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Age- and speed-related differences in walking smoothness

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ABSTRACT

Extensive literature documents age-related differences in spatiotemporal variables during gait. Recent efforts have focused on upper body control and balance during gait. Harmonic ratios, derived from trunk acceleration signals, measure smoothness of motion, and are an indication of dynamic balance during gait. Limited and conflicting information exists regarding the effect of walking speed on harmonic ratios, as well as age-related differences. This research extends previous literature by: 1) examining harmonic ratios across a range of self-selected speeds in young adults; 2) comparing harmonic ratios at preferred speed in young adults, healthy 60-year-olds, and healthy 80-year-olds; and 3) comparing harmonic ratios in all three age groups across a range of self-selected and paced speeds. In contrast to previous research, young adults and 60-year-olds had similar trunk smoothness during over ground walking at a preferred pace, but 80-year-olds exhibited reduced smoothness specifically in the anteroposterior direction. In contrast to previous research, clear optimization of trunk smoothness at preferred speed in young adults was not found; trunk smoothness was reduced at slower speeds, but was maintained at speeds faster than preferred. The two older groups showed this same pattern, although the 80-year-olds exhibited reduced anteroposterior and vertical smoothness across speeds. Together, these findings indicate that active healthy 80-year-olds exhibit changes in trunk control even during unobstructed walking at their preferred pace. But contrary to expectations, changes in speed did not differentially affect 80-year-olds, except at very fast walking speeds.

CHAPTER 1. INTRODUCTION

Substantial research documents age-related changes in spatiotemporal gait parameters. The most consistent findings have been that older adults walk more slowly, take shorter steps, and spend more time in double limb support (Imms & Edholm, 1981; Murray, Kory, & Clarkson, 1969). These changes have generally been interpreted as older adults adopting a more cautious, stable gait. However, these same gait changes are characteristic of fallers (Imms & Edholm, 1981; Guimaraes & Isaacs, 1980; Maki, 1997) and predictive of falls (bellan Van et al., 2009). Thus, reduced walking speed in older adults has been interpreted both as a predictor of falls and as a strategy to ensure stability. This problem arises because walking speed is an outcome measure, and may not adequately describe the organizational or essential features of walking.

In an effort to quantify dynamic walking balance, recent research has used tri-axial accelerations of the upper body to determine anteroposterior, vertical, and mediolateral harmonic ratios as a measure of the smoothness and rhythmicity of motion. Results have shown differences in harmonic ratios between younger and older adults (Kavanagh, Barrett, & Morrison, 2005; Mazza, Iosa, Pecoraro, & Cappozzo, 2008), and between stable and unstable older adults (Menz, Lord, & Fitzpatrick, 2003a; Yack & Berger, 1993). Though these studies provide evidence that trunk acceleration measures are discriminatory and offer insight into the underlying mechanisms of gait control, several issues need consideration.

Considerable research documents the relationship of cadence, stride length, and double support to walking speed, but the relationship between trunk acceleration measures and walking speed is not well understood. While some results have shown that harmonic ratios are optimized at preferred speeds in young adults (Menz, Lord, & Fitzpatrick, 2003b), other results have shown some speed-dependency in young adults, with lower harmonic ratios at speeds slower than preferred pace (Yack & Berger, 1993). This may make interpretation of differences in harmonic ratios difficult as older groups or clinical groups are expected to walk more slowly.

Further, understanding age-related locomotor changes and adaptations means assessing ‘real-world’ walking, which includes examining gait over a range of speeds and complexities. One recent area of research has focused on understanding the relationship between attentional mechanisms and gait by examining age-related differences in dual-tasking and walking (Chen et al., 1996; Woollacott & Shumway-Cook, 2002). One consistent finding from this literature is that both young and old participants slow down walking when performing a secondary task if there is sufficient cognitive load. If the use of harmonic ratios is going to extend beyond preferred pace walking, they need to be examined across a range of speeds in healthy older adults.

While age differences in harmonic ratios have been observed in several studies, the differences have not been directionally consistent (i.e., one study reported a difference in mediolateral harmonic ratios, while another found no differences in the mediolateral direction, but did find differences in the other directions). These discrepancies may in part be due to differences in the ages and health of the older adults in each study.

The primary purpose of this study was to replicate and extend previous research by: 1) examining harmonic ratios across a range of self-selected speeds in young adults; 2) comparing harmonic ratios at preferred speed in young adults, healthy 60-year-olds, and healthy 80-year-olds; and 3) comparing harmonic ratios in all three age groups across a range of self-selected and paced speeds. A secondary purpose was to examine the relationships between harmonic ratios, spatiotemporal variables, and strength measures. As clinicians routinely use spatiotemporal parameters and strength measures in assessment and intervention, it is important to determine how these measures relate to harmonic ratios, and if these relationships are directionally-specific.

Significance of Research

Studies have shown that 30% to 70% of falls in older adults occur during walking (Lord, Ward, Williams, & Anstey, 1993; Tinetti, Speechley, & Ginter, 1988). Among older adults, falls are the leading cause of injury deaths, and the most common cause of nonfatal injuries and hospital admissions for trauma (CDC, 2009). As any fall can seriously affect an older adults' quality of life, research examining age-related changes in walking is important. By using a range of self-selected and paced speeds, examining healthy 60-year-olds separately from healthy 80-year-olds, and including measurements of spatiotemporal parameters and strength, the information from this research will add to our understanding of harmonic ratios, and normal age-related changes in walking smoothness. In addition, knowledge of normal age-related changes in walking smoothness is important to better understand changes in frail or clinical populations.

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CHAPTER 2. REVIEW OF LITERATURE

Introduction

Walking is one of the most basic of all human movements, allowing us to move our bodies safely and efficiently across many different types of surfaces and terrains. It is an amazingly elegant and complex skill, requiring the entire body, and thus the continuous integration of multiple sensorimotor systems for successful performance. Considerable research has examined over ground walking on a level surface, and even during ordinary conditions, walking is an inherently unstable task. Navigating through complex environments such as malls or city streets requires the use of sensory inputs to assist in the control and adaptation of gait. Successful walking also requires adaptation of gait patterns to avoid or negotiate obstacles, uneven terrain, and changes in speed and direction. Because of these complexities, understanding both the control of normal gait and how control is altered in normal aging and pathology is an overwhelming task.

There are three main sections to this review. First, the basic task of walking will be described, followed by a brief overview of the sensory components and descending neural influences on gait. Second, the literature highlighting age-related changes in gait is presented together with biomechanical, sensory and attentional factors that impact gait in older adults. Lastly, the literature examining the use of accelerometry and harmonic analyses as a method a gait analysis will be presented.

The Task of Walking

Overview and Strategies of Control

The walking cycle, or stride, is defined as the period between successive heel contacts of the same foot, or two steps. The stride consists of right and left single-support phases, and two double-limb support phases. Single-support phase is the period when only one foot is in contact with the ground, and double support is when both feet are in contact with the ground. When one limb is in single support, the opposite limb is in swing phase. Stance phase for one limb combines both the time in single support, as well as the two double support phases. At preferred speed, adults usually spend 60% in stance phase and 40% in swing, or 80% of the time in single support and 20% in double support.

The center of mass of the body (COM) lies within the lower trunk, just posterior to the umbilicus and in front of the spine. The vertical projection of the COM to the ground is the center of gravity (COG). When we are quietly standing the COG falls within our base of support (BOS) as defined by the area encompassing both feet. However, during walking, our COM is moving, and the BOS is moving and changing size. During single-support, the BOS is equal to the area of one foot, while in double support the BOS is larger as it encompasses the area around both feet.

Walking has been described as controlled falling (Frank & Patla, 2003). For a given stride, the COM is within the BOS only during the double support phases, only 20% of the stride. During the single-support phases (80% of the stride), the vertical projection of the body's COM travels forward and medial to the inside border of the supporting foot, thus *outside* the BOS (Winter, 1991). As the single-support phase progresses, the COM moves ahead of the supporting foot and toward the anticipated landing of the swing foot.

An additional control challenge is due to the fact that two-thirds of total body weight is carried in the head-arm-trunk (HAT) segment, and this segment has to be controlled both in the sagittal and frontal planes to prevent falling and to stabilize the head for gaze orientation. The finding that trunk motion in the sagittal plane varies little (between +1 and -1 degree) indicates prioritization for trunk stabilization (Winter, MacKinnon, Ruder, & Wieman, 1993). As stated previously, in single-support the COM is medial to and outside the BOS created by the one foot. Though this is advantageous for weight transfer, it also creates mediolateral instability, as the gravitational moment makes the body fall toward the midline. To prevent this falling toward midline, a counterbalancing moment is produced by the hip abductors and trunk lateral flexors (MacKinnon & Winter, 1993). It has been shown that the most important factor affecting frontal plane balance is the mediolateral foot position relative to the COM at heel strike, which is turn dependent on hip abductor/adductor moments generated during single-support (MacKinnon & Winter, 1993). Step widths are generally selected to position the COM equidistant between the feet in double support (Redfern & Schumann, 1994).

In the sagittal plane, ground reaction forces at heel strike cause the hip to accelerate backwards, which in turn causes the HAT to lean forward. Similarly, at push-off, the ground reaction forces cause the hip to accelerate forward and HAT to lean backward. Due to changes in hip accelerations, the upper body would be unstable, oscillating back and forth if not for the counterbalancing torques produced by the hip and trunk (Winter, Ruder, & MacKinnon, 1990).

Considering the above discussion, the potential for loss of balance during walking is considerable. Patla (Patla, 2003) has stated "stability during locomotion is maintained through reactive, predictive, and anticipatory strategies and involves the control of the COM within the changing and moving base of support" (pg. 48). Predictive and anticipatory strategies together are termed proactive strategies. Reactive control is used for recovery from an unexpected disturbance, such as a slip or trip, predictive control refers to the estimation of expected perturbations generated by the ongoing movement, and anticipatory control is used for the identification of potential disruptions and hazards. Reactive control, or feedback control, is highly dependent on sensory systems, particularly the proprioceptive system, to detect and trigger recovery responses (Frank & Patla, 2003). Research has shown that spinal reflexes are integrated into the phases of gait to remain functionally adaptive. For example, when the support surface was suddenly translated backward during single support, there was a fast 100ms response in the ankle plantarflexors and postural muscles that helped to prevent the body from pitching forward and realign the COM with the supporting foot (Nashner, 1980).

Unlike reactive control, predictive control is feed-forward and thought to develop from experience and be shaped by sensory information. It is thought to be used continuously throughout gait to ensure control of the COM. For example, predictive control is presumably used to stabilize the trunk during heel strike and push-off; if absent, the trunk would pitch forward and backward when walking (Patla, 2003). Similarly, the trunk would fall towards midline in single support if not for counterbalancing strategies by the hip abductors and trunk lateral flexors. Anticipatory control is largely controlled by vision, whereby we scan our environment and make navigational decisions.

Globally, successful mobility depends on the interactions between the abilities of the performer, the requirements of the task, and challenges of the environment (Shumway-Cook et al., 2002). Walking is inherently a complex, unstable task and challenges to walking, particularly in the community, may be great. With the additional age-related changes in body systems, older adults face greater challenges during walking and are more susceptible to falling.

Studies have shown that 30% to 70% of falls in older adults occur during walking (Lord, Ward, Williams, & Anstey, 1993; Tinetti, Speechley, & Ginter, 1988). Among older adults, falls are the leading cause of injury deaths, and the most common cause of nonfatal injuries and hospital admissions for trauma (CDC, 2009). In 2000, the total direct cost of all fall injuries for people 65 and older exceeded $19 billion: $0.2 billion for fatal falls, and $19 billion for nonfatal falls, as this includes costs for services such as long-term care, medication, rehabilitation, and equipment (CDC, 2009). As any fall can seriously affect an older adults' quality of life, research examining age-related changes in walking is important.

Summary of Lower Extremity Kinematics and Kinetics

Extensive research has detailed lower body kinematics and kinetics during gait (Murray, 1967; Perry, 1992; Winter, 1983). An exhaustive presentation is beyond the scope of this review; however, major kinematic and muscle activation patterns at each joint/segment will be summarized. As this dissertation research involves trunk motion, more detail will be presented regarding trunk kinematics and kinetics.

Starting at the foot and ankle, the ankle is within a few degrees of neutral at initial contact with the ground. Following heel strike, the ankle plantarflexes bringing the forefoot to the ground. Moving to mid-stance, the tibia moves over the foot resulting in dorsiflexion, then moves back into plantarflexion during late/terminal stance to prepare for push-off. During swing phase, the ankle moves into dorsiflexion after which it remains neutral before the next heel contact. The foot is slightly supinated at initial contact, and generally maintains supination through mid-stance to toe-off. The knee is near full extension at initial contact, flexes during loading and early mid-stance, then extends during later mid-stance, then flexes and continues to flex reaching a peak in the early part of the swing phase. The knee then extends again prior to initial contact. The hip is maximally flexed at initial contact, then extends throughout the stance phase as the upper body moves over the limb. The hip reaches peak extension at the end of stance, then flexes until the next initial contact of the foot.

In general, muscles in the stance limb have two functions: 1) support, to stabilize the limb for weight acceptance and shock absorption, and to prevent collapse during single-support, and 2) propulsion, to propel the body forward to the next step. To accomplish the function of support, eccentric activation of the quadriceps at initial contact allows for a small amount of knee flexion that acts to absorb the impact of heel strike. Eccentric activation of ankle dorsiflexors allows for controlled lowering of the foot to the ground. Also at initial contact progressing through early stance and loading, hip, knee and ankle extensors are activated.

The sum of the hip, knee, and ankle moments is termed the “support moment” which is the net extensor moment that prevents the limb from vertical collapse while bearing weight (Winter, 1980). Though the support moment remains relatively invariant across strides, the individual moments at the hip and knee are more variable than the ankle, presuming to allow on-line adjustments to anteroposterior motion (Winter, 1983). As described previously, the hip abductors are active during mid-stance to prevent the HAT falling toward midline. The second function of the stance limb is propulsion, which is primarily accomplished by contraction of the ankle plantarflexors at the end of stance phase.

The goal of the swing phase is to accurately reposition the limb to allow for smooth forward progression. In early swing, contraction of the quadriceps and hip flexors causes the thigh to accelerate forward. The flexion of the knee in swing is generally accomplished passively as a result of the acceleration of the thigh. Similarly, the knee extension observed at the end of swing to prepare for initial contact is the result of passive forces (Winter, 1984).

As successful, adaptive gait requires changes of speed, there has been considerable examination of the impact of speed on lower extremity kinematics and kinetics. Winter (1983) summarized the findings of his research in healthy adults walking at slow, preferred and fast speeds as follows: 1) joint angle patterns remained generally invariant across speeds (though more variable at the hip), with correlations at .90 or greater between speeds. This ensures that positional feedback will be consistent across speeds, while at the same time velocity-related spindle information increases with increasing speeds; 2) moments of force at the ankle were the least variable and varied little across speeds; 3) the support moment (sum of hip, knee and ankle moments; a positive extension moment) was relatively invariant across speeds; 4) moments at the hip and knee were more variable than the ankle, but variability decreased as speed increased; 5) power patterns (torque x angular velocity) were invariant across speeds, though gain increased with higher power generation with increased speed; and 6) electromyography (EMG) profiles also showed consistent timing, with increased amplitude with increased speed.

More recently, researchers have examined individual muscle function using EMG across a wider range of speeds. Activity from eight lower extremity muscles was recorded in young adults walking at preferred to very slow (.06 m/s) speeds. Results showed that, overall, amplitudes were decreased but the phasic timing varied little across speeds. At the slowest speeds, there were unusual bursts of activity that were attributed to increased demands on postural control, as normal walking became more a series of static postures (Den Otter, Geurts, Mulder, & Duysens, 2004). Similarly, EMG patterns were examined in very slow (.03, .02 m/s) over ground, and treadmill walking (Nymark, Balmer, Melis, Lemaire, & Millar, 2005). Results showed that while amplitudes decreased for all muscles, phasic timing was preserved in distal muscles, but proximal muscles exhibited coactivation.

In a simulation study, researchers found that a combined set of muscle activations (gluteus maximus and medius, vasti, hamstrings, gastrocnemius and soleus) contributed to both support and progression across speeds ranging from very slow to fast, with increased activity as speed increased (Liu, Anderson, Schwartz, & Delp, 2008). Similarly, in a forward dynamics simulation study, results showed that the functional role of individual muscles did not change from preferred to fast speeds, as hip and knee extensors contributed to support, while propulsion was provided primarily by soleus and rectus femoris in late stance (Neptune, Sasaki, & Kautz, 2008).

Taken together, these data indicate that proximal hip and knee moments tend to be more variable than distal ankle moments. In addition, EMG amplitudes and powers increase with speed, while slower speeds result in disruption of normal timing with more coactivation of proximal muscles, but the functional role of muscles is consistent across speeds.

Summary of Trunk and Center of Mass Motion

There is considerably less research examining the motion and muscle activation patterns of the trunk during walking. Trunk movements in both the frontal and sagittal plane were examined in adults walking at speeds from 1.0 - 2.5 m/s on a treadmill. During one stride, the trunk exhibited two vertical oscillations, two anteroposterior oscillations, and one oscillation in the frontal plane. Peak trunk tilt toward the stance limb occurred in single-support, and peak forward inclination occurred at the start of single support (Thorstensson, Nilsson, Carlson, & Zomlefer, 1984). Using a camera system and force platforms, researchers identified trunk movements relative to the gait cycle during over ground walking at preferred pace (Sartor, Alderink , Greenwald, & Elders , 1999). Results showed that global trunk motion in the sagittal plane fluctuated near neutral throughout the gait cycle. In the frontal plane maximum lateral flexion of the trunk toward the stance limb occurred during limb loading, and flexion away from the stance limb at toe off (which is loading for the contralateral limb). In the transverse plane, trunk motion was opposite to pelvic motion (counter-rotation); maximal rotation toward the reference limb occurred at initial contact, and rotation away at toe-off. The authors suggest that these events, lateral flexion, counter-rotation and extension, aid in forward progression and may help to reduce energy costs.

A combination study examined both trunk kinematics and kinetics in subjects walking at 1.0 and 2.0 m/s on a treadmill (Saunders, Schache, Rath, & Hodges, 2005). The kinematic findings were similar to previous findings as maximal lateral flexion occurred shortly after initial contact, and the timing of this peak did not change with increased speed. In the transverse plane, peak counter-rotation occurred at initial contact and the amplitude of this counter-rotation also increased with speed. The EMG results showed that trunk muscle activity was generally low at both walking speeds, a few abdominal muscles were tonically active, while others and the paraspinals were phasically active, reaching peaks after contact of each foot. Additionally, EMG activity increased with speed.

Specifically examining movement of the COM, researchers found that while the COM translated forward during walking, it also moved in a sinusoidal pattern with two maxima in the vertical direction, and one major oscillation in the lateral direction, such that the COM reached its highest and most lateral point in mid-stance and its lowest point during double support (Farley & Ferris, 1998). Extending these findings, researchers examined COM motion across a range of speeds. Results showed the vertical excursion of the COM increased as speed increased, but the mediolateral excursion decreased as speed increased, thus at slow speeds there were large mediolateral displacements of the COM. At preferred speed, the average mediolateral excursion was 3.29 cm, while the average vertical excursion was 4.89 cm. The authors concluded that slow walking may place a greater demand on balance systems due to the larger mediolateral displacements of the COM (Orendurff et al., 2004).

Beyond kinematics and kinetics, researchers have examined the coordination between the trunk and pelvis using analyses of continuous and discrete relative phasing (van Emmerik & Wagenaar, 1996; Wagenaar & Beek, 1992). Their results showed that the pelvis and trunk changed from an in-phase pattern to an out-of-phase pattern with increasing speed (consistent with the previous discussion on counter-rotations findings), that optimal coupling of the trunk and pelvis occurred at velocities greater than .75 m/s, and that coordination patterns were unstable (highly variable as measured by standard deviations) at slow speeds between .5 and 1.0 m/s (van Emmerik & Wagenaar, 1996).

Researchers have suggested that control of the trunk (and thus head control) is prioritized by the nervous system (Patla, Adkin, & Ballard, 1999; Winter et al., 1993), as the HAT segment is kept virtually vertical throughout walking, despite the destabilizing forces. Recently, researchers have examined the sensitivity of body segments to local perturbations (local divergence exponents) and the ability of segments to return to their average pattern after a perturbation (maximum Floquet multipliers). Results of a recent study showed that in young adults the trunk is less sensitive to small perturbations; however the trunk is slower to return to the average pattern after a perturbation than the thigh, shank or foot (Kang & Dingwell, 2009b). The authors suggested that the results were possibly due to the greater inertia of the trunk, and that slower correction of the trunk would make feedback control difficult, and feed-forward control advantageous.

These data indicate that upright control of the trunk may be prioritized. Additionally, balance demands at slower speeds are greater as mediolateral excursions increase and coordination dynamics become progressively unstable.

Neural and Sensory Contributions to Gait

Walking is a complex motor act that requires interaction between the brain and spinal cord, with the final motor output being influenced by sensory feedback from the periphery (Rossignol, Dubuc, & Gossard, 2006). Globally, the spinal cord is critical for generating the rhythmical patterning of the lower extremities, while subcortical and cortical centers provide the drive, sensory integration, decision-making, and planning necessary for skilled walking (i.e. walking in real-world environments).

Results of studies in both animals and humans has shown that central pattern generators (CPGs) located in the spinal cord are responsible for the production of the rhythmic, alternating flexion/extension movements of the lower extremities (Grillner & Wallen, 1985). Patients with incomplete spinal cord injuries exhibit involuntary rhythmical movements of the legs (Bussel et al., 1988; Bussel, Roby-Brami, Neris, & Yakovleff, 1996; Calancie et al., 1994), and spontaneous motor patterns in the legs have been observed in complete spinal cord injuries (Nadeau, Jacquemin, Fournier, Lamarre, & Rossignol, 2010).

Additional evidence for human CPGs is shown by the presence of fetal movements (Rayburn, 1995) and infant stepping behavior (Forssberg, 1985). Briefly, CPGs are described as functional neural networks that connect right/left halves of the spinal cord, so that when the extensors are activated in single-limb support, flexion for the swing limb is facilitated (and extension of the swing limb inhibited) (Duysens & Van de Crommet, 1998). While CPGs are responsible for the rhythmic alternating activity, they are not able to function in isolation, receiving descending information from the brain, as well as rich sensory information from the periphery, allowing for modification of the final motor output (Van de Crommet, Mulder, & Duysens, 1998).

The roles of cortical and subcortical structures in walking are currently areas of extensive research, and much of what is known has been from animal studies or examination of individuals with gait disorders. The use of neuroimaging in gait research has been problematic, as these techniques (fMRI, PET,EEG) require no motion of the head. Paradigms have been developed that allow for estimation of cerebral activity during gait by recording activity of motor planning processes prior to gait initiation, by using tasks that share common processes with gait, such as motor imagery of gait, and by imaging patients with gait disorders during rest (Bakker, Verstappen, Bloem, & Toni, 2007). Using EEG, participants were imaged as they heard an auditory stimulus and were waiting for another auditory cue to start walking. Results showed strong activity in the medial frontal cortex, indicating frontal cortex has a role in gait initiation (Yazawa et al., 1997).

A motor imagery study compared cerebral activity during imagery of standing, gait initiation, walking, and walking over obstacles (Malouin, Richards, Jackson, Dumas, & Doyon, 2003). Primary findings included increased activity in pre-supplementary motor area in walking compared to standing, and increased activity in supplementary motor area (SMA), parietal cortex, and parahippocampal gyrus during obstacle negotiation compared to walking. These findings indicate that networks that support gait extend beyond the primary motor and sensory areas of the cortex. Another recent study presented subjects with photographs of corridors where the prescribed walking path varied in width and length (Bakker et al., 2008). Subjects were asked to imagine themselves walking those paths during imaging. Results showed significant differences in the narrow path condition, with an increase in activity in parietal and occipital areas. Together these studies demonstrate there is increased activation and recruitment of areas outside primary motor areas when the imagined walking task requires spatial accuracy or balance control.

There are neuroimaging techniques available that measure cerebral activity during actual gait, including nuclear scanning and near-infrared spectroscopy (NIRS). Nuclear imaging is an invasive technique involving injection of a radioactive labeled substance prior to walking, with imaging completed after walking. Using this approach, researchers found increased activity in supplementary motor area, primary sensorimotor areas, cerebellum, and visual cortex during actual walking compared to rest (Fukuyama et al., 1997). A later study using this approach found similar results, but also found increased brainstem activation compared to a rest condition (Shibasaki, Fukuyama, & Hanakawa, 2004).

The advantage of NIRS is that head movements are allowed and there is no radioactive tracer. Rather, sensors on the scalp detect changes in blood flow. Using this approach, cerebral activity was measured during actual gait, alternating foot movements, arm swings, and motor imagery of gait. Primary findings were increased activity in medial sensorimotor areas and the supplementary motor area (Miyai et al., 2001). A later study using

NIRS examined the effect of walking speed on cerebral activity. Results showed increased activity in prefrontal and premotor areas as speed increased, while activity in medial sensorimotor areas did not vary with speed (Suzuki et al., 2004).

Common areas of activation across studies were medial sensorimotor areas and supplementary motor area. Additionally, these studies consistently showed increased recruitment beyond primary motor and sensory areas when the difficulty of the gait task was increased.

The output from cortical primary motor areas descends to influence the spinal cord and CPGs. Though the cerebellum does not directly connect to the spinal cord, it has significant influence on spinal neurons. The cerebellum is a major processor of sensory information, comparing motor commands from the cortex about the intended movement with the sensory information it receives regarding the actual movement (Brooks, 1984).The cerebellum receives afferent proprioceptive information from the periphery, information regarding the output of the CPGs, as well as visual and vestibular inputs. Thus it plays an important role in error detection and correction and regulates activity in all the descending pathways to the spinal cord. The functional outcome of cerebellar input is improved inter- and intralimb coordination during walking (Timmann & Horak, 2001). Individuals with cerebellar lesions exhibit both spatial and temporal variability during stepping and poor balance (Timmann & Horak, 2001). Recently, a study examining locomotor adaptations in cerebellar patients found that cerebellar damage did not impair the ability to make reactive feedback-driven motor adaptations, but significantly disrupted predictive feedforward motor adaptations (Morton & Bastian, 2006).

The basal ganglia are a group of subcortical structures that receive input from most areas of the cortex and send output back to the motor cortex. The basal ganglia are thought to play an important role in initiating (due to connections with the limbic system) and regulating motor output involved in standing balance, gait, and also in overall fluidity and sequencing of movement (Marsden, 1982). The two primary diseases of the basal ganglia, Parkinson's disease and Huntington's chorea, lead to serious motor control disabilities and gait impairments. Studies have shown much greater stride-to-stride variability in both of these disease processes (Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998), and our own work has shown impairments in trunk coordination during gait even early in the Parkinson's disease process (Lowry, Smiley-Oyen, Carrel, & Kerr, 2009).

It has been hypothesized that walking is organized in a top-down mode, so that control of gaze is optimized (Di Fabio & Emasithi, 1997). During walking the relationship between the performer and the environment is ever-changing, and successful negotiation of different terrains and obstacles relies on continually updated visual information about the relationship of the body to objects in the environment. Patla and colleagues (Patla, 1997) have extensively studied the role of vision during walking. In a series of studies, subjects wore liquid crystal glasses that obscured their vision, but were able to press a switch to make the glasses transparent when they wanted to visually sample their environment. Subjects walked in a straight path, on a winding path, with varying foot placements and negotiated obstacles. In general, they found that as the walking task required greater spatial accuracy there was increased visual sampling (Patla, Adkin, Martin, Holden, & Prentice, 1996; Patla & Vickers, 1997).

Persons with visual impairments walk more slowly, exhibited reduced step length, and spent more time in double-support (Beggs, 1991; Nakamura, 1997; Spaulding et al., 1994). Overall, vision provides information about the environment and body position and self-motion that the nervous system uses both in a feed-forward and online manner to modulate walking behavior (Patla, 1997).

While visual information provides body position and self-motion information, the vestibular system generates information about head position during walking. The three main components of vestibular function are; 1) the vestibulo-ocular reflex (VOR), which serves to stabilize gaze during head movements, 2) vestibulo-collic reflexes, which initiates neck movements to keep the head vertical, and 3) vestibulo-spinal reflexes, and vestibulospinal tract input, which triggers musculature in the neck, trunk, and extremities to keep the head and body upright (Highstein, 1996). Investigations into the role of the vestibular apparatus during walking have compared vestibular patients to normal controls. In general, studies have shown reduced walking speed and step length, increased step width, increased variability, and difficulty traversing uneven surfaces (Gauthier & Vercher, 1990; Marchetti, Whitney, Blatt, Morris, & Vance, 2008).

In addition to visual and vestibular information, afferent proprioceptive information from cutaneous receptors and muscle receptors play critical roles in regulating the timing of the locomotor rhythm, responding to perturbations (reactive control), and implementing adaptive strategies for predictive control. Regulation of the rhythmic motion of the lower extremities requires constant monitoring of muscle length via muscle spindles, and muscle tension via golgi tendon organs (GTOs). For example, research has shown that spindle and GTO input play an important role in triggering swing phase during locomotion (Grillner &

Rossignol, 1978; Pearson, 1995). As the hip is stretched into extension in late stance, muscle spindle activity is increased while GTOs are deactivated, and this combination of inputs may trigger swing phase (Pearson, 1995). Additionally, in early stance when the tibia is rotated over the foot, the stretch reflexes in plantarflexors are low to permit forward progression, while in late stance when the COM is in front of the foot the reflex is large and contributes to the contraction of the plantarflexors for push-off (Stein RB, 1991).

The first response to a gait perturbation is the monosynaptic stretch reflex triggered by the muscle spindles. Research has shown that these earliest responses, for example during slips or trips, are not phase-dependent and serve to increase joint stiffness (Schillings, van Wezel, Mulder, & Duysens, 2000). Longer latency responses are then organized that involve muscles of the whole body to maintain stability (Eng, Winter, & Patla, 1997).

In summary, while the spinal cord has hard-wired circuitry that is capable of producing the basic lower extremity rhythm, it is continuously receiving direct descending input from primary motor cortical areas, indirect input from the cerebellum and basal ganglia, and afferent proprioceptive information from muscles that regulate and modulate its activity. Together, these interactions ultimately allow for flexible and adaptable gait behavior.

Age-Related Changes in Gait

Age-related Changes in Spatiotemporal Variables

There has been extensive literature examining age-related changes in gait, the majority of which has concentrated on outcome measures, for example, speed, step length, cadence, and more recently the variability of these measures (Imms & Edholm, 1981; Murray, Kory, & Clarkson, 1969). The most consistent finding is that walking speed decreases with age (Elble, Thomas, Higgins, & Colliver, 1991; Imms & Edholm, 1981; Oberg, Karsznia, & Oberg, 1993), though maximum gait speed declines more sharply than preferred gait speed (Bohannon, 1997). Research has shown that reduced step or stride length with increased time in double support explained the reduction in preferred and maximal speed in older adults (Elble et al., 1991; Oberg et al., 1993; Winter, Patla, Frank, & Walt, 1990). The results for step width are not as clear. Some results have shown increased step width with age (Murray et al., 1969) or no change in mean step width (Gabell & Nayak, 1984; Hernandez, Silder, Heiderscheit, & Thelen, 2009). These age-related changes in walking patterns have generally been interpreted as the adoption of a more conservative gait (Woollacott & Tang, 1997).

A common method to analyze motor control error is movement variability. Researchers have described stride length and time, and the variability of these measures as more automatic patterning mechanisms, but stride width and stride width variability as balance mechanisms (Gabell & Nayak, 1984). Stride-to-stride fluctuations in length, time, and width have been investigated with inconsistent results. Early research indicated there were no differences in any variability measure between young and healthy older adults (though step width variability was greater than stride time or length variability for both groups), and proposed that increased variability was indicative of pathology (Gabell & Nayak, 1984). In contrast, recent studies reported that older adults had greater step width variability, but not step time or step length variability compared to young adults during preferred speed (Owings & Grabiner, 2004a; Owings & Grabiner, 2004b).

In a larger community-dwelling sample of older adults who walked at preferred and fast paces, results showed that for the whole sample, step width variability was greater than step length variability (Brach, Berthold, Craik, VanSwearingen, & Newman, 2001). Step length variability was highest, while step width variability was lowest in those older adults who walked the slowest. Additionally, older adults who had only minimal increases in walking speed in the fast condition exhibited greater step length variability and lower step width variability, while those who were able to substantially increase their speed exhibited greater step width variability at the faster speed. These findings indicate there are complex relationships between speed, variability, and function. These measures can have opposing relationships, and some amount of increased variability (particularly step width variability) may be adaptive.

In a follow-up study, 503 older community dwelling older adults were examined during over ground walking at preferred speed, and retrospective information about fall history was gathered (Brach, Berlin, VanSwearingen, Newman, & Studenski, 2005). Results showed that step length and step time variability did not differ with respect to fall history, however step width variability did. In older adults who walked slowly (< 1.0 m/s), step width variability was not related to falls, but in older adults whose preferred pace was > 1.0m/s and near normal, both high and low levels of step width variability were related to falls. These results indicate that moderate amounts of step width variability (between 7 and 30%) in healthy older adults may be an adaptive strategy to maintain balance.

These same researchers recently examined relationships between cognitive function (Mini-Mental State Examination, Trail making A and B, Digit Symbol Substitution Test), sensory function (vibratory perception, visual impairment), strength (grip strength, repeated chair stands), and gait variability (Brach, Studenski, Perera, VanSwearingen, & Newman, 2008). In older adults that walked slowly, greater cognitive impairment was associated only with greater stance time variability. In older adults that walked at normal speeds, sensory impairment was related to step width variability; however, better sensory function was associated with greater step width variability. These results indicate that variability measures are not equivalent, and may have different underlying causes.

Examination of gait variability between young and older adults is problematic if older adults are walking more slowly. While stride time variability is low and usually below 3% among young healthy adults (Hausdorff, Zemany, Peng, & Goldberger, 1999), stride time variability also increases as young adults walk more slowly (Beauchet et al., 2009). Thus, when comparing young and older adults, the increased variability observed in older adults may be due to slower walking speeds alone. Two recent studies have investigated this premise. In the first study, young and healthy older adults walked on a treadmill at speeds of 80-120% of their preferred speed and variability of spatiotemporal parameters, joint angles, and trunk motion was examined (Kang & Dingwell, 2008). Walking speeds did not differ between the groups, and both groups exhibited greater spatiotemporal variability at slower speeds. Older adults, despite similar walking speeds, had greater variability of stride time, step length, and mediolateral trunk motion. They found that strength and range of motion explained the age differences in variability.

In a large population-based study, 412 adults (60 to 86-years-old) walked over ground at their preferred speed, and variability of step time, length and width was measured (Callisaya et al., 2009). They found significant age effects for all variability measures; however, when they adjusted for gait speed, the step time variability age-effect was significantly reduced, while the age effect for step width variability persisted.

Taken together, these results indicate that the relationships between age, variability, and function are quite complex. In general, the literature supports age effects for both temporal and spatial variability; however, these age-effects are reduced, particularly for stride time variability, when speed or strength is accounted for.

Age-related Changes in Kinematics and Kinetics

In general, minor differences in joint angle profiles during gait between young and older adults have been reported (Oberg, Karsznia, & Oberg, 1994; Winter, 1991). Older adults exhibited slightly reduced dynamic range of motion at both the knee (Judge, Ounpuu, & Davis, III, 1996) and ankle (Kerrigan, Todd, Della, Lipsitz, & Collins, 1998; Oberg et al., 1994), while studies have shown either increased range at the hip (Winter, 1991), reduced peak hip extension (Kerrigan et al., 1998), or no age differences at the hip (Oberg et al., 1994).

Less is known about age-related changes in upper body motion. Recently, upper body motion and coordination in young and older adults was examined during treadmill walking at speeds ranging from .2 m/s to 1.8 m/s (van Emmerik, McDermott, Haddad, & van Wegen, 2005). Main findings were that older adults exhibited reduced pelvic rotation in all three planes, and less counter-rotation between the pelvis and trunk as speed increased.

The most consistent kinetic findings have been reduced ankle plantarflexion power during late stance with concurrent increased hip flexor power (DeVita & Hortobagyi, 2000; Judge, Davis, III, & Ounpuu, 1996; Kerrigan et al., 1998). Reduced ankle plantarflexion power was found to be the primary factor responsible for reduced step length in older adults (Judge et al., 1996). Additionally, older adults were able to increase step length at faster paces, though they appeared to substitute increased hip flexion power to compensate for reduced ankle power. These results were confounded by the slower walking speed in the older group (Judge et al., 1996).

A later study examined joint torques and powers in young and older adults walking over ground at the same speed and found similar results; older adults performed significantly more work at the hip and less work at the knee and ankle, though the total support torque was the same between groups (DeVita & Hortobagyi, 2000). Extending these findings, a recent study examined joint powers in speeds that ranged from 0.5 to 1.3 m/s (slow to preferred ranges) in young and older adults (Monaco, Rinaldi, Macri, & Micera, 2009). Joint powers increased with speed for both groups, and similar to previous findings, older adults exhibited greater hip and knee concentric power during stance phase with reduced ankle plantarflexion power. Together these results suggest an age-related redistribution of joint power during gait.

System Changes in Older Adults Impacting Walking Performance

One of the more consistent findings in aging literature is that strength declines with age. With advancing age, there is a generalized decline in muscle mass, accounting for ~ 25% of total bodyweight, in contrast to 43% in young adults (Serratrice, Roux, & Aquaron, 1968). This decline is primarily due to a reduction in the size and number of type 2 (fast twitch) muscle fibers (Clarkson, Kroll, & Melchionda, 1981; Larsson, Sjodin, & Karlsson, 1978; Lexell, Taylor, & Sjostrom, 1988; Tomonaga, 1977). There is evidence that this decrement may not be uniform, but is particularly pronounced in the lower extremity (Jennekens, Tomlinson, & Walton, 1971). This loss of muscle mass is consistent with age-related losses in strength.

Usual walking does not tax lower extremity muscle groups to their full capacity. The only muscle group that approaches its maximal capacity is the ankle plantarflexors (Winter, 1991). As previously discussed, decreased ankle plantarflexion strength and power contribute to reduced step length and walking speed in older adults. Numerous studies have found that lower extremity strength measures positively correlated with walking speed in older adults (Callisaya et al., 2009; Ringsberg, Gerdhem, Johansson, & Obrant, 1999). While strength training programs have resulted in increased walking speed in older adults (Schlicht, Camaione, & Owen, 2001), the role of lower extremity strength in walking balance or stabilizing upper body motion is not clear.

In the somatosensory system, reduced vibration sense, proprioception, and tactile sensation have been found in older adults (De et al., 1991; Kokmen, Bossemeyer, Jr., Barney, & Williams, 1977). These reductions have been associated with increased postural sway in standing (Brocklehurst, Robertson, & James-Groom, 1982; Lord, Clark, & Webster, 1991), and changes in automatic postural responses in standing and walking (Allum, Bloem, Carpenter, Hulliger, & Hadders-Algra, 1998).

Age-related changes in visual function have been well documented, including losses of smooth pursuit eye movements (Moschner & Baloh, 1994), sensitivity under low luminance conditions (Jackson & Owsley, 2000), visual acuity (Haegerstrom-Portnoy, Schneck, & Brabyn, 1999), contrast sensitivity (Nomura, Ando, Niino, Shimokata, & Miyake, 2003), and motion sensitivity (Trick & Silverman, 1991). As discussed earlier, vision is critical to anticipatory control and navigation during walking, and these age-related losses make navigation more difficult. Young and older adults were asked to walk across a straight path with and without specific foot placements while wearing specialized glasses that provided a view of the path only when the subjects pressed a button. Results showed that while there were no group differences in the number of times the terrain was sampled, older adults increased the duration of viewing during each sample in each condition, indicating older adults were more reliant on vision (Patla, 1992).

There is little research examining the role of vestibular function in walking performance in older adults. With advancing age, there is a gradual loss of vestibular hair cells (Rosenhall, 1973), a reduction in the diameter of the vestibular nerve (Bergstrom, 1973), and reduced gain of the vestibular-ocular reflex (Wall, III, Black, & Hunt, 1984). Several studies have demonstrated that vestibular contributions to ocular function break down among older adults (Demer, 1994; Paige, 1994). These results suggest that older adults become more reliant on the visual guidance of gait due to a diminished sense of balance.

Attention and Walking

There is evidence that walking demands higher levels of control processing with increasing age; that is, gait becomes less automatic with age (Woollacott & Shumway-Cook, 2002). Studies examining automation and walking have typically measured auditory reaction times while walking or used simple concurrent mental tasks (Beauchet et al., 2002). Young and older adults performed an auditory reaction time (RT) task while sitting, standing with both a normal and wide base of support, and while walking (Lajoie, Teasdale, Bard, & Fleury, 1996). During walking, the auditory probe was presented in both the single limb support and double support phases of gait. While the reaction time for both groups increased from sitting to standing to walking, older adults had slower reaction times overall. The simple dual-task had no effect on the gait parameters for either group. The authors concluded that walking is more attentionally demanding for older adults. The results of these studies are consistent with the premise that a postural hierarchy exists and attentional demands increase from sitting to standing to walking (Shumway-Cook, Woollacott, Kerns, & Baldwin, 1997).

In a more complex experiment, the cognitive demands of walking were investigated using a foot-targeting task that required subjects to place their foot while walking within a designated place on the floor (Sparrow, Bradshaw, Lamoureux, & Tirosh, 2002). In addition, while walking in either this constrained or unconstrained manner, subjects had to respond to an auditory stimulus, a visual stimulus, or both at the same time. Older adults had significantly longer visual and auditory/visual RT's in both unconstrained and constrained walking, suggesting greater dual-task costs for older adults in both normal walking, and when stepping has to be modified.

Examining the cognitive demands of obstacle negotiation, young and older subjects were asked to avoid stepping on a band of light that was suddenly projected across their walking path (Chen et al., 1996). In addition, they were asked to verbally respond to the turning on of a red light that was either synchronous or asynchronous with the presentation of the virtual obstacle. They found that both young and older adults had increased obstacle-contact rates with the dual-task; however the performance of the older adults was degraded significantly more. The authors suggested that this diminished ability to respond to environmental hazards when attention is directed elsewhere may be a factor in the increased fall rate among some elderly.

Successful interaction with our environment often requires that we engage in talking while walking, or that we engage in some other form of cognitive effortful processing such as rehearsal or comprehension. One study examined memorizing while walking (Lindenberger, Marsiske, & Baltes, 2000). First, young, middle-aged and older adults were trained to criterion to use a mnemonic technique. Then, subjects encoded word lists while sitting, standing or walking around either a simple oval track, or walking on a highly complex aperiodic track. Dependent variables included recall, walking speed, and walking accuracy (steps outside of the path). They found that with increasing age, subjects showed poorer recall when they were encoding while walking, versus sitting or standing. In addition, the middle-aged group exhibited reductions in walking speed in the dual-task condition, while the older group showed decrements in both speed and accuracy while engaged in encoding.

Together, these studies indicate that walking does become more attentionally demanding with age. This suggests that even in normal, unobstructed walking, older adults are allocating more attentional resources to the control of walking. In light of the resource reduction model of cognitive aging, this implies that this allocation is occurring under an already reduced capacity. Thus, when older adults engage in a challenging concurrent task while walking (talking, memorizing, stepping over obstacles) that requires additional attentional demands, the limits of capacity may be stressed resulting in decrements in walking, the other concurrent task, or both.

The Use of Trunk Accelerometers in Gait Analysis

Overview of Accelerometry

Accelerometers have been used for decades to measure human movement. Recent advancements in accelerometer technology, such as smaller size, improved accuracy, portability and low cost have resulted in increased interest in their use for gait analysis. As accelerometers directly measure body accelerations, validity has not been a focus of research; however, one study compared output from several uniaxial accelerometers over the thigh and shank with accelerations determined from a 3D camera-based motion analysis system and found high correlation coefficients of .98 ± .02 between the two systems (Mayagoitia, Nene, & Veltink, 2002).

Trunk triaxial accelerometric gait analysis has shown high test-retest reliability with ICC (3,1) values ranging from .77-.96 for mean accelerations for the three axes (Henriksen, Lund, Moe-Nilssen, Bliddal, & nneskiod-Samsoe, 2004; Moe-Nilssen, 1998b) with no change in reliability when walking on an uneven surface (Moe-Nilssen, 1998b). Additionally, researchers found high inter- and intra-examiner and stride-to-stride reliability for trunk acceleration signals across a range of walking speeds (coefficient of multiple determination, a waveform repeatability statistic) ranged from 0.60-0.95 (Kavanagh, Morrison, James, & Barrett, 2006).

The acceleration signal can be analyzed in many different ways: 1) examination of the maxima and minima of the acceleration signal (Kavanagh, Barrett, & Morrison, 2004; Zijlstra & Hof, 2003); 2) RMS transformed data, which is essentially the standard deviation (Menz, Lord, & Fitzpatrick, 2003b); 3) harmonic analysis, which assesses movement smoothness or rhythmicity (Menz et al., 2003b; Yack & Berger, 1993); 4) power spectral analysis (Kavanagh, Barrett, & Morrison, 2005); 5) autocorrelation analysis, which measures signal repeatability (Moe-Nilssen & Helbostad, 2005); 6) approximate entropy analysis (ApEn), which assesses the degree of repeatable patterns or randomness in the signal (Kavanagh et al., 2005); 7) cross ApEn analysis, which measures the degree of synchrony or coupling between two acceleration signals (Kavanagh et al., 2005); 8) local divergence exponents, which assess the sensitivity of a segment to small, continuous local perturbations, and Floquet multipliers, which assess the ability of a segment to return to its average pattern following a perturbation (Dingwell & Cusumano, 2000; Dingwell & Kang, 2007).

Normal Acceleration Patterns during Gait

Recently, accelerometers have been used to directly measure anteroposterior (AP), vertical (VT) and mediolateral (ML) accelerations of the upper body during walking. In the AP and VT directions, accelerations of the head and trunk exhibit a highly repeatable, bi-phasic pattern over one stride (Kavanagh et al., 2004; Moe-Nilssen, 1998a; Menz et al., 2003b), with peak frequency at approximately 2 Hz, which coincides with step frequency and accounts for most of the total signal power (Kavanagh et al., 2005). In both AP and VT directions, basic patterns of positive and negative accelerations are observed. In the AP direction, prior to heel contact, the trunk has been accelerating in a positive anterior direction. Heel contact results in a brief, rapid backward or negative acceleration. Following heel contact, the trunk again experiences positive anterior acceleration until forefoot loading, when there is a rapid negative acceleration until heel lift of the opposite limb. Following heel lift, there is gradual anterior positive acceleration throughout swing as the body moves forward (Menz et al., 2003b; Zijlstra & Hof, 2003). In the VT direction, there is rapid upward positive acceleration at heel contact, which declines until forefoot loading. After loading there is smaller positive peak. From midstance to toe-off there is gradual negative acceleration. After toe-off there is gradual positive acceleration until the next heel contact (Menz et al., 2003b; Zijlstra & Hof, 2003).

In contrast, ML accelerations exhibit a mono-phasic pattern over one stride, with a dominant frequency of approximately 1 Hz. Additionally, the ML acceleration profile of the trunk is more complex, exhibiting multiple low amplitude peaks (Smidt, Arora, & Johnston, 1971; Kavanagh et al., 2005). In general, during support phase on one limb, the ML accelerations are to the opposite side (Zijlstra & Hof, 2003).

There is attenuation of both ML and AP accelerations from the trunk to the head, while VT accelerations are not as attenuated due to the rigid nature of the spine (Kavanagh et al., 2004; Menz et al., 2003b). In the ML direction, the complex patterns of accelerations are substantially reduced from the trunk to the head (Kavanagh et al., 2005). In addition to greater acceleration attenuation at the head, there is greater coupling of between acceleration directions at the head (Kavanagh et al., 2005). At the trunk, there is coupling between AP and VT accelerations, but not with ML accelerations, supporting independent mechanism of control in the ML direction (Bauby & Kuo, 2000).

In general, these findings indicate that the trunk acts as a low-pass filter to reduce and change accelerations before reaching the head, and that head stabilization may be prioritized to allow for optimal utilization of the visual and vestibular systems.

Age-related Changes in Acceleration-derived Measures

Older adults have generally exhibited reduced peak trunk and head accelerations (Kavanagh et al., 2004; Menz, Lord, & Fitzpatrick, 2003c). Specifically, results have shown reduced peak positive AP accelerations associated with push-off, and higher peak negative AP accelerations following heel contact. Cross-correlation analyses revealed shorter lag time between trunk and head accelerations in older adults (Kavanagh et al., 2004). The authors suggested these results are indicative of a more rigid trunk and the adoption of a cautious gait pattern.

Power spectral, ApEn and cross ApEn analyses for trunk and head accelerations revealed similar results for young and elderly groups (Kavanagh et al., 2005). Cumulative power plots revealed that both groups had rapid increases in power at 2 Hz for AP and VT accelerations, with a similar increase at 1 Hz for ML accelerations. Older groups did, however, exhibit greater power at higher frequencies in the trunk, while there were no differences for the head. While there were no age differences in ApEn at the head or trunk, and no differences for cross ApEn for paired trunk and head accelerations, there were age differences in directional coupling. Specifically, older adults had less directional coupling at the trunk, but greater directional coupling at the head than younger subjects. These results indicate that while there were structural differences in trunk acceleration patterns between groups, older adults maintained head stability.

Recently, results showed that the trunk was less sensitive to perturbations compared to distal segments in both older and young adults; however, older adults exhibited greater sensitivity (more local instability) at the trunk compared to young adults (Kang & Dingwell, 2009a). Taken together, these results indicate that trunk motion is a sensitive marker for the change in gait in older adults.

Harmonic Analysis of Acceleration Signals

Harmonic analysis of trunk accelerations was first used by Gage in 1964 (Gage, 1964), who used this method to examine the gait of healthy adults and amputees. Gage was the first to develop an *index of smoothness* of the walking pattern, which was later refined by Smidt et al (Smidt et al., 1971; Smidt, Deusinger, Arora, & Albright, 1977), who used this approach to compare normal gait, crutch-walking gait, and the gait of patients with orthopedic diagnoses.

As previously discussed, typical anterior-posterior (AP) and vertical (VT) acceleration patterns of the trunk (and head) during walking have repeatable biphasic signals that reflect the cyclical movement of the trunk during one stride. Due to the biphasic characteristic of the AP and VT acceleration patterns, frequency decomposition of the signal for each stride through Fourier analysis yields a dominance of the second harmonic and subsequent even harmonics. For the AP and VT planes, the even harmonics are the intrinsic harmonics (Zijlstra & Hof, 1997) and indicate the in-phase components of the signal; the even harmonics are the expected harmonics in a biphasic pattern. The odd harmonics comprise the extrinsic or out-of-phase components, representing irregular accelerations; odd harmonics should be minimal in the VT and AP planes in walking. A ratio can be calculated by dividing the even harmonics (summed amplitudes of the first 10 even harmonics) by the odd harmonics (summed amplitudes of the first 10 odd harmonics). Thus, for both the AP and VT planes, this ratio should be high if the even harmonics dominate the pattern and odd harmonics are small, which is expected in a healthy gait pattern.

Conversely, the mediolateral (ML) accelerations in walking is a monophasic pattern. Thus, what is expected is the dominance of the first harmonic and subsequent odd harmonics. In the ML plane, the odd harmonics are intrinsic and the even harmonics are extrinsic. Therefore, for the ML direction the harmonic ratio is calculated from a ratio of the odd harmonics divided by the even harmonics (Menz et al., 2003b).

As the even harmonics represent typical, regular accelerations, and odd harmonics represent irregular accelerations, the harmonic ratio quantifies the smoothness, or rhythmicity of trunk or head motion. By locating the accelerometer at the level of L2-3 (which has minimal angular displacement and transverse rotation) at the cross section of the midsagittal and axial planes, COM acceleration is approximated (Moe-Nilssen, 1998a). Thus, trunk harmonic ratios are an indicator of upper body control and provide information regarding dynamic balance during gait.

Harmonic Analysis in Young Adults

Harmonic ratios have been examined in young adults walking at varying speeds and across different terrains. There are limited and conflicting data exploring the relationship between speed and harmonic ratios. Menz et al. (Menz et al., 2003b) studied six young adults walking at speeds from very slow to very fast. They reported that trunk harmonic ratios were highest at preferred speed, concluding that dynamic stability is optimized at a person's preferred walking speed. In a follow-up study (Latt, Menz, Fung, & Lord, 2008) this optimization was re-examined in ten young adults with walking speeds ranging from very slow to very fast, and during different manipulations of step length and cadence. They concurred with previous findings; trunk harmonic ratios in all three directions of motion were highest at preferred speed. Even with manipulation of step length and cadence, the VT and AP harmonic ratios were again highest at preferred values, while the ML harmonic ratio was maximized at preferred cadence but shorter-than-preferred step length. As statistical comparisons between speed conditions were not reported for either study, it is unclear whether these findings are reflective of a true optimization at preferred pace.

In contrast, Yack and Berger (Yack & Berger, 1993) examined HRs derived from upper trunk accelerations in nine young adults walking at slow, preferred, and fast speeds. In the VT and AP directions, HRs generally increased as speed increased, with significant differences between slow and fast conditions for both. These findings indicate some speed-dependency; however, direct comparison with Menz et al. (2003b) and Latt et al. (2008) is difficult due to differences in accelerometer placement and the more narrow speed range (no inclusion of a very fast condition). Multiple trunk acceleration measures, including harmonic ratios at the lower trunk, upper trunk, and head during slow, preferred, and fast walking were examined in eight healthy men (Kavanagh, Barrett, & Morrison, 2006). Results showed that mean VT and AP harmonic ratios were highest during fast walking, while there was no apparent change in ML-HR across speed conditions. This study also did not include a very fast condition. The data from these latter studies challenge a true optimization of HRs at preferred pace.

Harmonic ratios have also been compared while walking over ground and over an uneven surface. Young adults exhibited reduced harmonic ratios at both the trunk and head, except for ML smoothness which did not change across surfaces. These differences were observed despite no changes in walking speed between the two surfaces (Menz et al., 2003b).

These data indicate that there is not a one-to-one relationship between walking speed and trunk/head smoothness, and that harmonic ratios can provide information about coordination strategies beyond the information conveyed by spatiotemporal parameters.

Age-related Differences in Harmonic Ratios

Trunk acceleration patterns have been examined in young and older subjects while walking on a level and irregular walking surface (Menz et al., 2003c). Results showed that though older adults walked more slowly on both surfaces, and the magnitudes of their trunk accelerations were smaller, there were no differences in smoothness between the two groups for either surface (both groups reduced smoothness on the uneven surface).

Kavanaugh et al. (Kavanagh et al., 2005) examined preferred gait in young and healthy older men, using several acceleration-derived measures. In contrast to Menz et al. (2003b), although there were no age differences in AP or VT harmonic ratios, older men exhibited lower smoothness values in the ML direction, suggesting a specific lateral instability. In this study, differences in harmonic ratios were found, even though the groups walked at similar speeds.

A recent study found a different pattern of results from both of the above studies (Mazza, Iosa, Pecoraro, & Cappozzo, 2008). Trunk and head harmonic ratios were examined in young and older women walking at preferred and fast speeds. Older women walked more slowly in both conditions. At the level of the trunk, older women exhibited reduced AP and VT harmonic ratios; there were no age differences in the ML direction. At the head, older women exhibited reduced VT harmonic ratios only. There were also condition effects, where both groups exhibited reduced harmonic ratios at the faster speed.

Several studies have shown trunk acceleration measures are able to discriminate between fit and frail or unstable older adults. Yack and Berger (1993) measured trunk accelerations during preferred walking in young, stable elderly (those who reported no history of falls or unsteadiness) and unstable elderly (individuals with self-report of falls or unsteadiness). They found no differences in smoothness between young and stable elderly, but smoothness was significantly lower in the unstable elderly compared to the other two groups; however, it should be noted that the unstable elderly walked at a slightly lower cadence than stable elderly, while there were no differences in cadence between stable elderly and young.

Older adults with a low and high risk of falling walked at preferred pace over a level and an irregular surface (Menz, Lord, & Fitzpatrick, 2003a). They found that high risk older adults walked more slowly and had lower AP and VT harmonic ratios at both the trunk and head on the level surface. These differences were further pronounced on the irregular surface. Interestingly, there were no group differences for the ML-HR at the head on either surface. These findings suggest that adopting a slower gait pattern may be only beneficial up to a point, and that slower gait speeds may not ensure adequate control of the trunk or head.

Overall, these studies have provided interesting and at times conflicting information regarding age differences in harmonic ratios. On one hand, results have shown no age differences in trunk/head smoothness in an older group who walked more slowly than the young, while another study revealed a specific loss of ML smoothness in a group of older men who walked at similar speeds. Additionally, there were different patterns of results in studies that compared young and older men, versus young and older women. These differences may be due to the age range of the older adult samples, gender differences, or the way harmonic ratios were derived in each study. The commonality across all the studies using harmonic ratios is that this measure appears to provide unique information in regards to upper body control and coordination during gait, and together with spatiotemporal parameters offers a more comprehensive gait analysis.

Summary

There is extensive literature documenting age-related differences in spatiotemporal variables during gait. Age differences in kinematics and kinetics are also well documented, though this literature has focused on lower extremity changes. More recently, efforts have been made to understand upper body control and balance during gait. Accelerometry offers an attractive method to investigate upper body control during gait, as accelerometers are relatively inexpensive and can be used outside of the laboratory. Harmonic ratios are derived from acceleration signals, measure smoothness of motion, and are an indication of dynamic balance during gait. Acceleration signals have documented reliability, and harmonic ratios have demonstrated discriminant validity by revealing differences between young and older adults, between stable and unstable older adults, and changes with walking speed.

Every day walking involves both intentional speed changes, as well as speed changes that are normal adaptations to the context and environemt, i.e. walking on different terrains or surfaces, and walking while performing tasks such as talking or carrying. The speed-smoothness relationship has only been studied in smaller samples of young adults, with results supporting either optimization of trunk smoothness at preferred speed, or sustained smoothness at faster walking speeds. Other than the study by Mazza et al. (2008), who compared trunk smoothness in young and older adults at preferred and fast speeds, all previous literature examining age-related differences in trunk smoothness has examined preferred pace only.

This series of studies was designed to extend previous research by examining the effect of speed on trunk smoothness in a larger sample of young adults during overground walking at five self-selected speeds from very slow to very fast, and examining the effect of speed on trunk smoothness in healthy older adults using the same self-selected speeds. To address the conflicting results of previous comparisons of walking smoothness between young and older adults, healthy 60-year-olds were examined separately from healthy 80-year-olds. Additionally, strength measures and typical spatiotemporal parameters were examined to explore the relationships between trunk smoothness and common clinical measures of gait and functional mobility.

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CHAPTER 3.TRUNK MOTION WHILE WALKING AT DIFFERENT SPEEDS: HARMONIC RATIOS AND SPATIOTEMPORAL VARIABLES

A paper to be submitted to *Human Movement Sciences*

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Abstract

Harmonic ratios (HRs), derived from trunk accelerations, assess smoothness of trunk motion during gait. There is limited and contrasting evidence examining the relationship of harmonic ratios and walking speed. Eighteen young adults walked overground at speeds ranging from very slow to very fast. Trunk acceleration signals were used to derive HRs in the anteroposterior (AP), vertical (VT) and mediolateral (ML) directions. Spatiotemporal parameters were also measured. Similar to previous research, lower VT and AP-HRs were found at slow speeds. In contrast to previous research, optimization of HRs at preferred pace was not found. Rather, no differences were noted among preferred, fast, and very fast conditions for either the VT or AP-HRs. The ML-HR exhibited a different pattern of response, changing much less across speeds. At slower speeds, the VT and AP-HRs were inversely related to COV of stride time and length, while the ML-HR exhibited minimal associations with spatiotemporal variables. These data demonstrate that trunk smoothness was sustained at higher speeds in young adults. One explanation for this preserved smoothness was the reduced stride time variability. That the ML-HR was less influenced by speed suggests a prioritization of control in this direction.

Introduction

Harmonic ratios (HRs), derived from trunk and head acceleration measures, offer insight into underlying mechanisms of balance control during gait. By measuring the rhythm of accelerations, HRs provide information on control of the trunk and head during walking (higher HRs correspond to greater smoothness and rhythmicity), thus indicating whole body balance and coordination during gait. Additionally, HRs offer unique information by distinguishing VT, AP ML directions of motion. As HRs become more widely used to examine gait in clinical populations, it is critical to better understand how HRs change across a wide range of walking speeds in younger, healthy individuals.

Harmonic ratios derived from lower trunk accelerations have effectively discriminated between healthy and unstable or frail older adults and between clinical populations and healthy controls during preferred walking (Lowry, K. A., Smiley-Oyen, A. L., Carrel, A. J., & Kerr, J. P., 2009; Menz, H. B., Lord, S. R., & Fitzpatrick, R. C., 2003a; Yack, H. J. & Berger, R. C., 1993). However, to expand the use of HRs beyond preferred pace walking, the relationship between speed and trunk motion requires further investigation.

There are limited and conflicting data exploring this relationship. Menz, Lord and Fitzpatrick (2003b) studied six young adults walking at speeds from very slow to very fast. They reported that mean lower trunk HRs were highest at preferred speed, concluding that dynamic stability is optimized at a person's preferred walking speed. In a follow-up study (Latt, M. D., Menz, H. B., Fung, V. S., & Lord, S. R., 2008) this optimization was re-examined in ten young adults with walking speeds ranging from very slow to very fast, and during different manipulations of step length and cadence. They concurred with previous findings; trunk HRs in all three directions of motion were highest at preferred speed. Even with manipulation of step length and cadence, the V- and AP-HRs were again highest at preferred values, while the ML-HR was maximized at preferred cadence but shorter-than-preferred step length. As statistical comparisons between speed conditions were not reported for either study, it is unclear whether these findings are reflective of a true optimization at preferred pace.

In contrast, Yack and Berger (1993) examined HRs derived from upper trunk accelerations in nine young adults walking at slow, preferred, and fast speeds. In the V and AP directions, HRs generally increased as speed increased, with significant differences between slow and fast conditions for both. These findings indicate some speed-dependency; however, direct comparison with Menz et al. (2003b) and Latt et al. (2008) is difficult due to differences in accelerometer placement and the more narrow speed range (no inclusion of a very fast condition). Kavanagh, Barrett and Morrison (2006) examined multiple trunk acceleration measures including HRs at the lower trunk, upper trunk, and head during slow, preferred, and fast walking in eight healthy men. They found that mean VT- and AP-HRs were highest during fast walking, while there was no apparent change in ML-HR across speed conditions. This study also did not include a very fast condition. The data from these latter studies challenge a true optimization of HRs at preferred pace.

Recently, lower body motion in healthy young adults was examined during slow, preferred, and fast walking using trunk acceleration-derived measures of signal regularity (approximate entropy) and signal repeatability (coefficient of multiple determination)(Kavanagh, J. J., 2009). Results showed that trunk accelerations were less repeatable and regular at slow speeds compared to preferred, but there were no differences in these measures between preferred and fast walking. The author suggests that motor control is altered at slow speeds, particularly in the ML direction, resulting in irregular accelerations, but that regularity and repeatability of trunk motion is sustained at faster paces. This study did not include an examination of HRs or include very slow or very fast speed conditions.

Findings from robotic and prosthetic research support the premise that stability is sustained, if not enhanced, at faster speeds. Defining stability as the standard deviation of relative phase, dynamical systems researchers found that interlimb coordination between extremities improved with increasing speed (Donker, S. F., Beek, P. J., Wagenaar, R. C., & Mulder, T., 2001). Using this same index, researchers also found improved interlimb coordination at increasing speeds for a prosthetic group (Donker, S. F. & Beek, P. J., 2002). These authors argued that walking at faster speeds is accompanied by a reduction in stride length and frequency combinations (i.e., as speed increases both parameters will increase), and it is this reduction in variation that increases stability. Researchers developing walking robots have found that faster paces are more stable than slow paces (Hobbelen, D. G. E. & Wisse, M., 2007; Wisse, M., Hobbelen, D. G. E., & Schwab, A. L., 2007).

Taken together, general agreement exists that walking at slower speeds results in lower HRs compared to preferred pace; however, how faster paces influence HRs is unclear, with studies supporting either an inverted-U optimization curve with loss of smoothness at speeds away from preferred, or supporting sustainability of smoothness at faster walking speeds. Considering these contrasting observations, the primary purpose of this study was to examine HRs during overground walking in a larger sample of healthy young adults walking at speeds from very slow to very fast. Slower speeds were predicted to result in lower HRs and loss of smoothness, while trunk smoothness would be sustained at faster speeds.

Additionally, typical spatiotemporal parameters were correlated to HRs. Clinicians routinely use spatiotemporal variables to characterize gait, thus it is important to determine how these outcome measures relate to HRs, and if these relationships are directionally-specific.

Methods

Participants

Eighteen young adults (9 women mean age 22.11 ± 0 .84, 62.78 kg ± 8.98, leg length 86 cm ± 0 .03; 9 men mean age 24.04 ± 3.15, 79.65 kg ± 11.36, leg length 94 cm ± 0 .05) with no history of neurological diagnoses, head trauma, significant heart disease, significant musculoskeletal impairments or symptoms such as recent injury or pain, participated in this study. All participants signed an IRB-approved consent form, and were recruited and tested according to institutional review board procedures.

Instrumentation

A triaxial accelerometer (Crossbow CXLO2LF3, range ±2g) mounted to a plastic base plate on a gait belt and secured and aligned with L3 (Smidt, G. L., Arora, J. S., & Johnston, R. C., 1971) measured trunk accelerations in AP, VT, and ML directions. Prior to each data collection, the accelerometer was statically calibrated on a flat surface, with the output corresponding to -1g for the vertical axis, and zero for the orthogonal axes. After positioning the accelerometer on the lower trunk, it was leveled in both the frontal and sagittal plane prior to each walking trial. Accelerometer data were sampled at 200 Hz using a portable data logger (Crossbow AD2000 Ready DAQ) that participants wore in a small backpack.

Procedures and Experimental conditions

Participants walked over a 19m x 0.6m paper-covered walkway at five speeds using the same cues as in previous studies (Menz, H. B., Lord, S. R., & Fitzpatrick, R. C., 2003a; Moe-Nilssen, R., 1998): 1) walk very slowly (VS) as if in an art gallery, 2) walk slower (SL) than normal, as if there were ample time, 3) walk at preferred, comfortable speed (PF), 4) walk faster (F) than normal but not maximal speed, and 5) walk as fast as is safe without running (VF). Participants did not fixate on a target, but were told to 'look ahead' and avoid looking around the laboratory. Two markings were made to designate the middle 12.5m of the walkway for determination of walking speed. Three consecutive trials were performed in each condition. Participants were first asked to walk at their preferred pace followed by SL, VS, F, and VF. We chose this order so that preferred pace would not be contaminated by the other conditions.

Gait Variables

The primary dependent variables were AP-, VT-, and ML-HRs. Full descriptions of HR theory and determination are reported elsewhere (Menz, H. B., Lord, S. R., & Fitzpatrick, R. C., 2003c; Yack, H. J. & Berger, R. C., 1993). Briefly, harmonic content of trunk accelerations signals was derived using Fast Fourier Transform, and HRs were calculated in the VT and AP directions as the ratio of the sum of the first 20 even harmonics/sum of the first 20 odd harmonics, and in the ML direction as odd/even. Harmonic analysis was applied to all acceleration data with custom Visual Basic software using National Instruments Measurement Studio™ 6.0 libraries. A low-pass second-order Butterworth filter with a cutoff frequency of 21 Hz was applied to the raw acceleration data prior to stride segmentation. Stride segmentation was determined by identifying local maximum deceleration points in the vertical axis. Maximum deceleration candidate points were determined from the negative-going zero-crossings in the first-order derivative of the filtered data. Each stride was classified as consecutive maximum deceleration points (heel strike to heel strike of the same foot). The 'true' heel strike points were then found using a localized search about each point for the maximum deceleration point in the original raw data. These strides were then used for determining an HR per stride.

In addition, the spatiotemporal variables measured were: *speed (m/s)*, using a stopwatch to record time to walk the middle 12.5m; *stride time (seconds)*, based on the number of samples in the acceleration signal between consecutive heel strikes of the same foot; *stride length (m)*, derived from the acceleration data using procedures outlined by Zijlstra (2004) using a general correction factor of 1.75; *step width (cm)*, measured in a manner consistent with previous research (Stolze, H., Kuhtz-Buschbeck, J. P., Mondwurf, C., Johnk, K., & Friege, L., 1998). A felt pad was fixed to the bottom of each shoe, placed consistently at the intersection of the midpoint of the widest measurement of the shoe and 25% of the length, generally at the ball of the foot. Prior to the last trial for VS, PF, and VF, the pads were inked red for right, and black for left, and step width was collected only for those trials. In addition, the *coefficient of variation* (*COV*, standard deviation/mean*100*)* was used to quantify the variability of stride time, stride length, and step width.

Data Reduction and Analysis

The first trial was considered practice and values were averaged across trials 2 and 3 (except for step width). Prior to calculation of means, the trials were visually inspected to determine if the program correctly selected strides. To avoid acceleration and deceleration effects, the first 2 and last 2 strides were removed from the acceleration data.

Four Repeated Measures MANOVAs were used to examine speed condition main effects for 1) HRs, 2) means of the spatiotemporal variables across five speed conditions, 3) COVs of stride length and time across five speed conditions, and 4) step width and COV of step width across three speed conditions. If sphericity was met (Huynh-Felt > .75), the multivariate Wilks' Lambda was examined for signficance. For Huynh-Felt < .75, the multivariate Pillai's Trace was interpreted. If the multivariate test was significant, all significant univariate ANOVAs were interpreted using Bonferroni-corrected pairwise comparisons. Pearson *r* correlations were used to examine the relationships between HRs and spatiotemporal parameters. All statistical analyses were performed using SPSS15. Due to conducting four MANOVAs, the alpha level was adjusted to .0125.

Results

Spatiotemporal Variables

Data for all spatiotemporal variables for each speed condition are presented in Table 1. The MANOVA for the means of the spatiotemporal variables across five speeds was significant, $F(12,6)=177.33$, $p<.001$, as were all univariate ANOVAs ($p<.001$). As expected, stride length increased with speed while stride time decreased. Pairwise comparisons revealed that all speed conditions were significantly different from one another, with the exception that the increase in stride length from F to VF was not significant. The MANOVA for COV of stride time and length was significant, $F(8,10)=10.891$, $p<.001$. Variability for both stride length and time decreased with increasing speed. Pairwise comparisons for both variables revealed that variability was significantly greater at VS and SL speeds compared to F and VF speeds.

The MANOVA for the mean and COV of step width was not significanct (*p*=.069). However, it is noteworthy that step width variability increased by 8-10% in VF. Thus, as speed increased, variability in stride length and time decreased while variability in step width tended to increase.

Harmonic Ratios

The MANOVA for HRs was significant F(12,6)=7.329, *p*=.011, as were the three univariate analyses (*p*<.001). Pairwise comparisons for both the AP- and VT-HRs revealed that the lowest HRs occurred in VS and SL, and these values were significantly lower than at PF, F, and VF (See Fig. 1). There were no differences between PF, F, and VF in either the AP or VT directions. Pairwise comparisons for the ML-HR revealed that VS and SL were significantly different from F and VF; however, PF was not different from any other condition. These analyses suggest that speed did not influence HRs in the same way. In both the AP and V directions, there was a large increase from VS to PF (138 and 143%, respectively), while the change from PF to VF averaged only 15-16%. For the ML direction, there was only a 36% increase from VS to PF, and a 29% increase from PF to VF. Using only the data from VS, PF, and VF, we conducted post-hoc analyses using paired sample t-tests on the difference scores (PF-VS and VF-PF) for each HR. For both the AP- and VT-HRs the comparisons were significant (*p*< .001), while the differences for the ML-HR were not (*p*=.122).

In summary, for the AP and VT directions, trunk motion was less rhythmical and smooth at speeds slower than preferred, but not at speeds faster than preferred. In contrast, the ML direction appears to be less influenced by speed condition, as PF values were not different from any other condition.

Relationships between gait variables

Pearson *r* correlations are presented in Table 2. In general, at the slower paces all three HRs were related to each other, with the highest correlations between VT and AP. At preferred speed, the VT and AP-HRs remain correlated, but the ML-HR was not. At very fast speeds, AP and ML-HRs were moderately correlated. Examination of the relationship between HRs and spatiotemporal variables revealed significant associations at very slow speeds for both the VT and AP-HRs, most notably an inverse relationship between COV of stride time and length and HRs. In addition, the ML-HR was minimally associated with any spatiotemporal variable.

These data indicate that the relationship between HRs changes with speed, that increased stride time and length variability is associated with reduced VT and AP trunk smoothness at slower speeds, and that the ML-HR in particular is capturing unique information.

Discussion

In view of limited and equivocal findings regarding the relationship between HRs and speed, the purpose of this study was to examine how a range of walking speeds influences trunk rhythmicity and smoothness in healthy young adults. Previous studies using trunk acceleration measures found either an inverted U-shaped response, with optimization at preferred pace and loss of trunk smoothness at speeds faster and slower than preferred (Latt, M. D., Menz, H. B., Fung, V. S., & Lord, S. R., 2008; Menz, H. B., Lord, S. R., & Fitzpatrick, R. C., 2003b) or observed no loss, or even smoother trunk motion at speeds faster than preferred (Kavanagh, J. J., 2009). Note that the spatiotemporal data indicate our cues were effective in producing the desired speed effects (i.e, we found the expected increases in cadence and stride length and a decrease in stride time with increased speed).

Our results indicate that slower speeds in healthy young adults caused loss of trunk smoothness in all three directions (although the decrease was not significant for the ML-HR), but there were no significant differences between preferred, fast, and very fast in any direction of motion. These results are in contrast to findings of optimization of HRs at preferred speed with loss of trunk smoothness at both slower and faster speeds, but concur with others who found no loss of smoothness (Yack, H. J. & Berger, R. C., 1993), or greater smoothness when walking faster (Kavanagh, J., Barrett, R., & Morrison, S., 2006).

One possible explanation for the different results could be that balance mechanisms in our participants were not sufficiently challenged because they may have walked at slower speeds in the F and VF conditions. However, our speeds in the faster paces were similar if not faster than Latt et al. (in m/s, our data then Latt et al., respectively -- preferred: 1.38, 1.2; fast: 1.77, ~1.7 estimated from a graph; very fast: 2.38, 2.1).

Another explanation for the differences in the patterns of HRs between our data and Menz et al. (2003b) and Latt et al. (2008) could be individual differences in patterns of responses. Although overground walking at a range of speeds has ecological validity, these methods allow for greater variability between participants. Upon examination of all 54 individual speed-HR curves (18 participants x 3 HRs), only four individuals optimized HRs in the ML direction at preferred speed, seven in the AP direction, and two in the V direction. In fact, there was no single participant who maximized all three HRs at PF. Rather, the distribution for the highest HR for each direction was spread between PF, F, and VF speeds, with the one consistent pattern being reduced smoothness at slow and very slow speeds.

These findings support the premise that trunk smoothness is sub-optimal at slower speeds, but there is no clear optimization of HRs at preferred pace.

A more plausible explanation for our finding is the relationship between stride time and length variability and speed. Latt et al.'s (2008) explanation for optimization of HR at preferred speed was the quadratic relationship they found between timing variability and speed, with the lowest timing variability at PF and increasing variability at non-preferred speeds. They proposed that greater variability at non-preferred speeds led to irregular accelerations, which resulted in lower HRs. However, we did not find this same relationship between stride time variability and speed (whether analyzing COV or SD of stride time). In contrast, we found the lowest stride time and length variability in VF, which was significantly lower than the slower speeds as well as PF. These results are in agreement with others who found that the gait cycle has less temporal variability as speed increases.(Jordan, K., Challis, J. H., & Newell, K. M., 2007; Winter, D. A., 1983) In addition, our correlational analyses revealed an inverse relationship between the VT- and AP-HRs and stride time and length variability. Thus, although we found different patterns of results for the relationships between HRs, stride time variability, and speed, we agree with Latt et al. that stride time variability provides some explanation. One possible reason we did not see a decrease in trunk smoothness in the faster speeds was due to continued increases in stride time and stride length consistency. Though the analyses for step width and step width variability did not reach significance, it is interesting to note that while average step width changed minimally across conditions, step width variability increased at very fast speeds. It is possible that this increase represents a normal strategy to assist in balance control at faster speeds, thus contributing to sustained trunk smoothness at these speeds.

The correlational analyses revealed that the primary associations between HRs and spatiotemporal variables occurred at the slower paces, where higher stride time and length variability was associated with reduced VT and AP trunk smoothness. Research has shown increased stride time variability to be a characteristic of older adult and pathological gait (Schaafsma, J. D. et al., 2003), and has shown to be an independent risk factor for falls in older adults (Hausdorff, J. M., Rios, D. A., & Edelberg, H. K., 2001), thus these associations are not unexpected and augment the view that HRs are a measure of whole body balance and coordination. Though these associations were apparent at slower speeds, there were minimal associations between HRs and spatiotemporal parameters at other speeds, with essentially no findings for the ML direction. This suggests that HRs, in particular the ML-HR, is capturing unique information regarding gait control.

We found the response of the ML-HR to changes in speed was different from the VT- and AP-HRs (see Figure 1). In the ML direction, trunk smoothness at preferred pace was not significantly different from any other condition. These results concur with Kavanagh (2006) who found less change in ML smoothness across speeds. We suggest these data indicate a prioritization for mediolateral control. Results from biomechanical modeling suggest that while control of AP dynamic stability is ‘passive’ with sensory information from the limbs and lower level control being sufficient to stabilize motion, lateral balance control is ‘active’ with input from higher centers necessary for lateral stabilization of gait (Bauby, C. E. & Kuo, A. D., 2000). As higher level integration of sensory input is presumably not degraded in healthy young adults, we suggest that active lateral control is sustained across speeds.

Previous research using trunk accelerations has supported independent mediolateral control at preferred pace. Hernandez, Silder, Heiderscheit and Thelen (2009) found that AP and VT center of mass accelerations were correlated to each other but not to ML accelerations across a range of speeds. Kavanagh (2009), using measures of signal regularity and repeatability, found that VT and ML signals were less coupled at preferred speed; however, they found increased coupling at a faster walking speed. We found a similar pattern of response for HRs. At preferred pace, the VT and AP-HRs were correlated to each other but not to the ML-HR. Similar to Kavanagh, we found that the relationship between HRs changed across speeds. At the slower paces, all 3 HRs were related to each other (highest correlations between VT and AP), but at the faster paces the only significant correlation was between the AP and ML-HRs in the very fast condition. Kavanagh suggested the "increased coupling with the ML direction at faster paces indicates the system placing greater importance on regulation of global motion of the trunk in the frontal plane." We extend this premise by suggesting this change in control also occurs at very slow speeds, as both very slow and very fast paces add challenge to postural and balance mechanisms.

In summary, our main finding was that trunk smoothness is preserved during fast and very fast walking in healthy young adults. One explanation for this sustained smoothness is that stride time and stride length variability decreased as speed increased. In addition, the ML-HR exhibits less change across speeds, and lacks association with typical spatiotemporal variables, emphasizing the uniqueness of this measure. Further research is warranted to confirm these findings and extend our understanding of trunk control during gait. It may be that HRs are reflective of the combined spatiotemporal, kinematic, and kinetic activity during gait. An interesting future direction would be to examine kinematic and kinetic data together with HRs, or other trunk acceleration-derived measures, to better understand how the system controls global body motion during walking.

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Tables

Table 1. Means (SD) of spatiotemporal variables for each speed condition. Dashed lines indicate data were not collected during these conditions.

	Very Slow	**Slow**	**Preferred**	**Fast**	**Very Fast**
Speed (m/s)	0.58 (0.22)	0.96 (0.26)	1.38 (0.28)	1.77 (0.29)	2.38 (0.34)
Stride time (sec)	1.92 (0.62)	1.34 (0.24)	1.09 (0.09)	0.98 (0.07)	0.83 (0.07)
Stride length (m)	0.82 (0.19)	1.09 (0.22)	1.35 (0.28)	1.60 (0.27)	1.70 (0.17)
Step width (cm)	13.07 (2.17)	----	13.10 (2.38)	----	12.56 (1.70)
COV stride time (%)	5.05 (2.38)	4.09 (1.80)	2.52 (0.82)	1.79 (0.67)	1.64 (0.72)
COV stride length (%)	5.71 (2.24)	4.30 (1.64)	3.35 (1.67)	2.70 (1.61)	2.54 (1.35)
COV step width (%)	13.46 (4.34)	----	14.30 (4.56)	----	21.77 (9.75)

Table 2. Pearson *r* correlations between harmonic ratios and spatiotemporal variables. VT-HR = vertical harmonic ratio, AP-HR = anterior-posterior harmonic ration, ML-HR = mediolateral harmonic ratio. Dashed lines for step width and COV step width indicate data were not collected during these conditions. *p < .05, ** p < .01

	AP-HR	VT-HR	ML-HR	Speed (m/s)	Stride Time (s)	Stride Length (m)	Step Width (cm)	COV Stride Time (%)	COV Stride Length (%)	COV Step Width (%)
AP-HR										
very slow	---	**.817****	**.637****	.571*	**-.694****	.571*	.169	**-.813****	**-.621****	-.016
slow	---	**.818****	**.590****	.286	-.318	.296	---	-.034	-.506*	---
preferred	---	.546*	.459	.095	.105	.258	.092	-.38	-.041	-.014
fast	---	.023	.007	.087	-.158	.055	---	-.205	-.128	---
very fast	---	.121	**.642****	.221	-.055	.060	.221	-.083	-.094	-.175
VT-HR										
very slow	**.817****	---	.495*	**.742****	**-.735****	**.709****	.125	**-.715****	-.563*	.025
slow	**.818****	---	.429	.470*	-.389	.416	---	-.327	-.571*	---
preferred	.546*	---	.327	.408	-.244	.444	.377	-.338	-.063	-.011
fast	.023	---	-.057	.167	-.102	.411	---	-.133	-.586*	---
very fast	.121	---	-.107	-.558*	.497*	.113	-.006	-.036	-.549*	.349
ML-HR										
very slow	**.637****	.495*	---	.162	-.425	.191	.405	-.405	-.168	-.270
slow	**.590****	.429	---	.093	-.149	.103	---	-.148	-.313	---
preferred	.459	.327	---	-.142	.056	-.028	.214	-.246	-.197	-.247
fast	.007	-.057	---	-.355	.158	-.482*	---	-.222	.169	---
very fast	**.642****	-.107	---	.104	-.151	-.313	.429	.099	-.019	-.217

Figures

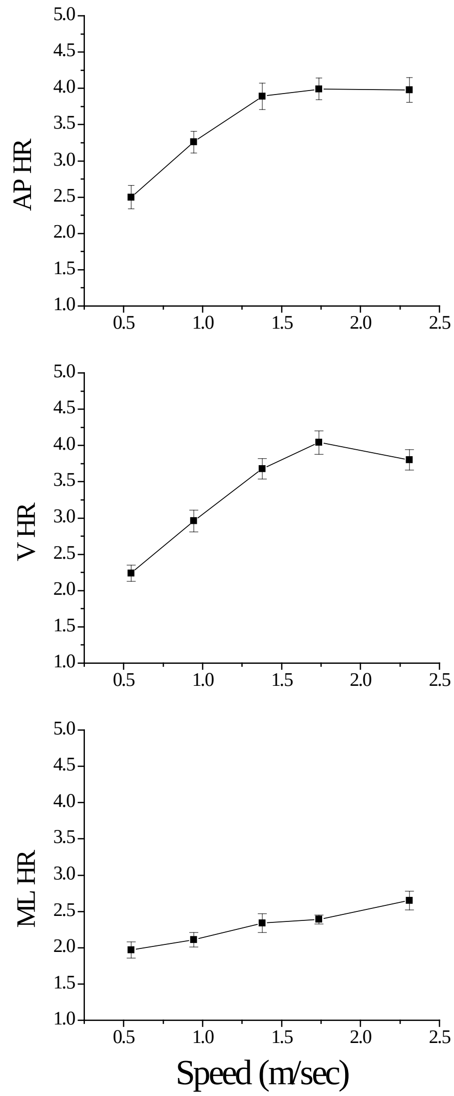

Figure 1. The effect of walking speed on harmonic ratios. AP-HR = anteroposterior harmonic ratio, VT- HR = vertical harmonic ratio, M- HR = mediolateral harmonic ratio. Bars represent standard error. The third data point indicates preferred walking speed.

CHAPTER 4. THE EXAMINATION OF TRUNK SMOOTHNESS DURING PREFERRED GAIT IN 20-, 60-, AND 80-YEAR-OLDS

A paper to be submitted to the *Journal of the American Geriatrics Society*

Kristin A. Lowry, Ann L. Smiley-Oyen, & Carolyn Siskoff

Abstract

Background: harmonic ratios (HRs), derived from trunk accelerations, measure trunk smoothness during gait; higher ratios indicate greater smoothness. Age-related differences in trunk smoothness have been equivocal, partially due to wide age-range in older adult samples. Objective: to examine age-related differences in HRs to better understand how aging affects trunk control during gait. Methods: anteroposterior (AP), vertical (VT) and mediolateral (ML) HRs and spatiotemporal variables were examined in 13 young adults (ages 20-23), 14 healthy older adults (ages 60-69), and 13 healthy old-old adults (ages 80-86) during preferred walking. Hip and knee strength testing was performed with a hand-held dynamometer. Results: there were no differences in HRs between young adults and 60-year-olds. Eighty-year-olds exhibited lower AP- and VT-HRs, but not lower ML-HRs. Both older groups had greater step width variability and reduced strength.

Conclusion: separation of age groups revealed different results for 60- and 80-year-olds. Sixty-year-olds maintained trunk smoothness, while 80-year-olds exhibited the greatest loss of smoothness in the AP direction, only partially explained by loss of strength. Lack of group differences in ML trunk smoothness suggests a prioritization of ML control. Greater step width variability in the older groups may be a strategy to maintain ML smoothness.

Introduction

Age-related changes in spatiotemporal gait parameters, including reduced speed and step length [1,2] are well-documented. Below certain criteria reduced walking speed is a powerful indicator of general ambulatory function and predictor of falls. But spatiotemporal measures do not provide insight into the underlying coordination of gait.

In contrast, examination of AP, VT, and ML trunk acceleration patterns provides insight into gait coordination. Harmonic content of trunk accelerations can be extracted using a frequency-domain analysis to measure smoothness or rhythmicity of gait [3]. This derived harmonic ratio represents the contribution of in-phase/regular accelerations divided by out-of-phase/irregular accelerations, with higher HRs corresponding to greater trunk smoothness and rhythmicity, thus providing an indication of whole body balance and coordination during gait.

No differences in HRs during preferred-pace gait were found between young and stable older adults (mean age 78 ±7) [4]. Similarly, no differences in HRs were found between young and low-fall risk community-dwelling adults (79.9 ±3) while walking on level and irregular surfaces [5], although the older adults did walk more slowly and with reduced magnitudes of accelerations. The authors concluded that older adults adopted a conservative gait pattern to maintain stability. In contrast, examination of preferred gait in young and community-dwelling older men (74 ±3) revealed no differences in VT- or AP-HRs, but lower ML-HRs [6]. Interestingly, HRs in healthy older women (72 ±4) compared to young women were lower in VT and AP directions, but *not* ML [7].

These equivocal findings may be due to differences in age as well as gender and health of the samples. There may be motor performance differences between community-dwelling older adults screened for major pathology [6] and those characterized as physically active [7]. Also, the age range of older adult samples may have spanned two to three decades. There is substantial evidence supporting somatosensory and neuromuscular decline with age, independent of pathology. For example, poorer clinical and laboratory-based balance scores were found in healthy old-old (88 ±5) compared to young old (72 ±3) participants [8]. Thus, interpretation of findings within and across these studies is difficult.

The primary purpose of this study was to examine age-related changes in HRs during preferred gait in young adults (YA), healthy older adults 60-69 years old (OA), and healthy old-old adults 80-89 years old (OOA). We hypothesized no differences in HRs between YAs and OAs, but lower HRs in all directions for OOAs compared to both YA and OA groups.

A secondary purpose examined the relationship between HRs, variability, and strength.

Since stride time variability is associated with fall risk [9], we hypothesized lower stride time variability would be associated with greater trunk smoothness. To date, no study has investigated the relationship between step width variability and HRs, or strength and HRs. The hip flexors/extensors are critical for maintaining dynamic balance of the head, arms, and trunk in the plane of progression, while the hip abductors are critical for frontal plane control [10]. As clinicians routinely use spatiotemporal parameters and strength measures in assessment and intervention, it is important to determine how these measures relate to HRs, if these relationships are directionally-specific, and if they change with age.

Methods

Participants

Thirteen healthy young, old, and old-old adults participated in this study (Table 1). Young adults were university students. Older adults were recruited from the local community. Reasons for exclusion included neurological diagnoses, history of head trauma, significant heart disease, significant musculoskeletal impairments, or symptoms such as recent fracture or joint replacement, severe chronic pain, peripheral neuropathy, cognitive decline, and use of a walking device. All participants signed an IRB-approved consent form, and were recruited and tested according to IRB procedures.

Instrumentation and Procedures

A triaxial accelerometer (Crossbow CXLO2LF3, range ±2g) mounted to a plastic baseplate on a gait belt and secured and aligned with L3 [3] measured trunk accelerations in three direction. The accelerometer was statistically calibrated on a flat surface, and once positioned on the lower trunk, was leveled prior to each trial. Data were sampled at 200Hz using a portable data logger (Crossbow AD2000 Ready DAQ) worn in a small backpack. Participants walked over a 19m x .6m paper-covered walkway at their self-selected preferred pace, and were told to ‘look ahead’ but were not given a fixation point. Three consecutive trials were performed, with trial 1 considered practice.

Maximal isometric strength of four lower extremity muscle groups (hip flexion, extension and abduction; knee extension) was assessed using a hand-held dynamometer (Lafayette MMT) using test positions and procedures as outlined in previous research [11].

To characterize the older groups, the Berg Balance Scale (BBS) [12], and Physical Activity Scale for the Elderly (PASE) [13] were administered.

Gait Variables

The primary dependent variables were AP-, VT-, and ML-HRs. Full descriptions of HR theory and determination are reported elsewhere [14-15] . Harmonic content of trunk accelerations signals was determined with Fast Fourier Transform using custom Visual Basic software from National Instruments Measurement Studio™ 6.0 libraries. HRs were calculated for each stride in the VT and AP directions as the ratio of the sum of the first 20 even harmonics/sum of the first 20 odd harmonics, and in the ML direction as odd/even.

Spatiotemporal variables were: *average speed (m/s)*, time to walk the middle 12.5m; *stride time (s)*, number of samples in the acceleration signal between consecutive heel strikes of the same foot*5ms/1000ms; *average stride length (m)*, calculated by multiplying the average speed by an averaged stride time; *step width (cm)*, measured as the distance between ink marks, consistent with previous research [16]. Felt pads were fixed to the bottom of each shoe, placed at the intersection of the midpoint of the widest measurement of the shoe and 25% of the length, generally at the ball of the foot. Pads were inked red for right, black for left; *coefficient of variation* (*COV*, standard deviation/mean*100*)* was used to quantify the variability of stride time and step width.

Data Reduction and Analysis

The first and last two strides were removed from each trial, and remaining middle-stride values were averaged across trials 2 and 3 (except for step width, which was only collected during the last trial). Two strength measurements were recorded for each muscle group and the maximal value was analyzed. Hip extension values are not reported due to failure of proper stabilization. Paired t-tests revealed no differences between left and right sides, so strength values were averaged and normalized for body weight and leg length.

Using SPSS 15, a series of one-way between group MANOVAs were conducted on HRs, spatiotemporal variables, and normalized strength data. Speed was used as a covariate in the first two analyses. If the significance level of Pillai's Trace was less than .05, the univariate analyses were interpreted and a significant Group effect was followed with Bonferroni adjusted pairwise comparisons. Partial eta-squared effect sizes were interpreted as .01(small), .06(medium), and .138(large) (17). Pearson *r* correlations tested relationships between variables.

Results

Participant Characteristics

There were no group differences for leg length or mass. (See Table 1 for all means). There were no differences in PASE scores between the older groups; both men and women scored above published norms for their age groups [13]. The OOAs exhibited lower scores on the BBS compared to OAs, but all scored above 50, indicating low fall risk. These data show both older groups were physically active, while the OOAs exhibited the expected age-related loss of dynamic standing balance.

Spatiotemporal Gait Variables

A univariate analysis on walking speed revealed no group differences *(P*= .704), though OOAs had the lowest average walking speed. The MANOVA on the remaining spatiotemporal variables was significant ($F_{10,64}$= 3.473, *P*= .001, Pillai's Trace= .704, η^2= .352). When considered separately, there were group differences for stride time ($F_{2,35}$= 8.65, *P*= .001, η^2= .331), average stride length ($F_{2,35}$= 6.386, *P*= .004, η^2= .267), and COV of step width ($F_{2,35}$= 10.972, *P*< .001, η^2= .385), where OOA exhibited shorter stride times and lengths. Both older groups exhibited greater step width variability compared to YA.

Harmonic Ratios

The MANOVA on the combined HRs was significant ($F_{6,68}$= 2.283, *P*= .046, Pillai's Trace= .335, η^2= .168). When considered separately, the only group difference was in the AP direction ($F_{2,35}$= 5.486, *P*= .008, η^2= .239) with OOAs exhibiting loss of smoothness compared to both other groups. There was a trend in the VT direction ($F_{2,35}$= 3.25, *P*= .051, η^2= .157), with the pattern of results similar to AP. No differences were found in the ML direction (*P*= .094). Interestingly, the OA group had the highest mean HR in the ML direction.

Normalized Strength

The MANOVA for strength measures was significant ($F_{6,70}$= 4.032, *P*= .002, Pillai's Trace= .514, η^2= .257) and univariate analyses revealed group differences for all muscle groups: hip flexion ($F_{2,36}$= 10.78, *P*< .001, η^2= .375); hip abduction ($F_{2,36}$= 12.53, *P*< .001, η^2=. 410); and knee extension ($F_{2,36}$= 5.60, *P*= .005, η^2= .237). For the hip muscle groups, pairwise comparisons revealed YAs had greater strength than both older groups. For knee extension, YA had greater strength than OOA.

Relationships between HRs and Strength

To determine if strength helped explain group differences in HRs, we conducted univariate analyses on each HR using a composite strength score (sum of the normalized strength measures) and speed as covariates (Table 2). Age effects were minimized for both VT- and AP-HRs when strength was added as a covariate; however, even after adjusting for both speed and strength, the effect of age remained significant for the AP-HR (*P*= .043). These findings suggest that age-related loss of AP smoothness can only be partially explained by losses in strength.

Relationships between HRs and Variability

With age, there were increased correlations between HRs, with OOAs showing significant correlations among all three directions (Table 3). Although not significant, all groups exhibited inverse relationships between stride time variability and all three HRs, with the strongest relationships with the AP-HR. Interestingly, there was an inverse relationship between the ML-HR and step width variability for YAs, while both older groups demonstrated a positive relationship.

Discussion

We investigated age-related changes in trunk smoothness during preferred gait. One unique aspect was separating healthy older adults into two groups, 60-yr-olds (OAs) and 80-yr-olds (OOAs). Consistent with our hypothesis, we found no reduction in trunk smoothness in OAs compared to YAs in any direction. In contrast, we found a loss of smoothness specifically in AP and VT directions in OOA.

The strongest age effect was in the AP direction, with hip and knee strength partially explaining the age differences. Coupling between head and trunk acceleration patterns was found to be highest in young adults and lowest in older adults in the AP direction, indicating a change in AP coordination [6]. Additionally, coupling between hip and knee moments was reduced in fit elderly, suggesting decreased ability to consistently control the trunk in the AP direction [18].

The effect of age was also significant for the VT-HR, but when speed was added as a covariate the age effect was minimized, and then lost when strength was added. In contrast, the significant age effect in AP trunk smoothness even after adjusting for speed and strength, suggests that lower AP-HRs may be partially due to declining sensorimotor control.

We found no age-related differences in trunk smoothness in the ML direction. One explanation is that aging does not affect ML smoothness; however, we think a better explanation is that ML control is prioritized. That OAs had the highest mean ML-HR, a pattern also found by others [7], may reflect an 'overcontrol strategy' [19], and supports prioritization of ML control.

In YAs we found higher associations between AP- and VT-HRs than with ML-HRs, supporting independent control of ML motion [20]. In contrast, OOA's HRs were highly correlated to each other, reflecting a more en bloc motion of the trunk. Young adults increased coupling between VT and ML accelerations at faster speeds, indicating a shift in strategy to ensure control of ML motion [21]. We suggest this strategy is employed by OOAs during preferred gait to help control ML motion. But this strategy may also degrade AP control, thus contributing to lower AP-HRs. Adjusting for speed and strength did not affect our ML results, and age explained the least amount of variance in the ML-HR. These data suggest that significant reductions in ML smoothness may be indicative of pathology [14,22].

Another unique aspect of this study was the examination of the relationship between HRs and variability measures. Consistent with previous research [23], we found no group differences in mean step width, but higher step width variability in both older groups. In older adults, a moderate amount of step width variability is thought to be adaptive to maintain dynamic balance [24], and has been related to higher scores of sensory function in older adults [25]. Interestingly, although not statistically significant, in YAs lower step width variability was associated with greater ML smoothness, while in both older groups the correlations were positive, indicating higher variability was associated with greater ML smoothness. We speculate that in the presence of age-related losses in hip strength, both older groups increased step width variability possibly to maintain ML smoothness.

While step width variability changed with age, stride time variability did not. There were modest inverse relationships (-.2 to -.4) between all HRs and stride time variability across ages. This indicates that higher stride time variability is associated with lower trunk smoothness in all directions of motion. This finding is consistent with our previous work [14] and consistent with findings that higher stride time variability is present in pathology [26] and predictive of fall risk [9].

A limitation of this study was the small sample size, thus lower statistical power, although we did find large, meaningful group effects in HRs and step width variability. Due to unequal number of men and women per group, we were not able to examine the effect of gender on trunk smoothness, thereby addressing different results between older men and women [6-7]. Interestingly, we had more women than men, and our OOA results were most similar to Mazza et al. [7] who tested only women. Future research is warranted to examine gender differences in HRs, particularly in advanced age.

We recently found evidence that HRs are responsive to specific gait cueing strategies in people with Parkinson's disease (Lowry, *in press*), suggesting that HRs are a promising research tool to investigate treatment efficacy in gait rehabilitation. And, given that gait speed improves after strength training [27], future research is warranted to investigate whether improvements in hip and lower extremity strength vs. specific sensorimotor gait retraining impacts trunk smoothness, and whether improved trunk smoothness reduces fall risk.

Key Points

1) Healthy 60-year-olds' trunk smoothness was similar to healthy 20-year-olds' while healthy 80-year-olds' smoothness was lower in the AP and VT directions.

2) Strength only partially explained group differences in AP trunk smoothness; strength and speed accounted for age-differences in VT smoothness.

3) Smoothness in the ML direction was not affected by age, indicating that control in the ML direction may be prioritized.

4) Both older groups exhibited greater step width variability, which may help to preserve ML smoothness.

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Tables

Table 1. Mean (SD) for subject characteristics, strength, spatiotemporal variables and harmonic ratios. * = group is significantly different from other two groups, $P < .05$; ƚ = groups are significantly different from each other, $P < .05$.

	Young Adults men (n=4) women (n=9)	**Old Adults** men (n=3) women (n=10)	**Old-Old Adults** men (n=4) women (n=9)
Age (yrs)	22.13 ± .9 (20-23)	66.34 ± 2.6 (60-69)	82.47 ± 2.2 (80-86)
Leg Length (m)	.90 ± .04	.93 ± .06	.88 ± .08
Mass (kg)	69 ± 12.9	77.6 ± 13.4	67.38 ± 10
Berg Balance Score	----	55.6 ± .8	52.9 ± 1.4
PASE	----	188 ± 75.4	173 ± 61.6
Normalized Strength(N/kg*m)			
hip flexion	3.34 ± .78 *	2.13 ± .92	2.04 ± .68
hip abduction	4.98 ± 1.0 *	3.57 ± 1.11	3.10 ± .87
knee extension	4.96 ± .86 ƚ	4.15 ± .92	3.78 ± .97 ƚ
Spatiotemporal Variables			
velocity (m/s)	1.33 ± .25	1.34 ± .13	1.28 ± .12
ave stride length (m)	1.42 ± .18	1.40 ± .10	1.30 ± .09 *
stride time (s)	1.09 ± .10	1.05 ± .07	1.02 ± .09 *
step width (cm)	13.38 ± 2.11	13.17 ± 3.13	13.61 ± 3.03
COV stride time (%)	2.48 ± .80	2.38 ± .72	2.94 ± 1.24
COV step width (%)	12.66 ± 3.27 *	22.90 ± 4.84	23.08 ± 9.58
Harmonic Ratios			
Anteroposterior	3.85 ± .81	3.80 ± .47	3.00 ± .75 *
Vertical	3.74 ± .54	3.65 ± .40	3.13 ± .83
Mediolateral	2.34 ± .49	2.58 ± .52	2.13 ± .43

Table 2. Comparison of univariate analyses using speed and strength covariates. ML-HR = mediolateral harmonic ratio, VT-HR = vertical harmonic ratio, AP-HR = anteroposterior harmonic ratio.

Effect	**ML-HR**	**VT-HR**	**AP-HR**
Age			
P value	.074	.034	.005
Adjusted R^2	8.7%	12.5%	20.9%
Age adjusted by speed			
P value	.094	.051	.008
Adjusted R^2	7%	19.7%	20.5%
Age adjusted by strength			
P value	.063	.138	.037
Adjusted R^2	11.3%	11.9%	24%
Age adjusted by speed and strength			
P value	.075	.123	.043
Adjusted R^2	8.7%	17.4%	22.5%

Table 3. Associations between HRs and variability measures. ML-HR = mediolateral harmonic ratio, VT-HR = vertical harmonic ratio, AP-HR = anteroposterior harmonic ratio. *p < .05, ** p < .01

	VT-HR	**ML-HR**	**COV stride time %**	**COV step width %**
AP-HR				
YA	.498	.275	-.299	-.048
OA	.058	.089	-.403	-.129
OOA	**.768****	**.630***	-.458	-.282
VT-HR				
YA		.040	-.389	.093
OA		**.574***	-.002	-.154
OOA		**.856****	-.298	.102
ML-HR				
YA			-.231	-.413
OA			-.142	.202
OOA			-.466	.293

CHAPTER 5. THE EFFECT OF WALKING SPEED ON TRUNK SMOOTHNESS IN HEALTHY 20-, 60-, AND 80-YEAR-OLDS

In preparation for *Journal of Gerontology: Medical Sciences*

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Abstract

Background: Harmonic ratios (HRs), derived from trunk accelerations, measure trunk smoothness during gait; higher ratios indicate greater smoothness. Previous research has shown that young adults optimize HRs at preferred pace, exhibiting reduced smoothness at speeds faster and slower than preferred. More recent studies using HRs and other trunk acceleration measures challenge this finding. It is not known how walking speed impacts HRs in healthy older adults. The objective of this study was to examine age-related differences in HRs across a range of self-selected overground walking speeds. *Methods*: Anteroposterior (AP), vertical (VT), and mediolateral (ML) HRs and spatiotemporal variables were examined in 13 young adults (ages 20-23), 13 healthy older adults (ages 60-69), and 13 healthy old-old adults (ages 80-86) while walking overground at very slow, slow, preferred, fast and very fast speeds. Participants also walked at speeds 20% faster and slower than preferred pace. *Results*: Young and older adults performed similarly across speeds, while old-old adults exhibited lower AP-and VT-HRs. All groups exhibited reduced HRs at speeds slower than preferred, with no differences in HRs between preferred and faster speeds, with the exception of lower VT-HRs in the very fast condition for the older groups. *Conclusion*: HRs did not show a clear pattern of optimization at preferred pace for any group. Though all groups had loss of trunk smoothness at slower speeds, older groups were not

disproportionately affected by walking more slowly, and all groups maintained trunk smoothness from preferred to fast speeds. HRs were related to stride time variability.

Introduction

Recently, trunk accelerometry has been used as an informative gait analysis technique. Various methods have been used to examine and compare acceleration signals, including analyses of peak values, normalized root mean square, autocorrelations, approximate entropy, and harmonic analyses. Harmonic content of trunk accelerations can be extracted using a frequency-domain analysis to measure smoothness or rhythmicity of gait (Smidt, Arora, & Johnston, 1971). The derived harmonic ratio represents the contribution of in-phase/regular accelerations divided by out-of-phase/irregular accelerations, with higher HRs corresponding to greater trunk smoothness and rhythmicity, thus providing an indication of whole body balance and coordination during gait. Additionally, HRs offer unique information by distinguishing VT, AP, ML directions of motion.

Understanding age-related locomotor changes and adaptations means assessing ‘real-world’ walking, which includes examining gait over a range of speeds and complexities. Previous studies examining the relationship between trunk harmonic ratios and walking speed in young adults found an inverted U-shaped response for all directions of motion, with optimization at preferred pace and loss of trunk smoothness at speeds faster and slower than preferred (Latt, Menz, Fung, & Lord, 2008; Menz, Lord, & Fitzpatrick, 2003a). The explanation for this optimization at preferred speed was the quadratic relationship found between timing variability and speed, with the lowest timing variability at PF and increasing variability at non-preferred speeds. Greater timing variability at non-preferred speeds led to irregular accelerations, which resulted in lower HRs.

Recently, these findings of optimization of harmonic ratios at preferred speed have been challenged. In a study examining young adults walking at self-selected speeds ranging from very slow to very fast, we found reduced HRs at slower speeds, but did not find optimization of HRs at preferred pace (*Lowry, in preparation*). Rather, we found no differences between preferred, fast, and very fast conditions for either the VT- or AP-HRs. The ML-HR exhibited a different pattern of response, changing much less across speeds. In addition, we did not find a quadratic relationship between stride time variability and speed; rather, we found the lowest stride time and stride length variability at the very fast speed. These results indicate trunk smoothness is sustained at higher speeds in young adults, and one explanation for this preserved smoothness is the reduced stride time and length variability.

Supporting our findings of sustained trunk smoothness at faster speeds, a recent study examined trunk acceleration-derived measures of signal regularity (approximate entropy) and signal repeatability (coefficient of multiple determination) in healthy young adults during slow, preferred, and fast walking (Kavanagh, 2009). Results showed that trunk accelerations were less repeatable and regular at slow speeds compared to preferred, but there were no differences in these measures between preferred and fast walking. In addition, findings from robotic and prosthetic research support the premise that stability is sustained, if not enhanced, at faster speeds. Interlimb coordination (defined as the standard deviation of relative phase) between extremities improved with increasing speed (Donker, Beek, Wagenaar, & Mulder, 2001) in young adults and for a prosthetic group (Donker & Beek, 2002). These authors argued that walking at faster speeds is accompanied by a reduction in stride length and frequency combinations (i.e., as speed increases both parameters will increase), and it is this reduction in variation that increases stability. Researchers developing walking robots have found that faster paces are more stable than slow paces (Hobbelen & Wisse, 2007; Wisse, Hobbelen, & Schwab, 2007). Taken together, there is general support that for young adults, walking stability is maintained at faster speeds.

Comparison of HRs at preferred speeds between healthy young and older adults has either revealed no differences (Menz, Lord, & Fitzpatrick, 2003b; Yack & Berger, 1993) or reductions in smoothness in specific directions of motion (Kavanagh, Barrett, & Morrison, 2005; Mazza, Iosa, Pecoraro, & Cappozzo, 2008). These equivocal findings may be due to sample differences in age range, gender, and health. Similar to Mazza et al., we recently found no differences in HRs between young adults and healthy 60-year-olds walking at preferred speed, but reduced AP and VT trunk smoothness in healthy 80-year-olds, suggesting that separation of older age groups is important for better understanding of age-related changes in gait (*Lowry, in preparation*).

To date, no study has compared HRs between young and older adults across a wide range of walking speeds. The primary purpose of this study was to examine HRs during overground walking across a range of self-selected speeds using a lifespan approach, by comparing young adults (YA) to healthy 60-year-olds (OA), and healthy 80-year-olds (OOA). Because we anticipated a wide range of self-selected walking speeds, we also compared HRs at speeds 20 % faster and slower than preferred. We hypothesized that 1) OAs would perform similarly to YAs at all speeds except at very fast and very slow speeds, where they would exhibit reduced smoothness, 2) OOAs would exhibit reduced trunk smoothness in general, with disproportionate decreases in smoothness at very slow and very fast speeds. Therefore, we expected to see the greatest group separation at very slow and very fast speeds.

Significant reductions in speed place more emphasis on balance, particularly lateral balance control, and may require more active control of the swinging leg (Den Otter, Geurts, Mulder, & Duysens, 2004), while very fast speeds necessitate the need for greater muscular force output. Decrements in the postural control and neuromuscular systems with age would result in greater challenge at these extreme speeds for both older groups with resulting loss of trunk smoothness, with OOAs exhibiting the greatest losses.

A secondary purpose examined the ability of spatiotemporal parameters and strength measures to predict HRs at certain speeds. As clinicians routinely use spatiotemporal parameters and strength measures in gait assessment and intervention, it is important to determine how these measures relate to HRs, if these relationships are directionally-specific, and if they change across speeds.

Methods

Participants

Thirteen healthy young, old, and old-old adults participated in this study. Young adults were university students. Older adults were recruited from the local community. Reasons for exclusion included neurological diagnoses, history of head trauma, significant heart disease, significant musculoskeletal impairments, or symptoms such as recent fracture or joint replacement, severe chronic pain, peripheral neuropathy, cognitive decline, and use of a walking device. All participants signed an IRB-approved consent form, and were recruited and tested according to IRB procedures.

Instrumentation and Procedures

A triaxial accelerometer (Crossbow CXLO2LF3, range ±2g) mounted to a plastic baseplate on a gait belt and secured and aligned with L3 (Smidt et al., 1971) measured trunk accelerations in three directions. The accelerometer was statistically calibrated on a flat surface, and once positioned on the lower trunk, was leveled prior to each trial. Data were sampled at 200Hz using a portable data logger (Crossbow AD2000 Ready DAQ) worn in a small backpack. Participants walked over a 19m x .6m paper-covered walkway at five speeds using the same cues as in previous studies (Menz et al., 2003a; Moe-Nilssen, 1998): 1) walk very slowly (VS) as if in an art gallery, 2) walk slower (SL) than normal, as if there were ample time, 3) walk at preferred, comfortable speed (PF), 4) walk faster (F) than normal but not maximal speed, and 5) walk as fast as is safe without running (VF). In addition, all subjects walked at paces 20% slower (20S) and faster (20F) than their preferred pace. Participants did not fixate on a target, but were told to 'look ahead' and avoid looking around the laboratory. Two markings were made to designate the middle 12.5m of the walkway for determination of walking speed. Three consecutive trials were performed in each condition. Participants were first asked to walk at their preferred pace followed by SL, VS, F, and VF, 20S and 20F. We chose this order so that preferred pace would not be contaminated by the other conditions.

Maximal isometric strength of four lower extremity muscle groups (hip flexion, extension and abduction; knee extension) was assessed using a hand-held dynamometer (Lafayette MMT) using test positions and procedures as outlined in previous research (Wang, Olson, & Protas, 2002).

To characterize the older groups, the Berg Balance Scale (BBS) (Berg, Wood-Dauphinee, Williams, & Maki, 1992), and Physical Activity Scale for the Elderly (PASE) (Washburn, Smith, Jette, & Janney, 1993) were administered.

Gait variables

The primary dependent variables were AP-, VT-, and ML-HRs. Full descriptions of HR theory and determination are reported elsewhere (Lowry, Smiley-Oyen, Carrel, & Kerr, 2009; Menz et al., 2003a) . Harmonic content of trunk accelerations signals was determined with Fast Fourier Transform using custom Visual Basic software from National Instruments Measurement Studio™ 6.0 libraries. HRs were calculated for each stride in the VT and AP directions as the ratio of the sum of the first 20 even harmonics/sum of the first 20 odd harmonics, and in the ML direction as odd/even.

Spatiotemporal variables were: *average speed (m/s)*, time to walk the middle 12.5m; *stride time (s)*, number of samples in the acceleration signal between consecutive heel strikes of the same foot*5ms/1000ms; *average stride length (m)*, calculated by multiplying the average speed by an averaged stride time; *step width (cm)*, measured as the distance between ink marks, consistent with previous research (Stolze, Kuhtz-Buschbeck, Mondwurf, Johnk, & Friege, 1998). Felt pads were fixed to the bottom of each shoe, placed at the intersection of the midpoint of the widest measurement of the shoe and 25% of the length, generally at the ball of the foot. Pads were inked red for right, black for left. Step width data was collected for one trial at very slow, preferred and very fast speeds only; *coefficient of variation* (*COV,* standard deviation/mean*100*)* was used to quantify the variability of stride time and step width.

Data Reduction and Analysis

The first trial was considered practice and values were averaged across trials 2 and 3 (except for step width variables that were collected for a single trial in very slow, preferred, and very fast only). Prior to calculation of means, the trials were visually inspected to determine if the program correctly selected strides. To avoid acceleration and deceleration effects, the first 2 and last 2 strides were removed from the acceleration data.

Two strength measurements were recorded for each muscle group and the maximal value was analyzed. Hip extension values are not reported due to inadequate stabilization during testing. Paired t-tests revealed no differences between left and right sides, so strength values were averaged and normalized for body weight and leg length.

Due to high skewness values of some variables, all data were log-transformed for statistical analyses. A Group (3) x Condition (3 or 5) repeated measures ANOVA was conducted for each dependent variable. Due to multiple primary comparisons of gait variables, the alpha level was set at .004 (.05/12). Significant interactions were followed up with between group one-way ANOVAs, and main effects were interpreted using Bonferroni-corrected pair-wise comparisons. Partial eta-squared effect sizes were interpreted as .01(small), .06(medium), and .138(large) (Cohen, 1988). Stepwise linear regressions were used to explore relationships between harmonic ratios and gait and strength variables. All statistical analyses were performed using SPSS15.

Results

Participant Characteristics

There were no group differences for leg length or mass (Table 1). There were no differences in PASE scores between the older groups; both men and women scored above published norms for their age groups (Washburn et al., 1993). The OOAs exhibited lower scores on the BBS compared to OAs, but all scored above 50, indicating low fall risk. One-way ANOVAs for normalized strength revealed significant group differences for all muscle groups (hip flexion: $F(2,36) = 10.777$, $p < .001$; hip abduction: $F(2,36) = 12.527$, $p < .001$; knee extension: F(2,36)= 5.599, p = .008). Post hoc tests revealed that YAs had greater strength than both older groups for hip flexion and abduction. For knee extension, YA had greater strength than OOA.

These data show while older groups were physically active, they exhibited the expected losses in strength, and in addition, OOAs exhibited the expected age-related loss of dynamic standing balance.

Spatiotemporal variables at self-selected speeds

Data for all variables for each speed condition are presented in Table 2. Group (3) x Condition (5) ANOVAs were conducted for speed, stride length, stride time, and COV stride time. The ANOVA for speed revealed only a significant condition effect (F (4,144) = 469.634, p < .001, partial eta squared = .929. Pair-wise comparisons revealed that all conditions were significantly different from one another, thus all groups complied and the cues were effective. The average speeds for our YA group were very similar to those reported by Latt et al. (in m/s, our data then Latt et al., respectively -- very slow: .56, .50; slow: .93, ~ .90 estimated from graph; preferred: 1.38, 1.2; fast: 1.77, ~1.7 estimated from graph; very fast: 2.38, 2.1). Though the interaction was not significant (p= .114), all groups had similar average speeds in the very slow, slow and preferred conditions, but, as expected, OOAs had lower average speed in the fast and very fast conditions (Figure 1A).

There was a significant interaction for stride length F (8,144)= 4.785, p= .001, partial eta squared = .210. Follow up one-way ANOVAs revealed group differences only in F (p< .001) and VF (p< .001) conditions. Post-hoc tests revealed OOAs had reduced stride length compared to both YA and OA in the fast condition, and all groups were significantly different from one another in the very fast condition (Figure 1B).

The ANOVA for stride time revealed only a significant Condition effect $F(4,144) = 290.711$, $p < .001$, partial eta squared $= .890$. Pair-wise comparisons revealed all conditions were significantly different from one another; stride time decreased as speed increased. The Group effect did not reach significance ($p = .032$), though YAs tended to have longer stride times, particularly at the slower speeds.

The ANOVA for COV stride time revealed significant effects for both Group ($F(2,36) = 6.719$, $p = .003$, partial eta squared $= .272$) and Condition ($F(4,144) = 35.924$, $p < .001$, partial eta squared $= .499$). Post-hoc tests for group revealed OOAs had higher stride time variability than the other two groups. Pair-wise comparisons for Condition revealed higher variability in very slow and slow, while preferred, fast and very fast were not significantly different from one another. Though the interaction was not significant ($p = .136$), OOAs exhibited a more distinct U-shaped curve with increased variability in the very fast condition (Figure 2).

A Group (3) x Condition (3: very slow, preferred, very fast) repeated measures ANOVA was conducted for both step width and COV step width. The ANOVA for step width revealed only a strong trend for Condition, $F(2,72) = 5.478$, $p = .006$, partial eta squared $= .132$, with increased step widths in the very slow condition. The ANOVA for COV step width revealed a significant interaction, $F(4,72) = 5.773$, $p < .001$, partial eta squared $= .243$. Follow up one-way ANOVAs revealed significant group differences in preferred only ($p < .001$) with both older groups exhibiting greater variability than YAs (Figure 3).

In summary, as expected, stride length increased and stride time decreased with increasing speed for all groups. The OOA group exhibited reduced stride length in fast and very fast, accounting for their slower walking speeds in these conditions. In general, stride time variability decreased with increasing speed, and OOAs exhibited greater temporal variability across all speeds. Similar to previous research, we found no group differences in step width (Owings & Grabiner, 2004); however, the results for step width variability revealed different group patterns across speeds. Young adults exhibited lower step width variability in very slow and preferred, then increased variability in very fast. The older groups also had lower variability in very slow, but exhibited an increase in preferred, with no change from preferred to very fast. Thus, with the exception of stride time variability, there were minimal group differences in spatiotemporal variables at slower speeds, while preferred and faster paces emphasized group differences.

Harmonic Ratios at Self-Selected Speeds

A Group (3) x Condition (5) ANOVA was conducted for each HR. The ML-HR analysis revealed only a significant Condition effect $F(4,144) = 16.766$, $p < .001$, partial eta squared = .318. Pair-wise comparisons revealed that ML-HR values in preferred, fast, and very fast were not significantly different from each other, and were higher than very slow and slow (Figure 4). While the Group effect did not reach significance ($p = .042$), it is interesting to note that OAs exhibited higher ML-HRs across speeds.

The ANOVA for the VT-HR revealed a strong trend for an interaction, $F(8,144) = 2.861$, $p = .006$, partial eta squared = .137. Follow up analyses revealed group differences in fast ($p = .002$) and very fast ($p = .001$). The OOAs exhibited lower VT-HRs compared to both other groups in fast, and lower values than YAs in very fast (Figure 5).

The ANOVA for the AP-HR revealed no interaction ($p = .615$), but significant effects for both Group ($F(2,36) = 9.619$, $p < .001$, partial eta squared = .348) and Condition ($F(4,144) = 28.065$, $p < .001$, partial eta squared = .438). Post- hoc tests for Group revealed

OOAs exhibited lower AP-HRs compared to both other groups. Similar to the results for the ML-HR, pair wise comparisons for Condition revealed that values in preferred, fast, and very fast were not significantly different from each other, and were higher than very slow and slow (Figure 6).

Harmonic Ratios at 20% Slower, Preferred, 20% Faster

There were no group differences in speed in the 20% slower, and 20% faster than preferred conditions ($p=.492$, $.480$ respectively). The speed ranges in these conditions were more narrow than for the self-selected slow and fast conditions, allowing for a comparison of harmonic ratios with less confounding influence of differences in walking speed.

The Group (3) x Condition (3) ANOVA for the ML-HR revealed results similar to the self-selected speed ANOVA, with a significant Condition effect, $F(2,72)= 22.185$, $p< .001$, partial eta squared = .381. The Group effect did not reach significance ($p= .022$), and OA's had the highest values across the conditions (Figure 7A).

Without the inclusion of the very fast condition, the strong trend for an interaction for the VT-HR was lost. Both VT- and AP-HRs exhibited the same pattern of results (Figures 7B,C). There were significant Condition effects (VT-HR: $F(2,72)= 17.046$, $p< .001$, partial eta squared = .321; AP-HR: $F(2,72)= 11.194$, $p< .001$, partial eta squared = .237) and Group effects (VT-HR: $F(2,36)= 7.521$, $p= .002$, partial eta squared = .237; AP-HR: $F(2,36)= 10.334$, $p< .001$, partial eta squared = .365). Pair-wise comparisons for Condition revealed lower HRs in the 20% slower condition compared to both preferred and 20% faster, which were not significantly different from one another. Post-hoc tests for Group revealed that YAs and OAs performed similarly, while OOAs exhibited reduced smoothness.

Taken together with the results from the self-selected conditions, the findings for the VT- and AP-HRs were similar; OOAs exhibited overall reduced trunk smoothness, YA and OAs performed similarly, and all groups exhibited loss of smoothness at speeds slower than preferred. In addition, while all groups maintained AP trunk smoothness at speeds faster than preferred, both older groups exhibited some loss of vertical smoothness in the very fast condition. The ML-HR exhibited a different pattern of results, with OAs having the highest ML smoothness, and less change across speeds for all groups.

Relationships between Variables

We ran a series of exploratory stepwise multiple regressions to examine prediction of HRs (Table 3). The predictor variables in the model were age, walking speed, COV stride time and COV step width, and a composite strength score (sum of the normalized strength measures). These variables were used to examine the prediction for each HR at very slow, preferred and very fast speeds. The variability measures were chosen because they have been related to falls and gait instability (Hausdorff, Rios, & Edelberg, 2001; Maki, 1997) and strength was included because we previously found that strength partially explained age-related loss of AP trunk smoothness (*Lowry, in preparation*).

The adjusted R^2 values show that ML trunk smoothness had the least amount of variance accounted for, and in the very slow condition walking speed accounted for a significant amount of the variance of all three HRs. In addition, stride time variability consistently explained a significant amount of variance for both VT-and AP HRs.

Discussion

We examined the impact of walking speed on age-related differences in trunk smoothness. We hypothesized that OAs would perform similarly to YAs except at very slow and very fast speeds, but OOAs would exhibit general reductions in smoothness, with further reductions during very slow and very fast walking. We found OOAs exhibited reduced smoothness, but in general, the pattern of smoothness across speeds was similar for all groups, with the exception of vertical smoothness in the very fast condition.

In general, across speeds, OOAs exhibited lower AP and VT trunk smoothness, which is consistent with previous research (Mazza et al., 2008). Coupling between hip and knee moments was reduced in fit elderly, suggesting decreased ability to consistently control the trunk in the AP direction (Winter, Patla, Frank, & Walt, 1990). Aging also caused a redistribution of joint torques and powers during gait, where older adults used more hip extension and less knee extension and ankle plantarflexion than young adults (DeVita & Hortobagyi, 2000). A redistribution towards proximal segments may result in irregular trunk accelerations and lower harmonic ratios. Additionally, altered visual, somatosensory, and vestibular integration, or central nervous system 'noise' may also result in decrements in trunk control. Thus, reduced AP and VT trunk smoothness may be reflective of changes both in central and peripheral control.

The results of the exploratory regression analyses revealed that both age and stride time variability (but not the composite strength score) were significant predictors of AP and VT HRs. Previous research suggested that stride time is determined predominantly by a gait-patterning mechanism involving repeated sequential contraction and relaxation of muscle groups (Gabell & Nayak, 1984), thus it may be that stride time variability better reflects the dynamic, phasic nature of muscle activation than a static isometric strength score. In the current study, higher stride time variability was associated with reduced smoothness. This finding is consistent with findings that higher stride time variability is present in pathology

(Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998) and predictive of fall risk (Hausdorff et al., 2001), and suggests stride time variability may be a good clinical marker for global walking stability.

All groups exhibited reduced trunk smoothness at speeds slower than preferred, and though OOAs exhibited the lowest smoothness values, particularly in the AP direction, they were not disproportionately affected by walking more slowly. Reductions in walking speed generally increase time spent in double support, and at extremely slow speeds, gait becomes a series of static postures, with postural muscular synergies dominating control. In addition, the trunk exhibits more in-phase (reduced counter rotation) movement (van Emmerik & Wagenaar, 1996), and typically larger lateral excursions (Orendurff et al., 2004). Using the standard deviations of discrete and continuous relative phasing of the trunk and pelvis to quantify stability, researchers found increased variability (lower stability) between .5 and 1.0 m/s (our very slow and slow speed ranges), with decreasing variability (greater stability) from 1.0 to 1.3 m/s. At slower speeds the typical coordination pattern between the trunk and pelvis was unstable, with a gradual change to a stable pattern as speed increased to preferred speed ranges (van Emmerik & Wagenaar, 1996). Similarly, optimal coupling of trunk and pelvic rotations was found to occur at speeds above .75 m/s (Wagenaar & Beek, 1992). Our data of reduced trunk smoothness at slower speeds for all age groups in all directions of motion is consistent with these findings of altered coordination dynamics at slower speeds. The results of our regression analyses strengthen this position, as speed and stride time variability (but not age) predicted HRs in the very slow condition.

Though the above discussion helps to explain why all groups exhibited reduced smoothness at speeds slower than preferred, it does not explain why older groups were not disproportionately affected at these speeds. Though there were no differences in stride length at slower speeds, stride time was longer for YAs. It may be that YAs and the older groups used different strategies to accomplish the same slower speeds (i.e. older groups may have spent more time in double support to gain stability). In addition, slower speeds provide more time for error detection and correction, possibly offsetting the age-related decrements in motor control.

In contrast to previous studies (Menz et al., 2003a), but similar to our findings with young adults (*Lowry, in preparation*), we found that both older groups maintained trunk smoothness at faster paces (with the exception of VT smoothness in the very fast condition). Thus there was no clear optimization at preferred speed for any age group. However, both older groups exhibited some loss of vertical smoothness in the very fast condition. Both groups exhibited less change in stride length from fast to very fast compared to YAs. This may possibly be due to reaching their limits of ankle plantarflexion power and/or their range of motion of the lower extremity. Loss of VT smoothness in the very fast condition may be related to these types of changes.

Mediolateral trunk smoothness exhibited a different pattern of results from AP and VT smoothness. In general, there was less change across speeds, and though OOAs exhibited lower average values, they were not statistically different from YAs. Interestingly, OAs had the highest values across speeds, which may be reflective of an 'overcontrol strategy' (Vernazza-Martin et al., 2008). Additionally, the ML trunk smoothness had the least amount of variance accounted for by the model, indicating that the ML-HR is capturing unique information regarding gait control.

Our findings of increased step width variability with age (Owings & Grabiner, 2004) and young adults increasing step width variability with speed (Sekiya, Nagasaki, Ito, & Furuna, 1997) are consistent with previous research. Step width variability has been related to lateral balance control (Bauby & Kuo, 2000); however, in our regression analyses step width variability was not a significant predictor of ML trunk smoothness. Further research is needed to determine contributions to the ML-HR.

Study limitations include small sample sizes, and limitations in variables. We estimated stride length from speed and stride time, therefore we could not determine stride length variability. Additionally, we did not have measures of hip extension or ankle strength, which may have offered more insight and improved prediction of HRs. A future research direction would be to examine kinematic and kinetic data together with HRs, and other trunk acceleration-derived measures, to better understand how the system controls global body motion during walking, and which measures are most sensitive to age-related changes in gait.

In summary, our main findings were the following. 1) YA and OAs performed similarly across speeds, while OOAs exhibited reduced trunk smoothness across speeds. This finding suggests that examining 60-year-olds separately from 80-year-olds is important, and that inclusion of older adult participants into one large group may mask or misrepresent age-related changes in motor performance. The functional meaning of this reduced smoothness in OOAs is not known. Harmonic ratios have effectively discriminated the gait patterns of high and low risk fallers (MEnz), but no study has investigated the ability of HRs to predict future fall risk. 2) All groups exhibited reduced smoothness at speeds slower than preferred. This finding indicates that HRs must be interpreted with caution when examining individuals who walk at slower speeds, or if using HRs to examine the effect of a dual task, as it is expected that given a certain cognitive load, people will walk more slowly. 3) HRs were not clearly optimized at preferred pace, rather, all groups maintained smoothness in the fast condition. These data indicate that healthy older adults are not compromised walking at brisk paces, at least for level, unobstructed walking. Slower preferred walking speeds (< 1.0 m/s) predict future falls and health status (Studenski et al., 2003), and functional decline (Brach & VanSwearingen, 2002). Walking briskly promotes cardiovascular health, and may help to prevent the decline in preferred speed. 4) HRs were most closely associated with stride time variability, where greater variability is associated with reduced smoothness. These data support the position that stride time variability is an important clinical marker for gait function.

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Table 1. Means (SD) of subject characteristics, PASE = Physical Activity Survey for the Elderly; * significant difference from other two groups at p< .001; ł two groups are different from each other at p= .008.

	Young Adults men (n=4) women (n=9)	**Old Adults** men (n=3) women (n=10)	**Old-Old Adults** men (n=4) women (n=9)
Age (yrs)	22.13(.9), 20-23	66.34 (2.6), 60-69	82.47 (2.2), 80-86
Leg Length (m)	.90 (.04)	.93 (.06)	.88 (.08)
Mass (kg)	69 (12.9)	77.6 (13.4)	67.38 (10)
Berg Balance Score	---	55.6 (.8)	52.9 (1.4) **ł**
PASE	---	188 (75.4)	173 (61.6)
Normalized Strength (N/kg*m)			
hip flexion	3.34 (.78) *****	2.13 (.92)	2.04 (.68)
hip abduction	4.98 (1.0) *****	3.57 (1.11)	3.10 (.87)
knee extension	4.96 (.86) **ł**	4.15 (.92)	3.78 (.97) **ł**

Table 2. Non-transformed means (SD) for all variables for each condition. ML-HR = mediolateral harmonic ratio, VT-HR = vertical harmonic ratio, AP-HR = anteroposterior harmonic ratio. Dashed lines indicate data were not collected during these conditions.

	Very Slow	Slow	Preferred	Fast	Very Fast	20% slower than Preferred	20% faster than Preferred
ML-HR							
YA	1.96(.43)	2.05(.38)	2.34(.49)	2.43(.25)	2.55(.47)	2.10(.34)	2.52(.43)
OA	2.09(.35)	2.26(.40)	2.58(.52)	2.75(.57)	2.59(.65)	2.26(.50)	2.88(.69)
OOA	1.89(.44)	1.93(.40)	2.13(.43)	2.23(.60)	2.35(.61)	1.96(.30)	2.23(.49)
VT-HR							
YA	2.19(.41)	2.91(.73)	3.74(.54)	4.09(.61)	3.97(.51)	3.16(.57)	3.95(.55)
OA	2.47(.48)	3.27(.79)	3.65(.40)	3.94(.59)	3.41(.91)	3.32(.77)	4.04(.66)
OOA	2.19(.49)	2.66(.57)	3.13(.83)	3.12(.99)	2.79(.73)	2.74(.44)	3.02(.93)
AP-HR							
YA	2.59(.71)	3.25(.70)	3.85(.81)	3.93(.54)	3.80(.51)	3.36(.62)	4.00(.38)
OA	2.86(.63)	3.44(.60)	3.80(.47)	3.90(.59)	3.79(.83)	3.51(.55)	3.92(.57)
OOA	2.33(.55)	2.60(.75)	3.00(.75)	3.03(.83)	3.00(.82)	2.73(.63)	3.11(.85)
Speed (m/s)							
YA	.56(.18)	.93(.19)	1.33(.25)	1.69(.26)	2.23(.26)	1.06(.20)	1.61(.29)
OA	.64(.17)	1.01(.11)	1.34(.13)	1.60(.10)	2.10(.28)	1.08(.09)	1.62(.16)
OOA	.59(.12)	.96(.15)	1.28(.12)	1.49(.15)	1.92(.18)	1.01(.10)	1.53(.14)
Stride length (m)							
YA	.99(.11)	1.20(.14)	1.43(.18)	1.65(.16)	1.89(.16)	1.29(.15)	1.61(.16)
OA	1.06(.13)	1.25(.10)	1.40(.10)	1.55(.11)	1.72(.21)	1.30(.09)	1.56(.11)
OOA	.92(.16)	1.15(.12)	1.30(.09)	1.39(.08)	1.52(.09)	1.18(.10)	1.41(.08)
Stride time (s)							
YA	1.94(.66)	1.32(.19)	1.09(.10)	.99(.08)	.85(.06)	1.24(.15)	1.02(.12)
OA	1.71(.29)	1.24(.09)	1.05(.07)	.97(.06)	.82(.08)	1.20(.08)	.97(.07)
OOA	1.59(.177)	1.21(.13)	1.02(.09)	.95(.08)	.79(.08)	1.17(.12)	.93(.08)
Step width (cm)							
YA	13.29(2.05)		13.38(2.11)		12.41(1.55)		
OA	14.61(3.56)	——	13.17(3.13)	——	14.12(2.45)	——	
OOA	15.03(3.36)		13.61(3.03)		12.83(2.31)		
COV stride time (%)							
YA	4.82(2.21)	4.36(1.99)	2.48(.80)	1.94(.69)	1.66(.74)	3.29(.68)	2.04(.87)
OA	4.79(1.97)	3.60(1.51)	2.38(.72)	2.08(.76)	2.28(1.40)	3.36(1.2)	2.25(.86)
OOA	5.67(2.03)	4.46(1.16)	2.94(1.24)	2.91(1.55)	3.51(1.76)	4.20(1.38)	2.72(1.16)
COV step width (%)							
YA	12.81(3.82)		12.66(3.27)		24.00(10.21)		
OA	14.52(5.11)	——	22.90(4.84)	——	20.28(4.70)		
OOA	15.62(4.97)		23.08(9.58)		23.03(9.05)		

Table 3. Results of stepwise regressions. Dependent variables: ML-HR = mediolateral harmonic ratio, VT-HR = vertical harmonic ratio, AP-HR = anteroposterior harmonic ratio. Independent variables in the model: age, speed, composite strength score, stride time variability, step width variability.

Stepwise Regressions			
	ML-HR	**VT-HR**	**AP-HR**
Very slow			
predictors	Speed	Speed, COV stride time	COV stride time, Speed
p value of model	< .001	< .001	< .001
Adjusted R^2	27%	60%	63%
Preferred			
predictors	COV stride time	Age	COV stride time, Age
p value of model	0.045	0.044	0.01
Adjusted R^2	8%	8%	18%
Very fast			
predictors	none	COV stride time, Age	COV stride time
p value of model		< .001	0.001
Adjusted R^2		39.50%	26%

Figures

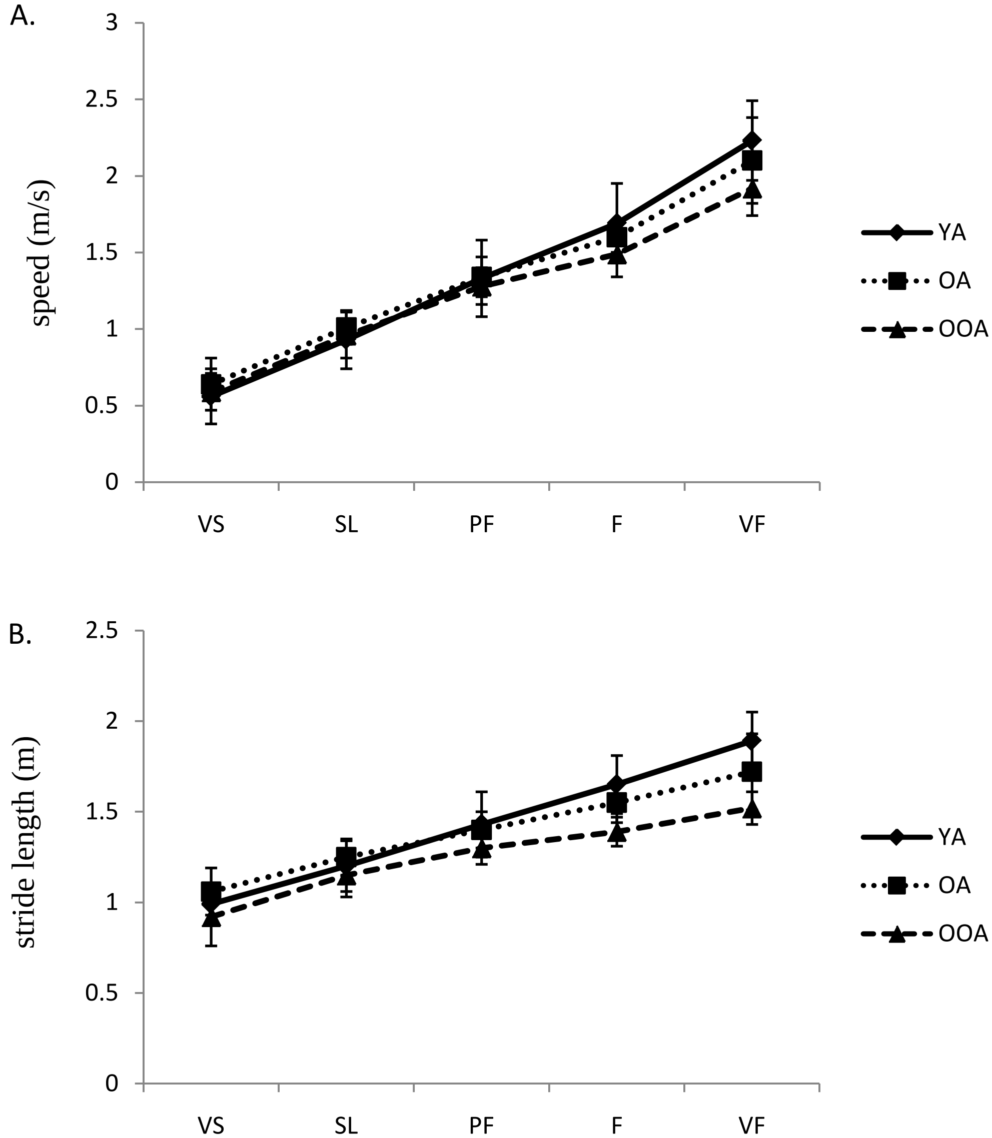

Figure 1. Walking speed in m/s (A.), Stride length in m (B.) for the five self-selected speed conditions. Error bars are SD. VS = very slow, SL = slow, PF = preferred, F = fast, VF = very fast.

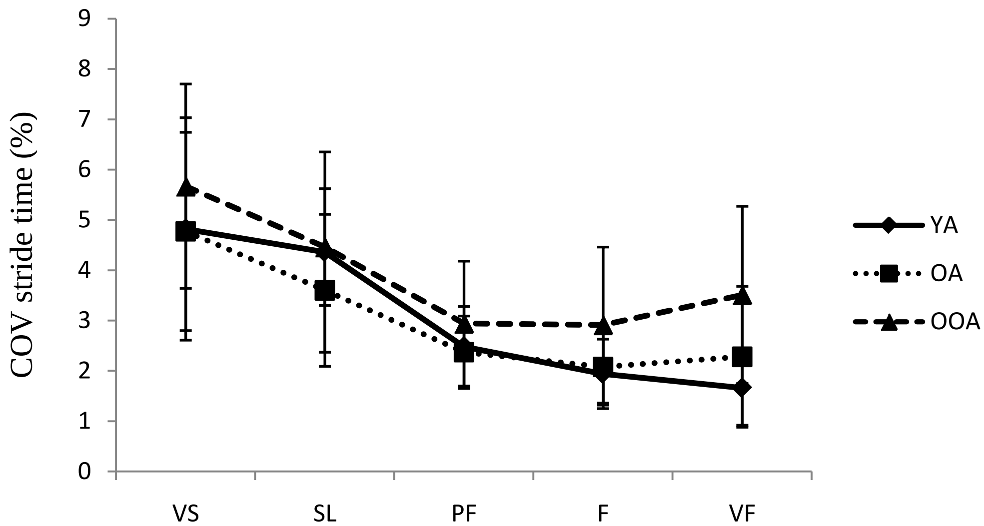

Figure 2. Coefficient of variation of stride time for the five self-selected speed conditions. Error bars are SD. VS=very slow, SL=slow, PF=preferred, F=fast, VF=very fast walking condition.

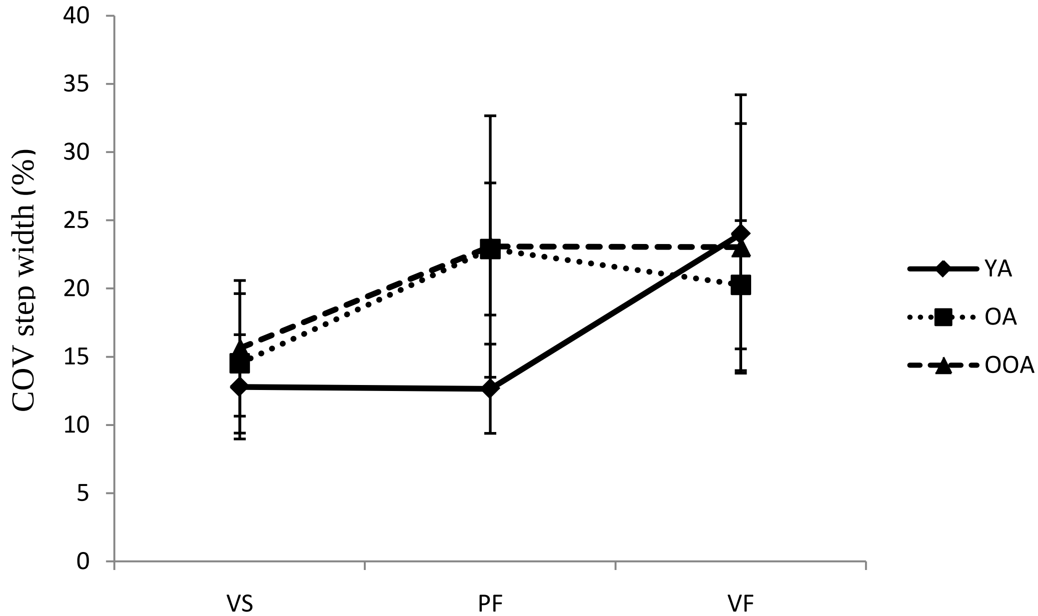

Figure 3. Coefficient of variation of step width for three self-selected conditions. Error bars are SD. VS=very slow, SL=slow, PF=preferred, F=fast, VF=very fast walking condition.

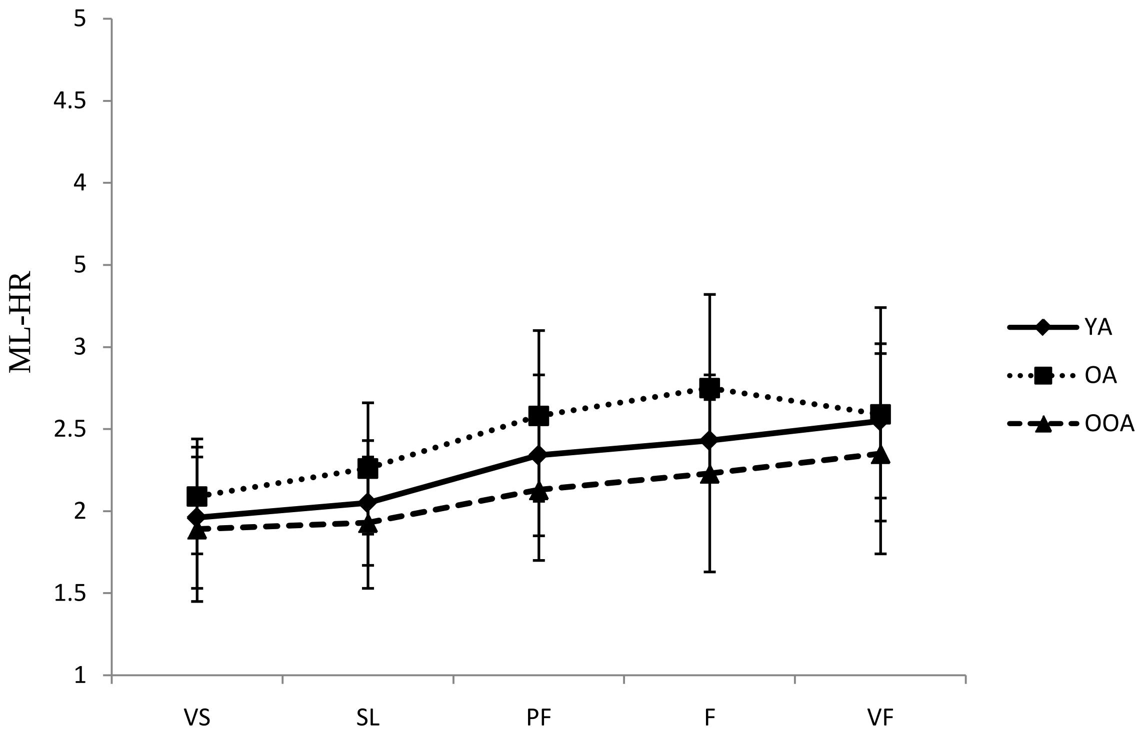

Figure 4. Mediolateral harmonic ratios of non-transformed data for the five self-selected speed conditions. Error bars are SD. VS=very slow, SL=slow, PF=preferred, F=fast, VF=very fast walking condition.

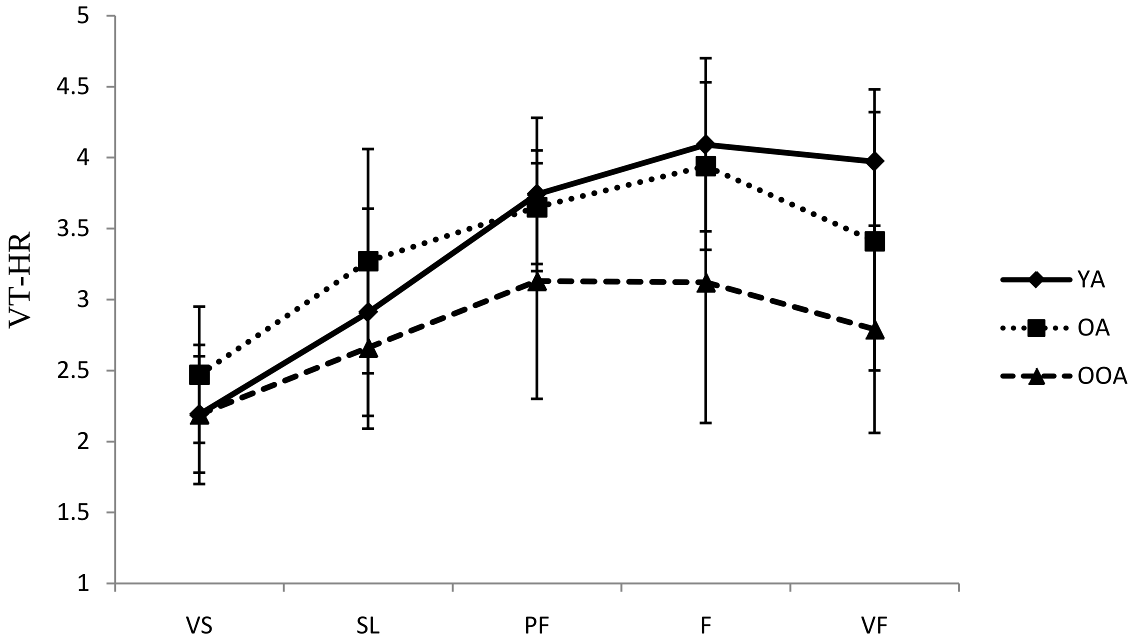

Figure 5. Vertical harmonic ratios of non-transformed data for five self-selected speed conditions. Error bars are SD. VS=very slow, SL=slow, PF=preferred, F=fast, VF=very fast walking condition.

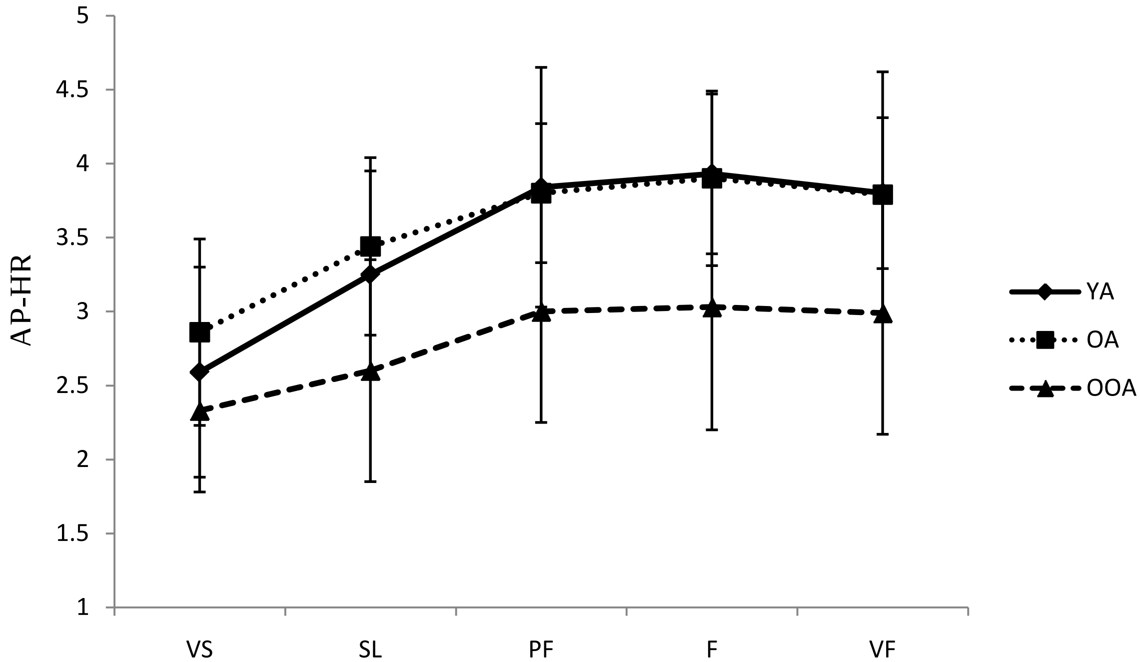

Figure 6. Anteroposterior harmonic ratios of non-transformed data for five self-selected speed conditions. Error bars are SD. VS=very slow, SL=slow, PF=preferred, F=fast, VF=very fast walking condition.

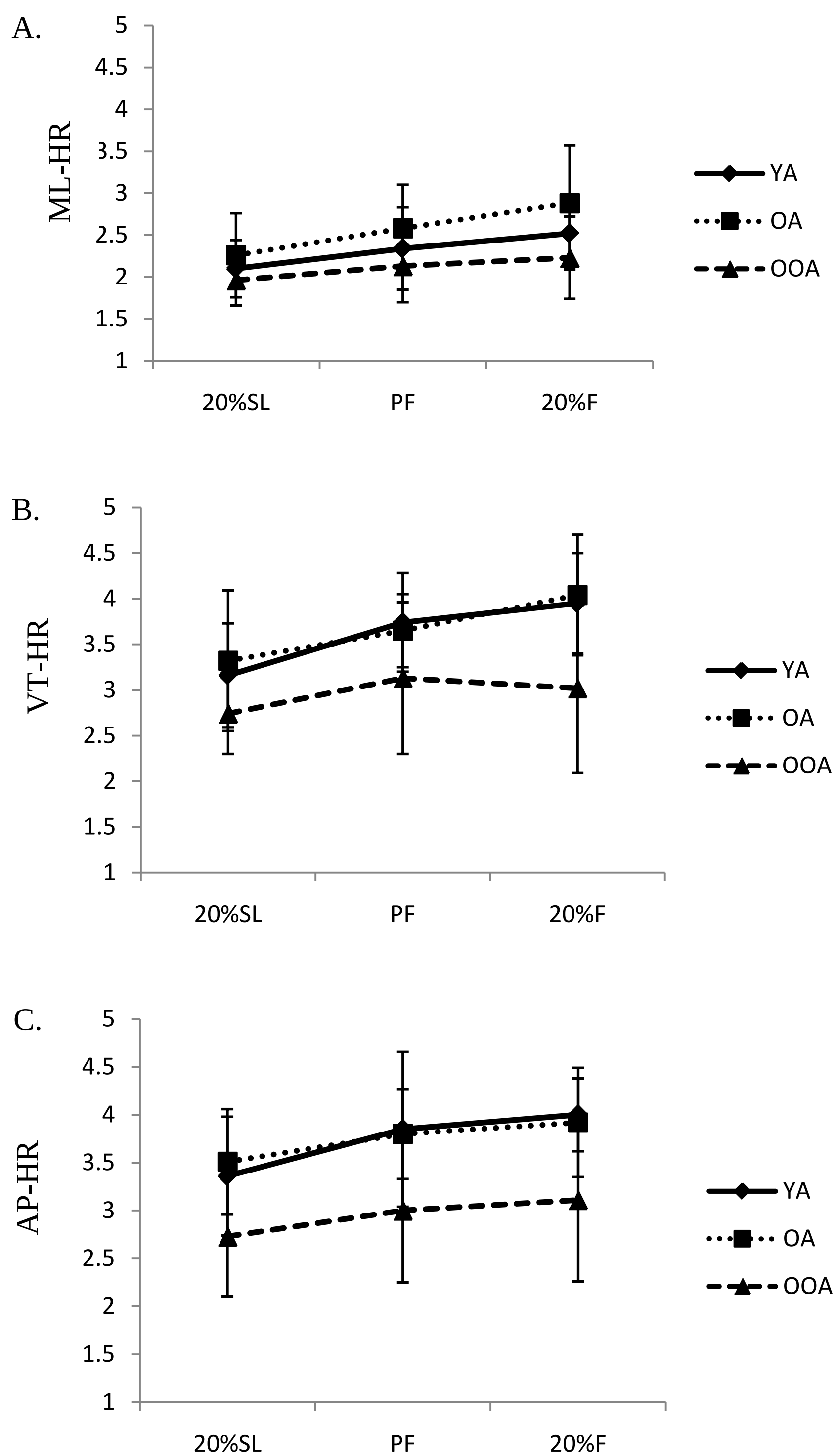

Figure 7. Mediolateral (A.), Vertical (B.), and Anteroposterior (C.) harmonic ratios of non-transformed data. Error bars are SD. 20%SL=20% slower than preferred speed, PF=preferred, 20%F = 20% faster than preferred.

CHAPTER 6. SUMMARY AND RECOMMENDATIONS

Summary

There is extensive literature documenting age-related differences in spatiotemporal variables during gait. Age differences in kinematics and kinetics are also well documented, though this literature has focused primarily on lower extremity changes. More recently, efforts have been made to understand upper body control and balance during gait. Accelerometry offers an attractive method to investigate upper body control during gait, as accelerometers are relatively inexpensive and can be used outside of the laboratory. Harmonic ratios are derived from acceleration signals, measure smoothness of motion, and are an indication of dynamic balance during gait. Acceleration signals have documented reliability, and harmonic ratios have demonstrated discriminant validity by revealing differences between young and older adults, between stable and unstable older adults, and changes with walking speed. However, the pattern of age differences and changes with speed have varied across studies, likely due to differences in sample sizes and the age ranges and health status of the older groups.

A study was presented that replicated and extended previous research by: 1) examining harmonic ratios across a range of self-selected speeds in young adults; 2) comparing harmonic ratios at preferred speed in young adults, healthy 60-year-olds, and healthy 80-year-olds; and 3) comparing harmonic ratios in all three age groups across a range of self-selected and paced speeds. In addition, spatiotemporal variables and strength measures were obtained and relationships between these measures and harmonic ratios were examined. These data are the first to separately examine harmonic ratios in 60- and 80-year olds both at preferred speed and across a range of speeds. Further, these are the first data to examine step-width variability, strength, and harmonic ratios in the same study.

In young adults, we found loss of trunk smoothness at speeds slower than preferred pace, but with trunk smoothness preserved during fast and very fast walking. These findings are in contrast to published data (Latt, Menz, Fung, & Lord, 2008; Menz, Lord, & Fitzpatrick, 2003b), but consistent with recent research using other acceleration-derived measures and coordination measures (Donker, Beek, Wagenaar, & Mulder, 2001; Donker & Beek, 2002; Kavanagh, 2009). An explanation for this sustained smoothness was that stride time and stride length variability decreased, while step-width variability tended to increase as speed increased. In addition, the mediolateral harmonic ratio exhibited less change across speeds, and lacked association with spatiotemporal variables, emphasizing the uniqueness of this measure.

Comparing harmonic ratios between the three age groups at preferred speed, we found that healthy 60-year-olds' trunk smoothness was similar to healthy 20-year-olds' while healthy 80-year-olds' smoothness was lower in both the anteroposterior and vertical directions. These findings are in contrast to previous research that found no differences in smoothness between young and older groups (Menz, Lord, & Fitzpatrick, 2003c; Yack & Berger, 1993), but consistent with a recent study that compared healthy young and older women (Mazza, Iosa, Pecoraro, & Cappozzo, 2008). Strength only partially explained group differences in anteroposterior trunk smoothness, while strength and speed accounted for age-differences in vertical smoothness. Smoothness in the mediolateral direction was not affected by age, indicating that control in this direction may be prioritized. Both older groups exhibited greater step width variability, which may help to preserve mediolateral smoothness.

Lastly, we examined harmonic ratios across a range of self-selected and paced speed conditions for all three age groups. The main findings were that young adults and 60-year-olds performed similarly across speeds while 80-year-olds exhibited reduced trunk smoothness across speeds. All groups exhibited reduced smoothness at speeds slower than preferred, and harmonics were not clearly optimized at preferred pace as all groups maintained smoothness in the fast condition. In addition, harmonic ratios were most closely associated with stride time variability, where greater variability was associated with reduced smoothness.

Together these findings suggest that harmonic ratios must be interpreted with caution when examining individuals who walk at substantially slower speeds, or who are walking more slowly do to dual task conditions. A previous study found no differences in harmonic ratios between young and older groups, though the older group walked more slowly (Menz et al., 2003c). However, our self-selected and paced slower speeds were substantially lower than the reported walking speed of this older group. Thus, there may be a criterion speed below which the coordination of walking deteriorates. Alternatively, there may be a difference between slower speeds that are 'preferred' and a ' required ' slow pace.

Additionally, these findings suggest that examining 60-year-olds separately from 80-year-olds is important for gait research, as inclusion of older adults into one large group may mask or misrepresent age-related changes in motor performance. Our results indicate loss of trunk smoothness in 80-year-olds, particularly in the anteroposterior direction. This loss is not fully explained by strength deficits, indicating other changes, such as degradation in sensory integration may be involved.

Further, we found that harmonic ratios were associated with stride time variability, where greater variability was associated with reduced walking smoothness. These findings are consistent with current literature (Hausdorff, Rios, & Edelberg, 2001) that support the use of stride time variability as a clinical marker for gait function.

Recommendations for Future Research

Harmonic ratios may be more appropriately defined as an integrated variable, reflective of the kinematic and kinetic output during walking, which in turn is dependent on the accuracy and integrity of afferent input and descending influences. An interesting future direction would be to examine kinematic and kinetic data together with harmonic ratios and other trunk acceleration-derived measures to better understand how the system controls global body motion during walking.

Together with our findings, the literature supports loss of walking smoothness in advanced age, and in older adults at a high risk of falling (Menz, Lord, & Fitzpatrick, 2003a; Yack & Berger, 1993).One question is whether reduced smoothness is a better predictor of falls than measures such as stride time variability. A second question is how smoothness changes with aging by following a sample across years; to date, all the studies using harmonic ratios have been cross-sectional designs. A third question is how responsive harmonic ratios are to a rehabilitation intervention. It is interesting information that harmonic ratios are reduced in advancing age, but this information in itself cannot guide clinical practice, i.e. how does a clinician rehabilitate reduced smoothness? A future direction would be a combination intervention and prospective study design, where harmonic ratios, spatiotemporal parameters, and functional measures would be examined at baseline and after specific gait interventions. Participants might then be followed for up to a year to gather fall history information. The information from this type of study would certainly enhance 'best-practice' guidelines, serving the ultimate goal of fall reduction in older adults.

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