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# Functional movement assessment for individuals with knee osteoarthritis

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**Functional movement assessment for individuals with knee osteoarthritis**

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# ABSTRACT

The relationships between dynamic stability, lower extremity strength, and functional performance are not well-understood for individuals with knee pain due to osteoarthritis. The series of studies presented in this dissertation incorporate the assessment of functional movements for healthy older adults, individuals with symptoms of knee osteoarthritis, and individuals following total knee arthroplasty. The functional activities investigated in these populations included stair descent and sit-to-stand. This research was completed through a combination of kinematic techniques of motion analysis and kinetic assessment using multiple force platforms to evaluate joint moments, center of pressure parameters and weight-bearing asymmetry. The findings can be summarized as follows: 1) Kinetic and kinematic indicators of sit-to-stand movement may be standardized for healthy individuals of various ages without regard to initial positioning; 2) Individuals post-total knee arthroplasty compensate for residual physical deficits by utilizing upper extremity support and altered joint loading to perform sit-to-stand successfully; and 3) Altered joint loading and compensatory weight-bearing asymmetry allows individuals with mild to moderate knee pain to perform sit-to-stand and stair descent while maintaining dynamic stability. Older adults with symptoms of osteoarthritis appear to compensate for physical limitations of reduced joint mobility, strength deficits, and pain by adjusting joint loads and altering patterns for movement.

# CHAPTER 1: GENERAL INTRODUCTION

## I. Organization & Introduction

Dissertation Organization

This document is organized in chapter format with five overall chapters. This first chapter presents the general introduction and a review of literature regarding the relationships between loading symmetry, postural stability, and osteoarthritis during functional performance in sit-to-stand and stair descent. The second chapter is a paper on an investigation into the kinematic and kinetic movement determinants of the sit-to-stand movement for both healthy young and older adults. The third chapter reviews a research project involving healthy older adults and individuals who underwent total knee replacement. This particular project investigated weight-bearing asymmetry and joint moment differentials as participants completed sit-to-stand utilizing different initial upper and lower extremity positions. The final research chapter presents a culminating study involving functional movement assessment for individuals with knee osteoarthritis. This study investigated stair descent and sit-to-stand movements using symmetrical and asymmetrical foot placements as a means to evaluate the potential role of strength in postural stability during functional tasks. The terminal chapter provides an overall conclusion summarizing the scope of this research line and future directions.

Introduction

The field of physical therapy involves the application of therapeutic exercise, modalities, and manual techniques to restore and improve function based on kinesiology to identify impairments in body structure and function, activity limitations, and participation restrictions. Historically, the profession of physical therapy dates back to the reconstruction aides of the World War I era. Although physical therapy has expanded in terms of its content and application, the profession continues to lag in standardization of techniques and measures. In the ongoing health care debate, physical therapy has provided insufficient evidence for the justification of its medical necessity, fiscal prudence, and contributions to societal wellness. With this background, the purpose for pursuing this line of doctoral research was to add to the body of kinesiological evidence to support clinically-applicable intervention methods for orthopedic-based physical therapists. More specifically, this line of research focused on functional mobility for older adults with osteoarthritis.

As the U.S. population is aging, older adults will be a larger percent of the population in 2030 (Aging Statistics, DHHS). Osteoarthritis is a chronic degenerative disease that is diagnosed in over 21% of the population or 46.4 million individuals (CDC, 2006). Approximately 85% of individuals over the age of 75 years of age experience some symptoms of the disorder. It is expected that this disorder will continue to pose limitations for individuals and many agencies have focused efforts to alleviate its impact. For example, the anticipated goals for Healthy People 2020 include reducing the average joint pain and activity limitations for individuals with osteoarthritis (Healthy People 2020, DHHS).

Anecdotally, individuals with osteoarthritis and individuals after total knee arthroplasty reported difficulty returning to their previous level of function following surgical interventions or invasive pain-relieving procedures. It appeared these individuals were being referred to physical therapy yet were unable to tolerate general weight-bearing activity or quadriceps strengthening due to pain levels. In physical therapy, these individuals continued to exhibit strength deficits, relied on assistive devices for ambulation, and experienced instability and occasional falls.

Individuals who exhibit lower extremity weakness have a 4.4x increased risk of falling, while factors such as osteoarthritis, use of an assistive device, and a gait or balance deficit more than double (2.4-2.9 odds ratio) fall risk compared to healthy individuals (Brewer *et al.*, 2007). Despite statistics such as these, there is poor documentation of falls in the orthopedic patient population. Although there is limited evidence for a correlation between osteoarthritis and falls in an orthopedic population, osteoarthritis has been indicated as a risk factor for falls (Nevitt *et al.*, 1989).

The association of fall risk with sit-to-stand performance and functional independence triggered the need to explore the relationship between sit-to-stand movement compensation and weight-bearing stability for individuals with osteoarthritis. This research was expanded to stair descent, a task that tends to give individuals with osteoarthritis significant challenge and affects their participation in daily life. Current literature states that knee loads are significantly higher with activities such as sit-to-stand and stair climbing as compared to gait (Hughes *et al*., 1996; Stacoff *et al*., 2007).

As previous research has established that strengthening knee musculature is equivocal in its benefit on slowing the progression of knee osteoarthritis and on symptom reduction, it may be beneficial to determine if the musculature at other joints may play a role in reducing osteoarthritic symptoms. Some authors have reported the hip's role in controlling the medial load at the knee joint (Chang *et al*., 2005). Therefore, by determining threshold levels of hip strength that minimize knee joint loading, it may be possible to reduce painful osteoarthritic symptoms and/or improve function.

A primary step in this process is to determine if strength measures are associated with deficits in motor control or postural instability during functional tasks in individuals with osteoarthritis. Hence, the goal of this dissertation research was to evaluate the postural stability of individuals with osteoarthritis while performing functional tasks such as sit-to-stand and stair descent. The primary aim of the first project was to establish standardized measures for assessment of sit-to-stand for the subsequent projects. The principal purpose of the second project was to evaluate the sit-to-stand movement for weight-bearing asymmetry in individuals with total knee arthroplasty. The purpose of the final project was to correlate lower extremity strength with kinetic/kinematic parameters of loading asymmetry and postural stability to predict performance during functional activities.

Based on previous research that suggested osteoarthritic individuals exhibit asymmetry during movement and reduced step lengths during gait, they likely also display weak hip musculature that may contribute to instability during functional movements. Weight-bearing asymmetry indicated by a shift from midline during a movement may also provide information about how individuals preferentially load the body either to maximize strength capabilities or to avoid pain. If movement demands exceed strength capability or if weight-shifting approaches stability boundaries, individuals may be at increased risk for falls, muscle failure, exceeding their pain tolerance, or task failure. By identifying clinical measures (strength or balance) which affect postural stability, opportunities to intervene and enhance safety could be determined.

## II. General Hypotheses

Based on the review of literature, I hypothesized that individuals with symptomatic knee pain would demonstrate weak hip musculature and weight-bearing asymmetry during functional task performance. In addition, these individuals would exhibit postural instability in the frontal plane (as measured by kinetic movement parameters) during functional tasks of sit-to-stand and stair descent due to weak hip abductors. An alternative hypothesis was that individuals with symptomatic knee pain or weakness would alter their strategy for performance by slowing their movement to maintain postural stability.

## III. Literature Review

### Falls & Older Adults

A common health concern for older adults is the risk of falling. One-third of older adults fall each year (Tinetti *et al.*, 1988). Many risk factors have been identified for falls in the older adult population. Individuals who exhibit lower extremity weakness have a 4.4x increased risk of falling, while factors such as osteoarthritis, use of an assistive device, and a gait or balance deficit more than double (2.4-2.9 odds ratio) fall risk compared to healthy individuals (Brewer *et al.*, 2007). Older adults with hip weakness and poor balance who use multiple prescription medications are at significantly increased fall risk (Robbins *et al.*, 1989).

Falls are associated with nursing home admission due to individuals requiring assistance for mobility (Tinetti & Williams, 1997). Previous research indicated that older adults tend to fall during mobility tasks. Older adults report a majority of falls occur during ambulation; however, over 20% of falls may be associated with changes in position or stair ascension (Gehlsen & Whaley, 1990). For individuals with neurological deficits, approximately one-third (37.2%) of falls occurred during transfers (Nyberg & Gustafson, 1995). Many researchers have reported that individuals with neurological deficits demonstrate asymmetric weight-bearing and impaired balance during functional tasks. Older adults may also demonstrate asymmetry with functional tasks due to postural imbalances, muscle weakness, reduced joint flexibility, or pain.

For older adults, balance deficits and weakness can lead to difficulties with daily functional tasks, such as walking, stair negotiation, and sit-to-stand transfers (STS). For individuals with pathology, physical limitations may lead to functional deficits and overall decreased activity. Disuse caused by inactivity may lead to additional muscle weakness and joint stiffness, which may cause dependence in activities of daily living and disability. Individuals with greater than four physical limitations tend to be inactive, which appears to increase disability (Shih *et al.*, 2006). The risk of disability is therefore high for individuals with osteoarthritis, as 37% of those diagnosed are inactive.

## Osteoarthritis

Osteoarthritis is a chronic degenerative disease that is diagnosed in over 21% of the U.S. population or 46.4 million individuals (CDC, 2006). The reported percentage of those affected may be underestimated as many individuals are not yet diagnosed. As the U.S. population is aging, older adults will be a larger percent of the population in 2030 (Aging Statistics, DHHS). Of the current U.S. population, 18.9 million individuals (8.8%) are functionally limited in daily activities due to osteoarthritis (CDC, 2006). The treatment of osteoarthritis also costs the U.S. significantly in terms of health care dollars and lost work productivity, lending support to the fact that osteoarthritis is a significant health problem for the U.S. and will continue to escalate.

Approximately 85% of individuals over the age of 75 years of age experience some symptoms of osteoarthritis. Of individuals with the disorder, 40% experience significant difficulties with daily activities to the point of interfering with work-related or social roles (CDC, 2006). Individuals report challenges with mobility activities including distance ambulation, stair climbing, and sit-to-stand performance. Activities of daily living (ADLs) pose problems for individuals suffering with osteoarthritis as well, especially the tasks of bathing, grocery shopping, and housekeeping.

Disability and functional limitations are based on physical changes experienced due to the disease process of osteoarthritis. Osteoarthritis is a degenerative process that typically affects the synovial joints of the body. Current explanations for the onset of knee osteoarthritis entail altered joint loading due to traumatic events or the cumulative effects of aging and repetitive use which leads to a shift in cartilage loading (Andriacchi *et al.*, 2004). Dehydrated cartilage is ill-prepared for excessive loading and begins to undergo repetitive loads at a level that it is unable to withstand. This leads to proteoglycan loss, fibrillation from compressive forces and the initiation of the inflammatory process due to shearing. In healthy cartilage, the body responds through the limited ability for self-repair. However, once the degenerative cascade has begun, the process is not reversible. As the cartilage begins to thin, the degenerative process accelerates and the joint space narrows. As this process occurs, typically mal-alignment and an additional increase in joint loading occurs. Further cartilaginous degradation ensues to the point where there is bone involvement as well. Bone may produce osteophytes in response to the increased stress, which eventually may break off and produce further irritation within the joint capsule. Throughout this process, the body responds by shifting kinematic patterns to avoid pain and to minimize the loads being placed on the knee joint (especially at the weaker locations in the cartilage).

There are many factors that can affect the disease progression of knee osteoarthritis. The degeneration of cartilage and bone can reduce the knee joint space and cause a reduction in ligamentous support leading to joint laxity. The increased laxity can produce further cartilage wear from shear forces. Alignment issues resulting from either degenerative changes or movement compensations due to pain can also increase disease progression. For example, varus alignment produces a four-fold increase in the rate of medial osteoarthritis progression, while lateral osteoarthritis increases at a five-fold rate with a valgus alignment

(Sharma *et al.*, 2001). Lower extremity strength deficits also result from osteoarthritis. There appears to be conflicting evidence regarding the quadriceps' role in progression. Some evidence suggests stronger quadriceps musculature is associated with disease progression while other researchers suggest attempts to strengthen the quadriceps may slow progression (Mikesky *et al.*, 2006). Hip strength may also play a role in osteoarthritis progression as weak hip abductors may lead to medial knee osteoarthritis progression, whereas strong hip abductors may theoretically help protect against osteoarthritis progression (Chang *et al.*, 2005) by maintaining the lateral shift of body weight during gait and decreasing the pelvic drop in single-leg stance.

As healthy individuals age, they experience physical declines that may affect motor control and impact gait stability. Sarcopenia, reduced muscle coordination, and reduced muscular force production limit the amount of muscle strength older adults are capable of producing (Spirduso et al., 2005, p. 110-115). They tend to ambulate more slowly with a reduced stride length and a slightly wider step width (Spirduso et al, 2005, p. 150). These adjustments may either be a result of weak hip musculature, may lead to hip weakness, or may be a compensation for balance deficits. There is little evidence to clarify the nature of the relationship between gait parameters and hip strength measures.

Knee osteoarthritis may compound the motor control issues for older adults and lead to additional balance concerns. Changes in the ligaments, muscles, and joints due to aging and osteoarthritis may contribute to altered knee joint loading during functional tasks, which may accelerate the process of osteoarthritis or may trigger the onset of osteoarthritis in another joint. Due to the ongoing degenerative process, osteoarthritic joints are painful and demonstrate reduced range of motion, weakness, and proprioceptive deficits (Messier *et al.*, 2002; Messier *et al.*, 1992). These factors in addition to increased postural sway are associated with subjective and objective function (Hurley *et al.*, 1997a). Much research has focused on osteoarthritic gait, emphasizing the adductor moment at the knee during the gait cycle (Chang *et al.*, 2004; Sharma *et al.*, 1998). However, osteoarthritic individuals report greater difficulty with functional movements that require extremes of knee flexion such as sit-to-stand, stair ascent/descent, and entering/exiting a car (Marsh *et al.*, 2003). Harrison

(2004) demonstrated that self-reported function was associated with physical performance for individuals with osteoarthritis, yet pain appeared to explain the differences in self-reports of function for individuals with differing radiographic measures of osteoarthritis.

When osteoarthritis progresses to the point of extreme pain or severe functional restrictions, individuals may elect to undergo a total knee arthroplasty or replacement procedure. Over 418,000 total knee arthroplasty (TKA) surgeries were done in 2003 (Defrances *et al.*, 2005), and as the population ages this number is expected to increase. After TKA, individuals tend to lose approximately 4˚ of range of motion and tend to have limited knee flexion to 100-115º post-surgery (Chiu *et al.*, 2002). In general, individuals require 105º of knee flexion to perform daily mobility following TKA, and a minimum of 95º differentiates individuals into higher and lower functional levels due to compensatory ability (Miner *et al.*, 2003). However, the majority of individuals report satisfaction relative to pain and functional levels (Anderson *et al.*, 1996). In this study, 89% of patients reported satisfaction with their TKA while 91% reported better function.

The goal of a TKA is to reduce pain, which appears to be a major factor in satisfaction. Fortin and colleagues (1999) reported high functioning individuals demonstrated less pain at 6 months after surgery, while low functioning individuals improved to a lesser extent. Older adults in the study reported pain reduction that was equivalent to younger individuals, although their functional improvements were notably less. The time frame for evaluation can affect satisfaction ratings as individuals attain only 62% of their uninvolved limb strength by 16 months post-op (Rossi & Hasson, 2004). TKA patients may improve functional range of motion (ROM) for gait and stair performance up to seven years after surgery (van der Linden *et al.*, 2007). However, functional scores such as the Knee Society Score or Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) appear to plateau around two years. Functional performance is correlated with the uninvolved quadriceps strength, likely due to compensatory movement patterns (Mizner & Snyder-Mackler, 2005). The compensatory pattern of relying on the uninvolved extremity may prevent the opportunity for strengthening of the involved limb. For example, TKA patients exhibit decreased hip and knee moments on the involved side during sit-to-stand. The asymmetric movement pattern may be influenced by preoperative compensations for osteoarthritic pain and thus, the lower extremity with the TKA may remain weaker than the uninvolved side. This asymmetry may affect postural stability during mobility tasks.

Postural Stability & Osteoarthritis

There is limited research on postural stability relating to falls during functional movements in orthopedic populations. Much of the evidence in the area of falls during functional tasks (sit-to-stand) has been done in a neurological population (individuals after stroke, Parkinson's disease, etc.) There is limited evidence for a correlation between balance measures and falls in an orthopedic population, although osteoarthritis has been indicated as a risk factor for falls (Nevitt *et al.*, 1989).

The ligamentous, proprioceptive, and strength changes in older adults with lower extremity osteoarthritis appear to affect balance control and postural stability. Stability refers to the ability to control the body's center of mass (COM) relative to the base of support (BOS). Stability is typically assessed by determining the center of pressure (COP) movement relative to the BOS, both in terms of magnitude and velocity (Popovic *et al.*, 2000). As an individual's COP moves towards the edges of the BOS, heel/toe lifts and upper extremity and whole body adjustments allow for stability to be maintained. When unstable regions (approximately 30% of the outer BOS) are reached by the COP, individuals are required to step to maintain an upright position without falling. The lateral sway of the COP appears to be influenced by aging, as some older adults exhibit increased sway velocity and displacement (Raymakers *et al.*, 2005). Stability is affected by the BOS, as maximal foot width and foot angle (out-toeing) provide a wider stance that may also alter foot mobility and provide additional proprioceptive input through the ankle/hip coupling (Chiari *et al.*, 2002). Individuals with a threat to stability (such as reduced somatosensation or strength) may unconsciously use a wider stance (Stevermer *et al.*, 2005) or increased step width to improve control during functional tasks.

Research indicates that osteoarthritic individuals have reduced postural control as they demonstrate increased COP sway measures during static stance (Masui *et al.*, 2006; Wegener *et al.*, 1997) and reduced dynamic standing balance (Hinman *et al.*, 2002; Messier *et al.*, 2002). Multiple investigations have identified postural instability in individuals with knee osteoarthritis, both through quasi-static measures (swaymeter or force platform) and through clinical measures of balance such as functional reach or a running figure-8 test.

Although research on falls is limited, there are significant associations between stability/balance and lower extremity strength measures in individuals with knee osteoarthritis (Hasson *et al.*, 2001; Jadelis *et al.*, 2001). Researchers have also identified an association between balance and difficulty with functional movements (Marsh *et al.*, 2003). However, the few intervention studies that have utilized strengthening exercises to enhance balance in individuals with osteoarthritis have not identified a significant improvement in balance despite an increase in function. Improvements in strength have translated into improved performance in the task of STS (Alexander *et al.*, 1997; Hinman *et al.*, 2007). Unfortunately, much of the intervention work in the area of improving strength has focused on the musculature of the sagittal plane (knee flexion and extension, hip flexion and ankle dorsiflexion/plantar flexion), rather than the frontal plane (hip abduction/adduction, ankle inversion/eversion). As mediolateral sway is increased in individuals with osteoarthritis, perhaps the frontal plane musculature of the ankle and hip should be investigated. Overall, there is limited evidence on the strength of hip musculature for individuals with knee osteoarthritis, as previous research has also focused on the musculature controlling sagittal plane movement (quadriceps and hamstrings), even though improvements in lower extremity strength have translated into improved functional performance and pain reduction for individuals with osteoarthritis (Alexander *et al.*, 1997; Hinman *et al.*, 2007; Bennell *et al.*, 2010).

The relationships between lower extremity strength, dynamic stability, and functional performance have yet to be clarified in terms of fall risk. Individuals with knee osteoarthritis appear to experience declines in static and dynamic balance which potentially contribute to increased fall risk and decrements in functional task performance during sit-to-stand and stair negotiation.

Sit-to-stand

The STS movement is a fundamental activity of daily living that is required for upright posture, gait initiation, and personal-care tasks. The capability to complete sit-to-stand movements is associated with a decreased risk of disability (Guralnik *et al.*, 1995) and independent living rather than institutionalization (Guralnik *et al.*, 1994). STS requires greater knee strength, knee range of motion, and hip moment than stair ascension or gait (Rodosky *et al.*, 1989). The movement of STS is a physically-challenging task for healthy older adults because it requires 97% of available knee strength whereas standing or walking require a much smaller proportion (Hughes *et al.*, 1996). Many older adults find STS to be challenging without additional support (Papa & Cappozzo, 2000) due to lower extremity limitations associated with pathology and/or reduced dynamic stability. Older adults, individuals with osteoarthritis and those who have undergone total joint replacement may have additional difficulty with STS motions.

Musculoskeletal impairments and disability may result from decreases in muscle force and range of motion due to disuse when individuals experience functional decline (Beissner *et al.*, 2000). Reduced strength and increased pain in the lower extremity joints may impede the completion of sit-to-stand transfers in older adults with knee osteoarthritis. Difficulty with STS as manifested by slow sit-to-stand performance (>2.0 s) has been associated with greater than two times the risk for falls in older adults (Nevitt *et al*., 1989). For older adults with osteoarthritis, fall risk was increased 2.7 times.

The STS task is a transitional movement requiring an individual to move the COM from a position of stability (sitting) to a more unstable BOS on two feet (Janssen *et al.*, 2002). Although some authors simplify the movement of STS into two phases (Nuzik *et al.*, 1986), Kralj and colleagues (1990) reported four components of the sit-to-stand movement for young adults: initiation, seat-off, ascension, and stabilization. Other researchers described how these components are incorporated into different phases, including the flexion-momentum phase, momentum transfer, extension, and stabilization (Schenkman *et al.*, 1990). The flexion-momentum phase is from movement initiation to seat-off, followed by momentum transfer which is the phase from seat-off to maximal ankle dorsiflexion. The third phase of movement is the extension phase (ascension) which progresses from maximal ankle dorsiflexion until hip extension is completed. The stabilization phase includes the portion of movement from full hip extension until postural stability is reached.

Authors have varied their indicators of these phases of the STS movement which complicates the comparisons between published reports. The variation in phase identification may affect reports of STS times, delineation of STS strategies, moment values, and measures of stability. Many authors evaluate STS from seat-off (as it is identifiable from force platforms under the feet) to the end of ascension (detected through kinematic measures) rather than assessing the movement from initiation through the completion of stabilization. Sit-to-stand, progressing through all four phases, can be accomplished in various ways as individuals have demonstrated multiple strategies for completing STS successfully.

*STS - Strategies*

By evaluating young healthy women, Schenkman and colleagues (1990) discussed two movement strategies for performing the STS task. Healthy younger adults utilize a momentum-transfer strategy where upper body momentum shifts to the lower body. During this strategy the center of mass is located posterior to the center of pressure at seat-off, requiring the maintenance of dynamic stability for successful performance. The authors suggest that the momentum-transfer strategy requires adequate levels of strength and coordination to control center of mass momentum and produce full extension. In an alternate strategy, additional trunk flexion prior to seat-off allows the center of mass to be within the base of support at seat-off. This strategy may be referred to as a stabilization strategy. A third sit-to-stand strategy, dominant vertical rise, can also be an identified pattern of task performance which requires additional knee extensor moments to perform (Scarborough *et al.*, 2007).

The STS strategy used may be based on the strength level and motor control ability of the performer (Schenkman *et al.*, 1996a; Schenkman *et al.*, 1996b). There appears to be a relatively constant lower extremity support moment for sit-to-stand completion, requiring 4.4-5.0 Nm/kg as the knee, hip, and ankle move into extension (Shepherd & Gentile, 1994). The support moment may be distributed across the lower extremity joints in various ways, such that a stabilization strategy may reduce the knee moment requirements (Doorenbosch *et al.*, 1994). Scarborough et al. (2007) noted that peak trunk flexion may differentiate the strategies in evaluating performance, and acknowledged that physical limitations do not explain the strategy used.

Many researchers have suggested that STS is a programmed movement, evidenced by a muscular activation pattern for STS performance. The sequence of muscle activation in healthy adults is the tibialis anterior, quadriceps, hamstrings, and soleus muscles when a momentum-transfer strategy is used (Cheng *et al.*, 2004). The executional muscles – the lumbar paraspinals, quadriceps, and hamstring musculature – appear to be representative of a programmed movement by their consistent activation (Goulart & Valls-Sole, 1999). The tibialis anterior, soleus, abdominal musculature, sternocleidomastoid, and trapezius muscles are utilized for postural control during the movement. Other researchers have also indicated that trunk muscles were not used for initiating STS movement through trunk flexion (Millington *et al.*, 1992). The consistent sequence of lower extremity muscle activation during ascension likely indicates that the STS movement is programmed to be active when the COM is over the BOS, as in the momentum-transfer strategy.

Horizontal momentum appears to be most important for successful performance and control of STS. This factor is relatively unchanged under a variety of environmental conditions and different lower extremity placements (Hanke *et al.*, 1995; Pai & Rogers, 1991b; Reisman *et al.*, 2002). The nature of the STS task limits horizontal momentum more than vertical momentum irrespective of the speed of performance (Pai & Rogers, 1990), and the head-arms-trunk (HAT) segment appears to be essential in controlling horizontal momentum independent of the height of the individual (Pai & Rogers, 1990, 1991a). This allows the other segments flexibility in their movement according to the different strategies used, yet allows the control of STS to be governed by horizontal momentum as it is critical for balance maintenance.

Failure to achieve sufficient forward horizontal or vertical momentum can lead to a sit-back failure or require an individual to step to maintain postural stability (Riley *et al.*, 1997). In older adults, poor momentum coordination or insufficient extensor moments may lead to a STS failure. Insufficient vertical momentum due to poor momentum transfer or weak lower extremities may lead to a sit-to-stand failure, as seen in neurologically-involved individuals (Zablotny *et al.*, 2003). The knee joint contributes more to horizontal momentum than the ankle or hip joints, and the hip contributes more to vertical momentum than the other lower extremity joints (Yu *et al.*, 2000). Differences in joint contributions to momentum are based on initial posture which may alter the lower extremity demands and/or selected movement strategy (Mathiyakom *et al.*, 2005).

Many factors affect the strategy used for sit-to-stand performance, including chair height, foot placement, upper extremity utilization, speed of movement, and environmental conditions (Janssen et al., 2002). The presence of neurological involvement or orthopedic pathology may also alter the STS strategy.

*STS - Chair height*

Researchers demonstrated that interactions between physical ability and environmental factors such as seat height determine movement strategies and upper extremity compensation techniques, thereby affecting sit-to-stand effectiveness (Mazza *et al.*, 2004). Altering seated posture by changing foot placement and/or chair height redistributes moment-generation between the ankle, knee, and hip joints (Mathiyakom *et al.*, 2005). In a review article, Janssen et al. (2002) reported that increasing seat height may lead to reductions in joint moments at the knee by 60% and at the hip by 50%.

In the community setting, chair heights average approximately 16 inches with a range of 12-18 inches in medical and household environments (Weiner *et al.*, 1993). Comparing younger and older adults, Schenkman *et al.* (1996b) observed that sit-to-stand movement coordination was disrupted for older adults at lower chair heights, altering the order of joint extension between the knee, hip, and trunk. At low chair heights, older adults utilize more of their strength capability. Impaired older adults used a knee moment equal to 97% of their strength at the lowest height from which they were capable of standing while young adults only used 39% of available strength (Hughes *et al.*, 1996). Wheeler *et al.* (1985) also reported that older adults require a higher percentage of maximal muscle activity from a standard chair, despite similarities with younger adults in terms of hand placement, knee flexion, and anterior-posterior positioning in the initial posture.

Alexander *et al.* (1997) reported a better correlation for strength as a predictor for STS performance than age when evaluating individuals at four seat heights and two speeds, in addition to assessing the effects of a strengthening intervention. It has been suggested that older adults require a seat height equal to 120% of their lower leg length (Janssen *et al.*, 2002). However, they are also able to accommodate alternate chair heights by changing their foot position or generating additional forward momentum. Impaired older adults increase their horizontal momentum for momentum transfer and increase their rise time when using the stabilization strategy to perform STS from lower seat heights (Hughes & Schenkman, 1996). These adjustments produce an inefficient movement pattern and require greater effort as they are effectively trying to use competing strategies.

*STS - Foot Placement*

Altered seat height can require a change in foot placement for STS success. Changing initial posture by modifying foot placement can also benefit individuals during STS from a surface of equivalent height. There are three categories of symmetrical foot placement which have been investigated. A foot-neutral placement is represented by the ankle joint being placed directly inferior to the knee joint with the thigh horizontal, creating a 90˚ angle at the knee and 0˚ at the ankle joint. The foot-back placement increases the knee flexion angle, requires dorsiflexion at the ankle, and moves the BOS closer to the COM before movement.

The last type, the foot-forward placement, demonstrated by reduced knee flexion and plantarflexion at the ankle, requires more trunk flexion to bring the COM over the BOS.

Kawagoe et al. (2000) suggested that using a foot-back placement made sit-to-stand transfers more efficient by reducing the anterior distance that the center of mass has to travel and decreasing the joint moments at the hip. Khemlani et al. (1999) also maintained that those with lower extremity strength deficits should use a foot-back placement and found that changing foot position altered the timing between onset of muscle activity and joint movement. The location of muscle weakness or range of motion deficits may determine the preferential foot placement based on capability. For example, using a foot-forward placement has been shown to increase hip extension moments and was not recommended for individuals with hip osteoarthritis or with total hip replacements (Fleckenstein *et al.*, 1988). However, individuals who underwent TKA tend to have a less-flexed initial knee posture (a foot-neutral or foot-forward placement) and use higher hip moments to complete STS (Su *et al.*, 1998).

A staggered foot placement involves placing one lower extremity in a foot-back position while the opposite limb is in either foot-neutral or foot-forward. Evaluation of asymmetric foot placements during STS has not yet been published, but clinical experience demonstrates that many individuals with pathology utilize an asymmetric placement to compensate for weakness or pain. Healthy individuals may use a staggered placement when preceding sit-to-walk transitions or for anticipated directional changes upon standing. An initial posture with a staggered or asymmetric foot placement may be used to compensate for pain or weakness in individuals with pathology. Likewise, an asymmetric foot placement may be utilized for training purposes to enhance strength or stability. Brunt *et al.* (2002) suggested staggering foot placement during rehabilitation to strengthen the weakened leg for individuals with hemiplegia. As STS has been indicated as sufficient stimulus to maintain lower extremity strength (Kotake *et al.*, 1993), asymmetry in STS performance could be a potential detriment to strengthening in individuals with osteoarthritis or TKA. Foot placement can alter the movement strategy of the sit-to-stand transfer for ease of movement or to challenge an individual, but foot placement also affects stability in a standing posture (Mazza *et al.*, 2005) or during dynamic tasks once upright.

*STS - Upper extremity usage*

It is common for individuals to spontaneously use their upper extremities when performing sit-to-stand (Wheeler *et al*., 1985). These authors reported 13/20 participants (80% of older adults and 50% of young adults) used upper extremity assistance for forward propulsion, elevation or balance. The use of armrests may alter trunk rotation for older adults (Alexander *et al.*, 1991), although other joint angles are unchanged. The use of arm support does change the location of the COM which may affect the STS movement requirements. Burdett et al. (1985) demonstrated 50% reduction in hip moments through upper extremity usage, with reductions in knee moments as well.

*STS - Speed*

Altering speed of STS performance by slowing the movement may also reduce joint moments. Variation in speed (fast, natural or slow) of sit-to-stand performance can be indicative of limitations in performance. Older adults tend to perform STS more slowly, which allows them to focus on dynamic stability (Kerr *et al.*, 1997). Hanke *et al.* (1995) reported that slow STS performance is more variable in terms of movement sequencing which may affect the results of kinematic and kinetic measures, thus complicating comparisons between researchers. Therefore, these researchers recommended referencing events from seat-off rather than initiation of movement. These authors and Pai & Rogers (1990) demonstrated that horizontal momentum and displacement are relatively stable across all speeds of performance. Hip flexion, knee extension, and ankle dorsiflexion increased with increasing speed although initial postures were unchanged (Pai & Rogers, 1991b).

Other researchers have reported that even with speed changes and visual manipulation, older adults tend to demonstrate a posterior COM position likely due to a fear of falling forward (Mourey *et al.*, 2000). They also reported that older adults start to ascend from a more vertical position and use less horizontal velocity. Older adults use increased trunk flexion for increasing momentum in a stabilization strategy, yet switch to a momentum-transfer strategy with increasing speed (Papa & Cappozzo, 2000).

*STS - Strength*

Older adults create additional horizontal momentum prior to seat-off and move their COM forward for postural stability. Other researchers have also suggested that older adults use upper body momentum as a compensation technique for weakness (Schenkman *et al.*, 1996b). With strengthening, individuals may shift to a modified stabilization strategy emphasizing both stability and speed as they demonstrate increased COM velocity in horizontal and vertical directions (Schot *et al.*, 2003). However, older adults continue to exhibit a delay in the transition from horizontal to vertical momentum compared to younger adults. Older individuals typically use increased time for rising, but are able to rise from lower seat heights with increased strength (Alexander *et al.*, 1991). Predictors of successful sit-to-stand transfers include a lower fear of falling and increased leg strength (Alexander *et al.*, 1997; Hughes *et al.*, 1996).

The ability to rise from varied household surfaces correlated with knee extensor strength and task difficulty (Corrigan & Bohannon, 2001). Perceived task difficulty also correlated with the duration of the task (STS time). Scarborough and colleagues (1999) demonstrated a correlation between STS time and dynamic stability measures of center of mass momentum as well as an inverse correlation between STS time and gait speed, stride length and knee strength. These authors suggested that STS is an indirect measure of knee strength which also correlates with dynamic stability measures required for STS and gait.

Some authors reported that the ankle joint strength can compensate for postural instability in an individual with osteoarthritis and knee weakness (Messier et al., 2002). For individuals with motion limitations at the knee joint, such as individuals following total knee replacement, previous research indicated reliance on the hip joint to perform STS (Stevermer *et al.*, 2005). Both the TKA population and individuals with osteoarthritis have increased hip moments to successfully perform STS (Su *et al.*, 1998). Neither of these investigations included an evaluation of lower extremity strength to determine the percent of capacity used at each joint.

In a study comparing young and elderly women, Gross *et al*. (1998) suggested that deficits in hip muscle strength were the most critical strength-based factor to maintain sit-to-stand performance. However, these authors did note that knee strength may have a threshold effect on STS, and that there is a minimum amount of knee extension strength that is required for performance. In contrast, hip strength likely has a "gradient effect," such that increasing strength improves performance as measured by STS time and the amount of trunk flexion.

Other researchers have reported that function is correlated with the uninvolved quadriceps strength for individuals after TKA as they rely on asymmetric weight-bearing for performance (Mizner & Snyder-Mackler, 2005). However, the residual weakness in the involved extremity may be due to compensatory movement patterns created to avoid osteoarthritic pain prior to the joint replacement. Therefore, individuals with TKA exhibit decreased hip and knee moments on the involved side during STS due to pain levels associated with osteoarthritis. Differences in lower extremity motor function, such as following TKA, after neurological insult or amidst osteoarthritis, may produce compensatory weight-bearing asymmetries during functional tasks.

Older adults have been observed to have higher hip joint moment asymmetries during sit-to-stand than younger adults, although younger adults also exhibit asymmetry of joint moments (Lundin *et al.*, 1995). Many authors have evaluated individuals unilaterally and assumed symmetry of performance in STS. This assumption was supported by Wheeler et al. (1985) based on equivalent motor activity in dominant and non-dominant limbs. Millington et al. (1992) also determined mediolateral forces were relatively equal in a healthy population and therefore, little symmetry concern should be present. However, bilateral evaluation indicates asymmetry in performance occurs in young adults and to a greater extent in older adults (Lundin *et al.*, 1995; Lundin *et al.*, 1999). Some of the explanation for asymmetry may be due to the lateral shifting of the shoulders over the pelvis seen in healthy elderly (Baer & Ashburn, 1995), although these authors did not evaluate movement kinetics.

Eng & Chu (2002) reported weight-bearing asymmetries during STS were due to pain, reduced balance, muscle weakness and other neurological limitations including motor unit output and muscle recruitment. Cheng and colleagues have determined an association between postural sway in STS and fall risk (Cheng *et al.*, 1998; Cheng *et al.*, 2001) in a population following cerebrovascular accident (CVA). Weight-bearing asymmetry and increased sway were associated with decreased overall function in a similar neurological population (Lee *et al.*, 1997). Functional capacity (according to Functional Independence Measure) correlated with symmetry and dynamic balance as measured by postural sway. Other researchers have noted this relationship during functional tasks of rising and standing. After CVA, individuals demonstrate asymmetric weight-bearing potentially leading to reduced ROM and strength due to disuse (Engardt & Olsson, 1992). With symmetry training, these individuals demonstrate improved performance and overall function in activities of daily living (Engardt *et al.*, 1993).

Increased asymmetry, increased sway and decreased rate of force development during STS were associated with an increase in fall occurrence in a neurological population (Cheng *et al.*, 1998). These individuals exhibited significantly increased mediolateral sway compared to healthy controls, who demonstrated increased anteroposterior sway compared to mediolateral sway. It has yet to be determined if weight-bearing asymmetry is associated with falls in an orthopedic population. Based on evidence in neurologically impaired individuals, it is likely that individuals with osteoarthritis would demonstrate asymmetries in weight-bearing as they also have pain, weakness, decreased joint position sense, and reduced balance as measured by sway.

*STS - Pathology*

Individuals may use an alternate movement strategy for STS completion due to physical limitations (weakness, pain, range of motion deficits, etc.). Strength and pain limitations lead to changes in the sit-to-stand movement pattern for older adults. It has been observed that older adults use greater trunk flexion during STS, but lean forward more slowly than young adults when developing linear momentum (Kerr *et al*., 1997).

Older adults, especially female, demonstrate reduced momentum coordination for transition and require additional time to lean forward and rise to standing due to the increased demand on the lower extremities (Kerr *et al.*, 1997). For older adults with a history of falls, this elevation time is further increased due to limitations of strength, power, and balance (Cheng *et al*., 1998; Kerr *et al.*, 1997).

Similarly, Mourey and colleagues (2000) observed lower horizontal center of mass velocities and further found that older adults kept their weight balanced further back on their heels (increasing the risk of falling backward). However, Papa and Cappozzo (2000) reported that older adults demonstrated higher trunk flexion angles and velocities than young adults while generating momentum potentially due to using less knee flexion. As noted previously, impaired older adults use increased momentum and increase their rising time for STS from lower seat heights (Hughes & Schenkman, 1996). These compensations are components from two different movement strategies, the momentum-transfer strategy and the stabilization strategy. The conflicting evidence regarding momentum generation and trunk flexion may be related to the type of strategy implemented.

Previous investigators have established that osteoarthritic individuals require higher chair heights (115% leg length) for successful sit-to-stand than individuals without symptoms due to pain and strength deficits. Before and after knee replacement, individuals with osteoarthritis tend to require additional time to perform the sit-to-stand movement from all heights compared to healthy older adults (Su *et al.*, 1998). These individuals are limited due to range of motion restrictions and strength deficits.

For individuals with osteoarthritis, STS and stair negotiation become even more difficult tasks as they are already restricted in terms of strength and range of motion due to the disease process. Often the disease process limits their physical activity tolerance because of pain, so in terms of cardiorespiratory status, they are further restricted. Therefore, from a functional capacity perspective, these individuals will need to use a higher percentage of their capability to perform these daily tasks.

Many individuals with physical limitations also require multiple attempts to complete the task (Najafi *et al.*, 2002). However, research has demonstrated that active older adults maintain the ability to create sufficient power to move from sitting to standing with equivalent STS times and sway (Feland *et al.*, 2005). Asymmetrical trunk rotations were small in a healthy older adult population (Baer & Ashburn, 1995). However, these authors reported trunk lateral flexion is often seen in patients with neuromuscular impairments. Difficulties completing sit-to-stand transfers for individuals with Parkinson's disease have been associated with slower or reduced development of hip flexion moments (Mak *et al.*, 2003). The wide range of findings for individuals with pathology is attributed to different factors, suggesting that an individual's physical capability in terms of motor control may determine STS success. The evidence from various studies also suggests that older adults may rely on different STS strategies than young adults due to weakness and increased strength asymmetry in the lower extremity joints.

## Stair Negotiation

Individuals with osteoarthritis report difficulty with functional tasks such as walking and stair negotiation due to pain (Guccione *et al.*, 1994). In general, older adults rate stair negotiation in a list of five most difficult tasks. Interestingly, 21.6% of non-disabled elderly reported difficulty with STS which was significantly associated with stair negotiation difficulty (Verghese *et al.*, 2008). Stair descent difficulty was also correlated with reduced capability for activities of daily living, specifically bathing, dressing, and in-home ambulation. These authors demonstrated that self-reported difficulty with stair descent is correlated with increased fall risk and gait abnormalities. A large proportion of falls in older adults occur during stair negotiation with a greater risk of a fall during stair descent.

Many authors have indicated that older adults use different techniques to descend steps compared to young adults. Knee strength is critical for stair negotiation, and older adults need to use a greater proportion of their knee extensor strength to descend steps (Reeves *et al.*, 2008). Additionally, older adults demonstrate lower ankle moments than young adults, although they operate at a similar proportion of available strength. The lower ankle moments may be due to the fact that elderly tend to use shorter steps, place the foot closer to the step, and ensure more of the foot is supported both during stair climbing and descent (Lythgo *et al.*, 2007). These authors also indicated that older adults land on their forefoot to assist in force attenuation and balance control.

Stair descent also places large forces on the lower extremities, approaching two times body weight at the knee joint (Stacoff *et al.*, 2007). For individuals with osteoarthritis or total knee arthroplasty, the increased muscular demand may require compensations to avoid pain or collapse. Many individuals with reduced motion or low knee extensor strength use upper extremity support to aid stair negotiation (Tiedeman *et al.*, 2007). Occasionally, osteoarthritic adults perform stair descent facing backward in an attempt to reduce the demands on the knee joint (Beaulieu *et al.*, 2007). However, this technique increases the requirements of the hip flexors and the ankle joint for weight-bearing.

Osteoarthritic adults require additional time for tasks such as stair climbing, rising to stance, and ambulation. In general, older adults perform stair descent more slowly than young adults, while stair climbing is even more deliberate (Mian *et al.*, 2007a; Mian *et al.*, 2007b; Stacoff *et al.*, 2007). Individuals who report difficulty with stair negotiation, such as individuals with osteoarthritis, further slow their descent speed (Verghese *et al.*, 2008).

Asymmetric knee loading occurs in individuals with osteoarthritis during stair descent (Liikavaino *et al.*, 2007). Faster loading occurred on the involved knee as individuals spent more time in single limb support on the uninvolved side. It has been suggested that the rapid loading rate of force while performing stairs may contribute to osteoarthritis progression. Older adults as a group tend to spend additional time in single limb support during stair descent (Lythgo *et al.*, 2007; Mian *et al.*, 2007b). An increase in the proportion of time on one lower extremity may require additional hip strength in the frontal plane to counter the hip adduction moment created in order to avoid instability.

Older adults demonstrate increased frontal plane pelvic and hip motion (Mian *et al.*, 2007a; Mian *et al.*, 2007b) as well as increased asymmetry in stair descent (Liikavaino *et al.*, 2007; Stacoff *et al.*, 2007). Healthy elderly demonstrate faster sway than younger adults in both stair ascent and descent; however, the greatest sway magnitudes were demonstrated during descent (Lee & Chou, 2007). The increased sway was especially noted in the mediolateral direction according to inclination angles of the COM relative to the COP. The authors suggested that this finding may indicate a greater balance challenge in the frontal plane for older adults. Individuals following total knee replacement exhibited moderate weight-bearing asymmetry during STS (Stevermer *et al.*, 2006) and during stair descent (Stacoff *et al.*, 2007) compared to healthy elderly. These findings in concert with those from the previous authors suggest that mediolateral stability may play a greater role in functional performance for individuals with osteoarthritis.

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# CHAPTER 2: KINEMATIC AND KINETIC DETERMINANTS OF SIT-TO-STAND

A paper to be submitted to *Journal of Applied Biomechanics*

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## Abstract

The purpose of this investigation was to select appropriate kinematic indicators for sit-to-stand movement analysis in young and older adults using kinetic measures as reference standards. Three-dimensional kinematic data were collected in conjunction with kinetic data from ground and seated force platforms. Eighteen older and seventeen younger healthy adults performed sit-to-stand from various initial seated postures. The sit-to-stand movement was divided into a momentum phase (initiation to seat-off), an ascension phase (seat-off to vertical posture) and a stabilization phase (vertical posture to termination). Sit-to-stand trials were evaluated for threshold values as indicators of movement. Movement indicators were consistent for both age groups and across initial seated postures. Without a seated force platform, vertical ground reaction forces may be used to detect seat-off. The shoulder marker may be used as a kinematic indicator of initiation and vertical posture, while the hip marker may be the best clinical indicator of seat-off.

## Introduction

The sit-to-stand (STS) movement is a fundamental activity of daily living that is required for upright posture, gait initiation and personal-care tasks. For these reasons, it is a task frequently utilized in a rehabilitation environment for screening or assessment purposes. Many researchers have investigated links between physical capability and performance environment to evaluate movement compensations in STS. However, there is a range of descriptions in STS performance and the manner of assessment.

The STS task is a transitional movement which requires an individual to move the center of mass (COM) from a stable position in sitting to a less stable base of support in stance (Janssen *et al.*, 2002). Although some authors simplify STS into two parts (Nuzik *et al.*, 1986), others have reported four components of the sit-to-stand movement: initiation, seat-off, ascension, and stabilization (Kralj *et al.*, 1990). Researchers incorporated these components into four phases, including a flexion-momentum phase, momentum transfer, extension, and stabilization (Schenkman *et al.*, 1990). The flexion-momentum phase occurs from movement initiation to seat-off, followed by momentum transfer, which is the phase from seat-off to maximal ankle dorsiflexion. The third phase is extension, progressing from this point until hip extension is completed. Finally, stabilization proceeds from full hip extension until postural stability is achieved and the termination of movement is denoted. In addition to varying the phase descriptions of STS, authors have varied the indicators for the beginning and ending points of each phase of STS movement which complicates the comparisons between published reports (Schenkman *et al.,* 1996b; Mourey *et al*., 2000; Gross *et al*., 1998).

There has also been variation across and between settings (laboratory and clinic) regarding the ability to evaluate movement indicators for each phase of STS movement. Depending on instrumentation, variation exists in the availability of kinematic and kinetic measurements of STS performance (Table 2.1). This difference in equipment availability may affect the assessment of STS components. Many authors evaluate STS only from seat-off as it is identifiable from force platforms under the feet or seat switches. Alternately, others collect data only through the end of ascension as it can be detected through kinematic measures, rather than assessing STS through the completion of the stabilization phase. Sit-to-stand, progressing through all phases, can be accomplished in various ways as individuals have demonstrated multiple strategies for completing STS successfully (Schenkman *et al.*, 1996b; Doorenbosch *et al*., 1994). The knowledge about the type of strategy may provide key information which can impact functional capability (Scarborough *et al*., 1999).

For older adults, slow STS performance (>2 seconds) is associated with an increased risk for falls (Nevitt *et al*., 1989) and may affect physical functioning. However, Kralj and colleagues (1990) indicated a range of 2.58 to 5.12 seconds for STS in healthy older adults. These timing differences may be due to different movement initiation and termination indicators. The variation in phase identification and movement indicators may affect descriptions of strategies, performance duration and curve normalization for biomechanical analysis of STS.

The purpose of this investigation was to determine standardized measures for STS assessment. Specifically, the goal was to select appropriate kinematic indicators for STS movement analysis in young and older adults using kinetic measures as reference standards. Using force platforms provides high accuracy ground reaction forces (GRF) as a reference standard. Kinematic indicators may be used with motion analysis systems, standard video or potentially even visual assessment. With this knowledge, clinicians and researchers may evaluate STS evidence and collaborate for rehabilitation purposes. For this study, four points of interest were identified: movement initiation, seat-off, vertical posture, and stable posture (termination). These points divided the STS movement into three phases: a momentum phase (initiation to seat-off), an ascension phase (seat-off to vertical posture) and a stabilization phase (vertical posture to termination).

It was expected that movement of the shoulder marker would be the most accurate indicator of the initiation of STS as it would identify initial anterior or posterior movement. Based on previous work, we hypothesized that the vertical position of the hip marker would be the best indicator of seat-off as it is related to the goal of leaving the seated support. We anticipated that trunk angular velocity would be the most accurate indicator for achievement of vertical positioning and movement termination in conjunction with previous authors.

Previous research has focused on assumptions of bilaterally equivalent anthropometrics, joint timing, and weight-bearing during STS. Although older adults have demonstrated slower performance and more horizontal COM displacement during STS performance compared to younger adults (Kerr *et al.*, 1997), the movement sequencing did not appear to be different in healthy elderly. Therefore, it was expected that movement indicators will be similar in both young and older adults across different initial seated postures (foot placements).

**Table 2.1: Author Variation in Movement Indicators**

| **Initiation** | **Seat-Off** | **Termination**<br>(Vertical or Stable Posture) |
|---|---|---|
| Vertical GRF<br>(Shepherd & Gentile, 1994) | Vertical GRF<br>(Schenkman *et al.*, 1996b; Lundin *et al*., 1995) | COM Position<br>(Cameron *et al*., 2003; Schenkman *et al*., 1996a, 1996b) |
| COM Velocity<br>(Pai & Rogers, 1990) | Seated GRF<br>(Pai & Rogers, 1990) | COM Velocity<br>(Pai & Rogers, 1990) |
| Trunk Angle<br>(Cameron *et al.*, 2003) | Max Horizontal GRF<br>(Doorenbosch *et al.*, 1994; Gross *et al*., 1998) | Hip Angular Velocity<br>(Mizner & Snyder-Mackler, 2005) |
| Trunk Angular Velocity<br>(Mourey *et al*., 2000; Gross *et al*., 1998) | Seat Switch<br>(Shepherd & Gentile, 1994) | Trunk Angular Velocity<br>(Mourey *et al*., 2000; Gross *et al*., 1998) |
| Hip Flexion<br>(Mizner & Snyder-Mackler, 2005) | Hip Vertical Position<br>(van der Linden *et al*., 2007; Schenkman *et al*., 1996a; Mourey *et al.,* 2000) | Hip Vertical Position<br>(van der Linden *et al*., 2007) |
| Head Movement<br>(van der Linden *et al*., 2007) | | Hip Marker Horizontal Velocity<br>(Shepherd & Gentile, 1994) |
| Forward Lean<br>(Kerr *et al.*, 1997) | | Backward Lean<br>(Kerr *et al*., 1997) |
| Body Movement<br>(Nuzik *et al*.,1986) | | Pelvic Position<br>(Nuzik *et al*., 1986) |

**Methods**

Eighteen healthy older adults and seventeen healthy younger adults participated in this study. Participants were recruited via community flyers and through word-of-mouth. A verbal review of medical history and physical activity was completed with each potential participant. Exclusion criteria included any physical impairment which limited an individual's ability to perform a sitting to standing movement without upper extremity assistance. The Human Subjects Research Compliance Office at Iowa State University approved the experimental protocol, and the research participants provided informed consent prior to participating in the study.

During the experimental session, height and weight were assessed. Retro-reflective markers were applied to participants for tracking by an eight-camera video system (Peak Performance, Englewood, Colorado) to measure the movements of anatomical landmarks during the sit-to-stand transfers in each session. The spherical markers (diameter of 2 cm) were mounted on a vinyl base and attached to the skin or snug-fitting clothing with double-sided tape. A static trial was collected with the participant standing with markers placed bilaterally on the toes, midfeet, heels, lateral malleoli, medial malleoli, shins, lateral knee joint lines, medial knee joint lines, thighs, greater trochanters, posterior superior iliac spines, acromion processes, upper arms, lateral elbow joints, forearms and ulnar styloids. Additional markers were placed at the suprasternale and sacrum. This marker set divided the body into eleven segments: right/left feet, right/left calves, right/left thighs, right/left upper arms, right/left forearms and a head/trunk segment. The video data were collected at a sampling rate of 120 Hz and noise was reduced at a cutoff frequency of 6 Hz with a symmetric, fourth-order Butterworth filter. Heel markers and medial markers were dynamically recreated during the sit-to-stand trials using transforms derived from the static trial.

Participants began each STS trial in a seated posture on a bench-mounted force platform at a height of 48.5 cm to measure seated reaction forces. With their feet at a comfortable width on separate force platforms (ATMI, Newton, Massachusetts) to record ground reaction forces, participants performed sit-to-stand with four initial foot placements. The initial foot placements included: foot-neutral (symmetrically 90º of knee flexion), foot-back (symmetrically 100º of knee flexion), right-staggered and left-staggered. The staggered foot placements entailed a combination of the foot-back and foot-neutral placements. For example, in the right-staggered placement, the right knee was flexed to 100º while the left knee was flexed to 90º. The force platform data were collected at 120 Hz and synchronized with the video data through the Peak Motus software.

Three repetitions of each foot placement were performed during an experimental session for a total of twelve trials for each individual during the testing session. The order of the trials within each session was balanced across subjects in an attempt to reduce the influence of learning and fatigue. An interval of at least one minute was allocated between trials to minimize fatigue and allow for repositioning. Multi-colored athletic tape was used to mark the three lower extremity placements and the location/depth of the participant's buttocks on the bench during initial positioning in order to ensure consistency during the experimental trials. Participants were verbally instructed to position their feet according to the color of athletic tape (blue, red or yellow) for each trial. A two-stage verbal command ("Ready, Go") was used to cue the participant to initiate the STS movement. For all trials, participants' arms remained crossed over their torso throughout the duration of the STS movement and they remained standing in their final position for the completion of the ten second data collection.

Centers of pressure (COP) were determined from ground reaction forces and moments measured by the force platforms. Calculations and filtering were programmed using MATLAB software (Natick, Massachusetts). Individual trials were evaluated using MATLAB for thresholds in kinematic variables and kinetic variables as indicators for STS movement initiation, seat-off, verticality and movement termination.

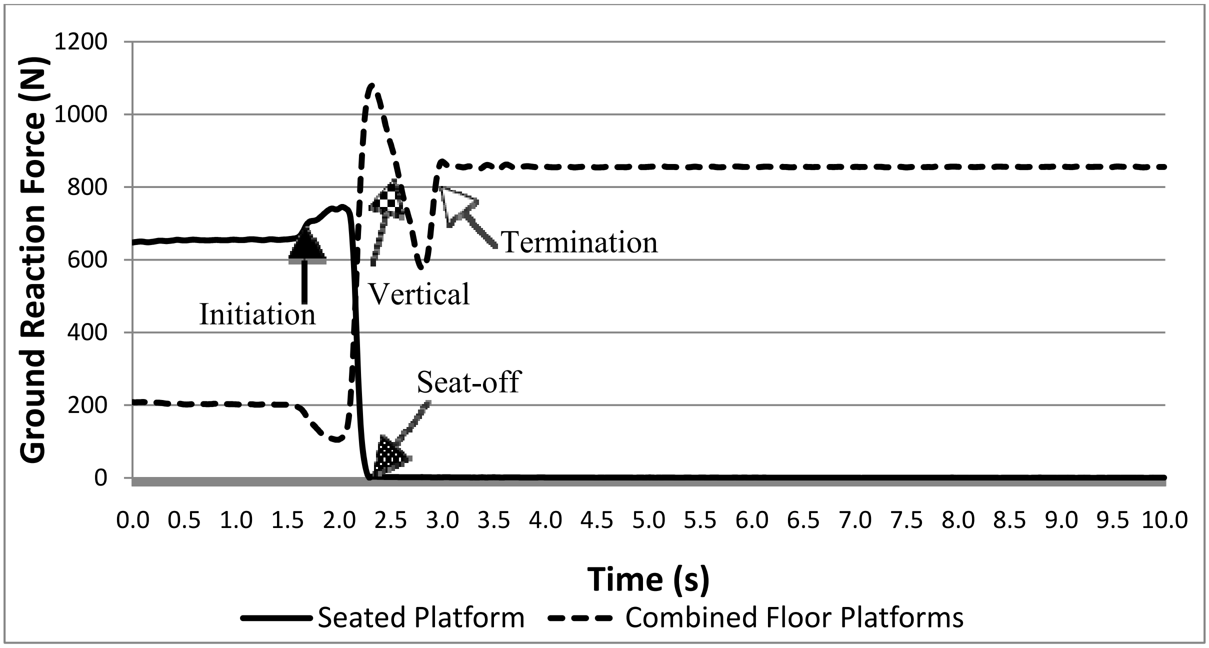


**Figure 2.1: Ground Reaction Forces during Sit-to-Stand from Seated and Floor Force Platforms**

Points of interest were initiation, seat-off, vertical posture, and termination. Trials were evaluated for 5% initial and 7.5% terminal thresholds in kinematic and kinetic variables as indicators for movement (Pai & Rogers, 1990). Thresholds were based on the maximum-to-minimum range of ground reaction force values. Reference values for initiation and seat-off were determined from seated forces, while vertical posture and termination were based on vertical GRF measures (Figure 2.1). Kinematic movement indicators included marker positions, marker velocities, joint angles, joint angular velocities, and virtual COM based on previous work (Table 2.1). Kinetic movement indicators included anterior-posterior GRF (AGRF), vertical GRF (VGRF), seated GRF (CGRF) and COP measures. Potential movement indicators were compared to reference values through calculations of time intervals of interest: momentum time (MT), ascension time (AT) and total sit-to-stand time (STS Time). Total sit-to-stand time was compared against different movement indicators of initiation and termination.

Multivariate analysis was conducted to screen for potential differences between groups (older/younger) and between foot positions, with follow-up univariate analyses of variance (ANOVA). A Bonferroni correction of the alpha level was made to account for multiple comparisons of timing indicators (0.05/12 = 0.004). SPSS software (SPSS, Inc., Chicago, Illinois) was used for statistical analysis. Non-parametric correlations were performed on each set of timing indicators to compare relationships with the reference times. Root mean square error (RMSE) was calculated for differences between times calculated with different indicators within each set of time points.

**Results**

Participant demographic information is presented in Table 2.2. The age groups were different based on t-test with $p<0.05$. There was also a significant difference in mass between the two groups ($p<0.05$). Based on MANOVA screening, no significant group or foot position differences were identified for STS Initiation, MT or AT measures. One potential group difference was identified in STS Termination.

**Table 2.2: Participant Demographics** (* indicates statistical significance at $p<0.05$)

| **Category** | **Older** (n=18) | **Younger** (n=17) | **p-value** |
|---|---|---|---|
| Age (years) | 68.2 ± 7.2 | 33.1 ± 4.0 | <0.001* |
| Height (meters) | 1.7 ± 0.1 | 1.7 ± 0.1 | NS |
| Mass (kg) | 70.2 ± 11.4 | 80.8 ± 19.7 | <0.05* |
| Body Mass Index (kg/m$^2$) | 24.6 ± 3.4 | 26.8 ± 5.0 | NS |
| Seat Depth (as % thigh length) | 76.3 ± 11.8 | 68.5 ± 16.6 | NS |

<u>STS Initiation</u>:

STS Time was calculated using the CGRF to detect initiation and the VGRF as the indicator of termination. Other STS times were compared to this measure using alternate indicators for initiation (Table 2.3). Across foot positions, the VGRF measurement had the lowest RMSE of kinematic and kinetic indicators for both groups. In terms of kinematic indicators, the shoulder indicator exhibited a low average RMSE across foot positions for both groups. For the young adult group, the trunk angular velocity time and hip angle time also exhibited low RMSE.

**Table 2.3: STS Initiation Indicators – RMSE averaged across foot placements** (bold font highlights the lowest RMSE of the kinetic indicators and the three lowest values for the kinematic indicators)

| **Indicators** | **Older Average** (s) | **Younger Average** (s) |
|---|---|---|
| **Kinetic Indicators** | | |
| VGRF | **0.089** | **0.101** |
| AGRF | 0.161 | 0.226 |
| COP Anteroposterior | 0.192 | 0.170 |
| COP Mediolateral | 0.195 | 0.271 |
| **Kinematic Indicators** | | |
| Trunk Angular Velocity | **0.220** | **0.151** |
| Shoulder Marker | **0.195** | **0.180** |
| Hip Angle | **0.227** | **0.187** |
| Body COM | 0.231 | 0.266 |
| Hip Marker Horizontal | 0.411 | 0.388 |
| Hip Velocity Vertical | 0.388 | 0.416 |
| Hip Marker Vertical | 0.416 | 0.422 |
| Hip Velocity Horizontal | 0.431 | 0.440 |

<u>Momentum Time:</u>

The reference for momentum time was determined from the initiation of movement using the CGRF measurement until the absence of force on the seated force platform, also from the CGRF measure. The other momentum times were calculated using alternate indicators for seat-off and compared to this reference measure (Table 2.4). The VGRF and AGRF measures had the lowest RMSE for kinetic indicators. The lowest RMSE for kinematic measures were trunk angular velocity, hip marker horizontal position and hip angle for both age groups.

**Table 2.4 STS Momentum Indicators – RMSE Averaged across foot placements** (bold font highlights the lowest RMSE of the kinetic indicators and the three lowest values for the kinematic indicators)

| **Indicators** | **Older Average** (s) | **Younger Average** (s) |
|---|---|---|
| **Kinetic Indicators** | | |
| VGRF | **0.063** | **0.068** |
| AGRF | **0.065** | **0.068** |
| COP Anteroposterior | 0.503 | 0.466 |
| COP Mediolateral | 0.505 | 0.580 |
| **Kinematic Indicators** | | |
| Trunk Angular Velocity | **0.174** | **0.051** |
| Hip Marker Horizontal | **0.263** | **0.175** |
| Hip Angle | **0.171** | **0.192** |
| Hip Velocity Vertical | 0.315 | 0.203 |
| Hip Velocity Horizontal | 0.275 | 0.208 |
| Hip Marker Vertical | 0.295 | 0.256 |
| Body COM | 0.341 | 0.274 |
| Shoulder Marker | 0.352 | 0.314 |

Ascension Time:

Ascension time was determined based on VGRF measures of achieving full weight-bearing and assumed upright posture. This time was calculated using the initiation of STS from the CGRF measurement through VGRF equivalent to body weight (after an initial overshoot due to the acceleration peak). Other times were compared to this reference time (Table 2.5).

Vertical indicators were highly correlated with the reference measure (Spearman correlation range = 0.260-0.853, $p<0.01$) except anteroposterior COP. Anterior GRF had a lower RMSE compared to the COP parameters as alternate kinetic measures. Many kinematic measures had lower RMSE compared to the kinetic measures. For both younger and older adults, the hip marker vertical position and hip angle indicators had low RMSE. The body COM indicator for younger adults had similar RMSE levels.

**Table 2.5 STS Ascension Indicators – RMSE Averaged across foot placements** (bold font highlights the lowest RMSE of the kinetic indicators and the three lowest values for the kinematic indicators)

| **Indicators** | **Older Average** (s) | **Younger Average** (s) |
|---|---|---|
| **Kinetic Indicators** | | |
| AGRF | **0.213** | **0.264** |
| COP Anteroposterior | 0.714 | 0.661 |
| COP Mediolateral | 0.711 | 0.764 |
| **Kinematic Indicators** | | |
| Hip Marker Vertical | **0.117** | **0.086** |
| Hip Angle | **0.109** | **0.109** |
| Shoulder Marker | **0.171** | 0.136 |
| Body COM | 0.194 | **0.102** |
| Hip Marker Horizontal | 0.197 | 0.198 |
| Trunk Angular Velocity | 0.317 | 0.203 |
| Hip Velocity Vertical | 0.470 | 0.365 |
| Hip Velocity Horizontal | 0.461 | 0.411 |

<u>STS Termination</u>:

As noted previously, STS Time was calculated using the CGRF to detect initiation and the VGRF as the indicator of termination. This reference measure was compared against the other calculated STS times using alternate indicators for the termination of STS (Table 2.6). All termination indicators were correlated with the reference measure of STS Time (Spearman correlation range = 0.364-0.950, $p<0.01$). During MANOVA screening, a potential group difference was noted ($p=0.001$) in STS Time. Follow-up univariate ANOVA identified the potential group difference in trunk angular velocity measures, however a Bonferroni correction due to multiple comparisons did not indicate a statistically significant difference between groups ($p=0.024$).

The AGRF measure had the lowest RMSE of all indicators for both older and younger adults. For kinematic measures, the lowest average RMSE were identified for the indicators of the hip angle, hip horizontal position, and trunk angular velocity for both groups.

**Table 2.6 STS Termination Indicators – RMSE Averaged across foot placements** (bold font highlights the lowest RMSE of the kinetic indicators and the three lowest values for the kinematic indicators)

| **Indicators** | **Older Average** (s) | **Younger Average** (s) |
|---|---|---|
| **Kinetic Indicators** | | |
| AGRF | **0.324** | **0.314** |
| COP Anteroposterior | 1.808 | 1.740 |
| COP Mediolateral | 1.705 | 1.205 |
| **Kinematic Indicators** | | |
| Hip Angle | **0.140** | **0.091** |
| Hip Marker Horizontal | **0.143** | **0.160** |
| Trunk Angular Velocity | **0.190** | **0.158** |
| Shoulder Marker | 0.191 | 0.220 |
| Body COM | 0.250 | 0.272 |
| Hip Marker Vertical | 0.286 | 0.305 |
| Hip Velocity Vertical | 0.790 | 0.487 |
| Hip Velocity Horizontal | 0.943 | 0.683 |

**Discussion**

The outcome of this project suggests that laboratory equipment incorporating force platform and videography provides a thorough three-dimensional assessment of human motion. However, in many clinical settings the elaborate set-ups and equipment are not available to analyze movement to the extent possible in a laboratory setting. The purpose of this project was to provide recommendations for alternative methods of STS assessment in the event laboratory equipment was unavailable for utilization.

Based on the results, it appears that the movement indicators for STS are consistent for healthy young and older adults through all phases of sit-to-stand (momentum, ascension and stabilization). Likewise, evidence suggests there does not appear to be an effect of foot placement in detecting the indicators of the three phases of the sit-to-stand movement. This would suggest that kinetic and kinematic indicators could be consistently utilized in a laboratory environment for the assessment of STS without specific requirements for initial seated posture to detect STS phases.

To detect the initiation of STS or seat-off without kinetic input from a force platform under the seat, the current data suggests an alternate kinetic source to be the most accurate method of detection. Therefore, the low RMSE of the vertical ground reaction force measures indicates that the use of an in-ground force platform is the preferred surrogate indicator for initiation and seat-off in the absence of seated reaction force measures for a healthy population.

In lieu of force platform availability, clinicians should observe shoulder movement to detect the initiation of STS movement. A limitation of this project was not placing a marker on the head to determine if head movement preceded shoulder movement for initiation. Without kinetic measures for seat-off detection, the recommended kinematic surrogate was trunk angular velocity for both groups. However, this measure may be impractical for clinicians to detect without utilizing motion analysis software. Thus, it may not be a clinically-relevant measure for determining seat-off. However, the determination of seat-off by a clinician may be pertinent for assessment of lower extremity strength, STS strategy or weight shifting capability (Alexander *et al.*, 1997; Scarborough *et al.*, 1999). Although the observation of seat-off timing relative to the initiation and termination points of STS is important in determining STS strategy, the specific determination of momentum time may be more relevant to laboratory-based activities.

Previous researchers have measured STS time by evaluating the ascension phase of STS, or the point at which individuals reach a vertical posture. Therefore, the point of verticality was determined by when individuals returned to body weight after the acceleration peak as measured via the VGRF. If the kinetic measure for this point of determination is not available, the vertical position of the shoulder marker appears to be the best surrogate measure. The shoulder marker may be the recommended clinical substitute, as it could be consistent with the recommendation for the initiation of STS movement, effectively determining one indicator for clinicians to observe for timing purposes. The other indicators with similar RMSE measures for vertical posture, the whole-body COM and horizontal hip marker position did not demonstrate the same level of consistency across age groups. In addition, whole body COM would require visual estimation or laboratory software for use.

In terms of detecting termination without kinetic input from in-ground force platforms, the current results suggest the best kinematic alternatives would be to utilize hip angle, trunk angular velocity or hip horizontal position indicators. The shoulder marker indicator ranked fourth compared to these other indicators, and is involved in the determination of two of these indicators. Therefore, it would appear that the shoulder indicator may be an appropriate overall clinical indicator to use when videography is not available.

Many other researchers have utilized a component of the hip marker to determine termination of STS (see Table 2.1). However, the VGRF was selected as the reference measure for termination because of occasional occlusion trouble with the hip marker during motion analysis, while the VGRF readings were consistently available. Additionally, the VGRF incorporated the stabilization phase of STS. The higher RMSE measures from the termination indicators may suggest high variability in the reference measure for termination due to oscillations during stabilization of STS. Further research should critically distinguish between the ascension and stabilization phases using kinetic measures. As 5% thresholds were used for initiation and 7.5% for termination, additional research comparing hand analysis of sit-to-stand assessment with automated methods of detection may determine different optimal thresholds for movement indicators.

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# CHAPTER 3: INVESTIGATION OF SIT-TO-STAND AFTER TOTAL KNEE ARTHROPLASTY

A paper to be submitted to *Clinical Biomechanics*

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## Abstract

Background: Mobility restrictions may persist following total knee arthroplasty due to range of motion and strength deficits. Weight-bearing asymmetries following surgery may affect joint loading and postural stability during sit-to-stand. The purpose of this investigation was to determine upper and lower extremity initial positioning which would enhance stability and performance during sit-to-stand following total knee arthroplasty.

Methods: Individuals with total knee replacements (n=7) and healthy adults (n=7) performed sit-to-stand with three different initial foot placements (foot-neutral, foot-back and dominant-staggered) combined with three different upper extremity conditions (no hand support, hands-separated and hands-on-bench). Motion analysis and kinetic assessments were completed and inverse dynamics was utilized to determine lower extremity joint angles, joint moments, weight-bearing asymmetry and upper extremity support. A 2x3x3 ANOVA was conducted to test for effects of group, foot placement postures and upper extremity conditions.

Findings: Individuals with total knee arthroplasty used more hip flexion range of motion and higher ankle dorsiflexion moments bilaterally to perform sit-to-stand than healthy controls ($P<0.001$). They also used more upper extremity assistance during sit-to-stand when permitted than healthy individuals ($P<0.001$). Both groups used less upper extremity force in the hands-separated condition than in the hands-on-bench condition ($P<0.001$).

Interpretation: Utilizing adjacent joint movement and assisting with the upper extremities are modifications individuals use following total knee arthroplasty to successfully perform sit-to-stand. This information may be beneficial to clinicians for determining appropriate therapy interventions, to improve safety during transfers and to maximize client independence with mobility training.

Keywords: Sit-to-stand, Knee Replacement, Kinetics

**Introduction**

Of the U.S. population, 18.9 million individuals are limited in daily activities due to osteoarthritis (CDC, 2006). When osteoarthritis progresses to the point of extreme pain or severe functional restrictions, individuals may elect a total knee arthroplasty procedure. Over 418,000 total knee arthroplasty (TKA) surgeries were performed in 2003 (Defrances *et al.*, 2005) and as the population ages, this number is expected to increase. Most individuals report post-surgical satisfaction relative to pain and functional levels following TKA (Anderson *et al.*, 1996). However, range of motion (Chiu *et al.,* 2002*)* and strength deficits (Rossi & Hasson, 2004) persist following surgery, which may limit performance of functional tasks such as sit-to-stand (STS).

The STS movement is fundamental for daily living, as it is required for upright posture, gait initiation and personal-care tasks. STS is a physically-challenging task requiring 97% of available knee strength (Hughes *et al.*, 1996) and more knee motion and hip moment than stair climbing or gait (Rodosky *et al.*, 1989). Because of these higher demands, older adults find STS transfers difficult due to limitations associated with pathology and/or dynamic stability (Papa & Cappozzo, 2000), thus older adults modify factors which may affect STS performance.

Janssen (2002) provided a review of STS research, which identified factors that influence STS performance including seat height, armrest usage, and foot placement. Higher seat heights and arm rest usage reduced hip and knee joint moments. Previous investigators established that individuals with osteoarthritis required higher seat heights for successful STS (Mazza *et al.*, 2004). Adjusting the position of the feet relative to the knee joint line also appeared to alter hip and knee moments (Janssen, 2002). For example, placing the feet posterior to the knee joint line (foot-back placement) required greater knee joint moments but lowered hip moments, while placing the feet directly under the joint line lowered knee moments and required higher hip moments. Researchers suggested individuals with lower extremity strength deficits should use a foot-back placement for STS (Khemlani *et al.*, 1999). However, as individuals after TKA may have difficulty adopting a foot-back placement for

STS due to limited knee flexion capacity, a foot-neutral placement (feet directly under the knee joint line) may be a preferred foot placement for TKA individuals for STS performance.

Besides altering foot placement, older adults adjust weight-bearing to accommodate strength capability. Older adults were observed to have higher hip joint moment asymmetries during STS than younger adults (Lundin *et al.*, 1995). Functional performance has been correlated with the uninvolved side knee extensor strength following TKA as individuals rely on asymmetric weight-bearing for performance (Mizner & Snyder-Mackler, 2005). Differences in lower extremity motor function following TKA or amidst osteoarthritis may produce compensatory weight-bearing asymmetries during functional tasks.

Weight-bearing asymmetries during STS were reportedly due to pain, reduced balance, muscle weakness, and neuromuscular deficits in a neurologically-affected population (Eng & Chu, 2002). Increased weight-bearing asymmetry and mediolateral sway during STS were associated with an increase in fall occurrence in a similar population (Cheng *et al.*, 1998). This evidence suggests that individuals with osteoarthritis may demonstrate weight-bearing asymmetries due to pain, weakness, and reduced static balance that may increase fall risk. Weight-bearing asymmetry for individuals following TKA may affect postural stability during mobility tasks such as STS.

The purpose of this investigation was to determine the upper and lower extremity initial positioning that will enhance postural stability and performance during STS following TKA. Based on previous evidence in osteoarthritic individuals and individuals post-TKA, we hypothesized that individuals following TKA would perform STS more slowly than healthy controls. It was expected that individuals following TKA would use more upper extremity assistance and higher hip extension moments to facilitate STS compared to healthy controls. Finally, it was expected that TKA individuals would avoid loading their TKA side (referred herein as non-dominant side) and demonstrate more weight-bearing asymmetry and higher instability measures.

**Methods**

Study participants were recruited via community flyers and through word-of-mouth. A verbal review of medical history, joint pain and physical activity was completed with each potential participant to screen for exclusion criteria. Exclusion criteria included any physical impairment that limited an individual's ability to perform a sitting to standing movement without upper extremity assistance. Potential participants were excluded due to impairments including: hemiparesis, knee pain or surgical complication, knee joint edema, and low back pain. Seven individuals with total knee replacements and seven healthy adults participated in this study, each attending two experimental sessions an average of 23 days apart. The Human Subjects Research Compliance Office at Iowa State University approved the experimental protocol, and the research participants provided informed consent prior to participating in the study.

During the initial experimental session, height, weight, and knee joint range of motion were measured. Body mass index calculations were determined from mass and height for each participant. The participants with total knee replacements were 1.7 (0.6) years post-surgery. Participant demographics for the TKA group and the healthy control group are listed in Table 3.1.

Reflective markers were applied to participants for tracking by an eight-camera video system (Peak Performance, Englewood, Colorado) to measure the movements of anatomical landmarks during the STS transfers in each session. The spherical markers (diameter of 2 cm) were mounted on a vinyl base and attached to the skin or snug-fitting clothing with double-sided tape. A static trial was collected with the older adult standing quietly with markers placed bilaterally on the toes, heels, lateral malleoli, medial malleoli, lateral knee joint lines, medial knee joint lines, greater trochanters, acromion processes, lateral elbow joints, and ulnar styloids. A single marker was placed at the suprasternale. This marker set divided the body into seven segments: right/left feet, right/left calves, right/left thighs, and a HAT (head/arms/trunk) segment. The video data were collected at a sampling rate of 120 Hz and noise was reduced at a cutoff frequency of 6 Hz with a symmetric, fourth-order Butterworth filter. Obscured markers (typically the medial knee, medial malleoli, and heel markers) were dynamically recreated during the STS trials using transforms derived from the static trial.

The older adults began each STS trial in a seated posture on a 46 cm (18 inch) high wooden bench, which was selected to correspond to the average seat height in public places. During the STS trials, the older adults placed each foot on a separate force platform (ATMI, Newton, Massachusetts, USA) mounted in the floor. Upper extremity support forces were measured by triaxial force transducers (Kistler, Winterthur, Switzerland) to determine the upper extremity contributions to the STS movement. The hand force sensors were mounted in a portable system to allow for placement on the wooden bench during the first experimental session and on a collapsible aluminum walker for the second session. The walker height with the hand force transducer bridge was 0.92 meters for all participants. The force platform and hand force transducer data were collected at 120 Hz and synchronized with the video data through the Peak Motus software.

Three initial lower extremity placements were identified for each participant: foot-back, foot-neutral, and dominant-staggered. The foot-back position corresponded to 100˚ of knee flexion bilaterally, determined goniometrically. The foot-neutral position corresponded to 90˚ of knee flexion bilaterally. The dominant-staggered position was selected by the subject with the constraint of the depth of the force platform and the guideline of placing the foot of the dominant leg more posterior. The dominant leg was determined to be the uninvolved lower extremity for participants with unilateral total knee arthroplasty. For the healthy older adults and individuals with bilateral total knee arthroplasties, the dominant leg was determined to be the more stable leg while performing single-leg stance. Multi-colored athletic tape was used to mark the three lower extremity placements and the location of the participant's buttocks on the bench during initial positioning in order to ensure consistency during the experimental trials.

In addition to varying lower extremity placement, three upper extremity positions were also tested: without arm assistance (no-hands), hands-on-bench, and separated hands. Without arm assistance, participants were required to cross their arms over their chest during the STS transfer. The hands-on-bench position involved placing the hands on the handles mounted on the bench to begin the movement and finishing the movement with their arms at their sides. With separated hands, the older adults began the task with their non-dominant hand (relative to their dominant leg) on a bench handle, and their dominant hand on a handle secured on the walker. The participants completed the movement with their non-dominant arm at their side in the hands-separated condition. For all trials, participants remained standing in their final position for 5 seconds after the completion of the data collection.

The three lower and three upper extremity positions were combined for a total of nine initial STS conditions. Three repetitions of six of the nine STS combinations were performed during an experimental session for a total of eighteen trials for each individual per session. During the first session, participants performed the three lower extremity positions without arm assistance and with their hands on the bench. For the second session, the three lower extremity positions were combined with the separated hands in addition to repeating sit-to-stand without arm assistance. The trial sequence within each session was balanced within subjects in an attempt to reduce the influence of learning, while the order of support conditions was balanced across subjects. An interval of at least one minute was allocated between trials to minimize fatigue and allow for repositioning. Participants were verbally instructed to position their arms by location (across chest, hands-on-bench or separated) and their feet according to the color of athletic tape (blue, red or yellow) for each trial. A two-stage verbal command ("Ready, Go") was used to cue the participant to initiate the STS movement.

The independent variables in this experiment were: initial lower extremity position, initial upper extremity position, and group (total knee arthroplasty or healthy older adult). The dependent measures of interest included: joint angles, joint moments, hand-support forces, loading asymmetry, and center of pressure (COP) parameters. Joint moments were calculated using inverse dynamic equations to combine anthropometric measures and body segment accelerations obtained from video data with ground reaction forces determined from force platform measurements. Hand support forces were calculated by using a calibrated regression equation with voltage measures from the force transducers. Maximum hip, knee, ankle, and combined support joint moments, as well as hand support forces provided an overall measure of the force generation demands on the body. The difference in ground reaction forces between the right and left feet as a percentage of body weight provided a measure of loading asymmetry during the STS transfers.

Center of pressure measures were determined from ground reaction forces and moments measured by the force platforms. Center of pressure area was calculated by determining the area surrounding the COP trace and was used as a measure of dynamic stability during the movement (Popovic, *et al*. 2000). COP excursions and maximum COP velocities in the mediolateral and anteroposterior directions were also used as indicators of stability. Calculations and filtering were programmed using MATLAB software (Natick, Massachusetts, USA).

Multiple univariate analyses of variance and Scheffe post-hoc tests were used to test for significant differences as a function of: total knee replacement versus healthy older adult, initial lower extremity position and initial upper extremity condition. SPSS software (SPSS, Inc., Chicago, Illinois, USA) was used for statistical analysis and statistical significance was set at $P<0.05$. Only the third trial of STS for each participant was used for this analysis.

**Results**

Upon cursory review of the data, there was minimal violation of the assumptions of parametric statistics required for ANOVA. Based on Shapiro-Wilk statistics of demographic data, the majority of the data were normally distributed. In some cases there was mild kurtosis present (around 1 or less on all variables), and very little skewness present in any variable. Using univariate ANOVAs, there were no statistically significant differences in demographics between groups (age, mass, height or BMI with $P$ values > 0.1).

**Table 3.1: Participant Demographics** (NS = non-significant; * = significant at $P<0.05$)

| **Category** | **TKA overall** (n=7) | **Healthy** (n=7) | ***P*-value** |
|---|---|---|---|
| Age (yrs) | 75.3 (5.5) | 71.4 (2.4) | NS |
| Mass (kg) | 82.9 (7.3) | 75.6 (12.2) | NS |
| Height (m) | 1.71 (0.07) | 1.72 (0.07) | NS |
| BMI ($kg/m^2$) | 28.3 (3.3) | 25.5 (4.6) | NS |
| Right knee ROM (˚) | 114.1 (9.8) | 134.7 (10.8) | 0.003* |
| Left knee ROM (˚) | 121.0 (12.4) | 135.1 (11.3) | 0.046* |

A one-way between-groups analysis of variance (ANOVA) was conducted to explore the impact of group on knee ROM. There was a statistically significant difference at the $P<0.05$ level in right knee ROM for the two groups: $F(1,12) = 13.904$, $P = 0.003$. There was also a difference in the groups on the left knee: $F(1,12) = 4.974$, $P = 0.046$. In addition to reaching statistical significance, the actual difference in mean scores between the groups was clinically important. The calculated effect sizes (using partial eta squared) were moderate to large for both sides: 0.29 on the left and 0.54 on the right. TKA participants exhibited less right knee ROM compared to healthy controls (Table 3.1, $P<0.005$). Of the TKA participants, all had a right knee replaced and 3 had also had a left knee arthroplasty (bilateral TKA). The range of motion difference on the left side was also statistically significant ($P<$ 0.05). There were no significant differences in demographics between bilateral TKA and unilateral TKA individuals ($P>0.05$). Range of motion differences between individuals with bilateral and unilateral TKA were not significantly different (Table 3.2, $P>0.005$).

**Table 3.2: Range of Motion for TKA**

| **ROM** (Degrees) | **Bilateral TKA** (n=3) | **Unilateral TKA** (n=4) | ***P*-value** |
|---|---|---|---|
| Right Knee ROM | 117 ± 12.5 (affected) | 112 ± 8.6 (affected) | 0.56 |
| Left Knee ROM | 122.7 ±20 (affected) | 119.8 ± 6.1 (unaffected) | 0.79 |

<u>Kinematic Variables:</u>

A 2(group) x 3(foot) x 3(hand) ANOVA was performed to investigate differences in joint angles in the sagittal plane. A Bonferroni correction of the alpha level for statistical significance was made to account for multiple comparisons such that the *P*-value was set at 0.001.

Group differences in range of motion (ROM) were identified in bilateral peak joint ankle dorsiflexion (*P*<0.001), knee flexion (*P*<0.001) and hip flexion (*P*<0.001). Controls exhibited higher peak ankle dorsiflexion and knee flexion compared to TKA participant across conditions, while peak hip flexion was higher in the TKA group compared to controls (Table 3.3).

**Table 3.3: Differences by Group** (** = significant differences between groups at *P*<0.001)

| Variables | TKA | Control |
|---|---|---|
| **Range of Motion (Degrees)** | | |
| Dominant Ankle Dorsiflexion ROM** | 22.9 (5.5) | 28.7 (4.2) |
| Non-dominant Ankle Dorsiflexion ROM** | 22.0 (5.3) | 25.2 (4.3) |
| Dominant Knee Flexion ROM** | 91.5 (4.9) | 96.2 (6.5) |
| Non-dominant Knee Flexion ROM** | 88.8 (5.6) | 92.8 (6.2) |
| Dominant Hip Flexion ROM** | 103.1 (11.3) | 95.0 (8.8) |
| Non-dominant Hip Flexion ROM** | 104.4 (11.8) | 97.3 (9.0) |
| **Joint Moments (Nm)** | | |
| Dominant Ankle Dorsiflexion Moment** | 2.7 (4.8) | 0.7 (3.2) |
| Non-dominant Ankle Dorsiflexion Moment** | 3.7 (4.4) | 1.2 (2.9) |
| Dominant Hip Extension Moment | 86.8 (32.9) | 83.2 (29.3) |
| Non-dominant Hip Extension Moment** | 114.6 (37.0) | 91.3 (36.5) |
| Dominant Support Moment** | 127.9 (25.4) | 149.9 (37.3) |
| Non-dominant Support Moment | 157.1 (35.2) | 138.5 (42.0) |

Differences were noted in terms of knee flexion range of motion (ROM) related to initial foot placement (*P*<0.001). Based on Scheffe post-hoc analysis, the dominant knee flexion ROM in the foot neutral placement was less than the other two placements (Table 3.4). For non-dominant knee flexion, the foot-back placement was larger than the other two placements.

**Table 3.4: Differences by Foot Placement Condition –** (** = differs from other dominant placements, $P<0.001$; * = differs from other non-dominant placements, $P<0.001$)

| Variables | Foot-Neutral | Foot-Back | Dominant Stagger |
|---|---|---|---|
| **Range of Motion (Degrees)** | | | |
| Dominant Knee Flexion ROM | 88.9 (4.6) ** | 95.8 (4.8) | 96.9 (5.9) |
| Non-dominant Knee Flexion ROM | 89.3 (4.4) | 95.8 (4.5) * | 87.4 (6.2) |
| **Joint Moment (Nm)** | | | |
| Dominant Ankle Dorsiflexion Moment | 3.7 (3.7) ** | 0.5 (3.6) | 0.8 (4.5) |
| Non-dominant Ankle Dorsiflexion Moment | 3.5 (3.7) | 0.9 (3.9) | 2.9 (3.6) |
| Dominant Knee Extension Moment | 43.7 (14.2) | 48.7 (16.2) | 57.2 (18.9) ** |
| Non-dominant Knee Extension Moment | 43.2 (12.4) | 47.0 (13.9) | 33.7 (14.8)* |

Differences were significant between hand support conditions in bilateral hip flexion ($P<0.001$). Increasing degrees of hip flexion were noted from the hand-split condition to the hands-on-bench condition to the no-hands condition for both dominant and non-dominant sides (Table 3.5).

**Table 3.5: Differences by Hand Placement Condition** (** = no hands condition greater than hands-separated or hands on bench, $P<0.01$)

| Variables | No Hands | Hands-Separated | Hands-on-Bench |
|---|---|---|---|
| **Range of Motion (Degrees)** | | | |
| Dominant Hip Flexion ROM | 102.1 (10.2) ** | 93.4 (9.6) | 98.3 (11.1) |
| Non-dominant Hip Flexion ROM | 104.0 (10.1) ** | 95.5 (10.9) | 99.5 (11.0) |
| **Joint Moment (Nm)** | | | |
| Dominant Support Moment | 148.5 (29.1) ** | 128.2 (34.1) | 131.7 (37.3) |
| Non-dominant Support Moment | 157.4 (38.7) | 139.5 (43.1) | 136.9 (34.6) |
| Total Support Moment | 302.6 (58.4) ** | 261.7 (70.8) | 263.7 (65.7) |

<u>Kinetic Variables - Joint Moments:</u>

Multiple univariate ANOVAs were performed to identify differences in joint moments between groups. Group differences were identified in ankle dorsiflexion moment ($P\leq0.001$), such that TKA individuals used more ankle dorsiflexion moment bilaterally (Table 3.3). Non-dominant hip extension moment was also increased for the TKA group ($P=0.001$).

Differences in ankle dorsiflexion moment were noted between foot placement conditions on the dominant side (*P*<0.001). For the dominant side, foot-neutral placement had more ankle dorsiflexion moment than the other two conditions (Table 3.4). On the non-dominant side, the foot-back placement trended (*P*<0.002) towards less ankle dorsiflexion moment than the other two placements.

Additional differences between foot placements were noted for bilateral knee extension moments (*P*<0.001). The dominant staggered placement differed from the other two foot placements with increasing extension moments on the dominant side and lower joint moments on the non-dominant side (Table 3.4). There were no statistically significant differences between hand placement conditions in terms of lower extremity joint moments.

Kinetic Variables - Support Moments, Symmetry & Upper Extremity (UE) Measures:

For bilateral support moments, weight-bearing asymmetry measures, and upper extremity support measures, the Bonferroni corrected *P*-value of 0.001 was also used. Individuals with TKA used less dominant-side support moment than the control group when averaged across conditions (Table 3.3, *P*<0.001). There was no statistical difference between groups in non-dominant support moment or total support moment. There was a statistically significant difference between hand conditions in terms of dominant and total support moment (*P*<0.001). The no-hand condition averaged 12.7% higher support moments than the other two hand conditions (Table 3.5). This trend was also evident in the non-dominant support moment but was not statistically significant (*P*<0.006).

There was a statistically significant interaction between group and upper extremity (UE) support measures (*P*<0.001). When permitted to use upper extremity assistance, the TKA group used more maximum UE force (*P*<0.001) than the control group (Table 3.6). In terms of UE force as a percentage of body weight, the TKA group used a higher percentage (TKA: 27.4(35.3)%; controls: 15.7(26.4)%, *P*<0.001). The amount of upper extremity assistance used was different in all three upper extremity conditions. Individuals used less

UE force in the separated-hand condition than in the hands-on-bench condition. Foot placement did not affect upper extremity assistance levels (*P*>0.001).

**Table 3.6: Upper Extremity Force Across Hand Placement Condition** (** = hands-on-bench greater than hands-separated, *P*<0.001; * = hands-separated greater than no hands, *P*<0.001)

| Group | No-Hands | Hands-Separated | Hands-On-Bench |
|---|---|---|---|
| Control | 0 N | 147.4 (107.7) N* | 269.6 (242.3) N** |
| TKA | 0 N | 359.2 (162.2) N* | 533.1 (319.1) N** |

In evaluating COP variables, healthy controls used a greater percentage of their base of support (BOS) in the mediolateral direction than individuals with TKA (TKA: 35(14)%; controls: 49(33)%, *P*<0.001). No differences were noted between groups in the size of the base of support. These results supported trends (*P*=0.02) of more symmetry in weight-bearing for TKA individuals during STS; however, both groups exhibited fairly high levels of asymmetry (19% vs. 23%) in terms of force distribution between dominant and non-dominant lower extremities across conditions during STS.

The hands-on-bench condition created a larger BOS than the other two conditions due to the placement of the hands lateral to the hips (*P*<0.001). There was also an effect of hand placement on the percent of the BOS used in the mediolateral direction (*P*<0.001), with 21% used in the hands-on-bench condition, 46% used in the no-hands condition and 55% used in the hands-separated condition.

For COP velocity measures, TKA individuals demonstrated slower peak velocities than healthy controls (Table 3.7), both in the mediolateral (ML) and anteroposterior (AP) directions (*P*<0.001). Overall, TKA individuals moved more slowly compared to controls as combined COP velocity was reduced (*P*<0.001). This was reflected in a trend towards slower STS times for individuals after TKA as well (1.7 ± 0.3 s vs. 1.5 ± 0.3 s, *P*=0.006).There was no effect of foot or hand placement on COP velocity measures or STS times (*P*>0.01).

**Table 3.7: Maximal Center of Pressure Parameters for Both Groups** (* = statistically significant difference between groups at $P<0.001$)

| COP Parameters | TKA | Control |
|---|---|---|
| Max COP Velocity-Anteroposterior | 0.46 (0.15) m/s* | 0.68 (0.28) m/s |
| Max COP Velocity-Mediolateral | 0.52 (0.23) m/s* | 0.88 (0.57) m/s |
| Max COP Velocity-Combined | 0.65 (0.24) m/s* | 1.12 (0.61) m/s |

**Discussion**

It has been suggested that after TKA, individuals compensate for range of motion deficits, weakness, and residual pain by altering movement patterns to accomplish functional tasks. The results of the current study are consistent with the findings of previous work, indicating that post-TKA individuals are limited compared to healthy adults in knee ROM (Chiu *et al*., 2002). The current data indicate that TKA individuals exhibited on average 13% (17˚) less knee flexion than healthy controls.

To accomplish STS, individuals are required to move from a position of seated stability to a vertical posture over a narrower BOS. As expected, TKA individuals utilized more ankle dorsiflexion moment in the foot-neutral placement to position the COM over the BOS, a key component of STS. Additionally, it appears individuals post-TKA used supplementary hip flexion for momentum generation and weight shifting over the BOS without regard to foot placement. Other explanations for the additional hip flexion included a slower speed of STS movement (Papa & Cappozzo, 2000) and the data from this investigation provide some support for this hypothesis as there was a trend for TKA individuals to have slower STS performance times.

Consistent with previous work indicating that UE assistance facilitates the COM shift over the BOS (Alexander *et al*., 1991), upper extremity assistance did reduce the amount of hip flexion range of motion utilized by both groups. Both groups used more hip flexion motion when no upper extremity assistance was permitted. However, the walker placement used in the hands-separated condition may have also provided a constraint to the STS movement, which limited the amount of hip flexion individuals were able to achieve during the movement.

It is common for individuals to spontaneously use their upper extremities for forward propulsion, elevation, or balance when performing STS (Wheeler *et al*., 1985). There was a distinct difference between groups in terms of the level of upper extremity support. As expected, the TKA group used a greater percentage of body weight to assist their STS performance when permitted than healthy controls. For both groups, individuals tended to use less upper extremity assistance in the hands-separated condition compared to the hands-on-bench condition. Therefore, as clinicians attempt to reduce reliance on upper extremity support for STS and progress clients towards full independence, it may be feasible to suggest a hands-separated starting posture. Previous work in our lab has indicated that the balance of AP and vertical hand support forces should be monitored to avoid "tipping" a walker as a safety concern.

The difference between the hand conditions also demonstrated how upper extremity assistance reduced lower extremity support moment as noted by previous authors. Burdett et al. (1985) demonstrated 50% reduction in hip moment through upper extremity usage with reductions in knee extension moment as well. In the current investigation, there was an average 12.7% reduction in dominant-side and total support moments across foot placements compared to the unassisted condition.

The hypothesis regarding TKA using greater hip extension moment was partially supported. Prior research established residual strength deficits following TKA (Rossi & Hasson, 2004), suggesting individuals may have difficulty producing the knee extension moment required during STS. In this study, TKA individuals only utilized greater non-dominant hip extension during STS. Therefore, TKA individuals tended to use more hip extension moment on the side of their least preferred knee, even for bilateral TKA participants. The dominant-side is typically associated as the more capable limb, although for individuals with TKA, the dominant-side as classified in this investigation may have symptoms of osteoarthritis which may include motion deficits, weakness or pain.

The dominant-staggered foot position did require higher dominant knee extension moment values than the other two foot placements since the dominant foot was placed more posteriorly. The additional knee extension moment was likely used to raise the COM in the ascension phase of STS, rather than for shifting the COM horizontally over the BOS. Speculatively, this posture may offer a greater shift in moment distribution to the opposite hip joint rather than the posteriorly-placed knee joint.

A staggered foot placement involves placing one lower extremity in a foot-back position while the opposite limb is either in foot-neutral or foot-forward. Healthy individuals may use a staggered placement preceding sit-to-walk transitions or to anticipate directional changes upon standing. Clinical experience demonstrates that an asymmetric foot placement may compensate for pain or weakness in individuals with pathology. Likewise, an asymmetric foot placement may be utilized for training purposes to enhance strength or stability (Brunt *et al.*, 2002).

In assessing postural stability, both groups exhibited equivalent levels of symmetry in their performance of STS, which is in contradiction to expected results and previous research (Su *et al.*, 1998). In the current study, it did appear that TKA individuals maintained their weight-bearing posture on their non-dominant (affected) side rather than preferably loading their dominant side. However, bilateral TKA, residual pain levels, and the potential for non-symptomatic osteoarthritis may confound the symmetry issue as weight-bearing asymmetry measures were not significantly different between groups.

As healthy controls shifted more within the BOS in the mediolateral direction than the TKA group, it suggests that healthy individuals were comfortable shifting weight between either lower extremity. TKA individuals may have had limitations either in strength, pain or balance that limited weight shifting in the mediolateral plane during STS. Across groups, individuals used more of the BOS in the mediolateral direction in the hands-separated condition, followed by the no-hands condition and then the hands-on-bench condition. A similar trend of weight-bearing asymmetry was noted in the data ($P$<0.02).

Individuals with TKA may compensate for this potential asymmetry by reducing movement velocity to maintain postural stability. The reduced peak velocity of the COP in both the anteroposterior and mediolateral direction may be an indicator of attempts to control stability. TKA individuals appeared to make slower corrections in dynamic balance as well as performing the STS movement more slowly. Similarity in knee extension moments and weight-bearing asymmetry across groups would suggest stability during STS. TKA individuals maintain stability through controlling weight-shifting by limiting extremes of motion and reducing speed.

The current findings suggest individuals with TKA may benefit from utilizing upper extremity assistance with both hands placed on the seated surface to facilitate shifting the COM over the BOS. This posture appears to provide a maximum level of postural stability as well as a moderate hip flexion requirement. As there were no significant differences between the two symmetrical foot placements in terms of knee or hip moments, it would suggest either position could be used as a symmetrical posture. Placing the feet posterior to the knee joint line may reduce the level of ankle dorsiflexion moment; however, as individuals after TKA may have difficulty adopting a foot-back posture due to limited knee flexion, a foot-neutral posture may be the preferred posture. Finally, the asymmetric placements (dominant-staggered foot placement and hands-separated for upper extremity assistance) may be utilized for more functionally advanced individuals when strengthening is desired as they provide a moderate level of joint moment requirements and a moderate level of assistance.

As the data support, post-TKA individuals use compensatory mechanisms to perform STS. Utilizing adjacent joint movement, assisting with the upper extremities and restricting the speed and range of movement are modifications to the STS task to assist individuals in successful performance. This information may provide beneficial information to rehabilitation professionals for determining appropriate therapy interventions, to improve safety during transfers and to maximize client independence with mobility training.

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# CHAPTER 4: FUNCTIONAL MOVEMENT ASSESSMENT IN OLDER ADULTS WITH KNEE PAIN

A paper to be submitted to *Clinical Biomechanics*

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## Abstract

Background: Individuals with knee osteoarthritis appear to experience declines in postural stability which may contribute to decrements in functional performance during sit-to-stand and stair negotiation. The purpose of this project was to evaluate the postural stability of individuals with knee pain during functional tasks such as stair descent and sit-to-stand, and to correlate lower extremity strength measures with performance during these activities.

Methods: Individuals with symptomatic knee pain and physician-diagnosed osteoarthritis (n=12) performed sit-to-stand with three different initial foot placements and descended a three-step staircase without upper extremity assistance. Motion analysis and kinetic assessments were completed and inverse dynamics was utilized to calculate lower extremity joint moments, center of pressure velocities and weight-bearing asymmetry. Lower extremity strength assessments were obtained with a hand-held dynamometer.

Findings: Unilaterally-affected individuals demonstrated less symmetry during sit-to-stand compared to those with bilateral symptoms ($P$=0.002). Affected hip extensor moment, affected knee extensor moment and bilateral hip abduction strength were correlated with weight-bearing symmetry during sit-to-stand ($P$<0.005). During stair descent, mediolateral center of pressure velocity and knee extension moment increased, while knee varus moment decreased on the third (floor) step ($P$<0.01).

Interpretation: Weight-bearing asymmetry and altered joint loading were factors in sit-to-stand for individuals with knee pain. For both groups, individuals attempted to minimize affected side knee extension moments. Weight-bearing asymmetry was more altered for those with unilateral symptoms during sit-to-stand, while bilaterally-affected individuals were more challenged by mediolateral stability during stair descent.

Keywords: Sit-to-stand, Stair descent, Osteoarthritis

**Introduction**

Postural stability refers to an individual's ability to control the body's center of mass relative to the base of support. Stability is often assessed by determining the center of pressure (COP) movement relative to the base of support, both in terms of magnitude and velocity (Popovic *et al.*, 2000). Many researchers have demonstrated that individuals with neurological deficits demonstrate asymmetric weight-bearing and instability during functional tasks, increasing fall risk. Older adults may also demonstrate weight-bearing asymmetry due to postural imbalances, muscle weakness, reduced joint flexibility or pain. Researchers have identified an association between instability and difficulty with functional movements for individuals with orthopedic disorders (Marsh *et al.*, 2003).

Although research on falls in an orthopedic population is limited, there are significant associations between postural stability and lower extremity strength measures in individuals with knee osteoarthritis (Hasson *et al.*, 2001; Jadelis *et al.*, 2001). Osteoarthritis is a common physical limitation, affecting at least 30% of the older adult population (Harrison, 2004). The knee joint is the most commonly affected weight-bearing joint, limiting functional activities such as ambulation, stair climbing and kneeling (CDC, 2006). Due to the ongoing degenerative process, osteoarthritic joints are painful and demonstrate reduced range of motion, weakness and proprioceptive deficits (Messier *et al.*, 2002; Messier *et al.*, 1992). Changes in the ligaments, muscles and joints due to aging and osteoarthritis may contribute to altered knee joint loading and potential weight-bearing asymmetry which may lead to instability.

Research indicates that individuals with osteoarthritis have reduced postural control as demonstrated by increased postural sway during static stance (Masui *et al.*, 2006; Wegener *et al.*, 1997) and reduced dynamic standing balance (Hinman *et al.*, 2002; Messier *et al.*, 2002). For older adults, balance deficits and weakness may lead to difficulties with daily functional tasks, such as walking, sit-to-stand transfers (STS) and stair negotiation. Osteoarthritic individuals report greater difficulty with STS and stair negotiation due to the extreme knee flexion involved (Marsh *et al.*, 2003). Difficulty with STS demonstrated by slow performance (>2.0 s) has been associated with increased fall risk in older adults with osteoarthritis (Nevitt *et al*., 1989). Researchers have suggested that STS is an indirect measure of knee strength which may correlate with dynamic stability required for STS and gait (Scarborough *et al.*, 1999).

Stair descent also places large forces on the lower extremity joints, up to 1.6-1.8 times body weight at the knee joint (Stacoff *et al.*, 2007). For individuals with osteoarthritis or total knee replacement, the increased muscular demand may require compensations to avoid pain or collapse. Knee strength is critical for stair negotiation, and older adults need to use a greater proportion of their knee extensor strength to descend steps than younger adults (Reeves *et al.*, 2008).

In general, older adults and individuals with osteoarthritis perform stair descent more slowly than young adults (Mian *et al.*, 2007a; Mian *et al.*, 2007b, Verghese *et al.*, 2008). Older adults demonstrate increased pelvic and hip motion in the frontal plane (Mian *et al.*, 2007a; Mian et al., 2007b) and individuals with osteoarthritis display asymmetric loading during stair descent (Liikavainio *et al.*, 2007; Stacoff *et al.*, 2007). Older adults also tend to spend more time in single limb support during stair descent than younger adults (Lythgo *et al.*, 2007; Mian *et al.*, 2007b). These findings suggest mediolateral stability may play a greater role in functional performance for individuals with osteoarthritis. An increase in the proportion of time on one lower extremity may require additional hip strength in the frontal plane to counter the external hip adduction moment created in order to avoid instability on stairs.

Although researchers have identified an association between postural stability, knee strength and difficulty with functional movements for individuals with knee pain, there are few links between osteoarthritis, fall risk and other lower extremity strength measures. Individuals with knee osteoarthritis appear to experience declines in stability which may contribute to increased fall risk and decrements in functional performance during STS and stair negotiation. Improvements in lower extremity strength have translated into improved function for individuals with osteoarthritis (Alexander *et al.*, 1997; Hinman *et al.*, 2007; Bennell *et al.*, 2010). Previous research has focused on the musculature controlling sagittal plane movement (quadriceps and hamstrings). However, there is limited evidence describing the strength of hip musculature for individuals with knee osteoarthritis.

The purpose of this project was to evaluate the postural stability of individuals with knee pain while performing functional tasks such as stair descent and STS, and to correlate isometric lower extremity strength measures with performance during these functional activities. Previous research suggests individuals that exhibit asymmetry and reduced step lengths during gait may exhibit weak hip musculature that contributes to mediolateral instability during functional movements. Knee joint loads are significantly higher during STS and stair climbing than gait. Thus, individuals with knee pain may exhibit asymmetry during functional tasks when redistributing joint loads to avoid pain or due to weak hip musculature. By identifying clinical measures (strength or balance) that affect stability, potential opportunities to intervene and enhance safety may be identified.

Based on previous research, we hypothesized that hip extension and hip abductor weakness would correlate with mediolateral instability during stair descent and STS. This would be supported by increased mediolateral sway measures and reduced weight-bearing symmetry. Asymmetry in joint moments related to altered weight-bearing would also be a potential source of instability if joint moments approach or exceed strength capacity or pain tolerance. Asymmetry was hypothesized to be more pronounced in the staggered foot placement STS postures that preferentially load the posterior leg. Finally, STS performance time was hypothesized to be negatively associated with knee and hip strength measures.

**Methods**

Study participants were recruited via community flyers and through word-of-mouth. A verbal review of medical history, joint pain and physical activity was completed with each potential participant to screen for exclusion criteria. Exclusion criteria included recent cortisone injection, previous or currently scheduled total joint replacement, or other physical impairment that limited an individual's ability to perform sit-to-stand or stair negotiation without upper extremity assistance. Potential participants were excluded due to impairments including: total joint replacement, morbid obesity, rheumatoid arthritis, neuropathy and low back pain. Twelve individuals with symptomatic knee pain and physician-diagnosed osteoarthritis participated in this study. The Human Subjects Research Compliance Office at Iowa State University approved the experimental protocol, and the research participants provided informed consent prior to participating in the project.

**Table 4.1: Participant Demographics**

| Variable | Mean (SD) |
|---|---|
| Age (years) | 62.1 (3.4) |
| Mass (kg) | 94.4 (26.1) |
| Height (cm) | 169.5 (10.5) |
| Comorbidities | 3.0 (1.7) |
| Medications | 2.7 (2.9) |
| Exercise (hrs/wk) | 3.3 (2.3) |
| Berg Balance Scale | 55 (1) |
| WOMAC – pain | 12.4 (4.1) |
| WOMAC – stiffness | 4.8 (1.6) |
| WOMAC - function | 31. 8 (9.3) |
| Pain Rating - affected | 2.8 (2.1) |
| Pain Rating - unaffected | 1.6 (1.9) |

During the experimental session, height, weight, and anthropometric measurements of ASIS width and thigh length were recorded. Knee joint range of motion was goniometrically determined (Rothstein *et al.*, 1983). Hand and foot dominance was determined by asking participants which upper extremity was used for writing tasks and which lower extremity was used for kicking tasks. Demographic characteristics of the participants are reported in Table 4.1. The calculated body mass index was 32.4 (4.1) kg/m$^2$. Self-reported disease status was measured using Western Ontario and McMaster Universities (WOMAC) Osteoarthritis Index with 17 functional items, 2 stiffness questions and 5 pain components (Bellamy *et al.,* 1988). Pain ratings via an 11-point visual analog scale were completed through all phases of data collection and accommodations made in testing as required. Clinical balance assessment was completed via the Berg Balance Scale (Berg et al., 1989).

Lower extremity strength was assessed using a hand-held dynamometer (Manual Muscle Tester by Lafayette Instrument Company, Lafayette, Indiana, USA). Participants were evaluated in sitting (knee flexion/extension), standing (hip extension) and supine (hip abduction/flexion, ankle plantarflexion/dorsiflexion/inversion/eversion). The testing sequence was consistent across participants to minimize position changes. A "make" test with a 5 second count was used by a single evaluator. Participants were instructed to reach maximal contraction by 3 seconds and hold for 2 additional seconds. Peak force values (in kg) were recorded for each trial from the digital readout on the dynamometer. Three consecutive measurements were obtained for each muscle group (Bohannon, 1997; Boone *et al.*, 1978) and the maximal value was used for analysis. Test-retest reliability was calculated for the first & second measurements obtained by the tester using intraclass correlation coefficients (ICC). The range of ICCs was from 0.773 to 0.971, depending on muscle action and side for the 18 assessments. Most values were greater than 0.876, with a median score of 0.908.

Retro-reflective markers were applied to participants for tracking by an eight-camera video system (Vicon, Englewood, Colorado, USA) to measure the movements of anatomical landmarks during the functional tasks. The spherical markers (diameter of 2 cm) were mounted on a vinyl base and attached to the skin or snug-fitting clothing with adhesives. A static trial was collected with the older adult standing quietly with markers placed bilaterally on the great toes, 5$^{th}$ metatarsal heads, heels, lateral malleoli, medial malleoli, mid-shank, lateral knee joint lines, medial knee joint lines, mid-thigh, greater trochanters, ASISs, PSISs, acromion processes, triceps, lateral elbow joints, forearms, and ulnar styloids. A single marker was placed at the suprasternale, sacrum, cervicale and forehead. This marker set divided the body into eight segments: right/left feet, right/left calves, right/left thighs, pelvis, and a HAT (head/arms/trunk) segment. The video data were collected at a sampling rate of 160 Hz and noise was reduced at a cutoff frequency of 6 Hz with a symmetric, fourth-order Butterworth filter. Medial knee, medial ankle and heel markers were dynamically recreated using transforms derived from the static trial.

The older adults performed stair descent trials from atop a three-step wooden stair module (step height = 19 cm, tread depth = 28 cm) equipped with handrails. Two portable force platforms (AMTI, Watertown, Massachusetts, USA) were placed on the second and third tread to collect ground reaction forces. Participants descended the stairs using a reciprocal (step-over-step) pattern at a self-selected pace. A two-stage verbal command ("Ready, Go") was used to cue the participant to initiate descending the stairs.

Participants performed the stair descent task with arms freely moving at their sides. Beginning at the top of the staircase, individuals descended three steps to floor level and walked an additional two meters, coming to a stop with their feet side-by-side until the completion of the 10 second data collection. Each participant completed two trials leading with the right foot and two trials leading with the left foot, for a total of four trials. The order of performance was balanced across subjects.

Sit-to-Stand:

Participants began each STS trial in a seated posture on a wooden bench equipped with a portable force platform (0.40 m wide x 0.60 m long x 0.08 m high) at a seat height of 48.5 cm. During the STS trials, the older adults placed each foot on a separate force platform (ATMI, Newton, Massachusetts, USA) mounted in the floor. The force platform data were collected at 160 Hz and synchronized with the video data through the Vicon Nexus software.

Three initial lower extremity placements were identified for each participant: self-selected symmetric, right-staggered and left-staggered. The self-selected symmetric position allowed the individual to place both feet at a comfortable position directly across from each other. The right-staggered position corresponded to a position where the right foot was placed posteriorly relative to the left foot. The left-staggered position was the opposite, where the left foot was posterior to the right foot. Multi-colored athletic tape was used to mark the three lower extremity placements, foot width and seat depth on the bench during initial positioning in order to ensure consistency during the experimental trials. Seat depth was set at 75% of thigh length from anthropometric measures.

Two repetitions of the three foot positions were performed for a total of six trials for each individual. Participants were verbally instructed to position their feet according to the color of athletic tape (blue, red or yellow) for each trial. The order of the trials was balanced within subjects. A two-stage verbal command ("Ready, Go") was used to cue the participant to initiate the sit-to-stand movement. An interval of at least one minute was allocated between trials to minimize fatigue and allow for repositioning. The participants completed the STS movement with their upper extremities crossed over their body to avoid using their upper extremities for assistance in rising. For all trials, participants remained standing in their final position until completion of the 10-second data collection.

Data Analysis:

Performance duration measures were calculated for STS time and stair stance durations. Joint moments were calculated using inverse dynamic equations to combine anthropometric measures and body segment accelerations obtained from video data with ground reaction forces determined from force platform measurements. The difference in maximum ground reaction forces between the right and left feet as a percentage of body weight provided a measure of loading symmetry during the sit-to-stand transfers. Centers of pressure were determined from ground reaction forces and moments measured by the force platforms. Center of pressure excursions and maximum COP velocities in the mediolateral and anteroposterior directions were used as indicators of stability. Univariate and multivariate ANOVA were utilized for statistical comparisons. Significance level of $P<0.05$ was set for multivariate ANOVA and $P<0.01$ for univariate comparisons. Correlational analysis using non-parametric statistics was completed to determine associations between strength measures and task performance variables. Calculations and filtering were programmed using MATLAB software (Natick, Massachusetts, USA).

## Results

Two participants related only right-sided osteoarthritic knee pain while three individuals relayed left side pain. Seven participants reported bilateral knee pain. Patient symptom levels were in the mild to moderate level based on WOMAC scores and visual analog pain ratings (Table 4.1). Overall, the participants were generally healthy, active participants, without clinical balance deficits based on Berg Balance Scale scores (Berg et al., 1989). Strength data were assigned into affected and unaffected limb categories. Paired t-tests were performed to compare affected and unaffected side strength assessments (Table

4.2). There were no differences noted between lower extremity strength measures based on paired t-tests ($P$>0.01).

**Table 4.2: Strength Assessments in Newtons –** Mean (SD)

| **Motion** | **Affected** | **Unaffected** | ***P*-value** |
|---|---|---|---|
| Hip Extension | 149.1 (25.5) | 152.0 (37.3) | 0.564 |
| Hip Flexion | 157.9 (38.2) | 154.9 (36.3) | 0.689 |
| Hip Abduction | 190.2 (36.3) | 182.4 (34.3) | 0.440 |
| Knee Extension | 267.7 (85.3) | 264.8 (55.9) | 0.833 |
| Knee Flexion | 176.5 (44.1) | 174.5 (37.3) | 0.658 |
| Ankle Plantarflexion | 372.6 (132.4) | 380.5 (110.8) | 0.714 |
| Ankle Dorsiflexion | 143.2 (44.1) | 131.4 (26.5) | 0.227 |
| Ankle Inversion | 70.6 (17.7) | 80.4 (22.6) | 0.135 |
| Ankle Eversion | 82.4 (21.6) | 82.4 (20.6) | 0.950 |

Sit-to-Stand:

Multiple ANOVAs were completed after screening to test for effects of knee pain symptoms (unilateral or bilateral) and STS condition (affected and unaffected). Statistical significance was identified at $P$<0.01 after Bonferroni correction for multiple variables. The variables assessed by ANOVA included: weight-bearing (WB) symmetry, mediolateral COP velocity, knee extension moment, hip extension moment, and hip abduction moment.

ANOVA revealed a difference between knee symptom groups in WB asymmetry ($P$=0.002). The unilateral group demonstrated less symmetry during STS performance compared to those with bilateral symptoms (Table 4.3). Bilateral differences in hip extension moment were also identified ($P$≤0.001). Individuals with bilateral symptoms exhibited higher hip extension moment across conditions compared to those with unilateral symptoms (Table 4.3). Finally, those with bilateral knee symptoms utilized higher knee extension moment on the affected side relative to the unilateral knee symptom group (Table 4.3, $P$<0.001).

**Table 4.3: Knee Symptom Group Differences** (**Statistical significance at *P*<0.01)

| Variables | Bilateral Symptoms | Unilateral Symptoms | *P*-value |
|---|---|---|---|
| Weight-bearing Symmetry | 89 (10) % | 80 (13) % | 0.002** |
| STS Time | 1.15 (0.15) s | 1.54 (0.81) s | 0.013 |
| Mediolateral COP Velocity | 4.95 (2.57) cm/s | 4.84 (2.64) cm/s | 0.887 |
| Affected Knee Extension Moment | 0.66 (0.10) Nm/kg | 0.52 (0.12) Nm/kg | <0.001** |
| Unaffected Knee Extension Moment | 0.70 (0.09) Nm/kg | 0.71 (0.17) Nm/kg | 0.636 |
| Affected Hip Extension Moment | 0.61 (0.18) Nm/kg | 0.48 (0.12) Nm/kg | <0.001** |
| Unaffected Hip Extension Moment | 0.64 (0.17) Nm/kg | 0.48 (0.14) Nm/kg | <0.001** |

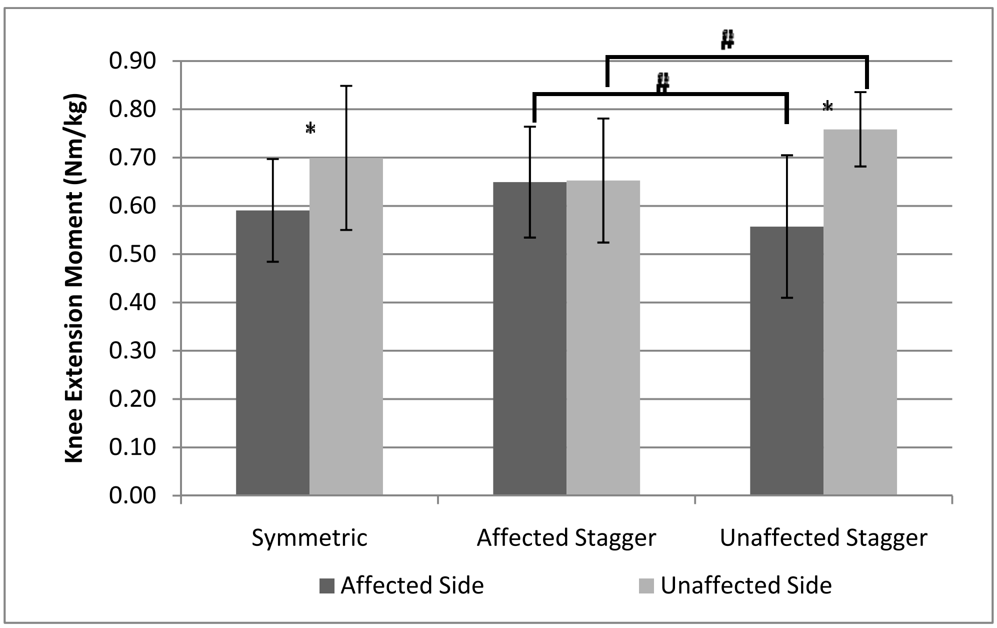


**Figure 4.1: Knee Extension Moment by Foot Placement Condition** (* indicates statistical significance within foot placement at *P*<0.01 with 2-way paired t-test; # indicates significance between foot placements at *P*<0.01 with ANOVA)

ANOVA detected an effect of foot placement (Figure 4.1) for bilateral knee extension moments (*P*<0.01). For both the affected and unaffected knee extension moments, Tukey post-hoc testing identified that the affected stagger placement differed from the unaffected stagger placement (*P*<0.02). In the affected stagger placement, the affected knee extension moment was increased while the unaffected knee extension moment was reduced compared to the unaffected stagger placement. Paired t-tests revealed lower knee extension moments on the affected side compared to the unaffected limb within the symmetric and unaffected stagger foot placements (*P*<0.01).

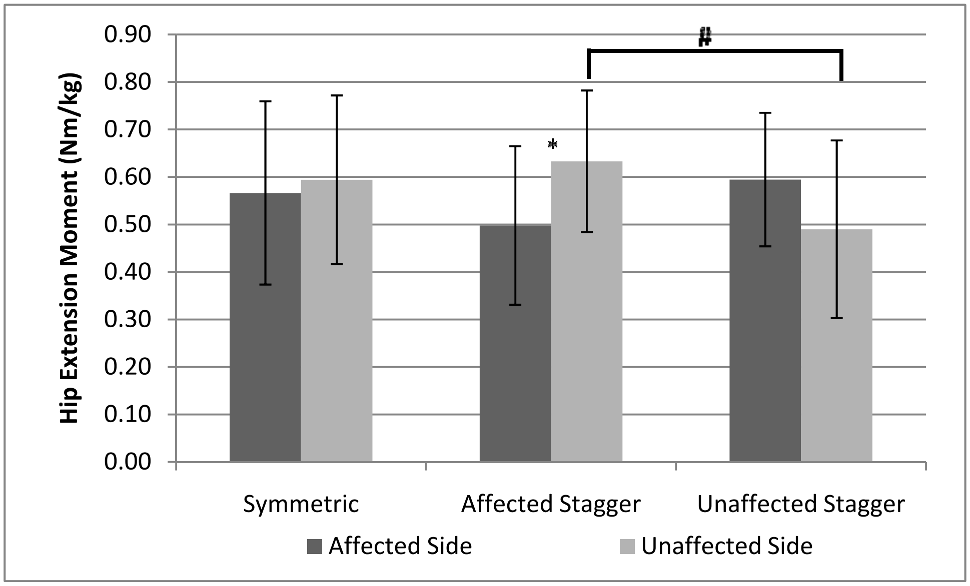


**Figure 4.2: Hip Extension Moment by Foot Placement Condition** (* indicates statistical significance within foot placement at *P*<0.01 with 2-way paired t-test; # indicates significance between foot placements at *P*<0.01 with ANOVA)

ANOVA revealed a significant difference in unaffected hip extension moment across foot placement conditions (Figure 4.2, *P*=0.005). Tukey follow-up testing indicated higher unaffected hip extension moment in the affected stagger placement compared to the unaffected stagger placement. Paired t-tests also revealed a difference between hip extension moments on the affected and unaffected sides within the affected stagger placement (*P*<0.01).

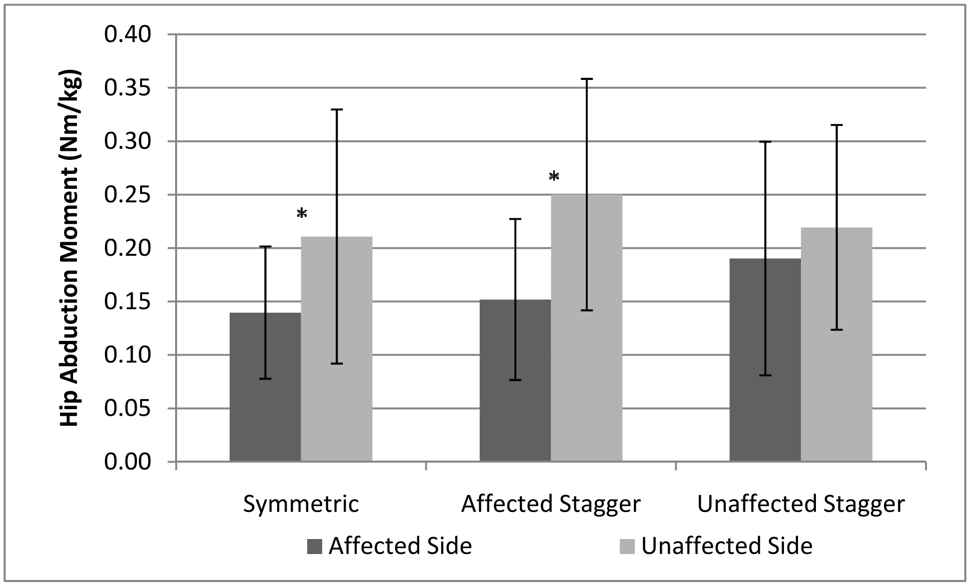


**Figure 4.3: Hip Abduction Moment by Foot Placement Condition** (* indicates statistical significance within foot placement at *P*<0.01 with 2-way paired t-test)

Univariate analysis did not indicate a difference in hip abduction moments across foot placement conditions (affected *P*=0.16, unaffected *P*=0.03). Paired t-tests indicated differences within affected and unaffected limb hip abduction moment during STS for symmetric and affected stagger foot placements (*P*<0.01). In both cases, the affected limb hip abduction moment was less than that of the unaffected limb (Figure 4.3).

Correlation analysis revealed positive correlations between STS time and bilateral knee extension strength (affected: 0.501, unaffected: 0.642, *P*<0.001) and ankle plantarflexion (affected: 0.363, *P*<0.005, unaffected: 0.625, *P*<0.001).

Both affected hip extensor moment and affected knee moment were correlated with WB symmetry (*P*<0.005), as was bilateral hip abduction strength (affected: 0.660 & unaffected: 0.473, *P*<0.001). Bilateral hip extension moments had positive correlations with bilateral hip extensor strength assessments (range 0.407-0.546, *P*<0.001). Strong correlations were found between bilateral hip abduction strength measures and affected hip extension moment (affected: 0.605; unaffected: 0.644, *P*<0.001).

Stairs:

One subject was removed from the stair analysis due to saturation of force platform signals. A mixed between-within subjects analysis of variance was conducted to assess the impact of stair descent condition on stance time, mediolateral COP velocity, knee varus moment, hip abduction moment and knee extension moment across three staircase steps. Statistical significance was determined by an alpha level of <0.01 based on multiple comparisons.

The mixed ANOVA identified a significant effect of step on stance time (*P*<0.001), such that each step had a shorter stance time than the one proceeding it. The first step (0.93±0.28 s) was longer than the second (0.78±0.14 s), with the final step having the shortest stance time (0.68±0.10 s).

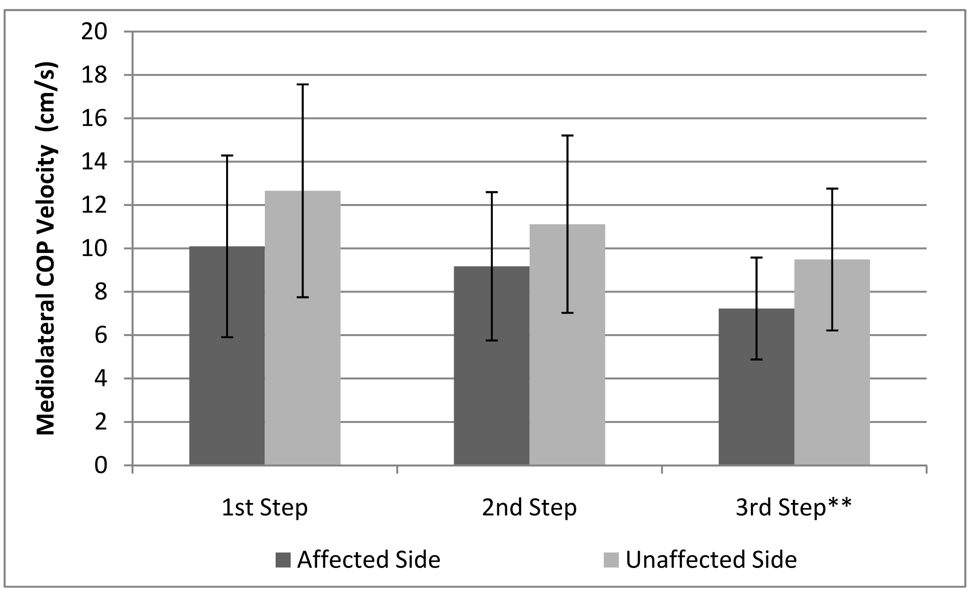


**Figure 4.4: Mediolateral Center of Pressure Velocity across Staircase Steps by Affected Limb** (** indicates statistical significance from other steps at *P*<0.01)

The ANOVA identified a significant interaction between the descent condition and steps on mediolateral COP velocity (*P*<0.001). There was a significant main effect of step on mediolateral COP velocity as well (*P*<0.001) with a large effect size (partial eta squared =

0.310) (Figure 4.4). Upon pairwise comparisons, the ground step (3rd step) was significantly lower in mediolateral COP velocity (8.33±3.03 cm/s) than the first two steps (1st step = 11.36±4.68 cm/s; 2nd step = 10.18±3.86 cm/s, $P<0.01$). There was no significant main effect of descent condition. There was a main effect of knee symptom group, such that the bilaterally-affected individuals demonstrated higher mediolateral COP velocities compared to the unilaterally-affected participants (bilateral: 11.0±4.4 cm/s, unilateral: 8.3±2.9 cm/s, $P=0.004$).

Affected hip abduction strength correlated with mediolateral COP velocity on the first and third steps (1st step: 0.422, 2nd step: 0.467, $P<0.005$), while unaffected hip abduction strength correlated with mediolateral COP velocity on the first step (1st step: 0.534, $P<0.001$). Other correlations were inconsistent between hip strength assessments and stair step variables.

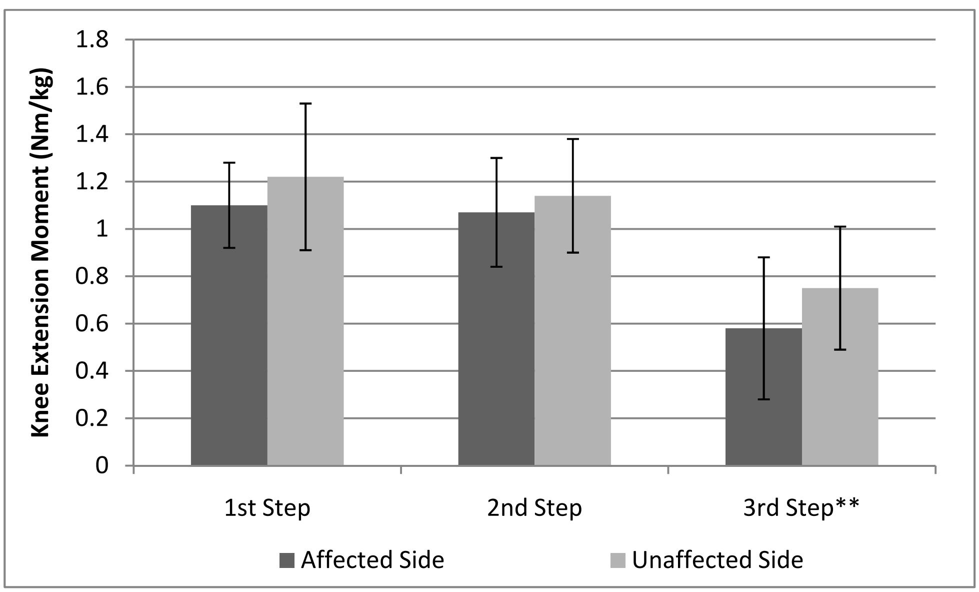


**Figure 4.5: Knee Extension Moments across Staircase Steps by Affected Limb** (** indicates statistical significance from other steps at $P<0.01$)

The mixed ANOVA identified a significant interaction between step and descent condition on knee extension moment (Figure 4.5, $P<0.01$). There was a significant effect of step on knee extension moment ($P<0.001$), with a strong effect size (partial eta squared =

0.811). Based on follow-up testing through pairwise comparisons, knee extension moment on the third step (0.66±0.29 Nm/kg) was significantly lower than the other two steps (1st step: 1.16±0.25 Nm/kg, 2nd step: 1.11±0.23 Nm/kg, *P*<0.001). The main effects comparing the descent conditions and knee symptom group were not significant.

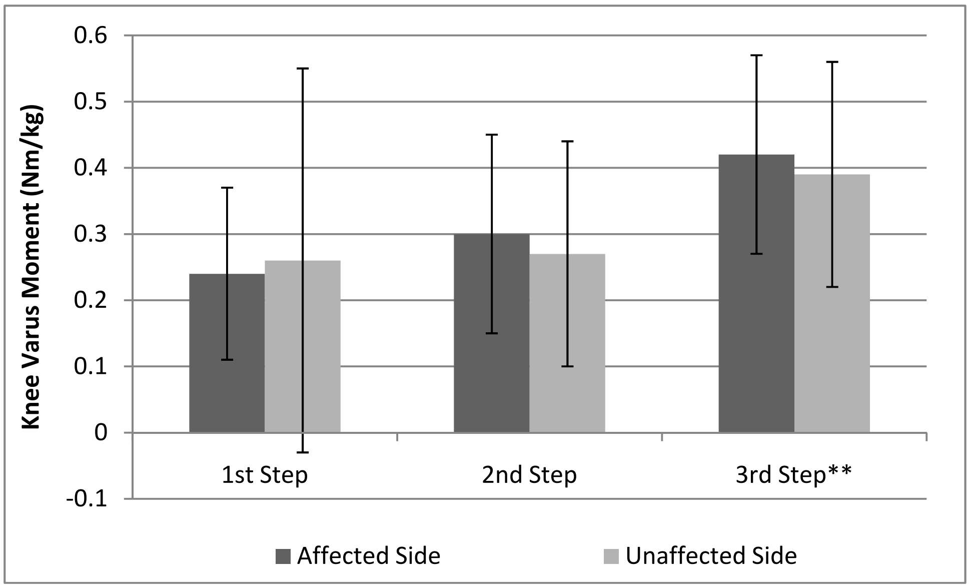


**Figure 4.6: Knee Varus Moments across Staircase Steps by Affected Limb** (** indicates statistical significance from other steps at *P*<0.01)

**Table 4.4: Knee Varus Moments (Nm/kg) by Staircase Step for Knee Symptom Groups**

| | **Unilateral Symptoms** | | **Bilateral Symptoms** | |
|---|---|---|---|---|
| **Staircase Steps** | Affected Side | Unaffected Side | Affected Side | Unaffected Side |
| 1st Step | 0.19 (0.10) | 0.28 (0.08) | 0.27 (0.14) | 0.25 (0.37) |
| 2nd Step | 0.24 (0.09) | 0.32 (0.14) | 0.34 (0.17) | 0.24 (0.18) |
| 3rd Step | 0.33 (0.12) | 0.47 (0.10) | 0.47 (0.14) | 0.34 (0.18) |

The mixed ANOVA identified a three-way interaction between step, knee symptom group and descent condition for knee varus moment (*P*=0.004). There was a main effect of step (Figure 4.6, *P*<0.001), with a large effect size (partial eta squared = 0.445). Based on follow-up analysis, the third step exhibited higher knee varus moments (0.41±0.16 Nm/kg) compared to the first and second steps (1st step: 0.25±0.22 Nm/kg, 2nd step: 0.28±0.16

Nm/kg, *P*<0.001). This pattern was consistent for both groups (Table 4.4). There was no main effect for descent condition or knee symptom group (*P*>0.01). A mixed ANOVA did not identify any significant findings for hip abduction moment (Figure 4.7).

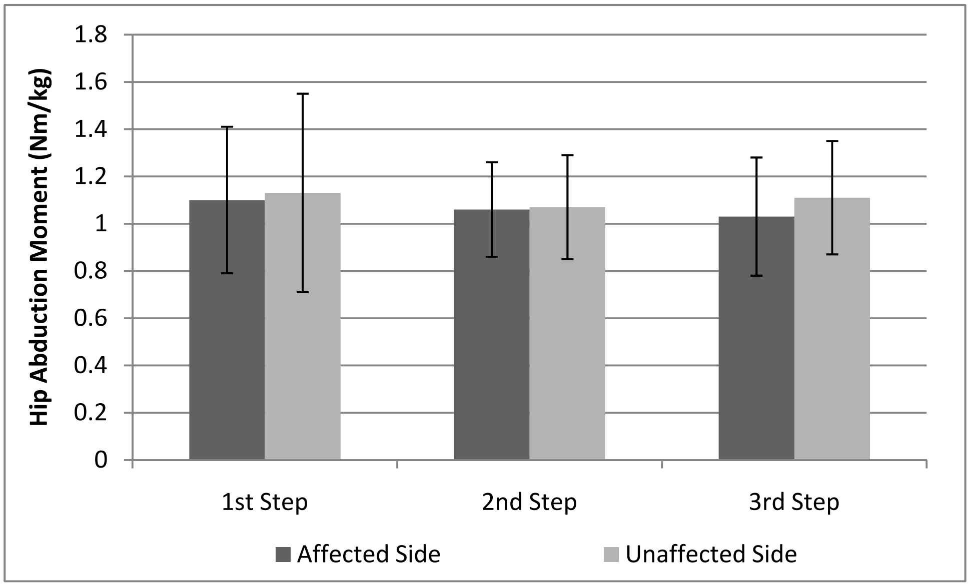


**Figure 4.7: Hip Abduction Moments across Staircase Steps by Affected Limb**

## Discussion

From the findings of this investigation, weight-bearing asymmetry and altered joint loading were factors in STS performance for individuals with knee pain. Weight-bearing asymmetry was more altered for those with unilateral symptoms compared to those participants with bilateral symptoms. As expected, those with unilateral symptoms tended to favor the more symptomatic limb and demonstrated increased asymmetry in STS (approximately 10% greater). It is likely these individuals utilized this weight shift to reduce knee extension moments on the affected side to reduce pain levels, as individuals with bilateral symptoms demonstrated higher levels of knee extension moments compared to the unilateral group. Weight-bearing symmetry was correlated with affected hip and knee extension moments, indicating that higher levels of joint moments may suggest that an individual is able to tolerate additional loading on the affected side and thus a more symmetrical STS performance.

For both knee pain groups, individuals attempted to minimize knee extension moments on the affected side to compensate for knee pain. For both the symmetric and unaffected stagger placements of STS, the unaffected limb provided more knee extension moment than the affected limb (denoted by ***** in Figure 4.1). In the affected stagger placement, the affected limb was placed posteriorly and the unaffected limb was not in a biomechanically advantageous position. In this condition, the affected knee extension moment was increased without exceeding symptom levels based on pain ratings while the unaffected knee extension moment was less than in the unaffected stagger position (denoted by **#** in Figure 4.1).

Bilateral hip extensor strength measures were highly correlated with STS hip extensor moments, indicating a role in performance of STS ascension. Higher unaffected hip extension moments were noted in the affected stagger placement compared to the unaffected stagger placement (denoted by **#** in Figure 4.2). Therefore, it appears that with a less flexed knee position in either of the staggered postures, additional hip flexion motion may be necessary to shift the center of mass over the base of support. In the affected stagger placement, individuals tended to use additional hip extension moments on the unaffected side compared to the affected hip (denoted by ***** in Figure 4.2). It is likely that the affected limb was unable to accommodate the additional demands for rising to stance in the affected stagger placement, while the unaffected limb was able to produce sufficient knee extension moments to achieve STS in the unaffected stagger position. Unilaterally-affected participants were likely able to compensate for the reduced affected knee extension moments by using unaffected knee extension moments. However, the bilaterally-involved individuals would likely experience added knee discomfort from such an accommodation. Instead, the bilaterally-affected participants utilized additional hip extension moments to complete STS. These findings are in agreement with other authors who noted compensations for reduced knee extension moments by altering foot placement or hip extension moments through trunk flexion (Fleckenstein *et al.*, 1988; Su *et al.*, 1998).

Previous authors have suggested posterior positioning for an affected extremity during STS for strengthening purposes in a neurologically-involved population, as the posterior placement requires increased knee extension moments (Brunt *et al.*, 2002). However, in an orthopedically-involved population, pain appears to control performance more than weakness and thus, a symmetric or unaffected stagger may be the preferred placement. This may lead to a compensatory pattern of weight-bearing asymmetry and subsequent lower extremity weakness (Mizner & Snyder-Mackler, 2005). As STS has been indicated as sufficient stimulus to maintain lower extremity strength (Kotake *et al.*, 1993), asymmetry in STS performance could be a potential detriment to strength in individuals with osteoarthritis.

As hypothesized, bilateral hip abduction strength measures were correlated with affected hip extension moments and with weight-bearing symmetry in STS, which may indicate that hip abductors play a major a role in stabilization during STS. Hip abductors may also be integral in controlling joint loading on the affected limb as individuals maintain a preferential load on the unaffected limb despite initial posture. Significant differences were noted between the affected and unaffected limbs within foot placements (Figure 4.3). In the symmetric and affected stagger conditions, the unaffected limb produced more hip abduction moment which may imply a role in balance stabilization and/or weight-bearing protection of the affected side. Interestingly, in the unaffected stagger condition, there was not a difference between affected and unaffected limbs in terms of the amount of hip abduction moment produced. Therefore, it suggests that when the unaffected limb is in the posterior position, the requirements for balance stabilization and weight-bearing protections are reduced.

In evaluating the stair descent task for both groups, individuals increased their speed of descent as they reduced their stance time on each subsequent step. The behavior was similar in many instances on the first and second steps; however, the third step included the progression from the last step to the ground. As there was not the constraint of step depth with the third and final step, movement behavior and joint loading were affected. Differences in balance and joint loading measures suggest that multiple steps should be monitored when assessing stair descent.

In terms of balance during stair descent, participants appeared to be more stable as indicated by mediolateral COP velocity on the ground (3$^{rd}$ step) compared to the staircase steps (1st and 2nd steps). This may be representative of a more “gait-like” pattern as individuals return to level-ground ambulation without the constraint of tread depth and additional descent. In addition, individuals who experience bilateral knee pain symptoms may be at more risk during stair descent due to higher average peak mediolateral COP velocities. This result suggests that the bilaterally-affected individuals rely on more rapid balance adjustments relative to the unilaterally-affected group. Other authors have suggested a faster sway on stairs in healthy elderly is indicative of a greater balance challenge (Lee & Chou, 2007). Affected hip abduction strength correlated with mediolateral COP velocity on stair descent during the weight-bearing steps. This supports the hypothesis that hip abductors on the affected side were associated with the peak mediolateral COP velocity shifts in single-limb support during stair descent.

There were no significant differences in joint moments between stair descent conditions as participants were required to perform the task in a step-over-step pattern without the use of upper extremity support as a compensatory technique. Many individuals with reduced knee range of motion or low knee extensor strength use upper extremity support (Tiedeman *et al.*, 2007) or other compensations to aid stair negotiation. Stair descent difficulty has been correlated with limitations in activities of daily living such as bathing, dressing and in-home ambulation for non-disabled older adults (Verghese *et al.*, 2008). Therefore, future protocols should address compensatory techniques for stair descent including upper extremity support, altered descent patterns and adjusted speeds.

The lower extremity strength of the participants appeared to be less than normative strength measures for healthy older adults without knee pain, although testing positions and joint motions assessed tend to vary between studies (Bohannon, 1997; Andrews *et al*., 1996; Bäckman *et al.*, 1995). Contrary to hypothesized strength deficits, no differences were noted between affected and unaffected lower extremities in terms of strength assessments. It is possible that the number of individuals with bilateral knee symptoms confounded the strength measures as “affected” and “unaffected” terminology may have been insufficient to differentiate the limbs for participants with low-level symptoms.

Along these lines, the classification and severity of symptoms for these individuals may be inadequate to address strength issues. As the nature of knee pain for the participants in this investigation was not specified beyond "physician-diagnosed osteoarthritis," there may have been inadequate categorization of the type of knee pain or osteoarthritis. Moreover, these participants may not exhibit sufficient weakness in lower extremity strength to be at a threshold for affecting functional performance or stability. Others have also noted that individuals with low-level knee osteoarthritis may have sufficiently strong hip abductor musculature to meet the control requirements during gait to avoid excessive knee varus moments (Mundermann *et al*., 2005). It is likely that task requirements for this investigation, namely stair descent and STS without upper extremity usage, may be threshold strength indicators for individuals with knee pain to achieve independent mobility.

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# CHAPTER 5: GENERAL CONCLUSIONS

As previously noted, the goal of this dissertation research was to further understand the relationships between postural stability, lower extremity strength, and functional performance for individuals with knee osteoarthritis. This was accomplished through a series of investigations assessing the functional movements of sit-to-stand (STS) and stair descent for healthy older adults, individuals with symptoms of knee osteoarthritis and individuals following total knee arthroplasty (TKA).

The primary aim of the first project was to establish standardized measures for assessment of STS for subsequent projects. In published literature, researchers have utilized a variety of different determinants for the initiation, seat-off, and termination components of STS (Schenkman *et al*., 1996; Pai & Rogers, 1990; Shepherd & Gentile, 1994; Mourey *et al.,* 2000). Previous work has noted the need to reference movement to seat-off for timing relationships between the different STS events (Hanke *et al*., 1995). Other researchers have evaluated portions of the STS movement phases using symmetrical initial foot placements and participant-accommodated seat heights (seat set at equal percentage of participants' heights). This first dissertation project involved motion analysis to identify kinematic determinants of movement and multiple force platforms to detect kinetic indicators of movement. This project evaluated STS using a standard height chair and investigated both symmetrical and asymmetrical foot placements. The findings from this study demonstrated that kinetic and kinematic indicators of STS movement may be standardized for healthy individuals of various ages without regard to initial positioning.

The principal purpose of the second project was to evaluate the STS movement for individuals with TKA compared to healthy older adults. This study utilized motion analysis and kinetic assessment to evaluate joint moments, center of pressure (COP) parameters and weight-bearing asymmetry. Individuals post-TKA compensate for residual physical deficits by utilizing upper extremity support and altered joint loading to perform STS successfully. This project was the first to evaluate asymmetrical (staggered) foot placements for STS in an orthopedic population, as previous work was completed with individuals with neurological deficits (Brunt *et al*., 2002). Previous authors noted TKA individuals compensated for quadriceps weakness through apparent weight-bearing asymmetry, which reduced the affected side hip and knee moments (Mizner & Snyder-Mackler, 2005). The dissertation project indicated that the affected side hip joint may actually increase its contribution to the overall support moment due to the altered foot positioning and weight-bearing asymmetry. This project also evaluated the role of upper extremity support in various postures that may be utilized by a clinical population. Other work has noted that upper extremity assistance reduces joint moments, and this project further identified the staggered-hand position as providing a moderate level of assistance without apparent postural stability issues.

The purpose of the final project was to correlate lower extremity strength with kinetic/kinematic parameters of postural stability while performing functional activities. Other researchers have identified differences in stair descent techniques (step-over-step, step-by-step, posterior descent) in younger populations by evaluating center of mass movement, joint moments, speed of descent, and stance time. Previous research demonstrated asymmetry on stairs for older adults and individuals with osteoarthritis, identifying faster loading on an involved limb and additional time in single limb support on the uninvolved limb (Liikavainio *et al*., 2007). In the final study presented in this dissertation, altered joint loading and compensatory weight-bearing asymmetry allowed individuals with mild to moderate knee pain to perform STS and stair descent yet maintain postural stability.

Cumulatively, older adults with symptoms of osteoarthritis appear to compensate for physical limitations of reduced joint mobility, strength deficits, and pain by adjusting joint loads and altering patterns for movement. As we have identified issues with weight-bearing asymmetry in both healthy and osteoarthritic populations, questions arise regarding the potential detriment of a chronic weight-bearing asymmetry on functional performance and long-term postural stability. It appears that loading asymmetry may produce a gradual strength deficit and potentially lead to postural stability concerns as an individual may approach their stability boundary with movement and increase their risk of falling. The interaction of these components continues to require additional understanding.

It is important that the outcomes from these investigations lead towards justification for physical therapy interventions. The findings from this dissertation research have clinical relevance because the movements investigated are functional tasks that affect daily mobility. From this work, clinicians may appreciate the necessity of documenting movement strategy, objective measures of performance, and compensatory movement patterns used by individuals with physical limitations. In the case of STS, accommodations may include altered foot placement, stance width, seat height, and upper extremity/external assistance. Movement compensation may be an increase in hip flexion/extension, asymmetrical weight-bearing or greater unaffected limb muscular effort (joint moments) to account for other deficits. For stair descent, it may be necessary for individuals to use a different movement pattern (such as step-over-step or step-by-step) or invoke a compensatory technique (upper extremity support, decreasing speed, increased single-leg support on the unaffected limb, or increased step width) to accomplish the task.

From a clinical perspective, the type of strategy used for movement performance may also indicate the level of motor control an individual possesses. From the current investigations, it appears that postural stability may serve as a control parameter for functional movements of sit-to-stand and stair descent. It is possible that strength capacities and pain levels interact to determine the most efficient and effective strategy for task completion. As a rehabilitation professional, it may be important to note if strategies and therapy interventions are effective not only from the standpoint of improving the ability to complete a task but also from the point of improving the quality of life. For example, a successful intervention may produce both an improvement in physical impairment and cause a shift in an individual's preferred strategy for performance. Previous research has indicated that physical limitations alone did not explain the STS strategy selected (Scarborough *et al*., 2007). Therefore, additional research may be warranted to investigate how individuals prioritize for strategy selection. If pain avoidance drives the choice of movement strategy, non-optimal techniques that result in loading asymmetry, unaffected joint degeneration, and increased risk of falls may be adopted. It should be a clinical research priority to determine strengthening and/or balance interventions that enable older adults to utilize more efficient movement strategies without exacerbating pain symptoms.

In terms of future work, this research continues to provide additional questions for exploration. From a standardization perspective of STS, I plan to continue analyzing the STS movement to determine appropriate thresholds for automated detection of movement indicators. There is also potential to continue exploring STS strategies in various populations to determine if it is possible to use the STS motion as a screening or assessment tool in evaluating functional status, pain levels, or pathology. It would be beneficial to correlate the laboratory measures of STS and stair descent into a functional mobility assessment for fall risk in orthopedic clients that could be shifted into a clinical setting. Finally, additional work on stability measures (COP parameters and time-to-contact index) for both STS and stair negotiation may increase the depth of understanding of dynamic stability control.

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APPENDIX A: CHAPTER 2 EXPANDED RESULTS

**Table 2.3: STS Initiation Indicators – RMSE averaged over foot placement in seconds** (bold font highlights the lowest RMSE of the kinetic indicators and the three lowest values for the kinematic indicators)

FN = Foot Neutral Placement  RS = Right Staggered Placement
FB = Foot Back Placement  LS = Left Staggered Placement

| **Indicators** | FN | FB | RS | LS | **Older Average** (s) | FN | FB | RS | LS | **Young Average** (s) |
|---|---|---|---|---|---|---|---|---|---|---|
| **Kinetic Indicators** | | | | | | | | | | |
| VGRF | 0.121 | 0.068 | 0.096 | 0.072 | **0.089** | 0.115 | 0.062 | 0.090 | 0.137 | **0.101** |
| AGRF | 0.108 | 0.155 | 0.218 | 0.162 | 0.161 | 0.241 | 0.223 | 0.209 | 0.232 | 0.226 |
| COP Anteroposterior | 0.194 | 0.143 | 0.253 | 0.178 | 0.192 | 0.115 | 0.174 | 0.200 | 0.190 | 0.170 |
| COP Mediolateral | 0.208 | 0.169 | 0.191 | 0.211 | 0.195 | 0.261 | 0.246 | 0.393 | 0.184 | 0.271 |
| **Kinematic Indicators** | | | | | | | | | | |
| Trunk Angular Velocity | 0.166 | 0.242 | 0.296 | 0.177 | **0.220** | 0.128 | 0.171 | 0.166 | 0.140 | **0.151** |
| Shoulder Marker | 0.176 | 0.183 | 0.222 | 0.198 | **0.195** | 0.183 | 0.187 | 0.179 | 0.171 | **0.180** |
| Hip Angle | 0.248 | 0.236 | 0.226 | 0.198 | **0.227** | 0.194 | 0.190 | 0.191 | 0.174 | **0.187** |
| Body COM | 0.197 | 0.275 | 0.237 | 0.215 | 0.231 | 0.196 | 0.486 | 0.193 | 0.187 | 0.266 |
| Hip Marker Horizontal | 0.400 | 0.424 | 0.430 | 0.389 | 0.411 | 0.411 | 0.342 | 0.353 | 0.443 | 0.388 |
| Hip Velocity Vertical | 0.399 | 0.400 | 0.414 | 0.339 | 0.388 | 0.417 | 0.480 | 0.341 | 0.426 | 0.416 |
| Hip Marker Vertical | 0.428 | 0.423 | 0.440 | 0.375 | 0.416 | 0.437 | 0.428 | 0.362 | 0.460 | 0.422 |
| Hip Velocity Horizontal | 0.277 | 0.319 | 0.709 | 0.417 | 0.431 | 0.738 | 0.416 | 0.271 | 0.334 | 0.440 |

**Table 2.4 STS Momentum Indicators – RMSE averaged over foot placement in seconds** (bold font highlights the lowest RMSE of the kinetic indicators and the three lowest values for the kinematic indicators)

FN = Foot Neutral Placement RS = Right Staggered Placement
FB = Foot Back Placement LS = Left Staggered Placement

| **Indicators** | FN | FB | RS | LS | **Older Average** (s) | FN | FB | RS | LS | **Young Average** (s) |
|---|---|---|---|---|---|---|---|---|---|---|
| **Kinetic Indicators** | | | | | | | | | | |
| VGRF | 0.059 | 0.064 | 0.065 | 0.063 | **0.063** | 0.070 | 0.069 | 0.067 | 0.067 | **0.068** |
| AGRF | 0.061 | 0.074 | 0.064 | 0.061 | 0.065 | 0.067 | 0.074 | 0.064 | 0.068 | **0.068** |
| COP Anteroposterior | 0.514 | 0.453 | 0.537 | 0.508 | 0.503 | 0.486 | 0.438 | 0.437 | 0.504 | 0.466 |
| COP Mediolateral | 0.508 | 0.506 | 0.505 | 0.501 | 0.505 | 0.604 | 0.532 | 0.633 | 0.552 | 0.580 |
| **Kinematic Indicators** | | | | | | | | | | |
| Trunk Angular Velocity | 0.040 | 0.301 | 0.304 | 0.050 | **0.174** | 0.050 | 0.054 | 0.052 | 0.051 | **0.051** |
| Hip Marker Horizontal | 0.169 | 0.360 | 0.353 | 0.169 | **0.263** | 0.193 | 0.176 | 0.148 | 0.182 | **0.175** |
| Hip Angle | 0.158 | 0.176 | 0.188 | 0.161 | **0.171** | 0.181 | 0.205 | 0.194 | 0.187 | **0.192** |
| Hip Velocity Vertical | 0.374 | 0.352 | 0.363 | 0.169 | 0.315 | 0.135 | 0.365 | 0.172 | 0.141 | 0.203 |
| Hip Velocity Horizontal | 0.199 | 0.338 | 0.358 | 0.205 | 0.275 | 0.240 | 0.212 | 0.179 | 0.201 | 0.208 |
| Hip Marker Vertical | 0.296 | 0.279 | 0.322 | 0.283 | 0.295 | 0.263 | 0.245 | 0.251 | 0.265 | 0.256 |
| Body COM | 0.317 | 0.414 | 0.335 | 0.297 | 0.341 | 0.282 | 0.258 | 0.270 | 0.287 | 0.274 |
| Shoulder Marker | 0.356 | 0.337 | 0.380 | 0.333 | 0.352 | 0.322 | 0.299 | 0.310 | 0.323 | 0.314 |

**Table 2.5 STS Ascension Indicators – RMSE averaged over foot placement in seconds** (bold font highlights the lowest RMSE of the kinetic indicators and the three lowest values for the kinematic indicators)

FN = Foot Neutral Placement RS = Right Staggered Placement
FB = Foot Back Placement LS = Left Staggered Placement

| **Indicators** | FN | FB | RS | LS | **Older Average** (s) | FN | FB | RS | LS | **Young Average** (s) |
|---|---|---|---|---|---|---|---|---|---|---|
| **Kinetic Indicators** | | | | | | | | | | |
| AGRF | 0.059 | 0.252 | 0.264 | 0.278 | **0.213** | 0.257 | 0.269 | 0.254 | 0.277 | **0.264** |
| COP Anteroposterior | 0.752 | 0.642 | 0.724 | 0.740 | 0.714 | 0.675 | 0.633 | 0.619 | 0.718 | 0.661 |
| COP Mediolateral | 0.745 | 0.689 | 0.693 | 0.717 | 0.711 | 0.777 | 0.718 | 0.795 | 0.768 | 0.764 |
| **Kinematic Indicators** | | | | | | | | | | |
| Hip Marker Vertical | 0.069 | 0.127 | 0.186 | 0.084 | **0.117** | 0.129 | 0.044 | 0.119 | 0.052 | **0.086** |
| Hip Angle | 0.089 | 0.102 | 0.164 | 0.079 | **0.109** | 0.132 | 0.088 | 0.088 | 0.130 | **0.109** |
| Shoulder Marker | 0.137 | 0.178 | 0.238 | 0.132 | **0.171** | 0.183 | 0.099 | 0.150 | 0.112 | 0.136 |
| Body COM | 0.099 | 0.369 | 0.205 | 0.104 | 0.194 | 0.149 | 0.057 | 0.124 | 0.079 | **0.102** |
| Hip Marker Horizontal | 0.164 | 0.207 | 0.253 | 0.165 | 0.197 | 0.239 | 0.191 | 0.200 | 0.163 | 0.198 |
| Trunk Angular Velocity | 0.249 | 0.387 | 0.413 | 0.219 | 0.317 | 0.208 | 0.205 | 0.190 | 0.211 | 0.203 |
| Hip Velocity Vertical | 0.530 | 0.475 | 0.510 | 0.367 | 0.470 | 0.297 | 0.489 | 0.367 | 0.308 | 0.365 |
| Hip Velocity Horizontal | 0.413 | 0.485 | 0.531 | 0.414 | 0.461 | 0.429 | 0.418 | 0.406 | 0.390 | 0.411 |

**Table 2.6 STS Termination Indicators – RMSE averaged over foot placement in seconds** (bold font highlights the lowest RMSE of the kinetic indicators and the three lowest values for the kinematic indicators)

FN = Foot Neutral Placement RS = Right Staggered Placement
FB = Foot Back Placement LS = Left Staggered Placement

| **Indicators** | FN | FB | RS | LS | **Older Average** (s) | FN | FB | RS | LS | **Young Average** (s) |
|---|---|---|---|---|---|---|---|---|---|---|
| **Kinetic Indicators** | | | | | | | | | | |
| AGRF | 0.283 | 0.345 | 0.342 | 0.327 | **0.324** | 0.342 | 0.352 | 0.282 | 0.281 | **0.314** |
| COP Anteroposterior | 1.605 | 1.335 | 2.350 | 1.941 | 1.808 | 1.829 | 1.430 | 1.360 | 2.341 | 1.740 |
| COP Mediolateral | 1.917 | 1.291 | 2.127 | 1.487 | 1.705 | 1.193 | 0.523 | 1.407 | 1.697 | 1.205 |
| **Kinematic Indicators** | | | | | | | | | | |
| Hip Angle | 0.248 | 0.095 | 0.130 | 0.086 | **0.140** | 0.104 | 0.079 | 0.103 | 0.078 | **0.091** |
| Hip Marker Horizontal | 0.150 | 0.178 | 0.139 | 0.107 | **0.143** | 0.184 | 0.133 | 0.173 | 0.149 | **0.160** |
| Trunk Angular Velocity | 0.142 | 0.284 | 0.205 | 0.129 | **0.190** | 0.118 | 0.132 | 0.276 | 0.106 | **0.158** |
| Shoulder Marker | 0.208 | 0.174 | 0.203 | 0.181 | 0.191 | 0.226 | 0.210 | 0.240 | 0.205 | 0.220 |
| Body COM | 0.269 | 0.241 | 0.257 | 0.235 | 0.250 | 0.277 | 0.269 | 0.292 | 0.252 | 0.272 |
| Hip Marker Vertical | 0.306 | 0.280 | 0.290 | 0.270 | 0.286 | 0.308 | 0.300 | 0.327 | 0.286 | 0.305 |
| Hip Velocity Vertical | 0.913 | 0.185 | 0.133 | 1.930 | 0.790 | 1.427 | 0.118 | 0.287 | 0.114 | 0.487 |
| Hip Velocity Horizontal | 1.108 | 0.637 | 0.224 | 1.800 | 0.943 | 0.685 | 0.258 | 0.395 | 1.394 | 0.683 |

# APPENDIX B: CHAPTER 3 EXPANDED RESULTS

**Table 3.3: Differences by Groups – Expanded** (* = statistically significant difference between groups at p<0.001)

| Variables | TKA | Control |
|---|---|---|
| **Range of Motion (Degrees)** | | |
| Dominant Ankle Dorsiflexion ROM (p<0.001) | 22.9 ± 5.5* | 28.7 ± 4.2 |
| Non-dominant Ankle Dorsiflexion ROM (p<0.001) | 22.0 ± 5.3* | 25.2 ± 4.3 |
| Dominant Knee Flexion ROM (p<0.001) | 91.5 ± 4.9* | 96.2 ± 6.5 |
| Non-dominant Knee Flexion ROM (p<0.001) | 88.8 ± 5.6* | 92.8 ± 6.2 |
| Dominant Hip Flexion ROM (p<0.001) | 103.1 ± 11.3* | 95.0 ± 8.8 |
| Non-dominant Hip Flexion ROM (p<0.001) | 104.4 ± 11.8* | 97.3 ± 9.0 |
| **Joint Moments (Nm)** | | |
| Dominant Ankle Dorsiflexion Moment (p=0.001) | 2.7±4.8* | 0.7±3.2 |
| Non-dominant Ankle Dorsiflexion Moment (p<0.001) | 3.7±4.4* | 1.2±2.9 |
| Dominant Knee Extension Moment (p=0.214) | 51.3 ± 17.7 | 48.5± 17.0 |
| Non-dominant Knee Extension Moment (p=0.274) | 40.2 ± 9.8 | 42.5 ± 18.5 |
| Dominant Hip Extension Moment (p=0.305) | 86.8 ± 32.9 | 83.2 ± 29.3 |
| Non-dominant Hip Extension Moment (p=0.001) | 114.6±37.0* | 91.3±36.5 |
| Dominant Support Moment (p<0.001) | 127.9±25.4* | 149.9±37.3 |
| Non-dominant Support Moment (p=0.014) | 157.1±35.2 | 138.5±42.0 |
| Total Support Moment (p=0.0249) | 285.5 ± 77.8 | 279.1 ± 51.9 |
| **Other Variables** | | |
| STS Times (p=0.006) | 1.7 ± 0.3 s | 1.5 ± 0.3 s |
| Max COP Velocity-Anteroposterior (p<0.001) | 0.46±0.15 m/s* | 0.68±0.28 m/s |
| Max COP Velocity-Mediolateral (p<0.001) | 0.52±0.23 m/s* | 0.88±0.57 m/s |
| Max COP Velocity-Combined (p<0.001) | 0.65±0.24 m/s* | 1.12±0.61 m/s |
| Weight-bearing Asymmetry (p=0.021) | 19.2 ± 11.3 % | 23.1 ± 15.8 % |
| Upper Extremity Force (p<0.001) | 219.0 ± 290.2 N* | 112.7 ± 180.7 N |
| Upper Extremity Assistance (p<0.001) | 27.4±35.3 %BW * | 15.7± 26.4 %BW |
| BOS width (p=0.011) | 0.371±0.10 m | 0.35±0.12 m |
| Mediolateral BOS Percentage (p<0.001) | 35±14 % * | 49±33 % |

**Table 3.4: Differences by Foot Placement Condition - Expanded** (** = differs from other dominant placements, p<0.001; * = differs from other non-dominant placements, p<0.001)

| Variables | Foot-Neutral | Foot-Back | Dominant Stagger | Tukey Follow-Up Testing |
|---|---|---|---|---|
| **Range of Motion (Degrees)** | | | | |
| Dominant Ankle Dorsiflexion ROM (p=0.001) | 22.9 ± 4.6 | 26.0 ±5.1 | 28.7 ± 5.4 | All 3 differ (p<0.005) |
| Non-dominant Ankle Dorsiflexion ROM (p<0.001) | 23.7±4.3 | 26.3±4.1 | 20.6±5.1 | All 3 differ (p<0.005) |
| Dominant Knee Flexion ROM (p<0.001) | 88.9±4.6** | 95.8±4.8 | 96.9±5.9 | 1 vs. 2 (p<0.001); 1 vs. 3 (p<0.001) |
| Non-dominant Knee Flexion ROM (p<0.001) | 89.3±4.4 | 95.8±4.5 * | 87.4±6.2 | 1 vs. 2 (p<0.001); 2 vs. 3 (p<0.001) |
| Dominant Hip Flexion ROM (p=0.004) | 102.7±10.5 | 96.7±10.4 | 97.5±10.8 | 1 vs. 2(p<0.01); 2 vs. 3 (p=0.039) |
| Non-dominant Hip Flexion ROM (p=0.015) | 103.1±10.8 | 97.4±10.7 | 101.7±11.0 | 2 vs. 1 & 3 (p<0.05) |
| **Joint Moments (Nm)** | | | | |
| Dominant Ankle Dorsiflexion Moment (p<0.001) | 3.7±3.7 ** | 0.5±3.6 | 0.8±4.5 | 1 vs. 2, 1 vs. 3 (p<0.001) |
| Non-dominant Ankle Dorsiflexion Moment (p=0.002) | 3.5±3.7 | 0.9±3.9 | 2.9±3.6 | 1 vs. 2 (p=0.001), 2 vs. 3 (p=0.012) |
| Dominant Knee Extension Moment (p=0.001) | 43.7±14.2 | 48.7±16.2 | 57.2±18.9** | 1 vs. 3 (p<0.001), 2 vs. 3 (p=0.024) |
| Non-dominant Knee Extension Moment (p<0.001) | 43.2±12.4 | 47.0±13.9 | 33.7±14.8* | 1 vs. 3 (p=0.001), 2 vs. 3 (p<0.001) |
| Dominant Hip Extension Moment (p=0.038) | 94.3±30.6 | 81.5±28.3 | 79.2±31.4 | 1 vs. 3 (p=0.025) |
| Non-dominant Hip Extension Moment (p=0.049) | 105.2±38.9 | 92.6±34.7 | 110.4±39.9 | 2 vs. 3 (p=0.025) |
| Dominant Support Moment (p=0.450) | 137.4 ± 31.9 | 136.5 ± 34.5 | 143.4 ± 35.0 | |
| Non-dominant Support Moment (p=0.864) | 147.6 ± 38.1 | 145.9 ± 37.2 | 149.3 ± 44.3 | |
| Total Support Moment (p=0.753) | 280.8 ± 65.2 | 279.1 ± 67.0 | 287.1 ± 67.5 | |
| **Other Variables** | | | | |
| STS Times in seconds (p=0.258) | 1.6 ± 0.4 | 1.6 ± 0.2 | 1.6 ± 0.2 | |
| Max COP Velocity-Anteroposterior in m/s (p=0.563) | 0.60±0.27 | 0.53±0.23 | 0.58±0.25 | |
| Max COP Velocity-Mediolateral in m/s (p=0.774) | 0.73 ± 0.49 | 0.64 ± 0.45 | 0.73 ± 0.48 | |
| Max COP Velocity-Combined in m/s (p=0.554) | 0.91 ± 0.53 | 0.82 ± 0.51 | 0.93 ± 0.52 | |
| Weight-bearing Asymmetry (%) (p=0.044) | 20.0 ± 14.3 | 18.9 ± 12.5 | 25.3 ± 14.0 | 2 vs. 3 (p=0.035) |
| Upper Extremity Force in N (p=0.868) | 172.9± 261.5 | 160.5±235.7 | 163.6±246.1 | |
| Upper Extremity Assistance (%BW) (p<0.870) | 22.4 ± 33.5 | 20.8 ± 30.2 | 21.2 ± 31.4 | |
| BOS width in m (p=0.910) | 0.36 ± 0.11 | 0.36 ± 0.11 | 0.36 ± 0.11 | |
| Mediolateral BOS Percentage (%) (p=0.857) | 43.1 ± 27.1 | 40.4 ± 25.7 | 42.3 ± 25.6 | |

**Table 3.5: Differences by Hand Placement Condition – Expanded** (** = no hands condition greater than hands-separated or hands on bench, p<0.01)

| Joint Motion | No Hands | Hands-Separated | Hands-on-Bench | Tukey Follow-Up Testing |
|---|---|---|---|---|
| **Range of Motion (Degrees)** | | | | |
| Dominant Ankle Dorsiflexion ROM (p=0.850) | 26.0 ± 5.6 | 25.5 ± 5.7 | 26.0 ± 6.0 | |
| Non-dominant Ankle Dorsiflexion ROM (p=0.230) | 23.4 ± 4.8 | 24.5 ± 5.2 | 23.0 ± 5.4 | |
| Dominant Knee Flexion ROM (p=0.874) | 93.7 ± 6.2 | 94.2 ±6.0 | 94.0 ±6.7 | |
| Non-dominant Knee Flexion ROM (p=0.161) | 90.6 ± 6.2 | 92.0 ± 6.1 | 90.3 ± 6.4 | |
| Dominant Hip Flexion ROM (p<0.001) | 102.1±10.2 ** | 93.4±9.6 | 98.3±11.1 | 1 vs. 2 (p<0.001); 1 vs. 3 (p=0.037) |
| Non-dominant Hip Flexion ROM (p<0.001) | 104.0±10.1** | 95.5±10.9 | 99.5±11.0 | 1 vs. 2,3 (p<0.02); 2 vs. 3 (p=0.05) |
| **Joint Moments (Nm)** | | | | |
| Dominant Ankle Dorsiflexion Moment (p=0.151) | 1.65 ± 4.30 | 2.60 ± 3.69 | 0.87 ± 4.38 | |
| Non-dominant Ankle Dorsiflexion Moment (p=0.212) | 2.73 ± 3.51 | 2.78 ±5.02 | 1.56 ± 3.30 | |
| Dominant Knee Extension Moment (p=0.425) | 50.2 ± 16.8 | 52.1 ± 16.5 | 47.4 ± 19.0 | |
| Non-dominant Knee Extension Moment (p=0.075) | 42.6 ± 14.5 | 43.3 ± 16.4 | 37.1 ± 12.9 | |
| Dominant Hip Extension Moment (p=0.020) | 91.6±29.3 | 77.8±28.9 | 79.4±32.8 | 1 vs. 2 (p=0.046) |
| Non-dominant Hip Extension Moment (p=0.011) | 111.4±39.4 | 93.6±38.1 | 95.3±33.8 | 1 vs. 2 (p=0.027); 1 vs. 3 (p=0.045) |
| Dominant Support Moment (p=0.001) | 148.5±29.1** | 128.2±34.1 | 131.7±37.3 | 1 vs. 2 (p=0.002); 1 vs. 3 (p=0.012) |
| Non-dominant Support Moment (p=0.006) | 157.4±38.7 | 139.5±43.1 | 136.9±34.6 | 1 vs. 2 (p=0.041); 1 vs. 3 (p=0.013) |
| Total Support Moment (p=0.001) | 302.6±58.4** | 261.7±70.8 | 263.7±65.7 | 1 vs. 2,3 (p<0.01) |
| **Other Variables** | | | | |
| STS Times in Seconds (p=0.084) | 1.6 ± 0.2 | 1.5 ± 0.2 | 1.7 ± 0.4 | |
| Max COP Velocity-AP in m/s (p=0.849) | 0.58±0.25 | 0.57±0.23 | 0.56±0.28 | |
| Max COP Velocity-ML in m/s (p=0.192) | 0.73 ± 0.47 | 0.75 ±0.48 | 0.60± 0.46 | |
| Max COP Velocity-Combined in m/s (p=0.720) | 0.90 ± 0.49 | 0.91 ± 0.49 | 0.84 ± 0.60 | |
| Weight-bearing Asymmetry (%) (p=0.032) | 20.4 ± 12.1 | 25.9 ± 16.7 | 18.7 ± 13.3 | 2 vs. 3 (p=0.039) |
| Upper Extremity Force in Newtons (p<0.001) | 0.282±6.87 | 253.173.1 | 392.6±307.7 | All 3 differ (p<0.001) |
| Upper Extremity Assistance (%BW) (p<0.001) | 0.0±0.95 | 33.0±22.4 | 50.7±38.6 | All 3 differ (p<0.001) |
| BOS width in meters (p<0.001) | 0.302±0.05 | 0.303±0.05 | 0.54±0.000 | 1,2 vs. 3(p<0.001) |
| Mediolateral BOS % (p<0.001) | 46.1±22.7 | 54.5±30.7 | 21.1±11.1 | 1,2 vs. 3(p<0.001) |

APPENDIX C: CHAPTER 4 EXPANDED RESULTS

**Table 4.3: Differences by Knee Symptom Group - Expanded** (**Statistical significance at p<0.01)

| Variables | Bilateral Symptoms | Unilateral Symptoms | Results |
|---|---|---|---|
| Weight-bearing Symmetry | 89 ± 10% | 80 ± 13% | p=0.002** |
| STS Time | 1.15±0.15 s | 1.54±0.81s | p=0.013 |
| Mediolateral COP Velocity | 4.95 ± 2.57 cm/s | 4.84 ± 2.64 cm/s | p=0.887 |
| Affected Knee Extension Moment | 0.66 ± 010 Nm/kg | 0.52 ± 0.12 Nm/kg | p<0.001** |
| Unaffected Knee Extension Moment | 0.70 ± 0.09 Nm/kg | 0.71 ±0.17 Nm/kg | p=0.636 |
| Affected Hip Extension Moment | 0.61 ± 0.18 Nm/kg | 0.48 ± 0.12 Nm/kg | p<0.001** |
| Unaffected Hip Extension Moment | 0.64 ± 0.17 Nm/kg | 0.48 ± 0.14 Nm/kg | p<0.001** |
| Affected Hip Abduction Moment | 0.15±0.09 Nm/kg | 0.17±0.08 Nm/kg | p=0.285 |
| Unaffected Hip Abduction Moment | 0.26±0.13 Nm/kg | 0.20±0.08 Nm/kg | p=0.024 |

**Table 4.5: Differences by Foot Placement Condition - Expanded** (**Statistical significance at p<0.01)

| Variable | Affected Stagger (1) | Unaffected Stagger (2) | Symmetric (3) | Tukey Follow-Up Testing |
|---|---|---|---|---|
| Weight-bearing Symmetry (p=0.998) | 85.5 ± 11.66 % | 85.07 ±11.98 % | 85.23 ± 12.87% | |
| STS Time (p=0.888) | 1.31 ± 0.54 s | 1.40 ± 0.75 s | 1.41 ± 0.68 s | |
| Mediolateral COP Velocity (p=0.027) | 4.89 ± 2.63 cm/s | 5.87 ± 2.97 cm/s | 3.91 ± 1.67 cm/s | 2 vs. 3 (p=0.030) |
| **Joint Moments (Nm/kg)** | | | | |
| Affected Knee Extension Moment (p=0.009) ** | 0.65 ± 0.11 | 0.56 ± 0.15 | 0.59 ± 0.11 | 1 vs. 2 (p=0.009) |
| Unaffected Knee Extension Moment (p=0.014) | 0.65 ± 0.13 | 0.76 ± 0.08 | 0.70 ± 0.15 | 1 vs. 2 (p=0.012) |
| Affected Hip Extension Moment (p=0.084) | 0.50 ± 0.17 | 0.59 ± 0.14 | 0.57 ± 0.19 | |
| Unaffected Hip Extension Moment (p=0.005) ** | 0.63 ± 0.15 | 0.49 ± 0.19 | 0.59 ± 0.18 | 1 vs. 2 (p=0.005); 2 vs. 3 (p=0.056) |
| Affected Hip Abduction Moment (p= 0.158) | 0.15 ± 0.08 | 0.19 ± 0.11 | 0.14 ± 0.06 | |
| Unaffected Hip Abduction Moment (p=0.387) | 0.25 ± 0.11 | 0.22 ± 0.10 | 0.21 ± 0.12 | |

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