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The effects of hallux valgus and walking speed on dynamic balance in older adults

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THE EFFECTS OF HALLUX VALGUS AND WALKING SPEED ON DYNAMIC BALANCE IN OLDER ADULTS

By

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Abstract

Hallux valgus (HV) contributes to deficits in static balance and increased fall risk in older adults. Very limited research has examined dynamic balance deficits in walking in this population. These individuals generally walk slowly, as balance challenge is lesser at slow speeds. The purpose of this study was to examine dynamic balance in older adults with HV compared to healthy controls at controlled slow and fast speeds. Nineteen older adults with HV and 13 healthy controls completed 5 continuous walking trials at 1.0 and 1.3 $m \cdot s^{-1}$ as whole body marker position and ground reaction force data were captured. Dynamic balance was evaluated using whole body center of mass (COM) and center of pressure (COP) inclination angles (IA) and duration of double support. There were no differences in measures of dynamic balance between older adults with and without HV at slow and fast speeds. At the faster speed, the peak sagittal plane COM-COP IA increased and the double support duration decreased, while the peak frontal plane COM-COP IA were not affected. Older adults with HV do not exhibit deficits in dynamic balance during continuous walking at comfortable speeds when compared to healthy older adults.

Keywords: Bunions, Center of mass, Center of pressure, Inclination angles

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Chapter I

Introduction

Older adults exhibit a decreased capacity to maintain balance during dynamic tasks such as walking [1]. This decline in dynamic balance may explain why older adults tend to have an increased risk for falls [2]. With advancing age, foot deformities such as hallux valgus (HV) also become increasingly prevalent [3,4], which further increases fall risk in older adults [5,6]. HV is also known to adversely affect static balance [7–9] and habitual walking speed [10,11]. These findings suggest that dynamic balance during walking may be affected in older adults with HV.

The relationship between HV and dynamic balance during walking in older adults has not been sufficiently investigated. Only two studies have examined the effect of HV on dynamic balance during walking in older adults. Mickle et al. [12] and Sadra et al.[13] both found that double support duration, a spatiotemporal gait parameter associated with dynamic balance in walking, was not different between HV subjects and controls. However, double support duration alone is not considered to be a sensitive measure of dynamic balance although the need for dynamic balance is greatest in the double support phases of walking, when weight is rapidly transferred from the trailing leg to the leading leg [14,15]. Double support duration is not a sensitive enough measure because it does not take into account the relationship between whole body center of mass (COM) and base of support. The interaction of whole body COM, center of pressure (COP), and the inclination angle (IA) between these two points (i.e. COM and COP) has been shown to be a more sensitive measure of balance in walking than double support duration [16,17]. Peak COM-COP IA occur near the instants of weight acceptance and transfer during the double support phases and thus provide a more precise assessment of dynamic balance. No studies to date have examined dynamic balance in HV using measures of COM and COP interactions.

Dynamic balance challenge increases with walking speed, and thus walking speed should be controlled when assessing dynamic balance. Research indicates that walking speed independently affects double support duration [18] and COM-COP IA [19,20]. As walking speed increases, the double support phase shortens in healthy subjects [18]. Similarly, peak anterior and posterior COM-COP IA increase while peak medial IA decreases with increasing walking speed [19,20]. Older adults with HV often walk at slower preferred speeds than healthy, age-matched controls [10,11]. The two studies which examined dynamic balance during walking in older individuals with HV assessed double support duration at slower, preferred speeds [12,13]. The effect of HV on walking balance at controlled, faster speeds has not yet been examined. The deficits in balance may become more apparent in individuals with HV if assessments are performed at faster walking speeds using more sensitive measures of analysis (i.e. anterior, posterior, and medial COM-COP IA).

There has been limited assessment of the cumulative effects of age and HV on walking balance in the gait of older adults. Deficits in dynamic balance during walking have not been examined in older adults with HV under experimentally-controlled slow and fast speed conditions using sensitive measures of balance such as peak COM-COP IA. Therefore, the purpose of this study was to evaluate dynamic balance evaluated via peak COM-COP IA during walking in older adults with HV under experimentally-controlled slow and fast walking speeds, specifically near the double support phase near the instants of weight acceptance and transfer. We hypothesized that individuals with HV compared to healthy age- and sex-matched controls would demonstrate greater deficits in dynamic balance, displaying increased medial and decreased anterior and posterior peak COM-COP IA for the fast speed condition, but that these peak IA would not be affected at slower speeds.

Chapter II

Methods

Participants

Nineteen older community-dwelling adults with HV and 19 controls participated in this study (age range 60-85 years). Since no previous research has examined dynamic balance using COM-COP IA measures in older adults with HV, a statistical power analysis was conducted using GPower 3.1 software to determine the appropriate sample size. To achieve a statistical power of 0.8 for a medium effect size (Cohen's f=0.25) age x speed interaction, a sample size of 34 participants (17 participants per group) was needed.

All participants were community-dwelling individuals capable of walking unassisted continuously for a minimum of 10 minutes without resting. Using the Manchester Scale, all participants (i.e. HV and controls) were evaluated to determine the presence and grade of HV [21]. Participants were included in the experimental HV group if the degree of deformity assessed was at either moderate or severe, class 2 or class 3, respectively. Prior research has shown that balance is adversely affected and fall risk is greater in older adults with a moderate-to-severe grade of HV [5,6,10]. HV participants were included in the study if they exhibited the deformity without the presence of disabling pain which affects walking. Prior research has shown that approximately 70%-88% of individuals with moderate-to-severe HV do not experience excessive pain [22,23]. Control participants were also evaluated using the Manchester Scale and included if they did not have HV, otherwise known as a class 0 categorization. Individuals with any orthopedic or neurological conditions affecting balance and walking were excluded from the study. Also, individuals who received surgical correction for HV were also excluded. Western Washington University Institutional Review Board approved the study and all participants provided a written informed consent prior to participation.

Data collection

All participants completed a single, one-hour testing session. Upon arrival at the lab, participants filled out a health history and physical activity questionnaire. Prior to trials, each of the subject's feet were visually inspected in a weight-bearing position by a single investigator (CB), and classified according to the Manchester Scale. Previous research has shown that the grade of HV can be reliably classified using this visual scale [24]. The Manchester Scale has been shown to be a highly reliable (i.e. high intra- and inter-rater reliability) and valid method for classifying HV for research purposes, and has been used frequently in the literature [3,21–28]. Studies have shown that the Manchester Scale has a high level of concurrent criterion validity when compared to criterion radiographic assessment [29]. After HV assessment, participants changed into spandex clothes provided by the investigators.

Anthropometric measurements of body height, body mass, inter-anterior superior iliac spine distance, and bilateral leg length, knee width, ankle width, shoulder offset width, elbow width, wrist width, and hand thickness were taken. Next, 39 reflective markers were placed on the participant's skin and spandex clothes according to the full body VICON PlugIn Gait model guidelines. VICON PlugIn Gait model was chosen for the current study because it has been extensively used to quantify whole body center of mass (COM) position [15,19,20,30–35]. A 10-camera VICON motion capture system (Centennial, CO, USA) and two AMTI in-ground forceplates (Watertown, MA, USA) captured three dimensional marker position and ground reaction force data synchronously at 100Hz and 1000Hz, respectively. A static calibration trial was completed during which the participants were instructed to stand stationary on a single forceplate in the "motorcycle pose" according to VICON PlugIn Gait model guidelines.

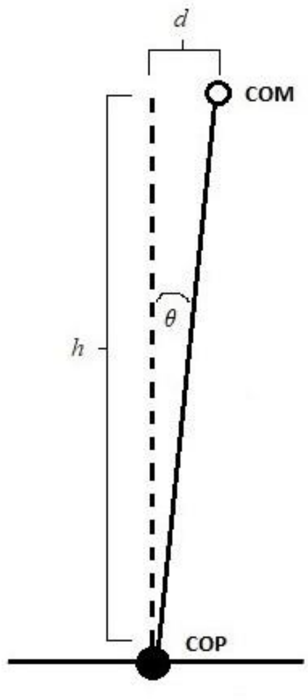
After the static trial, dynamic walking trials were completed at two experimentally controlled speeds of 1.0 $m \cdot s^{-1}$ and 1.3 $m \cdot s^{-1}$ in a random order. The slower speed, 1.0 $m \cdot s^{-1}$, was selected as it is reported to be the typical natural gait of older adults exhibiting moderate-to-severe HV [10–12]. Similarly, the faster speed, 1.3 $m \cdot s^{-1}$, was selected as it is the typical natural gait of healthy older adults [36–38]. Participants were given 3-5 familiarization trials to get acclimated to walking at each speed before trials were collected. After familiarization, data were collected for five trials for each walking speed condition. Walking speed was computed immediately after each trial (using the data for one-stride length as the participant walked across the capture zone) to check if it was within ±3% of the target speed. A trial was accepted if the subject was able to 1) achieve the targeted speed within the ±3% range of error with a natural gait, 2) strike the correct forceplate without making any visible alterations to stride pattern, and 3) strike the forceplates such that the leg exhibiting the greatest degree of HV was the leading leg [39].

Data analysis

Marker position and ground reaction force data were low pass filtered at 8 Hz using a 4th order Butterworth filter [40–42]. Basic spatiotemporal variables such as walking speed, stride length, step width, cadence, and double support duration were computed from the filtered data. Using marker position data and Dempster's equations [43], each segment's center of mass and the whole body COM were computed using VICON Plug-in gait post capture pipeline [15,19,20,29–34]. Instantaneous COM position coordinates were determined for the entire gait cycle. Center of pressure (COP) coordinates were computed from forceplate ground reaction force and moment data. Because each forceplate provides its own COP data, there was a need to determine resultant COP which combines data from both forceplates during the double-support phase. This was accomplished using the equation below [14–16,44]:

$$COP_R = COP_1 \frac{F_{Z1}}{(F_{Z2}+F_{Z1})} + COP_2 \frac{F_{Z2}}{(F_{Z1}+F_{Z2})}$$

where COP_R is the resultant COP, COP_1 and COP_2 are the COP coordinates from the respective forceplates, and Fz_1 and Fz_2 are the vertical ground reaction forces from each respective forceplate. Instantaneous COM and COP position data were then used to calculate the IA for the frontal (medial) and sagittal (anterior and posterior) planes. The calculation of these angles has been described previously [16]. Briefly, the equation used to compute AP and ML COM-COP IA is provided below (Figure 1):



$$\theta = tan^{-1}(d/h)$$

Figure 1. The calculation of the center of mass-center of pressure IA uses the height of the COM (*h*) and the horizontal separation of the center of mass and center of pressure (*d*) to calculate the angle (θ), as shown above. This model can be applied to either the sagittal (anterior-posterior) or the frontal (mediolateral) planes.

where θ is the IA, d is the horizontal separation between the COM and COP, and h is the vertical height of the COM from the ground. Instantaneous COM-COP IA were computed over a gait cycle, and peak anterior, posterior, and medial values were identified for further analysis.

Statistical analysis

The independent variables for this experiment were the presence of HV and walking speed. The primary dependent variables examined were peak anterior, posterior, and medial COM-COP IA, and the double support duration during which the foot exhibiting HV is the push-off leg. The secondary dependent variables were stride length and step width. A mixed model ANOVA with repeated measures for speed was used to examine the effect of HV on the dependent variables. Alpha was set a priori to 0.05. Effect sizes were calculated as partial eta squared (η_p^2), where, $\eta_p^2 > 0.01$ was small, $\eta_p^2 > 0.06$ was medium, and $\eta_p^2 > 0.15$ was large effect size [45]. All statistical procedures were performed using SPSS (Version 23).

Chapter III

Results

Demographic characteristics are presented in Table 1. The hallux valgus and control group participants were not significantly different in age, height, mass, and amount of moderate-to-vigorous physical activity they engaged in regularly on a weekly basis. A majority of the participants in the hallux valgus group (14/19) had grade 3 severity of the condition.

Table 1. Participant demographic characteristics

	Hallux Valgus (*n*=19)	**Controls (*n*=19)**	***t*-value**	***p*-value**
Sex	13 females	13 females		
Grade of Hallux Valgus	2.7 ± 0.5	0		
Age (years)	71.4 ± 5.6	68.6 ± 5.5	t_{36}=1.557	0.128
Height (cm)	164.6 ± 11.9	165.8 ± 11.0	t_{36}=-0.324	0.784
Mass (kg)	68.5 ± 10.3	71.4 ± 12.7	t_{36}=-0.784	0.438
Moderate to vigorous physical activity (minutes/week)	464.4 ± 382.0	455.1 ± 386.0	t_{36}=0.074	0.941

There were no group-speed interactions or differences between groups for any of the spatiotemporal variables and peak COM-COP IA. With an increase in speed there was an increase in stride length and anterior and posterior COM-COP IA, a decrease in double support duration, but no significant differences in step width and medial COM-COP IA (Table 2).

Table 2. Gait parameters, double support duration, and COM-COP IA

	Hallux Valgus		Controls		Group main effect	Speed main effect	Group-speed interaction
	$1.0\ m\cdot s^{-1}$	$1.3\ m\cdot s^{-1}$	$1.0\ m\cdot s^{-1}$	$1.3\ m\cdot s^{-1}$			
Stride length (m)	1.18 ± 0.09	1.33 ± 0.10	1.16 ± 0.10	1.30 ± 0.11	$F_{1,36}$=0.35; *p*=0.556; η_p^2=0.010	**$F_{1,36}$=437.07; *p*<0.001; η_p^2=0.924 ***	$F_{1,36}$=0.60; *p*=0.442; η_p^2=0.016
Step width (cm)	8.5 ± 2.5	8.2 ± 2.7	8.6 ± 3.1	8.8 ± 4.1	$F_{1,36}$=0.15; *p*=0.703; η_p^2=0.004	$F_{1,36}$=0.01; *p*=0.933; η_p^2=0.000	$F_{1,36}$=0.65; *p*=0.425; η_p^2=0.018
Double support duration (s)	0.17 ± 0.02	0.14 ± 0.02	0.17 ± 0.02	0.14 ± 0.02	$F_{1,36}$=0.01; *p*=0.939 η_p^2=0.000	**$F_{1,36}$=374.77; *p*<0.001; η_p^2=0.912 ***	$F_{1,36}$=0.99; *p*=0.755; η_p^2=0.003
Anterior COM-COP IA (º)	9.5 ± 1.1	11.5 ± 1.2	8.8 ± 1.2	10.7 ± 1.3	$F_{1,36}$=3.65; *p*=0.064 η_p^2=0.092	**$F_{1,36}$=387.14; *p*<0.001; η_p^2=0.915 ***	$F_{1,36}$=0.00; *p*=0.968; η_p^2=0.000
Posterior COM-COP IA (º)	6.7 ± 0.8	7.7 ± 1.1	7.0 ± 1.0	7.8 ± 1.1	$F_{1,36}$=0.38; *p*=0.543 η_p^2=0.010	**$F_{1,36}$=108.59; *p*<0.001; η_p^2=0.751 ***	$F_{1,36}$=1.06; *p*=0.311; η_p^2=0.028
Medial COM-COP IA (º)	2.9 ± 0.9	2.9 ± 1.1	3.3 ± 1.1	3.1 ± 1.3	$F_{1,36}$=0.89; *p*=0.351; η_p^2=0.024	$F_{1,36}$=1.063; *p*=0.309; η_p^2=0.029	$F_{1,36}$=0.44; *p*=0.512 η_p^2=0.012

*** statistical significance**

Chapter IV

Discussion

We hypothesized that older individuals with HV, when compared to healthy controls, would demonstrate greater deficits in dynamic balance for the faster walking speed condition but that dynamic balance deficits would not be observed at slower speeds. We, therefore, expected a group x speed interaction for the balance-related dependent variables. Our data did not support the hypothesis. Anterior, posterior, and medial IA and double support duration were not significantly different between groups for either speed condition (Table 2). We expected the group x speed interaction because dynamic balance challenge has been shown to increase with walking speed [19,20,46–49].

Consistent with previous research, our results showed that anterior and posterior IA increased significantly with increasing walking speed (Table 2). At faster speeds, the whole body COM moves more in the sagittal plane over the base of support [41], thus increasing the demand on dynamic balance to control the whole body COM with respect to the base of support. Also consistent with literature, our results further showed that double support duration decreased with increasing walking speed [18,46] (Table 2). A decrease in duration of double support phase that occurs with faster speed presents a greater challenge to dynamic balance**.** This decrease in double support duration reflects a reduced amount of time to transfer weight from one leg to the other and control the more rapid movement of the COM over the new leading limb [1,18,50]. Overall, our sagittal plane IA data and duration of double support show that dynamic balance challenge did indeed increase with increasing walking speed.

For the frontal plane IA, we did not observe the anticipated decrease in peak angles with increased speed. Medial IA has previously been shown to decrease with increasing walking speed [19,20]. Interestingly, step width, which provides an indication of the base of support breadth in the frontal plane, was also not significantly affected by speed (Table 2). Generally, with an increase in walking speed, step width decreases [46]. These results are aligned with previous research which showed that frontal plane

COM kinematics are related to step width [51]. COM-COP IA are computed from the whole body COM and the base of support (i.e., COP). An absence of speed effect on measures of the base of support such as step width could partially explain why changes were not observed in frontal plane COM-COP IA in this experiment. Perhaps the range of speeds selected in this study were not large enough to observe changes in frontal plane kinematics. The speed range in the current study was 1.0-1.3 $m \cdot s^{-1}$, whereas in previous research walking speeds ranged from 0.7-1.6 $m \cdot s^{-1}$ [19,20,46].

We did not observe dynamic balance deficits when comparing older adults with HV to healthy controls. When compared to healthy controls, older adults with balance deficits typically show increased peak medial IA and decreased peak anterior and posterior IA, which occur near the double stance phase of walking [16,17,30]. We did not observe such trends in our COM-COP IA results. Our data are comparable to previous results. When plotted over a gait cycle, our findings for peak COM-COP IAs are similar in profile but slightly lower in magnitude than those reported by Lee et al [16]. Although not consistent with our hypothesis, the lack of significant differences in dynamic balance between groups is consistent with previous findings on the balance of older adults with HV. Previous research that investigated differences in dynamic balance in older adults with HV and controls used less sensitive measures of dynamic balance, such as double support duration, and found no differences between groups [12,13]. Our data add to the literature on the effects of HV on balance control by examining differences in these groups using sensitive COM-COP IA measures of dynamic balance [16] and controlled, faster walking speeds. The observations from the current study and previous research cumulatively suggest that perhaps no differences in dynamic balance exist between older adults with HV and healthy older adults at comfortable speeds during continuous walking.

For the current study, we chose to control walking speed. Walking speed is known to be a confounding variable that independently influences spatiotemporal measures of gait and dynamic balance during walking [18–20]. We examined walking speeds of 1.0 and 1.3 m/s, as these speeds closely resemble self-selected comfortable walking speeds of older adults with (55-80 years) [10–12] and without

HV (60-80 years) [36–38], respectively. Although the comparison of these speeds may be more representative of these populations' comfortable walking habits, it is possible that these speeds may not have been challenging or fast enough to uncover dynamic balance deficits. Future studies investigating dynamic balance in older adults with HV and healthy controls should consider increasing balance challenge by choosing faster speeds.

Participant inclusion and exclusion criteria for this study were highly rigorous. The older adults with HV who were recruited for this study had only the two more severe grades of the deformity. We elected to exclude mild forms of the deformity to elicit the most significant possible effect on dynamic balance [5,9,10,52]. In addition, these individuals had no obvious age-related comorbidities that would influence balance control. These highly rigorous criteria were intended to isolate the effects of HV from other aging-related comorbidities. A possible unintended consequence of our recruitment process was that the older adults with HV who met the recruitment criteria tended to be very physically active individuals (Table 1). Generally, older adults who are very physically active tend to have less deficits in dynamic balance during walking [53–55]. The possible confounding effect of physical activity may further explain why we did not observe differences in dynamic balance between groups. Future research should consider investigating dynamic balance in older adults with HV who are less active and have no age-related comorbidities.

In conclusion, healthy older adults with moderate-to-severe HV do not exhibit deficits in dynamic balance during walking across a range of comfortable speeds when compared to healthy older adults without HV.

Chapter V

Review of Literature

Introduction

This review will broadly explore two main topics: hallux valgus (HV) and dynamic balance in older adults. Full comprehension of the proposed origins, treatments, and effects of HV is necessary to understand how this condition may impact factors of dynamic balance, particularly in walking. Dynamic balance is a complicated parameter, with a multitude of methods for calculation and interpretation that currently exist. This review will thoroughly appraise these methods of assessment and their accuracy in predicting dynamic balance in walking. Typical values expected of normal, healthy, subjects will also be reported and compared to those of older adults and those with unstable gait. Through this review, the author hopes to illuminate possible connections between HV and a decline in the dynamic balance of older adults.

Hallux Valgus

HV is a common foot deformity that can develop over the lifespan. The deformity is characterized by a deviation of the distal end of the great toe (i.e., hallux) away from the midline of the body [21]. Colloquially referred to as a bunion, HV is chronic and often progressive. Generally, a clinical diagnosis of HV is considered to be a deviation of 15 degrees or greater [4,56,57]. Most literature reports that HV incidence rates among mostly shod (i.e. shoe-wearing) populations fall between 20-45% [3,4,56,58,59]. These numbers are typically much higher in older adults beyond the age of 65 years, where incidence rates rise to 35-70% [23,58,60,61]. HV is more prevalent in women than in men (35-60%), particularly over the age of 60 years [4,23,56,58,61,62]. HV is usually a bilateral deformity, with only 17% of cases exhibiting the deformity in only one foot [57,59]. In the subsequent review, the causes, treatments, and effects of HV will be thoroughly explored.

Musculoskeletal and ligamentous contributions to hallux valgus

Despite the large proportion of the population that suffers from this condition, a conclusive etiology of HV has yet to be identified. Some theories attribute its development to weakness in the muscles and ligaments surrounding the hallux. There is evidence that HV was accompanied by hypermobility in the first ray [57,63]. Subjects with HV have decreased plantarflexion and abduction strength at the hallux [9,64]. Pes planus, otherwise known as flat-footedness, is commonly observed in conjuction with HV [3,58,65]. Pes planus indicates increased ligament laxity at the forefoot, resulting in excess pronation [66]. Patients displaying both pes planus and HV may have inherent ligament weakness in the forefoot that contributes to the progression of HV. Indeed, changes in plantar pressures with HV indicate a medially-shifted loading pattern that is associated with overpronation [67]. However, since not everyone with HV also displays flat-footed characteristics, it is not likely that ligament weakness at the forefoot is the sole cause. Research on the role of muscular and ligamentous deficiencies in the development of HV has been limited. While it is likely from the evidence that certain muscles and ligaments may contribute to HV, further examination is necessary to determine the significance of their contribution.

Morphological differences associated with hallux valgus

Many with HV claim the deformity is inherited from family members. This claim is supported by the literature, with more than 85% of people with HV self-reporting a family history [57]. Though scientists have yet to confirm this link conclusively, some research has illuminated inherited variation within the musculoskeletal structure of those with HV. HV patients typically have longer first metatarsal bones than their non-HV peers [57]. Radiographic assessments have shown that increased torsional angle at the first ray is linked to the development of HV [68]. A recent cadaveric study found that the accessory tendon of extensor hallucis longus muscle was more frequently found in conjunction with HV [69]. The presence of accessory tendons and altered bone structure may alter force vectors at the first ray, thus gradually contributing to the deformity over the lifespan. Simulation studies have shown that altered force vectors in the toes could lead to the development of progressive HV [70]. As with the theories on muscle and ligament weakness, however, these morphological differences are not prevalent enough to match the high prevalence of the deformity. The current evidence suggests that structural variation likely contributes, but cannot be confirmed as the sole cause.

The contributions of footwear to the development of hallux valgus

One of the more common theories on the etiology of HV is the contribution of footwear. Strong evidence suggests that long-term use of narrow, constrictive footwear such as high heels drastically increases the risk of developing HV [4,62,70,71]. This evidence may explain the higher incidence rates in female populations, who traditionally tend to wear narrower footwear and high heels. However, there is evidence that wearing shoes at all may increase risk. A study of a remote island community where many residents rarely wore shoes showed that HV incidence rates plummeted to 2% in traditionally-barefoot individuals [56]. It may be that wearing shoes, and especially those that are constrictive, exacerbates weaknesses and structural issues in the forefoot. Snijders and colleagues [70] proposed that the constriction of the toes by restrictive footwear may result in increased use of the first ray musculature through a deviant angle, possibly leading to the progressive development of the characteristic lateral deviation. A structural predisposition towards HV is likely exacerbated by footwear which inherently misaligns and detrains the musculature of the forefoot over the lifespan. While not significant on their own, these factors combined may explain the high prevalence rates of this deformity in shod populations.

Non-invasive treatment of hallux valgus

Due to the nature of the proposed causes of HV, prevention of the deformity is challenging. Treatment tends to focus on effective correction, though currently only a few methods of correction are well-supported. The nature of the musculature surrounding the first ray and the issues with force vectors makes re-strengthening the afflicted muscle groups difficult, though not impossible. Kim and colleagues [72] found that a toe-spreading training regimen lessened hallux angle and symptoms, as well as improved hallucis abductor muscle functioning. Corrective taping to straighten the hallux angle has been shown to affect certain aspects of balance in older patients with HV [73], and thus tends to be avoided. Still, no strengthening protocols have been shown to reverse the deviation. In milder cases, before patients reach the need for surgery, physicians recommend alterations to footwear selection or the addition of a custom-molded orthotic to current footwear [74]. Chadchavalpanichaya and colleagues [75] found that wearing a custom-molded toe separator overnight decreased hallux angle in subjects with moderate to severe HV. These forms of non-invasive treatment need further exploration to validate their effectiveness at reversing this condition. Though the study by Chadchavalpanichaya and colleagues showed an improvement in hallux angle in their treatment group, the change was minimal and still classified within the moderate to severe category. This study found that the control group hallux angle worsened while the treatment group improved over a 12-month period. This indicates that a custom orthotic, even worn at night, can potentially improve or at least halt the progression of HV. Currently, full reversal of the deformity is not possible outside of surgical measures.

Surgical treatment of hallux valgus

A vast number of surgeries are used to correct HV. Though several surgeries exist, no one procedure has been agreed upon as the gold standard for correcting HV [74,76]. Surgery becomes the primary correction of HV in cases where pain is debilitating and the hallux angle is severe enough to cause dysfunction. Most surgeries have fairly high success rates, but the rate of recurrence for nearly all of these procedures lies between 10 and 30% [57,77,78]. As there is currently no widely accepted surgical treatment and the chances of reversing the condition outside of surgery are minimal, prevention of the deformity is critical. Further research is needed to explore recent developments in HV treatment and prevention.

The grading of hallux valgus

While treatment and prevention pose a challenge to physicians, the classifications of severity within this deformity remain relatively easy. HV is a visually apparent deformity. For medical professionals, this makes HV an easy affliction to diagnose. The current gold standard for HV classification is radiographic assessment to measure hallux angle and observe any torsional occurrences within the metatarsals [29]. Radiographs are cumbersome and costly however, and most patients will not elect for radiographic evaluation until symptoms become severe. Considering the prevalence of HV, a more cost-effective and easily accessible method of evaluation is warranted. This endeavor has been limited by the difficulty of accurately measuring exact hallux angle without the use of x-rays. Goniometric measurements have been invalidated in recent research due to their inability to reliably measure hallux angle [57,79].

The Manchester Scale

The most convenient and most commonly recommended tool used by clinicians and researchers is a scaling system that classifies the hallux angle into categories of severity, rather than measuring the exact angle. The Manchester Scale was first developed in 2001 by Garrow and colleagues [21] using standard, accepted photographs of hallux angle with increasing severity to classify the deformity (see Figure 1). While this style of classification does not provide an exact angle, it does provide a reliable classification of severity. It has been validated against radiographic assessment [29], and also against the

American Orthopaedic Foot and Ankle Society Score index [28]. It is highly reliable for inter-rater assessment, and even reliable enough to be used without specific training [24]. The Manchester Scale is widely accepted as a reasonable alternative to radiographic assessment when exact hallux angle is not required.

Figure 2. The Manchester Scale for the grading of HV [21]. Each image indicates a category of increasing severity: 0) no HV present, 1) mild HV, 2) moderate HV, and 3) severe HV.

The impacts of hallux valgus on quality of life

HV is affiliated with a broad range of symptoms and conditions that are not fully understood. The ones that clearly impact quality of life, such as pain or a loss of function, are often the ones that drive patients to seek medical attention for their deformity. The symptoms that impact quality of life usually do not surface until a patient reaches the moderate to severe classifications. Mild HV is not correlated with pain or excessive dysfunction [80]. While milder forms of this deformity cause few issues, if any, HV is often progressive. The symptoms worsen as the deformity progresses, and virtually never recede on their own, often recurring even after intensive treatment measures are taken [81]. Many studies have shown that HV has a negative impact on quality of life. Patients with HV frequently complain of displeasure with the appearance of their feet and report difficulty selecting footwear [82]. HV has been associated increased hallux pain and decreased foot function [3,64], though this finding is more common in older adults and somewhat inconsistent. Other studies have shown that more severe HV is not always accompanied by debilitating foot pain [4]. HV is associated with arthritis in older adults however, which can produce pain symptoms separately. Kuryliszyn-Moskal and colleagues [83] found a correlation between increasing hallux angle and worsening osteoarthritis and rheumatoid arthritis. Some have suggested that HV may in fact worsen osteoarthritis, even in more proximal joints [84]. As mentioned earlier, HV produces a decline in toe strength, especially in severe HV [6,9,64]. HV also produces altered plantar pressure distributions [67]. Considering the progressive nature of HV, it is possible that the decreased functioning and pain in the hallux with HV could precede larger deficits up the kinetic chain.

The effect of hallux valgus on static balance

Several studies have quantified the effects of HV on static stability and postural sway. Menz and colleagues [85] first linked foot deformities such as HV to declines in performance during tests of postural sway in older adults. Menz and his associates later found [22] that mediolateral sway is significantly impacted by severity of HV deformity in similar populations. These results have more recently been observed in younger demographics as well. Healthy adults between the ages of 20-76 years show increased mediolateral sway in subjects with HV during double-leg stance [64] and single-leg stance [9]. Patients with surgical correction for HV still show increased postural sway when compared with their non-HV peers [13]. HV is often accompanied by other issues such as arthritis and pain, as discussed earlier. It could be argued that these confounding variables may influence these results. However, a recent study showed that even mild HV angles in otherwise asymptomatic adults still results in a loss of mediolateral and overall stability in single-leg stance [7]. HV appears to have impacts on functional abilities in otherwise healthy adults. These deficits likely extend beyond static balance into areas of movement, such as walking.

The effect of hallux valgus on spatiotemporal gait parameters

Study of HV impact on spatiotemporal gait parameters has yielded relatively inconsistent results. Stride length and step length have been reported to be shorter in HV patients [10,86]. There is also evidence that stance time with HV subjects increases compared to controls [86,87]. Two studies [10,22] found that gait speed is decreased in older adults presenting with HV. Walking speed has been observed to decline in younger adults with HV as well [86]. Mickle et al. [12] found that walking speed was not affected in older adults with HV compared to controls. There are some minor inconsistencies on whether gait speed is affected by HV in the literature, though the majority appear to support a decline in walking speed with HV. Decreased gait speed, increased stance time, and shorter strides are typical characteristics that indicate an adoption of a more stable gait pattern to compensate for the decline in balance ability in older populations [1]. These alterations to spatiotemporal gait patterns in older HV subjects may suggest that a decline in dynamic balance during walking is present.

The effect of hallux valgus on dynamic balance and fall risk

Despite the indications for a decline in balance that exists within altered spatiotemporal gait parameters with HV, only a few studies have examined performance in tests of dynamic balance in subjects with HV. Interestingly, these few have shown little indication of a decline in the dynamic balance of HV patients. A recent study observed no impact of HV on timed up-and-go performance nor in a step test [80]. Mickle and colleagues [12] found that gait variability during walking was not impacted by HV. Sadra and colleagues [13] found that trends of gait imbalance were not present patients with HV. However, some evidence that dynamic balance is affected still exits. Menz, Morris, and Lord [22] found that HV negatively impacted performance in sit-to-stand, alternating stepping, and coordinated stability performance. While the evidence suggests that HV has minimal, if unpredictable, impacts on dynamic balance, HV is known to be strongly associated with an increased risk for falls in older adults [5,6,12,85]. The gait of older adults was also found to be significantly less stable in older subjects presenting with moderate to severe HV, particularly on irregular surfaces [10]. It is unlikely that HV would affect static balance and fall risk so conclusively without also impacting dynamic balance, but no pattern has yet been discovered. The disconnect between these affiliated areas of literature within the study of HV suggests that an unknown mechanism behind the relationship between HV and dynamic balance that is not yet understood.

It should be noted that nearly all studies mentioned thus far examining dynamic balance in the gait of HV subjects did not control gait speed. To the author's knowledge, all studies thus far have examined HV subjects and controls who naturally walked at slower-than-normal speeds. Gait speed is widely known to have a large confounding effect on gait parameters, including measures of dynamic balance while walking [18,88]. Studies examining the impacts of HV on dynamic balance in walking tend to recruit older adult subjects, due to their inherent increased risk for falls. Older populations are known for slower self-selected walking speeds [89]. Older adults above the age of 60 years typically walk at a comfortable speed of 1.2-1.3 m/s, though this number declines as people age [90]. Small changes to gait speed produce large impacts on spatiotemporal gait parameters [18]. In all studies examining HV gait speed mentioned in this review, gait speed did not exceed 1.0-1.1 m/s, including those observing populations younger than 60 years [10,12,86]. The fact that all subjects, both HV and controls, in these studies walked slower than typical older adults indicates that some declines in gait velocity still exist.

Future studies of the spatiotemporal gait parameters of adults with HV should consider controlling gait speed.

Furthermore, no study on dynamic balance in subjects with HV has measured dynamic balance using calculation of center of mass or its relationship with the base of support. The only study to have currently examined anything resembling center of mass movement was Sadra and colleagues [13]. The researchers examined center of mass mediolateral sway during steady state walking and found no relationship between HV subjects and controls. However, this finding may have been the product of not controlling for subject age and height or improper calculation of the center of mass. Another large limitation of that study was the exclusion of center of pressure interaction. The kinematics of the center of mass and its interaction with the base of support is highly predictive of gait instability and imbalance [16,40,91]. These concepts will be explored further in this review. In regards to HV however, further research with greater control for confounding variables is necessary to elucidate the true effects of HV on dynamic balance in walking.

Dynamic Balance in Walking in Older Adults

Dynamic balance is the ability to maintain posture without falling while experiencing perturbations from gravity, momentum, or other outside forces. Dynamic balance is particularly important to walking, where the body is in a constant state of imbalance. Without the ability to maintain balance dynamically, walking would be impossible. As humans age, dynamic balance declines and falls become more likely [92]. Falls are of great concern in older populations as they tend to be associated with increased morbidity rates and the likelihood of sustaining a traumatic injury [93]. Falls are also more likely in older adults exhibiting HV foot deformity [5,6]. However, the link between HV, fall risk, and dynamic balance during walking has not yet been identified.

Quantifying dynamic balance in walking

Dynamic balance is complex and multifaceted. There are many different methods to quantify the variable ranging from simplistic spatiotemporal gait parameters to complicated center of mass calculations. There are currently five methods of quantifying dynamic balance during gait that are widely accepted, and these will be explored in the following subsections.

Initial base of support

Initial base of support is defined as the width between ankle joint centers upon heel strike [91]. It has also been referred to more generally in walking as step width [94], though initial base of support indicates an instantaneous value that occurs only at heel strike. Since dynamic balance is related to the ability to maintain the center of mass over a base of support [95], it seems reasonable that having a larger base of support might indicate greater stability. However, a number of studies have invalidated initial base of support as a mechanism for monitoring gait instability [94,96].

Double support duration

Double-support duration (DSD) is defined as amount of time when both feet are in contact with the ground during a full gait cycle [91,97,98]. Longer amounts of time spent in double-support indicate the adoption of a safer, more stable pattern of walking, and usually indicate instability. Longer DSD values are associated with an increased risk for falls in older adults [99]. DSD decreases with vestibular rehabilitation in people suffering from balance issues affiliated with bilateral vestibular hypofunction [100]. However, the validity of DSD on its own may not accurately reflect the true presence of instability. DSD is often paired with other more sensitive measures of gait imbalance, such as alterations to center of mass trajectory [100].

Center of mass velocity

The complexity of dynamic balance during walking stems from the complexity inherent in the kinematics of the whole body center of mass (COM) during gait. One method for monitoring center of mass movement as an indicator of imbalance during gait is maximum and average anteroposterior (A-P) COM velocity. Average A-P COM velocity is defined as the change in COM position over the course of a gait cycle, while maximum A-P COM velocity is the greatest instantaneous change in COM displacement over time [91]. COM velocity in the forward direction is indicative of the ability to control momentum during gait, thus maintaining balance [101]. There is little consensus in the literature on how to interpret COM velocity, and many studies show mixed results [40,101]. COM position has been found to have a greater effect on balance than the velocity of the COM [102], which may suggest that A-P COM velocity is not as reliable of a method to assess balance capacity in walking. These findings imply that A-P COM velocity may not be as sensitive of an indicator of imbalance during walking as was previously thought.

Center of mass-center of pressure separation

The interactions between COM and center of pressure (COP) show far more promise in accurately reflecting dynamic balance capacities. Some have referred to the maximum moment arm as a highly sensitive indicator of balance in walking [91]. The horizontal separation of the COM from the COP indicates how far an individual is willing to allow their COM to displace from their base of support. This method of analyzing dynamic balance has been used extensively with great results [15,34,40,91,103,104]. COM-COP separation distances are versatile. They can be used to calculate separation in all three planes of motion [33]. They can be presented as peak values for a gait cycle [40,103,104] or as instantaneous events of interest within a gait cycle, such as at heel strike and toe off [15,34]. Despite its sensitivity, versatility, and wide acceptance in the literature, COM-COP presents one major flaw. Subject height greatly impacts postural stabilization [95,105], yet COM-COP does not account for subject height. Without standardizing for intersubject height variability, COM-COP cannot accurately predict gait imbalance.

Center of mass-center of pressure inclination angles

COM-COP inclination angles were developed as a method to account for intersubject height variability. Inclination angles (IA) were conceptualized and introduced by Lee and colleagues [16], as an angle that can be calculated using basic trigonometry using the height of the COM and the instantaneous COM-COP separation distance. By using COM-COP separation distance as part of the calculation of an IA, this methodology reaps the benefits and sensitivity of COM-COP separation while still accounting for the influence of subject height. COM-COP is widely accepted in the most recent literature as the most valid method for quantifying gait imbalances [19,20,31,32,42,106,107].

When measuring dynamic balance in walking, it is always best to validate findings by using more than one method to quantify the results. Utilizing both a COM-related variable and a spatiotemporal gait parameter (DSD) may mitigate any loss of accuracy in determining gait imbalances. For the purposes of this review, we will explore DSD and COM-COP inclination angles in the anterior-posterior and mediolateral directions as complementary measurements of dynamic balance during walking.

Calculating COM-COP inclination angles

As was previously stated, COM-COP inclination angles (IA) are currently the most broadly accepted method for determining dynamic balance, particularly in gait. Often, IA are calculated using basic trigonometry [16,19]. The simplest method is the original method, which was developed by Lee and colleagues [16] and is demonstrated below (see Figure 2). Per this method, IA are calculated as follows:

$$\theta = tan^{-1}(d/h)$$

where θ is the IA, h is the instantaneous height of the COM, and d is the corresponding horizontal distance between the COM and the COP.

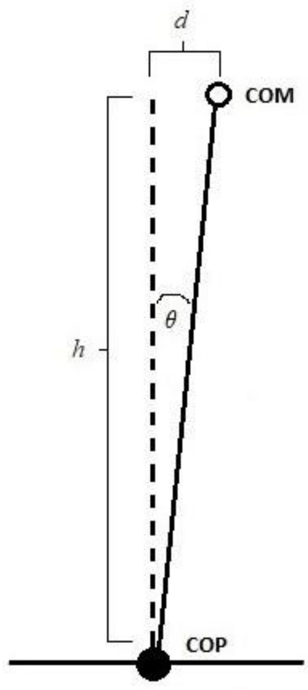


Figure 1. The calculation of the center of mass-center of pressure inclination angle uses the height of the COM (*h*) and the horizontal separation of the center of mass and center of pressure (*d*) to calculate the angle (θ), as shown above. This model can be applied to either the sagittal (anterior-posterior) or the frontal (mediolateral) planes.

Inclination angles are typically analyzed in the mediolateral (M-L) and anterior-posterior (A-P) directions. Typically, gait imbalance is associated with smaller peak A-P angles and greater peak M-L angles [16]. These peaks tend to occur at heel strike for the peak anterior COM-COP angle and at toe off for the peak posterior COM-COP angle [16], as weight is transferred from the trailing limb to the leading limb. Peak medial angles occur during double-support as well [16] The occurrence of these peaks at the start and end of the double-support phase indicate that the greatest area of interest for studying imbalance during gait is during the transfer of weight from one limb to the next.

COM can be calculated using a motion-sensing cameras and 39 reflective markers placed on a subject according to the VICON Plug-In Gait full body placement (see Figure 3) and two AMTI force plates. Segmental COMs are calculated and used to estimate whole body COM. The COP is recorded by each of the force plates when the foot is in contact with the plate. However, the area of interest occurs when one foot is on each force plate and the weight is distributed between both limbs. Therefore, COP must be calculated as a resultant COP using the following equation:

$$COP_R = COP_1 \frac{F_{Z1}}{(F_{Z2} + F_{Z1})} + COP_2 \frac{F_{Z2}}{(F_{Z1} + F_{Z2})}$$

where COP_R is the resultant COP, COP_1 and COP_2 are each COP for the two respective force plates, and F_{Z1} and F_{Z2} are the vGRF from each respective force plate [15,16,44].

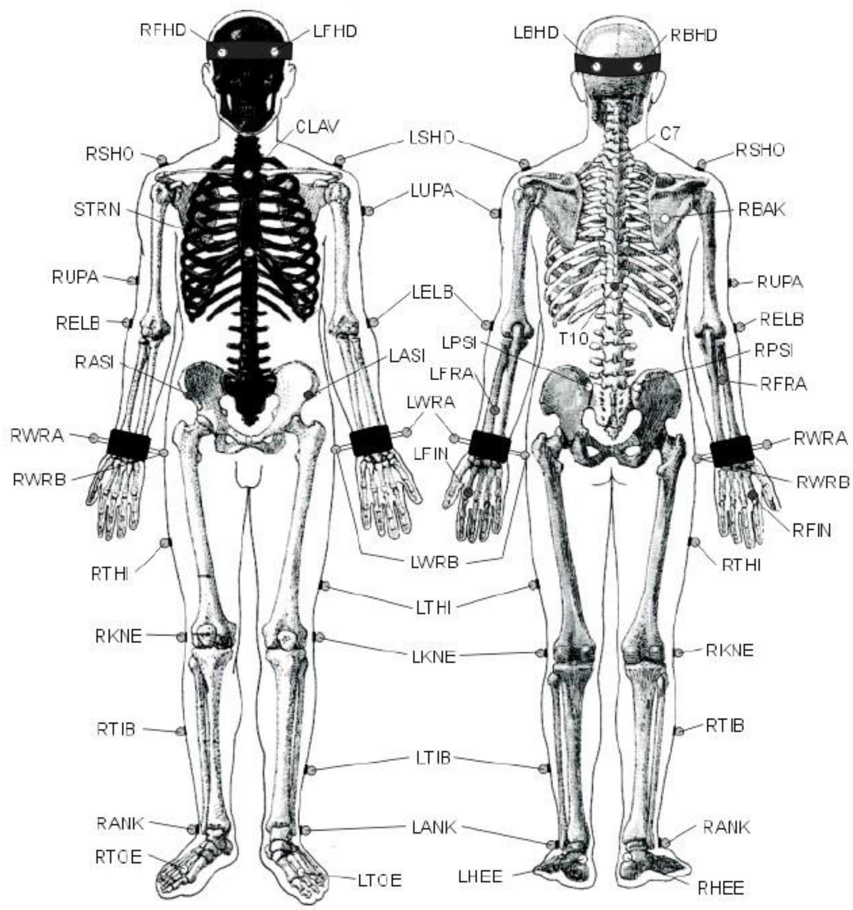


Figure 3. A diagram of the full-body marker placement as described by VICON Plug-In Gait (Centennial, CO, USA) as seen from an anterior (left) and posterior (right) view.

Acronyms: left/right forehead (LFHD/RFHD), left/right back of head (LBHD/RBHD), 7th cervical vertebrae (C7), 10th thoracic vertebrae (T10), clavicle (CLAV), sternum (STRN), right-side back (RBAK), left/right shoulder (LSHO/RSHO), left/right upper arm (LUPA/RUPA), left/right elbow (LELB/RELB), left/right forearm (LFRA/RFRA), left/right thumb-side wrist (LWRA/RWRA), left/right pinkie-side wrist (LWRB/RWRB), left/right 2nd metacarpal (LFIN/RFIN), left/right anterior superior iliac spine (LASI/RASI), left/right posterior superior iliac spine (LPSI/RPSI), left/right lateral mid-thigh (LTHI/RTHI), left/right lateral knee (LKNE/RKNE), left/right lateral mid-shank (LTIB/RTIB), left/right lateral malleolus of the fibula (LANK/RANK), left/right heel (LHEE/RHEE), and left/right 2nd metatarsal (LTOE/RTOE).

Typical values for COM-COP inclination angles

At preferred speeds, older adults with gait instability and a higher risk for falls tend to show an increase in medial or mediolateral (M-L) COM-COP peak IA while simultaneously showing a decrease in anterior and posterior (A-P) COM-COP peak IA (see Table 1). Fallers may not be willing to allow their COM to travel too far forward or back from their base of support, thus displaying a more cautious gait to prevent falls. This unwillingness to lean too far forward or back while walking may explain excess mediolateral movement, possibly resulting from a translation of momentum. The effects of HV on inclination angles have not yet been explored. Note that all studies listed in Table 1 did not control for gait speed, and that among those who reported gait speeds the older subjects with imbalance issues all walked at slower speeds. Gait speed has a large impact on gait parameters and COM movement during walking [18,46]. The possible consequences of not controlling for gait speed when studying dynamic balance in walking will be explored in a later section of this review.

	Self-Selected Gait Speed	Peak Anterior COM-COP IA	Peak Posterior COM-COP IA	Peak Medial COM-COP IA
Lee et al. 2006 [16]				
Healthy OA	1.16 m/s	13.18°	11.69°	4.06°
OA Fallers	1.01 m/s	10.31°	10.07°	5.89°
Lugade et al. 2008 [42][1] *	1.28 m/s	~13°	~13.5°	~4°
Healthy OA	1.00 m/s	~10°	~11.5°	~5.5°
OA requiring THA	1.14 m/s	~12.5°		4.98°
Mandeville et al. 2008 [108]*	0.94 m/s	~10.0°		5.13°
Healthy OA				
OA requiring TKR	Not reported	12.3° 10.3	11.3° 10.1°	4.6° 5.3°
Chen et al. 2010 [17]				
Healthy OA				
OA with Imbalance				

Table 3. A summary of results found by various studies examining peak A-P and M-L inclination angles when comparing healthy older adults against older adults with fall history, increased fall risk, or other conditions that may impact dynamic balance when walking. Some of those conditions included are subjects scheduled to receive a Total Hip Arthroplasty (THA) and Total Knee Replacement (TKR) surgery, which are a treatment for conditions associated with gait imbalance [109–111].

[1]indicates that exact inclination angles were not reported in the study, estimates are made from figures
*indicates that not all of the subjects in the study were within the ages of 60-80 years

Due to its complexity, dynamic balance is most frequently calculated using more than one method [91,97]. For this reason, this next section will cover a complementary method of calculating dynamic balance alongside COM-COP IA: double support duration.

Calculating double support duration

Double support duration (DSD) is calculated as the amount of time when both feet are in contact with the ground. Often this value is easily normalized for intersubject variability by presenting it as a percentage of the gait cycle. Force plates are commonly used to measure DSD due to their sensitivity. Two force plates are necessary for quantifying DSD to distinguish between each foot's contact. Using force plates, DSD has been calculated as the time when both force plates' vertical ground reaction forces exceed negligible values [19,31,107]. This method pairs well with measures of COM-COP interaction, such as IA, because COP coordinates can be calculated from force plate data. Generally, commercially-available force plate and motion capture software automatically calculate COP coordinates from forces and moments measured. DSD has been used previously to validate COM-COP IA as a measure of dynamic balance in walking [19,31,107].

Normal values for double support duration

DSD usually only lasts for about ~20% of the gait cycle, beginning at initial contact of the leading heel through pre-swing at the toe-off of the trailing leg [112]. Shorter DSD can indicate improvements in dynamic balance during walking [91,100]. Increased DSD is known to be affiliated with an increased risk for falls and gait disturbance [99,113]. Longer time spent in DSD may indicate the adoption of a more stable gait pattern to prevent falls and mitigate imbalance.

DSD increases drastically in older adults, showing DSD values ranging anywhere from 14% to 30% of the gait cycle [47,98,99]. Older adults with instability or increased fall risk tend to have these increased DSD values, rarely remaining in DSD for anything less than 20% of the gait cycle (see Table

2). Larger DSD values suggest the adoption of a safer, more stable walking pattern that indicate imbalance while walking. Older adults are known for adopting safer, more stable patterns of walking, which may explain the trend for increased DSD in older populations. It is reasonable to expect this type of pattern to be exacerbated in older adults also suffering from conditions that challenge balance such as HV.

Double support duration when hallux valgus is present

Very few studies have examined double support as a measure of dynamic balance in patients with HV [12,13]. Though the results were not statistically significant, Sadra and colleagues [13] did find an increasing trend in DSD when comparing patients without HV against those with preoperative and postoperative HV. This finding may suggest that dynamic balance may be affected in those suffering from the HV deformity. Mickle and colleagues [12] conversely found minimal to no impact of HV of DSD. Further evidence is necessary to better understand the true effect of HV on DSD and balance while walking.

	Self-Selected Gait Speed	Double Support Duration
Maki 1997 [98]		
Healthy OA/no fear of falling	0.79 m/s	14.1% GC
OA Fallers/with fear of falling	0.66-0.76 m/s	18.2-19.8% GC
Verghese et al. 2009 [99]		
Healthy OA	0.95 m/s	26.60-30.0% GC
OA Fallers	-0.1 m/s	+10% GC
Callisaya et al. 2011[114]		
Healthy OA	1.18 m/s	22.4% GC
OA Fallers	1.08-1.13 m/s	23.1-24.0% GC
Sadra et al. 2013 [13]*		
Healthy	1.25 m/s	18.30% GC
Preoperative HV	1.25 m/s	20.28% GC
Postoperative HV	1.06 m/s	23.03% GC
Mickle et al. 2011 [12]		
Healthy	1.00 m/s	23.6% GC
HV	1.03 m/s	24.0% GC

Table 4. A summary of results found in various studies examining DSD as a percentage of the gait cycle (GC) among older adult populations, populations of older adults experiencing imbalance or elevated fall risk during walking, and populations of older adults with hallux valgus (HV).

*indicates that not all of the subjects in the study were within the ages of 60-80 years
1. The researchers reported smaller than normal values because they chose to define DSD using only heel and metatarsal contact with the ground, excluding the period of toe contact to remove possible motion artifact that may have been caused by dragging the toes
2. This study evaluated the risk factors associated with falls. The researchers found that a 10% increase in DSD from the values reported for Healthy OA would result in a statistically significant increase in fall risk. This finding is reflected and supported by other studies [98,114].

The effect of walking speed on measures of dynamic balance

As mentioned previously in this review, walking speed has a large impact on parameters of gait. It is well known that walking speed influences spatiotemporal parameters, such as double support duration. In healthy adult subjects, double support duration decreases as walking speed increases [18]. As seen in Table 2, slower walking speed are correlated with longer DSD, though these values may also be impacted by age and instability. Other studies have reported that DSD values are longer at slower speeds and shorter at faster speeds [19,20]. COM movement is also affected by walking speed. Mediolateral COM movement decreases as walking speed increases in healthy, young adult subjects, though mediolateral COM excursion is smallest at preferred walking speeds [46]. If COM excursion is affected by walking speed, it seems likely that COM-COP inclination angles could possibly be impacted as well. Recent evidence has found that peak A-P COM-COP IA increase with increasing walking speed while M-L COM-COP IA decrease in younger adult subjects [19,20]. Considering this evidence, walking speed should be treated as a confounding variable when studying dynamic balance in walking.

There are counter-arguments that advocate for preferred walking speed. Preferred walking speed is more externally valid as it is more representative of an individual's natural gait pattern. However, considering the threat to the internal validity of any outcomes measured using preferred walking speed, it may be prudent to test both preferred and controlled speed to mitigate this issue when conducting research on walking imbalance. Controlled speed trials should be set at a speed most representative of the age group being studied. This can account for the effect of age on walking speed [115,116]. In this case, both controlled and preferred speed trials may result in very similar gait velocities for each subject as a result. Regardless, the inclusion of both a controlled and self-selected speed walking trial is essential to maintain both the internal and external validity of any study examining dynamic balance.

Conclusion

From the literature, it can be concluded HV is a condition with varying degrees of effect on foot strength, sensation, and function. The condition is progressive and irreversible without extreme treatment measures, and as a result tends to be more prevalent in older populations. This is concerning considering that the implications that HV has a detrimental effect on balance, both in standing and potentially also in walking. Balance is a key area of interest for older populations, who are at an increased risk for falls and often suffer from the serious complications that result from such incidents. Balance is difficult to measure, but there is a sensitive method of analysis that has yet to be used to verify the effects of HV on dynamic balance in walking. From this review, it is apparent that there is a need for research examining dynamic balance in walking in older adults with HV using more sensitive techniques.

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APPENDIX A

<u>Western Washington University Informed Consent</u>

The effects of bunions on walking balance in older adults at fixed speeds

We are asking you to be in a research study. Participation is voluntary. The purpose of this form is to give you the information you will need to help you decide whether to participate. Please read the form carefully. You may ask questions about anything that is not clear. When we have answered all of your questions, you can decide if you want to be in the study or not. This process is called "informed consent."

Purpose and Benefit:

This research aims to examine the effects of bunions on balance during walking in older adults at fixed walking speeds. The effects of this common foot condition on balance in older adults are not fully understood. This study will increase our current understanding of how bunions impact balance during walking in older adults.

I understand that:

• To be eligible for this research, participants should be:

  • between the ages of 60 and 85 years

  • able to walk continuously for at least 10 minutes

• Participant is aware that the entirety of the testing session will be conducted in bare feet while wearing spandex clothing and reflective markers placed on the clothes and the skin.

• Participation in this research will involve the completion of 5 walking trials at two controlled walking speeds (2.2 and 2.9 mph) for a total of 10 trials. Participation will require approximately a 1-hour time commitment.

• Participant's feet will be visually examined to determine if bunions are present on the feet or not. • There are minimal risks from participation in this research. Participant may experience some mild fatigue from the walking trials.

• The participant understands that there are no potential direct benefits from participation in this study.

• Participation is completely voluntary. Participants are able to withdraw from this research at any time without penalty or loss of benefits to which they are otherwise entitled.

• All information is confidential. Participants will be given an ID number for this study, which will be used to label their data. The link between this ID number and the participant's name and other identifying information will be stored separately. Only the primary investigator and graduate researcher will have access to the data collected from this study.

• Participant signature on this form does not waive any legal rights of protection.

• A physical copy of this informed consent form will be provided to the participant upon arrival at the laboratory.

This research is conducted by Carolyn Barbee under the supervision of Dr. Harsh Buddhadev. Any questions that you have regarding the study or your participation may be directed to Dr. Harsh Buddhadev at harsh.buddhadev@wwu.edu or to Carolyn Barbee at barbeec@wwu.edu.

If you have any questions about your participation or your rights as a research participant, you can contact the WWU Office of Research and Sponsored Programs (RSP) at compliance@wwu.edu or (360) 650-2146. If during or after participation in this study you suffer from any adverse effects as a result of participation, please notify the researcher directing the study or the RSP.

By signing below you are saying that you have read this form, that you have had your questions answered, that you understand the tasks involved, and volunteer to take part in this research.

I have read the above description and agree to participation in this study.

Participant's Signature:____ Date: _________

Participant's Printed Name:____

APPENDIX B: Gait & Posture Format Guidelines

GUIDE FOR AUTHORS

JOURNAL DESCRIPTION
Gait and Posture publishes new and innovative basic and clinical research on all aspects of human movement, locomotion and balance.

The topics covered include: Techniques for the measurement of **gait** and **posture,** and the standardization of results presentation; Studies of normal and **pathological gait**; Treatment of gait and **postural abnormalities**; Biomechanical and theoretical approaches to gait and posture; Mathematical models of **joint** and **muscle mechanics**; **Neurological** and **musculoskeletal** function in gait and posture; The evolution of **upright posture** and **bipedal locomotion**; Adaptations of carrying loads, walking on uneven surfaces, climbing stairs, running and performing other movements. Spinal biomechanics only if they are directly related to gait and/or posture and are of general interest to our readers; The effect of aging and development on gait and posture; Psychological and cultural aspects of gait; Patient education.

Submission checklist
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[3] W. Strunk Jr., E.B. White, The Elements of Style, fourth ed., Longman, New York, 2000.
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[4] G.R. Mettam, L.B. Adams, How to prepare an electronic version of your article, in: B.S. Jones, R.Z. Smith (Eds.), Introduction to the Electronic Age, E-Publishing Inc., New York, 2009, pp. 281–304.
Reference to a website:
[5] Cancer Research UK, Cancer statistics reports for the UK. http://www.cancerresearchuk.org/aboutcancer/statistics/cancerstatsreport/, 2003 (accessed 13 March 2003).
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[dataset] [6] M. Oguro, S. Imahiro, S. Saito, T. Nakashizuka, Mortality data for Japanese oak wilt disease and surrounding forest compositions, Mendeley Data, v1, 2015. https://doi.org/10.17632/xwj98nb39r.1.

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