Dynamic Stability and Variability of Perturbed Walking in Young Adults

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Abstract

Falls are the third major cause of inadvertent injury in young Australian adults aged between 18-35 years. The inability of an individual to respond to external perturbations due to walking on an inclined surface, or internal perturbations such as dual-task walking, are known to be associated with significantly higher risk of balance loss. Significant factors known to increase risk of balance loss during walking include performing an additional task requiring high motor-cognitive, sensory or cognitive load (internal perturbations), and walking on uneven surfaces such as sloped terrain (external perturbations). At present, however, dynamic balance of the entire body and the risk of balance loss during walking under such perturbations is not well understood. The objective of this study was to investigate dynamic stability and variability of the human body during walking, and assess the influence of external, motor-cognitive, sensory and cognitive perturbations on dynamic balance, including surface inclination, use of a cell phone, auditory and visual stimulation, and mental calculation.

Nineteen healthy young adult males were recruited. Three-dimensional joint kinematics were obtained using an optical motion capturing system as subjects walked at their self-selected speed on an instrumented treadmill. Dual-tasking was simulated by subjecting participants to motor-cognitive, visual, cognitive and auditory perturbations during walking including cell phone usage (talking, texting and reading), watching a video clip, listening to music, and performing numeric calculations mentally. External perturbations were also applied through alteration of surface inclination. Variability analysis was performed on spatiotemporal gait parameters using Detrended Fluctuation Analysis (DFA) and Standard Deviation. Dynamic stability was subsequently estimated for the entire body as well as the head, trunk and lower extremity joints using linear and nonlinear measures including Margin of Stability (MoS), Lyapunov Exponent (LyE) and Maximum Floquet Multipliers (MaxFM). A novel method was

devised to assess stability using Margin of Stability at heel contact (HC) and minimum foot clearance (MFC), gait events associated with backward and forward balance loss, respectively. Slip and trip propensity estimated using Required Coefficient of Friction (RCoF) and MFC height, respectively. Finally, the most destabilizing additional task while walking was determined using deviation of MoS and trip propensity values during dual-task trials from the corresponding values during baseline walking.

The results showed that dual-tasking during walking adversely affects balance in a direction specific-manner. Specifically, cell phone texting and reading while walking reduces balance in the mediolateral direction, while cell phone talking increases the risk of tripping in the anteroposterior direction. Upslope terrain increased the risk of balance loss in the anteroposterior and vertical directions and did not affect gait balance in the mediolateral direction, while walking down was associated with greater stability in the anteroposterior direction. Cognitive and sensory perturbations affected gait balance mostly in the anteroposterior and vertical directions rather than the mediolateral direction. Analysis of trip propensity showed that motor-cognitive dual-tasking due to cell phone usage while walking, cognitive and sensory perturbations due to performing additional auditory and visual tasks while walking are associated with greater risk of tripping, as measured by a lower MFC height. Particularly, talking while walking, and cognitive and sensory dual-tasking while walking may ultimately lead to an increase in risk of tripping in young adults. However, the risk of tripping in young individuals is not sensitive to external perturbations caused by sloped terrains. Participants mostly changed their step length and step time during walking under perturbations. Among the various measures used to determine the most destabilizing secondary task while walking, MFC height was more sensitive to the applied perturbations. Talking while walking was associated with the largest deviation from baseline condition.

The findings of this investigation confirmed that head stabilization during ambulation has higher priority compared to other segments, and individuals try to adopt different strategies to attenuate perturbations from the lower body to the head. The current data highlighted the importance of arm swing in balance maintenance during walking under perturbations, and demonstrated that individuals try to compensate restricted arm swing during walking by modulating step width. With respect to gait adaptations, the results of this research support the idea that individual's response to applied perturbations through dual-tasking while walking depend on the magnitude of the applied perturbation. The evidence from this study suggests that talking while walking is the most challenging secondary task during locomotion among the applied perturbations in this study, and additional sensory tasks are the least challenging one.

These findings have significant implications for development of a gait training protocol for more frail people to successfully address common perturbations arising from daily living activities during everyday life. These data also suggest that MFC height analysis and local stability analysis of the lower body should be performed to gain better understanding on the effect of additional concurrent task while walking on the risk of tripping and gait stability, respectively. The analysis of MoS presented extends knowledge of step-to-step balance changes during walking at different gait events associated with common fall patterns occurring at HC and MFC. Further work needs to be done to analyse the influence of similar attention-demanding secondary tasks or 'distractions' in more vulnerable populations, including the elderly, fallers, and individuals with sensory or motor impairments that affect locomotor control.

To my beloved wife and my parents without whom this journey would* not *have been possible.

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List of Abbreviations

MoS	Margin of Stability
LyE	Maximum Lyapunov Exponent
MaxFM	Maximum Floquet Multiplier
MSE	Mutiscale Entropy
SampEn	Sample Entropy
ApEn	Approximate Entropy
DFA	Detrended Fluctuation Analysis
STD	Standard Deviation
BoS	Base of Support
CoM	Centre of Mass
HC	Heel Contact
MFC	Minimum Foot Clearance
RCoF	Required Coefficient of Friction

List of Publications

Shahidian H., Nagano H., Begg R., Ackland D., Dynamic Stability of the Upper Body during Walking on Sloped Surfaces, *The 22nd Australian & New Zealand Orthopaedic Research Society (ANZORS) Conference*, Melbourne, 13-15th October 2016.

Shahidian H., Nagano H., Begg R., Ackland D., Stability of Human Walking on Sloped Surfaces, *The 10th Australian and New Zealand Society of Biomechanics (ABC) Conference*, Melbourne, 4-6th December 2016.

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1 Introduction

Walking is the most common physical activity during daily life. According to the Australian Institute of Health and Welfare, walking is the primary physical activity for more than 40% of Australians aged 18-64 years [1]. It has been suggested that taking 10,000 steps per day is associated with some advantages to body health including reduced body fat [2] and low blood pressure [3]. Among Australians, individuals aged 33-44 are the only group likely to reach the threshold [4]. Young males walk more compared to their female counterparts, while elderly females aged above 75 years old walk more than males from the same age group [4]. Surveys conducted by Australian Bureau of Statistics (ABS) have shown that healthy young adults aged 5-17 years took nearly 2000 steps more per day compared to adults aged 18-64 years [4].

Drury and Wolley (1995) described walking as a…. "series of controlled falls requiring a repeated pattern of losing and regaining a balanced stance" (p. 714) [5]. According to this definition, body progression during walking is linked to balance maintenance and controlling falls during ambulation. To preserve stability and maintain balance during walking, it has been shown that the Centre of Mass (CoM) should be located within border of Base of Support (BoS) formed by the feet [6]. At the end of a gait cycle, the CoM leaves the BoS area for propulsion; so, a new BoS area is created by the step taken for body progression, called the leading foot, and the trailing limb. Balance is maintained provided the CoM moves inside the new BoS area during each gait cycle. Thus far, several studies confirmed the effectiveness of the concept of BoS-CoM distance in balance assessment during double support phase. However, this approach is limited in balance measurement during other phases of gait e.g. single support phase. In fact, during single support phase, one leg would be off the ground in swing phase and the shape of BoS is changed by progression of the swing limb. In this project, the concept of

BoS-CoM is extended further, and an alternative approach is proposed to create BoS area and measure balance during the single support phase.

During daily life, people walk under various perturbations including internal and external perturbations. Using cell phone, listen to music, thinking and talking are only some examples of additional tasks which apply internal perturbations to human walking. To date, several studies have linked increased cognitive cost during dual-task walking with gait changes in young adults [7-11]. The additional task could be a motor task such as carrying a glass of water, a cognitive task such as counting down by three [12], or a combination of them such as cell phone usage while walking. In either case, the ambulation might be associated with compromised safety if both concurrent tasks share the same cognitive resources, according to the capacity-sharing model [10, 13]. To put it another way, if the cognitive load during simultaneous execution of both tasks is greater than that during performing a single task, performance would be reduced in either primary (e.g. walking) or secondary (e.g. cognitive) task [14]. The level of deterioration of performance depends on the nature and complexity of the additional task [15].

Everyday locomotion is not always over even and firm surfaces. Sloped, grassy, muddy, sandy and rocky surfaces are typical diverse terrains that apply external perturbations to gait. Among them, walking over hill type footpath and ramps are more common in urban environments. Data from several studies showed that walking over inclined surfaces is associated with gait changes in healthy adults [16-18]. During walking, information about position of the body relative to the walking environment is delivered to the brain through proprioceptive sensors. Perturbations associated with challenging environment e.g. upslope and downslope walking surfaces, can affect function of proprioceptive sensors and result in poor detection of surfaces demands and slower counteracting strategy to reduce risk of balance loss [19]. Depending on the magnitude of induced perturbation by the terrain, the risk of balance loss is varied.

A fall is defined as an unintentional descent of the body to the ground, floor or level of movement [20], and individuals unable to recover balance during walking are at risk of a fall. Falls primarily occur during walking, with higher prevalence of slip-induced falls in young adults [21]. They occur due to inability of an individual to respond to applied either external perturbations, which may arise, for instance, due to walking on an irregular surface (e.g. loose rocks) [22], or internal perturbations which may originate from dual-tasking while walking. Depending on the extent of the applied perturbation and neuro-musculoskeletal capacity of individuals, internal and/or external perturbations can affect function of the cognitive and/or sensory system and may ultimately increase the risk of balance loss [23]. Here we need to introduce two important definitions in the context of human movement: risk of falling and stability. While these two are being used interchangeably in some studies, they are neither similar, nor unrelated. In fact, both concepts provide us with different information about human walking. Risk of falling indicates the propensity of experiencing a fall or losing balance during locomotion, while stability evaluates the capacity of a system in responding to perturbations which make individual fall during movement. Stability of human walking, as an example of a dynamic system, can be examined in two ways: global stability analysis and local stability analysis. Dingwell et al. (2006) defined local stability as the ability of the system to recover from infinitesimally small perturbations that occur naturally during gait such as uneven surfaces, and global stability as the capability of the system to recover from large perturbations such as tripping or slipping [24]. In a recent study by Bruijn et al. (2013), stable gait of a passive dynamic walker was characterized by ability to recover from infinitesimal and large perturbations, given that the magnitude of perturbation that the subject encountered is not larger than the greatest recoverable perturbation of the walker [25].

How do applied perturbations affect likelihood of balance loss and dynamic stability during walking? How do individuals respond to applied perturbations and maintain balance? Which type of secondary tasks generate larger perturbation to gait patterns in young adults? This dissertation was conducted to answer these questions and enhance our understanding of the influence of internal and external perturbations arising from daily living activities on gait stability and the risk of balance loss in young individuals using a wide variety of linear and nonlinear techniques. The primary focus was on young healthy individuals to create a baseline level to evaluate gait changes under the same perturbations in more frail people in later studies. Moreover, the findings of this study may provide valuable data for future studies to develop gait training protocols to prevent falls in individuals who are more susceptible to fall e.g. patients and elderly people.

1.1 Dissertation Aims and Research Questions

This purpose of this dissertation was to investigate stability and variability of human walking in response to sensory and cognitive perturbations arising from daily living activities. The specific aims and hypotheses are as follows:

1.1.1 Aim 1: Determine how cell phone talking, reading and texting during walking influence variability and dynamic stability of human walking (see Chapter 3)

Cognitive load associated with dual-tasking while walking is known to affect balance and increase fall risk. One of the most common activities of daily living involving motor-cognitive dual-tasking is cell phone usage while walking, which is associated with visual distraction and cognitive load. However, little is known about the independent effects of cell phone reading, texting and talking as a secondary task on stability during walking, nor their influence on the trunk, lower limb joints and head segment, which plays an important role in gaze stabilisation and postural control. In Aim 1, experiments will be performed where individuals walk while engaging in a conversation, reading and texting a message on a cell phone. The first research question addressed in this chapter was how cell phone talking, reading and texting while walking influenced the risk of balance loss and stability of walking. The second research question addressed was how different body segments responded to perturbations arising during cell phone usage while walking. The third research question focused on the most destabilizing type of cell phone usage during walking.

The specific hypotheses are:

Hypothesis 1A: Cell phone usage while walking affects gait pattern and results in shorter step length, wider step width, shorter step duration and double support time.

Hypothesis 1B: Talking, texting or reading on a cell phone during walking decreases dynamic stability, increases gait variability and does not affect the risk of slipping and tripping.

Hypothesis 1C: Texting and reading on a cell phone during walking is more destabilizing compared to talking on a cell phone while walking.

1.1.2 Aim 2: Determine how surface inclination influences variability and dynamic stability of human walking (see Chapter 4).

Walking on an inclined surface is known to contribute to more physically demanding ambulation and increased fall risk relative to level-gradient walking; however, the influence of ground inclination on dynamic stability of the body during walking is not well understood. Additionally, little is known about the effect of surface inclination on stability of the head segment and gaze stability, which is necessary for postural control. In Aim 2, experiments will be performed where individuals will walk over inclined surfaces, and dynamic stability and variability will be quantified in these cases. The first research question addressed in this chapter was how dynamic stability and variability were affected during walking over inclined and declined surfaces. The second research question was how different body segments, especially the head responded to perturbations associated with sloped terrain. The specific hypotheses are:

Hypothesis 2A: Downhill walking is associated with shorter step length, step time and double support time compared to level-ground walking

Hypothesis 2B: Participants walk with longer step length, step time and double support time during uphill walking relative to level-ground walking.

Hypothesis 2C: Step width is not affected during walking on sloped surfaces.

Hypothesis 2D: Gait variability and risk of tripping is higher during level-ground walking compared to sloped walking conditions.

Hypothesis 2E: Stability of the head and trunk segments is more affected during downhill walking compared to uphill conditions, and in the anteroposterior and vertical directions rather than mediolateral direction.

1.1.3 Aim 3: Determine how dual-tasking due to performing an additional cognitive, auditory and visual task while walking influences variability and dynamic stability of human walking (see Chapter 5).

It has been shown that dual-tasking during locomotion can result in altered gait profiles characterized by slower walking and wider steps due to diversion of attention. If additional tasks undertaken during walking result in increased sensory and cognitive demand compared to walking alone, balance maintenance may ultimately be more challenging, and fall risk increased. At present, however, little is known about the influence of cognitive and sensory-based dual-tasking on gait balance. In Aim 3, experiments will be performed where individuals walk while listening to a piece of music, watching a video clip, or under cognitive load. The first research question addressed in this chapter was how cognitive, auditory and visual additional task influenced dynamic stability and variability in young individuals. The second research question addressed was how different body segments responded to perturbations arising during cell phone usage while walking. The third research question addressed was what type of concurrent additional task during walking destabilized young adults' gait the most. The specific hypotheses are:

Hypothesis 3A: Sensory and cognitive dual-tasking affects gait pattern and results in shorter step length, wider step width, shorter step duration and double support time.

Hypothesis 3B: Auditory, visual and cognitive secondary tasks during walking decrease dynamic stability, increase gait variability and the risk of slipping and tripping.

Hypothesis 3C: Cognitive secondary tasks during walking are more destabilizing compared to those of sensory secondary tasks.

2 Literature Review

2.1 Falls and its Associated Costs

Falls and their associated injuries are the secondary cause of inadvertent fatal injuries among older adults worldwide [26] after transport accidents, and the primary reason of non-fatal injuries [27]. In young individuals aged between 18-35 years, falls are also identified as the third major cause of inadvertent injury [8]. Elderly people are more susceptible to falls than younger individuals due to poor balance, muscle weaknesses, cognitive deficits, slow reaction times, and reduction in somatosensory acuity [28]. The proportion of female fallers is higher than the proportion of male fallers across different age groups [29]. The financial costs of falling in Australia, including hospitalization fees, medication, rehabilitation, is estimated at AUD$648 million annually [29]. On individual basis, the average costs of hospitalization due to falls was estimated to be about AUD$8000 per episode in Queensland during 2007-8 [30]. In Western Australia, the cost of hospitalization due to falls between 2000 and 2008 was about AUD$618 million [31]. However, these results were based upon data from the older adults and it is unclear how much would be the associated financial costs of falls in young individuals.

Falls are associated with different types of injuries ranging from minor bruises to severe soft tissue injuries [32]. Fractures are among the most common fall injuries [33]; however, the type of injury resulting from a fall depends on the orientation of the fall. Backward and sideway balance loss has been shown to be associated with greater risk of hip fracture [34, 35], while wrist fracture is most common during falls on an outstretched hand in forward or backward direction [36]. Head injuries due to falls are also prevalent in elderly individuals. In a study by Oxley et al. (2018), it has been demonstrated that head, face and neck injuries have the maximum rate of hospital admission for fall related injuries in older adults [33]. Owing to adoption of protective strategies such as utilization of arms and trunk flexion, head injuries due to falls rarely occur in young individuals [37]. Compared to elderly cohort, in a recent study conducted by Heijnen and Rietdyk (2016), it has been shown that contusions are the most frequent injury type in young adults [21].

While falls occur both indoors and outdoors, the majority of falls occur outdoors among most age groups [38]. In a recent study by Timsina et al. (2017), it has been demonstrated that outdoor falls are more prevalent in young adults, while indoor falls are more common in older women [39]. Previous research has established that walking is the most predominant activity which is associated with fall in young and elderly individuals [21, 33, 39]. Tripping or forward balance loss is more prevalent in older adults [40], while young individuals mostly slip or lose balance in backward direction [21]. Considering gender differences, tripping is also more common in women compared to men in young and old individuals [39]. It can therefore be assumed that a general fall prevention approach cannot be applicable for individuals from different gender and age group.

2.2 Fall Risk Assessment of Human Gait

Falls are a major global health issue. To reduce the number of falls through diagnosis of the likelihood of balance loss, detect the underlying causes of falls and provide individuals with a falls' prevention care plan, falling risk assessment is commonly performed. Generally, falling risk assessment can be divided into qualitative and quantitative approaches. Questionnaires have been used in clinical settings as the primary qualitative falling risk assessment tool. Individuals have been asked to answer several questions about their fall history, cognitive status, psychological conditions, medications and falling risk factors. Each question has a specific score and general score would be calculated for each participant. Falling risk assessment would be performed based on the calculated general score. While this tool has been widely used in hospitals around the world, it has been suggested that questionnaires could be associated with bias by history of falls [41]. Another example is Activities-Specific Balance Confidence Scale (ABC) which was introduced by Powell and Meyers (1997) to measure level of confidence in balance maintenance during performing daily living activities [42].

In addition to falling risk assessment tools, some quantitative tools have received attention during the last two decades. One of the most reliable and simplest quantitative tools is the Timed Up and Go (TUG) test [43]. Standing up from a chair, walking for 3 meters, turning around and walking back are four steps of the TUG test [44]. Assessment would be based on the spent time to perform this sequence of tasks. The main reason behind wide usage of the TUG test in clinical settings is its simplicity which has made it easy to run in a clinic [45]. Previous research has established that the TUG test can predict falling risks in older adults [46-49]. However, there has been no evidence that the TUG test can predict the likelihood of balance loss in young individuals aged between 20-40 years old [50]. One disadvantage of the TUG test is that its limited capability in differentiation between affected subcomponents of balance and gait [51].

Another example of quantitative falling risk assessment tool is Functional reach test (FRT) [52]. This test was developed by Duncan et al. (1990) to determine maximum intentional displacement while the participant tries to reach forward beyond arm’s length during standing without losing balance [52]. According to reported functional reach norms for individuals with different genders and age groups, the older people get, the less reach distance they have [52]. In addition, reach distance is limited in females compared to males in different age groups, suggesting higher risk of balance loss in females. It has been demonstrated that forward FRT is a reliable predictor of falling risks in individuals suffering from Parkinson’s Disease [53], and vestibular dysfunction [54]. However, compensatory mechanisms such as movement of trunk during the test limits its application in balance measurement in elderly individuals [55]. Adoption of different reach strategies by individuals performing FRT has convinced researchers to evaluate reach assessment and reach distance to get further information about falling risks assessment [56]. Recently, Maranesi and his colleague (2013) concluded that assessment of kinematic behaviour in addition to the FRT might be a useful tool in early fall detection [57].

While the above-mentioned examples are easy to perform, affordable, quick and have also been highlighted as reliable predictors of falling risks, their results are subjective, and level of deterioration or progress in balance maintenance cannot be measured using such approaches. On top of that, these assessment tools are unable to provide researchers and clinicians with underlying causes of high/low risk of falling in an individual. Technological advancement and enhancement of collaboration between clinicians and engineers in recent years, hold promise for development of alternative falling risk assessment tools which contribute to fall detection and fall prevention. Gait analysis, variability analysis using signal processing and wearable inertial sensors are some examples of approaches that have gained popularity in the last two decades. The following sub sections discuss these approaches and their application in falling risk assessment.

2.2.1 Gait Analysis

Data from several studies suggested that ambulation is the most common activity associated with falls [21, 33, 39]. Analysis of gait, therefore, can answer multiple questions about falls. Data can be captured through marker-based systems e.g. Vicon, Optotrak or maker-less systems e.g. GAITRite, Force Platforms. Typically, gait analysis has been performed based on three different perspectives [15]. The first one focuses on sequence of events during ambulation in which movement of each limb is divided to stance phase, when it is in contact with the walking surface and accepting body weight, and swing phase, when it is off the walking surface and moving forward. The second view identifies functionality of different events e.g. heel contact, toe off, minimum foot clearance etc. during a gait cycle. The third view aims spatiotemporal parameters during locomotion. These approaches improved our understanding with regards human walking in normal and pathological cases, contributed to identification of gait problems which may lead to fall in future [15].

Calculation of spatiotemporal variables is the most common approach in assessment of the risk of falling. Spatiotemporal variables represent temporal and spatial characteristics of a dynamic system. In human walking, spatial parameters are represented by step/stride length and step width, while step/stride time, double support time, stance time, swing time, cadence and speed represent temporal parameters. These gait variables have been used in understanding how individuals respond to the applied internal [58] and external [59] perturbations during walking. Walking with narrow steps has been shown to be associated with increased risk of falling in older adults [60, 61], since it results in smaller base of support in the frontal plane and makes controlling of CoM movement more challenging in the mediolateral direction. A number of studies have found that walking with wide steps reduces likelihood of falling in young [62], elderly [62, 63] and post stroke individuals [64, 65]. This strategy is frequently adopted by elderly and post-stroke patients to compensate for hip abductor weakness [66-68]; however, Perry et al. (2017) failed to observe significant changes in stability during walking with wider steps in young individuals [69]. More recently, it has been concluded that humans prefer to ambulate with the most metabolically optimal step width [70]. However, this conclusion has been updated in a recent review by Bruijn and Van Dieen (2018), suggesting that individuals walk with an average step width that corresponds to limited / minimized energy cost of ambulation [71], implying that foot placement in the mediolateral direction is actively coordinated with CoM movement.

Data from two studies have shown that walking with shorter step length is associated with increased risk of falling in the elderly [72, 73]. Increased step length has also been demonstrated as a strategy for increasing stability and preventing forward falls by increasing the base of support in young [74] and older adults [75-77]. It has been shown that having larger step length contributes to longer balance recovery time and larger anteroposterior force impulse during the landing phase in young healthy adults [78]. Conversely, a shortened step length during walking in young individuals has been linked to increased backward margin of stability [79]. Cham et al. (2002) also demonstrated that a shorter step length reduces the probability of falling in young adults during walking on a slippery surface, as this reduces the shear component of the ground reaction force [80]. Smaller distance between CoM and backward border of base of support in sagittal plane has been indicated as the major reason behind the positive effect of reduced step length on the risk of balance loss in young adults during walking on slippery surfaces [81]. It should be noted, however, that altering step length may not the main mechanism of balance maintenance and control in the anteroposterior direction, and CoP adjustment using the ankle may have a more prominent role [82].

Slower steps have been reported as another destabilizing factor during human walking in young and elderly individuals [83, 84]. Bruijn and his colleagues (2009) observed improved local stability in the mediolateral direction during fast walking, while they indicated the effect of gait speed changes on dynamic stability is direction-specific in young adults [85]. Hak et al. (2013) also suggested that increased walking pace in healthy young individuals can improve backward margin of stability [79]. In a recent comprehensive study which has been published in Nature, Buurke and his co-workers (2019) found that fast walking improves margin of stability in the mediolateral direction [86]. In a study conducted by England et al. (2007), however, it has been demonstrated that slow walking is associated with greater local stability in healthy young adults [87]. Schablowski et al. (2006) concluded that as a result of increases in instability during fast walking of healthy young individuals, individuals change their ambulation from walking to running [88]. Likewise, Kang and Dingwell (2008) held the view that fast walking decreases stability in young and elderly individuals [89]. Although extensive research has been carried out on the effect of gait speed on local and orbital walking stability, the generalisability of much published research in this area is problematic. For instance, Granata and Lockhart (2008) reported no significant difference in orbital stability between fast walking and walking at preferred speed in healthy and fall prone adults [90]. In another study, Hak et al. (2010) found no evidence that gait speed changes have an influence on local stability of healthy individuals during walking [79]. Debate on orbital stability and local stability results during walking suggest that the effect of gait speed on local and orbital stability is not fully understood and requires further research.

2.2.2 Variability Analysis

There is evidence to suggest that variability of spatiotemporal gait parameters is also associated with falling risk in older adults [91-94]. In the context of human movement, variability is a built-in property of all biological systems, detected from multiple repetitions of a task, for example, dissimilar footprints while repeatedly walking the same path on snow [95]. Consistent with this idea, Stergiou and Decker (2011) described movement variability as variations in motor performance while repeating a task [96] and in Generalized Motor Program Theory, movement variability is viewed as random or independent noise which contributes to variation in motor performance [97]. Similarly, from a Dynamical Systems Theory perspective, movement variability indicates reduced dynamic stability [98-100]. The aforementioned studies recognize variability as a destabilizing factor in motor performance [96]. This concept has, however, been challenged in studies demonstrating that increased variability is not positively correlated with reduced stability [101-103], for example, a skilled person’s variable responses to maintain stability on a wobbling pad or balancing a soccer ball on one foot. Similarly, when maintaining balance on a compliant surface, variation of centre of pressure is increased compared to standing on a firm surface. Granata and England (2007), for example, suggested that variability reflects the alibility of a system to function effectively under various environmental and task constraints [104]. Moreover, they argued that both reduced and increased variability could be indicators of pathological states. Rather than taking a position on the functional status of movement variability, Bruijn et al. (2013) argued that higher variability of a complex dynamic system like human gait with multiple degrees of freedom does not necessarily reflect instability of the systems since the measured variability is a function of deterministic dynamic of the system [25]. They concluded that gait variability has a complex relationship with stability which requires an understanding of the control strategy of the nervous system to interpret [25].

According to the Generalized Motor Program Theory, human gait variability could be measured using Standard deviation (STD) of time series [105-108]. The STD is calculated over entire gait cycles for discrete variables e.g. step length and time. For continuous variables such as joint kinematics, at first, time series data ought to be divided into gait strides, and each gait stride should be time normalized, for example, from 0 to 100% of one gait cycle. The STD is computed at each time point of the gait cycle across all strides, and then summed or averaged over all obtained variability values of the stride for further investigation [24, 25, 109]. Despite the simplicity of this technique, it suffers from several major drawbacks. A number of studies have reported that linear methods confound our perception of the underlying characteristics of movement variability [95, 110, 111]. In other words, the structure of movement variability and the temporal emergence of motor behaviour cannot be discerned using these methods [95, 112]. In the context of human movement, it has been shown that variability analysis using STD resulted in a belief among scientists that everything away from the average or mean of a data set is error [96]. From statistical point of view, variability analysis using conventional linear method relies on the assumption that all variations around the mean value are random, while it has been indicated in several studies that such variations have deterministic origin [95, 112, 113]. Hausdorff et al. (1999) argued that trends in stride time data, due to for instance changes of speed, have an adverse effect on STD and may give rise to a false estimation of variability [114]. From human movement point of view, it has been suggested that variability analysis using STD ought to be performed on at least 50 gait strides data, which has been shown on an 8-walk protocol to be more accurate and reliable [115].

Much uncertainty still exists about the applicability of the STD measure in quantification of data variability. A study by Stergiou et al. (2006) showed that linear evaluation of variability using STD, because of averaging out of dynamic data by normalization, conceals systematic variations of time series' patterns in time [111]. More recent studies offer contradictory findings about the deterministic origin of such variation that differs from random nature [96, 110-112, 116-119]. Such research has drawn attention to the significant contrast between the results of variability analysis using linear and nonlinear approaches [95, 110, 111].

Nonlinear methods in quantification of variability are based on mathematical concepts that underpin chaos and time series analysis, and have been used to assess temporal evolution of variables during a human movement task [96, 111]. Detrended Fluctuation Analysis (DFA) is one of the most practical ways of revealing information about long range correlation in a data set of discrete events e.g. stride time [114, 118-126]. In order to quantify DFA for a time series, first the cumulative sum of the deviation from the mean at each t is calculated using

$$y(i) = \sum_{t=1}^{i} [x(t) - \bar{x}] \qquad for\ i = 1,2,3, \ldots, N$$ Equation 2-1

where N is the length of time series. Then, $y(i)$ is separated into n non-overlapping windows of the same size (s). As N may or may not be a multiple of s there may be a small residual after dividing $y(i)$ to n congruent windows. In order to avoid this, the dividing process is repeated one more time from another side. Therefore, the number of intervals will be $2n$. In the next step, the local trend (y_n) at each interval is calculated using least square fitting of the data. The fluctuation function can be computed from

$$F(n) = \sqrt{\frac{1}{N}\sum_{i=1}^{N}[y(i) - y(i)_n]^2} \qquad \text{Equation 2-2}$$

where $y(i)_n$ is the local trend at each interval.

The slope of the graph of $F(n)$ as a function of n in log-log scale is called the scaling component or self-similarity exponent α. The scaling exponent is the indicator of long-range correlation. On the basis of different quantities of α, the correlation can be characterized. Specifically, $\alpha = 0.5$ when the time series is random or uncorrelated and behaves like Gaussian (white) noise. If $\alpha = 1$, the time series will behave like pink noise, or its power spectral density has an inverse relationship with frequency. The time series may be estimated as Brownian noise provided that $\alpha = 1.5$. As long as the alteration of α is limited between 0.5 and 1, the time series will have a long range correlation [126, 127].

Preliminary studies on DFA in gait analysis by Hausdorff (1995) showed that the stride interval time series of healthy young adults has a long range correlation and it is not random [119, 121]. Moreover, Hausdorff (1999) compared the fractal index in different age groups and concluded that an individual in the early and late stages of life has greater fractal index [114]. However, physiologic interpretation of DFA is still debated among researchers. In other words, it is not clear whether having long range correlated gait reflects a healthy or a pathologic case. An example of this is the study carried out by Dingwell et al. (2007) in which it has been determined that sensory loss does not affect long range correlation of stride intervals in individuals with peripheral neuropathy [124]. In another study, elderly fallers with a higher level of gait disorder showed low fractal index during walking compared to healthy older adults [128]. The uncertainty in interpretation of DFA has also been exemplified in another study by

Hausdorff (2009) which found white noise properties in gait of patients with Parkinson's Disease [129]. In a recent study, Dingwell and Cusumano (2010) suggested that interpretation of the fractal index should be within the context of undertaken control processes, as well as available redundancies to the system [130].

Entropy analysis is another well-established nonlinear variability modelling approach. This technique aims to quantify the probability of obtaining information about a configuration of an element from another in a given time series. It has been suggested that smaller entropy indicates more predictability, less complexity and less adaptivity in a time series [131], while larger entropy implies less predictability, and greater randomness in a time series. Three different approaches have been proposed for analysing entropy: Approximate Entropy (ApEn), Sample Entropy (SampEn) and Multiscale Entropy (MSE). ApEn is the most commonly used method and was introduced in 1991 by Pincus to analyse medical data sets [132]. Stergiou et al. (2003) reported that aging leads to less certainty in the neuromuscular system to choose proper velocity and configuration during gait [133]. Higher ApEn of MFC values has also been observed in elderly fallers compared to that of healthy older adults [134]. Compared to a control group, individuals with ACL-reconstruction have exhibited increased ApEn of knee joint flexion/extension angle during walking, indicating higher susceptibility of injury to the reconstructed knee [135]. Makikallio et al. (1996) concluded that ApEn reflects the degree of similarity of each sequence of a timeseries, however, it cannot provide any information about overall randomness of the time series [136].

Recent evidence suggests that Sample Entropy (SampEn) is a more appropriate choice for regularity analysis since it corrects the above-mentioned associated error with ApEn computation [131, 137-139]. If X_N is a time series which is made of S_i vectors of similar length (m), the probability of matching of a two vectors $(x(i), x(j)$ where $distance[x(i), x(j)] < r$, $(0.1 * STD(X_N) < r < 0.25 * STD(X_N)))$ can be calculated as

$$S_m = (N-m)^{-1} \sum_{i=1}^{N-m} S_m(i) \qquad \text{Equation 2-3}$$

where $S_m(i)$ is

$$S_m(i) = \frac{S_i}{N-m} \qquad \text{Equation 2-4}$$

By repeating the same procedure for $m+1$, T_m can be derived. SampEn can be measured as

$$\text{SampEn} = \lim_{N\to\infty} \ln\left(\frac{S_m}{T_m}\right) \ or \ \ln\left(\frac{S_m}{T_m}\right) \qquad \text{Equation 2-5}$$

The latter formula is for finite N and may be derived statistically [131, 139, 140]. This approach was developed by Richaman et al. (2000) for entropy analysis of physiologic datasets [139]. Despite higher accuracy compared to ApEn, there have been only few studies in the field of gait analysis that have utilized SampEn for complexity or variability analysis of gait under internal and/or external perturbations. Increased SampEn has been reported for the trunk [141-143], shank and centre of pressure motion [141, 144] during walking while performing an additional cognitive task in young and older adults, suggesting more complexity of gait under cognitive dual-tasking condition [143]. In a recent study, however, it has been shown that dual-tasking due to talking on a cell phone during walking in young individuals results in increased regularity of trunk motion [145]. While some research has been carried out on the effect of cognitive dual-tasking on gait regularity using SampEn, there is still very little scientific understanding of gait variability under sensory dual-tasking as well as external perturbations.

While variability measures have been shown to be reliable indicators of falling risks, they cannot tell us about the possible causes of high or low falling risks, nor about the response of human body to internal and/or external perturbations during walking. Having information about individuals' response to different types of perturbations and the reason behind increased/decreased likelihood of balance loss provides a useful account of how to reduce falling risks in different populations.

2.3 Common Causes of Balance Loss

Several lines of evidence suggest that tripping and slipping are the most common types of balance loss. In this section, these causes of balance loss will be described in detail.

2.3.1 Tripping

Tripping maybe defined as an event within the gait cycle in which the most distal surface of the swing foot, collides an object or walking surface. At the time of contact, the force of the swing foot due to impact results in forward angular momentum which tends to destabilize body in a forward direction. If the angular momentum is not arrested by adoption of a proper recovery strategy, tripping may result in fall [146]. One key strategy to reduce angular momentum during tripping is positioning of the contralateral foot anterior to the body CoM [147]. This contributes to generation of adequate force and moment by the contralateral limb to counteract applied angular momentum during tripping [148]. Pijnappels et al. (2004) showed that required time and clearance for proper positioning of the recovery limb after tripping are provided by strong pushing off of the support limb in young adults [146]. In another study by the same author, arm swing has also been highlighted as an important strategy to prevent falls after tripping in similar cohort [149]. Prolongation of arm swing at the initial phase of tripping results in reduction in amount of transferred momentum from arms to trunk and legs which facilitates further axial rotation of trunk and legs toward the contralateral side. Such rotation in the transverse plane contributes to larger length of recovery step which is important for positioning of the recovery limb and falls prevention [150].

Data from several studies suggest that tripping usually occurs at mid-swing (Figure 2-1). Low height of swing foot respect to walking surface (1-2 cm), and high foot velocity compared to the walking speed (about three times) have been linked to higher chance of tripping at mid-swing [151-154]. Since the foot clearance has it minimum value at mid-swing, this event is most often called minimum foot clearance (MFC).

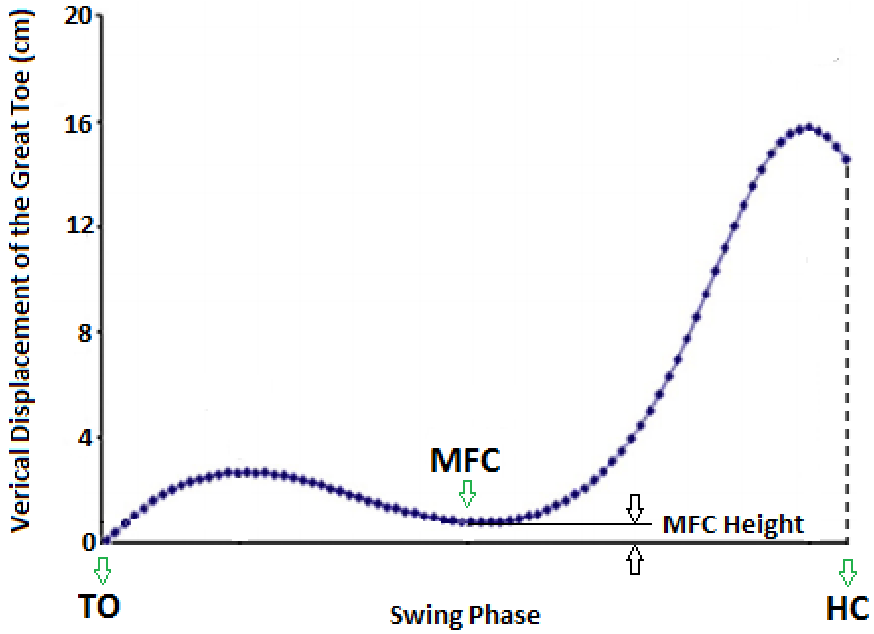

Figure 2-1: Vertical displacement of the great toe during swing phase from Toe Off (TO) to Heel Contact (HC). MFC occurs in the middle of swing phase.

The propensity to trip during walking can be estimated using MFC height. It has been demonstrated that higher mean and/or less variable MFC values are associated with lower risk of tripping [152]. However, greater MFC during walking requires much effort and energy expenditure [155]. Kinematic changes including knee flexion, hip flexion and ankle dorsiflexion have been identified as effective strategies to increase MFC height [156, 157]. Recently, Sato (2015) conducted a multiple regression analysis and revealed that hip flexion and ankle dorsiflexion have the most influence on MFC height in elderly individuals [158]. Accordingly, patients with limited muscle strength e.g. stroke patients would be more prone to trip during walking [159].

2.3.2 Slipping

Slipping occurs when the available friction at the interface of walking surface and foot is insufficient to prevent the foot from sliding [160]. During the stepping process, forward or rearward shear force would be applied at the interface due to the forces applied by the foot to the floor at the instant of touching the walkway [161]. An individual will slip, if the shear force exceeds the friction force at the interface [162]. Slips can either occur at the initial phase of the gait cycle shortly after HC, or at the end of the gait cycle prior to toe off [163, 164]. Friction loss at the foot/ground interface contributes to anterior slip at HC, while it results in posterior slip at toe off. In comparison, it has been suggested that anterior slipping at HC impose greater risk of balance loss, whereas the chance of balance recovery would be higher during posterior slipping [165, 166]. A short period (<70-120 milliseconds) after HC has been shown to be associated with the most dangerous slips [167]. Threshold of 10 cm for slip distance and 0.5 m/s for heel sliding velocity have been reported by previous studies for slip-induced falls during gait [168].

Slip propensity has been identified in the literature using Required Coefficient of Friction (RCoF) which is the minimum required coefficient of friction to prevent slipping [169]. During walking, instantaneous coefficient of friction (CoF) is defined as the ratio of anteroposterior component to vertical component ($\frac{F_{AP}}{F_V}$) of the GRF, and RCoF is one of the local maximum of instantaneous CoF [169, 170]. Slipping initiates once RCoF is greater than the available CoF between foot/shoe sole and walking surface. Available CoF is defined by the ratio of friction force to vertical component of GRF ($\frac{F_{friction}}{F_V}$) [171], and can be quantified using mechanical testing and different sophisticated instruments such as Drag-type meters and Articulated strut devices [160]. Instantaneous CoF pattern during walking consists of several peaks (Figure 2-2) [163]. Among them, peak 3 which is right after HC and has the highest positive CoF value is considered as the RCoF. At peak 3, the heel tends to slide forward due to exerted anterior force by the walking surface on the heel. Considering other peaks of CoF plot, peak 1,2 and 4 are most often ignored, but peaks in negative direction (peak 5 and 6) indicate potential of posterior slipping at gait termination [162, 169].

Slippery conditions are known to affect gait patterns, resulting in kinetic and kinematic changes. During slips, the ankle plantar flexion may be increased, the hip joint movement limited, and the knee joint more flexed to keep the CoM within the area of BoS reduce to risk of slip-induced fall [172]. GRF data during walking over slippery surfaces are also associated with some changes including reduced peak normal and shear forces, as well as incomplete body weight transferring phase [162]. Increased activation period and magnitude of muscle activity have also been reported for lower extremity muscles during slipping [172]. The general characteristics seen with gait kinematics under slippery condition are reduced step length, slower HC velocity and decreased friction demand for young and elderly individuals [173].

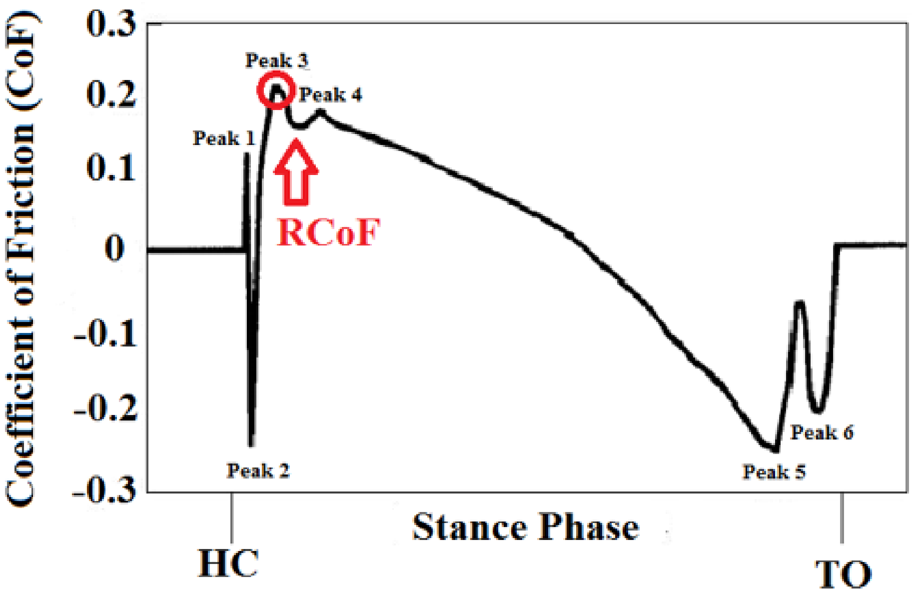

Figure 2-2: CoF changes during stance phase of walking. Peak 3 indicates the RCoF when there is a high risk of anterior slipping. Peak 5 and 6 near the end of the gait cycle show potential risk of posterior slipping at gait termination (adapted from Redfern et al. (2001)).

2.4 Stability of Human Walking

According to engineering dynamics, stability of a dynamic system can be defined by the response of a system to perturbations. Stability may be evaluated by determining whether trajectories of a system remain close to equilibrium when the system is subjected to small perturbations. As long as the trajectories remain close to each other about the equilibrium point, the system is stable. A dynamic system is considered asymptotically stable if its trajectories trend towards an equilibrium point as time approaches infinity. Stability of human walking, as an example of a dynamic system, can be examined in terms of global stability and local stability. Dingwell et al. (2006) defines local stability as the ability of the system to recover from infinitesimally small perturbation such as uneven surfaces, and global stability as the capability of the system to recover from large perturbations such as tripping or slipping [174]. In a recent study by Bruijn et al. (2013), stable gait of a passive dynamic walker was characterized by ability to recover from infinitesimal and large perturbations, as well as larger magnitude of the greatest recoverable perturbation compared to that the walker encountered [25].

In order to quantify gait stability, several approaches have been proposed including Maximum Lyapunov Exponent (LyE) [175], Maximum Floquet Multiplier (MaxFM) [176], Margin of Stability (MoS) [6], Stabilizing and Destabilizing Force [176], Foot Place Estimator [177] and Gait Sensitivity Norm [178]. Among all the techniques, three approaches have received the most attention in the literature: LyE, MaxFM and MoS [25].

2.4.1 The Maximum Lyapunov Exponent

The Lyapunov theory, developed in 1892 by the Russian mathematician Aleksandr Lyapunov, was initially used to analyse stability of dynamic systems. It was proposed that if solutions of a dynamic system start close to an equilibrium point and remain close to this point, the equilibrium point is Lyapunov stable or locally stable. Moreover, if all solutions converge to the point, the equilibrium point will be asymptotically stable. According to such theory, the largest Lyapunov exponent was defined to characterize Lyapunov stability. This quantity measures the rate of divergence of solutions from the equilibrium point. The more diverged solutions, the lower local stability, while the more converged solutions indicate greater local stability.

Various methods have been suggested for LyE estimation [179-181], but each of them had drawbacks. In 1993, Rosenstein and his colleagues proposed a new approach which has been widely adopted for LyE calculation [182]. For this method, the state space representation of a time series of gait kinematic data should be reconstructed. The rationale behind this reconstruction is to provide a sufficiently large Euclidian space to reveal the topological and geometric characteristics of a dynamical system. Differentiated position data and acceleration data are currently the most popular time series for calculation of LyE due to their stationarity [25]. Joint angles and trunk kinematic data have received greater attention compared to other time series for estimation of LyE, but there is no agreement among scholars that which time series is more sensitive to the applied perturbations [25]. Calculating of LyE using either of above-mentioned time series in the anteroposterior, mediolateral or vertical directions provide us with useful information about stability changes and the response of individuals to the applied perturbations during walking in that particular direction.

In order to perform reconstruction, two variables are required: time delay (τ) and embedding dimension (d_E) [183]

$$x(t) \rightarrow Scalar\ Time\ Series$$

$$S(t) = [x(t), x(t+\tau), x(t+2\tau), \dots, x(t+(d_E-1)\tau)] \rightarrow State\ Space\ Representation \quad \text{Equation 2-6}$$

Time delay can be calculated as the first zero of the autocorrelation function of a time series [184]. In addition, it can be estimated using Average Mutual Information (AMI) theory [185]. AMI investigates the level of shared information between two data sets. The minimum amount of shared information is at the first minimum of AMI, and this point will represent the appropriate time delay. The embedding dimension is the optimum required dimension to reveal dynamic characteristics of a dynamical system. The embedding dimension can be computed using Global False Nearest Neighbours [186] or Correlation Dimension [187].

Recently, Ihlen and colleagues proposed an alternative approach for reconstruction of state space to investigate intra-stride changes in local stability of walking [188]. In this method, three-dimensional position and velocity of the heel and toe markers, as well as the ankle, knee and hip joint centres, were used to reconstruct a 30-D state space for each leg during walking. The main advantage of this method compared to time-delay approach is separation of dynamics of single and double support phases.

The next step after reconstruction of the state space form of a time series is to find the initial distance between j^{th} point in a trajectory (solution) and $\hat{j}^{th}$ point on a nearest neighbour trajectory (solution)

$$d_j(0) = min\|j - \hat{j}\| \qquad \text{Equation 2-7}$$

where $d_j(0)$ is the initial naturally occurring perturbation.

The distance should then be estimated after i steps,

$$D_j(i) = C_j e^{(LyE)*t} \qquad \text{Equation 2-8}$$

where $D_j(t)$ is the distance between adjacent markers trajectories, C_j is initial separation and LyE is the largest Lyapunov Exponent.

Taking the Napierian logarithm of both sides and least-square curve fitting to the average line yields

$$\ln d_j(i) = \ln C_j + LyE(i.\Delta t)$$ Equation 2-9

$$y(i) = \frac{1}{\Delta t}\langle \ln d_j(i) \rangle = LyE.i + c, c = \frac{\ln C_j}{\Delta t}$$ Equation 2-10

where LyE is the slope of the fitted line and an indicator for stability analysis. Positive LyE indicates divergence of solutions (the level of chaos), while a negative LyE indicates convergence of solutions and the level of determinism of the dynamic system. This indicates whether a biological system is locally stable or unstable.

During the last two decades, utilization of LyE to examine local dynamic stability of human walking has received increasing attention. Preliminary work in this area was undertaken by Dingwell and Cusumano (2000) to investigate local dynamic stability of walking among peripheral neuropathic patients [112]. In this study, they reported that neuropathic patients walk with greater local stability than healthy individuals. Since then, a large and growing body of literature has evaluated the effect of various diseases or disability on local dynamic stability. For instance, it has been shown that patients suffering from transfemoral amputation [189], stroke [190], degenerative cerebellar ataxia [191], neurological disorders [192] and multiple sclerosis [193] have lower local stability compared to their healthy compartments. While there is a consensus among researchers that LyE is a reliable measure of stability in pathological gait [192, 194, 195], a recent systematic literature review by Mehdizadeh (2018) concluded that utilization of LyE for stability analysis of pathological gait requires a baseline LyE range for each segment calculated using a generally accepted method [196].

To date, several studies have investigated the influence of internal and/or external perturbations on local dynamic stability of walking. It has been demonstrated that walking while performing a verbal task results in reduced local stability in elderly people with and without dementia [142]. Rábago et al (2016) suggested that gait balance in healthy young individuals is resistant to perturbations associated with cognitive distractions [28]. In a more recent analysis of stability, it has been suggested that cognitive dual-tasking while walking can decrease local stability in young individuals [197]. Considering cognitive load associated with cell phone usage in elderly individuals, it has been argued that talking while walking gives rise to larger trunk LyE [142, 198]. In contrast, Kao et al. (2015) reported no significant difference in trunk stability during cell-phone usage while walking in healthy young and older adults [199].

It has been shown that visual dual-tasking during walking results in lower local stability in healthy young individuals [59, 200]. However, Sejdic et al. (2013) found no significant changes in local stability when performing secondary visual task in young adults [201]. With regards to the effect of external perturbations, reduced local stability of young individuals has been reported during walking under pseudo-random perturbations applied to walking surfaces [59, 200]. LyE analysis of the upper body in young adults using trunk motion trajectories has shown that walking uphill is associated with reduced local stability in all principal anatomic directions, whereas walking downhill is associated with decreased local stability in the vertical direction [202]. Although some research has been carried out on the influence of internal and/or external perturbations on local dynamic stability, there is no general agreement among researchers about the influence of cognitive, auditory and visual perturbations on local stability. Moreover, there is still very little scientific understanding of the response of the head segment to perturbations arising from walking under dual-tasking or walking on irregular surfaces.

2.4.2 Maximum Floquet Multiplier

The Maximum Floquet Multiplier (MaxFm) was introduced by the French mathematician, Henri Poincaré, to examine stability of periodic systems. This technique investigates how a system responds to perturbation from one cycle to the next cycle at a special point during a cycle e.g. heel strike during walking. This type of analysis is performed on a plane which is called Poincare' map. The Poincaré map, used to calculate MaxFM, is a section or plane which is perpendicular to all trajectories and maps r_n to r_{n+1} between two neighbouring trajectories.

$$r_{n+1} = P(r_n) \qquad \text{Equation 2-11}$$

If $r^* = P(r^*)$, r^* is called a fixed point of a trajectory on the map. Then, M will be defined as characteristic multiplier:

$$M = \frac{dP}{dr} @ r^* \qquad \text{Equation 2-12}$$

Stability is defined as follows:

- If $M > 1$ the system will be unstable,
- If $M < 1$ the system will be stable,
- If $M = 1$ the system will be partly stable and partly unstable.

Using Floquet theory, Hurmuzlu and Basdogan (1994) proposed a new approach for stability analysis of human walking [203]. The first step in this method was reconstruction of the state apace form of a time series. The state space can be reconstructed either by the aforementioned time-delay approach, or by alternative approaches proposed by Kang (2007) [94]. He argued that as the movement of a rigid body can be characterized by six degrees of freedom $(x, y, z, \alpha, \beta, \gamma)$, the dynamic characteristics of the body can be described by degrees of freedom and their derivatives. Further, in order to avoid non-stationarity (changes of joint probability distribution during time shifting), the first and second derivatives of the degrees of freedom can be used to construct the state space as follows

$$S = [x, y, z, \alpha, \beta, \gamma, \dot{x}, \dot{y}, \dot{z}, \dot{\alpha}, \dot{\beta}, \dot{\gamma}]$$ Equation 2-13

Or

$$S = [\dot{x}, \dot{y}, \dot{z}, \dot{\alpha}, \dot{\beta}, \dot{\gamma}, \ddot{x}, \ddot{y}, \ddot{z}, \ddot{\alpha}, \ddot{\beta}, \ddot{\gamma}]$$ Equation 2-14

Recently, Ihlen and colleagues proposed another approach for reconstruction of state space to intra-stride changes in local stability of walking [188]. In this method, three-dimensional position and velocity of the heel and toe markers, as well as the ankle, knee and hip joint centres were used to reconstruct a 30-D state space for each leg during walking. The main advantage of this method compared to time-delay approach is separation of single and double support phases.

The next step in estimation of MaxFM is construction of the Poincaré plot which maps state of the system at stride k to stride $k + 1$:

$$S_{k+1} = P(S_k)$$ Equation 2-15

Here, the fixed point is denoted S^* and can be calculated as the average of the trajectories of the system over many strides using

$$S^* = P(S^*)$$ Equation 2-16

The alterations of $[S - S^*]$ between stride k to stride $k + 1$ can be investigated using the Jacobian matrix

$$[S_{k+1} - S^*] = J(S^*)[S_k - S^*]$$ Equation 2-17

The eigenvalues of the Jacobian matrix are called floquet multipliers. The time series is said to be orbitally stable when the maximum floquet multiplier (MaxFM) is less than 1, otherwise (MaxFM > 1), the time series is orbitally unstable and perturbation will grow from cycle k to cycle $k + 1$. The assumption of performing such analysis is periodic behaviour of the dynamic system (gait cycle). In contrast, local dynamic stability analysis assumes the system is aperiodic.

The first systematic study of orbital stability was reported by Hurmuzlu and Basdogan in 1996 [204], in which polio patients were found to exhibited lower orbital stability of walking compared to their healthy counterparts. Later, some studies have attempted to evaluate the effect of various diseases or disability on orbital stability. In a contrasting study that set out to determine the periodicity of ankle clonus behaviour and its associated input in patients with spinal cord injury, Hidler et al. (2000) found no relationship between spasticity and orbital stability [205]. One study by Marghitu and Hobatho (2001) reported lower orbital stability in children with torsional anomalies in knee and ankle joints, compared to healthy group [206]. In an investigation by Dingwell and his colleagues (2007), it was shown that orbital stability was not affected in neuropathic patients during walking [207]. The experimental data, however, are rather controversial, indicating there is no general agreement on utilization of MaxFM to distinguish stability between healthy and pathological gaits [208]. One possible reason behind this controversy could be lack of methodological uniformity in calculation of MaxFM in previous studies [208].

A great deal of previous research into orbital dynamic stability has focused on the effect of internal and/or external perturbations during locomotion. In an investigation into the effect of cognitive dual-tasking while walking on orbital stability, it was reported that cognitive perturbation arising from dual-tasking does not affect orbital stability in young individuals [209]. In the same manner, Rabago (2015) argued that locomotion is robust to applied cognitive perturbations arising from dual-tasking in healthy young people [210]. It has been demonstrated that walking in a visually destabilizing environment can affect orbital stability in a direction specific manner in young adults, suggesting that orbital stability is lower in the direction of applied visual perturbation [200]. Although some research has been carried out on the influence of internal and/or external perturbations on orbital dynamic stability, there is still very little scientific understanding of the response of the human body to perturbations arising from walking under sensory and/or cognitive dual-tasking or walking on irregular surfaces.

MaxFM and LyE are stability measures derived from dynamical systems theory. The major benefit associated with these approaches is that they can be calculated for different joints and segments in principal anatomic directions, thus providing information about the response of individual joints and/or segment to small perturbations [25]. However, there is necessity for data collected over many continuous walking cycles (more than three minutes). This can be addressed from treadmill walking data; however, walking on a treadmill is associated with several advantages and disadvantages. It has been shown that treadmill walking improves reliability of data through collection of many consecutive strides in a short period of time [211]. During treadmill gait, walking speed can be controlled, and participants can wear safety harness to avoid unexpected injuries [212]. However, this may alter gait variability, including local and orbital dynamic stability. For example, recent evidence suggests that treadmills may artificially reduce natural variability during walking, and improve local and orbital stability [213]. Another limitation associated with nonlinear measures of gait stability is that they are not able to provide information about the stability of a joint and/or segment at an individual event e.g. minimum foot clearance within a gait cycle. In other words, these measures tell us about the overall response of a source of kinematic data to small perturbation over many continuous gait cycles. This limitation has, however, been addressed to some extent by phase-dependent analysis of local stability which was proposed by Ihlen and colleagues (2012) [188].

2.4.3 Margin of Stability

It has been shown that the stability of an individual during standing may be maintained as long as the vertical projection of the CoM is located inside the border of the base of support (BoS) [214, 215]. This concept is based on the inverted pendulum model, which has been described by Winter (1995) [216]. The applicability of this model in dynamic situations has been refuted by some authors, since it does not consider the effect of direction of velocity [217]. Hof et al. (2005) attempted to improve the simple inverted pendulum model by considering direction and magnitude of the velocity of the CoM [6]. He introduced a variable called Extrapolated Centre of Mass, X_{CoM} and showed that in order to maintain stability, the X_{CoM} should be located within BoS. X_{CoM} is defined using

$$X_{CoM,x} = x + \frac{\dot{x}}{\omega_0}$$ Equation 2-18

$$\omega_0 = \sqrt{\frac{l}{g}}$$ Equation 2-19

where x is the position of CoM, $\dot{x}$ is the velocity of CoM, ω_0 is the natural frequency of the inverted pendulum, l is the leg length and g is the universal gravitational constant.

In order to calculate MoS in the sagittal and frontal planes, X_{CoM} and BoS should be calculated in the anteroposterior and mediolateral directions. However, there is no general agreement about the definition of BoS in either direction. Hof et al. (2007) used centre of pressure (CoP) position as the BoS boundary [218]. A number of studies used the line between two great toes as the anterior boundary of BoS in the sagittal plane, and lateral malleolus or the 5th metatarsal as the mediolateral border [219-222], while some studies used an alternative approach which considers the lateral malleolus marker on the ankle joint as the posterior boundary in sagittal plane and mediolateral boundary in the frontal plane [59, 79, 223, 224]. The difference between discussed definitions of BoS in the mediolateral direction merely results in significant changes in MoS results due to very small distance between lateral malleolus and the 5th metatarsal in the frontal plane. In the anteroposterior direction, however, calculation of MoS with respect to the anterior and posterior edges of BoS could possibly elicits two different interpretations: forward and backward balance loss, respectively.

The spatial MoS can be estimated as follows:

$$MoS = BoS - X_{CoM} \quad \text{Equation 2-20}$$

where MoS is the minimum distance between BoS and X_{CoM} within each step [25].

The minimum required impulse to make an individual unstable can be derived using

$$I = m\omega_0 \cdot MoS \quad \text{Equation 2-21}$$

The minimum required impulse is the required impulse for the CoM to pass the border of BoS. Positive MoS indicates lower risk of loss, while negative MoS implies higher likelihood of balance loss. MoS could be calculated at different stages of gait cycle, but it has been mostly calculated at HC since it has high risk of balance loss [162, 225, 226]. Despite its simplicity, this technique has some notable benefits. Firstly, the possibility of stability analysis at a specific time in a gait cycle can be provided with this method. Therefore, this approach is called ‘instantaneous stability assessment’. Moreover, this method provides the opportunity of step-to-step stability analysis of human walking to evaluate the alterations of cycle to cycle stability. One limitation of such an approach is that all calculations are based on a passive dynamic system. Consequently, any changes resulting from muscle activities cannot be considered [227].

In a study by Hof (2007), it was shown that knee amputees exhibited larger margin of stability compared to a control group because of compensation strategies to maintain stability in the prosthetic foot [218]. In an analysis of the effect of surface of walking on margin of stability, Rosenblatt and Grabiner (2010) found that walking on a treadmill does not affect MoS compared to overground walking in young adults [35]. Hak et al. (2013) examined the stability of post-stroke patients compared to age-matched subjects during walking under continuous perturbation of walking surface, as well as performing an additional task [228]. It was concluded that mediolateral MoS was not affected in both groups in either testing condition, while backward MoS reduced in the patient's group during walking and performing gait adaptability task. Using MoS as a measure of step-to-step gait balance, it has also been demonstrated that young individuals maintain lateral balance by immediate correction of unstable steps by stable one [229].

A great deal of previous research into MoS has focused on the effect of internal and/or external perturbations during walking. Marone et al. (2014) demonstrated that texting can increase MoS in the frontal plane during walking in young individuals [230]. In another investigation into the effect of cell phone usage on dynamic stability, Kao et al. (2015) also showed that walking while texting on a cell phone can result in a greater mediolateral MoS in young and older adults [199]. In a recent study, Kao (2018) suggested that walking under cognitive perturbation does not affect MoS in healthy individuals [231]. Considering external perturbations, Menant et al. (2006) showed that walking on inclined surfaces results in reduced lateral and posterior MoS in young and elderly individuals [232]. It has also been demonstrated that a negative gradient walkways contributes to decreased mediolateral MoS in young adults [202]. While some research has been carried out on the effect of various types of perturbation on MoS, there is still very little scientific understanding of the influence of internal perturbation arising from sensory and cognitive dual-tasking as well as external perturbation due to irregular walking surfaces on instantaneous dynamic stability.

2.5 Balance Maintenance Strategies During Walking

Regulation of the Centre of Pressure (CoP) placement during ambulation helps an individual to keep their CoM within the border of BoS and maintain balance during gait. A number of studies have highlighted different adopted strategies to stabilize the body during locomotion. This section provides a brief summary of the literature relating to the most major gait balance maintenance strategies including ankle and hip strategies, stepping and push off strategies, as well as upper body strategies.

2.5.1 Lateral Ankle Strategy

Lateral ankle strategy was first been described as a technique to control quiet standing [233, 234]. Using this strategy, an individual rotates the whole body as a single-segment inverted pendulum over the ankle joint and moves the CoM horizontally forward in the sagittal plane in response to anteroposterior perturbations [235]. Under the risk of posterior balance loss, the ankle strategy may result in activation of ventral muscles like ankle dorsi-flexors, while anterior falls are associated with activation of dorsal muscles like ankle plantar-flexors because of ankle strategy [236]. In a study by Blenkinsop and his colleagues (2017), it has been suggested that the ankle strategy is adopted by individuals either under perturbation with low amplitude and speed, or no perturbation [237]. Effectiveness of the ankle strategy, however, relies on capability of the foot segment in exertion of torque while it is in touch with the support surface [235, 237]. Previous studies have indicated that people adopt similar ankle strategy during walking. Initially, Hof et al. (2007) found that ankle strategy is actively used by healthy individuals to correct CoP position and control balance during locomotion [218]. In another recent study by Reimann and his colleagues (2017), it has also been demonstrated that ankle strategy plays a vital role in whole body balance maintenance during walking in young adults [238]. To date, however, this strategy has received little attention in the gait research literature as a balance maintenance mechanism due to its limited functional range compared to the foot placement strategy [239].

2.5.2 Hip Strategy

Previous studies defined hip strategy as a mechanism to stabilize the body during quiet standing [236]. This strategy is mostly adopted by individuals either under larger perturbations, or malfunction of the ankle joint in generating required torque for balance maintenance through ankle strategy. In this strategy, the body behaves like an inverted pendulum with two segments: upper body and lower body [235]. The upper body rotates around the hip joint anteriorly while the hip joint moves backward, and the leg rotates around the ankle joint posteriorly to adjust CoM position. In contrast to the ankle strategy, the knee extensor and hip flexor muscles are mostly active during adoption of the hip strategy [237]. While the hip strategy is more powerful compared to the ankle strategy, it has been shown that pure hip or ankle strategy is not commonly employed and individuals adopt mixed hip and ankle strategy for postural control [235]. During walking, the hip strategy contributes to propulsion and balance maintenance [239]. However, one limitation for hip control could be to provide small range of CoM acceleration because it is associated with angular acceleration of the upper body that needs to be controlled before reaching the limits of joint range of motion [239, 240]. It has been, therefore, suggested that this strategy should be adopted mostly for short term purposes since excessive angular motion of the upper body, which is the heaviest part of the body, can increase the risk of whole body balance loss [239].

2.5.3 Stepping Strategies

Modulation of foot placement during walking has been identified as one of the most effective strategies to maintain gait balance. As discussed earlier in section 2.2.1, walking with wider/narrower step width, longer/shorter step length, and the effect of speed on them have been shown to be associated with alteration of gait stability [72, 73, 241]. To better understand the foot placement mechanism to stabilize walking, we will discuss them in different anatomic planes. In the sagittal plane, anteroposterior placement of the swing foot plays a key role in balance maintenance since the CoM is mostly out of the BoS border during single support phase [71]. It has been suggested that required foot placement to maintain balance in sagittal plane results from passive dynamics [242]. Under perturbation, however, planar passive dynamics will not be sufficient to maintain balance, and modulation of step length through active control might be a strategy to keep walking stable [71, 242]. In the frontal plane, due to side-to-side movement of CoM's vertical projection towards lateral border of BoS during single support phase, the need for active control through foot placement would be necessary. Widening or narrowing of the BoS area through modulation of step width has been shown to be the dominant strategy for active controlling of balance during gait [216]. Considering high metabolic cost associated with narrower step width, and high energy cost of ambulation during walking with wider step width, individuals tend to choose optimised step width values that minimize energy cost of locomotion [71].

2.5.4 Contribution of Upper body

Control of upper body movements (the head, trunk and arms) during walking plays a fundamental role in maintenance of gait balance. Since the upper body mass has considerable inertia located above the ground, at about two third of body height, its motions under certain conditions during ambulation can play a role in balance loss to the whole body [243, 244]. Contribution of the upper body in balance maintenance during locomotion can be classified into trunk strategies and arm swing [245]. Data from several studies suggest that increased trunk stiffness might be a mechanism to maintain gait balance in response to disturbances, especially in older adults and patient suffering from low back pain [246-249]. A stiffer trunk through antagonistic co-contraction of torso musculature contributes to effective reduction of the perturbation-induced trunk deviations, and ultimately balance maintenance [250]. However, it has been shown that a stiffer trunk during ambulation may affect performance of other adaptive responses of the trunk to the applied perturbation [251, 252]. Lateral trunk bending has also been identified as one of the compensatory mechanisms to keep walking stable in older adults and patients suffering from various diseases including cerebral palsy [253], spinal muscular atrophy [254] and dystrophy [255]. Through this strategy, the body CoM would be shifted toward the stance limb and aligned over the hip joint centre to compensate lack of body support in the single stance due to weakness of hip abductor muscles [253]. However, excessive lateral bending can increase shoulder oscillation, resulting in compromised gaze stability and whole-body balance in frontal plane [253].

Arm swing has also been highlighted in previous studies as an effective contributor of the upper body to balance maintenance in the frontal plane. This strategy can help to balance angular momentum and control CoM movement in the mediolateral direction [256]. Pendulum-wise motion of arms during locomotion has been shown to be associated with reduced metabolic costs to stabilize body in frontal plane through energetically expensive mechanisms e.g. step modulation [257]. In a study by Punt et al. (2015), arm swing was linked to increased local stability of walking in young healthy adults [258]. Effectiveness of arm swing on balance maintenance, however, has been challenged by some studies demonstrating no balance improvement due to arm movement. For instance, Collins et al. (2009) found no association between step-to-step balance and arm swing during gait following a passive modelling study [259]. In another major study, Bruijn et al. (2010) reported no significant difference in local stability of walking with and without arm swing in healthy adults [260]. Despite the existing debate in the literature on usefulness of swinging arm for balance maintenance, there is an agreement among scholars on functional role of arm swing in balance recovery from a perturbation [149, 258, 260].

2.6 Balance Hazard

As stated before, walking is one of the most complicated tasks of daily living which requires continuous processing and integration of provided information by various sensory modalities to be performed successfully by an individual [7]. Balance maintenance during such a complex cognitive type of movement can be challenging under altered environmental, cognitive or sensory circumstances that are likely to pose greater threat to whole-body stability [261]. Those disturbances are generally classified into two types according to their origin: internal perturbations, and external perturbations. The latter is most commonly associated with disturbances originated from environmental complexity e.g. uneven terrains and obstacle crossing, while the former covers those disturbances such as dual-task walking and aging, which may arise from internal sources e.g. neuromuscular noises [25].

2.6.1 Internal Perturbations

Dual-tasking while walking has been shown to be one of the main sources of internal perturbation associated with increased probability of falling [262]. During locomotion, performing an additional task with shared cognitive resources may results in performance reduction in either task and greater likelihood of balance loss [10, 13, 14]. The risk of falling would be varied depending on the demand of the concurrent task and neuro-musculoskeletal capacity of individuals [25]. Several studies have linked dual task walking with slow walking in healthy older adults and elderly patients [23, 263-269]. Considering temporal gait parameters, previous research has established that multitasking during walking results in longer temporal gait characteristics, especially stride and double support time [23, 264, 266, 267, 270]. In another study, Lee (2017) argued that performing an additional physical and cognitive task while walking decreases single limb stance time and step length in elderly people [271]. Hausdorff et al. (2008) also labelled longer swing time as one of the temporal decrements during dual task walking in older adults [268]. A number of studies have postulated that healthy elderly individuals have shorter stride length during dual task walking [272-274]. It has been shown that cognitive dual-tasking results in increased step width and shorter step length in healthy young and elderly individuals [58]. Further, it has been demonstrated that dual-tasking due to cell phone usage while walking contributes to wide steps gait pattern in young and elderly individuals [271]. These studies suggest that dual-tasking during walking may affect stepping strategies in young adults.

A large and growing body of literature has investigated the effect of dual-tasking while walking on gait stability. Considering cognitive load associated with multitasking during walking, compromised dynamic stability of walking has been reported in older adults [142, 275]. However, the generalisability of much published research on this issue in young adults is problematic. For instance, Longo et al. (2018) observed decreased local stability due to performing cognitively demanding additional task during walking [276], while Rabago et al. (2016) found no evidence for association between cognitive multitasking and local stability [210]. In a similar fashion, Dingwell reported that cognitive perturbation arising from dual-tasking doesn’t affect orbital stability in young individuals [209]. In a recent study, Kao and co-workers linked unaffected MoS and local stability with adopted conservative walking strategies by young individuals during cognitive multitasking [231]. Considering other types of dual-tasking while walking, there is little published data on additional visual and auditory tasks. These studies mostly found no significant reduction in gait stability due to performing concurrent visual or auditory task during locomotion [201, 277-279].

A great deal of previous research into gait stability analysis under multitasking conditions has focused on combined perturbations such as cell phone usage while walking. A few recent studies reported improvements in the mediolateral MoS during walking while texting on a cell phone in healthy young and older adults [199, 230]. Conversely, reduced trunk stability has been linked with talking or texting on a cell phone while walking in young adults [145]. In contrast to above studies, Pau and colleagues (2018) suggested that typing a text while walking is not a threat to gait stability in elderly individuals [280]. Likewise, several studies found no evidence that texting on a cell phone while walking has an influence on dynamic stability of gait in young people [281, 282]. Agostini et al. (2015) reported that adjustment of motor control strategy toward stabilization of the ankle joint during weight transferring phase might be an adaptive response by the central nervous system (CNS) during texting on a cell phone while walking to improve gait balance in young adults [283]. The inconsistency in reported findings reflects that much uncertainty still exists about dynamic stability changes under walking while using a cell phone.

2.6.2 External Perturbations

External perturbations can be imposed by various environmental resources on individuals during locomotion. Numerous studies have attempted to explain dynamic stability changes during walking over irregular terrains. Chang et al. (2010) showed that walking over compliant surfaces in young adults is associated with improved local stability compared to that during overground walking [284]. Reduced CoM height through increased angular displacement of the knee and hip joints in sagittal plane has been reported as the reasons behind improved gait stability during walking over compliant surfaces in healthy individuals [285, 286]. Menz et al. (2003) demonstrated that head stability is prioritized during walking on irregular surfaces through stepping strategies in young individuals [287]. Higher variability of MoS has also been reported during walking over loose rock surface compared to overground walking in unilateral transtibial amputees and healthy young adults [288]. In a recent study, LyE analysis of the upper body in young adults using trunk motion trajectories has shown that walking uphill is associated with reduced local stability in all principal anatomic directions, whereas walking downhill is associated with decreased local stability in the vertical direction [202].

To date, several studies have investigated gait balance changes during walking on moveable platforms. Reduced MoS in sagittal plane has been reported in young adults during walking under oscillated platform in the anteroposterior direction [289]. In a similar fashion, the mediolateral oscillation of the walking surface during walking has led to reduced local and orbital stability in unilateral transtibial amputees [109]. Conversely, Hak et al. (2012) reported that transtibial amputees responded positively to side-to-side mechanical perturbation of walking surface and showed increased MoS [59]. Considering young healthy cohorts, Beurskens et al. (2014) observed decreased local and orbital gait stability, while Hak et al. (2013) reported increased MoS during walking over mediolaterally translated walking platform [59, 109]. In another study by Sinitsky et al. (2012), it has been shown that orbital walking stability is negatively correlated with magnitude of mediolaterally applied surface perturbation [290]. A number of studies have also discussed the effect of visual field oscillation on dynamic stability during walking. It has been suggested that healthy young adults are more sensitive to translation of visual field in the mediolateral direction rather than anteroposterior direction [291]. Unilateral transtibial amputees have exhibited reduced local and orbital stability during walking under mediolaterally translated visual scene [109]. Likewise, Sinitsky and colleagues (2012) observed decreased dynamic stability of walking under continuous side-to-side visual perturbation of different amplitudes, and concluded that magnitude of mediolateral translation of visual field during locomotion is correlated negatively with local stability of walking in healthy young individuals [22].

Researchers have also attempted to evaluate the impact of slippery surfaces on gait stability during locomotion. According to a study by Lockhart et al. (2005), recovery from a slip would be much faster in young individuals compared to their elderly counterparts [292]. In another study conducted by Liu and colleagues (2015), it has been suggested that slip training in virtual reality environment can reduce the number of falls in older adults [293].

Yang and Pai (2014) applied a sudden slip during walking and found that elderly fallers can be distinguished from non-fallers through feasible-stability-region during locomotion [294]. Several studies also examined balance maintenance during obstacle crossing. High strength of the muscle groups responsible for push off and trunk control during trip recovery has been shown to be associated with reduced risk of balance loss in elderly individuals following a trip [147]. Prolonged arm swing at the instant of tripping has also been identified as one of the key factors of balance recovery in young individuals [149]. Wang et al. (2012) showed that balance maintenance during walking under consecutive trip-induced perturbations in young individuals, might be related to increased toe clearance and decreased forward velocity which was observed before and after trips [295].

So far, different measures of stability and falling risks assessment, common causes of balance loss and balance maintenance strategies, as well as potential hazards to gait balance, have been discussed in this thesis. Although extensive research has been carried out on the effect of potential disturbances on gait stability and probability of balance loss, no study exists that demonstrates that how internal and external perturbations arising from daily living activities affect dynamic stability and the risk of balance loss. Moreover, how individuals respond to such perturbations and control stability of gait, and which type of additional task while walking pose greater challenge to human gait, has not been determined. To answer these questions, in the next three chapters, dynamic stability analysis and falling risks assessment are performed for young individuals ' gait under internal perturbations originating from some of the most common additional daily living activities while walking including cell phone usage, listening to music, watching a video clip and performing mathematical calculation, as well as external perturbations associated with sloped terrains.

3 The influence of cell phone usage on dynamic stability and variability during walking

3.1 Summary

Cognitive load associated with dual-tasking while walking is known to affect balance and increase fall risk. The objective of this study was to quantify the effect of cell phone usage on dynamic balance during walking. Nineteen healthy male participants were recruited and asked to walk on an instrumented treadmill at their self-selected speed (baseline walking) as well as walking whilst simultaneously (i) reading on a cell phone (ii) texting (iii) talking on a cell phone. Trajectories of retro-reflective markers placed on the body were simultaneously recorded using an optical motion analysis system. Using these data, spatiotemporal measures and their variability were computed including Margin of stability (MoS) at HC and MFC, Maximum Floquet Multiplier (MaxFM) of the head motion, Maximum Lyapunov Exponent (LyE) of the head and trunk, hip joint, knee joint and ankle joint motions, slip and trip propensity. Detrended fluctuation analysis (DFA) was also performed on spatiotemporal time series. Finally, deviation of MoS and trip propensity values during dual-task trials from the corresponding values during baseline walking were calculated to identify the most destabilizing type of cell phone usage. Mean step length, step width and step time changed significantly in response to perturbations associated with talking while walking compared to baseline walking.

Participants walked with higher variability of step width during dual-task trial compared to single-task walking. Dual-tasking did not affect DFA results compared to baseline trial. Mediolateral MoS at HC was significantly increased during walking while talking compared to baseline walking. However, dual-tasking due to cell phone usage did not affect anteroposterior MoS at HC and MFC. Texting while walking caused a nearly significant increase in LyE of the head movement in the mediolateral direction compared to that during

baseline walking, but reading and talking on a cell phone did not affect local stability of the head. Trunk local stability was not significantly different between dual-task and single-task trials. Talking and texting while walking mostly affected local stability of the ankle, knee and hip joints in the mediolateral direction. Reduced mediolateral local stability of the ankle joint was observed during reading task, but other joints appeared unaffected. In the anteroposterior and vertical directions, local stability results of the ankle, knee and hip joints were mixed. In some joints we detected slight increase in LyE due to dual-tasking, while others demonstrated no significant changes during dual-task trials compared to baseline walking. Orbital stability of the head and trunk movements were not affected by performing a concurrent cell phone related task while walking in either anatomic direction, except reduced head orbital stability in the mediolateral direction due to reading on a cell phone while walking. Trip propensity significantly increased during talking while walking compared to single-task walking, but slip propensity remained unchanged during the other dual-task trials. Talking while walking resulted in the largest perturbation effects compared to other types of cell phone task while walking.

The findings suggest that individuals appeared to adopt cautious gait patterns in response to perturbations associated with cell phone usage while walking. It was also found that cell phone usage while walking affects balance in a direction specific manner. Texting while walking may increase the risk of side-to-side balance loss while talking during walking may increase likelihood of tripping. Orbital stability and anteroposterior MoS were less sensitive to perturbations associated with cell phone usage while walking compared to other stability measures. Talking while walking has also been identified as the most destabilizing type of cell phone usage while walking. One can conclude that stabilising strategies that target upper body configuration and stepping patterns have potential to reduce gait instability associated with cell phone usage while walking.

3.2 Introduction

Falls are the third major cause of inadvertent injury in young adults aged between 18-35 years [21], with greatest prevalence during walking. Performing cognitively demanding tasks simultaneously during walking can slow the processing of the visual and vestibular systems required in maintaining balance, and may ultimately increase the risk of a fall event [145, 296]. One of the most common activities of daily living involving dual-tasking while walking is cell phone usage, which is associated with visual distraction and cognitive load, and has been shown to increase risk of collision and falling [297, 298]. In the US, 1,500 injuries associated with cell phone usage during walking were reported in 2010; however, most dual-tasking-related injuries are unreported [299]. Greater cell-phone-related injury risk is associated with individuals aged between 16 and 25 yrs, who are known to be the heaviest users of this technology [299]. Shorter stride length, greater lateral deviation from a straight line and reduced gait speed have been reported during walking while texting and talking on a cell phone [300, 301], with these studies suggesting that stability of the body may be compromised.

In the context of human movement, variability is a built-in property of all biological systems which can be detected through multiple repetitions of a task, for example, dissimilar footprints while walking the same path on snow twice [95]. Stergiou and Decker (2011) described movement variability as occurred variations in motor performance while repeating a task [96]. According to the existing theoretical frameworks, the Generalized Motor Program Theory and Dynamical Systems Theory, variability is a destabilizing factor for motor performance [96, 98-100]. Gait variability can be measured using a wide variety of tools including linear e.g. standard deviation [108] and nonlinear e.g. Sample Entropy (SampEn) [131] and Detrended Fluctuation Analysis (DFA) techniques [114]. Linear approaches quantify magnitude of variability while nonlinear techniques provide information about changes in gait pattern variations over time [110]. DFA is one of the most practical methods of revealing information about long range correlation in a dataset made of discrete events e.g. stride time [114, 118-126]. The outcome of analysis for a set of data is DFA Exponent which quantifies the level of statistical persistency in the data set. For instance, DFA Exponent > 0.5 indicates persistency, and DFA Exponent < 0.5 implies anti-persistency. It has been suggested that DFA analysis is a reliable approach to get useful information about neuromuscular control system in response to perturbations during ambulation [321]. While variability provides useful information about the risk of balance loss during walking, it should be kept in mind that variability is different from gait stability, and uniform interpretation of both gait complexity and stability ought to be avoided [25, 302].

The influence of physical and cognitive perturbations on the risk of balance loss during walking have been previously quantified using margin of stability (MoS) [289, 303, 304], a measure used in human ambulation stability analysis that is most commonly calculated from the extrapolated centre of mass (XCoM) relative to the boundaries of the base of support (BoS) [6, 25]. Positive MoS indicates lower risk of balance loss while negative MoS implies higher likelihood of balance loss. Marone et al. (2014) demonstrated that texting can increase MoS in the frontal plane during walking [230]. Likewise, Kao et al. (2015) showed that cell phone usage can result in a greater mediolateral MoS [199]. These studies, and others in the gait literature [229, 305-309], compute MoS relative to the CoM position in either the frontal or sagittal plane, but do not explicitly consider the direction of motion most commonly associated with falls at a given point in the gait cycle. For example, slipping or backward balance loss most commonly occurs at heel strike due to insufficient friction at foot contact [162, 226, 310], while forward balance loss or tripping frequently occurs at MFC due to the collision of the swing foot with the ground or an obstruction [154]. Thus, MoS calculations in directions other than those associated with common fall patterns may potentially underestimate dynamic balance estimations and fall risk.

Stability of the human body during walking under different conditions has been described in terms of regularity and variability of trunk motion. The most likely reason behind this is the trunk's mass which can considerably affect dynamic stability of whole body [311]. Trunk local and orbital stability during ambulation has been quantified using nonlinear measures of stability including the Maximum Lyapunov Exponent (LyE) and the Maximum Floquet Multiplier (MaxFM), respectively [25]. Local stability estimates the logarithmic rate of divergence/convergence of neighbouring trajectories in the state-space [112], while orbital stability quantifies the rate of divergence/convergence of trajectories from one gait cycle to the other [203]. It has been shown that cognitive load associated with talking while walking gives rise to larger trunk LyE [142, 198]. More recently, it was demonstrated that texting or having a cell phone conversation during walking increases gait variability but decreases sample entropy of trunk velocity time series, suggesting more predictability and repetitiveness in pattern of trunk movement across time during talking/texting while walking. It has also been argued that cell phone talking or texting while walking decreases trunk stability [145], while Kao et al. (2015) failed to establish any relationship between cell phone usage while walking and trunk stability [199].

At present, little is known about the independent effects of cell phone reading, texting and talking on balance and stability during walking, nor their influence on the trunk, lower limb joints and head segment, which plays an important role in gaze stabilisation and postural control [312]. In addition, research to date has not yet determined the effect of cell phone usage while walking on persistency of spatiotemporal time series, and trip and slip propensity. Finally, it is not well understood which type of cell phone usage including talking, texting or reading during walking are more destabilizing. Our first aim was to use MoS, LyE and MaxFM to quantify the effect of cell phone talking, reading and texting while walking on balance of the body, considering stability of the head and trunk segments, as well as the ankle, knee and hip joints. Secondly, we aimed to investigate the relationship between gait between dual-tasking due to cell phone usage while walking and slip/trip propensity as well as persistency of stepping parameters. Finally, we aimed to evaluate which type of cell phone usage while walking has the greatest effect on gait pattern. According to previous studies, we hypothesized that: 1) cell phone usage while walking results in shorter step length, larger step width, shorter step duration and double support time [79, 145] ; 2) talking, texting or reading on a cell phone during walking decreases dynamic stability [145], increases gait variability [145] but does not affect the risk of slipping and tripping [58]; 3) texting and reading on a cell phone during walking is more destabilizing than talking on a cell phone while walking [145].

3.3 Methodology

3.3.1 Participants

Twenty healthy young adult males (age: 28.4 ± 4.5 years, height: 175.8 ± 5.3 cm, body mass: 74 ± 11.4 kg) were recruited from the University of Melbourne (Table 3-1). Exclusion criteria for participants were (i) previous lower extremity injury; (ii) previous history of lower limb surgery; and (iii) known neurological or musculoskeletal disorders. Ethical approval was obtained from the institution’s Human Ethics Advisory Group, and written informed consent was provided. Due to technical issues collected data from one participant were discarded.

Table 3-1: Demographic data of participants

Subject	*Height (cm)*	*Weight (kg)*	*Age (yrs)*	*BMI ($\frac{kg}{m^2}$)*
S1	175	65	22	21.2
S2	177	90	33	28.7
S3	173	84	33	28
S4	177	87	35	27.8
S5	183	68	26	20.3
S6	175	65	28	21.2
S7	186	90	28	26
S8	175	65	36	21.2
S9	176	76	26	24.5
S10	177	65	28	20.7
S11	160	67	28	26.2
S12	175	70	22	22.8
S13	174	75	25	24.8
S14	170	48	31	16.6
S15	180	80	32	24.7
S16	180	84	34	25.9
S17	177	80	25	25.5
S18	173	62	25	20.7
S19	178	85	22	26.8

Prior to data collection, the participants received an explanation of the project. Sample size was estimated based on the data from previous relevant studies in the literature [79, 87, 200, 209, 224, 230, 258, 290, 313-315]. Power analysis was also performed in G*Power (Figure 3-1) [316] for one group of participants and 10 different walking conditions with Effect size of 0.253 (suggested by Cohen (1969) [317]) and α= 0.05.

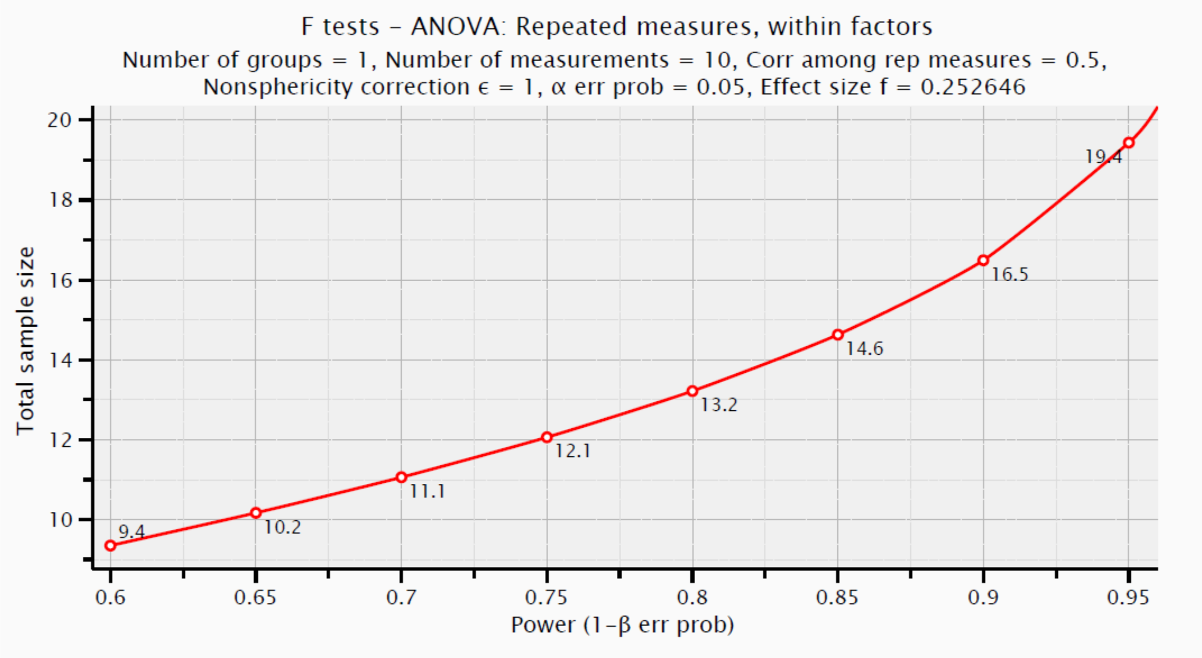

Figure 3-1: Distribution plot of power analysis; total sample size vs power.

3.3.2 Experimental Protocol

- Participant sign consent form
- Subject Preparation
- Marker attachment
- Determination of preferred walking speed (served also as a warmup on treadmill)
- Adjustment of the harness respect to participant's body size
- Participant perform walking trials, while data collected

3.3.2.1 Subject Preparation

Before starting data collection, participants were asked to wear skintight bike shorts and be topless. To reduce footwear effects, participants were also provided with standardized running shoes in different sizes (FILA, South Korea), and asked to wear the best-fitting pair.

3.3.2.2 Marker Attachment

Sixty retro-reflective markers were attached to bony landmarks using a modified Helen-Hayes marker-set [318] comprising four markers on the head segment, two on the shoulders, two on the sternum and C7, four on the lateral/medial epicondyles of the elbows, four on the ulnar/radius sides of the wrists, four on the anterior/posterior iliac spines, one on the sacrum, two on the greater trochanters, four on the thighs, four on the lateral/medial epicondyles of the knees, three on the shanks, four on the lateral/medial malleolus, four on the great toes and heels, and four on the second and fifth metatarsals.

3.3.2.3 Walking Speed

Preferred walking speed was determined for each participant prior to data collection according to the procedure used by the Dingwell and Marine (2006) [174]. Specifically, each participant was asked to walk on the treadmill at a very low speed, and then the treadmill speed was increased gradually by the investigator until the optimal walking speed was confirmed by the participant.

3.3.2.4 Gait Protocol

Participants performed gait experiments on a front-to-back dual belt motorized treadmill (AMTI, USA). The trajectories of attached markers were simultaneously acquired using an 11-camera video motion analysis system sampling at 100 Hz (Vicon, Oxford, UK). Participants wore a safety harness tethered to an overhead frame to avoid injury during testing in the event of a fall.

Participants walked at their self-selected speed (baseline walking) as well as walking whilst simultaneously (i) reading on a cell phone (ii) texting, and (iii) talking on a cell phone. Tasks were performed using a Samsung Galaxy Ace (Samsung Electronics Co. Ltd., South Korea), and participants were free to use their preferred cell phone usage strategy. Text provided for reading comprised 845 words and was designed to engage participants for approximately four minutes [319]. For the texting trial, participants were asked to manually enter text on their cell phone about a university-related activity they had recently engaged in. For cell phone talking, a series of generic questions were asked [320], and participants requested to respond frankly and without time limit (Table 3-2). Each task was performed over a maximum four minutes, with rest periods between tasks.

Table 3-2: Questions asked of participants during talking trials [320].

What were your favourite classes taken during your studies?
Which classes did you like least and why?
What classes are you taking next semester?
Have you studied previously? If so, did you enter college knowing your major?
What are your future career plans?
What kinds of jobs have you had?
What has been your favourite job and why?
What is your favourite sport?
Who is your favourite sports player?

3.3.3 Data Processing

Three-dimensional motion trajectories of attached retroreflective markers were reconstructed from the camera data in Vicon Nexus for all trials of each participant. All occluded signals were compensated using available gap filling approaches in the software. Kinematics were then filtered using a 4th order, low-pass Butterworth filter with a cut-off frequency of 6 Hz. Gait events were detected by adapting the procedure proposed by Zeni et al. (2008) [321]. According to this method, HC was defined when the horizontal distance between sacrum and heel markers has its maximum value, and TO was defined when the distance between sacrum and toe markers has its minimum value.

3.3.4 Modelling and Kinematic Analysis

A subject-specific skeletal model was developed in Visual 3D (C-Motion, USA) for each participant using the pre-processed static data in Vicon Nexus. Each model was then used for kinematic analysis to calculate joint angular displacements, velocity and acceleration for the hip, knee and ankle joints, as well as linear displacement, velocity and acceleration for the head and trunk segments for each walking trial.

3.3.5 Spatiotemporal Measures

Step length was defined as the anteroposterior distance between heel markers at the instant of HC. Step width was determined as the mediolateral distance between heel markers at the instant of HC. Step time was calculated as the duration between consecutive heel contacts, and double support time was defined as the period that both feet contact the walking surface. Mean and standard deviation of step length, step width and step time were quantified for each testing condition.

3.3.6 Gait Variability

Gait variability was measured using a Detrended Fluctuation Analysis of spatiotemporal variables. In order to quantify DFA for a time series, first the cumulative sum of the deviation from the mean at each t is calculated using

$$y(i) = \sum_{t=1}^{i} [x(t) - \bar{x}] \qquad for\ i = 1,2,3,\ldots,N$$ Equation 3-1

where N is the length of time series. Then, $y(i)$ is separated into n non-overlapping windows of the same size (s). As N may or may not be a multiple of s there may be a small residual after dividing $y(i)$ to n congruent windows. In order to avoid this, the dividing process is repeated one more time from another side. Therefore, the number of intervals will be $2n$. In the next step, the local trend (y_n) at each interval is calculated using least square fitting of the data. The fluctuation function can be computed from

$$F(n) = \sqrt{\frac{1}{N}\sum_{i=1}^{N}[y(i) - y(i)_n]^2} \qquad \text{Equation 3-2}$$

where $y(i)_n$ is the local trend at each interval.

The slope of the graph of $F(n)$ as a function of n in log-log scale is called the scaling component or self-similarity exponent α. The scaling exponent is the indicator of long-range correlation. On the basis of different quantities of α, the correlation can be characterized. Specifically, $\alpha = 0.5$ when the time series is random or uncorrelated and behaves like Gaussian (white) noise. If $\alpha = 1$, the time series will behave like pink noise, or its power spectral density has an inverse relationship with frequency. The time series may be estimated as Brownian noise provided that $\alpha = 1.5$. As long as the alteration of α is limited between 0.5 and 1, the time series will have a long range correlation [126, 127].

3.3.7 Margin of Stability

MoS for each test condition was calculated at HC and MFC, which are gait events that have been shown to be associated with backward and forward balance loss, respectively [154, 162]. HC was detected using the maximum horizontal distance between sacrum and heel markers [321], while MFC was identified at mid-swing phase when the vertical trajectory of the first distal phalange marker was minimum [40]. The extrapolated centre of mass (XCoM) was calculated in the transverse plane using the whole-body CoM position and velocity from

$$X_{CoM,x} = x + \frac{\dot{x}}{\omega_0}\,, \qquad \omega_0 = \sqrt{\frac{l}{g}} \qquad \text{Equation 3-3}$$

where x and $\dot{x}$ are the CoM position and velocity, respectively; ω_0 is the natural frequency of the inverted pendulum; *l* is the leg length, which was calculated as the vertical distance between the greater trochanter and the lateral malleolus markers; and *g* is the universal gravitational constant. MoS was defined as the distance between the XCoM and the BoS boundary in the mediolateral and anteroposterior directions.

Many gait stability models have maintained a focus on centre of mass control within a stability base defined by foot position during double support. The approach adopted in the present study *is important to consider centre of mass control dynamics throughout the gait cycle*. Destabilizing events arise continuously when walking and threats to stability during the swing phase, when there is only single-foot ground contact, are more likely to be associated with critical instability leading to a fall. This observation was made almost 20 years ago in a seminal research study by Patla and colleagues [322] who introduced the concept of a "projected" or "virtual" base of support to characterise swing phase stability. Their model [322] also defined a supporting base using the relative position of both feet but, most important, is that in their formulation swing phase foot position was "…projected to the ground to define this area" (p.613). Similar in concept to Patla et al. [322], in the model presented below, a single critical swing phase event has been identified (MFC) and at MFC the line between the first distal phalange marker of the stance foot and projection of similar marker of the swing foot was used to define the virtual anterior BoS boundary [322], while the position of heel marker of the leading foot in the sagittal plane at the instant of HC was used as the posterior BoS border of BoS. In the frontal plane, the mediolateral position of the lateral malleoli marker attached to the leading foot was used as the lateral border of BoS [277]. Since forward balance loss occurs along the direction of CoM velocity, MoS at MFC was defined as the distance between the XCoM and the anterior BoS boundary in the direction of CoM velocity (Figure 3-2A) [323]. At HC, anteroposterior MoS was defined as the distance travelled by the XCoM in the sagittal plane relative to the posterior BoS border (Figure 3-2B). Mediolateral MoS was also quantified as the distance between XCoM and lateral BoS boundary in the frontal plane (Figure 3-2B). Larger positive MoS values indicate lower risk of balance loss, while the larger negative values imply a higher risk of balance loss. MoS was calculated separately for each limb at HC and MFC.

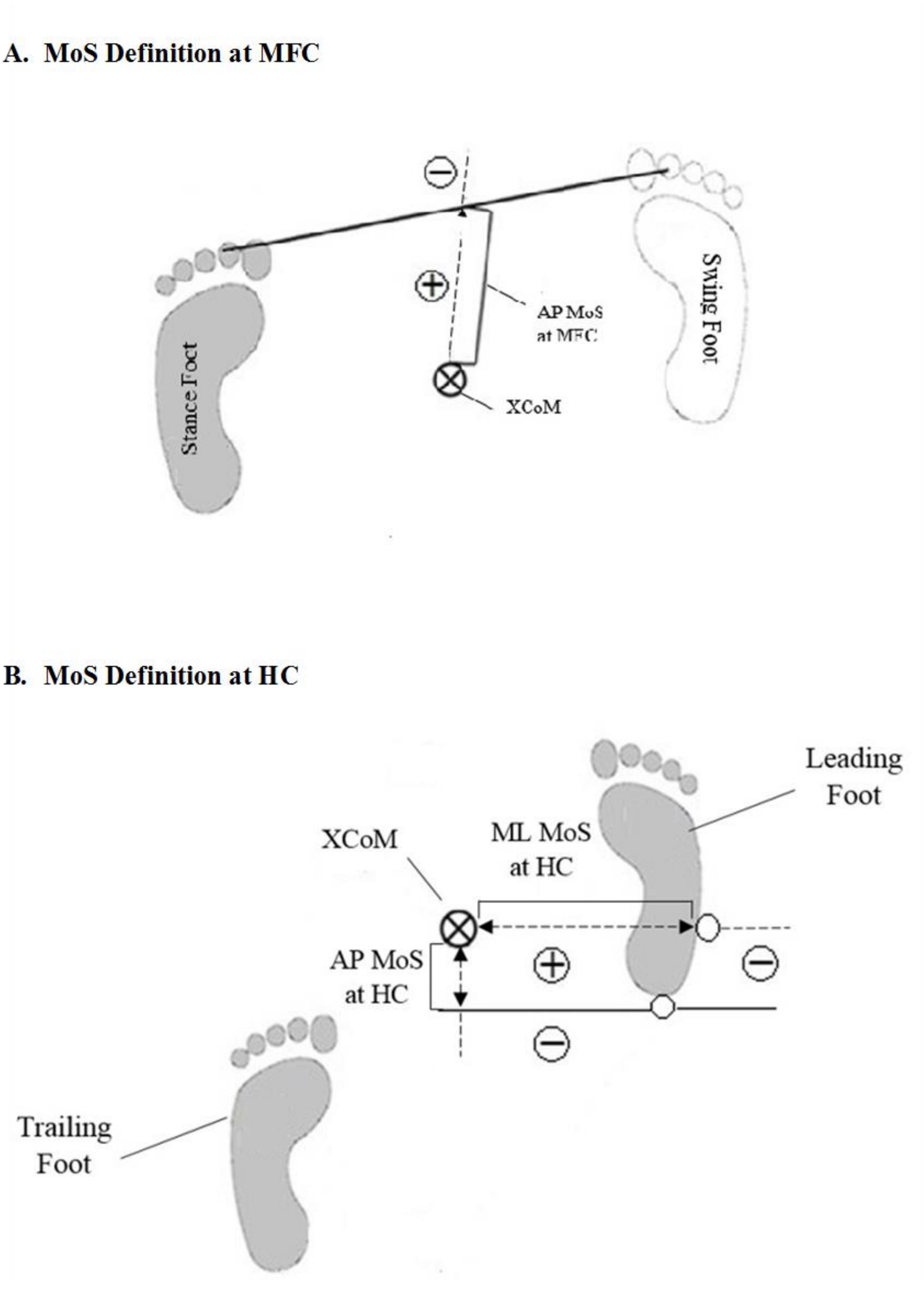

Figure 3-2: Schematic diagram illustrating definition of MoS at HC and MFC in the transverse plane view. At MFC, MoS is defined as the travelled distance by the XCoM along with the direction of velocity relative to the anterior border of BoS made by a line between two great toes. At HC, the anteroposterior excursion of the XCoM relative to the posterior border of BoS (the position of heel marker of the leading foot in the sagittal plane) defines the anteroposterior MoS, while the mediolateral excursion of the XCoM relative to the lateral border of BoS formed by lateral malleoli of the leading foot defines the mediolateral MoS.

3.3.8 Local Stability

Hip, knee and ankle joint kinematics were calculated using inverse kinematics (Visual 3D, C-Motion, USA). Local dynamic stability was subsequently evaluated using LyE by employing a reconstructed 5-dimensional time-delayed state space of the angular velocity of the hip, knee and ankle joints as well as the linear velocity of the head CoM and C7 marker (seventh cervical vertebra) in the anteroposterior, mediolateral and vertical directions using

$$S(t) = [x(t), x(t+\tau), x(t+2\tau), \dots \dots, x(t+(d_e-1)\tau]$$ Equation 3-4

where τ is the time delay and d_e is the embedding dimension.

The time delay for each set of data was estimated using the Average Mutual Information (AMI) function, as described previously [185]. The rate of divergence of nearest neighbouring trajectories was estimated using LyE with

$$D_j(i) = C_j e^{(LyE)*t}$$ Equation 3-5

where $D_j(t)$ is the distance between adjacent markers trajectories and C_j is the initial distance between the trajectories.

Taking the Napierian logarithm of both sides and least-square curve fitting to the mean data yields

$$y(i) = \frac{1}{\Delta t}\langle \ln d_j(i) \rangle = LyE.i + c, c = \frac{\ln C_j}{\Delta t}$$ Equation 3-6

where LyE is the slope of the fitted line and used as proxy for stability. The short-term LyE was calculated for 150 strides [25], with higher LyE values indicating more divergence of nearest neighbours and lower stability, and lower LyE values indicating more convergence of adjacent neighbours and higher stability.

3.3.9 Orbital Stability

Orbital dynamic stability was also calculated using MaxFM from the linear velocity of the head CoM and C7 marker, as well as the angular velocity of the hip, knee and ankle joints in the sagittal, frontal and transverse planes for each test conditions. The time of each walking stride was normalized to 101 points, and a Poincaré plot was constructed to map the state of the system at stride k into stride $k + 1$ with

$$S_{k+1} = P(S_k) \qquad \text{Equation 3-7}$$

The fixed point (S^*) was then calculated by averaging the trajectories of the system over all strides using

$$S^* = P(S^*) \qquad \text{Equation 3-8}$$

The changes in $[S - S^*]$ between stride k to stride $k + 1$ was computed using eigenvalues of a Jacobian matrix (J), calculated with

$$[S_{k+1} - S^*] = J(S^*)[S_k - S^*] \qquad \text{Equation 3-9}$$

The eigenvalues of the Jacobian matrix J are the Floquet Multipliers. When the maximum Floquet Multiplier is less than one, the system is considered stable [203].

3.3.10 Slip and Trip Propensity Measures

Backward slip propensity was quantified for each walking condition using the Required Coefficient of Friction (RCoF) which is the minimum required coefficient of friction to prevent slipping [58]. The anteroposterior and vertical components of ground reaction force were used to calculate RCoF:

$$RCoF = \frac{F_{Anteroposterior}}{F_{Vertical}} \qquad \text{Equation 3-10}$$

Higher RCoF indicates greater risk of slipping.

Trip propensity was evaluated for each testing condition using the height of foot clearance with respect to the walking surface at mid swing. Lower MFC height implied greater risk of tripping.

3.3.11 Detection of the Most Destabilizing Type of Cell Phone Usage during Walking

To detect which type of cell phone usage while walking affected gait pattern the most, a customized version of a newly developed approach by Roeles et al. (2018) was employed [277]. According to this method, the average of MFC height and MoS during baseline walking were first calculated for each participant. The level of distraction due to a secondary task during walking was then quantified by calculation of differences between MoS and spatiotemporal variables at each of the first four post-perturbation steps (right and left) during dual-task walking trials with and averaged values for right and left steps during baseline walking:

$$Deviation = \sum_{i=1}^{2}\sum_{j=1}^{4}\sqrt{((Ave(i) - DT(i + (j-1) * 2))^2} \qquad \text{Equation 3-11}$$

where Ave is the average value of MFC height and MoS during baseline trial for i^{th} side (with $i = 1$ representing the left side and $i = 2$ representing the right side), and DT is the value of MFC height and MoS during dual-task trials for each side at the j^{th} step.

Since accommodation of perturbation starts from the first steps of walking, and participants adapt to changes and challenges applied by each type of perturbation after a number of steps (which varies person to person), a fixed number of initial steps (the first four steps) at the beginning of each dual-tasking trial were selected to evaluate the initial response of individuals to each type of perturbation.

3.3.12 Statistical Analysis

A repeated measure analysis of variance (ANOVA) was performed to investigate the influence of cell phone usage on dynamic walking stability. The dependent variables were mean and variability of spatiotemporal measures, RCoF, MFC height, MoS, LyE, MaxFM, DFA scaling component and Deviation, while the independent factors were the test conditions which included baseline walking and walking while cell phone reading, texting, and talking, as well as anatomical movement directions (just in stability measures). Significance level was defined as $p \leq 0.05$. All statistical analyses were performed using SPSS 25 (SPSS Inc., Chicago, IL).

3.4 Results

3.4.1 Spatiotemporal Measures

In response to perturbations associated with talking while walking, participants tended to walk with significantly greater step width ($p=0.008$), step length ($p=0.007$) and step time ($p=0.014$) (Table 3-3). Texting and reading while walking were also associated with slightly increased step parameters except double support time compared to baseline walking. However, none of the observed differences were statistically significant ($p>0.05$). Double support time did not differ significantly between dual-task walking trials and baseline walking ($p>0.05$). Participants exhibited significantly greater step width variability during talking (mean difference: 4.2 mm, $p=0.001$), texting (mean difference: 4.4 mm, $p<0.001$) and reading (mean difference: 3.6 mm, $p<0.001$) while walking relative to baseline walking (Table 3-3). Higher non-significant variability of step length, step time and double support time were also found during dual-task walking trials compared to walking with no perturbation ($p>0.05$).

Table 3-3: Summary of the spatiotemporal parameters during walking trials, including Baseline walking, Talking, Reading and Texting while walking

	Baseline	*Talking*	*Reading*	*Texting*
Step length (mm)	515.4 ± 79.7	529.9 ± 73.2	519.9 ± 79.4	520.02 ± 75.6
Step length variability (mm)	19.9 ± 6.2	22.8 ± 7.8	21.5 ± 10.9	22.3 ± 7.5
Step width (mm)	84.5 ± 31.5	96.9 ± 28.7	85.8 ± 32.5	85.9 ± 29.3
Step width variability (mm)	21.6 ± 5	25.9 ± 4.9	25.2 ± 4.9	26 ± 5
Step time (ms)	609.9 ± 55.0	628.8 ± 52.5	615.2 ± 51.6	615.8 ± 51.2
Step time variability (ms)	23.2 ± 14	28.06 ± 13.2	24.7 ± 18.5	24.9 ± 13.2
Double support time (ms)	223.5 ± 45.7	235.9 ± 51.5	227 ± 40.5	221.2 ± 37.8
Double support time variability (ms)	40.8 ± 32.9	55.1 ± 39.03	50.2 ± 34.8	43.3 ± 31.7

3.4.2 Gait Variability

Step width, step time, double support time and step length time series exhibited strong statistical persistence during all walking conditions (DFA Exponent >> 0.5) (Figure 3-3). DFA Exponent of step time and step length time series did not differ significantly between dual-task trials and baseline walking (p>0.05). For step width, decreased DFA exponent was observed during texting, reading and talking while walking, but it was not statistically significant (p>0.05).

3.4.3 Margin of Stability

Increased mediolateral MoS was observed during talking while walking compared to baseline walking (mean difference: 0.003 m, p<0.05) (Figure 3-4A); however, MoS did not differ significantly in the mediolateral direction due to either texting or reading while walking compared to walking under no perturbation (p>0.05). Anteroposterior MoS at HC appeared to be unaffected during dual-tasking trials compared to baseline walking (p>0.05) (Figure 3-4B).

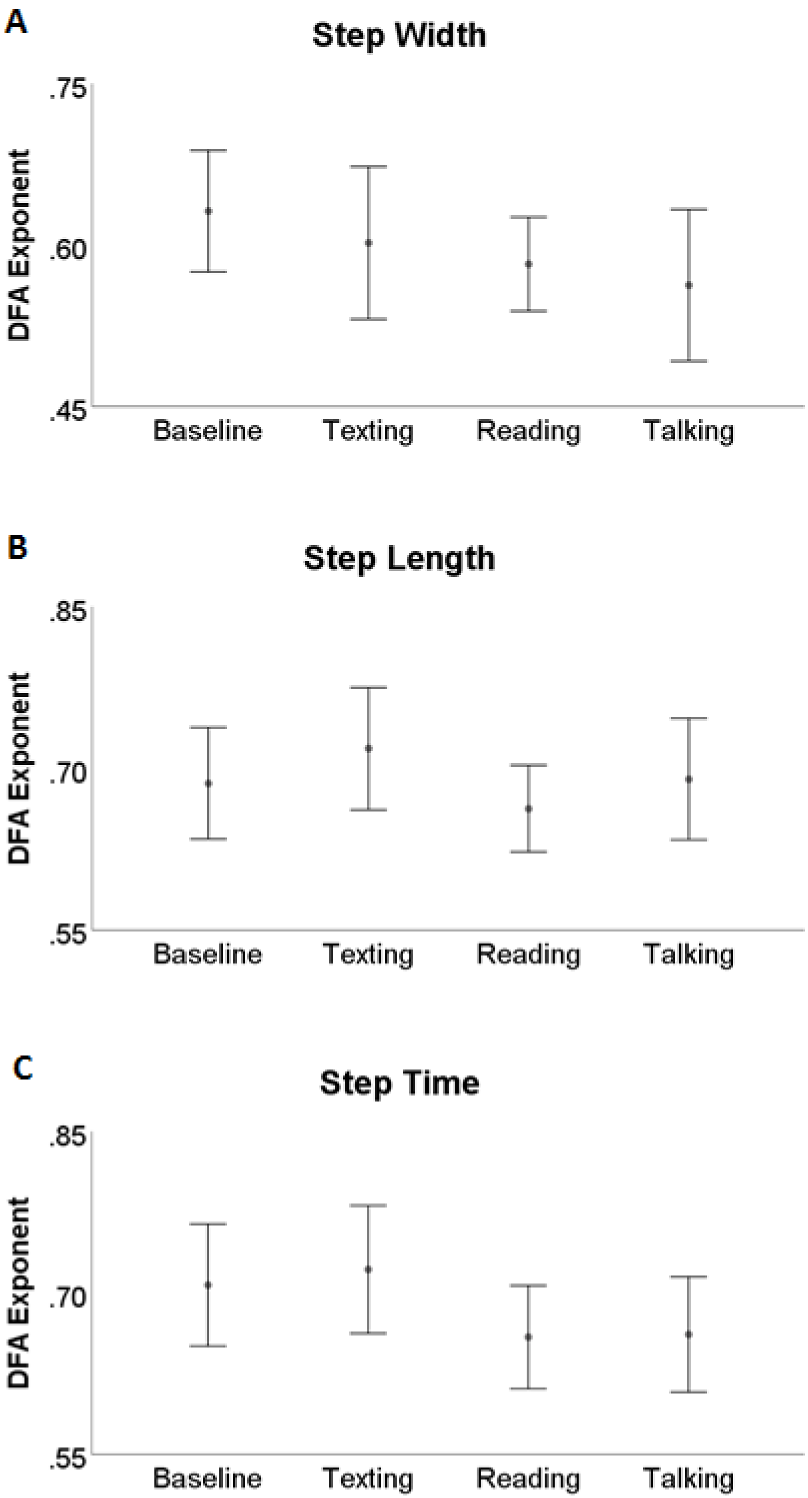

Figure 3-3: DFA Exponent values for time series of step width (A), step length (B) and step time (C) during baseline walking, texting, reading and talking. When the DFA Exponent falls between 0.5 and 1, the time series will have a long-range correlation. Error bars denote one standard deviation of DFA Exponent values in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$.

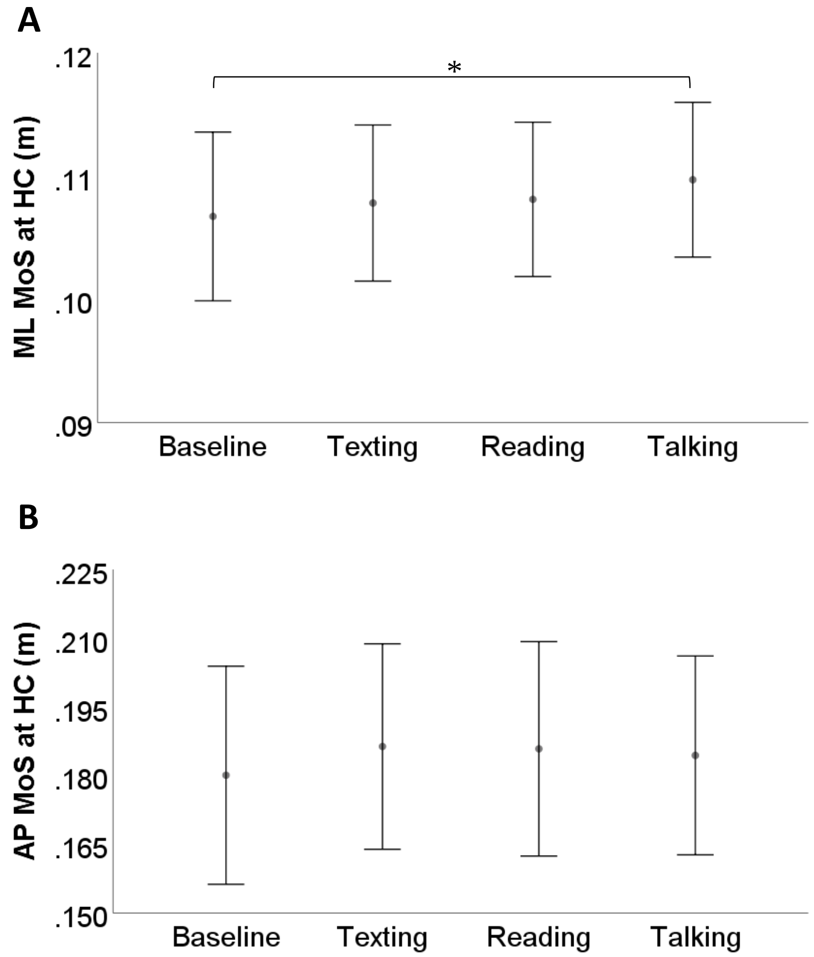

Figure 3-4: Mean mediolateral (A) and anteroposterior (B) MoS values (m) at HC during baseline walking, talking, reading and texting trials. Greater MoS values indicate less chance of backward balance loss. Error bars denote one standard deviation of MoS values during each condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral.

At MFC, MoS decreased in response to perturbations associated with talking while walking compared to baseline walking, but it was not statistically significant ($p>0.05$). Significantly smaller MoS values at MFC were detected during talking while walking compared to texting (mean difference: 0.008 m, $p<0.001$) and reading (mean difference: 0.007 m, $p<0.001$) while walking trials (Figure 3-5).

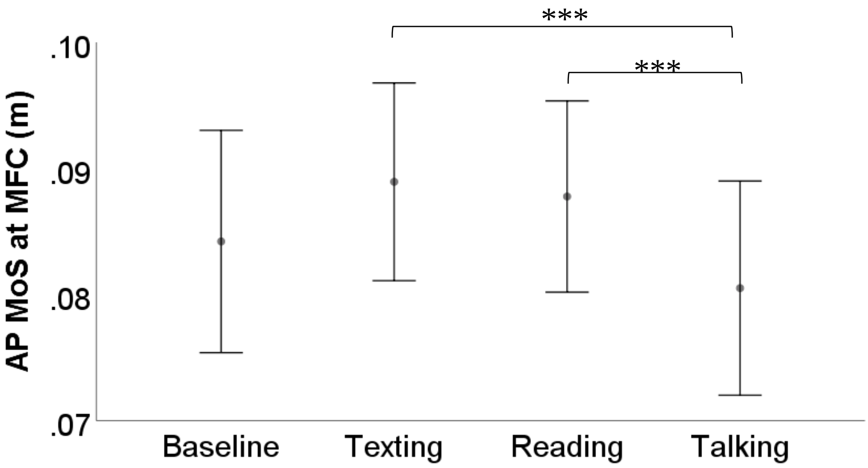

Figure 3-5: Mean anteroposterior (AP) MoS values (m) at MFC during baseline walking, talking, reading and texting trials. Greater MoS values indicate less chance of backward balance loss. Error bars denote one standard deviation of MoS values during each condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001.

3.4.4 Local Stability

Texting while walking caused increase in LyE of the head movement in the mediolateral direction compared to that during baseline walking and approached statistical significance (p=0.052) (Figure 3-6A). Reading and talking during walking also resulted in increased head LyE in the mediolateral direction, but was not statistically significant (p>0.05). There were no significant changes in LyE values of the head motion in the anteroposterior and vertical directions during dual-task trials relative to baseline walking (p>0.05) (Figure 3-6B & C).

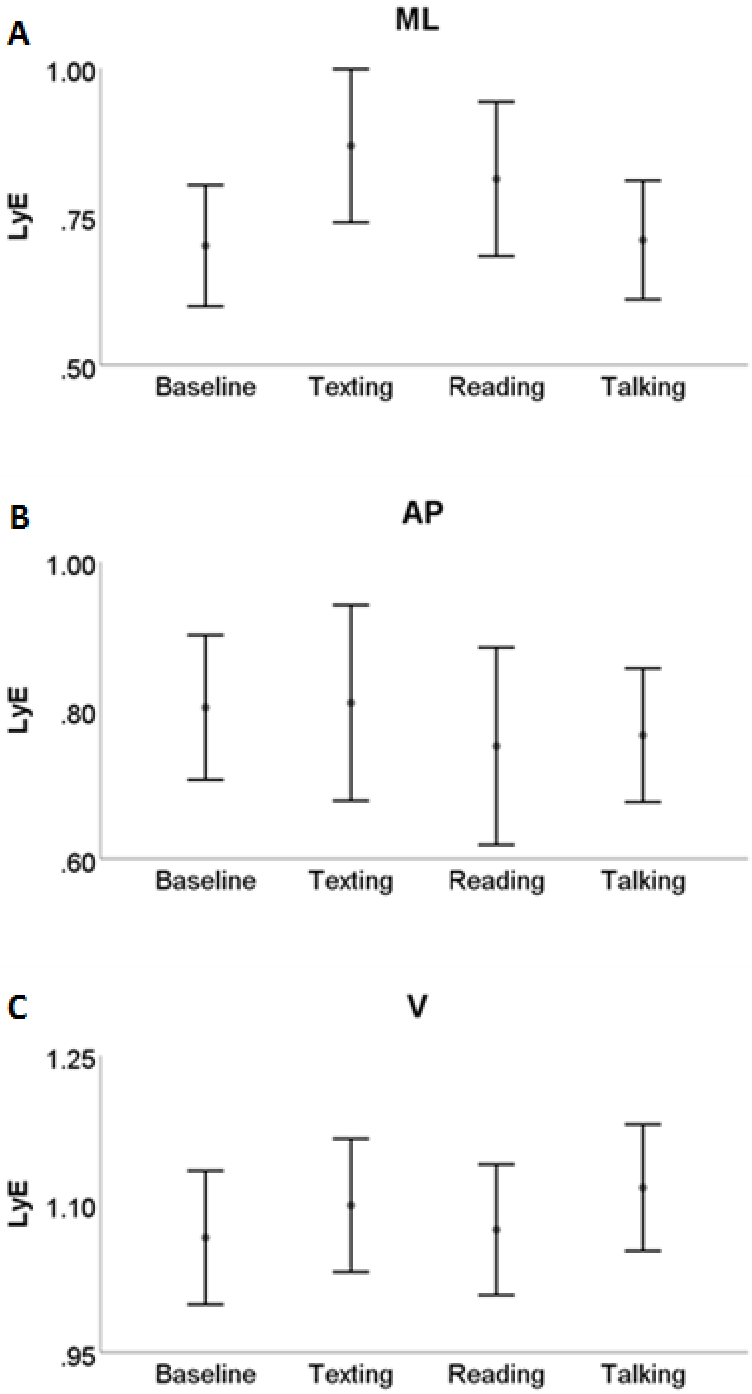

Figure 3-6: LyE values for the head in the mediolateral (A), anteroposterior (B) and vertical (C) directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

Considering local stability of the trunk segment, LyE values did not vary in the mediolateral, anteroposterior and vertical directions significantly by performing an additional cell phone related task during gait compared to walking under no perturbation ($p>0.05$), except significant increase in trunk LyE values in the vertical direction due to texting while walking($p<0.001$) (Figure 3-7A, B & C).

With respect the lower extremity joints (hip, knee and ankle), cell phone usage while walking resulted in significant changes in local stability compared to the baseline trial. In the right ankle joint, significantly larger LyE values were observed in the mediolateral direction during texting ($p<0.001$), reading ($p=0.028$) and talking ($p<0.001$) while walking compared to baseline walking (Figure 3-8A). For the right ankle joint in the anteroposterior direction, increased LyE values were also detected during texting ($p=0.007$) and talking ($p=0.004$) while walking compared to baseline walking (Figure 3-8B). Reading while walking was also associated with increased LyE of the right ankle joint in the anteroposterior direction, but was not statistically significant ($p>0.05$). Subjects' right ankle movements in the vertical direction were more locally unstable during texting ($p<0.001$) and talking ($p<0.001$) while walking than walking under no perturbation, but it was not changed significantly by reading while walking compared to baseline walking ($p>0.05$) (Figure 3-8C).

In the left ankle joint, LyE values were significantly larger in the mediolateral direction during texting ($p=0.004$) and talking ($p=0.005$) while walking compared to baseline walking (Figure 3-8D). However, reading during walking yielded no significant changes in LyE values in the mediolateral direction compared to baseline trial ($p>0.05$). In the anteroposterior direction, increased LyE values of the left ankle joint were observed during dual-task walking trials compared to single-task walking, but none of them were statistically significant ($p>0.05$) (Figure 3-8E). Subjects' left ankle movements in the vertical direction were more locally unstable during reading ($p=0.043$) and talking ($p=0.047$) while walking than walking under no perturbation, but local stability was not significantly affected by texting while walking compared to baseline walking ($p>0.05$) (Figure 3-8F).

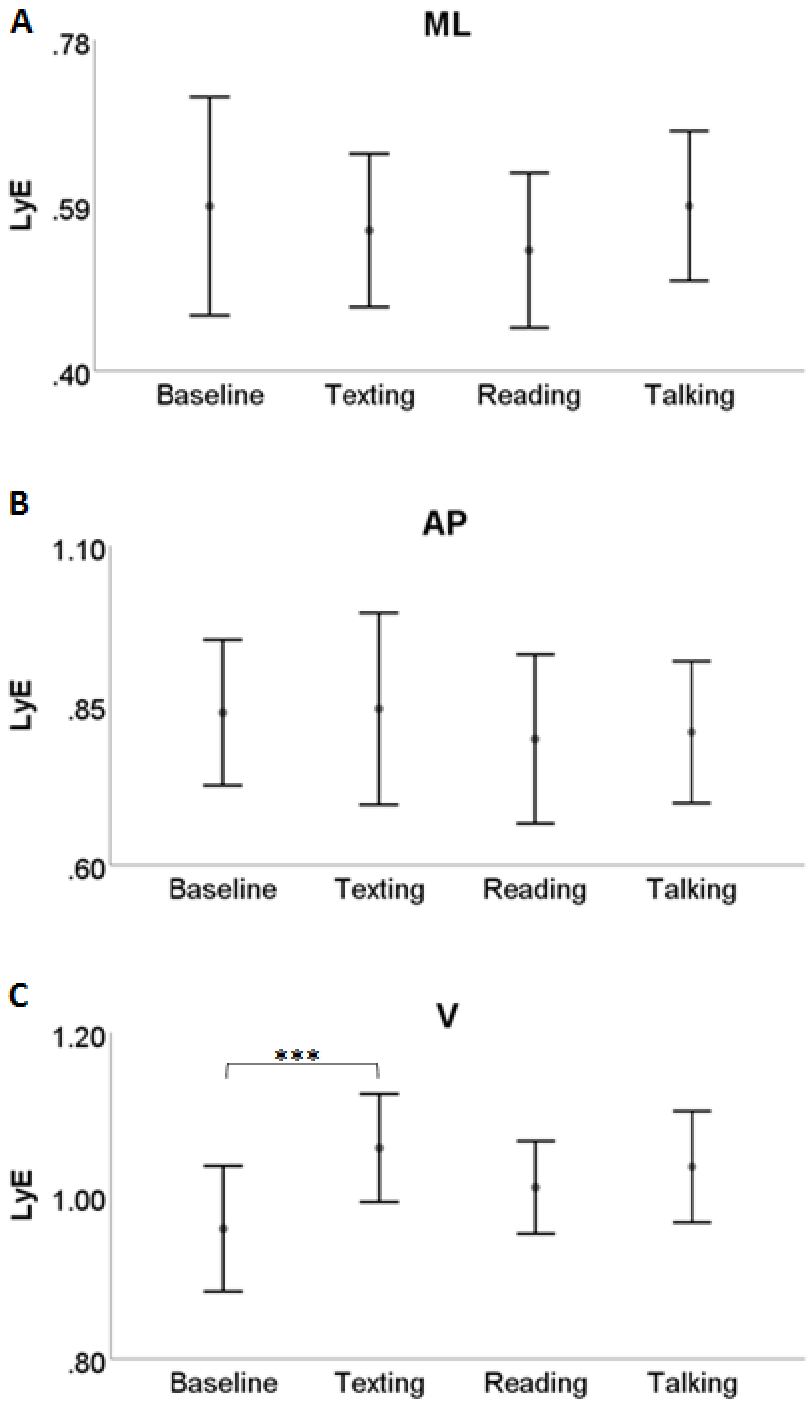

Figure 3-7: LyE values for the trunk in the mediolateral (A), anteroposterior (B) and vertical (C) directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

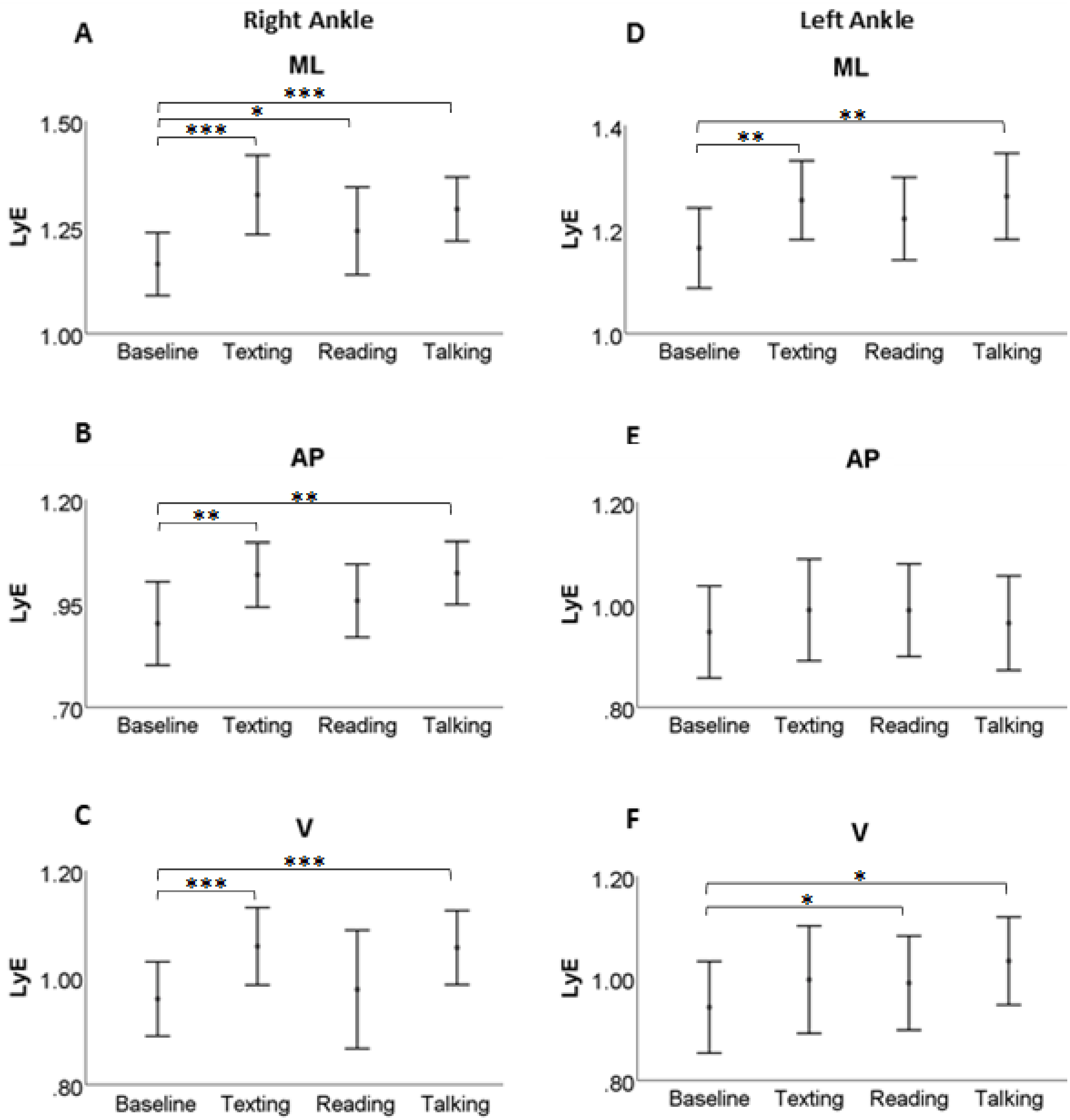

Figure 3-8: LyE values for the right (A, B, C) and left (D, E, F) ankle joints in the mediolateral, anteroposterior and vertical directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

In the right hip joint, participants exhibited significantly greater LyE values in the mediolateral direction during texting ($p=0.002$) and talking ($p=0.032$) while walking compared to baseline walking (Figure 3-9A). Reading while walking was also associated with increased

LyE of the right hip joint in the mediolateral direction, but was not statistically significant (p>0.05). In the anteroposterior direction, texting while walking resulted in nearly significant increases in LyE values of the right hip joint (p=0.056) compared to baseline walking (Figure 3-9B). However, there were no significant changes in LyE values of the right hip joint during reading and talking while walking compared to walking under no perturbation (p>0.05). Subjects' right hip movements in the vertical direction were more locally unstable during texting (p<0.001), reading (p=0.013) and talking (p=0.002) while walking than baseline walking (Figure 3-9C). In the left hip joint, increased LyE values in the mediolateral direction were detected in response to performing an additional task while walking compared to single-task walking, but none were statistically significant (p>0.05) (Figure 3-9D). For the left hip joint in the anteroposterior direction, participants demonstrated significantly larger LyE values during texting (p<0.001), reading (p=0.029) and talking (p=0.005) while walking compared to walking under no perturbation (Figure 3-9E). Subjects' left hip movements in the vertical direction were also more locally unstable during texting (p=0.003), reading (p=0.001) and talking (p<0.001) while walking than baseline walking (Figure 3-9F).

In the right knee joint, participant exhibited significantly greater LyE values in the mediolateral direction during texting (p=0.048) and talking (p=0.009) while walking compared to baseline walking (Figure 3-10A). Reading while walking was also associated with increased LyE of the right knee joint in the mediolateral direction, but it was not statistically significant (p>0.05). For the right knee joint in the anteroposterior and vertical directions, there were no significant differences between dual-task and single-task walking trials (Figure 3-10B & C).

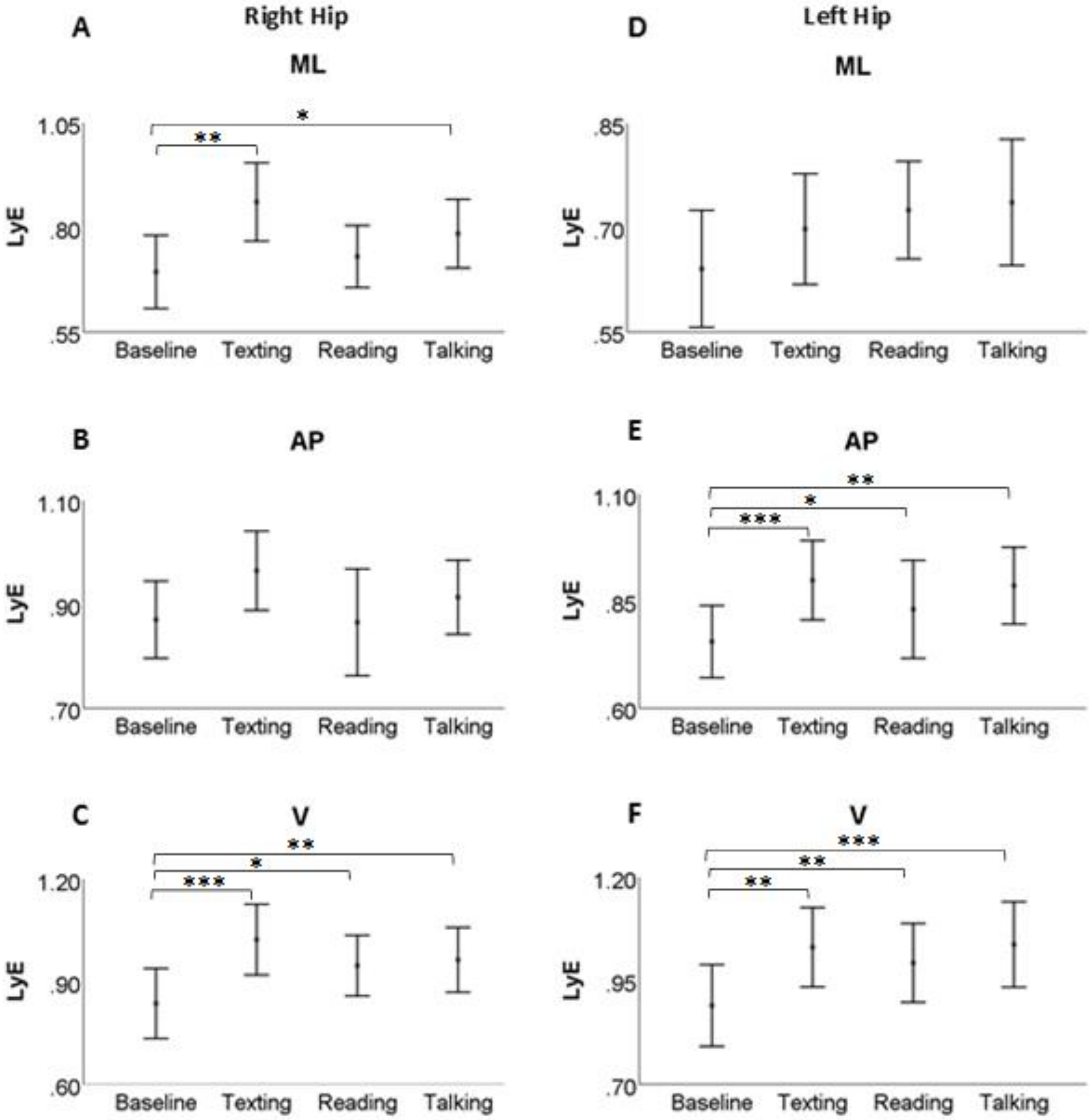

Figure 3-9: LyE values for the right (A, B, C) and left (D, E, F) hip joints in the mediolateral, anteroposterior and vertical directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

In the left knee joint, participants demonstrated significantly larger LyE values in the mediolateral direction during texting (p=0.01) and reading (p=0.037) while walking relative to baseline walking (Figure 3-10D). Talking while walking was also associated with increased LyE of the left knee joint in the mediolateral direction, but was not statistically significant (p>0.05). For the left knee joint in the anteroposterior direction, there were no significant differences between dual-task and single-task walking trials (p>0.05), except significantly greater LyE values during texting while walking compared to baseline walking (p=0.028) (Figure 3-10E). Subjects' left knee movements in the vertical direction were associated with nearly significant increases in LyE values during texting (p=0.057) and talking (p=0.055) while walking than baseline walking (Figure 3-10F). Local stability of the left knee joint motion in vertical direction was not changed during reading task compared to baseline trial (p>0.05).

3.4.5 Orbital Stability

Reading while walking resulted in increased head MaxFM in the mediolateral direction compared to baseline walking (p=0.001) (Figure 3-11A). In the anteroposterior and vertical directions, however, there were no significant changes in orbital stability of the head during dual-task trials compared to single-task walking (p>0.05) (Figure 3-11B & C). Similarly, orbital stability of the trunk segment in three anatomic directions were not significantly affected by cell phone usage during walking compared to baseline walking (p>0.05) (Figure 3-12A, B & C). For the hip, knee and ankle joints, MaxFM values in three anatomic directions did not differ significantly between dual-task walking trials and baseline walking (p>0.05), except increased MaxFM in the mediolateral direction for the right knee due to reading while walking (p=0.043) and the left hip due to talking while walking (p=0.028) (Figure 3-13, 3-14, 3-15).

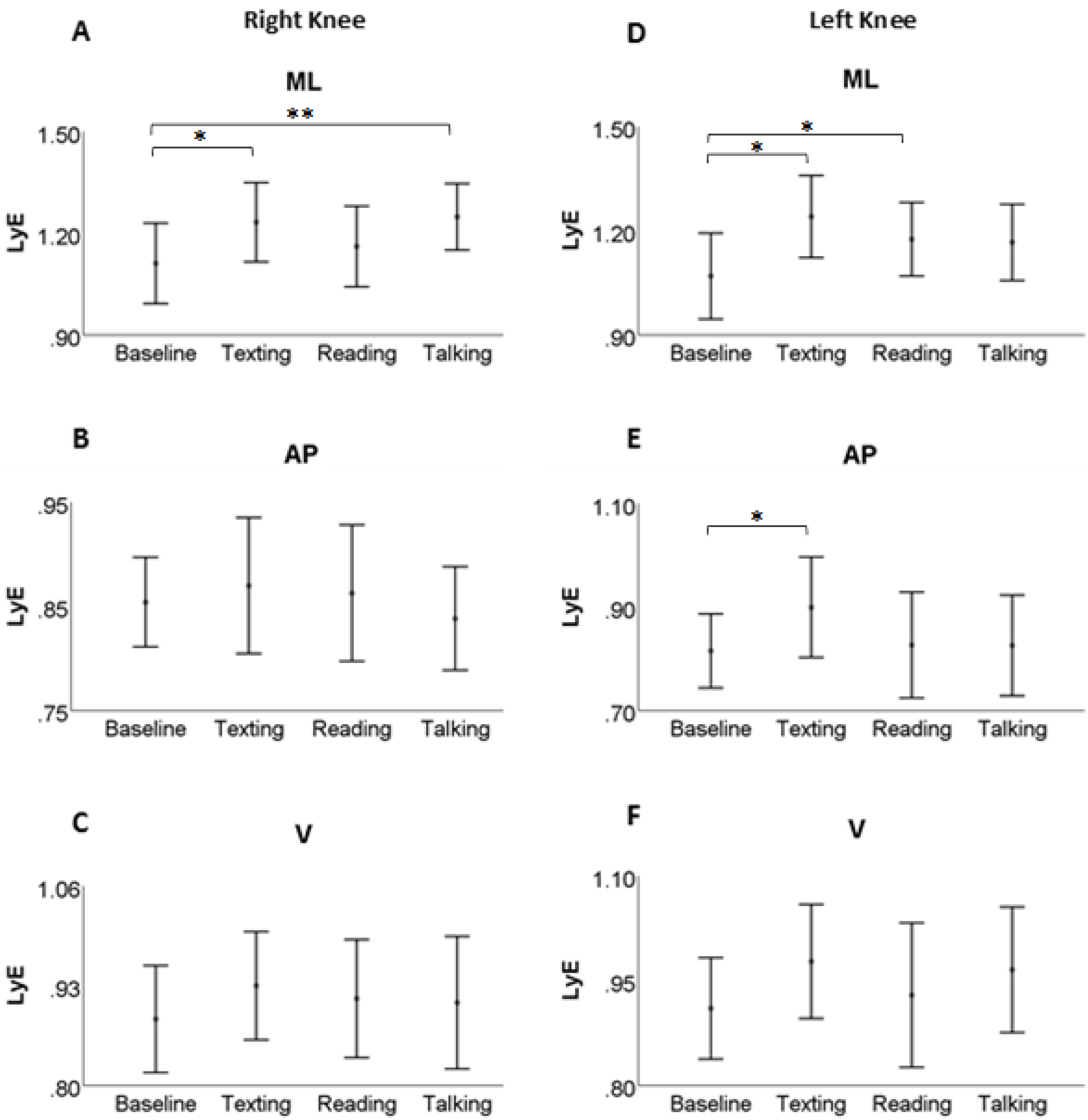

Figure 3-10: LyE values for the right (A, B, C) and left (D, E, F) knee joints in the mediolateral, anteroposterior and vertical directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

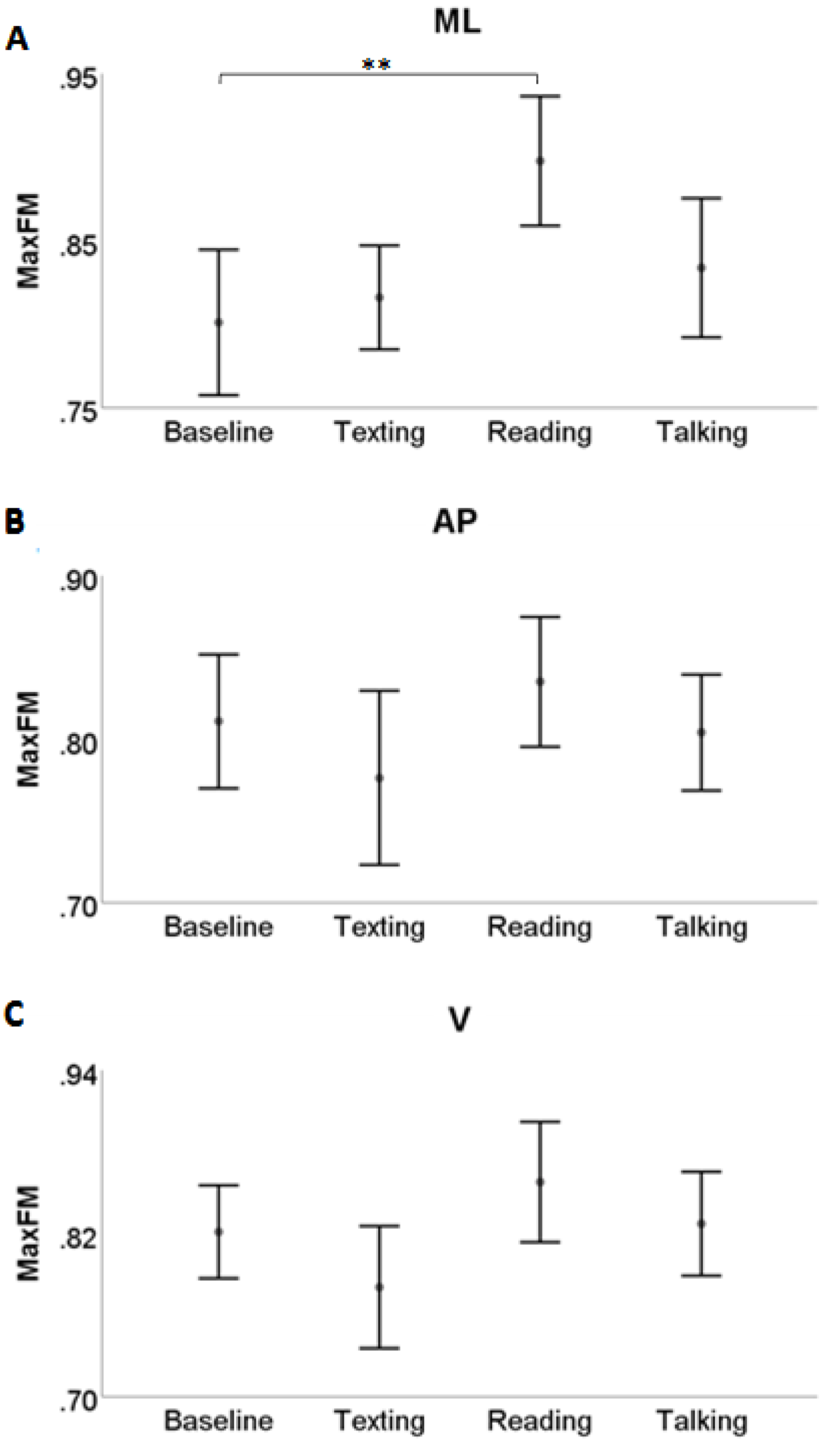

Figure 3-11: MaxFM values for the head in the mediolateral (A), anteroposterior (B) and vertical (C) directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

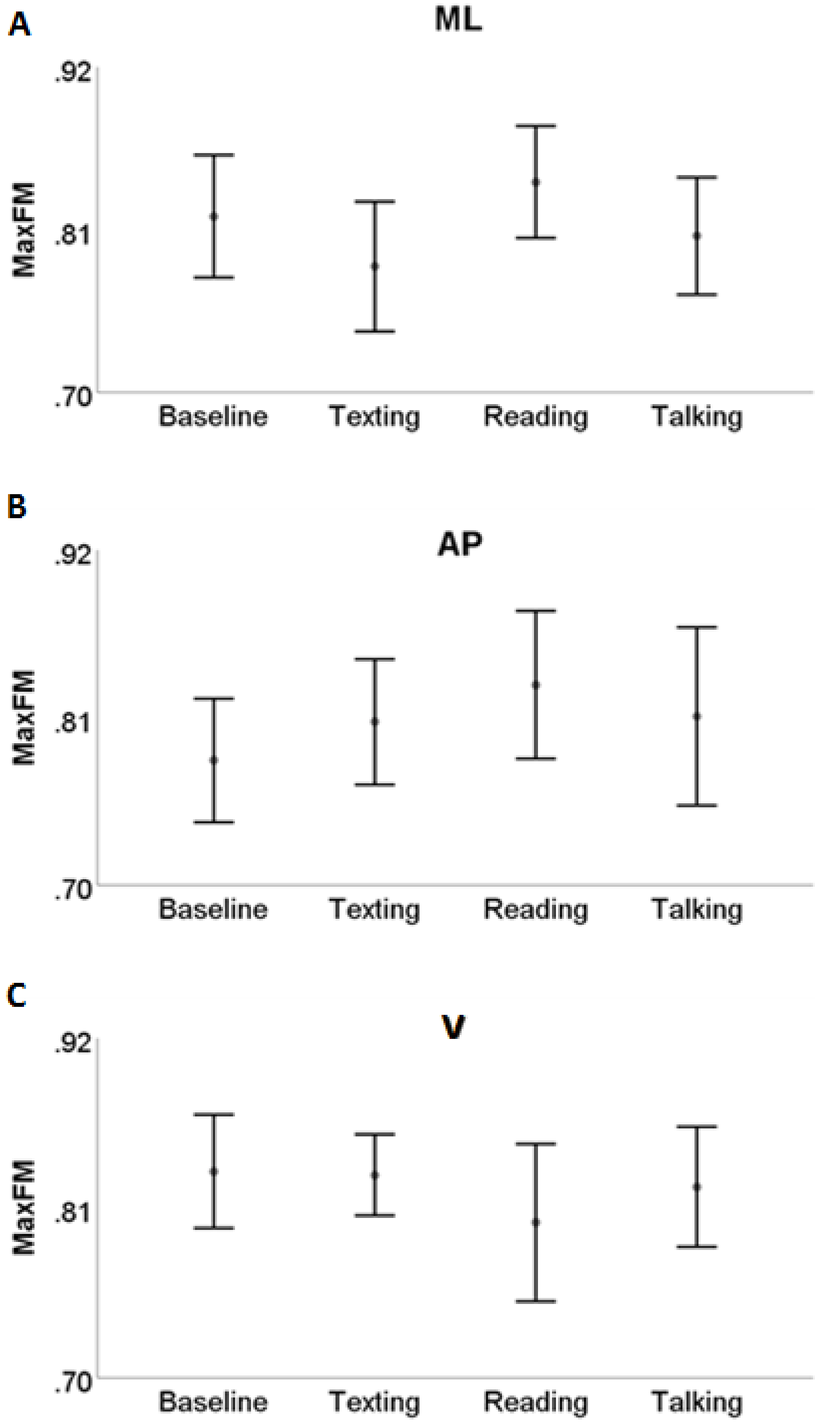

Figure 3-12: MaxFM values for the trunk in the mediolateral (A), anteroposterior (B) and vertical (C) directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

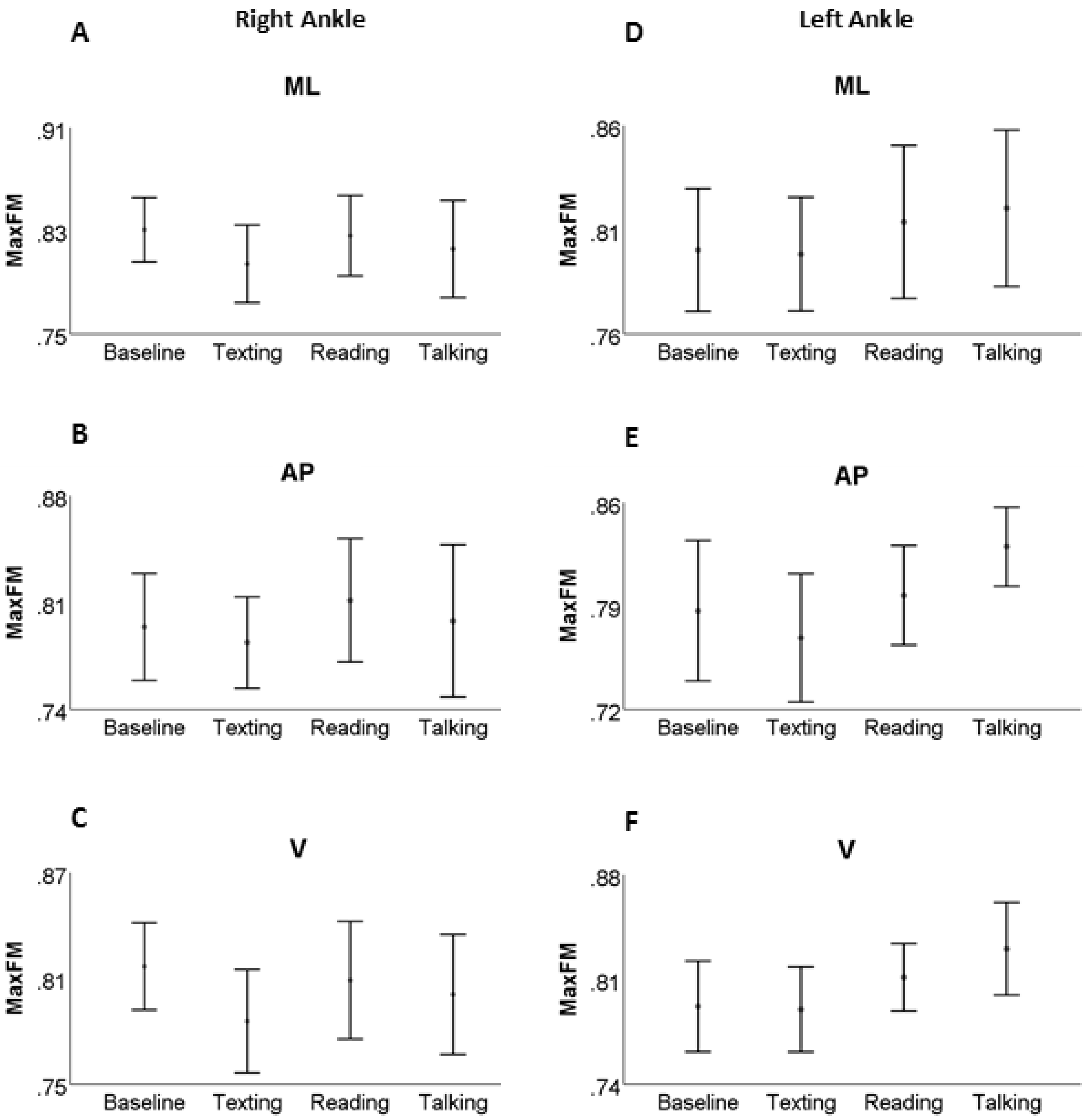

Figure 3-13: MaxFM values for the right (A, B, C) and left (D, E, F) ankle joints in the mediolateral, anteroposterior and vertical directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

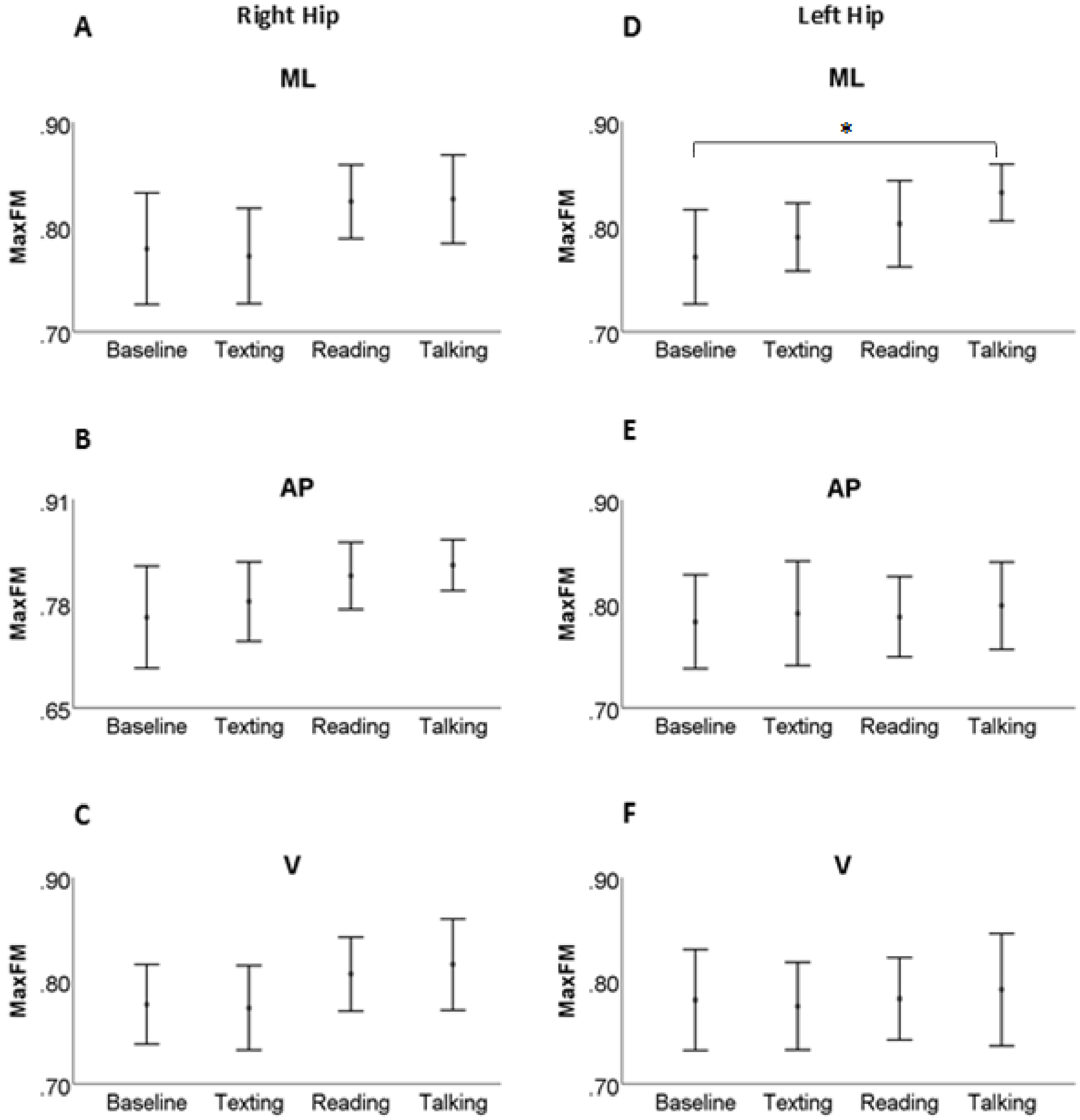

Figure 3-14: MaxFM values for the right (A, B, C) and left (D, E, F) hip joints in the mediolateral, anteroposterior and vertical directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

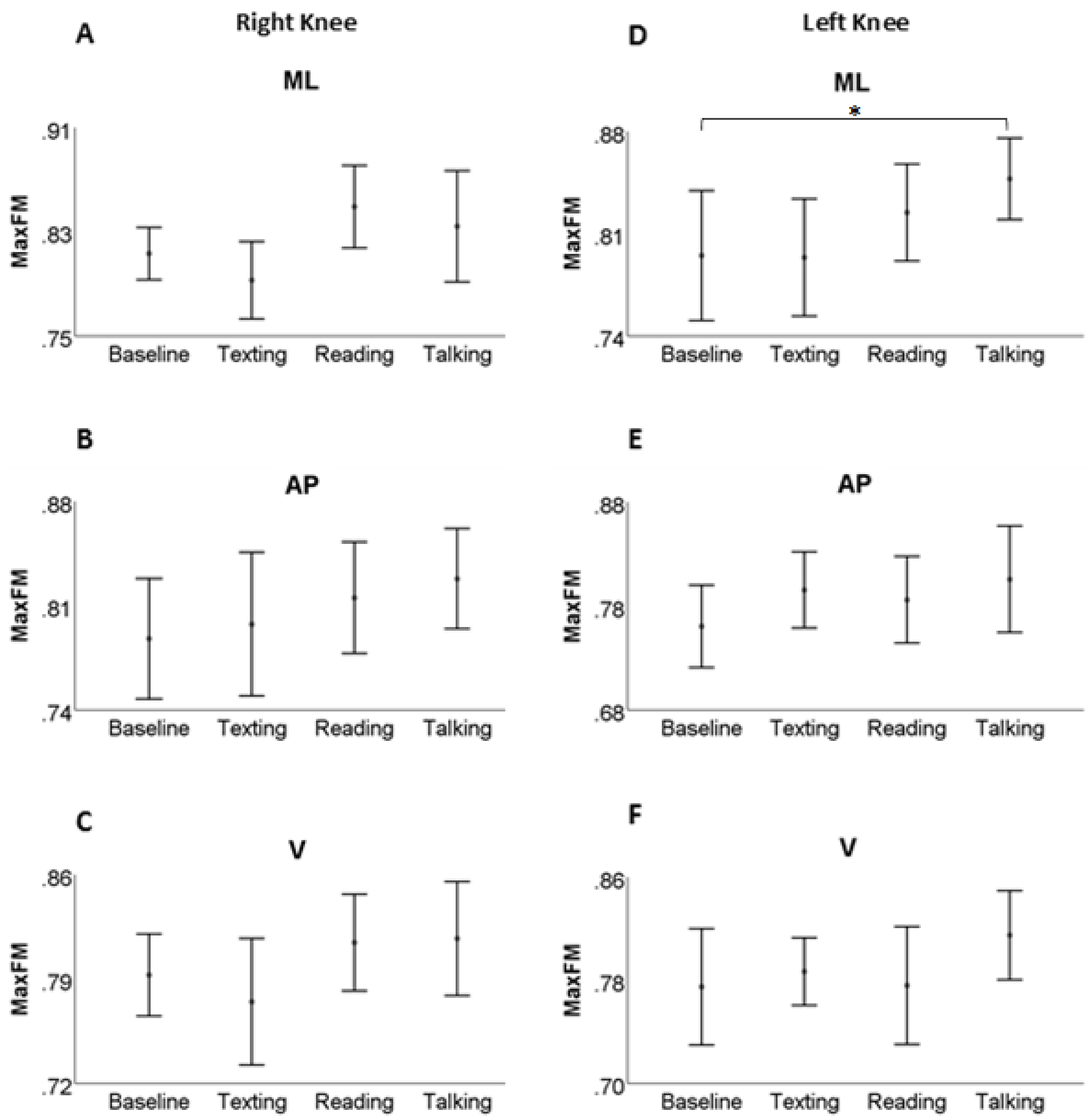

Figure 3-15: MaxFM values for the right (A, B, C) and left (D, E, F) knee joints in the mediolateral, anteroposterior and vertical directions during baseline walking, texting, reading and talking. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

3.4.6 Slip and Trip Propensity Measures

Participants exhibited smaller RCoF during texting and reading while walking compared to baseline walking, but none of these differences were statistically significant ($p>0.05$) (Figure

3-16). RCoF was exhibited no affect due to talking while walking compared to walking under no perturbation conditions (p>0.05).

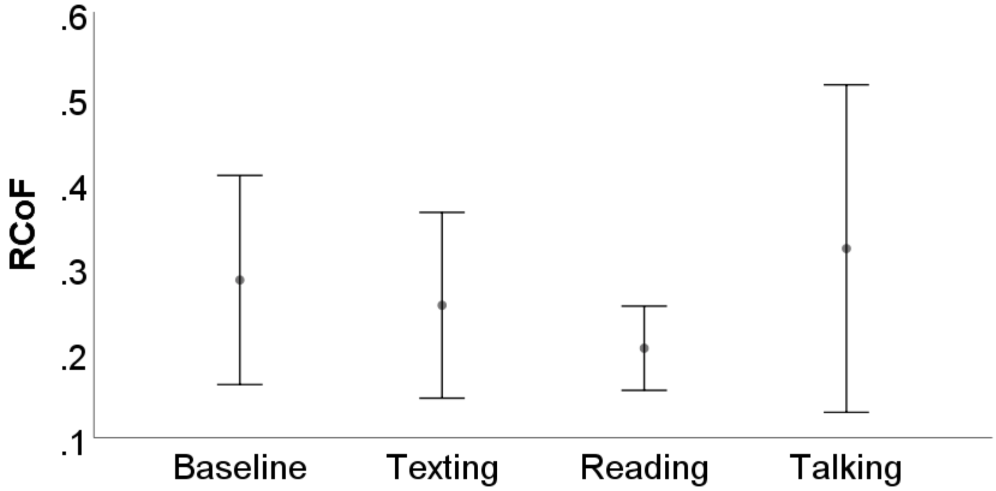

Figure 3-16: RCoF values during baseline walking, texting, reading and talking. Greater RCoF values indicate greater risk of slipping. Error bars denote one standard deviation of RCoF values in either condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001.

In response to perturbations associated with talking while walking, MFC height decreased compared to baseline walking (mean difference: 0.003 m, p=0.001) (Figure 3-17). Decreased MFC height were also observed during texting and reading while walking compared to baseline walking, but they were not statistically significant (p>0.05).

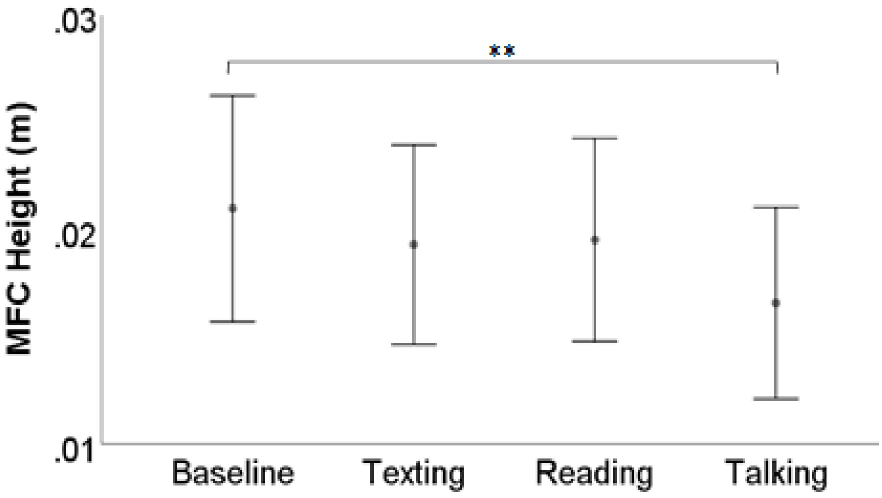

Figure 3-17: MFC height values during baseline walking, texting, reading and talking. Smaller MFC height values indicate greater risk of tripping. Error bars denote one standard deviation of MFC height values in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$.

3.4.7 Detection of the Most Destabilizing Type of Cell Phone Usage during Walking

Participants exhibited different gait patterns during walking under different types of perturbations. Statistical analysis revealed that talking while walking was associated with the largest deviation of MFC height from baseline walking compared to texting (mean difference: 0.002 m, $p=0.009$) and reading (mean difference: 0.003 m, $p=0.01$) while walking (Figure 3-18A). Likewise, slightly greater deviation of mediolateral MoS at HC and anteroposterior MoS at MFC from baseline walking were observed during talking while walking relative to texting and reading while walking, but none of them were statistically significant ($p>0.05$) (Figure 3-18C & D). Deviation of anteroposterior MoS at HC values from baseline walking also appeared to be unaffected by different types of cell phone usage while walking ($p>0.05$) (Figure 3-18B).

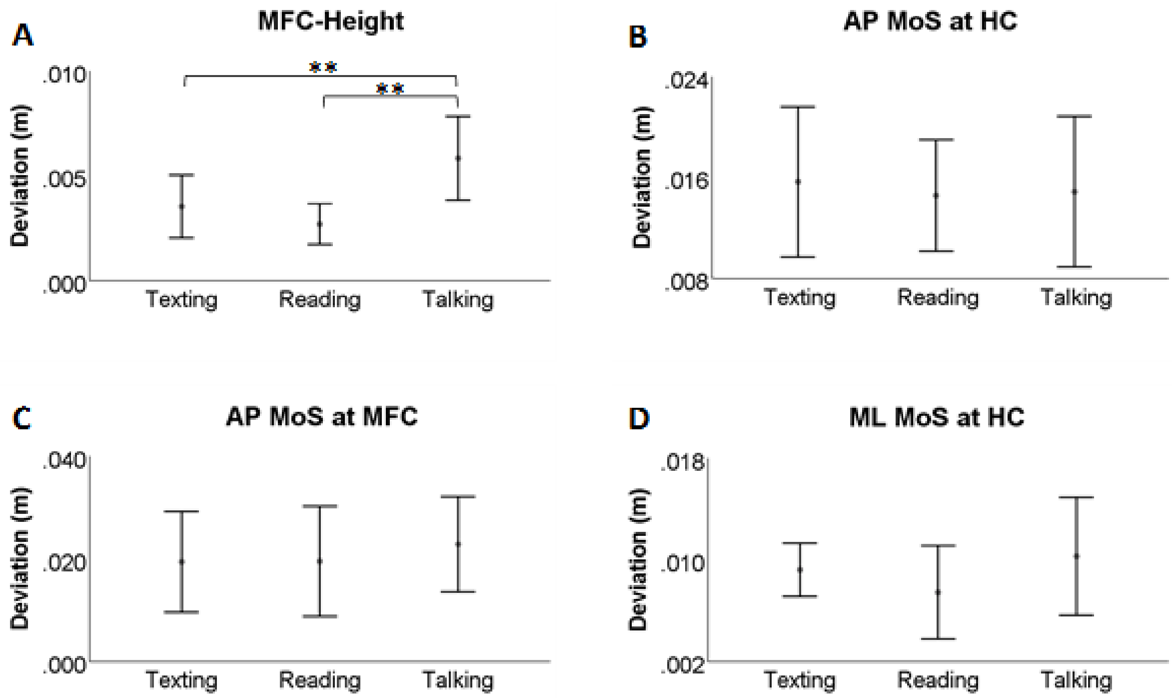

Figure 3-18: Mean deviation of MFC height (A), anteroposterior MoS at HC (B), anteroposterior MoS at MFC (C), and mediolateral MoS at HC (D) from baseline walking during texting, reading and talking while walking. Error bars denote one standard deviation of calculated deviation values in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral.

3.5 Discussion

The objective of this study was to evaluate the influence of cell phone usage on spatiotemporal measures, dynamic stability and variability of human walking, as well as associated slip and trip risk. The type of cell phone usage which affected gait stability the most was also studied. It was hypothesized that participants walk with longer and narrower steps with increased step time and double support time during dual-task trials. It was also hypothesized that individuals would exhibit a less stable, more variable gait pattern with higher risk of balance loss due to talking, texting and reading on a cell phone while walking. With regards local and orbital walking stability, it was hypothesised that stability measures in mediolateral direction would be higher during texting and reading tasks compared to talking task. Finally, it was hypothesised that texting and reading while walking would be more distracting compared to talking while walking. A summary of some of the important findings are shown in Table 3-4.

Table 3-4: Summary of the observed significant changes during talking, texting and reading while walking conditions.

	Talking	*Texting*	*Reading*
Step Width	Increased	-	-
Step Length	Increased	-	-
Step Time	Increased	-	-
Step Width Variability	Increased	Increased	Increased
Mediolateral MoS	Increased	-	-
Mediolateral LyE of Right Ankle	Increased	Increased	Increased
Mediolateral LyE of Left Ankle	Increased	Increased	-
Mediolateral LyE of Right Hip	Increased	Increased	-
Mediolateral LyE of Right Knee	Increased	Increased	-
Mediolateral LyE of Left Knee	-	Increased	Increased
MFC Height	Decreased	-	-

We showed that participants walked with longer, wider and slower steps that were also more variables in response to perturbations caused by cell phone usage while walking. Collectively, these findings, which partially support our hypothesis, suggest that participants were unable to consistently respond to distractions associated with cell phone usage while walking using their usual normal walking strategies. It has been suggested that reduced step length and/or increased step frequency are two important strategies to preserve backward and mediolateral MoS in young individuals [79]. In the current study, however, none of these strategies were observed during dual-task trials, especially talking while walking. This indicates that young individuals participating in this study prioritized performance on the cell phone task rather than balance maintenance. This is consistent with the proposed task-prioritization model by Yogev-Seligmann and colleagues (2012) which suggested that focus of attention in healthy individuals with intact postural reserve would be more on the cognitive secondary task during dual-task walking as long as there is low chance of balance loss [324].

During dual-task trials, we also found that participants walked with increased step width, which closely match previous findings by Magnani et al. (2017) [145]. This could be a strategy by individuals to improve MoS in the mediolateral direction and compensate their slower steps.

Supporting our hypothesis, the current study also found that variability of step width increased in response to perturbations induced by cell phone usage while walking, and this effect was more pronounced during texting and talking tasks. These findings are in agreement with recent studies indicating that cell phone usage while walking can be associated with motor impairment [145, 325]. This adaptation may also imply that participants attempted to reduce lateral instability due to limited arm swing during texting and talking while walking by changing foot placement [242]. We also observed great persistency in spatiotemporal variables during single and dual-task walking condition, which was not in line with our expectation. It has been shown that existence of long-range correlation reflects a natural healthy system [326]. However, Dingwell and Cusumano (2011) reinterpreted this phenomenon in a recent study and linked persistency of a data set with its level of control during walking [130]. In other words, the more tightly controlled a variable, the less persistency. Following this interpretation, the reduced long-range correlation of step width during the talking task compared to other conditions might be explained by its higher variability, suggesting that participants attempted to maintain balance during talking task by regulating and tightly controlling step width.

In the present study, we observed a slight tendency toward increase of MoS in the mediolateral direction during dual-task trials, but differences were only significant during talking tasks. According to a recent study by Vervoort and colleagues (2020), reducing the excursion of XCoM in the mediolateral direction through taking faster steps and/or extending of the BoS area in the mediolateral direction through taking wider steps are the two dominant mechanisms to increase MoS in the mediolateral direction [327]. In this study, participants took slower steps in response to perturbations applied through dual-tasking while walking, implying that individuals were not able to reduce XCoM excursion in the frontal plane to have greater mediolateral walking stability. Therefore, the observed increase in the mediolateral MoS during cell phone talking trial could be a direct consequence of enhanced BoS in that direction due to taking wider steps [229]. In the anteroposterior direction, however, MoS at HC and MFC appeared to be unaffected during dual-tasking compared to single task walking, which was not in agreement with our hypothesis. According to the study by Hak et al. (2013), increased walking speed or decreased stride length results in increased MoS at HC in the anteroposterior direction [79]. In this study, individuals walked with their preferred walking speed which was constant during all trials. In addition, stride length was increased in response to perturbation applied by talking while walking compared to baseline trial. These results further support the idea that balance maintenance in the anteroposterior direction has low level of sensitivity to visual and vestibular provided feedback compared to that in the mediolateral direction [242].

Stabilization of the head during human locomotion is a human postural control mechanism that involves the vestibular and visual systems, which are organs responsible for balance and orientation of the body during ambulation [312, 328]. Therefore, instability of the head may affect the feedback provided by the vestibular and visual systems required for gaze stabilization and postural control [312]. In response to perturbations associated with cell phone usage while walking, mediolateral local stability of the head motion decreased insignificantly, which was not in line with our expectation. While this reduction was more pronounced during texting and reading tasks, it may be associated with reduced capacity to maintain balance [329-331]. During texting and reading using a cell phone while walking, the head moves less in an attempt to focus on a screen undergoing motion. This may contribute to reduced capacity of the vestibular and visual systems to provide the required visual feedback for balance control, and may ultimately compromise stability during ambulation [145, 199, 332].

With regards local stability of the trunk motion, contrary to our hypothesis, we found no significant changes during walking while performing a concurrent cell phone related task compared to walking under no perturbation. These results seem to be consistent with other research which found no significant reduction in local stability of the trunk during cell phone usage while walking. Unaffected local stability of the trunk in the mediolateral direction during dual-task trials could be attributed to large BoS area that participants made by taking wider steps compared to baseline walking [59]. In the anteroposterior direction, the obtained results can be explained by the role of the trunk in stabilization of the head during walking. Since the trunk serves as a head stabilizer by providing a base of support for the neck and head through dissipation of acceleration from the lower body to head [312], it's stability plays an important role in head stabilization and gaze stability. It can thus be suggested that participants attempted to preserve trunk stability in the anteroposterior direction during dual-task trials to maintain gaze stability.

When considering local stability in the lower extremity joints, cell phone usage resulted in lower stability in the frontal plane compared to that in the sagittal and transverse planes, which was in agreement with our hypothesis. In addition, we found higher sensitivity of local stability in lower extremity joints to perturbations arising from texting during walking compared to those arising from reading and talking while walking in either direction. These results may be explained by increased variability of the step width during cell phone-based dual-tasking trials, which has been previously reported as well [145]. Limited arm swing as a consequence of cell phone usage may results in altered step width as a compensatory mechanism to maintain stability in the mediolateral direction, which was demonstrated by Kao et al. (2015), and is likely to have contributed to the high variability in step-width observed in the present study [199]. In fact, individuals' effort toward extending of the BoS area are driven by alteration of joint mechanics [333]. This may be associated with greater divergence of angular velocity trajectories in the mediolateral direction, and ultimately higher LyE values and lower local stability in that direction. These findings suggest that step width alteration may be a strategy adopted to maintain balance and stability during cell phone usage while walking.

Contrary to expectations, this study did not find a significant difference in dynamic orbital stability of walking between dual-task and single-task trials. This inconsistency which has also been observed in earlier studies [296, 334], might be related to violation of the periodicity assumption of walking during performing a concurrent secondary task, which may have contributed to failure of this approach in detecting stability changes during dual-task conditions compared to baseline trial [335]. In this study we did not assess periodicity of data using the index of periodicity, but it might be an interesting topic for future research. Another possible explanation for this could be compensating strategies adopted by individuals in response to perturbations associated with cell phone usage while walking which helped them to maintain orbital walking stability.

In the current study, talking while walking was associated with reduced MFC height, reflecting high sensitivity of trip propensity to perturbations caused by talking while walking. This finding further supports the idea that individuals are at higher risk of tripping during cell phone usage while walking. We also observed reduced MFC height during texting and reading while walking compared to baseline walking, but none of them were statistically significant. Given these results, together with the findings of larger MoS at MFC, and gait adaptations during texting and reading while walking, one may surmise that cell phone texting and reading while walking has a major effect on the risk of tripping during walking. With regards slip propensity, our findings do not support our hypothesis that cell phone usage while walking would lead to increased risk of slipping. It seems possible that these results are due to cautious gait pattern of individuals during dual-task trials [58]. Based on represented MoS results at HC, we can conclude that positive effects of gait adaptations outweigh negative impacts of dual-tasking on slip propensity during walking.

In the present study, we found talking on a cell phone to be the most challenging cell phone related concurrent task during walking. We observed greatest deviation of MFC height, mediolateral MoS at HC and anteroposterior MoS at MFC from baseline walking during talking on a cell phone while walking. Generally, cell phone usage while walking is a combination of cognitive, motor and visually demanding tasks. With regards the attentional demands, the main difference between talking and other cell phone related tasks is shorter processing time which makes it more cognitively demanding. This limitation is not highly pronounced during texting and reading while walking because an individual has enough processing time for reading or responding to a message. Regarding visual demands, it has been suggested that cell phone usage while walking is associated with altered vestibular and visual feedback due to limited head movement [145, 199, 332]. In terms of motor demands, cell phone usage is associated with limited function of the upper body in lateral stabilization through arm swing [58].

In this study, participants walked with longer, wider and slower steps in response to motor demands associated with talking task. Due to these adaptations, participants had greater MoS in the mediolateral direction, and slightly larger MoS in the anteroposterior direction at HC. In the anteroposterior direction at MFC, it seems that smaller MFC height and MoS also was recovered and stability was maintained by taking advantage of those adaptations. While talking during gait evoked to be more challenging than texting and reading tasks while walking, these results need to be interpreted with caution because distinguishing different types of cell phone usage in terms of the level of visual, vestibular and motor demands depends on the adopted strategies by individuals in using their cell phone (e.g. one hand or both hands). In the present study, participants were free to use their preferred cell phone usage strategy, and as a result, we observed mixed types of cell phone usage including one hand and two hands occupied. Therefore, generalization of these findings to different types of cell phone usage needs to be done with caution.

There are a number of limitations associated with this study. Data collection was performed using a motorized treadmill to obtain continuous data over many gait cycles, and to accurately speed-match walking trials; however, treadmill gait has been shown to result in different local stability and orbital stability compared to that during overground gait [336], and therefore the data collected may underestimate some stability outcome measures. In addition, the cohort in this study comprised healthy, young individuals, and the results may differ in other populations including older adults and individuals affected by neuromuscular conditions.

3.6 Conclusion

The present study showed that cell phone usage including texting, talking and reading during walking in healthy young individuals affects balance in a direction-specific manner. In response to perturbations associated with talking on a cell phone during walking, participants adopted a cautious gait pattern characterised by increased step parameters. Texting, however, increased the risk of side-to-side balance loss due to lower stability of the ankle, knee and hip joints. Reading while walking appeared to be less distracting for young individuals compared to talking and texting while walking. Orbital stability and anteroposterior MoS were less sensitive to perturbations caused by dual-tasking due cell phone usage while walking compared to other stability measures, indicating that participants attempted to attenuate the applied perturbations by gait adaptations and increase of mediolateral MoS at HC. Comparing different types of cell phone usage during walking, talking on a cell phone was identified as the most destabilizing type of cell phone usage during walking with higher risk of tripping during gait. We can conclude that strategies that target upper body configuration and stepping patterns have potential to reduce gait instability associated with cell phone usage while walking.

4 Dynamic stability and variability of the human body during walking on inclined surfaces

4.1 Summary

Walking on an inclined surface is known to contribute to more physically demanding ambulation and increase fall risk relative to level-ground walking; however, the influence of ground inclination on dynamic stability of the body during walking is not well understood. The aim of this study was to assess the effects of surface inclination on dynamic stability during walking. Gait experiments were performed on 19 male volunteers walking on an instrumented treadmill at their self-selected speed under the following conditions: level-ground walking, uphill walking ($+9.5^{\circ}$gradient) and downhill walking (-8.5° gradient). Trajectories of retroreflective markers attached to bony landmarks were simultaneously recorded using a video motion analysis system. Step length, step width, step time and double support time were calculated using the motion data. Gait variability was estimated using standard deviation as the linear method and Detrended Fluctuation Analysis (DFA) as the nonlinear method. Local stability analysis was performed on motion trajectory of the head and trunk segments. Trip propensity was measured using the height of MFC.

Participants walked with shorter step time and double support time during downhill walking. Step length increased during uphill walking and decreased during downhill walking, but none were significant. Gait variability remained unchanged during walking over sloped surfaces. Compared to level-ground walking, local stability of the head motion in the anteroposterior direction was significantly higher during downhill walking (p=0.001) and lower during uphill walking (p=0.013). Similarly, trunk motion in the sagittal plane was more locally stable during downhill walking (p=0.001) and less locally stable during uphill walking (p=0.048). In the vertical direction, head local stability was lower during downhill (p=0.035)

and uphill (p<0.001) walking conditions. Likewise, decreased local stability of the trunk motion in the vertical direction was observed during downhill (p<0.001) and uphill (p<0.001) walking trials. Participants also walked with higher and more variable MFC height during uphill (p<0.001) and downhill (p<0.001) walking conditions. Local stability analysis demonstrates that surface inclination affects balance in a direction-specific manner. Local stability analysis of the head and trunk segments revealed that stability is mostly affected in the anteroposterior and vertical directions. Uphill walking may pose greater hazard to local stability in the anteroposterior direction compared to downhill walking. Statistical persistence of stepping parameters was also observed during sloped walking conditions with increased DFA exponent of step width and step time, as well as unchanged DFA exponent of step length compared to level-ground walking. Sloped terrains do not affect gait linear variability and trip propensity in young adults. Taken together, the results of this study demonstrate that young individuals are capable of responding to perturbations associated with sloped terrains through adaptations in the anteroposterior and vertical directions, while maintaining balance over sloped surfaces.

4.2 Introduction

Walking over inclined surfaces has been shown to impose greater challenges in balance maintenance and stability than walking on level-ground , and contributes to more than 70% of falls [337]. Reduced step length, increased cadence and posterior tilt of the mid-upper torso has been reported during downhill walking, whereas increased step length, decreased cadence and anterior tilt of the pelvis and trunk has been observed during uphill walking [16-18]. Also during uphill walking, increased ankle dorsi flexion, greater knee flexion at HC and larger knee extension angles during mid-stance have been reported, while downhill walking has been associated with increased knee flexion, hip flexion, and ankle plantar flexion [16, 18]. These studies suggest that walking on inclined surfaces requires a degree of gait adaptation and increased power generation/absorption at the joint-level, which may ultimately influence dynamic balance of the body.

In the context of human movement, variability is a built-in property of all biological systems which could be easily detected through multiple repetitions of a task, for example, dissimilar footprints while walking the same path on snow twice [95]. Stergiou and Decker (2011) described movement variability as occurred variations in motor performance while repeating a task [96]. According to the existing theoretical frameworks, the Generalized Motor Program Theory and Dynamical Systems Theory, variability is a destabilizing factor for motor performance [96, 98-100]. Gait variability can be measured using a wide variety of tools including linear e.g. standard deviation [108] and nonlinear e.g. Sample Entropy (SampEn) [131] and Detrended Fluctuation Analysis (DFA) techniques [114]. Linear approaches quantify magnitude of variability while nonlinear techniques provide information about changes in gait pattern variations over time [110]. DFA is one of the most practical ways of revealing information about long range correlation in a data set made of discrete events e.g. stride time [114, 118-126]. The outcome of analysis for a set of data is DFA Exponent which quantifies the level of statistical persistency in the data set. For instance, DFA Exponent > 0.5 indicates persistency, and DFA Exponent < 0.5 implies anti-persistency. It has been suggested that DFA analysis is a reliable approach to get useful information about neuromuscular control system in response to perturbations during ambulation [321].

Balance during walking on inclined surfaces has been measured using local stability analysis of trunk motion by employing the Maximum Lyapunov Exponent (LyE) [202], a nonlinear approach to estimating the logarithmic rate of divergence/convergence of neighbouring motion trajectories in a state-space [112]. LyE analysis of the upper body using trunk motion trajectories has shown that walking uphill is associated with reduced local stability in all principal anatomic directions, whereas walking downhill is associated with decreased local stability in the vertical direction [202]. At present, however, most gait studies to date evaluate the effect of surface inclination on balance by employing trunk segment trajectories. While walking on inclined surfaces is associated with forward and backward tilt of the trunk segment to achieve power generation and absorption, respectively, such gait may also produce larger motions of the head relative to the trunk [338]. Balance of the body has been described in terms of both gaze stabilization and postural stabilization [339], and gaze stability, which is necessary for postural control, combines the visual, vestibular and proprioception systems, and is governed by a seamless interaction between ocular and head motion [340, 341]. Since the visual and vestibular systems reside within the head, stabilisation of the head segment is thought to be important in maintaining balance and spatial awareness during locomotion [312, 342]; however, little is known about the effect of surface inclination on stability of the head segment.

The aim of this paper was therefore to quantify gait variability using DFA, and dynamic stability of walking on inclined surfaces using LyE at the head and trunk segments. We also aimed to measure trip propensity during walking trials using MFC height. According to previous studies, we hypothesized that: 1) downhill walking is associated with shorter step length, step time and double support time compared to level-ground walking [16, 17]; 2) participants walk with longer step length, step time and double support time during uphill walking relative to level-ground walking [16, 17]; 3) step width would not be affected during sloped walking conditions [202]; 4) stability of the head and trunk segments would be more affected during downhill walking compared to uphill conditions, and in the anteroposterior and vertical directions rather than mediolateral direction [202]; 5) risk of tripping would be higher during level-ground walking compared to sloped walking conditions [343], and gait variability would not be affected by sloped terrains [344].

4.3 Methodology

4.3.1 Participants

Nineteen healthy young adult males (age: 28.4 ± 4.5 years, height: 175.8 ± 5.3 cm, body mass: 74 ± 11.4 kg) were recruited. Inclusion criteria were no musculoskeletal or neuromuscular impairment, and no history of lower-limb injury or prior surgery. Ethical approval was obtained from the Human Ethics Advisory Group at the University of Melbourne, and written informed consent was provided.

4.3.2 Motion analysis experiments

Sixty retro-reflective markers were attached to the bony landmarks using a modified Helen-Hayes marker-set [318] (see chapter 3 for more details). Marker trajectories were recorded using an 11-camera video motion analysis system sampling at 100 Hz (Vicon, Oxford, UK) as the subjects walked on an instrumented treadmill (AMTI, USA) at their self-selected speed (see chapter 1 for more details) under three conditions: (i) level-ground walking (ii) walking uphill on a +9.5º gradient slope [16, 345] (iii) walking downhill on a -8.5º gradient slope [16, 345]. These angles were chosen because they have been associated with significant differences in gait dynamics and body motion relative to those during level-ground walking [16, 345]. Participants wore a safety harness secured to an overhead frame, and rest breaks were granted on request. Marker trajectories were filtered using a 4th order, low-pass Butterworth filter with a cut-off frequency of 6 Hz.

4.3.3 Spatiotemporal Measures

Step length was defined as the anteroposterior distance between heel markers at the instant of HC. Step width was determined as the mediolateral distance between heel markers at the instant of HC. Step time was calculated as the duration between consecutive heel contacts, and double support time was defined as the period that both feet contact the walking surface. Mean and standard deviation of step length, step width and step time were quantified for each testing condition.

4.3.4 Gait Variability

Gait variability was measured using Detrended Fluctuation Analysis of spatiotemporal variables. In order to quantify DFA for a time series, first the cumulative sum of the deviation from the mean at each t is calculated using

$$y(i) = \sum_{t=1}^{i} [x(t) - \bar{x}] \qquad for\ i = 1,2,3, \dots, N$$ Equation 4-1

where N is the length of time series. Then, $y(i)$ is separated into n non-overlapping windows of the same size (s). As N may or may not be a multiple of s there may be a small residual after dividing $y(i)$ to n congruent windows. In order to avoid this, the dividing process is repeated one more time from another side. Therefore, the number of intervals will be $2n$. In the next step, the local trend (y_n) at each interval is calculated using least square fitting of the data. The fluctuation function can be computed from

$$F(n) = \sqrt{\frac{1}{N} \sum_{i=1}^{N} [y(i) - y(i)_n]^2}$$ Equation 4-2

where $y(i)_n$ is the local trend at each interval.

The slope of the graph of $F(n)$ as a function of n in log-log scale is called the scaling component or self-similarity exponent α. The scaling exponent is the indicator of long-range correlation. On the basis of different quantities of α, the correlation can be characterized. Specifically, $\alpha = 0.5$ when the time series is random or uncorrelated and behaves like Gaussian (white) noise. If $\alpha = 1$, the time series will behave like pink noise, or its power spectral density has an inverse relationship with frequency. The time series may be estimated as Brownian noise provided that $\alpha = 1.5$. As long as the alteration of α is limited between 0.5 and 1, the time series will have a long range correlation [126, 127].

4.3.5 Local Stability

Local dynamic stability was estimated using LyE from a 5-dimensional time-delayed state space representation of the linear velocity of the head CoM and C7 marker (seventh cervical vertebrate) in the anteroposterior, mediolateral and vertical directions using

$$S(t) = [x(t), x(t+\tau), x(t+2\tau), \dots\dots, x(t+(d_e-1)\tau]$$ Equation 4-3

where τ is the time delay and d_e is the embedding dimension [227, 229]. The time delay for each set of data was estimated using the Average Mutual Information (AMI) function [185]. The rate of divergence of the nearest neighbouring trajectories was estimated using LyE

$$D_j(i) = C_j e^{(LyE)*t}$$ Equation 4-4

where $D_j(t)$ is the distance between adjacent trajectories and C_j is the initial distance between the trajectories. Taking the Napierian logarithm of both sides and least-square curve fitting to the average line yields

$$y(i) = \frac{1}{\Delta t}\langle \ln d_j(i) \rangle = LyE.i + c, c = \frac{\ln C_j}{\Delta t}$$ Equation 4-5

where LyE is the slope of the fitted line and used as the indicator for stability. Smaller LyE values indicate more convergence of nearest neighbouring trajectories and higher stability, while larger LyE values indicate more divergence of adjacent trajectories and lower stability. LyE was estimated from 150 consecutive strides in each trial [25].

4.3.6 Trip Propensity Measure

Trip propensity was measured for each testing condition using the height of foot clearance respect to the walking surface at mid swing. Lower MFC height implied greater risk of tripping during walking.

4.3.7 Statistical Analysis

A repeated measure analysis of variance (ANOVA) was performed to quantify the influence of surface inclination on dynamic stability during walking. The dependent variables were spatiotemporal variables, DFA scaling component and trip propensity as well as LyE for the head and trunk, while the independent factors were test conditions, specifically, level-ground walking, uphill walking and downhill walking. Significance level was defined as $p \leq 0.05$. All statistical analyses were performed using SPSS 25 (SPSS Inc., Chicago, IL).

4.4 Results

4.4.1 Spatiotemporal Measures

Participants walked downhill with significantly smaller step time (mean difference: 51.3 ms, $p=0.029$) and double support time (mean difference: 154.8 ms, $p<0.001$) compared to level-ground walking (Table 4-1). There was, however, no significant changes in temporal gait parameters during uphill walking compared to level-ground walking ($p>0.05$). In response to perturbations caused by upslope and downslope terrains, participants also walked with larger step width, but none of them was statistically significant ($p>0.05$). Downhill walking was associated with a nearly significant decrease in step length compared to level-ground walking (mean difference: 74.8 mm, $p=0.052$), but there was an insignificant increase in step length during uphill walking ($p>0.05$). Variability of spatiotemporal measures did not change significantly during walking over sloped surfaces compared to level-ground walking condition (p>0.05).

Table 4-1: Summary of the spatiotemporal parameters during walking trials, including level-ground, uphill and downhill walking conditions

	Level-ground walking	*Uphill walking*	*Downhill walking*
Step length (mm)	515.4 ± 79.7	530.4 ± 90.8	455.6 ± 76.7
Step length variability (mm)	19.9 ± 6.2	18.2 ± 5.3	17.7 ± 5.2
Step width (mm)	84.5 ± 31.5	101.9 ± 34.6	94.7 ± 25.4
Step width variability (mm)	21.6 ± 5	21.0 ± 5.3	20.9 ± 5.0
Step Time (ms)	609.9 ± 55.0	614.8 ± 62.5	563.5 ± 54.2
Step time variability (ms)	23.2 ± 14	18.6 ± 6.6	16.8 ± 6.9
Double support time (ms)	223.5 ± 45.7	215.0 ± 39.2	60.2 ± 23.9
Double support time variability (ms)	40.8 ± 32.9	12.7 ± 4.1	11.2 ± 3.8

4.4.2 Gait Variability

Strong statistical persistence of stepping parameters was observed during level-ground and sloped walking conditions (DFA Exponent >> 0.5) (Figure 4-1). DFA exponent of step width increased significantly during uphill (p=0.003) and downhill (p=0.011) walking compared to level-ground walking. Participants also walked with increased DFA exponent of step time during uphill walking compared to level-ground condition (p=0.001). Similar trend was observed in DFA exponent of step time during downhill walking, but it was not statistically significant (p>0.05). For step length, DFA exponent did not differ significantly in response to perturbations caused by sloped terrains compared to level-ground walking (p>0.05).

4.4.3 Local Stability

Local stability of the head motion was significantly affected by sloped terrains. In the anteroposterior direction, significantly smaller head LyE was demonstrated during downhill walking compared to level-ground walking (p=0.001), while a significantly larger head LyE was observed during uphill walking compared to downhill walking ($p=0.013$) (Figure 4-2). Walking uphill ($p<0.001$) and downhill ($p=0.035$) also resulted in significantly larger head LyE values compared to that during level-ground walking in the vertical direction, while ground inclination did not significantly affect LyE values in the mediolateral direction ($p>0.05$). For C7 motion, participants exhibited significantly increased local stability during downhill walking ($p=0.001$), while uphill walking resulted in decreased local stability relative to level-ground walking ($p=0.048$) (Figure 4-3). C7 motion in the vertical direction were locally unstable during uphill ($p<0.001$) and downhill walking ($p<0.001$) compared to level-ground walking condition. In the mediolateral direction, however, sloped terrain yielded no significant changes in local stability of the C7 motion ($p>0.05$).

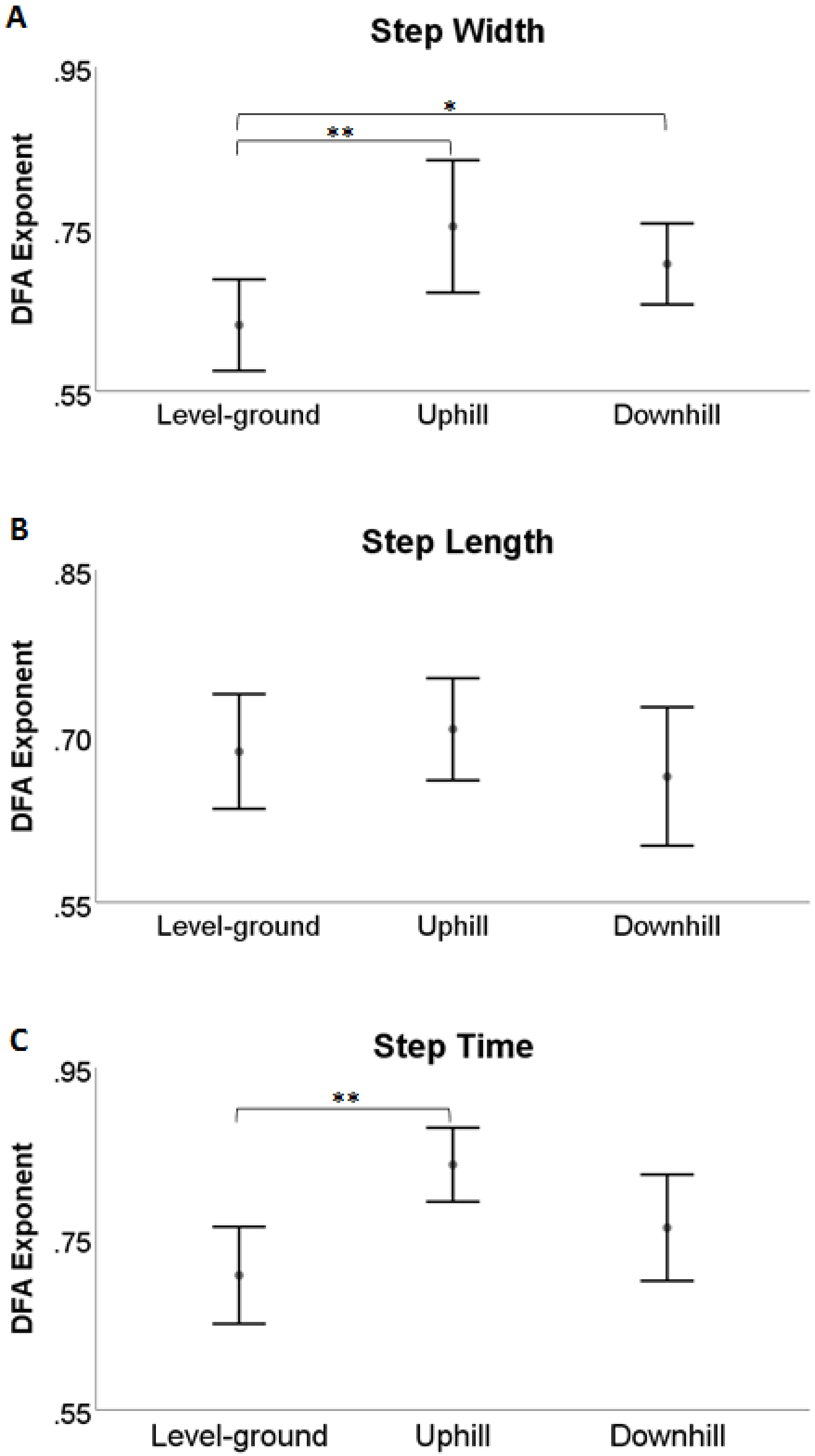

Figure 4-1: DFA Exponent values for step width (A), step length (B) and step time (C) during level-ground, uphill and downhill walking conditions. When the DFA Exponent falls between 0.5 and 1, the time series will have a long-range correlation. Error bars denote one standard deviation of DFA Exponent values in either condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001.

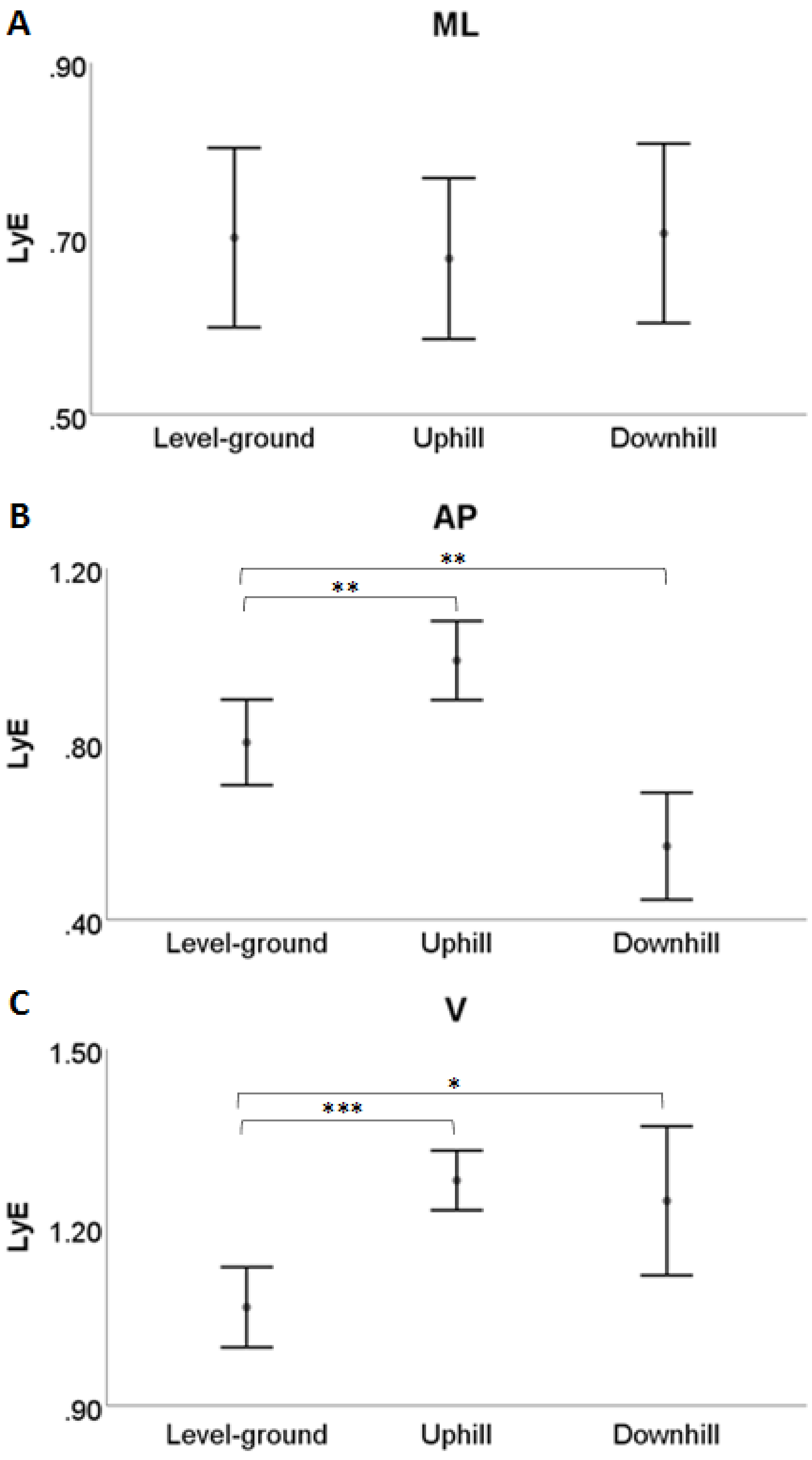

Figure 4-2: LyE values for the head in the mediolateral (A), anteroposterior (B) and vertical (C) directions during level-ground, uphill and downhill walking conditions. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

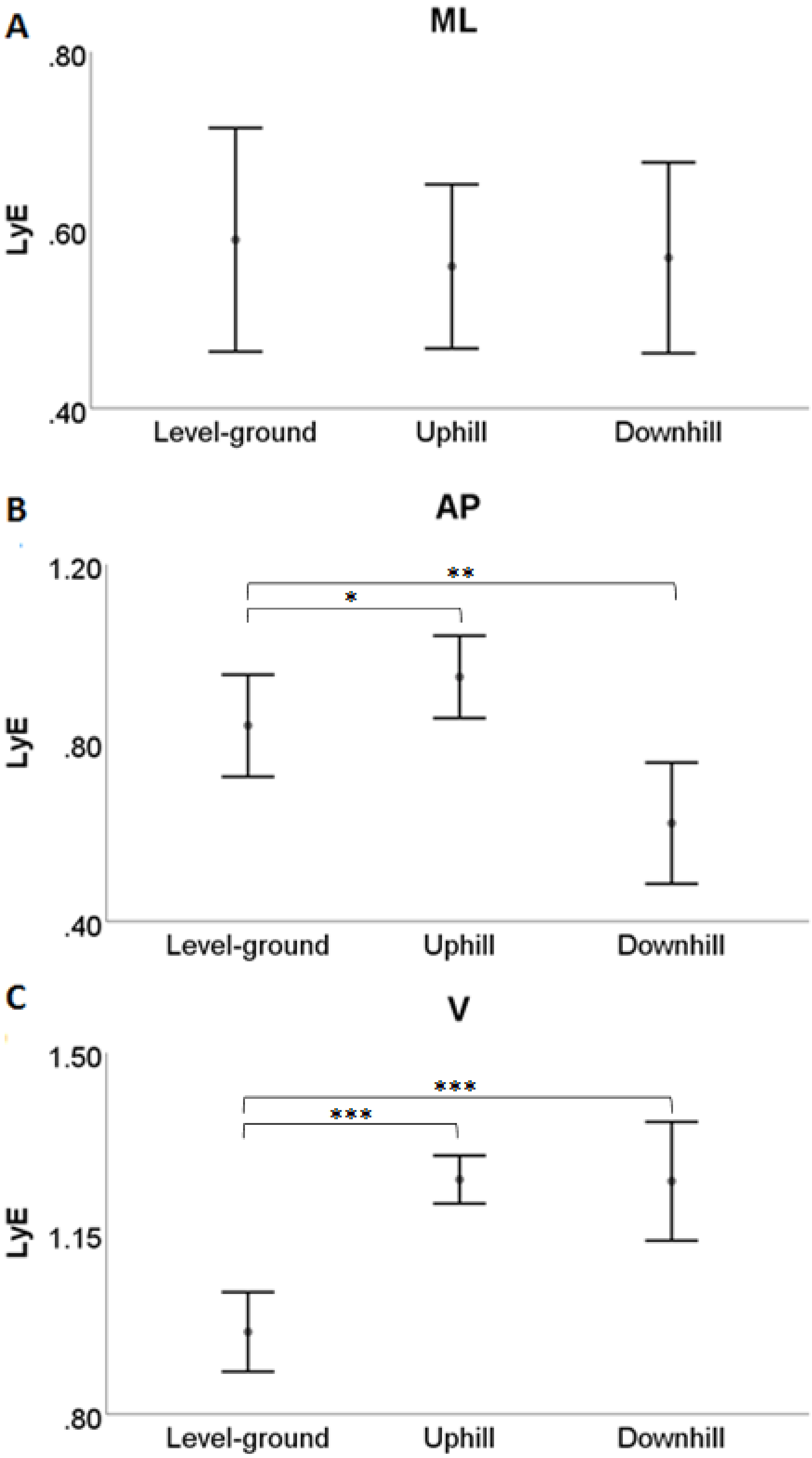

Figure 4-3: LyE values for the trunk in the mediolateral (A), anteroposterior (B) and vertical (C) directions during level-ground, uphill and downhill walking conditions. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

4.4.4 Trip Propensity Measure

Sloped terrain affected the propensity of tripping during walking. Compared to level-ground walking, MFC height was significantly higher during uphill walking (mean difference: 0.009 m, p=0.001) (Figure 4-4). Downhill walking also resulted in higher MFC height relative to level-ground condition, but it was not statistically significant.

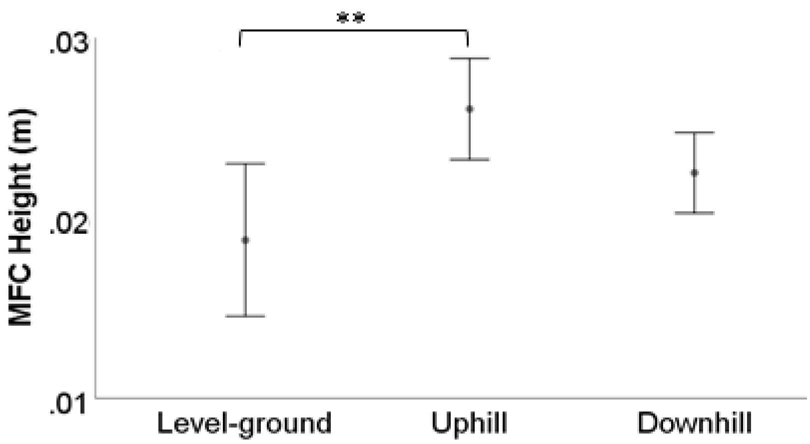

Figure 4-4: MFC height values during level-ground, uphill and downhill walking conditions. Smaller MFC height values indicate greater risk of tripping. Error bars denote one standard deviation of MFC height values in either condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001.

4.5 Discussion

Several studies have shown that walking over inclined surfaces poses unique biomechanical challenges during ambulation [16, 346-348], resulting in altered position and orientation of the head due to increased gravitoinertial forces, increased lower limb power generation and absorption, as well as adaptive step patterns to increase friction forces and reduce fall risk [18, 202]. These ambulation strategies may improve balance and stability of the body, but may ultimately result in different time-histories of lower-limb and whole-body COM dynamics as the body adapts to the increased physical demands of ground inclination. The first objective of the present study was to measure the stability of walking over the inclined surfaces using local dynamic stability of the head and trunk's movements. Secondly, we aimed to investigate the effect of surface inclination on gait variability and the risk of tripping using DFA and trip propensity measures. Uphill walking was associated with lower local stability, while downhill walking resulted in greater local stability of walking in the anteroposterior direction. Sloped terrains did not affect local stability in the mediolateral direction. Head and trunk segments responded similarly to the perturbations applied by the sloped terrains. The probability of tripping decreased during walking over uphill and downhill surfaces. A summary of some of the important findings are demonstrated in Table 4-2.

Table 4-2: Summary of the observed significant changes during uphill and downhill walking conditions.

	Uphill	***Downhill***
Step Time	-	Decreased
Double Support Time	-	Decreased
DFA Exponent of Step Width	Increased	Increased
DFA Exponent ST	Increased	-
Anteroposterior LyE of Head	Increased	Decreased
Vertical LyE of Head	Increased	Increased
Anteroposterior LyE of Trunk	Increased	Decreased
Vertical LyE of Trunk	Increased	Increased
MFC Height	Increased	-

In the current study, it was found that participants walked with shorter step time and double support time in response to perturbations caused by downslope terrain, which was in agreement with our hypothesis. Reduced step time during downhill walking was also been observed in an earlier study by Leroux et al. (2002), may be indicative of efforts made by individuals to increase frequency of steps during downhill walking and keep walking stable [17, 79]. Reduced double support time during downslope walking could also be attributed to increased step frequency which resulted in greater time spent in single support phase. Franz and Kram (2012) also observed similar trend in double support time during downhill walking in young adults, but it was not statistically significant [349]. These findings also reflect great confidence of young individuals during walking down over sloped surfaces, as well as their ability in preserving gait stability while they are mostly spending time at unstable phases of gait i.e. single support phases.

In agreement with previous studies, participants walked downhill with shorter step length and walked uphill with longer step length. While the observed differences were not statistically significant [16], it shows that movement of the CoM is tightly controlled in the sagittal plane. Consistent with the literature, this research found that sloped terrains have no significant influence on step width during walking. One reason behind this could be unrestricted arm swing during ambulation over sloped surfaces which contributes to trunk stability in the mediolateral direction, and ultimately limits movement of the CoM in the frontal plane [256]. As a result, individuals did not need to modify step width to enhance BoS area and maintain balance in the mediolateral direction.

Supporting our hypothesis, the present study showed that variability of spatiotemporal variables did not change during walking over sloped surfaces compared to level-ground walking, indicating that young individuals maintain the same level of stability during uphill and downhill walking as walking on flat surfaces. A possible explanation for this could be insufficient imposed challenge by the positive and negative gradients to elicit significant variation in spatiotemporal measures during sloped walking in healthy young adults. Another reason behind limited variation of spatiotemporal parameters during uphill and downhill walking conditions might be adaptation to sloped walking surfaces due to long period of treadmill walking in our study, which was not possible in previous studies that observed considerable changes in gait dynamics during walking at -8.5° and +9.5° grades [16, 345]. We also observed strong persistency of spatiotemporal measures during all walking conditions, with increased DFA exponent of step width and step time during sloped walking compared to level-ground walking. According to a recent study by Dingwell and Cusumano (2011), the observed increase in long-range correlation exponents could be an indicator of less efforts made by individuals to control step width and step time during sloped walking [130]. With regards the step width, this seems to be consistent with other research which found more control of the balance in the sagittal plane compared to frontal plane during walking over sloped surfaces [202]. Considering step time, this finding matches previous studies which found increase of step length as a main adaptation strategy in stepping parameters in the sagittal plane during uphill walking, with no significant changes in step time [16-18].

In the present study, local stability analysis during walking on inclined surfaces was performed on the head segment, which has not been undertaken to date. Stabilization of the head plays an important role in human postural control, since the vestibular and visual systems, which are required to maintain balance during walking, reside within the head and must adapt to the motion and loads transmitted through the body during ambulation [312, 342]. Reduced stability of the head has been shown to result in reduced capacity of the vestibular-visual system to provide the required feedback for stabilization of optical flow and balance maintenance [312, 342]. We showed higher local stability of the head and trunk movements in the sagittal plane during downhill walking compared to level-ground walking conditions. This may be due to greater attenuation of trunk and head motion in order to maintain the center of mass within the border of the BoS, which may ultimately reduce the degree of gaze perturbation due to the gravity-driven forward momentum of the body [338, 350]. In this manner, attenuation of the head's angular motion may minimize energy expenditure and reduce gravity-driven loads at the head, contributing to ocular compensation and steady gaze during walking [351].

In line with previously reported results by Vieira and colleagues (2017), local stability of the head and trunk movements in the anteroposterior direction were lower during uphill walking compared to level-ground condition [202]. These results are likely to be related to larger movements of the upper body in the sagittal plane during uphill walking. It has been shown that increased inclination of the upper body during uphill walking contributes to generation of more momentum by the lower body to counteract gravity-driven resistance of upslope terrain [17]. In the current study, reduced local stability of the head and trunk movements in the vertical direction was shown during uphill and downhill walking compared to level-ground walking condition. This might be explained by the function of the upper body in power generation and absorption by raising and lowering the CoM during ambulation over sloped surfaces to maintain balance and facilitate propulsion [18, 202].

As hypothesized, sloped terrains did not affect local stability of the head and trunk movements in the mediolateral direction. These findings are in line with those of previous studies [202], suggesting that young adults effectively attenuate mediolateral acceleration of the body in the frontal plane during walking [352]. In general, balance is actively controlled in the mediolateral direction during level-ground walking [242]. In the current study, however, balance in the frontal plane during sloped walking was unchanged and actively controlled in the anteroposterior direction and to some extent in the vertical direction. These findings might be explained by the fact that most of the adaptations during sloped walking occur in the sagittal and transverse planes rather than the frontal plane [202]. Besides, trunk stabilization strategies like arm swing limits the effect of sloped terrain on stability of the upper body in the frontal plane.

The current study found that participants walked with increased MFC height in response to perturbations caused by sloped terrains. These findings are in agreement with those obtained by Thies et al. (2011) [343], suggesting that young individuals have greater risk of tripping during walking over flat surfaces relative to inclined surfaces. However, the average MFC height values in this study were smaller compared to Thies' estimate of MFC height during uphill and downhill conditions. This discrepancy could be attributed to adaptation to the slope over long period of treadmill walking in our study (about 150 strides in four minutes) which was not possible in Thies' study due to short period of data collection (two consecutive strides during walking over 1.2 meter custom made ramp). However, the observed larger MFC height during uphill walking compared to level-ground condition is contrary to that of Khandoker et al. (2010), who found decreased MFC height during upslope walking [353]. One reason behind this discrepancy could be very small range of surface inclination in his study (±3°) compared to our study (-8.5° and +9.5°).

There were a number of limitations associated with this study. Gait data were collected using an instrumented treadmill to facilitate collection of large continuous walking dataset, and to assist in speed-matching gait in subjects; however, it has been shown that treadmill walking is associated with different trunk local stability data compared to those during over-ground walking [336], which may have the effect of underestimating LyE for the head and trunk segments. The scope of this study was also limited in terms of the surface inclinations. Specifically, -8.5° and +9.5° gradients were employed as these angle have been associated with significant differences in gait dynamics and body motion relative to those during level-ground walking [16, 345]. While other ground inclinations may produce different results, the general positive/negative gradient-specific trends observed in the present study are likely to reflect the dynamic balance strategies that occur during day to day ambulation over inclined surfaces. In this study, experimental conditions were not randomized and some adaptation to tasks may have occurred, which may have biased the results.

4.6 Conclusion

The present study showed that walking on inclined surfaces in healthy young individuals affects balance in a direction-specific manner. Local stability analysis of the head and trunk segments revealed that stability is mostly affected in the anteroposterior and vertical directions during walking on inclined surfaces. Downhill waking was associated with higher stability in the anteroposterior direction and lower stability in the vertical direction, whereas uphill walking resulted in lower stability in the anteroposterior and vertical directions. We also found that sloped terrains do not affect gait variability. With respect to trip propensity, this study demonstrated that sloped terrains do not increase the risk of tripping in young adults. Further research is required to understand the response of young individuals to upslope terrain with a larger range of gradients. To develop a full picture of dynamic stability changes during walking over sloped surfaces, additional studies should be conducted focusing on stability metrics based on inverted pendulum model such as foot placement estimate. Overall, the results of this study demonstrate that young individuals are capable of responding to perturbations associated with sloped terrains through adaptations in the anteroposterior and vertical directions, while maintain balance over sloped surfaces.

5 Dynamic stability and variability of walking during sensory and cognitive dual-tasking

5.1 Summary

Dual-tasking during locomotion has been found to result in an altered gait profile characterized by slower walking, increased step width and greater step duration. The addition of a cognitively demanding task, or an activity that requires additional sensory input, may ultimately affect balance and increase the risk of balance loss during walking. The objective of this study was to investigate the influence of performing additional auditory, visual and cognitive tasks during walking on dynamic stability. Nineteen healthy, young adults walked on a motorized treadmill at their preferred walking speed under different conditions which included baseline walking and walking while (i) watching a video clip on a screen (ii) performing low and high load cognitively demanding mathematical tasks, and (iii) listening to a common popular music song. Trajectories of retro-reflective markers attached to bony landmarks were simultaneously measured using an optical motion analysis system. Using these data, spatiotemporal measures and their variability, Margin of stability (MoS) at HC and MFC, Maximum Floquet Multiplier (MaxFM) of the head motion, Maximum Lyapunov Exponent (LyE) of the head, trunk, hip, knee and ankle joints motions, slip and trip propensity, were evaluated for each test condition. Finally, deviation of MoS and trip propensity values during dual-task trials from the corresponding values during baseline walking were calculated to identify the most destabilizing type of secondary task during walking. Mean step length changed significantly in response to perturbations associated with dual-tasking while walking. Step time also increased significantly during walking under cognitive perturbations.

Participants walked with higher variability of step width during sensory dual-tasking trials compared to single-task walking. MoS in the anteroposterior and mediolateral directions

remained unaffected during dual-task trials. Head and trunk local stability were not significantly different in either principal anatomic direction during dual-tasking conditions compared to baseline walking. Local stability in the sagittal, frontal and transverse planes significantly changed across lower extremity joints during dual-task trials compared to single-task walking, and most of the changes were observed in the ankle joint. Participants walked with significantly greater orbital stability of the head motion in the mediolateral and anteroposterior directions during cognitive perturbations compared to walking under no perturbations. Dual-tasking did not affect orbital stability of the head in the vertical direction, nor orbital stability of the trunk motion. Considering the ankle, knee and hip joints, dual-tasking mostly resulted in significantly higher orbital stability compared to single-task trial. Trip propensity significantly increased during walking under cognitive and auditory perturbations, and insignificantly increased during walking under visual perturbation. Slip propensity, however, was not significantly different during dual-task trials compared to single-task walking. Comparing different types of perturbations, gait in young people appeared to be more sensitive to cognitive perturbations rather than sensory perturbations.

These findings suggest that young individuals adopt different strategies in response to cognitive and sensory perturbations to preserve gait balance. Performing additional cognitive and sensory demanding tasks while walking increases the risk of tripping. Results from this study suggests that gait patterns in young adults are robust to sensory perturbations but more sensitive to cognitive perturbations during walking. Further experiments, using a broader range of sensory and cognitive perturbations with higher intensities, could shed more light on gait pattern changes under perturbations caused by dual-tasking while walking in young adults and other populations which are more prone to balance loss.

5.2 Introduction

Performing additional activities during walking has been shown to make ambulation a more demanding activity [354-357], depending on the complexity of the secondary task [358], and may ultimately result in the adoption of a more cautious gait pattern to avoid a fall event [58]. This cautious gait pattern may include walking with increased step width, decreased step length and decreased gait velocity [58]. It has been shown that walking is a cognitively demanding process [359], and in order to maintain balance, the visual and vestibular systems provide feedback for gaze stabilization which is necessary for whole-body balance control [340, 341]. Performing additional cognitive and/or sensory demanding tasks while walking can increase load on the brain and result in capacity sharing in attentional resources and delayed processing of either task. The more difficult the additional task, the more allocated share of sensory and/or cognitive system's capacity to the additional concurrent task, the less allocated share of sensory and/or cognitive system's capacity to balance maintenance, and the greater the risk of balance loss [360].

The visual system has a major role in postural control during walking through providing required feedback about the walking environment and potential hazards, as well as orientation and movement of the body to the CNS [361]. Compared to cognitive dual-tasking, however, few scholars have been able to draw on any systematic research into gait changes during performance of a visual concurrent task while walking. In one early example of research in this area, Bock and Beurskens (2011) reported that a visual secondary task can affect gait pattern in older adults, but young individuals are more resistant to visual perturbations caused by dual-tasking while walking [362]. Cell phone usage while walking is another common example of an additional visually demanding task during ambulation, which has been demonstrated to be distracting in some studies. Magnani et al. (2017), for instance, found that cell phone texting while walking increases gait variability but decreases trunk stability [145]. In a recent study, Prupetkaew and colleagues (2019) demonstrated that visual and cognitive demands associated with cell phone usage while walking affects gait pattern in older adults [363]. However, a note of caution is due here since cell phone usage while walking is a visually, cognitive and motor demanding task and the observed gait changes in the study by Magnani et al. (2017) and similar studies cannot be necessarily linked to visual demands. In a recent investigation into the effect of visual dual-tasking on gait balance, it has been shown that watching TV with subtitles is associated with smaller stride interval variability compared to watching TV with sound and without subtitle in young individuals [201]. These studies report alterations of gait profiles due to the act of performing an additional visually demanding task during walking, which may compromise gait balance. At present, however, the experimental data are controversial, and there is no general agreement about the influence on dynamic stability and gait variability of performing an additional visually demanding task.

Performing auditory additional tasks during walking is one of the most common types of sensory-based dual-tasking while walking. In a recent review, Ghai et al. (2018) demonstrated that auditory additional task can improve gait velocity and stride length, while it has negative impact on cadence in Parkinson's disease [364]. In Alzheimer's disease, however, it has been shown that auditory dual-tasking while walking results in greater spatial gait variability [365]. Considering healthy cohorts, no changes have been reported in the elderly individuals due to performing additional auditory task during walking compared to single-task walking [366]. A popular example of auditory demanding dual-tasking activity performed while walking is listening to music. Thompson et al. (2012) showed that listening to music during walking increases walking speed and results in decreased street-crossing accuracy in pedestrians through reduced ability to watch for traffic [367]. It has also been demonstrated that adults that walk attempt to synchronize their walking pace with the song's tempo, which results in increased variability of the stride interval [368]. Ultimately, these studies suggest that listening to music while walking can alter gait patterns and have the potential to influence balance and stability of the body. A recent study by Sejdic and colleagues (2013), however, found no significant changes in gait stability and variability due to walking and listening to music in young adults compared to single-task walking [201]. Due to these controversial findings, it seems at present that there is no general agreement about the effect of listening to music on gait patterns, and how each population responds to specific types of music.

Dynamic stability of human gait under cognitive, visual and auditory perturbations caused by dual-tasking has been expressed in terms of local and orbital stability of the trunk segment [296, 369]. Trunk local and orbital stability during walking have been determined using two well-known criteria from dynamical system theory: the Maximum Lyapunov Exponent (LyE) and the Maximum Floquet Multiplier (MaxFM), respectively [25]. Local stability analysis determines whether nearest neighbours in consecutive trajectories diverge or converge over the time in the state-space, through measurement of the LyE [112], whereas orbital dynamic stability estimates the rate of divergence or convergence of trajectories between consecutive gait cycles in the Poincare' section, using the MaxFm criteria [203]. Dingwell et al (2008) showed that the effect of applied distractions due to a cognitive additional task on walking stability is not clear [296]. In a more recent analysis of stability, Rábago et al (2016) suggested that the gait balance in healthy young adults is resistant to perturbation due to cognitive distractions [210]. In response to visual perturbations caused by sensory dual-tasking, increased stride variability was observed in young adults, but gait stability appeared to be unaffected [369]. The MoS analysis has also been used in previous studies to investigate the effect of dual-tasking on gait stability which is most commonly calculated from the extrapolated centre of mass (XCoM) relative to the boundaries of the base of support (BoS) [6, 25]. In comparison to normal walking, Kao et al. (2018) found no significant changes in MoS due to cognitive and auditory dual-tasking while walking [231]. The influence of visual additional task through cell phone texting has been assessed in several studies, suggesting an increase in mediolateral MoS in young adults [230, 370].

At present, there is no consensus on the impact on dynamic stability of performing additional auditory, visually or cognitively demanding task during walking, nor the impact of dual-tasking on the trunk, lower limb joints and head segment, which plays a significant role in providing stable gaze and balance control [352]. In addition, research to date has not yet determined the effect of cognitive and sensory dual-tasking on persistency of spatiotemporal time series, trip and slip propensity. Finally, it is still not known which type of sensory and cognitive additional tasks during walking is more destabilizing. The first aim of this study was to employ different measures including MoS, LyE and MaxFM to assess the impact of performing additional auditory, visual and cognitive tasks while walking on variability and dynamic stability of gait. Secondly, the study aimed to investigate the relationship between performing additional concurrent sensory and cognitively demanding task during walking on the propensity of slipping and tripping. The last objective of this study was to identify the most destabilizing type of sensory and cognitive additional task during walking. According to previous studies, we hypothesised that: 1) sensory and cognitive dual-tasking affects gait pattern and results in shorter step length, larger step width, shorter step duration and double support time [79]; 2) dynamic stability in young adults would be resistant to perturbations arising from auditory, visual and cognitive dual-tasking during walking [58, 201, 296] ; 3) gait variability would be increased due to auditory and cognitive dual-tasking [368, 371] and decreased due to visual dual-tasking while walking [369]; 4) auditory, visual and cognitive secondary tasks would not affect the risk of slipping and tripping [58]; 5) an additional cognitive task during walking is more destabilizing to gait compared to a secondary sensory task.

5.3 Methodology

5.3.1 Participants

Nineteen young healthy males (age: 28.4 ± 4.5 years, height: 175.8 ± 5.3 cm, body mass: 74 ± 11.4 kg) were recruited for this study. Inclusion criteria for all participants were no history of (i) lower extremity injury; (ii) lower limb surgery; (iii) neurological or musculoskeletal disorder. Participants were selected among the students and graduates of the University of Melbourne. Ethical approval was obtained from the Human Research Ethics Committee at the University of Melbourne.

5.3.2 Motion analysis experiments

Participants were asked to walk on a motorized treadmill at their preferred walking speed under five conditions which included baseline walking and walking while (i) watching a video clip on a screen (visual stimuli) (ii) performing low (serial threes from 200) and high (serial threes from 300) load cognitively demanding mathematical tasks, and (iii) listening to a piece of music (auditory stimuli). An 11-camera video-based motion analysis system sampling at 100 Hz (Vicon, Oxford, UK) was used to acquire trajectories of sixty retro-reflective markers (see chapter 3 for more details) attached to the body with a modified Helen-Hayes marker-set [318]. Preferred walking speed for each participant was determined by adjusting treadmill speed until adults indicated their most comfortable walking speed [174]. All participants wore a standardized running shoe during testing. For safety in the event of a fall, participants wore a safety harness tethered to an overhead frame during all trials. Motion data were filtered using a low-pass 4th order Butterworth filter with a cut-off frequency of 6 Hz. Inverse kinematics was employed to quantify kinematics of the head segment (Visual 3D, C-Motion, USA).

As part of standard attention tests, serials were selected for the cognitive dual-tasking trials [372]. It has been widely used in previous studies to assess the effect of cognitive dual-tasking on gait pattern [373-375]. The cognitive trials consist of two relatively effortful mathematical tests of different levels of difficulty: counting down from 300 by 3 representing a more challenging additional cognitive task, and counting down from 200 by 3 representing a less challenging additional cognitive task. During cognitive tasks, participants were asked to count as accurately as possible. For the auditory stimuli trial, the popular track by Pharrell Williams ("Happy") was played using a Samsung Galaxy Ace mobile phone (Samsung Electronics Co. Ltd., South Korea). Various musical instruments in this song are known to generate different cueing modalities [376]. For the visual stimuli trial, a video clip of the best commercial advertisements in 2014 was played without sound and English subtitles [377]. Each task took four minutes, and rest periods were given on request.

5.3.3 Spatiotemporal Measures

Step length was defined as the anteroposterior distance between heel markers at the instant of HC. Step width was determined as the mediolateral distance between heel markers at the instant of HC. Step time was calculated as the duration between consecutive heel contacts, and double support time was defined as the period that both feet contact the walking surface. Mean and standard deviation of step length, step width and step time were quantified for each testing condition.

5.3.4 Margin of Stability

MoS for each test condition was calculated at HC and MFC, which are gait events that have been shown to be associated with backward and forward balance loss, respectively [154, 162]. HC was detected using the maximum horizontal distance between sacrum and heel markers [321], while MFC was identified at mid-swing phase when the vertical trajectory of the first distal phalange marker was minimum [40]. The extrapolated centre of mass (XCoM) was calculated in the transverse plane using the whole-body CoM position and velocity from

$$X_{CoM,x} = x + \frac{\dot{x}}{\omega_0}\,, \qquad \omega_0 = \sqrt{\frac{l}{g}} \qquad \text{Equation 5-1}$$

where x and $\dot{x}$ are the CoM position and velocity, respectively; ω_0 is the natural frequency of the inverted pendulum; l is the leg length, which was calculated as the vertical distanced between the greater trochanter and the lateral malleolus markers; and g is the universal gravitational constant. MoS was defined as the distance between the XCoM and the BoS boundary in the mediolateral and anteroposterior directions.

Many gait stability models have maintained a focus on centre of mass control within a stability base defined by foot position during double support. The approach adopted in the present study is important to consider centre of mass control dynamics throughout the gait cycle. Destabilizing events arise continuously when walking and threats to stability during the swing phase, when there is only single-foot ground contact, are more likely to be associated with critical instability leading to a fall. This observation was made almost 20 years ago in a seminal research study by Patla and colleagues [322] who introduced the concept of a "projected" or "virtual" base of support to characterise swing phase stability. Their model [322] also defined a supporting base using the relative position of both feet but their formulation for the swing phase foot position was "…projected to the ground to define this area" (p.613). Similar in concept to Patla et al. [322], in the model presented below, a single critical swing phase event has been identified (MFC) and at MFC the line between the first distal phalange marker of the stance foot and projection of similar marker of the swing foot was used to define the virtual anterior BoS boundary [322], while the position of heel marker in the sagittal plane at the instant of HC was used as the posterior BoS border of BoS. In the frontal plane, the mediolateral position of the lateral malleoli marker attached to the leading foot was used as the lateral border of BoS [277]. Since forward balance loss occurs along the direction of CoM velocity, MoS at MFC was defined as the distance between the XCoM and the anterior BoS boundary in the direction of CoM velocity (Figure 5-1A) [323]. At HC, anteroposterior MoS was defined as the distance travelled by the XCoM in the sagittal plane relative to the posterior BoS border (Figure 5-1B). Mediolateral MoS was also quantified as the distance between XCoM and lateral BoS boundary in the frontal plane (Figure 5-1B). Larger positive MoS values indicate lower risk of balance loss, while the larger negative values imply a higher risk of balance loss. MoS was calculated separately for each limb at HC and MFC.

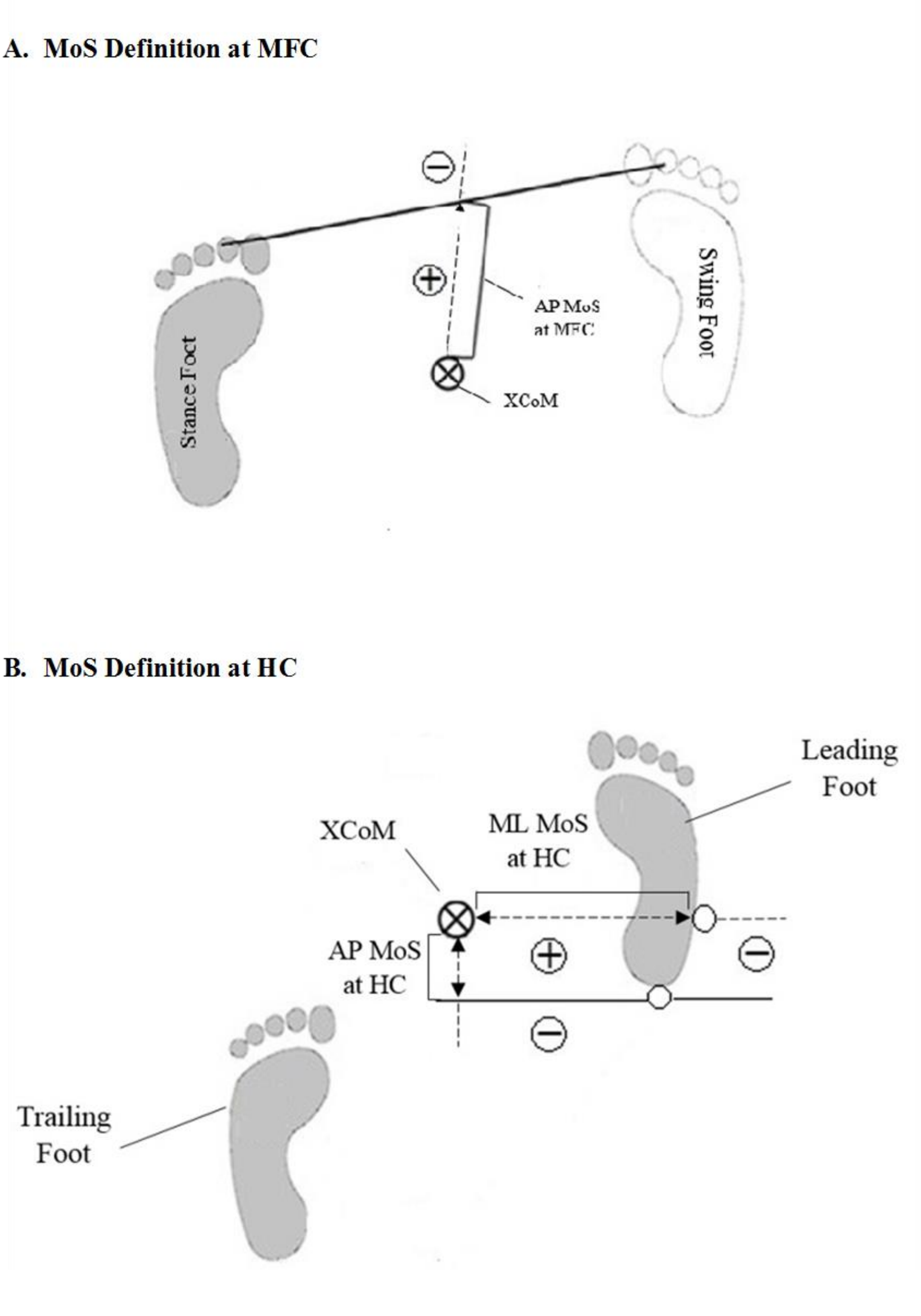

Figure 5-1: Schematic diagram illustrating definition of MoS at HC and MFC in the transverse plane view. At MFC, MoS is defined as the travelled distance by the XCoM along with the direction of velocity relative to the anterior border of BoS made by a line between two great toes. At HC, the anteroposterior excursion of the XCoM relative to the posterior border of BoS (the position of heel marker of the leading foot in the sagittal plane) defines the anteroposterior MoS, while the mediolateral excursion of the XCoM relative to the lateral border of BoS formed by lateral malleoli of the leading foot defines the mediolateral MoS.

5.3.5 Local Stability

Local dynamic stability during walking under five different test conditions was calculated using a reconstructed 5-dimensional time delayed state space of the angular velocity of the hip knee and ankle joints as well as the linear velocity of the head CoM and C7 marker (seventh cervical vertebra) in the anteroposterior , mediolateral and vertical directions for each test condition using

$$S(t) = [x(t), x(t+\tau), x(t+2\tau), \dots \dots, x(t+(d_e-1)\tau]$$ Equation 5-2

where τ is the time delay and d_e is the embedding dimension. The time delay for each set of data was estimated using the Average Mutual Information (AMI) function [185]. Briefly, the AMI function investigates the level of shared information between general data set and its corresponding time-delayed data set, and the maximum amount of shared information occurs at the first minimum of the AMI and correspondent to the chosen time delay. The embedding dimension of 5 was considered for all data set based on the previous studies [227, 229]. The rate of divergence of nearest neighbouring trajectories was estimated using the Maximum Finite Time Lyapunov Exponent (LyE) as follows:

$$D_j(i) = C_j e^{(LyE)*t}$$ Equation 5-3

where $D_j(t)$ is the distance between neighbouring trajectories and C_j is the initial distance between the trajectories. Taking the Napierian logarithm of both sides and least-square curve fitting to the average line yields:

$$y(i) = \frac{1}{\Delta t}\langle \ln d_j(i) \rangle = LyE.i + c, c = \frac{\ln C_j}{\Delta t}$$ Equation 5-4

where LyE is the slope of the fitted line and indicator for short-term local stability, and is a value between 0 and 1. LyE was estimated for 150 consecutive strides [25]. Higher LyE values indicate more divergence of nearest neighbours and poorer stability, whereas lower LyE values imply more convergence of adjacent neighbours and higher stability.

5.3.6 Orbital Stability

Orbital dynamic stability was also calculated through estimation of the MaxFM using angular velocity of the hip knee and ankle joints as well as the linear velocity of the head CoM and C7 marker in the sagittal, frontal and transverse planes for each test conditions. Time normalization of each stride to 101 points was performed using the cut up reconstructed state spaces into strides. A Poincaré plot was constructed to map the state of the system at stride k into stride $k + 1$ with

$$S_{k+1} = P(S_k) \qquad \text{Equation 5-5}$$

The fixed point (S^*) was then calculated by averaging the trajectories of the system over all strides using

$$S^* = P(S^*) \qquad \text{Equation 5-6}$$

The alterations of $[S - S^*]$ between stride k to stride $k + 1$ was investigated using eigenvalues of a Jacobian matrix (J), calculated with

$$[S_{k+1} - S^*] = J(S^*)[S_k - S^*] \qquad \text{Equation 5-7}$$

The eigenvalues of the Jacobian matrix (J) are called Floquet multipliers [203]. A stable system is characterized by a MaxFM of less than one.

5.3.7 Slip and Trip Propensity Measures

Backward slip propensity was quantified for each walking condition using Required Coefficient of Friction (RCoF) which is the minimum required coefficient of friction to prevent slipping [58]. The anteroposterior and vertical components of ground reaction force were used to calculate RCoF:

$$RCoF = \frac{F_{Anteroposterior}}{F_{Vertical}} \qquad \text{Equation 5-8}$$

Higher RCoF indicates greater risk of slipping.

Trip propensity was measured for each testing condition using the height of foot clearance with respect to the walking surface at mid swing. Lower MFC height implied greater risk of tripping during walking.

5.3.8 Detection of the Most Destabilizing Secondary Task among the Performed Cognitive and Sensory Additional Tasks during Walking

To detect which type of perturbation affected gait the most, a customized version of a newly developed approach by Roeles et al. (2018) was employed. According to this method, the average of MFC height and MoS during baseline walking were first calculated for each participant [277]. The level of distraction applied by a secondary task during walking then was quantified by calculation of differences between MoS and spatiotemporal variables at each of the first four post-perturbation steps (right and left) during dual-task walking trials with and averaged values for right and left steps during baseline walking:

$$Deviation = \sum_{i=1}^{2} \sum_{j=1}^{4} \sqrt{((Ave(i) - DT(i + (j-1) * 2))^2} \qquad \text{Equation 5-9}$$

where Ave is the average value of MFC height and MoS during baseline trial for i^th side (with i=1 representing the left side and i=2 representing the right side), and DT is the value of MFC height and MoS during dual-task trials for each side at the j^th step.

Since accommodation of perturbation starts from the first steps of walking, and participants adapt to changes and challenges applied by each type of perturbation after a number of steps (which varies person to person), a fixed number of initial steps (the first four steps) at the beginning of each dual-tasking trial were selected to evaluate the initial response of individuals to each type of perturbation.

5.3.9 Statistical Analysis

A repeated measure analysis of variance (ANOVA) was performed to investigate the influence of dual-tasking on gait stability and variability. The dependent variables were mean and variability of spatiotemporal measures, RCoF, MFC height, MoS, LyE, MaxFM, DFA scaling component and Deviation, while the independent factors were the test conditions i.e. baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli. Anatomical movement directions (anteroposterior, mediolateral, and vertical) were also included as independent variables (just in stability measures). Significance level was defined as $p \leq 0.05$All statistical analyses were performed using SPSS 25 (SPSS Inc., Chicago, IL).

5.4 Results

5.4.1 Spatiotemporal Measures

Step length and step time were significantly affected by tasks performed concurrently during walking. For example, step length was significantly larger during low cognitive task (mean difference: 1.2 cm, p=0.017), high cognitive task (mean difference: 2.4 cm, p=0.001), auditory stimuli (mean difference: 1 cm, p=0.036) and visual stimuli (mean difference: 1 cm, p=0.044) compared to baseline walking (Table 5-1). Step time also increased under perturbations caused by performing low cognitive task (mean difference: 1.4 cm, p=0.039) and high cognitive task (mean difference: 2.9 cm, p=0.003) during walking compared to single-task walking. Step time was longer during walking under auditory and visual perturbations, but it was not statistically significant (p>0.05). There were no significant changes in step width and double support time during dual-task trials compared to and baseline walking (p>0.05). Participants exhibited significantly greater step width variability during auditory (mean difference: 0.3 cm, p<0.001) and visual (mean difference: 0.2 cm, p=0.003) stimuli trials relative to baseline walking (Table 5-1). A similar trend was observed during walking under cognitive perturbations, but it was not statistically significant (p>0.05). Variability of step length, step time and double support time did not differ significantly between dual-task and single-task trials (p>0.05).

Table 5-1: Summary of the spatiotemporal parameters during walking trials, including Baseline walking, and walking under Low Cognitive Load, High Cognitive Load, Auditory Stimuli and Visual Stimuli. Acronyms are as follows: cm, centimetre; cs, centisecond.

	Baseline	*Low Cognitive Load*	*High Cognitive Load*	*Auditory Stimuli*	*Visual Stimuli*
Step length (cm)	51.5 ± 8.0	52.8 ± 8.5	53.9 ± 7.8	52.6 ± 8.1	52.5 ± 7.6
Step length variability (cm)	2.0 ± 0.6	2.0 ± 0.7	1.9 ± 0.6	2.2 ± 0.8	2.0 ± 1.1
Step width (cm)	8.5 ± 3.2	8.9 ± 3.2	8.9 ± 3.0	8.6 ± 3.2	8.4 ± 3.2
Step width variability (cm)	2.2 ± 0.5	2.2 ± 0.4	2.3 ± 0.4	2.5 ± 0.6	2.4 ± 0.5
Step Time (cs)	61.0 ± 5.5	62.4 ± 7.1	63.9 ± 6.4	62.2 ± 5.8	62.2 ± 5.7
Step time variability (cs)	2.3 ± 1.4	2.2 ± 0.9	2.1 ± 0.9	2.6 ± 1.1	2.1 ± 1.1
Double support time (cs)	22.4 ± 4.6	22.6 ± 5.1	23.5 ± 5.2	23.1 ± 4.7	23 ± 5.0
Double support time variability (cs)	4.1 ± 3.3	3.9 ± 3.2	4.8 ± 3.1	5.1 ± 3.4	4.9 ± 3.8

5.4.2 Margin of Stability

Mediolateral and anteroposterior MoS appeared to be unaffected during dual-tasking trials compared to baseline walking ($p>0.05$)

5.4.3 Local Stability

Statistical analysis showed no significant difference in LyE values of head and trunk motions between dual and single-task walking trials in either anatomic direction ($p>0.05$). With respect to local dynamic stability across the lower extremity joints, the ankle joint was particularly affected by perturbations caused by dual-tasking during walking, whereas there were no significant changes in LyE values of the hip and knee joints due to cell phone usage while walking compared to baseline trial ($p>0.05$). In the right ankle joint, significantly larger LyE values were observed in the mediolateral direction during high cognitive load trial compared to baseline walking ($p=0.005$) (Figure 5-2A). Low cognitive load, auditory and visual perturbations were also associated with increased LyE of the right ankle joint in the mediolateral direction, but none of the observed differences were statistically significant ($p>0.05$). For the right ankle joint in the anteroposterior direction, increased LyE values were also detected during performing low cognitive load ($p=0.036$), high cognitive load ($p=0.015$) and visual ($p=0.022$) additional task while walking compared to baseline walking (Figure 5-2B). Auditory perturbation also resulted in local instability of the right ankle joint in the anteroposterior direction, but it was not statistically significant ($p>0.05$). Subjects' right ankle movements in the vertical direction were more locally unstable during walking under low cognitive load ($p=0.001$), high cognitive load ($p=0.008$), auditory ($p=0.046$) and visual ($p<0.001$) perturbations than walking under no perturbation (Figure 5-2C). In the left ankle joint, LyE values did not differ significantly in the mediolateral and anteroposterior directions between dual and single-task walking trials ($p>0.05$) (Figure 5-2D & E). For the left ankle joint in the vertical direction, LyE increased in response to perturbations caused by performing low cognitive load task during gait compared to baseline walking ($p<0.001$) (Figure 5-2F). We also observed non-significant increase of LyE for the left ankle joint in the vertical direction in response to high cognitive load, auditory and visual perturbations compared to walking under no perturbation ($p>0.05$).

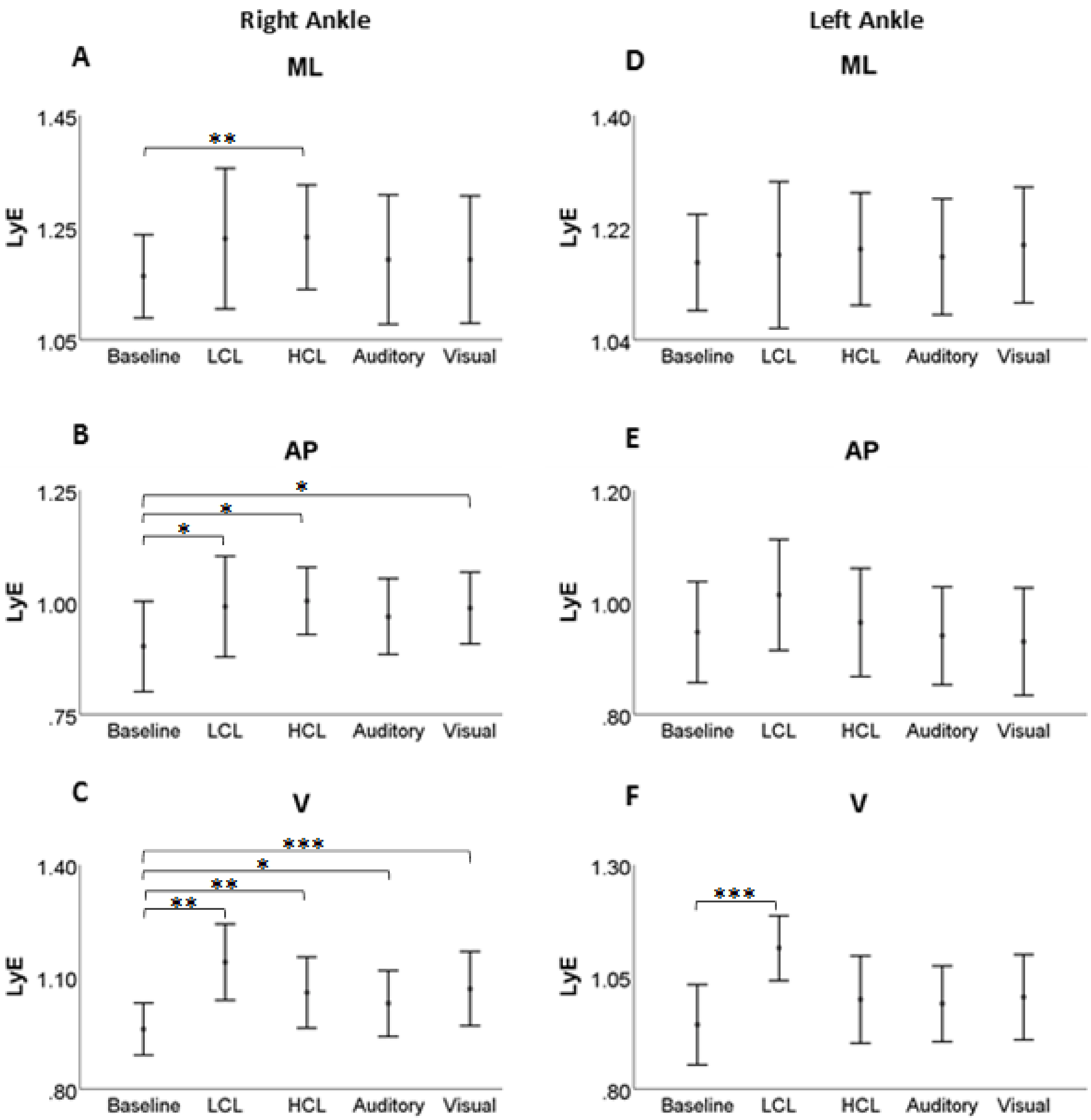

Figure 5-2: LyE values for the right (A, B, C) and left (D, E, F) ankle joints in the mediolateral, anteroposterior and vertical directions during baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli trials. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

5.4.4 Orbital Stability

Walking under high cognitive load perturbation resulted in larger head MaxFM in the mediolateral direction compared to baseline walking ($p=0.038$) (Figure 5-3A). When participants walked under low cognitive load, auditory and visual perturbations, no significant difference in orbital stability of the head motion in the mediolateral direction relative to the baseline trial was detected ($p>0.05$). In the anteroposterior direction, MaxFM of the head decreased in response to low ($p=0.015$) and high cognitive load ($p=0.012$) perturbations compared to walking under no perturbation (Figure 5-3B). We also observed smaller anteroposterior head MaxFM during walking under auditory and visual perturbations compared to baseline walking, but none of them were statistically significant ($p>0.05$). In the vertical direction, subject's head movement exhibited smaller MaxFM values during high cognitive load perturbation compared to baseline walking ($p=0.003$) (Figure 5-3C). Low cognitive load, auditory and visual perturbations did not have a significant effect on orbital stability of the head in the vertical direction ($p>0.05$). For the trunk segment, orbital stability in three anatomic directions was not affected significantly by dual-tasking while walking compared to baseline walking ($p>0.05$) (Figure 5-4A, B & C).

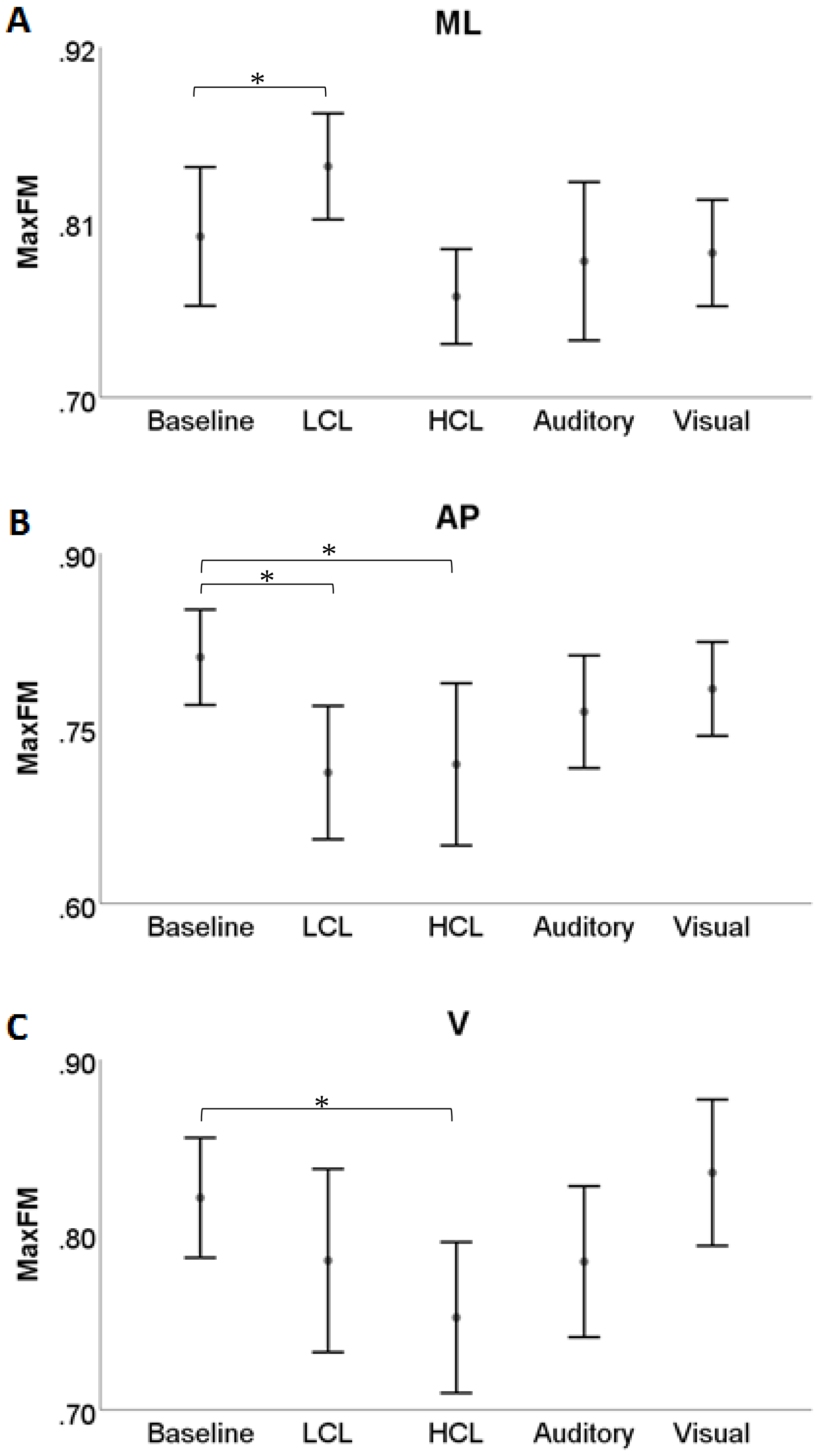

Figure 5-3: MaxFM values for the head in the mediolateral (A), anteroposterior (B) and vertical (C) directions during baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli trials. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

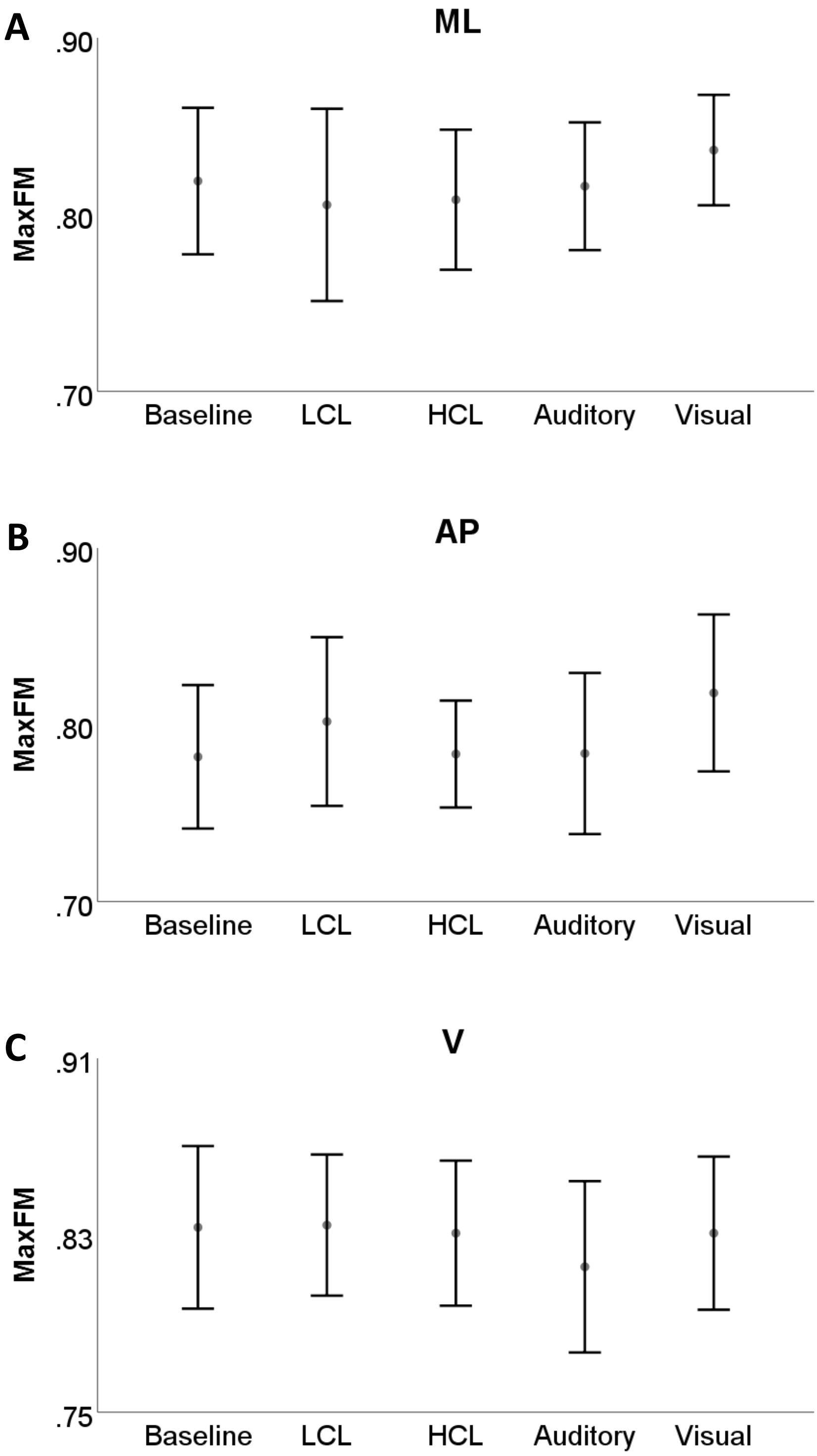

Figure 5-4: MaxFM values for the trunk in the mediolateral (A), anteroposterior (B) and vertical (C) directions during baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli trials. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

Considering orbital stability across the lower extremity joints, dual-tasking during walking affected the ankle joint more than the knee and hip joints. In the right ankle joint, MaxFM decreased in response to high cognitive load (p=0.001) and visual (p=0.009) perturbations compared to baseline walking (Figure 5-5A). We also observed decreased MaxFM during walking under low cognitive load and auditory perturbations relative to baseline walking, but none of them were statistically significant (P>0.05). For the right ankle joint in the anteroposterior direction, walking under high cognitive load was associated with decreased MaxFM compared to baseline walking (p=0.044) (Figure 5-5B). Low cognitive load and visual perturbations resulted in smaller MaxFM, and auditory perturbation resulted in larger MaxFM values compared to baseline trial, but none of them were statistically significant (P>0.05). Subjects' right ankle movements in the vertical direction were more orbitally stable during walking under low cognitive load (p=0.010), high cognitive load (p=0.005), auditory (p>0.05) and visual (p=0.006) perturbations than walking under no perturbation (Figure 5-5C). In the left ankle joint, low cognitive load (p=0.025) and visual (p=0.05) perturbations resulted in decreased MaxFM in the mediolateral direction compared to baseline walking, but orbital stability remained unchanged during walking under high cognitive load and auditory perturbations (p>0.05) (Figure 5-5D). In the anteroposterior and vertical direction, orbital stability of the left ankle joint appeared to be unaffected by dual-tasking while walking compared to single-task walking (p>0.05) (Figure 5-5E & F).

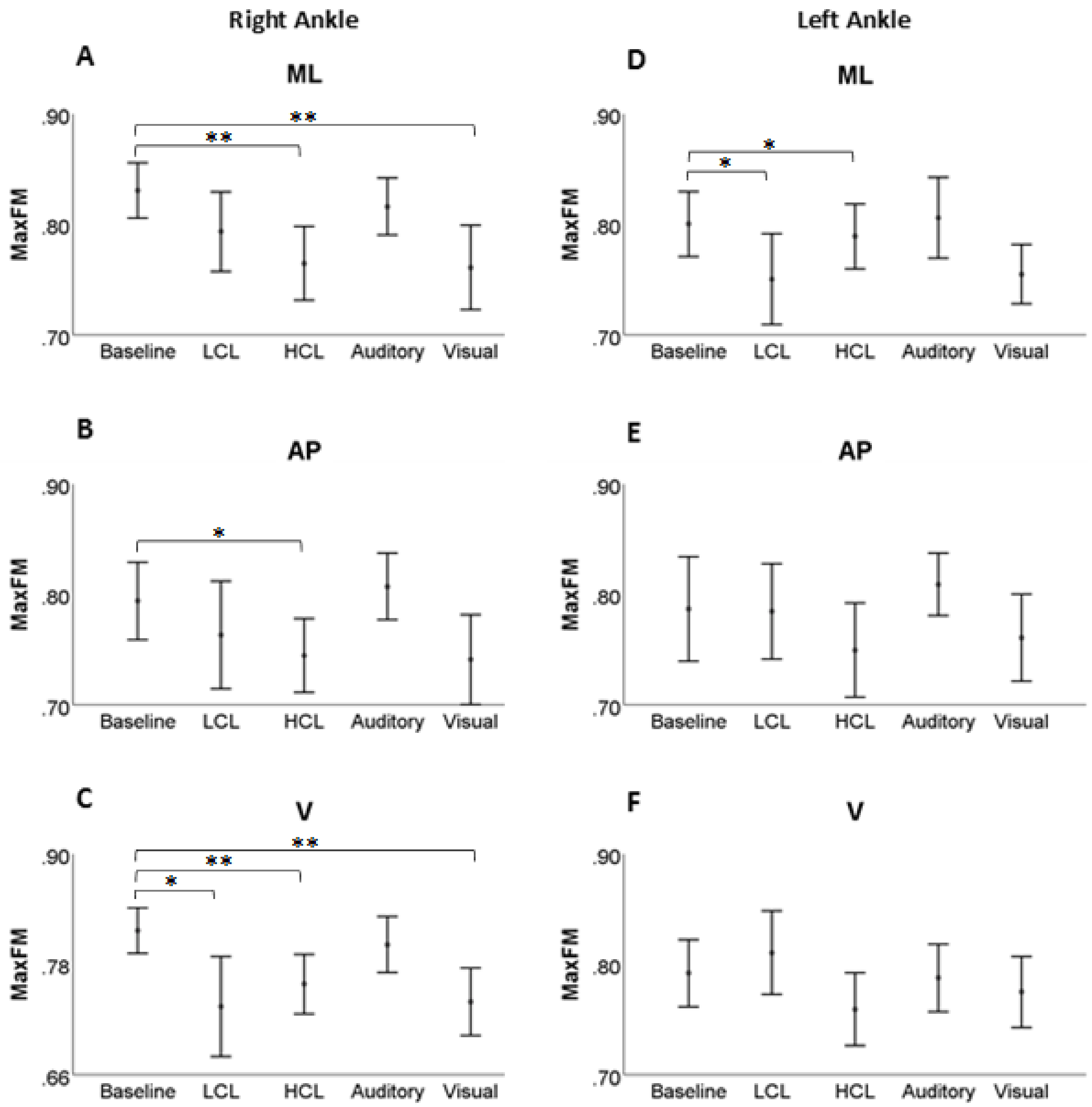

Figure 5-5: MaxFM values for the right (A, B, C) and left (D, E, F) ankle joints in the mediolateral, anteroposterior and vertical directions during baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli trials. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

In the right hip joint, there were no significant changes in orbital stability in the mediolateral and anteroposterior direction during dual-task trials compared to baseline walking (p>0.05) (Figure 5-6A & B). In the vertical direction, subjects exhibited smaller MaxFM of the right hip joint during walking under low cognitive load perturbations compared to baseline trial (p=0.03) (Figure 5-6C). We also detected decreased MaxFM of the right hip joint due to high cognitive load, auditory and visual perturbations, but none of them were statistically significant (p>0.05). In the left hip joint, no significant differences were found in orbital stability in the mediolateral and vertical directions between dual-task and single task trials (p>0.05) (Figure 5-6D & F). In the anteroposterior direction, subjects demonstrated significantly smaller MaxFM values during walking under low cognitive load compared to baseline walking (p=0.05) (Figure 5-6E). High cognitive load, auditory and visual perturbations also resulted in decreased MaxFM of the left hip joint in the anteroposterior direction, but none of them were statistically significant (p>0.05). Orbital stability of the right knee joint in anatomic directions appeared to be unaffected by dual-tasking compared to single-task walking (p>0.05), except decreased MaxFM in the vertical direction during walking under low cognitive load compared to baseline walking (p=0.014) (Figure 5-7A, B & C). For the left knee joint, there was no evidence that dual-tasking has an influence on orbital stability in the mediolateral, anteroposterior and vertical directions relative to baseline walking (p>0.05) (Figure 5-7D, E & F).

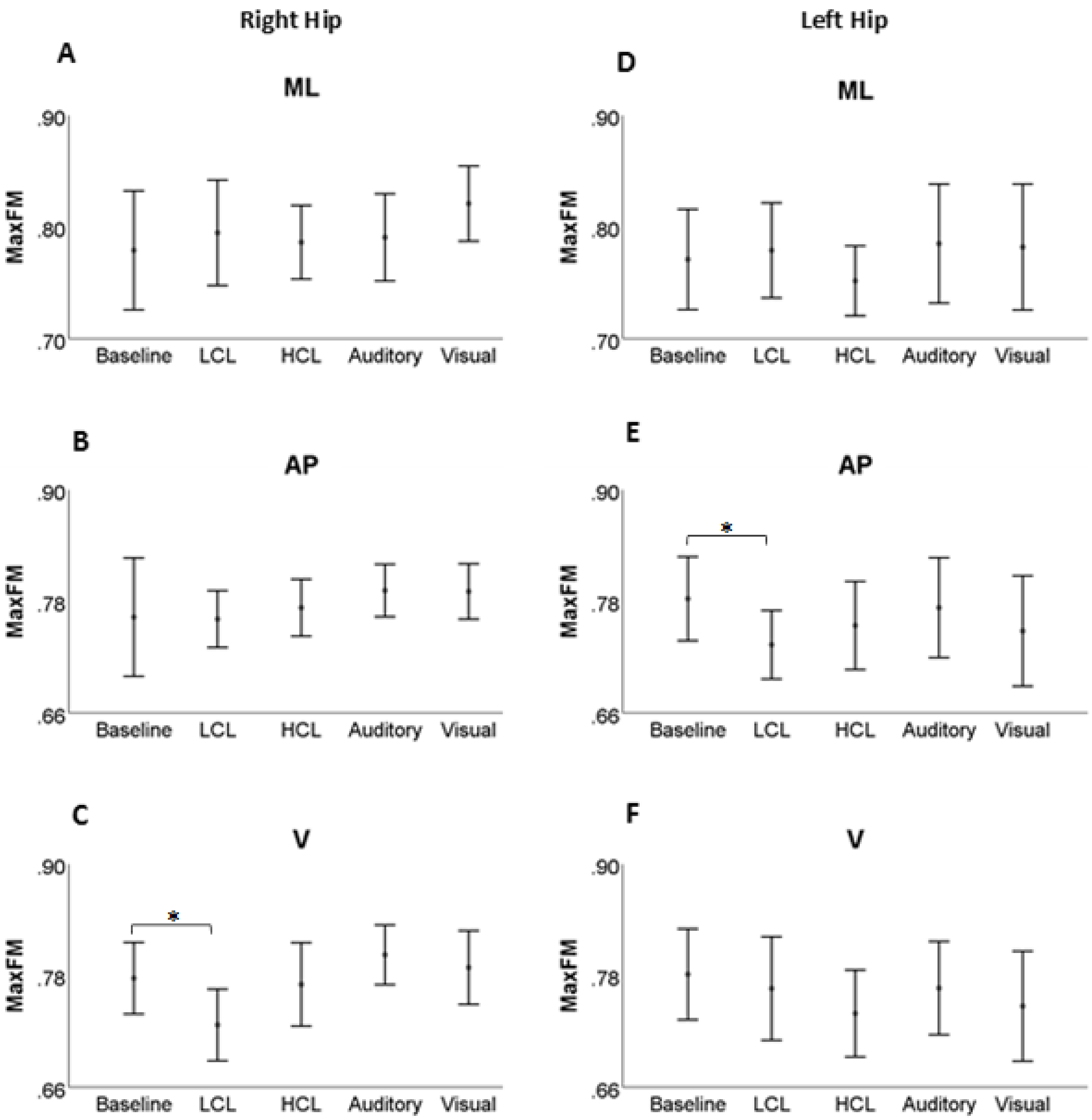

Figure 5-6: MaxFM values for the right (A, B, C) and left (D, E, F) hip joints in the mediolateral, anteroposterior and vertical directions during baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli trials. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

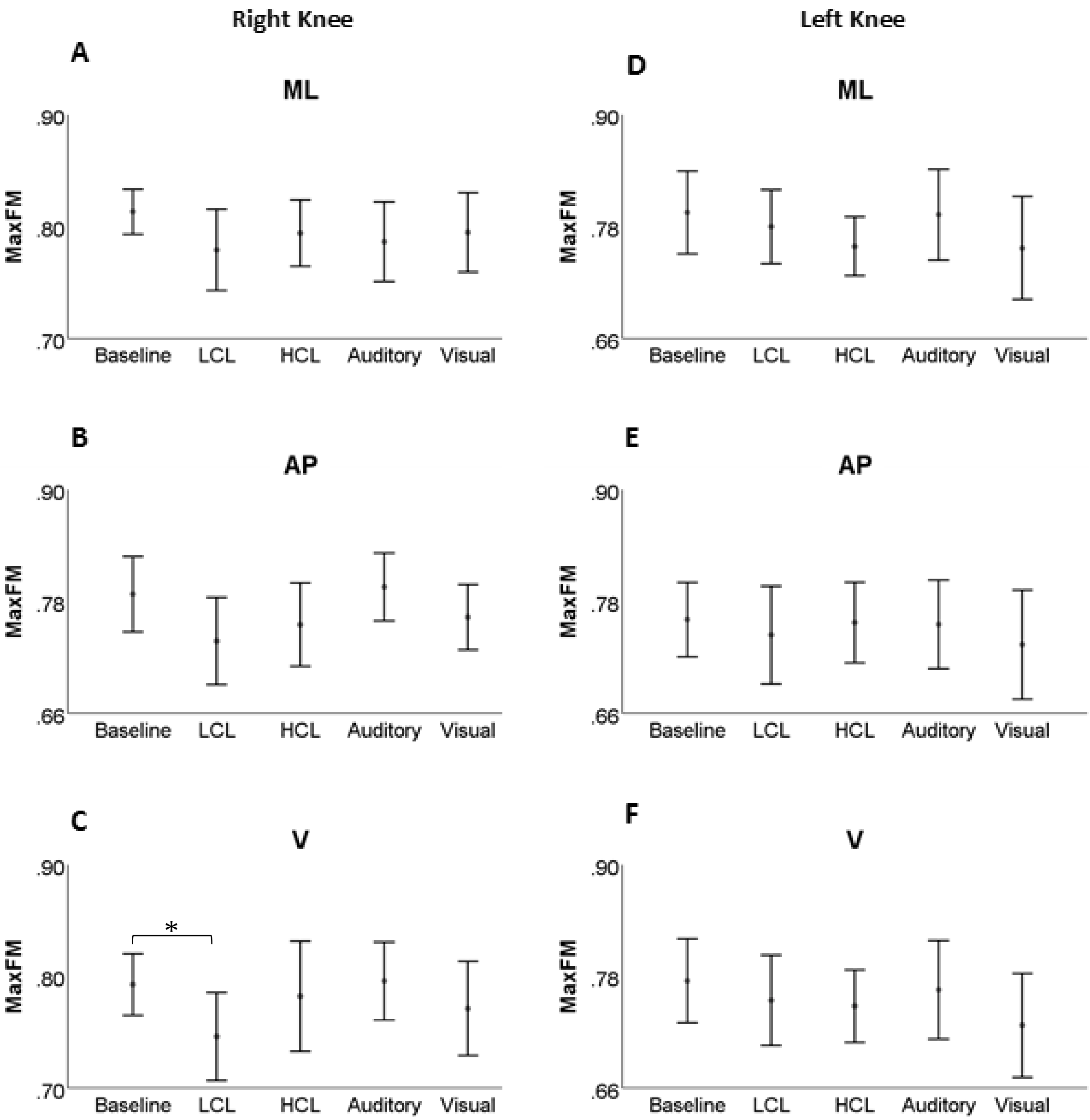

Figure 5-7: MaxFM values for the right (A, B, C) and left (D, E, F) knee joints in the mediolateral, anteroposterior and vertical directions during baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli trials. Greater LyE values indicate less local stability of the head. Error bars denote one standard deviation of LyE values within each direction in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral; V, vertical.

5.4.5 Slip and Trip Propensity Measures

Compared to baseline walking, participants exhibited slightly increased RCoF during walking under high cognitive load and visual perturbations, and decreased RCoF during walking under low cognitive load and auditory perturbations, but none of them were statistically significant (p>0.05) (Figure 5-8).

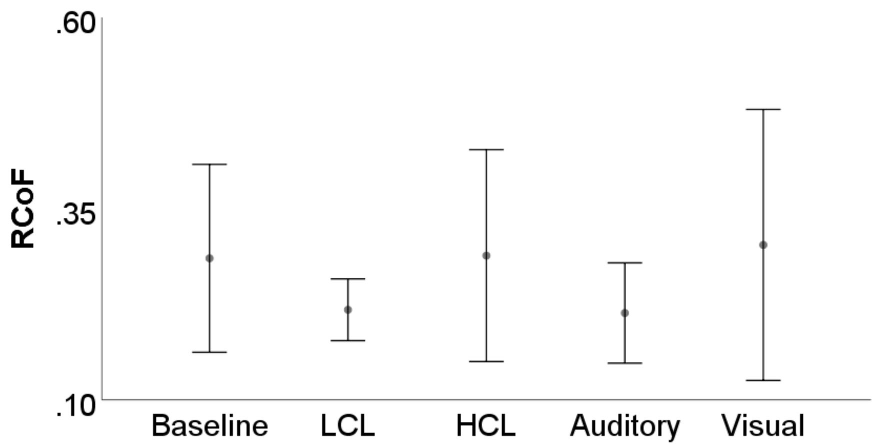

Figure 5-8: RCoF values during baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli trials. Greater RCoF values indicate greater risk of slipping. Error bars denote one standard deviation of RCoF values in either condition. One asterisk indicates p<0.05, two asterisks indicate p<0.01 and three asterisks indicate p<0.001.

MFC height decreased significantly in response to low cognitive load (mean difference: 0.001 m, p=0.026), high cognitive load (mean difference: 0.003 m, p=0.006) and auditory perturbations (mean difference: 0.002 m, p=0.015) while walking compared to walking under no perturbation (Figure 5-9). Visual perturbations also caused a non-significant decrease in MFC height relative to that during baseline walking (mean difference: 0.002 m, p=0.051).

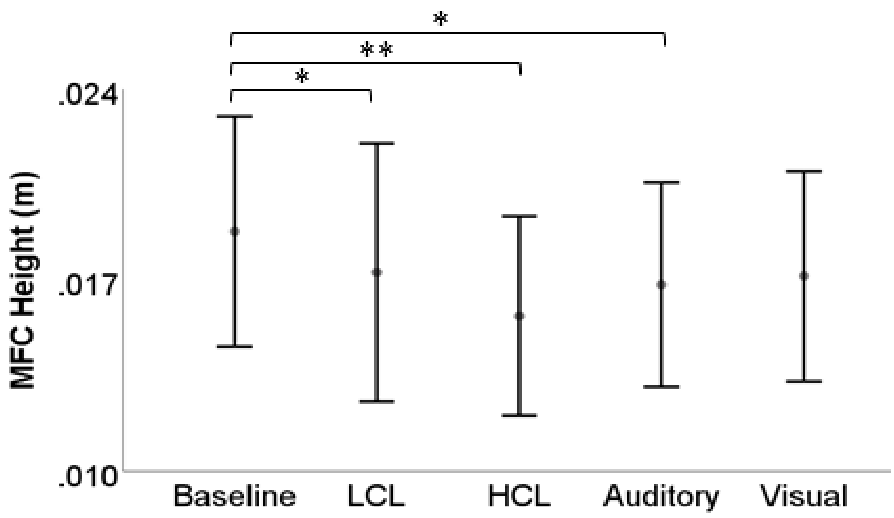

Figure 5-9: MFC height values during baseline walking, low cognitive load (LCL), high cognitive load (HCL), auditory stimuli and visual stimuli trials. Smaller MFC height values indicate greater risk of tripping. Error bars denote one standard deviation of MFC height values in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$.

5.4.6 Detection of the Most Destabilizing Secondary Task among the Performed Cognitive and Sensory Additional Tasks during Walking

Participants exhibited different gait patterns during walking under different types of perturbations. High cognitive load perturbation caused the greatest deviation in MFC height, and ML MoS at HC compared to other types of perturbations, but it was not statistically significant ($p>0.05$) (Figure 5-10A & B). Considering AP MoS at HC and MFC, low cognitive load perturbation was associated with larger deviation compared to other types of perturbations, but these were also not statistically significant ($p>0.05$) (Figure 5-10C & D).

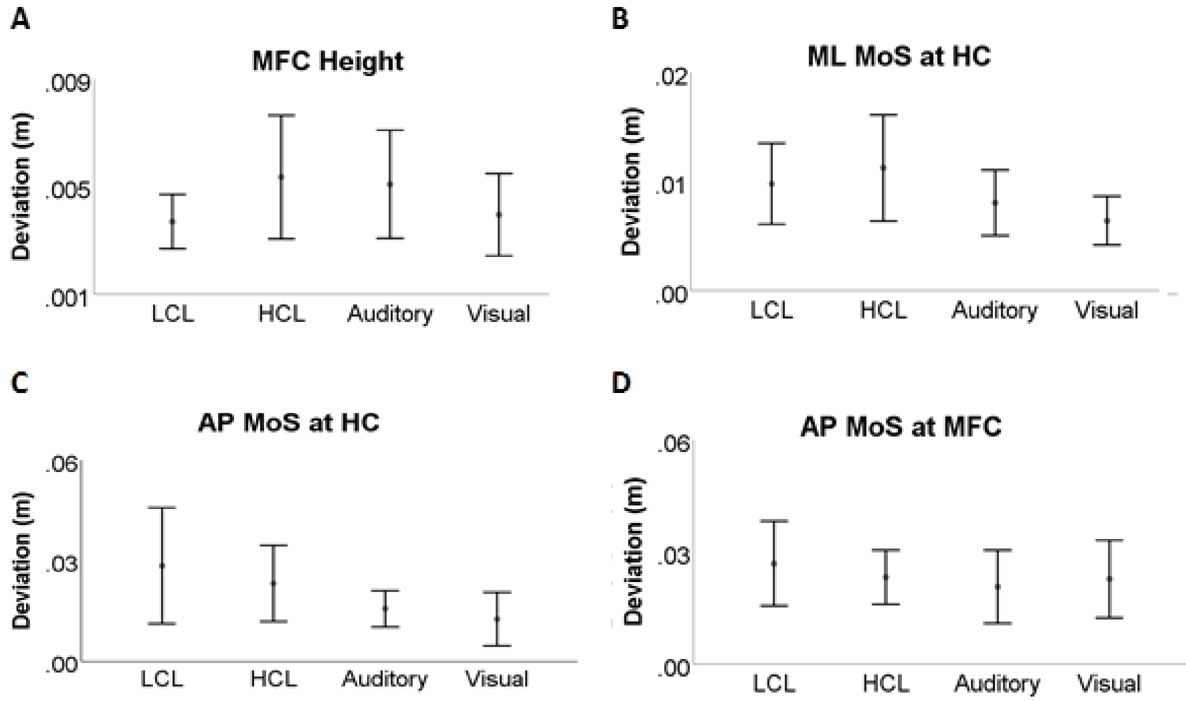

Figure 5-10: Mean deviation of MFC height (A), mediolateral MoS at HC (B), anteroposterior MoS at HC (C) and anteroposterior MoS at MFC (D) from baseline walking during walking under low cognitive load (LCL), high cognitive load (HCL), auditory and visual perturbations. Error bars denote one standard deviation of calculated deviation values in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral.

5.5 Discussion

Data from several studies have identified decreased balance of the body associated with performing an additional task during walking [354-357]; however, the influence of performing additional sensory and cognitive tasks during walking on dynamic stability is not well understood. The objective of the present study was to quantify the effect of auditory, visual and cognitive tasks undertaken while walking on stepping parameters and their variability, MoS, local stability, orbital stability, slip and trip propensity using motion trajectory of the head and trunk, as well as hip, knee and ankle joint. We also aimed to detect which type of sensory or cognitive perturbation is distracting the most during walking. A summary of some of the important findings are demonstrated in Table 5-2.

Table 5-2: Summary of the observed significant changes during low cognitive load (LCL), high cognitive load (HCL), auditory stimuli (Auditory) and visual stimuli (Visual) trials. Acronyms are as follows: AP, anteroposterior; ML, mediolateral.

	LCL	*HCL*	*Auditory*	*Visual*
Step Length	Increased	Increased	Increased	Increased
Step Time	Increased	Increased	-	-
Step Width Variability	-	-	Increased	Increased
ML LyE of Right Ankle	-	Increased	-	-
AP LyE of Right Ankle	Increased	Increased	-	Increased
ML LyE of Right Knee	Decreased	-	-	-
ML MaxFM of Head	-	Increased	-	-
AP MaxFM of Head	Decreased	Decreased	-	-
ML MaxFM of Right Ankle	-	Decreased	-	Decreased
AP MaxFM of Right Ankle	-	Decreased	-	-
ML MaxFM of Left Ankle	Decreased	-	-	Decreased
AP MaxFM of Left Hip	Decreased	-	-	-
MFC Height	Decreased	Decreased	Decreased	-

The present study showed that participants walked with longer and slower steps in response to cognitive perturbations and took longer steps during walking under sensory perturbations. Collectively, these findings, which are contrary to our hypothesis, suggest that participants were unable to employ their usual normal walking strategies to consistently respond to distractions associated with dual-tasking while walking. Hak et al. (2013) suggested that taking shorter and/or more frequent steps are two prominent balance maintenance strategies in young individuals [79]. In this study, however, participants did not appear to adopt these strategies during walking under perturbations, indicating that secondary task was prioritized during dual-task trials by young individuals [324]. This implies that perturbations associated with dual-tasking increased the risk of balance loss. At the same time, we did not observe any significant changes in MoS at HC and MFC during dual-task trials compared to single-task walking. A possible explanation for this contradiction could be sufficiently large BoS area that participants made during dual-task trials which allowed increased step length and step time without increased probability of balance loss. This supports evidence from previous observations (e.g. Hak et al., 2013; Hof et al., 2007), who found contradictory trends in stability measures and/or stepping parameters during ambulation [59, 218].

The results of this study showed that participants walked with high variability of step width while performing an additional sensory task, and no significant difference while performing additional cognitive tasks. This finding is consistent with that of O'Connor and colleagues (2012) who reported 65% increase in step width variability during walking under visual perturbation [378], and those of Terrier (2012) who found more variable step width during slow walking under auditory perturbations [379]. The observed changes in step width in response to sensory perturbations suggest that participants adopted a cautious gait pattern and attempted to preserve lateral stability by modifying placement of swing foot [58]. One can surmise that increased variability could be an indicator of instability and higher risk of side-to-side balance loss [380]. Since we did not observe any significant changes in the mediolateral MoS, the former notion that linked higher step width variability with a strategy toward gait stabilization seems to be more realistic.

In contrast to our hypothesis, in the current study, it was found that none of the sensory and cognitive perturbations affect MoS in the anteroposterior and mediolateral directions at HC and MFC. One reason behind unaffected MoS in the mediolateral direction could be increased step time during walking under perturbations compared to baseline walking, indicating that individuals prioritized the secondary task during ambulation rather than balance maintenance [79]. It has also been shown that modulating step width with respect to CoM during walking is an optimal strategy for active balance control in the mediolateral direction [71, 381]. Hence, the observed high variability of step width in our study could be another reason for limited changes of MoS in the transverse plane. In the anteroposterior direction, it has been suggested that increased step length and decreased step time are two important strategies for balance maintenance [224]. In this study, however, none of these adaptations were observe during walking trials. This could further support the idea that preservation of balance in the sagittal plane relies on adjustment of centre of pressure through the ankle moment rather than foot placement strategy [71, 82].

Head stability has been previously shown to be important for whole body balance maintenance, since it contributes to navigational control and gaze stability through preservation of visual acuity during walking [312, 342, 382]. The present study showed that there were no significant changes in local stability of the head due to cognitive and sensory dual-tasking while walking in young adults. A possible explanation might be stability of the trunk motion during walking trials regardless of additional tasks difficulty. It has been shown that the trunk plays a critical role in stabilization of head motion during ambulation by providing a stable platform for the head and neck [312]. This stable platform dissipates all oscillations from the lower limbs and pelvis to the head and facilitates smooth movements of the head which is necessary for gaze stabilization. Another possible factor for unaffected head stability could be free movement of the arms during locomotion which contributes to limited head yaw and gaze stability [383].

In the current study, it was found that performing additional cognitive and sensory tasks during walking do not affect local dynamic stability of trunk motion in young adults. These results are in accordance with recent studies by Soangra et al. (2017) and Roeles et al. (2018) indicating robustness of trunk stability to cognitive, auditory and visual perturbations while walking [58, 277]. One reason behind unchanged trunk stability due to dual-tasking could be unrestricted arm swing which contributes to regaining balance after a perturbation during walking [257]. Another possible explanation for this could be the observed adaptations e.g. increased step length and high variability of step width, during dual-task trials which helped subjects to maintain approximately similar level of balance to single-task walking condition

[296]. The unaffected trunk stability may also be related to a trunk stiffening strategy to keep head motion stable during ambulation under perturbations [384]. However, we did not quantify trunk stiffness in this study, and this ought to be a focus for future research.

Local stability of the ankle, knee and hip joint motions appeared to be mostly affected by performing cognitive additional task rather than sensory secondary tasks while walking. This suggests that the auditory and visual stimuli were not challenging enough to distract the subject's motion during walking compared to cognitive concurrent task. Further research should be undertaken to investigate the effect of larger sensory perturbations through dual-tasking while walking on dynamic stability. These findings also provide further support to the idea of sensory reweighting by the central nervous system to preserve balance during walking under sensory perturbations [71, 385]. We also observed most of the local stability changes in response to sensory and cognitive perturbations in the ankle joint rather than the hip and knee joints. These results are in agreement with those of Kang and Dingwell (2009) who found that inferior segments are less stable compared to superior segments [311]. Taking these findings together with the obtained results for the local stability of the head and trunk motions, we can conclude that the upper body is more robust to cognitive and sensory perturbations compared to the lower body, and whole body stability analysis can be estimated solely based on the upper body. These also match those observed in earlier study by Kang and Dingwell (2009) [311].

In the current study, different local stability values of the ankle, knee and hip joint motions were observed in the right and left sides during all walking conditions. While limb effect analysis was out of focus of this study since it has not been recommended by previous similar studies, a possible explanation for the observed asymmetry might be different roles of each limb during gait. It has been shown that locomotion is associated with 'functional asymmetry', in which the dominant leg is responsible for forward progression while the nondominant leg is required for support [386]. Since all participants were right limb dominant in this study, the observed higher LyE of the right ankle, knee and hip joints in the mediolateral direction compared to that in the left side may be attributed to utilization of the right limb joints for making kinematic adjustments in response to the applied perturbations and the left limb joints for preserving stability during walking under those perturbations. These results reflect those of Segal and colleagues (2008) who also found greater local stability in the left side lower body joints compared to that in the right side during walking under perturbation [387]. These findings imply that each limb has a different contribution in maintaining local stability during walking while engaging in an additional concurrent task.

In the present study, orbital stability analysis revealed that performing additional concurrent cognitive and sensory tasks is associated with greater movement stability. This finding was unexpected and suggests low sensitivity of MaxFM to cognitive and sensory perturbations as observed in an earlier study by Dingwell and colleagues [296]. A possible explanation for these results may be the lack of adequate information about periodicity of human gait as a dynamic system before conducting orbital stability analysis. Since periodicity is the main criterion of a system to be eligible for orbital stability analysis [25], and it is still not clear whether human gait meets this condition or not, these data must be interpreted with caution. Another source of uncertainty is the required number of strides to achieve more reliable results from orbital stability analysis. In the current study, analysis was completed on 150 consecutive gait cycles in accordance with a study by McAndrew and co-workers (2011) [291], but another study by Bruijn et al. (2009) suggested that MaxFM estimation should be performed within 300 strides to be more accurate [388]. These results, therefore, should be interpreted with caution and cannot be generalised to global stability changes under dual-tasking conditions.

In the current study, slip propensity remained unaffected during walking while performing additional sensory and cognitive tasks, which was contrary to our expectations. These results are in agreement with Soangra and Lockhart's (2017) findings that showed no significant changes in RCoF due to dual-tasking in young individuals [58]. The observed minimal change to slip propensity during dual-task trials might be attributed to increased step time which facilitated control of foot placement through receiving greater range of information in a longer duration [389]. In contrast to slip propensity, participants walked with greater trip propensity in response to sensory and cognitive perturbations caused by dual-tasking while walking. These results, which are in line with our hypothesis, reflect those of Hamacher et al. (2019) who also found that performing a secondary task during walking is associated with reduced MFC height in young individuals [390]. However, this outcome is contrary to that of Soangra and Lockhart's (2017) who found no significant changes in trip propensity during dual-tasking compared to single-task walking [58]. This discrepancy could be attributed to limited sample size of their study (n=7) compared to this study (n=19). Combining trip and slip propensity results during dual-task walking trials, it can be concluded MFC height is more sensitive to perturbations applied by additional concurrent task compared to RCoF.

Comparing different types of secondary tasks while walking, the present study found that ambulation in young individuals is more sensitive to perturbations associated with cognitively demanding concurrent tasks rather than additional sensory-based tasks. This might be attributed to difficulty level of cognitive secondary tasks compared to sensory additional tasks during walking which limited the ability of young individuals in shifting attention during walking [8, 391]. In this study, we did not measure the level of accuracy in the secondary cognitive tasks, but we asked participants to do their best. Compared to sensory perturbations, this may have motivated them to pay more attention to the cognitive task. Another possible explanation for these findings could be lower magnitude of the applied sensory perturbations through dual-task walking compared to threshold magnitude of accommodation for the same type of perturbation in healthy people [296]. One important area for future research could be determination of the threshold magnitude for different types of sensory perturbation to assess whether each person can tolerate and accommodate the applied perturbation without losing balance. In the current study, we also observed different levels of sensitivity for MFC height, MoS and spatiotemporal measures to cognitive and sensory perturbations. To be more precise, a frequency count of individuals' responses to cognitive and sensory perturbations was obtained (Table 5-3). It can be seen from the data in the table that some of the measures including double support time and MFC height were more sensitive to sensory perturbations, while other measures exhibited greater sensitivity in response to cognitive perturbations. These results are similar to those observed in an earlier study by Roeles et al. (2018) [277], suggesting that a mixed analysis of different gait measures are required to gain more reliable perspective of an individual's responses to different types of perturbations.

Table 5-3: Frequency count of the observed changes in MFC height, Double Support Time, Step Width, Step Length, Step Time, AP and ML MoS at HC and AP MoS at MFC in response to sensory and cognitive perturbations compared with the baseline trial. The larger / smaller value of a measure under each perturbation type indicates more / less sensitivity of that measure to the corresponding type of perturbation.

	Sensory Perturbations (%)	*Cognitive Perturbations (%)*
MFC Height	68.42	31.57
Double Support Time	57.89	42.10
Step Width	42.10	57.89
Step Length	21.05	78.94
Step Time	10.52	89.47
AP MoS at HC	31.57	68.42
ML MoS at HC	38.88	61.11
AP MoS at MFC	42.10	57.89

A limitation of this study is in the fact that data collection was performed using an instrumented treadmill to facilitate collection of a large data set of consecutive gait strides, as well as speed-matching of all walking trials for each participant. However, it has been suggested that treadmill walking can affect measurement of local and orbital dynamic stability indices [392], and may ultimately result in underestimation of LyE and MaxFM values. This study was also constrained by data concerning the quality of the additional cognitive task, which limits ability to quantify the level of cognitive distraction during walking. Having such information in the future may be useful to quantify the response of the human body to cognitive perturbations of different levels. Another variable that was not explored in our experiments was the type of song used to measure the effect auditory dual-tasking on dynamic stability, specifically, the music tempo. It has been shown that synchronization of walking pace with music beats during dual-task walking occur close to 120 BPM and individuals walk faster during walking while listening to a piece of music with higher tempo [393]. Only one music tempo was explored in the present study (160 BPM), which may have contributed to the non-significant outcome of music influence on gait. Finally, being limited to young participants makes these results less generalisable to other populations such as older adults or patients with musculoskeletal or neurological conditions. However, the techniques presented in this study provide a blueprint for future studies in other cohorts.

5.6 Conclusion

Young individuals adopt different strategies to cope with sensory perturbations caused by performing concurrent auditory and visual tasks while walking compared to cognitive perturbations arising from cognitive dual-tasking while walking. Performing additional concurrent cognitive and sensory tasks while walking may increase the risk of tripping. Gait patterns in young adults are more robust to sensory perturbations and more sensitive to cognitive perturbations during walking. Compared to the lower body, the upper body movement appears to be more controlled during cognitive and sensory dual-task trials in either principal anatomic direction, confirming that head stability is the main priority of an individual for postural control during walking, and human body attempts to adopt different strategies to attenuate the applied perturbation from the lower body toward the head.. Further research ought be undertaken to investigate threshold magnitudes for accommodation of cognitive and sensory perturbations in young adults as well as other populations which are more prone to balance loss during walking.

6 Conclusion

This dissertation was conducted to investigate the influence of internal and external perturbations arising from daily living activities on dynamic stability and variability in young adults. In the first research question, we attempted to understand the effect of the applied perturbation on the likelihood of balance loss and dynamic stability during walking. Our tools to assess the risk of balance loss were standard deviation, DFA, propensity of tripping and propensity of slipping. Using standard deviation, it was found that motor-cognitive dual-tasking due to cell phone usage while walking, and sensory perturbations due to performing additional auditory and visual tasks while walking are associated with greater risk of balance loss. Moreover, it was found that the risk of balance loss in young individuals is not sensitive to external perturbations caused by sloped terrains, as well as internal perturbations due to cognitive dual-tasking while walking. Using the nonlinear technique, it was found that statistical persistency of stepping parameters is not sensitive to internal perturbations, and somehow increased in response to external perturbations. So, the DFA technique did not find any changes in the likelihood of balance loss during walking under perturbations relative to walking under no perturbation. Using propensity of tripping, it was found that motor-cognitive perturbations caused by talking while walking, and cognitive and sensory perturbations increase the risk of tripping, while external perturbations due to sloped terrains do not increase tripping risk in young adults. Using the propensity of slipping, it was found that internal perturbations during walking does not have an effect on the risk of slipping in young individuals.

With respect to stability changes under the applied perturbations, our tools were a combination of one linear (MoS) and two nonlinear techniques (local and orbital stability). It was found that MoS in the sagittal plane is not sensitive to the applied internal perturbations during locomotion in young adults. In the frontal plane, however, MoS was only sensitive to cognitive-motor perturbations applied by talking while walking. Using local and orbital stability analysis of the head and trunk movements during walking under perturbations, it was found that upper body stability is mostly resistant to the applied internal perturbations. In regard to external perturbations, local stability of the upper body decreased during uphill walking and increased during downhill walking. Participants walked with decreased local stability of the ankle, hip and knee joints in response to internal perturbations, and most of the changes were observed in the ankle joint. Orbital stability of the ankle, hip and knee joints remained unaffected during walking under motor-cognitive perturbations, but it increased in response to cognitive and sensory perturbation, and most of the changes were observed in the ankle joint.

In the second research question, I assessed how individuals respond to applied internal and external perturbations, and maintain balance during walking. Participants took longer steps in response to motor-cognitive, cognitive and sensory perturbations during walking. Subjects also walked with longer step time during walking under motor-cognitive and cognitive perturbations. Wider steps were taken by individuals only in response to perturbations caused by talking while walking, and step width remained unaffected during walking under other types of motor-cognitive, cognitive and sensory perturbations. Participants walked with shorter step time and double support time during downhill walking, but they did not change their gait pattern substantially during uphill walking.

In the third research question, we attempted to investigate which type of secondary tasks while walking poses greater challenge to balance in young individuals. For this purpose, deviation of MoS and trip propensity values during dual-task trials from the corresponding values during baseline walking were calculated for each testing condition. With respect to motor-cognitive perturbations, talking while walking was found to be the most destabilizing task compared to texting and reading while walking conditions. Comparing sensory and cognitive perturbations, it was found that ambulation in young individuals is more sensitive to perturbations associated with cognitively demanding concurrent tasks rather than additional sensory-based tasks, though this finding was not significant.

In this section, we compared all perturbations including motor-cognitive, cognitive and sensory perturbations to investigate which one is more destabilizing during walking in young adults. The largest deviation of MFC height relative to baseline walking was attributed to perturbations caused by talking while walking, whereas the smallest deviation was observed during reading while (Figure 6-1A). In this regard, the only significant differences were found between Talking-Visual (mean difference: 1.88 mm, p=0.025), Talking-Texting (mean difference: 2.30 mm, p=0.009), Talking-Reading (mean difference: 3.14 mm, p=0.01), and Auditory-Reading (mean difference: 2.39 mm, p=0.026) conditions. The observed difference between talking and low cognitive load conditions was also approaching significance (mean difference: 2.13 mm, p=0.056).

With regards MoS in the mediolateral direction at HC, it was found that high cognitive load perturbation results in the largest deviation, while visual perturbation was associated with the smallest deviation from walking under no perturbation (Figure 6-1B). However, the observed differences were not statistically significant, but approaching significance (Talking-Visual: p=0.061, HCL-Visual: p=0.051, Texting-Visual: p=0.064). The smallest deviation of MoS values in the anteroposterior direction at HC was observed during visual perturbation, while the largest deviation was associated with low cognitive load perturbation with respect to corresponding values during baseline walking (Figure 6-1C). However, the only significant difference was found between high cognitive load and visual perturbations (mean difference: 10.6 mm, p=0.045). The observed difference between low cognitive load and visual perturbations was also approaching significance (mean difference: 15.9 mm, p=0.055). Using deviation of MoS at MFC, no significant differences were found between different perturbations relative to corresponding values during baseline trial (p>0.05) (Figure 6-1D).

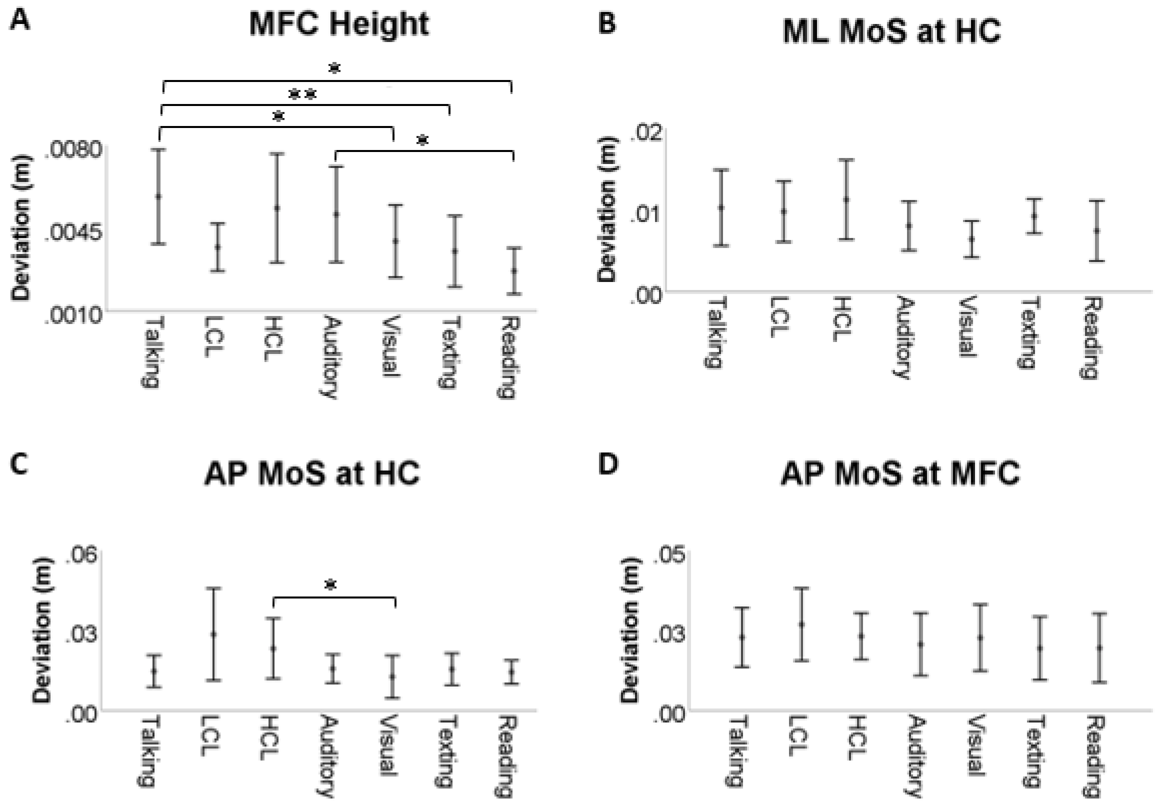

Figure 6-1: Mean deviation of MFC height (A), mediolateral MoS at HC (B), anteroposterior MoS at HC (C) and anteroposterior MoS at MFC (D) during walking under perturbations caused by cell phone talking (Talking), low cognitive load additional task (LCL), high cognitive load additional task (HCL), auditory additional task (Auditory), visual additional task (Visual), cell phone texting (Texting) and cell phone reading (Reading) from baseline walking. Error bars denote one standard deviation of calculated deviation values in either condition. One asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$ and three asterisks indicate $p<0.001$. Acronyms are as follows: AP, anteroposterior; ML, mediolateral.

What can be clearly seen in these results is direction-specific influence of perturbations applied through dual-tasking on gait in healthy adults. Some of them tend to increase instability in the mediolateral direction and the rest are more destabilizing in the anteroposterior direction. Taking all the results together, it was shown that perturbations caused by talking and performing additional cognitive task during walking posed greater challenge to gait in young adults compared to that caused by sensory additional concurrent tasks and cell phone texting/reading while walking. On the other hand, reading and performing additional visual task while walking appeared to be less challenging for young individuals compared to that caused by cell phone talking/texting, and cognitive/auditory dual-tasking while walking. To further investigate these findings, we can look at step length, step width and step time during walking while performing these secondary tasks. During cognitive load trials, step length and step time deviation from baseline is higher compared to talking trial, indicating that participants did not need to make substantial changes to their gait pattern to maintain balance through taking shorter and faster steps (Figure 6-2A & B) [79]. It can thus be suggested that talking while walking was more destabilizing compared to cognitive dual-tasking while walking. This finding can also be inferred by looking at larger deviation of step width during talking trial compared to cognitive trials, suggesting greater effort by individuals to maintain balance in the mediolateral direction (Figure 6-2C). The observed large deviation of step length and step time while having control on balance during cognitive trials could be either due to inadequate challenge applied through cognitive secondary tasks to elicit deterioration of gait stability, or insensitivity of deviation measures to cognitive perturbations in in young adults [296].

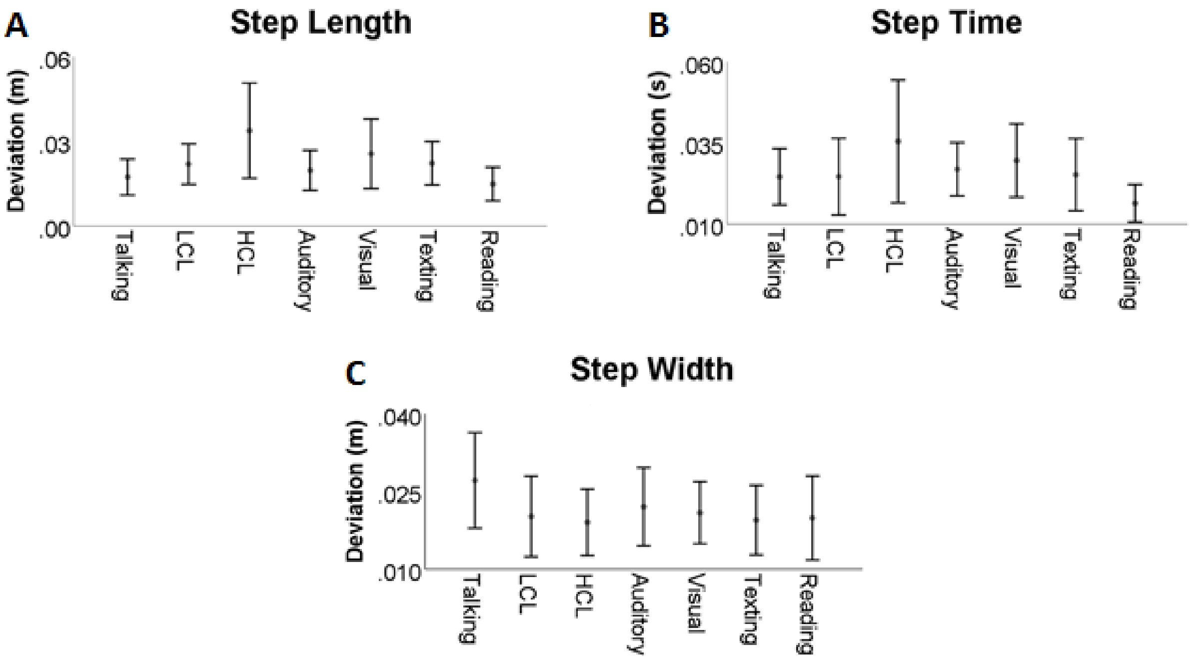

Figure 6-2: Mean deviation of Step Length (A), Step Time (B) and Step Width (C) during walking under perturbations caused by cell phone talking (Talking), low cognitive load additional task (LCL), high cognitive load additional task (HCL), auditory additional task (Auditory), visual additional task (Visual), cell phone texting(Texting) and cell phone reading (Reading) from baseline walking. Error bars denote one standard deviation of calculated deviation values in either condition.

Focusing on all dual-task walking conditions, participants tended to walk with increased step length, step width and step time relative to that during baseline walking. These findings might be attributed to insufficient taxing of motor, cognitive or sensory demands associated with secondary tasks in this study to elicit significant changes in gait adaptations, providing further support to the idea that individuals' response to the applied perturbations through dual-tasking while walking depends on the magnitude of the applied perturbation. The more distracting the secondary task during walking, the larger the impact on locomotion [15]. However, it is still not clear how much perturbation can be accommodated by individuals during walking while preserving global stability. One important line of future research could be investigation of the resilience of human gait to much larger perturbations (resulting in a fall) than what we employed in the present study.

In the current study, participants walked under different types of perturbations including external perturbations, as well as motor-cognitive, cognitive and sensory perturbations. Under some of the applied perturbations, participants were able to swing their arm during walking, while under a group of perturbations caused by cell phone usage while walking and somehow auditory dual-tasking, arm swing was relatively restricted. We expected to see greater reduction of the stability measures in the mediolateral direction during walking trials under arm swing restriction compared to walking under those perturbations without arm swing restriction. As expected, step width variability increased significantly during cell phone usage while walking, and average step width increased significantly during talking while walking. During other types of perturbations, there was no significant changes in average step width, and significant increase of step width variability was only observed during walking under sensory perturbations compared to single-task walking. Increased average and variability of step width during cell phone usage and auditory dual-tasking while walking shows participants' reliance on arm swing during walking to maintain balance in the mediolateral direction. During walking under cognitive and external perturbations, however, free arm swing contributed to greater stability in the frontal plane and less need of individuals for step width modulation. The observed changes in step with variability during walking under visual perturbation might be attributed to larger effect of visual perturbations in the mediolateral direction rather than the anteroposterior direction [71].

In this study, a novel method was devised to assess stability that measured MoS at HC and MFC, gait events associated with backward and forward balance loss, respectively. The major advantage of the approach employed was that it incorporates the direction of motion most commonly associated with falls at specific events in the gait cycle. A further advantage of the tested method is that it considers CoM velocity when calculating MoS relative to the BoS anterior boundary. As forward falls occur in the direction of the velocity vector, which is not necessarily perpendicular to the anterior BoS boundary, calculation of the distance between XCoM and the BoS anterior border of in the direction of CoM velocity may improve stability analysis. The current version of this method can only be used to assess gait stability under internal perturbations. Further work is needed to establish a more general version for dynamic stability analysis in more complicated conditions.

Besides linear stability analysis technique, in the present study, nonlinear approaches were also used to investigate the effect of internal and external perturbations on dynamic stability of walking. Local and orbital stability analysis was performed on the head and trunk segments, as well as the hip, knee and ankle joints during dual and single task trials, while stability changes during walking over sloped surfaces was assessed only by local stability analysis of the head and trunk segments. It was found that stability changes in the lower body motions is more sensitive to the applied internal perturbations compared to the upper body motions. It can thus be suggested that stability analysis should be performed on the upper and lower bodies to provide us with a deeper understanding on the influence of perturbations on gait balance. Comparing two nonlinear measures of stability, local stability appeared to be more promising method of stability analysis. The reason behind unreliable outcome of orbital stability analysis could be either aperiodicity of human's gait under perturbations, or lack of uniformity in the reported calculation methods of MaxFM in the literature. In this study, for instance, analysis was completed on 150 consecutive gait cycles in accordance with a study by McAndrew and co-workers (2011) [291], but another study by Bruijn et al. (2009) suggested that MaxFM estimation should be performed within 300 strides to be more accurate [388].

In the present study, local stability analysis was performed on the head segment which to date has not been undertaken in the context of walking stability under perturbation. Stabilization of the head plays an important role in human postural control because the vestibular and visual systems are critical to balance and orientation of the body during ambulation. Head instability may, therefore, influence the feedback provided by the vestibular and visual systems required for gaze stabilization and postural control. In response to perturbations caused by dual-tasking while walking, stability of the head motion remained unchanged relative to single-task walking. This confirms that head stability is the main priority of an individual for postural control during walking, and human body attempts to adopt different strategies to attenuate the applied perturbation from lower body toward the head. Besides, the observed changes in local stability of the head and trunk movements during walking under external perturbations were similar in either anatomic direction, while they were different during walking under internal perturbations. These results suggest that rigidity of the upper body during walking increases under external perturbations and decreases in response to internal perturbations. From this, it can also be concluded that the applied internal perturbations were more destabilizing for the upper body compared to the applied external perturbations since increased flexibility of the upper body is a strategy that individuals adopt to attenuation more perturbations from the lower body toward the head and increase head stability [394].

In this study, we estimated RCoF and MFC height to understand the risk of slipping and tripping during walking under internal and external perturbations in young adults. With regards slip propensity, it was found that walking under internal perturbations does not have an effect on the risk of slipping in young adults. Considering the propensity of tripping, however, significant changes were observed during walking under internal perturbations, suggesting high sensitivity of MFC height values to the applied perturbations. We also calculated deviation of MFC height for each dual-task condition from baseline trial and observed significant differences between different perturbations. These results further support the idea that MFC height is an independent risk factor of falling that needs to be taken into account for further investigation of the risk of balance los during dual task walking [395].

Overall, the findings of this dissertation suggest that motor-cognitive dual-tasking due to cell phone usage while walking, cognitive and sensory perturbations due to performing additional auditory and visual tasks while walking may be associated with greater risk of tripping, as measured by a lower MFC height. Particularly, talking while walking, and cognitive and sensory dual-tasking while walking may ultimately lead to an increase in risk of tripping in young adults. The results of this investigation showed that internal and external perturbations during walking affect balance during gait in a direction-specific manner in young adults. Aim 1 showed that talking while walking increased the risk of forward balance loss, while texting and reading during walking were associated with higher risk of side-to-side balance loss. Aim 2 showed that up-sloped terrain increased the risk of balance loss during gait in the anteroposterior and vertical directions but not in the mediolateral direction. Aim 3 showed that cognitive and sensory perturbations affect gait balance mostly in the anteroposterior and vertical directions.

The findings of this investigation confirmed that head stabilization during ambulation has higher priority compared to other segments, and individuals attempted to adopt different strategies to attenuate perturbations from the lower body toward the head. The upper body was found to have more rigidity in response to external perturbations and more flexibility in response to internal perturbations. The data reported in this thesis highlight the importance of arm swing in balance maintenance during walking under perturbation. It suggests that individuals try to compensate for restricted arm swing during walking with a phone by modulating step width. With respect to gait adaptations, the results of this research support the idea that individuals' response to the applied perturbations through dual-tasking while walking depends on the magnitude of the applied perturbation. The evidence from this study suggests that talking while walking is the most challenging secondary task during locomotion among the applied perturbations in this study, and additional sensory tasks are the least challenging one.

These findings have significant implications for the understanding of how able-bodied individuals respond to different types of perturbations arising from daily living activities. Using the obtained results of this study, one may develop a gait training protocol for patients to successfully address common perturbations during daily life. The analysis of MoS presented extends knowledge of step-to-step balance changes during walking at different gait events associated with common fall patterns occurring at HC and MFC. Another important implication of the results of this study is that stability analysis should be performed on the lower body movements as well as the upper body movements to provide us with deeper understanding of the effect of applied perturbations on human gait. While the upper body plays a more important role in whole body balance maintenance, stability analysis of the lower body can reveal some subtle information about the influence of the secondary task on gait balance. This study is the first to comprehensively examine dynamic stability of the head segment during walking under perturbation. Knowledge of head stability during walking may ultimately enhance our understanding of upper body stability and the role of the head in whole body balance maintenance. With respect to stability of the upper body, the obtained results of this study indicate that local stability analysis of the upper body using head and C7 motion trajectories results in similar outcome in response to external perturbations during walking, while it might not give rise to similar results in response to internal perturbations. Finally, these data suggest that MFC height analysis should be performed to gain better understanding on the effect of additional concurrent task while walking on the risk of tripping.

A natural progression of this work would be to analyse the influence of similar attention-demanding secondary tasks ('distractions') in more vulnerable populations, including the elderly, fallers, and individuals with sensory or motor impairments that affect locomotor control. In addition, experiments using a broader range of perturbations could further reveal the balance maintenance strategies adopted in response to attention demanding secondary tasks. More broadly, research is also needed to determine gait stability and variability changes during long term overground walking under internal and external perturbations, using newly developed motion capturing systems e.g. inertial sensors to address limitations associated with treadmill walking. Further studies out to extend the application of the new MoS application proposed here, and potentially use this measure, together with machine learning, as a tool to assess the extent to which the variations of step parameters, gait speed and MFC height could affect forward and backward balance during walking under internal and external perturbations in healthy individuals, as well as patients suffering from movement disorders. Future research ought to examine body stability under different walking conditions to confirm the extent to which head stability is related to upper body stability.

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