THE EFFECT OF ARM SWING AND ROCKY SURFACE ON DYNAMIC STABILITY IN HEALTHY YOUNG ADULTS

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CHAPTER 0: GENERAL ABSTRACT

0.1 Abstract

There are millions of fall-related injuries worldwide requiring medical attention on a yearly basis. These falls place a financial burden on the healthcare system. These falls can occur in the event of disruption in the postural control system and/or a loss of balance while walking. Previously, most gait studies have focused on the assessment of the lower extremities while neglecting the contribution of arm swing as it was believed to be a passive motion.

However, it has been shown that there is an active component to arm swing. Moreover, these arm movements have been shown to affect the motion of the center of mass when walking. Therefore, arm swing could mitigate the destabilizing effects of perturbations caused by challenging surfaces. Additionally, no studies have examined the effect of arm swing when walking on a rocky surface. This type of surface causes perturbations in the anteroposterior and mediolateral directions simultaneously, leading to uneven center of mass displacement and spatiotemporal modifications.

Hence, the present study assessed the effect of normal arm swing, held arm swing and active arm swing on postural control and dynamic stability when walking on regular and rocky surface. We hypothesized that active arm swing will have a negative impact on postural control and gait dynamics on a regular surface, while rocky surface walking will decrease stability and increase spatiotemporal variability. Additionally, we expect active arm swing to attenuate the negative effects of the rocky surface.

Fifteen healthy young adults from the University of Ottawa community (mean age 23.4 ± 2.8 years) were recruited to participate in this study. They were asked to walk using three different arm conditions (normal, held and active arm swing) on the dual-belt CAREN-Extended System (Motek Medical, Amsterdam, NL) on simulated regular and rocky surface. This last is generated using the "Rumble" module (maximum range of ±2 cm at 0.6 Hz vertically, ±1° at 1 Hz pitch, and ±1° at 1.2 Hz roll). Mean, standard deviation and maximal values of trunk linear and angular velocity were calculated in all three planes. Moreover, step length, time and width mean and coefficient of variation as well as margin of stability mean and standard deviation were calculated. A mixed linear model was performed to compare the effects of the arm swing motions and surface types. The arm and surface conditions were set as fixed effects, while the walking speed was set as a covariate.

Active arm swing increased trunk linear and angular velocity variability and peak values compared to normal and held arm conditions. Active arm swing also increased participants' step length and step time, as well as the variability of margin of stability. Similarly, rocky surface walking increased trunk kinematics variability and peak values compared to regular surface walking. Furthermore, rocky surface increased the average step width while reducing the average step time.

The spatiotemporal adaptations show the use of "cautious" gait to mitigate the destabilizing effects of both the active arm swing and rocky surface walking and, ultimately, maintain stability.

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ABBREVIATIONS AND OPERATIONAL DEFINITIONS

Active arm swing	Arms actively swinging to shoulder height
AP	Anteroposterior
AV	Angular velocity
BOS	Base of support
C7 marker	Marker placed at the 7th cervical vertebrae
COM	Center of mass
COV	Coefficient of variance; (standard deviation/mean) x 100
Dynamic stability	Maintenance of the extrapolated center of mass within the moving base of support
HAT	Head arms and trunk model
Held arm swing	Arms held alongside the thighs and secured in the harness
LV	Linear velocity
Max	Maximal values
ML	Mediolateral
MOS	Margin of stability
Normal arm swing	Participants' natural arm motion
Postural control	Maintenance of relative segmental positioning to ensure a reliable reference frame
SD	Standard deviation
xCOM	Extrapolated center of mass; position of center of mass considering its velocity
V	Vertical

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CHAPTER 1: GENERAL INTRODUCTION

1.1 Introduction

Postural control allows an individual to perform all of their activities efficiently and safely. This is achieved with the help of the postural control system whose main roles are to (1) build up a resistance against gravity by regulating tonic muscle activity (Ivanenko & Gurfinkel, 2018), and (2) fix the orientation of segments in space to maintain a reliable reference frame, thus allowing to resist perturbations from external and internal origins (Ivanenko & Gurfinkel, 2018; Massion, 1994). A disruption in this control can cause falls that have medical consequences as well as financial strain on healthcare systems (Ministry of Labour Inspection Blitzes, n.d.; Yeoh et al., 2013). Fall-induced injuries are a leading cause of injuries in the workplace, with a majority resulting from challenging same-level surfaces that increase the difficulty of foot placement (Bell et al., 2013; Drebit et al., 2010; Merryweather et al., 2011; Moyer et al., 2006; Parijat & Lockhart, 2008; Yeoh et al., 2013). Therefore, identifying the association between postural control, dynamic stability and aspects of walking in challenging environments could prevent these injuries in the workplace as well as reduce fall occurrence in known fallers.

Proper trunk postural control is a key aspect of postural balance and in avoiding falls. Normally, the neuromuscular system determines foot placement and timing according to trunk's future position (Brenière & Do, 1991; Rankin et al., 2014; Siragy et al., 2019; Winter, 1987; Winter, 1995) and, ultimately, a person's dynamic balance (the ability to maintain the center of mass (COM) within a moving base of support (BOS) (Hof et al., 2005; Winter, 1995)). When examining gait, studies have traditionally employed the inverted pendulum paradigm to model human walking (Winter, 1995). In this model the head, arms and trunk (HAT) are considered to be a rigid-body which accounts for two-thirds of the body's mass (MacKinnon & Winter, 1993;

Winter, 1987; Winter, 1995). Therefore, the contribution of the arms is neglected. In addition, most models consider arm swing passive, i.e. the result of trunk movement, gravity and inertia (Winter, 1995). Thus, despite the large number of studies investigating falls and fall related injuries, the prevalence of occupational falls has remained stable (Burns & Kakara, 2018; Chang et al., 2016; MacKinnon & Winter, 1993; Marigold, 2002; Marigold & Patla, 2002; Winter, 1995).

Unlike the assumptions made in the development of the HAT model, studies have shown that arm swing is generated from both passive control and active neuromuscular input arising from shoulder joint muscular activity and torques (Cimolin et al., 2012; Collins et al., 2009; Kuhtz-Buschbeck & Jing, 2012; Meyns et al., 2013; Sylos-Labini et al., 2014). While only few studies have looked explicitly into the role of arm motion in normal and challenging environments on gait stability, it has been shown that arm swings help to initiate gait motion (Collins et al., 2009) and reduce the overall energetic costs associated with walking (Collins et al., 2009) as it generates opposing torque to the legs and trunk about the vertical axis (Meyns et al., 2013). Furthermore, arm motion has been shown to improve postural control and dynamic balance during steady-state walking (Meyns et al., 2013; Wu et al., 2016) as well as when recovering from perturbations (Bruijn et al., 2010). Yet, current literature is conflicting as some evidence suggests that walking without arm swing increases trunk inertia and therefore enhances resistance to perturbation (Bruijn et al., 2010; Pijnappels et al., 2010).

The metrics used to quantify dynamic balance could be among the factors explaining these apparent conflicting results as the relationship between arm swing and gait balance has only been examined using one metric of stability at a time (Wu et al., 2016). This has been shown to limit the evidence for the overall contribution of arm swing towards gait postural control as each measure reflects unique neuromuscular control components (Siragy & Nantel, 2018). Recently, our group reported an increase in trunk velocity variability and whole-body angular momentum, as well as a reduced anteroposterior (AP) and mediolateral (ML) harmonic ratio (a measure for evaluating smoothness and consistency within the walking pattern (Bellanca et al., 2013; Latt et al., 2008)) when walking was combined with actively swinging arms to shoulder height among healthy young adults (Siragy et al., 2019). These are all indicators of a more variable trunk motion and, ultimately, a more variable COM displacement. As a result, there was an increase in spatial-temporal variability as a compensation mechanism to maintain an adequate margin of stability (MOS) (Siragy et al., 2019). In other words, spatiotemporal parameters adaptations were required to keep the trunk within a moving BOS. This also supports the relationship between the stepping parameters and this stability marker as the wider steps increase the distance between the edge of the BOS and the position of the COM, leading to favourable MOS values.

In addition, there is a limited number of studies focusing on gait in challenging terrains and their effects on gait stability. Challenging terrains such as asymmetric walking increases energy consumption during gait (Ellis et al., 2013) as it causes an imbalance between the mechanical energy at push-off and at heel contact, therefore increasing the metabolic work (Ellis et al., 2013). Challenging terrains also stray away from the preferred walking parameters (McAndrew Young et al., 2012). During asymmetric walking, an unequal distribution of weight bearing occurs on the lower extremity which impairs COM trajectory and challenges foot placement (Lewek et al., 2014; Patterson et al., 2008). This causes non-uniform accelerations at the trunk level which impairs its rhythmic movement; this results in an increase of fall risk (Winter, 1987). A recent publication from our group concluded that asymmetric walking increased ML MOS and reduced AP and ML harmonic ratios (Siragy et al., 2019). Though the former is an indicator of improved stability, the latter shows a more variable gait, which is linked to poorer postural control and stability. In fact, the increase in MOS in the ML direction was a direct result from adopting a wider step to mitigate the perturbing effects of asymmetric walking. Similarly, walking on a rocky surface, which generates minor perturbations in both the AP and ML directions has been shown to increase variability of spatial-temporal parameters and MOS in the ML direction in both young adults and individuals with transtibial amputation (Gates et al., 2012; Gates et al., 2013). As arm motion has been shown to control the translation of COM (Wu et al., 2016), it is possible that arm swing may play a pivotal role in maintaining trunk postural control and dynamic balance on destabilizing terrains.

The aim of this study is to compare the effect of different arm motions on postural control and dynamic stability in symmetric and asymmetric walking conditions. This will be done by examining key variables including trunk velocities, angular momentum, spatial-temporal parameters and margin of stability.

CHAPTER 2: LITERATURE REVIEW

2.1 Human gait

Human locomotion consists of assuring safe and efficient forward progression (Saunders et al., 1953). Over the years, different measures have been proposed to assess dynamic stability. Hoff et al (2005) proposed the MOS which evaluates the distance between the extrapolated, i.e. velocity-adjusted, position of the COM in relationship with the BOS. In this measure, gait dynamic stability is considered achieved as long as the extrapolated position of the COM lies within the limits of the BOS (Hof et al., 2005). Other variables such as linear and angular segmental kinematic measures have also been proven valuable to assess gait dynamic stability because they comment on the variability of gait and could even be used as predictors for fall risks (Schooten et al., 2011). Finally, metrics such as spatial and temporal variability have been used to infer dynamic stability during gait, with an increase in variability in trunk kinematics being linked to reduced dynamic stability in healthy young adults (Schooten et al., 2011).

Among spatiotemporal parameters, stride length and width are of particular interest when assessing dynamic balance as they directly affect the BOS (McAndrew Young & Dingwell, 2012; Sivakumaran et al., 2018). Studies reported that walking at preferred stride length allowed to minimize both spatial and temporal variabilities (Sekiya et al., 1997; Latt et al., 2008). This decrease in variability was later associated with improved gait stability (Bruijn & van Dieën, 2018). Reducing stride length was also reported to improve gait stability (Sekiya et al., 1997; Latt et al., 2008; Espy et al., 2010; McAndrew Young & Dingwell, 2011; Hak et al., 2013; Hak et al., 2015). However, these modifications in strides lengths yielded contradictory results depending on the metrics used to assess dynamic stability. Using the harmonic ratios, Latt et al., (2008) showed an increase in mediolateral dynamic stability when decreasing stride length in healthy young adults. Similarly, backwards MOS in healthy participants was improved by reducing stride length (Hak et al., 2013). Even when recovering from large external perturbation such as a slip, decreasing stride length yielded improved stability as participants had better control of their COM within their BOS (Espy et al., 2010). In contrast, McAndrew Young & Dingwell (2012) revealed an increase in stability with increasing stride length when analyzing the velocity of the C7 marker in healthy participants. Conversely, the authors suggested that the increase in stability was most likely due to the increased walking speed rather than solely the effect of increased stride length (Hak et al., 2013).

Similar to stride length, the impact of stride width on dynamic stability led to contradictory outcomes. Taking larger steps broadens the BOS and allows larger displacement of the COM before moving over the limits of the BOS (O'Connor et al., 2012). However, it was also shown that walking with wider steps led to more variability in foot placement (Perry & Srinivasan, 2017) and less control over the trunk motion among healthy young adults (McAndrew Young & Dingwell, 2012). This variability has been associated with a decrease in postural control throughout the walking cycle (Bruijn & van Dieën, 2018). While the increased step width variability may not have a significant impact on fall risks in healthy young adults, this could be of significant importance for those with an impaired gait, such as post-stroke patients and older adults with mobility issues.

2.2 Arm swing

The movement of the arms during walking has been shown to reduce the overall energetic expenditure of walking (Meyns et al., 2013). This is possibly due to a reduction in angular momentum about the vertical axis and the decreased vertical ground reaction through the anti-phase movement of the arms (Meyns et al., 2013). Ortega et al. (2008) calculated an overall reduction in energy consumption by approximately 7% when using arm swing compared to walking without arm swing. Work by Umberger (2008) showed a similar decrease in energy consumption when comparing walking with and without arm swing.

Other advantages have been reported from swinging the arms during walking. Arm swing allows to initiate the forward motion of walking and provides increased torque when walking at higher speeds (Collins et al., 2009). Moreover, arm swing allows to correct errors (such as neuromuscular misfires) as they arise. This motion also contributes towards gait stability as arm motion counteracts the rotation of the torso around the vertical axis (Collins et al., 2009). Finally, the contralateral arm swing movements counteract the angular momentum about the vertical axis generated by the legs when walking (Ballesteros et al., 1965; Collins et al., 2009; Elftman, 1939; Ortega et al., 2008). This reduction in momentum helps to limit the COM's lateral displacement and maintain its intended course during normal, steady-state walking (Ortega et al., 2008). In terms of postural control and stability, these findings can hold positive consequences: the limited mediolateral COM displacement could reduce trunk kinematics variability in this direction (leading to improved postural control) while simultaneously increasing an individual's mediolateral MOS as the distance between the extrapolated position of the COM (xCOM) and the edges of the BOS is increased (leading to improved stability).

Advantages associated to normal arm swing have even been shown in non-steady-state conditions. In fact, when assessing the impact of arm motion before and following a large perturbation, results by Bruijn et al., (2010) and Pijnappels et al. (2010), showed that arm swing almost exclusively benefit dynamic stability in recovery phase following the perturbation. Consequently, normal arm swing plays multiple roles aimed at facilitating bipedal locomotion.

Interestingly, arm swing can be modified to give similar stability-related advantages through different mechanisms.

On study by Collins et al. (2009) examined the impact of both walking with and without arm swing. The authors reported an increase in metabolic rate, vertical angular momentum and whole-body angular momentum when arms were either bound or held to the sides compared to normal arm swing. These outcomes showed both metabolic and mechanical negative impacts of restricting arm swing. Conversely, restricting arm swing has also been found to contribute to improving dynamic balance following a large perturbation. This is due to the higher trunk inertia caused by the additional weight of the arms. This could limit the displacement of the COM within the BOS, (Bruijn et al., 2010; Pijnappels et al., 2010). Another study by Punt et al. (2015) compared the use of four different arm swing strategies and their effects on dynamic stability of the trunk. The arm strategies were normal arm swing, in-phase with each other, ipsilaterally and active arm swing (participants were instructed to swing excessively) during steady-state walking. They reported that using active arm swing increased local dynamic stability (the body's ability to maintain a smooth gait pattern despite the presence of infinitesimally small and naturally occurring perturbation (Josinski et al., 2019)) in the ML direction specifically. They also suggested that those with an increased risk of falls should employ this strategy, which is in line with results by Hu et al. (2012) who conducted a similar study with older adults. Similarly, Wu et al. (2016) investigated the effects of active arm swing (which they defined as "voluntary arm swings actively driven by shoulder or arm muscles") on local dynamic of the trunk and stride-to-stride variability during steady-state walking as well. Their results showed that active arm swing decreased gait variability, leading to improved dynamic stability at the level of the trunk (Wu et al., 2016). More specifically, there was an increase in trunk stability in the ML direction when using active arm swing. These studies provided insight into the possibility of using human movements to improve gait stability and postural control.

However, recent results by our group comparing different arm swing strategies showed that active arm swing led to increased trunk linear and angular velocity variability compared to normal and held arm swing, reflecting poorer postural control (Siragy et al., 2019). Moreover, these results showed a decrease in gait stability as witnessed by decreased MOS and harmonic ratios and increased the coefficient of variance of stride length width and time when using active arm swing (Siragy et al., 2019). These results conflict with those of Wu et al. (2016), showing a gap in literature that needs to be addressed.

2.3 Perturbed walking

When walking in a changing environment or uneven surface, both spatial and temporal aspects of the gait pattern have to be modified to maintain postural stability and avoid falling (Hawkins et al., 2018). Often, these adaptations lead to walking patterns resembling those resulting from musculoskeletal injuries and gait-impairing conditions or pathological conditions such stroke or Parkinson’s disease. In most cases, these adaptations lead to asymmetrical gait patterns, which induce a minor mechanical perturbation (perturbation that induces an involuntary displacement, followed by corrective movements to maintain stability (Terry et al., 2012)) in the AP direction.

Asymmetric walking has been reported to increase energy consumption (Ellis et al., 2013) as shown by expired gas analysis in symmetric and asymmetric walking conditions at different speeds. More specifically, asymmetric walking showed an increase in both energy consumption and mechanical power production as a direct result of the loss of symmetry throughout the pattern and the modification in stride time (Ellis et al., 2013). It was recently shown that a mild asymmetric gait pattern is associated with a decrease in dynamic stability as evidenced by increased variability of spatial-temporal parameters (Siragy et al., 2019). Additionally, it was determined that both symmetry and stride time are independently optimized during normal walking to minimize energy consumption (Ellis et al., 2013). These findings support results from O'Connor et al. (2012) and Rock et al. (2018), which demonstrated that deviations from the preferred walking parameters and increased gait variability led to increased energy consumption.

Another factor that could induce gait asymmetry is the type of terrain, such as walking on a rocky road, cause perturbations both in the ML and AP directions, a more severe perturbation than asymmetric walking. Protocols simulating challenging terrains interfere with normal gait by challenging the body COM's trajectory. In healthy young adults, this walking condition has been shown to increase step length, step width and step time variability when compared to a normal surface (Gates et al., 2012). According to Gates et al., (2013), these spatial adaptations increased the base of support and therefore reduce postural instability. This strategy also allowed participants to maintain levels of MOS in the ML direction similar to that of walking on a regular surface (Gates et al., 2013). McAndrew Young et al. (2012) and Onushko et al. (2019) drew similar conclusions, showing that continuous pseudo-random oscillation (analogous to walking on a rocky surface) in the AP and ML directions led to increased foot placement variability, and sometimes negative MOS (an indicator of loss of stability where the extrapolated COM position is outside of the BOS) in the ML direction.

To our knowledge no studies investigated the impact of arm motion on terrain-inducing perturbations in both the ML and AP direction. As arm swing has been shown to affect the displacement of the COM, it is possible that this arm motion can mitigate the mechanical destabilization caused by this challenging terrain. Furthermore, this type of terrain can be more representative of daily walking: most outdoor surface are uneven, especially those found in a natural environment like parks and forests. Even some man-made surfaces are not perfectly even. For example, sidewalks are not perfectly leveled; therefore, this type of surface can allow to evaluate a more “natural” walking condition.

CHAPTER 3: RESEARCH OBJECTIVES AND HYPOTHESES

3.1 Research question

Does arm swing motion have an impact on dynamic stability during gait when walking on a challenging terrain?

3.2 Purpose

The aim of this study is to compare the effect of different arm motions on dynamic stability when walking on a normal surface and a rocky surface. More specifically, we aim at identifying if dynamic stability is affected by the combined effect of arm swing and gait surface.

3.3 Research hypothesis

Based our group's previous study (Siragy et al., 2019), we hypothesize that active arm swing will have a negative impact on postural control and gait dynamics when walking on regular surface (increased trunk velocity variability and coefficient of variation (COV), as well as decreased MOS). Following the studies of Gates et al. (2012; 2013), the rocky surface is believed to have a negative impact on both postural control and dynamic stability (increased trunk velocity variability and coefficient of variation (COV), as well as decreased MOS). However, in accordance with the studies by Bruijn et al. (2010) and Wu et al., (2016), we expect an interaction will exist between arm swing and surface conditions as active arm swing will have a positive impact and attenuate the negative effects of the rocky surface (decreased trunk velocity variability and coefficient of variation (COV), as well as increased MOS).

CHAPTER 4: GENERAL METHODS

4.1 Participants

Fifteen healthy young adults from the University of Ottawa community aged 18-30 years old were recruited to participate in this study. The sample size is based on similar protocols performed by Bruijn et al. (2010) and Gates et al. (2012) in which healthy young adults completed both normal and perturbed walking trials. Exclusion criteria included any physical discomfort using a virtual reality system, any reported injuries and/or orthopedic surgeries in the previous 12 months that could interfere with gait. All participants provided informed written consent and the study was approved by local ethics and scientific committees. Using the *International Physical Activity Questionnaire*, seven participants were identified as having a moderate physical activity level and six participants identified as having a high level of physical activity. Two participants did not complete the questionnaire.

4.2 Instrumentation

3D motion capture was completed using a virtual park scenario within the CAREN-Extended System (Motek Medical, Amsterdam, NL), which combined a six degree-of-freedom motion platform with embedded dual-belt instrumented treadmill, 12 camera Vicon motion capture system, 180-degree projector screen, and safety harness. A 57-marker set was used to track full body kinematics (Sinitski et al., 2015). Kinematic data was collected at 100 Hz and ground reaction forces (GRF) at 1000 Hz. Vicon Nexus (Nexus 2.6, Oxford, UK) were used to process markers and GRF data before exporting to Visual3D v6 (C-Motion, Germantown, MD) for 3D kinematic and kinetic calculations.

4.3 Procedures

Participants completed three-minute trials of steady-state walking at a speed of 1.2 m/s on a dual-belt treadmill using a virtual park scenario within the CAREN-Extended System (Motek Medical, Amsterdam, NL). Participants then walked on a dual-belt treadmill at a self-paced speed. Walking trials consisted of steady-state walking over 20 meters, followed by 20 meters on a "rocky" surface and then another 20 meters of steady-state walking. We fixed the walking speed on the regular surface to allow for more consistency between trials and ensure true steady-state conditions. The rocky surface is simulated through the pseudo-random oscillation of the platform in three directions simultaneously using the CAREN "Rumble" module with a maximum range of ±2 cm at 0.6 Hz vertically, ±1° at 1 Hz pitch, and ±1° at 1.2 Hz roll (Sinitski et al., 2015). This was repeated three times, once for each of the different arm conditions which included: (1) **Normal** - participants' natural arm motion, (2) **Held** - arms held alongside the thighs and secured in the harness, and (3) **Active** - arms actively swinging to shoulder height on each surface condition (regular and rocky surfaces). Participants completed a total of six trials of three minutes each. For each trial, participants walked for 25 seconds to reach steady-state walking before beginning data collection. This time was chosen as some studies showed that the minimal time to reach steady-state of kinematic data and spatiotemporal data ranged from 10 to 30 seconds (Meyer et al., 2019; Van de Putte et al., 2006; Wall & Charteris, 1980), though true steady-state gait requires at least 10 minutes (Meyer et al., 2019; Van de Putte et al., 2006; Wall & Charteris, 1980). Participants wore a safety harness attached to an overhead structure throughout the entire procedure. They also had the possibility to rest as necessary to minimize the effect of fatigue.

4.4 Data processing

Twenty consecutive steps were taken at random from the steady-state trials to compare to 20 consecutive steps from the "rocky" terrain. Spatial-temporal data (step length, width, time), and peak trunk angular and linear velocities for each stride were extracted and analyzed using custom scripts in Matlab (Mathworks, Natick, MA). A 4^{th} order, low-pass Butterworth filter with a 12 Hz cut-off frequency was used to filter kinematic data.

4.5 Independent variables

The independent variables for this study were the three arm conditions (held, normal and active arm swing) and the treadmill conditions (regular terrain and rocky surface).

4.6 Dependant variables

The dependent variables for this study were trunk linear and angular velocity mean, standard deviation (SD) and maximal values. Larger SD of trunk linear and angular velocities indicate poorer postural control (Siragy et al., 2019). Spatial-temporal variability for step length, width and time for both legs were calculated as the COV calculated as follow (SD / Mean) x 100. Larger COV represents decreased gait dynamic balance (Siragy et al., 2019).

The MOS, defined as the Extrapolated Center of Mass's distance (xCOM) to the right/left lateral heel marker with $xCOM = COM_{position} + (COM_{velocity}/\varpi_{\Theta})$, where $\varpi_{\Theta} = \sqrt{g/l}$. In this term, $g = 9.81 m/s^2$ and l is the inverted pendulum length calculated as the average distance of the right/left lateral heel marker to the COM at respective heel-strikes. A larger MOS values indicates greater dynamic balance as the xCOM is further away from the BOS's edge (Hak et al., 2013; Hof et al., 2005; Hof, 2008; McAndrew Young et al., 2012). MOS was calculated for both legs at respective heel-strikes. The COM velocity was calculated as the first central difference of the COM's position. However, only the ML MOS was calculated as previous research suggests that this metric is only valid in this direction (Bruijn et al., 2013). A baseline was established using the walking on regular surface with normal arm swing. This baseline was used as a reference for comparing the effects of both the terrain and arm conditions on our dependent variables.

4.7 Statistical analyses

Data was analyzed using SPSS 23.0 and $p<0.05$ was considered statistically significant. The Shapiro-Wilk test was used to verify normality of variables. A mixed linear model was generated to test for an interaction with arm swing and surface conditions set as fixed effects while walking speed was used as a covariate. In the event that an interaction was not found (p-value greater than 0.05), a test for the main effects of arm swing and walking conditions was performed. Post-hoc with a Bonferroni correction was used to compare all main effects and interactions when applicable.

CHAPTER 5: MANUSCRIPT

The Effect of Arm Swing and Rocky Surface on Dynamic Balance
in Healthy Young Adults

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Abstract

Fall-induced injuries can stem from a disruption in the postural control system and place a financial burden on the healthcare system. Most gait research focused on lower extremities and neglected the contribution of arm swing, which have been shown to affect the movement of the center of mass when walking. This study evaluated the effect of arm swing on postural control and stability during regular and rocky surface walking. Fifteen healthy young adults (age = 23.4 ± 2.8) walked on these two surfaces with three arm motions (normal, held, and active) using the CAREN Extended-System (Motek Medical, Amsterdam, NL). Mean, standard deviation and maximal values of trunk linear and angular velocity were calculated in all three axes. Moreover, step length, time and width mean and coefficient of variation as well as margin of stability mean and standard deviation were calculated. Active arm swing increased trunk linear and angular velocity variability and peak values compared to normal and held arm conditions. Active arm swing also increased participants' step length and step time, as well as the variability of margin of stability. Similarly, rocky surface walking increased trunk kinematics variability and peak values compared to regular surface walking. Furthermore, rocky surface increased the average step width while reducing the average step time. Though this surface type increased the coefficient of variation of all spatiotemporal parameters, rocky surface also led to increased margin of stability mean and variation. The spatiotemporal adaptations showed the use of "cautious" gait to mitigate the destabilizing effects of both the active arm swing and rocky surface walking and, ultimately, maintain dynamic stability.

5.1 Introduction

Falls and fall-related injuries that require medical attention are common debilitating issues (Bergen et al., 2016) that place a financial burden on the healthcare system (Ministry of Labour Inspection Blitzes, n.d.; Yeoh et al., 2013). Falls can result from a disruption in the postural control system (Hof et al., 2005), a system tasked with the maintenance of relative segmental positioning to ensure a reliable reference frame (MacKinnon & Winter, 1993; Massion, 1994; Ivanenko & Gurfinkel, 2018). Normally, the center of mass (COM) translates smoothly in a sinusoidal trajectory in the walking direction and is kept within the base of support (BOS) by the safe placement of the foot on the ground during double support (Winter, 1995). This dynamic between the COM and the BOS allows for a safe and efficient gait (Lugade et al., 2011), which can be evaluated using spatiotemporal parameters, linear and angular segmental kinematics to assess postural control (Schooten et al., 2011), while margin of stability (MOS; distance between the edge of the BOS and the extrapolated COM position) (Hof et al., 2005) and coefficient of variation (COV) (Siragy et al., 2019) have been shown relevant to evaluate gait dynamic stability (the ability to maintain one's COM within a moving BOS (Siragy & Nantel, 2018)). As each measure gives insight into a particular neuromuscular control component (Siragy & Nantel, 2018), it is essential to consider the objective of the study when selecting metrics to assess human locomotion.

The selection of anthropometric models is also critical when studying gait. Most gait studies assessing stability and fall risk tend to focus on the lower extremities (Choi & Kim, 2015; Gates et al., 2012) and rely on the inverted pendulum model (Winter, 1995), thereby neglecting the contribution of the arms. Therefore, the potential impact of arm motion on trunk stability was not taken into account as it was considered a passive product of trunk motion (Meyns et al., 2013; Winter, 1995). However, electromyography studies reported an active contribution of the shoulder muscles to arm swing (Collins et al., 2009; Meyns et al., 2013), which could actively contribute to gait stability (Collins et al., 2009).

Current literature remains conflicting with regards to the impact of arm motion on postural stability when walking. Arm motion during gait has been shown to aid dynamic stability by counteracting the lower body's angular momentum (Angelini et al., 2018; Nakakubo et al., 2014; Ortega et al., 2008; Punt et al., 2015). Yet other studies (Bruijn et al., 2010; Pijnappels et al., 2010) showed that walking with restricted arm motion improved dynamic stability through increased trunk inertia, which reduced COM displacement. Some examination of active arm swing have also demonstrated a positive association between active arm swing and dynamic stability based on local divergence exponent (Hu et al., 2012; Punt et al., 2015; Wu et al., 2016). Contrarily, our group showed that active arm swing led to increased gait variability of spatiotemporal parameters and decreased stability based on the harmonic ratios (a measure of a signal's periodicity), which stemmed from the increase in trunk kinematics variability (Siragy et al., 2019). Nonetheless, arm motion has been shown to affect the COM's trajectory in a steady-state condition.

However, in the case of perturbations, the gait pattern is altered and becomes more variable (Madehkhaksar et al., 2018). Perturbations in the anteroposterior (AP) direction, such as adopting an asymmetric gait pattern, can impair normal COM motion, increase trunk movement and spatiotemporal variability (Madehkhaksar et al., 2018; Siragy et al., 2019). Perturbations in the mediolateral (ML) direction, such as lateral tugging, can generate similar adaptations, but require greater active control (Madehkhaksar et al., 2018; McAndrew et al., 2010; McAndrew et al., 2011) as ML direction has been shown to be more unstable during biped walking (Kuo, 1999). When walking on a rocky surface, which presents perturbations in both the AP and ML directions simultaneously, spatiotemporal parameters (step length, width and time) are altered and become more variable as a response to the deviation of the body's COM from its intended path (Gates et al., 2012; Hawkins et al., 2018; McAndrew et al., 2010; McAndrew et al., 2011). McAndrew Young et al. (2012) and Onushko et al. (2019) showed that perturbations such as continuous pseudo-random oscillations in the AP and ML directions led to increased foot placement variability, and sometimes, negative MOS_{ML} (an indicator of loss of stability where the extrapolated COM position is outside of the BOS in the ML direction). As arm motion has been shown to affect COM motion, it is then possible that the previously mentioned arm swing strategies (normal, held and active) can mitigate the mechanical destabilisation caused by this challenging terrain and reduce the gait variability (Wu et al., 2016).

Therefore, the present study assessed the effect of normal arm swing, held arm swing and active arm swing on postural control and dynamic stability when walking on regular and rocky surface. To our knowledge no studies investigated the impact of arm motion in terrain inducing perturbation in both the AP and ML direction. We hypothesized that active arm swing will have a negative impact on postural control and gait dynamics on a regular surface, while rocky surface walking will decrease stability and increase spatiotemporal variability. Moreover, we expected to see an interaction between arm swing and surface type. We hypothesized that postural control and stability will be increased with normal and active arm swing while walking on a rocky surface compared walking without arm swing.

5.2 Methodology

5.2.1 Participants

A convenient sample of fifteen healthy young adults from the University of Ottawa community (eight males, seven females; mean age 23.4 ± 2.8 years; mean height 170.2 ± 8.1 cm; mean weight 72.3 ± 13.5 kg) were recruited to participate in this study. Exclusion criteria were any physical discomfort using a virtual reality system, any reported injuries and/or orthopedic surgeries in the previous 12 months that could interfere with gait. All participants provided informed written consent and the study was approved by the University of Ottawa's Institutional Review Board and the Ottawa Hospital Research Ethics Board.

5.2.2 Experimental protocol

Participants completed three-minute trials of steady-state walking at a speed of 1.2 m/s on a dual-belt treadmill using a virtual park scenario within the CAREN-Extended System (Motek Medical, Amsterdam, NL). This system includes 12 Vicon cameras for motion capture and an instrumented platform to capture kinematic and kinetic parameters. One trial was performed for each arm swing type: (1) **Normal** – participants' natural arm motion, (2) **Held** - arms held alongside the thighs and secured in the harness, (3) **Active** - arms actively swinging to shoulder height. These were completed in a random order. Afterwards, participants were asked to walk at a self-paced speed using the same virtual park scenario. This was also repeated three times in a random order, once for each of the different arm conditions. These last walking trials consisted of steady-state walking over 20 meters, followed by 20 meters on the "rocky" surface and then another 20 meters of steady-state walking. Using the CAREN "Rumble" module, the rocky surface was simulated through the pseudo-random oscillation of the platform in three directions simultaneously with a maximum range of ±2 cm at 0.6 Hz vertically, ±1° at 1 Hz pitch, and ±1° at 1.2 Hz roll (Sinitski et al., 2015). We fixed the walking speed on the regular surface to allow for more consistency between trials and ensure true steady-state conditions. For each trial, participants walked for 25 seconds to reach steady-state before beginning data collection and before the first set of 20 meters steady-state walking in the rocky surface trials. Participants wore a safety harness attached to an overhead structure throughout the entire procedure. They also had the possibility to rest as necessary to minimize the effect of fatigue.

5.2.3 Data analysis

Twenty consecutive steps were taken at random from the steady-state trials to compare to 20 consecutive steps from the “rocky” terrain. The independent variables for this study were the three arm conditions (normal, held, and active arm swing) and the treadmill conditions (regular terrain and rocky surface). A baseline was established using the walking on regular surface with normal arm swing. The dependent variables for this study were trunk linear and angular velocity mean, standard deviation (SD) and maximal values. Additionally, spatiotemporal (step length, step time and step time for both legs) mean and variability were evaluated. Spatiotemporal variability were measured as the COV calculated as (SD / mean) x 100.

The MOS was calculated for both legs at respective heel-strikes. The MOS is defined as the extrapolated center of mass’s distance (xCOM) to the right/left lateral heel marker with xCOM = $COM_{position} + (COM_{velocity}/\varpi_{\Theta})$, where $\varpi_{\Theta} = \sqrt{g/l}$. In this term, $g = 9.81 m/s^2$ and l is the inverted pendulum length calculated as the average distance of the right/left lateral heel marker to the COM at respective heel-strikes. The COM velocity was calculated as the first central difference of the COM’s position. The MOS was only calculated in the ML direction as previous research suggests that this metric is only valid in this direction (Bruijn et al., 2013).

Spatiotemporal data (step length, width, time), and peak trunk angular and linear velocities for each stride were extracted and analyzed using custom scripts in Visual3D (C-Motion, Germantown, MD) and Matlab (Mathworks, Natick, MA). A 4^{th} order, low-pass Butterworth filter with a 12 Hz cut-off frequency was used to filter kinematic data.

Data was analyzed using SPSS 23.0 (IBM Analytics, Armonk, USA) and $p<0.05$ was considered statistically significant. The Shapiro-Wilk test was used to verify normality of variables. A mixed linear model was generated to test for an interaction with arm swing and surface conditions set as fixed effects while walking speed was used as a covariate. In the event that an interaction was not found (p-value greater than 0.05), a test for the main effects of arm swing and walking conditions was performed. Post-hoc with a Bonferroni correction was used to compare all main effects and interactions when applicable.

5.3 Results

Descriptive statistics for linear and angular velocities are presented in

Table 1 and Table 2 respectively, while spatiotemporal and MOS data means and variability are reported in Table 3 and Table 4 respectively.

5.3.1 Sagittal plane

In the sagittal plane, no interactions were detected between arm and surface conditions. A main effect for arm swing existed for linear velocity max ($F(2,70.573)=46.661$, $p<0.001$), angular velocity SD ($F(2,70.484)=16.212$, $p<0.001$) and angular velocity max ($F(2,70.000)=17.356$, $p<0.001$). Post-hoc comparisons revealed that active arm swing increased these parameters compared to normal and held arm swing conditions (Figure 1 and Figure *2*). Additionally, a main effect for surface condition was detected for linear velocity SD ($F(1,48.523)=74.949$, $p<0.001$), angular velocity SD ($F(1,71.151)=84.382$, $p<0.001$) and angular velocity max ($F(1,70.000)=61.496$, $p<0.001$). Post-hoc revealed that rocky surface increased these parameters compared to regular surface (Figure 1 and Figure *2*).

5.3.2 Frontal plane

Within the frontal plane, statistical analyses revealed a main effect for arm swing in linear velocity SD ($F(2,68.850)=4.045$, $p<0.05$) and max ($F(2,70.000)=8.655$ $p<0.001$) values. Post-hoc tests demonstrated that active arm swing was significantly higher than normal arm swing for linear velocity SD (Figure 3). Furthermore, for linear velocity max, active arm swing was significantly higher than the normal and held arm conditions (Figure 3). A main effect was also detected for surface condition for linear velocity mean ($F(1,56.650)=6.195$, $p<0.05$), SD ($F(1,42.469)=11.694$, $p<0.01$) and max ($F(1,70.000)=42.016$, $p<0.001$). Post-hoc analysis showed that rocky surface led to larger values compared to regular surface (

Table 1 and Figure 3). As for angular velocity, an interaction was identified for the mean ($F(2,70.000)=3.251$, $p<0.05$). The relationship between arm swing and surface conditions are shown in Figure 7A. Active arm swing led to significantly larger values compared to normal arm swing when walking on a regular surface and walking on a rocky surface led to larger values compared to a regular surface when using a held arm swing strategy (Figure 7A). A main effect for surface conditions were found for angular velocity SD ($F(1,70.00)=49.570$, $p<0.001$) and max ($F(1,71.514)=10.067$, $p<0.01$). In all cases, rocky surface led to larger values when compared to regular surface (Figure 4).

5.3.3 Transverse plane

No interactions between arm swing and surface conditions were detected in this plane of motion. Along the vertical axis, a significant difference for arm swing existed for linear velocity SD ($F(2,70.000)=4.746$, $p<0.05$), where active swing led to larger values compared to normal swing (Figure 5). Moreover, a significant difference for arm swing existed for angular velocity mean ($F(2,70.000)=6.370$, $p<0.01$), SD ($F(2,70.000)=9.279$, $p<0.001$) and max

($F(2,70.000)=27.862$, $p<0.001$) values. Active arm swing led to larger values when compared to normal and held arm conditions for all angular velocity parameters (Table 2 and Figure 6). In terms of surface condition, a significant difference was detected for linear velocity SD ($F(1,70.000)=366.015$, $p<0.001$), linear velocity max ($F(1,72.784)=40.601$, $p<0.001$), as well as angular velocity mean ($F(1,70.000)=4.326$, $p<0.05$), SD ($F(1,70.000)=9.062$, $p<0.01$) and max ($F(1,70.000)=10.154$, $p<0.01$) values. Once again, rocky surface led to larger values when compared to regular surface in all cases (Table 2, Figure 5 and Figure 6).

5.3.4 Spatiotemporal

There was an interaction detected for the average step time on the left leg ($F(2,70.104)=4.592$, $p<0.05$) with the relationship between the two independent variables shown in Figure 7C. In general, active arm swing led to significantly larger step times than normal and held arm swings on both walking surfaces (Table 3). In addition, a regular walking surface led to larger values compared to walking on a rocky surface (Table 3). Statistical analysis results also revealed an arm swing main effect for average step length of the left ($F(2,70.302)=28.032$, $p<0.001$) and right ($F(2,70.079)=27.666$, $p<0,001$) leg, as well as right average step time ($F(2,69.845)=64.961$, $p<0.001$). Post-hoc revealed that active arm swing increased values in all cases compared to the normal and held arm conditions (Table 3). Furthermore, a surface main effect was detected for average step width following a left heel strike ($F(1,70.000)=72.343$, $p<0.001$) and a right heel strike ($F(1,70.000)=74.455$, $p<0.001$), as well as average right step time ($F(1,71.870)=11.111$, $p<0.01$). As for step width, rocky surface led to larger values (Table 3); while for step time, rocky surface led to smaller values as compared to regular surface walking (Table 3).

In terms of the COV of spatiotemporal parameters, interactions were revealed for right step length $F(2,70.828)=3.540$, $p<0.0.5$) and step width at both left ($F(2,69.369)=8.634$, $p<0.001$) and right ($F(2,70.000)=5.699$, $p<0.01$) heel strikes. These interactions are displayed in Figure 7D, E and F. The right step length, active arm swing only led to larger values compared to normal and held when walking on a rocky surface. This challenging surface led to larger values despite the arm strategy used by the participants (Figure 7D). As for step width, active arm swing led to larger values compared to the other two arm swing strategies only when walking on a rocky surface at both heel strikes (Figure 7E and F). Also, in general, rocky surface led to significantly larger values than regular surface walking at both heel strikes no matter the arm swing used (Figure 7D, E and F). The only exception to this is the COV of step width at left heel strike when using the held arm swing; rocky surface still led to larger values, but the difference yielded a p-value greater than 0.05 (Table 9B in the appendix) In addition, statistical analyses did not indicate any arm swing main effect. However, a surface main effect was shown for left step length ($F(1,75.286)=142.059$, $p<0.001$) and step times on both left ($F(1,73.759)=136.657$, $p<0.001$) and right ($F(1,76.622)=157.699$, $p<0.001$) legs. Post-hoc analysis indicated that rocky surface led to larger values compared to regular surface (Table 4).

5.3.5 Mediolateral margin of stability

For the standard deviation of the MOS of the right leg, an interaction was detected ($F(2,64.871)=4.211$, $p<0.05$) and this relationship can be seen in Figure 7B. Overall, active arm swing led to larger values compared to normal arm swing when walking on a regular surface and led to larger values compared to both normal and held arm strategies when walking on a rocky surface. Also, rocky surface led to larger values compared to regular surface no matter the arm strategy used (Figure 7B). Statistical analyses showed only a surface main effect on MOS mean for both the left ($F(1,57.186)=99.104$, $p<0.001$) and right ($F(1,56.282)=126.489$, $p<0.001$) legs, where rocky surface led to larger values overall compared to regular surface for both legs (Table 3). For the SD of this metric, both an arm and surface main effect were detected for the left leg ($F(2,69.601)=7.550$, $p<0.01$ and $F(1,46.588)=166.696$, $p<0.001$ respectively). Post-hoc showed that active arm swing led to larger values than normal and held arm swing and rocky surface led to larger values than the regular surface (Table 4).

Table 1

Trunk linear velocities (x10^{-3} m/s) according to surface type and arm swing strategy in all three directions

	Regular surface			Rocky surface		
	Normal	Held	Active	Normal	Held	Active
AP	4.5 ± 3.3	4.3 ± 4.8	3.7 ± 3.2	4.4 ± 2.1	3.7 ± 2.4	5.5 ± 4.1
ML*	1.6 ± 1.0	1.8 ± 1.3	1.3 ± 0.9	2.0 ± 1.4	2.9 ± 2.0	3.1 ± 2.5
V	1.9 ± 0.4	2.1 ± 0.3	1.9 ± 0.5	2.5 ± 1.1	2.3 ± 1.0	2.4 ± 1.2

Note: * shows that rocky surface led to larger values than regular surface at $p<0.05$. AP represents the anteroposterior direction, ML represents the mediolateral direction and V represents the vertical direction.

Table 2

Trunk angular velocities (x10^{-2} °/s) according to surface type and arm swing strategy in all three directions

	Regular surface			Rocky surface		
	Normal	Held	Active	Normal	Held	Active
AP	7.7 ± 5.7	10.2 ± 7.6	12.6 ± 9.7	15.9 ± 10.4	12.8 ± 12.0	20.6 ± 13.2
ML	12.4 ± 9.0	16.6 ± 19.4	34.9 ± 40.4	27.4 ± 22.5	41.9 ± 27.6	33.1 ± 25.8
V*	11.7 ± 6.4	12.7 ± 10.2	19.6 ± 16.9 ‡	12.3 ± 6.9	19.0 ± 22.3	36.34 ± 33.4 ‡

Note: * shows that rocky surface led to larger values than regular surface at $p<0.05$ and ‡ shows that active arm swing led to larger values than normal and held at $p<0.05$. AP represents the anteroposterior direction, ML represents the mediolateral direction and V represents the vertical direction.

Table 3

Average step length (cm), width (cm), time (ms) and margin of stability (cm) for both left and right heel strikes according to the surface and arm swing conditions

		Regular			Rocky		
		Normal	Held	Active	Normal	Held	Active
Step length	Left	56.2 ± 3.7	56.1 ± 4.0	61.3 ± 3.4 ‡‡‡	58.2 ± 9.8	53.9 ± 6.3	67.7 ± 5.8 ‡‡‡
	Right	56.6 ± 3.4	55.9 ± 3.8	60.9 ± 4.6 ‡‡‡	57.5 ± 9.8	52.0 ± 7.5	66.7 ± 7.0 ‡‡‡
Step width	Left***	17.7 ± 3.9	17.3 ± 3.4	18.9 ± 4.0	21.4 ± 3.6	22.8 ± 5.4	22.2 ± 4.0
	Right***	17.7 ± 3.8	17.3 ± 3.5	19.0 ± 3.9	21.7 ± 3.7	22.9 ± 5.2	22.3 ± 3.8
Step time	Left	531 ± 24	531 ± 34	577 ± 36	488 ± 49	483 ± 48	547 ± 48
	Right##	528 ± 27	524 ± 28	584 ± 50 ‡‡‡	490 ± 52	485 ± 45	548 ± 48 ‡‡‡
MOS	Left***	12.0 ± 1.8	12.1 ± 1.5	12.4 ± 2.2	19.9 ± 4.4	20.0 ± 5.3	20.8 ± 5.7
	Right***	10.9 ± 1.8	11.1 ± 1.9	11.3 ± 1.9	17.0 ± 2.6	18.2 ± 3.6	17.8 ± 5.4

Note: ## shows that the regular surface led to larger values compared to the rocky surface at $p<0.01$; *** shows that the rocky surface led to larger values compared to the the regular surface at $p<0.001$, while ‡‡‡ shows that active arm swing led to larger values than normal and held at $p<0.001$.

Table 4

Coefficient of variance of step length, step width and step time, as well as the standard deviation of the margin of stability ($x10^{-2}$ cm) for both heel strikes according to surface type and arm swing strategy

		Regular			Rocky		
		Normal	Held	Active	Normal	Held	Active
Step length	Left***	1.9 ± 0.5	2.4 ± 0.8	3.0 ± 1.0	5.8 ± 1.9	7.1 ± 2.8	5.5 ± 2.2
	Right	2.4 ± 0.7	2.3 ± 0.5	3.2 ± 1.3	5.7 ± 3.4	8.1 ± 3.8	5.2 ± 3.3
Step width	Left	8.8 ± 3.9	10.2 ± 4.6	8.7 ± 3.1	13.2 ± 3.9	12.6 ± 5.3	19.6 ± 7.0
	Right	7.9 ± 2.5	9.6 ± 4.4	9.4 ± 3.4	11.3 ± 3.9	12.8 ± 4.9	18.8 ± 6.6
Step time	Left***	1.9 ± 0.5	2.2 ± 0.6	2.5 ± 0.8	4.8 ± 1.6	5.4 ± 1.8	4.9 ± 1.3
	Right***	1.7 ± 0.7	2.0 ± 0.6	2.5 ± 1.1	4.6 ± 2.0	5.8 ± 1.8	4.9 ± 1.4
MOS	Left	1.3 ± 0.3	1.3 ± 0.4	1.7 ± 0.6	3.9 ± 1.4	3.7 ± 1.1	5.1 ± 1.5

	Right	1.3 ± 0.4	1.2 ± 0.4	1.4 ± 0.5	3.7 ± 1.4	3.8 ± 1.3	5.6 ± 2.5

Note: *** shows that rocky surface led to larger values than regular surface at p<0.001.

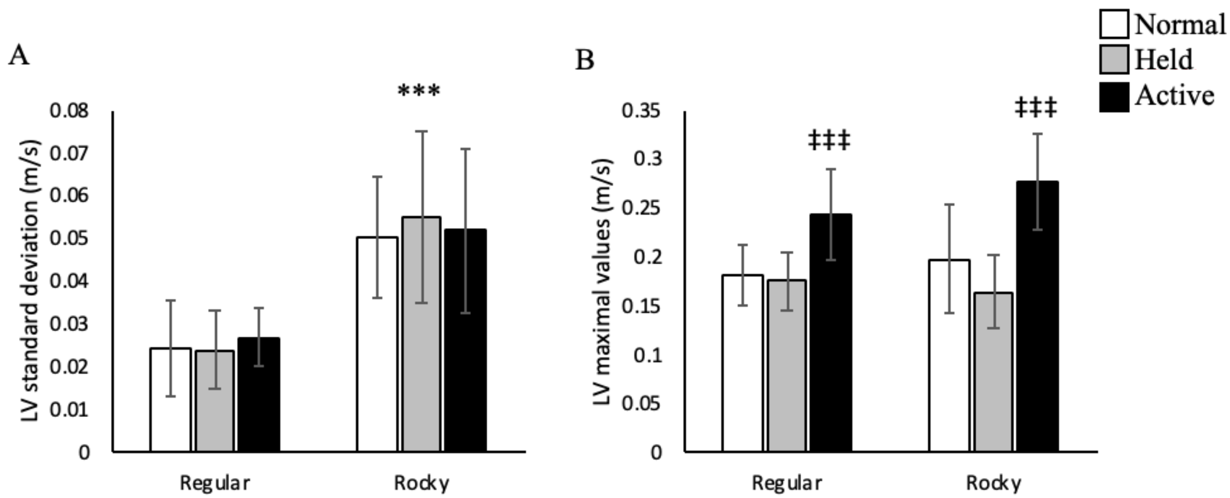

Figure 1. Trunk linear velocity (A) standard deviation (m/s) and (B) maximal values (m/s) according to surface type for each arm condition in the anteroposterior direction. *** shows that rocky surface led to larger values than regular surface at p<0.001, while ‡‡‡ shows that active arm swing led to larger values than normal and held at p<0.001.

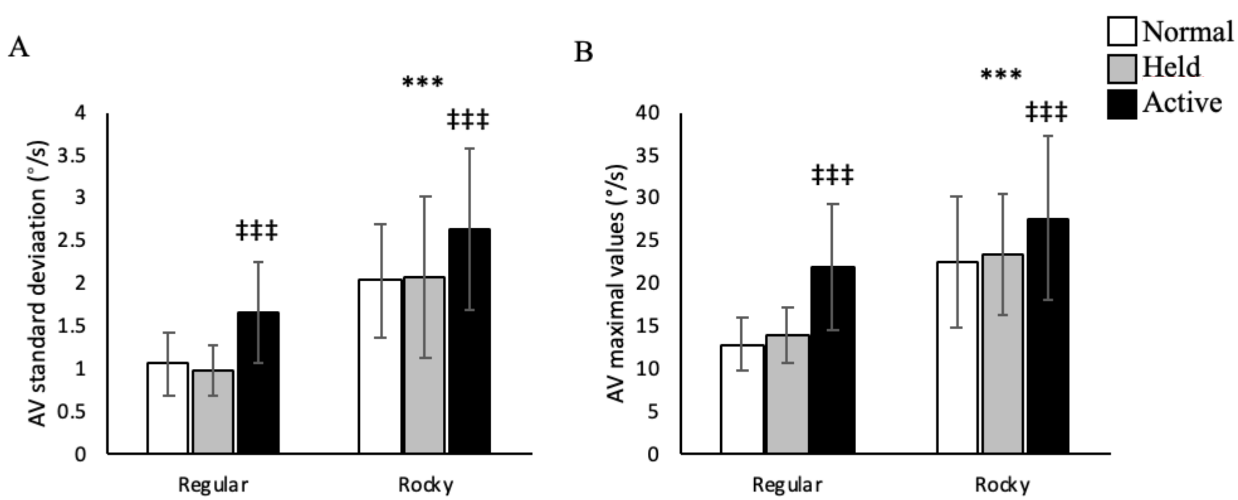

Figure 2. Trunk angular velocity (A) standard deviation (m/s) and (B) maximal values (m/s) according to surface type for each arm condition about the anteroposterior axis. *** shows that rocky surface led to larger values than regular surface at p<0.001, while ‡‡‡ shows that active arm swing led to larger values than normal and held at p<0.001.

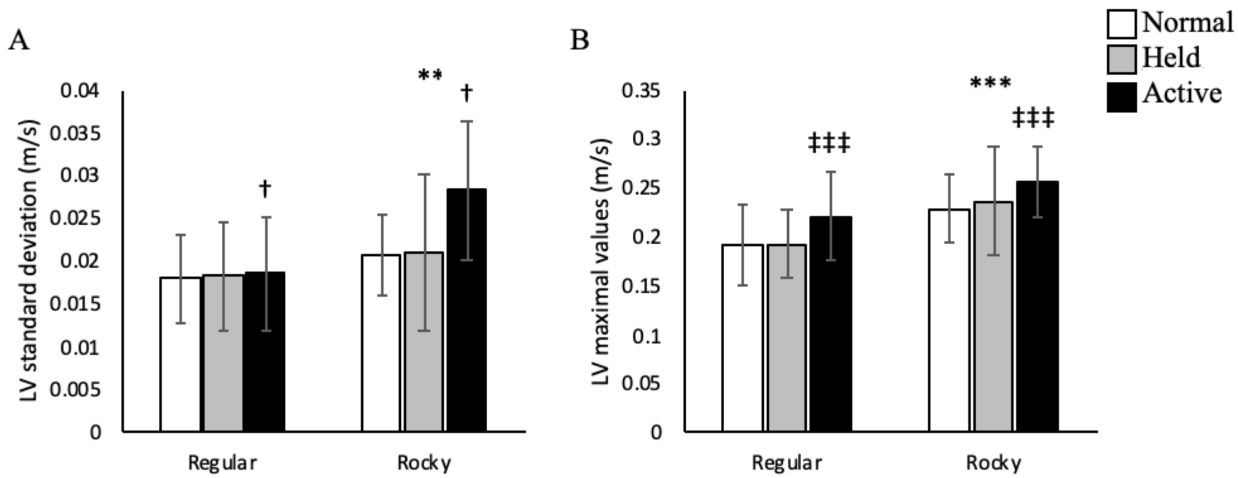

Figure 3. Trunk linear velocity (A) standard deviation (m/s) and (B) maximal values (m/s) according to surface type for each arm condition in the mediolateral direction. ** and *** shows that rocky surface led to larger values than regular surface at p<0.01 and p<0.001, while † shows that active arm swing led to larger values than normal arm swing at p<0.05 and ‡‡‡ shows that active arm swing led to larger values than normal and held at p<0.001.

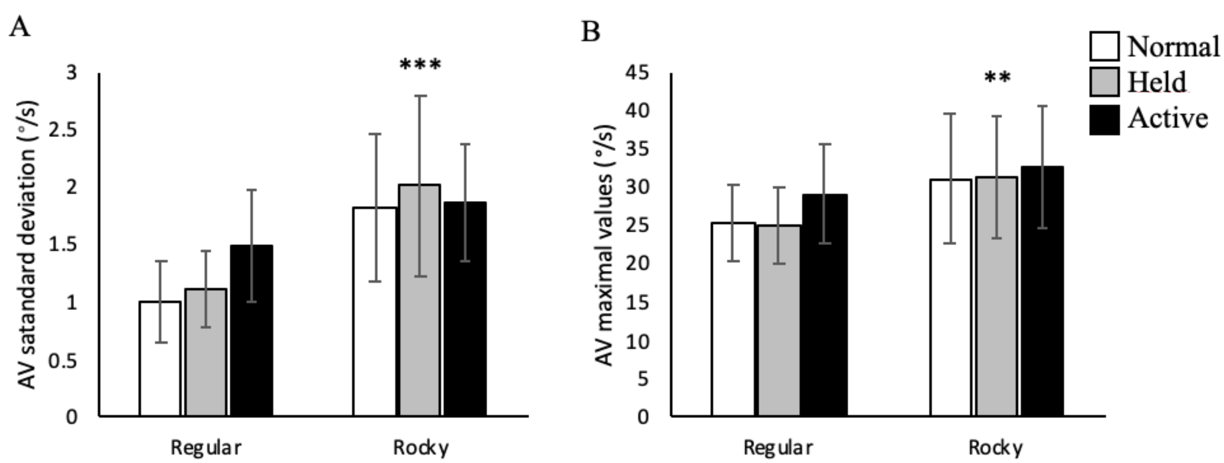

Figure 4. Trunk angular velocity (A) standard deviation (m/s) and (B) maximal values (m/s) according to surface type for each arm condition about the mediolateral axis. ** and *** shows that rocky surface led to larger values than regular surface at p<0.01 and p<0.001.

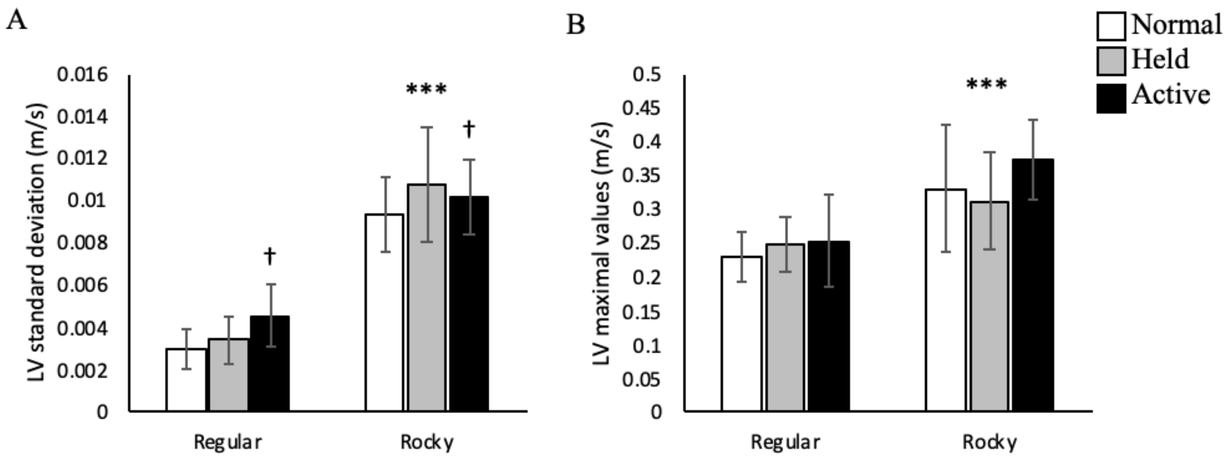

Figure 5. Trunk linear velocity (A) standard deviation (m/s) and (B) maximal values (m/s) according to surface type for each arm condition in the vertical direction. *** shows that rocky surface led to larger values than regular surface at p<0.001 while † shows that active arm swing led to larger values than normal arm swing at p<0.05.

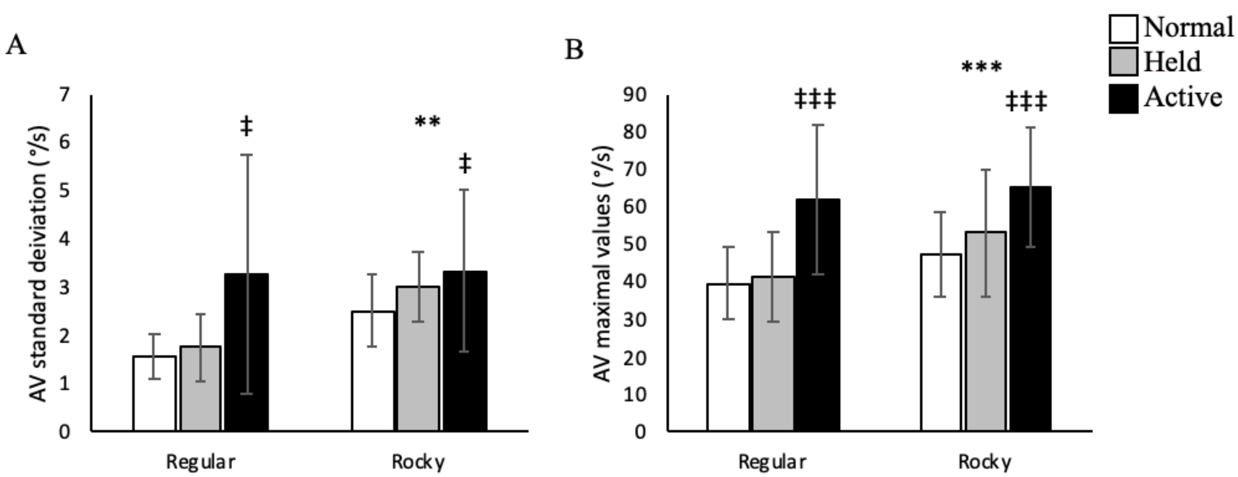

Figure 6. Trunk angular velocity (A) standard deviation (m/s) and (B) maximal values (m/s) according to surface type for each arm condition about the vertical axis. ** and *** shows that rocky surface led to larger values than regular surface at p<0.01 and p<0.001, while ‡ and ‡‡‡ shows that active arm swing led to larger values than normal arm swing at p<0.05 and p<0.001.

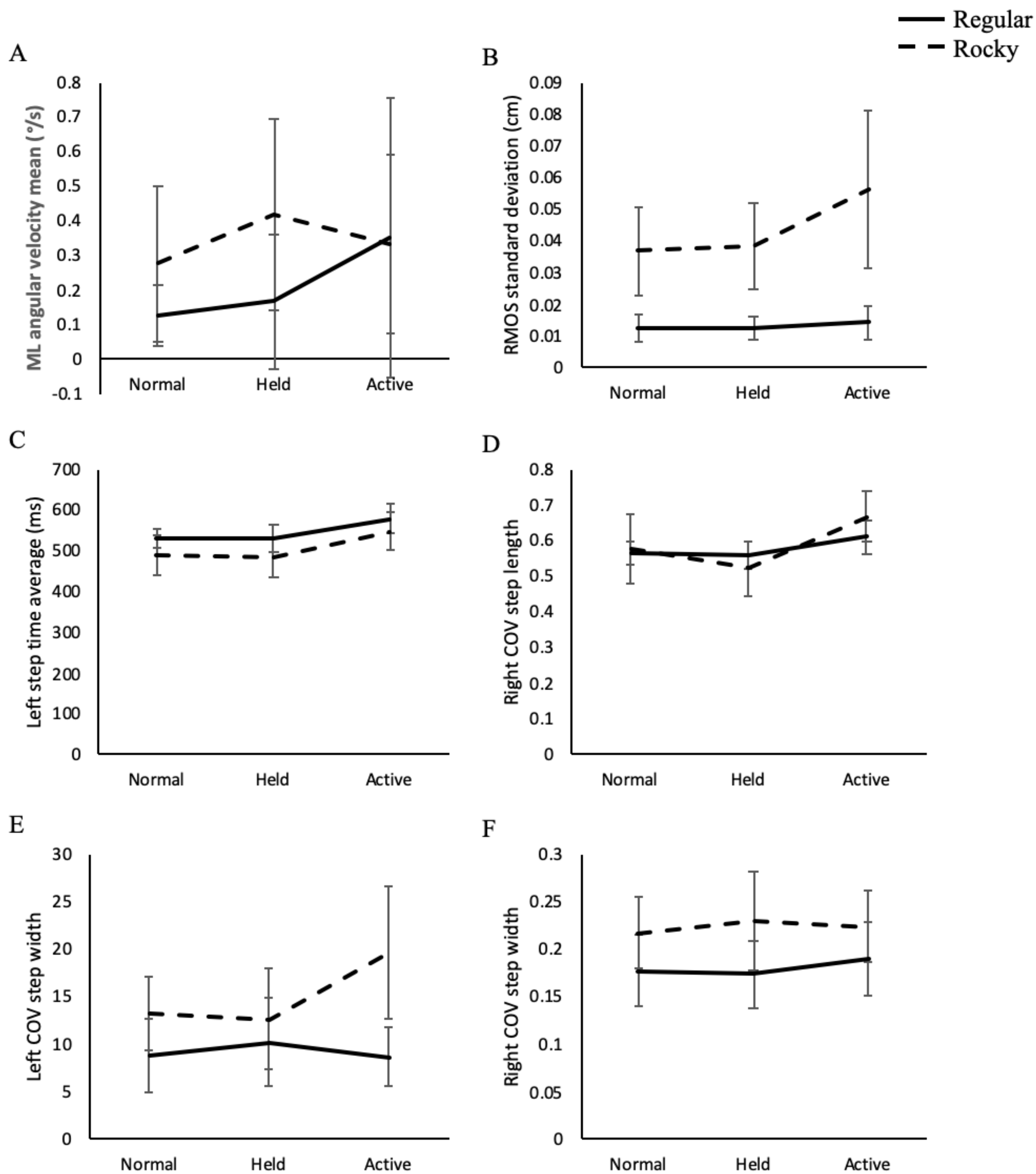

Figure 7. Interaction figures according surface type and arm swing strategy used for (A) angular velocity mean (°/s) about the mediolateral axis, (B) MOS standard deviation at right heel strike (cm), (C) step time average (ms) at left heel strike, (D) COV of step width at left heel strike, (E) COV of step length (cm) and (F) step width (cm) at right heel strike.

5.4 Discussion

5.4.1 Main findings

This study examined the effects of different arm swing conditions (normal, held and active) on postural control and dynamic stability in healthy young adults when walking on even and rocky surfaces. Our hypotheses were partially supported as overall our results demonstrated that, compared to normal and held arm swing, active arm swing (1) increased trunk kinematics variability and peak values, (2) increased the average step length and step time while increasing step width COV. As for the effect of terrains, results showed (1) increased trunk kinematics variability and peak values as well as larger mean MOS and MOS variability when walking on a rocky surface compared to regular surface, and (2) increased average step width and decreased average step time when walking on rocky compared to regular surface. Finally, rocky surface led to increased COV of all spatiotemporal values (length, width and time).

5.4.2 Arm swing

Overall and in accordance with our hypothesis, walking while actively moving the arms had a destabilizing effect compared to normal or held arm swing. When using the active arm swing strategy, our participants displayed increased mean and peak values of trunk angular velocities in the frontal and transverse planes as well as increased peak values for trunk linear velocities compared to the normal and held arm conditions. The control of the trunk is critical for regulating postural control (Menz et al., 2013; Tucker et al., 2008). Studies using motion capture and inertial sensors placed at the trunk level reported that the increase in linear velocity and angular velocity were indicators of diminished postural control and increased risk of falls (Arvin; Mazaheri;

Hoozemans; Pijnappels; Burger; Verschueren; van Dieën, 2016; Arvin, van Dieën, & Bruijn, 2016; Gill et al., 2001; Goutier et al., 2010; Siragy et al., 2019). This was shown through a variety of tasks, including steady-state gait and obstacle crossing protocols, single-legged and double-legged standing with open and closed eyes (Arvin; Mazaheri; Hoozemans; Pijnappels; Burger; Verschueren; van Dieën, 2016; Arvin, van Dieën, & Bruijn, 2016; Gill et al., 2001; Goutier et al., 2010; Siragy et al., 2019). In these studies, the more challenging tasks, such as standing with closed-eyes and obstacle gait trials led to higher trunk sway measures, which according to the authors were indicative of inferior postural stability.

Furthermore, when using the active arm swing, our participants exhibited larger trunk kinematics variability compared to both the normal and held arm conditions. While the COM displacement normally follows a smooth path along the ML and AP directions when walking (Tesio & Rota, 2019; Winter, 1995), the larger trunk kinematics variability found in our results indicated that actively moving the arms disrupted the expected COM's smooth trajectory. Our results contradicted previous findings by Jurčević Lulić et al., (2008) and Nakakubo et al. (2014), who both suggested that active arm swing strategy improved trunk stability among young adults (Jurčević Lulić et al., 2008) and older adults (Nakakubo et al., 2014). Although the active arm swing led to changes in trunk control in our young healthy adults, this condition was not challenging enough to disrupt the completion of the walking task, nor to cause a fall. However, this could be of importance in populations such as older adults, known to walking with larger trunk angular velocity mean and variability measurements (Goutier et al., 2010) and therefore with a more unstable gait pattern compared to their younger counterparts (Goutier et al., 2010). In this population, the additional disruption to the COM's trajectory could increase the difficulty of the task and lead to falls.

In response to the larger trunk velocities and variability in the active arm swing condition, participants modified the BOS through spatiotemporal adjustments such as an increase in step length and step width variability. This combination could have been used to ensure proper foot placement and to adopt a more cautious gait (Lamoth et al., 2011) in order to maintain adequate levels of stability (Siragy et al., 2019). Interestingly, despite all the kinematic indicators showing decreased postural control when using the active arm swing strategy, no main effect for arms on average MOS were detected, therefore showing no disruption to postural stability during the active arm swing condition. These results suggested that, while active arm swing led to diminished postural control, the spatiotemporal adjustments performed by our participants sufficed to maintain adequate levels of stability (Siragy et al., 2019). However, active arm swing increased MOS variability compared to the other arm conditions, which illustrated poorer control of the trunk and foot placement (Onushko et al., 2019).

Consistent with some of our group's previous works, no significant differences were detected between the normal and held arm swing conditions for all parameters studied (Hill & Nantel, 2019; Siragy et al., 2019). An explanation for this outcome lies within the observations made by Bruijn et al. (2010). When using the held arm strategy, the weight of the arms contributed to the total weight of the trunk thereby increasing trunk inertia. The increased trunk inertia effectively limited the displacement of the COM as well as the trunk linear and angular velocity. Further, the increased trunk inertia reduced the xCOM, resulting in larger, and therefore favourable, MOS values. This could explain the lack of significant differences in step length, width and time between the normal and held arm swing strategies as no further spatiotemporal adaptations would have been required. Alternatively, restraining the arms could have triggered a whole-body compensation mechanism to appropriately respond to the changing environment

(Marigold & Misiaszek, 2009). These compensations included increased activation of the muscles at the trunk and hip levels which could limit the displacement of the COM to maintain levels of stability comparable to those when using a normal arm swing.

5.4.3 Rocky surface

Walking on the rocky surface increased the trunk linear velocity and angular velocity averages in the frontal and transverse planes when compared to walking on a regular surface. Furthermore, the rocky surface led to larger peak values for the trunk linear velocity in the frontal plane and angular velocity in all three planes, indicating poorer postural control (Goutier et al., 2010). Finally, this condition resulted in greater variability in trunk linear and angular velocity in the sagittal, frontal and transverse planes. The oscillations caused by this terrain disrupted the trunk’s intended path (the sinusoidal path described by Winter (1995)) and displaced the COM more than the steady-state condition (Brenière & Do, 1991; Gates et al., 2013; Rankin et al., 2014; Winter, 1987). Normally, a healthy gait possesses relatively small amount of variability (Hausdorff, 2005; Siragy et al., 2019). However, the more variable trunk motion displayed by our participants indicated decreased postural control (Siragy et al., 2019) and most likely explained their more variable foot placement (Brenière & Do, 1991; Rankin et al., 2014; Siragy et al., 2019; Winter, 1987).

Contrary to our hypothesis, walking on the rocky surface led to larger MOS mean and larger MOS SD compared to walking on the regular surface. This contradicted the increase in kinematic data in the frontal plane as a larger MOS is indicative of an improvement in gait stability (Siragy et al., 2019). The larger MOS data also showed that the participants were compensating for the perturbation caused by the surface type by modifying their normal gait pattern to maintain stability. This compensation was shown through our spatiotemporal data as the rocky surface walking led to increased step width. Adopting wider steps would increase the BOS within the frontal plane, leading to an increase in the distance between the xCOM and the edge of the BOS (Rosenblatt & Grabiner, 2010). This allowed for larger displacement of the COM before moving over the limits of the BOS (O'Connor et al., 2012) and therefore improved MOS. Our spatiotemporal results were consistent with findings by Gates et al. (2013) in participants walking the rocky surface as well as with Onushko et al. (2019) who reported similar spatiotemporal adjustments when participants walked on a surface oscillating in the sagittal and frontal planes. However, our participants did not adopt an increase in step length, likely because the other adaptations were enough to respond to the surface and maintain adequate postural control and stability.

Though widening the steps had its benefits, it was also shown to lead to increased trunk velocities and accelerations (Rosenblatt & Grabiner, 2010; Siragy & Nantel, 2018). Therefore, it seemed that the increased step width served to maintain the already present stability levels (Siragy & Nantel, 2018) rather than to improve stability. Furthermore, it was also shown that walking with wider steps led to more variability in foot placement (Perry & Srinivasan, 2017) and less control over the trunk motion among healthy young adults (McAndrew Young & Dingwell, 2012). This foot placement variability was made evident through the increased coefficient of variance of all spatiotemporal parameters studied, which were indicators of decreased gait (Siragy et al., 2019), as well as through the increased MOS SD.

When walking on the rocky surface, participants also reduced their step time. This could have been done to increase the ratio of double support to single support stance in a challenging environment. Voloshina et al. (2013) also reported decreased step time when walking on an uneven surface. However, as gait speed was controlled, the observed reduction in step length caused the decrease in step time (Voloshina et al., 2013). As our participants walked using a self-paced speed, our observations could have stemmed from a reduction in gait speed when walking on the rocky surface compared to the regular surface. Otherwise, this observation could have been due to the participants feeling an increased risk of falling due to the destabilizing terrain (McAndrew et al., 2010).

5.4.4 Interaction

Active arm swing did not counteract the negative effects of walking on a rocky surface; it enhanced the destabilizing effect of the rocky surface on the COM displacement. Rocky surface walking led to larger trunk kinematics and spatiotemporal variability than regular surface despite the arm swing strategy used. The combination of active arm swing and rocky surface led to larger values for all parameters showing a significant interaction effect. Consequently, our results did not support our hypothesis that active arm swing would attenuate the effects of the rocky surface on postural control and stability.

There were two exceptions to these findings. First, step time of the left leg was decreased when walking on rocky compared to regular surface. Though, active arm swing still increased this parameter compared to the other arm swing strategies on both regular and rocky surfaces. Lastly, active arm swing reduced trunk angular velocity in the frontal plane when walking on a rocky surface to similar values when walking on a regular surface, showing an improvement in postural control (Goutier et al., 2010; Siragy et al., 2019).

Since active arm swing was shown to benefit stability in steady-state and perturbed walking (Jurčević Lulić et al., 2008; Nakakubo et al., 2014; Punt et al., 2015; Wu et al., 2016), and our results showed otherwise, these advantages could be situational or based on the perturbation type. Therefore, more research is required to fully comprehend the effect of arm swing when walking.

5.4.5 Limitations

Firstly, participants were secured in a harness throughout the trials. This could have created a sense of confidence and, ultimately, modified the participants' gait pattern. Secondly, the use of a treadmill may not always be representative of over-ground walking as some use a more "cautious gait" when using a treadmill (Yang et al., 2016). Thus, future research should examine these variables during over-ground walking using a protocol similar to Gates and collaboration (Gates et al., 2012; Gates et al., 2013) where their participants walked on a flat surface and on a surface covered with rocks. Finally, the attentional demands associated to modifying arm swing strategies can affect gait stability. Mofateh et al. (2017) showed an improvement in gait stability as shown through reduced gait spatiotemporal variability while Chow et al. (2018) demonstrated that adopting an internal focus was detrimental to motor performance. Therefore, future studies should ensure maintenance of inter-limb coordination when modifying arm swing.

5.5 Conclusion

In summary, active arm swing and rocky surface walking increased the variability of trunk kinematics and their peak values. Our spatiotemporal data showed that our participants responded to the active arm swing and rocky surface conditions with a larger BOS and the adoption of a "cautious" gait to maintain sufficient stability levels. The findings of this study could improve our overall understanding of the role of arm motion on gait stability during normal and challenging environments. These results could have a potential interest for individuals with gait impairments, individuals with an increased risk of falls, or individuals who work on challenging surfaces, such as construction workers.

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CHAPTER 6: GENERAL CONCLUSION

6.1 Conclusion

In conclusion, significant modifications to the walking pattern were required in order to diminish the destabilizing effects of the active arm swing and rocky surface conditions. These independently increased trunk linear and angular velocities variability and peak values while simultaneously increasing spatiotemporal variability. These increases in variability showed that using an active arm swing strategy and walking on a rocky surface decreased both postural control and dynamic stability. Participants adapted by increasing their BOS: they increased their step length in response to the active arm swing and they increased their step width as a reaction to the rocky surface, which assured that the trunk remained within the BOS and maintained adequate levels of stability, as indicated by the larger MOS. Our interactions results showed that using an active arm swing strategy did not attenuate the destabilizing effects of the rocky terrain. Rather, this arm swing strategy enhanced the spatiotemporal and MOS variability. Contrary to our results, some studies have shown that using active arm swing can improve stability in steady-state and perturbed walking. Therefore, it is possible that these advantages are context-dependent; among others, factors such as walking speed (preferred versus. set speed), age and surface type (smooth, uneven, over-ground, treadmill, etc.) contribute to the overall gait pattern. This could explain the difference between our results and those of Jurčevic Lulić et al. (2008), Nakakubo et al. (2014), Punt et al. (2015) and Wu et al., (2016). More studies are required to understand the contribution of active arm swing towards postural control and dynamic stability. Future studies should also consider examining the inter-limb coordination and how this last is modified when changing arm swing strategies.

Our results support a cautious use of traditional gait models such as the inverted pendulum which considers the HAT segment to be a rigid body or models that consider arm motion to be passively achieved. Rather, our results demonstrate the importance of the contribution of arm swing towards postural control and dynamic stability in both steady-state and challenging conditions. Therefore, future research should include the arms and arm motion in their models when investigating gait.

6.2 Limitations

Firstly, participants were secured in a harness throughout the trials. This could have created a sense of confidence and, ultimately, modified the participants' gait pattern. Secondly, the use of a treadmill may not always be representative of over-ground walking as some use a more "cautious gait" when using a treadmill (Yang et al., 2016). Finally, the attentional demands related to changing arm swing strategies from the normal strategy can affect gait stability (Hill & Nantel, 2019). Mofateh et al. (2017) reported that a cognitive task can improve gait stability as shown through reduced gait spatiotemporal variability whereas Chow et al. (2018) demonstrated that motor performance was impaired when adopting an internal focus.

6.3 Significance

In addition to improving our overall understanding of the role of arm motion on gait stability during normal and challenging environments, these results could have a potential interest for individuals with gait impairments and increased risk of falls. Using an active arm swing can enhance their risk of falls through an increase in gait variability, leading to reduced postural control; this observation is more evident on a rocky surface. Instead, it would be more beneficial to employ a normal or held arm swing on a regular and on this challenging terrain. The present findings could also be of interest to those working on challenging surfaces such as construction workers, as on many occasions they must walk on an uneven terrain while carrying equipment, thus limiting their ability to modify their arm swing.

CHAPTER 7: REFERENCES

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CHAPTER 8: APPENDIX

8.1 Additional tables

Table 5

Comparison of (A) arm swing strategies for a fixed surface type and comparison of (B) surface for a fixed arm swing strategy for mean angular velocity in the mediolateral direction

A

Larger values	Smaller value	p-value
Reg-held	Reg-normal	$p>0.05$
Reg-active	Reg-normal	$p<0.05$
Reg-active	Reg-held	$p>0.05$
Rocky-held	Rocky-normal	$p>0.05$
Rocky-active	Rocky-normal	$p>0.05$
Rocky-held	Rocky-active	$p>0.05$

Note: Reg represents the regular surface.

B

Larger values	Smaller value	p-value
Normal-rocky	Normal-reg	$p>0.05$
Held-rocky	Held-reg	$p<0.01$
Active-reg	Active-rocky	$p>0.05$

Table 6

Comparison of (A) arm swing strategies for a fixed surface type and comparison of (B) surface for a fixed arm swing strategy for MOS standard deviation at right heel strike

A

Larger values	Smaller value	p-value
Reg-held	Reg-normal	$p>0.05$
Reg-active	Reg-normal	$p<0.05$
Reg-active	Reg-held	$p>0.05$
Rocky-held	Rocky-normal	$p>0.05$
Rocky-active	Rocky-normal	$p<0.001$
Rocky-active	Rocky-held	$p<0.01$

Note: Reg represents the regular surface.

B

Larger values	Smaller value	p-value
Normal-rocky	Normal-reg	$p<0.001$
Held-rocky	Held-reg	$p<0.001$
Active-rocky	Active-reg	$p<0.001$

Table 7

Comparison of (A) arm swing strategies for a fixed surface type and comparison of (B) surface for a fixed arm swing strategy for mean step time at left heel strike

A

Larger values	Smaller value	p-value
Reg-normal	Reg-held	$p>0.05$
Reg-active	Reg-normal	$p<0.001$
Reg-active	Reg-held	$p<0.001$
Rocky-normal	Rocky-held	$p>0.05$
Rocky-active	Rocky-normal	$p<0.001$
Rocky-active	Rocky-held	$p<0.001$

Note: Reg represents the regular surface.

B

Larger values	Smaller value	p-value
Normal-reg	Norm-rocky	$p<0.001$
Held-reg	Held-rocky	$p<0.001$
Active-reg	Active-rocky	$p<0.01$

Table 8

Comparison of (A) arm swing strategies for a fixed surface type and comparison of (B) surface for a fixed arm swing strategy for COV of step length at right heel strike

A

Larger values	Smaller value	p-value
Reg-normal	Reg-held	p>0.05
Reg-active	Reg-normal	p>0.05
Reg-active	Reg-held	p>0.05
Rocky-held	Rocky-normal	p<0.05
Rocky-normal	Rocky-active	p>0.05
Rocky-held	Rocky-active	p<0.01

Note: Reg represents the regular surface.

B

Larger values	Smaller value	p-value
Normal-rocky	Normal-reg	p<0.001
Held-rocky	Held-reg	p<0.001
Active-rocky	Active-reg	p<0.05

Table 9

Comparison of (A) arm swing strategies for a fixed surface type and comparison of (B) surface for a fixed arm swing strategy for COV of step width at left heel strike

A

Larger values	Smaller value	p-value
Reg-held	Reg-normal	p>0.05
Reg-normal	Reg-active	p>0.05
Reg-held	Reg-active	p>0.05
Rocky-normal	Rocky-held	p>0.05
Rocky-active	Rocky-norm	p<0.001
Rocky-active	Rocky-held	p<0.001

Note: Reg represents the regular surface.

B

Larger values	Smaller value	p-value
Normal-rocky	Normal-reg	p<0.01
Held-rocky	Held-reg	p>0.05
Active-rocky	Active-reg	p<0.001

Table 10

Comparison of (A) arm swing strategies for a fixed surface type and comparison of (B) surface for a fixed arm swing strategy for COV of step width at right heel strike

A

Larger values	Smaller value	p-value
Reg-held	Reg-normal	p>0.05
Reg-active	Reg-normal	p>0.05
Reg-held	Reg-active	p>0.05
Rocky-held	Rocky-normal	p>0.05
Rocky-active	Rocky-norm	p<0.001
Rocky-active	Rocky-held	p<0.001

Note: Reg represents the regular surface.

B

Larger values	Smaller value	p-value
Normal-rocky	Normal-reg	p<0.05
Held-rocky	Held-reg	p<0.05
Active-rocky	Active-reg	p<0.001

Table 11

Walking speeds (m/s) of each participant according to the experimental conditions

Surface condition	Regular	Rocky		
Arm swing condition	A*	Normal	Held	Active
1	1.20	1.20	1.19	1.20
2	1.20	1.61	1.47	1.70
3	1.20	1.23	1.17	1.31
4	1.20	1.31	1.10	1.23
5	1.20	1.26	1.34	1.40
6	1.20	1.07	1.15	1.14
7	1.20	1.25	1.34	1.47
8	1.20	1.21	1.04	1.43
9	1.20	1.28	1.21	1.27
10	1.20	1.56	1.27	1.37
11	1.20	1.18	1.13	1.17
12	1.20	1.08	1.17	1.43
13	1.20	1.64	1.07	1.58
14	1.20	1.54	1.31	1.62
15	1.20	1.66	1.27	1.72

Note: A* represents all of the arm condition; the regular surface walking trials were completed using a fixed-speed of 1.2m/s across all arm conditions.