by eight months post-operative. Patients in this study tended to use less hip and ankle moments for propulsion, which may have been a strategy for the reduction in hip compressive loads, thereby reducing pain. The hip abduction moment, however, was not significantly different between limbs or across time points. The use of a cane decreased the hip abduction moment on the affected side, but may have increased the hip abduction moment on the contralateral side. Therefore, the use of a cane reduced hip loading on the healing hip, but may have increased the loads placed on the good hip, which had the pootential to accelerate joint disease in that hip. At four months post-operatively, the duration of the tensor fascia lata muscle crossing the affected hip was increased, potentially, as a compensation for the surgically damaged gluteus medius.

Future studies should determine if the effects seen in this study are generalizable to a larger population of THR patients. A larger sample size of THR patients with a broader etiology would be needed to determine these effects. A study examining the effects of the use of a cane on the affected and contralateral hip loads would clarify the role of a cane in hip disease management. A study relating hip muscle recruitment patterns and gait to surgical approach and rehabilitation protocols would reveal which techniques would most benefit THR patients' gait performace.

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Appendix

INFORMED CONSENT FORM

PROJECT:	Examination of changes in muscles electrical activity and walking patterns associated with the use of a cane following total hip replacement.
INVESTIGATORS:	Deborah Thon, BSc.P.T., Peter Clare, M.A., BSc.P.T. Lauri Kaul, BSc.P.T. (Foothills Hospital Physical Therapy) Gary Hughes, MD, Ronald Zernicke, Ph.D., Stan Ajemian, BSc. (Joint Injury and Arthritis Research Group, Faculty of Medicine) Manuel Hulliger, Ph.D. (Department of Clinical Neurosciences and Medical Physiology, Faculty of Medicine)
LOCATION:	McCaig Centre for Joint Injury and Arthritis Research, University of Calgary, Heritage Medical Research Building, 3330 Hospital Drive N.W., Calgary, Alberta, T2N 4N1
FUNDING AGENCY:	Canadian Orthopaedic Foundation Hip Hip Hooray (Calgary and National)

This consent form, a copy of which has been given to you, is only part of the process of informed consent. It will give you the basic idea of what the research project is about and what your participation will involve. If you would like more detail about something mentioned here, or information not included here, feel free to ask. Please take the time to read this carefully and to understand any accompanying information.

The purpose of this research project is to find out more about how the muscles that are used during walking change as a result of your total hip replacement. We think that because of pain that gradually increased in the years prior to your surgery, you learned to walk differently to avoid the pain. This changed walking pattern may include using your muscles differently than persons that do not have hip pain. We are very interested in the muscle-activity patterns after surgery when the pain is gone.

We will examine the activity of eight muscles in your legs and lower back. For each muscle, we will shave and clean with rubbing alcohol a small area (2 cm by 2 cm), then tape a pair of electrodes to the cleaned skin. Each electrode is a small cup (8 mm in diameter) that is filled with a gel and attached to small wires. After the eight pairs of electrodes are attached, the wires will be coiled together and taped to your shorts to keep them from swinging as you walk. The muscles that we are interested are: for both hips, the muscle found just below your waist in the back and the muscle that runs along the outside of your hip joint; the muscles that lie on each side of your spine at the level of your waist; and one muscle on the front and one muscle on the back of the thigh on the same side as your operated hip. To record the movement of your body during walking, we will videotape you while you walk along a straight path. In order to measure the body movements, we will tape small reflective balls (1 cm in diameter) on the skin at your shoulders, the outside surfaces of both hips, knees, and ankles, and your waistline in the back.

We will be comparing your data collected on three occasions–before your surgery and four and eight months after your surgery. We are only interested in understanding how your data change over time, so we will not be comparing it with the data collected from other people.

The placement of the electrodes and the marking of your joints will require that you wear shorts and a blouse/shirt that we can roll up. This may be somewhat embarrassing to you although we will try to make you feel comfortable by providing you with privacy during testing. Only the researchers and technicians directly related to the study will be present during the data collection sessions, and only those persons will be allowed to view the videotape after the session is completed. The Human Motion Analysis Laboratory is in a private room of the McCaig Centre.

It is possible that the gel that fills the electrodes or the tape used to hold the electrodes and joint markers in place may cause some minor skin irritation but this risk is very slight. We will minimize these possibilities by thoroughly cleaning the skin areas after the electrodes and markers are removed. The risk of any long-term irritation is very small. During the actual walking trials, you could accidently trip and fall. We will minimize this risk by making sure that the lead wires from the electrodes or any other electrical cords are not in your way, and an investigator will walk beside you to prevent your fall if you stumble.

The data collection involves the use of complex electronics equipment, and because of equipment problems, you might experience delays in data collection.

than the investigators will view your videotaped walking trials. After completion of data collection and analysis, you will be welcome to review our findings and conclusions and to view your videotape.

We will reimburse you for mileage and parking at the end of your participation. If data collection extends over a mealtime, we will reimburse you for a meal. Please save your receipts and return them to us for the reimbursment.

Your signature on this form indicates that you have understood to your satisfaction the information about your participation in the research project and agree to participate as a subject. In no way does this waive your legal rights or release the investigators, sponsors, or involved institutions from their legal and professional responsibilities. You are free to withdraw from the study at any time without jeopardizing your health care. Your continued participation should be as informed as your initial consent, so you should feel free to ask for clarification or new information throughout your participation. If you have further questions, please contact: Lauri Kaul, BSc. P.T (670-2454) or Ronald Zernicke, Ph.D. (220-8666)

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Name

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Signature of subject

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Name of witness

____

Signature of witness

____

Date

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