The effect of aging on movement characteristics and postural control during stooping and crouching tasks

by

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AUTHOR’S DECLARATION

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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ABSTRACT

Stooping and crouching (SC) postures are integral to many daily tasks, such as retrieving objects from the floor and reaching to low shelves, yet nearly one in four community-dwelling older adults (24%) report having difficulty or being completely unable to perform SC movements. While limited research has identified physical (e.g., lower extremity strength and joint immobility) and behavioural (e.g., obesity and balance confidence) determinants of SC difficulty, little is known about how aging affects the manner in which SC tasks are performed. The objective of this thesis was to describe age-related differences in movement kinematics and balance control during stooping and crouching tasks.

Healthy younger (n = 12) and older (n = 12) participants performed a series of object-retrieval tasks – varying in initial lift height, precision required, and duration – that required them to bend over or reach toward the floor. In addition to kinematic and postural control measures describing the movements, measures of lower limb isometric strength, passive range of motion (ROM), and balance confidence were obtained for each participant.

Compared to younger, older participants moved slower into and out of self-selected postures, which were characterized by higher whole-body centre of mass (COM) vertical positions. Specifically, older adults exhibited lower vertical COM linear velocities and lower hip, knee, and ankle joint angular velocities during transitions, and higher COM heights achieved through comparatively less flexion in the hip, knee, and ankle joints during object retrieval. Older participants also displayed smaller, more centralized anterior-posterior (AP) COM excursions and lower COM velocities, but higher centre of pressure (COP) activity compared to younger participants, demonstrated through increased COP velocity (relative to COM velocity) and more frequent COP adjustments aimed at regulating COM position. Changing task constraints (i.e., lower initial lift height or longer duration) elicited greater postural changes in younger compared to older participants, potentially reflecting a diminished ability in older adults to make appropriate task-specific adaptations. In particular, younger participants were 4 times more likely than older participants to use a lower to the floor, forefoot crouching posture, especially during longer duration tasks. Older participants also had decreased leg strength and less passive range of motion compared to younger participants.

Overall, the results of this thesis demonstrate that despite moving slower through shorter distances, older adults displayed higher COP activity, which may have reflected a heightened effort to control COM position, during SC tasks. This compliments existing works describing age-related differences in movement strategies and balance control during lifting and sit-to-stand tasks. Further work exploring relationships between specific physiological and behavioural factors and SC task performance measures is needed to inform therapeutic intervention strategies.

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Chapter 1 Introduction

1.1 Aging and Functional Decline

Worldwide, the number of persons over 65 years is growing faster than any other age group (WHO, 2007). In Canada, specifically, this cohort accounted for 15% of the population in 2011 and is expected to increase to 25% by the year 2036 (Statistics Canada, 2010). Accompanying this demographic shift is a growing concern over the social and health care needs of the aging population. There is particular alarm over the human and economic resources that will be needed to support individuals who lose their functional independence and begin relying on help from either informal or formal caregivers to perform essential daily tasks such as bathing, ambulating, and transferring from one position to another (Dunlop et al., 1997; Ferrucci et al., 1997; Fried et al., 2001).

Approximately 12% of community-dwelling Canadians aged 75 and older experience a decline in functional status every year (Hebert, 1977). An individual's functional status is determined by their ability to perform activities of daily living (ADL), such as eating, dressing, transferring, bathing, ambulating, and toileting (Kleinpell et al., 2008). Functional decline describes the process of losing the ability to perform one or more ADL independently (Fried et al., 2001), and is often accompanied by a host of symptoms that can lead to loss of motivation, withdrawal from social activities, and cognitive decline (Hebert, 1997). Interestingly, functional decline – caused directly by weakness, diminished muscle strength, and reduced exercise capacity – often results from deconditioning and immobility associated with hospitalization for acute or chronic disease (Wu et al., 2006).

1.2 Balance, Mobility, and Fall Risk

The relationship amongst measures of functional decline, balance and mobility, and fall risk is troubling. Since balance and mobility are critical to many ADLs, such as ambulating and transferring, it is not surprising that declines in mobility lead to a three to five times increased risk for dependency in ADLs (Hirvensalo et al., 2000). Declines in ADLs and mobility are also linked to a deterioration of balance ability and heightened fall risk (Bloch et al., 2010; Era et al., 1997; Yokoya et al., 2007). Unfortunately, individuals with declining functional capabilities may have to avoid desired activities or rely on help performing them in order to reduce their risk of falling (Tinetti and Kumar, 2010). Relying on help and/or avoiding daily activities, however, means surrendering functional independence, reducing activity levels, and potentially accelerating the rate of functional decline. Older adults often choose to continue their regular activities despite declining functional capabilities and heightened fall risk, so that they can keep their independence and avoid institutionalization (Quine and Moreell, 2007; Salvage et al., 1989; Saulkeld et al., 2000). Toward maximizing the functional capabilities of older adults and preserving their independence, a better understanding of the mechanisms underlying age-related limitations in daily task performance is required.

1.3 Stooping and Crouching Difficulties in Older Adults

Stooping and crouching (SC) movements are integral to many daily living tasks, including picking up objects from the floor, reaching to low-lying shelves, and gardening. Recent epidemiologic data suggests that almost one quarter of community-dwelling older adults (24%) have considerable difficulty or are completely unable to perform these types of movements (Hernandez et al., 2008; Taylor et al., 1997). Limitations in stooping and crouching ability are associated with an increased likelihood of limitations in other lower-body functional tasks such as lifting and prolonged standing (Long and Pavolko, 2004), and are associated with increased fall risk (O'Loughlin et al., 1993). Although SC difficulty can significantly impact the overall mobility and functional independence of older adults, few works have explored the mechanisms underlying diminished stooping and crouching abilities in this population.

Significant coordination, physical strength, and balance ability are required to control the centre of mass through the wide range of postures characteristic of stooping and crouching movements

(Hemmerich et al., 2006; Hernandez et al., 2008, 2010, 2013; Kuo et al., 2011). Thus, it is not surprising that SC movements in older adults may be limited by obesity, decreased lower limb strength, pain or stiffness-induced leg and upper back limitations, and low balance confidence (Edmond et al., 2003; Han et al., 1998; Hernandez et al., 2008, 2010; Janssen et al., 2002). While researchers have established links between SC difficulty and these physiologic and behavioural traits, little is known about how aging affects movement characteristics and postural control strategies during SC task performance.

Preliminary research has demonstrated interesting trends in quantitative measures describing stooping and crouching movements in older adults (Kuo et al., 2011; Hernandez et al., 2013). Kuo et al. (2011) reported lower maximum joint angles, slower lower limb velocities, longer movement times, and higher relative muscle activity in older compared to younger adults during constrained squat-to-reach movements. This suggests that while older participants were less willing and/or unable to move quickly into deeper squatting postures, they required a greater proportion of their muscular capacity to control their movements. In a similar scenario, Hernandez and colleagues (2013) found that the base of support (BOS) size with which older participants were willing to stoop down and touch their toes was 50% larger, and their maximum forward floor-level reach distance was 22% shorter, compared to younger participants. Despite reaching to these shorter distances with a larger base of support, older participants were twice as likely to lose their balance as younger participants while performing a rapid forward floor-level reach (Hernandez et al., 2013). Although limited, these early works complement existing data describing age-related differences in sit-to-stand (STS) and lifting tasks (Hughes and Schenkman, 1996; Mourey et al., 1998, 2000; Puniello et al., 2001; Schultz et al., 1992). Broadly, these works suggest that, due primarily to strength limitations (in addition to deficits in neuromuscular control, range of motion, and sensory function), older adults tend to avoid high knee-flexion postures and prioritize stability control over movement speed and efficiency (Hughes et al., 1996; Hughes and Schenkman, 1996; Puniello et al., 2001; Scarborough et al., 2007).

1.4 Thesis Purpose and Goals

The majority of existing works describing age-related differences in stooping and crouching movement characteristics involve constrained scenarios in which postures were predefined and movement speeds were manipulated by the experimenters (Hernandez et al., 2013; Kuo et al., 2011). The focus of this thesis was to describe how aging and task requirements (i.e., duration of task) affect sagittal plane movement characteristics and balance control activity during *natural* performance of goal-oriented tasks that require stooping and crouching. Further, I wanted to determine whether 'conservative' strategies observed in older adults while performing similar activities, such as lifting tasks and the sit-to-stand, are present during stooping and crouching movements. Toward achieving these goals, I observed the performance of twelve young (mean age = 23 years) and twelve older (mean age = 70 years) adults, during a range of stooping and crouching tasks varying in initial lift height, precision demand, and duration. Based on the previous works outlined above, I postulated that, compared to younger, older adults would employ more conservative movement strategies and heightened postural control activity, as demonstrated by lower whole body linear and hip, knee, and ankle joint angular velocities when transitioning into and out of postures used for the tasks, less flexion in the hip, knee, and ankle joints, and greater anterior-posterior (AP) centre of pressure activity (velocity and frequency of COM crossings) despite lower COM velocities and smaller ranges of COM motion in the AP direction.

Chapter 2 Literature Review and Background Information

2.1 Aging Population

2.1.1 Scope of the Problem

Canada's population is aging rapidly. In 2011, older adults – aged 65 or older – accounted for approximately 14.8% of the Canadian population (Statistics Canada, 2010). It is projected that this proportion will reach 25% by 2036 (Statistics Canada, 2010), with the fastest growing demographic being adults over the age of 80 (Health Canada, 2002; Seidel et al., 2009). Accompanying this demographic shift is an increasing number of seniors unable to independently perform activities of daily living such as bathing, dressing, transferring, toileting, and feeding (Dunlop et al., 1997). When these abilities deteriorate due to age and age-related chronic disease, people become dependent on help from either informal (family members or friends) or formal (paid) caregivers (Dunlop et al., 1997). The consequence of this growing number of older adults with declining functional abilities is substantial on several levels.

In 2009, 45% of Canadian provincial and territorial governments' health care expenditures were spent on seniors; a disproportionate amount given that this age cohort made up only 14% of the population at the time (CIHI, 2011). Several studies have demonstrated that declining functional status is directly linked to increased hospital use and health care expenditure in older adults (Ferrucci et al., 1997; Fried et al., 2001; Mor et al., 1994). In a sample of community-dwelling seniors aged 72 or older, the 20% of participants who were functionally dependent – meaning they required some form of assistance – at baseline or who developed functional dependence within 2 years accounted for almost 50% of hospital, outpatient, home health, and nursing home expenditures (Fried et al., 2001). A better understanding of specific mechanisms underlying functional decline is necessary in order to devise and improve interventions aimed at maximizing independence in the aging population.

2.1.2 Functional Decline in Older Adults

Approximately 12% of community-dwelling Canadians aged 75 and older experience a decline in functional status every year (Hebert, 1997). This group is at a three times greater risk of mortality than their functionally independent counterparts and suffers drastic reductions in quality of life (Fried et al., 2001; Hebert, 1997).

Functional status has been traditionally determined by the ability to perform activities of daily living (ADLs) – eating, dressing, transferring, bathing, ambulating, and toileting – and instrumental ADLS (iADLs) – shopping for groceries, meal preparation, housework, laundry, getting to places beyond walking distance, managing medications, managing finances, and using a telephone (Kleinpell et al., 2008). While specific definitions vary (Cornette et al., 2006; Ferrucci et al., 1997; Hebert, 1997; Hoeymans et al., 1996), functional decline is broadly defined as the deterioration of ability to perform one or more ADL and/or iADL independently. While the formal definition focuses on ADL performance, functional decline is often accompanied by physical, psychological, social, and cognitive symptoms (Fried et al., 2001).

Physically, individuals may complain of fatigue, weakness, loss of appetite, weight loss, falls, and incontinence. Psychological symptoms can include loss of attention, interest, initiative, and motivation. Cognitive declines are sometimes also present. Socially, the individual may withdraw from his or her usual activities and become more isolated, perhaps neglecting housekeeping and grooming habits. Functional manifestations are much easier to identify since the person has progressively lost their capacity to perform iADL and basic ADL (Hebert, 1997). The inability to perform routine tasks and care for oneself can lead to drastic reductions in quality of life with affected persons sometimes being neglected by family members, and having to move out of their homes and into long-term care institutions (Fried et al., 2001).

2.1.3 Balance, Mobility, and Fall Risk

Several research groups have identified links between an individual's ability to perform activities of daily living and their physical mobility (Guralnik et al., 1995; Hirvensalo et al., 2000; Jette and Branch, 1981; Seidel et al., 2011). One group in particular has demonstrated that declines in mobility lead to a three to five times increased risk for dependency in ADLs (Hirvensalo et al., 2000). This is not surprising, considering that mobility is a critical component of many ADLs and iADL's, such as shopping for groceries, performing basic home maintenance, or getting up from a bed (Frank and Patla, 2003). Declines in ADLs and mobility have also been linked to a deterioration of balance ability and heightened fall risk (Bloch et al., 2010; Era et al., 1997; Yokoya et al., 2007). The relationship amongst these measures is concerning, as individuals with declining ADLs and mobility may have to avoid desired activities or rely on help when performing them in order to reduce their risk of falling (Tinetti and Kumar, 2010). Doing this, however, means forfeiting one's independence, reducing activity levels, and potentially accelerating the rate of functional decline. Older adults often choose to continue their regular activities despite their declining mobility and heightened fall risk, as they list 'loss of independence' amongst their most feared consequences of aging (Quine and Morrell, 2007; Salvage et al., 1989; Saulkeld et al., 2000). The topic of falls and fall-risk in older adults thus becomes an important one, as it can be both a source and a byproduct of functional decline.

Although some falls in older adults have a distinct cause, most result from complex interactions among a host of intrinsic and extrinsic risk factors. Intrinsic factors include predisposing biological elements such as reduced muscle strength and joint range of motion, sensory degradation (e.g., visual defects, reduced somatic sensitivity, and vestibular disorders), and cognitive impairments, as well as behavioural elements, which include reduced physical fitness and depression (Campbell et al., 1989; Fernie et al., 1982; Maki et al., 1994; Tinetti et al., 1988; Tinetti and Kumar, 2010). Extrinsic factors tend to compound the effect of any intrinsic issues and may include items within the physical environment (e.g., inadequate lighting, loose carpets, and unmarked steps), pharmacologic drugs (e.g., four or more prescription medications), and the use of assistive devices (Bateni and Maki, 2005). Predictably, the risk of falling consistently increases as the number of these risk factors increases (Nevitt et al, 1989; Tinetti et al., 1988). For example, in a cohort of community-dwelling older adults, the risk for falling increased from 8 percent among those with no risk factors to 78 percent among those with four or more risk factors (Tinetti et al., 1988). Encouragingly, a number of investigations have shown that, with appropriate interventions, the impact of many of these risk factors can be diminished, leading to reductions in fall rates (Campbell et al., 1997; Robertson et al., 2001; Sherrington et al., 2011; Tinetti et al., 1994; Wagner et al., 1994). However, in order to enhance the effectiveness of any intervention strategy, screening procedures aimed at identifying individuals at risk of falling must be improved.

2.2 Clinical Assessment of Balance and Mobility

A number of assessment approaches are used in clinical settings to predict fall risk and evaluate balance and mobility in older adults. In general, clinical balance assessments can be divided into three categories: (1) functional assessments, (2) systems assessments, and (3) quantitative assessments (Horak, 1997). Functional and quantitative assessments are typically used to identify whether or not a balance problem exists, while a systems approach combined with a quantitative assessment may be used to pinpoint the underlying cause of the balance problem (Horak, 1997).

2.2.1 Functional Assessment

Because of the time and equipment limitations associated with a typical clinical assessment scenario, such as a doctor's visit, functional assessments are the most commonly used approach in clinical practice. These tests usually involve a clinician rating the patient's performance on a series of 'functional tasks', which are meant to provide an indication of overall balance ability.

One of the most popular functional balance and mobility assessment tools is the Berg Functional Balance Scale (BBS), which consists of 14 tasks that the patient completes while the examiner rates their performance on a five-point scale (Berg et al., 1989, 1992). It involves functional tasks such as standing with different foot placements, rising from a chair, turning 360 degrees, and picking up an object from the floor. The BBS shows very high inter-rater (98%) and intra-rater (99%) reliability, reflecting its consistency as a measurement tool (Berg et al., 1989). It has also been evaluated for its predictive validity for falls, in comparison to self-reported fall rates in older adults. For example, Shumway-Cook et al. (1997, 2000) reported that this instrument had high sensitivity (91%) and specificity (82%) in dichotomously classifying fall-risk. In contrast, while Bogle et al. (1996) also found sufficient specificity to classify non-fallers (96%), they reported poor sensitivity for identifying fallers (53%). Interestingly, Muir et al. (2008) assert that the BBS was developed as a clinical measure of functional balance with suggested applications of comparing balance between groups of people, describing balance of an individual, and evaluating treatment effectiveness; not predicting fall risk. These authors performed a prospective assessment of the predictive validity of the BBS and found this tool to have insufficient sensitivity (25%) to justify its use as a dichotomous scale to predict fall risk.

The "functional reach" (FR) test is another commonly used functional assessment tool. It provides an approximation of the margins of stability in the anterior-posterior direction by requiring patients to reach forward with their arms extended as far as possible without adjusting their base of support (Duncan et al., 1990). The FR shows very high inter-rater (98%) and intra-rater (92%) reliability, and has also been validated for its ability to predict multiple falls in older adults (Duncan et al., 1990, 1992). While the authors reported that decreased FR distance was associated with increased odds-ratios for multiple falls, they did not evaluate the validity of the tool as a dichotomous predictor of fall risk (Duncan et al., 1992). Interestingly, several groups have subsequently found no difference in

FR distance between older adult fallers and non-fallers, questioning the predictive validity of this assessment tool (Franzen et al., 1998; Wallman, 2001; Wernick-Robinson et al., 1999). Furthermore, one group has questioned the construct validity of this tool, as it relates to an individual's stability limits, citing a low correlation between centre-of-pressure (COP) excursions and reach distance (Jonsson et al., 2002).

The inability of these and many other functional assessment tools to predict future falls in older adults is likely related to the multifactorial etiology of fall events. As mentioned previously, factors such as age, vision, muscle force, flexibility, sensory function, number and type of medications, and cognitive impairment can all interact to affect fall risk (Bergland and Wyller, 2004; Fernie et al., 1982; Tinetti et al., 1988, 1995). It is impossible for functional assessments, which must often be completed in a short period of time using simple instruments, to evaluate the influence of all of these factors and accurately predict future falls. While these tools may provide an indication of a patient's general balance ability, more comprehensive evaluations are needed to identify specific factors underlying balance and mobility impairments in older adults.

2.2.2 Systems Approach to Balance Assessment

A systems approach to balance assessment tries to identify the set of disordered subcomponents underlying functional balance limitations in order to focus potential treatments on these subcomponents. The subcomponents are classified as (1) biomechanical, (2) motor-coordination, or (3) sensory organization (Figure 2.1). Balance impairments often occur when these subcomponents are affected by problems resulting from pathophysiology such as poor sensation or loss of muscle strength. A systematic, clinical assessment attempts to specify the constraints or limitations in each of the subcomponents underlying control of postural stability (Horak, 1997).

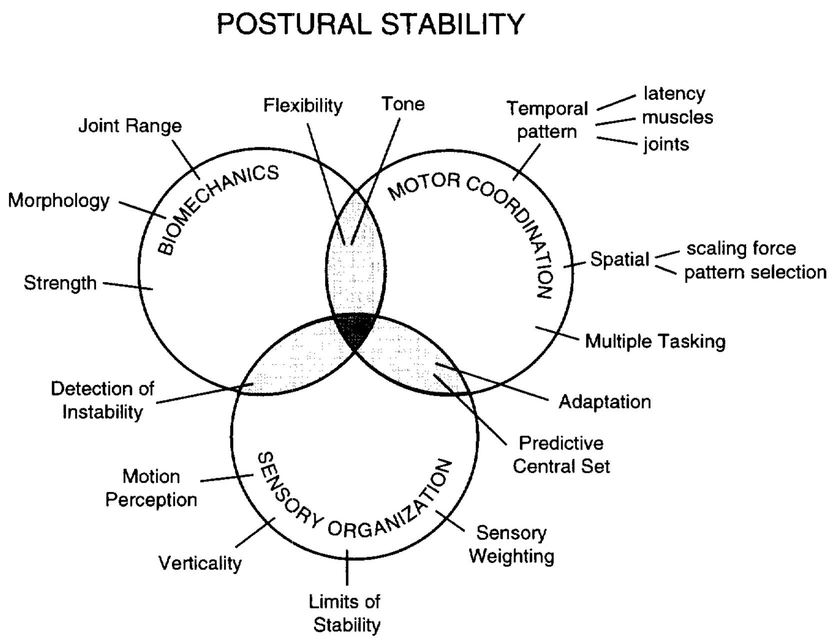


Figure 2.1 A breakdown of the biomechanical, motor coordination, and sensory organization subcomponents underlying control of postural stability (Horak, 1997).

An assessment of the biomechanical subcomponents involves determining whether individuals have the musculoskeletal capacity to accomplish desired motor tasks requiring balance control. In addition to an evaluation of isolated joint and muscle function, overall strength, range of motion, flexibility, and alignment in functional postures such as standing and sitting is important. In particular, clinicians must determine whether the musculoskeletal elements are able to align the body segments over the base of support (BOS) such that the projection of the body's centre-of-mass (COM) is safely within the limits of stability.

An assessment of motor coordination involves determining whether patients have adequate postural movement strategies for a variety of conditions, and if they can rapidly adapt these strategies in response to a change in condition. Specifically, clinicians are interested in determining whether an individual can maintain stability during three types of perturbations: (a) in response to external perturbations such as a nudge or support surface translation; (b) in anticipation of a voluntary limb movement such as a rapid arm or leg raise; and (c) during voluntary motions of the whole body COM such as leaning and locomotion. During each of these scenarios, patients are evaluated for their coordination of movements using measures such latency of response, spatial and temporal sequencing of limb movements, and appropriate scaling of response to the magnitude of the perturbation (Horak, 1997).

An assessment of sensory organization involves determining whether an individual uses visual, vestibular, and somatosensory information appropriately under various sensory conditions in order to maintain stability. Additionally, clinicians will evaluate an individual's perception of their limits of stability, ability to orient to vertical, and capacity to accurately differentiate self-motion from environmental-motion.

Evaluating each of these postural subcomponents requires expensive equipment, sophisticated testing protocols, and a substantial amount of time. Because of these demands, a comprehensive systems approach to balance assessment is impractical for most clinical settings.

2.2.3 Quantitative Posturography

Quantitative posturography uses technology to measure a range of variables that describe the performance of the balance control system. Force transducers measure external forces and moments, electromyographic (EMG) electrodes record muscle activity, and motion capture systems track the movement of an individual's body segments. Information derived from these instruments explains the timing, magnitude, and spatial coordination of postural movements, and can aid in identifying impairments in balance and mobility.

While quantitative posturography involves significant costs in terms of time, money, and technical complexity, it provides detailed, objective information regarding the underlying mechanisms of postural control. Further, this approach provides superior resolution in detecting small changes in balance performance, and eliminates the possibility of subjective error associated with qualitative tests used in the functional and systems approaches to balance assessment. This assessment approach has been underused in the study of stooping and crouching movements, and may contribute to understanding the mechanisms underlying self-reported and functionally assessed stooping and crouching difficulty in older adults.

2.3 Control of Balance during Static Upright Stance

Control of balance during standing depends on a complex interaction among several physiological mechanisms (Horak, 2006). These mechanisms (i.e., biomechanical, sensory, motor-coordination) must continuously act together to control a system of linked segments that is inherently unstable (Winter, 1995). From a biomechanical perspective, the goal is to ensure that the vertical projection of the whole body COM onto the support surface – referred to as the centre-of-gravity (COG) – is contained within the geometric limits of the BOS defined by the anterior, lateral, and posterior borders of the feet (Winter, 2009). If the COG is allowed to move beyond these limits, the stability of the system is compromised (Horak, 2006). In this scenario, a fall will occur unless the person is able to rapidly move their limbs (i.e., take a step or grasp an object for support) and alter their BOS such that the COG is once again contained within its limits (Maki et al., 2003). Further complicating the control problem is the fact that an individual's functional stability limits are governed not only by the geometric constraints of the base of support, but also by an individual's motor and sensory capacities, and fear of falling (Binda et al., 2003; King et al., 1994). Knowledge of the mechanisms that control the COG, and prevent it from exceeding the functional limits of the BOS, is critical to understanding postural control during upright stance.

In seeking to understand how humans maintain upright stance, researchers have traditionally focused on global measures of balance, such as the location of the centre-of-pressure (COP), using a single force platform. The COP is the point location of the vertical ground reaction force vector and represents the weighted average of all the pressures over the surface of the area in contact with the ground (Winter, 1995). While measurement of the time-varying COP displacement has at times been referred to as 'postural sway', Winter (1995) asserts that this term should be reserved for the movement of the COG, and should not be confused with the COP. He further suggests that authors who have misinterpreted the COP displacement as 'postural sway' have likely ignored the earlier work of Murray et al. (1967), who may have been the first to show a distinct difference between the COP and COG trajectories. Murray et al. (1967) and later Prieto et al. (1993) observed that excursions of the COP were always greater than the COG and that the COP signal oscillated on either side of the COG at a higher frequency. These works led to the suggestion that variations in the position of the COP represent muscular contractions that serve to accelerate the COG and regulate its position with the area of the BOS (Geursen et al., 1975; Murray et al., 1967; Prieto et al., 1993). Thus, while COP measures can be used to make inferences about postural control, a more complete analysis involves concurrently observing the COP (controlling variable) and the COM (controlled variable) and their relationship with the BOS.

2.3.1 Quantitative Assessment of Balance during Quiet Stance

The most common method of assessing balance during static conditions involves using a force platform to measure indices of postural sway (Bagchee et al., 1998; Prieto et al., 1996; Winter et al., 1995). Most studies evaluating age-related differences in postural stability focus on measurement of the centre-of-pressure under both feet. Since the COP position is much easier to obtain than the COM, and is believed to represent the postural activity required to control the COM, magnitude and displacement-based measures of the COP are often used to describe an individual's stability. Commonly reported COP measures include sway path length, root mean square (RMS) of the COP from its mean position, sway area, mean velocity, range, and mean power frequency (Hufschmidt et al., 1980; Palmieri et al., 2002; Prieto et al., 1996). Older adults often display increased COP displacement (Era and Heikkinen, 1985) and velocity (Fernie et al., 1982; Prieto et al., 1996) compared to healthy, young individuals; presumably reflecting increased muscular activity required to control a greater relative displacement of the COM from its equilibrium position. However, several authors emphasize that an increase in COP displacement and/or velocity in isolation is not sufficient to describe an individual's stability (Maki and McIlroy, 1996; Patla et al., 1990). Furthering this point, Hof et al. (2005) described a scenario in which a broomstick standing on its end exemplifies perfect balance if the only metric describing its stability is COP or COM motion. Interestingly, displacement-based COP measures actually decrease in patients with neurological disorders such as Parkinson's disease (Dimitrova et al., 2004; Horak et al., 1992), or when individuals perceive a threat to stability (Carpenter et al., 2006). Hof et al. (2005) and others have appropriately suggested that, rather than describing the movement of individual control variables, such as the COP, balance measures should focus on the "margins of stability" by relating COP or COM measures to the BOS area (Patton et al., 2000; Popovic et al., 2000; Van Wegen et al., 2002). Accordingly, researchers have defined a host of measures to describe the spatial and temporal behavior of the COP and/or COM relative to the BOS (see section 2.4.1).

2.4 Control of Balance during Volitional Movements

Maintaining balance during movement, whether it is spontaneous (i.e., in response to an external disturbance) or volitional (i.e., moving deliberately from one posture to another), requires complex control of a moving body COM (Horak, 2006). Biomechanically, the condition for dynamic stability depends on the movement context. If the base of support is fixed during the movement, the goal of the control system is the same as for quiet upright stance; to regulate the position of the COG within the stability limits of the BOS. Of course, compared to quiet stance, the movement of the COM is greater when transitioning from one posture to another; activities such as bending over or rising from a chair require much higher levels of postural coordination and muscular effort to control the COM (Hughes and Schenkman, 1996; Mourey et al., 2000). There are also many movements that require maintenance of stability despite a changing BOS. For example, the COG during walking is actually located outside of the BOS 80% of the time (Azevedo et al., 2007). In this case, dynamic equilibrium is still possible providing the horizontal velocity of the COM is large enough and directed toward the BOS (Iqbal, 2011). Several movement strategies are typically employed in order to stabilize the body COM when exposed to an external balance perturbation or performing of volitional movements.

Early studies utilized perturbation paradigms, by which the support surface was unexpectedly translated, to probe the response of the balance control system. Nashner (1977) observed that, in response to an unexpected horizontal force platform perturbation in the sagittal plane, there was a fixed pattern of postural response, whereby muscular contractions began distally at the ankles and moved in sequence proximally to the knees, hips, and trunk. This distal to proximal muscle activation pattern suggests the CNS recognizes the need to stabilize the joint closest to the perturbation first, and explains the organization of fixed-support reactions to external perturbations (Nashner, 1982).

Three main types of movement strategies can be used to return the body to equilibrium when exposed to a sagittal plane balance disturbance in the stance position: two which keep the feet in place and one which changes the base of support through stepping or reaching (Horak, 2006). The 'ankle strategy' involves generation of torque by the ankle plantar/dorsiflexors to maintain balance when exposed to relatively slow and small perturbations while standing on a firm surface capable of resisting ankle torques (Horak et al., 1989). The 'hip strategy', in which the body exerts torque at the hips, involves abdominal/lumbar and anterior/posterior thigh muscular activation to rapidly stabilize the body COM. This strategy is typically employed in response to relatively large perturbations that cannot be stabilized using the ankle strategy, or in situations that limit the effectiveness of torque generation at the ankles, such as when standing on narrow or compliant surfaces (Horak and Nashner, 1986). Higher level responses involve taking a step or grasping an object to increase the base of support (Do et al., 1982; Maki and McIlroy, 1997, 1999a, 2003). It was traditionally believed that these “change in support” reactions occurred as a last resort, only when earlier fixed-support reactions (i.e., the ankle and hip strategies) failed to keep the COG within the stability limits of the BOS (Maki and McIlroy, 2003). However, several authors have demonstrated that compensatory stepping or grasping are often initiated very early, with the COG well within the stability limits, indicating that it may actually be the preferred response in many situations (Maki and McIlroy, 1997; Mille et al., 2003; Pai et al., 1998, 2000).

In addition to maintaining stability during upright stance and stabilizing the COM in response to external perturbations, the balance control system must also have the capacity to regulate the COM position while performing voluntary movements. Numerous studies have demonstrated that, compared to standing, people are more likely to fall during activities involving larger displacements of the COM, such as walking, stepping up or down, or standing up (Gryfe et al., 1977; Overstall et al., 1977; Robinovitch et al., 2009; Tinetti et al., 1988). Such voluntary movements are considered ‘internal’ or ‘self-imposed’ balance perturbations, as they involve a volitional disturbance to the relationship between the COG and BOS (Horak et al., 1989; Massion, 1992; Patla et al., 1990).

Understanding postural activity associated with voluntary movements has been the focus of many researchers ever since Babinski’s observation of “axial synergies” during trunk movement (Babinski, 1899). He reported that when a standing subject voluntarily executes trunk extension, the knees and hips move in the opposite direction. This postural adjustment is necessary to compensate for the backward shift of the body COG associated with the focal movement – the trunk extension (Massion, 1992). Martin (1967) illustrated a similar example in which the head and trunk are displaced backwards when the arms are raised. In this case, the backwards shift of the head and trunk compensate for the forward shift of the arms to ensure that the body COG remains safely within its stability limits. These axial ‘synergies’ were aptly renamed ‘kinematic strategies’ by Massion (1994) in light of their similarity to the 'hip' strategy used to stabilize the COM during unexpected perturbations to upright stance.

Postural adjustments usually precede, accompany, and follow the intentional movement, and their coordination is critical to stabilizing the COM during movement (Bouisset, 2008). The majority of research has focused on the postural adjustments that precede voluntary movement, which counteract the expected perturbing forces associated with the impending movement and have been appropriately termed 'anticipatory postural adjustments' (Belenkiy et al., 1967; Cordo and Nashner, 1982; Massion, 1992). Anticipatory postural adjustments (APAs) have been studied in a range of focal movements such as arm raises, pushing and pulling tasks, and trunk bending (Aruin and Latash, 1995; Belenkiy et al., 1967; Bouisset and Zattara, 1981; Cordo and Nashner, 1982; Oddsson and Thorstensson, 1986). In a healthy system, these adjustments occur in the postural musculature shortly prior (50 – 100 ms) to the onset of muscle activity responsible for the focal movement (Belenkiy et al, 1967). APAs can be measured using electromyography (EMG) to monitor timing of muscle activation or force platforms to detect changes in the COP or kinetic signals prior to the intended movement (Bouisset and Zattara, 1981; Cordo and Nashner, 1982; Maki, 1993; Riach et al., 1992). A number of factors are likely to play a role in the process of APA generation. Among them are the magnitude and direction of the expected perturbation, properties of the voluntary movement, and features of the posture required to achieve the movement, in particular, body configuration (Aruin and Latash, 1995; Massion, 1992). All of these factors must be integrated by the CNS in order to properly plan and coordinate an APA that is appropriately scaled to the movement (Horak, 2006).

2.4.1 Quantitative Assessment of Dynamic Stability

Assessing dynamic stability is critical to determining the risk of falls during functional activities such as walking, turning, rising from a chair, and bending over. Large movement amplitudes and changes in body configuration make dynamic stability more complex to evaluate than in a standing posture

(Duclos et al., 2009). Many researchers have recognized that global parameters used to describe postural sway during upright stance (i.e., COP displacement-based measures) are likely inappropriate for dynamic scenarios. Pai and colleagues (Iqbal and Pai, 2000; Pai and Patton, 1997; Patton et al., 1999) addressed this issue by introducing a model of dynamic stability that incorporates the instantaneous position and horizontal velocity of the COM within the base of support to define 'feasible stability regions' from which equilibrium may be restored. While this model has been validated experimentally for its ability to predict loss of balance during balance recovery tasks (Patton et al., 1999) and volitional sway (Hof et al., 2005), it may be unsuitable for voluntary movements with large COM translations, such as stooping and crouching. In fact, the authors warn that tasks involving large multi-joint movements may violate a key assumption of the model (which is based on the inverted pendulum model of human upright standing): the distance of the COM from the axis of rotation must remain constant (Geursen et al., 1975; Patton et al., 1999; Winter, 1995). Nevertheless, several recently developed measures may be applicable for quantifying dynamic stability during stooping and crouching movements.

Postural control measures that may be relevant during volitional movements include the COP-COM error signal, and parameters that describe the spatial and temporal behavior of the COM relative to the BOS limits. The biomechanical variable COP-COG, which represents the distance between the COP and COG at a point in time, is proportional to the horizontal acceleration of the COM and represents the 'error signal' in the control system (Winter, 1995). A lower COP-COM value is postulated to reflect tighter stability control. Indeed, one group has reported that the root mean square of COP-COG during quiet stance is larger in older adults who have neurological impairments and in stroke patients, when compared to healthy age-matched controls (Corriveau et al., 2001; Corriveau et al., 2004). Age-related differences in COP-COG have also been reported, with healthy older adults exhibiting larger values than healthy young adults (Mesani et al., 2007). As this variable represents the relationship between the controlling variable (COP) and the controlled variable (COG), it may provide important insight into the online stabilization of the COM during voluntary movements such as stooping and crouching, in which the goal is to control the COM within a fixed base of support, often while attending to a separate goal-directed task such as grasping an object.

Parameters that directly relate the position of the COG (or COP) to the stability boundary describe how close an individual is to initiating a fall or requiring a change-in-support response to arrest the outward horizontal velocity of the COM. The COP safety margin is a commonly used measure, which is obtained by calculating the horizontal distance between the COP and the stability limit at an instant in time. For an idealized rigid body at rest, this stability limit is equivalent to the perimeter of the base of support. In humans, however, various factors such as decreased muscle strength and reaction time, impaired coordination of movement, and sensory deterioration reduce the 'functional stability boundary' (FSB) to an area much smaller than that outlined by the BOS (Binda et al., 2003; King et al., 1994). In light of the transient nature of these factors, which are commonly associated with aging and pathology, researchers have proposed various approaches to defining functional stability limits. As mentioned above, Pai and his group (Iqbal and Pai, 2000; Pai and Patton, 1997; Patton et al., 1999) used a modeling approach to define a 'feasible stability region', governed by the instantaneous position and horizontal velocity of the COM, within which a restoration of postural equilibrium is possible. Other groups have used an experimental approach, recording the maximum displacement of the COP during voluntary sway in multiple directions to define the FSB (Bagchee et al., 1998; Blaszczyk et al., 1994; Hof et al., 2005). Nevertheless, due to time and equipment constraints, researchers often resort to using the perimeter of the BOS to define the stability boundary (Bhattacharya et al., 2009; Hughes and Schenkman, 1996; Patton et al., 1999). While this approach ignores the potential effect of different FSB's among individuals, the COP-BOS safety margin has shown concurrent validity with physiologically based models of stability (Patton et al., 1999).

Unfortunately, individual measures of dynamic stability, which appropriately describe the state of equilibrium in all movement contexts, do not exist. Nevertheless, by combining some of the measures outlined above, different aspects of dynamic stability control (i.e., the proximity of the COM or COP to the stability boundary and ability to tightly regulate the COM movement) may be quantified.

2.5 Effect of Aging on Volitional Movement Control

Volitional movements involving large ranges of motion have been studied in a number of different scenarios (Hernandez et al., 2013; Hughes and Schenkman, 1996; Mourey et al., 2000). These movements are particularly challenging, as they demand simultaneous coordination of balance and body configuration, often while performing a goal-oriented task. For many older adults with diminished functional strength, sensory capability and coordination ability, daily tasks that involve large volitional movements, such as bending over to retrieve an object from the floor, are especially challenging (Hernandez et al., 2008, 2010; Long and Pavalko, 2004; O'Loughlin et al., 1993; Puniello et al., 2001). Recent data suggests that 24% of community-dwelling older adults (age 65 or older) have significant difficulty, or are completely unable to stoop, crouch, or kneel (Hernandez et al., 2008; Taylor et al., 1997). While the effect of aging on stooping and crouching movements, specifically, has only recently received attention in the literature, there are numerous works describing age-related differences in a similar movement: the sit-to-stand (Akram and McIlroy, 2011; Hughes, 1996; Hughes and Schenkman, 1996; Ikeda et al., 1991; Mourey et al., 1998; Schultz et al., 1992).

2.5.1 Sit-to-stand

The ability to rise unassisted from a seated to a standing position is integral to many activities of daily living, and losing this ability greatly compromises an individual's functional independence. The movement itself involves a controlled transfer of the COM from one stabilized posture (sitting) to another (standing) with movements of all body segments except the feet (Mourey et al., 1998). Thus, both coordinated movement of the different body segments that contribute to the change of posture, and equilibrium control during a significant displacement of the COM, are required. Numerous groups have investigated the age-related deterioration of sit-to-stand (STS) ability using kinematics and dynamics to describe characteristics of postural control (Akram and McIlroy, 2011; Hughes and Schenkman, 1996; Hughes et al., 1996; Mourey et al., 1998; Shultz et al., 1992).

While clinical assessments of sit-to-stand performance focus primarily on chair rise success or failure, detailed evaluations of the process of attaining the upright position have been useful in identifying specific subcomponents of the motor control system that are impaired (Scarborough et al., 2007). Three distinct chair rise strategies have been observed through such analyses. The momentum transfer (MT) strategy, used primarily by young subjects and healthy older adults, involves converting forward linear momentum generated by flexing the trunk into vertical momentum, which aids in rising (Hughes and Schenkman, 1996; Scarborough et al., 2007; Schenkman et al., 1990). This appears as a smooth movement with simultaneous back and knee extension after lift off, and is considered the least conservative and most efficient STS strategy (Scarborough et al., 1999). On the other end of the spectrum is the stabilization strategy, often used by older adults with functional limitations, in which the trunk is flexed to first place the COM over the feet before lifting off the seat (Scarborough et al., 2007). In this case, lift off from the seat is accomplished without the assistance of vertical momentum, and the movement is considered much less efficient, but more conservative with regards to maintaining postural stability (Schenkman et al., 1990; Hughes and Schenkman, 1996). This strategy has also been termed the “exaggerated flexion strategy”, based on the excessive trunk flexion observed for most of the chair rise. A third strategy, called the “dominant vertical rise”, involves limited trunk motion and rising predominantly in the vertical direction (Scarborough et al., 1999). Several authors have described this strategy as the least energetically efficient, requiring substantially higher leg muscle effort compared to the other strategies in order to complete the chair rise (Scarborough et al., 1999; 2007).

The idea that older adults, especially those with functional limitations, use more conservative movement strategies has also received support in constrained STS scenarios. Hughes and Schenkman (1996) demonstrated that older adults compensate for increased STS difficulty associated with progressively lower chair heights by simultaneously increasing both momentum and stability. The authors noted that increased momentum generation and increased stability are generally at odds as together they produce an inefficient strategy. Nevertheless, older adults placed more value on their perceived stability than on successfully rising from the chair (Hughes and Schenkman, 1996). Schultz et al. (1992) reported similar findings, with older participants prioritizing postural stability over movement efficiency (i.e., reducing joint torque requirements).

Several authors have identified lower limb strength, particularly of the knee extensors (i.e., quadriceps), as the strongest independent determinant of STS performance in older adults (Bohannon, 2009; Hughes et al., 1996; Scarborough et al., 1999). Hughes et al. (1996) reported that knee extensor strength was a limiting factor in determining the minimum chair height from which subjects could independently rise. Further, because of their decreased strength capacity and less efficient movement patterns, older adults are likely to use greater relative muscle activity levels when rising from a chair (Hughes et al., 1996; Papa and Cappozzo, 2000). Strength limitations, combined with diminished sensory and neuromuscular control capabilities, lead to increased STS difficulty which is often manifested by slower STS times, decreased body segment velocities, and less efficient movements (Akram and McIlroy, 2011; Hughes et al., 1996; Hughes and Schenkman, 1996; Mourey et al., 1998; Shultz et al., 1992). While an understanding of age-related differences during sit-to-stand movements is helpful, the constraints of goal-directed stooping and crouching movements present distinct balance control and strength challenges that may provide complimentary information about the functional capacity of older adults.

2.5.2 Stooping and Crouching Tasks

Stooping and crouching postures are fundamental components of many daily tasks such as picking up objects from the floor, tying shoelaces, gardening, and reaching to low-lying shelves. The inability to maintain and/or transition to and from these postures is related to limitations in other lower body functional tasks such as lifting and prolonged standing (Long and Pavolko, 2004), and is associated with increased fall risk (O'Loughlin et al., 1993). Specific factors associated with stooping and crouching difficulty include decreased trunk and lower extremity muscle strength, sarcopenia, pain or stiffness-induced leg joint immobility, obesity, and low balance confidence (Janssen et al., 2002; Han et al., 1998; Hernandez et al., 2008; Hernandez et al., 2010). While existing reports have established links between physiological risk factors and stooping and crouching difficulties, little is known about age-related differences in kinematics and balance control during SC tasks.

Stooping and crouching postures present the control system with scenarios that are vastly different from upright stance. Both postures involve lowering the centre-of-mass, which should theoretically increase postural stability by reducing the magnitude and frequency of COM oscillations (Winter, 1995). However, these postures also require a reconfiguration of the muscles at the hip and ankle such that their ability to generate forces required to stabilize the COM may be affected by operating at non-optimal lengths. A crouching posture significantly reduces the base of support, as only the forefeet are in contact with the ground while the ankles are plantarflexed. Significant flexion of the hip and knees in this posture may also affect the ability of the lower limb musculature to generate force. Stooping, in contrast, is characterized by high flexion at the hips and relatively little flexion at the knee and ankle joints. This posture leaves the base of support unchanged, but requires considerable trunk flexion, which may threaten stability by altering the performance of stabilizing muscles at the hip, and placing greater demand on the back extensors. Furthermore, stooping tasks naturally induce an inversion of the head (i.e., when bending over to retrieve an object from the floor), which may affect cerebral blood flow, vestibular, and visual function; thereby providing an even greater challenge for the balance control system (Buckley et al., 2005; Paloski et al., 2006; Johnson and Van Emmerik, 2011, 2012). The biomechanical and neurophysiological constraints introduced by these postures likely influence the overall performance of the balance control system and modulate strategies used to control body sway.

Moving to and from stooping and crouching postures requires coordination of the whole body COM while maintaining balance over a typically unchanging base of support. Such complex movements require sufficient sensory organization to monitor the state of equilibrium throughout the movement, motor-coordination to interpret sensory information and plan appropriate movement strategies, and muscular strength and flexibility to execute motor commands. Older adults often suffer from degradation to one or all of these physiological subcomponents, which predisposes them to an increased risk of falling when performing such movements (Horak, 2006).

Despite the significant impact of stooping and crouching difficulty on the overall mobility and independence of older adults, no studies to date have examined the underlying mechanisms of postural control in these postures, or compared both kinematic and stability measures between healthy young and older adults in a quantitative manner. Accordingly, the purpose of this thesis was to identify age-related differences in movement kinematics and postural control during stooping and crouching tasks.

Chapter 3 Laboratory Preparation and Equipment Quality Assurance

3.1 Methods – Laboratory Preparation

Before each participant arrived, several procedures were performed in order to prepare the laboratory equipment for collection. All movement-related data was collected through NDI First Principles software (Northern Digital Instruments, Inc., Waterloo, ON). Analog signals representing ground reaction forces and moments from the force platform (Advanced Medical Technologies, Inc., Newton, MA, USA) were synchronized with kinematic data from the four Optotrak Certus motion sensors in the System Control Unit (SCU). A diagram of the laboratory set-up is shown in Figure 3.1.

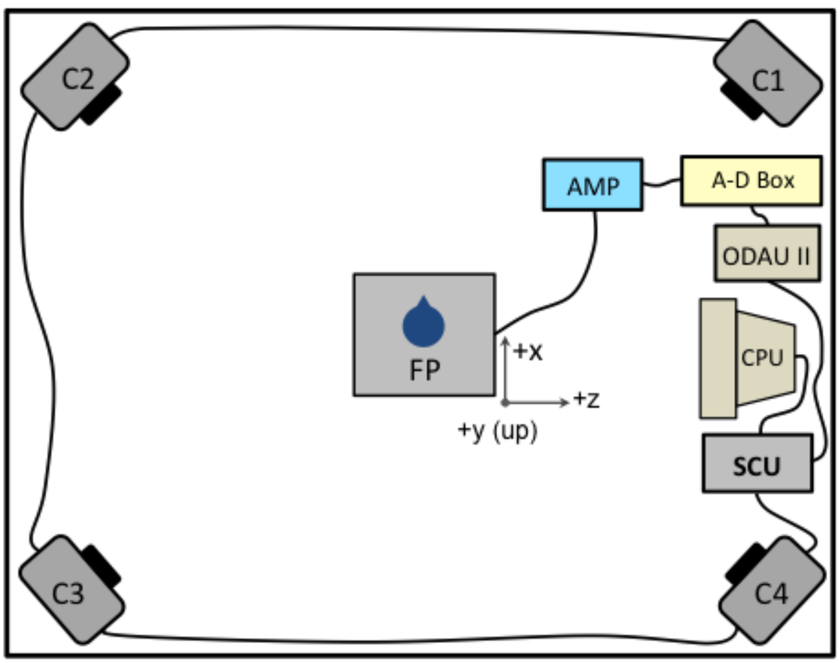


Figure 3.1 Laboratory set-up and equipment arrangement. Abbreviations on the schematic represent the following: C1-C4 are the four motion sensors; FP is the force platform; AMP is the force platform amplifier; ODAU II is the data acquisition system, SCU is the system control unit, and CPU is the data collection computer.

3.1.1 Force Platform Quality Control Tests

The force platform was used to measure the centre-of-pressure (COP) under the participants' feet. The location of the COP is determined by the relative forces recorded by transducers located at each corner of the force platform, under the top plate (Browne and O'Hare, 2000). If these transducers provide inaccurate and/or unreliable measurements, the overall performance of the force platform may be compromised, leading to incorrect calculations of the COP location. In order to assess the performance of the force platform (46.4 cm x 50.8 cm, Model OR6-7-2000, Advanced Medical Technology, Inc., Watertown, MA, USA), a one-time quality control test was performed to characterize the drift, linearity, and spatial accuracy of the force platform signals.

Drift is defined as "an undesirable change in the output signal over time that is not a function of the measured variable" (Szepessy and Zoltan, 2002). In order to assess whether electronic drift would be an issue during the data collections, the force platform was left unloaded and its signals were sampled over a 3-hour period immediately after the amplifier had been turned on. Each signal (F_x, F_y, F_z, M_x, M_y, M_z) was collected for 30 seconds at 64 Hz, every 20 minutes for the first two hours, and every 10 minutes during the last hour. Comparisons were made between each signal value at the beginning and end of the test, and during the last hour of the test. An inverse sensitivity matrix provided by the force platform manufacturer was used to convert each signal from volts into SI units.

Linearity represents how closely force platform readings match the actual values of the applied mass over a range of known weights. In order to assess the linearity of the laboratory force platform, calibrated weights, ranging in mass from 7.71 kg to 22.73 kg, were incrementally added to the geometric centre of the platform. The exact range of mass tested was 0 to 142.2 kg – well above (and below) the mass of the heaviest participant (116 kg) tested in the current study. At each weight increment, data was collected for 5 seconds at 512 Hz. The average vertical force (F_z) was then obtained for each weight, converted to kilograms, and compared to the known masses across the range.

Spatial accuracy refers to the consistency of a force platform's measurement regardless of the location of the applied load. A 21.07 kg weight was placed in the geometric centre and systematically at eight known co-ordinate points within a ±10 cm area of the origin in both the x- and y-directions, as recommended by Bizzo et al. (1985). At each location, force platform data was collected at 300 Hz for 5 seconds, and the resulting signals were converted into SI units using the inverse sensitivity matrix. Average vertical force (F_z) and COP location was then compared between recorded and known values at each location.

Each of these quality control measures is expressed in units consistent with the technical specifications provided by the force platform manufacturer (Advanced Medical Technology, Inc., Watertown, MA, USA). Drift is expressed as a percentage of full scale output (% FSO); linearity as a percentage of the range of known weight tested, also termed % FSO; and uniformity as a percent difference from the central location. Results of the force platform quality control tests are presented in Section 3.2.1.

3.1.2 Force Platform and Motion Capture Congruence

Errors can arise if there is any discrepancy in the location of the centre of pressure (COP) between the motion capture system and the force platform data. Therefore, in order to determine the congruence of these two systems, a one-time test was performed according to Holden et al. (2003) using the CalTester (C-Motion, Inc., Kingston, ON, Canada). The test involved using a calibrated CalTester rod, instrumented with 5 markers, to apply a point load to the force platform. Using the CalTester software, the location of the tip of the device was calculated from motion capture data through transformation of the marker coordinates, resulting in an estimate of COP location based on kinematic data. The location of the COP was also calculated from the force platform data. A comparison was then made between the two COP locations derived from the motion tracking and force platform systems to provide an estimate of the amount of error. Results of the CalTester procedure are detailed in section 3.2.2.

3.1.3 Motion Capture Calibration and Registration

An Optotrak Certus Motion Capture System (Northern Digital Instruments, Inc., Waterloo, ON, Canada) was used for to collect kinematic data during all experiment sessions. This system has a reported accuracy of up to 0.1 mm and a resolution of up to 0.01 mm at a distance of 2.25 m (http://www.ndigital.com). Four Certus sensors were arranged as shown in Figure 3.1. Version 1.2.3 of the NDI First Principles software (Northern Digital Instruments, Inc., Waterloo, ON, Canada) was used to collect kinematic and kinetic data. Prior to each collection, the four Optotrak sensors were calibrated for the motion capture volume with a calibration cube instrumented with 16 infrared emitted diode (iRED) markers. First, a dynamic calibration procedure was performed to register all sensors to a single global coordinate system (GCS). Once this was complete, a static calibration or "alignment" procedure was performed to specify the location of the origin for the GCS. The collection volume was confined to the area above and surrounding the force platform. The root mean square error (*rms*) for both dynamic and static calibrations was recorded on the data collection sheet for each session. Collections proceeded only if the *rms* error for the dynamic calibration was less than 0.50 mm (NDI technical support - http://www.ndigital.com). The GCS origin was placed at the right rear corner of the force platform, with the ***x-axis*** (+ve) pointing anterior, ***z-axis*** (+ve) pointing to the participant's right, and the ***y-axis*** (+ve) pointing upwards (Figure 3.1).

3.2 Results – Laboratory Preparation

3.2.1 Force Platform Quality Control Tests

The force platform had low levels of *drift* over the 3-hour test. The output data from the 6 channels, converted into SI units using a calibration matrix provided by the manufacturer, are displayed below in Table 3.1. The bottom of the table summarizes, in bold, the change in the output signals between the value measured at hour 3 (time 180), and hours 2 (time 120) and 0 (immediately after turning on the amplifier). It is evident that there was a larger relative degree of drift from the time the amplifier was turned on until the end of the test than there was during the last hour of the test. Thus, it was important to allow the amplifier and associated electronics sufficient time to warm-up and stabilize prior to data collection. These results provided the rationale for selecting a 2-hour amplifier warm-up period adopted for all collections.

Table 3.1 Drift test data for force platform

Time on (min)	F_x (N)	F_y (N)	F_z (N)	M_x (N-m)	M_y (N-m)	M_z (N-m)
0	-0.59	-0.96	-1.29	-1.33	-0.42	-0.78
60	-0.80	-1.27	-0.23	-0.90	-0.21	-0.80
120	-0.94	-1.40	0.42	-0.70	-0.25	-0.77
150	-0.92	-1.42	0.58	-0.66	-0.30	-0.78
180	-0.92	-1.46	0.75	-0.60	-0.33	-0.77
Change last 60	**0.01**	**-0.06**	**0.34**	**0.10**	**-0.08**	**0.00**
Change from 0	**-0.34**	**-0.50**	**2.05**	**0.72**	**0.09**	**0.01**

The *linearity* test results, (Table 3.2), showed a small and consistent difference (average = 1.16 %) between the actual and recorded values across the range of weights tested. While this is above the +/- 0.2 %FSO value specified by the manufacturer (Advanced Medical Technology, Inc., Watertown, MA, USA), it is consistent across the range and likely did not affect comparisons made between participants.

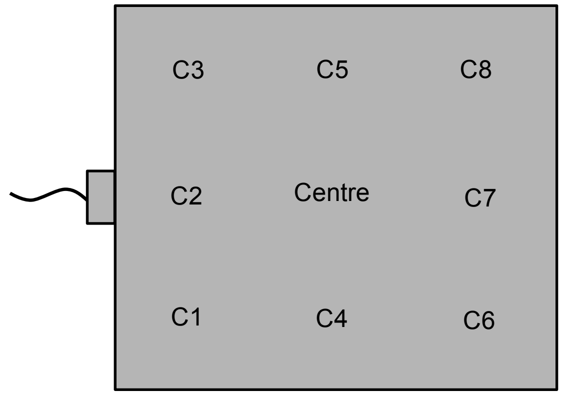
Table 3.2 Force plate linearity test

Actual Weight (kg)	0.00	22.73	39.99	62.26	84.49	92.20	102.21	112.19	122.19	142.20
Recorded Weight (kg)	0.18	22.44	39.52	61.52	83.49	91.14	101.04	110.91	120.81	140.62
Difference (%)	**--**	**1.25**	**1.18**	**1.19**	**1.18**	**1.15**	**1.14**	**1.14**	**1.13**	**1.11**

The results of the *spatial accuracy* test for the force platform are summarized in Table 3.3. Differences between the actual and recorded weight were minimal and consistent (average = 0.93 %), regardless of the location at which the calibration weight was placed. This indicates that the F_z contribution to the calculation of COP location was accurate and consistent, regardless of location.

Table 3.3 Force plate spatial accuracy test results

Location	Actual Weight (kg)	Recorded Weight (kg)	Difference (%)
Centre	10	10.05	0.54
C1		10.11	1.11
C2		10.08	0.82
C3		10.08	0.76
C4		10.10	1.04
C5		10.10	0.96
C6		10.11	1.05
C7		10.09	0.90
C8		10.12	1.19



3.2.2 Force Platform and Motion Capture Congruence

The force platform was in good agreement with the motion capture system for the determination of the centre of pressure, as evidenced by minimal errors in the anterior-posterior (COP_x) and medial-lateral (COP_z) coordinates of the COP (Table 3.4). The larger error in COP_y was expected, as the CalTester pivoting jig caused the tip of the rod to be slightly above the force-platform surface.

Table 3.4 Reported error in COP from CalTester

Δ COP_x (mm)	Δ COP_y (mm)	Δ COP_z (mm)
0.03 (0.9)	-4.7 (0.1)	1.4 (0.7)

3.2.3 Motion Capture Calibration and Registration

Registration and alignment were performed within the NDI First Principles software using the built-in wizard. The average *rms* error for the dynamic registration was 0.27 ± 0.02 mm and 0.09 ± 0.03 mm for the static alignment. These values were within the range of acceptable limits recommended by NDI technical support (http://www.ndigital.com), given the number of sensors (4) used in the collections.

Chapter 4 The effect of aging on movement kinematics and postural control during stooping and crouching tasks

4.1 Introduction

Stooping and crouching (SC) movements require significant coordination, physical strength, flexibility, and balance (Hemmerich et al., 2006; Hernandez et al., 2008, 2010, 2013; Kuo et al., 2011). As individuals age, many of these capabilities deteriorate (Raj et al., 2010; Samuel and Rowe, 2009; Tinetti et al., 1995) leading to limitations in daily tasks such as picking up items from the floor or reaching to low-lying shelves (Han et al., 1998; Hernandez et al., 2008; Janssen et al., 2002). Stooping and crouching postures are integral to these and many other daily tasks, yet nearly one quarter (24%) of community-dwelling older adults (age 65 or older) report having significant difficulty or being completely unable to stoop, crouch or kneel (Hernandez et al., 2008; Taylor et al., 1997). Despite this knowledge, few studies have investigated the mechanisms underlying SC difficulty in older adults. As aging demographics continue to grow in developed nations (Statistics Canada, 2010), a better understanding of these mechanisms is essential to devising interventions aimed at maximizing functional independence in older adults.

Volitional movements involving large ranges of motion in the hip, knee, and ankle have been studied in a number of different contexts (Hernandez et al., 2013; Hughes and Schenkman, 1996; Mourey et al., 2000). Such movements present a challenge, particularly for older adults, as they demand simultaneous coordination of balance and body configuration, often while performing a goal-oriented task such as grasping an object. One area that has garnered substantial attention – and involves movements similar to those required for SC tasks – is the sit-to-stand (STS). While surprisingly few kinematic differences have been reported between young and older adults when performing the STS movement (Akram and McIlroy, 2011; Ikeda et al., 1991; Mourey et al. 1998), several researchers have noted that older adults, especially those with physical limitations, employ a more conservative movement strategy; prioritizing stability control over speed and energy efficiency (Hughes and

Schenkman, 1996; Mourey et al., 1998, 2000; Schultz et al., 1992). This strategy, often referred to as the 'stabilization strategy', involves exaggerated flexion of the trunk and legs prior to liftoff to ensure the centre of mass (COM) is located safely within the foot support area before rising to a standing posture. This leads to an inefficient transfer of horizontal to vertical momentum, which in turn places greater demand on the knee extensor musculature to raise the COM and extend the hip into an upright standing posture (Schultz et al., 1992; Hughes and Schenkman, 1996). While an understanding of age-related differences during sit-to-stand movements is helpful, the constraints of goal-directed stooping and crouching movements present distinct balance control and strength challenges that may provide complimentary information about the functional capacity of older adults.

Knowledge of postural control during stooping and crouching movements is limited. Nevertheless, preliminary investigations have provided important information regarding performance and stability in these postures. DiDomenico et al. (2010, 2011a, 2011b) have published several works describing perceived and objectively measured stability associated with postures commonly used by construction workers. Specifically, they reported that workers' self-perceived instability was highest upon standing up from stooping, crouching, and kneeling postures compared to a range of other working positions (DiDomenico et al., 2010). Such subjective ratings of stability have been verified by a range of objective measures, including increased COP velocity, shorter time to BOS boundary contact (TtC) times, and smaller COP to BOS distances during the stabilization phase that immediately follows standing (Bagchee et al., 1998; Bhattacharya et al., 2008; Chiou et al., 1998; DiDomenico et al., 2011a). While these reports provide insights into the challenges faced by individuals in the young, working population, an understanding of age-related declines in stooping and crouching performance is needed.

Hernandez and colleagues (2008, 2010) have reported that older individuals who struggle with SC tasks are likely to have self-reported leg joint limitations, decreased knee extensor and plantar flexor strength, and low balance confidence. These findings are consistent with the STS data (Hughes and Schenkman, 1996; Hughes et al., 1996; Scarborough et al., 1999) and help explain why older adults with SC difficulty are also likely to experience limitations in other lower body tasks such as lifting and prolonged standing (Long and Pavolko, 2004), and may be at a higher risk of falling (O'Loughlin et al., 1993). In addition to establishing links between physical characteristics and SC difficulty, Hernandez et al., and others, have described age-related differences in kinematics, muscle activity, and postural stability during controlled downward reaching scenarios (DiDomenico et al., 2011b; Hernandez et al., 2013; Kuo et al., 2011). DiDomenico et al. (2011b) reported no differences in joint angular kinematics or COP to BOS margins across a range of different working-aged individuals during a knee-push squat, although they acknowledge this was likely because their oldest participant was younger than 65 years old. Kuo et al. (2011) performed a similar experiment, but compared younger adults to a group of older participants with mean age: 77 ± 4.6 years, and found that older adults displayed smaller maximal angular displacements of the head, knee, and thigh, and lower knee angular velocities, while using a higher proportion of their available muscle capacity. These results suggest that while older participants were less willing and/or unable to move as quickly into deeper squatting postures, they required more of their available muscle capacity to control stability. Hernandez et al. (2013) found that the BOS size with which older women were willing to stoop down and touch their toes was 50% larger, and their maximum forward floor-level reach distance was 22% shorter, compared to younger women. Despite these more conservative movement and postural characteristics, the older women were twice as likely to lose their balance while performing a challenging rapid forward reach to a target placed on the floor at their maximum forward reach distance (Hernandez et al., 2013), indicating that they were indeed more unstable than their younger counterparts. The results of these studies complement the STS data by revealing kinematic, muscle activity, and stability control differences, which collectively suggest a more conservative, but less efficient, stability control strategy adopted by older adults during volitional movements requiring large ranges of motion.

The majority of existing stooping and crouching research involves constrained scenarios in which postures are predefined and movement speeds are controlled by the experimenters (DiDomenico et al., 2011b; Hernandez et al., 2013; Kuo et al., 2011). To my knowledge, no existing works describe age-related differences during *natural* performance of daily tasks that require SC postures. Accordingly, the focus of this study was to describe how aging and task constraints affect movement characteristics and balance control during object-retrieval tasks that require stooping and crouching postures.

Toward achieving this goal, I postulated that, compared to younger, older adults would employ more conservative kinematic strategies, and comparatively higher postural control activity when performing a range of goal-directed stooping and crouching tasks. Further, I predicted that as tasks became more challenging through (1) lower initial lift height, (2) increased precision demand, and (3) increased duration, age-related differences in kinematic and balance control measures would increase as a byproduct of older adults adapting less to the changing demands of the tasks. The following hypotheses form the basis for this line of reasoning:

1) Maximum Joint Flexion Angles

I hypothesized that compared to older, younger participants would display greater maximum hip, knee, and ankle joint flexion angles, leading to lower COM heights at the moment of object retrieval across all tasks. I also hypothesized that age-related differences would increase as tasks became more challenging (i.e., via lower initial lift height, higher precision demand, and increased task duration).

2) Maximum Velocities

I hypothesized that younger participants would exhibit greater maximum vertical COM velocities, achieved through higher angular velocities in the hip, knee, and ankle joints during transitions into and out of postures chosen to perform each task. I also predicted that age-related differences would increase as tasks became more challenging.

3) BOS Condition

I hypothesized that, compared to older, younger participants would be more likely to use a smaller forefoot BOS, which would bring them closer to floor level, especially for longer duration tasks. This was based on the work of Hernandez et al. (2013), which demonstrated that the minimum BOS that older women were willing to use while reaching down to touch their toes was 50% larger than that used by younger women. That this posture would be preferred for longer tasks was based on the findings of Gallagher and colleagues (2011), which showed that muscle activity is reduced in high knee flexion postures, such as the deep squat, thus providing a postural option that may minimize lower limb muscle activity, and ultimately reduce fatigue during longer tasks.

4) COM to BOS Margin of Safety

I hypothesized that older participants would constrain their anterior-posterior (AP) COM movement within a centralized region of their base of support (BOS) to a greater extent than younger participants. This would be reflected by greater COM to BOS margins of safety (minimum horizontal distances) in the anterior and posterior directions for older compared to younger participants throughout all of the tasks, with age-related differences increasing as tasks became more challenging.

5) COM and COP Horizontal Velocities

I believed that while younger participants' COM velocities would be faster than their older counterparts, the opposite would be true of their COP velocities, in the AP direction. This was based on previous work showing that older adults increase the number of COP submovements, despite having slower focal COP movements, in an attempt to maintain accuracy during targeted COP movements in upright stance (Hernandez et al., 2012a). I further hypothesized that age-related differences in COM and COP mean AP velocity would increase as tasks became more challenging.

6) COM to COP Interaction

Extending from 5), I hypothesized that, compared to young, older adults would exhibited a greater number of COP to COM crossings in the AP direction across all tasks, reflecting increased muscle activity aimed at continually controlling the COM position throughout the movement. I also hypothesized that the root mean square distance between the COP and COM in the AP direction would be greater in older compared to younger participants, representing an unintended 'overshooting' of the COM by the COP.

7) Finally, I predicted that younger participants would have greater lower limb isometric strength and range of motion, and higher balance confidence than older participants.

4.2 Methods

4.2.1 Participants

The study population consisted of twelve healthy young adults and twelve healthy community-dwelling older adults, with equal numbers of males and females in each age group. Participant characteristics are summarized in Table 4.1. Young participants were recruited from the university student population, while older participants were drawn primarily from the Waterloo Research in Aging Participant (WRAP) Pool. All subjects were interviewed prior to participation to ensure they were free of anatomical, neurological, or cognitive impairments that may have affected their balance and/or mobility. Initial telephone interviews with older candidates were used to ensure they were confident in their ability to independently perform tasks that required stooping and crouching postures, were not using any psychotropic medications, did not rely on any ambulatory aids, and did not have any prosthetics or joint replacements. Once participants arrived, they completed a health status form to confirm they were eligible for participation in the study (see Appendix A). Each participant provided informed written consent, and the study was approved by the Office of Research Ethics at the University of Waterloo.

Table 4.1 Mean (SD) subject characteristics

	Young Participants (n=12)	Older Participants (n=12)
Mean age (y) [a]	22.8 (2.4)	69.5 (6.9)
Height (m)	1.73 (0.1)	1.75 (0.1)
Mass (kg) [a]	73.5 (15.6)	87.6 (17.7)
Body mass index (kg/m^2) [a]	22.8 (2.4)	28.4 (6.3)

[a]Indicates significant age-effect ($p < 0.05$)

4.2.2 Instrumentation

Four Optotrak Certus motion capture banks utilizing 12 cameras (Northern Digital, Inc., Waterloo, ON, Canada) were used to track the motion of body segments during the experiment. Infrared diode emitting (iRED) markers arranged in 'rigid' clusters of four were placed on the posterior aspects of the sacrum and thorax, left side of the head, and bilaterally on the feet, lower legs, thighs, upper arms, and lower arms (see Appendix B). A digitizing probe was then used to identify anatomical landmarks representing relevant bony landmarks and segment endpoints corresponding to each 'rigid' cluster. All activities were performed while standing on a force platform (46.4 cm x 50.8 cm, Model OR6-7-2000, Advanced Medical Technology, Inc., Watertown, MA, USA) embedded in the laboratory floor. Kinematic and kinetic data were sampled at 32 Hz and 512 Hz, respectively.

The force platform and amplifier were turned on and allowed to warm up for 2 hours prior to each collection, as determined by the drift test performed in Section 3.2.1. The amplifier was zeroed prior to collection to remove any bias. Once the force platform and motion capture systems were calibrated, spatial synchronization was achieved between the two by identifying each force platform corner with a calibrated digitizing probe (Northern Digital Instruments, Inc., Waterloo, ON, Canada).

The 3-D coordinates for each force platform corner in the GCS were recorded and exported to an ASCII file.

4.2.3 Experimental Protocol

Upon arriving in the laboratory and completing an informed consent form, the following measures were obtained from each participant: total body height, shoulder height, greater trochanter (of the right femur) height, and total body mass. A health status questionnaire was then verbally administered, in which participants were asked to report details regarding fall history, musculoskeletal and/or other acute injuries, medical conditions and/or illnesses, symptoms and/or feelings experienced on a daily basis, recent hospitalizations and surgeries, medications used, and physical exercise habits (see Appendix A).

After being instrumented with iRED markers, participants performed a series of functional tasks that required them to bend over and/or reach downwards to pick up objects in scenarios that isolated three distinct elements of stooping and crouching tasks: (1) lift height, (2) precision required, and (3) task duration. In the 'lift height' scenario, participants picked up a round, 3.5 cm diameter plastic poker chip from the following heights using their dominant hand: floor level, or a height-adjustable shelf positioned at 10, 20, 30, 40, and 50% of their hip height. Hip height was defined as the height of the greater trochanter of their right femur above the ground. 'Precision demand' was modified by asking participants to pick up a light plastic dustpan from the floor in front of them using their dominant hand. Because of the large, graspable handle on the dustpan, this task was considered to require less precision than picking up the smaller poker chip from the floor (which was the same as the 'floor level' trial in the 'lift height' scenario). Finally, the 'task duration' scenario involved a bimanual task in which participants held a paper cup in their non-dominant hand while using their dominant hand to pick up and place either 1, 4, 8, or 12 poker chips from the floor into the cup, before ascending back to a standing position. Each of these task elements is summarized in Figure 4.1. For single-object retrieval tasks and the 4 chip bimanual task, the object(s) was (were) placed approximately 20 cm in front of the participant's toes, centred between their feet. This was considered to be a natural distance away from the participant, representing a real-life scenario that does not require excessive forward reaching or awkward postures associated with retrieving objects either too far away from, or too close to, one's base of support. For the 8 and 12 chip bimanual tasks, the chips were arranged in rows of 4, with the closest row approximately 15 cm, and farthest row approximately 25 cm, anterior to the participant's toes. Participants were instructed to perform each task at their own pace as they naturally would if they had to retrieve the object(s) in their daily lives. There were no specific instructions regarding speed of movement or postures that should be used to perform the tasks. The only requirement was for participants to remain on the force platform at all times, and assume a quiet upright standing posture before and after they performed each task. Each task was performed only once in order to elicit a natural response that was not influenced by previous trials of the same task. Participants removed their footwear for all tasks and wore tight-fitting, flexible shorts and a tee shirt. Task order was randomized for each participant.

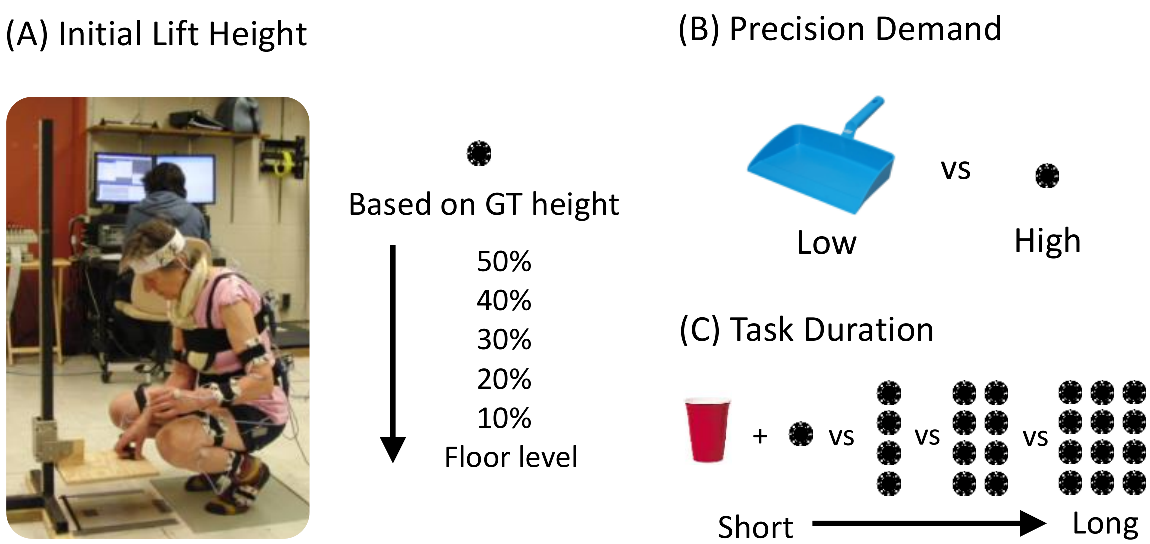


Figure 4.1 Depiction of the three elements of stooping and crouching tasks examined in the present study: (A) initial lift height, (B) precision demand, and (C) task duration.

At the end of the session, we assessed each participant's lower limb isometric strength, passive range of motion, and balance confidence. Isometric strength and range of motion were assessed only for the dominant limb, which was determined by asking participants which leg they would use to kick a soccer ball or stomp a fire out.

Isometric peak torques of the knee extensors and flexors, as well as the ankle plantar and dorsiflexors, of the dominant lower limb were evaluated using a Cybex II dynamometer (Cybex International Inc., Medway, MA, USA). After a practice trial, participants exerted maximal contractions for 2 trials of 3 seconds using each of the muscle groups mentioned above. Participants received approximately 1 minute of rest in between trials. The maximum torque value of the two trials was recorded as the peak torque for that particular muscle group. Knee extensor and flexor isometric strength was evaluated in a seated position with the knee flexed at 90 degrees. The thigh was strapped onto the seat pan, arms were folded across the chest, and participants were allowed to lean on the backrest for support and stabilization. Participants exerted either a maximal knee extension or flexion contraction, depending on the group of muscles being tested, against a pad located at the distal tibia. Ankle plantar flexor and dorsiflexor strength was evaluated with participants in a semi-reclined position, the foot and tibia at approximately 90 degrees to each other, and the knee and hip at 30 and 60 degrees of flexion, respectively (Hernandez et al., 2010). The distal tibia was strapped to the sitting surface, arms were folded across the chest, and back was again leaning against a support. Participants pushed maximally into a padded footplate either upward and toward (dorsiflexors) or downward and away (plantar flexors) from their tibia. Specific instructions were given to direct the force about the ankle joint rather than pushing or pulling with the entire leg. Verbal encouragement was provided by the experimenters to encourage maximum effort in all of the strength trials. Each peak torque value was normalized by body weight, and expressed in Nm/kg (Bohannon, 2009).

Passive ranges of motion about the flexion/extension axes of the hip, knee, and ankle joints of the dominant leg were assessed using a Leighton flexometer according to the measurement guidelines described in MacDougall et al (1991). Hip flexion was assessed with the participant lying supine on a massage table with their knee extended and the flexometer fastened to the lateral side of the upper thigh. The experimenter moved the participant's leg in an arc upward and toward the forehead as far as possible by applying a constant lifting force to the calf region while the pelvis was stabilized on the table. Hip extension was evaluated with the same leg and flexometer orientation, but with the participant lying prone. The experimenter lifted the leg in an arc upward and backward as far as possible, applying a constant force to the thigh region, with the leg straight and pelvis secured to the table. Knee flexion and extension were evaluated with the participant lying prone with their knees at the end, and lower legs extending beyond the end of the table. For knee flexion, the flexometer was fastened to the lateral side of the lower leg, which was moved by the experimenter about the knee in an arc upward and backward toward the buttocks. Knee extension was evaluated by moving the lower leg about the knee in an arc downward toward the floor as far as possible. The instrument was zeroed before each movement with the lower leg hanging off the end of the table in a relaxed position. Finally, ankle plantar flexion and dorsiflexion were assessed with the participant in a sitting position, their dominant leg resting on, and their foot projecting over, the end of the table. The non-dominant leg was allowed to rest on the floor to the side of the table. With the instrument fastened to the medial side of the foot, the ankle was rotated either downward and away (plantar flexion), or upward and toward (dorsiflexion) the lower leg as far as possible. The instrument was zeroed before each movement with the ankle in a relaxed position, projecting over the end of the table as described above. For each joint, the experimenter terminated the movement if the participant felt any discomfort or the experimenter felt significant resistance to joint rotation.

Finally, balance confidence was assessed using the shortened version of the Activities-specific Balance Confidence (ABC) scale, known as the ABC-6 (Schepens et al., 2010). The ABC-6 is a 6-item questionnaire that asks respondents to rank their “self-confidence”, from 0 to 100%, that they will “not lose their balance or become unsteady” when performing a series of tasks (see Appendix C). The shortened version includes only the most challenging tasks from the original ABC, and has shown excellent concurrent validity with, and a stronger relationship to falls than, the original (Schepens et al., 2010). It was administered verbally to the participants at the end of the experimental session.

4.2.4 Data Processing and Analysis

All raw kinetic and kinematic data were processed using custom pipelines in Visual3D motion capture analysis software (Visual3D v4, C-Motion, Germantown, MD, USA). Data points where marker signals were occluded were interpolated using a cubic spline. Based on the results of residual analyses and consultation with previous works examining similar movements, 4th order dual pass Butterworth filters with 4 Hz cutoff frequency for marker data and 10 Hz cutoff frequency for force plate data were used to attenuate noise in the signals (Hernandez et al., 2012 and 2013; Mourey et al., 2000). It was not anticipated that using different cutoff frequencies to filter the kinematic and kinetic data would induce significant artifacts. While a previous report suggested that using different cutoff frequencies can induce discrepancies in the relation between calculated peak forces and moments at the knee during the impact phase of a jump (Bisseling and Hof, 2006), the movements in the current study were of considerably lower frequency, and were thus unlikely to be affected in the same manner. The entire body was modeled as a rigid system of 15 independently tracked segments. Segment masses and centre of mass (COM) locations were estimated using de Leva's body segment parameters (de Leva, 1996). The trunk was modeled as two distinct segments: one representing the abdomen and pelvis region, the other representing the thorax. The abdomen and pelvis were modeled using a hybrid of the lower and middle portions of de Leva's trunk segments, while the thorax was modeled using only the upper portion (de Leva, 1996). The endpoints of the hybrid abdomen and pelvis segment for the present study were represented by the midpoint of the hip joint centres proximally, and the midpoint of a vector connecting the xiphoid process and the T10 vertebra distally. Average segment lengths of a sample of males and females from Zatsiorsky, Seluyanov, and Chugunova (1990) were used to calculate the length of the hybrid abdomen and pelvis segment. These segment lengths, in conjunction with the relative segmental COM locations of the individual lower- and middle-trunk segments, were then used to compute the hybrid abdomen and pelvis segment COM location which was to lie on a longitudinal axis running from the proximal to distal segment endpoints. The total body centre of mass position was calculated as a weighted average of all body segment COM locations, where each segment was weighted according to its mass proportion. Final COM location was referenced to the COP position derived from the force platform, such that the mean location of both variables was the same in the AP direction. Whole body COM and COP locations were expressed in the global coordinate system described in section 3.1.3.

Joint angles were calculated for the left ankle, left knee, and the hip (spanned by the thorax and left thigh segments), using the Cardan sequence Z-X-Y, corresponding to flex/extension – abd/adduction – axial rotation. In order to simplify the analysis, the left lower limb was chosen over the right. Moreover, during processing, it was noted that the rigid body placed on the right thigh became loose after calibration for one of the participants, rendering unrealistic joint angles that did not represent true movements. Nevertheless, differences in flexion between left and right sides were minimal (mean difference = 8% during the floor level poker chip retrieval task), as all participants were approximately symmetrical in their movements. Only flexion angles were considered in the present study to describe the postures used for the tasks, as this is the primary axis of movement for forward stooping and crouching movements. In defining the hip angle, the thorax was chosen over the hybrid abdomen and pelvis segment as it better represented the full extent to which participants were bending downwards with their torso. Centre of pressure (COP) location was calculated and spatially synchronized with kinematic data using Visual3D (Visual3D v4, C-Motion, Germantown, MD, USA). Time-series data representing individual marker data, the joint angles outlined above, total body COM, and COP were then exported to Matlab (MATLAB r2011b, Mathworks Inc., Natick, MA, USA) for further analysis.

Each trial was divided into distinct movement phases. For all tasks that required picking up only one object (all varying lift height tasks involving a single poker chip, the dustpan task, and the 1-chip bimanual task), trials were divided into two phases: the transition down into the pick-up posture, and the transition back up from the posture to standing. For the longer bimanual tasks (4, 8, and 12 poker chips), a static phase was also defined in between the transition down and transition up phases, as participants had to maintain the posture they chose to use while picking up the required number of chips. The onset and end of movement for all trials were defined using the vertical velocity of the vertex (top of the head) marker. Movement onset was defined by tracing backwards from the sample with the peak downward vertical vertex velocity to locate the first sample at which the velocity exceeded 5% of the maximum (Teasdale et al., 1993). The end of the movement was defined in a similar fashion, but by tracing forwards from the sample with the peak upward vertical vertex velocity and locating the first frame that was below the 5% threshold. For all single object retrieval tasks, the transition-down phase was separated from the transition-up phase by the moment of object retrieval, which coincided with the frame at which the vertical position of the vertex marker was closest to the ground (minimum value). This approach was verified visually in Visual3D. The longer bimanual tasks (4, 8, and 12 chips) were considered to have a 'static' phase, as mentioned above, which was defined using an algorithm similar to that used for movement onset determination. In this case, the start of the static phase coincided with the first sample at which the vertical vertex velocity slowed below the 5% of peak downward velocity threshold, and the end of the static phase coincided with the last sample before the vertical vertex velocity increased above the 5% threshold.

The minimum vertical height of the COM was calculated for each trial in order to describe the proximity of each participant's total body mass to the ground while picking up the required object(s). Maximum ankle, knee, and hip joint angles were calculated about the flex/extension axis to describe body configuration when retrieving the object(s) during each task. Maximum joint angular and vertical COM velocities were also calculated during the transition down and transition up phases of each task, describing how fast participants were moving individual segments and their total body COM. Additionally, the vertical positions of the calcaneus markers were monitored to determine whether participants were lifting their heels up into a "forefoot crouch" base of support configuration. The frames at which the heels lifted off (and subsequently returned back down to) the support surface were determined using the method described above for determining whole body movement onsets and offsets, but the calcaneus markers were used instead of the vertex marker. This algorithm was only invoked if the mean vertical position of the calcaneus marker, during the movement phase, was at least 1.5 cm above its resting position during the first second of the trial, indicating that the heel was raised above the floor. The relatively small 1.5 cm vertical threshold was used because there were a few cases in which the heel was raised only minimally above the floor. Furthermore, all cases in which the algorithm identified a 'heels up' condition were verified by inspecting video recordings of these trials. Base of support conditions used for each task were then described as either "flatfoot" or "forefoot", depending on whether the heels were lifted above, or remained on, the floor.

In addition to the kinematic measures outlined above, a number of balance control measures were calculated in order to describe the behavior of the COM and COP independently, their interaction with one another, and their proximity to the physical boundaries of the base of support (BOS). All measures were calculated in the anterior-posterior (AP) direction only, as the tasks required movements occurring primarily in the AP (sagittal) plane.

The minimum margin of safety (MMOS) was defined by computing the minimum horizontal distance between the COM and the physical boundary of the BOS in the anterior and posterior directions during each trial. The anterior BOS boundary was represented by the position of the tip of the 2nd metatarsal marker on the foot located farthest forwards in the anterior direction, while the posterior BOS boundary was defined by either the position of the calcaneus (in the case of a flatfoot BOS) or the head of the 5th metatarsal (if the heels were above the ground, in a forefoot BOS) marker on the foot located farthest backwards in the posterior direction. MMOS values were divided by the length of the BOS in the AP direction in order to account for differences in foot size. Minimum margin of safety values in the anterior (*MMOS_Ant*) and posterior (*MMOS_Post*) directions provided a measure of the range of the COM excursion during the movements relative to the available BOS. Moreover, these values described how close individuals allowed their whole body COM to approach their AP BOS boundaries (i.e., their stability limits).

Centre of mass and centre of pressure mean velocities (*COM_vel* and *COP_vel*, respectively) were calculated in the AP direction for each trial as the average of the absolute instantaneous velocities at each frame within the trial. Instantaneous velocity was calculated using a forward finite difference technique. COM mean velocity provided an indication of how fast participants were translating their body mass in the AP direction during the trial, while mean velocity of the COP represented how fast the COP moved in order to control the COM.

The final two measures sought to describe the relationship between the COM and COP. The number of crossings (*numCross*) defined the number of times the COP signal crossed over the COM in the AP direction. This value was normalized by the time taken to perform each task, and expressed as the number of crossings per second. The root mean square distance between the COP and COM (*COPtoCOM_rms*) was calculated as the root mean square average of the COP to COG separation distance in the AP direction throughout each trial. Where *numCross* described the frequency with which the COP crossed over the COM in an attempt to control its location, *COPtoCOM_rms* provided a measure of the average distance between the COP and COM during the trial; reflecting the potential error or 'overshoot' during scenarios in which the COM position had to be tightly regulated. Each balance control measure was calculated for the entire movement phase – that is, from movement onset until end of movement, as defined above – of each task. The independent and dependent variables comprising this study are summarized in Table 4.2 and Table 4.3, respectively.

Table 4.2 Description of the independent variables in the study

Independent Variables	**Definition**
Age	Between-subjects factor and distinguishing characteristic of the two groups tested.
Initial Lift Height [a]	Within-subjects factor: initial height from which the object (a poker chip) was retrieved. Heights tested: 50, 40, 30, 20, and 10 % of hip height, and floor level.
Precision [a]	Within-subjects factor: relative precision demand of the task. Two levels were tested: dustpan at floor level (low precision due to large, easy-to-grasp handle), and poker chip at floor level (described above, requiring more precision to grasp than the dustpan).
Task Duration [a]	Within-subjects factor: time required to complete the task. Modified by increasing the number of poker chips (from 1 to 4, 8, and 12) that had to be retrieved from the floor.

[a] These are the ‘elements’ modified in order to provide more challenging variations of stooping and crouching tasks

Table 4.3 Table summarizing the dependent variables examined in this study

Kinematics	**Definition**
COM Ht [a] (m)	Minimum height of the whole-body COM.
Hip Angle [a] (°)	Maximum flexion angle of the hip joint – spanned by the thorax and thigh segments.
Knee Angle [a] (°)	Maximum flexion angle of the knee joint – spanned by the lower leg and upper leg (thigh) segments.
Ankle Angle [a] (°)	Maximum dorsiflexion angle of the ankle joint – spanned by the lower leg and foot segments.
COM vertical vel [b] (m/s)	Maximum linear vertical velocity of the whole-body COM.
Hip vel [b] (°/s)	Maximum angular flexion velocity of the hip joint.
Knee vel [b] (°/s)	Maximum angular flexion velocity of the knee joint.
Ankle vel [b] (°/s)	Maximum angular flexion velocity of the ankle joint.
Balance Control	**Definition**
MMOS_Ant (% BOS)	Minimum horizontal distance between the COM and the anterior boundary of the BOS, expressed as a percentage of AP BOS length.
MMOS_Post (% BOS)	Minimum horizontal distance between the COM and the posterior boundary of the BOS, expressed as a percentage of AP BOS length.
COM_vel (mm/s)	Absolute mean velocity of the COM in the AP direction.
COP_vel (mm/s)	Absolute mean velocity of the COP in the AP direction.
numCross (#/s)	Number of times the COP crossed over the COM in the AP direction during the entire movement, divided by the duration of the movement.
COPtoCOM_rms (mm)	Root mean square distance between the COP and COM in the AP direction during the entire movement.
Subject Characteristics	**Definition**
Age (yrs)	Participant age, in years.
Height (m)	Participant height, in metres.
Mass (kg)	Participant mass, in kilograms.
Body mass index (kg/m^2)	Participant body mass index, calculated as body mass divided by height squared.
Isometric strength (Nm/kg)	Peak torque output of the knee extensors and flexors, and ankle dorsi flexors and plantar flexors, as measured by a Cybex II Dynamometer.
Passive ROM (°)	Maximum passive range of motion of the hip, knee, and ankle joint, as measured by a Leighton flexometer.
Balance Confidence (%)	Results of the ABC-6 balance confidence scale.

[a] These measures describe the postural configuration at the moment of object retrieval
[b] Calculated for transitions down (TD) into and transitions up (TU) out of pick-up postures

4.2.5 Statistics

Assumptions of normality were confirmed using the Shapiro-Wilk test for each dependent variable. Independent *t*-tests were performed to assess differences in age, height, weight, BMI, isometric strength, range of motion, and balance confidence between young and older participants. Mixed-model repeated measures ANOVAs were performed on each kinematic and balance control measure (see Table 4.2), to investigate main effects of age and task element (initial lift height, task duration, and precision required), as well as potential interactions between these variables. In cases where significant differences were not observed, trends and mean differences were reviewed and commented on to support inferred potential effects. Mauchly's sphericity test was used to assess the homogeneity of variance for each repeated measures ANOVA. In cases where sphericity was violated ($p < 0.05$), the estimate of departure from sphericity, denoted by epsilon (ε), was evaluated. As recommended by Girden (1992), if $\varepsilon > 0.75$, the Huynh-Feldt correction was used; whereas if $\varepsilon < 0.75$, the Greenhouse-Geisser correction was used. If significant interaction effects were present and ordinal in nature, they were reported in concert with main effects and a description of the manner in which the independent variables 'age' and 'task element' jointly combined to influence the dependent measure (Howell, 2002). If interactions were significant and disordinal in nature, main effects of age and/or task element were not interpreted independently. The Firsher exact test was used to determine group differences in BOS condition (flatfoot or crouch). All analyses were performed with statistical analysis software (SPSS Version 18.0, SPSS Inc., Chicago, IL, USA) using a significance level of $\alpha = 0.05$.

4.3 Results

The study results are separated into two sections summarizing the kinematic (section 4.3.1) and balance control (section 4.3.2) measures independently. Each section is further subdivided into the three distinct task elements being examined in the study: initial lift height, precision demand, and task duration.

Physical Characteristics and Balance Confidence

Younger participants exhibited 44% greater knee extensor, 46% greater knee flexor, and 39% greater ankle plantar flexor strengths, compared to their older counterparts. They also displayed passive ranges of motion that were 38% higher in hip extension and 13% higher in knee flexion. While differences in balance confidence were not significant (t=1.86, p = 0.076), older adults were, on average, 7% less confident in their ability to maintain balance than to their younger counterparts. Age-related differences in isometric strength, range of motion, and balance confidence are summarized in Table 4.4 below.

Table 4.4 Mean (SD) and t-test results for physical characteristics and balance confidence scores

		Younger (n=12)	Older (n=12)	t-value	p-value
Strength (Nm/kg)	Knee extensor[a]	2.28 (0.75)	1.58 (0.67)	2.393	0.026[a]
	Knee flexor[a]	1.05 (0.37)	0.72 (0.24)	2.616	0.016[a]
	Ankle plantar flexor[a]	1.21 (0.38)	0.87 (0.33)	2.404	0.025[a]
	Ankle dorsiflexor	0.59 (0.21)	0.58 (0.26)	0.082	0.936
Range of Motion (degrees)	Hip flexion	97.5 (18.8)	88.4 (14.8)	1.314	0.202
	Hip extension[a]	54.4 (22.7)	39.4 (7.7)	2.164	0.042[a]
	Knee flexion[a]	144.5 (11.5)	128.7 (15.3)	2.863	0.009[a]
	Knee extension	7.6 (5.7)	4.3 (1.8)	1.873	0.074
	Ankle dorsiflexion	51.3 (13.4)	46.3 (11.3)	1.006	0.325
	Ankle plantarflexion	25.8 (9.2)	23.5 (2.4)	0.849	0.405
Balance Confidence (ABC-6 scores)		91.1 (7.3)	83.8 (11.6)	1.863	0.076

[a]Indicates significant age-effect ($p < 0.05$).

4.3.1 Kinematics

4.3.1.1 Initial Lift Height

Posture Description

Across all the varying initial lift height tasks, younger participants positioned their COM 7 cm (10%) lower than their older counterparts ($F_{1,22}$ = 7.33, p = 0.013; Fig. 4.2A). Not surprisingly, there was a main effect of lift height ($F_{5,22}$ = 256.48, p < 0.001), with participants in both age groups gradually lowering their COM closer to the floor as lift height decreased. A significant ordinal interaction effect between age and lift height was also observed, by which the difference in COM height between young and older participants increased as lift height decreased toward floor level ($F_{5,22}$ = 3.35, p = 0.046). Younger participants achieved this lower COM position by exhibiting 10% more hip flexion ($F_{1,22}$ = 13.19, p = 0.001; Fig. 4.2B), 84% more knee flexion ($F_{1,22}$ = 12.86, p = 0.002; Fig. 4.2C), and 146% more ankle dorisflexion ($F_{1,22}$ = 9.14, p = 0.006; Fig. 4.2D) compared to older participants. Significant main effects of lift height were also observed for the maximum joint angles representing postural configuration during the moment of chip retrieval, with flexion increasing by 182% in the hip ($F_{5,22}$ = 649.96, p < 0.001; Fig. 4.2B), 275% in the knee ($F_{5,22}$ = 50.37, p < 0.001; Fig. 4.2C) , and 465% in the ankle ($F_{5,22}$ = 26.54, p < 0.001; Fig. 4.2D), as lift height decreased from 50% of hip height toward floor level. While no significant interaction effects between age and lift height were observed for any of the above joint angles, knee flexion ($F_{5,22}$ = 2.87, p = 0.069; Fig. 4.2C) and ankle dorsiflexion ($F_{5,22}$ = 2.16, p = 0.111; Fig. 4.2D) angles tended to increase more in younger compared to older participants as lift height decreased toward floor level (Fig. 4.2C and D).

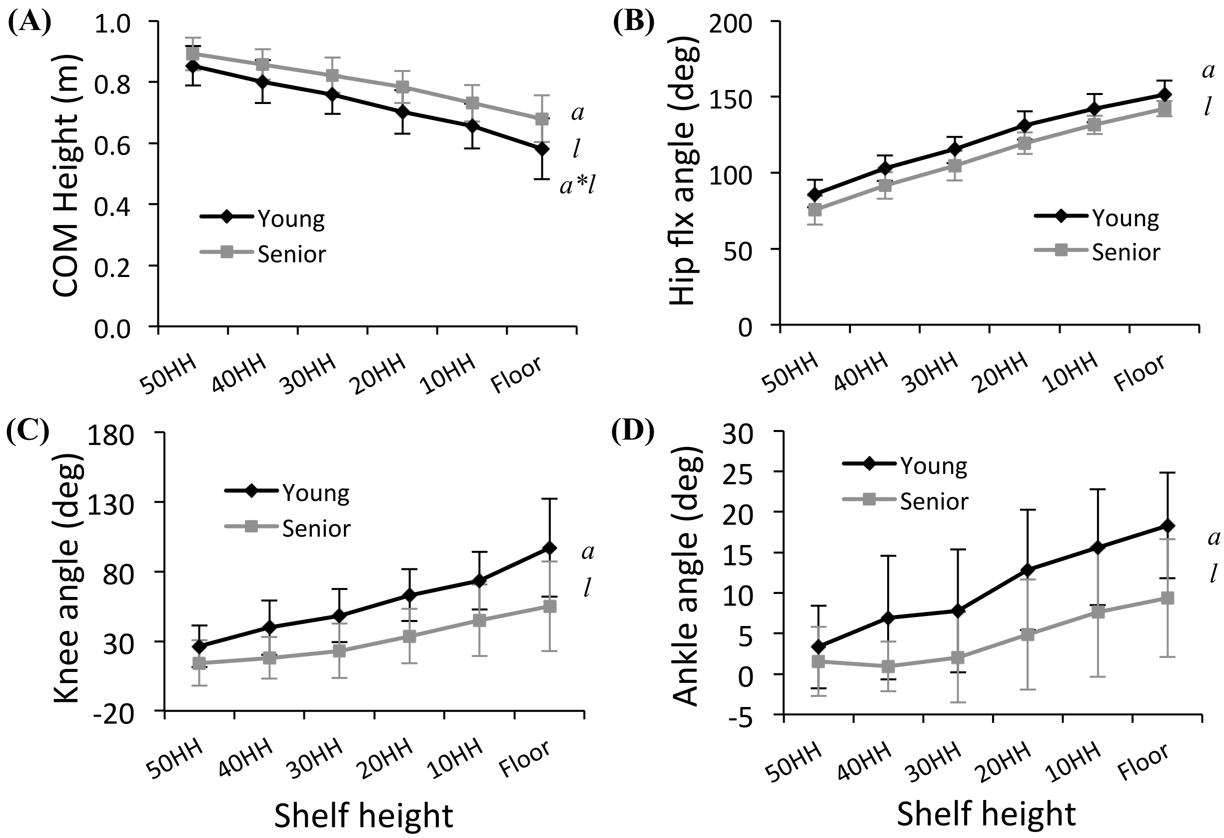


Figure 4.2 Kinematic descriptors of the postures used by older (grey line) and younger (black line) participants during the varying lift height tasks: (A) centre of mass height, (B) hip (thorax-thigh) flexion angle, (C) knee flexion angle, and (D) ankle dorsiflexion angle. Error bars represent +/- one standard deviation from the mean. Significant age (*a*), lift height (*l*), and interaction (*a*l*) effects are indicated on the figure.

Speed of Movement

In addition to positioning their COM closer to the ground by increasing hip, knee, and ankle flexion, younger participants moved faster than older participants, achieving greater maximum angular velocities in each joint and in their vertical COM during movements into (Fig. 4.3) and out of (Table 4.5) each posture. Specifically, during the transition down phase, younger participants achieved an average maximum downward COM velocity of 0.39 (SD = 0.17) m/s across all tasks, which was 41% faster than the 0.28 (SD = 0.10) m/s of the older cohort ($F_{1,22}$ = 13.38, p = 0.001; Fig. 4.3A). Both age groups also increased their maximum downward COM velocities as they reached to lower shelf heights ($F_{5,22}$ = 90.89, p < 0.001; Fig. 4.3A), although younger participants did so to a greater extent than their older counterparts ($F_{5,22}$ = 7.76, p < 0.001; Fig. 4.3A).

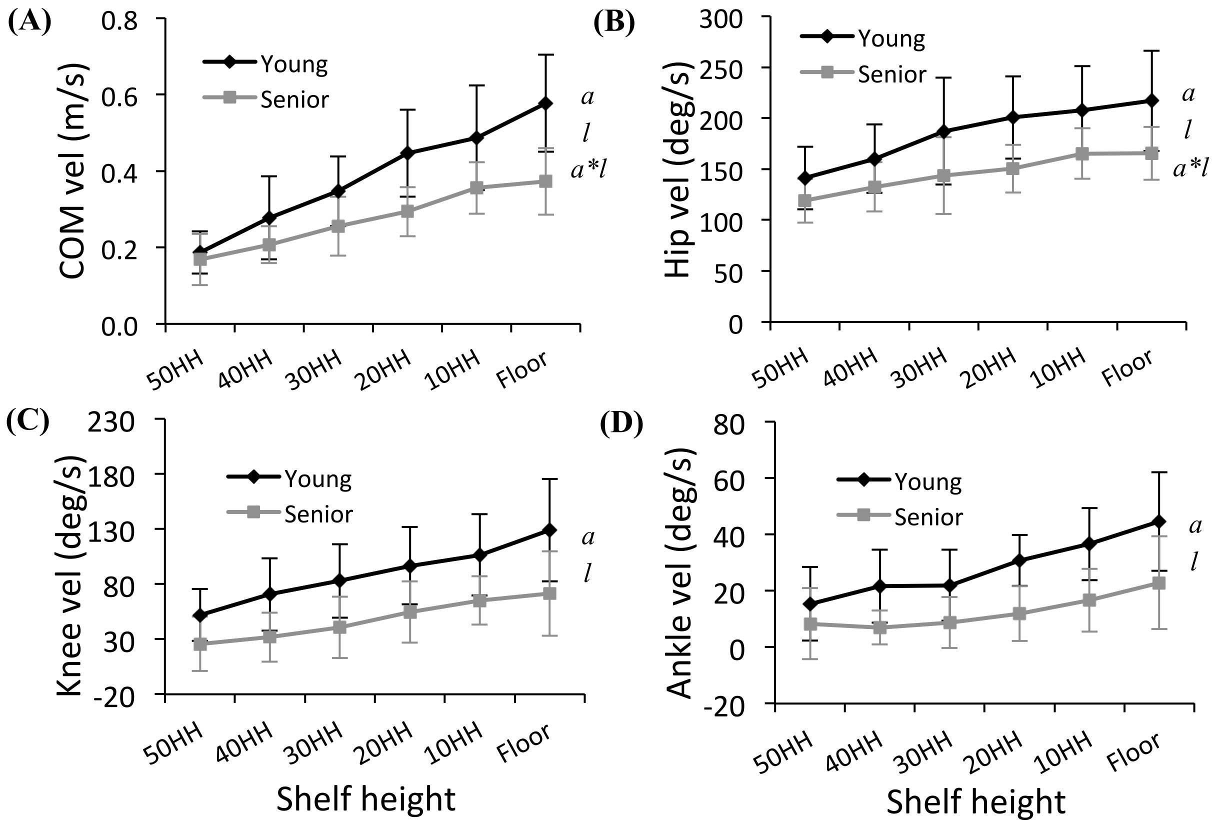


Figure 4.3 Vertical centre of mass and joint angle maximum velocities for older (grey line) and younger (black line) participants during the transition-down phase of each movement in the varying lift height tasks: (A) centre of mass maximum downward vertical velocity, (B) maximum hip (thorax-thigh) flexion velocity, (C) maximum knee flexion velocity, and (D) maximum ankle dorsiflexion velocity. Error bars represent +/- one standard deviation from the mean. Significant age (*a*), lift height (*l*), and interaction (*a*l*) effects are indicated on the figure.

Similar age-effects were observed for maximum joint angular velocities, with younger participants achieving values that were on average, 27%, 86%, and 127%, faster in the hip ($F_{1,22}$ = 9.79, p = 0.005; Fig. 4.3B), knee ($F_{1,22}$ = 15.84, p = 0.001; Fig. 4.3C), and ankle ($F_{1,22}$ = 23.99, p < 0.001; Fig. 4.3D) joints, respectively, compared to older participants. Significant main effects of lift height were also observed for the maximum hip ($F_{5,22}$ = 38.86, p < 0.001), knee ($F_{5,22}$ = 28.67, p < 0.001), and ankle ($F_{5,22}$ = 15.91, p < 0.001) joint angular velocities, with both age groups moving faster as lift height decreased toward the floor. A significant ordinal interaction effect between age and lift height was observed only for the maximum angular velocity of the hip joint ($F_{5,22}$ = 2.51, p = 0.034), with younger participants increasing their maximum velocities to a greater extent than their older counterparts as lift height decreased toward the floor. While no interaction effects were present for the maximum knee ($F_{5,22}$ = 1.40, p = 0.255) or ankle ($F_{5,22}$ = 1.66, p = 0.187) joint angular velocities, graphical trends appear similar to that of the hip, with younger participants tending to increase their knee and ankle joint angular velocities disproportionately more than older participants as lift height decreased.

Kinematic trends during the transition-up phase of the movement were similar to those described above for the transition-down phase, with younger participants achieving greater vertical COM, hip, knee, and ankle joint angular maximum velocities than their older counterparts and significant interactions occurring for the COM upward and knee angular velocities. A summary of the ANOVA results is provided in Table 4.5 below.

Table 4.5 Age group means (SD) and differences for maximum upward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during transition-up phase (TU) of the varying lift height tasks. *F*-ratios (*p*-value) from the repeated measures ANOVA tests are shown for each velocity measure.

	Group Means			ANOVA		
TU Max Velocity	Younger	Older	Mean diff	Age	Lift Height	Interaction
COM vert (m/s)	0.417 (0.175)	0.317 (0.119)	0.1[a]	12.88 (0.002)[a]	117.1 (<0.001)[b]	4.17 (0.015)[ab]
Hip vel (°/s)	-174.3 (40.2)	-144.1 (32.4)	-30.2[a]	8.67 (0.008)[a]	58.56 (<0.001)[b]	0.61 (0.639)
Knee vel (°/s)	-91.9 (48.6)	-48.1 (29.9)	-43.9[a]	17.14 (<0.001)[a]	29.75 (<0.001)[b]	3.36 (0.024)[ab]
Ankle vel (°/s)	-25.9 (15.0)	-11.1 (10.4)	-14.8[a]	24.73 (<0.001)[a]	18.14 (<0.001)[b]	1.26 (0.295)

[a] significant age effect ($p < 0.05$)
[b] significant lift height effect ($p < 0.05$)
[ab] significant interaction (age*lift height) effect ($p < 0.05$)

BOS Condition

Participants in both age groups used a flatfoot BOS for all lift-height conditions except floor level. When retrieving the chip from the floor, 3 (25%) younger participants and 1 (8.3%) older participant used a forefoot crouch with their heels above the support surface, although this difference in proportions was not significant (p = 0.295).

4.3.1.2 Precision Required

Posture Description

Main effects of age were consistent with trends observed during the varying lift height tasks, as younger participants used postures characterized by 12% lower COM heights ($F_{1,22}$ = 6.89, p = 0.015), achieved through 6% more hip flexion ($F_{1,22}$ = 5.01, p = 0.036), 63% more knee flexion ($F_{1,22}$ = 9.10, p = 0.003), and 79% more ankle dorsiflexion ($F_{1,22}$ = 7.74, p = 0.011) during both 'precision' tasks (Fig. 4.4). Main effects of precision were also observed ($F_{1,22}$ = 28.53, p < 0.001; Fig. 4.4A) with both age groups positioning their COM at an average height of 0.63 (SD = 0.10) metres above the ground when picking up the chip (more precision required), which was 6% lower than the 0.67 (SD = 0.08) metre COM height when retrieving the dustpan (less precision required). A significant ordinal interaction effect between age and precision was also observed ($F_{1,22}$ = 5.52, p = 0.028; Fig. 4.4C), by which young participants increased knee flexion by 26% (from 76.9 (SD = 25.5) for the dustpan to 97.1 (SD = 35.2) degrees for the chip) when retrieving the chip compared to the dustpan, whereas older participants increased knee flexion by only 6.7% (from 51.5 (SD = 19.6) to 55.0 (SD = 32.3) degrees). While this interaction was not significant for the hip ($F_{1,22}$ = 2.39, p = 0.137; Fig. 4.4B) and ankle ($F_{1,22}$ = 4.19, p = 0.053; Fig. 4.4D) angles, or for the minimum COM height ($F_{1,22}$ = 3.85, p = 0.063; Fig. 4.4A), similar trends to those observed for the knee angle were apparent, with age-related differences tending to increase when the task demanded greater precision (i.e., while retrieving the poker chip). Specifically, younger participants increased hip flexion by 5% and ankle dorsiflexion by 17%, while older participants increased by only 3% in both the hip and ankle joints when picking up the chip, compared to the dustpan. Similarly, minimum COM height decreased by 9% in young participants compared to only 4% in their older counterparts when precision demands increased.

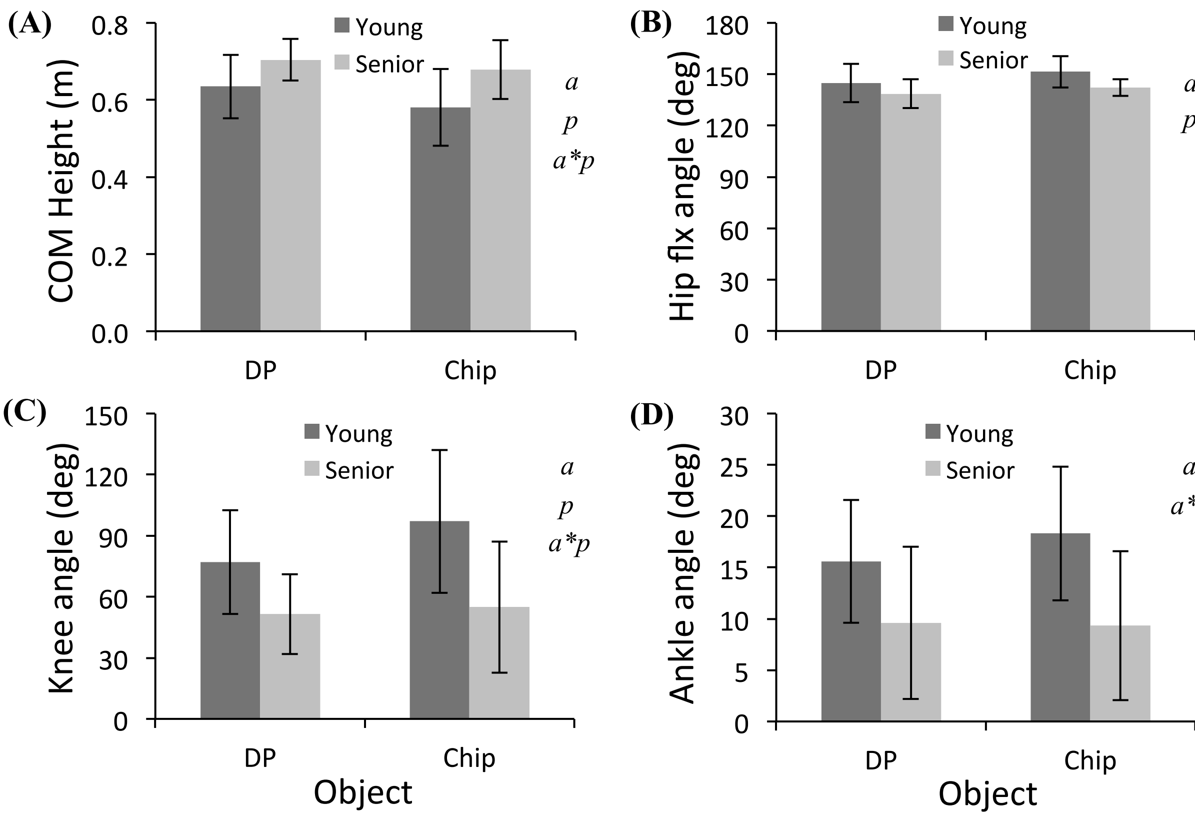


Figure 4.4 Kinematic descriptors of the postures used by older (light grey bars) and younger (dark grey bars) participants during the varying level of precision tasks: (A) centre of mass height, (B) hip (thorax-thigh) flexion angle, (C) knee flexion angle, and (D) ankle dorsiflexion angle. The dustpan ('DP') condition was considered to require less precision than the poker chip ('Chip'). Error bars represent +/- one standard deviation from the mean. Significant age (*a*), precision (*p*), and interaction (*a*p*) effects are indicated on the figure.

Speed of Movement

Consistent with the varying lift-height condition results, younger participants again moved faster than their older counterparts, achieving greater maximum vertical COM velocities and greater maximum hip, knee, and ankle joint angular velocities during movements into (Fig. 4.5) and out of (Table 4.6) postures used for the 'precision' tasks. Specifically, during the transition-down phase, younger participants achieved an average maximum downward COM velocity of 0.546 (SD = 0.141) m/s across both tasks, which was 42% faster than the 0.385 (SD = 0.077) m/s of the older cohort ($F_{1,22}$ = 13.25, p = 0.001; Fig. 4.5A). Similar age-effects were observed for the maximum hip ($F_{1,22}$ = 6.98, p = 0.015; Fig. 4.5B), knee ($F_{1,22}$ = 10.89, p = 0.003; Fig. 4.5C), and ankle ($F_{1,22}$ = 10.05, p = 0.004; Fig. 4.5D) joint angular velocities, with younger participants exhibiting velocities that were, on average, 24%, 65%, and 75% faster in the hip, knee, and ankle joints, respectively, compared to older participants. Significant main effects of 'precision' were observed only for the maximum ankle joint angular velocity ($F_{1,22}$ = 13.33, p = 0.001; Fig. 4.5), with values increasing in both age groups by approximately 35%, from an average of 25.0 (SD = 10.3) °/s for the dustpan (less precision required) retrieval task to 33.7 (SD = 20.0) °/s for the chip (more precision required) retrieval task.

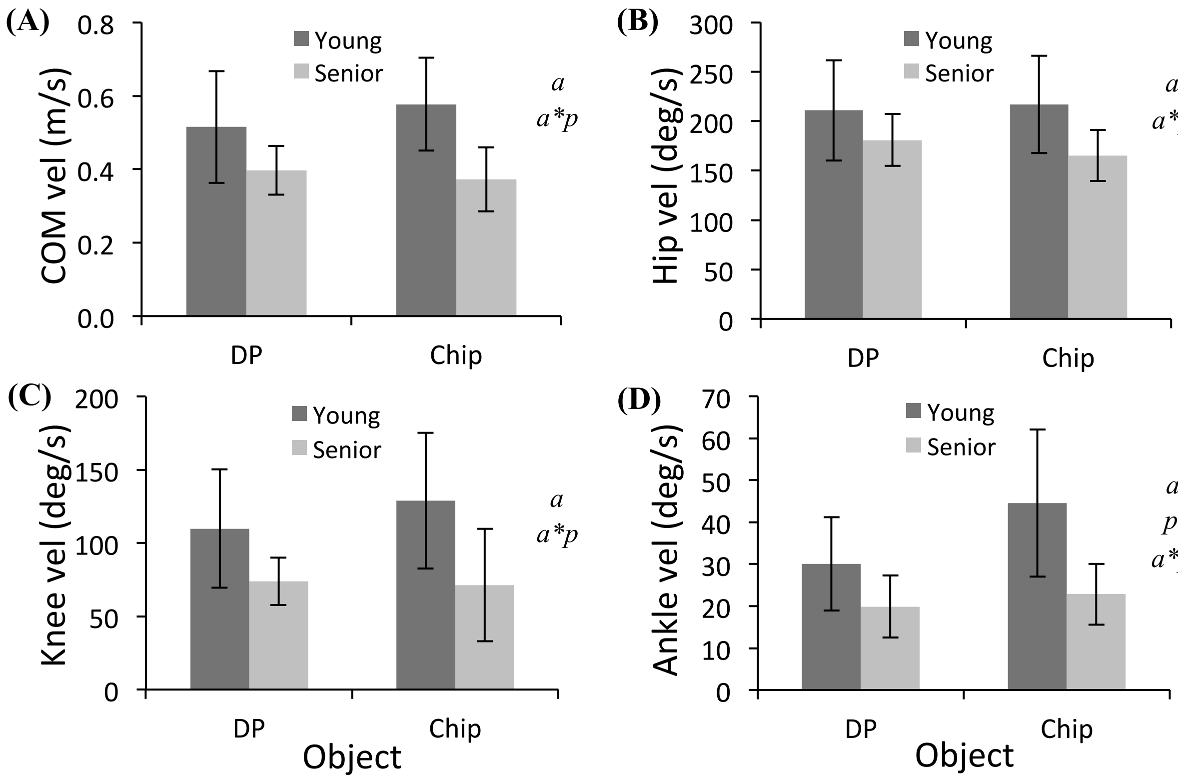


Figure 4.5 Vertical centre of mass and joint angle maximum velocities for older (light grey bars) and younger (dark grey bars) participants during the transition-down phase of each movement in the varying level of precision tasks: (A) centre of mass maximum downward vertical velocity, (B) maximum hip (thorax-thigh) flexion velocity, (C) maximum knee flexion velocity, and (D) maximum ankle dorsiflexion velocity. The dustpan ('DP') condition was considered to require less precision than the poker chip ('Chip'). Error bars represent +/- one standard deviation from the mean. Significant age (*a*), precision (*p*), and interaction (*a*p*) effects are indicated on the figure.

Significant ordinal interaction effects between age and precision were observed for the maximum downward COM velocity ($F_{1,22} = 10.65$, $p = 0.004$; Fig. 4.5A) and the maximum hip ($F_{1,22} = 4.35$, $p = 0.049$; Fig. 4.5B), and ankle ($F_{1,22} = 5.86$, $p = 0.024$; Fig. 4.5D) joint angular velocities, with age-related differences increasing when precision demands were higher (i.e., during the chip retrieval task). Interestingly, the nature of the interaction for the maximum downward COM and hip angular velocities was such that increased precision demand affected each age group oppositely; older participants seemed to slow down when the task required more precision, while younger participants sped up. Specifically, while maximum downward COM velocity increased in younger participants by 12%, from 0.515 (SD = 0.152) m/s when retrieving dustpan to 0.577 (SD = 0.127) m/s when retrieving the chip, it decreased by 6% in older participants, from 0.397 (SD = 0.066) m/s for the dustpan to 0.373 (SD = 0.087) m/s for the chip. Similarly, maximum hip angular velocity increased by 3% in young participants but decreased by 9% in older participants as a result of the increased precision demand associated with the chip task. The nature of the interaction on maximum ankle angular velocity during the transition down phase was ordinal and unidirectional, with velocity increasing in both age groups, but to a greater extent in younger compared to older participants (35% vs 15%) as precision demands increased. Finally, while the interaction between age and precision on maximum knee angular velocity was not significant ($F_{1,22} = 4.16$, $p = 0.054$; Fig. 4.5C), trends were similar to those observed for the maximum downward COM and hip angular velocities, with values increasing in younger participants by 18% and actually decreasing in older participants by 9% as precision demands increased.

Speed of movement results during the transition-up phase of the movement were similar to those described for the transition-down phase, with younger participants achieving greater vertical COM, hip, knee, and ankle joint angular maximum velocities than their older counterparts. Main effects of precision were detected for upward COM and knee angular maximum velocities, while interactions between age and precision were not observed for any of the velocity measures. ANOVA results are summarized in Table 4.6.

Table 4.6 Age group means (SD) and differences for maximum upward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during the transition-up (TU) phase of the varying precision tasks. *F*-ratios (*p*-value) from the repeated measures ANOVA tests are shown for each velocity measure.

	Group Means			ANOVA		
TU Max Velocity	Younger	Older	Mean diff	Age	Precision	Interaction
COM vert (m/s)	0.593 (0.128)	0.443 (0.070)	0.15[a]	15.09 (0.001)[a]	12.74 (0.002)[b]	2.57 (0.123)
Hip vel (°/s)	-201.8 (35.5)	-167.3 (25.0)	-34.5[a]	8.57 (0.008)[a]	0.09 (0.773)	1.02 (0.323)
Knee vel (°/s)	-131.7 (48.1)	-69.8 (25.6)	-61.8[a]	18.44 (<0.001)[a]	6.35 (0.020)[b]	3.99 (0.058)
Ankle vel (°/s)	-33.2 (11.7)	-20.4 (8.4)	-12.8[a]	12.45 (0.002)[a]	0.41 (0.527)	0.06 (0.808)

[a] significant age effect ($p < 0.05$)
[b] significant precision effect ($p < 0.05$)
[ab] significant interaction (age*precision) effect ($p < 0.05$)

BOS Condition

When retrieving the dustpan (less precision required), only one (8%) younger participant lifted her heels off the ground into a forefoot BOS configuration, while all older participants remained flatfooted. For the chip task (more precision required), 3 (25%) younger participants and 1 (8%) older participant used a forefoot crouch with their heels above the support surface. Neither difference was significant (dustpan: $p = 0.500$, chip: $p = 0.295$).

4.3.1.3 Task Duration

Posture Description

Age-effects followed trends observed for posture-describing measures in previous task conditions, with younger participants employing 15% lower COM heights ($F_{1,22} = 6.42$, $p = 0.019$) , 6% more hip flexion ($F_{1,22} = 6.53$, $p = 0.018$), 69% more knee flexion ($F_{1,22} = 9.67$, $p = 0.005$), and 61% more ankle dorsiflexion ($F_{1,22} = 6.42$, $p = 0.019$) compared to their older counterparts across the varying duration tasks (Fig. 4.6).

Main effects of task duration were also observed for each measure, with both age groups decreasing their COM height by 10% ($F_{3,22} = 11.20$, $p < 0.001$), and increasing hip flexion by 5% ($F_{3,22} = 24.68$, $p < 0.001$), knee flexion by 25% ($F_{3,22} = 8.48$, $p < 0.001$), and ankle dorsiflexion by 13% ($F_{3,22} =$

3.81, $p = 0.030$) as task duration increased from the 1 chip to the 12 chip condition (Fig. 4.6). No interactions between age and task duration were observed for any of the posture-describing measures.

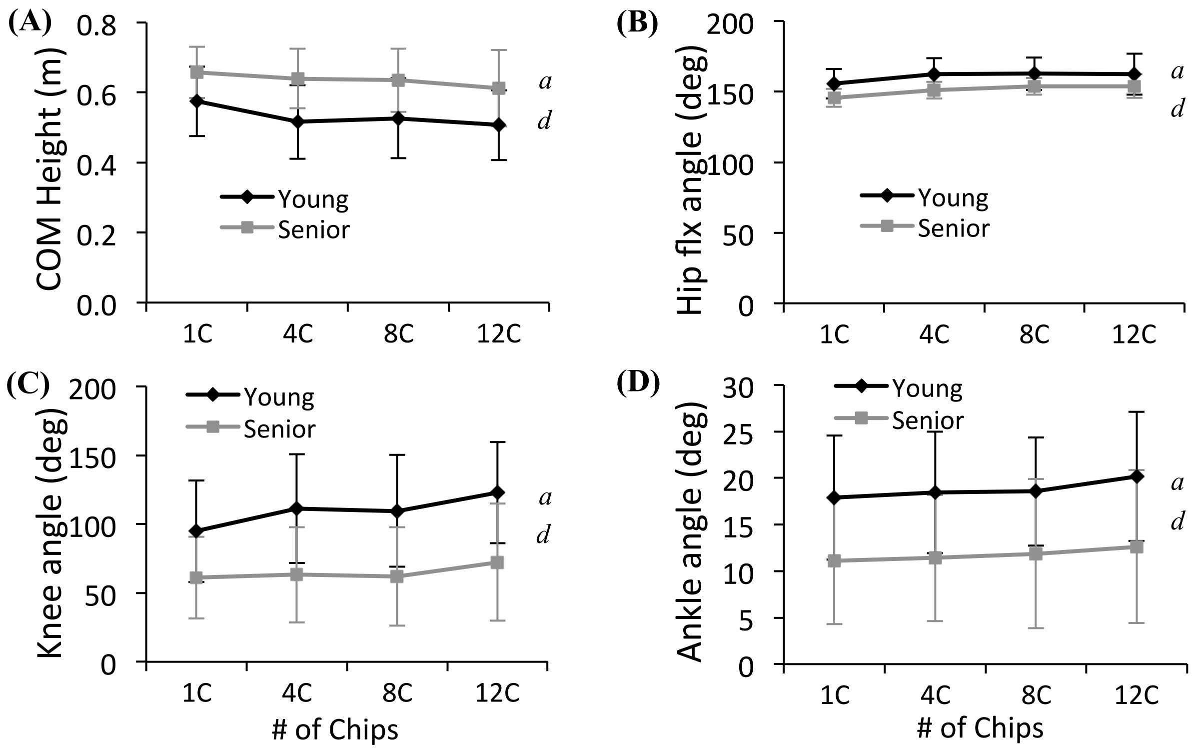


Figure 4.6 Kinematic description of the postures used by older (grey lines) and younger (black lines) participants during the varying duration tasks: (A) centre of mass height, (B) hip (thorax-thigh) flexion angle, (C) knee flexion angle, and (D) ankle dorsiflexion angle. Error bars represent +/- one standard deviation from the mean. Significant age (*a*) and duration (*d*) effects are indicated on the figure.

Speed of Movement

Consistent with the varying lift-height and precision conditions, younger participants moved faster than their older counterparts when performing the varying duration tasks, achieving greater maximum vertical COM velocities and greater maximum hip, knee, and ankle joint angular velocities during movements into (Fig. 4.7) and out of (Table 4.7) postures used for the tasks. Specifically, during the transition-down phase, younger participants achieved an average maximum downward COM velocity of 0.563 (SD = 0.157) m/s across all tasks, which was 35% faster than the 0.416 (0.093) m/s average maximum velocity of the older cohort ($F_{1,22} = 8.50$, $p = 0.008$; Fig. 4.7A).

Younger participants also achieved maximum angular velocities that were 22%, 77%, and 102% faster in the hip ($F_{1,22} = 7.02$, $p = 0.015$; Fig. 4.7B), knee ($F_{1,22} = 11.55$, $p = 0.003$; Fig. 4.7C), and ankle ($F_{1,22} = 12.09$, $p = 0.002$; Fig. 4.7D) joints, respectively, compared to older participants.

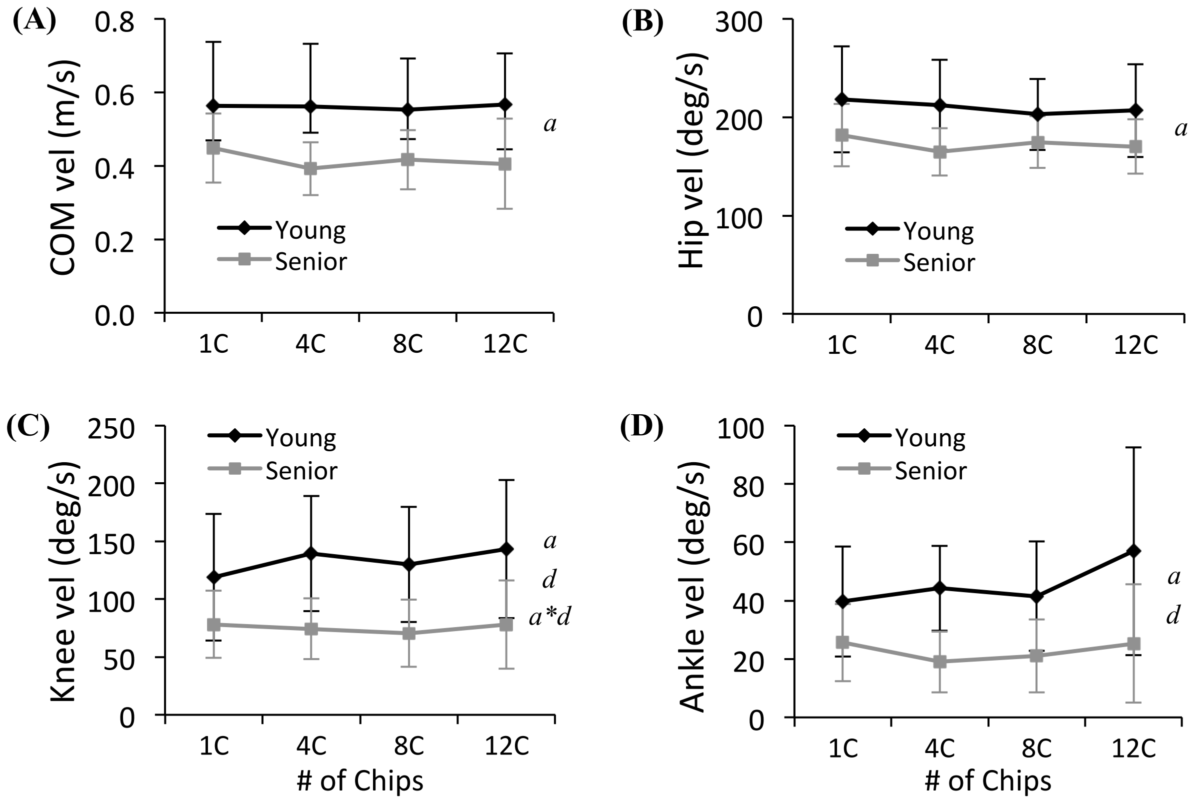


Figure 4.7 Vertical centre of mass and joint angle maximum velocities for older (grey lines) and younger (black lines) participants during the transition-down phase of each movement for the varying duration tasks: (A) centre of mass maximum vertical downward velocity, (B) maximum hip (thorax-thigh) flexion velocity, (C) maximum knee flexion velocity, and (D) maximum ankle dorsiflexion velocity. Error bars represent +/- one standard deviation from the mean. Significant age (*a*), duration (*d*), and interaction (*a*d*) effects are indicated on the figure.

While significant main effects of task duration were observed for the knee ($F_{3,22} = 2.88$, $p = 0.043$) and ankle joint ($F_{3,22} = 3.28$, $p = 0.046$) angular velocities during the transition-down phase, they appear to be driven primarily by increased velocities in younger participants. From the 1 chip condition (shortest duration) to the 12 chip condition (longest duration), younger participants increased maximum angular velocities in their knee and ankle joints by 21%, from 118.9 (SD = 54.9) to 143.3 (SD = 59.7) °/s, and 43%, from 39.8 (SD = 18.9) to 57.0 (SD = 35.7) °/s, respectively, while older participants saw virtually no change in their knee (0.3% increase) and ankle (1.2% decrease) joint angular velocities. The nature by which task duration caused these disproportionate changes, at least in the maximum knee angular velocity, was captured by a significant interaction effect ($F_{3,22}$ = 3.06, p = 0.034). While the same significant interaction was not present for the ankle ($F_{3,22}$ = 2.06, p = 0.139, the trend appears similar (Fig. 4.7D). Task duration did not affect maximum downward COM ($F_{3,22}$ = 0.87, p = 0.462) or maximum hip angular ($F_{3,22}$ = 2.20, p = 0.119) velocities, and no interaction effects between age and duration were observed for these measures (COM velocity: $F_{3,22}$ = 1.01, p = 0.393; hip angular velocity: $F_{3,22}$ = 1.07, p = 0.354).

Speed of movement results during the transition-up phase of the movement were slightly different from those described for the transition-down phase. While age-effects were similar, with younger participants achieving greater maximum vertical COM, hip, knee, and ankle joint angular velocities than their older counterparts across the varying duration tasks, main effects of duration were detected only for the maximum hip angular velocity, with maximum values decreasing (i.e., slowing down) as duration increased. Further, no interactions between age and duration were observed for any of the velocity measures. ANOVA results are summarized in Table 4.7 below.

Table 4.7 Age group means (SD) and differences for maximum upward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during transition-up (TU) phase of the varying duration tasks. *F*-ratios (*p*-value) from the repeated measures ANOVA tests are also shown for each velocity measure.

	Group Means			ANOVA		
TU Max Velocity	Younger	Older	Mean diff	Age	Duration	Interaction
COM vert (m/s)	0.578 (0.153)	0.419 (0.106)	0.16[a]	10.56 (0.004)[a]	0.25 (0.865)	0.72 (0.547)
Hip vel (°/s)	-185.9 (38.0)	-157.9 (23.0)	-28.0[a]	7.13 (0.014)[a]	2.78 (0.048)[b]	0.11 (0.953)
Knee vel (°/s)	-139.3 (54.3)	-69.0 (34.9)	-70.3[a]	15.18 (0.001)[a]	2.05 (0.116)	0.40 (0.752)
Ankle vel (°/s)	-43.8 (21.0)	-20.4 (12.8)	-23.4[a]	14.73 (0.001)[a]	0.99 (0.387)	0.96 (0.398)

[a] significant age effect ($p < 0.05$)
[b] significant duration effect ($p < 0.05$)
[ab] significant interaction (age*duration) effect ($p < 0.05$)

BOS Condition

Younger participants were 4 times more likely to use a 'forefoot' BOS than their older counterparts during the varying duration tasks, as they raised their heels off the support surface in 50% of the trials, compared to only 13% for older participants ($p < 0.001$; Fig. 4.8). Longer tasks seemed to encourage the forefoot BOS to a greater extent in younger participants than their older counterparts. In particular, during the 8-chip retrieval task, 8 out of 12 (75%) younger, compared to only 1 out of 12 (8%) older, participants used a forefoot BOS crouching posture. Age-related differences were significant in the 8 ($p = 0.005$) and 12 ($p = 0.050$), but not the 1 ($p = 0.295$) and 4 ($p = 0.077$) chip retrieval tasks.

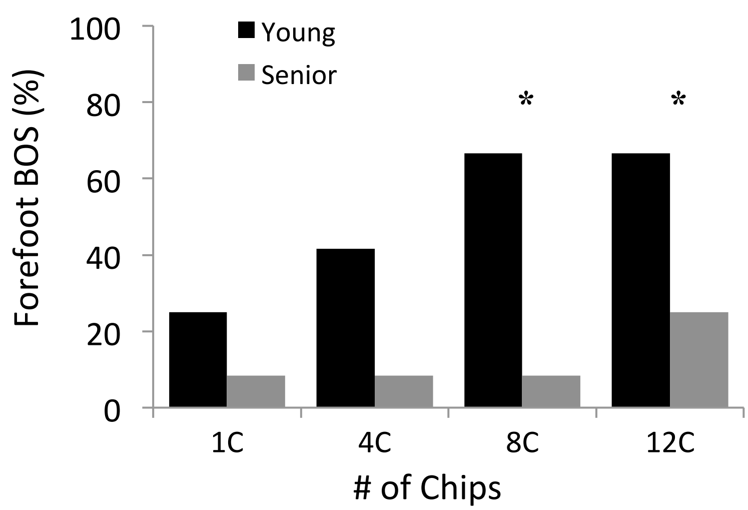


Figure 4.8 Comparison of BOS conditions between older (grey bars) and younger (black bars) participants. Data are displayed as proportions of participants in each age cohort (n=12 for both) using a heels up, forefoot BOS during the task. Significant age-related differences in proportions of participants using this posture are denoted by '*' for each level (i.e., number of chips) of the varying duration tasks.

4.3.2 Balance Control

4.3.2.1 Initial Lift Height

Independent COM and COP Behaviour

Age-effects were not significant for *MMOS_Ant* ($F_{1,17}$ = 0.00, p = 0.995; Fig. 4.9A) or *MMOS_Post* ($F_{1,17}$ = 3.67, p = 0.072; Fig. 4.9B) during the varying lift height tasks. Nevertheless, mean differences indicated that older participants displayed 12% larger (i.e., farther away from the posterior BOS boundary) *MMOS_Post* values (42.4 (SD = 7.1) % BOS) than their younger counterparts (38.0 (SD = 8.1) % BOS). Significant effects of lift height were observed for *MMOS-_Ant* ($F_{5,17}$ = 9.42, p < 0.001), with both age groups generally approaching the anterior BOS boundary as shelf height decreased toward the floor. Specifically, at floor level both age groups' *MMOS_Ant* average was 39.5 (SD = 10.0) % of BOS length, which was 21% closer than the 50.1 (SD = 5.8) % BOS length value at 50% of hip height. Main effects of lift height were not observed for *MMOS_Post* ($F_{5,17}$ = 0.54, p = 0.559), and no interactions between age and lift height were present for *MMOS_Ant* ($F_{5,17}$ = 0.18, p = 0.968) or *MMOS_Post* ($F_{5,17}$ = 0.14, p = 0.840).

A significant age-effect was observed for *COM_vel* ($F_{1,17}$ = 6.66, p = 0.019; Fig. 4.9C), with younger participants moving at an average velocity of 30.3 (SD = 13.4) mm/s, which was 36% faster than the 20.2 (SD = 10.2) mm/s of the older participants. No age-effect was observed for *COP_vel* ($F_{1,17}$ = 0.31, p = 0.583; Fig. 4.9D). Significant lift-height effects were observed for *COM_vel* ($F_{5,17}$ = 9.94, p < 0.001) and *COP_vel* ($F_{5,17}$ = 12.93, p < 0.001), with velocities increasing in both age groups as lift height decreased. Specifically, *COM_vel* increased by 83% from the 50% hip height (19.8 (SD = 12.5) mm/s) condition to floor level (36.3 (SD = 16.6) mm/s), while *COP_vel* increased by 53% from the 50% hip height (58.6 (SD = 21.6) mm/s) condition to floor level (89.7 (SD = 27.3) mm/s). No interaction effects between age and lift height were observed for *COM_vel* ($F_{5,17}$ = 1.44, p = 0.218) or *COP_vel* ($F_{5,17}$ = 0.25, p = 0.940).

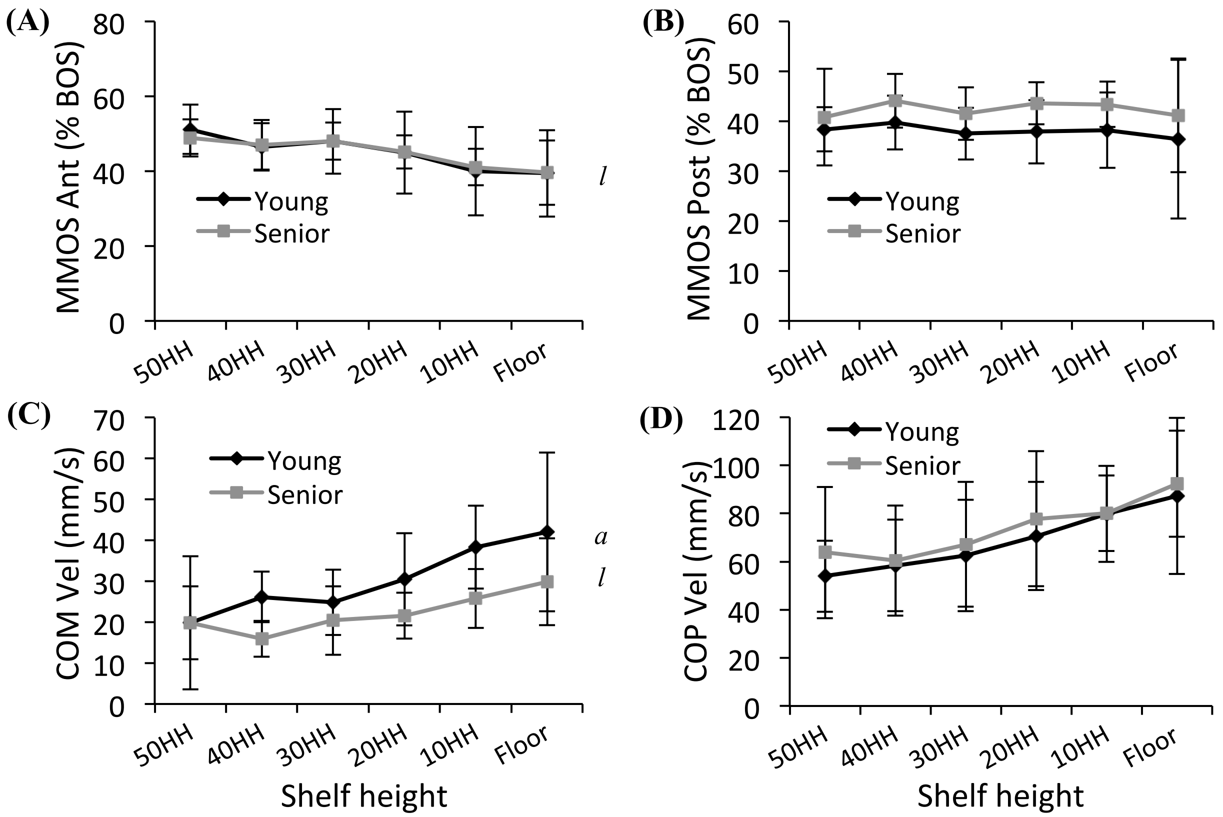


Figure 4.9 Balance measures describing COM and COP behavior of older (grey lines) and younger (black lines) participants during the varying lift height tasks: (A) COM minimum margin of safety (MMOS) to anterior BOS boundary, (B) COM minimum margin of safety (MMOS) to posterior BOS boundary, (C) mean COM anterior-posterior velocity, and (D) mean COP anterior-posterior velocity. Error bars represent +/- one standard deviation from the mean. Significant age (*a*) and lift height (*l*) effects are indicated on the figure.

Interaction between COM and COP

Age-effects were not significant for *numCross* ($F_{1,17} = 2.25$, $p = 0.152$; Fig. 4.10A) or *COPtoCOM_rms* ($F_{1,17} = 0.10$, $p = 0.752$; Fig. 4.10B). Nevertheless, mean differences indicated that older adults (1.76 (SD = 0.74) cross/sec) had approximately 21% more COP-COM crossings per second than younger participants (1.46 (SD = 0.63) cross/sec). While lift height did not affect *numCross* ($F_{5,17} = 0.09$, $p = 0.994$), it had a significant effect on *COPtoCOM_rms* ($F_{5,17} = 7.71$, $p <$ 0.001), with both age groups increasing by approximately 42% from the 50% hip height condition (10.7 (SD = 4.0) mm) to the floor (15.1 (SD = 5.4) mm). Finally, while interactions between age and lift height were not observed for *numCross* ($F_{5,17}$ = 1.99, p = 0.089) or *COPtoCOM_rms* ($F_{5,17}$ = 0.18, p = 0.970), age-related differences in *numCross* appeared to increase as lift height decreased.

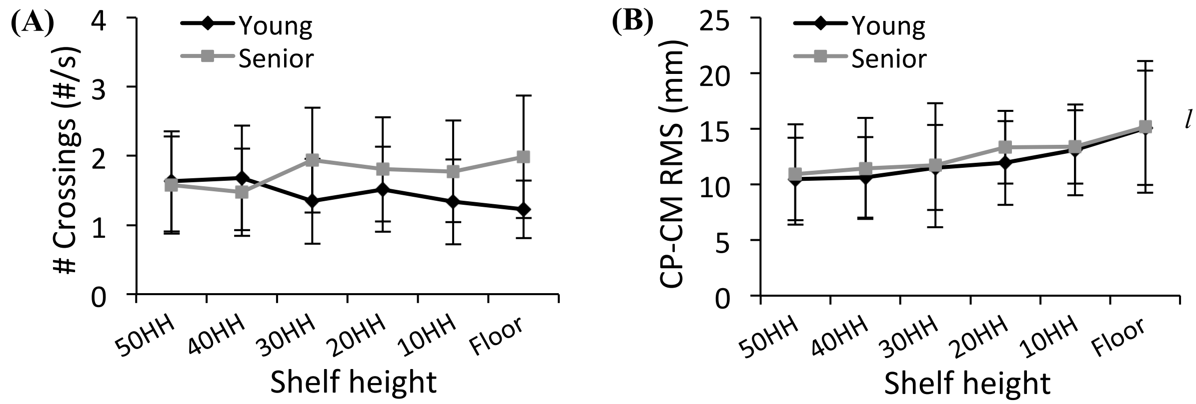


Figure 4.10 Relationship between COP and COM in older (grey lines) and younger (black lines) participants during varying lift height tasks: (A) number of anterior-posterior COP-COM crossings (*numCross*) normalized by time, (B) root-mean-square distance between COM and COM in the anterior-posterior direction (*COPtoCOM_rms*). Error bars represent +/- one standard deviation from the mean. Significant lift height (*l*) effects are indicated on the figure.

4.3.2.2 Precision Required

Independent COM and COP Behaviour

Age-effects were not significant for *MMOS_Ant* ($F_{1,17}$ = 0.05, p = 0.832; Fig. 4.11A) or *MMOS_Post* ($F_{1,17}$ = 1.35, p = 0.260; Fig. 4.11B) during the precision tasks. Nevertheless, mean differences demonstrated 14% greater *MMOS_Post* values in older (43.1 (SD = 9.0) % BOS) compared to younger (37.8 (SD = 13.3) % BOS) participants. While precision-effects were also not observed for *MMOS_Ant* ($F_{1,17}$ = 0.24, p = 0.632) or *MMOS_Post* ($F_{1,17}$ = 1.48, p = 0.240), *MMOS_Post* values decreased by about 8%, from 42.0 (SD = 9.1) % BOS when retrieving the dustpan to 38.7 (SD = 13.8) % BOS when precision demands increased during the chip retrieval task. No interactions between age and precision were observed for *MMOS_ Ant* ($F_{1,17}$ = 0.07, p = 0.800) or *MMOS_Post* ($F_{1,17}$ = 0.05, p = 0.831).

No age-effects were present for *COM_vel* ($F_{1,17}$ = 3.24, p = 0.090; Fig. 4.11C) or *COP_vel* ($F_{1,17}$ = 0.55, p = 0.470; Fig. 4.11D) either, although mean differences indicated that while older participants' mean *COM_vel* (40.0 (SD = 17.1) mm/s) was 27% slower than younger participants' (29.0 (SD = 9.0) mm/s), their mean *COP_vel* was actually 9% faster (older *COP_vel*: 97.7 (SD = 25.7) mm/s; young *COP_vel*: 89.4 (SD = 27.8) mm/s). Main effects of precision were not observed for *COM_vel* ($F_{1,17}$ = 1.57, p = 0.227) or *COP_vel* ($F_{1,17}$ = 1.74, p = 0.205). Interactions between age and precision were also not present for *COM_vel* ($F_{1,17}$ = 0.27, p = 0.610) or *COP_vel* ($F_{1,17}$ = 0.32, p = 0.578).

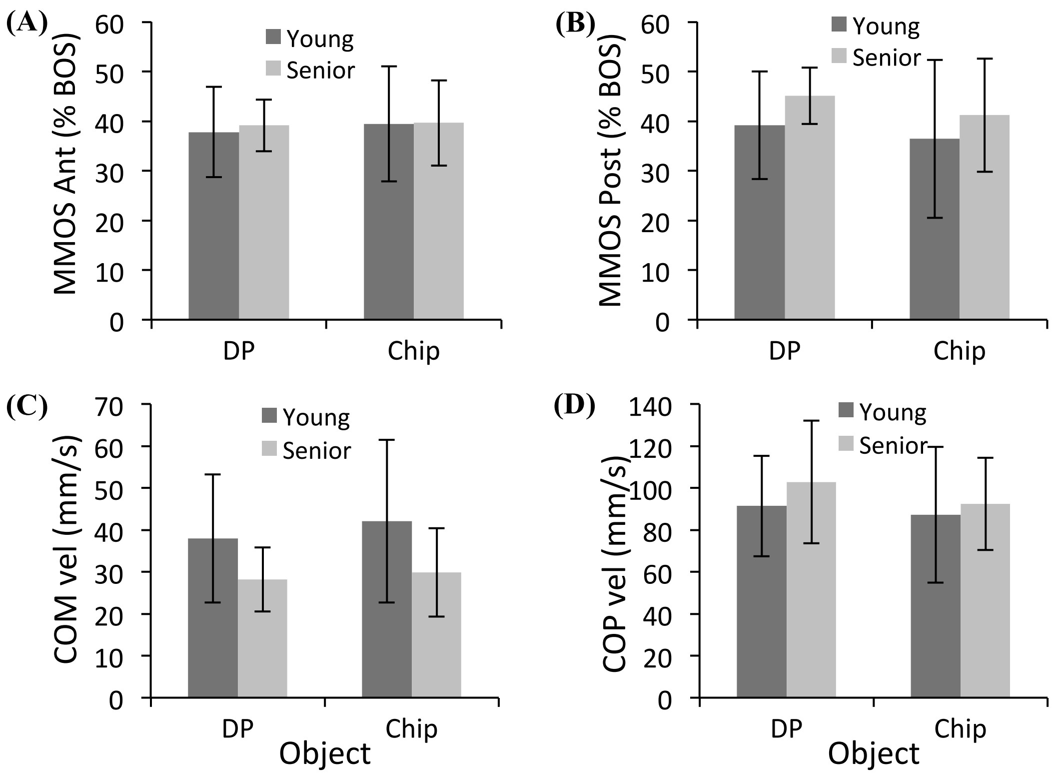


Figure 4.11 Balance measures describing COM and COP behavior of older (grey lines) and younger (black lines) participants during varying level of precision tasks: (A) COM minimum margin of safety (MMOS) to anterior BOS boundary, (B) COM minimum margin of safety (MMOS) to posterior BOS boundary, (C) mean COM anterior-posterior velocity, and (D) mean COP anterior-posterior velocity. Error bars represent +/- one standard deviation from the mean. No significant age, precision, or interaction effects were observed.

Interaction between COM and COP

Interestingly, a significant age-effect was observed for *numCross* ($F_{1,17}$ = 6.31, p = 0.022; Fig. 4.12A), by which older participants (1.89 (SD = 0.77) cross/sec) had approximately 47% more COP-COM crossings per second than their younger counterparts (1.29 (SD = 0.43) cross/sec). An age-effect was not present for *COPtoCOM_rms* ($F_{1,17}$ = 0.23, p = 0.640; Fig. 4.12B). Precision effects were significant for *COPtoCOM_rms* ($F_{1,17}$ = 4.54, p = 0.048), but not for *numCross* ($F_{1,17}$ = 0.02, p = 0.882). Specifically, *COPtoCOM_rms* decreased in both age groups by approximately 12% when precision demand increased from the dustpan (17.13 (SD = 5.45) mm) to the chip (15.12 (SD = 5.36) mm). Although interaction effects between age and precision were not observed for *numCross* ($F_{1,17}$ = 1.04, p = 0.322) or *COPtoCOM_rms* ($F_{1,17}$ = 1.11, p = 0.307), increased precision demands tended to cause greater age-related differences in *numCross* and smaller age-related differences in *COPtoCOM_rms*.

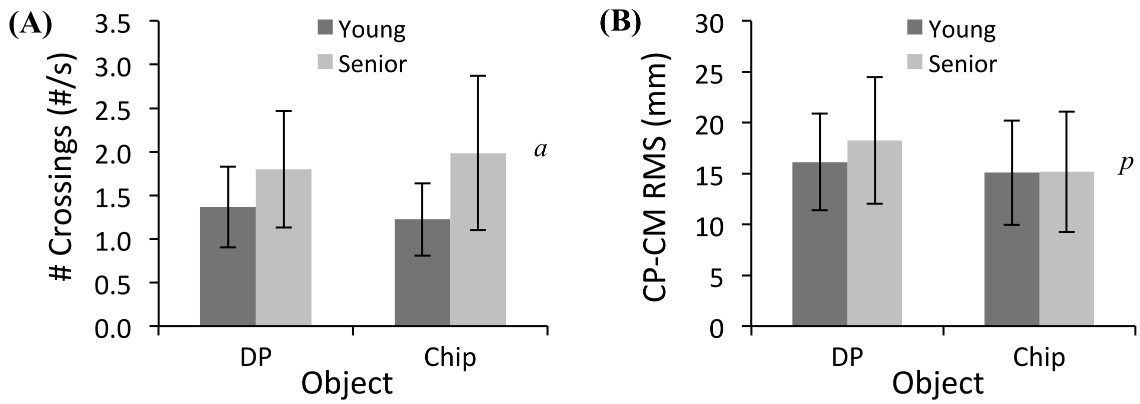


Figure 4.12 Relationship between COP and COM in older (grey lines) and younger (black lines) participants during varying level of precision tasks: (A) number of anterior-posterior COP-COG crossings normalized by time, (B) root-mean-square distance between COP and COM in the anterior-posterior direction. Error bars represent +/- one standard deviation from the mean. Significant age (*a*) and precision (*p*) effects are indicated on the figure.

4.3.2.3 Task Duration

Independent COM and COP Behaviour

While age did not affect *MMOS_Ant* ($F_{1,17}$ = 0.25, p = 0.624; Fig. 4.13A), it had a significant effect on *MMOS_Post* ($F_{1,17}$ = 5.09, p = 0.038; Fig. 4.13B). Older participants (37.7 (SD = 11.5) % BOS) exhibited *MMOS_Post* values that were 34% higher than their younger counterparts (24.8 (SD = 18.1) % BOS), indicating that they stayed closer to the centre of their BOS throughout the varying duration tasks. While task duration effects were not observed for *MMOS_Ant* ($F_{3,17}$ = 0.20, p =

0.896) or *MMOS_Post* ($F_{3,17} = 2.76$, $p = 0.052$), *MMOS_Post* values across both age groups tended to decrease by about 25% from the shortest (35.6 (SD = 14.3) % BOS) to the longest duration task (26.8 (SD = 16.7) % BOS). Interactions between age and task duration were not present for *MMOS_Ant* ($F_{3,17} = 0.89$, $p = 0.451$) or *MMOS_Post* ($F_{3,17} = 2.38$, $p = 0.081$), although age-related differences in *MMOS_Post* tended to increase as the tasks became longer.

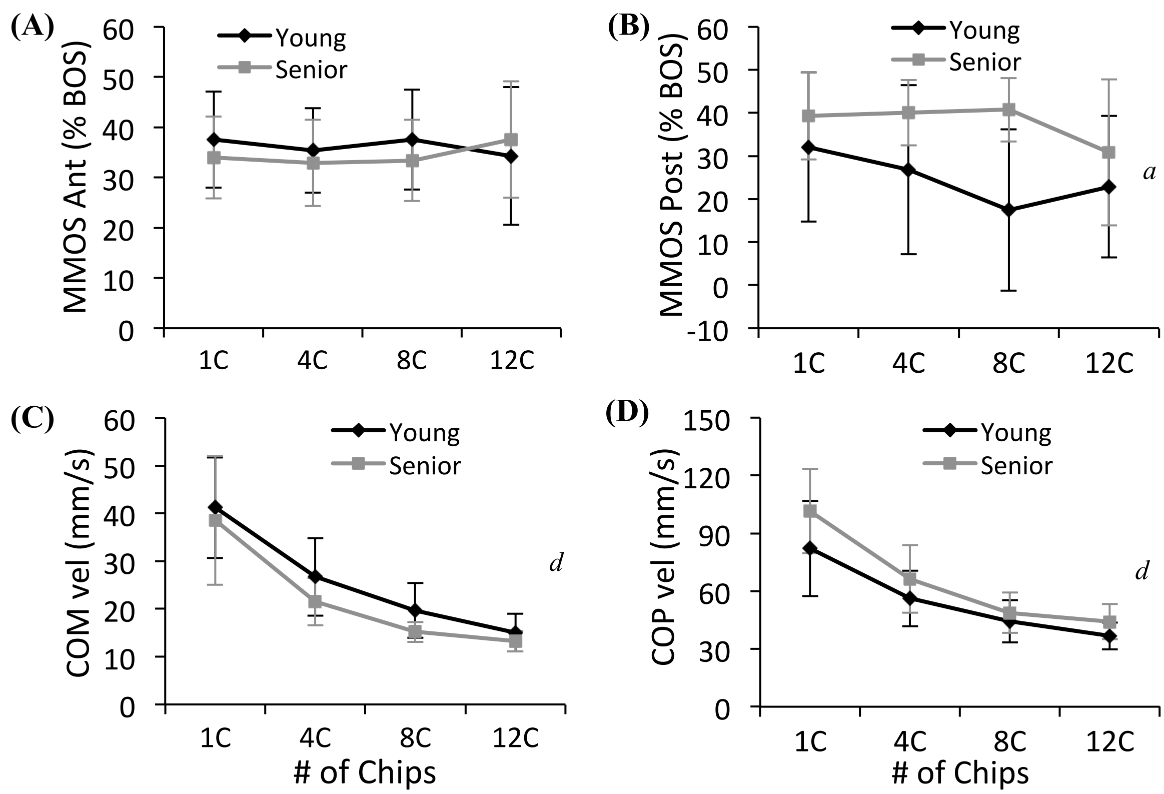


Figure 4.13 Balance measures describing COM and COP behavior of older (grey lines) and younger (black lines) participants during the varying duration tasks: (A) COM minimum margin of safety (MMOS) to anterior BOS boundary, (B) COM minimum margin of safety (MMOS) to posterior BOS boundary, (C) mean COM anterior-posterior velocity, and (D) mean COP anterior-posterior velocity. Error bars represent +/- one standard deviation from the mean. Significant age (*a*) and duration (*d*) effects are indicated on the figure.

No age-effects were present for *COM_vel* ($F_{1,17}$ = 1.94, p = 0.183; Fig. 4.13C) or *COP_vel* ($F_{1,17}$ = 3.31, p = 0.088; Fig. 4.13D), although age-related differences displayed opposite trends for each variable. Specifically, while older participants' (22.1 (SD = 12.3) mm/s) *COM_vel* was 14% slower than younger participants' (25.66 (SD = 12.3) mm/s), their *COP_vel* was actually 19% faster (older: 65.2 (SD = 27.3) mm/s; young: 54.9 (SD = 23.0) mm/s). Main effects of duration were observed for both *COM_vel* ($F_{3,17}$ = 67.49, p < 0.001) and *COP_vel* ($F_{3,17}$ = 70.33, p < 0.001), with mean velocities decreasing in both age groups as tasked task duration increased. Specifically, from the shortest (1 chip) to the longest (12 chips) duration task, *COM_vel* decreased by 65%, from 39.9 (SD = 11.8) mm/s to 13.1 (SD = 3.2) mm/s), and *COP_vel* slowed by 56%, from 91.8 (SD = 24.7) mm/s to 40.5 (SD = 8.7) mm/s. No interactions between age and task duration were observed for *COM_vel* ($F_{3,17}$ = 0.33, p = 0.666) or *COP_vel* ($F_{3,17}$ = 1.43, p = 0.256).

Interaction between COM and COP

No age-effects were present for *numCross* ($F_{1,17}$ = 3.02, p = 0.101; Fig. 4.14A) or *COPtoCOM_rms* ($F_{1,17}$ = 0.37, p = 0.551; Fig. 4.14B). Nevertheless, mean differences indicated that older participants (2.29 (SD = 0.78) cross/sec) had approximately 19% more COP-COM crossings per second than younger participants (1.93 (SD = 0.61) cross/sec). Task duration affected both *numCross* ($F_{3,17}$ = 5.45, p = 0.016) and *COPtoCOM_rms* ($F_{3,17}$ = 26.42, p < 0.001), causing opposite trends in each measure. Specifically, from the shortest (1 chip) to the longest (12 chips) duration task, *numCross* across all participants increased by 32%, from 1.79 (SD = 0.93) cross/sec to 2.37 (SD = 0.41) cross/sec, while *COPtoCOM_rms* decreased by 48%, from 14.1 (SD = 5.6) mm to 7.3 (SD = 1.2) mm. Interactions between age and task duration were not present for *numCross* ($F_{3,17}$ = 0.87, p = 0.407) or *COPtoCOM_rms* ($F_{3,17}$ = 1.63, p = 0.219).

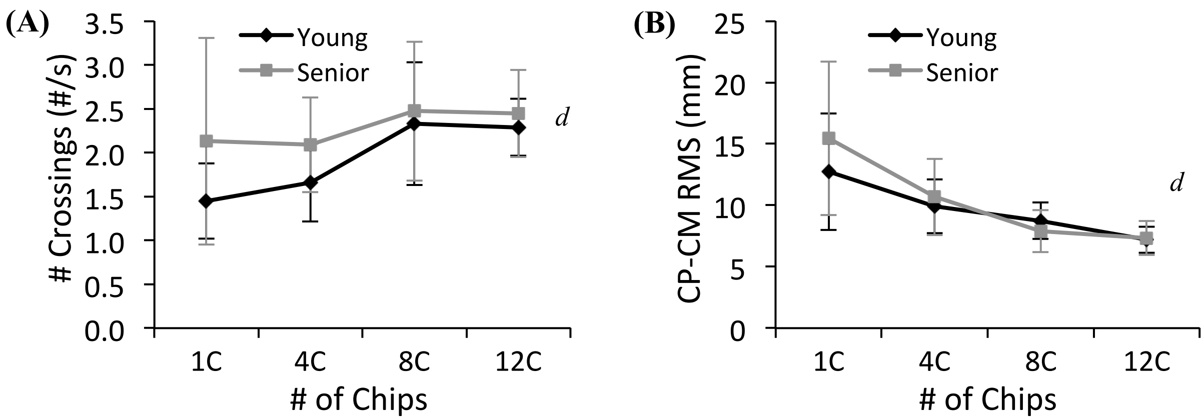


Figure 4.14 Relationship between COP and COM in older (grey lines) and younger (black lines) participants during varying duration tasks: (A) number of anterior-posterior COP-COM crossings normalized by time, (B) root-mean-square distance between COP and COM in the anterior-posterior direction. Error bars represent +/- one standard deviation from the mean. Significant duration (*d*) effects are indicated on the figure.

4.4 Discussion

The purpose of this thesis was to describe age-related differences in movement kinematics and postural control during stooping and crouching (SC) tasks. My prediction that, compared to younger, older adults would demonstrate more conservative movement strategies was in large part supported by the kinematic measures. Specifically, the first two hypotheses were supported by the findings that older participants (1) did not position their COM as low to the ground (due to less hip, knee, and ankle flexion), and (2) moved slower into and out of postures used to retrieve the object(s) in each SC task variation examined (lift height, precision required, and task duration). The third hypothesis, that younger participants would be more likely than older participants to use a smaller, forefoot BOS, crouching posture, especially during longer duration tasks, was also supported by the results.

The prediction that these reductions in movement characteristics (i.e., lower velocities and smaller amplitude movements) would be accompanied by greater postural control activity in older compared to younger adults was only partially supported by the balance control measures in this study. Specifically, the fourth hypothesis, that older adults would display smaller ranges of anterior-posterior (AP) COM motion within a more centralized area of the BOS – as evidenced by larger COM to BOS margins of safety in both the anterior (*MMOS_Ant*) and posterior (*MMOS_Post*) directions – was only partially supported. While *MMOS_Post* values were significantly larger in older compared to younger adults during the varying duration tasks, and showed similar, non-significant trends in the varying initial lift height and precision tasks, *MMOS_Ant* values were not different between the age groups in any of the task variations. The fifth hypothesis, that older adults would display lower COM velocities (*COM_vel*), but higher COP velocities (*COP_vel*), was also only partially supported. While *COM_vel* was significantly lower in older compared to younger adults during the varying initial lift height tasks, and showed a similar, non-significant trend in the varying precision tasks, *COP_vel* tended to be higher (not significantly) only in the varying duration tasks. The sixth hypothesis, that older adults would exhibit a higher frequency of COP to COM crossings (*numCross*), while also displaying an increased root mean square separation distance between these variables (*COPtoCOM_rms*), was again only partially supported by the results. While *numCross* was significantly higher in older compared to younger adults during the varying precision tasks, and showed similar (though not significant) trends in the varying lift height and duration tasks, *COPtoCOM_rms* was not different between the age groups in any of the SC tasks. Finally, the seventh hypothesis, that younger participants would exhibit greater lower limb isometric strength and range of motion, and higher balance confidence was mostly supported by the results, with younger participants displaying significantly higher knee extensor, flexor, and ankle plantar flexor strengths, significantly larger passive ranges of hip extension and knee flexion, and potentially higher (though not significant) balance confidence scores.

Finally, the secondary hypothesis that age-related differences would increase when tasks became more challenging (i.e., via lower initial lift heights, higher precision demands, and longer tasks) was supported only by the kinematic measures in this study. Specifically, interactions between age and task element on maximum velocities and joint flexion angles demonstrated that younger participants had a higher propensity to alter their movement characteristics in response to changing task demands.

This section is separated into the constituent SC task elements evaluated in this study (lift height, precision demand, and task duration), so that kinematic and balance control results can be discussed within the context of each experimental condition. The overall study results are then synthesized at the end of the discussion, with specific commentary regarding the contribution of this work to existing literature, a brief discussion of limitations and future study suggestions, and a concluding statement.

4.4.1 Initial Lift Height

Kinematics

This series of tasks was designed to become more challenging as initial lift height decreased from 50% of hip height toward floor level, requiring participants to progressively increase their downward reach distance in order to retrieve a plastic poker chip. Predictably, in response to decreasing initial lift height, participants in both age groups increased flexion in their hip, knee, and ankle (Fig. 4.2), thereby lowering their COM to a position from which they could complete the task. Interestingly, older adults performed all tasks with a significantly higher mean COM position than their younger counterparts (achieved through comparatively less flexion in the hip, knee, and ankle joints). Moreover, as initial lift height decreased, younger participants translated their COM disproportionately lower to the ground than their older counterparts (Fig. 4.2A). This may have reflected a reluctance on behalf of the older participants to lower their COM into a position from which they might have difficulty getting back up. Several research groups have demonstrated that older adults are unable to rise from lowered chair heights because the knee extensor moments needed to stand from such positions are beyond what they are capable of generating (Hughes et al., 1996; Rodosky et al., 1989). Differences in isometric knee extensor strength and mean knee flexion angles used during the tasks in the present study may provide evidence of this reluctance. Across all of the varying lift height trials, younger participants exhibited mean knee flexion angles of 58° – nearly 84% greater than older participants (mean knee flexion angle: 32°). This comparatively straighter-legged posture used by older participants meant they relied less on their weaker quadriceps (by approximately 30%; Table 4.4) to lower (and raise) themselves by bending their knees, and more on bending their trunk forward and downward from the waist. Potential reductions in balance confidence, as indicated by differences (though not significant) in ABC-6 scores, combined with smaller passive knee range of motion (ROM) values in older adults (Table 4.4) may have also contributed to the avoidance of deeper, flexed-knee postures, although there are conflicting opinions regarding the relationship between passive ROM and mobility ranges used during functional activities (Beissner et al., 2000; Moreside and McGill, 2013; Shrier, 2004).

In addition to not moving as far down as their younger counterparts, older adults were considerably slower when transitioning into and out of each posture, as evidenced by lower maximum vertical velocities of the COM and lower maximum angular velocities of the hip, knee, and ankle joints (Fig. 4.3, Table 4.5). Moreover, while decreasing initial lift height was associated with increased maximum velocities in both age groups, the effect was stronger in younger participants. The disproportionate increase in maximum velocities in younger compared to older participants was likely linked to the progressively lower postures used by younger participants, and their superior lower limb strength and perhaps muscle coordination. The progressively lower postures meant younger adults were translating their COM farther downward than their older counterparts, allowing them to achieve higher velocities over the greater distance travelled. A similar effect was observed within subjects, with decreasing lift height leading to increased downward displacements and subsequently higher maximum velocities. Several groups have reported a similar phenomenon during volitional COP movements, with larger movement amplitudes leading to faster movements (Duarte and Freitas, 2005; Hernandez et al., 2012b). Quadriceps strength is also known to correlate with maximum upper body vertical and anterior-posterior linear momentum and stability in older adults during the sit-to-stand (Hughes et al., 1996; Scarborough et al., 1999). As stooping and crouching tasks often require flexed-knee movements that are similar to those employed in the sit-to-stand, it is likely that older participants' strength deficits (Table 4.4) contributed to their lower movement velocities in a similar manner. Thus, as a consequence of their weaker knee extensor (quadriceps) strength, and possible age or subclinical pathology-related degradation of sensory function and muscle control (Lord et al., 1994; Schupert and Horak, 1999; Tinetti et al., 1995), older adults moved less rapidly despite translating their COM through shorter vertical displacements.

Another trend that emerged in these tasks relates to BOS condition. While the difference was insignificant, 25% (3 out of 12) of younger, compared to 8.3% (1 out of 12) of older participants used a forefoot crouching posture during the floor level task. A change in BOS area for only this task and none of the other, higher initial lift height, conditions is not surprising as it was considered the most challenging – requiring the greatest downward COM displacement to retrieve the chip from the lowest vertical level. A deep knee flexion, forefoot crouch posture allows participants to achieve a lower final COM position than a stoop, and leaves the head in a relatively upright position, closer to the ground (Burgess-Limerick, 2001). This may aid in visual control of the task, and reduce the likelihood of diminished vestibular inputs from an inverted and/or extended head and neck position, which could lead to postural instability (Johnson and Van Emmerik, 2012). That older participants were less likely to decrease their BOS, and adopt this posture, also supports the idea that their movements were more conservative (i.e., slower and smaller). Transitioning into a forefoot crouching posture requires controlling the downward and forward displacement of the COM while changing from a flatfoot (larger) to a forefoot (smaller) BOS, with the knees fully flexed and heels above the support surface in the final posture (Fig. 4.15). This demands significant postural control, muscle coordination, and leg strength – especially in the knee extensors (which were 44% stronger in young compared to older adults) – in order to lower oneself onto (and raise oneself from) a reduced, forefoot BOS. Further, maintaining 'static' stability once in the posture leaves less room for error, as the COM is closer to the BOS boundary by virtue of being contained within a much smaller foot support area. Lower linear vertical COM and joint angular displacements and velocities, combined with an increased likelihood of using a flatfoot BOS, provide kinematic evidence of a more conservative movement strategy used by older adults during the varying initial lift height tasks.

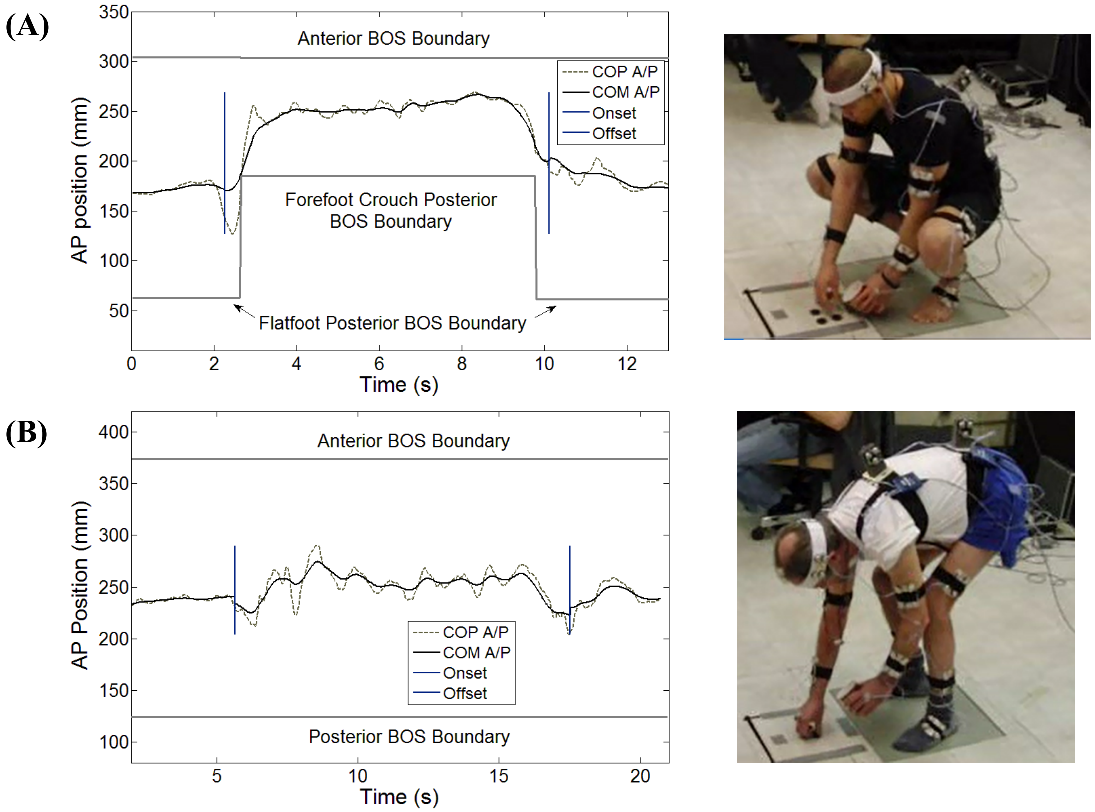


Figure 4.15 COP and COM AP positions and images of representative (A) forefoot crouch (younger) and (B) flatfoot stoop (older) postures used during the 8 chip bimanual task. Note the reduced BOS length when the heels are above the ground in the forefoot crouch posture.

Balance Control

The balance control measures evaluated in the present study were meant to collectively describe whether older adults exhibited heightened COM control activity compared to younger participants. Although age-related differences were statistically significant only for COM AP mean velocity, graphical trends in several key variables suggest that compared to younger, older adults utilized greater COP activity to control the COM position within a more centralized area of the BOS.

It is well known that in order to satisfy static equilibrium, the horizontal position of the COM must remain within the geometric limits of the BOS (Maki and McIlroy, 1996; Winter, 1995). If the COM moves horizontally away from the centre of the support surface, perhaps as a result of an external disturbance (e.g., push from behind) or volitional movement (e.g., forward reach), postural stability may be compromised unless the COM is decelerated before it reaches the BOS limits (Horak, 2006). There are two solutions to ensure stability is maintained in this scenario. The first involves generating a large enough vertical ground reaction force in front of the moving COM – thereby producing sufficient torque to stop the forward horizontal momentum of the COM and reverse it toward the centre of the BOS. The torque in this scenario is typically generated by the ankle plantar flexors while the BOS remains fixed. The second solution involves taking a step or reaching to a support, effectively increasing the available BOS area within which the COM may be regulated (Maki and McIlroy, 1996). Since all tasks in the present study were performed with a fixed BOS, the horizontal position of the COM was regulated exclusively by the COP, which represents the location of the ground reaction force generated by the net muscle activity of the legs (Winter, 1995). Thus, examining the independent behaviour of the COP and COM, in addition to their interaction with one another and the BOS, provided important insights into the postural control strategies used during the tasks.

While neither *MMOS_Ant* nor *MMOS_Post* values were significantly different between the age groups, graphical trends indicated that older adults did not allow their COM to approach their posterior BOS boundary to the same extent that younger adults did. These notably larger *MMOS_Post* values (Fig. 4.9B) in older participants may have reflected a conscious effort to maintain the COM AP position within a tighter band in the centre of the BOS. Older adults also exhibited lower COM AP mean velocities throughout the tasks. As mentioned above, if the COM is allowed to move away from the BOS centre, especially at a high velocity, greater muscle force (perhaps beyond that which older adults are capable of producing) is required to generate torques sufficient to counter the momentum of the COM and return it to a stable position. It is possible that older adults were safeguarding against potential strength, sensory, and coordination deficits by avoiding such scenarios of high COM displacement and velocity in which postural stability might be vulnerable.

While older adults displayed lower COM activity (i.e., movement amplitude and velocity) than their younger counterparts, their COP activity actually tended to be higher. First, although mean COP AP velocity was not significantly different between the age groups, it appeared slightly faster in older compared to younger participants, especially when considering the opposite trend observed in COM velocities (Fig. 4.9D). This observation suggests that, per unit of COM velocity, older participants displayed considerably higher COP velocities compared to younger participants. Second, while *numCross* and *COPtoCOM_rms* differences, which describe the interaction between the COP and COM, were also not significant, graphical trends in *numCross* suggested the COP crossed the COM more frequently in older compared to younger adults (Fig. 4.10A). This is consistent with previous findings showing that although older compared to younger adults moved 27% slower during targeted COP movements, they utilized nearly twice as many submovements to maintain accuracy (Hernandez et al., 2012a). Collectively, these variables indicate that despite slower COM AP velocities and narrower ranges of COM excursion, older adults tended to exhibit greater COP activity, reflected by comparatively faster COP velocities (relative to their own COM velocity), and a higher frequency of COP to COM crossings.

4.4.2 Precision

Because age-related differences during the precision tasks were similar to those described for the varying lift height tasks, this section will focus primarily on main effects of increasing precision demand and their interactions with age. The relative level of precision demand was modified by changing the object that participants had to retrieve from the floor. The poker chip (same trial as the 'floor level' condition in the varying lift height tasks) was considered to require more precision than the dustpan, which had a large, easy-to-grasp, handle positioned slightly above the floor.

Kinematics

In response to increased precision demands, both age groups adopted postures that were characterized by 12% lower COM heights, achieved primarily through increased flexion in the hip, knee, and ankle joints (Fig. 4.4). Interestingly, postural adaptations were more pronounced in younger participants. Berthier et al (1996) have demonstrated that reach to grasp tasks become more difficult as objects are reduced in size and placed farther from the individual. It is possible that because retrieving the poker chip required more precision than the dustpan, a lower posture with the head and eyes closer to the ground was preferred to optimize visual control. While younger participants were able to achieve these lower postures, making their focal grasp task easier, older participants' physical limitations (e.g., strength, functional ROM, and coordination) likely precluded them from doing the same (Hughes et al., 1996).

Increasing precision demands also differentially affected maximum velocities, but, interestingly, only during transitions down into the postures (Fig. 4.5; Table 4.6). Specifically, while maximum downward COM velocity increased in younger participants, it actually appeared to decrease in older adults as precision demands increased from the dustpan to the chip (Fig. 4.5A). Similar interactions were observed in the hip, knee, and ankle joint angular velocities, with older adults showing no difference or slight decreases in the increased precision condition, while younger participants displayed increased velocities. These opposite trends are likely related to a number of underlying factors.

That older adults displayed no difference or even slight decreases in maximum velocities when precision demands increased may reflect a heightened concern over the posture they were moving toward. Maximum flexion angles and minimum COM height indicated that older participants were not getting as low as their younger counterparts, perhaps indicating they were approaching the lowest posture they were willing to use. In such extreme postures, physical strength and coordination requirements are amplified, making it more difficult to control stability (Bhattacharya et al., 2009; Hughes et al., 1996). Thus, despite actually translating their COM farther downward in the chip compared to the dustpan condition, older adults' potentially increased concern over their ability to maintain stability in this posture may have prompted them to slow down.

Another factor may have been the object itself. Smaller objects, such as the chip in the present study, require more precision to successfully grasp (Berthier et al., 1996). Because precision was important in this task, older adults likely relied on feedback control to guide their movements. Feedback control involves using available information (i.e., visual, somatosensory, and kinesthetic) to continuously update one's estimation of the object's location relative to the moving hand so that a precise and successful grasp can be made (Fitts, 1954; Hoff & Arbib, 1993; Seidler et al., 2004). Asymmetrical hand speed profiles, reduced velocities, and slower movement times are often observed as a result of the increased processing demands associated with feedback control (Berthier et al., 1996). Conversely, reaching for larger objects that require less precision, such as the dustpan handle, can typically be accomplished in a feedforward manner, with pre-planned hand-speed profiles and little to no online-control (Berthier et al., 1996; Marteniuk et al., 1987). It may be that the increased precision demand of the poker chip provoked a shift from feedforward to feedback control in older adults, which, compounded by potential deterioration of strength and coordination, led to slower velocities during the downward movement (Morgan et al., 1994). Contrarily, younger participants may have been less affected by the increased precision demands of the chip retrieval task, and thus increased their downward velocity accordingly.

That these precision by age interactions were observed only during transitions down to, and not back up from, the postures further supports the hypothesis that older adults were more concerned about stability in lower postures and the precision requirements of the chip retrieval task. This is in agreement with the work of Mourey et al. (1998), which demonstrated age-related differences in the kinematics of sitting down, but not standing up. Older adults made distinct velocity adjustments toward the end of the sitting movement when accuracy became important to safely place the buttocks onto the seat. These postural control differences were manifested by disproportionately slower movement times and lower maximum knee and trunk angular velocities during sitting down versus standing up in older compared to younger adults (Mourey et al., 1998). The same velocity adjustments were not observed for standing back up; a movement to a more familiar standing posture with much lower precision demands. Although velocity adjustments within movements were not examined in the present study, significant interactions between age and precision on maximum velocities during downward movements only, likely reflect the age-related differences in movement control noted by Mourey et al. (1998).

Differences in BOS condition were again not significant, although the increased precision demand of the poker chip task appeared to encourage forefoot crouching more than the easier dustpan task, with 4 out of 24 (3 young, 1 older) participants using a forefoot crouch during the chip task compared to only one (young) in the dustpan task. This also highlights that although a lower posture, which can be achieved via forefoot crouching, is likely preferred for optimal visual control of the higher precision task, older participants were less likely us use one due to potential limitations in neuromuscular function.

Balance Control

While no significant age effects were observed for *MMOS_Ant* or *MMOS_Post,* similar trends to those observed during the varying initial lift height tasks were apparent, with older adults tending to keep their COM within a tighter band near the centre of their BOS (Fig. 4.11A & B). The greatest differences were again observed in the *MMOS_Post* values, with older adults' average margin of safety being 14% larger than that of younger participants, which may suggest that older adults were safeguarding against posterior instability. Interestingly, several groups have demonstrated this phenomenon in which older adults, especially those with functional limitations, are likely to guard against posterior instability by ensuring their COM is anterior compared to younger adults (Hernandez et al., 2012a; Schultz et al., 1992). Both age groups' *MMOS_Post* values also tended to decrease when retrieving the chip compared to the dustpan (Fig. 4.11B). This decrease may be directly linked to the increased number of individuals using a forefoot BOS crouching posture when retrieving the chip. During the downward transition into this posture, the COM comes very close to the posterior BOS boundary as it accelerates toward a more anterior position while the BOS shifts from a flatfoot to a forefoot configuration (Fig. 4.15A). As more individuals exhibited this postural strategy during the chip compared to the dustpan condition, their considerably smaller *MMOS_Post* values likely decreased group means.

The relationship between the COM and COP was similar to that observed during the varying initial lift height tasks. While neither variable showed significant differences between the age groups, older adults' *COM_vel* was 27% slower, while their *COP_vel* was 9% faster, than young participants (Fig. 4.11C & D). Further, *numCross* was 47% higher in older compared to young adults. These results are consistent with the varying lift height findings, which indicated that COP activity was greater (relative to COM activity) in older compared to young adults. Significant precision effects were observed only for *COPtoCOM_rms*, with values decreasing in the chip compared to the dustpan condition (Fig. 4.12B). The difference between the COP and COM has been referred to as the 'error' between the controlling (COP) and controlled (COM) variables, which is responsible for the horizontal acceleration of the COM (Winter, 1995). Further, it has been validated as an accurate and reliable measure of postural stability in older adults, with larger values representing greater instability (Corriveau et al., 2001, 2004). It is possible that the increased precision demand of the poker chip prompted participants to tighten up control of their COM, reflected by lower

COPtoCOM_rms values. This may have represented a conscious effort to minimize the error signal, thereby stabilizing the individual while attending to the focal chip retrieval task.

4.4.3 Task Duration

Kinematics

Age-related differences in movement kinematics during the varying duration tasks were similar to those reported for the previous two elements (lift height and precision) of SC tasks examined in this study. However, several interesting task duration effects were observed. As the perceived time of task completion increased[1], participants in both age groups adopted postures that were closer to the floor, with gradually lower COM heights achieved through increased flexion at the hip, knee and ankle (Fig. 4.6). Moreover, the longer duration bimanual tasks seemed to encourage the use of a forefoot BOS crouching posture to achieve this lower COM position, especially in young participants (Fig. 4.8). The number of forefoot crouching participants in the young cohort increased from 3 in the 1-chip task, to 5 in the 4-chip task, and 8 in both the 8- and 12-chip tasks. Conversely, only one older participant crouched during the 1-, 4-, and 8-chip tasks, while a total of only 3 crouched in the 12-chip task. Two logical questions arise. First, why do longer tasks encourage lower, crouching postures? And second, why don't older adults use these postures? The second question has been largely addressed already (see sections 4.4.1 and 4.4.2), as it relates to decreased strength capacity, coordination, and balance confidence in older adults. The first question as to why these postures are preferred for longer duration tasks has several potential answers.

One possible explanation involves minimizing muscle activity. Although safely transitioning down into and up from a crouching posture actually requires significantly higher knee extensor moments, and therefore more muscle activity, than would be required for a stoop (Burgess-Limerick, 2001; Giat and Pike, 1992), once the body is fully lowered into the crouch posture, muscle demands are

[1] Perceived time of task increased as the participant was presented with more chips, and was likely used to preselect the posture used to perform the task.

relatively low (Dionisio et al., 2008; Gallagher et al., 2011). Commonly referred to as the 'full squat', this posture involves maximum flexion of the knee joint with the buttocks effectively resting on the calf region, which functions as a seat (Sriwarno et al., 2008). This passive posture allows the knee extensor muscles to relax, with the head and torso in a relatively neutral, upright position (Fig. 4.15A). Conversely, maintaining a stoop or semi-squat posture (with knees slightly bent) requires activation of the knee extensors to keep the legs still and support the upper body (Dionisio et al., 2008). Further, significant extensor moments are required about the joints of the vertebral column to overcome the flexor moment caused by the weight of the forward leaning head, neck and torso (Burgess-Limerick, 2001). Injury to vertebral and ligamentous structures occur as a direct consequence of the high forces involved in creating these extensor moments (Burgess-Limerick, 2001). It is likely that participants elected to lower themselves into the passive crouching posture for longer duration tasks in order to minimize the physical effort required to maintain the posture, thereby reducing the chance of muscle fatigue and potential injury.

Another potential explanation may involve optimal head and trunk position and orientation. Within the head, the eyes provide visual feedback about the location and movement of the body in space. The vestibular system, in the inner ear, provides feedback about the head's acceleration and its orientation relative to gravity. Postural muscles in the neck provide proprioceptive feedback that translates lengthening and shortening of the muscles into sensory information regarding muscle tone. The integration of feedback from each of these sensory organs facilitates regulation of postural control (Johnson and Van Emmerik, 2012). Interfering with these feedback components by changing the head and trunk orientation can lead to changes in postural control that may impact stability. Several researchers have demonstrated that in young, healthy individuals with eyes closed, flexing the head forward on the trunk or tilting it laterally does not increase postural sway as much as extending it does (Johnson and Van Emmerik, 2012; Paloski et al., 2006). Other groups reported that head flexion and extension are equally destabilizing in a population that is already posturally challenged, such as the elderly, (Buckley et al., 2005), or when there are added sensory challenges (Ledin et al., 2003). In any case, researchers agree that proprioceptive feedback from the neck muscles contributes to the regulation of posture, and it can be disturbed in non-neutral head postures (Buckley et al., 2005; Paloski et al., 2006; Johnson and Van Emmerik, 2011,2012). It is likely that the combination of minimizing muscle activity in the lower limbs and optimizing sensory feedback from the head and neck contributes to the selection of a crouch over a stoop posture for longer duration tasks, at least for younger participants.

Age effects on maximum COM vertical and maximum joint angular velocities were similar to those observed in previous conditions, with younger adults generally moving faster than older adults into and out of each posture (Fig. 4.7 and Table 4.7). Interestingly, as tasks became longer, participants increased their maximum knee and ankle joint angular velocities during transitions down (Fig. 4.7C & D), and decreased their maximum hip angular velocity when transitioning back up (Table 4.7). Further, a significant interaction between age and task duration was observed, by which maximum knee angular velocity increased disproportionately in younger compared to older participants during transitions down into postures (Table 4.5).

That longer duration tasks elicited increased knee and ankle joint angular velocities during the transition down phase was likely a function of the increased proportion of participants using a crouch posture for these tasks (Fig. 4.8). As mentioned above, the crouching posture involves resting in a position near the end range of motion of the knee and ankle joints, with the knees fully flexed, and heels raised above the ground (Fig. 4.15A). Bending the knees and ankles through such large ranges of motion involves much faster maximum angular velocities than would be expected for a stoop posture in which the legs are relatively straight throughout the movement. Since a significantly higher proportion of young compared to older participants used the crouch posture for the longer duration tasks (Fig. 4.9 & 4.15A), their group mean values of maximum knee angular velocity would increase disproportionately more than that of older adults. The same tendency of participants to favour a crouch posture for longer duration tasks explains why hip angular velocity tended to decrease (Table 4.7). While the knee and ankle joints exhibit fast angular velocities in reaching their end range of motion positions during a crouch, the trunk remains relatively upright. By comparison, a stoop posture involves substantial forward bending of the trunk – and therefore greater hip angular velocities – but little motion in the lower extremities. The crouch posture is hence characterized by lower hip angular velocities. As a greater proportion of individuals crouched during longer compared to shorter duration tasks, it is not surprising that hip angular velocity decreased.

Balance Control

Age-related trends in COM safety margins were similar to those observed during the varying lift height and precision tasks, although *MMOS_Post* in this case was actually significantly smaller in young compared to older participants (Fig. 4.13B). *MMOS_Post* values also decreased as task duration increased, but to a greater extent in young compared to older participants. This trend is likely directly linked to the increased number of younger participants transitioning into a forefoot BOS during longer tasks (Fig. 4.8). As noted previously, the COM comes very close to the posterior BOS boundary during this movement, as it accelerates forward while the BOS shifts from a flatfoot to a forefoot stance with the heels lifting off the floor (Fig. 4.15A). Since the proportion of young compared to older participants using this posture increased for longer duration tasks, disproportionate decreases in *MMOS_Post* values were observed. Additionally, though not significant, *COM_vel* appeared faster in young compared to older participants, with increased task duration causing decreases in both age groups. Task duration-related increases in *COM_vel* may be explained by the fact that as the number of chips increased, the duration of the 'static' phase increased, in which the

COM was relatively still while the chips were being retrieved (Fig. 4.15). The lower COM velocities of this static phase effectively decreased the mean values calculated over the entire task.

Similar to previous task conditions, even though older participants tended to have slower overall COM velocities, their COP tended to move faster (Fig. 4.13D). This increased COP velocity was likely required to achieve the higher number of COP to COM crossings (*numCross*), in order to regulate the COM position. Interestingly, while *COP*_vel slowed down significantly as task duration increased, *numCross* increased. This is likely because the frequency of COP to COM crossings actually increases in static compared to dynamic scenarios. When the goal is to move the COM from one location to another (such is the case when initiating a downward reach movement), the number of COP to COM crossings is typically pretty low. In this scenario, the COM is accelerated forward by a downward ground reaction force (acting through the COP) located posterior to the COM. The forward movement of the COM must then be stopped when the desired position is reached. This is achieved in a manner similar to the initiation of movement, with the COP now racing in front of the COM to an anterior position at which a downward ground reaction force will slow the COM to a new static position. In a well-coordinated movement, few COP to COM crossings should theoretically take place, as the goal *is* to move the COM freely (Winter, 1995). Conversely, in static postures the goal is to maintain the COM position within a confined area. This requires the COP to make small but frequent adjustments by oscillating on either side of the COM (Winter, 1995). Thus, as the number of chips increased in the current study and the static phase got longer, *numCross* increased (Fig. 4.14A). Accompanying this task duration-related increase in *numCross* was a decrease in the root mean square separation distance between the COP and COM (*COPtoCOM_rms*). This follows from the above explanation that static scenarios, which comprised a greater proportion of longer tasks, require more frequent, but smaller in amplitude, adjustments to regulate the COM position.

4.4.4 Relevance

The results of this study complement the limited existing body of literature related to stooping and crouching performance in older adults. To my knowledge, this is the first work comparing measures of both kinematics and balance control between young and older adults during natural performance of common SC tasks. In general, the age-related trends observed in the current study are in agreement with existing works. In particular, the finding that aging leads to more conservative movement strategies aimed at tightly regulating postural stability during SC tasks is consistent with findings in other volitional movements such as the sit-to-stand. Several researchers have noted that older adults are more likely to adopt the more conservative “exaggerated flexion” strategy when rising from a chair, which involves fully flexing the trunk forward to ensure the COM is located safely within the foot support area before rising to a standing position (Hughes and Schenkman, 1996; Mourey et al., 1998, 2000; Schultz et al., 1992). This is in contrast to the more efficient, but perhaps less controlled, “momentum-transfer” strategy, which involves flexing the upper body during lift-off and continuing through the initiation of knee extension, with a smooth transfer of horizontal to vertical linear momentum ending in simultaneous back and knee extension (Scarborough et al., 1999, 2007). The results of this study suggest that a similar, conservative strategy with associated increases in postural control activity, was employed by older participants when performing SC movements. The specific finding that older adults moved slower into and out of, and not as far down into, their SC postures further supports this idea, and is in agreement with the findings of Kuo and colleagues (Kuo et al., 2011), which demonstrated decreased lower limb velocities and less flexion in older compared to young adults when performing a constrained full-squat movement. Furthermore, the tendency of older adults to use a relatively straight-legged, flatfoot BOS stooping posture compared to a flexed-knee, forefoot BOS crouching posture, used primarily by younger participants (especially for tasks of longer duration and those that required higher precision), was consistent with several works examining lifting strategy and forward reach capabilities in older adults (Hernandez et al., 2013; Punielo et al., 2001). Finally, that weaker knee extensor strength was likely an important factor in limiting the functional range of postures that older adults used in this study was also consistent with numerous reports evaluating similar large range of motion movements (Hernandez et al., 2008, 2010, 2013; Hughes et al., 1996; Punielo et al., 2001; Schultz et al., 1992).

Although they moved slower through shorter ranges of motion than their younger counterparts, older participants appeared less efficient in controlling postural stability, as evidenced by a tendency to exhibit greater sagittal plane COP activity (i.e., faster COP velocities relative to COM velocity, and higher frequency of COP to COM crossings), despite smaller COM excursions and slower COM velocities. This finding is in agreement with numerous works reporting increased relative muscle activity in older compared to younger adults while squatting, and a disproportionately higher number of COP submovements during targeted volitional movements (Kuo et al., 2011; Hernandez et al., 2012a, 2012b).

Collectively, the results of this study provide evidence of a more conservative movement strategy – perhaps related to the 'stabilization strategy' often observed in other movement domains such as the sit-to-stand (Hughes and Schenkman, 1996; Mourey et al., 1998, 2000; Schultz et al., 1992) – employed by older adults during stooping and crouching movements. This work also provides the first data describing age-related differences in movement strategies and postural control during *natural* SC task performance. Existing works have examined only constrained stooping and crouching scenarios in which participants were given explicit instructions and/or movement targets that likely affected natural behaviour (Hernandez et al., 2013; Kuo et al., 2011). Although causal relationships between physiological traits (i.e., strength, range of motion, and coordination) and measures of SC performance (i.e., speed of movement, postures used, or COP activity) were not established in the present study, the results provide an important initial summary of the differences between healthy young and older adults. Such data may be useful in guiding therapeutic interventions towards optimal postural control strategies in older adults with SC difficulty.

4.4.5 Limitations

This study involved several notable limitations. First, the relatively small sample size, combined with instructions to perform tasks naturally, led to high within-group variability and ultimately, relatively low statistical power (see Appendix D for a summary of ANOVA results and effect sizes). Because of this, only graphical trends and/or mean differences could be used to make inferences regarding age- and task constraint-related effects for many of the dependent measures (balance control variables in particular). While a primary motivation for undertaking this study was to examine individuals' *natural* behaviour during stooping and crouching tasks, further work using larger sample sizes and more homogeneous study populations would decrease within-group variability and likely lead to increased power and stronger effects. Second, although participants were allowed to pick up the objects however they wanted to, they were not allowed to step off of the 46.4 cm x 50.8 cm force platform. This spatial constraint limited them from using a range of other posture such as kneeling, lunging, or wide base squatting. Nevertheless, most participants reported that they would likely use postures similar to those demonstrated in the experiment when faced with similar tasks in their daily lives. Third, kinematic and postural control analyses were restricted to the sagittal plane (i.e., anterior-posterior direction). It is important to recognize that kinematic asymmetries and/or postural instabilities often occur in the lateral direction as a consequence of anterior-posterior movements (Kuo et al., 2011; Maki & McIlroy, 1999b; Singer et al., 2012). Nevertheless, as movements required to complete SC tasks occur primarily in the sagittal plane, it is considered the most relevant plane for assessing age-related differences in movement strategies. Fourth, only mean values describing gross characteristics (i.e., maximum joint angles and mean velocities) of the entire movement were examined in the present study. While these were insightful in describing general age-related differences in postures used to perform SC tasks, a more detailed description of distinct movement phases and intersegmental coordination, similar to that of Schenkman and colleagues detailing sit-to-stand strategies (Schenkman et al., 1990), would provide additional detail regarding the entire movement process. Further, a deeper analysis of specific physical, neurological, and behavioural characteristics responsible for reduced stooping and crouching performance would help inform interventions aimed at improving functional mobility. While several of these measures were collected, only lower limb isometric strength, passive range of motion, and balance confidence scores were included in the current study, and causal relationships between these traits and measures of SC performance were not investigated. A much larger sample size is likely required to render such analyses meaningful (i.e., a multiple regression between physiological predictor variables and some measure of SC performance). Finally, the older adult study population consisted of healthy individuals with little to no observable or self-reported stooping and crouching difficulty. In order to truly evaluate kinematic and postural control mechanisms underlying SC difficulty and/or inability in older adults, a study population that better represents the 25% of older adults with SCK difficulty is required (Hernandez et al., 2008; Taylor et al., 1997). Nevertheless, investigating the manner in which healthy aging affects these tasks is an important first step, and provides insights into the progression of age-related declines in stooping and crouching performance.

4.4.6 Conclusion

This thesis sought to understand age-related differences in postural kinematics and stability control during natural performance of stooping and crouching tasks. In general, a more conservative movement strategy was observed in older adults, with members of this cohort moving slower into and out of postures that were characterized by lower COM heights and less flexion in the hip, knee, and ankle joints. These reductions in mobility may have reflected a heightened concern over postural stability, demonstrated through a tendency to increase COP activity (i.e., COP velocity and number of COP to COM crossings), despite smaller ranges of COM movement and lower COM velocities in the sagittal plane. Moreover, decreased lower limb strength and passive range of motion, combined with potential age-related degradation of sensory function and motor coordination, may have contributed the reductions in stooping and crouching mobility ranges observed in older participants. This work provides initial insights into age-related differences in stooping and crouching task performance, and highlights the need to identify specific determinants of SC difficulty so that therapeutic interventions aimed at maximizing functional mobility and independence in older adults can target relevant components of the neuromuscular system.

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APPENDIX A – HEALTH STATUS FORM

Date (DD/MM/YY): ______________ Participant code: _______ Age: ______ Gender (M/F): ____
Mass (kg): _______ Height (m): ______ Ambulatory function: ☐ community (independent) ☐ other________
Previous falls within the past year: ☐ 0 ☐ 1 ☐ 2 ☐ 3 ☐ >3

Circumstance of fall(s):

Fall	Date	How did the fall occur?	Did you suffer any injuries during the fall?
1.			
2.			

Have you suffered any other musculoskeletal injuries in the past 5 years (types/dates)?

☐ Fracture	Type/Date: ____________________	☐ Ligament sprain	Type/Date: ____________________
☐ Muscle strain	Type/Date: ____________________	☐ Joint dislocation	Type/Date: ____________________
☐ Ligament sprain	Type/Date: ____________________	☐ Head injury	Type/Date: ____________________
☐ Tendinitis	Type/Date: ____________________	☐ Other	Type/Date: ____________________

Have you had any surgeries in the past 5 years?

____

Do you ever experience any of the following symptoms/feelings? Check all that apply.

☐ Dizziness	☐ Confusion or disorientation	☐ Shortness of breath
☐ Vertigo	☐ Fear of falling	☐ Migraine headaches
☐ Nausea/light headedness	☐ Anxiety	☐ Depression
☐ Blurred vision	☐ Diarrhea	☐ Numbness (where: _________)
☐ Tinnitus (ringing in the ear)	☐ Rapid heart beat	☐ Pain (where: ______________)

Do you have or have you had any of the following medical conditions or illnesses? Check all that apply.

☐ Chest pain, angina	☐ Heart murmur	☐ Glaucoma	☐ Drug/alcohol dependency
☐ Heart attack	☐ Hypertension	☐ Cataracts	☐ Other ____________
☐ Stroke	☐ Diabetes (type: ______)	☐ Macular degeneration	
☐ Parkinson's disease	☐ Respiratory disease	☐ Ear infection	
☐ Epilepsy	☐ Arthritis (type: ______)	☐ Depression	
☐ Alzheimer's disease	☐ Osteoporosis	☐ Memory loss	

Are you currently taking any medications or herbal supplements?

☐ Sedatives (tranquilizers)	Name: _______________	Dose/freq: ________________	Length: ______________
☐ Antipsychotics	Name: _______________	Dose/freq: ________________	Length: ______________
☐ Antidepressant	Name: _______________	Dose/freq: ________________	Length: ______________
☐ Antihypertensives	Name: _______________	Dose/freq: ________________	Length: ______________
☐ Diuretics	Name: _______________	Dose/freq: ________________	Length: ______________
☐ Pain medications	Name: _______________	Dose/freq: ________________	Length: ______________
☐ Other	Name: _______________	Dose/freq: ________________	Length: ______________

Exercise:

Do you currently engage in any form of physical activity? ☐ Yes ☐ No
1. Activity: ____________________________________ ☐ <15 min / day ☐ 15-30 min / day ☐ > 30 min/day
2. Activity: ____________________________________ ☐ <15 min / day ☐ 15-30 min / day ☐ > 30 min/day

*No participants were excluded from the study as they were all pre-screened via interview (and re-screed using this form) to ensure they did not meet the following ***exclusion*** criteria: one or more accidental falls in the previous year; anatomical, neurological, visual, or cognitive impairment; history of balance or coordination problems; use of psychotropic medication(s); dependence on ambulatory aids; or reliance on prosthetic devices.

APPENDIX B – MARKER SET AND BODY SEGMENT PARAMETERS

Figure A1 Illustration of marker placement on anterior (left) and posterior (right) aspects of the body. Each rigid cluster (beige-yellow) consisted of 4 active markers (blue). Green markers illustrate marker locations digitized using the probe.

Table A1 Body segment parameters by gender. Segment masses are relative to total body mass and segment COM positions are referenced to the origin of the segment. See Table A2 for endpoint definitions.

Segment	Origin	Endpoint	Mass (%)		COM position (%)	
			F	M	F	M
Head	MIDG	VERT	6.68	6.94	41.06	40.24
Thorax	XP	SS	15.45	15.96	79.23	70.01
Pelvis/Abd	XP	MIDH	27.12	27.5	47.96	50.2
Thigh	HJC	KJC	14.78	14.16	36.12	40.95
Shank	KJC	AJC	4.81	4.33	44.16	44.59
Foot	Heel	Toe Tip	1.29	1.37	40.14	44.15
Upper Arm	SJC	EJC	2.55	2.71	57.54	57.72
Forearm	EJC	WJC	1.38	1.62	45.59	45.74
Hand	WJC	MET3	0.56	0.61	74.74	79.00

Table A2 Segment endpoint definitions (de Leva, 1996).

Endpoint	Definition
VERT	Vertex - the most superior point of the head
MIDG	Mid-gonion - the point midway between the gonions (the most lateral point of the posterior angle of the mandible)
MIDS	Mid-shoulder - the point midway between the shoulder joint centres
XP	Midpoint between the xiphoid process and T10 vertebra.
SS	Midpoint between the suprasternal notch and the T2/T3 vertebral space
MIDH	Mid hip - the point midway between the hip joint centres
HJC	Hip joint centre - The centre of the femoral head, estimated by projecting the greater trochanter marker medially by 25% of the distance between it and the GT marker on the other leg (Bell et al., 1990).
KJC	Knee joint centre - the midpoint between the lateral and medial femoral condyles
AJC	Ankle joint centre - the center of the transverse section of the talus, approximately at the level of the distal tip of the fibula, estimated by taking the midpoint between the lateral and medial malleoli of the tibia
Heel	The most posterior-most point of the calcaneus
Toe Tip	The tip of the 2nd metatarsal
*SJC	Shoulder joint centre – center of the humeral head (60 mm caudal to acromion (Nussbaum and Zhang, 2000)).
EJC	Elbow joint centre – the centre of the transverse section of the humerus, at the level of the greatest projection of the medial humeral epicondyle, estimated by taking the midpoint between the medial and lateral epicondyles.
WJC	Wrist joint centre – the centre of a transverse section of the capitate bone, at the level of the palpable groove between the lunate and capitate bone, estimated by taking the midpoint between the ulnar and radial styli.
MET3	Base of the 3rd metacarpal (knuckle)

APPENDIX C – THE 6-ITEM ACTIVITIES-SPECIFIC BALANCE CONFIDENCE (ABC-6) SCALE

Instruction to Participants:

For each of the following, please indicate your level of confidence in doing the activity without losing your balance or becoming unsteady by choosing one of the percentage points on the scale from 0% to 100%. If you do not currently do the activity, try and imagine how confident you would be if you had to do the activity.

For each of the following activities, please indicate your level of self-confidence by choosing a corresponding number from the following rating scale:

0%	10	20	30	40	50	60	70	80	90	100%
no confidence								**completely confident**		

"How confident are you that you will not lose your balance or become unsteady when you…"

1. … stand on your tiptoes and reach for something above your head? _____%
2. … stand on a chair and reach for something? _____%
3. … are bumped into by people as you walk through the mall? _____%
4. … step onto or off an escalator while you are holding onto a railing? _____%
5. … step onto or off an escalator while holding onto parcels such that you cannot hold onto the railing? _____%
6. … walk outside on icy sidewalks? ____%

APPENDIX D – ANOVA SUMMARY

KINEMATICS – Lift Height

Table A3 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for minimum COM height and maximum hip, knee, and ankle joint angles during the varying initial lift height tasks.

Max ht/angle	Age	η_p^2	Lift Height	η_p^2	Interaction	η_p^2
COM ht (m)	7.33 (0.013)[a]	0.250	256.48 (<0.001)[b]	0.921	3.35 (0.046)[ab]	0.132
Hip (°)	13.19 (0.001)[a]	0.375	649.96 (<0.001)[b]	0.967	0.208 (0.892)	0.009
Knee (°)	12.86 (0.002)[a]	0.369	50.37 (<0.001)[b]	0.696	2.87 (0.069)	0.115
Ankle (°)	9.14 (0.006)[a]	0.294	26.54 (<0.001)[b]	0.547	2.16 (0.111)	0.089

[a] significant age effect ($p < 0.05$)
[b] significant lift height effect ($p < 0.05$)
[ab] significant interaction (age*lift height) effect ($p < 0.05$)

Table A4 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for maximum downward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during transition-down (TD) phase of the varying initial lift height tasks.

TD Max Velocity	Age	η_p^2	Lift Height	η_p^2	Interaction	η_p^2
COM vert (m/s)	13.38 (0.001)[a]	0.378	90.89 (<0.001)[b]	0.805	7.76 (<0.001)[ab]	0.261
Hip vel (°/s)	9.79 (0.005)[a]	0.308	38.86 (<0.001)[b]	0.639	2.51 (0.034)[ab]	0.102
Knee vel (°/s)	15.84 (0.001)[a]	0.419	28.67 (<0.001)[b]	0.566	1.40 (0.255)	0.060
Ankle vel (°/s)	23.99 (<0.001)[a]	0.522	15.91 (<0.001)[b]	0.420	1.66 (0.187)	0.070

[a] significant age effect ($p < 0.05$)
[b] significant lift height effect ($p < 0.05$)
[ab] significant interaction (age*lift height) effect ($p < 0.05$)

Table A5 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for maximum upward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during transition-up (TU) phase of the varying initial lift height tasks.

TU Max Velocity	Age	η_p^2	Lift Height	η_p^2	Interaction	η_p^2
COM vert (m/s)	12.88 (0.002)[a]	0.369	117.1 (<0.001)[b]	0.842	4.17 (0.015)[ab]	0.159
Hip vel (°/s)	8.67 (0.008)[a]	0.283	58.56 (<0.001)[b]	0.727	0.61 (0.639)	0.027
Knee vel (°/s)	17.14 (<0.001)[a]	0.438	29.75 (<0.001)[b]	0.575	3.36 (0.024)[ab]	0.133
Ankle vel (°/s)	24.73 (<0.001)[a]	0.529	18.14 (<0.001)[b]	0.452	1.26 (0.295)	0.054

[a] significant age effect ($p < 0.05$)
[b] significant lift height effect ($p < 0.05$)
[ab] significant interaction (age*lift height) effect ($p < 0.05$)

KINEMATICS - Precision Required

Table A6 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for minimum COM height and maximum hip, knee, and ankle joint angles during the varying precision demand tasks.

Ht/Angle	Age	η_p^2	Precision	η_p^2	Interaction	η_p^2
COM ht (m)	6.89 (0.015)[a]	0.238	28.53 (<0.001)[b]	0.565	3.85 (0.063)[ab]	0.149
Hip (°)	5.01 (0.036)[a]	0.185	27.37 (<0.001)[b]	0.554	2.39 (0.137)	0.098
Knee (°)	9.10 (0.006)[a]	0.292	11.06 (0.003)[b]	0.335	5.52 (0.028)[ab]	0.201
Ankle (°)	7.74 (0.011)[a]	0.260	2.79 (0.109)	0.113	4.191 (0.053)[ab]	0.160

[a] significant age effect ($p < 0.05$)
[b] significant precision effect ($p < 0.05$)
[ab] significant interaction (age*precision) effect ($p < 0.05$)

Table A7 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for maximum downward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during transition-down (TD) phase of the varying precision demand tasks.

TD Max Velocity	Age	η_p^2	Precision	η_p^2	Interaction	η_p^2
COM vert (m/s)	13.25 (0.001)[a]	0.376	2.12 (0.160)	0.088	10.65 (0.004)[ab]	0.326
Hip vel (°/s)	6.98 (0.015)[a]	0.241	0.91 (0.349)	0.040	4.35 (0.049)[ab]	0.165
Knee vel (°/s)	10.89 (0.003)[a]	0.331	2.44 (0.133)	0.100	4.16 (0.054)[ab]	0.159
Ankle vel (°/s)	10.05 (0.004)[a]	0.313	13.33 (0.001)[b]	0.377	5.86 (0.024)[ab]	0.210

[a] significant age effect ($p < 0.05$)
[b] significant precision effect ($p < 0.05$)
[ab] significant interaction (age*precision) effect ($p < 0.05$)

Table A8 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for maximum upward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during transition-up (TU) phase of the varying precision demand tasks.

TU Max Velocity	Age	η_p^2	Precision	η_p^2	Interaction	η_p^2
COM vert (m/s)	15.09 (0.001)[a]	0.407	12.74 (0.002)[b]	0.367	2.57 (0.123)	0.105
Hip vel (°/s)	8.57 (0.008)[a]	0.280	0.09 (0.773)	0.004	1.02 (0.323)	0.044
Knee vel (°/s)	18.44 (<0.001)[a]	0.456	6.35 (0.020)[b]	0.224	3.99 (0.058)	0.154
Ankle vel (°/s)	12.45 (0.002)[a]	0.361	0.41 (0.527)	0.018	0.06 (0.808)	0.003

[a] significant age effect ($p < 0.05$)
[b] significant precision effect ($p < 0.05$)
[ab] significant interaction (age*precision) effect ($p < 0.05$)

KINEMATICS - Task Duration

Table A9 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for minimum COM height and maximum hip, knee, and ankle joint angles during the varying duration tasks.

Ht/angle	Age	η_p^2	Duration	η_p^2	Interaction	η_p^2
COM ht (m)	6.42 (0.019)[a]	0.234	11.20 (<0.001)[b]	0.348	1.09 (0.359)	0.049
Hip (°)	6.53 (0.018)[a]	0.229	24.68 (<0.001)[b]	0.529	0.88 (0.419)	0.038
Knee (°)	9.67 (0.005)[a]	0.305	8.48 (<0.001)[b]	0.278	1.89 (0.140)	0.079
Ankle (°)	6.42 (0.019)[a]	0.226	3.81 (0.030)[b]	0.147	0.22 (0.805)	0.010

[a] significant age effect ($p < 0.05$)
[b] significant duration effect ($p < 0.05$)
[ab] significant interaction (age*duration) effect ($p < 0.05$)

Table A10 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for maximum downward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during transition-down (TD) phase of the varying duration tasks.

TD Max Velocity	Age	η_p^2	Duration	η_p^2	Interaction	η_p^2
COM vert (m/s)	8.50 (0.008)[a]	0.288	0.87 (0.462)	0.040	1.01 (0.393)	0.046
Hip vel (°/s)	7.02 (0.015)[a]	0.242	2.2 (0.119)	0.091	1.07 (0.354)	0.046
Knee vel (°/s)	11.55 (0.003)[a]	0.344	2.88 (0.043)[b]	0.116	3.06 (0.034)[ab]	0.122
Ankle vel (°/s)	12.09 (0.002)[a]	0.355	3.28 (0.046)[b]	0.130	2.06 (0.139)	0.086

[a] significant age effect ($p < 0.05$)
[b] significant duration effect ($p < 0.05$)
[ab] significant interaction (age*duration) effect ($p < 0.05$)

Table A11 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for maximum upward vertical velocity of COM, and maximum hip, knee, and ankle joint angular velocities during transition-up (TU) phase of the varying duration tasks.

TU Max Velocity	Age	η_p^2	Duration	η_p^2	Interaction	η_p^2
COM vert (m/s)	10.56 (0.004)[a]	0.335	0.25 (0.865)	0.012	0.72 (0.547)	0.033
Hip vel (°/s)	7.13 (0.014)[a]	0.245	2.78 (<0.05)[b]	0.112	0.11 (0.953)	0.005
Knee vel (°/s)	15.18 (0.001)[a]	0.408	2.05 (0.116)	0.085	0.40 (0.752)	0.018
Ankle vel (°/s)	14.73 (0.001)[a]	0.401	0.99 (0.387)	0.043	96 (0.398)	0.042

[a] significant age effect ($p < 0.05$)
[b] significant duration effect ($p < 0.05$)
[ab] significant interaction (age*duration) effect ($p < 0.05$)

BALANCE CONTROL – Lift Height

Table A12 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for the balance control measures during the varying lift height tasks.

Bal Var	Age	η_p^2	Lift Height	η_p^2	Interaction	η_p^2
MMOS_Ant	0.00 (0.995)	0.000	9.42 (<0.001)[b]	0.356	0.18 (0.919)	0.011
MMOS_Post	3.67 (0.072)	0.178	0.54 (0.559)	0.031	0.138 (0.840)	0.008
COM_vel	6.66 (0.019)[a]	0.281	9.94 (<0.001)[b]	0.369	1.44 (0.218)	0.078
COP_vel	0.314 (0.583)	0.018	12.93 (<0.001)[b]	0.432	0.25 (0.940)	0.014
numCross	2.25 (0.152)	0.117	0.09 (0.994)	0.005	1.99 (0.089)	0.105
COPtoCOM_rms	0.10 (0.752)	0.006	7.71 (<0.001)[b]	0.312	0.18 (0.970)	0.010

[a] significant age effect ($p < 0.05$)
[b] significant lift height effect ($p < 0.05$)
[ab] significant interaction (age*lift height) effect ($p < 0.05$)

BALANCE CONTROL – Precision

Table A13 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for the balance control measures during the varying precision demand tasks.

Bal Var	Age	η_p^2	Precision	η_p^2	Interaction	η_p^2
MMOS_Ant	0.05 (0.832)	0.003	0.24 (0.632)	0.014	0.07 (0.800)	0.004
MMOS_Post	1.35 (0.261)	0.074	1.48 (0.240)	0.080	0.05 (0.831)	0.003
COM_vel	3.24 (0.090)	0.160	1.57 (0.227)	0.085	0.27 (0.610)	0.016
COP_vel	0.55 (0.470)	0.031	1.74 (0.205)	0.093	0.32 (0.578)	0.019
numCross	6.31 (0.022)[a]	0.271	0.02 (0.882)	0.001	1.04 (0.322)	0.058
COPtoCOM_rms	0.23 (0.640)	0.013	4.54 (0.048)[b]	0.211	1.11 (0.307)	0.061

[a] significant age effect ($p < 0.05$)
[b] significant precision effect ($p < 0.05$)
[ab] significant interaction (age*precision) effect ($p < 0.05$)

BALANCE CONTROL – Task Duration

Table A14 *F*-ratios (*p*-value) and partial eta squared (η_p^2) measures of effect size, from the repeated measures ANOVA tests for the balance control measures during the varying duration tasks.

Bal Var	Age	η_p^2	Duration	η_p^2	Interaction	η_p^2
MMOS_Ant	0.25 (0.624)	0.015	0.20 (0.896)	0.012	0.89 (0.451)	0.053
MMOS_Post	5.09 (0.038)[a]	0.241	2.76 (0.052)	0.147	2.38 (0.081)	0.130
COM_vel	1.94 (0.183)	0.108	67.48 (<0.001)[b]	0.808	0.33 (0.666)	0.020
COP_vel	3.31 (0.088)	0.171	70.33 (<0.001)[b]	0.815	1.43 (0.256)	0.082
numCross	3.02 (0.101)	0.159	5.45 (0.016)[b]	0.254	0.87 (0.407)	0.052
COPtoCOM_rms	0.37 (0.551)	0.023	26.42 (<0.001)[b]	0.623	1.63 (0.219)	0.093

[a] significant age effect ($p < 0.05$)
[b] significant duration effect ($p < 0.05$)
[ab] significant interaction (age*duration) effect ($p < 0.05$)

APPENDIX E – MEAN (SD) VALUES FOR ALL KINEMATIC AND BALANCE CONTROL VARIABLES

Table A15 Mean (SD) joint flexion angles and minimum vertical centre of mass position during the varying initial lift height and precision tasks

Position	Cohort	50HH	40HH	30HH	20HH	10HH	Floor level	Dustpan
COM Ht (m)	Young	0.85 (0.06)	0.80 (0.07)	0.76 (0.06)	0.70 (0.07)	0.66 (0.07)	0.58 (0.10)	0.63 (0.08)
	Senior	0.89 (0.05)	0.86 (0.05)	0.82 (0.06)	0.78 (0.05)	0.73 (0.06)	0.68 (0.08)	0.70 (0.05)
Hip (°)	Young	85.9 (9.4)	102.8 (8.7)	115.6 (8.2)	131.3 (9.3)	142.3 (9.4)	151.4 (9.1)	144.9 (11.1)
	Senior	75.4 (9.5)	91.8 (8.8)	104.5 (9.7)	119.4 (7.0)	131.5 (6.0)	142.3 (5.0)	138.7 (8.5)
Knee (°)	Young	26.4 (15.1)	39.7 (19.5)	48.4 (19.2)	62.9 (18.6)	73.3 (20.7)	97.1 (35.2)	76.9 (25.5)
	Senior	14.2 (16.3)	18.1 (15.0)	23.1 (19.6)	33.7 (19.7)	45.0 (25.6)	55.0 (32.3)	51.5 (19.6)
Ankle (°)	Young	3.3 (5.1)	7.0 (7.6)	7.8 (7.6)	12.8 (7.4)	15.6 (7.2)	18.3 (6.5)	15.6 (6.0)
	Senior	1.6 (4.3)	0.9 (3.1)	2.1 (5.6)	4.9 (6.8)	7.6 (8.0)	9.3 (7.3)	9.62 (7.4)

Table A16 Mean (SD) joint flexion angles and minimum vertical centre of mass position during the varying duration tasks

Position	Cohort	1 Chip	4 Chips	8 Chips	12 Chips
COM Ht (m)	Young	0.57 (0.10)	0.52 (0.11)	0.53 (0.11)	0.51 (0.10)
	Senior	0.66 (0.07)	0.64 (0.08)	0.64 (0.09)	0.61 (0.11)
Hip (°)	Young	155.7 (10.5)	162.5 (11.0)	162.6 (11.5)	162.3 (14.7)
	Senior	145.6 (6.6)	151.1 (5.9)	153.8 (6.0)	153.9 (8.4)
Knee (°)	Young	94.7 (36.8)	111.0 (39.5)	109.6 (40.5)	122.9 (36.8)
	Senior	61.2 (29.7)	63.3 (34.7)	62.1 (35.8)	72.4 (42.5)
Ankle (°)	Young	17.9 (6.7)	18.5 (6.5)	18.6 (5.8)	20.2 (6.9)
	Senior	11.1 (6.8)	11.4 (6.8)	11.9 (8.0)	12.6 (8.2)

Table A17 Mean (SD) of the maximum joint angular and vertical centre of mass velocities while transitioning down into postures used for the varying initial lift height and precision tasks

TD Vel	Cohort	50HH	40HH	30HH	20HH	10HH	Floor level	Dustpan
COM vel (m/s)	Young	0.19 (0.06)	0.28 (0.11)	0.35 (0.09)	0.45 (0.11)	0.49 (0.14)	0.58 (0.13)	0.52 (0.15)
	Senior	0.17 (0.07)	0.21 (0.05)	0.25 (0.08)	0.29 (0.06)	0.36 (0.07)	0.37 (0.09)	0.40 (0.07)
Hip vel (°/s)	Young	141.3 (30.8)	160.1 (33.6)	187.3 (52.4)	200.8 (40.5)	207.5 (43.3)	216.9 (49.2)	211.1 (50.9)
	Senior	119.0 (21.6)	132.4 (24.2)	143.5 (37.5)	150.4 (23.4)	165.1 (24.9)	165.4 (25.9)	181.0 (26.5)
Knee vel (°/s)	Young	51.9 (23.6)	70.6 (33.0)	82.9 (33.4)	96.6 (35.0)	106.6 (37.0)	129.0 (46.4)	109.8 (46.4)
	Senior	25.6 (24.7)	31.9 (22.3)	40.6 (27.7)	54.5 (27.7)	65.3 (22.1)	71.3 (38.3)	73.8 (16.2)
Ankle vel (°/s)	Young	15.3 (13.0)	21.7 (13.0)	22.0 (12.7)	30.8 (9.1)	36.6 (12.8)	44.5 (17.5)	30.1 (11.1)
	Senior	8.3 (12.7)	6.9 (6.1)	8.8 (9.1)	12.0 (9.7)	16.6 (11.2)	22.8 (16.4)	19.9 (6.6)

Table A18 Mean (SD) of the maximum joint angular and vertical centre of mass velocities while transitioning down into postures used for the varying duration tasks

TD Vel	Cohort	1 Chip	4 Chips	8 Chips	12 Chips
COM vel (m/s)	Young	0.56 (0.18)	0.56 (0.17)	0.55 (0.14)	0.57 (0.14)
	Senior	0.45 (0.09)	0.39 (0.07)	0.42 (0.08)	0.41 (0.12)
Hip vel (°/s)	Young	218.2 (53.9)	212.5 (45.6)	202.9 (36.2)	206.7 (47.3)
	Senior	181.9 (31.8)	164.9 (24.2)	174.7 (26.5)	170.2 (27.6)
Knee vel (°/s)	Young	118.9 (54.9)	139.4 (49.8)	130.1 (49.9)	143.3 (59.7)
	Senior	78.3 (29.3)	74.4 (26.5)	70.5 (29.0)	78.0 (38.1)
Ankle vel (°/s)	Young	39.8 (18.9)	44.3 (14.6)	41.6 (18.7)	57.0 (35.7)
	Senior	25.6 (13.2)	19.0 (10.4)	21.0 (12.5)	25.3 (20.3)

Table A18 Mean (SD) of the maximum joint angular and vertical centre of mass velocities while transitioning back up to standing from the postures used for the varying initial lift height and precision tasks

TU Vel	Cohort	50HH	40HH	30HH	20HH	10HH	Floor level	Dustpan
COM vel (m/s)	Young	0.22 (0.06)	0.28 (0.11)	0.37 (0.08)	0.47 (0.10)	0.54 (0.12)	0.63 (0.13)	0.55 (0.12)
	Senior	0.18 (0.06)	0.23 (0.06)	0.28 (0.07)	0.35 (0.06)	0.41 (0.08)	0.46 (0.07)	0.43 (0.07)
Hip vel (°/s)	Young	-132.1 (19.7)	-157.3 (34.6)	-166.1 (30.1)	-190.1 (35.2)	-196.4 (38.2)	-203.6 (34.2)	-200.1 (38.2)
	Senior	-108.4 (23.7)	-127.8 (27.2)	-138.1 (28.8)	-159.7 (22.2)	-166.5 (28.9)	-164.1 (19.0)	-170.5 (30.4)
Knee vel (°/s)	Young	-51.5 (19.5)	-71.8 (39.0)	-74.4 (33.2)	-95.4 (34.1)	-113.2 (44.3)	-145.0 (54.5)	-118.3 (38.6)
	Senior	-28.8 (24.9)	-34.9 (23.6)	-37.0 (23.4)	-51.9 (22.8)	-64.5 (30.5)	-71.4 (31.2)	-68.3 (19.8)
Ankle vel (°/s)	Young	-12.9 (12.7)	-21.4 (14.1)	-21.1 (10.5)	-30.0 (11.8)	-36.4 (15.7)	-33.6 (12.6)	-32.8 (11.3)
	Senior	-4.8 (7.4)	-5.1 (6.9)	-6.3 (6.4)	-11.1 (6.4)	-17.8 (11.3)	-21.4 (10.7)	-19.5 (5.5)

Table A19 Mean (SD) of the maximum joint angular and vertical centre of mass velocities while transitioning back up to standing from the postures used for the varying duration tasks

TU Vel	Cohort	1 Chip	4 Chips	8 Chips	12 Chips
COM vel (m/s)	Young	0.57 (0.15)	0.56 (0.18)	0.57 (0.13)	0.60 (0.14)
	Senior	0.42 (0.12)	0.43 (0.12)	0.41 (0.08)	0.42 (0.12)
Hip vel (°/s)	Young	-192.5 (46.9)	-190.4 (40.3)	-180.6 (34.6)	-180.3 (31.7)
	Senior	-165.3 (19.7)	-164.3 (28.9)	-154.1 (16.7)	-147.9 (22.9)
Knee vel (°/s)	Young	-138.6 (57.0)	-136.2 (56.5)	-132.9 (49.5)	-149.6 (59.6)
	Senior	-70.2 (28.9)	-69.9 (37.9)	-63.5 (34.5)	-72.5 (41.1)
Ankle vel (°/s)	Young	-42.0 (19.4)	-40.4 (16.0)	-46.4 (27.2)	-46.3 (22.2)
	Senior	-21.6 (9.3)	-19.3 (9.9)	-16.4 (7.8)	-24.1 (20.5)

Table A20 Means (SD) for the balance control variables during the varying initial lift height and precision tasks

Bal Var	Cohort	50HH	40HH	30HH	20HH	10HH	Floor level	Dustpan
MMOS Ant (mm)	Young	51.1 (6.6)	46.5 (6.4)	48.0 (8.6)	44.9 (11.0)	40.0 (11.9)	39.4 (11.6)	37.8 (9.1)
	Senior	48.9 (5.0)	47.0 (6.6)	48.1 (5.0)	45.1 (4.4)	41.1 (4.8)	39.6 (8.6)	39.1 (5.2)
MMOS Post (mm)	Young	38.4 (4.5)	39.7 (5.4)	37.5 (5.2)	37.9 (6.3)	38.2 (7.6)	36.4 (15.9)	39.2 (10.9)
	Senior	40.8 (9.7)	44.1 (5.4)	41.5 (5.3)	43.6 (4.2)	43.4 (4.5)	41.2 (11.4)	45.1 (5.7)
COM AP vel (mm/s)	Young	19.8 (8.9)	26.2 (6.2)	24.8 (8.0)	30.5 (11.3)	38.3 (10.1)	42.0 (19.4)	37.9 (15.3)
	Senior	19.8 (16.3)	15.9 (4.4)	20.4 (8.4)	21.6 (5.6)	25.8 (7.2)	29.9 (10.6)	28.2 (7.7)
COP AP vel (mm/s)	Young	53.9 (14.8)	58.4 (19.1)	62.5 (23.2)	70.6 (22.5)	79.8 (20.0)	87.3 (32.4)	91.4 (24.0)
	Senior	63.8 (27.3)	60.4 (22.9)	67.2 (26.0)	77.8 (28.1)	80.1 (15.7)	92.4 (22.1)	102.9 (29.3)
numCross AP (#/s)	Young	1.63 (0.72)	1.68 (0.76)	1.34 (0.61)	1.52 (0.62)	1.33 (0.61)	1.22 (0.42)	1.36 (0.46)
	Senior	1.58 (0.70)	1.48 (0.63)	1.94 (0.76)	1.81 (0.75)	1.78 (0.74)	1.99 (0.88)	1.80 (0.67)
COP-COM RMS (mm)	Young	10.5 (3.7)	10.6 (3.6)	11.5 (3.8)	11.9 (3.8)	13.1 (4.1)	15.1 (5.1)	16.1 (4.8)
	Senior	10.9 (4.5)	11.4 (4.6)	11.7 (5.6)	13.4 (3.3)	13.4 (3.3)	15.2 (5.9)	18.3 (6.2)

Table A21 Mean (SD) for the balance control variables during the varying duration tasks

Bal Var	Cohort	1 Chip	4 Chips	8 Chips	12 Chips
MMOS Ant (mm)	Young	37.6 (9.6)	35.4 (8.4)	37.6 (10.0)	34.28 (13.7)
	Senior	34.0 (8.1)	32.9 (8.6)	33.4 (8.1)	37.6 (11.6)
MMOS Post (mm)	Young	32.1 (17.3)	26.8 (19.6)	17.4 (18.7)	22.9 (16.5)
	Senior	39.3 (10.1)	40.0 (7.6)	40.8 (7.4)	30.8 (16.9)
COM AP vel (mm/s)	Young	41.2 (10.5)	26.7 (8.1)	19.7 (5.8)	15.0 (3.9)
	Senior	38.5 (13.5)	21.5 (5.0)	15.2 (2.1)	13.3 (2.1)
COP AP vel (mm/s)	Young	82.1 (24.7)	56.2 (14.4)	44.4 (11.0)	36.8 (7.0)
	Senior	101.6 (21.9)	66.3 (17.5)	48.7 (10.5)	44.1 (9.1)
numCross AP (#/s)	Young	1.45 (0.43)	1.66 (0.44)	2.33 (0.70)	2.29 (0.32)
	Senior	2.13 (1.18)	2.09 (0.54)	2.47 (0.79)	2.45 (0.50)
COPtoCOM RMS (mm)	Young	12.74 (4.75)	9.92 (2.21)	8.73 (1.49)	7.20 (1.06)
	Senior	15.46 (6.25)	10.68 (3.11)	7.91 (1.72)	7.34 (1.40)