Effect of arm motion on postural strategies during uphill and downhill walking

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Abstract

The aim of this study was to investigate the effects of arm motion and surface slope on postural strategies and gait stability. We hypothesized that active arm swing would increase postural control compared to walking with arms held and normal arm swing, and that holding the arms would lead to an increasing number of compensatory gait strategies with the aim of increasing balance, both uphill and downhill.

We tested fifteen healthy, young adults (age 23.4 ± 2.8 years) using the Computer-Assisted Rehabilitation ENvironment (CAREN) using a simulated rolling-hills condition under 3 arm swing conditions: held, normal, and active. Outcome measures included spatiotemporal gait parameters and postural stability measures in the 3 planes of motion (anterior-posterior, medial-lateral, and vertical).

No significant interaction effects between arm swing and surface slope were found. However, results showed main effect for arms (held, normal, active) and slope (uphill versus level walking, downhill versus level walking) conditions. Stepping and postural strategies when walking uphill compared to level were opposite to those used in downhill walking compared to level. Participants adopted an overall more cautious strategy when walking downhill, as seen by a combination of decreased cadence and increased double-support time, while the opposite strategies were seen in uphill walking. Effects of arm swing remained relatively consistent for both uphill and downhill walking conditions. Both uphill and downhill, holding the arms led to stability-seeking measures in the form of increasing base of support (double-support time), and increased control (decreased vertical accelerations of the head and trunk compared to normal and active arm swing).

These results substantiate the destabilizing effects of walking without arm swing and the usefulness of active arm swing for enhancing gait stability on minor slopes. This research also provides insight into the control mechanisms regulating dynamic balance in healthy young adults, which can be used to inform protocols and develop models aimed at preventing occupational health and safety hazards in challenging environments (e.g. construction workers).

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Abbreviations and operational definitions

Active arm swing	“Anterior arm swing should peak when arms are roughly shoulder height”
AP	Anterior-posterior
Cadence	The rate of walking measured in steps per minute
CoM	Centre of mass; 3D position of body’s centre of mass
CoV	Coefficient of variation; standard deviation / mean * 100
DH	Downhill walking (decline slope condition; -1 to -3º)
DST	Double-support time; when two feet are in contact with the support surface
Held arms	Arms volitionally held still in a relaxed manner at the participant’s sides
Level	Level walking
ML	Medial-lateral
Alignment	Postural alignment; position of C7 marker relative to pelvis
Step time	Time from heel strike of one foot to heel strike of the contralateral foot
Step width	Medial-lateral distance between subsequent (contralateral) heel strikes
SST	Single-support time; when only one foot is in contact with the support surface
UH	Uphill walking (incline slope condition; +1 to +3º)
VT	Vertical

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CHAPTER 1: INTRODUCTION

1.1 Introduction

The primary goals of human gait are forward progression and maintenance of postural stability (Woollacott & Tang, 1997). To accomplish these goals, different factors have to be controlled to allow for efficient walking. First, the body's arrangement above the legs is one of the critical factors in producing efficient gait. Keeping the body's centre of mass (CoM) as aligned as possible above the base of support reduces the amount of mechanical work required to move forward. Another apparent benefit of this alignment is the head's placement above the torso, as vision is an important factor in maintaining postural stability while we move around (Patla, 1991; Hallemans *et al.*, 2010; Logan *et al.*, 2010). Primarily, forward progression is driven by the legs; However, there is a natural anti-phase motion between the lower limbs and the trunk. While pelvis rotation adds to the movement of the legs by extending the stride length, the torso rotates in the opposite direction to create a dampening effect (Tomohisa, 2014; Stokes, Andersson & Forssberg, 1989). This leads to a lower energy expenditure and a smoother forward translation of the CoM. The upper limbs act as an extension of this anti-phase action of the trunk, and have been shown to improve mechanical stability (Ortega, Fehlman & Farley, 2009) and reduce vertical ground reaction moments (Li *et al.*, 2001). Physiological benefits have also been evidenced by findings of increased energetic cost of walking without arm swing (Ortega *et al.*, 2009; Umberger, 2009). Studies examining walking with the arms bound found similar increases in metabolic cost, as well as links to decreased postural stability (Collins, Adamczyk & Kuo, 2009; Yang *et al.*, 2015). Further, some have found active arm swing to increase gait stability (Nakakubo *et al.*, 2014; Punt *et al.*, 2015; Wu *et al.*, 2016).

A number of studies have been done to examine the effect of arm motion on postural stability during level walking (Elftman, 1939; Umberger, 2008; Collins *et al.*, 2009; Bruijn *et al.*, 2010). However, daily walking often involves changes in elevation that impose different demands on the body's postural system and walking stability. For example, uphill walking requires the body to pull itself upward against the force of gravity, consuming more energy than level and downhill walking (Minetti *et al.*, 2002). Conversely, downhill walking requires coping with inertial forces acting upon the body (Gottschall & Kram, 2005). To ensure walking stability in such situations, postural strategies also need to be modified when walking on unleveled grounds. During level walking the trunk and head have a coupled relationship (Menz, Lord & Fitzpatrick, 2003; Tucker *et al.*, 2008; Latt *et al.*, 2009). This head-trunk pattern has also been found during uphill and downhill walking (Cromwell, 2004). Synchronous movements of the head and trunk in the same direction are evidence of this relationship, which serves to maintain alignment of the head over the trunk (Cromwell, 2004).

Still, it is the legs which carry the entire upper body and interact with the walking surface. Adaptations in walking pattern can be defined using spatiotemporal gait parameters. These parameters capture the general walking characteristics of stride length, stride width, coefficient of variation, single and double support time, and cadence. Certain changes in these characteristics can be expected as a means of coping with specific demands. For example, double-support time is often favoured over single-support time when balance is challenged by uphill or downhill surface slopes (Kawamura, Tokuhiro & Takechi, 1991; Redfern & DiPasquale, 1997). This ratio of single to double-support time is expected to vary throughout the gait cycle. While this small amount of variance a necessary component of adaptability, large amounts of variation indicate problematic coping strategies (Vieira *et al.*, 2017). Decreased gait variability has been found in level walking with active arm swing as compared to normal or no arm swing, indicating increased walking stability (Wu *et al.* 2016). Lastly, changes in speed may be a result of changes in stride length, cadence, or both. Kawamura *et al.* (1991) found that steeper slopes resulted in decreased walking speed for both uphill and downhill conditions. This decreased speed primarily resulted from increased double support time when walking uphill, and decreased step length when walking downhill. Decreased speed corresponding with decreased step length was similarly seen by Kimel-Naor *et al.* (2017). While no differences were found in walking speed or stride length between normal and no arm swing (Bruijn *et al.*, 2010; Collins *et al.*, 2009; Nakakubo *et al.*, 2014), faster walking resulted from walking with active arm swing (Nakakubo *et al.*, 2014).

Certain limitations have been present in previous studies of uphill and downhill walking, such as being performed in restricted walking space which limits the number of continuous gait cycles (Gottschall *et al.*, 2011; McIntosh *et al.*, 2006; Prentice *et al.*, 2004; Redfern & DiPasquale, 1997). Additionally, use of a fixed-speed treadmill may restrict natural walking patterns. Instead, a self-paced treadmill may retain more of the integrity of variable gait characteristics, such as the fluctuations that occur over successive strides. Allowing for this freedom in gait characteristics ultimately better reflects overground walking (Dingwell *et al.*, 2001; Sloot, van Der Krogt & Harlaar, 2013).

CHAPTER 2: PURPOSE AND HYPOTHESES

2.1 Purpose

The purpose of this study is twofold: 1) To examine the effect of arm motion on spatial-temporal gait parameters during uphill and downhill walking; 2) To investigate the postural strategies adopted under different arm swing conditions during uphill and downhill walking.

2.2 Independent variables

1) Arm swing
    i. Held
    ii. Normal
    iii. Active (arm swing peak when arm approximately horizontal)
2) Surface conditions
    i. Level as a baseline
    ii. UH (uphill, +1 to +3º)
    iii. DH (downhill, -1 to -3º)

2.3 Dependent variables

1) Spatial-temporal gait parameters:
    i. Step time
    ii. Step width
    iii. Cadence (steps / min)
    iv. Single-support time (SST)
    v. Double-support time (DST)
2) Coefficients of variation (CoV, standard deviation / mean * 100)

   i. Step time CoV
   ii. Step width CoV
   iii. DST CoV

3) Kinematics in the three planes of motion (ML, AP, VT):
   i. Mean postural alignment (position of C7 marker relative to pelvis)
   ii. Mean centre of mass position (CoM)
   iii. Mean acceleration of head and trunk

2.4 Hypotheses

H1: It is hypothesized that active arm swing will be most helpful to walking stability, and held arm swing will lead to the adoption of an increasing number of compensatory measures in the form of altered gait strategy with the aim of increasing balance. More specifically,

H1a: In uphill: Step time, step width and DST will increase while cadence and SST will decrease.

H1b: In downhill: Step time , step width and SST will decrease while cadence and DST will increase.

H2: It is hypothesized that slopes will lead to the adoption of an increasing number of compensatory stability measures with increasing slope both UH an DH compared to level.

H2a: Alignment and CoM will be shifted forward UH compared to level

H2b: Postural alignment and CoM will be shifted backward DH compared to level

H2c: Acceleration of the head and trunk will be larger in both UH and DH than level.

H3: Gait will be least stable when walking without arm swing. Arms held condition will increase the coefficient of variability both UH and DH compared to level.

CHAPTER 3: REVIEW OF LITERATURE

3.1 Human gait

Primary goals of human locomotion include forward progression and maintenance of upright posture. For the purposes of analyzing the walking gait, the human body has frequently been divided into two sub-systems, the lower limbs and the upper part of the body. The upper body including the head, arm, and trunk (often referred to as HAT), with the lower limbs supporting and transporting the HAT (Winter, 1995; Cappozzo, 1991). As human gait originates from the legs, it is bipedal. Thus, gait can be further broken down into the two primary phases of stance (both feet on the ground) and swing (one foot). Stance and swing can be further subdivided into initial contact, early stance, mid stance, late stance, initial swing, mid swing, and terminal swing.

However, the bipedal nature of human walking poses inherent challenges such as the considerable distance of one's centre of mass from the supporting surface as well as the marked single-limb support phase in which the centre of gravity is passing outside the base of support. For this reason, postural alignment is a crucial aspect of balance while walking, as the HAT segment needs to be positioned as closely as possible above the base of support.

Tucker *et al.* (2008) began to bridge this gap by inducing voluntary sway movements and comparing static and dynamic conditions in both the young and the elderly. Participants were asked to either stand quietly (static) or to perform a voluntary anteroposterior or mediolateral sway (dynamic) prior to an auditory directional cue. Centre of pressure (CoP) as well as acceleration of the head and lower trunk in three planes were collected. They examined CoP-trunk-head coupling as the dependent measure. Results revealed that the elderly exhibited a smaller phase difference between CoP and head motion compared to the young, indicating a more rigid movement pattern. Most investigations of standing postural control have used CoP measures alone which has limited the information on intersegmental control. The addition of head and trunk measures provides more information regarding coordination of postural control. In examining phase differences for CoP measures of the trunk, head, and upper body as a whole in the dynamic reaction, a bottom-up coordination sequence was identified in the young. Alternately, the synchronous movement pattern seen in the elderly means they moved as a rigid unit. While this may seem to be a stable strategy in maintaining a standing position, the reduction in joint mobility and flexibility is limiting in dynamic conditions such as walking. Such reductions in mobility likely hinders one's ability to execute voluntary or compensatory responses.

Trunk control is critical not only to postural stability, but also for head stabilization to better govern vision (Menz, Lord & Fitzpatrick, 2003; Tucker *et al*., 2008). Menz (2003) used an irregular surface to investigate postural strategies used by healthy young individuals respond to challenging walking conditions. In investigating spatiotemporal gait parameters on the irregular surface compared to flat, participants were able to maintain velocity but had slower cadence and a significantly longer stride length. This adaptation is likely a result of having less interaction with potential tripping hazards. While the magnitude of pelvis accelerations increased, head accelerations remained unaffected by walking surface. This finding suggests that healthy, young individuals modify their stepping pattern on irregular surfaces to ensure head stability. Thus, a tighter coupling of the head, trunk, and pelvis segments during walking may contribute to more perturbed visual input and decreased ability to accurately respond to environmental changes while walking (Latt *et al.*, 2009). Further, less flexible postural strategies are linked to decreases in the body control necessary for initiating gait and gait changes.

3.2 Arm swing in gait

Arm swing during walking is a widely-experienced natural occurrence and has been observed in children as young as 18 months (Sutherland, 1980). Though, as the use of arm swing is not inherent to the definition of bipedal locomotion, contributions toward gait stability by the arms have gone relatively unexplored. Furthermore, two conflicting schools of thought have emerged. The first being that the arms act primarily in a pendulum-like manner and thus have negligible active role in gait, if any **(**Pijnappels *et al.*, 2010; Bruijn *et al.*, 2012). Conversely, many conclude that the arms do play an active role in gait (Elftman, 1939; Collins *et al.*, 2009; Ortega *et al.*, 2009; Umberger, 2009; Kuhtz-Bushbeck & Jing, 2012; Goudriaan *et al.*, 2014).

Within the literature demonstrating that the arms are active in gait and in gait stability, arm swing has been found to reduce net metabolic energy expenditure during walking by 5-7.7% as compared to walking with no arm swing (Ortega *et al.*, 2009; Umberger, 2008). Similarly, Collins *et al.* (2009) found significant increases in metabolic rate (7-12%) and increased vertical angular momentum when comparing bound arm swing and held arm swing to normal arm swing. Elftman first showed that arm swing acts to reduce angular momentum about the vertical axis, as well has having beneficial effects on the other two axes (Elftman, 1939). The anti-phase swing of the arms thus likely acts in part by reducing whole-body angular momentum. More recent experimental evidence also supports Elftman's results, showing that lack of arm swing and ipsilateral arm swing significantly increase whole-body angular momentum as compared to normal arm swing (Collins *et al.*, 2009). One conclusion from this evidence is that this overall reduction in angular momentum achieved in normal arm swing aids in maintaining upright posture with less muscular effort by the torso.

Further, Elftman successfully measured muscle activation at the shoulder complex, concluding that arm swing does not solely arise from passive, pendulum-like oscillations. This is further supported in work by Ballestros *et al.* (1965) which measured muscle activity with electromyography (EMG) and found muscle activation between 4-15% maximal voluntary contraction (MVC) among the Deltoids, Latissimus dorsi, Supraspinatus, and Trapezius muscles. Beyond this, (Kuhtz-Bushbeck *et al.*, 2012, Goudriaan *et al.*, 2014) shoulder muscle activation continued to be seen while walking even when arms were bound. This implies a corticospinal contribution to the shoulder muscle activation seen in arm swing.

Additionally, there is evidence suggesting links between arm swing and stability measures. Yang *et al.* (2015) examined these relationships by linking them to CoM measurements. Increased vertical CoM displacement indicates less steady gait, as it shows a poorer control of CoM momentum and requires more mechanical work from the body. In the arms-constrained condition as compared to normal arm swing, vertical CoM displacement was significantly greater and participants adopted shorter stride length. Reduction in stride length is one compensatory strategy for decreased stability. This response indicates that there is some effect on balance by the arms that is missing in the arms-constrained condition. Bruijn *et al.* (2010) similarly found that when a perturbation occurs to the upper body with the arms constrained, it will be less able to adequately respond to perturbation.

The growing evidence of decreased walking and postural control without arm motion led researchers to examine the effects of purposefully swinging the arms during walking (Hu *et al.*, 2012; Nakakubo *et al.*, 2014; Punt *et al.*, 2015; Wu *et al.* 2016). Hu *et al.* (2012) compared normal arm swing with restricted (bound) and active arm swing. They aimed to assess the contribution of arm swing to human dynamic stability in relation to age. Human dynamic stability during walking is defined as the ability to maintain balance during locomotion. Within this definition, local dynamic stability is the ability to attenuate minor perturbations that occur during steady state walking (Bruijn *et al.*, 2012; Bruijn *et al.*, 2013). This excludes major external events such as slips or trips. To quantify the local dynamic stability, the maximum finite Lyapunov exponent was used. This measures the sensitivity to minor perturbations that could disrupt local dynamic stability. No significant difference was found in the local dynamic stability between subjects walking with normal arm swing and with restricted arm swing. This is consistent with previous literature that found the effect of normal arm swing to contribute to overall stability of human gait rather than local dynamic stability (Collins *et al.*, 2009; Bruijn *et al.*, 2010; Pijnappels *et al.*, 2010). However, the local dynamic stability of all motion segments was significantly increased while walking with active arm swing compared to normal arm swing and restricted arm swing. Further, in analyzing the percentage decrease of the Lyapunov exponent while walking with active arm swing, they also found improvement in local dynamic stability to be even greater in the middle-aged group than in the young group. This finding further evidences the potential role of active arm swing to compensate for decreased human dynamic stability that occurs with aging.

Nakakubo *et al.* (2014) extended the work by Hu *et al*. (2012) with a focus on the trunk stability in elderly individuals while walking. Since trunk and arm movements interact with each other, they wanted to see the impact of deliberately altering arm swing. They used the same arm swing conditions of normal, held, and active. Acceleration and angular velocity of the trunk were measured while walking at a preferred walking speed. Trunk stability, particularly in the mediolateral direction, was significantly enhanced when participants walked with active arm swing. Participants also walked faster in the active arm swing condition as compared to normal arm swing and arms held. This response is in agreement with findings by Behrman, Teitelbaum and Cauraugh (1998) which showed that verbal instructions to emphasize arm swing increased walking speed, even in healthy elderly individuals. Consistent with previous findings by Bruijn *et al.* (2010) and Collins *et al.* (2009), no differences were found in the walking speed or stride length between normal arm swing and arms held conditions.

Lulic, Susic and Kodvanj (2008) had complementary findings to those of Nakakubo *et al.* (2014) in young individuals. They found a decreased displacement of the CoM in the mediolateral direction when young individuals walked with active arm swing. This decreased displacement of the CoM contributed to greater control of CoM momentum and thus greater walking stability. Therefore, active arm swing was successful in maintaining and improving walking stability in the mediolateral direction (Punt *et al.*, 2015; Nakakubo *et al.*, 2014; Hu *et al.*, 2012).

Where previous studies by Nakakubo *et al.* (2014) and Punt *et al.* (2015) focused on the stability of the trunk segment during walking, Wu *et al.* (2016) aimed to investigate the effects of active arm swing on local dynamic stability of the lower extremity joints in addition to the trunk. The two arm swing conditions used were normal arm swing and active arm swing. They also quantified the amount of gait variability to provide further insight into the difference between walking under the two different arm swing conditions. They hypothesized that using active arm swing, the dynamic stability of the trunk and lower extremity joints would be improved as seen through decreased variability. While no effects of arm swing condition were seen in lower limb joints, decreased gait variability was observed with active arm swing. This supports their hypothesis that active arm swing would improve dynamic stability of the trunk. Findings were consistent with previous studies (Nakakubo *et al.*, 2014; Punt *et al.*, 2015). They found increased trunk stability in the mediolateral direction with active arm swing as compared to normal arm swing.

3.3 Walking on sloped surfaces

The surfaces we interact with in daily life are not perfectly level. Indeed, both natural and built environments contain many surface variations including the presence of slopes. Increases in slope cause the generation of higher shear forces which also increase the potential for falls (Redfern & DiPasquale, 1997). For this reason, as well as the significant contribution of falls in the workplace, home, and public accessible areas to injury, changes in elevation are identified as a risk factor and major occupational hazard. This definition includes ramps, that present a falling hazard due to slip potentials and loss of balance. With increased ramp angle, step length and step period are found to significantly decrease, and patterns of increased anterior pelvic tilt coinciding with single-leg stance phases have emerged (McIntosh *et al.*, 2006; Redfern & DiPasquale, 1997). These findings suggest that different gait kinematics may be required with increasing surface slope.

Kawamura *et al.* (1991) conducted a study to determine spatiotemporal gait parameters (step length, stride width, cadence, speed, and single and double support time) during uphill and downhill walking. In both uphill and downhill conditions, a steeper slope resulted in decreased cadence. Moreover, at the steepest conditions for both uphill and downhill walking (9° and 12°) gait speed was found to decrease significantly. The main contributing factors for this decrease in speed in the uphill condition was a decrease in cadence and increased double support time. In the downhill condition, decreased step length led to decreased walking speed. Conversely, stride length increased significantly with increasing slope between 0 and 9 degrees as an attempt to maintain constant speed. No differences were found in stride width for uphill or downhill as compared to level walking.

Sun, Walters & Svensson (1996) also found increased step length with increasing incline, but only compared to downhill walking. Their observational study aimed to examine walking speed, cadence, and step length in an urban setting. Findings were in line with Kawamura *et al.* (1991) and Wall, Nottrodt & Charters (1981). Uphill walking was characterized by significant decreased mean walking speed, cadence, and step length with increasing slope. Due to the large sample size (n=1200 female and 1200 male) and range of pedestrians observed (10-75 years), they were also able to see that older individuals (55-75 years) had greater reductions in step length when walking downhill compared to younger individuals. Redfern & DiPasquale (1997) took a closer look at the downhill walking in healthy young adults. They primarily examined sagittal plane body movements. Step length and period decreased as ramp angle increase; gait speed did not significantly change. This confirmed previous findings showing that stride length, step period, and walking velocity while walking downhill at a self-selected pace decreased with increasing ramp angle.

As no previous investigations had been done on how the trunk and pelvis assist the lower limbs in the process of adaptation to incline walking, Leroux, Fung, and Barbeau (2002) investigated postural adaptation to walking on inclined surfaces. The pelvis and trunk represent a large portion of body mass, which interact with the lower limbs to accomplish efficient gait. It had been previously shown in level walking that the pelvis and trunk rotate in opposite directions during most of the gait cycle, creating a damping effect that creates a smooth gait cycle (Stokes *et al.*, 1989). This relative coordination pattern observed in level walking did not change with inclined walking (Leroux *et al.*, 2002). Data recorded from subjects walking on a treadmill with up to 20° incline or decline showed that pelvic rotation was the same as that on level surface (Wall *et al.*, 1981). As the trunk and pelvis assist lower limbs in the gait adaptations inherent to daily life (e.g. stair climbing, changes in walking speed or slope), they are important considerations in such models and calculations. In the sagittal plane, movements were greater for the pelvis than for the trunk. Postural strategies included a forward tilt of the trunk and pelvis during uphill walking and a backward tilt during downhill walking, as compared to level walking.

The increase in stride length during uphill walking reported by Kawamura (between 0 and 9 degrees) was also seen by Leroux *et al.* (2002). At the same time, a progressive forward tilt of the pelvis and trunk was observed with increasing uphill slope during walking. Both uphill and downhill walking elicited changes in trunk position and pelvis postural alignment in the sagittal plane. However, they also included a quiet standing measure of postural control on the same uphill and downhill slopes as a comparison. When standing, the trunk and pelvis remained aligned with respect to the Earth's vertical regardless of surface incline. Uphill walking and downhill walking causing changes not seen in standing on the same slopes show that postural adaptations are task-specific. In walking, the CoM is repositioned to minimize mechanical work.

During downhill walking the CoM is moved backward to decrease the forward and downward momentum produced by the slope and gravity. The forward trunk tilt observed in uphill walking moves the centre of gravity ahead of the base of support to aid in forward propulsion. A similar increase in pelvic tilt during uphill walking was observed by McIntosh *et al.* (2006). McIntosh *et al.* (2006) identified a pattern of increased anterior pelvic tilt that coincided with single support stance on each side. This is also in accordance with previous findings (Leroux *et al.*, 1999, 2002) and further evidences the role of pelvic tilt in assisting trunk positioning.

Kimel-Naor, Gottlieb, and Plotnik (2017) included both upper body kinematics and gait coordination parameters in their self-paced treadmill study of the effect of uphill and downhill walking. They used the CAREN system and the self-paced treadmill mode to test 11 young, healthy participants. Stride length, gait speed, and cadence were decreased in the downhill condition as opposed to level walking. Gait speed and cadence were increased in downhill walking compared to uphill, but not compared to level walking. During uphill walking, an average decrease in walking speed occurred. Trunk and pelvic sagittal angles were different in both uphill and downhill walking as compared to level in accordance with previous findings (McIntosh *et al.*, 2006; Leroux *et al.*, 1999, 2002). Unlike previous studies, they determined uphill walking to have a larger impact on gait kinematics than downhill walking. This was the first study to show that upper limb kinematics are systematically affected during walking on inclined surfaces. They propose that these discrepancies arise from the variable walking speed used in this study. Still, they are unable to reach a conclusion as to the relative effect of walking speed and incline on specific gait parameters. This further highlights the need to facilitate increasingly natural walking conditions as technology allows. In this way, more potentially interacting factors (e.g. gait speed and slope) are accounted for.

CHAPTER 4: METHODOLOGY

4.1 Subjects

Fifteen healthy, young adults (female = 7, male = 8) were recruited using word of mouth. This sample size was based on similar protocol used by Bruijn *et al.*, 2010. Inclusion criteria were as follows:

- Between the ages of 18-30
- No neurological or orthopedic disorders that could affect gait
- No musculoskeletal injury within the past six months

Prior to recruitment and study commencement, approval of the study by the University of Ottawa Research Ethics Board and Ottawa Hospital Research Ethics Board was completed.

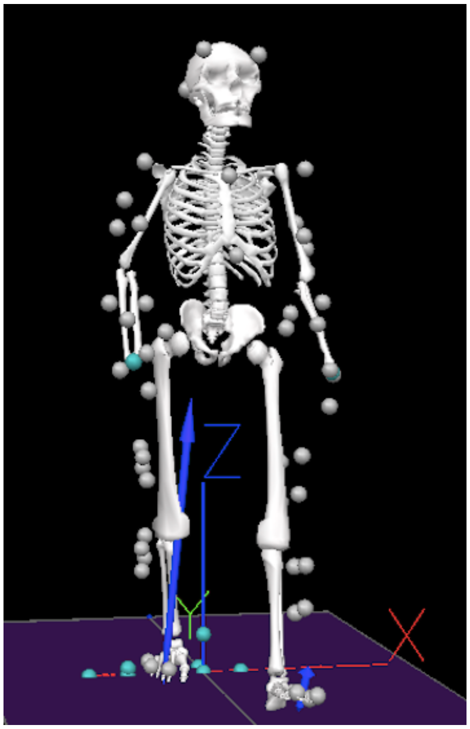

Figure 1. 57-marker set

4.2 Instrumentation

Participants were tested using the Computer Assisted Rehabilitation ENvironment (CAREN, CAREN-Extended, Motek Medical, Amsterdam, The Netherlands) located at the Ottawa Hospital Rehabilitation Centre. This system combines a 6 degree of freedom (6DOF) platform with integrated split-belt treadmill (Bertek Corp., Columbus, OH), as well as a 12-

camera Vicon motion capture system (Vicon 2.6, Oxford, UK), and 180° virtual reality screen. Motion data was gathered at a rate of 100 Hz, and force data gathered at 1000 Hz.

A 57-marker set (Figure 1) was used to capture motion data (Wilken *et al.*, 2012). The park terrain conditions application within the CAREN system was used (Lemaire *et al.*, 2012).

4.3 Experimental protocol

At the beginning of the testing session, informed consent was gathered. Participants first completed a 5-minute acclimatization period. After this, participants walked 20 m on a flat section of the park application. Then walked over 20 m of 3 different types of terrain in the Park – an ML translational perturbation, an AP rotational perturbation (hilly perturbation) both a sum of sin waves, and a pseudorandom perturbation in 3 different axes (rocks – vertical pitch and roll). Between each terrain tile they walked 40 m on a flat surface. The terrain was programmed such that it posed a balance challenge for the participants. This protocol was completed with self-selected, self-propelled walking speed and under each of the three arm swing conditions: held, normal, and active. The held arm swing condition was set as arms volitionally held still in a relaxed manner at the participant's sides. Subjects were instructed to ensure that anterior arm swing should peak when arms are roughly shoulder height for the active arm swing condition. Participants were tested wearing their personal footwear (running shoes). The 3 trials took approximately 10 min to complete.

4.4 Data analyses

All data was imported into Visual3D (C-Motion, Germantown, MD). Raw data was filtered at 12 Hz using a 4th order, zero-lag low-pass Butterworth filter (Winter, 2009). Heel strike gait events were manually identified using ground reaction forces.

The following variables were then calculated:

1) Spatial-temporal gait parameters:
    i. Step time
    ii. Step width
    iii. Cadence (steps / min)
    iv. SST
    v. DST
2) Coefficients of variation (CoV, standard deviation / mean * 100)
    i. Step time CoV
    ii. Step width CoV
    iii. DST CoV
3) Kinematics in the three planes of motion (ML, AP, VT):
    i. Mean alignment
    ii. Mean CoM
    iii. Mean acceleration of head and trunk

4.5 Statistical analyses

A three (arm swing) by two (slope) experimental design was be performed to quantify gait patterns and postural variables amongst healthy, young adults and assess the effects of arm motion on postural strategies during uphill and downhill compared to level walking. Three arm swing conditions included: held, normal, and active; two surface conditions comprised of one sloped condition (UH or DH) compared to level. The dependent variables being examined were 1) spatial-temporal gait parameters of gait: step time, cadence, step width, SST, and DST, as well as coefficients of variation for step time, step width, and DST and 2) kinematics in the ML, AP, and VT directions of motion: alignment, CoM, and accelerations of the head and trunk. Due to the varying changes of horizontal distance resulting from the AP fluctuations of the platform that comprised the rolling hills condition, step time was used rather than step length.

Two-way repeated measures ANOVAs were used to test for significant differences between and within surface conditions (UH, DH) each compared to level and arm conditions (held, normal, active) using IBM SPSS Statistics 25 (IBM Analytics, Armonk, USA). Shapiro-Wilks was used to test for normal distribution. Significance level was set at $p < .05$. A Bonferroni correction was used as needed.

One-way repeated measures ANOVAs were used within each sloped condition (UH, DH) to characterize the effects of arm swing. Significance level was set at $p < .05$. A Bonferroni correction was used as needed.

CHAPTER 5: MANUSCRIPT

Effect of arm motion on postural strategies during uphill and downhill walking

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Abstract

The aim of this study was to investigate the combined effects of arm motion and surface slope on postural strategies and gait stability in a healthy population.

The Computer-Assisted Rehabilitation ENvironment was used to test participants (n = 15) in a simulated rolling-hills environment which imparted both incline and decline conditions (-3 to +3º). The protocol was completed under three arm swing conditions: held, normal, and active. Outcome measures included spatiotemporal gait parameters and postural control measures in the anterior-posterior (AP), medial-lateral (ML), and vertical (VT) directions. Gait parameters included step time, step width, cadence, single-support time, and double-support time, as well as coefficient of variation for step time, step width, and double-support time; Postural measures included mean postural alignment (position of C7 relative to pelvis), mean center of mass position, head acceleration, and trunk acceleration. No significant interaction effects between arm swing and surface slope were found. However, results showed main effect for arms and for slope (uphill versus level and downhill versus level). Stepping and postural strategies when walking uphill compared to level were opposite to those used in downhill walking compared to level. Participants adopted an overall more cautious strategy when walking downhill, as seen by a combination of decreased cadence and increased double-support time, while the opposite strategies were seen in uphill walking. As for the effect of arm motion, walking with arms held resulted in participants increasing double-support time and increasing control of the head and trunk in VT compared to normal and active arm swing. Conversely, active arm swing decreased double-support time and caused a forward shift in postural alignment. These results substantiate the destabilizing effects of walking without arm swing and the usefulness of active arm swing for enhancing gait stability on minor slopes.

Keywords: Stability, Posture, Gait, Arm swing, Uphill, Downhill

5.1 INTRODUCTION

The primary goals of human gait are efficient forward progression and maintenance of postural stability (Woollacott & Tang, 1997). While the legs are the primary contributor to forward progression, natural anti-phase motion between the lower limbs and the trunk contributes to both progression and stability (Stokes, Andersson & Forssberg, 1989). The pelvis rotation adds to the movement of the legs by extending the stride length while the torso rotates in the opposite direction to create a dampening effect (Tomohisa, 2014; Stokes *et al.*, 1989). This leads to a lower energy expenditure and a smoother translation of the center of mass (CoM) (Lin, Gfoehler & Pandy, 2014).

Often overlooked when assessing gait, the arms act as an extension of this anti-phase action of the trunk, and have been shown to improve mechanical stability (Ortega, Fehlman & Farley, 2008) and reduce vertical ground reaction moments (Li *et al.*, 2001; Yang *et al.* 2015). Because arm swing serves to further counteract angular momentum about the vertical axis in this way, less neuromuscular effort is required to maintain balance when walking with natural arm swing (Ortega *et al.*, 2008; Collins, Adamczyk & Kuo, 2009; Yang *et al.*, 2015). Physiological benefits have also been evidenced by findings of increased energetic cost of walking in the absence of arm motion (Ortega *et al.*, 2008; Umberger, 2008). Without arm swing, this cancellation effect is absent and vertical ground reaction force is amplified (Yang *et al.*, 2015). Studies examining walking with the arms bound found similar increases in metabolic cost, as well as links to decreased postural stability (Collins *et al.*, 2009; Yang *et al.*, 2015). The growing evidence of decreased walking and postural stability when walking without arm swing led researchers to examine the effects of purposefully swinging the arms during walking (Hu *et al.*, 2012; Nakakubo *et al.*, 2014; Punt *et al.*, 2015; Wu *et al.*, 2016). Hu *et al.* (2012) looked specifically at local dynamic stability, which is the ability to attenuate minor perturbations that occur during steady state walking (Bruijn *et al.*, 2012; Bruijn *et al.*, 2013). Consistent with previous literature, no significant difference was found in the local dynamic stability between subjects walking with normal arm swing and arms bound (Collins *et al.*, 2009; Bruijn *et al.*, 2010; Pijnappels *et al.*, 2010). Active arm swing, however, significantly increased the local dynamic stability of all motion segments compared to natural arm swing and bound arms (Hu *et al.*, 2012). Studies extending this work found that trunk stability, particularly in the mediolateral direction, was significantly enhanced when participants walked with active arm swing (Lulic *et al.*, 2008; Hu *et al.*, 2012; Nakakubo *et al.*, 2014; Punt *et al.*, 2015; Wu *et al.*, 2016).

However, daily walking often involves changes in elevation that impose different demands on the body's postural system and therefore on walking stability. For example, uphill walking requires the body to pull itself upward against the force of gravity, consuming more energy than level and downhill walking (Minetti *et al.*, 2002). Conversely, downhill walking requires coping with inertial forces acting on the body (Gottschall & Kram, 2005). To ensure walking stability in such situations, postural strategies need to be modified when walking on unleveled grounds. During level walking the trunk and head have a coupled relationship (Menz, Lord & Fitzpatrick, 2003; Tucker *et al.*, 2008; Latt *et al.*, 2009). This head-trunk pattern has also been found during uphill and downhill walking (Cromwell, 2004). Synchronous movements of the head and trunk in the same direction are evidence of this relationship, which serves to maintain alignment of the head over the trunk (Cromwell, 2004).

Responses to challenging terrain can be seen in modifications to the base of support via spatiotemporal gait parameters (step length, cadence, step width, and single and double support time). Certain changes in these characteristics can be expected as a means of coping with specific demands. For example, increasing time spent in double-support when balance is challenged by uphill or downhill surface slopes (Kawamura, Tokuhiro & Takechi, 1991; Redfern & DiPasquale, 1997). As environmental demands are continually changing, a small amount of variability is a necessary component of adaptability, yet large amounts of variability may indicate poorer gait stability and instead reveal inadequate coupling strategies (Vieira *et al.*, 2017). Steeper slopes were shown to result in decreased walking speed for both uphill and downhill conditions (Kawamura *et al.*, 1991). This decreased speed primarily resulted from increased double support time when walking uphill (Kawamura *et al.*, 1991), and decreased step length when walking downhill (Kawamura *et al.*, 1991; Kimel-Noar *et al.*, 2017).

The purpose of this study was to examine the effect of arm motion on spatiotemporal gait parameters and postural strategies during uphill and downhill walking. We hypothesized that active arm swing would be most helpful to walking stability, and that held arm swing would lead to the adoption of an increasing number of compensatory gait strategies with the aim of increasing balance both uphill and downhill.

5.2 METHODOLOGY

A convenience sample of 15 healthy, young adults (eight male, seven female; age 23.4 ± 2.8 years; height 170.2 ± 8.1 cm; weight 72.3 ± 13.5 kg) were recruited from the University of

Ottawa. Participants had no neurological or orthopedic disorders that could affect gait and no musculoskeletal injury in the 6 months previous. Prior to study commencement, the study was approved by the Institutional Review Board (University of Ottawa) and the Ottawa Hospital Research Ethics Board and all participants provided written informed consent.

5.2.1 Data collection

3D motion capture analysis was completed using the Park setting (Lemaire et al., 2012) of the Computer-Assisted Rehabilitation Environment (CAREN) (CAREN-Extended, Motek Medical, Amsterdam, The Netherlands). This system combines a 6 degree of freedom platform with integrated split-belt treadmill (Bertek Corp., Columbus, OH), 12-camera VICON motion capture system (Vicon 2.6, Oxford, UK), and 180° projector screen. Platform motion was tracked by three markers, and full body kinematics were collected using a 57-markers set (Wilken et al., 2012; Sinitski et al., 2015). Motion data were gathered at a rate of 100 Hz, and force data collected at 1000 Hz.

5.2.2 Experimental protocol

Participants first completed a 5-minute acclimatization period before walking 20 m on a flat section of the Park application. Then they walked over 20 m of 3 different types of terrain in the Park: a medial-lateral translational perturbation, an anterior-posterior rotational perturbation (hilly perturbation) both a sum of sin waves, and a pseudorandom perturbation in 3 different axes (rocks – vertical pitch and roll). The terrain was programmed such that it posed a balance challenge for participants. Between each terrain tile they walked 40 m of flat section. This protocol was completed with self-paced treadmill mode under each of the three arm conditions: Held, Normal, and Active. Arms held was set as arms volitionally held in a still, relaxed manner at the participant's sides. For the active arm swing condition, participants were instructed that anterior arm swing should peak when arms are roughly shoulder height. Participants were tested wearing their personal footwear (running shoes). The three trials took approximately 10 minutes to complete. This study focused on the hilly perturbation and used the flat section immediately previous as a level baseline comparison.

5.2.3 Data analyses

Data was imported into Visual3D (C-Motion, Germantown, MD). Raw data was filtered at 12 Hz using a 4th order, zero-lag low-pass Butterworth filter (Winter, 2009). Heel strike gait events were manually identified using ground reaction forces Visual3D v6. Spatiotemporal gait parameters (step time, cadence, step width, single-support time, and double-support time) and their coefficients of variation (CoV) were calculated, as well as kinematics (Mean postural alignment, CoM, mean accelerations of the head and trunk) in the three directions of motion (ML, AP, VT). Uphill walking was defined as steps occurring between surface angles of +1 and +3 degrees; Steps were considered downhill if they took place at surface angles between -1 and -3 degrees.

5.2.4 Statistical analyses

A three (arm swing) by two (surface slope) experimental design was performed to quantify gait patterns and postural variables amongst healthy, young adults and assess the effects of arm motion on postural strategies during uphill and downhill (each compared to level) walking. Three arm swing conditions included: held, normal, and active. Dependent variables examined included: 1) spatiotemporal parameters of gait: step time, cadence, step width, single-support time, double-support time, and coefficients of variation for step time, step width, and double-support time and 2) kinematics in ML, AP, and VT directions: mean postural alignment (C7 marker to pelvis), mean CoM position, and mean accelerations of the head and trunk. Due to the varying changes of horizontal distance resulting from the AP fluctuations of the platform that comprised the rolling hills condition, step time was used rather than step length.

Two-way repeated measures ANOVAs were used to test for significance between and within surface conditions (uphill vs level, downhill vs level) using IBM SPSS Statistics 25 (IBM Analytics, Armonk, USA). Additionally, one-way repeated-measures ANOVAs were used to test for significance within each slope condition (uphill, downhill). Shapiro-Wilks test was used to test for a normal distribution. Significance level was set at $p < .05$. A Bonferroni correction was used for post-hoc tests.

5.3 RESULTS

A two-way repeated measures ANOVA revealed no significant interaction effects between arm swing and surface slope. However, results showed main effect for arms (held, normal, active) and slope (uphill versus level walking and downhill versus level walking) conditions.

5.3.1 Arm swing

Main effects of arm swing remained relatively consistent for both uphill and downhill walking conditions. Both uphill and downhill, arm swing significantly affected step time, cadence, single-support time, and double-support (Table 1). Significant differences in AP postural alignment, vertical CoM, and vertical accelerations of the head and trunk were also seen for both uphill and downhill walking.

5.3.1.2 Arm swing during uphill walking

Active arm swing significantly increased step time ($F_{(2,15)} = 15.293$, $\eta_p^2 = .522$, $p < .001$) compared to arms held ($p = .006$) and natural arm swing ($p < .001$). Active arm swing also decreased cadence ($F_{(2,15)} = 14.015$, $\eta_p^2 = .500$, $p < .001$) compared to arms held ($p = .007$) and natural arm swing ($p < .001$). A main effect of arm swing was found for both single-support ($F_{(2,15)} = 12.354$, $\eta_p^2 = .469$, $p < .001$) and double-support time ($F_{(2,15)} = 12.354$, $\eta_p^2 = .469$, $p < .001$). Post-hoc analysis revealed significantly shorter single-support time, and longer double-support time, with the arms held compared to normal ($p = .026$) and active arm swing ($p = .001$). No significant differences were found for step width ($F_{(2,15)} = 1.338$, $\eta_p^2 = .087$, $p = .259$).Main effects of arm swing for step width CoV ($F_{(2,15)} = 3.998$, $\eta_p^2 = .222$, $p = .030$) were found, but no significant post-hoc effects. No significant differences were found for step time CoV ($F_{(2,15)} = 0.301$, $\eta_p^2 = .021$, $p = .743$) or double-support time CoV ($F_{(2,15)} = 0.202$, $\eta_p^2 = .014$, $p = .818$).

Active arm swing resulted in an anterior shift in postural alignment ($F_{(2,15)} = 10.787$, $\eta_p^2 = .435$, $p < .001$) compared to arms held ($p = .002$) and natural arm swing ($p = .021$). Main effects of arm swing for AP acceleration the trunk ($F_{(2,15)} = 3.414$, $\eta_p^2 = .196$, $p = .047$) were found. Post-hoc analysis revealed significantly greater AP acceleration of the trunk with normal arm swing compared to active arm swing ($p = .046$). CoM ($F_{(2,15)} = 19.735$, $\eta_p^2 = .535$, $p < .001$) was significantly higher under active arm swing compared to arms held ($p = .008$) and normal arm swing ($p < .001$). Lastly, holding the arms led to significant decreases in the magnitude of vertical accelerations of the head ($F_{(2,15)} = 6.166$, $\eta_p^2 = .306$, $p = .006$) and trunk ($F_{(2,15)} = 6.006$, $\eta_p^2 = .300$, $p = .007$) with arms held compared to active arm swing ($p < .05$). No main effects were found in the ML direction for postural alignment ($F_{(2,15)} = 1.219$, $\eta_p^2 = .080$, $p = .311$), CoM

($F_{(2,15)} = 1.827$, $\eta_p^2 = .115$, $p = .179$), head acceleration ($F_{(2,15)} = 0.546$, $\eta_p^2 = .038$, $p = .585$), or trunk acceleration ($F_{(2,15)} = 0.667$, $\eta_p^2 = .045$, $p = .521$).

5.3.1.3 Arm swing during downhill walking

Active arm swing significantly increased step time ($F_{(2,15)} = 13.974$, $\eta_p^2 = .500$, $p < .001$) compared to arms held ($p = .006$) and normal arm swing ($p = .001$). Active arm swing also decreased cadence ($F_{(2,15)} = 12.607$, $\eta_p^2 = .474$, $p = .001$) compared to arms held ($p = .009$) and normal arm swing ($p = .001$). A main effect of arm swing was found for both single-support ($F_{(2,15)} = 13.941$, $\eta_p^2 = .499$, $p < .001$) and double-support time ($F_{(2,15)} = 13.941$, $\eta_p^2 = .499$, $p < .001$). Post-hoc analysis revealed significantly longer single-support time, and shorter double-support time, with the active arm swing compared to arms held ($p = .002$) and normal arm swing ($p = .005$). No significant differences were found for step width ($F_{(2,15)} = 0.623$, $\eta_p^2 = .043$, $p = .544$).

Main effects of arm swing for step width CoV ($F_{(2,15)} = 3.676$, $\eta_p^2 = .208$, $p = .038$) were found, but no significant post-hoc effects. No significant differences were found for step time CoV ($F_{(2,15)} = 2.471$, $\eta_p^2 = .150$, $p = .103$) or double-support time CoV ($F_{(2,15)} = 0.016$, $\eta_p^2 = .001$, $p = .984$).

Active arm swing resulted in an anterior shift in postural alignment ($F_{(2,15)} = 12.179$, $\eta_p^2 = .465$, $p < .001$) compared to arms held ($p = .005$) and natural arm swing ($p = .001$). CoM ($F_{(2,15)} = 15.139$, $\eta_p^2 = .520$, $p < .001$) was significantly lower with natural arm swing compared to arms held ($p = .038$) and active arm swing ($p = .002$), and significantly higher under active arm swing compared to arms held ($p = .009$). Lastly, holding the arms led to significant decreases in the magnitude of vertical accelerations of the head ($F_{(2,15)} = 6.812$, $\eta_p^2 = .327$, $p = .004$) and trunk

($F_{(2,15)} = 6.369$, $\eta_p^2 = .313$, $p = .005$) with arms held compared to normal arm swing ($p < .05$). No main effects were found in the ML direction for postural alignment ($F_{(2,15)} = 0.010$, $\eta_p^2 = .001$, $p = .944$) , CoM ($F_{(2,15)} = 1.405$, $\eta_p^2 = .091$, $p = .262$), head acceleration ($F_{(2,15)} = 0.758$, $\eta_p^2 = .051$, $p = .051$), or trunk acceleration ($F_{(2,15)} = 0.354$, $\eta_p^2 = .025$, $p = .705$).

Table 1
Comparison of the three arm swing conditions during uphill and downhill walking. All data are represented as the mean values averaged for all 15 participants, mean (standard deviation). One-way repeated measures ANOVA results with statistical significance set at *p* < .05 with Bonferroni correction.

	Uphill			**Downhill**		
Outcome measure	**Held**	**Normal**	**Active**	**Held**	**Normal**	**Active**
Step time (s)	0.496 (0.054) [a]	0.495 (0.033) [a†]	0.551 (0.044)	0.511 (0.055) [a]	0.516 (0.035) [a†]	0.579 (0.061)
Cadence (steps/min)	123 (12.585) [a]	122 (8.620) [a†]	110 (9.084)	119 (14.144) [a]	117 (8.608) [a†]	105 (10.958)
Step width (cm)	20.72 (4.336)	19.87 (3.438)	20.71 (4.564)	23.17 (4.863)	22.26 (4.328)	23.00 (5.797)
Single-support time (s)	0.691 (0.038)	0.713 (0.044) [b]	0.726 (0.044) [b†]	0.664 (0.040) [a†]	0.673 (0.040) [a]	0.698 (0.040)
Double-support time (s)	0.309 (0.038)	0.287 (0.045) [b]	0.274 (0.044) [b†]	0.336 (0.040) [a†]	0.327 (0.039) [a]	0.302 (0.040)
Coefficient of variation						
Step time (%)	5.453 (1.477)	5.557 (1.821)	5.181 (1.304)	7.067 (3.413)	6.406 (2.730)	5.106 (2.157)
Step width (%)	8.668 (4.117)	10.321 (5.368)	13.826 (7.509)	8.212 (3.518)	8.898 (4.418)	13.725 (9.910)
DST (%)	11.326 (2.391)	11.169 (3.412)	10.650 (3.442)	9.488 (3.274)	9.544 (1.985)	9.344 (4.286)
ML direction						
Alignment (m)	0.001 (0.013)	0.004 (0.013)	0.001 (0.016)	0.001 (0.014)	0.001 (0.014)	0.001 (0.016)
CoM (m)	-0.001 (0.023)	-0.006 (0.020)	-0.010 (0.023)	-0.002 (0.021)	-0.011 (0.021)	-0.004 (0.024)
Head acceleration (m/s^2)	-0.040 (0.192)	-0.017 (0.163)	-0.085 (0.200)	-0.047 (0.183)	-0.003 (0.162)	-0.070 (0.240)
Trunk acceleration (m/s^2)	-0.038 (0.156)	-0.017 (0.143)	-0.078 (0.176)	-0.038 (0.145)	-0.018 (0.158)	-0.065 (0.226)
AP direction						
Alignment (m)	-0.033 (0.070) [a]	-0.038 (0.068) [a]	-0.056 (0.066)	-0.016 (0.070) [a]	-0.019(0.069)[a†]	-0.040 (0.069)
CoM (m)	0.140 (0.063)	0.152 (0.050)	0.129 (0.061)	0.095 (0.058)	0.0968 (0.049)	0.097 (0.055)
Head acceleration (m/s^2)	0.008 (0.067)	-0.021 (0.056)	0.035 (0.088)	-0.014 (0.097)	-0.054 (0.142)	-0.069 (0.155)
Trunk acceleration (m/s^2)	-0.032 (0.056)	-0.067 (0.043) [a]	-0.027 (0.069)	0.046 (0.077)	0.065 (0.082)	0.022 (0.094)
VT direction						
Alignment (m)	0.447 (0.030)	0.445 (0.031)	0.445 (0.032)	0.448 (0.029)	0.446 (0.030)	0.446 (0.031)
CoM (m)	0.941 (0.051) [a]	0.938 (0.052) [a†]	0.947 (0.051)	0.940 (0.052) [a]	0.937 (0.051) [ab]	0.945 (0.052) [b]
Head acceleration (m/s^2)	-0.073 (0.052) [a]	-0.105 (0.078)	-0.134 (0.099)	0.080 (0.091)	0.166 (0.121)	0.144 (0.107) [b]
Trunk acceleration (m/s^2)	-0.069 (0.048) [a]	-0.102 (0.083)	-0.127 (0.093)	0.076 (0.084)	0.160 (0.116) [b]	0.134 (0.100)

[a] Different from Active

[b] Different from Held

[†] Significant at $p \leq 0.001$

5.3.2 Slope

Main effects of slope were found for vertical and AP acceleration magnitudes of head and trunk when walking either uphill or downhill compared to level ($p < .05$). Results of pairwise comparisons of slope are presented in Table 2.

5.3.2.1 Uphill versus level

Uphill walking significantly decreased step time ($F_{(1,15)} = 5.414$, $\eta_p^2 = .279$, $p = .036$), increased cadence ($F_{(1,15)} = 7.724$, $\eta_p^2 = .356$, $p = .015$) and single-support time ($F_{(1,15)} = 1.597$, $\eta_p^2 = .102$, $p = .227$), and decreased double-support time ($F_{(1,15)} = 1.597$, $\eta_p^2 = .102$, $p = .227$) compared to level walking. Uphill walking also significantly increased step time CoV ($F_{(1,15)} = 27.147$, $\eta_p^2 = .660$, $p < .001$) and double-support time CoV ($F_{(1,15)} = 13.342$, $\eta_p^2 = .488$, $p = .003$), but not step width CoV ($F_{(1,15)} = 2.450$, $\eta_p^2 = .149$, $p = .140$). Additionally, walking uphill led to a significant forward shift in postural alignment ($F_{(1,15)} = 6.814$, $\eta_p^2 = .327$, $p = .021$) and significantly increased the magnitude of AP trunk acceleration ($F_{(1,15)} = 15.617$, $\eta_p^2 = .527$, $p = .001$). Decreased vertical alignment (distance) also occurred in uphill walking compared to level ($F_{(1,15)} = 6.094$, $\eta_p^2 = .303$, $p = .027$).

5.3.2.2 Downhill versus level

Downhill walking significantly increased step time ($F_{(1,15)} = 6.317$, $\eta_p^2 = .311$, $p = .025$) and step width ($F_{(1,15)} = 47.803$, $\eta_p^2 = .773$, $p < .001$), decreased cadence ($F_{(1,15)} = 4.100$, $\eta_p^2 = .227$, $p = .062$) and single-support time ($F_{(1,15)} = 15.928$, ηp2 = .532, $p = .001$), and increased double-support time ($F_{(1,15)} = 30.912$, $\eta_p^2 = .688$, $p < .001$) compared to level walking. Downhill walking also significantly increased step time CoV ($F_{(1,15)} = 23.898$, $\eta_p^2 = .631$, $p < .001$), but not double-support time CoV ($F_{(1,15)} = 3.965$, $\eta_p^2 = .221$, $p = .066$) or step width CoV ($F_{(1,15)} = 3.898$,

$\eta_p^2 = .218$, $p = .068$). A significant posterior shift in postural alignment ($F_{(1,15)} = 28.154$, $\eta_p^2 = .668$, $p < .001$) and increased magnitude of AP trunk acceleration ($F_{(1,15)} = 6.295$, $\eta_p^2 = .310$, $p = .025$) were also found in downhill walking compared to level.

Table 2

Pairwise comparison p-values of slope conditions (Uphill vs Level and Downhill vs Level). Statistical significance set at $p < 0.05$ with Bonferroni correction

			Uphill	Downhill
Spatiotemporal	Step time		0.036*	0.025*
	Cadence		0.015*	0.062*
	Step width		0.992	0.000*
	SST		0.227	0.001*
	DST		0.227	0.001*
CoV	Step time		0.000*	0.000*
	Step width		0.140	0.066
	DST		0.003*	0.068
ML	Alignment		0.904	0.360
	Center of mass		0.940	0.930
	Acceleration	Head	0.127	0.282
		Trunk	0.126	0.210
AP	Alignment		0.021*	0.000*
	CoM		0.731	0.006*
	Acceleration	Head	0.474	0.135
		Trunk	0.001*	0.025*
VT	Alignment		0.027*	0.897
	CoM		0.288	0.537
	Acceleration	Head	0.000*	0.000*
		Trunk	0.000*	0.000*

* Different from Level walking

5.4 DISCUSSION

In this study, we investigated the effect of various arm swings on gait parameters and postural strategies during uphill and downhill walking. Outcome measures of uphill and downhill walking were each compared to level walking, to look at effects of both slope and arm swing. Sloped conditions (uphill, downhill) were also examined separately to characterize the effects of arm swing alone in each walking condition. While stepping and postural strategies adopted when walking uphill directly opposed those used when walking downhill, the effects by arm swing were consistent between uphill and downhill walking. Arm swing effects withstood adaptations made between sloped and level conditions but did not result in any significant interaction effects.

5.4.1 Sloped walking

Compared to level walking, uphill walking significantly decreased step time and increased cadence. Downhill walking resulted in the opposite. The opposite spatiotemporal strategies for navigating uphill and downhill slopes underlines the direct opposition in the joint kinematics of the two tasks (Kuster, Sakurai and Wood, 1995). While uphill walking strategy uses a greater hip extensor moment, as well as greater knee and ankle extensor moments with increasing slope, downhill walking involves a delayed hip extensor moment and lesser knee extensor and ankle extensor moments (Kuster, Sakurai and Wood, 1995). Sun, Walters and Svensson (1996) found increased step length and decreased cadences when walking uphill (2º to 9º), and the reverse for downhill walking. Interestingly, their findings displayed an interaction of slope and cadence occurring at approximately 3º, before which the cadence of uphill walking was greater and downhill walking lesser. As we investigated slopes between 1º and 3º (incline and decline), our findings align with these results for both uphill and downhill walking.

Kawamura (1991) also found cadence to decrease with increasing uphill slope, and increase with increasing downhill slope. Overall, results from previous studies demonstrate that the slopes used in our study are at an intersect that results in transitional effects on cadence.

Interestingly, results by Kawamura (1991) showed that double-support time increased as cadence decreased when walking uphill. Similar to the transitional point found for cadence at 3º slope, it has been suggested that a transition grade exists between 6º and 9º of inclination where the movement strategy changes to prepare the limb for an elevated heel strike with increased propulsive requirements (Prentice, 2004; Lay ,2006). Therefore, the decreased double-support time in uphill compared to level walking found in our results, suggests that our slope was slight enough that the effort used to overcome it remained to advance forward progression rather than having to pull oneself upward, as is required when walking on steeper slopes.

In downhill walking, increased step time and decreased cadence was accompanied by an increase in double-support time and decrease in single-support time. This suggests that participants adopted an overall more cautious strategy (Menz, Lord, and Fitzpatrick, 2003) when walking downhill compared to level walking. Additionally, the increased step width during downhill walking, is in line with a strategy to enhance postural stability through increasing base of support (Hak et al., 2013). This is somewhat different from Kawamura (1991) in which increase step width did not reach significance in downhill walking. As participants had to walk on a relatively narrow ramp without railing, it is likely that this affected their ability to increase step width. However, Gottschall (2011) found increased step width in the transition from level to downhill walking, which is both in-line with and likely more representative of the current findings due to the rolling-hills nature of the terrain condition used.

Step time CoV was significantly higher during both uphill and downhill walking compared to level, which confirms our hypothesis that sloped walking would lead to greater temporal variability. As the protocol emulated a rolling-hills (alternating between uphill and downhill), the slopes varied in range and happened in short durations. Using a similar rolling-hills protocol, Sinitski (2019) reported increased foot clearance in healthy adults, similar to strategies used when walking over small obstacles (Shulz, 2011). Step length variability was also previously reported to be significantly greater when transitioning to either uphill or downhill walking (Gottschall, 2011). Thus, it is probable that participants navigated the rolling hills as an obstacle and attempted various strategies to maintain anterior-posterior balance through the brief uphill and downhill sections.

Anterior-posterior alignment was shifted forward in uphill walking and backward in downhill walking. This postural strategy serves to keep the CoM more directly above the moving base of support, minimizing mechanical work (Leroux et al., 2002). In uphill walking, the forward shift of alignment also led to a significant decrease in the vertical distance between the pelvis and C7 (vertical alignment). The implementation of this postural strategy at such low surface angles likely influences the temporal changes in step time, cadence, double-support and single-support times that initially appeared inconsistent with other accounts of sloped walking (Sun et al., 2006; Kawamura, 1991). Additionally, the magnitude of antero-posterior trunk acceleration increased significantly in both sloped conditions compared to level walking while AP head acceleration did not change significantly from level walking in either sloped condition. Previous research has identified that acceleration is attenuated by the trunk primarily in the antero-posterior direction to help stabilize the head (Kavanagh, 2004), which is supported by these findings. Our results are also similar to Sinitski et al., (2019) who found increased magnitude of trunk acceleration in the rolling-hills condition compared to level walking.

As for the larger magnitude of vertical accelerations of both the head and trunk found in uphill and downhill walking compared to level, these findings are also in accordance with Kavanagh (2004). The authors suggested that the trunk does not aid in shock-absorption in the vertical direction since similar vertical forces were experienced by both the head and trunk (Kavanagh, 2004). The vertical components of propulsive force in uphill walking, and braking force in downhill walking, contribute to increases in their respective vertical force components (Gottschall & Kram, 2005; Lay, 2006).

5.4.2 Arm Swing in uphill and downhill walking

Within the opposing strategies of uphill and downhill walking, effects by arm swing created matched gradients within both tasks.

Consistent with existing literature, no significant differences were found in step time between arms held and normal arm swing (Bruijn et al., 2010; Collins et al., 2009; Nakakubo, 2014). However, we found significantly longer step times and decreased cadence with active arm swing compared to held and normal arm swing. This potentially conflicts with findings of increased walking speed under active arm swing in level walking found in Wu et al. (2016), and from verbal instructions to emphasize arm swing (Behrman, Tietlbaum and Cauragh, 1998). Our results suggest that active arm swing may contribute to forward progression, as evidenced by the decrease in double-support time and forward shift in postural alignment. The present study was conducted in a rolling-hills terrain which required participants to navigate changes in surface height. Sinitski (2019) found that healthy adults modified their base of support to minimize interaction with the rolling-hills terrain compared to level walking. Therefore, increased step time and decreased cadence are likely an artifact of increased vertical displacement and is emphasized by the upward motion of the arms in active arm swing.

Furthermore, the decreased double-support time indicates overall stable gait with active arm swing. Therefore, the forward shift in postural alignment is likely attributable to the interaction of trunk and arms rather than a postural strategy or stability measure. This view is strengthened by the more forward postural alignment during normal arm swing compared to the arms held condition, wherein minimal arm mass is advancing forward. CoM was significantly higher with active arm swing both uphill and downhill compared to arms held and normal arm swing. Arms comprise about 10% of the bodyweight (Winter, 2009) which is shifted forward and upward when peak-anterior arm swing is set to shoulder height, as in the active arm swing condition. Yet, CoM was lowest with normal arm swing, indicating a more upright posture with arms held than with normal arm swing.

Holding the arms significantly decreased the magnitude of vertical accelerations of both the head and trunk, uphill and downhill. We hypothesized that vertical accelerations of the head and the trunk would be larger with arms held compared to normal and active arm swings, due to the decreased stability inherent to challenging walking conditions (Menz, 2003). Instead, a tighter control seems to have been adopted when walking with the arms held. Yang et al. (2015) similarly found that constraining arm motion was sufficient to decrease vertical ground reaction forces.

Still, the combination of strategies used with arms held indicate poorer balance or perception of balance, as it lead to the greatest number of compensatory measures with the aim of increasing balance. Therefore, our hypothesis that active arm swing would be most helpful to walking stability was supported. Arm swing was found to withstand the effects of slope, contributing equivalent changes in all surface conditions. This signifies that active arm swing is a robust strategy for increasing gait and postural stability at minor slopes (-3 to -1º, +1 to +3º). Conversely, lack of arm swing threatened walking stability in these slope conditions and healthy individuals overcame the destabilizing effects of absent arm swing through increased control.

5.4.3 Limitations

The "held" and "active" arm swing conditions both required attention, which may have been at a level similar to what is required in dual-tasking. While the use of a harness was necessary to assure participant's safety, it might have altered participants' perceptions safety and therefore modulate their response to the hill conditions. All participants wore running shoes, but we did not measure the heel-to-toe drop or take into account the type of cushioning of the shoe. However, we are confident that the within-participant comparisons used to analyze our data did not allow any differences in footwear to affect our findings. Finally, the rolling-hills condition did not allow participants to reach a steady state within each sloped condition. Further, a range of angles were used rather than specific slope angles. While this does capture a more natural terrain condition, it cannot provide insight to the strategies used to overcome specific surface angles or the extent of the adaptations.

5.5 CONCLUSION

This results of this study corroborated previous accounts of postural strategies during uphill and downhill walking, while extending the investigation to arm swing. No interaction effects between slope and arm swing were found, yet, significant effects for both slope and arm swing were seen. Moreover, arm swing effects withstood effects by slope, indicating that arm swing is a robust balance strategy when walking on minor slopes (-3 to +3º). Active arm swing was seen to have stabilizing effects while arms held was particularly destabilizing. The information gathered from this adds to the body of knowledge evidencing the role of arm motion in the gait of healthy, young adults on terrain beyond level walking. Future study should focus on sloped walking in populations with gait impairments (e.g. Parkinson's Disease) to investigate the strategies used in destabilizing environments when "increasing control" and/or active arm swing control is not a viable option.

5.6 REFERENCES

Woollacott, M., Shumway-Cook, A. (2002). Attention and the control of posture and gait: a review of an emerging area of research. *Gait & Posture, 16*, 1-14.

Stokes, V.P., Andersson, C. and Forssberg, H. (1989). Rotational and translational movement features of the pelvis and thorax during adult human locomotion. *Journal of Biomechanics, 22*, 43-50.

Tomohisa K. *et al.* (2014) Anti-phase action between the angular acceleration of trunk and leg is reduced in the elderly. *Gait & Posture*, *40*(1), 107-12

Lin, Y.C., Gfoehler, M., Pandy, M.G. (2014). Quantitative evaluation of the major determinants of human gait. *Journal of Biomechanics, 47*(6), 1324-31.

Ortega, J.D., Fehlman, L.A., Farley, C.T., (2008). Effects of aging and arm swing on the metabolic cost of stability in human walking. Journal of biomechanics. 5;41(16):3303-8.

Li, Y., Wang, W., Crompton, R.H. and Gunther, M.M. (2001). Free vertical moments and transverse forces in human walking and their role in relation to arm-swing. *Journal of Experimental Biology, 204*(1), 47-58.

Yang, H.S., Atkins, L.T., Jensen, D.B. and James, C.R. (2015). Effects of constrained arm swing on vertical center of mass displacement during walking. *Gait & Posture, 42*, 430-34.

Umberger, B.R. (2008). Effects of suppressing arm swing on kinematics, kinetics, and energetics of human walking. *Journal of Biomechanics, 41*(11), 2575-80.

Collins, SH., Adamczyk, PG., Kuo, AD., (2009). Dynamic arm swinging in human walking. Proceedings of the Royal Society B: Biological Sciences. 29;276(1673):3679-88.

Hu F., Gu D.Y., Chen J.L., Wu Y., An B.C. and Dai K.R. (2012). Contribution of arm swing to dynamic stability based on the nonlinear time series analysis method. *2012 Annual International Conference of the IEEE Engineering and Medicine and Biology Society*, 4831-4

Nakakubo, S., Doi, T., Sawa, R., Misu, S., Tsutsumimoto, K. and Ono, R. (2014). Does arm swing emphasized deliberately increase trunk stability during walking in the elderly adults? *Gait & Posture, 40*(4), 516-20.

Punt, M., Bruijn, S.M., Wittink, H., and van Dieën, J.H. (2015). Effect of arm strategy on local dynamic stability of human gait. *Gait & Posture, 41*(2): 504-509.

Wu, Y. *et al.* (2016). Effect of active arm swing to local dynamic stability during walking. *Human Movement Science, 45*(Complete), 102-09.

Bruijn, SM., Meijer, OG., Beek, PJ., Van Dieën, JH.,2010. The effects of arm swing on human gait stability. Journal of experimental biology. 1;213(23):3945-52.

Bruijn, S.M., Bregman, D.J., Meijer, O.G., Beek, P.J. and van Dieën, J.H. (2012). Maximum Lyapunov exponents as predictors of global gait stability: A modelling approach. *Mechanical Engineering & Physics, 34*(4), 428-36.

Bruijn, S.M., Meijer, O.G., Beek, P.J. and van Dieën, J.H. (2013). Assessing the stability of human locomotion: A review of current measures. *Journal of The Royal Society, Interface, 10*(83), 20120999.

Pijnappels, M., Kingma, I., Wezenberg, D., Reurink, G., van Dieën, J.H., (2010). Armed against falls: the contribution of arm movements to balance recovery after tripping. Experimental Brain Research. 201:689–99.

Lulic, T.J., Susic, A. and Kodvanj, J. (2008) Effects of arm swing on mechanical parameters of human gait. *Collegium Antropologicum, 32*(3), 869-73.

Minetti, A.E., Moia, C., Roi, G.S., Susta, D. and Ferretti, G. (2002). Energy cost of walking and running at extreme uphill and downhill slopes. *Journal of Applied Physiology, 93*(3), 1039-46.

Gottschall, J.S. and Kram, R. (2005). Ground reaction forces during downhill and uphill running. *Journal of Biomechanics, 38*, 445–52.

Tucker, M., Kavanagh, J., Barrett, R. and Morrison, C. (2008). “Age-related differences in postural reaction time and coordination during voluntary sway movements.” *Human Movement Science, 27*(50), 728-37.

Cromwell, R.L. (2004). Movement strategies for head stabilization during incline walking. *Gait & Posture, 17*, 246-53.

Kawamura K., Tokuhiro A. and Takechi H. (1991). Gait analysis of slope walking: a study of step length, stride width, time factors and deviation in the center of pressure. *Acta Medica Okayama, 45*(3), 179-84.

Redfern, M.S. and DiPasquale, J. (1997). Biomechanics of descending ramps. *Gait & Posture, 6*, 119-25.

Vieira, M.F. *et al.* (2017). Gait stability, variability and complexity on inclined surfaces. *Journal of Biomechanics, 54*, 73-79.

Kimel-Naor, S., Gottlieb, A. and Plotnik, M. (2017). The effect of uphill and downhill walking on gait parameters: A self-paced treadmill study. *Journal of Biomechanics, 60*, 142-149.

Kuster, M., Sakurai, S., Wood, G.A. (1995). Kinematic and kinetic comparison of downhill and level walking. *Clinical Bimechanics, 10*(2), 79-84.

Sun, J., Walters, M., Svensson, N. and Lloyd, D. (1996). The influence of surface slope on human gait characteristics: a study of urban pedestrians walking on an inclined surface. *Ergonomics, 39*(4), 677-692.

Prentice, S.D., Hasler, E.N., Groves, J.J. and Frank, J.S., (2004). Locomotor adaptations for changes in the slope of the walking surface. *Gait & Posture, 20*, 255–65.

Lay, A.N., Hass, C.J., Gregor, R.J. (2006). The effects of sloped surfaces on locomotion: A kinematic and kinetic analysis. *Journal of Biomechanics 39*, 1621-28.

Hak, L., Houdijk, H., Beek, PJ., van Dieen, JH., 2013a. Steps to take to enhance gait stability: the effect of stride frequency, stride length, and walking speed on local dynamic stability and margins of stability. PLoS One 8: e82842. https://doi.org/10.1371/journal.pone.0082842

Gottschall, J.S., Okorov, D.Y., Okita, N. and Stern, K.A. (2011). Walking strategies during the transition between level and hill surfaces. *Journal of Applied Biomechanics, 27*, 355-61.

Sinitski, E.H., Lemaire, E.D., Baddour, N., Besemann, M., Dudek, N., & Hebert, J.S. (2019). Maintaining stable transtibial amputee gait on level and simulated uneven conditions in a virtual environment. *Disability and Rehabilitation: Assistive Technology*; doi: 10.1080/17483107.2019.1629186

Leroux, A., Fung, J. and Barbeau, H. (2002). Postural adaptation to walking on inclined surfaces: I. Normal strategies. *Gait & Posture, 15*, 64-74.

Kavanagh, J. J., Barrett, R. S., & Morrison, S. (2004). Upper body accelerations during walking in healthy young and elderly men. Gait & Posture, 20, 291–298.

Behrman, A.L., Teitelbaum, P. and Cauraugh, J.H. (1998). Verbal instruction sets to normalise the temporal and spatial gait variables in Parkinson's Disease. *Journal of Neurology, 65*(4), 580.

Winter, D.A. (2009). *Biomechanics and Motor Control of Human Movement.* 4th ed. Hoboken, New Jersey: John Wiley & Sons, Inc.

Wilken, J. M., Rodriquez, K. M., Brawner, M. and Darter, B.J. (2012). Reliability and minimal detectible change in values for gait kinematics and kinetics in healthy adults. *Gait & Posture, 35*(2), 301-07.

CHAPTER 6: CONCLUSION

The information gathered from this study adds to the body of knowledge evidencing the role of arm motion in the gait of healthy, young adults on terrain beyond level walking. Additionally, these results demonstrated that significant postural changes are made even at slight slopes (1 to 3º incline/decline). The healthy population tested adequately adapted to the slopes but the postural strategies used caused larger spatiotemporal changes than expected which warrants further investigation in populations with gait impairments. Arm swing caused equivalent changes in all terrain conditions. Active arm swing contributed to steady gait and an increased forward drive. Conversely, holding the arms while walking was found to be destabilizing. The healthy, young adults in the present study coped with this condition through increasing postural control. As this study used healthy, young participants, the current findings can be used as a baseline comparison in future investigations of other populations.

6.1 Impact

The results of this study reinforce the use of sloped walking in therapeutic contexts due to its inevitable presence in daily life and it's challenging nature. Regarding arm swing, the current findings confirm the useful of active arm swing in meeting certain rehabilitative aims, such as progressing the right and left sides of the body evenly and leading the CoM forward. We recommend walking without arms constrained (i.e. leaving arms free to swing) for individuals at higher risk of falling (e.g. elderly, stroke patients, individuals with PD). Future study should focus on investigating strategies used in populations of people unable to exert increased control during challenging conditions.

6.2 Limitations

The “held” and “active” arm swing conditions both required attention, which may have been at a level similar to what is required in dual-tasking. While the use of a harness was necessary to assure participant’s safety, it might have altered participants’ perceptions safety and therefore modulate their response to the hill conditions. All participants wore running shoes, but we did not measure the heel-to-toe drop or take into account the type of cushioning of the shoe. However, we are confident that the within-participant comparisons used to analyze our data did not allow any differences in footwear to affect our findings. Finally, the rolling-hills condition did not allow participants to reach a steady state within each sloped condition. Further, a range of angles were used rather than specific slope angles. While this does capture a more natural terrain condition, it cannot provide insight to the strategies used to overcome specific surface angles or the extent of the adaptations.

REFERENCES

1) Ballestros, M.L.F., Buchtal, F. and Rosenfalck, P. (1965). The pattern of muscular activity during the arm swing of natural walking. *Acta Physiologic Scandinavica, 63*(3), 296-310.

2) Behrman, A.L., Teitelbaum, P. and Cauraugh, J.H. (1998). Verbal instruction sets to normalise the temporal and spatial gait variables in Parkinson's Disease. *Journal of Neurology, 65*(4), 580.

3) Bruijn, S.M., Meijer, O.G., Beek, P.J. and van Dieën, J.H. (2010). The effects of arm swing on human gait stability. *The Journal of Experimental Biology, 213*, 3945-52.

4) Bruijn, S.M., Bregman, D.J., Meijer, O.G., Beek, P.J. and van Dieën, J.H. (2012). Maximum Lyapunov exponents as predictors of global gait stability: A modelling approach. *Mechanical Engineering & Physics, 34*(4), 428-36.

5) Bruijn, S.M., Meijer, O.G., Beek, P.J. and van Dieën, J.H. (2013). Assessing the stability of human locomotion: A review of current measures. *Journal of The Royal Society, Interface, 10*(83), 20120999.

6) Cappozzo, A. (1991). The mechanics of human walking. *Adaptability of Human Gait, 78*, 167-86.

7) Collins, S.H., Adamczyk, P.G. and Kuo A.D. (2009). Dynamic arm swinging in human walking. *Proceedings of the Royal Society of London B: Biological Sciences, 276*(1673), 3679-88.

8) Cromwell, R.L. (2004). Movement strategies for head stabilization during incline walking. *Gait & Posture, 17*, 246-53.

9) Dingwell J.B., Cusumano J.P., Cavanagh P.R. and Sternad D. (2001). Local dynamic stability versus kinematic variability of continuous overground and treadmill walking. *Journal of Biomechanics Engineering, 123*(1), 37-32.

10) Elftman, H. (1939). “The function of the arms in walking.” *Human Biology; Baltimore, 11*(4), 529-35.

11) Gottschall, J.S. and Kram, R. (2005). Ground reaction forces during downhill and uphill running. *Journal of Biomechanics, 38*, 445–52.

12) Gottschall, J.S., Okorov, D.Y., Okita, N. and Stern, K.A. (2011). Walking strategies during the transition between level and hill surfaces. *Journal of Applied Biomechanics, 27*, 355-61.

13) Goudriaan, M., Jonkers, I., van Dieën, J.H. and Bruijn, S.M. (2014). Arm swing in human walking: what is their drive? *Gait & Posture, 40*(2), 321-26.

14) Hallemans, A., Ortibus, E., Meire, F. and Aerts, P. (2010). Low vision affects dynamic stability of gait. *Gait & Posture, 32*(4), 547-51.

15) Hu F., Gu D.Y., Chen J.L., Wu Y., An B.C. and Dai K.R. (2012). Contribution of arm swing to dynamic stability based on the nonlinear time series analysis method. *2012 Annual International Conference of the IEEE Engineering and Medicine and Biology Society*, 4831-4.

16) Kawamura K., Tokuhiro A. and Takechi H. (1991). Gait analysis of slope walking: a study of step length, stride width, time factors and deviation in the center of pressure. *Acta Medica Okayama, 45*(3), 179-84.

17) Kimel-Naor, S., Gottlieb, A. and Plotnik, M. (2017). The effect of uphill and downhill walking on gait parameters: A self-paced treadmill study. *Journal of Biomechanics, 60*, 142-149.

18) Kuhtz-Bushbeck, J.P. and Jing, B. (2012). Activity of upper limb muscles during human walking. *Journal of Electromyography and Kinesiology, 22*(2), 199-206.

19) Logan, D., Kiemel, T., Dominici, N., Cappellini, G., Ivanenko, Y., Lacquaniti, F. and Jeka, J. (2010). The many roles of vision during walking. *Experimental Brain Research, 206*(3), 337-50.

20) Latt, M.D., Menz, H.B., Fung, V.S. and Lord, S.R. (2009). Acceleration patterns of the head and pelvis during gait in older people with Parkinson's disease: a comparison of fallers and non-fallers. *Journal of Gerantology, 64*(6), 700-06.

21) Lemaire, E. *et al.* (2012) Rehabilitation Virtual Reality: Improving Clinical outcomes and advancing rehabilitation research. (2012) Rehabilitation research news. *Health Services, The Ottawa Hospital, The Ottawa Hospital Foundation and the community*, 22-23. http://www.ohri.ca/rehabilitation/Resources/2012%20Research%20News%20(English).22-23.pdf

22) Leroux A., Fung J. and Barbeau H. (1999). Adaptation of the walking pattern to uphill walking in normal and spinal-cord injured subjects. *Experimental Brain Research, 126*, 359-68.

23) Leroux, A., Fung, J. and Barbeau, H. (2002). Postural adaptation to walking on inclined surfaces: I. Normal strategies. *Gait & Posture, 15*, 64-74.

24) Li, Y., Wang, W., Crompton, R.H. and Gunther, M.M. (2001). Free vertical moments and transverse forces in human walking and their role in relation to arm-swing. *Journal of Experimental Biology, 204*(1), 47-58.

25) Lulic, T.J., Susic, A. and Kodvanj, J. (2008) Effects of arm swing on mechanical parameters of human gait. *Collegium Antropologicum, 32*(3), 869-73.

26) McIntosh, A.A., Beaty, K.T., Dwan, L.N. and Vickers, D.R. (2006). Gait dynamics on an inclined walkway. *Journal of Biomechanics, 39*, 2491-502.

27) Menz, H., Lord, S. and Fitzpatrick, R. (2003). Acceleration patterns of the head and pelvis when walking on level and irregular surfaces. *Gait & Posture, 18*(1), 35-46.

28) Minetti, A.E., Moia, C., Roi, G.S., Susta, D. and Ferretti, G. (2002). Energy cost of walking and running at extreme uphill and downhill slopes. *Journal of Applied Physiology, 93*(3), 1039-46.

29) Nakakubo, S., Doi, T., Sawa, R., Misu, S., Tsutsumimoto, K. and Ono, R. (2014). Does arm swing emphasized deliberately increase trunk stability during walking in the elderly adults? *Gait & Posture, 40*(4), 516-20.

30) Ortega, J.D., Fehlman, L.A. and Farley, C.T. (2009). Effects of aging and arm swing on the

metabolic cost of stability in human walking. *Journal of Biomechanics, 41*(16), 3303-08.

31) Patla, A.E. (1991). Visual control of human locomotion. *Adaptability of Human Gait, 78,* 55-95.

32) Pijnappels, M., Kingma, I., Wezenberg, D., Reurink, G. and van Dieën, J. H. (2010). Armed against falls: the contribution of arm movements to balance recovery after tripping. *Experimental Brain Researcher, 201*, 689-699.

33) Prentice, S.D., Hasler, E.N., Groves, J.J. and Frank, J.S., (2004). Locomotor adaptations for changes in the slope of the walking surface. *Gait & Posture, 20*, 255–65.

34) Punt, M., Bruijn, S.M., Wittink, H., and van Dieën, J.H. (2015). Effect of arm strategy on local dynamic stability of human gait. *Gait & Posture, 41*(2): 504-509.

35) Redfern, M.S. and DiPasquale, J. (1997). Biomechanics of descending ramps. *Gait & Posture, 6*, 119-25.

36) Sloot, L.H., van Der Krogt, M.M. and Harlaar, J. (2013). Self-paced versus fixed speed treadmill walking. *Gait & Posture, 39*, 474-84.

37) Stokes, V.P., Andersson, C. and Forssberg, H. (1989). Rotational and translational movement features of the pelvis and thorax during adult human locomotion. *Journal of Biomechanics, 22*, 43-50.

38) Sun, J., Walters, M., Svensson, N. and Lloyd, D. (1996). The influence of surface slope on human gait characteristics: a study of urban pedestrians walking on an inclined surface. *Ergonomics, 39*(4), 677-692.

39) Sutherland, D.H., Olshen, R., Cooper, L. and Woo, S.L. (1980). The development of mature gait. *The Journal of Bone and Joint Surgery, 62*(3), 336-53.

40) Tomohisa K. *et al.* (2014) Anti-phase action between the angular acceleration of trunk and leg is reduced in the elderly. *Gait & Posture*, *40*(1), 107-12.

41) Tucker, M., Kavanagh, J., Barrett, R. and Morrison, C. (2008). Age-related differences in postural reaction time and coordination during voluntary sway movements. *Human Movement Science, 27*(50), 728-37.

42) Umberger, B.R. (2008). Effects of suppressing arm swing on kinematics, kinetics, and energetics of human walking. *Journal of Biomechanics, 41*(11), 2575-80.

43) Vieira, M.F. *et al.* (2017). Gait stability, variability and complexity on inclined surfaces. *Journal of Biomechanics, 54*, 73-79.

44) Wall, J.C., Nottrodt, J.W. and Charters, J. (1981). The effects of uphill and downhill walking on pelvic oscillations in the transverse plane. *Ergonomics, 24*(10), 807-16.

45) Wilken, J. M., Rodriquez, K. M., Brawner, M. and Darter, B.J. (2012). Reliability and minimal detectible change in values for gait kinematics and kinetics in healthy adults. *Gait & Posture, 35*(2), 301-07.

46) Winter, D.A. (1995). Human balance and posture control during walking. *Gait & Posture, 3*(4), 193-214.

47) Winter, D.A. (2009). *Biomechanics and Motor Control of Human Movement.* 4th ed. Hoboken, New Jersey: John Wiley & Sons, Inc.

48) Wu, Y. *et al.* (2016). Effect of active arm swing to local dynamic stability during walking. *Human Movement Science, 45*, 102-09.

49) Woollacott, M., Shumway-Cook, A. (2002). Attention and the control of posture and gait: a review of an emerging area of research. *Gait & Posture, 16*, 1-14.

50) Yang, H.S., Atkins, L.T., Jensen, D.B. and James, C.R. (2015). Effects of constrained arm swing on vertical center of mass displacement during walking. *Gait & Posture, 42*, 430-34.