THE EFFECTS OF FEMORAL OFFSET ON HIP ABDUCTOR MUSCLE FUNCTION

by

SAMATCHAI CHAMNONGKICH

(Under the Direction of Kathy J. Simpson)

ABSTRACT

The purpose of this study was to determine if the magnitude of femoral offset of a total hip arthroplasty (THA) influences hip abductor muscle strength as well as kinematics and electrical activation of hip abductor muscles during stepping over an obstacle.

Twenty participants with a unilateral THA, 11 categorized as high femoral offset (HI-FO) and 9 as low femoral offset (LO-FO), participated in the study. Maximal isometric and concentric torques of hip abductor muscles were measured using a Cybex 6000™ dynamometer. Three-dimensional kinematic and electromyography (EMG) data were collected while the participant walked on a level surface, and while stepping over obstacles of 10% and 25% of each participant’s leg length. Comparisons of the hip abductor torques between limbs within each FO group and between LO-FO and HI-FO operated limbs were made by paired \( t \)-tests and Student’s \( t \)-tests, respectively. A 2 (FO group) x 3 (obstacle height) analysis of variance with obstacle height as a within-subject factor and FO group as a between-subjects factor was employed to test hypotheses related to the stepping over an obstacle task.

Isometric and concentric abductor torques tended to be higher for the HI-FO than LO-FO group. Compared to the LO-FO group, the HI-FO group exhibited significantly less magnitude of hip abductor strength deficit of the THA limb compared to the nonoperated limb, less mediolateral sway of the body’s center of mass, less EMG activation of the THA limb gluteus
medius muscle, and greater foot-obstacle clearance during negotiating an obstacle. These results confirm the use of an increased femoral offset as an effective choice to enhance hip abductor muscle functioning and promote balance control during ambulation.

INDEX WORDS: Total hip arthroplasty, Hip abductors, Strength, Gait, Obstacle, Kinematics, Electromyography
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CHAPTER 1
INTRODUCTION

Background

The development of modern hip replacement surgery began in the early 1960’s with Sir John Charnley, an English orthopedic surgeon (Charnley, 1979). Since then, it is estimated that more than 168,000 total hip arthroplasty (THA) surgeries are performed annually in the United States (AAOS, 2002). Degenerative joint disease, i.e., osteoarthritis (OA) of the hip, accounts for approximately 70% of elective THA cases (Siopack & Jergesen, 1995). Other conditions, such as rheumatoid arthritis, avascular necrosis, injury, and bone tumors also may lead to the need for hip replacement surgery (NIH, 1994).

It has been reported that patients with OA of the hip joint demonstrate asymmetric gait patterns, such as decreased stride length, decreased hip motion, and as well as reduced hip joint muscle moments on the affected side (Hurwitz et al., 1997). These altered gait characteristics reflect compensations by the patient to reduce the hip joint loading due to pain and limited mobility of the affected hip. Furthermore, weakness of the hip abductor muscles is commonly found in patients with hip disease as a result of debilitating hip pain and disuse atrophy (Arokoski et al., 2002).

Pain relief of after recovery from a THA allows patients to perform more of their daily and recreational activities (Brown et al, 1980). Consequently, the increased physical activity should promote the restoration of muscle function. However, several investigators have reported that residual gait impairments still persist up to two (Loizeau et al., 1995; Perron et al., 2000)
or even four years (Brown et al., 1980; Tanaka, 1998) following the surgery. Several constraints, such as age, preexisting muscle weakness, sedentary lifestyle, obesity, and insufficient rehabilitation time, might also contribute to a poor functional recovery of the THA patients (Munin et al., 1998).

Weakness of hip abductor muscles following THA has been found to be associated with poor gait patterns, with patients demonstrating abnormalities such as a decreased single limb stance time, decreased hip extension, and a positive Trendelenburg gait1. These gait alterations and/or the mechanics underlying these alterations may contribute to loosening of the implant components (Long et al., 1993). In addition, it has been reported that muscle activation patterns of the hip abductor muscles of individuals with hip OA or hip arthroplasty are atypical. Sims et al. (2002) reported that older adults with hip OA exhibited higher gluteus medius activation than healthy older adults during the single limb stance of gait. Consequently, Long et al. (1993) reported that THA patients with poor prosthetic fixation demonstrated postoperative limp that was associated with an absence of EMG activity of the gluteus medius muscle.

It has been suggested that one of the most effective methods for a surgeon to improve the mechanical leverage and, consequently, partially offset the reduced strength, i.e., reduced muscle moment production, of the hip abductor muscles is to increase the ‘femoral offset’ of the hip prosthesis (Steinberg & Harris, 1992). ‘Femoral offset’ (FO) is defined as the perpendicular distance from the long axis of the femur to the center of rotation of the femoral head (Figure 1). It can be seen from Figure 1 that, geometrically, if the FO is greater, then the leverage, i.e., the moment arm, of the resultant force of the hip abductor muscles also should increase.

1 The Trendelenburg test is a test for the stability and abductor strength of the hips. A positive sign is indicated if the pelvis drops laterally downward over the uninvolved side while the patient is standing on one foot on the involved side, presumably due to weak hip abductors. Hence, during the single support phase of the gait cycle of the involved limb, the pelvis will drop laterally over the swing (uninvolved) limb. This is called a positive Trendelenburg gait (Hardcastle & Nade, 1985).
proportionately. However, it is not known if the increase in femoral offset will significantly enhance the physical function of the hip abductors and improve movements of the lower extremities in individuals with THA during locomotion.

Figure 1.
Illustration of the THA components.

Note: Frontal view of the left femur and pelvic structures are shown. The femoral bone has been cut to show the femoral THA component.
A person’s gait must be adjusted when confronted by varying heights of the surface of the physical environment during daily activities, such as stepping up and over a door threshold, walking up stairs, and stepping down off a walkway curb. Maintaining stability of the whole body during stepping over obstacles is more difficult than level walking and involves precise control of the center of mass of the body (Bauby & Kuo, 2000). Therefore, for THA individuals with decreased hip abductor strength, these functional daily tasks can involve a higher risk of falling than level walking as active control of the hip abductor muscles of the stance limb is important in controlling the pelvis and contributes to maintaining balance of the trunk (MacKinnon & Winter, 1993).

In addition, during walking, the lower extremities provide a relatively narrow base of support as the body mass is transported forward when supported by one foot during the single limb support of gait. However, there is a lack of knowledge about how dynamic stability of the whole body is maintained during balance-challenging ambulatory tasks, such as obstacle crossing by individuals who have had total hip replacement.

Therefore, the purpose of this study was to determine if the magnitude of femoral offset of a total hip arthroplasty (THA) influences hip abductor muscle strength as well as the kinematics and the electrical activation of hip abductor muscles during stepping over an obstacle of different heights in individuals with THA.

Significance of the study

Recently, the Centers for Disease Control and Prevention (CDC, 2003a) reported that, in the United States, 70 million adults, equivalent to one in three adults, suffer from arthritis and/or chronic joint symptoms. Arthritis prevalence increases with age, affecting approximately 60% of the U.S. population aged over 65 years. According to the United States Administration on Aging
(AoA, 2003), the number of older Americans over 65 years of age reached 35.6 million in 2002, reflecting an increase of 3.3 million or 10.2% from 1992. The number of persons aged 65 years or older is expected to increase to an estimated 71 million in 2030, and the number of persons aged 80 years or older is expected to increase from 9.3 million in 2000 to 19.5 million in 2030 (CDC, 2003b).

As of 1999, more than 168,000 total hip arthroplasties were performed annually in the United States (AAOS, 2002). Therefore, it can be assumed that there will be a continued increase in the number of THA procedures in the United States, as the number of OA cases is expected to increase due to a larger proportion of the older population who are more likely to have some form of arthritis or related conditions. In fact, based on the National Hospital Discharge Survey 1996-1999, the American Academy of Orthopaedic Surgeons (AAOS, 2002) has estimated that over 184,000 total hip arthroplasty cases will occur in year 2010. It was estimated that the number of THA procedures will increase to 272,000 cases per year by 2030. Therefore, further research regarding THA surgical techniques and improvement of functional performance of individuals with THA is warranted.

Improvement in strength of the hip abductor muscles is important for the successful outcome of THA because, postoperatively, patients must be able to generate the hip muscle moments required to perform a variety of activities. However, THA limb abductor strength remains less than the nonoperated limb (Shih et al., 1994; Sicard-Rosenbaum et al., 2002). Therefore, it is considered desirable to optimize the femoral offset and the abductor lever arm at the time of hip reconstruction to promulgate greater magnitudes of hip abductor muscle moments (Johnston et al., 1979). Kaplan et al. (1987) reported that the distance from the center of the prosthetic hip to the tip of the greater trochanter, which is a measurement comparable to FO of
the current study, was less in THA patients with unstable implant components. Therefore, failure to identify and correct a potential decrease in femoral offset of the prosthesis may lead to instability of the hip joint, a visible limp, and increased hip joint reaction force (Steinberg & Harris, 1992).

It is essential to understand the movement strategies used by THA individuals to maintain balance during obstructed gait in order to prevent serious injuries that could result from falls. However, the existing biomechanical research regarding femoral offset and the effects on abductor muscle activity during functional activities is nearly nonexistent for THA population is still limited. Thus, this is, to our knowledge, the first study to examine functional effects of the femoral offset on the function of the hip abductor muscles as related to postural stability and controlling the pelvis while negotiating obstacles during locomotion. Ultimately, the results from the present study could thus provide insight into several clinical concerns associated with design of the hip prosthesis and the technique of total hip replacement surgery. In addition, the information may aid therapists in planning therapeutic exercises to enhance physical function of individuals with THA who might be at risk of injury due to balance or gait problems.

Premises of the study

The underlying premises of the study are explained with respect to the following two topics: hip abductor muscle strength and postural stability during stepping over obstacles.

Hip abductor muscle strength

Although controversial, it has been thought that the most effective method readily available to the surgeon to improve the abductor lever arm is to increase the offset of the femoral prosthesis during the hip reconstruction (Steinberg & Harris, 1992). McGrory et al. (1995) observed that an increase in FO was positively correlated with hip abduction range of motion and
hip abductor muscle strength. However, the effect of increased on the hip abductor muscle strength has not been conclusively shown. Recently, Asayama et al. (2002) used the proportion between femoral offset and the body weight lever arm to obtain a femoral offset ratio (%FO), in order to account for differences in the magnification of the radiographs and to allow for comparisons of patients of different heights and pelvic size. The authors reported that the restoration of %FO greater than 20% of the body weight lever arm was an important factor in obtaining a negative Trendelenburg sign after THA. The relationship between the femoral offset ratio and the hip abductor strength, however, was not investigated.

Maintaining the geometry of the reconstructed hip joint closely to that of the normal hip joint is considered one of the goals to a successful THA, because it is more likely that normal hip abductor function after THA can be regained. In theory, an increase in the offset of the length of the femoral component will increase the moment arm of the resultant abductor muscle force vector, and this will result in greater moment-generating capacity of the hip abductor muscles. In this study, therefore, it was surmised that individuals with high femoral offset (HI-FO), i.e., those individuals whose postoperative femoral offsets are greater than those of the nonoperated hips would be able to generate greater maximum hip abductor moment than individuals with low femoral offset (LO-FO), i.e., those individuals whose femoral offsets are less than the values of their nonoperated hips.

We anticipated that not only would peak isometric and concentric torque of the hip abductor group be less for the LO-FO compared to the HI-FO group, we also expected that the angle to peak hip abductor torque would be different between the two FO groups. Based on the results of mathematical modeling (Delp & Maloney., 1993; Vasavada et al., 1994), it was predicted that when FO increased, then the length of various abductors would increase, thereby,
allowing the hip abductor to be at more optimal lengths that would increase the force-generating capacity of the hip abductor. In this study, therefore, it was surmised that individuals with HI-FO, would be able to reach the maximum hip abductor moment at a smaller angle than individuals with LO-FO.

Stepping over an obstacle

Lower extremity strength is an important component for mobility-related activities for older adults (Riley et al., 2001). Gradual declines in muscle mass and strength are demonstrated by elderly persons as a result of the aging process (Frontera et al., 1991). It is therefore even more difficult to improve strength and function of the hip muscles of older patients who have had a THA, particularly those with low FO. In addition, as reported by several investigators, significant residual muscle weakness of hip abductor muscles of THA limb persists for several years following THA surgery. With deficits in the strength of the hip abductor muscles on the THA side of the body, it is common to find that, during gait, the pelvis drops down on the contralateral side during the THA limb single support phase (Magee, 2002). Therefore, if the FO of the prosthetic hip was not restored to the same value as that of the natural hip joint, it was expected that the individual would exhibit some form of kinematic adaptations in order to compensate for the weakness of the hip abductor muscles of the THA limb while crossing over obstacles.

Compared to walking on level ground, stepping over an obstacle requires greater hip abductor muscle force of the supporting limb. Therefore, stepping over an obstacle may be an ideal task for use in identifying abnormalities in hip abductor muscle function and postural stability control in THA recipients. Compared to level walking, an increase in duration of the single limb support phase during stepping over an obstacle is necessary for the swing limb to
clear the obstacle, and consequently, hip abductor muscle force of the support limb may need to be generated for a longer period of time. Therefore, without adequate hip abductor muscle moment-generating capacity of THA individuals with low FO to counteract the gravitational torque of the body weight, it is surmised that differences in kinematics variables would be found between individuals with high FO and those with low offset.

Therefore, it was expected that as obstacle height increases, THA individuals with high FO would exhibit no changes in magnitude of mediolateral displacement of the body COM and the level of EMG activity of the hip abductor muscles compared to the corresponding values from unobstructed level walking. In addition, for THA individuals with low FO, without an adequate THA hip abductor muscle moment being generated to counteract the body weight moment, it was expected that, during the stance period of the THA limb, the pelvis would drop laterally over the swing (uninvolved) limb resulting in a greater pelvic obliquity displacement, compared to individuals with high FO. Furthermore, THA individuals with low FO, compared to individuals with high FO, would perform the walking and stepping over an obstacle tasks with greater muscular effort as reflected by greater EMG activity of the hip abductor muscles. It was also expected that while stepping over the obstacle, THA individuals with low FO, compared to individuals with high FO, would exhibit an increased sway of the mediolateral displacement body COM due to excessive movement of the upper body and the swing limb.
Hypotheses

During maximal effort strength tests, the HI-FO group compared to the LO-FO group would exhibit:

Hypothesis 1: Greater maximum isometric and concentric hip abductor muscle moments for THA limb.

Hypothesis 2: Smaller abduction angle at peak concentric hip abductor muscle moment for THA limb.

While walking over an obstacle, the HI-FO group, compared to the LO-FO group, would exhibit:

Hypothesis 3: Less pelvic displacement.

Hypothesis 4: Less mediolateral displacement of the body center of mass (COM).

Hypothesis 5: Less magnitude of EMG activity of the THA limb hip abductor muscles.

Hypothesis 6: No change in mediolateral displacement of the body COM, and magnitude of EMG activity of the THA limb hip abductor muscles, as obstacle height increases from 0% to 10% or 25% obstacle height conditions.
CHAPTER 2
REVIEW OF LITERATURE

The literature review begins with the basic anatomy of the hip joint and the hip abductor muscles followed by kinematics and kinetics of the normal hip joint. This review also addresses the gait-related changes in the healthy elderly to better understand the effect of normal aging for comparison with gait deviations related to orthopedic disorders. A review of gait characteristics in individuals with osteoarthritis before and after having total hip arthroplasty will then be discussed. In the last section, gait characteristics during walking and stepping over an obstacle will be covered.

Anatomy of the hip

The hip is one of the largest weight-bearing joints of the body classified as an unmodified ovoid synovial joint. It is formed by the femoral head cupped into the acetabular cavity of the innominate (Clemente, 1985). During the THA procedure, the two parts of the hip joint, the femoral head and the acetabulum, are removed and replaced with a metallic prosthesis (Booth et al., 1988). The intrinsic stability of the hip joint is due to its ball-and-socket configuration. The normal hip joint, which is capable of circumductive motion, allows for a wide range of motion required for such diverse activities as walking, sitting, bending, and squatting. To accomplish such daily activities without difficulty, it is required that the acetabulum remain precisely aligned with the femoral head (Tronzo, 1984).
The proximal end of the femur includes the head, the neck, and the trochanters. The neck, which is a continuation of the shaft, joins the latter at an angle which varies from 125° to 135° (MacConaill & Basmajian, 1977). The angle between the plane of the femoral condyles and the axis of the femoral neck is the torsion angle of the femur. It shows a wide degree of variation from anteversion to retroversion with an average angle of 14° anteversion (Tronzo, 1984). The head of the femur forms roughly two-thirds a sphere. Except for a small fovea, the head of the femur is covered by hyaline cartilage which decreases in depth toward the periphery of the surface. The greater trochanter which projects from the expanded junction of neck and shaft is a typical traction epiphysis for insertion of the abductor muscles (Tronzo, 1984).

The acetabulum is formed from the fusion of the three bones, the ilium, the ischium, and the pubis, which make up the innominate (Lee, 1999). The acetabulum is roughly the shape of a hemisphere and projects in an anterolateral and inferior direction. The lunate surface of the acetabulum, which represents the articular portion of the acetabulum, is lined with hyaline cartilage. The non-articular portion of the acetabulum, the acetabular fossa, is filled with loose areolar tissue and covered with synovium. The apex of the labrum is lined with articular cartilage and lines inside the hip joint as a free border. The articular capsule encloses the joint and most of the femoral neck (Clemente, 1985).

**Hip abductor muscles**

There are three muscles identified as the primary hip abductor: gluteus medius, gluteus minimus, and tensor fascia latae (Tronzo, 1984). Previous research regarding the hip abductor functions often assessed the activity of the gluteus medius because it is the largest of the abductor group and also most accessible for surface electromyographical investigation.
The gluteus medius occupies about 60% of the total abductor cross-sectional area (Clark & Haynor, 1987). The gluteus medius is described as a board, thick, radiating muscle on the outer surface of the pelvis. Its proximal attachment is from the anterior superior iliac spine (ASIS), along the outer edge of the iliac crest to the posterior superior iliac spine. The line of attachment is approximately 1 cm broad and limited to the iliac crest. The muscle fibers converge to a strong-flatted tendon which is inserted into the anterosuperior portion of the greater trochanter, with a bursa separating the tendon from the surface of the trochanter (Clemente, 1985).

The gluteus medius muscle bulb is made up of three distinct parts: anterior, middle, and posterior (Gottschalk et al., 1989). The three parts are approximately equal in volume but reveal distinct muscle fiber directions. The fibers of the anterior part run almost vertically from the anterior of the anterior iliac crest to the top of the trochanter. The middle part also has fibers that are more vertically oriented whereas the fibers of the posterior part run almost parallel to the neck of the femur. The superior gluteal nerve provides separate branches to innervate each of the three parts of the gluteus medius. The middle and posterior fibers of gluteus medius muscle play an important role to stabilize the pelvis on the femoral heads from heel strike to midstance of the gait cycle (Soderberg & Dostal, 1978).

The gluteus minimus, placed immediately beneath the gluteus medius, is also an important hip abductor muscle. The muscle originated from the outer surface of the ilium, between the anterior and inferior gluteal lines, and behind, from the margin of the greater sciatic notch. The fibers converge to the deep surface of a radiated aponeurosis, and this ends in a tendon which is inserted into an impression on the anterior border of the greater trochanter, and gives an expansion to the capsule of the hip-joint (Clemente, 1985). The gluteus minimus
supports the function of the gluteus medius in preventing the sagging of the pelvis on the unsupported side during gait (Soderberg & Dostal, 1978).

The tensor fascia latae arises from the anterior part of the outer lip of the iliac crest and is inserted between the two layers of the iliobial band of the fascia latae about the junction of the middle and upper thirds of the thigh (Clemente, 1985). The tensor fascia latae supports the abduction function by continuing the abduction process initiating by the gluteus medius during initial double and single support (Gottschalk et al., 1989). In addition, during late stance, the tensor fascia latae with the anterior fibers of gluteus minimus also act as a medial rotator of the hip (Winter, 1991).

**Kinematics of the hip**

Hip joint motion occurs in all three planes but is greatest in the sagittal plane where the motion ranges from 20º extension to 100º flexion. In the frontal plane the motion ranges from 30º adduction to 45º abduction, whereas in the transverse plane, approximately 30º to 40º of internal rotation and 60º of external rotation are possible (Lee, 1999).

In the normal gait pattern, the hip is maximally flexed during the late swing phase as the leg moves forward for heel strike. The hip joint then moves from a flexed to an extended position as the body is propelled forward at the beginning of stance phase. Maximum extension is reached just prior to toe off (Winter, 1991). The other conjoined motions at the hip include adduction/abduction, and internal/external rotation which occur at different phases of the gait cycle. Sutherland et al. (1994) described that the hip, which is neutral at foot strike, rapidly adducts during first double support, and rapidly abducts during second double support. In swing phase, the hip adducts from near maximum abduction at toe off to neutral relative to the pelvis.
just before foot strike. Internal rotation of the hip occurs in mid-swing phase until opposite foot strike. The hip then externally rotates until mid swing of the next gait cycle.

Kinetics of the hip joint

The force acting on the femoral head is dependent on the external forces acting on the limb and the internal forces primarily generated by muscle contraction. During erect standing on both legs, the superincumbent body weight is distributed equally through the pelvic girdle to the femoral heads and necks. Each hip joint supports approximately 33% of the body weight (BW) which subsequently produces a bending moment between the neck of the femur and its shaft (Singleton & Leveau, 1975). Standing on one limb dramatically increases the magnitude of forces acting on the hip joint. Especially during walking, each hip alternately supports the mass of the head, trunk, upper limbs, and swinging leg. Based on a mathematical model, the force exerted on the hip has been reported to range from 138% to 432% BW during the single limb support period of the gait cycle (Maquet, 1985).

Hip joint forces during daily activities have been estimated indirectly using musculoskeletal and analytical methods (Bergmann et al., 2001; Lueponsak et al., 2002), and measured directly in vivo with implanted transducers (Bergmann et al., 1993; Bergmann et al., 1997). Validation of the loading conditions at the hip joint as predicted a musculoskeletal model has been made (Heller et al., 2001). It was reported that the calculated hip contact forces and those measured in vivo during daily activities were somewhat different. Heller et al. (2001) reported that the calculated peak hip contact forces differed from the measured forces by a mean of 12% during walking and 14% during stair climbing. The component acting along an idealized femoral midline showed best agreement, while the results for the significantly smaller forces in the transverse plane were less accurate. Different in patient ages and the anteversion angle of the
implantation did not affect the patterns and magnitudes of the calculated hip contact forces (Heller et al., 2001).

Although no criterion exists for normal contact force patterns, hip contact forces with a double peak curve during walking, similar to the ground reaction force, are usually regarded as normal (Bergmann et al., 2001). The force at the hip joint during gait reaches an initial peak in early stance phase and a second peak in late stance phase. Single peak contact forces have been observed in some patients, however, while the belonging ground reaction force showed the usual double peak pattern (Bergmann et al., 2001).

Using an inverse dynamics approach, Lueponsak et al. (2002) evaluated net forces and torques at the hip during basic daily activities in 132 healthy elderly adults with mean age of 75 years. During upright standing, peak contact force was about 32% BW. The highest peak contact force was noted during stair descent (108.74% BW) followed by walking (88.75% BW). The peak contact force was approximately 40% BW during chair rise and bending to reach an object on the floor. Stair descending yielded the highest peak adduction torque of 8.40% BW \times \text{height}. Bending activity yielded the highest peak flexion torque of 6% BW \times \text{height}.

Bergmann et al. (2001) measured the in vivo hip contact force via an instrumented femoral prosthesis. The average peak forces of the patients during normal walking at 1.1 m\cdot s^{-1} were between 211% and 285% BW. The average peak forces were slightly less during standing on one leg (230 - 290% BW). During climbing upstairs the joint contact force was 251% BW which is less than 260% BW when going downstairs. The average peak forces were between 175 and 225% BW and between 125 to 180% BW during standing and sitting down, respectively. The peak torsional implant moment was between 1.72 and 1.91% BW \times \text{m for walking, the moment was higher when going upstairs (2.25% BW \times \text{m})}. During all types of activities, the
direction of the peak force in the frontal plane changed only slightly when the force magnitude was high. Perpendicular to the long femoral axis, the peak force acted predominantly from medial to lateral. The component from ventral to dorsal increased at higher force magnitudes.

The peak contact force in the contralateral hip in patients with antalgic gait pattern has previously been found to be as much as four times the body weight during walking (Bergmann et al., 1993). The result supports the notion that dysfunction of one muscle increases the joint contact force, because a part of the required joint moment is taken over by other muscles with unfavorably short lever arms and therefore higher forces.

Age-related changes in gait characteristics of healthy elderly

Several studies have revealed significant changes in gait patterns associated with advancing age (Elble et al., 1991; Himann et al., 1988; Lord et al., 1996). The most consistent finding in the research on gait in the elderly is that walking velocity decreases with age. The rate of decline in walking speed has been reported to be 2% per decade to age 63 and 16% per decade after age 63 in healthy older adults (Himann et al., 1988). The gait cadence of the elderly has been reported to be the same as that of healthy young adults (Winter et al., 1990). The age-related decline in gait speed has been found to be a function of both a shorter stride length (Elble et al., 1991; Lord et al., 1996) and an increase in double limb support duration (Lord et al., 1996; Winter et al., 1990).

Several cross-sectional studies report that gait speed in older adults is correlated with strength of the lower extremity muscles. Bendall et al. (1989) investigated walking speed in a group of 67 women and 58 men aged between 65 and 90 years living independently and found a significant association between strength of the calf muscles and walking speed, both of which decreased with increasing age. Brown et al. (1995) examined the relationship of lower extremity
strength and the ability to accomplish selected functional activities in 16 healthy but frail older adults ranging in age from 75 to 88 years. It was reported that there was a significant association ($r = .636$) between gait speed and the sum of the strength measures of the hip extensors, knee extensors, and planter flexors. As nonsignificant associations of lower extremity muscle strength and other daily activity tasks such as time used to rising from a chair were found, it was proposed that other factors such as proprioceptive input, impaired balance and joint range of motion may potentially play a role (Brown et al., 1995).

The relative important of strength, aerobic capacity, overall physical health status, and depressive symptoms in explaining changing in gait speed was assessed by Buchner et al. (1996). There were 152 community-dwelling adults aged 68-85 participating in a 6-month exercise program involving either endurance and/or resistance exercise training. It was reported that strength of the knee and ankle muscles, VO2max, and body mass were significant independent predictors of gait speed. The results suggested that changes in gait speed were related to changes in depressive symptoms and physical health status, with more depressive symptoms and worsening health status associated with slower gait.

In the majority of previous studies investigating the effect of aging on gait, differences in the joint angle profiles between the young and the elderly adults have been reported to be minimal. However, the findings of previous studies regarding the aging on joint kinematics during walking are not consistent. For example, Judge et al. (1996) reported greater hip flexion at heel contact in the elderly than the young adult. Kerrigan et al. (1998) reported no differences in maximum hip flexion during late swing, but a decrease in peak hip flexion in the elderly, while others (Elble et al., 1991; Winter, 1991) reported a greater range of motion at the hip for the elderly than the young. According to Kerrigan et al. (1998), many of the differences in joint
kinematics reported between the young and elderly are attributed to differences in the gait velocities and/or stride lengths.

Inadequate hip extension and reduced ankle plantarflexion was reported to be primary responsible for the shorter step length in older adults (Judge et al., 1996; Riley et al., 2001). Judge et al., (1996) reported that reduced ankle joint power during late stance in the elderly compared with young adults has been found to be associated with slower walking speed. In addition, the older adults generated more hip flexor power to compensate for a reduction in plantarflexion power. However, in the recent study of 14 healthy elderly, Riley et al. (2001) found that when the elderly was asked to walk faster to match the walking speed of younger participants, the ankle plantar flexor power increased, while maximum hip extension remained reduced. The authors also observed that for the elderly walking at fast speed, there were significantly increases in hip and knee moment contributions to hip linear power to levels comparable to those of young participants at similar speed. Riley et al. (2001) pinpointed that kinematic alterations, particularly reduced maximum hip extension, act to limit walking speed in the elderly.

Gait analysis of individuals with hip osteoarthritis

Little information exists on the movements of the pelvis in individuals with osteoarthritis of the hip joint. Thurston (1985) examined the angular displacement of the pelvis and lumbar spine in 19 patients with unilateral hip OA and 10 age-matched individuals. The mean age of the hip OA group was 65.1 years. The most noticeable different movement of the pelvis between the hip OA patients and the healthy controls was in the sagittal plane along with significant restriction of the affected hip joint range of motion. The angular displacement of the pelvic tilt for the hip OA patients (8.6°) was two times that of the age-matched control participants (4.3°).
The authors also reported a significant decrease in the pelvic obliquity angular displacement for the hip OA patients (4.0º) compared to that of the control participants (5.7º).

More recently, evaluation of the pelvic movement was done by Watelain et al. (2001). Gait characteristics, pelvic motions, and muscle powers of the lower limb joints of seventeen patients with the early stage of hip OA were evaluated and compared with an age-matched control group. The authors found that the hip OA participants walked 12.4% slower and exhibited the upward tilt of the pelvis 2.5 times more than that of the control group. The pelvis also dropped 2.4 times more on the unsupported limb of the hip OA group at push-off. The authors reported a decreased hip flexion moment and less muscle power generation to pull the thigh at push off. In addition, the patients with hip OA demonstrated an increased power absorption in the frontal plane and an increased power generation in the transverse plane of the knee, and an increase in frontal power absorption of the ankle. The authors concluded that, for individuals with hip OA, movement patterns at the pelvis and other joints of the affected lower limb were altered to compensate for lack of hip extension, maintain lower limb alignment, and help with body weight transfer.

Changes in gait parameters, lower limb kinematics, muscle moments or muscle powers have been widely investigated in patients with hip OA as well as those who received total hip replacement resulting from hip arthritis. Prior to THA, individuals diagnosed with OA of the hip have an altered gait pattern characterized by asymmetry of the kinetics and kinetics variables of the lower extremities (Hurwitz et al., 1997; Kyriasis & Rigas, 2002).

Hurwitz et al (1997) performed gait analysis on 19 hip OA participants who were preoperative for unilateral THA surgery. Kinematic and kinetic measures were compared to an age- and gender-matched control group as well as between the affected and unaffected limbs.
The authors observed that hip OA patients who had hip flexion contracture demonstrated a hesitation or reversal in motion of the hip joint as it extended during the late stance phase. The hip range of motion, and the hip extension, adduction, and rotation moments on the affected side of the hip OA group were significantly less when compared to the unaffected side and the control group. An increased level of hip pain reported by the patients was also significantly correlated with the decreased hip extension moment of the affected side.

Comparisons of the pre- and postoperative temporal gait parameters of patients with a unilateral cementless total hip replacement have been shown by Kyriazis and Rigas (2002). Twenty-five females with mean age of 51 years were tested preoperatively as well as at 1 year and 8-10 years postoperatively. Twenty healthy females participated as a control group. A conductive walkway system was used to obtain the gait cycle duration, the single and double support and the single step duration and the gait speed. The gait parameters (%gait cycle) were reported for both the involved and uninvolved limbs. All of the parameters assessed at 1 year postoperative were differed from the preoperative values except gait cycle duration and velocity. The mean relative single support duration of the involved limb increased from 30.2% to 33.6% at one year evaluation. The asymmetry of the relative double support period between the involved and uninvolved limbs still persisted at the 8-10 years evaluation. Although a gender-matched control group was also studied and their gait parameters were reported, no statistical comparisons were made with the THA group.

Gait evaluation after total hip arthroplasty

Several researchers have reported improved measurable gait parameters from under one year to over five years post operation. The only study investigating the movement patterns the pelvis after THA was by Perron et al. (2000). Perron et al. (2000) evaluated gait characteristics
were evaluated in 18 women who had unilateral THA at a mean postoperative time of 3.9 years (Perron et al., 2000). Three-dimensional kinematic data of the pelvis and the lower extremities were collected synchronously with ground reaction force data. Fifteen healthy age-, height-, and weight-matched women were participated as a control group. It was reported that gait speed of the THA group was 14% slower than the control group due to a slower cadence and a shorter stride. The single limb support (%gait cycle) on the operated limb was 4.3% less compared to the control group. Impairments at the pelvis were more noticeable in the sagittal plane. The pelvic anterior tilt of the THA participants was increased by 63% of the value of the upright position during early push-off phase. In addition, the THA participants increased their lateral bending of the trunk toward the THA stance limb. The study by Perron et al. supports the opinion that after THA individuals may use a trunk bending strategy as a mean to assist the weaker hip abductor muscles in the control of balance and decrease the hip joint loading.

Long et al. (1993) evaluated gait pattern of 18 patients with cementless hip arthroplasty preoperatively and at three months, six months, one year and two years after surgery. All patients had unilateral hip arthritis and ranged in age from 35 to 85. Gait velocity, cadence, and stride characteristics were obtained using foot switches. Hip joint displacement was obtained using film analysis. Fire wire electrodes were used to evaluate EMG activity of the hip abductor, hip adductor and knee extensor muscles. It was reported that stride characteristics such as cadence and single limb support duration and the hip range of motion of the affected limb improved to normal compared to the uninvolved limb within two years post surgery. At one and two year evaluation, irregular loading pattern of the vertical ground reaction force of the involved limb revealed weakness of the hip muscles in all patients. Postoperative abnormal EMG activity of the
hip abductor muscles was revealed in four patients who developed aseptic loosening of the replaced hip.

Berman et al. (1991) performed gait evaluation in 31 patients with unilateral THA. All patients had cemented hip arthroplasty through an anterolateral approach. Participants were evaluated within one month before the surgery and within four months, five to eight months, nine to 12 months, and 13 to 18 months after the surgery. Age-matched healthy adults served as a control group. Preoperatively, participants with hip arthritis walked significantly slower than the control group and also demonstrated a decreased step length, and prolonged stance time and double-support time. The authors reported that improvement in these spatial temporal gait parameters was minimal at the zero to four month postoperative interval. Gait velocity of the THA participants approached that of the control during the five-to-eight month postoperative.

McCrary et al. (2001) evaluated the vertical ground reaction forces (VGRFs) in 27 individuals with unilateral hip arthroplasty and compared with a control group of 35 healthy subjects. VGRF measures collected using a treadmill instrumented with two force plates were used for calculation of a symmetry index in order to compare bilateral limb loading. There were significant differences in the symmetry indices in the parameters of the first and second peak forces, impulse, loading rate, stance time, and time to first peak force on the operated limb when compared to either the nonoperated limb or to the control group. The results indicated that during walking the THA group tended to put more weight on the nonoperated limb. The authors pointed out that with the presence of an asymmetric gait, the condition of the nonoperated limb may be at risk of developing joint arthritis if it continues to be placed under increased loading.

In summary, several studies involving pre- and post-THA gait analysis revealed that gait temporal and spatial characteristics improve after surgery. However, the movements of the pelvis
and lower extremity of the surgical limb still persists with deficits in both sagittal and frontal plane. In addition, recovery of walking velocity and symmetry of both lower extremities remain controversial several years post surgery.

Evaluation of hip abductor muscles after total hip arthroplasty

Postoperative weakness of the hip abductor muscles is considered one of the major concerns after THA (Borja et al., 1985; Downing et al., 2001). Several investigators have measured strength of the muscles around the hip in an attempt to evaluate the hip muscle function after THA (Shih et al., 1994; Sicard-Rosenbaum et al., 2002; Trudelle-Jackson et al., 2002; Vaz et al., 1993;).

Trudelle-Jackson et al. (2002) evaluated functional outcomes in eleven female and four male THA recipients one year after surgery. Assessment variables consisted of self-assessment of function and measures of postural stability, lower-extremity muscle strength, and hip ROM. The authors reported significant differences between the THA limb and the uninvolved limb in measures of postural stability in both medial-lateral postural stability and anterior-posterior postural stability were reported. One year after surgery, the participants had improved strength measures in the THA limb but not equal to the uninvolved side. The THA hip flexor, abductor, and extensor strength was 82% to 90% of the uninvolved hip. The THA knee flexor and extensor strength was 87% and 97% of the uninvolved hip, respectively. There was a moderate correlation between scores of self-assessed function and hip abductor and knee extensor strength. The authors suggested that strength of the hip abductor and knee extensor muscles may be used for determining the patients’ perception of their function.

Sicard-Rosenbaum, Light, & Behrman (2002) measured the lower extremity muscle strength and temporal-spatial gait parameters in 15 adults at 9 months to 6 years after THA
surgery and compared with an age- and gender-matched control group. Strength of the hip abductors and extensors, knee extensor, and ankle dorsiflexors was assessed using a handheld dynamometer. The authors reported that the THA group walked significantly more slowly than the control group when asked to walk at their maximum walking speed. The overall lower extremity strength of the affected limb was less compared to the nonaffected limb and the control group. The hip extensors and abductors were the most affected muscle groups with a strength deficit of 13% and 16%, respectively. However, several factors affected the results of this study. The causes lead to hip surgery were fracture from a fall (3 cases) and car accident (2 cases), and arthritis of the hip (10 cases). In addition, THA surgeries were performed by different surgeons.

Vaz et al. (1993) examined changes in hip abductor muscle strength after total hip arthroplasty in twenty-three males and twenty females with unilateral primary or secondary OA of the hip. Evaluation of isometric strength of the hip abductor muscles was performed before the surgery and at 1, 6, 12 and 24 weeks after surgery. A hand-held dynamometer was used for strength testing. Functional tests including the Harris hip rating, the d’Aubigne scales, a functional difficulty scale, and the distance covered during a 6-minute walk were also assessed at 12 and 24 weeks after surgery. The isometric hip torque was found to drop significantly from the preoperative torque at the one-week evaluation. However, the participants regained their hip strengths at the 6-week evaluation. In addition, all subsequent torques were significantly greater than the pre-operative torques. All functional performance was also significantly improved at each assessment. However, the authors reported a low correlation between hip abductor strength and functional performances.

Muscular recovery around the hip joint in THA patients was evaluated by Shih et al. (1994). Twenty males with median age of 49 years diagnosed with osteonecrotic of the femoral
head and twenty females with median age of 55 years with hip osteoarthritis were evaluated preoperatively and postoperatively at six months and one year. Maximal isometric torque of the hip flexors, extensors, and abductors for both limbs were measured using a Cybex 340 dynamometer. Preoperatively, all muscles of the involved limb of both the male and female groups were weaker than the uninvolved limb. After surgery, recovery of the hip muscle strengths was noted in both involved and uninvolved hips for both the males and females, but females had slower recovery rate compared to males. Compared to the preoperative condition, changes in muscle torques in females did not reach significantly until at one-year postoperative evaluation, whereas the hip strengths of the male started to improve significantly at six month. In neither group, however, the hip muscle strength of the involved side gained to the level of the uninvolved side. As reported by the authors the hip strength by the one-year evaluation was 84% - 89% of the uninvolved hip in the male group and 79% - 81% of the uninvolved hip in the female group.

In summary, relieving the hip pain after THA should allow for the improvement of the muscle function resulting in a significant improvement in the muscle strengths in both hips; however, continue weakness on the side of the operated hip has been shown in all studies several years postsurgery.

Gait analysis during stepping over an obstacle

Gait adjustments during stepping over obstacles have widely been examined by several researchers. Patla and colleagues (Patla et al., 1991; Patla & Rietdyk, 1993) have focused on the effects of obstacle properties on the locomotor control.

Patla and Rietdyk (1993) examined gait characteristics in six young healthy adults. Two dimensional trajectories of the markers at shoulder, hip, knee, ankle, heel, metatarsal, and the toe
were collected during level unobstructed gait and crossing the obstacle of different heights and widths (6.7 cm, 13.4 cm, and 26.8 cm). The results indicated that several aspects of the gait patterns during obstructed gait were different from level unobstructed walking. The differences in limb trajectory between obstructed gait and level walking take place during at the swing phase beginning at toe-off. The average clearance of the toe over the obstacle was about 10 cm. While there was an increase in vertical trajectories of all the markers as the obstacle height increased, the limb trajectories were minimally changed for the different obstacle widths. As the limb trajectory was adjusted at the initiation of the swing phase, the authors noted that control of the gait patterns was actively modulated at the end of stance phase and also at the swing phase.

In addition to varying the obstacle height, the time of appearance of obstacles of varying heights and their position within the step cycle, and cue lights for direction change that given to participants were also varied (Patla et al., 1991). Patla and colleagues (1991) reported that gait changes such as step length and step width regulation, and low obstacle avoidance can be safely implemented within the same step cycle. However, direction change must be planned in the previous step to reduce the acceleration of the body center of mass toward the landing foot due to the incapacity of muscles to rotate the body to different axis within the same step and (Patla et al., 1991).

Austin et al. (1998) examined the kinematics of the swing phase limb in 15 female young participants with mean age of 26.3 years during level walking and stepping over obstacle of different heights of 3.1 cm, 7.6 cm, and 12.6 cm. Two-dimensional video analysis was used to collect the angular displacements of the lower extremity, toe clearance, and the heel distance from the obstacle during toe-off and heel contact. The authors reported a significant increase in both toe and heel clearance as obstacle height increased. However, a plateau of toe and heel
clearance was observed at the greater obstacle heights. The angular displacements of hip, knee, and ankle joints also significantly increased as the obstacle height increased. As with toe and heel clearances, an increase in knee and hip range of motion was cease and did not follow the linear trend as the obstacle height reached the extreme of 12.6 cm. The authors proposed that there exists a transition phase characterized by invariance in clearance distances with maintain a margin of safety and efficiency of movement during stepping over an obstacle.

Besides the study by Patla and colleagues (1993), it is only recently that the kinematic and kinetics characteristics of obstructed gait have been more intensively investigated (Begg et al., 1998; Chou & Draganish, 1997; Sparrow et al., 1996) to detect the difference in limb trajectories of the of the lead and the trail limbs during stepping over obstacles. Sparrow et al. (1996) investigated gait characteristics of healthy young adults during stepping over obstacles. Six males with mean age of 40.8 years and six females with mean age of 36.3 years participated in the study. Kinematic characteristics of the lead foot and the trailing foot during stepping over obstacle of height adjusted to 10%, 25% and 40% of each participant’s leg length were calculated from there-dimensional video analysis. The authors reported that lead foot clearance was relatively uninfluenced by obstacle height whereas trail foot clearance increased systematically and always cleared the obstacle by a greater margin. The male participants cleared the obstacle by a greater margin than the female participants (by approximately 5 and 10 cm for the lead and trail foot, respectively). Obstacle crossing stride did not change as the step height increased. The obstacle crossing time decreased as the step height increased. A decrease in crossing time was a result of the shorter crossing time of the lead foot.

Chou and Draganish (1997) performed gait analysis in 14 young adults with mean age of 23 years. Toe-obstacle clearance and 3-D motions and moments of the trail limb were obtained
while the participants stepped over obstacles of heights of 5.1, 10.2, 15.3 and 20.4 cm. The authors reported that toe clearance of the trail foot at any obstacle height was significantly greater than that of the unobstructed gait. However, no significant differences in toe clearance were found among different obstacle heights. The authors also reported that, for the sagittal plane, the hip and knee flexion angles of the trailing limb increased as the obstacle height increased. There were no significantly changes in the range of motion of the hip, knee, or ankle in the transverse or coronal plane. At the hip, there were a 28% decrease and a 27% increase in the hip extension and adduction moment, respectively, when stepping over the highest obstacle compared to level walking. The adduction and external rotation moments at the knee and ankle at any obstacle height condition were significantly greater than those during the unobstructed gait, but not significantly different as the obstacle height increased.

Begg et al. (1998) examined gait characteristics during stepping over an obstacle in twelve healthy adults. There were six males with mean age of 28.4 years and six females with mean age of 24.4 years participating in the study. Ground reaction forces as well as kinematics data of both the lead and trail limbs were collected. The obstacle height conditions were 10%, 20%, and 30% of each participant’s leg length. The authors reported no significant differences in ground reaction forces and gait temporal parameters between genders. Both male and female participants increased obstacle crossing step lengths and reduced crossing speed as a function of obstacle height. The maximum propulsive force increased as the obstacle height increased. The results revealed that the lead foot produced lower vertical impulse for all obstacle heights. In addition, the trail foot produced greater braking and propulsive impulses. The authors proposed that stepping over obstacles demanded increased vertical and anterior-posterior push off forces of the trail limb to provide stable support for the lead limb (Begg et al., 1998).
Begg et al. (1998) and Chou and Draganish (1997) raised an important issue regarding the important of the trailing limb while crossing over an obstacle. Increased adduction moments at all three joints would place greater demands on the abductor muscles and may be a contributing factor to sideways instability in individuals with total hip arthroplasty who might have decreased hip abductor strengths.

Although stepping over obstacle is one of the activities of daily living and the potential consequences of tripping are high for the elderly, little attention has been paid into investigating the kinematic and kinetic characteristics of the lower extremities during stepping over obstacles of different heights in older adults in order to better understand the changes in gait control with age.

Comparison of kinematics of obstacle avoidance in healthy young and older adults was first made by Chen et al. (1991). In the study by Chen et al. (1991), the trajectories of the lead foot while stepping over obstacles were examined in 24 young adults with mean age of 21.6 years and 24 older adults with mean age of 71.3 years. Equal numbers of male and female participants were presented in each group. Testing conditions included level walking, and stepping over an obstacle height of 2.5, 5.1, and 15.2 cm. It was reported that crossing speed of the older participants was slower than the young adults as a result of shorter step length, for example at the 2.5 cm-obstacle height, the averaged older adults’ crossing speed was 1.16 m·s⁻¹ compared to 1.30 m·s⁻¹ for the young adults. After crossing an obstacle, foot placement of the older adults was significantly closer to the obstacle than the young adults. Although the authors reported that age had no effect on minimum foot clearance over an obstacle, the older adults exhibited less of an increase in step length over the obstacle and tended to produce lower foot clearances and placing the trail foot farther form the obstacle. The results reported by Chen et al.
(1991) are in agreement with McFadyen and Prince (2002) who reported that when negotiating a raised surface vertical toe clearance of the older adults were significantly lower than that of the young adults.

In the study by Patla and colleagues (1996), elderly adults were reported to step over obstacles with higher foot clearance than younger adults and place the foot farther from the obstacle. The mean age of the participants was not reported. It should be noted that higher obstacle heights were used in this study (7, 14, and 27 cm). The authors described that higher toe clearance observed in older adults resulted from hip hiking of the swing leg upwardly not by increasing the degree of hip and knee flexion of the swing limb. As reported by the authors, hip hiking also occurred in the young adult group, however, the relative contribution of hip hiking to toe clearance is considerably higher in the older adults and it increased as a function of obstacle height. Similar to Chen et al. (1991) have reported, the older adults placed their trail foot significantly farther away from the obstacle at take-off compare to the young adults.

In summary, the majority of the previous studies points out that the healthy older adults exhibited a more conservative strategy when crossing obstacles. Shorter crossing step lengths, slower crossing velocities, shorter post-obstacle heel strike distances, longer pre-obstacle toe approach distances, and lower foot-obstacle clearance were demonstrated by the older adults compared to the young adults during obstacle negotiation tasks. These findings suggest that older adults are at a greater risk for tripping during obstacle negotiation tasks than young adults because the probability for obstacle contact is enhanced by the low clearance height.
Control of the body center of mass during stepping over an obstacle

Approximately, one-third of the community dwelling elderly population aged over 65 years experience falls more than once a year. Age-related changes in the balance control system can be a factor contributing to the risk of imbalance during gait and tripping over obstacles. Dynamic stability is required during gait as a person’s ability to control the whole body center of mass during walking or stepping over obstacles is crucial to the prevention of fall, especially in the elderly.

The effects of age on body dynamic stability have been evaluated by Chou and colleagues (Chou et al., 2000; Chou et al., 2001). Chou et al. (2001) investigated the effect of obstacle height on the motion of the whole body's center of mass and its interaction with the center of pressure (COP) of the stance foot while negotiating obstacles. Six healthy young adults (3 males and 3 females) with mean age of 30 years participated in the study. Each participant performed unobstructed level walking and stepped over obstacles of heights corresponding to 2.5%, 5%, 10%, and 15% of his or her height. The authors reported that stepping over the higher obstacles resulted in significantly greater displacements of the COM in the anterior-posterior and vertical directions, a greater velocity of the COM in the vertical direction, and a greater anterior-posterior distance between the COM and COP. In contrast, the motion of the COM in the medial-lateral direction was not affected when negotiating obstacles of different heights.

The effects of age on body dynamic stability have been evaluated by Chou et al. (2000). Six healthy young adults with mean age of 30 year and six older adults with mean age of 70 years participated in the study. Gait analysis was performed at the unobstructed level walking and at the obstacles of heights of 2.5%, 5%, 10%, and 15% of each participant’s body height. It was reported that the healthy elderly adults crossed the obstacles with slower speed and took a
shorter step compared the young adults. The authors found that for the young adults, the displacements of the COM were greater in the anterior-posterior and medial-lateral directions for unobstructed level walking and all obstacle heights, except the 15% height. At the highest obstacle, the elderly adults had a significantly greater COM displacement in the medial-lateral direction.

Summary

Total hip arthroplasty is a common elective orthopedic procedure for the treatment of the painful arthritis of the hip. Patients who undergo total hip replacement can expect gains in walking ability as a result of gains in joint range as well as gain in speed of motion.

Previous research has examined how function of the hip is affected by the THA surgery, much of it focusing on general surgical outcomes such as range of motion, strength, kinetics, and joint kinematics. Based on research regarding gait analysis following the surgery, gait patterns of individuals with THA have demonstrated improvements in some aspects, but recovery to normal symmetric patterns continues to be limited. The literature points out that residual gait impairment persists in some patients for a number of reasons including maintenance of the preoperative gait pattern, continued pain, and decreased strength of the muscles around the hip.

Restoration of the hip abductor muscle strength is crucial to obtain a normal function and longevity of a total hip replacement. Although an increased femoral offset has been surmised to partially offset the weakness of the hip abductor muscles, its effect on physical function has not been conclusively established. Thus, it is necessary to better understand the effect of the femoral offset on the function of the hip abductor muscles when negotiating obstacles similar to those encountered in daily life.
CHAPTER 3

METHODS

Participants

Participants were recruited from a group of healthy patients who had undergone primary unilateral THA resulting from hip osteoarthritis (OA) at least 18 months previously from the same surgeon. Potential participants were prescreened by their personal physician or an orthopedic surgeon and excluded from the study if they had any of the following health related problems: hip revision surgery; previous hip fracture; hip dysplasia in either limb; the existence of a THA or other prosthetic implant in the opposite lower extremity; diagnosed neurological disorders, such as stroke or transient ischemic attacks; uncontrolled hypertension; cardiac disease; diabetes; diagnosed psychiatric disorders or use of psychotrophic drugs; diagnosed vestibular problems, such as vertebrovascular insufficiency, and bilateral vestibular loss; severe arthritis resulting in joint deformities of the knees (genu varus or genu valgus) or feet (hallux valgus, clubfoot, or equinus deformity); current or prior injuries or deformities in the lower extremities or the trunk that could influence lower extremity function, such as severe knee or foot injuries; lumbar spondylolthesis; or current symptoms of pain during practice trials of the test.

Twenty individuals with unilateral THA met the criteria and agreed to participate in the study. All participants signed a written informed consent approved by the Human Subjects Institutional Review Board at the University of Georgia. The participants were assigned into one
of two groups based on the difference of the femoral offset (FO) between their operated and non-operated hips. Standard hip and pelvic radiographs (Figures 2 and 3) of each THA participant taken on the day of examination were used to measure the FO of both hips (Steinberg & Harris, 1992). Using a 5% FO difference as a cutoff for grouping, eleven participants were classified into the high-offset (HI-FO) group (FO of the operated hip was greater than 105% of FO of the non-operated hip). The other nine participants were classified into the low-offset (LO-FO) group (FO of the operated hip was less than 95% of FO of the non-operated hip).
Participants’ characteristics and hip prosthesis information of both groups are summarized in Table 1. All participants had a reconstructed hip joint with cementless acetabular and femoral components. The femoral prosthesis was either a Secure-Fit™ stem (Stryker Howmedica Osteonics, Mahwah, N.J.) with a 132º neck-shaft angle, or an Accolade™ stem (Stryker Howmedica Osteonics, Mahwah, N.J.) with 132º or 127º degree neck-shaft angle.

Figure 3.

A schematic diagram shows the measurement of the femoral offset.
Instrumentation

- A Peak Motus Motion Measurement System® (Peak Performance Technologies, Inc., Englewood, CO) was used to collect the kinematic data. The system included four video cameras (Pulnix™, Model TM640) with a sampling rate of 60 Hz and a shutter speed of 1/1000 s, four VCRs (JVC™, Model BR-5378U), Peak Performance™ 25-point calibration frame (2.2185 m × 1.5830 m × 1.9110 m), Peak Motus Event & Video Control Unit® (EVCU) and 16 channel A/D interface unit, and Peak Motus® (v. 4.3.1) software.

- A Myopac EMG system™ (Run Technologies, Inc., Mission Viejo, CA) was used to collect the electromyographic (EMG) signals during gait and a stepping over an obstacle task. The EMG signals (sampling rate of 1080 Hz) from the electrode leads were sent to an EMG belt unit (selectable gain settings: 1,000 to 10,000, CMRR: 90 dB min at 60 Hz, and input impedance: 1MΩ), and transmitted to a receiver unit via a fiber optic transmission cable. A laptop computer was used to collect the EMG signals from the receiver via a 12-bit A/D board.

- The obstacle consisted of two upright frames and a movable 0.025-m wide x 0.60-m long cardboard strip. The strip was placed on the two frames with slots spaced in millimeters allowing the obstacle height to be adjusted relative to the participant’s averaged leg lengths. The obstacle was placed in the middle of a 10-m walkway.

- Photocells with 2 splits were used to monitor walking speed.

- A Cybex 6000 Dynamometer (Lumex Inc., Ronkonkoma, NY) was used for the test of isometric and concentric hip abductor muscle strength.
Experimental setup

Video Analysis

The four cameras were genlocked and arranged as shown in Figure 4. Video data and EMG signals were synchronized through the EVCU using a manual trigger switch.

Figure 4.
Experimental setup for the stepping over an obstacle task.
**Electromyography**

Following adequate skin preparation, three pairs of silver-silver chloride surface electrodes (Blue Sensor™, Medicotest, Inc.) were placed over the gluteus medius (GM), the rectus femoris (RF), and the biceps femoris (BF) muscles of THA limb with an inter-electrode distance of 2 cm. The ground electrode was placed over the tibial tuberosity of the right leg. Electrode placement locations were determined by one examiner following a standard procedure (Cram et al., 1998).

**Strength test**

For testing of the hip abductor muscle strength, the participant was positioned on a testing table in a side lying position facing away from the head of the dynamometer. The testing leg was positioned with hip and knee angles of 0° extension and neutral rotation. The non-testing leg was positioned in approximately 30° of hip flexion and 30° of knee flexion. Pillows were used to support and keep the test leg in neutral position. The location of the axis of rotation of the hip joint on the anterior aspect of the trunk was defined to lie about 1 cm medial to the anterior-posterior iliac spine (ASIS) and marked with a circular adhesive marker. The marker on the hip was used to guide the examiner and the participant while aligning the hip joint axis of rotation with the rotational axis of the dynamometer head. The lever arm length was adjusted to be equal to 80% of the participant’s upper leg length (upper leg length: greater trochanter to lateral femoral condyle).
General Protocol

- Each participant signed a consent form and answered the Pretest and the Hip Rating Questionnaires\(^2\) (Appendices A and B) regarding his or her general health, physical function, and activity level. Next, the participant’s body mass and height, leg lengths were determined.
- Prior to the stepping over an obstacle testing, spherical light-reflective markers were placed on locations of the participant’s body or clothing using a minimum of three markers for each segment (pelvis, thigh, lower leg, and foot) for the designated marker set for kinematic data collection (Appendix C). Next, the participant was videotaped while quietly standing in a natural posture as close to neutral as possible. The spatial locations of the reflective markers during natural standing were used in order to later generate segmental reference positions. The participant then performed the gait task.
- Following passive stretching and active-assistive warm-up exercise for the hip joints, the strength testing was administered.
- After completing all tests, the participants filled out the Posttest Questionnaire (Appendix D).

Protocol for the stepping over an obstacle tests

Participants were instructed to wear their customary shoes used most often for walking-related activities. The participant wore a pair of tight shorts and a sleeveless shirt for ease and accuracy of placing the EMG electrodes and reflective markers. A transfer belt was placed around the participant’s waist below the EMG belt pack to assist the participant in maintaining balance in the rare event that the participant began to lose balance. The tasks included stepping over obstacles corresponding to 0\% (i.e. level walking), 10\%, and 25\% of each participant’s

\(^2\) The Hip Rating Questionnaire (Johanson et al.,1992) uses a 100-point scale in which equal weight is given to the domains of the over all impact of arthritis, pain, walking, and function. The total score of the questionnaire correlated well ($r = -0.71$) with the score from the standardized test such as the arthritis impact-measurement scales. Reproducibility tested with the use of the kappa statistic was found to be good or excellent, $r = 0.58$ to 0.72 (Johanson et al.,1992).
averaged leg lengths. The averaged obstacle heights were $8.7 \pm 0.7$ cm and $21.8 \pm 1.7$ cm for the 10% and 25% conditions, respectively. The 0% obstacle was 0.025 m-wide strip of adhesive tape fixed to the floor. The selected obstacle heights correspond to those encountered during daily activities, such as walking across a floor or door threshold, and up a standard stair step (Austin et al., 1999).

Sufficient practice trials (up to three trials) were performed just prior to performing a new obstacle condition. The 0% obstacle condition was performed first, followed by the 10% and 25% conditions. During data collection, the participant was instructed to walk along the walkway at a natural, but consistent pace, step over the obstacle with the leading foot, and continue walking to the end of the walkway before stopping. Three trials were performed at each obstacle height with the nonoperated limb as the lead limb (first limb to step over the obstacle) and the THA limb as the support limb. Data from any trial during which the participant’s foot contacted the obstacle was not saved, and an additional trial was performed. Two trials with the closest average crossing speeds were selected for further analysis.

**Protocol for the strength tests**

After five minutes of supervised passive stretching and active-assistive warm-up exercise of the lower extremity, the strength testing began. The nonoperated limb was tested first for all tests. For each limb, two practice trials were performed at submaximal load in order to allow the participant to become familiarized with the test. Then, the participant was asked to perform two maximal isometric trials. The maximum isometric peak force for each trial was recorded. Preload for the concentric trials was equal to 80% of the maximum isometric peak force. The participant performed three single repetitions at maximum effort ($30^\circ\cdot{s}^{-1}$) with one minute of rest between
repetitions. The trial of the greatest peak concentric torque and the corresponding hip abduction angle was selected for later analysis.

Data Reduction

Kinematic data reduction

Following frame calibration and digitization of the 2-dimensional coordinates of the markers from the four cameras, a modified direct linear transformation (DLT) method was used to reconstruct 3-dimensional marker trajectories by use of the Peak Motus™ (V. 4.3.1) software. The raw data were then smoothed using a self-optimizing, quintic spline method (Jackson, 1979).

A 13-link rigid segment model, consisting of three links for each of the two lower extremities, two links for each of the upper extremities, one for the pelvis, one for the trunk, and one for the head, was used to calculate the whole body center of mass (COM) (Appendix E). Anthropometric estimates of each segment’s mass, center of mass location, and moments of inertia were obtained using the tables modified from Winter (1990) and de Leva (1996). The 3-dimensional trajectory of the body COM was calculated from the weighted sum of the COM from each body segment (Winter, 1990). The displacements of the COM (maximum value – minimum value achieved during the crossing stride) in the mediolateral, anteroposterior, and vertical directions were determined.

The kinematic data were generated based on one complete gait cycle (crossing stride) that begins with heel contact of the operated limb before crossing the obstacle and ends at the next heel contact of the operated limb after crossing the obstacle.

Angular displacement of the pelvis (pelvic obliquity, tilt, and rotation) and the hip joint of the THA limb (hip abduction, flexion, and rotation) were computed from the 3-dimensional quintic spline coordinates of the markers on the pelvis and the lower extremity (Appendices F
The knee and ankle joint flexion displacements were calculated using the Peak Motus® (v. 4.3.1) software. Maximum angular displacements of the pelvis, and the THA hip, knee, and ankle were calculated from the differences between the maximum and minimum angles during the gait cycle averaged over two testing trials. Foot-obstacle clearance was defined as the vertical distance between the toe marker and the marker placed on the midpoint of the obstacle bar at the point in time when the toe was directly over the obstacle. The following kinematics measures were generated to define participants’ gait characteristics: obstacle crossing speed, stance and swing durations, single and double limb support durations, step and stride lengths, and step width.

**EMG data reduction**

EMG activity of the GM, RF, and BF muscles recorded during stance phase of a crossing stride were selected for data analysis. Using Datapac 2000 software® (Run technologies, Inc., Mission Viejo, CA), root-mean-square (RMS) values were generated to quantify the level of muscular activity. The time constant used for the RMS processing was 0.053 second. Prior to the generation of RMS value, raw EMG activity was divided by gain factor used to magnify the EMG signal during data collection, notch-filtered (60 Hz), and bandpass- filtered (10-450 Hz). RMS values of the GM, RF, and BF muscles at the 10% and 25% obstacle heights were normalized to the averaged RMS values for the stance phase of the 0% walking condition (Yang & Winter, 1984). The normalized RMS of each muscle was then averaged separately over the three stance subphases: first double limb support, single limb support, and second double limb support.
Statistical Analysis

Comparisons of the abductor peak torque per body mass between the operated and nonoperated limbs within group and between the LO-FO operated and HI-FO operated limbs were made by paired $t$-tests and Student’s $t$-tests, respectively.

A 2 (FO group) x 3 (obstacle height) analysis of variance (ANOVA) with obstacle height as a within-subject factor and FO group as a between-subjects factor was performed to determine the differences between the two FO groups for the variables of the stepping over an obstacle task. Posthoc comparisons between obstacle height conditions were made using the least significant difference (LSD) adjustment method. Effect size estimation (Eta squared, $\eta^2$) is based on planned contrasts (Cohen, 1992a). Partial Eta squared ($\eta_p^2$) was reported for comparisons associated with ANOVA tests (Green et al., 2000). Traditionally, $\eta^2$ values of 0.01, 0.06, and 0.14 representing small, medium, and large effect sizes, respectively.

An alpha level of 0.05 was used for all statistical tests of significance. Statistical analyses were undertaken using the Statistics Package for Social Sciences (SPSS v.11 for Windows, SPSS, Inc., Chicago).
CHAPTER 4
RESULTS

Twenty participants with unilateral total hip arthroplasty, including seven men and 13 women, volunteered to participate in the study. There were eleven participants in the HI-FO group and nine participants in the LO-FO group. All participants performed three conditions of obstacle gait while being videotaped and EMG signals collected. Hip abductor strength data were collected separately from the walking and stepping over an obstacle data.

Participant characteristics and hip prosthesis information of both FO groups are summarized in Table 1. There were no significant differences in age, body height, body mass, and time since surgery between the HI- and LO-FO groups. The mean of the FO of the HI-FO group was significantly greater than that of the LO-FO group ($p < 0.001$). The hip-rating score was significantly higher for the HI-FO group compared to the LO-FO group, for the total score ($p = 0.012$) due to the higher score of the walking domain ($p = 0.002$).

Hip abductor strength measures

For strength testing, complete data for hip strength variables were available for ten and seven participants of the HI-FO group and the LO-FO group, respectively. Incomplete hip strength data for one HI-FO participant and one LO-FO participant were due to technical difficulties with the strength testing equipment. The strength data were not available for another participant of the LO-FO group due to self-reported discomfort during the test.
Comparisons for the hip abductor peak torque variables between the operated and nonoperated hips and between the two FO groups are shown in Table 2. For comparisons between the operated and nonoperated hips, the results indicated that the operated hip was significantly weaker than the nonoperated hip for both isometric ($p < 0.05$, $\eta^2 = 0.58$) and concentric test modes ($p < 0.05$, $\eta^2 = 0.51$) for both the LO- and HI-FO groups. No significant difference in the mean angle to peak concentric torque was found between the operated and nonoperated hips for either group.

Table 1.
Characteristics of participants (mean ± standard deviation).

<table>
<thead>
<tr>
<th>Characteristics</th>
<th>LO-FO group ($n = 9$)</th>
<th>HI-FO group ($n = 11$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (yr)</td>
<td>68 ± 7</td>
<td>65 ± 5</td>
</tr>
<tr>
<td>Body Height (m)</td>
<td>1.72 ± 0.14</td>
<td>1.78 ± 0.09</td>
</tr>
<tr>
<td>Body Mass (kg)</td>
<td>82.7 ± 14.8</td>
<td>98.4 ± 20.5</td>
</tr>
<tr>
<td>Postoperative time (mo)</td>
<td>51 ± 27</td>
<td>41 ± 14</td>
</tr>
<tr>
<td>FO THA side (mm)</td>
<td>38 ± 9*</td>
<td>48 ± 6</td>
</tr>
<tr>
<td>Hip rating score (Total)</td>
<td>89 ± 7*</td>
<td>97 ± 3</td>
</tr>
<tr>
<td>Overall impact</td>
<td>22 ± 3</td>
<td>24 ± 3</td>
</tr>
<tr>
<td>Pain</td>
<td>23 ± 2</td>
<td>24 ± 1</td>
</tr>
<tr>
<td>Walking</td>
<td>20 ± 2*</td>
<td>24 ± 2</td>
</tr>
<tr>
<td>Function</td>
<td>24 ± 1</td>
<td>25 ± 1</td>
</tr>
<tr>
<td>Gender</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Number of females</td>
<td>4</td>
<td>3</td>
</tr>
<tr>
<td>Number of males</td>
<td>5</td>
<td>8</td>
</tr>
<tr>
<td>Surgical Approach</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Number of Antero-Lateral</td>
<td>4</td>
<td>11</td>
</tr>
<tr>
<td>Number of Posterior</td>
<td>5</td>
<td></td>
</tr>
</tbody>
</table>

Note: * LO-FO group significantly different from HI-FO group ($p < 0.05$).
Table 2.

Peak isometric and concentric torques (mean ± standard deviation) of the hip abductor muscles between LO- FO and HI-FO groups, inter-limb comparisons within each FO group, and between the corresponding limbs of the FO groups.

<table>
<thead>
<tr>
<th>Variable</th>
<th>FO group</th>
<th>LO-FO (n = 7)</th>
<th>Inter-limb comparison</th>
<th>HI-FO (n = 10)</th>
<th>Inter-limb comparison</th>
<th>Between FO group comparison</th>
</tr>
</thead>
<tbody>
<tr>
<td>Isometric torque/body mass</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(Nm/kg)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Operated hip</td>
<td></td>
<td>1.04 ± 0.22</td>
<td>p = 0.047</td>
<td>1.34 ± 0.34</td>
<td>p = 0.003</td>
<td>p = 0.059</td>
</tr>
<tr>
<td>Nonoperated hip</td>
<td></td>
<td>1.38 ± 0.26</td>
<td></td>
<td>1.49 ± 0.33</td>
<td></td>
<td>p = 0.492</td>
</tr>
<tr>
<td>Operated:Nonoperated (%)</td>
<td></td>
<td>75.34 ± 10.3</td>
<td></td>
<td>90.52 ± 16.1</td>
<td></td>
<td>p = 0.045</td>
</tr>
<tr>
<td>Concentric torque/body mass</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(Nm/kg)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Operated hip</td>
<td></td>
<td>0.75 ± 0.25</td>
<td>p = 0.037</td>
<td>1.05 ± 0.34</td>
<td>p = 0.005</td>
<td>p = 0.070</td>
</tr>
<tr>
<td>Nonoperated hip</td>
<td></td>
<td>1.13 ± 0.36</td>
<td></td>
<td>1.20 ± 0.36</td>
<td></td>
<td>p = 0.713</td>
</tr>
<tr>
<td>Operated:Nonoperated (%)</td>
<td></td>
<td>67.1 ± 11.5</td>
<td></td>
<td>88.1 ± 15.1</td>
<td></td>
<td>p = 0.007</td>
</tr>
<tr>
<td>Angle of peak torque (°)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Operated hip</td>
<td></td>
<td>9 ± 5</td>
<td>p = 0.757</td>
<td>10 ± 4</td>
<td>p = 0.512</td>
<td>p = 0.653</td>
</tr>
<tr>
<td>Nonoperated hip</td>
<td></td>
<td>10 ± 4</td>
<td></td>
<td>12 ± 4</td>
<td></td>
<td>p = 0.203</td>
</tr>
</tbody>
</table>
Comparisons were made between FO groups for isometric and concentric strength values for both the operated and nonoperated hips. For the nonoperated hip, the group means for isometric peak torque did not differ by more than 0.11 Nm/kg between FO groups, and as such, were not statistically different. For the operated hip, the HI-FO group demonstrated a tendency ($p = 0.059, \eta^2 = 0.22$) to exhibit a greater mean isometric peak torque. Based on Cohen's $d$, low observed power (0.62) was detected (Cohen, 1992b) for FO group comparisons. However, a significantly greater operated:nonoperated isometric torque ratio ($p = 0.045, \eta^2 = 0.24$) was exhibited by the HI-FO THA limb strength (90.1% of nonoperated limb), compared to the LO-FO group (75.3% of nonoperated limb). Thus the strength of the HI-FO THA limb was more closely approximated the strength of the nonoperated limb.

For strength testing in an isokinetic test mode, for the nonoperated hip, no significant FO group difference for mean concentric peak torque was found within either FO group. For the THA limb, difference of mean concentric peak torque between groups also approached significance ($p = 0.070, \eta^2 = 0.20$, observed power = 0.58). However, when expressed as the ratio of operated:nonoperated limb, the 21% difference between groups was significant ($p = 0.007, \eta^2 = 0.39$). No significant FO group differences for the mean angle to peak concentric torque were found for either the THA or nonoperated hip as the hip abduction angle varied by less than two degrees. Similarly, no interlimb differences existed for either FO group.
Kinematics measures

Spatiotemporal measures of obstacle crossing

Gait variables of the two FO groups during walking and stepping over an obstacle with the THA limb as the support limb are presented in Table 3. The FO group means for gait velocity were not statistically different with the groups differing by no more than 0.05 m-s$^{-1}$ at any obstacle height. For the gait phase durations, a FO group x obstacle height interaction was detected for the first double limb support phase ($p = 0.027$, $\eta^2_p = 0.67$), with an increase in the duration of this subphase was observed for the LO-FO group compared to the HI-FO group at the 25% obstacle height condition. However, group differences in single and second double limb support duration, and swing duration were not significantly different. No statistically significant differences were detected for the spatial step and stride measures.

For obstacle height comparisons, the results showed that obstacle height had a significant effect on gait variables. As obstacle height increased, mean crossing speed significantly decreased for both FO groups ($p < 0.001$, $\eta^2_p = 0.73$). For both FO groups, increases in obstacle height led to significant increases in single support duration, and swing duration ($p < 0.001$), but had no effect on any double support durations. Increases in obstacle height led to significant increases in step length ($p = 0.001$, $\eta^2_p = 0.34$) but did not lead to increases in stride length or step width.
Table 3.

Gait kinematics (mean ± standard deviation) for LO-FO and HI-FO groups during stepping over an obstacle at different obstacle heights.

<table>
<thead>
<tr>
<th>Variable</th>
<th>FO group</th>
<th>Obstacle height condition</th>
<th>p values</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>0%</td>
<td>10%</td>
</tr>
<tr>
<td>Gait velocity (m·s⁻¹)</td>
<td>LO-FO</td>
<td>1.06 ± 0.25</td>
<td>0.96 ± 0.29</td>
</tr>
<tr>
<td></td>
<td>HI-FO</td>
<td>1.09 ± 0.19</td>
<td>0.94 ± 0.16</td>
</tr>
<tr>
<td></td>
<td>(M \pm SD)</td>
<td>1.08 ± 0.21[^a]</td>
<td>0.95 ± 0.22[^bc]</td>
</tr>
<tr>
<td>Stance time (s)</td>
<td>LO-FO</td>
<td>0.17 ± 0.04</td>
<td>0.17 ± 0.06</td>
</tr>
<tr>
<td></td>
<td>HI-FO</td>
<td>0.18 ± 0.04</td>
<td>0.17 ± 0.05</td>
</tr>
<tr>
<td></td>
<td>(M \pm SD)</td>
<td>0.46 ± 0.05[^ab]</td>
<td>0.62 ± 0.08[^bc]</td>
</tr>
<tr>
<td>1st double support (s)</td>
<td>LO-FO</td>
<td>0.46 ± 0.04</td>
<td>0.62 ± 0.08</td>
</tr>
<tr>
<td></td>
<td>HI-FO</td>
<td>0.45 ± 0.06</td>
<td>0.62 ± 0.09</td>
</tr>
<tr>
<td></td>
<td>(M \pm SD)</td>
<td>0.46 ± 0.05[^a]</td>
<td>0.62 ± 0.08[^bc]</td>
</tr>
<tr>
<td>2nd double support (s)</td>
<td>LO-FO</td>
<td>0.15 ± 0.04</td>
<td>0.15 ± 0.04</td>
</tr>
<tr>
<td></td>
<td>HI-FO</td>
<td>0.16 ± 0.04</td>
<td>0.15 ± 0.04</td>
</tr>
<tr>
<td>Swing time (s)</td>
<td>LO-FO</td>
<td>0.45 ± 0.04</td>
<td>0.54 ± 0.08</td>
</tr>
<tr>
<td></td>
<td>HI-FO</td>
<td>0.42 ± 0.04</td>
<td>0.55 ± 0.05</td>
</tr>
<tr>
<td></td>
<td>(M \pm SD)</td>
<td>0.43 ± 0.04[^a]</td>
<td>0.55 ± 0.06[^bc]</td>
</tr>
<tr>
<td>Step width/inter-ASIS</td>
<td>distance</td>
<td>LO-FO</td>
<td>0.36 ± 0.26</td>
</tr>
<tr>
<td></td>
<td>HI-FO</td>
<td>0.31 ± 0.15</td>
<td>0.43 ± 0.19</td>
</tr>
<tr>
<td>Step length/leg length</td>
<td>LO-FO</td>
<td>0.80 ± 0.07</td>
<td>0.85 ± 0.08</td>
</tr>
<tr>
<td></td>
<td>HI-FO</td>
<td>0.77 ± 0.11</td>
<td>0.82 ± 0.08</td>
</tr>
<tr>
<td></td>
<td>(M \pm SD)</td>
<td>0.78 ± 0.09[^ab]</td>
<td>0.83 ± 0.08[^a]</td>
</tr>
<tr>
<td>Stride length/leg length</td>
<td>LO-FO</td>
<td>1.54 ± 0.16</td>
<td>1.61 ± 0.16</td>
</tr>
<tr>
<td></td>
<td>HI-FO</td>
<td>1.52 ± 0.23</td>
<td>1.50 ± 0.33</td>
</tr>
</tbody>
</table>

Note: \(p\) values: \(p_g\) represents overall FO group effect; \(p_h\) represents obstacle height effect; \(p_{gh}\) represents group \(x\) obstacle height interaction. Pooled mean and standard deviation \((M \pm SD)\) of LO- and HI-FO groups are shown where significant obstacle height effect is present. For a given variable, means sharing a letter \((e.g., a, b, c)\) differ significantly \((p < 0.01)\) by the least significant difference (LSD) pairwise comparisons.
Pelvic and lower extremity joint displacements

Typical angular position-time profiles for pelvic obliquity, tilt, and rotation of representative participants are shown in Figure 5. Angular displacements of the pelvis in all three directions are shown in Figure 6. No statistical differences were detected for FO group x obstacle height interactions or between FO groups for pelvic angular displacements for any angular direction (observed power = 0.23). However, posthoc pairwise comparisons revealed that the LO-FO group exhibited a greater forward tilt displacement (7º) than the HI-FO group (5º) during level walking (p = 0.043).

However, for the obstacle height main effect, both FO groups demonstrated an increase in the magnitude of pelvic obliquity, tilt, and rotation displacements as obstacle height increased (p < .001). For pelvic obliquity, both FO groups demonstrated a pattern of ‘hip hiking’, in which the swing-side (nonoperated) hip was elevated above the support-side (THA) hip while crossing the obstacle. During the 25% obstacle height condition, the pelvic obliquity displacement increased 2.1 and 2.5 times the values of the level walking condition for the LO- and HI-FO group, respectively. Compared to the values displayed during the level walking condition, the pelvic tilt displacement during crossing the 25% obstacle height increased 3.1 and 4.0 times, respectively, for the LO- and HI-FO groups. At the 25% obstacle height, the pelvic rotation displacement increased to about 1.6 times the value exhibited during level walking.
Pelvic obliquity, tilt, and rotation at different obstacle heights of a representative THA participant for one gait cycle.
Figure 6.

Total angular displacements of the pelvis (mean ± standard deviation) of LO-FO and HI-FO groups at different heights during stepping over an obstacle.

**Note:** For a given variable, means sharing a letter (e.g., a, b, c) differ significantly \((p < 0.005)\) by the LSD pairwise comparisons for the obstacle height main effect.

Pictures illustrating pelvic displacement are modified from Perry (1992).
Means and standard deviations of hip joint angular displacements in all three directions are presented in Figure 7. No statistical differences were detected for FO group x obstacle height interactions for hip displacements for any direction. For the FO group main effect, however, a significant group difference was detected for hip abduction displacement ($p = 0.039$, $\eta_p^2 = 0.22$). Posthoc pairwise comparisons revealed that the hip adduction displacement was significantly greater for the HI-FO than the LO-FO group at the 10% obstacle height condition. There were no statistically significant group differences for hip flexion or rotation displacement during level walking or stepping over an obstacle at any height. Increases in obstacle height led to significant increases in the displacement of the hip joint for abduction ($p = 0.014$, $\eta_p^2 = 0.21$) and flexion ($p < 0.001$, $\eta_p^2 = 0.71$). Hip rotation displacement was not significantly altered as obstacle height increased.

Only the flexion/extension movements are reported for the knee and ankle joints (Figure 8). A significant FO group x obstacle height interaction ($p < 0.001$, $\eta_p^2 = 0.39$) was detected for the knee joint. While the magnitude of the knee flexion displacement was fairly equal for both FO groups during the level walking condition, the HI-FO group exhibited knee flexion displacement value that was approximately 11º greater than that of the LO-FO group for both the 10% ($p = 0.047$) and 25% ($p = 0.041$) obstacle height conditions.

No significant FO group differences were found for the ankle joint. However, increasing obstacle height from level walking (0%) to 25% obstacle height significantly increased the magnitude of dorsiflexion displacement ($p = 0.038$).
Figure 7.

Total angular displacements of the hip (mean ± standard deviation) of LO-FO and HI-FO groups at different heights during stepping over an obstacle.

Note: * LO-FO group significantly different from HI-FO group at a given height ($p < 0.05$).
For a given variable, means sharing a letter (e.g., a, b) differ significantly ($p < 0.05$) by the LSD pairwise comparisons for the obstacle height main effect.
Figure 8.

Total angular displacements of the knee and ankle (mean ± standard deviation) of LO-FO and HI-FO groups at different heights during stepping over an obstacle.

Note: LO-FO group significantly different from HI-FO group at a given height ($p < 0.05$).

*a 25% obstacle height condition significantly different from level walking ($p < 0.05$) for the obstacle height main effect.
Foot obstacle clearance

Vertical toe clearances of the lead (nonoperated) foot and trail (THA) foot when crossing the obstacle for both FO groups are summarized in Figure 9. Both FO groups displayed the same mean value of 0.15 m for lead foot-obstacle clearance, therefore no significant group difference was found. However, a significant FO group difference was observed for the trail foot-obstacle clearance ($p = 0.029$, $\eta_p^2 = 0.24$). Posthoc pairwise comparisons revealed that the HI-FO group cleared the obstacle by approximately 0.05 m more than the LO-FO group at the 10% obstacle height condition.

As obstacle height increased, lead and trail foot clearances increased for both FO groups ($p < 0.001$). Posthoc pairwise comparisons revealed that, for both FO groups, there were significant increases in lead foot and trail foot-obstacle clearances when crossing the obstacle at any height, compared to level walking condition.
Figure 9.

Foot-obstacle clearance (mean + standard deviation) of LO-FO and HI-FO groups at different heights.

Note: * LO-FO group significantly different from HI-FO group at a given height (\(p < 0.05\)). For a given variable, means sharing a letter (e.g., a, b) differ significantly (\(p < 0.001\)) by the LSD pairwise comparisons for the obstacle height main effect.
Displacement of the body center of mass (COM)

Means and standard deviations of the displacement of the body COM in the mediolateral direction relative to participant’s hip width and the anteroposterior and vertical directions relative to participant’s leg length during crossing an obstacle are presented in Figure 10. A significant group x obstacle height interaction was detected for the mediolateral direction ($p = 0.036$). Although group differences were not statistically significant for level walking and the 10% obstacle height condition, the LO-FO group exhibited a significantly greater mediolateral displacement than the HI-FO group ($p = 0.034$) for the 25% obstacle height condition. No significant group differences for the anteroposterior and vertical displacements were detected at any obstacle height as the overall group differences were less than 1% for both directions.

For obstacle height effects, a significant increase in the magnitude of the body COM displacement occurred for both the anteroposterior and vertical directions ($p < 0.001$). There was approximately 14% increase in the anteroposterior and a 7% increase in the vertical displacements, from the level walking to the 25% obstacle height conditions.
Figure 10.

Relative body COM displacements (mean + standard deviation) of LO-FO and HI-FO groups at different heights during stepping over an obstacle.

Note: * LO-FO group significantly different from HI-FO group.

For a given variable, means sharing a letter (e.g., a, b, c) differ significantly ($p < 0.005$) by the LSD pairwise comparisons for the obstacle height main effect.
Muscle activation

Typical RMS-time profiles the muscles observed in this study are shown in Figure 11. Qualitatively, during stepping over an obstacle, activation of the GM typically was demonstrated throughout the stance phase. The RF typically had two major periods of activity: at the beginning of the single support phase and during the early swing phase. The BF was activated later in the gait cycle during the second double support phase and at the end of swing phase.

Averaged RMS EMG activity of each subphase of the stance period for the GM, RF, and BF of the THA limb are presented in Figure 12. No significant FO group x obstacle height interactions were detected for the RMS EMG values of any muscles. For FO group comparisons, the averaged RMS EMG of the GM of the LO-FO group was significantly greater than that of the HI-FO group during the single limb support phase ($p = 0.012$, $\eta_p^2 = 0.30$).

Although no significant FO group differences were detected for the RF due to between-group variability, during the single limb support phase, based on posthoc analysis, the LO-FO group generated 33% higher RMS EMG activity at the 10% obstacle height ($p = 0.025$) and 36% higher activity at the 25% obstacle height conditions, although nonsignificant ($p = 0.294$). There were no statistically significant differences between FO groups in the averaged RMS EMG activity of the BF at any subphase of the stance period.

An increase in obstacle height from the 10% to 25% height condition did not affect the activation of GM of the HI-FO group, based on posthoc analysis. However, the averaged RMS EMG of the GM, for the LO-FO group, significantly increased by 20% - 30% ($p < 0.05$) for all three of the stance subphases. For the RF, during single limb support, obstacle height led to an increase of 50% for the averaged RMS EMG ($p = 0.005$) for both FO groups. The EMG activity
of the BF significantly increased by 44% as the obstacle height increased during the second limb support phase ($p = 0.043$).

![Graphs showing EMG activity]

Figure 11.

Typical RMS-time profiles of the EMG activity of the gluteus medius, rectus femoris, and biceps femoris at different heights during stepping over an obstacle.

Note: Average RMS EMG (%) values are normalized to the level walking (0%) condition.
Figure 12.

Normalized RMS EMG (mean + standard deviation) of the gluteus medius, rectus femoris, and biceps femoris of LO-FO and HI-FO groups at different heights during stance phase.

**Note:**
* LO-FO group significantly different from HI-FO group.
# 25% obstacle height significantly different from 10% obstacle height.

Average RMS EMG (%) values are normalized to 0% condition.
CHAPTER 5
DISCUSSION

The purpose of this study was to compare biomechanical characteristics of THA individuals with high and low femoral offset during stepping over an obstacle. Variables compared between FO groups were hip abductor strength, displacements of the pelvis and lower extremity joints, displacement of the body COM, and hip abductor muscle EMG activity. Therefore, in the first part of discussion, the effect of increasing femoral offset (FO) on hip abductor strength will be addressed. Differences between FO groups for kinematics and EMG activity of lower extremity muscles while crossing an obstacle of different heights will be discussed in the second section.

Hip abductor strength

The ability to generate sufficient moments of the hip abductor muscles to counteract the frontal plane torque resulting from body weight during walking or performance of other activities of daily life is critical for individuals who undergo hip replacement. However, weakness of the abductor muscles has been reported as a result of the underlying disease process of osteoarthritis and may be due to neuromuscular damage of the muscles during surgery (Arokoski et al., 2002; Baker & Bitounis, 1989). Therefore, modifying the implant and/or its location to create a greater hip abductor muscle moment may potentially compensate reduced muscle force.

Previous research has shown that an increase in femoral offset is associated with an increase in the hip abduction muscle moment arm and abduction range of motion (McG Rory et
al., 1995) that could lead to better mechanical advantage of the hip abductors, and consequently, greater hip abductor muscle moments. Therefore, for this study, it was hypothesized that the participants with high FO, compared to those with low FO, would exhibit greater hip abductor muscle moments during maximal effort strength testing.

Indeed, the findings of this study suggest that participants with high FO exhibited less strength deficit of the hip abductor muscles than participants with low FO. For the maximum isometric and concentric torques, compared to the LO-FO group, the HI-FO group had a significantly higher operated:nonoperated abductor strength ratio. For each participant, hip abductor torques of the operated limb were normalized to the corresponding values of the nonoperated hip and used for representing the magnitude of strength deficit of the hip abductor muscles. It is believed that this method was appropriate and more meaningful to compare participants, as each participant’s nonoperated limb served as a control limb.

Furthermore, for THA limb comparisons between FO groups, there was a tendency for the HI-FO group to generate greater isometric and concentric torques compared to the LO-FO group. These absolute differences in THA limb abductor strength, although not quite achieving statistical significance are worth noting because it is likely that lack of statistical power largely precluded achieving significance.

Previous research has shown that muscle strength of the operated hip may not improve to the level of the nonoperated hip several years after surgery. For the current study, the THA limb isometric strength deficit (9% for the HI-FO; 15% for the LO-FO group) was less than those previously reported by Sicard-Rosenbaum et al. (2002) and Shih et al. (1994). Sicard-Rosenbaum et al. (2002) reported a deficit in isometric hip abductor strength of 26% at two years
after surgery. However, for Shih et al.’s study, a deficit in the operated hip abductor strength of 19%, compared to the nonoperated hip, was observed at one year after surgery.

According to Brown et al. (1980), after THA, the absence of joint pain should allow for increased participation in daily activities and consequently allow for the restoration of muscular function in the operated and unaffected hips. For this study, although the hip abductor strength of the nonoperated limb was not statistically different between the FO groups (less than 0.11 Nm/kg), it appears that the THA limb of the LO-FO group did not have the same mechanical advantage, compared to the HI-FO group, in gaining strength of the hip abductor muscles, due to a shorter FO. It is more likely that significant differences in strength measures between the LO- and HI-FO groups found in this study could be largely attributed to the difference in FO than the HI-FO group just being a stronger group overall. Thus, these findings further support the surmise that a relatively longer FO would provide a better mechanical advantage of the hip abductor muscles for the HI-FO group, compared to the LO-FO group, to reduce the hip abductor strength deficit and allow for greater hip abductor muscle moment.

For the averaged angle of peak torque, the results did not support the surmise that the angle to peak hip abductor torque would be different between the two FO groups as the angle of peak concentric torque (approximately 10°) did not differ by more than 2° between the FO groups. One explanation for lack of a group difference in the angle of peak torque could be that an increase in FO affects only the moment generating capacity of the muscles being generated not the angle-moment relationship.

To our knowledge, the angle of peak torque during isokinetic hip abduction for older adults of similar age has not been established. However, the averaged angle of peak torque observed in this study is different from the averaged angle observed in healthy young adults
Donatelli et al. (1991) reported the angle of peak torque during concentric hip abduction in 42 young adults (mean age 26.3 yrs) to be at 0°. It is not known why peak torque occurred at a slightly abducted position compared to neutral position of young individuals. It is possible that differences in hip abductor muscle condition of the replaced hip and/or isokinetic test protocols might contribute to varying results.

**Stepping over an obstacle task**

No significant group differences in gait speed were detected at any obstacle height condition. This result suggested that the overall stepping over an obstacle task was equivalent for both FO groups. Thus, kinematics differences between the FO groups were more likely due to FO, not walking speed. For this study, gait speed of 1.07 m·s⁻¹ during level walking for both FO groups was comparable to the value previously reported for THA patients by Perron et al., (2000) but was found to be significantly slower than their healthy control group of similar age (Kyriazis & Rigas, 2002; Vogt et al., 2003). For Perron et al.’s study, a shorter stride length and a decreased cadence characterized the slower walking speed of THA patients. Similar observation was made in patients with OA prior to hip replacement surgery (Watelain et al., 2001).

It was predicted that hip abductor strength deficits would cause biomechanical differences during stepping over an obstacle. In addition to the differences in strength measures, the findings of this study suggest that, kinematic and EMG differences between FO groups also support the premise that less abductor strength deficit found in the HI-FO participants, compared to the LO-FO participants, would promote their performance while stepping over an obstacle.

First, for kinematic measures, an increase in the duration of the first double limb support phase was demonstrated by the LO-FO group, compared to the HI-FO group. This result was
consistent with past gait literature. Perron et al. (2000) reported that compared to age-matched healthy older adults, their THA participants exhibited an increase in the duration of the first double limb support phase that was associated with a significant reduction in abductor muscle moment. In addition, it has been reported that hip abductor muscles play a crucial role in the production of lateral ground reaction forces during stance phase (Rogers & Pai, 1993). Thus, the finding of this current study indicated that compared to the HI-FO group, the LO-FO group, potentially, tended to need more time to generate lateral ground reaction forces while stepping over an obstacle of higher height.

Second, the results of this study indicated that the presence of an obstacle in the travel path increased the challenge to stability of both FO groups, especially in the mediolateral direction. For this study, as obstacle height increased, both FO groups increased their mediolateral COM displacement compared to the values exhibited for the unobstructed level walking condition. However, the HI-FO group, compared to the LO-FO group, demonstrated significantly less mediolateral displacement of the body COM when encountering the 25% obstacle height condition. Thus, from the findings of this study, it is suggested that the increased mediolateral displacement at the 25% obstacle height condition demonstrated by the participants with low FO indicates increased potential for difficulties in maintaining dynamic stability in the frontal plane and, possibly, a greater risk of falling, compared to those individuals with high FO.

According to Winter et al. (1993), the hip abductor muscles and medial linear acceleration of the hip joint are partly responsible for controlling the mediolateral balance of the upper body during single limb support. Control of the body COM motion in the frontal plane has been used as a functional indicator of balance maintenance during walking. It was observed that elderly patients with balance disorders demonstrated greater lateral displacement of the COM
when crossing an obstacle compared to healthy elderly adults (Chou et al., 2002). It has also been observed that individuals whose their body weight is shifted further away from the medial border of the foot during gait may be demonstrating a less stable pattern of dynamic stability (Bauby & Kuo, 2000). This might increase the risk of falling to the side in frail older adults (Greenspan et al., 1998). In addition, Trudelle-Jackson (2002) reported that THA patients with decreased hip and knee muscle strength exhibited postural instability for both the mediolateral and anteroposterior directions during single leg stance of the operated limb compared to that of the nonoperated limb.

Third, the hypothesis was supported that HI-FO participants, compared to the LO-FO participants, would exhibit lower activation of hip abductor muscles during the single limb support phase while stepping over an obstacle. EMG has been utilized as a quantitative method for determining muscle force (Conwit et al., 1999; Suzuki et al., 2002). Thus, the increased EMG activity in the THA limb of the LO-FO group reflected a greater demand placed on the hip abductor muscles measured to compensate for reduced muscle force and/or leverage to prevent the pelvis from dropping laterally over the swing (nonoperated) limb while stepping over the obstacle. Furthermore, the EMG activity of the rectus femoris during single limb support tended to be higher for the LO-FO group than that exhibited by the HI-FO group (p = 0.104), indicating that more activation from other THA limb muscles, i.e., knee extensor muscles, was needed for the LO-FO participants, compared to the HI-FO participants.

In conjunction with greater displacement of the body COM, the results suggest that the increased activation of the hip abductor and knee extensor muscles of the stance limb of the LO-FO group was needed to generate sufficient muscle moments in an attempt to reduce the excessive displacement of the body center of mass during stance phase. This recruitment also
helped to maintain the line of progression within the base of support. These strategies may enable individuals with low offset to employ a safer, more stable gait pattern to accommodate for diminished hip abductor muscle strength on the operated side.

It was also hypothesized that displacements of the pelvis, particularly pelvic obliquity, would be less for the HI-FO group than the LO-FO group. This hypothesis was not supported by the results for any of the obstacle height conditions, as none of the pelvic displacements were statistically different between FO groups. In addition, no excessive pelvic obliquity displacement (pelvic drop) was observed for either the LO- or HI-FO group. The stepping over an obstacle task requires that a person must generate hip and knee flexion in order to safely cross the obstacle. Indeed, all participants of both FO groups successfully performed the stepping task without contacting or stepping onto the obstacle. Moreover, during stepping over a higher obstacle, the hip and knee joints of the nonsupport limb (the nonoperated limb) needed to be flexed more to raise the foot over the obstacle. Thus, it is likely that these movements initiated by the nonoperated hip muscles could have constrained any excessive movement of the pelvis in the frontal plane.

However, it should be noted that the HI-FO group had significantly greater knee displacement of the THA limb and tended to demonstrate a higher hip elevation and, as such, contributed to a higher foot-obstacle clearance of the operated limb while stepping over an obstacle. These movement patterns found in the participants with high offset are similar to the strategies used by healthy older adults to avoid contact with an obstacle (Patla et al., 1996). These authors noted that foot clearance is considered the most critical parameter for safe obstacle crossing. The LO-FO participants cleared the obstacle with less margin for error than the HI-FO
group, and this deficit could be a disadvantage in terms of avoiding a trip when stepping upwards onto a curb or a raised surface.

The last evidence of reduced walking ability of the LO-FO group, compared to the HI-FO group, is the results of the hip rating questionnaire. The LO-FO group scored their perceived walking ability lower than the HI-FO group on the hip rating questionnaire, reflecting that the participants with a low FO perceived that they had a decreased mobility level compared to those with high FO. Specifically, the majority of the participants with high FO (7 out of 11) indicated that they did not have difficulty walking an unlimited distance, i.e., they could walk more than 20 city blocks, while only 1 out of 9 individuals from the LO-FO group reported this ability. This is consistent with our findings of greater hip abductor strength deficit of the LO-FO group.

Residual strength deficits of the hip muscles have been shown to contribute to impairment in functional performance of THA patients. Furthermore, Vaz et al. (1993) stated that hip abductor torque could be used as a predictor of walking ability. They found that hip abductor torques were modestly correlated ($r = 0.51$) to distance covered during a 6-minute walk. In addition, Trudell-Jackson also noted a moderate association ($r = 0.56$) between hip abductor strength and hip functional assessment scores.

Other factors besides femoral offset minimally affected the observed differences between FO groups. Variations in the surgical approach, i.e., different incisions in different muscles, have raised some concern regarding subsequent effects on postoperative recovery of the hip abductor muscles (Booth et al., 1988). For the current study, two surgical approaches for total hip replacement, the anterolateral and posterior approaches, were used. To determine the effect of surgical approach on our data, comparisons between the two approaches for the strength measures was performed using only strength data from the LO-FO group ($n = 3$ for anterolateral
approach; \( n = 4 \) for posterior approach). No significant differences between the two surgical approaches were found on any strength measure. These results are consistent with previous research by Barber et al. (1996) and Downing et al. (2001). These investigators reported that THA patients receiving either the lateral or posterior approach demonstrated similar strength, functional, and radiological outcomes.
CHAPTER 6
SUMMARY AND CONCLUSIONS

Summary

The purpose of this study was to determine if THA individuals with high femoral offset compared to those with low femoral offset would exhibit greater hip abductor strength during isometric and concentric tests, and display differences for pelvic and lower extremity joint displacements, body COM displacement, and hip abductor muscle EMG activity while stepping over an obstacle. The current study is the first, to our knowledge, to compare hip abductor muscle function of individuals with short femoral offset to those with relatively longer femoral offset, and to determine biomechanical differences due to femoral offset for such a functionally demanding task as stepping over an obstacle.

In summary, during maximal effort strength testing, individuals with high femoral offset, compared to those with low femoral offset, demonstrated:

1. less strength deficit of the hip abductor muscles and a tendency to generate greater hip abductor moments for the THA limb during maximal effort strength testing.

In addition, the findings on kinematic parameters point out that the hip abductor strength may affect obstacle crossing performance of THA participants. It was observed that while negotiating an obstacle, participants with high femoral offset, compared to those with low femoral offset, demonstrated:

2. increased gait stability, indicated by less mediolateral displacement of the body’s center of mass;
3. increased ability to clear the obstacle with higher toe clearance of the THA limb due to increased knee flexion;

4. less muscular effort, as reflected by less EMG activation of the gluteus medius muscle of the THA limb, to stabilize the pelvis during the stance phase.

Conclusions

In conclusion, from the results of this study, some new and important findings of hip abductor muscle functioning with respect to the femoral implant offset are revealed. It is confirmed that there are differences in the manner in which locomotion is achieved between individuals with high and low femoral offset. An increase in femoral offset was found to be beneficial to individuals who received total hip arthroplasty and, at least partially, compensate for weakness of the hip abductor muscles.

Due to the possibility of the persistence of hip abductor weakness after surgery, this study confirms the use of an increased femoral offset as an effective choice to enhance hip abductor muscle function and promote balance control during ambulation. The information from the current study may also assist orthopaedic surgeons in making decisions about the most appropriate implant to use with a given patient. Most importantly, the information may aid future THA individuals to achieve better outcomes from hip replacement surgery and allow functional performance similar to healthy individuals of the same age.
REFERENCES


APPENDIX A

PRETEST QUESTIONNAIRE

I.D. ____________________
Date __________ ___

Have you started taking medications or changed the dosages of any medications that have the following side effects? Circle Yes or No

a. Balance problems       Yes / No
b. Produces nausea during physical activity Yes / No
c. Dizziness              Yes / No
d. Vision                 Yes / No
e. Coordination           Yes / No

Comments
________________________________________________________________________
________________________________________________________________________

Have you participated in any exercise or physical therapy program for your hip? Yes / No
For how long? __________________________________________________________________

Comments
________________________________________________________________________
________________________________________________________________________

Physical activity

Mark the activities in which you currently participate:

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<th>Activity</th>
<th>How long have you been participating?</th>
<th>Times/week</th>
<th>Duration</th>
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<tr>
<td>Yardwork</td>
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<tr>
<td>Others (please indicate)</td>
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</tbody>
</table>
APPENDIX B

HIP RATING QUESTIONNAIRE

Which hip did you have replacement surgery? (circle one)

Left  Right

Please answer the following questions about the hip you have just indicated.

1. Consider all of the ways that your hip affects you, mark (X) on the scale for how well you are doing (25 pts).

<table>
<thead>
<tr>
<th>0</th>
<th>25</th>
<th>50</th>
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<tr>
<td>very poor</td>
<td>poor</td>
<td>fair</td>
<td>well</td>
<td>very well</td>
</tr>
</tbody>
</table>

2. During the past month, how would you describe the usual pain in your hip?
   a) Very Severe (2 pts)
   b) Severe (4 pts)
   c) Moderate (6 pts)
   d) Mild (8 pts)
   e) None (10 pts)

3. During the past month, how often have you had to take medication for your hip pain?
   a) Always (1 pt)
   b) Very often (2 pts)
   c) Fairly often (3 pts)
   d) Sometimes (4 pts)
   e) Never (5 pts)

4. During the past month, how often have you had severe pain in your hip?
   a) Every day (1 pt)
   b) Several days per week (2 pts)
   c) One day per week (3 pts)
   d) One day per month (4 pts)
   e) Never (5 pts)

5. How often have you had hip pain at rest, either sitting or lying down?
   a) Every day (1 pt)
   b) Several days per week (2 pts)
   c) One day per week (3 pts)
   d) One day per month (4 pts)
   e) Never (5 pts)

6. How much assistance do you need for walking?
   a) Unable to walk (1 pt)
   b) Walk only with someone's help (2 pts)
   c) Two crutches or walker (need every day) (3 pts)
   d) Two crutches or walker (need several days/week) (4 pts)
   e) Two crutches or walker (need once/week or less) (5 pts)
   f) Cane or one crutch (need every day) (6 pts)
   g) Cane or one crutch (need several days/week) (7 pts)
   h) Cane or one crutch (need once/week) (8 pts)
   i) Cane or one crutch (need once/month) (9 pts)
   j) No assistance (10 pts)

7. How far can you walk without resting because of your hip condition?

I.D._____________

Date____________
8. How much difficulty do you have going up or down one flight of stairs because of your hip condition?
   a) Unable (1 pt)
   b) Require someone’s assistance (2 pts)
   c) Require crutch or cane (3 pts)
   d) Require banister (4 pts)
   e) No difficulty (5 pts)

9. How much difficulty do you have putting on your shoes and socks because of your hip condition?
   a) Unable (1 pt)
   b) Require someone’s assistance (2 pts)
   c) Require long shoehorn and reacher (3 pts)
   d) Some difficulty but no devices required (4 pts)
   e) No difficulty (5 pts)

10. When you bathe – either a sponge bath or in a tub or shower – how much help do you need?
    a) No help at all (1 pt)
    b) Help with bathing one part of your body like back or leg (2 pts)
    c) Help with bathing more than one part of your body (3 pts)

11. Are you able to use public transportation?
    a) No, because of my hip condition (1 pt)
    b) No, for some other reason (2 pts)
    c) Yes, able to use public transportation (3 pts)

12. If you had the necessary transportation, could you go shopping for groceries or cloths?
    a) Without help (taking care of all shopping needs yourself) (3 pts)
    b) With some help (need someone to go with you to help on all shopping trips) (2 pts)
    c) Completely unable to do any shopping (1 pt)

13. If you had household tools and appliances (vacuum, mops, and so on), could you do your own housework?
    a) Without help (can clean floors, windows, refrigerator, and so on) (3 pts)
    b) With some help (can do light housework, but need help with some heavy work) (2 pts)
    c) Completely unable to do any housework (1 pt)

14. How well are you able to move around?
    a) Able to get in and out of bed or chairs without the help of another person (3 pts)
    b) Need the help of another person to get in and out of bed or chair (2 pts)
    c) Not able to get out of bed (1 pt)

This is the end of the Hip Rating Questionnaire. Thank you for your cooperation.

The maximum score is 100 points and the minimum is 16 points. The point values of the answers are not shown in the questionnaire that was administered to participants.
## APPENDIX C

### REFLECTIVE MARKER PLACEMENTS

<table>
<thead>
<tr>
<th>Markers</th>
<th>Anatomical landmarks</th>
<th>Description of placement locations</th>
</tr>
</thead>
<tbody>
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<td>A</td>
<td>Head</td>
<td>Anterior to the ear (approximate the center of mass of the head)</td>
</tr>
<tr>
<td>B</td>
<td>Shoulder joint</td>
<td>The acromion process</td>
</tr>
<tr>
<td>C</td>
<td>Elbow joint</td>
<td>The humeral lateral epicondyle</td>
</tr>
<tr>
<td>D</td>
<td>Hand</td>
<td>Head of 5&lt;sup&gt;th&lt;/sup&gt; metacarpal bone</td>
</tr>
<tr>
<td>E</td>
<td>ASIS</td>
<td>Anterior superior iliac spine</td>
</tr>
<tr>
<td>F</td>
<td>Hip joint</td>
<td>Greater trochanter of the femur</td>
</tr>
<tr>
<td>G</td>
<td>Mid PSIS</td>
<td>The midpoint between the right and left posterior superior iliac spine (PSIS)</td>
</tr>
<tr>
<td>H</td>
<td>Mid thigh</td>
<td>Anterior mid thigh</td>
</tr>
<tr>
<td>I</td>
<td>Lateral epicondyle</td>
<td>Center of lateral femoral epicondyle</td>
</tr>
<tr>
<td>J</td>
<td>Medial epicondyle</td>
<td>Center of medial femoral epicondyle</td>
</tr>
<tr>
<td>K</td>
<td>Lateral malleolus</td>
<td>Center of the lateral malleolus</td>
</tr>
<tr>
<td>L</td>
<td>Medial malleolus</td>
<td>Center of the medial malleolus</td>
</tr>
<tr>
<td>M</td>
<td>5&lt;sup&gt;th&lt;/sup&gt; metatarsal head</td>
<td>Lateral edge of head of the fifth metatarsal</td>
</tr>
<tr>
<td>N</td>
<td>Heel</td>
<td>Lateral calcaneus and 2 cm inferior and posterior to the center of lateral malleolus</td>
</tr>
</tbody>
</table>

![Image of reflective marker placements]

- **Dynamic markers**
- **Anatomical markers**
APPENDIX D

POSTTEST QUESTIONNAIRE

I.D. __________________
Date _________________

Warm-up:

1. Did you feel the warm-up period was adequate to effectively perform the tasks? Yes / No
   Comments _______________________________________________________________
   ______________________________________________________________________

2. Did you experience any discomfort during the warm-up period? Yes/No
   Comments _______________________________________________________________
   ______________________________________________________________________

Strength Test:

1. Did you understand the instructions given to you during the strength test? Yes / No
   Comments _______________________________________________________________
   ______________________________________________________________________

2. Did you feel comfortable with the atmosphere in the testing room? Yes / No
   Comments _______________________________________________________________
   ______________________________________________________________________

3. Did you feel comfortable with the apparatus for testing leg strength? Yes / No
   Comments _______________________________________________________________
   ______________________________________________________________________

4. Did you feel that straps were too tight on the thigh? Yes/No
   Comments _______________________________________________________________
   ______________________________________________________________________

5. In each of the muscle strength tests, did you push as hard as you could?
   *Right leg* Yes/No
   If No, explain _____________________________________________________________

   *Left leg* Yes/No
   If No, explain _____________________________________________________________

EMG:

Did you feel the electrodes wires hampered your movement for any of the tests? Yes / No
Comments _______________________________________________________________
   ______________________________________________________________________
# APPENDIX E

## CALCULATION OF TOTAL BODY CENTER OF MASS LOCATION

Anthropometric data and equations for calculation of total body center of mass (COM) location.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Proximal to Distal</th>
<th>COM/segment length from proximal end (%)</th>
<th>Segment Mass (%)</th>
<th>COM/segment length from proximal end (%)</th>
<th>Segment Mass (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Head</td>
<td>Vertex to Base of Neck</td>
<td>48.41</td>
<td>6.9</td>
<td>50.02</td>
<td>6.7</td>
</tr>
<tr>
<td>2. R upper arm</td>
<td>R Shoulder to R Elbow</td>
<td>57.54</td>
<td>2.7</td>
<td>57.72</td>
<td>2.6</td>
</tr>
<tr>
<td>3. L upper arm</td>
<td>L Shoulder to L Elbow</td>
<td>57.54</td>
<td>2.7</td>
<td>57.72</td>
<td>2.6</td>
</tr>
<tr>
<td>4. R Forearm &amp; Hand</td>
<td>R Elbow to 2nd R Metacarpal head</td>
<td>42.77</td>
<td>2.2</td>
<td>43.58</td>
<td>1.9</td>
</tr>
<tr>
<td>5. L Forearm &amp; Hand</td>
<td>L Elbow to 2nd L Metacarpal head</td>
<td>42.77</td>
<td>2.2</td>
<td>43.58</td>
<td>1.9</td>
</tr>
<tr>
<td>6. Trunk</td>
<td>Mid-Shoulder to Mid-ASIS</td>
<td>49.64</td>
<td>31.1</td>
<td>51.38</td>
<td>30.4</td>
</tr>
<tr>
<td>7. Pelvis</td>
<td>Mid-ASIS to Mid-Hip</td>
<td>49.64</td>
<td>12.4</td>
<td>51.38</td>
<td>12.2</td>
</tr>
<tr>
<td>8. R Thigh</td>
<td>R Hip to R Knee</td>
<td>36.12</td>
<td>14.2</td>
<td>40.95</td>
<td>14.8</td>
</tr>
<tr>
<td>9. L Thigh</td>
<td>L Hip to L Knee</td>
<td>36.12</td>
<td>14.2</td>
<td>40.95</td>
<td>14.8</td>
</tr>
<tr>
<td>10. R Shank</td>
<td>R Knee to R Ankle</td>
<td>44.16</td>
<td>4.3</td>
<td>44.59</td>
<td>4.8</td>
</tr>
<tr>
<td>11. L Shank</td>
<td>L Knee to L Ankle</td>
<td>44.16</td>
<td>4.3</td>
<td>44.59</td>
<td>4.8</td>
</tr>
<tr>
<td>12. R Foot</td>
<td>R Toe to R Heel</td>
<td>40.14</td>
<td>1.4</td>
<td>44.15</td>
<td>1.3</td>
</tr>
<tr>
<td>13. L Foot</td>
<td>L Toe to L Heel</td>
<td>40.14</td>
<td>1.4</td>
<td>44.15</td>
<td>1.3</td>
</tr>
</tbody>
</table>

**Note:** Anthropometric data are modified from Winter (1990) and de Leva (1996). All segments were used for calculation of the total body COM.
Calculation of segment COM coordinate \((x, y, z)\) (Winter, 1990).

The location \((x, y, z)\) of the segment COM were calculated as follows:

\[
x_i = x_p - (\text{length of the segment in x-direction} \times d_i)
\]
\[
= x_p - ((x_p - x_d) \times d_i)
\]
\[
y_i = y_p - (\text{length of the segment in y-direction} \times d_i)
\]
\[
= y_p - ((y_p - y_d) \times d_i)
\]
\[
z_i = z_p - (\text{length of the segment in z-direction} \times d_i)
\]
\[
= z_p - ((z_p - z_d) \times d_i)
\]

Where \(x_i\) is the x-coordinate of the segment \(i\) COM,
\(y_i\) is the y-coordinate of the segment \(i\) COM,
\(z_i\) is the z-coordinate of the segment \(i\) COM,
\(x_p\) is the x-coordinate of the segment proximal end,
\(x_d\) is the x-coordinate of the segment distal end,
\(y_p\) is the y-coordinate of the segment proximal end,
\(y_d\) is the y-coordinate of the segment distal end,
\(z_p\) is the z-coordinate of the segment proximal end,
\(z_d\) is the z-coordinate of the segment distal end, and
\(d_i\) is the COM/segment length of the segment.

Calculation of total body COM coordinate \((X, Y, Z)\) (Winter, 1990).

The products of the coordinate of the segment COM and the segment mass for each segment were used to determine the total body COM location. In algebraic terms:

\[
X_{\text{com}} = \frac{\sum_{i=1}^{n} m_i x_i}{M}
\]
\[
Y_{\text{com}} = \frac{\sum_{i=1}^{n} m_i y_i}{M}
\]
\[
Z_{\text{com}} = \frac{\sum_{i=1}^{n} m_i z_i}{M}
\]

where \(X_{\text{com}}\) is the x-coordinate of the total body COM,
\(Y_{\text{com}}\) is the y-coordinate of the total body COM,
\(Z_{\text{com}}\) is the z-coordinate of the total body COM,
\(n\) is the total number of segments,
\(m_i\) is the mass of the \(i^{th}\) segment, and
\(M\) is the total body mass.
APPENDIX F

CALCULATION OF PELVIC MOTION

Pelvic obliquity, tilt and rotation were calculated from the three-dimensional coordinates of the two ASIS and mid-PSIS markers. The room coordinate system is shown. X, Y, and Z axes represent the mediolateral, anteroposterior, and vertical directions, respectively.

Figure G.1

Directions of the pelvic obliquity, tilt, and rotation displacements relative to the room coordinate system (RCS).

Note: Pictures illustrating pelvic displacement are modified from Perry (1992).
1. Pelvic obliquity was calculated as the angle in the coronal plane formed between the line connecting the two ASIS markers (McGibbon & Krebs, 1999).

\[
\text{Pelvic obliquity (°) } = \frac{180}{\pi} \times \arctan \left( \frac{(Z_{\text{NON}} - Z_{\text{OP}})}{\sqrt{(Y_{\text{NON}} - Y_{\text{OP}})^2 + (X_{\text{NON}} - X_{\text{OP}})^2}} \right)
\]

where \((X_{\text{OP}}, Y_{\text{OP}}, Z_{\text{OP}})\) and \((X_{\text{NON}}, Y_{\text{NON}}, Z_{\text{NON}})\) are the 3-D coordinates of the operated- and nonoperated-side ASIS markers, respectively.

2. Pelvic tilt was calculated as the angle in the sagittal plane formed between the line connecting the midpoint of the two ASISs and the mid-PSIS markers (Sanders & Stavrakas, 1981).

\[
\text{Pelvic tilt (°) } = \frac{180}{\pi} \times \arcsin \left( \frac{(Z_{\text{PSIS}} - Z_{\text{MidASIS}})}{\sqrt{(X_{\text{PSIS}} - X_{\text{MidASIS}})^2 + (Y_{\text{PSIS}} - Y_{\text{MidASIS}})^2 + (Z_{\text{PSIS}} - Z_{\text{MidASIS}})^2}} \right)
\]

where \((X_{\text{PSIS}}, Y_{\text{PSIS}}, Z_{\text{PSIS}})\) and \((X_{\text{MidASIS}}, Y_{\text{MidASIS}}, Z_{\text{MidASIS}})\) are the 3-D coordinates of the mid-PSIS and the midpoint between the two ASIS markers, respectively.

3. Pelvic rotation was defined as the angle in the transverse plane formed between the line connecting the two ASIS markers projected onto the XY plane of the RCS (Taylor, Goldie, & Evans, 1999).
APPENDIX G

CALCULATION OF THE HIP JOINT COORDINATE SYSTEM

Pelvic and thigh segments are modeled as rigid bodies and the relative rotation is assumed to take place about a fixed point in the proximal segment (Kadaba et al., 1990). A right-handed room coordinate system (RCS) was defined with the X axis is perpendicular to the walking direction, the Y axis is parallel to the walking direction, and the Z axis is orthogonal to the X and Y axes. The joint coordinate system (JCS) for the hip (Figure G.1) was established based on the relative motions between the coordinate system of the pelvic and thigh segments (Davis et al., 1991; Grood & Suntay, 1983).

Figure H.1

Illustration of the Pelvic Coordinate System (PCS), and the Femoral Coordinate System (FCS). The corresponding unit vectors for the pelvis and femur are \(\langle i_p, j_p, k_p \rangle\) and \(\langle i_f, j_f, k_f \rangle\), respectively. The Joint Coordinate System (JCS) for the right hip joint is represented by the unit vector \(\langle e_1, e_2, e_3 \rangle\).
Description of the segment and joint coordinate systems

Pelvic Coordinate System (PCS) \(<i_p, j_p, k_p>\)

Let \(P_1\), \(P_2\), and \(P_3\) represent the pelvic segment markers of the right anterior superior iliac spine (ASIS), left ASIS, and mid posterior superior iliac spine, respectively.

\[
i'_{p} = \frac{P_{12}}{|P_{12}|} \\
j_{p} = \frac{P_{123}}{|P_{123}|} \\
k_{p} = i'_{p} \times j_{p} \\
i_{p} = j_{p} \times k_{p} \text{ (corrected for nonorthogonal marker placement)}
\]

where \(P_{12}\) is the line parallel to a line connecting the right and left ASISs, \((P_1 - P_2)\)

\(P_{123}\) is the line parallel to a line lying in the plane defined by the two ASISs and midpoint of the PSISs, \((0.5 * (P_1 - P_2) - P_3)\)

Femoral Coordinate System (PCS) \(<i_f, j_f, k_f>\)

Let \(F_1\), \(F_2\), and \(F_3\) represent the femoral segment markers of the greater trochanter, lateral femoral condyle, and medial femoral epicondyle, respectively.

\[
k_{f} = \frac{F_{12}}{|F_{12}|} \\
i'_{f} = \frac{F_{23}}{|F_{23}|} \\
j_{f} = k_{f} \times i'_{f} \\
i_{f} = j_{f} \times k_{f} \text{ (corrected for nonorthogonal marker placement)}
\]

where \(F_{12}\) is the line parallel to a line connecting the greater trochanter and lateral femoral condyle, \((F_1 - F_2)\)

\(F_{23}\) is the line parallel to a line joining lateral femoral condyle, and medial femoral epicondyle, \((F_2 - F_3)\).
Hip Joint Coordinate System

\[ e_3 = k_f \]
\[ e_1 = i_p \]
\[ e_2 = \text{floating axis, } e_3 \times e_1 \]

Euler angles for the right hip (°)

\[ \text{Hip flexion (+)/extension (-)} = \arccos (e_2 \cdot k_2) \times \frac{180}{\pi} \]
\[ \text{Hip internal (+)/external rotation (-)} = -\arcsin (e_2 \cdot i_f) \times \frac{180}{\pi} \]
\[ \text{Hip adduction (+)/abduction (-)} = \frac{\pi}{2} - (\arccos (i_p \cdot k_f) \times \frac{180}{\pi}) \]

Euler angles for the left hip (°)

\[ \text{Hip flexion (+)/extension (-)} = \arccos (e_2 \cdot k_2) \times \frac{180}{\pi} \]
\[ \text{Hip internal (+)/external rotation (-)} = \arcsin (e_2 \cdot i_f) \times \frac{180}{\pi} \]
\[ \text{Hip adduction (+)/abduction (-)} = -\frac{\pi}{2} + (\arccos (i_p \cdot k_f) \times \frac{180}{\pi}) \]