THE ROLE OF COGNITION IN LOCOMOTOR BALANCE CONTROL ACROSS THE LIFESPAN

By

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This journey is dedicated to my mom, who spent a lifetime telling me I was smart enough to do anything, to Lyle for being my biggest cheerleader, to all the mentors that believed in me more than I ever did, and to all of my mentees who made coming to work a joy. Finally, this is dedicated to everyone that doubted me, including myself.
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4-8 Step width variability DTC represented as mean and standard deviation for three cognitive tasks. Significant differences ($p < 0.05$) are marked with an asterisk (*). ............................................................... 111
Walking involves widespread neurological activation, especially for older adults (OA), but the role of cognitive processes in walking balance is unclear. The purpose of this dissertation was to determine age-differences in the effect of mediolateral stability and upright posture against gravity on cognitive-motor interference during walking.

Young adult (YA) and OA dual-task performance costs (DTC) were quantified during more, or less stable, walking with three cognitive tasks: Digit Span Back (DSB), Alphabet Recitation (Recitation), and Verbal Fluency (Fluency). Experiment 1 (StepRx) evaluated walking with a preferred and narrow step width. Experiment 2 (BWS) manipulated walking with (BW+S) and without balance support (BW) with a walking harness. Gait (StepRx: Step length (SL) DTC, Step length variability (SLVar) DTC; BWS: Step width (SW) DTC, Step width variability (SWVar) DTC) were analyzed in 2 x 2 x 3 RM MANOVA (Age: Old, Young; Walk: StepRx: Narrow and Preferred/BWS: BW and BW+S; Cognition; DSB, Recitation, Fluency), significance was $p < 0.05$. In StepRx, 22 YA (22.0 (2.0) years, 15 women) and 20 OA (71.0 (8.0) years, 12 women) participated. Added cognitive tasks destabilized balance and decreased SL DTC ($p = 0.015$) more in narrow walking, and SLVar DTC ($p = 0.029$) revealed greater
interference in hard cognitive tasks (DSB, Fluency). For BWS, 20 YA (22.0 (4.0) years, 16 women) and 20 OA (72.0 (7.0) years, 10 women) revealed YA reduced SW DTC from BW to BW+S ($p = 0.003$), while Recitation led to wider steps ($p = 0.026$), and SWVar increased ($p = 0.002$) in challenging tasks (DSB, Fluency). For cognition, covarying executive interference in RM ANCOVA and MANCOVA showed YA reduced performance in DSB, decreased Recitation rate while increasing accuracy, and both groups increased Fluency accuracy in narrow walking ($p = 0.009$). In BWS, participants maintained cognition across all tasks (all $p > 0.05$). Working memory (WM) affected step variability and may be involved in step control, while rhythmic and simple tasks could facilitate walking. Age-differences were limited; participants changed cognition in unstable walking and maintained cognition when balance was supported.
CHAPTER 1
INTRODUCTION

Advances in modern medicine have increased lifespans, but the quality of longer life for older adults is often dampened by an inability to perform activities of daily living and reduced independence [1]. Emerging aging research has shown cognitive decline is related to degradations in mobility and costly injurious falls [2], motivating scientists to investigate the associations between cognition, mobility, and balance across the lifespan. The integral role of cognition in walking has been recently recognized, but mixed methods and results prevent clear understanding of the factors that contribute to mobility impairment in older adults. Scientists across fields have pursued the interplay between cognition and mobility in older adults with various approaches, often using a dual-task paradigm.

A dual task is defined as performing two concurrent goal-oriented activities simultaneously, such as walking while speaking with a friend. Dual-task experiments quantify differences in cognitive/motor performance when two tasks are performed concurrently. The outcome of interest is termed ‘cognitive-motor interference’, which quantifies resultant performance costs when two tasks are performed together, considered indicative of shared resources between tasks [3,4]. The magnitude of cognitive-motor interference in dual tasks has demonstrated salience in predicting fall risk, mobility-disability, frailty, and mortality [2,5–7]. A growing body of evidence supports that tasks that increase motor complexity [i.e. standing to walking to crossing obstacles] exaggerate the effects of cognitive-motor interference [8–10] indicating a greater role of cognition in complex motor tasks. Unfortunately, the specific contribution of cognitive processes to more fundamental aspects of mobility, such as dynamic
balance, remains inconclusive. Moreover, the rate of age-related changes to gait and balance control across the lifespan is unknown.

Maintaining balance to prevent a trip or fall is a primary goal of locomotion and mobility. Successful mobility depends on the ability to maintain upright posture by preventing collapse against gravity [11]. Specifically, coordination of internal joint moments is needed to produce force to prevent collapse from external gravitational and ground reaction forces and maintain forward velocity [11–13]. The biomechanics and motor control of balance during walking is a result of the dynamics between the position of center of mass (COM), the balancing point at which all external forces acting on the body equal zero, and the base of support (BOS), the area in contact with the support surface [11,14–16]. Stepping width and length (the BOS) are adjusted to maintain the position of the COM within the BOS. The length and the width of steps is determined by integrating neurological feedback of the COM position in relation to the support surface in each step. Therefore, dynamic balance is often considered ‘actively controlled’ [15,17].

Controlling the COM and the BOS during walking requires the integration of visual, vestibular, and somatosensory information, and the subsequent execution of motor commands. As motor task difficulty increases, or when disease/age-related deterioration occurs, compensatory processes contribute to walking and dynamic balance. For example, recent research has shown that young adults experience declines in cognitive performance [18] and recruit additional higher-level cognitive networks during single support (i.e. when only one foot is on the ground) and gait is the most unstable [19]. Older adults demonstrate considerably higher and more widespread
prefrontal activity than young adults during typical and complex (crossing obstacles) walking tasks [20]. Despite older adults recruiting additional resources, age-related behavioral differences persist in complex walking task performance.

A growing body of behavioral evidence has demonstrated a strong association between cognition and walking ability in older age [6,7,21–23]. The age-related changes to gait suggest that older adults compensate for difficulty controlling the COM and BOS during walking, but the underlying mechanisms at fault are unknown. Existing evidence suggests that when balance demands increase (from sitting to standing, or standing to walking), or when added cognitive and motor coordination is required (e.g. added dual task), older adults experience greater degradations to gait performance than young adults. For instance, older adults experience a greater impact of cognitive load on standing balance (greater movement of the COM within the BOS) than young adults [10,24,25] and these deficits are exaggerated as the cognitive task becomes more challenging [26]. Thus, performance declines in older adults appear susceptible to basic balance demands and may require increasingly higher-level cognitive resources with age.

During walking, older adults adopt compensatory strategies to control the COM such as a forward trunk lean, or taking shorter and wider steps [12,27,28]. When older adults walk with a dual task, they attempt to stabilize the COM within the BOS with even wider and shorter steps, exhibiting variable stepping patterns [29–31]. The negative effect of a dual task on walking performances is greater in unstable walking situations [32–35]. Moreover, evidence of more cognitive-motor interference during tasks that involve balance alludes to a greater role of higher-order cognitive resources to maintain
balance for older adults, especially in those at risk for falls or mobility impairment. Older adults who are cognitively impaired or at fall-risk exhibit more cognitive-motor interference than their age-matched peers, and such declines are greater in complex motor tasks [8,9,22,36]. Decreasing performance and corresponding increases in balance demands during complex motor tasks suggest that dynamic balance control may be an underlying precipitant of cognitive-motor interference in aging [6,7,37,38]. Therefore, the purpose of this study is to evaluate the age-related changes to dynamic balance in walking [11,12,27,39,40] and their association with higher-level cognitive processes.

The role of controlling the BOS and the COM during walking in cognitive-motor interference has recently drawn attention in aging research, but few studies examine the unique contributions of dynamic balance in walking [10,24,25]. One recent study manipulated mediolateral balance control (step width) during dual-task walking and the findings suggested that step width adjustments were not cognitively demanding. Contrary to expectations, Mazaheri et al., reported that stabilizing the COM and manipulating the mediolateral BOS with narrow and normal step widths did not impact cognitive-motor interference in young or older adults [41].

While performance on the cognitive or motor task did not worsen in unstable walking conditions, the null results may be related to several methodological considerations. First, the researchers utilized a discrete reaction time (RT) task where participants responded to a proprioceptive stimulus (vibration on the calf) during narrow verses normal treadmill walking with lateral stabilization [41]. However, in a rehabilitative setting, access to lateral stabilization is limited and is generally non-
exist in daily life. To promote interpretation, replicability, and accessibility, increasing
stability with wider steps would provide a more ecologically valid paradigm for clinicians
and patients. Additionally, rather than allowing more ‘natural’ behavior, participants were
instructed to prioritize walking, which has been shown to eliminate changes to walking
during dual tasks [42,43]. Therefore, this approach may not best represent cognitive-
motor interference in daily walking.

The ecological validity of the walking task was further hindered by unusual
proprioceptive stimuli that is not common to typical complex walking tasks. A discrete
RT task is simple to administer, but only momentarily divides attention, whereas
dynamic balance is challenged continuously in every step. Dividing attention
intermittently means the cognitive load was not constant, thus higher-level cognitive
processes were not challenged in conjunction with dynamic balance control across the
gait cycle. Further, the investigators limited the time participants spent in unstable gait
phases (single support) under a cognitive load by presenting stimuli at heel strike, the
initiation of the most stable phase of a step (double support) [12]. Imposing intermittent
attentional demands in double support may neglect cognitive contributions to
mediolateral balance control in single support [18,19]. Cumulatively, the approaches the
authors used to examine cognitive-motor interference in mediolateral balance may have
confounded the results.

Another recent study reported benefits to dual-task performance with body
weight support provided via an overhead walking harness, which is a common
rehabilitative tool in clinical settings. The authors report a preservation of cognitive
performance (lower performance costs) and a relative increase in brain activation of the
frontal lobe during dual-task walking with upright body weight support (partially alleviating body weight) [44]. Although the results indicate that balance support benefits dual-task ability (reduced cognitive costs, maintained walking speed, and increased brain activation), the study compared a small and ungeneralizable sample of multiple sclerosis patients (n=10) and older adults (n=12) [44]. Thus, it remains unknown if the reported benefits are meaningful for older adults or whether these findings extend to other populations.

Additionally, the duration of the walking tasks in this study was limited to a single thirty-second trial and prefrontal activation levels were significantly lower in the first 10 seconds versus the last 10 seconds. Longer duration of testing time across several trials may eliminate performance changes because of initial learning and task adaptation and therefore represent behavior more similar to real-world walking. Finally, the only motor outcome reported was decreased gait speed on a self-paced treadmill from single to dual tasks, and gait speed was maintained from dual-task walking to dual-task walking with body weight support. Thus, the interpretation of benefits to dual-task performance with body weight support may be confounded by reduced forward velocity from single to dual-task walking [11], rather than a direct result of the balance support provided. Comprehensively measuring biomechanical measures on a treadmill with fixed speed, rather than a self-paced treadmill, would reveal stability-related adjustments to gait (double support, step length, step width, and variability) when gait speed is held constant.

Although the aforementioned studies have examined the relationship between cognition and dynamic balance in aging, several important questions remain
unanswered. First, both studies only used one type of cognitive task with little difficulty. Rather, comparing the influence of several different cognitive tasks on locomotor balance could reveal associations between specific higher-order cognitive processes and balance while walking. Second, measures of physical function, balance, and cognitive status are inconsistently reported in the existing literature and would better frame the functional consequences of cognitive-motor interference in dynamic balance. Third, carefully manipulating balance demands with ecologically valid and accessible paradigms would significantly increase the reliability and interpretation of cognitive-motor performance outcomes. Using clinically accessible approaches to provide balance support, such as wide steps or body weight support with a harness, would improve the generalizability and impact of the results. Cumulatively, existing evidence prevents clarity in understanding the contributions of cognition to the control of dynamic balance.

Overall, convincing evidence implies that older adults require more cognitive resources to manage cognitive-motor interference with increasing balance challenge. However, typical aging research draws conclusions from comparisons between motorically disparate and arbitrary age groups, neglecting the progressive nature of the aging process. Consequently, the way that cognitive-motor interference develops with age is unknown. Understanding changes to mobility across the lifespan is important because the global aging population is increasing exponentially [1]. Thus, this proposed dissertation project will address this limitation by observing cognitive-motor performance in a wide range of aged individuals. Examining individuals from a wide age-range will characterize the chronology of complex walking performance declines. The findings
derived from the proposed work will provide important evidence to inform and improve the implementation of walking interventions.

This dissertation proposes to fill the existing gaps in the literature by examining changes to dynamic balance during walking and evaluate their relationship with higher-order cognitive resources across the lifespan. Understanding the influence of fundamental aspects of dynamic balance on cognitive resources (and vice versa) will evidence the development of age-related changes to cognitive-motor processes. To accomplish this overarching goal, I propose two specific aims:

**Specific Aim One**

To determine age-differences in the effect of controlling mediolateral stability on cognitive-motor interference during walking. In young and older adults, changes in cognitive (i.e. accuracy and response rate) and walking (i.e. step length, and step length variability of each) performance (cognitive-motor interference) during dual-task walking with more stability (i.e. preferred step) or less stability (i.e. narrow step) were quantified. Measuring performance during three concurrent cognitive tasks revealed the influence of separate higher-order cognitive processes on step-to-step walking control: the Digit Span Back (DSB); Alphabet Recitation (Recitation); and Verbal Fluency (Fluency). The resultant cognitive-motor interference (i.e. performance costs) reflect the shared resources between cognition and the neurological control of mediolateral walking stability.

The hypotheses predicted that older individuals would demonstrate more cognitive-motor interference in cognitive (i.e. more errors and slower rate of response) and gait performance (i.e. shorter steps and increased variability) than young adults in
less stable walking conditions (narrow), and this would be exaggerated in challenging tasks (DSB and Fluency). Participants would maintain gait and cognitive performance from single to dual-task walking during more stable walking (preferred). Finally, challenging cognitive tasks (DSB) would lead to more interference than simple (Recitation) and familiar (Fluency) cognitive demands, especially during narrow walking.

**Specific Aim Two**

To determine age-differences in effect of maintaining body weight and upright posture against gravity on cognitive-motor interference while walking. In young and older adults, changes in cognitive (i.e. accuracy and response rate) and walking (i.e. step width and step width variability) performance were measured during dual-task walking with a walking harness with full body weight (BW) and with partial body weight support (BW+S). Performing three concurrent cognitive tasks revealed the influence of separate higher-order cognitive processes on step-to-step control: the Digit Span Back (DSB); Alphabet Recitation (Recitation); and Verbal Fluency (Fluency). The resultant cognitive-motor interference (i.e. performance costs from single to dual task) reflect the shared resources between cognition and maintaining upright posture during walking.

The hypotheses predicted that older individuals would demonstrate greater cognitive (i.e. more errors and slower rate of responses) and motor interference (increased step width variability) than young adults without balance support (BW), especially in challenging concurrent tasks, but match young adults in BW+S. Further, individuals would reduce cognitive-motor interference (i.e. little or no change to performance) in BW+S, as compared to BW (i.e. greater performance costs). The changes to walking performance would reflect reduced balance demands (wider steps)
with balance support, and increased step width variability would indicate a negative impact of cognitive task on walking performance. Challenging cognitive demands would impose greater DTC (DSB) than familiar tasks (Fluency), and cognitive-motor facilitation would occur during easy concurrent tasks (Recitation), especially in BW+S.
CHAPTER 2
LITERATURE REVIEW

The Biomechanics and Motor Control of Gait

The ability to walk bipedally, a uniquely human skill, is deceptively simple and most individuals perceive their walking as ‘automatic’. In reality, the mechanics and motor control of human gait are complex and activate the entire neuromuscular system. Investigating both motor and mechanical aspects of gait control through neuromechanics, the study of neural and biomechanical aspects of human movement, provides a fascinating and relevant framework from which to understand development and mobility. This dissertation will use perspectives from neuromechanics to provide insight into cognitive contributions to dynamic locomotor balance control across the lifespan.

**Major motor functions of gait.** Human gait activates the entire nervous system and is thought to require both automatic and volitional motor processes. Five major motor functions of gait have been defined by Winter et al. [11]

- To prevent collapse of the lower limb in stance and support the upper body.
- To maintain upright posture and balance of the total body.
- To control of foot trajectory for safe ground clearance and heel or toe landing.
- To generate mechanical energy to initiate the present forward velocity.
- To absorb mechanical energy for shock absorption and stability or to decrease the forward velocity of the body.

**The Gait Cycle**

The gait cycle is broken into seven phases by seven critical events occurring within a stride. The stride cycle is defined as foot contact of one foot until that foot
contacts the ground again in the path of progression, or one step on each foot [11,12]. Each phase of the gait cycle is described with an overview of joint position, moments, and powers of the trunk, hip, knee, and ankle.

Herein, a full stride or 100% of the stride cycle begins at initial contact of the right foot and ends at the subsequent initial contact of the right foot, requiring a step for each foot (Figure 2-1). At 10% of the stride cycle, the opposite foot will push off (opposite toe off), and at 60% of the stride cycle the opposite foot will again contact the ground (opposite initial contact). This leads the right leg to leave the ground at 60% of the stride until the next heel contact, completing the stride. When the right leg is in contact with the ground (initial contact to toe-off), the right foot is in ‘stance’ and when the right leg is in flight this phase is termed ‘swing’ (toe-off to next initial contact). One stride is approximately 60% stance and 40% swing. When considering both legs, double support is the percent of the stride cycle in which both feet are in contact with the ground, about 20% of the stride cycle. Percent single support quantifies the part of a stride where only one foot (right and subsequent left) is in contact with the ground at a time, about 80% of a stride (Figure 2-1) [11].

Figure 2-1. Phases of the stride cycle represented from 0-100% of one stride. Events are marked by black diamonds, and double support phases are distinguished by grey shading.
Gait is often characterized by its temporal and spatial components. Gait speed reflects a summation of temporal (cadence, stride time, and step time) and spatial (stride length, step length, step width) aspects of gait. An example of spatiotemporal variables of gait are provided using a healthy young adult (HYA) example (Table 2-1). These characterize the BOS while walking, sometimes called the ‘walking base’, the area in contact with the support surface. The support surface is considered the area under one foot in single support and the distance between the two feet in double support, typically measured at heel strike.

Table 2-1. Spatiotemporal outcomes of example healthy young adult walk

<table>
<thead>
<tr>
<th></th>
<th>Speed (m/s)</th>
<th>Cadence (steps/min)</th>
<th>Stride L (m)</th>
<th>Step L (m)</th>
<th>Stride T (s)</th>
<th>Step T (s)</th>
<th>Step W (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right foot</td>
<td>1.28</td>
<td>114.29</td>
<td>1.31</td>
<td>0.61</td>
<td>1.05</td>
<td>0.53</td>
<td>0.13</td>
</tr>
<tr>
<td>Left foot</td>
<td>1.26</td>
<td>114.29</td>
<td>1.32</td>
<td>0.64</td>
<td>1.05</td>
<td>0.53</td>
<td>0.11</td>
</tr>
</tbody>
</table>

Note: Presented as average values. L=length, T=time, m=meters, s=second, min=minute

Overground walking data were collected from a 20-year-old HYA man (height: 1.9 m, weight: 77.4kg) using motion capture technology (Vicon, v 2.6). The subject was barefoot and instructed to walk at a comfortable pace across an eight-meter walkway. The marker position data was filtered with a 6 Hz fourth-order Butterworth filter. Note that the measures are similar, but may not match, previously published norms [12] because of inter-individual variability.

Four critical phases within the stance phase of the gait cycle detailed below include: (1) loading response, (2) mid-stance, (3) terminal stance, and (4) pre-swing. The five critical events of the stance phase include: (1) initial contact, (2) opposite toe-off, (3) heel rise, (4) opposite initial contact and (5) toe-off. The three phases of the
swing phase of gait include: (1) initial swing, (2) mid-swing, and (3) terminal swing. Two main events in the swing cycle include: (1) feet adjacent, and (2) tibia vertical [12].

**The stance phase**

The stance phase of this stride is outlined in Figure 2. Critical events within the stance phase are marked across the ankle, knee, and hip angles, moments and powers (left to right). In the HYA example (Figure 2-2), the stance phase was 60.16% of the stride.

![Figure 2-2. Stance phases of the stride cycle represented within the dashed lines on hip, knee, and ankle (top to bottom) angles, moments, and powers (left to right).](image)

**Initial contact.** The first critical event of a stride, initial contact, begins the loading response at the instant the foot contacts the floor. In healthy walking, the direction of the ground reaction force (GRF) vector projects generally upwards and immediately moves upwards and backwards during the loading response. The GRF vector projection is posterior of the ankle, and anterior of the hip and knee joint. The
GRF vector rapidly moves further backward within the first 10-20 milliseconds of the stride at the beginning of the loading response (Figure 2-2) [12].

The trunk is behind the leading foot (about half a stride length) in the sagittal plane at initial contact and begins to move medially towards the stance foot. Trunk lean is angled approximately 1.5° behind vertical and the trunk angle moves posteriorly at initial contact [13]. The trunk is twisted about the vertical axis in the transverse plane, with the greatest frontal excursion of the opposite shoulder and arm [12], approximately 5° with respect to the stance limb [13]. The amount of arm swing varies greatly between individuals [11] but is tightly linked to the anti-phasic rotation of the trunk and pelvis about the vertical axis [13,45].

The hip is in a neutral position at initial contact in the frontal plane. In the transverse plane, the hip is slightly externally rotated about the vertical axis, approximately 5° from neutral [13]. Previous literature reports the sagittal angle of the hip at initial contact is between 20° [11] and 30° degrees [12,13] and varies because of inter-individual differences. Figure 2-2 demonstrates a similar sagittal attitude of 21° at the hip in the example HYA. The hip extensors produce an internal extensor moment at the hip at initial contact that opposes the external flexor moment generated by the GRF, positioned anterior of the hip. The hips generate an internal extensor moment to prevent the hip from flexing and collapsing, acting concentrically and generating power. The hip extensors act at initial contact to control the forward rotation of the pelvis, slow the forward velocity of the trunk, and begin hip extension through the loading response [11,12].
The knee has extended rapidly just before initial contact and is almost straight. The knee flexors are active in the latter part of the swing phase, act eccentrically as a braking mechanism and prevent hyperextension of the knee at initial contact. At initial contact, the GRF vector projection moves anterior to the knee joint and the resultant external extensor moment is opposed by an internal flexor moment that stabilizes the knee. Concentric action of the knee flexors combined with the release of the stored elastic energy leads to a short burst of power generation at initial contact (Figure 2-2) [11,12].

The ankle is near its neutral position at initial contact, slightly supinated. Horizontal ankle velocity has dropped rapidly to near zero just before initial contact. The dorsiflexors are only slightly active in swing to clear the ground. To oppose the external plantarflexor moment from the posterior position of the COM, the dorsiflexors generate an internal moment to slowly lower the foot. Eccentric action of the dorsiflexors prevent ‘foot slap’ at initial contact (Figure 2-2) [11,12].

**Loading response.** The loading response phase is the period of double support (when both feet are on the ground) between initial contact and opposite toe-off (10-12% of the stride cycle). The HYA example underwent the loading response for 10.2% of the stride (Figure 2-2). The loading response is characterized by the rapidly increasing magnitude of the GRF, moving upwards and backwards [12]. The GRF vector is positioned upward, anterior to the hip and posterior to the knee and ankle joint by 20ms after initial contact (Figure 2-2).

The trunk is at its lowest position, approximately 20mm below its mean while its forward velocity is at its highest (about 10% faster speed than the average). The trunk is
moving toward the stance foot and the arms are beginning to rotate internally from a maximally rotated attitude. Throughout the loading response, the hip rotates about the vertical axis (from $5^\circ$ to $2^\circ$ by terminal stance) to bring the upper body back to a neutral position [13]. The hip extensors produce internal extensor moments to oppose an external flexor moment from the anteriorly positioned GRF, concentrically acting in an important phase of power generation during the loading response noted as H1 (Figure 2-3). Despite a brief drop in internal hip moments, the HYA example shows increased extensor moments that extend the hip during the loading response.

The knee starts to flex during the loading response, termed ‘stance phase flexion’. In the HYA example, the peak stance phase flexion angle was approximately $20^\circ$. To oppose the external flexor moment acting on the knee joint from a posteriorly projected GRF vector, the knee extensors begin to act eccentrically to prevent collapse of the knee. Knee extensors absorb energy throughout the loading response in the first phase of energy absorption at the knee, referred to as K1, from initial contact to about 20% of a stride (Figure 2-3).
Figure 2-3. Joint power phases for the hip, knee, and ankle across the stance phase of a stride represented by the dashed lines. Gait events are marked as diamonds and circles represent important phases of power for the hip, knee, and ankle (top to bottom) across the gait cycle.

The loading response involves plantarflexion at the ankle. This is sometimes called the ‘initial rocker’, ‘heel rocker’, or ‘heel pivot’, in which the foot pronates and the tibia rotates internally in plantarflexion. Internal dorsiflexor moments opposes external plantarflexor moments caused by the posterior orientation of the GRF throughout the loading response phase. The dorsiflexors control the loading response by slowing the translation of the tibia over the stationary foot eccentrically.

**Opposite toe-off.** The second event of the stance phase is opposite toe-off. Opposite toe-off marks the end of the double support period in the loading response
and begins mid-stance, the first single support period in a stride. In the example HYA, opposite toe-off occurred at 10.2% of the stride cycle, similar to examples provided by Whittle (2012) showing opposite toe-off occurs at 7% of the gait cycle (Figure 2-2) [12].

The GRF vector projection is oriented backward and upward, acting on the COM. The GRF projection is positioned anterior of the hip and posterior to the knee and ankle joint. The shoulder and arm are rotating about the vertical axis toward the medial aspect of the body. The pelvis begins to twist back to neutral. The vertical position of the trunk begins to move from its lowest position in the loading response and increases its height while the vector projection of the GRF acts directly on the COM and slows forward speed.

Beginning its extension at initial contact, the hip is still flexed at opposite toe-off. The HYA example shows a hip flexion angle of 20° at this event, similar to the example in Whittle of 25°. The hip continues to extend with an internal extensor moment acting concentrically to extend the hip and protect collapse against the external flexor moment generated by the GRF that is slightly anterior to the hip joint (Figure 2-2).

The knee is continuing to flex, reaching the approximate peak of stance phase knee flexion at opposite toe off and begins extension. The HYA example showed peak stance phase knee flexion (21.6°) which was reached at or slightly after opposite toe-off. The amount of knee flexion experienced at this time is sensitive to gait speed: slower speed leads to minimal knee flexion. The GRF is behind the knee producing an external flexor moment which is opposed by an internal extensor moment. The knee acts as a spring, storing energy in the knee extensors at opposite toe-off to prevent the GRF from...
building vertical force too quickly. This is part of the first major energy absorption phase at the knee, termed K1 (Figure 2-3).

The foot is flat on the ground at opposite toe-off. The ankle joint then moves from plantarflexion to dorsiflexion as the tibia translates over the stationary foot. The foot pronation and tibial rotation reach their peak at this moment and begin to reverse. The GRF moves anteriorly, initially producing internal dorsiflexor moments that then switch to internal plantarflexor moments at opposite toe-off. The dorsiflexors cease to activate and plantarflexors begin to act eccentrically initiating the A1 phase of energy absorption at the ankle joint.

**Mid-stance.** The second phase of the stance portion of gait is termed mid stance, beginning at opposite toe-off and ending at heel rise. Mid-stance occurs at approximately 7-32% of the stride cycle and composes 18% of the total stride [12]. The HYA example showed a mid-stance phase that occurred between 10 and 28% of the stride, totaling 17.4% of the stride cycle. This period is characterized by the swing leg passing the stance leg. The upwards acting GRF vector has translated anterior of the ankle and posterior to the knee and hip joint (Figure 2-2).

The trunk reaches its highest vertical position, approximately 20 millimeters (mm) above the mean in mid-stance. Maximum mediolateral displacement occurs in this phase as well where trunk position is approximately 20 mm beyond the mean over the stance foot. The shoulders and hips are in neutral as the arms pass each other in mid-stance, following the opposite leg. The GRF is moving backward and upward, acting on the COM.
At this point, the pelvis will drop toward the swing leg, sometimes referred to as ‘pelvic drop’. The hip is rotating back to the midline about the vertical axis to within 2° of a neutral position [13]. In the sagittal plane, the hip flexors act concentrically to extend the hip, moving from flexed to extended positions. Hip movement in the mediolateral plane relies on the concentric actions of the hip abductors of the stance limb to support the trunk and pelvis on one leg. The internal extensor moment generated at the hip during opposite toe-off generates a burst of power at the hip that declines throughout mid-stance. At the peak trunk position in mid-stance, concentric action of the hip extensors has ceased and is replaced with eccentric energy absorption. From there, the hip absorbs energy as the hip joint extends passively with inertia and gravity (potential energy).

The knee reaches its peak flexion in mid-stance, 21.6° in the HYA example. The peak typically occurs at 15-20% of the gait cycle. In alignment, the example in Whittle (2012) showed peak flexion angle at 20% of the stride cycle. The knee then begins to extend again in mid-stance while the GRF vector projects posterior to the knee joint causing an external flexor moment. When extension begins, an internal extensor moment acts as a spring to extend the knee, represented as K2, the second important phase of knee power during walking (Figure 2-3). Representing about 10-15% of the energy generated in normal walking, K2 is the only energy generated by the knee extensors in a stride [11].

At the ankle, the foot has hit the ground at ‘foot flat’ when opposite toe-off occurs. When the foot is on the floor, the tibia is rotating about the ankle joint. The ‘mid-stance rocker’ or ‘ankle rocker’ characterizes the forward translation of the tibia that change the
ankle position from plantarflexion to dorsiflexion. As the ankle moves from plantarflexion into dorsiflexion, an increased plantarflexor moment builds throughout the stance phase. The plantarflexors act eccentrically and absorb energy as the tibia moves over the flat foot. As the tibia rotates, the foot moves into peak supination, and the GRF vector moves forward and is on the forefoot by heel rise.

**Heel rise.** The third event of the stance phase, heel rise, marks the transition from mid-stance to terminal stance. At this time, the heel of the stance foot begins to lift. The timing of this event varies between people, Whittle (2012) showed this occurred at 32% of the stride while the HYA example lifted the heel at 28% of the stride cycle. The GRF vector projection is acting almost directly upwards, anterior of the ankle and knee but posterior to the hip joint (Figure 2-2).

The trunk is falling from its highest point, reached in mid-stance. The trunk begins to move back toward the opposite leg, transferring weight in preparation for the swing phase. As the stance hip extends, the hip rotates posteriorly. This transverse rotation is opposed with forward rotation of the ipsilateral shoulder and arm.

The hip continues to extend from heel rise. Eventually, peak extension is reached in initial swing. The hip is rotated about the vertical axis to its maximum of 15º of external rotation. The hip abductors must still support the stabilization of the pelvis. The GRF is positioned posterior to the hip joint, causing an external extensor moment. This is opposed by a small, but increasing, internal flexor moment at the hip is thought to be a result of stored energy from the hip and knee ligaments [12]. The hip flexors act eccentrically to absorb energy, decelerating the leg as it rotates posteriorly [11].
resultant energy absorption at the hip continues through terminal stance until toe-off, noted as the H2 phase of hip power (Figure 2-3).

The knee continues to extend, moving towards its peak extension. When extension begins, concentric actions of the extensors act as a spring to extend the knee. The GRF is moving to the forefoot and anterior to the knee joint, causing an external extensor moment that is opposed by an internal flexor moment at the knee. Because the plantarflexors are two-joint muscles that cross the knee joint, the plantarflexors at the ankle then generate internal flexor moments which translates to the knee- slowing the forward motion of the body about the foot. This is termed the ‘plantarflexion/knee’ coupling mechanism that activates the plantarflexors and prevents hyperextension by acting as a knee flexor.

Ankle dorsiflexion angles are steadily increasing at heel rise. The GRF has moved to the forefoot of the stance foot, in front of the ankle joint, causing an external dorsiflexor moment. This is opposed by a strong internal plantarflexor moment. As the heel rises, the metatarsophalangeal joints provide the axis of rotation known as the ‘metatarsal break’ or ‘toe break’, about which the heel rises, and the toes remain flat on the ground. Heel rise occurs as the dorsiflexors are absorbing energy rapidly to slow the forward translation of the tibia about the stationary ankle joint.

**Terminal stance.** Terminal stance, the third part of the stance phase prepares for the contralateral (swing) foot to contact the ground. This phase occurs between heel rise and opposite initial contact, between 27.5% and 50.4% of the HYA example stride. The GRF is projecting upwards and backwards, oriented in front of the ankle joint and behind the knee and hip (Figure 2-2).
The hip continues to extend as the thigh rotates backwards. To prevent collapse against an external flexor moment, the hip absorbs energy in the hip flexors through terminal stance up until toe-off. This energy absorption phase (H2) is important for initiating ‘pull-off’ of the hip at toe-off (Figure 2-3). The knee moves from peak extension in mid-stance and begins to flex the knee for toe-off. Initially, the knee extensors act concentrically to extend the knee against an external flexor moment. Then the knee extensors then absorb energy in preparation for toe-off. The ankle is continuing to dorsiflex, but begins plantarflexion towards the end of terminal stance, just before opposite initial contact. The force production and energy absorption at the ankle is primarily achieved in terminal stance, in the A2 phase (Figure 2-3). The GRF projection acts on the ankle joint with an external dorsiflexor moment that is opposed by a strong plantarflexor moment. The ankle plantarflexors transition from absorbing energy from the forward rotating shank to generating force to propel the foot forward and initiate toe-off.

**Opposite initial contact.** The fourth event of the stance phase is the instant that the opposite foot contacts the ground at approximately 50% of the stride cycle. This event occurs at 50.4% in the HYA example. The GRF is positioned in front of the stance limb ankle, but posterior to the knee and hip (Figure 2-2).

Opposite initial contact corresponds with the maximum extension at the hip in the stride cycle. Maximum hip extension angle in the example provided by Whittle was -19º which is similar to the example provided in Figure 2-1. The hip generates an internal flexor moment to oppose the external extensor moment acting on the hip joint.
The knee has already begun to flex at opposite initial contact. The stance knee of the HYA example was flexed at an angle of 13.6° when the opposite foot hit the floor. The knee generates an internal extensor moment that opposes the external flexor moment from the posteriorly oriented GRF. The knee extensors act eccentrically at opposite initial contact to prevent the knee from flexing too rapidly. This marks the beginning of a critical phase of energy absorption at the knee known as K3 (Figure 2-3).

The ankle reaches peak energy generation at opposite initial contact, marked as the A2 phase (Figure 2-3). The force and energy generated at the ankle in A2 is the largest source of energy produced in the stride cycle and a primary determinant of step length [46].

**Pre-swing.** Pre-swing is the fourth and final part of the stance phase, and the second period of double support in the stride cycle. The pre-swing phase begins at opposite initial contact and ends at toe-off, comprising approximately 10% of the stride cycle. The HYA example experience pre-swing from 50.4-59.8% of the stride, in alignment with previously reported norms (Figure 2-2).

In the pre-swing phase, the vertical position of the trunk begins to climb again. The trunk then moves medially from the stance leg toward the new supporting foot. The rotation of trunk about the vertical axis is approximately neutral as the arms meet again in the midline. The hip angle begins to move positive as its position transitions from peak extension into flexion. The eccentric actions of the hip flexors are then replaced by concentric action of the hip flexors. Internal joint moments are generated by the hip flexors as they prepare for ‘hip-off’ or ‘pull-off’ and complete toe-off. The GRF has moved from in front of the knee to behind it, changing the internal joint moments from
flexor to extensor. The knee extensors absorb power to limit the rate of knee flexion. Pre-swing is marked by K3, a period of energy absorption followed by a short burst of power generation at the knee (Figure 2-3). This is likely a consequence of the strong plantarflexion at the ankle that engages knee flexors during A2, generating a flexor moment that ultimately aids in pull-off.

The ankle rapidly moves from dorsiflexion in terminal stance to plantarflexion in pre-swing, to push off the ground. The foot undergoes maximal inversion that is coupled with external tibial rotation, locking the tarsal joints and structurally stabilizing the foot for weight bearing. The GRF is located anterior to the ankle joint, imposing an external dorsiflexor moment that must be opposed by the plantarflexor moment. The ankle moments and powers are beginning to decelerate as the ankle prepares for passive motion in swing. The ankle generates the most energy at opposite initial contact and initiates lift-off of the swing limb (A2) (Figure 2-3).

**Toe off.** The fifth and final event in the stance phase is toe-off, the moment the stance foot leaves the ground to initiate the next heel contact. Toe-off occurs at approximately 60% of the gait cycle. The HYA example experienced toe-off at 59.8% of the stride, similar to 57% in the example provided by Whittle (2012). The GRF projection is oriented upwards and forwards and is positioned anterior of the ankle joint and posterior of the knee and hip joints (Figure 2-2).

Toe-off occurs just after peak extension of the hip, but after the hip begins to move into flexion. The hip is maximally externally rotated about the vertical axis at 15º. The hip continues to flex as the foot leaves the support surface. External extensor forces are opposed by an internal flexor moment at the hip that help initiate knee
flexion. Internal flexor moments are a product of gravity, tissue elasticity, and adductor activation and gradually increase after toe off in the third important power phase at the hip: H3 (Figure 2-3). The concentric action of the hip flexors reaches its peak of energy generation at toe-off. To initiate hip pull off, the HYA example demonstrated a peak power of 1.46 W/kg, which occurs immediately after the toe leaves the ground.

The knee is flexing rapidly to pull the foot off the ground but does not reach its peak until after toe-off. As the hip flexes, the shank is behind the hip and the knee flexes because of the forward inertia of the body. When the knee is flexed it begins acting as a double pendulum in preparation for the swing phase. The GRF vector is behind the knee joint, opposed by eccentric actions of the knee extensors that generate internal extensor moments. The knee extensors store energy during the eccentric action, absorbing power until the GRF moves forward along the foot during the phase of knee power known as K3 (Figure 2-3). The short power burst at the knee at toe-off is attributed to the concentric action of the plantarflexors that transmit forces up the shank that act as knee flexors. The HYA example exhibited a similar pattern of energy absorption at the knee and a small power burst just after toe-off in K3 (Figure 2-3).

Peak ankle plantarflexion occurs just after the foot is lifted from the ground. The example HYA reach a peak plantarflexion angle of -11.6°, but the example provided by Whittle reached a peak of -25° [12]. The reason for this difference may be a result of different methods or inter individual differences. Note, the example provided by Whittle is a woman and the HYA example provided here is a man, which could influence the differences in gait outcomes. The external plantarflexor moment imposed by the GRF is controlled eccentrically by the dorsiflexors to lower the foot to the ground gently.
The swing phase

The swing phase is the period of walking in which only one foot is in contact with the ground, approximately 40% of a stride (Figure 2-4). The HYA example was in the swing phase 42% of the stride. The swing phase of gait is outlined below.

**Initial swing.** The first of the three phases of the swing cycle are initial swing, beginning at toe-off and ending at feet adjacent (Figure 2-4).

At toe-off, the hip has reached maximal external rotation of 15º, which begins to internally rotate in initial swing [13]. As the stance leg moves into the swing phase of gait, the foot pulls off the ground initializing the transition into potential energy in swing. The hip is moving gradually into a positive flexion angle. Angular velocity increases as the hip flexors bring the leg up. Internal flexor moments balance the torso and begin to
act concentrically to flex the limb. This power is generated in part by gravity and ligament elasticity. Power profiles show that the hip then undergoes a transition from positive to negative power as hip extensors begin to absorb energy (H3) at the midpoint of initial swing (Figure 2-5).

At this time, the knee is flexing and begins to extend passively. The extension the knee undergoes is part of the ‘pendulum’ action coupled at the hip and knee joints. Peak swing phase flexion of the knee occurs in initial swing. The example in Whittle (2012) reached a peak knee flexion angle of approximately 49° and the HYA example here reached a peak swing phase flexion angle of 61.9°. A small knee extensor moment exists throughout initial swing and absorbs energy to prevent the knee from flexing too much. The K3 power phase extends into initial swing as the knee extensors absorb energy and transition to generating energy by mid-swing (Figure 2-5). The swinging foot moves into a neutral position and remains passive through the initial swing phase.

Feet adjacent. The first event of the swing cycle is feet adjacent, the point in time that divides initial swing and mid-swing. Feet adjacent occurs when the swing leg passes the stance leg and the two limbs are side by side. The HYA example of feet adjacent at 78% of the gait cycle (Figure 4) was virtually identical to the example provided in Whittle (2012) whose feet were adjacent at 77% of the stride (Figure 2-4) [12].

The trunk is in its highest vertical position (and lowest forward velocity), located at its maximum medial position over the (left) stance foot. The arms are ‘level with each other’, meeting at the midline, as one is moving anteriorly and the other posteriorly. When the feet are adjacent, the hip has almost reached peak flexion. The hip angle is in
peak flexion just after this event. The HYA example showed a hip flexion angle of 18.5° at feet adjacent that increases until tibia vertical (Figure 2-4). Internal joint moments are generated by the hip flexors as they become more involved to control the acceleration of the swing leg. At feet adjacent, flexor moments begin to transfer to extensor moments as the swinging leg power transfers from potential to kinetic energy. Feet adjacent marks the transition from potential to kinetic energy for the hip, where kinetic energy is transferred to the trunk and the eccentric action of the flexors is replaced by the extensors.

Knee flexion is primarily driven by flexion of the hip in the swing phase. Peak knee flexion has already been reached and the knee joint is beginning to move into extension by feet adjacent. In pre-swing, the knee generates a small internal extensor moment to limit the magnitude of knee flexion caused by the hip and ankle at toe off. Feet adjacent marks the transition from knee extensor to flexor moments, and the corresponding transition from energy generation by the extensors, to absorption by the flexors.

The ankle is moving from plantarflexion at toe-off to a neutral position at feet adjacent. Dorsiflexion is at its peak in the swing phase to ensure the toe clears the ground safely. However, the majority of toe clearance is achieved at the hip and knee joint, only dorsiflexing the ankle to ~3.5°. Controlling toe clearance is critical at this moment, clearing the ground by only 1.4 cm.
**Mid-swing.** Appropriately, the middle of the swing phase is called ‘mid-swing’, characterized by the swing leg transition from potential to kinetic energy. Mid-swing begins at feet adjacent and ends at tibia vertical (Figure 2-4).

Few key things happen during mid-swing. The trunk is at its highest vertical position and horizontal velocity has dropped to a minimum at the peak of trunk position. The hip angle moves into peak flexion. Internal joint moments are generated by the hip extensors absorb energy in mid-swing. Hip powers are positive as the flexors begin to act concentrically to perform ‘hip-off’, the flexing of the hip that initiates toe-off [11].

The rapidly extending knee angle is a result of the passive extension of the shank, acting as a double pendulum from the knee and hip [13]. Knee moments are close to zero as the knee flexors act eccentrically to prevent the knee from disruptions to hyperextension just before the next surface contact. Peak velocities of the foot can build to up to 4.5 m/s by mid-swing with a toe clearance as low as 0.87 cm [11]. A slight dorsiflexor moment flexes the foot into slight dorsiflexion so the forefoot clears the ground.

**Tibia vertical.** The second event of the swing phase and the final event of the stride, tibia vertical, divides mid-swing and terminal swing. The HYA example underwent tibia vertical at 85.8% of the stride, almost identical to the example provided by Whittle (2012) of tibia vertical occurring at 86% of a stride cycle (Figure 2-4) [12].

The trunk has begun to decline from its maximum vertical height, increasing velocity and lateral displacement, and translating medially back toward the midline. The right pelvis is anterior to the left, which is opposed by the left arm, positioned ahead of the right. At tibia vertical, the hip has reached peak flexion, approximately 21° in the
HYA example and 27° in the example provided by Whittle (2012). The hip then begins to extend in preparation for the next initial contact. Increasing internal moments at the hip are generated by the hip extensors which transfer energy to the trunk. The knee is extending rapidly toward its maximum extension angle at the next initial contact. This extension is achieved passively with the double pendulum action at the hip and knee joint [15]. Eccentric action of the knee flexors prevents hyperextension of the knee (Figure 2-5).

Figure 2-5. Joint power phases for the hip, knee, and ankle across the swing phase of a stride cycle. Gait events are marked as diamonds and circles represent important phases of power for the hip, knee, and ankle (top to bottom) across the gait cycle.
At the ankle, the toe has just cleared the floor and the angle of the ankle joint becomes less consequential. The HYA example remains slightly dorsiflexed (2.8º) at tibia vertical and begins to move into plantarflexion in preparation for the next foot contact (Figure 2-4). The ankle primarily moves through the swing phase without major forces produces at the ankle. At tibia vertical, the dorsiflexors are producing a small moment to maintain the position of the ankle. Energy is not generally exchanged in the swing phase at the ankle, thus power profiles remain at nearly zero (Figure 2-5).

**Terminal swing.** The third and final period of the swing phase of the stride cycle is terminal swing, also known as ‘reach’, approximately 85-100% of the stride that prepares for the next initial contact (Figure 2-4). The terminal swing phase begins at tibia vertical and ends at the next initial contact.

In preparation for the next contact, the hip switches from a passive extension to a concentric extensor moment. The final phase of knee power levels occurs in terminal swing, noted as K4 (Figure 2-5). K4 marks the final major absorption of energy by eccentric work of the knee flexors. Toward the end of terminal swing, the knee generates a flexor moment during K4 that increases until the next initial contact. In the last 10% of the stride cycle, the vertical and horizontal velocity of the foot decrease rapidly to zero to gently lower the foot to the ground at the next initial contact [11].

**Dynamic locomotor balance**

Patla has defined stability as the maintenance of the vertical projection of the COM within the BOS [11,15,17,47]. If the COM moves beyond the BOS, the individual will fall or need to reactively adjust steps to prevent a fall [48]. In walking, the COM is continuously oscillating atop a BOS that changes in size (on one foot or two feet). The only way to ensure upright balance as the body progresses forward is to appropriately
place the swing limb to rebalance the COM during double support as needed. The COM oscillates between 37mm and 39mm up and down and between the feet two times during each stride [47]. Stability depends on the characteristics of the position and velocity of the COM at the time of foot placement [15]. Within each step, foot placement keeps the COM within the BOS [15]. To keep the COM inside the BOS, one must sense current and future COM position relative to the direction and magnitude of the GRF and the boundaries of the BOS and direct foot placement accordingly [15]. The purpose of this study is to determine if age-related changes to the motor control of upright posture and dynamic balance involve additional cognitive resources [15,17,47].

Early models of human walking use an inverted pendulum model, a simplified interpretation of the COM atop the limbs acting as pendulums [11,17,47,49]. The inverted pendulum model suggests that energy efficient forward walking can be passively accomplished through capitalizing on gravity and inertia of the COM [17]. Thus, gait is sometimes described as continuously ‘falling forward’ between the two feet. More modern models that include the width of the pelvis and collisions with the ground as a determinant of walking behavior have revealed that maintaining the COM within the BOS incurs motor, metabolic, and mechanical costs.

The primary objective of dynamic balance control is to prevent falling by keeping the COM atop the vertical GRF vector and within the BOS [11,15,49]. Foot placement provides supporting forces at initial contact and drives COM movement between the two feet [50,51]. In the sagittal plane, adjusting step length, the horizontal distance between the two feet in the path of the progression, is required to redirect the forward velocity of the COM from step-to-step [52]. Additionally, if step length increases, the metabolic
costs and mechanical work to walk increase significantly [53]. Studies show that the controller of the COM trajectory in the frontal plane while walking is step width, the horizontal (mediolateral) distance between the two feet at initial contact [15,50]. In single support, the COM moves outside of the BOS, thus mediolateral balance ability is particularly important during the single support phases of gait because the COM must negotiate a small BOS (one foot). In single support, the stance limb must support the 2/3 of the body’s mass, acting downward through the pelvis and while the COM is moving at its highest forward velocity. Further, the stance limb must stabilize the pelvis to limit ‘pelvic-drop’, when the contralateral pelvis lowers and rotates anteriorly as the swing leg moves from behind to in front of the COM. Thus, the ability to ‘catch’ the COM and keep it in the BOS is reliant on foot placement of the swinging limb. Previous research shows that mediolateral foot placement requires added neurological feedback of the position of the BOS relative to the COM at heel strike [49]. Thus, dynamic balance during locomotion requires goal-oriented adjustments and complex sensorimotor processes.

Controlling the BOS relies on sensory input from the center of pressure (COP), the average point location of the forces acting on the on the ground [49,50]. Proprioceptive sensation of the COP under each foot provides feedback for the neural control of postural and limb muscles in walking [49]. Previous researchers have suggested that the relationship between the COP and COM is uniquely perceivable and is critical to maintain stability and keep the COM within the BOS [49,52]. Additionally, when visual input is removed, mediolateral walking variability increases, suggesting that foot placement is ‘actively controlled’ to maintain frontal plane balance and requires
constant neurological feedback [49,54]. In alignment, recent research of brain activation during treadmill walking in young adults demonstrated an increase in cortical activity in the single support phases of gait [19]. In the most destabilizing phases of gait, increased activation of cortical networks implies that maintaining dynamic balance requires higher-level resources. Yet to be determined are the age-related changes to neurological control of dynamic balance and the degree balance control interferes with cognitive processes, and perhaps vice versa.

The relationship between the COM and the BOS in dynamic balance control has been recently examined by alleviating body weight (stabilizing the COM) and manipulating step width (BOS). Dragunas et al. alleviated varied levels of body weight via a ceiling-mounted walking harness to observe the effects of supporting body weight on dynamic balance [51]. Dragunas et al. reported that when body weight was lower (weight was alleviated by the harness), step width variability decreased, suggesting that the demands of mediolateral balance were reduced. Simultaneously, step width increased, which may have been a compensatory strategy to ensure balance in the face of reduced feedback from lateral supporting forces [51]. Dragunas et al. concluded that the magnitude of GRFs at heel contact may provide critical neurological feedback for the control of mediolateral balance [51].

Donelan et al., investigated the influence of the BOS on dynamic balance by observing walking at a narrow step width and a preferred step width [34]. Additionally, participants walked with and without lateral stabilization. With lateral balance support, individuals were better able to narrow their step width and chose a smaller step width. Further, step width variability decreased when balance was supported. These findings
led Donelan et al. to suggest that mediolateral balance is actively controlled [52]. Dean et al., followed up on the experiment by Donelan et al. and examined lateral balance in young and older adults. Older adults performed significantly worse than young adults in narrow walking, but both groups were equally able to reduce step width by 58% with lateral stabilization. However, older adults did reduce step width variability more than young when lateral stabilization was provided, alluding to age-related effects of dynamic balance on walking. The authors agree that lateral stabilization provides neurological feedback that reduces the sensorimotor integration demands of dynamic balance [34].

Considered together, dynamic locomotor balance control appears to involve the nervous system, be highly influenced by body weight and mediolateral step width, and susceptible to age-related changes. This dissertation will determine if the motor control of the COM and BOS in dynamic balance are associated with higher-level neurological processes in aging. This will be accomplished by observing subjects of all ages walk in stable and unstable constraints at a fixed treadmill speed, which will eliminate biomechanical changes to dynamic balance as a result of speed. Rather, changes to gait and cognitive performance will be a direct result of the effect of controlling the COM and BOS.

**Age-related changes to gait and dynamic balance**

Overcoming age-related physiological and neuromuscular changes makes safe walking increasingly difficult for older adults. Arguably, a high prevalence of devastating falls for older adults [55] is a result of an accumulation of several factors that impair balance ability. The source of age-related changes to gait is multifactorial, but older adults consistently demonstrate an overall slowing of gait [40]. The following will review
existing evidence of age-related changes to walking and dynamic balance control, the focus of the proposed dissertation.

The physiological changes that affect gait in older adults include a loss of muscle tissue, motor unit losses and re-innervation, decreased aerobic capacity, and poor skeletal health [56–58]. Researchers have reported more force variability [59,60], increased reaction times [61], and reduced proprioception [62] in older adults as a result of age-related changes at the musculoskeletal level [56,57,63,64]. Such consequences of aging lead to difficulty generating and absorbing the energy needed for optimal walking [27,40]. Age-related changes to muscle fiber size and activation patterns from apoptosis (death of unused motor units) disrupt oxygen consumption and production at the muscle level [69]. Such degradation in muscle tissue function translates to greater energetic costs of walking [70,71] for older adults than young adults when walking at the same pace [34,40].

Older adults must also overcome changes to the skeletal system that may inhibit locomotor movement at the hip, knee, and ankle. Increased tissue stiffness, particularly in cartilage [72,73], leads to a smaller joint range of motion and reduced transmissions of force across the joints with age [13,40,73]. Additionally, older joints bear the burdens of a lifetime of joint wear and tear, which often leads to painful osteoarthritis of the lower limbs. Older individuals diagnosed with symptomatic osteoarthritis demonstrate lower range of motion, reduced joint moments and powers, and a longer stance phase [74,75].

The biomechanics of age-related changes to walking are apparent in many aspects of gait performance. First, where young adults demonstrate an ability to
dampen vertical GRFs that produce external forces up the kinetic chain and resultant head accelerations, older adults do not similarly attenuate vertical forces. Damping the vertical GRF that acts on the pelvis is critical to stabilizing ‘the visual and vestibular platform’ and depends on healthy structural and neuromuscular properties of the trunk. Without effective damping of head acceleration, older adults could impair sensorimotor integration in the visual and vestibular system, and hinder their critical role in locomotor control [27,40].

Older adults demonstrate changes to force production that lead to imbalanced support moments at the hip, knee, and ankle. Detailed above, intra-limb and inter-limb coordination of the moments at the hip, knee, and ankle are required to prevent vertical collapse [11]. The majority of the energy generated in the stride cycle is to accomplish propulsion at the ankle to push the toe off the ground [11,13]. Winter showed that older adults produce significantly lower propulsive forces at the ankle than young adults (older: 2.48 W/kg; young: 3.27 W/kg) [27]. Later, Judge et al., showed that age-related changes to kinematics increased flexor power at the hip (young: -0.87 W/kg and older: -0.92 W/kg), to compensate for similar energy absorption at the knee in stance (young: 0.16 W/kg and older: 0.17 W/kg), but lower energy production at the ankle (young: 3.5 W/kg and older: 2.9 W/kg) during toe-off [27,28]. The result is difficulty producing the energy needed to initiate toe-off, ultimately shortening steps [27,28,40]. Peak ankle power predicts step length, supporting that lower ankle propulsion in older adults leads to shorter steps and slower walking [28]. This is further supported by evidence that walking cadence does not change with age [27,39]. Disruptions to joint powers may also relate to a higher horizontal heel velocity at initial contact in older (1.15 m/s) than
younger (0.87 m/s) adults, making older adults dependent on friction to reduce horizontal heel velocity to zero by mid-stance [11,16,76]. In the case that the ground is slippery and cannot provide a stabilizing coefficient of friction, a higher horizontal velocity at heel strike increases risk of slips in older adults than young individuals [11,40,76]. Thus, the slowing of walking speed may further aid older adults in overcoming increased horizontal velocities and prevent slips.

Potentially related to decreased strength and imbalanced support moments, older adults exhibit increased trunk angles when walking (i.e. a forward lean) [11,27]. A forward lean may be a strategy to increase stability by lowering the height of the COM and reducing the distance between the COM and the BOS boundaries [77,78]. However, this compensation may lead to imbalance from an anteriorly displaced COM that is closer to the anterior boundary of the BOS. If the COM moves beyond the BOS, the individual will fall or need to reactively take a step to prevent a fall [16,48]. If the COM is already at risk of moving beyond the critical BOS boundary, an older adult can modulate gait speed to mitigate this risk. Therefore, some suggest that slowing the forward velocity of the COM is a strategy to increase stability of the COM with regard to the BOS [27,40,55].

Age-related declines in proprioceptive and vestibular sensory systems reduce the ability to sense the point of COP application and body orientation [52]. Inability to perceive the position of the COP in relation to the COM increases risk of moving beyond the BOS. Reduced sensory input may lead to more 'cautious walking' during unstable phases of gait in older age. For instance, older adults tend to limit their time in single support and increase time in double support [27,39]. Reducing gait speed
correspondingly limits single support time, minimizing the risk of imbalance. Thus, slower speed may be a consequence of the risk to stability in single support that older adults attempt to avoid. Cautious walkers make further efforts to increase the BOS mediolaterally (wider steps) to promote stable posture [14,15,34,34,49,50] or shorten steps to keep the COM position within a stable support [14,15,17,27].

Cumulatively, the evidence shows that older adults adopt a more cautious strategy to maintain dynamic balance that may be a factor in slower walking. Interestingly, a cognitive load exaggerates the symptoms of a more cautious gait strategy [6,38]. Moreover, the magnitude of changes to walking with a cognitive load predicts cognitive impairment, fall-risk, frailty, and mortality in older adults [2,6,7,32]. This association between cautious gait and cognitive decline may be a consequence of involving higher-order cognitive processes to adapt to age-related changes to the control of dynamic balance.

**Neural Control of Gait**

Postural control and locomotion were once considered reflexive, requiring minimal cognitive processes [58]. Recent evidence to the contrary shows that higher-order cognitive processes are involved in postural control [24,26] and locomotion [7,20,22,79]. Scientists now recognize that the control of locomotion involves complex processes in both central and peripheral nervous systems and the coordination of each [80]. The contribution of the peripheral nervous system to human movement is beyond the scope of this review. The central nervous system plays a large role in movement and is the focus of this dissertation project. Hence, the focus of this review will be
‘higher-order’ cognitive processes involved in gait. Herein, higher-order cognitive systems are defined as structures and networks within the midbrain and cerebral cortex.

**Mid-brain**

From the spinal cord, the midbrain is the first anatomical stop to the higher-order resources involved in gait control. The mid-brain acts as a super-speed highway between the cortex and spinal cord [58,80]. Here the mid-brain is considered the brain stem composed of the mid-brain, pons, and medulla. Each structure of the midbrain is part of intricate networks that help initiate and regulate human movement by coordinating between the spinal cord and cortical networks. Often characterized as the ‘fight or flight’ network, the midbrain coordinates the delicate balance between inhibitory and excitatory projections to postural muscles and limbs critical to upright posture and locomotor control. The midbrain mediates muscle tone with specialized structures that carry information between cortical and spinal structures. These structures help initiate and maintain gait rhythmicity [81]. Noteworthy, these structures also connect directly to the cerebral cortex (frontal and parietal lobes) [58].

**Cerebellum**

The cerebellum is a remarkable multi-faceted structure involved in most aspects of cognition and movement [82,83]. The cerebellum is housed under the occipital cortex of the human brain and divided into two lateral hemispheres that contralaterally receive feedback and project output [58]. Relatively large compared to other species, the cerebellum is a specialized structure that contributes to muscle tone, anticipatory postural adjustments, emotional regulation, monitoring, correcting, and learning
movement through experience [82,83]. The cerebellum is specially designed to constantly monitor movement while integrating input from sensory and motor neurons. Specialized Purkinje cells rapidly receive feedback and transmit these signals to deep cerebellar nuclei. In parallel, the cerebral cortex relays feedback of intended movements. Integrating cortical and spinal information, the cerebellum monitors ongoing movement and compares the movement outcomes to intended movements to correct errors [82,83]. This information is relayed through direct projections to the basal ganglia and cortical structures that work together to adjust movements accordingly.

**Cerebral cortex**

More complex movements, executed to accomplish a particular goal (i.e. to reach a chair to sit down, to drive a grocery cart, etc.), activate widespread cortical networks [79,84]. Gait is often considered a volitional and goal-oriented task, typically initiated to move to a planned location. Thus, gait is well-practiced, but goal driven, and seems to recruit structures and complex networks throughout the cerebral cortex. As the task becomes more complex (i.e. obstacle crossing), additional cortical networks become involved [85,86].

The networks and regions of the brain involved in walking are only beginning to be discovered [87], but several have been recently identified in the control of movement reviewed below: the basal ganglia, posterior parietal cortex, sensorimotor cortex, primary motor cortex, premotor cortex, supplementary motor area [19,58,79,80,84,86]. Overall, the cerebral cortex plays a critical part in the control of movement.
**Basal ganglia**

The basal ganglia are integral in movement and tightly linked to the cerebellum and the cerebral cortex. The basal ganglia work with the cerebellum and cortical structures critical for initiating motion, smoothing movements by balancing excitatory and inhibitory motor control. The basal ganglia continuously integrates information of the body schema from the cerebellum and updates the cortical areas executing a motor task [80,88]. The basal ganglia network is directly involved in the excitation and inhibition of postural muscles critical to balance and upright posture. Additionally, the basal ganglia are involved in generating gait and maintaining gait rhythmicity through projections to the cerebellum and midbrain (i.e. the pedunculopontine nuclei) [81,88,89]. The basal ganglia are responsible for smoothing movements through direct projections from the primary motor cortex and cerebellum. Such direct projections onto the primary motor cortex and the limbic system also implicate the basal ganglia network in volitional motor control, cognitive processes, and the influence of emotion in human movement [58,80,81,88,89].

**Posterior parietal cortex and sensorimotor cortex**

Two major structures involved in voluntary human movement are located adjacent to the parietal lobes of the cerebral cortex: the posterior parietal cortex (PPC) and the sensorimotor cortex (SMC) [85,86,90]. The PPC is immediately posterior to the SMC and is rich in dense grey matter that enables fast communication between the PPC and SMC, as well as throughout a complex network throughout the brain [58]. The SMC simultaneously integrates sensory information from visual, auditory, and
proprioceptive feedback projecting to the brainstem and spinal cord. The SMC relays sensorimotor information to and from the basal ganglia and cerebellum [80,81].

Particularly important to locomotion, the adjacent vestibular cortex supplies orientation and positional information [81]. Paired with the dominant visual cortex, these structures are intricately related to perceptions of the self and the environmental factors involved in the planning and control of movement [91,92]. The PPC uses sensory information to coordinate with the basal ganglia and cerebellum to develop and maintain a body schema, or internal representation of oneself within the environment [91,93]. This schema originates in the sensorimotor, temporoparietal, vestibular, and PPC and is maintained in the dorsolateral prefrontal cortex for short-term retention and manipulation [92]. The PPC ensures the maintenance of the body schema relative to a changing environment, which is necessary for successful locomotion. When adaptive gait is required (i.e. stepping over an obstacle) greater sensorimotor processing in the frontal lobe is accomplished through connections from the PPC to frontal regions (i.e. anterior cingulate cortex, prefrontal cortex, posterior parietal lobe, and sensorimotor cortex) [84,91–94]. These experiences are stored in the cerebellum and used to learn and develop movement skills [91–93].

**Frontal lobe structures**

The anterior cerebral cortex (frontal lobe) is composed of several regions that are dedicated to movement control [19,58,80,81,84]:

- Primary Motor Cortex (M1)
- Premotor Cortex (PMC)
- Supplementary Motor Area (SMA)
The M1 is a large region of the frontal lobe just behind the central sulcus. The M1 organized somatotropically so that larger areas control fine movements and sensation (i.e. the hand and mouth). M1 directly projects to the spinal cord, midbrain, and is part of several cortical networks critical to movement (i.e. the fronto-parietal and fronto-cerebellar networks). Movements are achieved by activating motor neurons that control the direction and amplitude of movements [58,84]. Working in harmony, the PMC and SMA project to M1 to anticipate, initiate, direct, and plan motion [58,81,84,90]. These structures are also constantly receiving and sending feedback to and from SMC, PPC, basal ganglia, as well as midbrain and spinal systems.

**Age-related changes to the neural control of gait**

Overall, the networks involved in movement are remarkably complex and activate the entire brain [80,81,84]. In older adults, age-related changes in the integrity of cortical networks can impair mobility. Older adults exhibit reduced cortical volume [95], white matter [66], grey matter [96], and microstructural integrity [63,65] as well as increased amyloid plaques [68]. In movement, such cortical degradation has also been linked to increased force variability [63] and difficulty walking [65–68,96].

Longstreth et al. conducted an early study that revealed the link between cortical function and mobility in older adults [65]. Magnetic resonance imaging (MRI) of 3300 older adults showed that reduced white matter integrity (intra-cortical connectivity) was correlated with walking speed, static balance, the sit to stand, falls, and dysfunction in activities of daily living [65]. Later investigation using diffusion tensor imaging (DTI) revealed that intra-cortical neural connectivity was related to gait speed, stride length, cadence, and stride width in 429 older adults [66]. Brujn et al., used DTI to compare
young and older adults, confirming that older adults were susceptible to reduced white matter connectivity that was associated with degraded walking stability (step width and safety margin). Complementary evidence was provided by a recent study showing that reduced grey matter volume was related to greater instability while standing in older adults [96]. Evidence from Positron Emission Tomography (PET) revealed the influence of beta-amyloid plaques in older adult brains on mobility [68]. Wennberg et al., showed that PET could predict gait outcomes (speed, cadence, double support, stance time) in older adults such that worse hemodynamic flow in regions of interest led to worse gait outcomes [68]. Nadkarni et al., (2016) used similar methods to investigate the effect of cortical function on dual-task walking performance in older adults [97]. The authors reported that dual tasks impaired gait in those with worse cortical integrity [97].

Considered together, the evidence shows that older adults demonstrate reduced cortical integrity that often leads to difficulty in daily mobility tasks. Thus, the relationship between the brain and mobility has significant implications for older adults, warranting further investigation of the resultant impacts on daily life. This experiment proposes to evaluate the behavioral effects of such age-related neurological changes on the association between cognition and fundamental aspects of mobility. The results will represent tangible evidence to clarify which aspects of mobility may be affected by age-related changes to cortical function.

**Neural and Behavioral Aspects of Executive Function, Working Memory, and Language**

The neural networks involved in higher-order cognitive control are the focus of ongoing research and the source of unanswered questions in many scientific fields. Of
particular interest to the control of functional behavior are those cognitive and motor processes that might overlap in mobility. Three such cognitive domains are identified in the existing literature: executive function (EF), working memory (WM), and language.

**Executive Function**

EF is the term for higher order cognitive processes and sub-processes that control a multitude of complex and goal-oriented human behaviors in novel circumstances [98]. Often considered an exclusively ‘frontal lobe’ function, in reality EF is metabolically expensive operating at a high energy cost throughout the brain. EF behavior involves complex networks with functional pathways to and from anterior, posterior and parietal parts of the brain [98,99]. Thus, EF is thought to only ‘turn on’ in novel environments when automatic responses are no longer sufficient [98,99]. Some argue that isolating the anatomical structures involved in EF is less salient than understanding the emergent behavior of this complex system that activates the majority of the cerebral cortex [99,100]. In agreement, the proposed dissertation will use behaviorally based approached to evaluate tangible consequences that emerge from age-related cognitive changes.

Researchers across fields have attempted to define and operationalize EF anatomically [98,99], theoretically [100], and experimentally [101]. Although no clear consensus of EF and its specific terminology has been adopted across fields, scientists generally agree that EF controls three basic components of cognition; set switching (shifting), set maintenance (updating, task-setting), and inhibition (inhibiting prepotent responses/irrelevant information) [98,101–103]. Each is considered to be related to several other identified sub-processes such as attention [99], metacognition and
integration [98], action initiation [101], and processing speed [101]. Inherent issues in EF research confound accurate representations of such cognitive processes. For instance, EF is intricately related to speech production and many other layers of cognitive processes, making ‘pure’ EF difficult to test [101].

Because EF is accepted as the controller of novel and adaptive behavior, it is no surprise that walking requires EF processes. Research suggests that EF behaviors suffer as a result of the aging process [104]. This assumption is further supported by greater difficulty performing complex tasks in older adults with cognitive impairment [105] and worse performance on tests of EF [22,23]. EF scores have also been related to balance ability [106], gait speed [22,65], gait variability [107], and instrumental activities of daily living [65,108].

**Attention**

Attention is critical sub-process in EF implicated in a dual task. More specifically, divided attention is considered the independent variable in dual-task investigations. Most tasks common to daily life involve the division of attention. Examples of these behaviors range from easy, such as ensuring your walking path is safe or planning a route, to more complex dual tasks such as simultaneously walking and talking to a friend or navigating an unfamiliar setting.

Attention is studied across a wide variety of disciplines and is considered a domain of EF [102,109]. Kahneman (1973) postulates that dual-task declines are as a result of limited attentional capacity to devote to each task or potentially the time delays of the involved cortical networks [110]. The level of cognitive or motor complexity appears to affect attentional control, resulting in more impaired performance [35,111].
However, what aspects of cognitive or motor tasks drive the management of divided attention in mobility is unclear. Despite the relevance of attentional control, the manner individuals utilize cognitive resources to perform two simultaneous tasks remains appears to be sensitive to the type of cognitive or motor task (reviewed below). The effect of cognitive and task type is explored in dual-task experiments and is detailed later.

The purpose of this dissertation is to determine if higher-level cognitive processes coordinate dynamic balance control across the lifespan. Part of this question will be answered with a commonly implemented cognitive task called the “Alphabet Recitation Test”, which requires subjects name alternating letters of the alphabet while walking. This test is regularly used in dual-task studies and is significantly associated with disability and fall-risk [2,112]. However, some argue that the task is relatively ‘automatic’, and thus may not reflect the role of EF in typical tasks that involve cognitive and motor resources. The proposed dissertation will use Alphabet Recitation to examine the effect of instability on a cognitive task that depends on simple aspects of EF such as processing speed and divided attention. If compromised balance interferes with the performance of the Recitation task, we can deduce that processing speed may share higher-level cognitive resources with dynamic locomotor balance. Comparing performance with two other cognitive tasks focused on working memory and language skills will further clarify which aspects of cognition control balance in aging.

**Working Memory**

Working memory (WM) is often considered a facet of EF and is fundamental to two of the three primary domains of EF: set-maintenance and inhibition of prepotent
responses [113–115]. Indeed, investigators have proposed that working memory should instead be termed ‘executive attention’ because of the substantial overlap between cortical regions and networks in EF and working memory [113]. Control of WM has been identified in the frontal lobe, particularly the dorsolateral prefrontal cortex [114,116] that project onto the hippocampus and limbic systems in the central cortex [58]. Kane and Engle identified the dorsolateral prefrontal cortex as most active in set-maintenance in WM tasks [113]. This result led the investigators to propose WM be considered as ‘executive attention’, an aspect of EF responsible for short-term retention of information [113].

Behaviorally, Baddeley defines WM as the storage and manipulation of information in a task, where the simple storage of information is considered short-term memory. The ‘multicomponent’ theory splits WM into three basic parts; 1) central executive (CE) 2) visuospatial sketchpad and 3) phonological loop, considered independent and fluid systems that require only temporary activation [117]. The CE is described by Baddeley as a ‘homunculus’ that can direct attention as needed, store information, and make decisions, but the hypothesized complexity of the CE is difficult to explore empirically. The CE is thought to only engage in new behaviors, rather than well-practiced tasks[115]. The visual sketch-pad controls the storage and retention of visual stimulus only. Easier-to-implement phonological experiments show the phonological loop is related to vocabulary and language learning in long-term memory [118]. Baddeley’s model is mediated by an ‘episodic buffer’ assumed to hold integrated (bound multi-sensory information) representations in memory and bi-directionally
interact with crystallized systems of visual semantics and language (permanent knowledge) [117].

In older adults, memory and WM are colloquially thought of as ‘the first thing to go’ in aging [95]. Note that the frontal and central cortex areas identified in EF and WM [113] are particularly susceptible to losses in cortical volume with age [63,95]. WM also plays a part in many daily functional tasks, potentially leading to difficulties observed in older adults during tasks taxing WM and locomotion (i.e. reciting an address while walking to the mailbox). The involvement of WM in so many cognitive processes confound the understanding of the association of WM with older adult mobility and performance of activities of daily living. The proposed project aims to examine the role of WM in dynamic balance control by administering a challenging memory task while walking in stable and unstable constraints. To accomplish this efficiently, WM will be evaluated by applying two overarching principles from Baddeley’s behavioral models.

First, WM requires storying and using information. That is, measuring the amount of information one can retain and manipulate is the truest measure of WM capacity. Popular methods include the Reading Span and Digit Span test, each are validated assessments of WM capacity. Thus, this dissertation will use the Digit Span Back (DSB), requiring that the participant listen to a list of numbers, and repeat them back in the reverse order [119]. While many experiments perform the Digit Span Forward (DSF), the DSB challenges the manipulation of information by reciting number lists backward [119]. Many young and middle-aged adults would find the DSF to be too simple, but the advantage to the DSB is that it is challenging for everyone.
Second, visual and auditory stimulus involve different WM processes. Baddeley’s principles of WM support that auditory and visual stimuli rely on different WM storage systems. Currently, the presentation of visual stimuli while walking is difficult to test behaviorally. Given the wide age range of subjects expected to participate in this experiment, interrupting sensory feedback with a visually-based cognitive task could disrupt dual-task performance irrespective of cognitive demand. Moreover, presenting auditory stimulus in this task can help strengthen the interpretation and replicability of the results by further focusing on the capacity of phonological loop of WM. To mitigate confounds from age-related changes to sensory systems, the cognitive tests are carefully administered with clear auditory and visual feedback from an experienced test administrator, similar to real life. If the results from this experiment show stability negatively effects WM capacity, then we can deduce that WM shares resources with dynamic locomotor balance control.

**Language**

Communication is the foundation of understanding and participating in the world around us, enabling us to develop inner narratives and interact with people and the environment. Similar to locomotion, speaking is a goal-oriented skill that is learned early in development and seems simple but is actually incredibly complex. The relationship between language and movement is often alluded to in existing dual-task research, but most paradigms do not directly investigate the influence of language formulation and comprehension on dual-task performance. Language formulation and comprehension cannot be localized to any specific region of the brain because of the complexity of communication skills [120]. Further, both aspects of communication overlap with several
cognitive domains such as EF, attention, and WM [100,118]. Importantly, recent evidence supports that language systems overlap with those involved in movement control [121–123].

Speech production and understanding rely on EF and WM networks to constantly store nodes, syllables and semantics during speech production [124], as well as word meanings, semantics, and sentence structure in comprehension [100,124–126]. These aspects of language depend on EF and WM to maintain what has already been said, switch between planned speech and maintaining Gricean maxims (i.e. taking turns, listening, and addressing relevant information), to converse and inhibiting prepotent or irrelevant responses [101,124,127]. The critical contribution of WM is notable in those with frontal lobe dysfunctions, such as Parkinson’s disease, as they experience difficulty with planning sentences [128–130]. Clinical populations (e.g. stroke survivors and TBI patients) also demonstrate difficulty with word recognition, semantic retrieval, and regulating behavior to maintain conversation [125,131]. This evidence has supported that challenges to planning or understanding speech are related to poor EF or WM.

Language formulation is organized hierarchically, beginning with lexical retrieval and ending with articulation [120,126]. The formulation of speech begins with conceptualization. First, one must choose what words to speak in a semantically and syntactically appropriate way, described by the Lemma model [126]. Within milliseconds, lemma nodes are activated and stored in WM to construct logical and semantically appropriate sentences for communication [124–126]. Although several models have been proposed to define the neurophysiology of speech, the most widely accepted and empirically supported is the Directions Into Velocities of Articulators
The DIVA model focuses on the control of the articulators at the syllable and lower motor levels. The DIVA model is broken into predictive and reactive systems. Of particular interest in the DIVA model are the parts of the brain involved: the premotor cortex, cerebellum, and motor cortex. Note that each of these areas of the brain is implicated in motor control and EF [84,98]. For these reasons, older adults have more difficulty retrieving and verbalizing words than their younger peers [124,133,134].

The intricate link between experience and language formulation supports a bidirectional relationship between movement and language. This dissertation proposes to apply a neuromechanical perspective to provide unique and useful insight into the influence of cognitive-linguistics on cognitive-motor control. By administering a lexical retrieval task (Verbal Fluency) during walking, we can determine the effects of dynamic locomotor balance on language formulation. The Verbal Fluency task involves generating words from a randomly assigned word category (i.e. animals, words that begin with ‘F’, etc.). Verbal Fluency prompts challenge two types of word retrieval: semantic (associative) and phonemic (structural). If unstable walking impairs word retrieval, the evidence would suggest that communication ability is intertwined with mobility for older adults. The implications of this association could explain difficulty during daily tasks like walking while talking, which can predict mobility impairment and falls for older adults [5,7,21].

**Age-Related Changes to Neural and Behavioral Control of Dual Tasks**

A common paradigm to understand the relationship between cognition and gait control utilizes a dual-task methodology, quantifying performance in two goal-oriented tasks when executed alone and together. The recent interest in dual-task behavior has
uncovered the salience of dual tasks in understanding cognitive-motor control across the lifespan. The degradation of one or both tasks when performed concurrently is termed ‘cognitive-motor interference’. Generally, greater cognitive-motor interference in dual tasks has been linked to frailty [2], fall-risk [5,7,135], mobility-disability [2], and cognitive impairment [6]. Given this evidence and the ease of administering a dual task, evaluating cognitive-motor interference proves to be an appealing approach to understand locomotor control in aging. Exploration of underlying neurological systems in dual-task motor control is important to interpret the mechanisms related to diverse existing evidence. The following will briefly summarize the role of higher-level cognitive processes related to neurological contributions to dual-task walking in aging.

**Evidence from Brain Imaging**

The neural foundations and behavioral theories of dual-task performance across the lifespan have been recently explored through advances in brain imaging technology. Although these methods have inherent limitations, this technology has shed light on previously elusive information that is considered to compare well between real and imagined gait [136]. A systematic review by Hamacher et al. (2015) described results from imagined walking in magnetic resonance imaging (MRI) and positronic emissions tests (PET) and real walking with electroencephalography (EEG) and functional Near-Infrared Spectroscopy (fNIRS) [79]. For young adults, cortical activation increased as mobility tasks became more complex: standing to walking [137], double support to single support of a stride [19], and from single to dual tasks [79,138]. As compared to young adults, older adults showed increased activation in the dorsolateral prefrontal cortex, the pre-motor area, supplementary motor area, posterior parietal cortex, the
insula, and the anterior cingulate cortex in imagined [139] and real dual-tasks [20,139,140,140]

Clark et al. (2014, 2015) measured brain activation during increasingly difficult mobility tasks using functional near infra-red spectroscopy (fNIRS) in young and older adults [20,141]. Clark reported changes to a measure called total oxygen index (TOI) that determines oxygen intake by the brain, considered an index of brain activation while performing overground walking in several complex tasks. Clark observed greater activation of the frontal lobes in older adults during many tasks as compared to young adults. Of particular relevance was the finding that older adults continually increase activation more during a dual task than obstacle crossing, wearing a weighted vest, and walking in dim lighting [20]. This suggests that in older adults, dual tasks are unique in that they continually challenge CNS resources. This conclusion has gained further credence by reports from Holtzer et al., that demonstrate increased activation of the prefrontal cortex in older adults during a dual task with verbalization (Alphabet Recitation) [139,142–144].

Another compelling piece of evidence in the neural control of dual tasks in aging was recently reported by Malcom who measured cortical activity in young and old during dual tasks in treadmill walking with EEG [140]. Event related potentials showed an increase in the N2 component, an indicator of inhibition, as well as P3, which is typically detected in decision-making. Importantly, Malcom found older adults increased activity in frontal areas during dual tasks in older adults and this activation remained in the frontal areas of the brain throughout the trial. In contrast, young adults showed a shift in activation to more posterior aspects of the brain across time [140]. Evidence of greater
frontal activation and indications of increased inhibition processes during dual tasks supports that older adults use higher-order cortical processes to perform complex walking tasks. Overall, the literature demonstrates an increase in older adult brain activation during movement, especially as the walking task is more difficult.

**Theoretical implications.** Considering the evidence leads to several potential theories of the influence of neurological processes on dual-task behavior in older adults. Clark et al., proposes the ‘compensation theory’ [141]. The compensation theory is based on a loss of sensory-motor skills with age that leads to more reliance on high-level resources, increasing overall brain activation. Along similar lines, Clark adds the ‘de-differentiation’ theory to explain increased activation, suggesting that older adults lose the ability to selectively recruit higher-level resources needed for dual tasks. Increased brain activation to perform movement is a consistent finding across aging studies, suggesting that older adults overcome age-related changes to the neurological system with broader cortical activation. Evidence of greater cortical involvement in locomotion supports that higher-level cortical networks may be involved in mobility and fundamental dynamic balance with age. Using a dual-task methodology, the results of this study will reveal the behavioral implications of increased involvement of the aging brain in dynamic balance control during walking.

**Age-Related Consequences of Dual Tasks on Mobility**

The negative consequences of dual tasks on walking in aging individuals were first noted in a report from Lundin-Olssen (1997) who observed frail older adults arrest walking to speak [5]. Those that stopped walking to talk were found to have significantly
higher odds of falling within the next six months. Lundin-Olssen termed this the 'Stops Walking While Talking' test and promoted this evaluation as a predictor of fall-risk in aging [5]. Because fall-risk is a major cause of disability in later life, this sparked an extensive line of research that pursued the link between dual tasks and falls in older adults [7,32,135,145,146].

The consequences of aging on gait lead to slower gait speeds and this is typically exaggerated when concomitantly performing a dual task [6,7,21,38,147–151]. Mixed methods and results confound a clear understanding of the relative change to gait parameters, what drives these changes across ages, and the influence of various motor and cognitive tasks on dual-task performance.

A remarkably consistent consequence of a dual task on gait performances in older adults is an overall slowing of gait speed [6,7,21,38,147–151] that worsens with functional [21] and cognitive impairment [6,38,146]. Gait speed is simple to measure and represents a summation of spatial and temporal parameters. Thus, speed is most commonly reported in the literature [for review and meta-analyses see [7,152,153]]. Because of the summative nature of walking speed, the relative change to various aspects (i.e. spatial, temporal, or rhythmic) of dual-task gait remains unclear. However, a range of contradictory results sometimes show no change to gait speed with a dual task [9,37,154–156] and others have shown that the variability of gait speed increases [147,150,151,157] and decreases [158–160].

More refined biomechanical measures of have added important information about the effects of a dual task on walking in older adults. Those that have quantified spatial and temporal gait outcomes show individuals who have greatest difficulty with a
dual task stabilize the COM in relation to the BOS through wider [32] and shorter steps [7] and longer time in double support [107]. A study by Ayers et al., (2013) shows that shortening step length and slowing cadence during walking while talking predicts falls in a large sample (n=646) of older adults [7]. Martin et al., (2013) reported corroborating evidence that double support and step time during dual tasks are fall-predictors for older adults [107], and other studies suggest that changes in step width can best predict falls [32]. Asai (2013) reports that while gait speed was unaffected by a dual task, young adults were able to adjust their trunk movements to meet dual-task demands, but older adults did not [156]. De Hoon et al., (2003) measured trunk kinematics and discovered that older individuals that are worse at walking while talking demonstrate greater pitch and roll angles of the trunk [161].

The question remains as to which task characteristics precipitate change in gait performances under cognitive and motor demands. Differences in previous results may be related to treadmill versus overground walking [162,163] because some studies observe walking at fixed treadmill speeds [140,154,155,159,160,164] and others measure walking overground [32,107,146,149,157,165]. Alternatively, performance declines to dual-task walking may be a result of the cognitive processes involved in locomotor coordination, or the accompanying increase to balance demands in overground gait.

Unfortunately, studies rarely evaluate changes to gait parameters with a cognitive load across the lifespan. Few studies include middle-aged participants and existing studies group participants arbitrarily by age. Hollman et al. evaluated dual-task walking in three age groups (old, middle-aged, and young) reporting decreased speed
and increased stride time variability with a dual task in all groups, and the effect was greatest in older adults [149,157]. Because all age groups demonstrated a similar reduction in gait speed, but only older adults increased variability, Hollman suggests that variability in step timing may be an additional consequence of dual-task walking for older adults [157]. However, the group comparisons the investigators used overlooks the progressive nature of the aging process. Thus, the rate of age-related changes to gait under a cognitive load remains unknown. To address this limitation, the proposed dissertation will avoid disparate age comparisons by including participants at a range of ages and analyses that examine the gradual aging process.

To better understand the cognitive involvement in the control of stepping and balance while walking across the lifespan, this dissertation will examine the specific influence of increased balance demands by keeping speed constant while manipulated COM stability treadmill walking. Therefore, changes to gait and cognitive performances will result from controlling the COM, rather than speed-related adjustments. Additionally, a range of aged participants will reveal important information about the progression of age-related changes to cognitive-motor systems. This approach will identify the biomechanical contributors to cognitive-motor interference in dual-task walking across the lifespan.

**Methodological Considerations in Dual-Task Studies**

Studies that progressively increase motor demands report declines in dual-task performance while walking fast [8,23,32] crossing an obstruction to the walkway [37,76,166,167], narrowing the path [168] or dimming the lighting [20] Generally, the more complex the locomotor task, the more drastic the changes to performance.
Additionally, the more complex the cognitive task, the more costs to performance older adults experience. Each method has limitations and implications that were considered in the development this dissertation experiment and are reviewed below.

**Motor task type**

In older adults, when motor task complexity increased from cycling to standing and walking, performance was reported to progressively decline [10]. Yogev-Seligmann et al (2013) suggested that cycling was not as cognitively demanding as walking, but more demanding that standing. The author asserts that the results indicate that rhythmicity (not balance) is the most challenging aspects of locomotor control [10]. However, several dual-task studies report improved performance in dual tasks while stationary cycling [169,170] particularly in the acute-exercise literature [171]. A stationary cycle guides two fundamental aspects of motor control, balance and rhythmicity, by providing a stable support for the COM and guiding bilateral movements with coupled foot pedals. The benefits to dual-task performance was attributed to the cortical networks involved in supporting balance and stability, a proposal that aligns with previous studies [19,172]. The evidence suggests that differences in dynamic balance demands alone affect dual-task performance.

Perhaps because of the involvement of dynamic balance, and in agreement with Yogev-Seligmann et al., [173], walking does seem to lead to more impaired dual-task performance. Initial investigations of cognition and dynamic balance showed young adults experienced cognitive-motor interference from standing to walking, that was exaggerated in unstable phases of gait, evidenced by worse RT in single support (when only one foot is on the ground) [18]. Because cognitive tasks were more disrupted
during single support [18], when the COM moves outside of the BOS [11], this suggested that dynamic balance control involves high-level cognitive processes and is a factor in cognitive-motor interference. Recently, more direct support for this idea was provided by Gwin et al., who revealed increased activation of frontal and cortical regions during the single support phase of gait in young adults [19].

Kelly et al. [168] extended this evidence to a behavioral study of a gait targeting task that imposed an unstable walking pattern in young and older adults. The investigators quantified changes to dual-task performance during narrow walking verses usual walking [168]. The authors reported that a smaller BOS led to trade-offs between step accuracy and gait speed that was not apparent in usual walking [168]. This suggests that a narrow base of support (increased balance demands) could motivate older adults to sacrifice speed to focus on step accuracy to tightly control the dynamics of the BOS and COM. Simultaneously, this task involved a visual stimulus, a new goal, and a stepping target. Thus, this additional sensory input may have influenced the results more than only balance demand [168].

Existing evidence confirms that dual-task performance is highly affected by biomechanical and sensorimotor aspects of locomotor tasks [9,166,167]. However, the extant literature would be enriched by determining if the corresponding increase to fundamental dynamic balance demands leads to degrades performance in complex locomotion. To address this question, this study will hold the complexity of foot placement constant during treadmill walking and manipulate only dynamic balance demands during dual-task walking.
Cognitive task type

The type of cognitive task observed may also explain the differences in results between dual-task studies. The dual-task literature utilizes a variety of cognitive tasks which contribute to existing mixed results. Several studies have determined that the type of cognitive task implemented affects the magnitude of costs to gait performance [8,174,175].

The simplest cognitive task employed in dual-task studies is reaction time (RT), usually incongruent RT. Although the method of delivery varies from verbal responses to auditory cues or button pushing after a proprioceptive stimulus [9,18,41,166,176], RT tasks are assumed to challenge inhibition and processing speed. Unfortunately, these RT approaches utilize a discrete task, and even when randomly administered only intermittently draws cognitive resources away from locomotion. To efficiently measure the effects of dynamic balance control, a continuous cognitive task would better reveal the cognitive and motor response to stability throughout the stride cycle. Thus, the RT task was not included in this study in favor of other options.

Perceived to be more challenging than RT, arithmetic tasks require individuals to add or subtract by a designated number (a higher number is more difficult to manipulate). Unfortunately, without any other information regarding cognitive status, individuals that have better math skills would perform better on an arithmetic task, independent of cognitive load. Further limitations were proposed by Beauchet et al., who suggested that serial subtraction may impose a rhythmicity to the task that could influence coordination and walking parameters regardless of cognitive load [175]. Given the biomechanical focus of this experiment (and variability as an outcome measure), the limitations of the Arithmetic task were considered a poor fit for this study. Thus, as
common as the Arithmetic task is in dual-task experiments, this test was excluded from the proposed dissertation protocol.

Another popular task is Alphabet Recitation [165] which requires that the subject to recite alternating letters of the alphabet. While this task is continuous, unlike RT, reciting the well-practiced alphabet bears limitations. Alphabet Recitation is likely accomplished through a ‘subvocalization’ strategy; saying every other word quietly to oneself between responses. In lieu of significant cognitive effort, the nature of Alphabet Recitation promotes subvocalization that is guided by the harmony of the alphabet song (and ‘Twinkle Twinkle Little Star’). Further, the alphabet is well-learned and could be considered an ‘automatic’ response. Thus, Alphabet Recitation is likely a measure of processing speed rather than the higher-level cognitive processes involved in more challenging daily cognitive tasks [100]. Despite its limitations, the Alphabet Recitation test is included in this dissertation. The purpose of including the Alphabet Recitation task is to evaluate the influence of a simple EF test of processing speed on dynamic balance. Further, a major strength of this study is that the inclusion of this test will enable clear comparison with a previous study that examined the effects of body weight support on walking while performing the Alphabet Recitation task [44].

Finally, the Verbal Fluency or spontaneous speech tasks are considered the most challenging and ecologically valid [10,21]. Verbal Fluency assigns a word category to the subject and requires that they list as many words as possible from the assigned category, an EF and language task [177]. While this task is ideal because conversation is continuous, arrhythmic, and well-practiced, the performance of speech is overlaid with articulation and respiration. Those that investigate the influence of articulation alone find
that greater articulatory demands lead to greater costs in dual-task performance [178,179]. Thus, declines in dual-task performance with verbal responses may be a function of the motor coordination required to speak, irrespective of cognitive load. Because respiration is intimately related to speech production [180] and the energetics of walking [70], the evidenced relationship between motor control and the cardiovascular system [181] could drive changes to performance rather than language formulation demands.

Nevertheless, out of the existing commonly implemented concurrent tasks, language tasks are the most ecologically valid and similar to daily life. Further, the cognitive demand in language tasks is continuous, ideal for examining dynamic balance across strides. The study will address the interaction between the cardiovascular demands and speech production. The protocol includes regular measures of perceived exercise exertion and heart rate. Thus, if necessary, the results can be statistically corrected for the effects of respiratory demands.

In summary, the various motor and concurrent cognitive tasks observed in the literature affect dual-task performance differently. To address limitations of concurrent cognitive tasks, three types of task will be performed to compare the effects of walking instability on distinct cognitive processes. All tasks require a verbal response to an auditory stimulus to avoid changes in performance only because of age-related sensory declines. Additionally, each concurrent task is continuous to avoid intermittent interruptions and maintain a consistent cognitive load throughout the entire gait cycle. To control for performance changes because of speed, walking performance will be
evaluated on a fixed-speed treadmill. Therefore, speed will be fixed and changes to gait performance will be a result of manipulations to stability or cognitive demand.
CHAPTER 3
METHODS

Participants

To understand the pattern of age differences in cognitive-motor performance as walking balance demands changed, a young and older adult group of participants were recruited from the surrounding community. Using a protocol approved by the University of Florida Institutional Review Board, data collection for both Experiment 1 (Step Rx) and Experiment 2 (BWS) took place between December 2017 to June 2018. A random sample of volunteers were recruited via a 30-day newspaper advertisement in May 2018, social media digital flyers, flyers at local community centers, as well as senior recreation and residential communities. Stratified randomization was used to sort half of the participants to either Step Rx (Figure 3-1), or BWS (Figure 3-2) by age group, then gender.

![Figure 3-1. Protocol design for Experiment 1: Step Rx.](image)
All participants were able to walk unassisted on a treadmill and had vision and hearing corrected to normal. All participants’ native language was English. Participants were excluded if they had a history of non-medically stable cardiovascular disease, or the presence of a vestibular or neurological disorder that affected balance or walking ability. Participants were ineligible if they experienced any discomfort when walking. A total of 98 older and younger adults volunteered as participants for the study. Participants first provided informed consent approved by the IRB, and no monetary compensation was provided. Young adults earned extra credit in courses in exchange for their participation if approved by their instructor.

![Figure 3-2. Protocol design for Experiment 2: Body Weight Support.](image)

**Procedures**

**Walking Tasks**

Individual measurements of joint widths at the ankle, knee, elbow, wrist, hand, and shoulder were gathered, as well as weight and height, for each person. Light-weight reflective spheres were placed over bony landmarks on the head, torso and limbs in
accordance with the Vicon Plug in Gait model. Eight motion capture cameras recorded marker position at 120 Hz as the participant performed the experimental tasks. The participant was fitted with a heart rate monitor around the chest.

Prior to each condition the participant first determined their comfortable walking speed using an established method in the field [182]. The procedure required that the participant begin walking at 0.5 m/s, and the speed was gradually increased by increments of 0.02 m/s. The participant was instructed to tell the researcher when the speed is “faster than your comfortable walking speed”. The treadmill speed was subsequently decreased until the participant reported a speed at which they were walking slower than their comfortable pace. This process was repeated three times and averaged to determine a comfortable walking speed.

**Step Rx**

Each step width manipulation, or ‘prescription’, was cued with lightweight and tippable PVC pipes that acted as walking ‘guides’ intended to direct the width of stepping. The guide height was set just above the ankle and parallel along the length of the treadmill. The width of guides were adjusted for the morphology of each participant's pelvic width, based on previous methods used to scale step width manipulations [168]. In preferred stepping, the distance of the guides was placed at hip-width with reference to the distance between the left and right greater trochanters (prominent bony landmarks on the proximal femur). For narrow stepping, the guides were set at 80% of pelvic width, measured as the left and right anterior superior iliac spine (ASIS) of the pelvis. Step Rx conditions were administered in a randomized order. The participants walked at 90% of their comfortable walking speed in all walking conditions.
Half of participants were randomized to walk with (BW+S) and without partial body weight support (BW) provided by a walking harness. The harness was fitted to the participant and fixed to a ceiling-mounted trolley centered over a treadmill [Hacoma, Switzerland]. The walking conditions were administered in a random order: The BW served as a control condition that requires the participant walk with the harness, but with no body weight support. The BW+S condition required participants to walk while 50% of their body weight was alleviated [51,183] given previous studies of young adults that show changes to gait biomechanics indicative of reduced balance demands with 50% [183] to 60% [51] of body weight alleviated. The participants walked at their comfortable walking speed in all conditions.

Single-task walking performance was measured with motion capture technology while participants walked for two minutes in each condition without an additional cognitive task. After the single-task condition, the participant’s walking performance was quantified while performing three concurrent tasks in a random order; the DSB, Recitation, and Fluency, (described below). Instructions were administered while the treadmill was stopped and did not mention task prioritization. Heart rate was recorded after every task and participants provided a rating of perceived exercise exertion on the Borg scale from 6-20 [184].

Cognitive Tasks

Participants performed three cognitive tests presented in a random order to evaluate WM, EF, and language formulation (Figure 3-1 and Figure 3-2). Each cognitive test was performed while seated to quantify single-task performance, and during
walking to assess dual-task behavior. To mitigate the effects of learning, separate versions of each cognitive test were administered in each experimental condition. Specific stimuli changed, but each version maintained the same level of difficulty. Tests were administered verbally and in visual range of participants by the same trained researcher.

The Digit Span Back (DSB) test measures WM capacity and required the participant to listen to increasingly long lists of numbers and recite them backwards, in accordance with traditional methods [185]. The Alphabet Recitation (Recitation) task measures basic EF skills of set-switching, set-maintenance, and inhibition of prepotent responses [100], and is commonly used to assess cognitive-motor interference and identify fall risk [2,165]. Participants were assigned a random letter of the alphabet and instructed to recite the rest of the alphabet, alternating letters, until they reached the end of the alphabet. For each Recitation condition, the participant was assigned four random letters for a total of four separate Recitation trials. The trial was measured from the moment the prompt (letter of the alphabet) was delivered, until the participant’s last response. Measures of cognitive and motor performance in the four trials were combined to best represent Recitation performance. Verbal Fluency (Fluency) quantified complex EF and language skills [186,187]. The Fluency task required that the participant recite as many words as possible in a minute from an assigned word category (e.g., a semantic prompt such as ‘animals in the zoo’ or phonemic prompt such as ‘words beginning with a certain letter’). The participant was presented two separate word categories, semantic (associative) and one phonemic (structural), for a total of two minutes of Fluency performance in each walking condition. Cognitive and motor
performance outcomes from both categories were combined to characterize Fluency performance.

**Single-task cognitive performance.** The order of single task administration was randomized and counterbalanced with the performance of dual-task walking. To assess single-task cognitive performance, the three cognitive tests (detailed above) were administered while seated comfortably. In a quiet and well-lit room, the participant completed the DSB, Recitation, and Fluency (described above) in a random order.

**Functional Evaluations and Clinical Cognitive Assessments**

Both experimental sessions concluded with a series of cognitive and functional measures. Three surveys, two tests of physical function, and three common clinical tests of cognition were administered.

**Surveys of physical activity and balance efficacy.** First, participants filled out validated surveys to quantify physical activity, balance confidence, and fall history. Participant completed the Community Healthy Activities Model Program for Seniors (CHAMPS) survey [188] to assess physical activity levels and reported the time spent performing various activities of both leisure and exercise types. Participants then completed the Activities Balance Confidence scale (ABC scale) and rated their level of comfort performing various daily tasks (i.e. reaching for an item above you while standing on a stool) on a scale from 10-100 [189]. Additionally, the participant reported their history of falls in the previous six months. In this study, a fall is defined as an accidental change in position causing one to land on a lower level. This excluded near
falls, incidents occurring as a result of overwhelming external force (hit by a car) or a fall as a result of a loss of consciousness [55]. If the individual has fallen, they enumerated how often, if the fall led to serious injury, and under what circumstance the fall occurred [55].

**Physical function tests.** The researchers then administered two common validated measures of physical function: 1) Dynamic Gait Index (DGI), and 2) Berg Balance Scale (BBS). The DGI is a 20-item evaluation that scores one’s ability to walk, walk while turning one’s head, walk and turn to stop suddenly, walk while crossing and avoiding obstacles in the path, and climb stairs [190]. The BBS is a 14-item evaluation that involves sit to stand ability, standing balance (eyes open and eyes closed), turning 360 degrees, touching feet onto a step, standing with feet in a line and standing on one leg [191]. These evaluations quantified balance and walking ability.

**Clinical assessments of cognition.** Finally, to better interpret the results, baseline cognitive ability was quantified. The participants completed three validated neuropsychological measures of global cognition previously linked to dual-task costs in older adult gait performance [6,23,192,193]: 1) Montreal Cognitive Assessment, 2) Stroop test, and 3) Trail Making Test (TMT).

The Montreal Cognitive Assessment is a measure of EF sensitive to mild cognitive impairment and executive dysfunction [194]. The Stroop tasks are common clinical measures of inhibition and WM related to visual processes, important aspects of many cognitive behaviors [195,196]. The first part of the Stroop task, Stroop
Congruent, presented five columns of 20 rows of XXX’s, printed in three colors (blue, green, red). The participant was required to name the color of each row of X’s as quickly as possible in 45 seconds. The Stroop Incongruent presented the participant five columns and 20 rows of color-words (blue, green, red) printed in incongruent colored ink (e.g. the word ‘blue’ was printed in green ink). Participants were challenged to name color of the ink, rather than the color-word itself. The TMT is a two-part paper and pen test uniquely evaluating visuo-motor control, inhibition, and WM related to letters and numbers. Without picking the pen up from the page, the participant draws a line connecting a series of numbers (Trails A) as well as connect letters to numbers in order (Trails B) without picking up their pen or erasing any marks [197,198].

**Data Analysis**

The change in performance in biomechanical and concurrent task outcomes (dual-task costs) were computed as: \( DTC = \left( \frac{\text{dual task} - \text{single task}}{\text{single task}} \right) \times 100 \), to represent the magnitude and direction of the change from single to dual tasks, or cognitive-motor interference.

**Walking Performance**

A customized biomechanical model was used to calculate gait parameters. Trajectories were filtered with a 6Hz fourth order low-pass Butterworth filter and a custom Matlab (v 2.3) script calculated gait cycle parameters: step length, step width, and the variability of each (represented as standard deviation). Distance parameters were determined with reference to the toe marker that is placed on the base of the
second meta-tarsal. Step length measures the horizontal distance in the sagittal plane between one heel strike to heel strike on the opposite foot. Step width was calculated as the horizontal distance between the feet in the frontal plane between one heel strike and the heel strike of the following foot. Step outcomes for the right and left limb were averaged for each step.

Cognitive Performance

The changes from single to dual-task performance for each outcome (DTC) were computed. For ease of interpretation, the costs to cognitive outcomes were calculated so that a higher score is indicative of better cognitive ability, such that a higher value corresponds to better cognitive task performance in dual-task walking. In harmony with traditional measures of DSB performance, the score was quantified as the number of correct responses for digit recall prompts. For DSB, higher scores of correct responses indicate a greater WM capacity.

For Recitation and Fluency, a measure of accuracy \( \frac{Correct\ Response\ Ratio}{\text{Correct\ #\ Responses}} \) and time \( \frac{Response\ Rate}{\text{Total\ #\ Responses}} \) were calculated for Recitation and Fluency performance. A higher correct response ratio indicates more correct responses per trial and a higher response rate reflects faster responses.

To use as potential covariates, three measures of cognitive acumen were calculated. First, a composite ‘speed’ factor was calculated by averaging the standardized values of Stroop Congruent and Trails A performance. Second, executive visual processes, inhibition, set-maintenance, and set-switching were represented with scores for the Stroop \( \Delta\text{Stroop} = \text{Stroop congruent-Stroop incongruent} \) and TMT.
(\(\Delta TMT = \text{Trails A-Trails B}\)) [199,200]. Third, a composite ‘interference’ factor was calculated by averaging standardized values (z-scores) representing the percent change for \(\Delta\) Stroop and \(\Delta TMT\).

Measures of physical function, balance confidence, and cognitive acumen were quantified with their respective validated methods.

**Statistical Analyses**

An a priori significance level of \(\alpha < 0.05\) was applied to each analysis for both experiments. Each experimental sample and all analyses were treated independently, and the findings from Step Rx and BWS were not compared. First, all outcomes were examined to eliminate outliers and ensure normal distributions across variables. Then, independent t-tests determined differences in demographic variables between groups.

**Step Rx**

For Step Rx, to understand the interactions between age, cognition, and walking stability, a \(2 \times 2 \times 3\) (Age Group: Old, Young; Walk: Preferred, Narrow; Cognitive Task: DSB, Recitation, Fluency) repeated measures MANOVA compared performance changes (DTC) to step length and step length variability. Three separate analyses compared cognitive performance DTC (DSB: number correct, Recitation and Fluency: correct response ratio, response rate) in a \(2 \times 2\) (Age Group: Young, Old; Walk: Preferred, Narrow) repeated measure (RM) MANOVA. Participant characteristics (Table 4-1) were included in as covariates in the analyses as needed.
Body Weight Support

In BWS, $2 \times 2 \times 3$ (Age Group: Old, Young; Walk: BW, BW+S; Cognitive Task: Recitation, Fluency, DSB) RM MANOVA compared the DTC to step width and step width variability from single to dual tasks in each walking condition [201]. Three separate $2 \times 2$ (Age Group: Young, Old; Walk: BW, BW+S) RM MANOVA compared the cognitive performance DTC for each of the three cognitive tests (DSB: number correct, Recitation and Fluency: correct response ratio, response rate) [201,202]. When appropriate, post hoc comparisons were corrected with Bonferroni adjustments. Participant characteristics were included (Table 4-3) in as covariates in the analyses as needed.
CHAPTER 4
RESULTS

A total of 82 participants were eligible for inclusion in the analysis. The 17 excluded participants were removed from the analysis for the following reasons. In Step Rx, six participants were excluded after pilot testing because of protocol changes. One participant had a BMI over 30, preventing a true measurement of hip and pelvic width and accurate placement of the walking guide. Technical difficulty led to the exclusion of an additional young subject. For BWS, four participants were excluded due to changes in the protocol. Two participants completed the experiment with less than 50% of their body weight given equipment limitations in accommodating their height and morphology. Equipment failure (power outage and computer malfunction) led to missing data in three walking trials for two participants (both single-task walking trials for one participant and BW DSB for another) therefore these two participants were ineligible for the walking analyses. Additionally, one participant was excluded because they self-selected a speed that was too fast and they surpassed the walk/run transition in the second walking condition. After all exclusions were accounted for, 22 young and 20 older adults were included in Step Rx, and 20 young and 20 older were included in BWS (Table 4-1, Table 4-3).

Both experiments were designed to understand the overarching aim of the dissertation, to identify age-differences to the influence of fundamental aspects of locomotor balance of cognitive-motor interference. However, each experiment and their respective samples were treated independently and analyzed separately.
Step Rx

Twenty-two young (15 women, age 22 ± 2 years) and 20 older adults (12 women, age 71 ± 8 years) completed the Step Rx protocol (Table 4-1). Two older adults self-selected a comfortable walking speed that was not sustainable in the experiment, the walking speed for these participants was 80% of their self-selected comfortable pace. High variability of DTC results was expected; thus, examination of the results confirmed no extreme outliers were included, defined here as values ≥ three standard deviations of the mean.

Participant Demographics

Detailed in Table 4-1, independent t-tests were Bonferroni corrected for multiple comparisons (significance set at $p = 0.0025$) and showed that there were no significant differences between groups in morphological measures of height and body mass. Additionally, preferred walking speed was not different between groups, ensuring that differences in gait speed could not bias comparisons of gait outcomes. Further, most measures of physical function showed no age differences, but lower scores on the BBS forward lean, a clinical measure of static balance, suggested older adults exhibit difficulty in some aspects of balance. Several measures of cognitive function were significantly worse in older adults, the MoCA, Stroop Incongruent, $\Delta$ Stroop, TMT Part A and TMT Part B, but no other significant age-differences were detected in this sample. Overall, the demographic measures showed that few age-differences were present in this sample (Table 4-1).
Table 4-1. Sample demographics and group differences between young and older adults in the Step Rx experiment

<table>
<thead>
<tr>
<th></th>
<th>Young</th>
<th>Old</th>
<th>p-values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total N</td>
<td>22</td>
<td>20</td>
<td>--</td>
</tr>
<tr>
<td>Gender (Women/Men)</td>
<td>15/7</td>
<td>12/8</td>
<td></td>
</tr>
<tr>
<td>Age (years)</td>
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<td>71.00 (8.0)</td>
<td>0.000*</td>
</tr>
<tr>
<td>Height (meters)</td>
<td>1.75 (0.11)</td>
<td>1.69 (0.17)</td>
<td>0.141</td>
</tr>
<tr>
<td>Body mass (kilograms)</td>
<td>73.78 (16.76)</td>
<td>72.60 (11.77)</td>
<td>0.796</td>
</tr>
<tr>
<td>Preferred Speed (meters/second)</td>
<td>0.84 (0.18)</td>
<td>0.84 (0.17)</td>
<td>0.991</td>
</tr>
<tr>
<td>Education (years)</td>
<td>14.64 (1.7)</td>
<td>17.25 (2.3)</td>
<td>0.000*</td>
</tr>
<tr>
<td>Montreal Cognitive Assessment Score</td>
<td>28.18 (1.97)</td>
<td>25.25 (3.3)</td>
<td>0.001*</td>
</tr>
<tr>
<td>Stroop Congruent Score</td>
<td>81.37 (12.97)</td>
<td>69.42 (14.49)</td>
<td>0.011</td>
</tr>
<tr>
<td>Stroop Incongruent Score</td>
<td>62.42 (10.38)</td>
<td>38.58 (13.21)</td>
<td>0.000*</td>
</tr>
<tr>
<td>Δ Stroop Score</td>
<td>19.84 (9.31)</td>
<td>30.84 (6.87)</td>
<td>0.000*</td>
</tr>
<tr>
<td>Trail Making Test Part A (seconds)</td>
<td>21.11 (6.30)</td>
<td>33.08 (10.33)</td>
<td>0.000*</td>
</tr>
<tr>
<td>Trail Making Test Part B (seconds)</td>
<td>56.13 (34.05)</td>
<td>89.21 (36.43)</td>
<td>0.005</td>
</tr>
<tr>
<td>Δ Trail Making Test (seconds)</td>
<td>35.02 (30.50)</td>
<td>56.13 (34.08)</td>
<td>0.046</td>
</tr>
<tr>
<td>Number of Falls in Previous Six Months</td>
<td>0.36 (0.85)</td>
<td>0.45 (0.60)</td>
<td>0.708</td>
</tr>
<tr>
<td>Activities Balance Confidence (% confidence)</td>
<td>94.60 (6.11)</td>
<td>92.15 (6.66)</td>
<td>0.227</td>
</tr>
<tr>
<td>CHAMPS (hours/week)</td>
<td>14.81 (8.72)</td>
<td>12.08 (4.99)</td>
<td>0.236</td>
</tr>
<tr>
<td>Dynamic Gait Index Score</td>
<td>23.86 (0.35)</td>
<td>22.75 (1.74)</td>
<td>0.006</td>
</tr>
<tr>
<td>Dynamic Gait Index stair climb (seconds)</td>
<td>12.04 (2.53)</td>
<td>13.45 (2.58)</td>
<td>0.084</td>
</tr>
<tr>
<td>Berg Balance Scale Score</td>
<td>55.82 (0.66)</td>
<td>53.55 (3.95)</td>
<td>0.011</td>
</tr>
<tr>
<td>Berg Balance Scale Forward Lean (meters)</td>
<td>0.44 (0.05)</td>
<td>0.31 (0.07)</td>
<td>0.000*</td>
</tr>
</tbody>
</table>

Note: Values presented as mean (standard deviation) CHAMPS: Community Healthy Activities Model Program for Seniors, p-values represent between group differences, significant p < 0.0025 marked by asterisk (*)

Walking Performance Outcomes

Step length for the right and left legs were averaged to best represent walking performance. Evaluating characteristics of the distributions for step length variables showed that the means for each value were more skewed and kurtotic than the medians. For example, mean values for step length in the narrow single-task resulted in a skewness statistic of 0.722 (0.34) and a kurtosis statistic of -0.039 (0.67), where the median values resulted in a skewness of 0.703 (0.34) and a kurtosis of -0.015 (0.67). High variability in DTC were expected, thus, step length DTC were calculated from the
median value for step parameters from each trial to reduce deviations from the mean.

The means and standard deviations for each outcome are presented in Figure 4-1.

Figure 4-1. Results from Step Rx experiment. Walking performance costs (%DTC) to step length (top) step length variability (bottom) in young and older adults during the DSB, Recitation, and Fluency.

To better understand the outcomes, the DTC for step length and step length variability were analyzed in a 2×2×3 (Age Group: Young, Old; Walk: Preferred, Narrow; Cognitive Task: DSB, Recitation, Fluency) repeated measures (RM) MANOVA. Box’s M
value was 151.717, associated with a p-value of 0.037, above the recommended threshold of p < 0.005, confirming that the assumptions of homogeneity of variance and covariance (correlation of any two dependent variables within each group are similar) were met between the dependent variables [203]. Mauchly’s test of sphericity showed all variables met assumptions of sphericity (all p-values > 0.106) [204]. Levene’s test confirmed equal between-group variance (all p > 0.075) in all but two included variables: Preferred DSB Step Length \( F(1, 40) = 5.065, p = 0.030 \) and Narrow DSB Step Length Variability \( F(1, 40) = 5.058, p = 0.030 \). Therefore, Pillai’s trace statistic \( V \) was used to identify significant main effects because this technique is conservative, but overcomes limitations of unequal sample size and potential confounds of unequal variance [201].

The analysis showed that the main effect of walking condition significantly influenced DTC to walking performance \( V = 0.194, F (2, 39) = 4.699, p = 0.015 \), and a significant main effect of Cognitive Task \( V = 0.129, F (4,37) = 2.768, p = 0.029 \) and supported that walking stability and cognitive processes influenced step length and step length variability DTC. However, the main effect of Age Group was not significant \( V = 0.004, F (2,39) = 0.069, p = 0.934 \) and three-way interactions or two-way interactions were present. The omnibus effects validated the overall hypotheses of the experiment; complex cognitive processes influenced step-to-step locomotor balance control. Follow-up univariate ANOVA were used to understand the omnibus effect of walk and cognition on walking DTC.

The analysis revealed that the effect of walking stability on step length DTC was significant \( F(1,40) = 5.672, p = 0.022 \); participants took shorter steps with an added cognitive task in narrow than preferred walking (Figure 4-2). During narrow walking,
both groups showed a tendency to seek additional control of the COM in the A/P direction, thereby decreasing step length when under added cognitive demand. The findings suggest a concurrent cognitive task further destabilized the control of mediolateral balance.

Figure 4-2. Step length DTC represented as mean and standard deviations for each walking condition. Significant differences ($p < 0.05$) noted with an asterisk (*).

The effect of walking condition and cognitive task showed the opposite effects in step length variability DTC (Figure 4-3). The main effect of Cognitive Task on step length variability DTC was significant ($F(2,80) = 4.796$, $p = 0.011$). Bonferroni corrected pairwise comparisons confirmed that the step length variability DTC in Recitation was less than during Fluency ($p = 0.003$), but DTC did not differ between DSB and Fluency ($p = 0.311$) or DSB and Recitation ($p = 0.167$).
In sum, participants decreased step length and step length variability in narrow walking, demonstrating an attempt to increase stability to perform a concurrent cognitive task. While DSB and Fluency equally changed step variability (remaining the same or declining), Recitation decreased step length variability the most, and even led to motor facilitation in dual-task walking (increased step length and decreased step length variability).
Cognitive Performance

For comprehensive understanding of dual-task performance, the costs (% DTC) to cognitive outcome measures were compared between younger and older adults during the Step Rx experiments, the results are detailed in Table 4-2.

Table 4-2. Step Rx cognitive performance results (%DTC) for young and older adults in narrow and preferred walking

<table>
<thead>
<tr>
<th>Cognitive Condition</th>
<th>Performance Outcome</th>
<th>Preferred</th>
<th>Older</th>
<th>Narrow</th>
<th>Young</th>
<th>Older</th>
</tr>
</thead>
<tbody>
<tr>
<td>DSB</td>
<td>%DTC Capacity</td>
<td>11.56 (38.91)</td>
<td>6.48 (55.26)</td>
<td>-2.29 (30.94)</td>
<td>10.62 (42.25)</td>
<td></td>
</tr>
<tr>
<td>Recitation</td>
<td>%DTC Rate</td>
<td>-1.95 (36.31)</td>
<td>5.34 (29.22)</td>
<td>-5.61 (31.06)</td>
<td>0.46 (20.85)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>%DTC Ratio</td>
<td>-0.84 (9.75)</td>
<td>0.62 (8.59)</td>
<td>2.60 (11.54)</td>
<td>-0.40 (6.69)</td>
<td></td>
</tr>
<tr>
<td>Fluency</td>
<td>%DTC Rate</td>
<td>14.78 (27.63)</td>
<td>-4.71 (26.16)</td>
<td>9.89 (33.05)</td>
<td>-0.61 (23.89)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>%DTC Ratio</td>
<td>0.17 (8.13)</td>
<td>0.12 (4.52)</td>
<td>1.09 (7.08)</td>
<td>2.43 (4.21)</td>
<td></td>
</tr>
</tbody>
</table>

Note: Dual-task cost (%DTC) values presented as mean (standard deviation), Rate: Response rate (words per second), Rate: Correct response ratio (ratio of words to correct to total words)

A separate 2 × 2 (Age Group: Young, Old; Walk: Preferred, Narrow) RM ANOVA analyzed age differences in DSB performance costs from walking (Table 4-2). Box’s M value of 6.377 was associated with a p-value of 0.110, confirming equal covariance among the dependent variables. Bartlett’s test of sphericity showed the variables did not meet the assumptions of sphericity (p< 0.001), thus Greenhouse-Geisser adjustments were used to adjust degrees of freedom. Levene’s test verified equal between-group variance (p-values > 0.306). Performance costs to DSB showed no significant effect of Age Group \((F(1,40) = 0.196 \ p = 0.660)\), or Walk \((F(1,40) = 0.886 \ p = 0.352)\), and a trend toward a main effect of Walk × Age Group \((F(1,40) = 3.184, \ p = 0.082)\) (Table 4-2).

A separate 2 × 2 (Group: Young, Old; Walk: Preferred, Narrow) RM MANOVA analyzed DTC in response rate and correct response ratio during Recitation (Table 4-2).
Box’s M value of 17.68 was associated with a $p$-value of 0.107, confirming equal covariance among the dependent variables. Levene’s test verified equal between-group variance (all $p$-values > 0.305). No main effects of Age Group ($V = 0.020, F(2,39) = 0.383, p = 0.685$), or Walk ($V = 0.69, F(2,39) = 1.34, p = 0.273$) were detected on Recitation DTC, but the Age Group × Walk interaction showed a trend toward significant differences ($V = 0.146, F(2,39) = 2.84, p = 0.071$).

A $2 \times 2$ (Age Group: Young, Old; Walk: Preferred, Narrow) RM MANOVA analyzed age differences in Fluency performance using two outcomes, response rate (responses per second) DTC and correct response ratio (ratio of correct to incorrect responses) DTC. Box’s M value of 20.253 was associated with a $p$-value of 0.054, confirming equal covariance among the dependent variables. Bartlett’s test of sphericity showed that sphericity was violated ($p < 0.001$), thus univariate follow-ups were corrected with Greenhouse-Geisser degrees of freedom. Levene’s test confirmed the variance between groups was equal (all $p$-values > 0.130). Pillai’s trace statistic ($V$) was used to identify main effects because this technique accounts for unequal sample size. A significant omnibus main effect of Walk supported that altering mediolateral stability led to differences in performance DTC during Fluency performance ($V = 0.163, F(2,39) = 3.795, p = 0.031$), but the main effect of Age Group ($V = 0.090, F(2,39) = 1.934, p = 0.158$) and the Age Group × Walk interaction was not significant ($V = 0.071, F(2,39) = 1.710, p = 0.194$) (Table 4-2).

Follow-up univariate analyses explored the characteristics of the omnibus effect of walking task on measures of Fluency performance. The main effect of walking condition significantly impacted the Fluency correct response ratio DTC; $F(1,40) =$
7.654, \( p = 0.009 \)); correct response ratio DTC were higher while walking in a narrow step than preferred. To the contrary, the response rate was not significantly influenced by walking condition \( (F(1,40) = 0.009, p = 0.925) \). These findings uncovered surprising results that the correct response ratio was improved in unstable walking conditions; narrow walking led to more correct responses (Table 4-2).

The researchers were motivated to perform additional analyses that considered cognitive covariates that could contribute to the results. Two young participants did not complete the TMT; thus 20 young and 20 older adults were included in a 2 \( \times \) 2 (Walk: Preferred, Narrow; Age Group: Young, Old) RM ANCOVA. In DSB performance, using \( \Delta \) TMT as a covariate (evaluated at 45.577) removed the effect of executive interference related to numbers and letters from the analysis. An association between the \( \Delta \) TMT and the statistical model were confirmed with two-tailed bivariate correlations showing the \( \Delta \) TMT was significantly associated with Age Group \( (r^2 = 0.317, p = 0.046) \) and correct response ratio DTC in the preferred walking condition \( (r^2 = 0.348, p = 0.028) \) but was not associated with other dependent variables. Tests of between subjects effects confirmed that \( \Delta \) TMT was not different between groups \( (F(1, 37) = 0.002, p = 0.966) \), and the interaction of Age Group \( \times \) \( \Delta \) TMT was not significant \( (F(1, 37) = 2.935, p = 0.095) \) confirming the covariate was appropriate for the ANCOVA because it met the assumptions of homogeneity of regression [204,205]. The Box’s M value of 5.959 was associated with a \( p \)-value of 0.132, verifying equal covariance among the dependent variables. Bartlett’s test of sphericity showed the variables did not meet the assumptions of sphericity \( (p < 0.001) \), thus Greenhouse-Geisser adjustments were used to adjust the degrees of freedom. Levene’s test of equal variances showed equal error variance

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between groups ($p > 0.383$). The RM ANCOVA revealed a significant interaction of Age Group $\times$ Walk on DSB performance, $(F(1, 37) = 5.050, p = 0.031)$, but did not detect a main effect of Age Group $(F(1, 37) = 0.078, p = 0.781)$ or Walk $(F(1, 37) = 0.734, p = 0.397)$. When the variability from executive processes were considered, the interaction confirmed that young adults exhibited worse DSB performance in narrow walking than preferred walking, while older adults showed the same performance costs in both walking conditions. This implied that young adults may prioritize walking while walking in unstable situations, allowing DSB scores to decline. Older adults did not allow the DSB task to decline, reflecting possible cognitive prioritization, despite challenges to walking stability.

In Recitation, covariation by a composite interference factor significantly affected Recitation performance costs. The composite interference factor was determined for 19 young and 19 older adults because three young and one older participant did not complete the Stroop task. The association between executive interference measures and the statistical model were confirmed with two-tailed bivariate correlations that showed a strong association between interference and Age Group ($r^2 = 0.551, p = 0.000$), but no other dependent variables. Between-subjects tests confirmed that the interference factor (evaluated at -0.0197) was not different between groups for recitation rate $(F(1, 35) = 0.315, p = 0.578)$ or ratio $(F(1, 35) = 3.488, p = 0.070)$, and the interaction of Walk $\times$ Interference was not significant $(F(2, 34) = 2.616, p = 0.088)$, confirming the interference factor was appropriate to include as a covariate because the MANCOVA met the assumptions of homogeneity of regression [204,205]. Box’s M test statistic was 12.175 and associated with a $p$-value of 0.344, confirming equal
covariance among the dependent measures. Bartlett’s test of sphericity was significant ($p < 0.001$), thus univariate follow-ups used Greenhouse-Geisser adjusted degrees of freedom. Levene’s test of equal variance was not significant for any variable (all $p < 0.111$). Therefore, Hotelling’s trace (Hotelling’s $T^2$: $T$) statistic was used to identify main effects because the analysis met assumptions of equal variance and sample size was equal. When the effect of interference was removed from the analysis, the $2 \times 2$ (Age Group: Young, Old; Walk: Preferred, Narrow) RM MANCOVA showed the effect of Age Group was not significant ($T = 0.051, F(2,34) = 0.919, p = 0.409$), and a marginal, but not significant, main effect of Walk ($T = 0.193, F(2,34) = 3.287, p = 0.050$) on Recitation DTC. However, the interaction of Age Group $\times$ Walk was significant ($T = 0.260, F(2,34) = 4.42, p = 0.020$).

Exploring the interaction with Bonferroni corrected post-hoc comparisons showed that young adults increased the accuracy of their responses from preferred to narrow walking; the correct response ratio DTC were greater in narrow than preferred walking ($p = 0.002$). Conversely, their rate of responses was greater during preferred walking compared to narrow ($p = 0.030$). To the contrary, older adults exhibited no changes to cognitive performance across walking tasks. These findings may reflect a flexible speed-accuracy trade-off between number of responses, and accuracy in the Recitation task with cognitive-motor challenge. That is, young adults were willing to respond more in preferred walking (a low-risk walk), which increased the likelihood of incorrect responses. In narrow walking, young adults reduce the number of responses, consequently increasing accuracy during Recitation. Alternatively, older adults
maintained cognitive performance, allowing gait to change to manage the cognitive-motor interference when mediolateral stability was compromised (narrow walking).

Cognitive performance outcomes provided further insight to dual-task behavior and revealed that performance on the cognitive tasks were affected by manipulating mediolateral stability and age. After removing the effects of baseline executive interference (set-switching, set-maintenance, inhibition), cognitive performance costs during the DSB and Recitation showed interactions with age. In the DSB, challenging WM encouraged young adults to allow declines to DSB to manage mediolateral instability (narrow walking). Older adults preserved cognitive performance in both walking tasks and did not demonstrate differences in DSB DTC. Young adults exhibited an ability to flexibly allocate resources between walking and cognition: they maintained stability by allowing trade-offs to cognitive task performance (increased rate of responses and decreased correct response ratio) in DSB and Recitation. Fluency performance was improved in narrow walking, which may be a result of the familiarity of language production and the involvement of added cognitive networks.

The influence of age on the main effects were only detectable in cognitive performance costs. While young adults allowed cognitive costs to in challenging tasks during narrow walking, older adults showed no changes to cognitive performance. Assuming young adults represent the ‘gold standard’, older adults seemed to adopt a maladaptive strategy to cope with mediolateral instability by focusing on performance of challenging cognitive tasks (DSB and Fluency). Together, the findings supported that controlling walking balance involves higher-order cognitive processes. Further, the
cognitive-motor interference caused by mediolateral balance demands may be different with age.

**Body Weight Support**

A separate sample participated in Experiment 2: Body Weight Support (BWS). A total of 20 young adults (16 women, 22 ± 4 years) and 20 older adults (10 women, 72 ± 7 years) were included in the study (Table 4-3).

**Participant Demographics**

Participant characteristics and results from a series of Bonferroni corrected (significance set at $p = 0.0025$) independent t-tests are presented below (Table 4-3). Independent t-tests showed that older and younger adults were significantly different in few measures of cognitive and physical function. Comparing the morphology and physical function of young and older adults revealed no significant between group differences. Regarding cognition, most measures showed no significant age-differences. The only cognitive measures that were significantly worse in older adults were Stroop Incongruent and TMT part A. Additionally, significant differences between groups in the DGI confirmed that older adults had more difficulty in some clinical measures of walking balance in this experiment (Table 4-3).
Table 4-3. Sample demographics and group differences for young and older adults in BWS experiment

<table>
<thead>
<tr>
<th></th>
<th>Young</th>
<th>Old</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total N</td>
<td>20</td>
<td>20</td>
<td>--</td>
</tr>
<tr>
<td>Gender (Women/Men)</td>
<td>16/4</td>
<td>10/10</td>
<td>--</td>
</tr>
<tr>
<td>Age (years)</td>
<td>22.00 (4.00)</td>
<td>72.00 (7.00)</td>
<td>0.000*</td>
</tr>
<tr>
<td>Height (meters)</td>
<td>1.67 (0.12)</td>
<td>1.67 (0.16)</td>
<td>0.942</td>
</tr>
<tr>
<td>Body mass (kilograms)</td>
<td>69.77 (18.01)</td>
<td>76.33 (13.22)</td>
<td>0.193</td>
</tr>
<tr>
<td>Preferred Speed (meters/second)</td>
<td>0.94 (0.20)</td>
<td>0.97 (0.27)</td>
<td>0.659</td>
</tr>
<tr>
<td>Education (years)</td>
<td>15.21 (1.68)</td>
<td>16.55 (2.28)</td>
<td>0.045</td>
</tr>
<tr>
<td>Montreal Cognitive Assessment Score</td>
<td>27.56 (2.23)</td>
<td>26.10 (2.65)</td>
<td>0.051</td>
</tr>
<tr>
<td>Stroop Congruent Score</td>
<td>77.43 (12.18)</td>
<td>64.55 (15.29)</td>
<td>0.005</td>
</tr>
<tr>
<td>Stroop Incongruent Score</td>
<td>59.33 (17.52)</td>
<td>39.40 (10.75)</td>
<td>0.000*</td>
</tr>
<tr>
<td>△ Stroop Score</td>
<td>18.10 (20.14)</td>
<td>25.15 (12.07)</td>
<td>0.184</td>
</tr>
<tr>
<td>Trail Making Test Part A (seconds)</td>
<td>26.43 (12.60)</td>
<td>41.81 (16.87)</td>
<td>0.002*</td>
</tr>
<tr>
<td>Trail Making Test Part B (seconds)</td>
<td>56.75 (24.92)</td>
<td>79.12 (37.53)</td>
<td>0.030</td>
</tr>
<tr>
<td>△ Trail Making Test (seconds)</td>
<td>30.31 (21.56)</td>
<td>37.30 (34.56)</td>
<td>0.440</td>
</tr>
<tr>
<td>Number of Falls in Previous Six Months</td>
<td>0.90 (2.17)</td>
<td>0.15 (0.49)</td>
<td>0.136</td>
</tr>
<tr>
<td>Activities Balance Confidence (%)</td>
<td>94.24 (6.03)</td>
<td>95.24 (3.81)</td>
<td>0.108</td>
</tr>
<tr>
<td>CHAMPS (hours/week)</td>
<td>9.27 (6.64)</td>
<td>10.53 (7.51)</td>
<td>0.562</td>
</tr>
<tr>
<td>Dynamic Gait Index Score</td>
<td>24.0 (0.0)</td>
<td>22.90 (1.21)</td>
<td>0.000*</td>
</tr>
<tr>
<td>Dynamic Gait Index stair climb (seconds)</td>
<td>12.09 (2.10)</td>
<td>13.32 (2.47)</td>
<td>0.095</td>
</tr>
<tr>
<td>Berg Balance Scale Score</td>
<td>55.71 (1.10)</td>
<td>54.20 (2.51)</td>
<td>0.003</td>
</tr>
<tr>
<td>Berg Balance Scale Forward Lean (meters)</td>
<td>0.40 (0.05)</td>
<td>0.34 (0.34)</td>
<td>0.006</td>
</tr>
</tbody>
</table>

Note: Values presented as mean (standard deviation), CHAMPS: Community Healthy Activities Model Program for Seniors, p-values represent between group differences, p < 0.0025 marked by asterisk (*)

Walking Performance Outcomes

Step width for the right and left limbs for each step were averaged to best represent walking performance. Evaluating the distributions for each variable showed that the means were slightly more skewed and kurtotic than the medians. For example, mean values for step width in BW DSB resulted in a skewness statistic of 0.229 (0.37) and a kurtosis statistic of -0.155 (0.72), where the median values resulted in a skewness of 0.210 (0.37) and a kurtosis of -0.111 (0.72). Because high variability in DTC was expected, step width DTC were calculated from the median value for step...
parameters from each trial to reduce deviations from the overall means. The means and standard deviations for each outcome are presented in Figure 4-4.

Figure 4-4. Results from BWS experiment. Walking performance costs (%DTC), represented as mean and standard deviation, to step width (top) and step width variability (bottom) in young and older adults during the DSB, Recitation, and Fluency.

Step width and step width variability DTC were analyzed in a 2 × 2 × 3 (Age Group: Young, Old; Walk: BW, BW+S; Cognitive Task: DSB, Recitation, Fluency) repeated measures MANOVA. The Box’s M test value of 167.08 was associated with a p-value of 0.011, above the recommended p-value of <0.005 [203] confirming equal

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covariance among the dependent variables in each group. Bartlett’s test of sphericity showed the main effects violated assumptions of sphericity (all $p < 0.001$), therefore univariate follow-up tests were adjusted with Greenhouse-Geisser degrees of freedom. Levene’s test confirmed equal between-group variance on all measures (all $p > 0.05$). Mauchly’s test of sphericity confirmed all variables met within-subject assumptions of sphericity except one (step width in the main effect of Cognitive Task, $p = 0.016$). Therefore, Pillai’s trace ($V$) statistic was used to detect significant main effects because this method accounts for any violations of sphericity [201].

The omnibus multivariate analysis revealed a significant interaction between Age Group $\times$ Walk ($V = 0.184$, $F(2,37) = 4.168$, $p = 0.023$), and Walk $\times$ Cognitive Task ($V = 0.297$, $F(4, 35) = 3.704$, $p = 0.013$) on step width and step width variability DTC. Finally, the analysis revealed a significant main effect of Cognitive Task ($V = 0.515$, $F(4,35) = 9.287$, $p < 0.001$). The significant two-way interactions suggested age differences may indirectly influence the association between cognition and walking balance. However, no main effect of Age Group ($V = 0.096$, $F(2,37) = 1.963$, $p = 0.155$) or Walk ($V = 0.041$, $F(2,37) = 0.793$, $p = 0.460$) was detected.

The significant interaction between Age Group $\times$ Walk on walking DTC was explored further with univariate follow-ups. The interaction of Age Group $\times$ Walk significantly influenced step width DTC ($F(1,38) = 8.505$, $p = 0.006$), confirming that young adults have a significantly wider steps with a dual task in BW than BW+S ($p = 0.009$), but older adults did not significantly change step width ($p = 0.179$) (Figure 4-5). Regarding step width variability, no significant influence of Age Group $\times$ Walk was detected ($F(1,38) = 0.344$, $p = 0.561$).
Figure 4-5. Step width DTC represented as mean and standard deviation in young and older adults without (BW) and with balance support (BW+S). Significant differences ($p < 0.05$) are marked with an asterisk (*).

The effects of the interaction between Walk $\times$ Cognitive Task on step width variability support the influence of cognitive processes on walking balance (Figure 4-5). Although the interaction did not influence step width DTC significantly ($F(2,76) = 2.314$, $p = 0.106$), to the contrary, step width variability was significantly influenced by the interaction of Walk $\times$ Cognitive Task ($F(2,76) = 5.091$, $p = 0.009$). Bonferroni corrected post-hoc comparisons showed that while performing the DSB, participants increased step width variability in BW+S ($p = 0.006$) (Figure 4-6). With balance support, a challenging cognitive task imposed greater variability in step-to-step control, reflecting a negative impact of complex cognitive processes on mediolateral balance perception and foot placement.
Finally, the significant main effect of Cognitive Task on walking performance DTC was further clarified with univariate follow-ups and Bonferroni corrected post-hoc comparisons. The results indicated a significant effect of cognition on step width DTC ($F(2,76) = 5.736, p = 0.005$). Follow up pairwise comparisons showed that step width DTC were greater in Recitation than Fluency ($p = 0.026$), but DSB step width DTC were not greater than Fluency ($p = 0.104$), or Recitation ($p = 0.285$) (Figure 4-7).

![Figure 4-6. Step width variability DTC represented as mean and standard deviation without (BW) and with balance support (BW+S) in three cognitive tasks. Significant differences ($p< 0.05$) are marked with an asterisk (*).]
The main effect of Cognitive Task significantly impacted step width variability \( (F(2,76) = 14.309, p < 0.001) \). Bonferroni corrected post-hoc comparisons revealed that step width variability DTC were lower in Recitation than DSB \( (p < 0.001) \), and Fluency \( (p = 0.001) \) (Figure 4-8). These findings showed that cognitive demands influenced the variability of walking patterns such that challenging concurrent tasks impacted walking more than simple tasks. Decreased step width variability DTC in the Recitation task, without changing step width, indicated motor facilitation when walking with balance support in easy tasks. Together, these results suggested that unloading body weight enabled changes to gait performance that benefitted cognitive-motor processes. The link between cognition and upright postural control against gravity was further supported
by increased step width variability when cognitive demand was high (DSB) and decreased variability in simple tasks (Recitation) (Figure 4-8).

![Step Width Variability is Higher in Challenging Cognitive Tasks](image)

Figure 4-8. Step width variability DTC represented as mean and standard deviation for three cognitive tasks. Significant differences ($p < 0.05$) are marked with an asterisk (*).

**Cognitive Performance Outcomes**

A $2 \times 2$ (Age Group: Young, Old; Walk: BW, BW+S) RM ANOVA analyzed age differences in DSB performance costs from walking without support to with balance support (Table 4-2). Box’s M test statistic was 4.410 and associated with $p$-value of 0.245, confirming the equal covariance between dependent variables for each group. Levene’s test of between-subject effects showed both DSB DTC variables met the assumption of equal variance across groups (BW $p = 0.327$, BW+S $p = 0.484$). Pillai’s trace statistic ($V$) was used to detect significant effects because of the unequal sample included in this analysis. No main effect of Age Group ($F(1,38) = 0.319$, $p = 0.575$), or Walk ($V = 0.000$, $F(1,38) = 0.007$, $p = 0.984$), or Walk × Age Group ($V = 0.056$, $F(1,38) = 2.257$, $p = 0.141$) effects were detected in DSB performance DTC (Table 4-4).
Table 4-4. BWS cognitive performance results (%DTC) for young and older adults in without (BW) and with body weight support (BW+S) provided by a walking harness

<table>
<thead>
<tr>
<th>Cognitive Condition</th>
<th>Performance Outcome</th>
<th>BW Young</th>
<th>Older</th>
<th>BW+S Young</th>
<th>Older</th>
</tr>
</thead>
<tbody>
<tr>
<td>DSB Capacity</td>
<td>-0.15 (29.82)</td>
<td>-5.03 (34.23)</td>
<td>-5.92 (36.21)</td>
<td>3.25 (41.29)</td>
<td></td>
</tr>
<tr>
<td>Recitation Rate</td>
<td>0.98 (28.21)</td>
<td>1.40 (21.12)</td>
<td>0.31 (29.79)</td>
<td>1.96 (24.96)</td>
<td></td>
</tr>
<tr>
<td>Ratio</td>
<td>-1.38 (6.87)</td>
<td>3.45 (8.22)</td>
<td>1.94 (5.19)</td>
<td>3.65 (12.86)</td>
<td></td>
</tr>
<tr>
<td>Fluency Rate</td>
<td>-8.55 (20.60)</td>
<td>4.52 (24.78)</td>
<td>-0.21 (18.74)</td>
<td>10.11 (24.63)</td>
<td></td>
</tr>
<tr>
<td>Ratio</td>
<td>0.13 (2.35)</td>
<td>-1.27 (5.06)</td>
<td>-0.45 (4.41)</td>
<td>-2.11 (7.13)</td>
<td></td>
</tr>
</tbody>
</table>

Note: Dual-task cost (%DTC) values presented as mean (standard deviation), Rate: Response rate (words per second), Rate: Correct response ratio (ratio of correct words to total words)

Another 2 × 2 (Age Group: Young, Old; Walk: BW, BW+S) RM MANOVA analyzed age differences in DTC to response rate and correct response ratio during Recitation. Three participants (two young and one older adult) were missing time data to calculate response rate because of equipment difficulty, thus 20 YA and 19 OA were included in the omnibus MANOVA. Box’s M test statistic was 20.456 and associated with p-value of 0.054, confirming equal covariance between dependent variables for each group. Levene’s test of between-subject effects showed all variables met the assumption of equal variance across groups except for correct response ratio in BWS (F(1, 37) = 4.501, p = 0.041). Pillai’s trace statistic (V) was used to detect significant effects because of the unequal sample included in this analysis and the technique accounts for violations of equal variance. No effect of Age Group (V = 0.043, F(2,36) = 0.807, p = 0.454), Walk (V= 0.042, F(2,36) = 1.106, p = 0.464) or Walk × Age Group was detected in Recitation performance DTC (V= 0.031, F(2,36) = 0.579, p = 0.565).
A 2 × 2 (Age Group: Young, Old; Walk: BW, BW+S) RM MANOVA analyzed age differences in Fluency performance costs (% DTC) in two outcomes, response rate (number of responses per second) and correct response ratio (% correct responses out of total responses). Box’s M test statistic was 23.608 and associated with $p$-value of 0.022 which is below the recommended significance level of $p < 0.001$, confirming equal covariance between dependent variables for each group. Bartlett’s test of equal variance was significant ($p < 0.05$), thus Greenhouse-Geisser adjustments were applied as needed. Levene’s test of between-subject effects showed all variables met the assumption of equal variance across groups except the correct response ratio in BW ($p = 0.018$). Thus, Pillai’s trace statistic ($V$) was used to detect significant effects because of violations of equal variances. No main effect of Age Group ($V = 0.092, F(2,37) = 1.879, p = 0.167$) or significant differences to Fluency performance in the Age Group × Walk interaction were detected ($V = 0.005, F(2,37) = 0.089, p = 0.915$), and the main effect of Walk only showed a trend toward significance ($V = 0.127, F(2,37) = 2.694, p = 0.081$).

The cognitive performance results in BWS showed a consistent response to dual-task walking demands in both groups; both groups maintained cognitive performance in all walking conditions. Harnessed walking enabled a preservation of complex cognitive performance, removing cognitive costs to dual-task walking performance. Overall, the resistance to cognitive-motor interference demonstrated in this study provided promising evidence of cognitive benefits during dual-tasks when walking balance demands are reduced.
The overarching aim of this project was to determine age-differences in the effect of fundamental aspects of locomotor balance control (mediolateral stability and maintaining upright posture against gravity) on cognitive-motor interference during walking. The results from two studies confirmed higher-order cognitive processes are involved in step-to-step control. Herein, the following reviews the primary findings from both studies and focuses the interpretation of the results within three topic areas; 1) the influence of age, 2) dual-task prioritization, 3) the influence of cognitive task.

**Purpose and Primary Findings**

The findings from both experiments reflected shared resources between higher-order cognition, walking balance, and step-to-step control, such that added cognitive challenge negatively impacted fundamental aspects of locomotor balance. Conversely, providing balance support through body weight unloading seemed to benefit concurrent cognitive-motor performance. Previous studies reveal an association between cognition and controlling gait speed [21,87,206], step length [207], and width [208,209], but the degree that cognitive processes are actively involved in controlling steps independent of gait speed previously remained unclear.

**Step Rx**

The purpose of the Step Rx manipulations was to evaluate age differences in the effect of mediolateral balance control on cognitive-motor interference while walking. Cognitive and motor performance DTC were quantified in young and older adults while
walking in normal (preferred) and narrow step constraints (more unstable) during three cognitive tasks: the DSB, Recitation, and Fluency. The a priori hypotheses were only partially substantiated by the results.

The hypotheses predicted that compared to healthy young adults, older adults would demonstrate greater DTC to gait and cognition, especially under decreased mediolateral stability (i.e. a narrow walking pattern) and with greater cognitive difficulty (DSB and Fluency). However, few age-differences were detected, and none in gait performance. Significant interactions in cognitive performances revealed that YA decreased DSB cognitive performance more than OA in narrow walking. Further, the Fluency task was predicted to increase walking DTC more than Recitation and DSB; but Fluency DTC were equal to DSB and only greater than Recitation. Additionally, cognitive performance DTC were predicted to decline in narrow walking, and that the greatest declines would occur in Fluency performance, but only the DSB showed declines during unstable walking.

During narrow walking, both groups showed a tendency to seek additional control of the COM in the A/P direction [209–211], thereby decreasing step length when under added cognitive demand. The findings suggested a concurrent cognitive task further destabilized the control of mediolateral balance. Further support for this conclusion is provided by a significant main effect of a cognitive task on step length variability DTC, indicative of motor interference in step-to-step control [49,209]. DSB and Fluency equally changed step variability (remaining the same or declining), which may be a result of shared cognitive networks, specifically WM. Recitation decreased step length
variability the most, and even led to motor facilitation in dual-task walking (increased step length and decreased step length variability).

Cognitive performance outcomes provided further insight to dual-task behavior and revealed that young adults flexibly allocated resources between walking balance and cognition; maintaining walking by allowing trade-offs to cognitive task performance (increased rate of responses and decreased correct response ratio) in DSB and Recitation. Fluency performance was improved in narrow walking, despite motor interference (shorter steps), perhaps because language is a familiar task that combines multiple cognitive processes.

**Body Weight Support**

The purpose of the BWS experiment was to determine age-differences in the effect of maintaining upright postural control against gravity on cognitive-motor interference during walking. Walking and cognitive performance costs (DTC) were quantified as young and older adults walked without (BW) and with balance support (BW+S) provided by a walking harness during three cognitive tasks (DSB, Recitation, Fluency). Comparing group differences to cognitive and walking DTC only partially supported the a priori hypotheses.

The hypotheses predicted that decreased upright balance demands (BW+S) would reduce cognitive-motor interference for older adults in walking and cognitive performance. However, few age differences were significant in gait DTC between BW and BW+S, only that young adults reduced step width in BW+S. Further, the Fluency task was predicted to increase walking DTC more than Recitation and the DSB, but the DSB changed walking equal to Fluency in BW and BW+S. Cognitive performance was
maintained as expected, but no age-differences were detected. Finally, the hypotheses predicted that less challenging cognitive task performance (Recitation) would be facilitated when balance was supported (BW+S). This hypothesis was supported by the results; while participants demonstrated motor facilitation (reduced step variability), cognitive performance was maintained.

The findings from this study are difficult to directly compare with previous literature because few studies observe healthy older adults [44,212,213]. Importantly, the results aligned with previous research showing that body weight unloading can maintain cognitive performance, even in healthy older adults [44]. However, gait DTC revealed interactions between cognitive task and age, walking and age, and walking and cognition, suggesting that cognition influenced walking balance in a way that is indirectly influenced by age. Overall, walking with a harness and balance support led to stepping adjustments (increased step width and step width variability DTC in BW+S) that preserved cognitive performance, and the degree and direction of these changes were related to cognitive and balance demands.

In BWS, increased step width and variability while dual-tasking in BW+S supported that loading (maintenance of upright posture against gravity) provided neurological feedback with each step that involves higher order cognitive processes [34,49,51,52,209]. Providing body weight support reduces sensory feedback from restoring forces from each step that guide stepping behavior [34,51,52,213], and increased step width is a commonly reported consequence of added balance support [51,213]. A concurrent cognitive task may further impair processing of sensory feedback from reactive forces in each step, leading to wider steps as a means of gaining
additional somatosensory information [34,49,51,209]. Young adults decreased step width with a cognitive task in BW+S, but older adults did not, supporting that gravitational loading may be related to sensory processes that are disrupted with additional cognitive demands. Young adults may not rely as heavily on sensory information from gravitational forces for step-to-step control, acting in line with the a priori hypotheses; wider steps in BW, minimal step width DTC in BW+S, and maintained cognitive performance across all cognitive tasks.

Increasing step width seemed to compensate for reduced sensory information, but complex cognitive processes such as WM and Verbal Fluency further interfered with the neurological processes involved in planning each step, causing step variability. Challenging concurrent tasks (DSB and Fluency) incurred the greatest walking DTC to variability, demonstrating an influence of the type of cognitive task on dual-task walking and step-to-step control. Simultaneously, cognitive performance costs did not change in any walking condition, supporting that body weight unloading facilitated changes to gait that alleviated balance demands and maintained concomitant cognitive task performance. Support for this conclusion can be drawn from significantly greater step width variability during DSB performance from BW to BW+S (Figure 4-5). Cumulatively, the findings supported that higher order cognitive processes are involved in controlling foot placement against the demands of gravity in young and older adults.

**The Influence of Age**

In most cases, older adults responded to the manipulations like young adults. Although previous research shows that older adults have shorter step lengths during unstable dual-task walking than young adults [33], healthy older adults have previously
responded like young adults in dual-task treadmill walking [41]. Indeed, each sample of older adults were able to adapt to a cognitive load while negotiating mediolateral instability and altered upright balance demands. Step Rx showed older adults did not adjust cognitive or walking performance when mediolateral balance is challenged, where young adults flexibly allocated cognitive-motor resources. Alternatively, BWS demonstrated that older adults preserve cognition in low-risk settings by changing walking.

In relation to the existing literature, contradictory findings (a lack of statistical age differences) in the Step Rx study may be a result of different methods of narrowing steps; which have included constraining an overground walking path (without constraining speed) [33], aiming steps using visual cues (laser-projected lines) [41,207,214], or walking on a balance beam [215]. Mazaheri et al. reported no age-differences to dual-task performance (step width DTC) with and without lateral balance support [41]. However, unlike this study, they failed to detect any effect of a concurrent task, which was likely a result of a different sensory-based concurrent task, a proprioceptive stimulus on the calf, thus the findings are difficult to directly compare [41]. Perhaps previous studies that report age-differences impose certain sensorimotor demands to step width constraints that affect dual-task walking performance. Given that this study constrained speed, navigation, rhythmicity, and visual demands, other studies may observe larger differences in age groups when these factors are not controlled, such as during overground walking. For example, declines in older adult dual-task walking performance are consistent during overground walking [7,20,21,173].
In BWS, visual inspection of the graphs shows that BW and BW+S changed walking behavior differently for young and old, but few age-differences were statistically significant. The primary age-difference was reduced step width in BW+S for young adults, where older adults showed greater increases to step width during a dual-task. Increasing step width with body weight unloading indicates reduced neurological feedback and alleviated balance demands [51] demonstrated by healthy adults that increase step width with body weight support [34,51]. Although both groups demonstrated increased step width variability in challenging cognitive tasks, previous studies report that older adults increase step width variability more than young adults with body weight unloading [34,216]. Such age-differences to walking performance were predicted to extend (and increase) during dual-task walking performance, but the results did not support this prediction. Thus, young adults may not experience similar neurological effects from added balance support, where older adults must increase step widths to enhance/maintain cognitive performance. Nonetheless, when a higher-order cognitive task is added to walking, the variability of these stepping patterns was similar across ages.

Minimal age-differences revealed in this study could be related to the cognitive or motor benefits of a walking harness (even without support) that enabled older adults to walk more like young adults. Although young and older adults self-reported equal levels of balance confidence and number of falls, the placebo effect of the harness could increase perceived balance ability for older adults and encourage maintenance of the cognitive task in BW and BW+S [4,105,217]. Furthermore, this sample of older adults was physically active and highly educated (see demographics Tables 4-1 and Table 4-
2), which may have limited any detectable changes or age-differences in dual-task performance.

Although both experiments confirmed the influence of higher order cognitive networks on step-to-step dynamic balance control, limited age-differences were revealed. The findings suggested the cognitive processes that control fundamental aspects of locomotor balance are largely preserved in healthy aging. Large age-differences previously reported during dual-task walking [7,10,148] may arise from other aspects of the cognitive or motor tasks.

**Dual-Task Prioritization**

Maintaining cognition and allowing gait to change is often referred to as 'prioritization'; where one component of a dual task is allowed to decline, while the other is maintained [4,148,218]. This strategy is thought to depend on postural reserve, physical ability to recover from a perturbation in the motor task, as well as cognitive task difficulty [4,148,218]. Prioritization is considered important for older adults to successfully negotiate balance and walking performance along with concurrent cognitive and motor demands [4,148,219]. Alternatively, young adults do not exhibit prioritization until task demands are highly challenging [8,148,173]. The prioritization behavior in each experiment reflects the manner that resources are allocated between fundamental locomotor balance and completing each cognitive task.

The two experiments revealed contrary patterns of prioritization. In Step Rx, prioritizing gait was more important, forcing cognitive outcomes to change. The opposite is true with a walking harness in BWS; participants prioritized cognitive processes and allowed walking performance to change. Together, these experiments showed
participants prioritized the cognitive task when in low-risk walking settings (i.e. preferred step, BW+S) and prioritize gait when balance is most disrupted (narrow walking).

In Step Rx, participants maintained the cognitive task when the risk of falling was low (i.e. preferred walking) by allowing changes to walking but focused on the walking task in unstable situations (narrow walking). Young adults exhibited an ability to flexibly allocate resources between walking and cognition; they maintained stability by allowing trade-offs to cognitive task performance (increased rate of responses and decreased correct response ratio) in Recitation, and reduced DSB performance in narrow walking. The participants exhibit motor interference to gait performance (decreased step length and step length variability. Interactions with age in cognitive outcomes demonstrated surprising cognitive-task prioritization; older adults maintained or improving cognition in challenging tasks (DSB and Fluency, respectively) no matter the walking condition.

Although the participants were healthy older adults, the tendency to prioritize a cognitive task in destabilizing walking situations may put older adults at a higher risk of falls [219]. Alternatively, young adults exhibited predictable prioritization patterns when their gait was destabilized, prioritizing gait when stability was needed and allowing the cognitive task to decline [4,218]. This result is unsurprising given previous studies that show healthy adults only prioritize the gait task when both cognitive and motor task difficulty are high [148].

In BWS, both groups consistently prioritized cognition with a walking harness, only allowing changes to gait and enhancing cognitive performance [4,148,218]. Walking with a harness reduces risk of postural destabilization, likely encouraging participants to preserve cognition. When their body weight was unloaded and balance
demands reduced, they exhibited wider and more variable steps to meet task demands when cognitive processes were disrupted [51,67,209]. Changing step width variability in challenging tasks suggests that cognitive-motor interference is allowed only in walking so that cognition is maintained (prioritized). BW+S may promote cognitive performance by alleviating the continuous neurological demands of step width control [209] and facilitate performance of the cognitive task. Ultimately, unloading body weight allowed participants to prioritize cognition and increase step width variability in challenging tasks, revealing benefits to cognitive performance in dual-task walking with balance support.

**The Influence of Cognitive Task**

Most previous studies administer one type of cognitive task to evaluate dual-task walking performance [41,44,143,220]. However, the results from this dissertation aligned with some previous research suggesting that different types of cognitive tasks influence gait performance uniquely [8,175,221]. In both experiments, a significant main effect of cognitive task was revealed in walking DTC, and the step-to-step control of mediolateral and upright balance were similarly affected by the type of cognitive task. The methods maintained the type of stimuli (auditory) and response (verbal) across all task types, fortifying the assertion that influence of cognitive task demand is represented by the findings presented here.

The three cognitive tasks challenged different cognitive processes and elicited different responses in all stepping constraints. Generally, DSB and Fluency performance led to greater increases to walking variability DTC than Recitation. Interestingly, the effect of cognitive task was only present in the variability of step width
and step length but did not significantly impact overall step width or step length outcomes. Variability in foot placement signaled some difficulty (i.e. resource interference) in perceiving and planning the next step, as evidenced by studies reporting increased step length variability in individuals with reduced cortical integrity [67,215,222]. Together, the results underscored the important role of complex cognitive domains such as WM in dual-task walking.

The challenge to WM during DSB negatively interfered with walking balance control in both experiments, leading to increased variability. In Step Rx, participants exhibited added walking variability during performance on DSB in low risk (preferred) walking settings. In narrow walking, young adults prioritized walking by sacrificing DSB performance, showing significant declines to WM capacity instead of decreasing step length and variability. A similar pattern of DTC in BWS further supported that step-to-step control involves complex cognitive processes such as WM; step widths were more variable in challenging tasks (DSB and Fluency). Maintaining cognition in BW+S required significant increases to step width variability DTC during DSB in both groups, reflecting some cognitive interference with WM in upright balance during walking [183]. Participants further increased step width variability DTC in BW+S, demonstrating motor interference during DSB that required more variable steps to maintain performance on the DSB.

Here, increased step variability in DSB and Fluency further evidenced the link between step placement, cortical attention, and WM complex cognitive networks [207], showing EF may play a key role in foot placement that maintains walking stability. The results of this study are supported by recent brain research asserting that various
cortical regions are involved in step-to-step control. Neuroimaging evidence shows that different cortical networks control gait speed and gait speed variability in older adults [87], where gait speed variability [223], and step length variability [207] are related to activation of the dorsolateral prefrontal cortex (dIPFC) [34,49,67,207]. The dIPFC controls attention and WM [79,113] and demonstrates increased involvement in dual tasks in older adults [79,224,225]. Previous theories suggest that the neurological feedback gained with each step is actively used to plan the placement of the next step [209], and WM may be involved in maintaining the memory of position in previous steps. Altogether, the findings from this study supported that WM is involved in step-to-step balance control.

In both studies, Fluency performance was prioritized and even improved, despite the complexity of language formulation [120] or resultant motor costs (increased step variability). The results aligned with a previous study reporting that language is prioritized despite motor task difficulty in young adults [226], and the findings from both Step Rx and BWS suggest prioritizing language may occur in healthy older adults. In Step Rx, Fluency performance was improved in narrow walking, evidenced by an increased correct response ratio. However, participants decreased step length with a cognitive task to gain stability and had to increase variability DTC in narrow walking to maintain cognitive performance. The prioritization of Fluency extended to BWS, where young adults were able to maintain Fluency while reducing step width with support, demonstrating cognitive-motor benefits of body weight unloading in BW+S. Although older adults had a greater step width than young adults in both walking tasks, they maintained Fluency performance during this familiar, but challenging cognitive task.
Altogether, despite the resultant cognitive-motor costs in unstable walking, participants focused on Fluency performance, perhaps because of their familiarity with language production.

Language production incorporates many executive cognitive processes, including WM [100,120,126]. The findings suggested that motor interferences attributed to WM may contribute to the influence of language processes on dual-task walking. Equal gait DTC in DSB and Fluency in both studies suggested each may negatively influence walking with the heavy involvement of WM in both tasks [100,113]. However, the findings suggested that language processes that can be prioritized despite balance disturbances. Further, when balance support is provided, and WM networks may be under less demand, language processes could be facilitated. Given the large role of language and mobility in everyday life, future research should examine what aspects of language ability underlies cognitive-motor DTC in language-based dual tasks.

In both experiments, the Recitation task revealed some motor facilitation in both groups. Assuming decreased variability is a benefit to walking stability in both experiments; the combination of increased step length/width or maintained step length/width (Figure 4-1, 4-3) while decreasing step length/width variability (Figure 4-2, 4-4) indicated motor facilitation during the Recitation task. In Step Rx, both groups decreased their step variability during Recitation in both walking conditions (while maintaining cognition), indicating a facilitation of motor performance (i.e. balance recovery) in unstable situations. Further, young adults showed a flexible trade-off in cognitive task performance to accomplish Recitation in narrow walking; responding less frequently—but more accurately. A trade-off between cognitive and gait accuracy has
been previously reported in healthy adults during narrow walking [168], interpreted as a flexible allocation of resources between cognition and motor control [4,168]. In BW, participants demonstrated a large increase in step width during Recitation, which may be a result of a trade-off between step width and its variability to produce more verbal responses without balance support [53]. In BW+S, gait variability uniquely declined during Recitation without increasing step width or reducing cognitive performance, reflecting a benefit to walking performance during an easier concurrent task.

Recitation may facilitate performance as a byproduct of the harmony and rhythmicity inherent to the Recitation task. Previous evidence shows decreases to walking variability and benefits to walking performance in clinical rehabilitation when listening to music or cueing walking a metronome [227,228]. Reciting the alphabet is guided by a familiar harmony and beat, and the Recitation task may impose a rhythm to walking that affects step variability independent of complex cognitive processes [175,227,228]. When speed and step placement is constrained, motor benefits may occur because of the cue provided by Recitation, entraining the rhythmicity of steps to accomplish the dual task; demonstrated here by little or reduced step length and width variability when performing the Recitation task in both groups. Although this project was not designed to answer this question, understanding the potential for motor facilitation with simple and rhythmic dual tasks could inform and improve rehabilitative strategies. Future research should determine what mechanisms contribute to the unique cognitive-motor response to dual-task walking during Recitation.
CHAPTER 6
CONCLUSION

Cumulatively, the evidence demonstrated a link between fundamental walking balance and higher-order cognitive processes. Participants prioritized walking by adjusting steps to gain stability and allowed cognitive changes to cope with mediolateral instability in Step Rx. In BWS, the harness and balance support benefitted the cognitive task and only gait changed. The challenging concurrent tasks (DSB and Fluency) led to step adjustments that increased stability and increased variability, indicating cognitive loads further destabilized walking and cortical processes were involved in gait stabilization [8,221,229]. To preserve cognitive performance, both groups changed step variability (decreasing in Step Rx and increasing in BWS) in cognitive tasks that challenged WM and language (DSB and Fluency). Recitation, a simple and rhythmic task, decreased variability and showed the potential to improve cognitive and motor performance during dual-task walking.

In conclusion, complex cognitive processes contributed to two fundamental aspects of walking balance: mediolateral balance and maintaining upright balance against gravity. Increased complexity of the cognitive task negatively influenced dual-task walking performance in addition to locomotor balance demands. Reciprocally, providing balance support while walking prevented declines to concurrent cognitive tasks. Thus, dual-task declines in walking and cognitive performance may be a consequence of balance demands in step-to-step locomotor control. The cognitive control of step length and width likely impacted previous associations between cognitive function, decreased gait speed [21,22,44], and increased gait variability [87,207]. Altogether, the results suggested that cognitive involvement in mediolateral balance, or
managing the demands of gravity, could contribute to reduced dual-task gait speed during overground walking in young and older adults [22,148,206,230,231]. Few age-differences in performance provided encouraging evidence of proficient cognitive-motor function in healthy aging.

**Limitations**

There were several limitations to this study. Our sample of unpaid volunteers consisted of a group of older adults motivated to prove their physical and cognitive acumen, and a group of young adults motivated to gain extra credit. The results may have been affected by the motivation of the subjects, limiting detectable age-differences. Future studies would benefit from incentivizing participants or recruiting seniors with a wide range of education and socioeconomic backgrounds. Additionally, the added safety of harness (for older adults) may have had some placebo effect that prevented large age-differences. Further, testing took place on a split-belt treadmill and both groups may have widened step widths because of the two-belt design, which has been identified as a potential limitation in previous studies of age-differences in mediolateral walking balance [34].

Each study only evaluated one level of stability manipulation: destabilizing by imposing one level of step narrowing in Step Rx, and stabilizing with one level of body weight unloading in BWS. The results may be better understood with multiple levels of stability demands. Future research should determine the influence of progressively stabilizing or destabilizing walking on cognitive-motor demand. Additionally, added measures that more directly measure the characteristics of the COM may enrich the findings and interpretations of the current study. Finally, the focus of the investigation limits the generalizability of the results to daily walking activities. Future studies should
investigate cognitive-motor interference to step-to-step balance control in a relevant mobility behavior such as overground walking or navigating an unpredictable environment.

**Implications**

The findings from this study provided initial evidence of a link between step-to-step locomotor control and higher-order cognitive processes. Several important implications arose from the results.

First, the Step Rx experiment demonstrated that controlling rhythmicity, speed, and constraining foot placement may promote cognitive-motor performance when walking balance is disrupted alongside cognitive demands. Participants were able to perform dual-task walking when disruptions to walking balance were isolated, perhaps because of the guidance provided by treadmill walking and a defined walking path. In a clinical setting, this information could benefit individuals suffering from movement disorders that degrade movement rhythmicity and timing, especially under cognitive-motor demand, such as Parkinson’s disease [173]. Future research should determine if mobility therapies could improve by including cognitive-motor training that guide walking speed and rhythmicity.

Second, in the BWS experiment, the evidence (wider steps with a dual-task for older adults) suggested that gravitational demands may affect higher order sensorimotor processes involved in step-to-step control. Nonetheless, cognitive performance benefitted; body weight unloading seems to manipulate sensorimotor demand in a way that facilitates cognition. Body weight unloading is already a common rehabilitation tool for those with neurological deficits that affect walking such spinal cord injury, multiple sclerosis, or stroke [232–234]. Importantly, the findings from this study
could inform existing rehabilitative approaches to body weight unloading. For instance, walking therapies focused on peripheral neurological and sensory systems may better benefit step control by including a cognitive task while unloading body weight. Furthermore, the results added to existing evidence of benefits to walking as a result of unloading body weight in individuals with cortical/central nervous disorders (i.e. Parkinson’s disease, Huntington’s disease) [235,236], and suggested there may be some added benefit to cognitive processes with this type of walking rehabilitation. Future research should investigate the potential cognitive-motor benefits of body weight unloading with a concurrent cognitive task in such clinical populations.

Third, the findings suggested that Alphabet Recitation could facilitate walking when rhythmicity and speed are controlled. Several previous studies have reported dual-task declines to overground walking or self-paced treadmill walking [44,237,238]. To the contrary, reduced step variability during Recitation in both studies suggested a unique influence of rhythmical dual-tasks on step control when stepping rhythmicity and gait speed were constrained. The findings provided encouraging evidence that existing cognitive-motor rehabilitation therapies could be enhanced by targeting and training rhythmicity. Future research should explore the underlying mechanisms of Recitation in relation to gait speed and rhythmicity.
APPENDIX

AN ODE TO WALKING BY DAVID WINTER, PHD

Walking is for moving from one place to another- to go to table for breakfast, to climb stairs to bed, to meet a friend, to walk to aisles of the food mart.

Walking is for enjoying from one day to another- to play hopscotch, to play jump rope, to play cops and robbers, to go dancing through the trees in the park, at parties for weddings and birthdays, with the beloved at nightfall, to meet the returning child.

Walking is very important for meeting the world, for growing up, for retreating to solitude, for returning to join again, for carrying the day’s tasks, for belonging.

Walking makes a big difference in how one’s life turns out, in whether one has a life at all, so it’s very important for almost everything-like breathing.

Walking can make the differences whether we ever stand up straight or not.

But walking does not come automatically like breathing. It must be learned. It begins while we are still very small, lying alone in our bed quite dependent on the others we see walking.

We see the others walking- the mother who nurses and feeds us, the father who holds and plays with us, the sister or brother or neighbor who walks toward us, smiles to us, then walks away, leaving us there alone on our back in our bed.

We see others walking, see what they can do with their walking, what we cannot do, and we want to do it too.

So lying on our back or face soon is not enough. We struggle to crawl-then we crawl everywhere we can. Then we try pulling ourselves to a stand at a table led, at father’s leg, at the stairsteps, by holding the drapes, by hanging to tablecloth.

We grunt and push and pull and fall and roll and bump, then try again and keep it up over and over again, and never quit in spite of face-falls and nose bruises- all because we want to be what we feel persons come to be by walking.

Walking is expensive. It is not learned without risking, without falling so hard it hurts, without slipping on the stairs so suddenly it frightens us and our parents, keeps us from trying again-for a while at least.
LIST OF REFERENCES


BIOGRAPHICAL SKETCH

Tiphanie E. Raffegeau explores the relationship between gait and cognitive control across the lifespan, focusing on mobility and fall prevention in older adults. Specifically, she conducts translational research studies that combine expertise from cognition, motor development, and gait biomechanics to study the association between cognition and locomotor control. Primarily through a series of dual-task experiments, Tiphanie investigates the complex interaction of cognitive and locomotor control resources required to walk while performing daily activities and how this changes with age. She has investigated dual-task performance while walking, crossing obstacles, cycling, and elliptical exercise and the findings from this work have challenged commonly held theories governing dual-task performance. Further, her background and collaboration with Speech, Language, and Hearing Sciences enables novel experimental designs and analysis to examine cognitive and linguistic contributions during dual-task performance. During her graduate career she published two manuscripts, was involved in numerous research projects, and delivered academic presentations at international and national conferences. In 2018, Tiphanie completed her doctoral degree in health and human performance with a certificate in gerontology at the University of Florida.