THE EVALUATION OF VERTICAL POLE CONFIGURATION AND LOCATION ON ASSISTING
THE SIT-TO-STAND MOVEMENT IN OLDER ADULTS WITH MOBILITY LIMITATIONS

by

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Abstract

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Grab bars and transfer poles are common sit-to-stand aids for older adults with mobility limitations. This study investigates differences in kinematics and kinetics in the lower limbs across different transfer pole configurations and positions. Configurations tested are a single pole, two poles and a pole with a horizontal grab-bar at near and far positions. Three-dimensional biomechanics were used to calculate kinetics and kinematics of the lower extremities. Forces were also recorded from the pole. This study found horizontal pole forces were an effective replacement for trunk generated horizontal momentum. Reduced vertical pole forces were applied by participants using the ‘far’ poles which resulted in increases in non-dominant hip moments when using the single and double pole configurations. Horizontal pole use introduced non-dominant directed lateral COM trajectory throughout the movement. This coincided with increased horizontal forces in the dominant foot to balance moments about COM in the transverse plane as a motor control strategy.

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Chapter 1

Introduction

1.1 Motivation

The sit-to-stand (STS) transfer is described as one of the most biomechanically challenging functional tasks [1]. It has been reported that standing up from a seated position is associated with falls in older adults [2] and that 8% of community-dwelling older adults over 65 years of age show problems in rising from a chair or bed [3]. The movement requires peak joint moments greater than other movements such as stair climbing [4] and yields peak hip joint contact pressure higher than jogging or jumping [1]. Standing up and sitting down are important and common activities of daily life [5]. The STS transfer is a determinant of independence as it is a precursor to many activities of daily living (ADLs) and instrumental activities of daily living (IADLs) such as toileting, dressing, food preparation, and shopping.

There are two biomechanical requirements for slowly rising from a chair. The first is bringing the horizontal component of total body center-of-mass (COM) to within the standing base-of-support (BOS), defined as the area encompassing the feet during standing. The second is producing the joint torques required to rise [4,6,7]. The main challenges with the transfer are categorized as effort or safety related. Challenges with safety are related to stability and balance and originate from the inability to bring the COM into the BOS when transitioning from the seated posture into the standing position. Standing up is the second highest event leading to a fall event in community-dwelling older adults, next to walking [8]. With respect to effort, the challenge is having the muscle strength required to produce sufficient joint torques in the lower extremities to successfully move the COM vertically upwards into the standing position.

Assistive devices such as grab-bars enable the use of the upper-extremities to aid the STS transfer [9–13]. Upper-extremity assistance, increases the success of a STS task in individuals without adequate lower

extremity strength or coordination [14]. In addition, they increase BOS, thereby reducing the risk of suffering a fall when standing up. Grab-bars and other assistive supports are an effective means of preserving STS ability in those with a limited functional capacity. The transfer pole is a common grab-bar used in the home to aid in STS. Transfer poles are vertically oriented poles that pressure fit between the ceiling and floor (for ceiling heights up to 10 feet) and may be installed without requiring permanent home modification. They are typically placed beside a bed, chair or toilet and provide support for performing activities of daily living including standing up, sitting down and toileting. They are designed for a mobility-limited population that frequently includes individuals who have difficulty performing unassisted STS. Transfer poles are commonly sold as simple vertical poles, as shown in Figure 1.1. Some are also sold with flip-down horizontal bars that allow for a wider two-handed grip (see Figure 1.1); and it is also possible to place transfer poles on either side of the user, allowing for a two-handed vertical grip.

There is limited biomechanical data published to define the effects of grab-bar support on the STS transfer [9–12]. In addition, there is an even smaller body of literature focused on the effects of different configurations and positions of the grab-bars [9–11] and only one study includes a mobility limited population [11], however they assumed bilateral asymmetry in their analysis. Current practice for occupational therapists (OT) as determined from clinical observerships is to go the patient’s home and mark where the pole should be placed relative to the seat/bed/toilet with a piece of adhesive tape. The supplier of the pole comes in at a later date and installs the pole where the OT has suggested. In a follow-up visit, the OT checks to make sure that the patient can use the pole effectively and if not, the supplier must come in and make adjustments. The OT will draw on previous experience to best estimate where the pole should be ideally placed. This practice is particularly difficult for new OTs who have no experience setting up poles. Moreover, if a family member wants to set up the pole in the absence of an OT, doing so without adequate recommendations will likely result in poor pole placement or discouragement from purchasing the pole altogether. General recommendations for pole installation will be very valuable to new OTs and family members of older adults living at home with disability.

Lastly, adding to the current research on grab-bars is important to ensure technology being developed is effective at assisting users in the target activity. Equipment that is effective and fulfills the persons needs will have an impact on their level of independence, improve self-care skills, and may reduce the assistance required of caregivers [15, 16].

The purpose of this study was to examine different configurations including single pole, two poles and pole with a horizontal grab-bar positioned at two locations near and far from the user. The kinetics and kinematics of the movement were studied to determine the effects of the different transfer pole positions on characterization of movement and lower limb physical exertion during STS.

Figure 1.1: Common transfer pole configurations found in the community and clinical environment

Chapter 2

Review of Literature

The following literature review will describe the normal STS transfer in terms of kinetics and kinematics and review the effects of ageing, posture, and assistive devices on the task.

2.1 Phases of the STS transfer

Methods for describing the STS transfer vary widely between studies [17]. The following section will describe some common variability found in event identification and provide a summary of the different phases of STS used in the reviewed literature.

2.1.1 Variability in event identification

There is some disagreement in terms of what measures should be used as events to delineate the STS movement into appropriate phases [18–21]. The events used consistently across the literature reviewed include the beginning and endpoints of the movement, typically termed initiation and standing, as well as the point of seat-off, where the buttocks and thighs leave the chair.

The majority of reported studies define the beginning of STS as the initiation of trunk flexion displacement, velocity or momentum [22–31]. However, the initiation of horizontal displacement of the head, vertebra, or shoulder [32–35] and less commonly, the change in anterior-posteriorly (AP) directed ground reaction force (GRF) [19, 21, 36, 37] have also been used to define initiation of movement. Defining the end of movement include a fully upright standing (such as end $0^\circ/\text{s}$ flexion velocity), constant flexion angle (of the hip/knee) [6, 7, 19, 23–25] or the maximum height of a shoulder marker or COM [28, 32, 33]. Others defined the end of movement as the point in which movement in the horizontal direction stabilized [27, 31, 34–36, 38, 39].

Please note that momentum and velocity will be used interchangeably as some studies will use a normalized momentum which computes to velocity.

In addition to the beginning and end of STS, point of seat-off is often characterized. The ideal method for defining seat-off is using a seat-switch or a force transducer under the seat [19,26,28,34–36,40]. However other methods include initiation of vertical GRF under the feet [23,24], thigh separation from the seat [25,32,33], initiation of knee flexion [27] and peak horizontal GRF under the feet [22]. Given the variability in the definition of seat-off, comparisons between studies is difficult. Seat-off is a critical point in the movement where all of the load is transferred to the feet, resulting in large moments in the lower-extremities. Inaccuracies in the temporal definition of seat-off could result in underestimation of these moments.

2.1.2 Phases of STS from the literature

The STS transfer brings the body's COM from a stable seated position into a less stable standing position, requiring a forward and upward displacement of the COM. In its most simplest description, the STS movement can be divided into two phases: 1) flexion and 2) extension. The flexion phase consists of predominantly forward deceleration while the extension phase occurs during forward movement deceleration and is dominated by vertical movement caused by large knee and hip extension. Kelley and colleagues were among of the first to describe the movement in these two phases [26], delineating flexion from extension as the point in which maximum horizontal momentum of the COM is reached. Similarly, Nuzik and colleagues also described the movement in terms of flexion and extension phases [41], using the reversal in neck angular movement from flexion to extension as a surrogate marker to determine where forward acceleration ends and vertical movement begins. While it is reasonable marker that shows the individuals intent to transition from a horizontal movement and begin extending vertically upward, it is an indirect measurement of the transition from forward to upward movement and was not adopted in subsequent STS research.

More detailed than simply flexion and extension, Pai and colleagues have closely studied the control of body mass transfer during the STS movement, dividing the movement into 4 phases defined by COM momentum [38,40,42,43]. The trajectory of COM momentum during the initial forward movement is parabolic reaching its maximum at roughly 28% (0.43 s) of the STS transfer (before seat-off) [40]. At the point of maximum horizontal COM momentum, vertical COM momentum begins to increase and reaches a maximum at approximately 64% (0.79 s) of the STS movement [40].

More recently, Etnyre and colleagues used ground reaction force to identify STS events [21]. They investigated 100 subjects and 4 different conditions (arms-free, arms-crossed, hands-on-knees, hands-on-arm rests) and found invariate STS events. The study defined six events from the ground reaction force (GRF)

data. The first event, initiation, is defined as the first deflection from baseline of the force platform recording. The counter force event followed and is observed to be a reduction of the vertical force following initiation. Seat-off was identified with a pressure switch under the seat. Peak vertical force was defined as the maximum vertical force and rebound force was defined as the minimum force value following the vertical peak. Finally, standing determined as the first instant when the force recording settled to within normal postural sway.

These kinetic and kinematic delineations will be used throughout this thesis, however there are many other methods of delineating the STS movement. Events such as peak ankle dorsiflexion [23, 24] or reversal of trunk motion from flexion to extension. Additional methods of delineating STS movement were reported, but never adopted in subsequent literature [19, 20].

2.2 Temporal characteristics of normal STS

The total time to complete the movement component of STS (from initiation of movement to standing) from the articles reviewed ranges from 1.5 to 2.25 seconds, with one study [19] reporting a mean time of 3.33 seconds to transfer into standing. Seat-off typically occurs between 27% to 41.2% [19, 21, 23, 24, 27, 33, 40, 41, 44, 45] of STS, corresponding to a time range of 0.50 to 0.88 seconds.

2.3 Normal kinematics of the STS transfer

Kinematics refers to the spatial movement of a body or systems of bodies independent of the forces that cause the movement [46]. Variables commonly used to describe motion include linear and angular positions, displacements, velocities and accelerations of the body segments. During the STS transfer, joint kinematics define the phase of the STS transfer, position of the segments in space, direction of movement and speed of movement. The ankle, knee, hip, and trunk angles/velocities have been well described for normal, healthy individuals and will be summarized in the following section.

Initial joint angles in the seated position vary based on the height of the seat and placement of the feet. Seat height is either set based on anthropometrics or standard seat heights are used. Higher chair heights result in greater initial hip extension angles, whereas a lower seat will decrease hip extension angle. Foot placement may also affect initial posture. A more posterior foot position will increase ankle dorsiflexion and knee flexion angle, whereas an anteriorly placed foot will decrease these angles. Some studies will use a standard initial posture (typically knee angle) to control for its effect on the STS task [18, 22, 23, 47]. However, this controlled posture, may not correspond to the position from which the subject is comfortable standing and thus the resulting movement may not be typical [21, 48]. The effect of chair height, foot placement and

initial posture on the STS transfer is discussed further in section 2.5.

Reporting joint angles from the literature is difficult given the variety of methods used to define the angles between segments. Joint angles will be reported as the relative angles between segments, resolved in the distal segment coordinate system, as they are defined in the methods section (Chapter 3). Any knee and hip angles relative to the lab coordinate system will not be presented as they are neither comparable to the joint angles measured in this thesis nor the majority of literature reviewed. Ankle joint angles from the literature will only be reported if the foot segment is corrected for the apparent flexed posture inherent in a foot segment that spans the lateral/medial malleolus to the first and fifth metatarsal. Pelvis and trunk angles will be reported as relative to the lab coordinate system unless otherwise specified.

The first segment to initiate movement in the STS transfer is typically the trunk in flexion [23,24]. Peak trunk flexion angle (relative to the horizontal) occurs at, or immediately after seat-off at angles ranging from 30° to 61.1° [24,27,36,37,41]. Trunk angular displacement relative to the pelvis peaks prior to seat-off and at a smaller angle (15°) [24]. The head-arms-trunk (HAT) segment is considered the major contributing segment in generating horizontal momentum of the body's COM [23,38,44]. At seat-off, COM creates an external moment at the knee, tending to flex the knee and oppose the rise [49]. The moment must be overcome and COM brought to its final position over the BOS to achieve a quasi-static standing position. The role of initial maximum horizontal momentum is to oppose the gravitational moment arm of the COM about the knee joint to achieve this goal. Peak horizontal momentum of the COM occurs before seat-off at 0.43 s to 0.63 s into STS [37,38,44] with a magnitude ranging from 29.8 kg-m/s to 33 kg-m/s [38,44]. Trunk flexion angular velocity peaks just prior to maximum horizontal momentum [38].

Beginning shortly after initiation of trunk flexion, hip flexion begins and is initially a reflection of the forward rotation of the pelvis. Peak hip flexion velocity typically occurs before maximum trunk flexion velocity, with a magnitude of approximately $50^\circ/\text{s}$ [44]. The hip joint reaches its peak flexion angle approximately at seat-off with magnitude ranging from 98.9° to 156.5° of flexion [27,33,37,41].

Ankle angle remains fairly constant up to the point of maximum horizontal momentum when it experiences a small increase in dorsiflexion as the body begins to move horizontally, bringing the shank towards the foot. The ankle is the last joint to flex maximally (7° when foot segment is defined from metatarsal to the heel, and 10.7° when measuring the angle the shank makes with the vertical), reaching peak dorsiflexion shortly after seat-off [50].

Similar to the ankle, knee angle also remains fairly constant, extending slightly at seat-off at an angle of 100.4° [33]. At the point of maximum horizontal momentum, the knee begins extending to transition the body towards an upright position. At this point, the hip, knee and trunk all rapidly extend to bring the body into the upright standing position. Concurrent trajectory of the COM is predominantly vertical,

while vertical momentum of the COM is also rapidly increasing. Vertical COM momentum peaks at 51% to 67.4% (0.79 s to 1.15 s) of STS [37, 40, 44] with a magnitude of 44 kg-m/s [36, 40]. Much like the trunk segment is the main contributor to horizontal momentum, the rotation of the thigh about the knee joint is the main contributor to vertical momentum generation [23, 40, 44]. Hip, knee and trunk extension velocity are temporally aligned, reaching their maximum shortly after the point of maximum vertical momentum of COM. Corresponding magnitudes are $161.5^{\circ}/s$ [45], $122^{\circ}/s$ [45] and $147^{\circ}/s$ [31], respectively. Similarly, trunk, hip and knee reach full extension at or near standing, with angles near 0° .

2.4 Normal kinetics of the STS transfer

Kinetics is the branch of mechanics referring to the forces and moments that cause movement. Calculating joint kinetics requires the capture of both magnitude and direction of ground reaction forces. The combination of these external forces, anthropometrics data (mass, geometry, moment of inertia and centre of gravity for body segments) and kinematics allows for development of an inverse dynamic link-segment model. This biomechanical model enables the calculation of instantaneous moments of force, joint mechanical powers and work about each joint during the STS transfer [46].

Kinetic information for the STS transfer can be derived from a force platform under the feet and the seat. While rarely captured, forces under the seat are typically obtained from a single force platform, making it impossible to tell if the force from the chair is acting on the left or right thigh. As a result, a majority of studies will calculate joint kinetics from forces under the feet and exclude kinetics in the hip until after seat-off. To date, two studies have assessed left and right forces using a modified seat constructed from two force platforms [37, 50].

2.4.1 Reaction forces

The reaction forces under the feet and buttocks describe how the horizontal propulsive impulse of the COM is generated and controlled. The initial increase of forward acceleration of the body is predominantly generated by an anteriorly directed force under the buttocks peaking just prior to seat-off. Backward forces underneath the buttocks are generated in response to inertial forces caused by trunk flexion acceleration. Peak horizontal force (resultant force of AP and ML force vectors) of 6.2% and 5.7% of body weight for the left and right buttock, respectively [37], has been reported. In another study, peak AP forces under the seat were found to be 47.6 N (authors did not normalized to body weight) [50]. The forces beneath the feet do not contribute to the generation of a propulsive impulse, but rather act as a braking mechanism [37, 40]. At approximately the point of maximum AP buttocks force, a braking impulse sets in to control the body's horizontal momentum

and keep the COM within the BOS. This is represented by a predominantly posteriorly directed force under the feet that peaks shortly after the peak AP force under the buttocks and just prior to seat-off. Hirschfeld and colleagues reported a peak horizontal force (resultant force of AP and ML force vectors) of 4% and 4.4% of body weight for the left and right foot, respectively [37]. Etnyre and colleagues report a peak AP braking force of 9.7% of bodyweight and Mak et al. reported peak AP forces under the feet of 47.6 N [50]. AP force under the feet gradually reduces until the point of full upright standing where the force remains around 0 N [19,21,37,50]. ML forces increased medially during extension, suggesting that the body is trying to ensure lateral stability during forward and vertical displacement of COM [37]. Peak ML forces of approximately 3.1% and 0.9% body weight for the right and left foot, respectively, have been reported [21]. This corresponds to an outward pushing force applied by the feet to the floor.

There are a number of force events in the vertical direction under the feet, which have been discussed. An initial dip in vertical GRF, known as the counter force is the first change in force experienced under the feet. It was observed to be associated with hip flexion during lifting of the thighs, trunk extension, or ankle dorsiflexion occurring prior to generation of horizontal momentum. Leading up to seat-off, vertical forces under the feet begin to increase rapidly. Shortly after seat-off, peak vertical force is achieved with a magnitude greater than body weight (116.2% of bodyweight). Magnitude of peak vertical force is proportional to horizontal momentum generation and the associated downward-directed momentum generated during initial trunk flexion. This is followed by a rebound force in which the vertical ground reaction force reduces to below bodyweight (79.6% of bodyweight). After the rebound force, vertical forces under the feet stabilize to 100% of bodyweight [21].

2.4.2 Joint moments

Moment of force is defined as the product of a force acting at a perpendicular distance about the axis of rotation [46]. These moments of force about a joint reflect the net effect of all agonist and antagonist muscles acting across a joint and supply information about which muscle group is providing the movement (extensors or flexors).

The hip generates a small initial internal flexion moment of approximately 22 Nm at the early stages of the movement, before seat-off [50] (incorporated seat forces into their model). Initial flexion in the hip reflects the use of hip moments in producing horizontal momentum at the beginning of movement [50]. Ankle moments act as part of the body's control system to maintain COM to within BOS after seat-off [42,50]. The duration and magnitude of the internal ankle moment varies across studies, with some demonstrating small but consistent dorsiflexion moments peaking after seat-off [42,50,51], and others reporting no dorsiflex moment

Normalization	Ankle	Knee	Hip
Bodyweight (N-m/kg) [36]	1.11	1.19	3.05
Bodyweight*segment (N-m/kg-m) [52] length	0.132	0.158	0.271
Bodyweight*height (N-m/kg-m) [50]	0.64	1.170	0.914

Table 2.1: Summary of extension moments in the lower extremities as reported in literature

at all [4, 44]. Peak dorsiflexion moments ranged from 43 Nm to 50 Nm from the studies reviewed [42, 50, 51].

At approximately the point of maximum horizontal momentum, hip and knee extension moments and ankle plantarflexion moments begin to increase, peaking after seat-off and before maximum vertical momentum. Knee and hip moments reflect the effort used to bring the person into upright standing. Ankle moment decreases minimally after peaking, maintaining a plantarflexion moment during standing, unlike knee and hip moments which settle to zero. Peak knee and hip extension and ankle plantarflexion moments from literature are summarized in Table 2.1.

2.5 Determinants of STS

2.5.1 Speed of rising

Pai and colleagues evaluated the effect of speed of ascent on control of COM momentum. Changes in speed of ascent (from normal to fast as measured by total duration of movement) were attributed to changing peak vertical COM momentum, while maximum horizontal momentum was invariant across all speeds [39, 40, 43]. The explanation is seeded in the mechanical constraints of the control of balance and the acknowledged fact that movement of the COM outside of the BOS will result in a fall. Horizontal momentum is responsible for moving the COM forward in an effort to lift-off from the chair. It is a destabilising component as it generates a velocity of the COM directed towards the anterior limits of the standing base-of-support. Maintaining horizontal momentum despite speed requirements is a constraint required to meet the demands of maintaining upright stance and avoiding a forward fall. Vertical momentum generation is not destabilising and as a result increases with speed of ascent [40, 43, 45].

Increasing speed of movement was found to decrease hip flexion and employ a more upright trunk position at seat-off in young and elderly subjects [39, 45, 53]. Conversely, Vander Linden and colleagues found that speed of ascent had no effect on angular displacement of the joints [32].

In terms of ground reaction forces, both young and older adults generate significantly greater vertical ground forces under the feet when speed of ascent increases [45]. Ground reaction forces in the posterior

direction (representing the braking impulse), increase significantly with speed of ascent in young and old despite constant horizontal momentum generation [45]. Increasing the speed of the STS movement also results in increases in hip flexion, knee extension and ankle dorsiflexion joint moments [42].

2.5.2 Foot Placement

As previously mentioned, kinetics and kinematics may be influenced by initial posture. Preferred foot position was compared to posterior and anterior foot position prior to the start of STS by Shepherd and colleagues [34]. They found that an initial posture with anteriorly located feet resulted in longer duration of STS. This is contrary to others who compared an initial doriflexed angle (posterior position) to initial plantarflexed angle (anterior position) and did not find significant difference in task duration at both normal and fast STS transfers [39]. A more anterior position of the feet results in an increased distance from COM to the BOS and a more posterior COM position at seat-off [39, 54]. Increased hip and trunk flexion displacement reflect increasing horizontal momentum generation required to overcome this greater distance and increased gravitational moment arm at seat-off [32, 34, 39, 54].

Hip flexion moments increase as a result of increased propulsive force applied to the chair prior to seat-off. Knee extension moments increase to overcome the increased gravitational moment arm as a result of a more posterior COM at seat-off [54]. Anterior foot location require generation of a posteriorly directed force by the feet to the floor [48, 54]. When adopting a less optimal foot position, the feet apply additional posteriorly directed force (anteriorly directed reaction force) to the floor to help generate momentum in the forward direction.

2.5.3 Chair height

Lowering chair height increases the biomechanical demands of the STS movement [49, 55–60] and in some cases leads to failure to stand-up [49, 60]. The lowest chair height for successful rising in community-dwelling and nursing home residents over the age of 65 was 120% of lower leg length, according to one study [61]. At a lower chair height, maximal angular velocities for the trunk, thoracic spine, lumbar spine and hip were greater before and after seat-off [4, 49, 52, 55, 60, 62]. As a result, rising from a lower chair height leads to increased horizontal momentum generation. For example, rising from a lower chair height (65% knee height) resulted in a doubling of maximum horizontal momentum compared to a hight chair (115% knee height).

Increased maximum vertical COM momentum with decreasing chair height among the healthy normal subjects has also been reported. Similarly, maximum lower-limb joint extension angular velocity has been shown to increase significantly with chair height in young and old [55].

Hughes and colleagues describe three chair rise strategies, 'momentum transfer', 'stabilization', and some combination of the two. In the momentum transfer strategy, horizontal momentum is generated to bring the COM over the BOS. Once the COM is over the BOS, forward motion of the trunk decelerates and the musculature in the knee extend the knee to transition to standing. Due to the large generation of momentum while seated, this strategy requires fine postural control. The stabilization strategy uses little horizontal momentum and instead postural adjustments are made prior to standing to shorten the distance between the COM and BOS, diminishing the need for horizontal momentum. Adjustments include sliding the buttocks forward and moving the feet more posteriorly. In general, this stabilization strategy results in slower trunk movement, decreased horizontal momentum generation and an increased time to rise. Furthermore, for the momentum transfer strategy, as chair height is reduced, so did the generation of horizontal velocity (14.3 cm/s at the lowest chair height versus 11.7 cm/s at the highest chair height) [49].

Seat height also impacts the moment requirements in the lower extremities. Rodosky and colleagues reported increased moments at the knee as chair height decreased from 115% to 65% of knee height [4], and this has been confirmed by a number of other studies, reporting differences in the knee and hip moments as large as 50% to 60% [52, 56, 58, 63, 64].

2.5.4 Assistive devices

Assistive devices such as arm rests and grab-bars are useful aids in STS. They enable upper-extremity use to reduce moments required in the lower-extremities, improve stability by increasing the BOS and reduce the need to generate high horizontal momentum. The section below outlines the current literature in assisted STS.

Arm rests

Compared to a no arm rest condition (in a standard chair height), arm rests significantly reduce maximal hip extension moments (upwards of 50% reduction [63]), and reduce forces at the knee [6, 7, 52, 56, 63, 65]. Knee joint force reductions of 38% axially and 56% in shear have been reported when comparing with and without arm rests in the low chair (below knee height) condition [56]. Interestingly, less knee and hip moments were needed to stand up from the high chair without arm rests than a standard chair with arm rest, demonstrating that increasing chair height is a more useful chair adaptation [52, 63]. Wretenberg and colleagues evaluated a number of configurations including high, ordinary and low arm rests, as well as a no arm rest control. Ordinary arm rests included hand position variables of forward, rear, self-selected and rear using double power. Low and high configurations incorporated rear, self-selected and rear double power hand position

conditions. No differences were found between arm rest configurations for all heights and hand placement conditions. However, regardless of height and hand position, the use of arm rests gave significantly lower knee and hip moments compared to no arm rests. Doubling the force exerted on arm rests reduced hip and knee moments significantly compared to normal force exerted across all arm rest heights [66]. It was surprising to see the same reduction in hip and knee moments in low and high arm rests because it was hypothesized that higher arm rests could be used for a longer period of the STS transfer which could translate to reduced hip and knee moments. However, peak moments for hip and knee occur in the first one-third of the movement, therefore the need for arm support to reduce peak hip and knee moments is greatest at the beginning of the movement which both the low and high arm rests accommodate [66].

In terms of forces on the arm rests, vertical forces were several times larger than horizontal forces across all conditions (162 N versus 26 N for ordinary self-selected condition, respectively) [66]. Schultz and Alexander and colleagues found a greater horizontal force component on the arm rests (150 N at a 50 degree angle from the horizontal). The reason is that they incorporated a more anterior grab position that resulted in a larger horizontal pulling force to propel the body forward. In addition, functionally limited older adults demonstrated significantly greater horizontal and vertical components of hand force compared to healthy older and healthy young adults (0.29 N/kg, 0.24 N/kg and 0.21 N/kg, respectively). This shows that there is a difference in arm rest use among participants using arm rests out of necessity compared to those using them based on instruction.

In summary, arm rests reduce the burden on the lower-extremities during rising, as evidenced by reduced hip and knee forces and extension moments. However, increasing chair height is more effective at reducing burden on the lower-extremities than arm rests. There is no advantage to higher arm rests versus lower arm rests because peak moments in the hip and knee occur in the first one-third of movement which both low and high arm rests accommodate. However, increasing the applied force on arm rests does further reduce hip and knee torques.

Grab bars

Grab bars provide an alternative method of assistance during the STS movement. To date, only a few studies have investigated the STS task with and without the use of an assistive grab-bar [9–12]. O’Meara et al [9, 10] have provided the most comprehensive work to date. In a series of two papers from a single study, the authors evaluated effects of grab-bars positioned laterally and anteriorly in a population of healthy older adults, examined grab-bar forces, ground reaction forces, lower limb joint forces/moments/powers and measures of whole body COM. Comparisons were made between an unassisted condition [10] and several different grab-bar configurations tested [10]. In terms of grab-bar forces, significantly greater grab-bar force

was applied when grab-bars were placed laterally compared to anteriorly. Further, when the grab-bar was placed laterally, the user was able to apply greater posteriorly directed force when oriented horizontally. Not surprisingly, grab-bar force also increased with use of two hands compared to single-hand use. In all configurations, large horizontal forces are applied to the grab-bar alongside reductions in COM velocity. The authors [10] suggest that propulsion afforded by use of the grab-bar replaces momentum typically generated by the COM in unassisted STS [40].

The use of grab-bars also introduced an asymmetrical loading of the joints, resulting in an increase in moments in the hip and decrease in knee moments contralateral to the dominant hand with unilateral grab-bar assistance compared to unassisted [9]. In contrast, the ipsilateral net joint moments decreased in the ankle and hip, while actually increasing in the knee [9]. When comparing a laterally positioned grab-bar to one that is placed anterior to the individual, a reduction in peak joint forces in the lower limbs and knee joint moments was also detected [10]. These observations were coupled with the unilateral reaction force from the grab-bar and a posteriorly directed ipsilateral ground reaction force. Thus, observed asymmetry in joint kinetics occurred without obvious changes to whole body movement. Assisted propulsion forwards with the unilateral grab-bar is countered by the motor control system to balance moments about the COM in the transverse plane and achieve symmetric movement of the body [10] and conserve dynamic stability.

In a more recent study, Kinoshita et al. [11] evaluated grab-bar position in assisting the STS transfer in a sample of healthy young, healthy older and functionally limited older adults. Testing conditions included unassisted, two-handed high (90-110 cm) grab-bar, two-handed low (60-80cm) grab-bar and a combined low/high grab-bar design. Symmetry was assumed, therefore dominant side only was tested. All configurations resulted in a reduction in lower limb joint moments compared to unassisted rising. However, the low/high grab-bar configuration demonstrated the greatest reduction in joint moments. Results are consistent with other studies which have reported normal assistance and strong assistance (ie. putting additional force on the grab-bar) in healthy individuals using a grab-bar compared to unassisted STS [12].

2.6 Purpose of the study

The STS transfer has been studied extensively, however research on the effects of grab-bars is still limited. The literature described clearly demonstrates a biomechanical advantage of grab-bar use when compared to no assistance. However, comparisons across different types of grab-bars is limited to two studies [10,11] and only a single study included functionally limited older adults [11]. The present study sought to develop a better understanding of the effects of different types of grab-bar configurations on functionally limited older adults.

The purpose of this study was to determine how different grab-bar configurations and positions affected kinetics and kinematics during the STS transfer. The objectives of the study were:

1. Characterize differences in kinematics across different transfer pole configurations (single vertical pole, single pole with horizontal grab-bar, and double vertical poles) and two positions with respect to chair placement.
2. Compare differences in reaction forces at pole, seat and floor across different grab-bar configurations and positions.
3. Compare differences in physical exertion in the lower extremities across different configurations and positions.

Chapter 3

Methods

3.1 Study design

This study was a single group, repeated measures design testing 6 independent variables (3 pole configurations at two locations) using within-participant differences to determine the effect of different grab bar conditions on joint kinetics, kinematics and reaction forces at the seat, floor and pole.

3.2 Subjects

A convenience sample of nine subjects were recruited over a ten month period by the physical therapist(s) in the Geriatric Day Hospital and Geriatric Rehabilitation at the Toronto Rehabilitation Institute - University Hospital Network (TRI-UHN).

Inclusion criteria

1. Over the age of 65 years old
2. Unable to stand up without assistive technology (ie. grab bars, arm rests, etc.)
3. Able to understand 2-3 step instructions
4. Able to participate in a single 2 hour session

Exclusion criteria

1. Amputee

2. Requires human assistance to stand
3. Unable to use the grab bar due to upper-extremity disability

3.3 Experimental set up

Figure 3.2 shows a snapshot of the experimental apparatus and a plan view schematic of the force platform configuration, camera set up and laboratory coordinate system (LCS). The chair used for this study was a firm, backless, armless seat with a 43 cm standard seat height that was fastened to a force platform (Advanced Mechanical Technology Inc, Massachusetts, USA). Two smaller force platforms were positioned anterior to the seat such that each foot could be comfortably placed on their respective force platform while seated. The pole was constructed of 1.67 inch steel tube, assembled into an adjustable pole using Kee-Klamps (Simplified Building, New York, USA) and mounted on a fourth force platform. The pole accommodated up to 1.85 m grab height and it could be brought as close to 15 cm and as far as 1.2 m from the chair location.

Active motion capture (PhoeniX Technologies Incorporated, British Columbia) was used to track positions of the joints in three-dimensional (3D) space. Infrared emitting diodes (IREDs) were placed on the body using rigid body marker clusters and attached to the segments using neoprene straps, reinforced with skin tape to prevent slipping. Figure 3.1 shows the marker clusters attached to a subject. Foot clusters were taped to the shoe using duct tape. This permitted tracking of the trunk, pelvis, left and right foot, shank, and thigh. Landmarking trials were conducted using an instrumented probe after the movement trials to locate additional anatomical markers required to define the segment endpoints and the joints to input into the biomechanical model. The LCS was set such that the positive x-axis was the direction of forward movement (seat to pole), positive y-axis was directed to the non-dominant (ND) side and positive z-axis directed up. Figure 3.2 shows the LCS in the experimental setting.

3.3.1 Definition of the dominant side

The dominant (D) side was defined to be the same as the dominant hand. Referring to the dominant side in the lower limbs refers to the side ipsilateral to the dominant hand. Conversely, the ND side refers to the contralateral side. All figures are shown with the right hand as dominant.

3.3.2 Test conditions

A total of 6 independent variables were tested consisting of three pole configurations at two positions. Pole conditions tested included the single vertical pole, single pole with the horizontal grab bar attachment and

Figure 3.1: Rigid body clusters strapped on to a subject

Figure 3.2: Left, STS test apparatus set up in StairLab; Right, modelled overhead view of the experimental set up illustrating the locations of the force platforms and cameras. ND side = left and D side = right

Variable	Summary
Single vertical pole	Anterior to the user centred between left and right acromion.
Single vertical pole with a horizontal bar	Anterior to the user with the vertical bar aligned with the acromion of the dominant hand and horizontal bar set to a comfortable height based on user preferences. A 30cm horizontal bar aligned 90° to the vertical and parallel to the user
Double vertical pole	Anterior to the user with each bar aligned to the left and right acromion
Near	Set at a distance from the acromion equal to 75% of thumb-tip reach
Far	Set at a distance from the acromion equal to 75% of thumb-tip reach plus the equivalent distance of 20° trunk flexion ($\sin(20^\circ)$)

Table 3.1: Summary of independent variables

Figure 3.3: Top-view of the pole configurations relative to the chair that accompany 3.1. The large wooden textured block represents the chair, the grey blocks represent the left and right force platforms and the circle represents the pole. ND side = left and D side = right

the double vertical pole. Positions tested included a near position and a far position. Details about the variables are summarized in Table 3.1 and illustrated in Figure 3.3. Subjects were directed to grab the pole with both hands and their dominant hand over their non-dominant hand. Height of the horizontal bar was self-selected. The order of configurations changed with every subject to balance for the effects of fatigue.

3.4 Procedure for recording STS movement

Data was collected in the StairsLab, one of the laboratories in the Challenging Environment Assessment Laboratory (CEAL) at TRI-UHN.

Once consented, subjects were scheduled for a single test session that lasted approximately 2-hours in duration. Before entering the lab, height, thumb-tip reach, and distance from iliac crest to acromion (trunk height) were measured in the preparation room. Participants were then brought to the lab and prepared for motion capture. All 8 marker clusters were strapped onto the appropriate segments and plugged into the transmitter/power source components which were velcroed to a belt and strapped around the midsection. Once prepared subjects were assisted into the chair. The poles were adjusted to the first condition to be tested for that day.

Once the pole was adjusted, subjects were asked to stand up once and sit back down again with the assumption that they would be in a more natural position. This also allowed the subject to experience the assisted STS. Measurements of the pole relative to the user were taken again to ensure distance from the pole was the same and if not, adjustments were made. Once the final, self-selected seated position was established, tape was used to border the feet and buttocks to ensure when they were in the same initial starting position for each trial.

Participants were asked keep their hands to their side, then grab the pole with both hands and stand up on the researchers mark of “OK, please stand up”. Participants were asked to remain standing for 20 seconds to capture stabilization data. After two trials of the same condition, the conditions were changed. This was process was repeated until all six conditions were tested.

Landmark trials and static pose were recorded in a neutral standing posture once the movement trials were complete. Instrumentation was then removed from the subjects and they were escorted back up to the clinic or the lobby.

3.5 Data processing

A 3-D eight segment model was developed using biomechanical analysis software (Visual3D, Maryland). All segment motion was tracked using rigid body clusters consisting of four non-collinear markers. Rigid body tracking was selected over skin mounted markers to reduce soft-tissue artifact. A study by Angeloni and colleagues observed statistically greater displacements in skin mounted markers compared to markers mounted on plates. They concluded that markers mounted on rigid surfaces were preferable to skin mounted markers in terms of accuracy and practicality [67].

Rigid body clusters were designed based on the criteria put forward by Capozzo and colleagues [68]. Briefly, four markers or more is recommended for redundancy and accuracy. Mean radius of the cluster must be greater than ten times the assessed standard deviation of the experimental errors. Three dimensionality has no effect on the accuracy of tracking. Lastly, 2D isotropy index should be greater than 0.5. Figure 3.1 shows the rigid body clusters attached to a subject.

3.5.1 Segment definition

Landmarking trials identified virtual locations approximating the first and fifth metatarsal, lateral and medial malleolus, lateral and medial epicondyle, greater trochanter, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), and the acromion. Midpoints between the medial and lateral malleoli and the medial and lateral epicondyles defined the ankle and knee joint, respectively. Hip joint centres were estimated

using the CODA pelvis model developed by Visual3D [69]. The pelvic segment is defined using the left and right greater trochanter, ASIS and PSIS. Given these inputs the right and left hip joint centres are estimated using formulas 3.1 and 3.2 adapted from [70, 71]. The foot segment was defined by the first and fifth metatarsals and the medial and lateral malleoli virtual landmarks. The shank was defined by medial and lateral malleoli and medial and lateral epicondyle. The thigh segment was defined by the medial and lateral epicondyle and the greater trochanter and hip joint centre.

$$RHJC = (0.36 * d_{ASISx}, -0.19 * d_{ASISy}, -0.3 * d_{ASISz}) \quad (3.1)$$

$$LHJC = (-0.36 * d_{ASISx}, -0.19 * d_{ASISy}, -0.3 * d_{ASISz}) \quad (3.2)$$

Where RHJC represents the right hip joint centre coordinates, LHJC represents the left hip joint centre coordinates and d_{ASISx} , d_{ASISy} and d_{ASISz} represent the coordinate location of the ASIS relative to the origin.

Segment coordinate system

Methods for defining the segment coordinate system (SCS) vary based on the landmark locations available. The segment coordinate system definition used here was adopted from Visual3D which develop their biomechanical model in accordance to the National Institutes of Health (NIH) biomechanical standards [69]. Using the shank as an example, four landmarks are available to establish the coordinate system. The first plane developed is the frontal plane, where a plane is fitted (using a least squares fit) between the points defined by the medial and lateral malleoli and the medial and lateral epicondyles. The plane is defined as x-z plane. The origin is located at the proximal end. The axial axis, defined as the z-axis, passes through the joint centres (midpoints between medial and lateral landmarks) of each segment endpoint. The z-axis is directed from the distal to the proximal endpoint. Finally, the y-axis is a vector orthogonal to the x-z plane, starting at the origin and directed anteriorly. The x-axis is calculated as the cross-product of the z- and y-axes directed from the origin to the lateral landmark. The same principle is applied to foot, thigh, and trunk. Trunk landmarks used to establish frontal plane include left and right acromion and greater trochanter.

Pelvic segment coordinate system is calculated slightly different. The transverse plane (x-y plane) is defined using a least squares fit between the left and right PSIS and ASIS. The origin is calculated as the midpoint between the ASIS and PSIS. The z-axis is orthogonal to the x-y plane directed upwards. The x-axis is directed from the origin to the right ASIS. The y-axis is calculated as the cross-product of the x- and z-axes. Figure 3.5 illustrates the segment coordinate systems in the sagittal and frontal planes on the skeletal model.

Joint angle definition

Sagittal plane movement is rotation about the y-axis, frontal plane movement is rotation about the x-axis and transverse plane movement is rotation about the z-axis. Sagittal plane sign convention: dorsiflexion, knee and hip flexion, pelvic forward tilt, and trunk flexion are positive. Frontal plane sign convention: non-dominant(ND)-directed lateral trunk displacement are positive. Joint segment coordinate systems and definitions for trunk angles are displayed in the skeletal model shown in Figure 3.5.

The joint angles calculated are described by the rotation matrix defined between two segment coordinate systems. For example, the rotation matrix defined by transformation between the foot and shank coordinate systems describes the ankle angle. All joint angles are resolved in the coordinate system of the distal segment. The exception is the trunk coordinate system which is calculated relative to the laboratory coordinate system (LCS). Joint angles are calculated using the Cardan sequence (x, y, z; representing flexion/extension, abduction/adduction and intal/external rotation, respectively). Static posture captured at the end of data collection established neutral joint angles.

3.5.2 Joint moment calculation

Joint moments were computed using the Newton-Euler approach of inverse dynamics [72]. Figure 3.4 shows the free-body diagram (FBD) of the lower leg. Moments at the knee are derived in equations 3.3 to 3.5 from this FBD to exemplify moment computation using the Newton-Euler approach of inverse dynamics.

$$\sum M_{knee} = I_{COM} \cdot \alpha_{COM} \quad (3.3)$$

$$I_{COM}\alpha_{COM} = M_{knee} + R_{ankle_y} \cdot l_{leg} \cos(\theta) - R_{ankle_x} \cdot l_{leg} \sin(\theta) - mg \cdot l_p \cos(\theta) + M_{ankle} \quad (3.4)$$

$$M_{knee} = I_{COM} \cdot \alpha_{COM} - R_{ankle_y} \cdot l_{leg} \cos(\theta) + R_{ankle_x} \cdot l_{leg} \sin(\theta) + mg \cdot l_p \cos(\theta) - M_{ankle} \quad (3.5)$$

Where M_{knee} and M_{ankle} are the moments at the knee and ankle, respectively. I_{COM} is the moment of inertia about the COM, α_{COM} is the angular acceleration about the knee, m and g and denote mass of the segment and gravity respectively and a_{com} is the linear acceleration of the COM. R_{ankle_x} , R_{ankle_y} , R_{knee_x} and R_{knee_y} are the x- and y-components of the reaction forces at the knee and ankle, respectively. θ is the ankle of the leg relative to the horizontal, l_{leg} is the length of the leg and l_p is the length of proximal portion of the leg measured from the knee to the COM.

Figure 3.4: Free-body diagram of the lower leg for the inverse dynamic calculation represented by equations 3.3 to 3.5

3.5.3 Work and angular impulse calculation

Work was calculated from the point of seat-off as this was when the knee and hip underwent joint angular displacement using equation 3.6. Angular impulse was calculated from initiation to standing using equation 3.7.

$$W = \int_{\theta_{stand}}^{\theta_{SO}} M_{joint} d\theta \quad (3.6)$$

$$H = \int_{t_{stand}}^{t_i} M_{joint} dt \quad (3.7)$$

Where H and W are symbols representing work and angular impulse. M_{joint} denotes the moment of the joint in question. Time point t_i and t_{stand} denote time of initiation of movement and standing, respectively. θ_{SO} and θ_{stand} are joint angular displacement between seat-off and standing.

3.5.4 Force vector sign convention

The force vectors use the same sign convention as the LCS, that is, anterior, left-directed and up-directed reaction force vectors are positive. For example, posterior forces applied to the pole are positive as they

Figure 3.5: Skeletal model displaying segment coordinate systems for the sagittal plane and frontal plane (trunk). θ_{flex} and θ_{lat} represent the trunk angle in flexion and bending, respectively

generate an anteriorly directed reaction force vector. This study defines left as the ND side, therefore ND reaction force vectors are positive.

3.5.5 Model COM calculation

COM was calculated automatically by the Visual3D software using the following methods. To compute total COM of the model, a weighted average of the segments that make up the body are calculated using equations 3.8 and 3.9 below:

$$X_{total} = \sum_{s=1}^L P_s \cdot x_{cg} \quad (3.8)$$

$$Y_{total} = \sum_{s=1}^L P_s \cdot y_{cg} \quad (3.9)$$

Where L is the total number of segments x_{total}, y_{total} are the coordinates of the model COM, x_{cg}, y_{cg} are the coordinates of the segments' COM, P_s represent the mass proportion of segment s is the segment index. Segment mass is determined from the total body weight and Dempster's anthropometric data [73].

3.5.6 Event identification

All events were calculated automatically by algorithms developed in matlab. All identified events were checked manually to ensure accuracy. Events identified are summarized below.

Initiation of movement

Initiation of movement is typically defined as the point in which trunk flexion begins or AP force under the feet deviates from resting force. However, in some cases, very little trunk flexion was experienced when arms began to rise to grab the poles. A more appropriate event was found to be the point in when subjects began pulling on the pole(s). This was defined as the point in which AP pole force was equal to 10% of bodyweight. Unfortunately, this is an unconventional start point, therefore duration of the assisted STS in this study will not be comparable to literature values. Figure 3.6 shows defined the events on their respective curves.

Seat-off

Seat-off was established as the point in which z-directed force in the seat went to zero. The z-directed seat force dataset was treated as a first-order step response. Using the matlab function `stepinfo` with a 2% settling time threshold, settling time was found to accurately find the point in which z-directed forces at the seat approached zero. Figure 3.6 (top-right) shows the event on the curve representing z-directed seat forces as a function of time.

Standing

The end of STS was defined as the point in which the subject was in the final standing posture. This was defined by the stabilization of the knee or hip joint angle. The algorithm developed calculated a 95% confidence interval ($2 \times$ standard deviation) for knee and hip angle at the last 2-seconds (120 frames) of the trial. Starting from seat-off, standard deviation of knee and hip angle was calculated using a moving window consisting of 120 frames (2-seconds). Standing was defined as the point when calculated variability was within

Figure 3.6: Event identification for (moving across from top left to bottom right) initiation of movement, seat-off, ND and D hip flexion angles and vertical COM velocity. Vertical line denotes the identified event. Right hip and knee angles end at their identified events and therefore do not include a vertical line

the 95% confidence interval. Figure 3.6 (bottom-left) shows the event on the curve representing the hip joint angle.

Maximum horizontal and vertical COM velocity

These points were calculated as the maximum points on the x- and z-directed COM velocity data sets bound by initiation of movement and standing, as shown in Figure 3.6 (bottom-right). Manual inspection of the data ensured the appropriate peak was used to define the event. A second peak in maximum horizontal velocity was often observed near the end of the movement. If the algorithm found the second peak, data from the appropriate peak was selected manually.

3.5.7 Outcome variables

All outcome variables were calculated for the STS movement, defined by initiation and standing events.

Temporal parameters

STS duration from initiation to standing and from initiation to seat-off was calculated. Time of kinetic and kinematic parameters listed below were also determined.

Kinematic parameters

Mean and standard deviation of peak joint angles and ranges were calculated to describe the movement strategy from initiation to standing. The following section summarizes the joint metrics calculated and relates them back to the research objectives.

Ankle In the AP direction, maximum (dorsiflexion) and minimum (plantarflexion) angles were determined. The purpose of these metrics are to characterize the movement of the ankle during the movement.

Knee In the AP direction, maximum angle, angle at standing (minimum angle) and range between them were calculated to characterize movement.

Hip In the AP direction, maximum and minimum angles as well as the range between these values were output.

Trunk In the AP direction, maximum (flexion) angle was output. In the ML direction, peak ND and D directed lateral bending and range between these metrics were calculated to characterize the movement.

Figure 3.7: Schematic of the base of support identified the anterior and posterior borders and D and ND borders in the D and ND foot. The numbers 1 to 8 represent sequentially, ND fifth metatarsal, ND first metatarsal, ND lateral malleolus, ND medial malleolus, D fifth metatarsal, D first metatarsal, D lateral malleolus, D medial malleolus,

COM displacement In the AP direction, maximum COM displacement was calculated relative to the midpoint between ankle joints. The measure will aid in the discussion of trunk flexion and generation of COM velocity during STS. In the ML direction, peak ND and D COM displacement was calculated relative to the midline of the BOS of the feet. Figure 3.7 illustrates the perimeter of the BOS of the feet defined by the ND fifth metatarsal, ND malleolus, D fifth metatarsal and the D malleolus. The line connecting the midpoints of the ND and D malleolus and the ND and D fifth metatarsal defined the midline of the BOS. COM displacement in the AP direction was described relative to the posterior border of the BOS defined by the ND and D malleolus. A negative value corresponds to a COM position posterior to the posterior border. AP COM displacement was normalized to BOS length defined by the length of the midline described above. COM displacement in the ML direction was described relative to the midline of the BOS, where a negative value corresponds to a COM position on the D side and a positive value corresponds to a COM position on the ND side. ML COM displacement was normalized to BOS width defined by length of the line connecting the midpoints of the ND fifth metatarsal and malleolus and the ND fifth metatarsal and malleolus.

COM velocity In the AP direction, maximum (horizontal) COM velocity was calculated to determine the level of COM momentum generated during the movement. In the vertical direction, maximum vertical COM velocity was calculated to determine the level of COM momentum generated in the vertical direction during the movement. COM velocity was normalized to height.

COP displacement COP displacement was only calculated under the feet and is described relative to the BOS described under COM displacement. The COP displacement is reported for the ND foot (COPND), D foot (COPD) and both feet. Total COP displacements in the ML direction were calculated to further characterize the movement. AP displacement of COPND and COPD at the point of maximum hip extension moment was calculated to determine where the cumulative force vector was located under each foot and describe its location relative to ankle, knee and hip joint centers. Total COP was calculated at seat-off in the AP and ML directions to help characterize the movement. COP is described relative to BOS coordinate system displayed in 3.7. Formulas for calculating COP are displayed in equations 3.10 and 3.11.

$$COP_x = \frac{-(M_y + F_x \cdot z_{diff})}{F_z} \quad (3.10)$$

$$COP_y = \frac{(M_x + F \cdot z_{diff})}{F_z} \quad (3.11)$$

Where COP_x and COP_y are the x- and y-coordinates of the COP. M_x , M_y , F_x and F_y are the moments about the x- and y-axis and forces in the x- and y-directions. Z_{diff} is the distance from the top of the force platform to its origin as defined by the manufacturer.

Kinetic parameters

All kinetic parameters were normalized to bodyweight.

Ankle moment In the AP direction, maximum (dorsiflexion) and minimum (plantarflexion) moments were output to characterize the movement, evaluate effort and examine the role of plantarflexors during STS.

Knee moment In the AP direction, minimum (extension) moment was output to examine physical exertion by the lower limbs.

Hip moment In the AP direction, maximum (flexion) and minimum (extension) moments were output to examine physical exertion by the lower limbs.

Seat force Propulsive AP seat force was the peak located between maximum horizontal momentum and seat-off. Seat force was output to determine its contribution to forward propulsion across different conditions.

Pole force Peaks in the AP, ML (ND and D directed) and vertical direction were output to characterize pole use.

Ground forces In the vertical direction, peak vertical ground reaction force in the ND and D side, as well as the sum, were determined. In the AP direction, peak posteriorly directed ground reaction force were determined for ND and D feet, as well as the sum of both feet. In the ML direction peak laterally directed and medially directed ground reaction forces were determined. Ground reaction forces are calculated to determine their contribution to moments and characterize the movement.

Work done Work done was calculated at the knee and hip from seat-off to standing to help characterize physical exertion

Angular impulse Angular impulse in the knee and hip were calculated from initiation to standing

3.5.8 Conditioning the data

Filtering

Motion and force data were filtered using a 2nd order dual-pass Butterworth filter with 6 Hz cutoff frequency [9]. The cutoff frequency was confirmed to retain 99% of the data in the power spectrum analysis performed. All of the filtered data were visually reviewed in graphical form while simultaneously observing movement of the skeleton to ensure the graphical data appropriately represented human movement.

Interpolation

Motion data were interpolated using a 2nd-order spline. Up to 10 missing data points were interpolated. All data were inspected visually as described above.

3.5.9 Statistical analysis

Means and standard deviations were calculated for every metric under every pole condition. Kinetic and kinematic metrics in D and ND sides were analysed using repeated measures ANOVA to determine significant within participant differences. One main effect, condition, with 6 categories, one for each pole type and distance combination, was specified in the model. Statistical significance was determined using the traditional p -value < 0.05 . Nine contrasts were constructed for the ANOVA, comparing differences between near (3 comparisons) versus far (3 comparisons) locations for each configuration, and differences between each pair of configurations at each location (3 comparisons). If the condition was statistically significant, as per the ANOVAs F-test, the contrasts were examined to determine the effects of pole type and distance. Statistical analysis was conducted using SAS 9.3 (SAS Institute Inc, Cary, North Carolina).

Chapter 4

Results

A total of 9 subjects (2 females, 7 males) admitted to the geriatric day hospital or the geriatric rehab program at the Toronto Rehabilitation Institute (TRI) University Centre (UC), were recruited for this study. The range and mean age of the subjects were 67 to 90 years and 77.8 years, respectively. Subjects were admitted to the rehab clinic for a variety of ailments which effected function of upper- and lower-extremities and resulted in limited mobility. Demographics of the participants are listed in Table 4.1.

Table 4.1: Summary of subject demographics

Subject	Age	Height (m)	Weight (kg)	Dominant side
1	79	1.7	70	R
2	90	1.76	66.6	L
3	79	1.85	113	R
4	81	1.88	93.7	R
5	67	1.6	56.2	R
6	68	1.8	77.3	L
7	82	1.68	77.3	R
8	79	1.78	78	R
9	78	1.75	87	R
Mean	78.1	1.8	79.9	
SD	7.0	0.1	16.5	

L, left; R, right

4.1 Grab height

Median grab height for the all of the pole conditions tested with standard deviations in brackets are listed in Table 4.2. Differences between grab height were not analysed statistically. Observationally, the D hand consistently grabs the pole at greater height than the ND hand in the horizontal and single pole configurations

at near and far positions. Conversely, ND and D hand height in the double poles were similar. Mean distance of the pole from the subjects was 56.1 (2.61) cm for the near condition and 71.15 (3.02) cm for the far condition.

Table 4.2: Mean grab heights with the non-dominant and dominant hands across all grab-bar conditions

Parameter	Near			Far		
	Horizontal	Single	Double	Horizontal	Single	Double
ND hand	1.07 (0.12)	1.05 (0.12)	1.11 (0.07)	1.05 (0.12)	1.04 (0.12)	1.09 (0.12)
D hand	1.15 (0.08)	1.14 (0.12)	1.10 (0.07)	1.13 (0.09)	1.11 (0.08)	1.10 (0.08)

ND, non-dominant; D, dominant
Statistical analysis not conducted on grab height differences

4.2 Temporal parameters

Table 4.3 summarizes the temporal parameters for each configuration. No statistical differences were found in total time or identified events. In general, time of maximum horizontal velocity took place between 22% and 32% of the total movement time, similar to time of seat-off which occurred between 24% and 27% of total movement. Maximum vertical velocity occurred slightly later, between 41% and 44% of total movement.

Table 4.3: Temporal parameters of identified events in the STS movement

Events	Near			Far		
	Horizontal	Single	Double	Horizontal	Single	Double
Max horizontal velocity (s)	0.66 (0.3)	0.68 (0.5)	0.79 (0.5)	0.83 (0.4)	1.11 (1)	1.09 (1.2)
% STS	22.5 (10.5)	23.7 (15.1)	27.3 (15.4)	28.4 (12.0)	31.7 (14.2)	32.0 (17.5)
Seat-off (s)	0.79 (0.4)	0.72 (0.5)	0.76(0.6)	0.66 (0.3)	1 (1.1)	0.98 (1)
% STS	24.9 (8.2)	22.6 (9.4)	24.0 (9.4)	21.5 (6.5)	26.4 (12.0)	26.5 (12.3)
Max vertical velocity (s)	1.37 (0.8)	1.36 (1)	1.24(0.7)	1.39 (1.1)	1.58 (1.5)	1.52 (1.2)
% STS	42.8 (13.7)	41.6 (15.9)	40.6 (11.6)	41.8 (14.6)	43.7 (13.8)	42.8 (12.8)
Total (s)	3.06 (0.8)	3.04 (1.2)	2.96(1)	3.04 (1.1)	3.27 (1.6)	3.25 (1.6)

STS, sit-to-stand
No statistical differences

Figure 4.1 summarizes the mean timing of kinematic events during STS for one configuration (horizontal pole in the near position) on the ND side. Temporal parameters were included to determine when kinematic and kinetic measures occurred and contribute to the characterization movement. Finding differences in the timing of key events across pole conditions was beyond the scope of the thesis and as a result, statistical analysis was not conducted to determine differences in temporal parameters across test conditions. The sequence order of kinematic measures were generally the same across all configurations and any differences will be described. In addition, detailed timing data can be found in Tables A.1 and A.2 in Appendix A.

Investigation of the mean values to characterize the temporal events demonstrated that maximum hip flexion occurred first at approximately the point of seat-off (24%), which also coincided with maximum horizontal COM velocity and maximum ankle dorsiflexion (28%), followed by maximum trunk flexion (38%). Maximum hip and knee extension occurred at the end of the movement, as they are markers of upright standing. Timing of peak COM displacement in the ML axis was highly variable across configurations. The horizontal pole configuration at the near and far position demonstrated a consistent pattern where ND-directed lateral COM movement occurred first between seat-off and maximum vertical COM velocity. D-directed movements followed, occurring later in the movement as the participant approached standing. In the other configurations, timing of ND- and D-directed COM movement demonstrated the opposite trend, moving sequentially from D to the ND side during movement.

Figure 4.1: Temporal characteristics of kinematics data for the horizontal pole in the near position on the ND side

Figure 4.2 summarizes the mean timing of kinetic events during STS with the horizontal pole at the far position. Maximum AP seat force (18%) occurred prior to maximum ankle dorsiflexion moment (27%) prior to seat-off. This was followed by seat-off which coincided with maximum AP pole force and peak hip extension moment and then maximum horizontal velocity. Maximum knee extension moment occurred next (27%), coinciding the with peak vertical force in the pole. Maximum vertical velocity and finally maximum plantarflexion occurred last. The sequence order of all kinetic events were similar across all configurations and positions, see Appendix A for more detailed temporal measures.

Figure 4.2: Temporal characteristics of the kinetic data for the horizontal pole in the near position

4.3 Kinematics

4.3.1 Joint angles

Mean and standard deviations for kinematic measures of STS are presented in Table 4.5 for all configurations. Ankle, knee and hip joint angles from a representative subject (subject 8) are displayed in Figure 4.3. Table 4.4 display F-values and p-values of the main effects for all joint kinematic metrics. The ANOVA revealed main effects in maximum ankle dorsiflexion angles on the ND side, D and ND side peak hip flexion angles, trunk flexion and ND- and D-directed lateral bending. Dominant peak ankle dorsiflexion angle and the maximum knee flexion, extension and range of movement did not show main effects across pole conditions (see Table 4.4).

Contrasts revealed the greatest maximum ND ankle dorsiflexion in subjects using the double pole configuration in the far position compared to the double configuration in the near position ($p = 0.002$) and the single pole in the far position ($p = 0.0091$). Magnitudes of difference are very small (less than 4°) across significantly different conditions.

Examining post-hoc contrasts of hip joint flexion angles revealed that the use of the double and single pole configurations increased peak hip flexion angle in comparison to the same configurations in the near position ($p = 0.017$ and 0.04 , respectively). In addition, maximal trunk flexion was greatest in the far configuration compared to near, demonstrating statistical significance for all three pole configurations ($p = 0.05$, 0.05 and

0.009, for horizontal, single and double pole, respectively).

In terms of trunk movement in the frontal plane, differences in magnitude across all configurations were small, however, statistical differences emerged. Subjects using the near pole demonstrated significantly greater ND-directed trunk bending in the single and the horizontal pole compared to the double pole configuration ($p = 0.0006$ and 0.04 , respectively). Comparing the near pole with the far pole position, the double pole in the far position demonstrated greater ND directed trunk bending compared to the near position ($p = 0.02$). Examining D-sided bending, no differences were found between the near and far pole positions. In the near position, subjects using the horizontal and double pole configurations demonstrated significantly greater D-directed trunk bending compared to the single pole ($p = 0.03$ and 0.03 , respectively). In the far positions, users of the double pole demonstrated greater D-directed lateral bending compared the single pole ($p = 0.04$).

Table 4.4: F-values and main effect p-values for joint kinematics

Parameter	Side	F-value	Pr > F
Ankle	ND	2.66	0.038
Max DF (°)	D	1.37	0.26
Ankle	ND	3.46	0.012
Min DF (°)	D	1.65	0.17
Knee	ND	1.23	0.31
Peak to peak (°)	D	1.07	0.39
Hip	ND	3.18	0.016
Max flex (°)	D	3.06	0.020
Hip	ND	1.52	0.21
Min flex (°)	D	1.26	0.30
Trunk			
Max flex (°)		3.26	0.015
Max lateral (°)	ND directed	2.18	0.076
Max lateral (°)	D directed	5.35	0.0007

4.4 Center of mass and center of pressure

Mean and standard deviation for all COM and COP metrics are summarized in the Table 4.7 and 4.8. Results of the ANOVA are summarized in Table 4.6. Main effects were observed for COM displacement at seat-off in the AP direction, ND-directed ML COM displacement and COP displacement in the AP direction in the ND foot (see Table 4.6). No main effects were found for COM displacement at seat-off in the ML direction, ML COM velocity at peak ML COM displacement, maximum AP and vertical COM velocity, D-directed ML

Table 4.5: Summary of sagittal plane joint angles for all pole-use conditions

Parameter	Side	Near				Far			
		Horizontal	Single	Double	Horizontal	Single	Double	Horizontal	Double
Ankle	ND	21.48 (7.57)	21.26 (6.68)	19.60 (7.88)*	21.65 (5.38)	20.20 (6.23)†	23.18 (7.13)*†		
Max DF (°)	D	20.29 (8.73)	22.51 (6.50)	20.60 (8.62)	21.73 (7.33)	21.61 (6.06)	22.64 (6.71)		
Ankle	ND	7.79 (6.47)	7.79 (4.44)	8.38 (5.92)*	8.93 (4.42)	8.36 (5.46)†	10.00 (5.22)*†		
Min DF (°)	D	7.24 (6.15)	8.66 (5.26)	8.51 (5.98)	8.49 (4.95)	8.41 (5.30)	9.92 (5.83)		
Knee	ND	96.98 (10.42)	95.85 (9.42)	95.92 (9.80)	95.77 (8.18)	95.60 (8.75)	98.85 (14.82)		
Max flex (°)	D	96.37 (9.59)	97.87 (7.56)	96.85 (8.30)	96.54 (8.75)	96.57 (8.00)	96.75 (8.97)		
Knee	ND	24.40 (7.44)	24.94 (8.10)	23.87 (7.48)	23.41 (6.22)	23.90 (7.47)	24.22 (6.44)		
Min flex (°)	D	23.65 (7.67)	25.50 (8.75)	24.12 (8.45)	23.81 (6.88)	25.73 (5.62)	24.92 (6.84)		
Hip	ND	76.92 (12.11)	75.26 (11.87)†	74.94 (10.39)*	79.44 (12.39)	79.05 (14.29)†	78.13 (12.62)*		
Max flex (°)	D	77.06 (11.74)	76.62 (10.97)†	75.58 (10.31)*	78.75 (11.74)	79.94 (13.17)†	79.31 (13.70)*		
Hip	ND	16.14 (3.08)	14.70 (6.66)	13.67 (5.48)	14.36 (5.48)	13.72 (4.22)	14.78 (3.55)		
Min flex (°)	D	17.01 (3.72)	16.25 (5.40)	14.70 (5.77)	16.65 (4.82)	15.45 (4.34)	15.93 (4.37)		
Trunk									
Max flex (°)		29.14 (8.15)*	28.18 (9.23)†	28.24 (12.10)†	32.68 (10.19)*	31.87 (12.39)†	33.19 (13.71)†		
Max lateral (°)	ND directed	2.08 (6.49)*	4.44 (5.53)†	-1.08 (4.92)†	3.43 (9.93)*†	4.17 (11.36)	0.58 (6.17)†		
Max lateral (°)	D directed	-3.83 (7.51)*	0.25 (6.60)*†	3.84 (6.46)†	4.02 (8.17)	1.18 (5.86)†	4.64 (10.04)†		
Range trunk lateral (°)		9.68 (6.64)	8.08 (5.02)	6.33 (2.90)	11.46 (8.64)	8.60 (6.91)	8.78 (8.81)		

DF, dorsiflexion; PF, plantarflexion; Flex, flexion; Ext, extension ND, non-dominant; D, dominant;

Positive = ND-directed lateral trunk bending

Symbols denote $p < 0.05$

Figure 4.3: Angles and moments in the AP axis at the ND and D ankle, knee and hip with the corresponding skeletal postures of a representative subject (subject 8)

COM displacement and D-directed AP COP displacement in the D foot.

Table 4.6: F-values and main effect p-values for COM and COP metrics

Parameter	Component	F-value	Pr > F
COM displacement	AP	5.20	0.0009
at seat-off	ML	0.98	0.44
COM Velocity at max ND side	ML	2.01	0.098
COM Velocity at min D side	ML	0.32	0.90
COM displacement- ND	ML	3.49	0.010
COM displacement - D	ML	0.76	0.58
Max COM velocity	AP	0.89	0.49
Max COM velocity	Vertical	1.05	0.40
Min COP displacement - ND	AP	4.88	0.0014
Min COP displacement - D	AP	1.86	0.12

COP metrics represent COP under the ND, D and both feet (total COP), rather than the whole system of feet and pole(s). In general, at seat-off during the assisted STS, the COM is posterior to the BOS under the feet, delineated by the left and right malleoli and the left and right fifth metatarsal. Examining post-hoc contrasts revealed subjects using the far pole position demonstrated a COM located closer to the posterior edge of the BOS at seat-off when using the single and double pole configurations compared to the near position ($p = 0.04$ and 0.0001 , respectively). Qualitatively, the COP under the feet also moved more anteriorly with the COM in the far position. In addition, subjects demonstrated a more anterior COP position at seat-off in the far pole position compared to the near pole position for the horizontal, single and double pole configurations (qualitative observation, not examined with an ANOVA). Examining peak COP position under each foot, COP location in the ND foot was more anterior in the far position when using the single and double pole configurations compared to the far position ($p = 0.006$ and 0.03). In addition, when using the far poles, subjects' peak COP was more anterior in the double and single poles compared to the horizontal pole ($p = 0.02$ and 0.001). In the dominant foot, qualitatively, users of the far poles demonstrated a more anterior peak COP displacement compared to the near position.

The magnitude of COM displacement toward the ND-side was greatest when using the horizontal pole in comparison to the single and double pole configurations ($p = 0.05$ and 0.022 , respectively). As shown in Table 4.8, the COP shows the same pattern in the ML direction, where horizontal pole use results in a more ND-directed COP than the single and double pole configurations in the near position (qualitative observation not examined with an ANOVA). In the far position, qualitatively, the ND-directed ML COP displacement was greater in subjects using the horizontal pole versus the double pole. The magnitude of the difference in

Figure 4.4: COM displacement in the ML axis over time, the sagittal plane and the transverse plane with corresponding skeletal postures of a representative subject (subject 1)

COM movement is quite small (approximately 2.8 cm for the greatest COM displacement calculated with a 36 cm BOS). However, the movement also accompanies a significantly greater D-directed ML force applied to the pole ($p = 0.0003$ and $p < 0.0001$) (see section 4.5.2). Figure 4.4 illustrates a representative subject (subject 1) performing this ND-directed movement as a function of time (top-left), in the context of the transverse plane (bottom-left) and associated ML pole forces (bottom-right).

Table 4.7: Summary of center-of-mass metrics for all pole use conditions

Parameter	Component	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
COM displacement at seat-off	AP	-1.02 (0.38)	-1.04 (0.27)*	-1.04 (0.39)†	-0.94 (0.25)	-0.93 (0.26)*	-0.86 (0.33)†
	ML	0.02 (0.11)	-0.01 (0.13)	-0.04 (0.08)	0.04 (0.1)	0.09 (0.1)	-0.01 (0.10)
ND-dir COM displacement D-dir COM displacement	ML	0.10 (0.11)*†	0.05 (0.12)*	0.02 (0.07)†	0.11 (0.09)	0.07 (0.07)	0.07 (0.08)
	ML	-0.08 (0.07)	-0.07 (0.14)	-0.09 (0.09)	-0.04 (0.06)	-0.07 (0.12)	-0.07 (0.12)
Max COM velocity Max COM velocity	AP	0.13 (0.05)	0.11 (0.05)	0.11 (0.04)	0.12 (0.04)	0.12 (0.05)	0.12 (0.05)
	Vertical	0.21 (0.05)	0.21 (0.05)	0.22 (0.05)	0.19 (0.05)	0.21 (0.05)	0.21 (0.04)

COM, center of mass; ND, non-dominant side; D, dominant side; AP, anterior-posterior; ML, medial-lateral

AP COM values are relative to posterior border of base of support; positive = anterior displacement of COM; ML COM values are relative to midline of BOS; positive = ND

directed displacement

AP COM values normalized to foot length; ML COM values normalized to width of BOS; COM velocity normalized to height

Symbols denote $p < 0.05$

Table 4.8: Summary of center-of-pressure metrics for all pole use conditions

Parameter	Component	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
Total COP displacement at seat-off	AP	0.22 (0.20)	0.24 (0.22)	0.31 (0.22)	0.29 (0.21)	0.31 (0.18)	0.34 (0.22)
	ML	-0.02 (0.16)	-0.001 (0.10)	0.007 (0.1)	0.04 (0.10)	-0.001 (0.1)	-0.063 (0.06)
Peak COP displacement - ND foot Peak COP displacement - D foot	AP	0.21 (0.28)	0.20 (0.26)*	0.25 (0.27)†	0.21 (0.29)†●	0.27 (0.29)*●	0.30 (0.28)†‡
	AP	0.30 (0.27)*	0.30 (0.24)	0.33 (0.28)	0.36 (0.27)*	0.35 (0.22)	0.36 (0.30)
Max total COP displacement Min total COP displacement	ML	-0.26 (0.15)	-0.20 (0.15)	-0.20 (0.20)	-0.23 (0.14)	-0.20 (0.13)	-0.1 (0.11)
	ML	0.18 (0.18)	0.17 (0.12)	0.13 (0.09)	0.14 (0.18)	0.19 (0.13)	0.21 (0.12)

COP, center of pressure; ND, non-dominant side; D, dominant side; AP, anterior-posterior; ML, medial-lateral

Total COP represents COP under both feet and COP - ND/D represent COP under respective foot

COP values are relative to posterior border of base of support; positive = anterior displacement of COP (AP) or ND directed displacement of COP (ML)

AP COP values normalized to BOS length; ML COP values normalized to width of BOS; COM velocity normalized to height

Symbols denote $p < 0.05$; ANOVA only conducted on selected COP metrics (see Table 4.6)

4.5 Kinetics

4.5.1 Joint Moments

Mean and standard deviation of sagittal plane joint moments across all configurations are summarized in Table 4.10. Ankle, knee and hip joint moments from a representative subject (subject 8) are displayed in Figure 4.3. The ANOVA revealed main effects for ankle dorsiflexion, plantarflexion and hip extension moments in the ND side across the different pole use conditions, as shown in Table 4.9. Conversely, the knee and D sided hip remained unaffected by pole configuration and position (see Table 4.9).

Table 4.9: F-values and main effect p-values for joint moment data

Parameter	Side	F-value	Pr > F
Ankle	ND	10.12	<.0001
Max DF (Nm/kg)	D	3.20	0.016
Ankle	ND	10.84	<.0001
Max PF (Nm/kg)	D	6.57	0.0002
Knee	ND	0.88	0.50
Max ext (Nm/kg)	D	1.70	0.16
Hip	ND	2.96	0.023
Max ext (Nm/kg)	D	1.52	0.21

In general, maximum ankle dorsiflexion moments were very small throughout the STS task and in some cases the moment generated at the ankle remained negative (ie. plantarflexed). The ND peak ankle dorsiflexion moments were all positive with the exception of the single and double poles in the far configuration which were 0 Nm/kg and -0.03 Nm/kg, respectively. Conversely, the D sided peak ankle dorsiflexion moments were primarily negative (ie. plantarflexion), save for horizontal and single pole configuration in the near position which were 0.01 Nm/kg and 0 Nm/kg, respectively. Post-hoc contrasts revealed, in the near pole positions, subjects exhibited a significantly greater maximum ND ankle dorsiflexion moment when using the horizontal pole configuration compared to the single and double pole configurations ($p = 0.045$ and 0.018 , respectively). In the far position, horizontal pole use also resulted in significantly greater ND maximum ankle dorsiflexion moments compared to the double pole configuration ($p = 0.001$). Comparing near versus far, all near pole configurations experienced greater peak dorsiflexion moments ($p = 0.003$, 0.03 and 0.0001). On the D side comparing near and far pole position, contrasts did not reveal significant differences in maximum ankle dorsiflexion moment across different pole configurations.

Maximum ankle plantarflexion moment corresponds to the absolute maximum moment at the ankle. This typically occurs in the extension phase of the task and is significantly greater in the far position compared

to the near position across all configurations in D ($p = 0.0004, 0.003$ and 0.01 , respectively) and ND ankles ($p = 0.002, 0.0002$ and <0.0001 , respectively).

Finally, subjects experienced significantly greater hip extension moments in the ND side when using the single pole configuration compared to the horizontal pole configuration in the far position ($p = 0.03$). In addition, single pole use required greater ND hip extension moments in the far position compared to the same pole in the near position ($p = 0.03$). ND hip extension moments were also greater when using the double pole configuration in the far position compared to the near position, however this was only approaching significance ($p = 0.06$). No significant differences were found in the D hip (see Table 4.9).

Joint moment asymmetry

Table 4.11 summarizes the asymmetries in joint kinetics between the dominant and non-dominant side. A value greater than one corresponds to ND sided kinetic measures which are greater than those on the D side. An ANOVA was not completed on asymmetry values as the primary focus was to identify the presence of asymmetry, rather than compare relative differences between conditions. Based on visual inspection of the asymmetry ratios, asymmetries of peak extension moments were not evident to a great degree in the knee joints. The horizontal pole configuration in the far position revealed the greatest asymmetry taking into account mean D and ND knee moments and asymmetry ratios. Examining the hip joint, across all configurations and locations, hip joint asymmetry values were all greater than one. The greatest asymmetry was a 36% mean increase in hip extension moments in the ND side when using single pole configuration in the far position. The least asymmetry was seen in the horizontal pole configuration in the near position, which demonstrated an 8% mean increase in maximum ND hip moments.

4.5.2 Reaction Forces

Mean and standard deviation of reaction forces from the ground, seat and pole in AP, ML and vertical directions across all configurations are summarized in Table 4.13. Reaction force plots for a representative subject (subject 3) are displayed in Figure 4.5. The ANOVA identified main effects in GRFs on the D side in the AP direction and vertical direction and on both sides in the medial and lateral direction as shown in Table 4.12. P-values were just over 0.05 for vertical GRFs in the ND side (see Table 4.12). The ANOVA also identified main effects in ML and vertically directed pole forces (see Table 4.12). No main effects were identified in maximum AP forces in the ND foot, peak AP pole forces and seat forces (see Table 4.12).

Table 4.10: Summary table of joint moments in the sagittal plane for all pole-use conditions

Parameter	Side	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
Ankle	ND	0.04 (0.08)*†‡	0.02 (0.09)*•	0.02 (0.09)†^	0.01 (0.07)†~	0.00 (0.07)•i	-0.03 (0.07)^~i
Max DF (Nm/kg)	D	0.01 (0.12)*	0.00 (0.09)	-0.01 (0.09)	-0.02 (0.10)*	-0.02 (0.08)	-0.03 (0.08)
Ankle	ND	-0.22 (0.14)*	-0.21 (0.10)†	-0.24 (0.10)†	-0.29 (0.12)*	-0.29 (0.11)†	-0.32 (0.12)†
Max PF (Nm/kg)	D	-0.21 (0.18)*	-0.21 (0.16)†	-0.22 (0.17)†	-0.29 (0.17)*	-0.27 (0.17)†	-0.28 (0.16)†
Knee	ND	-0.92 (0.24)	-0.90 (0.16)	-0.87 (0.17)	-0.86 (0.15)	-0.88 (0.16)	-0.89 (0.16)
Max ext (Nm/kg)	D	-0.93 (0.21)	-0.84 (0.28)	-0.89 (0.20)	-0.91 (0.19)	-0.90 (0.18)	-0.88 (0.18)
Hip	ND	-0.60 (0.24)	-0.60 (0.28)*	-0.55 (0.19)	-0.60 (0.27)†	-0.68 (0.31)*†	-0.62 (0.25)
Max ext (Nm/kg)	D	-0.55 (0.17)	-0.50 (0.26)	-0.49 (0.22)	-0.55 (0.19)	-0.53 (0.17)	-0.48 (0.21)

DF, dorsiflexion; PF, plantarflexion; Flex, flexion; Ext, extension ND, non-dominant; D, dominant;

Moments normalized to body weight

Symbols denote $p < 0.05$

Table 4.11: Summary of ND:D asymmetry ratios of maximum extension moment in the knee and hip

Parameter	Near			Far		
	Horizontal	Single	Double	Horizontal	Single	Double
Knee						
Max ext (Nm/kg)	0.96 (0.30)	0.99 (0.20)	0.97 (0.21)	0.94 (0.18)	0.96 (0.17)	0.98 (0.21)
Hip						
Max ext (Nm/kg)	1.29 (0.86)	1.33 (0.77)	1.31 (0.69)	1.08 (0.61)	1.36 (0.75)	1.34 (0.76)

AP, anterior-posterior; Ratio greater than one represents ND > D
 Statistical analysis was not conducted on asymmetry ratios

Table 4.12: F-values and main effect p-values for pole, seat and ground reaction force data

Parameter	Side	F-value	Pr > F
Max ground reaction force	ND	1.56	0.19
AP direction	D	2.80	0.029
Max ground reaction force	ND	5.16	0.0010
Medially directed	D	2.78	0.030
Min ground reaction force	ND	4.12	0.0041
Laterally directed	D	4.01	0.0048
Min ground reaction force	ND	2.37	0.057
Vertical direction	D	2.56	0.042
Min pole force - AP		1.99	0.10
Max pole force - ND		2.80	0.029
Min pole force - D		14.49	<.0001
Min pole force - vertical		4.99	0.0012
Seat force - AP		0.54	0.75

Ground reaction forces

In the near pole position, subjects experienced a lower AP directed ground reaction force while using the horizontal pole in comparison to the double pole on the D side ($p = 0.01$). Similarly, on the D side in the far position, when using the horizontal pole, peak AP GRF was significantly lower in comparison to use of a single pole ($p = 0.02$). Pole position (ie. comparing near versus far) did not affect peak AP GRFs.

Analysis of the ML directed ground reaction forces revealed the peak medially-directed GRFs differed significantly across all pole configurations in the near position on the D side. Examining contrasts showed that horizontal pole use resulted in significantly greater peak medially-directed ND force applied to the feet compared to the double pole ($p = 0.05$). Similarly, in the far position, peak medial GRFs applied to the ND foot were greater while using the horizontal pole compared the double pole ($p = 0.004$). On the D side, horizontal pole use resulted in the greatest peak medial GRFs applied to the D foot compared to the single and double pole configurations ($p = 0.0001$ and 0.0006). Comparing near and far positions, the horizontal pole used in the near position resulted in greater peak medial GRFs applied to the D foot ($p = 0.009$). Similar trends were found in the peak laterally-directed GRFs applied to the feet. Use of the double pole resulted in the greatest peak lateral GRFs applied to the ND foot compared to the single and horizontal pole configurations in the near position ($p = 0.012$ and 0.014). In the far position, use of single and double pole configurations resulted in application of greater peak lateral GRFs applied to the ND foot compared to the horizontal pole ($p = 0.009$ and 0.014). Finally, the ND foot experienced significantly greater peak lateral GRFs in the far position compared to the near position when using the single pole configuration ($p = 0.02$). On the D side, horizontal pole use in the near position resulted in greater peak laterally-directed GRFs applied to the D foot compared to the single and double pole configurations ($p = 0.001$ and 0.002). Comparing near versus far, peak lateral GRFs applied to the D foot were greatest when using the horizontal pole configuration in the near position compared to the far position ($p = 0.05$).

In terms of vertically directed GRFs, post-hoc contrasts revealed greater peak vertical GRFs applied to the D foot when using the single pole in the far position compared to the near position ($p = 0.05$). Lastly, subjects using the horizontal pole in the far position experienced greater vertical GRFs applied to the D foot compared to the double pole ($p = 0.03$). Qualitatively, peak vertical GRFs applied to the ND foot demonstrated a trend for increasing peak vertical GRFs when using the single and double pole configurations in the far position compared to the near position.

Table 4.14 summarizes asymmetries in the GRFs across all pole use conditions. Asymmetry ratios greater than one demonstrate a increased GRF applied to the ND foot compared to the D foot. Horizontal pole use in the near and far position resulted in the greatest asymmetrical AP GRF application to the feet. These

ratios describe a greater AP GRF applied to the D foot compared to the ND foot which is also shown in the mean peak GRFs (see Table 4.10). In the vertical direction asymmetry ratios were close to one, however from the mean peak vertical GRFs in Table 4.10, users of the horizontal pole experienced greater vertical GRFs in the D foot compared to the ND foot.

Pole forces

Post-hoc contrasts revealed that subjects exerted the greatest vertical force on the double poles compared to the horizontal and single pole in the near position ($p = 0.007$ and 0.03 , respectively). Comparing near versus far pole position, greater vertical forces were exerted in the near position when using the single and double pole configurations ($p = 0.04$ and 0.004 , respectively). D-directed ML force applied to the horizontal pole was greater than the single and double configurations in the near position ($p = 0.0003$ and $p < 0.0001$, respectively), followed by the single pole which was significantly greater than double pole ($p = 0.015$). A similar trend was found in the far position where subjects applied the greatest maximal ND-directed forces to the horizontal pole compared to the single and double pole configurations for both near ($p = 0.05$ and $= 0.01$, respectively) and far pole position ($p = 0.03$ and 0.03 , respectively). No statistically significant differences were found in ML pole forces comparing near to far pole position.

4.6 Work and Angular Impulse

Mean and standard deviations for work and angular impulse are summarized in the Table 4.15. Pole configuration and position had no effect on work done and the angular impulse in the knee and hip joints in the sagittal plane, as shown in Table 4.16. Angular impulse and work done was greater in the knee compared to the hip. Visual inspection of asymmetry revealed a greater magnitude of work done in the ND hip when using the single and horizontal pole in the far position.

Table 4.13: Summary of reaction forces in the pole, seat and feet in the AP, ML and vertical axes

Parameter	Side	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
Peak ground reaction force	ND	-0.99 (0.21)*	-1.04 (0.19)	-1.10 (0.19)*	-0.97 (0.18)†	-1.07 (0.23)†	-1.04 (0.23)
AP direction	D	-1.13 (0.23)	-1.06 (0.20)	-1.12 (0.18)	-1.09 (0.12)	-1.06 (0.16)	-1.02 (0.14)
Peak ground reaction force	ND	-0.36 (0.24)*	-0.33 (0.28)	-0.30 (0.24)*	-0.36 (0.28)†	-0.33 (0.27)	-0.27 (0.23)†
Medially directed	D	0.46 (0.22)*††	0.39 (0.17)*	0.40 (0.19)†	0.42 (0.20)†	0.40 (0.18)	0.39 (0.18)
Peak ground reaction force	ND	0.33 (0.18)*	0.33 (0.20)††	0.37 (0.18)*†	0.33 (0.18)●^	0.37 (0.20)†●	0.37 (0.18)^
Laterally directed	D	-0.45 (0.24)*††	-0.39 (0.17)*	-0.39 (0.21)†	-0.42 (0.20)†	-0.39 (0.20)	-0.38 (0.20)
Peak ground reaction force	ND	4.39 (0.58)	4.66 (0.61)*	4.50 (0.55)†	4.57 (0.63)	4.83 (0.72)*	4.72 (0.55)†
Vertical direction	D	4.80 (0.63)	4.60 (0.38)*	4.41 (0.85)	4.85 (0.47)†	4.72 (0.40)*	4.62 (0.41)†
Peak pole force - AP		2.00 (0.39)*	2.13 (0.52)	2.26 (0.51)*	2.17 (0.36)	2.27 (0.61)	2.20 (0.48)
Peak pole force - ND-dir		-0.10 (0.18)*†	-0.20 (0.19)*	-0.23 (0.12)†	-0.10 (0.13)†●	-0.23 (0.31)†	-0.23 (0.15)●
Peak pole force - D-dir		0.41 (0.22)*†	0.23 (0.16)*†	0.12 (0.08)††	0.36 (0.18)●^	0.19 (0.16)●	0.13 (0.08)^
Peak pole force - vertical		1.67 (0.62)*	1.72 (0.49)††	1.90 (0.67)*†●	1.58 (0.65)	1.54 (0.69)†	1.66 (0.68)●
Peak seat force - AP		0.18 (0.16)	0.19 (0.14)	0.14 (0.13)	0.19 (0.14)	0.19 (0.15)	0.21 (0.12)

Ground and seat reaction forces: positive = anterior, upward and ND directed forces applied to the segment; ND- and D-directed forces on the pole describe direction of applied force to the pole

AP, anterior-posterior; ND, non-dominant; D, dominant;
 Symbols denote $p < 0.05$

Table 4.14: Summary of ND:D AP and vertically directed ground reaction force asymmetries

Parameter	Near			Far		
	Horizontal	Single	Double	Horizontal	Single	Double
Max ground reaction force AP direction	0.82 (0.19)	1.04 (0.22)	0.96 (0.15)	0.82 (0.11)	0.95 (0.22)	0.96 (0.20)
Min ground reaction force Vertical direction	0.92 (0.13)	1.05 (0.12)	0.97 (0.20)	0.99 (0.14)	1.06 (0.14)	1.04 (0.10)

AP, anterior-posterior; Ratio greater than one represents ND > D
Statistical analysis was not conducted on asymmetry ratios

Table 4.15: Summary table of work done and angular impulse for the knee and hip joints in the sagittal plane

Parameter	Side	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
Knee	ND	-0.11 (0.049)	-0.11 (0.052)	-0.098 (0.053)	-0.097 (0.048)	-0.10 (0.051)	-0.11 (0.058)
Angular impulse (Nms/kg)	D	-0.11 (0.047)	-0.10 (0.044)	-0.10 (0.039)	-0.10 (0.031)	-0.10 (0.036)	-0.10 (0.045)
Knee	ND	3.30 (1.34)	3.33 (1.35)	3.18 (1.23)	3.06 (1.32)	3.13 (1.27)	3.21 (1.27)
Work done (J/kg)	D	3.41 (1.31)	3.18 (1.40)	3.25 (1.14)	3.16 (0.96)	3.15 (1.05)	3.06 (0.97)
Hip	ND	-0.063 (0.031)	-0.054 (0.047)	-0.052 (0.031)	-0.056 (0.036)	-0.067 (0.048)	-0.063 (0.036)
Angular impulse (Nms/kg)	D	-0.066 (0.045)	-0.056 (0.057)	-0.054 (0.047)	-0.065 (0.053)	-0.066 (0.052)	-0.060 (0.051)
Hip	ND	1.31 (0.66)	1.17 (0.85)	1.16 (0.64)	1.28 (0.78)	1.34 (0.91)	1.22 (0.67)
Work done (J/kg)	D	1.32 (0.74)	1.11 (0.74)	1.20 (0.80)	1.21 (0.70)	1.25(0.60)	1.12 (0.67)

ND, non-dominant side; D, dominant side

No statistical differences found for measures of work or angular impulse

Figure 4.5: Reaction forces at the pole and the ND and D foot in the AP, ML and vertical directions with corresponding skeletal postures of a representative subject (subject 3)

Table 4.16: F-values and main effect p-values for work done and angular impulse at the knee and hip

Parameter	Side	F-value	Pr > F
Knee	ND	0.76	0.5838
Angular impulse	D	0.26	0.9307
Knee	ND	1.09	0.3796
Work done	D	1.29	0.2883
Hip	ND	1.73	0.1497
Angular impulse	D	1.15	0.3495
Hip	ND	1.45	0.2262
Work done	D	1.52	0.2052

Chapter 5

Discussion

Grab bars and poles are assistive devices that enable the use of the upper body to offset loads in the lower limbs and increase stability during the STS movement. Current literature describing the mechanics of rising from a chair with assistance from different configurations and positions of grab-bars and vertical poles is limited [10–12], particularly for individuals with impaired mobility [11]. This is the first study to provide a comprehensive evaluation of bilateral STS biomechanics related to the use of different vertical pole configurations and locations in older adults with limited mobility. The main findings are horizontal force application on the pole replaces trunk generated horizontal momentum. In addition, the far pole position increased plantarflexion moments which contributed to controlling forward movement. The far positioned pole resulted in reduced vertical pole forces, increased vertical forces and increased exertion of lower limbs evidenced by increased non-dominant hip moments. The horizontal pole configuration caused an asymmetric movement of the COM toward the non-dominant side and asymmetric forces applied to the floor by the dominant foot. The asymmetry in GRFs countered the moment about the COM in the transverse plane imparted by the use of a unilaterally placed pole (horizontal pole configuration). Lastly, hip moment asymmetry was evident with a greater moment about the non-dominant hip. The results of this study will assist in understanding mechanics of the assisted STS task and how it varies with pole position and configuration. This will help build an evidence base to provide recommendations for grab-bar design, installation in the home and clinic and strategy for use to assist clinicians, caregivers and persons with impaired mobility.

5.1 COM position and momentum during assisted rising

Increasing hip and trunk flexion in the unassisted STS is common in older adults who aim to bring their COM over their BOS prior to seat-off to ensure a stable posture is achieved at the point of the STS that is considered the most unstable [7,49]. This also enables the individual to rise without the use of high horizontal momentum that may also threaten balance. With pole assisted STS, hip and trunk flexion increased significantly in the far condition when compared to poles placed closer to the individual. This was a consequence of reaching for a pole positioned farther away. As a result, the COM location at seat-off was closer to the posterior border of the BOS compared to the near condition. Without considering the additional support provided by the pole, this is considered a more stable position from a biomechanical perspective. Previous literature investigating the unassisted STS has suggested that a farther COM relative to posterior border of the BOS in the AP direction at seat-off and/or lower AP COM velocity at seat-off puts the individual at risk for sit-back failure [74,75]. Pai and colleagues [43] state that balance during STS is based on COM velocity-position limits, where a COM located posterior to BOS can still be tolerated without a fall if COM velocity is sufficiently large in the direction of the BOS. Conversely, a high COM velocity near the limits of the BOS puts the individual at risk of falling. However, when a grab-bar is introduced that enables force application by the upper limbs. This model, which describes the risk of a fall due to inadequate maintenance of the COM within the BOS, is no longer suitable. The grab-bar, similar to handrail use during standing: (1) increases the size of the BOS, (2) enables the user to generate stabilising forces and moments by the upper body to counteract instability and (3) provides a rigid support for the user to grasp and anchor the body [76]. Therefore, in terms of stability, although rising from a chair while grasping a far pole may bring the COM closer to the BOS, it is unknown whether this translates into a particular advantage for the user. It should be noted, however, that the use of a pole placed farther from the individual increases the joint extension moments in the non-dominant hip and both ankles. Therefore, the far pole position provides no particular advantage or disadvantage in terms of stability, but increases hip extension and ankle plantarflexion moments, placing greater demand on the musculature, a drawback of a grab-bar placed further away from the user when rising from a chair.

Horizontal velocity of subjects in this study were less than half the magnitude of horizontal velocity reported in the literature for the unassisted STS [23,40,77]. Low-momentum strategies during grab-bar assisted STS has been documented in previous literature [10,12]. Low horizontal momentum indicates reduced trunk generated horizontal momentum, which is not a requisite during the assisted STS because it is replaced by horizontal forces applied to the pole. This is evident as the horizontal forces in the pole approach their peak prior to seat-off, when forward propulsion is required to bring COM into the BOS to then successfully

stand up. The control task in the unassisted STS involves the generation of enough horizontal momentum to ensure that the COM moves into the BOS after seat-off (avoiding sit-back failure), followed by decelerating the COM to keep it from overshooting the anterior border of the BOS and causing a forward fall [23, 43]. In the assisted STS movement, the control task changes from managing rapid increases in momentum to coordinating coupling forces of the hand at the pole and the feet on the ground [10].

5.2 Medial-lateral COM and kinetic asymmetry

A common movement strategy during the assisted STS transfer involved a COM displaced to the non-dominant side of the midline of the BOS in the ML direction. In the context of the STS transfer users would begin by (a) leaning toward the non-dominant side, (b) move farther in the direction of the non-dominant side for a short period after seat-off and then (c) begin moving back toward the midline as full standing is approached. The strategy was particularly evident when using the horizontal pole which demonstrated the greatest ML COM displacement directed toward the non-dominant side in the near and far pole position. This asymmetrical movement is accompanied by significantly greater ML forces applied laterally to the pole in the non-dominant direction. Figure 5.1 illustrates the movement of the COM during this strategy and the associated lateral forces in the pole and Figure 5.2 illustrates the balance of forces about the COM in the transverse plane.

This strategy was associated with asymmetries in GRFs under the feet corresponding to greater posteriorly- and laterally directed GRFs applied to the dominant lower limb. Similar findings have been reported in healthy older adults when using a unilateral grab-bar positioned on the dominant side [10]. The authors explain that when forces are applied to a unilateral grab-bar, larger forces applied to the feet are a motor control strategy used to counter the moments about the COM in the transverse plane generated by forces applied to the grab-bars to maintain a linear COM trajectory.

Application of force on the horizontal component of the horizontal pole configuration can aid in balancing the moments about the COM in the event that the user does not have the musculature to apply them to the floor. However, a single vertical pole placed laterally may require greater strength in the lower limbs to apply adequate horizontal forces to the floor. This may have implications for frail older adults with severe disabilities or joint pain. The additional musculature required to apply these horizontal forces at the feet and accompanying joint contact forces may result in STS failure. Further study is required to assess the effects of increased lateral distance of a pole on the STS movement.

Figure 5.1: COM displacement and corresponding pole forces in the ML direction

Figure 5.2: Transverse plane representations of force application at the pole and feet during horizontal pole use at the near condition. Arrows represent mean forces at the pole and feet for the horizontal pole configuration the near position

5.3 Physical exertion of the lower extremities

The STS movement requires large internal extension moments in the knee and hip at the point of seat-off to extend the knees and hip and bring the individual into the standing position. Internal plantarflexion moments oppose forward movement during the extension phase and aid in deceleration of the COM.

For this study, physical effort was characterized through variables of maximum extension moment, work done and angular impulse. No changes in angular impulse and work done in the hip and knee were found across different pole conditions tested in the near and far position. In addition, bilateral maximum knee and dominant hip extension moments were unaffected by pole configurations in the near and far positions. The findings agree with O'Meara and colleagues who found no changes in bilateral reductions in hip and knee extension moments comparing laterally versus anteriorly placed grab-bars [9]. Wretenberg and colleagues also found no difference in peak extension moments in the hip and knee comparing different arm rest heights and grab-positions [66]. Kinoshita and colleagues [11] tested a low (60 - 80 cm), high (90 - 110 cm) and combined low/high grab-bar and found very little difference between the three configurations among healthy older adults. However, among physically impaired older adults, the high/low grab-bar configuration reduced total moments substantially by 67% and 60% compared to the low and high grab-bars, respectively.

Pole position affected plantarflexion moment where a far pole position resulted in significantly greater plantarflexion moment. During the STS transfer the plantarflexors in act to maintain postural stability at the termination of STS [43]. Activation of the ankle plantarflexors are required to eccentrically control the forward translation of the body during the extension phase of the STS transfer, to ensure COM does not fall outside the anterior border of the BOS. Increases in plantarflexion moments are characteristically a result of increased speed of rising which requires a greater plantarflexion moment to decelerate the COM [42]. In this study, horizontal COM velocity did not change with pole position. Therefore, it is likely caused by the more anteriorly positioned COP under the feet as a result of reaching for a far pole, which in turn activates plantarflexors and placing additional demand on the distal musculature.

The far pole condition resulted in significant increases in non-dominant hip extension moment when using the single pole and a trend to increasing non-dominant hip moments when using the double pole configuration. Compared to isometric strength values reported in the literature, voluntary hip moments are well below of total available strength in hip extensors [78]. Among the population assessed small increases in hip moment, as well as ankle plantarflexor moments, are not likely to cause unsuccessful rising in a population of older adults with functional impairments. However, these findings highlight the notion that pole position can affect the demands on the muscles required to rise. Increase in pole distance was considerably small in the far compared to near conditions (mean difference of 15.1 cm). A more exaggerated distance would likely

induce greater demands on extension moments in the lower limbs. In this vein, far pole positions may limit very frail older adults with joint pain from successful rising.

It is important to understand why a farther pole may result in increased extension moments in the lower extremities and the corresponding clinical implications related to demands on the muscles. At the point of seat-off, the load has been transferred from the seat to the feet and vertical GRFs under the feet are at a maximum. This is combined with maximal hip and knee joint flexion angles. From a biomechanics perspective, the force vector lies posterior to the knee and anterior to the hip, at the greatest distance to the knee and hip joint centers, respectively. As a result, this is the point of maximum knee and hip extension moments. Given the location of the vector at this point, a change in joint moments by modifying the location of the vector relative the joint centers results in a trade-off between peak hip and knee extension moments, where the force vector closer to the hip joint center would reduce the hip moments but result in an increased knee moment and vice versa. Therefore, the most effective way to reduce both hip and knee extension moments is to reduce the magnitude of the vertical GRF vector during this highly flexed posture following seat-off. Grab-bars and poles enable transfer of vertical loads under the feet to the assistive device and reduce vertical component of the GRFs and corresponding extension moments in the knee, hip and ankle proportional to the applied pole forces. Peak vertical forces in the single and double pole in the far position reduced significantly compared to corresponding near poles by 10% and 13%, respectively. This corresponded to an increase in GRFs under the feet in the single and double pole configuration (significant for the D side) which are the likely cause of the increase in hip moments in the non-dominant side in corresponding pole configurations. Similar findings were seen in previous literature [12] which compared natural grab-bar use to strong grab-bar use (ie. individuals were instructed to rise from the chair placing greater force through the grab-bar). In this particular study, the authors found that greater application of force to the pole resulted in a reduction in the moments in the hip and knee substantially (greater than 50%).

5.4 Comparison of grab-bars and poles to arm rests

Unassisted STS was not tested in the current study because the sample population was specifically recruited based on their inability to stand without assistance. Previous work has investigated grab-bar use compared to unassisted STS, with varying results. O'Meara et al. found no differences among healthy older adults in bilateral hip and knee extension moments when individuals used a grab-bar compared to rising from a chair unassisted [9]. Bahrami et al. [12] reported reductions of 13% and 17% in maximum knee and hip, respectively, among healthy adults with grab-bar use. In the latter study, it should be noted that symmetry of the lower limbs was assumed, and therefore only dominant side data are reported.

Interestingly, differences in lower limb extension moments from grab-bar studies are much smaller than those reported in studies investigating the effects of arm rests on STS [7, 52, 63, 66], which report substantial reductions of greater than 50% in maximum hip and knee extension moments when rising from a chair with arm rests compared to unassisted. The likely explanation for the benefit of arm rests relates to the magnitude of vertical force applied to the respective STS aid. As shown in Table 5.1, vertical forces applied to arm rests [66] are greater than vertical force applied to the poles in this study and the grab-bars reported in the literature [10, 12]. This is potentially related in part to the distance of the shoulder from the point of application of the force on the STS aid. Pushing down on arm rests located below your shoulders results in a smaller moment arm relative to the shoulder joint center compared to that of applying a vertical force to an anteriorly positioned pole, thereby providing an apparent biomechanical advantage for the use of arm rests. It follows that with the ability to apply greater vertical force to the assistive device, a concomitant decrease in the vertical force applied at the feet occurs, which in turn results in reduced lower limb moments (as previously discussed above). The current study did not investigate chair rise with arm rests, thereby limiting definite conclusions. However, comparisons with published literature are interesting enough to warrant further study.

There is a likely trade-off with arm rest use, which is reduced stability. The user has to eventually let go of the arm rest and maintain balance with abductors and adductors in the legs as opposed to using the forces on the pole to help balance. In addition, horizontal forces applied to arm rests are lower, reducing their contribution to horizontal propulsion. The ideal STS aid should incorporate the advantages of stability contributed by the transfer pole and the biomechanical advantages for upper limb force application of arm rests. Interestingly, Kinoshita and colleagues tested a device similar to this description. It included a lower bar positioned similar to an arm rest held by the dominant hand and a higher bar, placed at approximately shoulder height, held by the non-dominant hand. This configuration incorporated the biomechanical advantages of a grab-bar placed close to the trunk and the stability benefits gained from holding a grab-bar for the entire duration of the STS movement. As previously mentioned, among physically impaired older adults, this high/low grab-bar configuration reduced total support moments substantially by 67% and 60% compared to the low and high grab-bar conditions tested, respectively.

5.5 Hip moment asymmetry

Across all configurations asymmetries in maximum extension moments were observed. Hip extension moments in the non-dominant side were consistently greater by an average of 30% across all pole configurations and positions with the exception of the horizontal configuration in the far position, which only showed an 8% asymmetry. Biomechanically, the changes in kinetics that contribute to a reduced hip moment include a

Table 5.1: Summary of forces applied to grab-bars, arm rests and poles from the literature

Assistance type	Vertical force (N/kg)	Vertical force (N)
Single-far	1.54	116
Single-near	1.72	129
Arm rest (low) [66]	3.25	—
Arm rest (high) [66]	2.17	—
Arm rest (anterior) [7]	—	92
Grab bar (horizontal) [12]	0.79	—
Grab bar (anterior) [10]	0.39	—
Grab bar (lateral) [10]	0.59	—

decrease in the vertical GRF, an increase in posteriorly directed (AP) GRFs and a more posteriorly located COP. The effects of changing the GRFs and COP locations in isolation is illustrated in Figure 5.3. The FBD represents the static situation (ignoring inertia), in the sagittal plane, when all other GRFs and the COP location are kept constant. For example, in the far left image in Figure 5.3, the resultant vector shown as the dotted blue line is a result of increasing the vertical GRFs while keeping AP force and COP location constant.

Posteriorly located COP and increases in posterior GRFs will both direct the force vector closer to the hip joint center and result in reduced hip moments. In this study, hip moment asymmetries in subjects using the horizontal pole are likely due to the increase in AP force experienced in the dominant limb. O’Meara and colleagues report similar asymmetries in grab-bar assisted STS, with a reduction in hip moments in the dominant hip [10]. In single and double pole configurations in the far position, greater vertical GRF in the non-dominant side likely contributed to increases in non-dominant hip moment and the corresponding asymmetry. It is difficult to identify the cause of hip asymmetries in the remaining configurations as none of the above mentioned kinetic variables that contribute to hip moment asymmetry present solely. Some combination of changes in multiple factors are likely responsible.

Asymmetrical moments in the lower extremities during the assisted STS implies increased burden on one side and/or reduced burden to the opposite side. This may have implication for pole installations and/or strategy for use in different clinical populations. For example, in individuals post-hip fracture, excessive loading of the non-affected hip is associated with development of osteoarthritis in the non-affected leg [79]. Understanding how asymmetries are developed with pole or grab-bar use is important for preventing long term health implications such as osteoarthritis due to pole related asymmetrical movement. Reducing hip moments in one side may also serve as temporary solution for individuals with severe joint pain or in individuals post-operative who are recommended to reduce loading on the affected lower limb for a period. Alternatively, in a stroke population, additional loading through a lower limb can act as a training mechanism for the paretic

limb. Understanding asymmetries will provide insight on setting up poles in the home/clinic, or strategy when using a pole to stand up for a particular clinical population.

Figure 5.3: Effects of changing GRFs on moments at the ankle, knee and hip. The FBD illustrates the static model at the point of seat-off when all reaction forces are kept constant, except the force in question being varied. Changes in the resultant force vector are shown as the dotted line

Chapter 6

Conclusions and recommendations

The purpose of this study was to examine the various pole conditions (including a single pole, two poles and pole with a horizontal grab-bar positioned at near and far locations from the user) and the effects on the STS transfer. The kinetics and kinematics of the movement were studied to characterize movement and lower limb physical exertion during STS related to the differing vertical pole configurations. In general, horizontal forces applied to the pole during assisted STS replace trunk generated horizontal momentum and reduce the horizontal momentum required to bring the COM into the BOS at seat-off. This has implications for stability, changing the control task from one of managing rapid increases in momentum to that of coordinating coupling forces at the pole and the ground. In terms of pole configuration, the horizontal pole in near and far positions, resulted in a laterally displaced COM trajectory directed toward the non-dominant side and increased AP and ML directed forces in the dominant foot to balance the moment about the COM in the transverse plane. Pole use also introduced asymmetries in hip moments with increased moments consistently experienced in the non-dominant hip. This has implications for a clinical population where pole set up and strategies can be varied to increase or decrease moments on a particular side based on the clinical indications of the population. The distance of the pole was found to have a significant effect on STS biomechanics. The far pole position resulted in significant reduction in vertical force applied to the pole and consequently, increased moments in the non-dominant hip when using the single and double poles compared to the near pole position. A potential cause of reduced vertical force applied to the far pole potentially relates to the biomechanical disadvantage of the shoulder when pulling downwards at a greater distance to the shoulder joint centre. This brings to light the effect of increased demands on the shoulder during upper limb use in assisted STS and the implications for physical exertion of the lower extremities. More research is needed to confirm if laterally placed poles require increased strength to balance forces about the COM and if far pole

configurations increase moments in the lower extremities.

The important clinical outcome is a recommendation that the pole centrally positioned between both shoulders or bilaterally, aligned with each shoulder at a distance of 75% of thumb-tip reach. Place the pole further forward or relying on single pole to one side, will result increased physical demands on the lower limbs.

6.1 Limitations

Variability in kinematics was introduced as a result of non-standardized seated posture. This was mitigated by suggesting participants sit at the edge of the seat with their knees over their toes as recommended by occupational therapists. The trade-off, however, was that the lower limb joint angles were not standardized at initiation of STS. In addition, the experimental set up was such that the force platforms for the left and right feet were located in front of the seat force platform. This limited the degree to which participants could bring their feet under their knees and as a result participants did not stand up from the most biomechanically efficient posture. Literature indicates that a constrained posture can result in an unnatural STS [19, 21], however using an initial controlled angle between the shank and the foot that is biomechanically efficient would be an effective compromise. The initial posture used in this study may impact the translation of these results to individuals rising with a biomechanically efficient posture.

The number of subjects enrolled was lower than anticipated. Enrolling additional subjects may have yielded greater statistical differences, particularly in measures which appeared to be approaching statistical significance. In addition, although the population studied was mobility limited, a more functionally impaired population may have relied on the pole more and demonstrated greater differences between pole conditions.

Unassisted STS was not measured because participants were recruited on the basis of not being able to stand up without assistance. Despite this criteria, some subjects stood up with less apparent effort than expected while using the pole. Harnessing participants and asking them to stand up unassisted would be a useful addition to the protocol to gain greater insight into the effect of grab-bar use in individuals who require assistance. The inclusion of an arm rest condition would also be effective for comparing transfer pole/grab-bar assistance to a more ubiquitous form of assistance in arm rests. Lastly, additional repeated measures would have reduced variability and would have likely been tolerated by participants. Physical therapists consulted for this study suggested keeping the number of STS transfers performed during testing below 10. As a compromise, repeated measures and the arm rest condition initially included in the protocol were removed. None of the participants complained about fatigue at the end of the trial and a greater number of STS trials would likely have been tolerated by participants.

The experimental set up was designed and constructed over a very short period to collect data as soon as possible in anticipation of a slow recruitment of the clinical population. As a result, the transfer pole test bed was not designed for quick transition between conditions and when testing a new condition, changing configuration and position of the pole took between 5 and 10 minutes. A testing session typically lasted two hours; testing 6 conditions translates to up to an hour consumed on transition between conditions alone. Time spent on transition between conditions can limit the number of conditions tested in a single test session. In addition, waiting for the new condition would sometimes cause the subject to become impatient. Lastly, spacers between the force platforms set up in the lab for other experiments are necessary for easy transfer from the entrance to the chair. Ideally, the lab floor should be level such that the participant can wheel or use their walker to get to the chair. Instead, participants were required to step up and down from force plates which was very difficult for all participants tested.

Measures of balance were not calculated for this thesis. The data were collected, however given time constraints it did not make it into the report. Publications from this study will compare different balance metrics across pole conditions. In addition, forces on the thighs from the seat were not considered which effects the accuracy of hip moment data prior to seat-off. Very little forces were applied from the seat as most of the forward propulsion came from the pole, but I cannot say with certainty how this effects hip moments prior to seat-off.

Kinetics and kinematics of the upper limbs were not calculated and forces in the dominant and non-dominant hands were not collected. Upper limb mechanics would generate additional insight into the effect of increased shoulder moments on production of vertical and horizontal pole force. In addition, whether or not particular movement strategies are performed in an effort to bring the shoulder close to the pole and reduce moments in the shoulder.

Kinetic asymmetries calculated in this study were evaluated based on ratios. Future work with this dataset will be to conduct another ANOVA and incorporate the dominant and non-dominant sides as dependent variables to output more accurate measures of asymmetry. This will enable further investigation into the causes and effects of kinetic asymmetry.

6.2 Recommendations

Examination of individual participant data and visual inspection of the STS movement revealed that some participants applied greater forces on the pole as a possible means of offsetting load under their feet and thereby reducing extension moments in the hip, knee and ankle. However, others applied smaller forces to the pole and appeared use the pole mainly to stabilize COM and increase their base of support. It

would be interesting to separate participants based on balance and strength deficits and evaluate pole use and corresponding markers of exertion in the lower extremities. Studies that evaluated grab-bars [9, 10] and found little differences in extension moments across different grab-bar types and no assistance state that participants only use the grab-bar to increase BOS and stabilize COM to within the BOS. However these studies use a sample population of healthy older adults who likely utilized the pole out of instruction rather than necessity. Separating populations into strength- and balance-deficient groups would determine how the different groups use the pole and properly address the question of whether or not poles are used predominantly for balance. A normal control population would also be required for this study. The current dataset is lacking the clinical data to make this comparison, however STS research will continue to be studied in this lab. Future work will incorporate this comparison into the study design to investigate differences in assisted STS based on balance- and strength-deficits.

The participant population enrolled in this study could not stand without assistance. However watching participants rise, it appeared as though some participants may successfully rise unassisted if there was a safety measure in place, such as wearing a harness. Harnessing participants and asking them to try their best to rise will provide comparison data to the assisted condition and provide new insight into the role of the STS aid.

Upper limb biomechanics during grab-bar use is nearly absent in the literature. Many of the questions that arose in this study would have benefited from understanding the mechanics in the upper limbs. Future work will adapt the biomechanical model to include upper limbs and additional STS work will study the changes in upper limb kinematics and kinetics with varying forms of assistance.

As mentioned in limitations time required to switch between test conditions was long and as a result a substantial proportion of the test was consumed by switching between conditions. Experimental test beds should be designed to minimize time between test conditions to enable more test conditions and a more pleasant experience for the participants.

Participants in this study moved the COM away from the midline of the BOS significantly more than the double and single pole configurations in the near position. However, the implications this has on balance is unknown because the effects of grab-bars on stability during STS is unknown. Most of the literature will reference models of stability developed by investigating the unassisted STS [75] however these models do not consider the added stability of grabbing a pole. Future work will be conducted using the data collected in this study to compare the change in COP under the feet to the change in COP of the entire system to determine the relative contribution of the pole in maintaining stability. Calculating the total path of the forces applied to the pole at standing and comparing that to COP movement under the feet might also be an interesting metric of balance. In addition, looking at high frequency content of the pole force might also

Figure 6.1: CAD drawing of the arm rest test bed assembly currently being built. Note that accommodations will be made to position small force platforms under the chair to allow for a biomechanically favourable chair rise

relate to contribution of the upper body and the pole on balance. Finally, investigating the effects of ML perturbations on stability during assisted STS and how this effects individuals rising with kinetic asymmetries applied to the feet to balance for the moment about the COM caused by unilateral pole forces.

An interesting question approached in this thesis is that of poles/grab-bars versus arm rests. As mentioned previously, arm rests are able to unload the lower extremities and thereby reduce peak extension moments in the hip and knee to the same, if not a greater degree than grab-bars and poles. The trade-off, however, is that at some point the user lets go of the arm rest and this might put a balance deficient individual at risk of a fall. Perhaps there is a configuration of arm rest that would bring the user to full standing while still holding onto the arm rest. A custom arm rest that can be attached to chairs may provide advantages over transfer poles in terms of price and ease of installation. In addition as a smaller device, it would be more aesthetically pleasing and potentially more cost effective than a transfer pole. A study evaluating the optimal position of arm rests and the comparison to grab-bars and poles would provide important insight into this topic. This work is already under way and we are close to finishing the arm rest test bed that is capable of testing nearly any desirable arm rest configuration as shown in Figure 6.1. The apparatus is mechanized and can switch between configurations quickly with a hand crank or drill. Construction of the apparatus is currently ongoing and testing of optimal arm rest configuration will begin in soon.

This study found that vertical forces decreased as the pole moved more anteriorly as a result of the mechanical disadvantage in the shoulder, however reductions in force were quite small. It would be interesting to evaluate greater distances for the far pole conditions and determine if vertical forces are further reduced with greater mechanical disadvantage. The effects of reduced vertical pole forces on vertical forces under the feet and corresponding moments can also be assessed. In addition, differences in mean timing of peak vertical pole force and peak hip and knee moments in this study were insignificant. However, examining

individual subjects, it was noted that some far pole position trials demonstrated differences in the timing of peak vertical pole force and the corresponding timing of peak vertical GRFs applied to the feet and peak hip and knee moments. It is possible that due to the mechanical disadvantage of the far pole position, applied vertical pole forces did not peak until the participant was closer to the pole. Therefore when using a far pole, later peak vertical force might contribute to the greater hip moments experienced. Furthermore, these differences may increase as the pole moves farther away. This same study could act to answer both questions of magnitude and timing of vertical force with increasing distance of the pole from the participant.

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Appendix A

Temporal kinetic and kinematic parameters

The thesis report included a representative pole condition to describe timing of kinetic and kinematic measures. Tables A.1 and A.2 below are included to provide additional detail on the timing of kinetic and kinematic measures across all pole conditions.

Table A.1: Temporal paramaters for kinetic measures across all pole configurations and positions

Parameter	Side	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
Ankle	ND	24 (13)	26 (19)	17 (11)	18 (10)	20 (15)	16 (12)
Max DF (% STS)	D	23 (22)	19 (17)	17 (12)	12 (09)	16 (13)	18 (15)
Ankle	ND	71 (24)	78 (20)	78 (24)	79 (17)	76 (19)	79 (21)
Max PF (% STS)	D	60 (34)	68 (25)	61 (33)	71 (24)	77 (20)	75 (23)
Knee	ND	27 (10)	30 (14)	27 (09)	25 (07)	30 (12)	31 (12)
Max ext (% STS)	D	27 (08)	31 (15)	27 (09)	26 (08)	29 (11)	30 (11)
Hip	ND	31 (14)	29 (25)	26 (14)	23 (12)	24 (11)	25 (14)
Max ext (% STS)	D	29 (10)	30 (18)	28 (10)	26 (06)	30 (11)	31 (13)

STS, sit-to-stand; DF, dorsiflexion; PF, plantarflexion; Flex, flexion; Ext, extension ND, non-dominant; D, dominant;
Statistical analysis not performed

Table A.2: Temporal paramaters for kinematic measures across all pole configurations and positions

Parameter	Side	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
Ankle	ND	28 (17)	29 (19)	33 (24)	34 (18)	37 (26)	29 (20)
Max DF (% STS)	D	25 (17)	25 (20)	31 (29)	29 (25)	32 (22)	25 (19)
Ankle	ND	75 (32)	70 (36)	68 (32)	77 (31)	69 (33)	66 (35)
Min DF (% STS)	D	75 (30)	73 (39)	75 (31)	73 (33)	71 (36)	69 (36)
Knee	ND	95 (07)	95 (11)	94 (09)	95 (06)	96 (05)	95 (06)
Min flex (% STS)	D	95 (06)	95 (11)	95 (12)	93 (08)	94 (08)	96 (08)
Hip	ND	27 (09)	25 (11)	27 (11)	24 (07)	28 (12)	29 (12)
Max flex (% STS)	D	24 (08)	21 (12)	22 (15)	21 (10)	25 (16)	25 (13)
Hip	ND	95 (09)	93 (12)	95 (05)	92 (09)	98 (04)	91 (09)
Min flex (% STS)	D	94 (10)	94 (12)	95 (05)	92 (09)	97 (05)	92 (08)
Trunk							
Max flex (% STS)		38 (11)	32 (11)	37 (15)	35 (08)	38 (13)	37 (12)
Max lateral (% STS)	ND directed	32 (24)	40 (26)	26 (20)	31 (18)	41 (28)	33 (23)
Max lateral (% STS)	D directed	58 (35)	63 (39)	56 (33)	70 (34)	55 (36)	54 (37)
COM displacement- ND	ML	44 (29)	59 (31)	66 (33)	39 (26)	59 (4)	61 (4)
COM displacement - D	ML	48 (42)	33 (37)	33 (34)	58 (4)	32 (36)	3 (35)

STS, sit-to-stand; DF, dorsiflexion; PF, plantarflexion; Flex, flexion; Ext, extension ND, non-dominant; D, dominant;
 Statistical analysis not performed

Appendix B

Joint moments and reaction forces not normalized

Tables B.1 and B.2 were generated to present joint moments and reaction force data without normalization, respectively.

Table B.1: Summary table of non-normalized joint moments in the sagittal plane for all pole-use conditions

Parameter	Side	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
Ankle	ND	3.5 (6.5)	2.3 (7.3)	1.8 (7.3)	1.0 (5.6)	-0.4 (5.6)	-2.2 (5.3)
Max DF (Nm)	D	1.2 (9.4)	-0.9 (7.9)	-1.0 (6.3)	-2.5 (7.5)	-2.7 (6.2)	-2.5 (6.1)
Ankle	ND	-16.6 (9.4)	-16.5 (6.7)	-18.4 (6.8)	-22.2 (8.5)	-23.0 (7.9)	-25.2 (8.4)
Max PF (Nm)	D	-16.3 (12.9)	-15.3 (11.0)	-17.2 (11.6)	-23.0 (12.5)	-21.3 (11.7)	-21.5 (11.4)
Knee	ND	-72.9 (19.1)	-73.0 (22.8)	-69.9 (20.4)	-69.8 (21.2)	-71.2 (22.0)	-71.5 (21.5)
Max ext (Nm)	D	-75.8 (25.4)	-68.5 (29.6)	-72.1 (23.2)	-74.3 (23.2)	-72.7 (22.8)	-71.3 (21.4)
Hip	ND	-46.8 (18.2)	-48.1 (25.0)	-43.9 (17.8)	-47.8 (23.4)	-54.6 (26.7)	-50.2 (23.6)
Max ext (Nm)	D	-43.3 (15.1)	-39.5 (19.6)	-39.5 (20.7)	-44.2 (17.7)	-42.2 (14.9)	-38.5 (17.2)

Nm, Newton-meters; DF, dorsiflexion; PF, plantarflexion; Flex, flexion; Ext, extension ND, non-dominant; D, dominant;
 Symbols denote $p < 0.05$

Table B.2: Summary of non-normalized reaction forces in the pole, seat and feet in the AP, ML and vertical axes

Parameter	Side	Near			Far		
		Horizontal	Single	Double	Horizontal	Single	Double
Max ground reaction force AP direction (N)	ND	-76.9 (14.8)	-82.4 (20.0)	-86.6 (15.4)	-75.7 (13.5)	-83.7 (19.8)	-82.0 (20.3)
	D	-90.8 (27.1)	-84.5 (21.8)	-88.8 (20.9)	-86.9 (18.0)	-84.2 (17.6)	-81.5 (17.5)
Max ground reaction force Medially directed (N)	ND	26.4 (19.5)	22.3 (24.8)	18.6 (20.7)	26.7 (21.3)	19.3 (26.5)	16.7 (22.2)
	D	-37.6 (28.3)	-31.5 (19.1)	-32.2 (26.8)	-34.1 (27.2)	-31.9 (25.7)	-31.3 (26.2)
	ND	-27.2 (18.3)	-27.8 (18.1)	-23.1 (15.1)	-27.6 (20.0)	-27.7 (16.8)	-23.5 (14.3)
	D	38.1 (21.6)	31.1 (14.7)	31.5 (17.8)	35.3 (18.1)	30.8 (17.0)	31.5 (17.2)
Min ground reaction force Vertical direction (N)	ND	412.8 (70.8)	425.1 (92.7)	421.2 (87.0)	417.0 (82.2)	437.5 (108.6)	443.6 (118.5)
	D	472.6 (139.6)	441.1 (91.8)	438.9 (96.8)	441.2 (100.4)	433.7 (88.7)	434.5 (96.4)
Min pole force (N) - AP		158.4 (35.6)	169.0 (46.4)	178.3 (48.7)	171.9 (35.4)	180.1 (57.2)	174.1 (43.9)
Max pole force (N) - ND		-6.3 (13.3)	-11.4 (21.0)	-5.6 (18.4)	-2.1 (12.7)	-7.7 (33.9)	-6.6 (20.1)
	Min pole force (N) - D	29.1 (25.2)	8.8 (21.4)	8.3 (8.7)	24.1 (22.4)	12.2 (15.2)	7.9 (10.2)
Min pole force (N) - vertical		129.5 (40.9)	135.4 (38.0)	148.0 (51.0)	122.0 (42.6)	117.8 (42.2)	127.3 (41.9)
Seat force (N) - AP		12.2 (11.5)	11.6 (11.4)	8.9 (11.2)	14.2 (12.3)	12.4 (13.0)	15.2 (11.0)

N, Newtons; AP, anterior-posterior; ND, non-dominant; D, dominant;
 Ground and seat reaction forces: positive = anterior, upward and ND directed forces applied to the segment; ND- and D-directed forces on the pole describe direction of
 applied force to the pole
 Symbols denote $p < 0.05$