Gait Adaptability and Biofeedback in Older Adults with Diabetes

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Doctor of Philosophy

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Keywords

Gait, walking, gait adaptability, older, elderly, ageing, diabetes, biofeedback, obstacle.

Abstract

In response to changes in the environment, the central nervous system frequently updates motor commands using sensory feedback related to the requirements of movements and adapts the internal model of gait. Therefore, gait adaptability is ability to modulate gait parameters in order to adjust trajectories of feet in the anatomical planes to avoid obstacles or step on targets. Although older adults with diabetes mellitus (diabetes) report falls more frequently than healthy older adult, a few studies have been conducted to investigate the effects of diabetes on gait adaptability in which participants with diabetes could plan their performance a few steps ahead. However, falls occur when participants have not enough time to plan and respond.

This thesis had three aims: (i) to investigate the effects of diabetes on gait adaptability, (ii) to investigate the effects of biofeedback (visualised real-time performance) on foot displacement adjustments in the sagittal plane, and (iii) to investigate the agreement between foot displacement adjustments quantified by the treadmill and overground adaptability tests for future application of biofeedback tools for assessing foot displacement adjustment.

To address all aims, 16 young adults (Group I), 16 healthy older adults (Group II) and 16 older adults with diabetes (Group III) were recruited. Exclusion criteria were musculoskeletal injury, uncorrected vision, a fall within a year before participation, cognition issues, and diabetic- or ageing-related neuropathy. To address the first aim, participants walked in baseline and then completed overground gait adaptability tests (40 trials) with four random conditions: step shortening, step lengthening, obstacle avoiding, and walking through. Step length targets were 40% of the baseline step length longer or

shorter than the mean baseline step length. The obstacle presented a 5 cm height across the walkway. A Vicon three-dimensional motion capture system was used to quantify spatiotemporal parameters of a gait cycle. To address the second aim, participants walked on a motorised treadmill for a few minutes. Preferred speed for each participant was determined. Customised MATLAB programs used marker trajectory data collected by a three-dimensional motion capture system streamed by a Visual3D Server. They then computed mean step length and minimum toe clearance of each participant from three 60-second trials collected during a 10-minute walking at preferred speed. Four subject-specific targets included two step length targets (baseline step length ± 10% baseline step length) and two minimum toe clearance targets (2.5 cm and 3.5 cm higher than the mean minimum toe clearance). Targets values for each participant were entered in the biofeedback system before the participant completed adaptability tests. Participant could see continuously a graphical display of step length or vertical trajectory of the toe mark in real time on a monitor installed in front of the treadmill. Targets were randomly presented as horizontal lines discretely appeared every 10 steps and disappeared after 10 steps three times. Participants adapted their step lengths and minimum toe clearance heights with presented targets on the monitor using biofeedback (real-time distances between their step length/ minimum toe clearance heights and presented targets) without being aware of the order of targets. Overground gait adaptability parameters (step velocity, stance time, swing time, double support time, and step length) and errors of foot displacement adjustments (differences between real and desired step lengths/ minimum toe clearance heights) in the overground and treadmill gait adaptability tests were quantified. For statistical analyses of data related to the first two aims, analysis of variance (ANOVA) was used to test the main effects of group and condition at a significance level

of 0.05. To address the third aim, Bland and Altman plots, scatter plots of difference and mean errors quantified by two treadmill and overground gait adaptability tests and the limits of agreement, were used in response to each condition to investigate the agreement between the tests.

Sixteen young adults (Group I), 14 healthy older adults (Group II) and 13 older adults with diabetes (Group III) completed both overground and treadmill tests. Groups were not significantly different in gait spatiotemporal parameters (step length, stance time, swing time, double support time, step velocity) when they walked normally at their preferred speed. However, they were different in gait spatiotemporal parameters when they tried to meet goal-tasks in adaptability tests. In Group III, stance and double support times significantly increased when they adapted the trajectory of their feet to step length targets and the obstacle height. Increased stance and double support times did not increase the accuracy of foot displacement adjustments, as older adults with diabetes in Group III showed the greatest errors of step length and minimum toe clearance adjustments. The participants in Group III could use visual feedback to significantly reduce errors of their step length and minimum toe clearance adjustments during an online correction. However, they had the greatest errors in their responses with and without biofeedback compared with other groups. Errors of step length and minimum toe clearance adjustments without biofeedback in the treadmill adaptability tests were in agreement with those in the overground gait adaptability tests. This may suggest efficacy of the application of the treadmill adaptability tools to assess precise foot displacement adaptation in a feedforward model for fall prevention in older adults with diabetes.

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Nomenclature

MTC	Minimum toe clearance
SL	Step length
OGA	Overground gait adaptability
TGA	Treadmill gait adaptability
DS	Double support
AP	Anterior-posterior
DPN	Diabetic peripheral neuropathy
LSL	Long step length
SSL	Short step length

Statement of authorship

"I, Suzanne Martin (Mahboobeh Mehdikhani), declare that the PhD thesis entitled Gait Adaptability and Biofeedback in Older Adults with Diabetes is no more than 80,000 words in length including quotes and exclusive of tables, figures, appendices, bibliography, references, and footnotes. This thesis contains no material that has been submitted previously, in whole or in part, for the award of any other academic degree or diploma. Except where otherwise indicated, this thesis is my own work."

"I have conducted my research in alignment with the Australian Code for the Responsible Conduct of Research and Victoria University's Higher Degree by Research Policy and Procedures."

Suzanne Martin 19/03/2021

Declaration regarding Ethics Approval

"All research procedures reported in the thesis were approved by the Victoria University Human Research Ethics Committee, HRE17-194."

Suzanne Martin 19/03/2021

Preface

A number of published works have resulted from this thesis. Because the research student changed her name in 2020, almost all publications show her previous name Mahboobeh Mehdikhani.

Her publications are listed below for reference and added as Appendix A.

Peer-reviewed journal article

Mehdikhani, M., Taylor, S., Shideler, B.L., Ogrin, R., Begg, R. (2020). Age effects on step adaptation during treadmill walking with continuous step length biofeedback, *Gait & Posture*, 80: 174-177.

Conference proceedings

Mehdikhani, M., Taylor, S., Shideler, B.L., Ogrin, R., Begg, R. (2019). A flexible real-time biofeedback tool that trains gait adaptability, *The XXVII Conference of the International Society of Biomechanics and the American Society of Biomechanics*, July 31 – August 4, Calgary, Canada.

HDR Student Conferences in iHeS

Mehdikhani, M., Taylor, S., Ogrin, R., Begg, R. (2020). Gait adaptability in Older Adults with Diabetes Mellitus during overground walking, *The HDR Student Conference, The Institute for Health and Sport*, Melbourne, Australia.

Mehdikhani, M., Taylor, S., Shideler, B.L., Ogrin, R., Begg, R. (2019). Aging and diabetes' effects on gait adaptability during treadmill walking, *The HDR Student Conference, The Institute for Health and Sport*, Melbourne, Australia.

Online publication

Martin, S. (2020). Threats and opportunities for a research student during COVID-19 restrictions, *isbWb.org, Dec 2020, Biomechanics in a COVID world-Student reflections and stories from 2020*, International Society of Biomechanics.

A number of scholarships were awarded to provide financial assistance during my study. They are acknowledged below:

- Doctoral Industry Placement Scholarship, 2020.
- COVID-19 Research Small Grant, 2020.
- COVID-19 Student Support Fund, 2020.
- Travel Grant, 2019.
- Australian Postgraduate Award, 2017-2019.
- International Postgraduate Research Scholarship (IPRS), 2016-2017.

Dedication

I dedicate this thesis to my family who consistently encouraged me to keep studying.

I also dedicate this thesis to my fellow researchers and friends. I have made great friends who helped make my life at Victoria University rewarding. I would specially like to thank my friends and fellow researchers Sharon, Robert, Rowan, Tahmineh, Shabnam, Sima, Nafiseh, Jack, Jora, Matthew, Suzanne, Cameron, Lisa and Anushka who encouraged me to keep going and perform to the best of my ability. I sincerely hope that our friendships continue.

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I would like to thank Blynn Shideler for coding MATLAB tools of the biofeedback system in my project. I also acknowledge the invaluable advice provided by Dr. Rajna Ogrin and Dr. Tony Sparrow. I am honoured to have worked in such a great team.

I must thank Rhett Stephens and John Izzard for helping me to design a safe obstacle in the overground test, and Dr. Petre Santry and Dr. Sharon Mcdowall for copyediting and Robert Steadman for proof readying.

Finally, I would like to thank the young volunteers at Victoria University and older participants from different clubs and groups including Yarraville Life Activities Club, PROBUS, and the Essendon Diabetes Support Group.

Chapter 1: General introduction

1.1. Motivation

Falls are common in older people and can cause mild to severe injuries. A fall is defined when a person comes to rest on the floor (Lamb et al., 2005). One in three older adults above age 65 years falls at least once a year (Ambrose et al., 2013). In the United States, falls are the seventh leading cause of death among older people (Rubenstein, 2006). Thirty percent of people over 65 years and 42% of people over 75 years fall every year (Tilling et al., 2006). The Australian Bureau of Statistics reported that 125,000 Australians aged 65 and over were hospitalised in 2016-2017 because of serious injuries to head (26%) and hip and thigh (22%) (AIHW, 2019).

Diabetes is one of the most common chronic conditions in the world (Corriere et al., 2013). Diabetes is a metabolic condition in which defective insulin secretion, insulin resistance or both cause hyperglycaemia (Gavin et al., 2000). Impairment of pancreatic ß-cells which produce insulin causes type 1 diabetes whereas insulin deficiency causes type 2 diabetes. In 2015, it was estimated that 415 million people aged 20-79 years lived with diabetes in the world. This is expected to rise to 642 million people by 2040 (Ogurtsova et al., 2017). In the United States, among a sample of 58186 adults responded to a survey in 2016-2017, 6317 people (10.9%) reported that they had been diagnosed with DM (Xu et al., 2018). Over 25% of people with DM are older adults (≥ 65 years) (Samos and Roos, 1998). The Australian Bureau of Statistics reported that 1.2 million Australians (4.8% of the population) lived with diabetes in 2017-18; one in seven of these was 65 years and older (AIHW, 2020).

Older people with diabetes fall more frequently (Yang et al., 2016). The incidence of falls has been reported to be 39% in older people with diabetes (Tilling et al., 2006) who were using either insulin or oral medications (Schwartz et al., 2008). Over 50% of older people with diabetes report at least one injurious fall or two non-injurious falls a year (De Mettelinge et al., 2013). In a follow-up study 87% of older participants with diabetes fell (Rivera et al., 1997). Around 30-50% of these falls caused minor injuries, and 5-10% caused major injuries including fractured neck of femur (Goldacre et al., 2002). In a study about the consequences of a hip fracture, one quarter of older people died in a year, 50% had reduced daily living activities, and 22% moved into a nursing home (Ambrose et al., 2013). Given that older adults with diabetes have many complications, fall-related injuries are more serious (Crews et al., 2013).

Apart from fall-related injuries, falls greatly increase the cost of national health care. The reported direct annual medical cost was US$23.3 billion in the United States and US$1.6 billion in the United Kingdom (Heinrich et al., 2010). Fall-related medical events account for 40% of nursing home placements, and contribute to further increases in healthcare costs (Masud and Morris, 2001). In Australia and Finland, the cost of hospitalisation for a fall-related injury is between US$6,646 – 17,483 (Fu, 2006). In 2000, the Australian Institute of Health and Welfare has predicted that the cost of fall-related hospitalisation will increase, requiring an estimated 3,320 residential care places in 2051 (Bradley and Harrison, 2007). Therefore, investigating the factors that cause falls in the elderly can reduce both costs and fall-related injuries.

Falls mostly occur because of incorrect shift of body mass following a perturbation (e.g. obstacle), elaborating the lack of enough ability to adapt gait and accurately respond to

changes in environment. Incorrect shift of body weight and tripping over obstacles are responsible for 59% and 41% of fall incidents in older adults, respectively (Robinovitch et al., 2013). Apart from control of step initiation (Rietdyk and Drifmeyer, 2009), quick control of foot trajectory is required to navigate in dynamic environments. To avoid an obstacle during walking, individuals must adjust step length (SL) in the anterior-posterior (AP) direction prior to the obstacle and minimum toe clearance (MTC) height in the vertical direction while stepping over the obstacle, and finally adjust the landing of the leading foot in the AP direction after avoiding the obstacle (Figure 1.1). Any maladjustment of sagittal foot displacement, before, while, or after negotiating an obstacle (i.e. misplaced steps and lack of foot clearance in the vertical direction) may lead to tripping, and a fall if recovery attempts are insufficient.

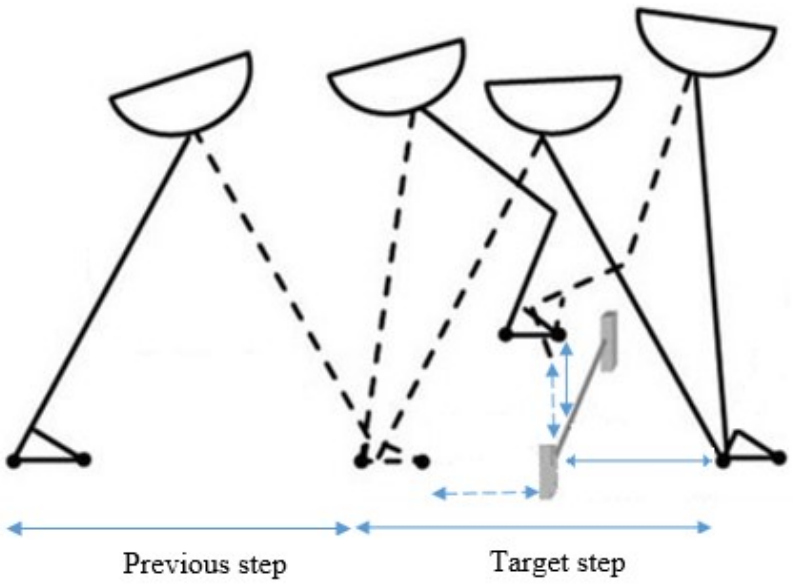


Figure 1.1. Sagittal foot displacement in response to a low-height obstacle. The leading (—) and trailing (--) feet are adjusted to avoid contacting the obstacle.

Step adjustments and obstacle avoidance studies have shown that older adults had impaired gait adaptability that is defined as inability to modulate gait parameters in response to goal-oriented tasks (Mazaheri et al., 2014, Lowrey et al., 2007, Caetano et al., 2016, Tseng et al., 2009). Impaired gait adaptability caused that the foot of older participants contacted obstacle more frequently when they negotiated obstacles.

Some other investigators also demonstrated that older adults had increased failure rates when negotiating obstacles, compared with young adults (Chen et al., 1994b, Brown et al., 2006, Chen et al., 1994a, Chen et al., 1996). However, a few studies found that many older adults were able to avoid obstacles, and averted any contact (Brown et al., 2005, Hahn and Chou, 2004, Mcfadyen and Prince, 2002).

Older adults touch obstacles more frequently than young adults while responding to obstacles under time-constrain conditions (Galna et al., 2009). Older people need more time to respond to an unpredictable task such as a sudden decrease or increase in a step length without compromising walking balance (Tseng et al., 2009, Mazaheri et al., 2014, Caetano et al., 2016); however, young adults are better able to modulate the ongoing movement during stepping. Older adults adapt the lengths of their steps a few steps ahead to be able to step on targets or avoid obstacles (Caetano et al., 2016). Although older adults have impaired gait adaptability, they may be able to respond to changes in environment if obstacles or hazards are visible a few steps ahead (i.e., no time constrain conditions) (Lowrey et al., 2007, Di Fabio et al., 2003, Hahn and Chou, 2004). Because falls occur while modulating ongoing movement suddenly, the assessment of SL adaptations can help to identify older adults who may need to enter into a training program.

Step length adaptations by either step lengthening or step shortening are effective strategies to prevent falls and a critical determinant of safe navigation especially for the elderly. Compared with young adults, older adults demonstrated poor stepping accuracy; they had greater errors (differences between the centre of the landing foot and the centre of the stepping targets) when they were required to step on the centre of stepping targets (Caetano et al., 2016, Mazaheri et al., 2015, Mazaheri et al., 2014). They could not adjust their step lengths and matched them with desired step lengths. Inability to adjust step lengths prior to an obstacle can affect the foot-obstacle clearance.

Reduced abilities to adapt SL were found to be associated with falls in older adults and people with stroke and Parkinson's disease (Geerse et al., 2019), and with increased trip incidences (Chen et al., 1991). Again, step adaptation in the AP direction as well as MTC adaptation in the vertical direction under time constrained conditions are insufficient in older adults due to ageing that reduces lower limb muscle strength, proprioception, visual perception and cognition (Hahn and Chou, 2004, Van Dieen et al., 2005, Pieruccini-Faria et al., 2019).

Despite the prevalence of falls in older adults with diabetes mellitus, few studies have investigated the effects of diabetes on gait and sagittal foot trajectory adjustments in response to the sudden appearance of goal-oriented tasks in older adults. At the time of writing the proposal of the current thesis, the knowledge about the effects of diabetes on obstacle crossings was limited to a few studies (Liu et al., 2010, Hsu et al., 2016, Richardson et al., 2005). These studies reported that diabetes reduced toe-obstacle clearances in older adults; however, obstacles were not touched (Hsu et al., 2016, Liu et al., 2010). This implies that the task of obstacle avoidance was not completed under time

pressure because the obstacle was always visible even before participants started walking towards it. Static obstacles obstructed the walkway, so participants could adapt the lengths of steps a few steps ahead to place obstacles in the middle of their crossing steps and easily avoided them. Moreover, in these studies, participants were aged under 65 years so they did not investigate the effects of diabetes on gait adaptability in older adults.

The present study investigated gait adaptability, errors of foot displacement adjustments and the effects of biofeedback on errors of foot displacement adjustment in three groups of participants: young, healthy older and older diabetes.

Firstly, given that diabetes per se (without diabetic complications such as neuropathic foot) does not affect gait adaptability, we hypothesized that ageing is responsible for impaired gait adaptability in older adults with diabetes. Secondly, we hypothesized that older people with diabetes are able to benefit from treadmill gait adaptability training with biofeedback. This thesis tested these hypotheses using two protocols that quantify gait adaptability and the accuracy of foot displacement adjustments (the distance between the toe marker on the first toe and targets in the sagittal plane) under time-constrained conditions and train a more adaptable foot displacement using biofeedback.

1.2. Research Questions

In light of limited information about the effects of diabetes on gait adaptability and the feasibility of using a targeted biofeedback for training gait adaptability, the three following research questions were addressed:

Research Question 1: What are the effects of ageing and diabetes on gait adaptability?

Research Question 2: What are the effects of biofeedback on sagittal foot displacement adaptation in older adults with diabetes compared with the young and healthy older groups?

Research Question 3: Are the treadmill and overground gait adaptability tests in agreement for quantifying precise foot displacement adaptation in the sample size?

1.3. Contribution of the thesis

For the first time, the effects of diabetes on gait adaptability in older adults were investigated. In order to investigate the effects of ageing and diabetes on gait adaptability three groups of participants (young, healthy older, and older diabetes) were compared with each other. Participants with neuropathy, history of falls, impaired cognition, and uncorrected vision were excluded from the study. Previous research reported that these factors increase the risk of falls up to three times (Deandrea et al., 2010). Therefore, the consideration of the exclusion criteria reduced the effects of confounder.

This thesis investigated the natural occurrence of trips over low-height obstacles under time constrained conditions without raising any ethical issues. The design of the real obstacle was novel; it had a height of 5 cm, extended across the walkway, so participants had to cross the obstacle and could not use any other strategy such as stepping outside the obstacle for obstacle avoidance. Excluding older participants with a history of falls, the addition of a real obstacle, the method of presentation of stepping targets and limitation of compensatory strategies were key differences to previous work (Caetano et al., 2016) on investigating step adjustment.

A novel biofeedback system was developed to address research questions. During the candidature, the system was developed, tested and finalised and then it was used to collect

data to complete the current PhD thesis. For the first time, the custom-made MATLAB program, which was coded by a visiting student (Shideler et al., 2017), was used to investigate whether targeted biofeedback would assist older people with diabetes to adapt foot displacements in the sagittal plane and reduce errors. The system presented discrete targets randomly on a monitor. Participants reacted to the sudden presentation of targets and tried to adapt their sagittal foot displacement with the presented target during online self-correction. For the first time, participants could use meaningful visual feedback to adapt step lengths or MTC heights with targets on a monitor without the involvement of an instructor. The study investigated the reduction in absolute errors, differences between real and presented (targeted) step lengths and MTC heights during online-correction in all three groups of participants. The results will inform future research of gait adaptability training in people with diabetes with biofeedback.

1.4. Outline of the thesis

Chapter 2: Background reviews the background knowledge related to gait, gait adaptability, and the accuracy of foot displacement adaptation in response to goal-oriented tasks. This chapter also presents the effects of ageing on gait adaptability reported in previous research. It then illustrates the systems that have been used for assessing and training foot displacement adaptation. Following this, the aims and hypotheses to answer the research questions are presented.

Chapter 3: Methods describes the research participants, apparatus, experimental setup, protocol, variables, and statistical tests. Participants included the following three groups of volunteers: (i) young adults, (ii) healthy older adults, and (iii) older adults with diabetes. Two overground and treadmill gait adaptability systems were used to present

conditions and collect kinetic and kinematic gait data during walking under time constrained conditions in which targets/obstacles were not always seen and were invisible when participants started walking. The overground gait adaptability (OGA) assessment system included virtual stepping targets and a real obstacle with 5 cm height. The treadmill gait adaptability (TGA) system presents continual real time SL and MTC and discrete targets (SL and MTC targets) on a monitor. During treadmill walking, tasks were: adapting real time step lengths with virtual SL targets (short and long) and adapting real time MTC heights with virtual MTC targets (high and higher). Targets were subject-specific which were determined based on the mean SL and the mean MTC of each participant. Errors in response to the sudden appearance of targets and in on-correction with biofeedback were compared within and between groups. The end section of this chapter used a method to investigate the agreement between the results of OGA and TGA tests when targets/obstacles were suddenly presented during walking.

Chapter 4: Results presents the collected data and statistical analyses. Descriptive statistics of the three groups of participants are presented with inferential statistics being used to determine any significant differences between and within the groups. Within groups, the effects of visual feedback on errors for sagittal foot displacement adaptation were investigated. The scatter plots of differences and means of values computed in the OGA and TGA tests present the agreement of two tests in the sample size.

Chapter 5: Discussion presents a discussion of the major findings in Chapter 4. These results are then compared with previous research where it is applicable. The primary limitations of the thesis and some suggestions for future investigations are presented in the last sections of this chapter.

Chapter 2: Background

2.1. Gait

Gait is a single repetitive sequence of walking to support and propel the body (Kharb et al., 2011). The gait cycle begins when the heel of one limb contacts the ground and terminates when the same heel contacts the ground for the second time (Figure 2.1).

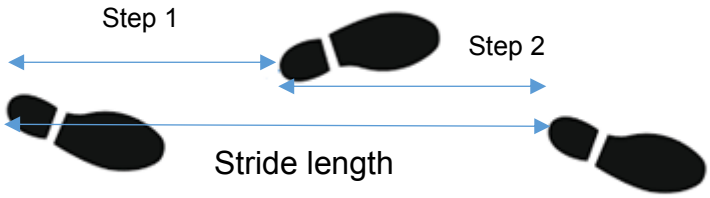


Figure 2.1. A gait cycle includes one stride (two steps).

Each gait cycle includes two steps, which in this study are called the previous and target steps. Stance is the period of a gait cycle when the foot is in contact with the ground, from heel contact to toe-off.

Swing is a period of time in a gait cycle when the foot is in the air, from the toe-off of one foot to the heel contact of the same foot. In the mid-swing phase of the gait, there is a moment when the vertical distance between the lowest point of the foot (toe) and the ground (Figure 2.2) is minimum (10-20 mm) and the speed of foot is three times of walking speed (Mills and Barrett, 2001).

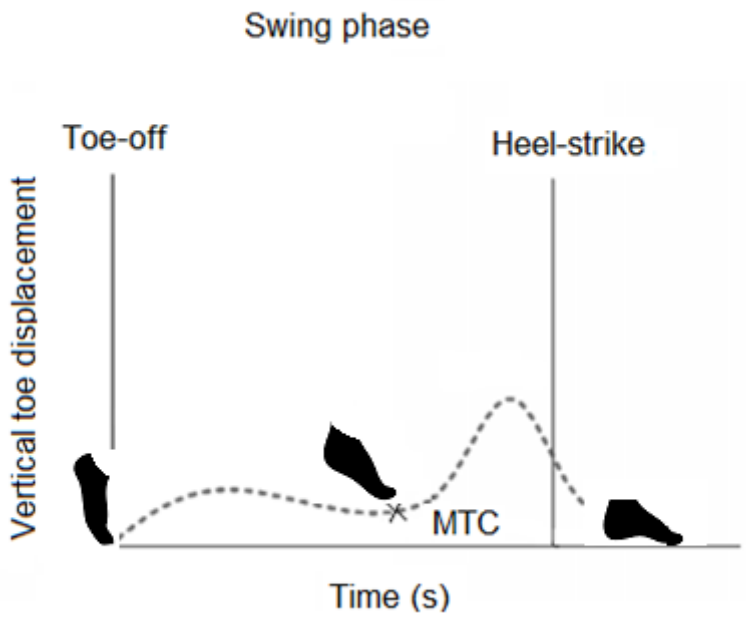


Figure 2.2. Trajectory of the toe marker in the vertical direction. The cross shows the minimum toe clearance (MTC) where the vertical distance between the big toe and the ground is a minimum.

There is a transition between the swing and stance phases called double support (DS) phase, when the heel of one foot and the toe of the other foot are in contact with the ground at the same time.

2.2. Effects of ageing and diabetes on gait

Ageing affects gait spatiotemporal parameters. Ageing reduces the speed of walking (Salzman, 2010) and increases the width of walking and the DS time (Winter et al., 1990).

Diabetes has been shown to impact gait parameters. When people with diabetes walked on level surfaces at self-selected speeds, their walking speed and SL were lower, but stance time was greater, compared with age-matched controls (Petrofsky et al., 2005).

The group with diabetes walked with the speed of 0.74 m/s whereas the control group walked with the speed of 1.19 m/s.

2.3. Gait adaptability

Gait adaptability relates to modulations of gait parameters in order to mobilise limb through foot displacement in the AP direction and the toe trajectory in the vertical direction. This is best examined through the context of walking tasks (cooperation between support and swing limbs) rather than static balance tasks, which only focus on the support limb. Gait adaptability refers to altering gait according to changes in the environment to adjust the three dimensional displacement of the swing foot in response to goal-oriented tasks such as stepping on projected targets on the ground or on the belt of a treadmill (Caetano et al., 2016, Mazaheri et al., 2015, Mirelman et al., 2016). In this thesis, gait adaptability refers to accurately adjusting the foot displacement in the AP and vertical directions in order to match the actual SL or MTC height with presented targets, or cross a presented low-height obstacle.

2.4. Impaired gait adaptability is associated with falls

The majority of falls in older people occur during obstacle avoidance because of inability to adjust foot displacement (impaired gait adaptability) (Berg et al., 1997). Reaction time increases with ageing and older adults are slower to adapt to external perturbation such an obstacles while walking (Fernández-Ruiz et al., 2000, Seidler, 2007). Obstacle avoidance while walking is a part of everyday life and executing a fast voluntary step while modifying gait to adjust foot displacement are essential to navigate in dynamic environments.

To avoid obstacles successfully, the sensory-motor system modulates the ongoing normal gait according to changes in the environment. Otherwise, a person is likely to stumble over the obstacle and potentially fall. Insights into the strategies that older people use during obstacle avoidance could relate to the effects of ageing on the sensory-motor system.

Any impairment to adapting foot displacement can cause falls (Berg et al., 1997, Geerse et al., 2019). Gait is regularly adapted during walking on obstructed terrains to secure adequate foot displacement in relation to local environmental features. For example, while crossing a river by stepping on stones, accurate adjustments of foot displacement in the AP direction prevent us from falling into the river. However, while walking and crossing over a low-height obstacle, accurate adjustments of foot displacement in the vertical direction prevent us from tripping over the obstacle. Therefore, gait adaptability is essential to navigate, as any deficits in adapting foot displacement in either direction can cause a falling incident.

Gait adaptability is controlled by the sensory-motor control system that adjusts gait to environmental circumstances. All elements of the sensory-motor control system work together to produce and update motor commands (Figure 2.3). When a person walks on a smooth surface at a preferred speed without any challenge, the central nervous system adapts gait in a way that muscles consume the minimum amount of energy. Since using receiving afferents can take too much time in keeping track of the current state, the system uses efferent copies of locomotion at the level of the spinal cord as a baseline for information about the task (Bari and Robbins, 2013, Day and Brown, 2001).

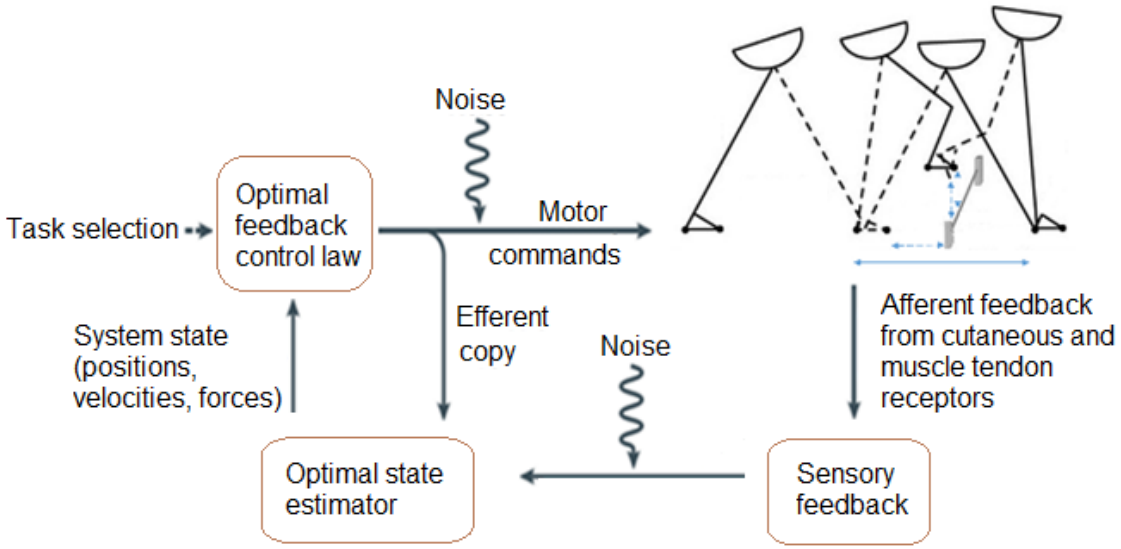


Figure 2.3. Optimisation of motor commands by the sensory-motor system. The sensory-motor control system produces and modulates locomotion using afferent feedback to adapt efferent copies of motor commands to external requirements for safe navigation.

Efferent copies are old memories that have been previously stored by synaptic modifications, which may need to be updated using afferent feedback from several receptors (Figure 2.4). These copies are mostly stored in the cerebellum which has a crucial role for internal model calibration. Indeed, the cerebellum has the internal models and makes predictions. It allows us to make accurate movements by comparing the predicted and the observed movement to adapt our subsequent movement more accurately (minimising error of predictions) (Shadmehr et al., 2010). Afferent information from muscles and skin reach the spinal cord; however, visual, auditory, somatosensory and proprioception afferents reach the telencephalon and brain stem.

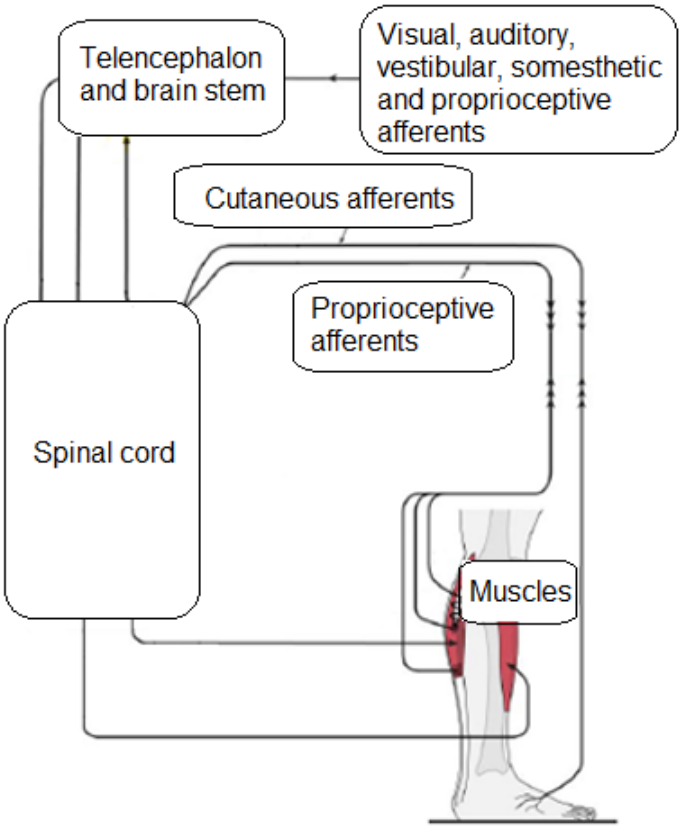


Figure 2.4. Sensory information is sent to the central nervous system from different locations. The central nervous system uses afferent feedback (sensory information) to update efferent copies of motor commands.

Sensory feedback control plays an important role in adjusting stride-to-stride limb trajectories to maintain balance, and in smoothing unintended irregularities during walking (Gandevia and Burke, 1992). Deficient peripheral sensory feedback control affects the stride-to-stride variability of gait more than it affects the mean locomotor pattern (Dingwell et al., 2000). As sensory feedback control of locomotion is dependent on the walking speed, the impact of a sensory loss or perturbation on gait decreases with a faster speed (Wuehr et al., 2014).

Functional imaging has confirmed that activity of the sensory cortex control of the brain reduces while walking faster, so fast locomotion is controlled more automatically without using sensory feedback (Schniepp et al., 2012). In this thesis, all tests were conducted while walking at preferred speed to exclude the effect of speed variability on visual information.

2.5. Effects of ageing and diabetes on gait adaptability

Timing of the detection an obstacle determines the result of obstacle avoiding and the strategies used to avoid such obstacles. The rate of successful obstacle avoidance is 100% when a young person has more than 450 milliseconds while walking at a self-selected speed (Chen et al., 1994b). This means that about one-step time is adequate to plan and avoid an obstacle in the following step. The time of an obstacle detection also determines strategies that young adults use to avoid an obstacle. In long step strategy (LSS), the crossing step length increases, whereas in the step-shortening strategy the crossing step length reduces (Figure 2.5). Young adults use a short step length (SSL) strategy when the time between the detection of an obstacle and responding to the obstacle reduces (Chen et al., 1994b, Weerdesteyn et al., 2005b). Therefore, the less time planning a performance the higher the chance of failure due to reduced foot displacement accuracy.

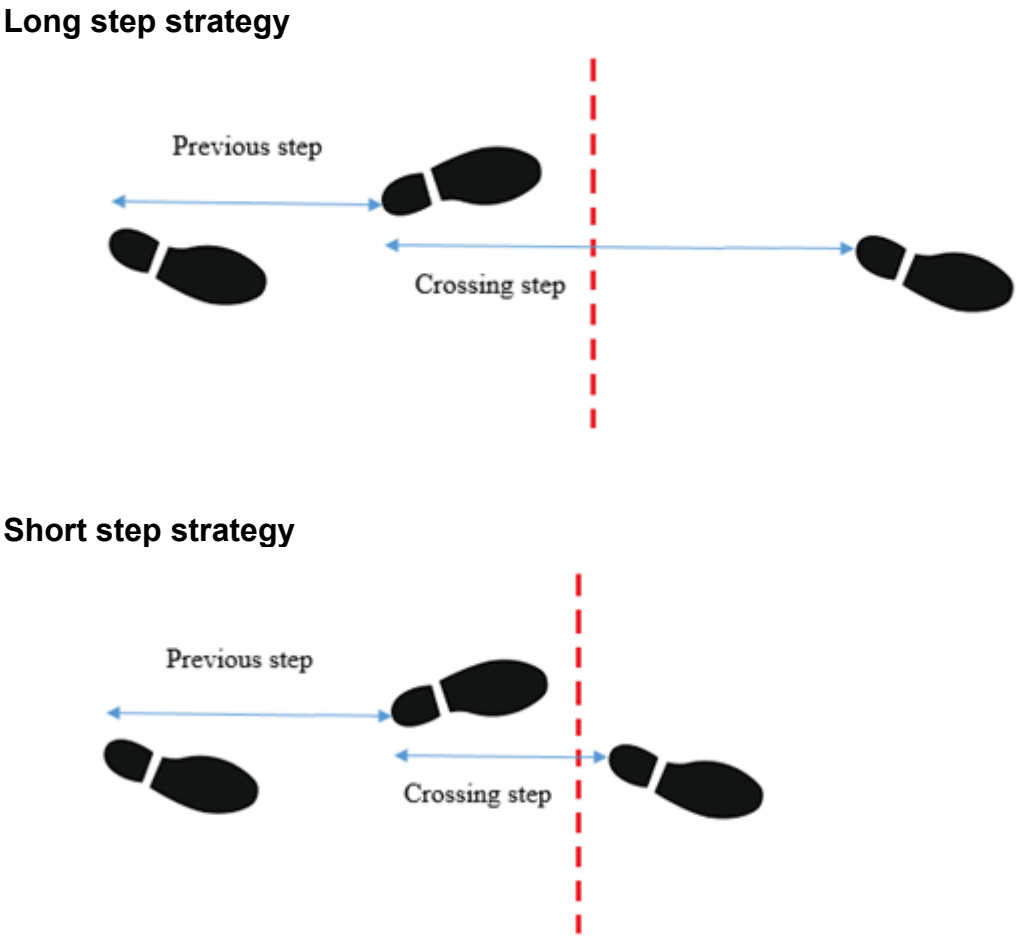


Figure 2.5. The overhead view of step strategies while crossing an obstacle.

Obstacle negotiation tasks involving older people have shown higher chance of contacting obstacles that suddenly appear, compared with fixed obstacles that are always visible. Table 2.1 presents a summary of previous research involving the number of obstacle contacts and the visibility of obstacle and limb clearance.

Table 2.1. Summary of obstacle crossing tests in previous research.

Author	Obstacle contact	Visibility of obstacle	Leading foot-obstacle vertical distance	Trailing foot-obstacle vertical distance
(Brown et al., 2005)	No obstacle contacts	Always	-	-
(Brown et al., 2006)	Older adults touched obstacles more frequently	Sudden	-	-
(Chen et al., 1991)	Four adults stepped on virtual obstacle	Always	No effects of age	-
(Chen et al., 1994a)	Four trips were reported	Sudden	-	-
(Chen et al., 1994b)	Older adults touched obstacles more frequently	Sudden	-	-
(Chen et al., 1996)	Older adults touched obstacles more frequently	Sudden	-	-
(Hahn and Chou, 2004)	No obstacle contacts	Always	-	-
(Hahn et al., 2005)	No obstacle contacts	Always	-	-
(Lowrey et al., 2007)	One older adult contacted	Always	No effects of age	No effects of age
(Mcfadyen and Prince, 2002)	No obstacle contacts	Always	Lower in older adults	No effects of age
(Caetano et al., 2016)	Twenty two percent of older adults made at least one mistake	Sudden	-	-

As early step adaption reduces the chance of touching the obstacle, older adults adapt strategies to cross the obstacle successfully when they have enough time to plan their performance. For instance, if older adults saw an obstacle at the start, they modulated their step lengths while approaching obstacles. They shortened their step lengths and took a final short step prior to crossing the obstacle. However, young adults adjusted only their obstacle crossing step to negotiate obstacles. Young adults did not reduce step lengths on approach like the older adults; they primarily modulated the time and length of one step, the crossing step. (Chen et al., 1991).

When an obstacle avoidance appears unpredictably, older adults reduce their SL (Caetano et al., 2016, Brown et al., 2006, Chen et al., 1994a, Chen et al., 1994b, Chen et al., 1996, Di Fabio et al., 2003) and add one short step before crossing the obstacle (Caetano et al., 2016). This implies that older adults increase the number of steps between the location where they see a target and the real location of the target (Figure 2.6).

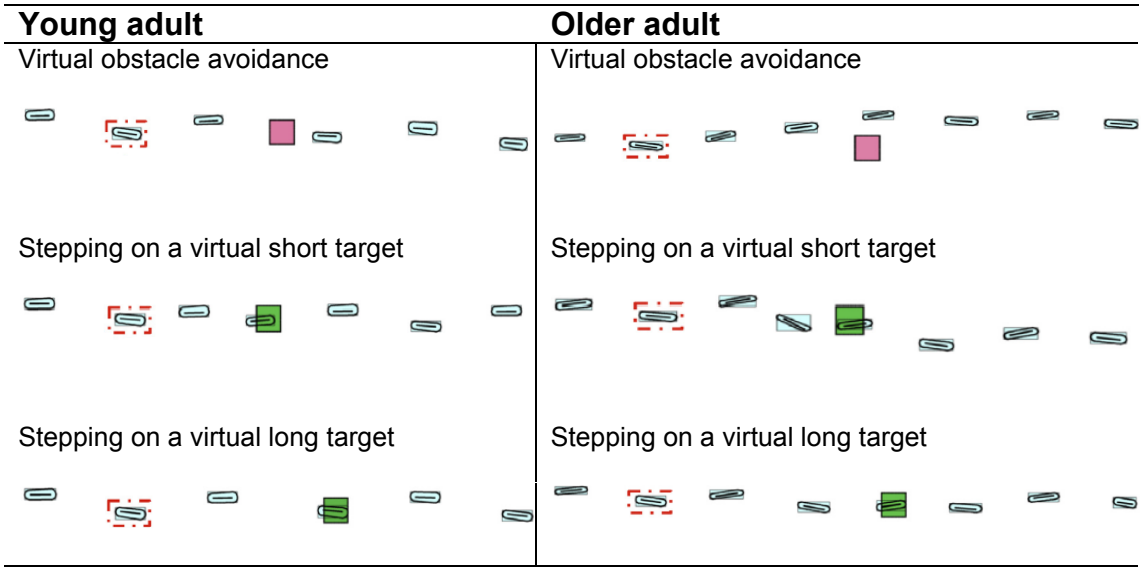


Figure 2.6. Foot displacement adjustments in typical older and young adults during avoiding an obstacle and stepping over targets. The red broken rectangle shows the time when a long, short stepping target or an obstacle was presented. Green squares are stepping targets and pink squares are obstacles, which were avoided. The figure was retrieved from Caetano et al (2016).

As shown in Figure 2.6, the older adult avoided the virtual obstacle (pink light projected on the ground) by placing their foot outside the light; instead, the young adult went over the obstacle and avoided it. Hence, the obstacle's characteristics (virtual or real, extended across or along the walkway, height) may impact strategies that are used by older adults.

Results of studies on reduced toe-obstacle distance in older adults during obstacle crossing appear to conflict. For example, in some studies vertical toe-obstacle distance while avoiding real obstacles was unaffected by age (Chen et al., 1991, Lowrey et al., 2007). In other studies, toe-obstacle distances on leading foot (Di Fabio et al., 2003), but not trailing foot (Mcfadyen and Prince, 2002, Draganich and Kuo, 2004), were lower in older adults compared with younger adults. Thus, the detection of a reduced vertical toe-obstacle clearance is dependent on whether obstacles were always visible.

Results of studies on the effects of ageing on the speed and length of crossing steps were also inconsistent. Some studies found that the speed and length of crossing steps in older adults were the same as those in young adults (Brown et al., 2006, Draganich and Kuo, 2004, Brown et al., 2005, Hahn and Chou, 2004, Hahn et al., 2005). However, other studies noted a slower obstacle crossing speed and shorter crossing step in older adults when compared with young adults (Di Fabio et al., 2003, Mcfadyen and Prince, 2002, Chen et al., 1991, Chapman and Hollands, 2007, Lowrey et al., 2007, Mckenzie and Brown, 2004, Caetano et al., 2017).

Ageing also increases the DS time (Caetano et al., 2017). In response to SSL and LSL targets, the DS times were 31.4% and 30% of gait cycle in older adults, compared with 28.9% and 26.3% of gait cycle in young adults (Caetano et al., 2017). Besides impairment of the sensory system, deficits in motor adaptation can originate in degeneration of the cerebellum in older adults (Seidler, 2007, Bernard and Seidler, 2014, Boisgontier and Nougier, 2013) to cause impaired gait adaptability.

Older adults were found to shorten their SL prior to crossing obstacles (Chen et al., 1994b). When the available response time was reduced, a virtual obstacle appeared earlier

than one-step time, which increased the chance of shortening the step prior to crossing on obstacle in older adults. Although the reduction of available time from 1000 milliseconds to 450 milliseconds did not affect the strategies which older and young adults used to cross an obstacle, 8-9 % of older adults were found to shorten their step prior to obstacles more often (Chen et al., 1994b). Therefore, the available response time plays an important role in the selection of strategy when responding to goal tasks.

Older adults have reduced success rates during avoidance of virtual obstacles under time constrained conditions compared with younger adults (Potocanac et al., 2015). In response to virtual targets during visually cued treadmill walking, older adults made more mistakes and stepped on a virtual obstacle more often than young adults did (Potocanac et al., 2015). Older adults had a 65-75% success rate in a study when the available response times were less than two steps (Potocanac et al., 2015), whereas they had a 95% success rate in response to virtual obstacles that were suddenly projected two steps ahead (Chen et al., 1994b). Individuals look on the landing target, on average two steps ahead, for visuomotor pre-planning during their stepping movements (Patla and Vickers, 2003). This would provide enough time to use feedforward information about how far the limb is from the location of the target for applying appropriate changes needed in subsequent steps to land on the target or to avoid it. Thus, a shorter available response time of less than two steps increased the chance of shortening the step prior to a virtual obstacle in older adults by 8-9 %; however, none of the young adults shortened their step prior to the obstacle (Chen et al., 1994b, Potocanac et al., 2015). Older adults may be more conservative when obstacles are real and the available time is inadequate.

Besides a decline in the rate of success in avoiding obstacles, ageing was found to reduce the accuracy of foot displacement adjustments (reduced step length adaptation) when stepping targets were shifted in the AP direction (Mazaheri et al., 2015, Caetano et al., 2016). The overall stepping errors were negative and significantly greater in older adults compared with young adults during continuous treadmill walking and responding to SSL and LSL targets. The absolute errors were larger in older than in young adults when a stepping target was shifted posteriorly to apply a SSL target. Other characteristics of obstacle avoidance in older adults reported in the literature included reduced heel-obstacle horizontal distance after crossing obstacles (Mcfadyen and Prince, 2002, Lowrey et al., 2007), reduced toe-obstacle horizontal distance between the toe of the trailing foot and the obstacle prior to an obstacle avoidance, and increased heel-obstacle horizontal distance between heels of leading feet and obstacles after crossing obstacles (Liu et al., 2010). It can be concluded that ageing reduces the accuracy of foot displacement in older adults.

Although gait adaptability of older people has been intensively investigated, the varying results may reveal several methodological issues. Despite the influence of time-constrained conditions, it is unlikely that all participants were free from age-related conditions that affect responses to tasks, so it is important to investigate gait adaptability in cohorts whose characteristics are homogenous. Without any other chronic condition such as diabetes mellitus, ageing may impair vision, cognition, and peripheral nerves. Reduced gait adaptability can place adults with age-related conditions at increased risk of falling when negotiating unexpected hazards. Compared with young adults, older people alter their gait parameters by taking more steps and spending more time in DS when approaching targets. This results in poorer stepping accuracy and failure to hit

stepping targets and avoid obstacles (Caetano et al., 2016). Furthermore, certain chronic diseases such as stroke in elderly people can further impair their ability to walk in dynamic environments (Tirosh et al., 2013, Begg et al., 2014a). Studies of gait adaptability in people with stroke have shown that nerve damage reduces the accuracy of their walking while responding to goal-oriented tasks (Roerdink et al., 2007, Van Swigchem et al., 2013). Results showed that stroke survivors demonstrated markedly decreased obstacle avoidance success rates. Therefore, it is postulated that the impairment of sensory afferents caused by diabetes can increase the risk of falling in elderly people with diabetes.

No previous studies have measured the accuracy of foot displacement adjustments in older adults with diabetes while walking overground and treadmill walking. However, there is evidence that diabetes increases the risk of falls by 30% in older adults (one in three persons with diabetes falls every six months) (Meyer et al., 2020). Our knowledge about the effects of diabetes mellitus on gait adaptability is limited to only two studies in which obstacles were visible always (Liu et al., 2010, Hsu et al., 2016). Some participants had mild diabetes-related neuropathy that could affect the reported results (Liu et al., 2010). Thus, they were deemed as having methodological issues related to the presentation of obstacles as earlier mentioned, and inclusion of participants with neuropathy.

Neuropathy develops in 65% of people with diabetes (Dyck et al., 1986) and is diagnosed on average about 8 years after the onset of diabetes (Savettieri et al., 1993). However, a later study (Cheing, 2010) found that neuropathy may be diagnosed up to 10 years in 25% and 20 years in 50% after onset of diabetes. Studies regarding the pathophysiology of

diabetes indicate the role of longstanding hyperglycaemia in the development of neuropathy (Said, 2007). Some investigators have confirmed the role of microvascular changes for distal nervous fibre loss in diabetic patients caused by microvascular changes in their distal nerves (Malik et al., 2005). The glycation of nerve tissues compromises the vascular supply by forming non-degradable advanced glycation products on the extracellular connective tissue matrix of nerves. This reduces the blood flow and nerve conduction velocity (Sugimoto et al., 1997), degenerates distal nerve fibres, and prevents nerve regeneration (Duran-Jimenez et al., 2009). Metabolic changes in the nerve tissues (e.g. accumulation of sorbitol and formation of free radicals) can also lead to the development of neuropathy in diabetic patients. In this case the accumulation of sorbitol inside cell membranes occurs during episodes of hyperglycaemia (Said, 2007), which causes the enzyme hexokinase to become saturated and allows excess glucose to enter the polyol pathway where it changes into sorbitol (Hotta et al., 2008). This leads to hyperactivity of the pathway and increased intracellular sorbitol concentrations, which then reduce nerve conduction velocity (Pop-Busui et al., 2006). All these changes in people with diabetes-related neuropathy might lead to impaired gait adaptability when avoiding obstacles. Reduced speed, reduced SL, and reduced toe-obstacle vertical distance indicate impaired gait adaptability (Hsu et al., 2016, Liu et al., 2010), which can lead to tripping and falling if compensatory and recovery strategies are insufficient.

It remains unknown whether people with long term diabetes but without neuropathy would show similar impaired gait adaptability. Previous studies investigating gait in response to obstacles did not find that the reduced toe-obstacle vertical distance increased the failure rate of obstacle avoidance (Hsu et al., 2016, Liu et al., 2010) because the obstacles were always visible. Thus, the participants could see the obstacles from the

starting point and plan their responses well ahead and navigate them accurately. Until now, only a few systems have been developed to present goal-tasks under time constrained conditions. These systems which were used for investigating reactive responses to unpredictable tasks and for training more accurate responses with biofeedback are presented in the following section.

2.6. Instruments to evaluate and train foot displacement

Inability to adapt step-to-step foot displacement according to sudden changes in the environment is associated with falls (Geerse et al., 2019). Therefore, evaluating and training precise foot displacement can reduce falls in pathological populations rather than older adults with diabetes (Caetano et al., 2016, Mirelman et al., 2016, Begg et al., 2014a, Potocanac et al., 2015, Houdijk et al., 2012)..

Previous research used various visual cues to perturb normal gait and assess gait adaptability. As Table 2.2 presents, there are two types of visual cued instruments: the first presents visual cues on the ground or belt of a treadmill (Caetano et al., 2016, Mazaheri et al., 2015, Geerse et al., 2019, Dingwell and Davis, 1996); the second presents visual cues in the form of non-immersive virtual targets on a monitor, which is installed in front of participants (Mirelman et al., 2016, Begg et al., 2014a, Tirosh et al., 2013).

Table 2.2. Visual-cued instruments for quantifying foot displacement adaptation during walking in response to goal-oriented tasks.

Type	Description	Application
Projected visual cues on the ground or the treadmill's belt	Participants walked without or with visual feedback. In instruments without visual feedback, continual visual cues were used to guide stepping on lights when one light was suddenly shifted (Geerse et al., 2019), changed its colour (Potocanac et al., 2015, Mazaheri et al., 2015), or was suddenly appeared while walking (Caetano et al., 2016). Instruments with visual feedback continually projected ground-cues to train step adaptation while walking (Dingwell and Davis, 1996).	Participants were used for evaluation (Geerse et al., 2019, Mazaheri et al., 2015), and evaluating/training step adaptation (Dingwell and Davis, 1996).
Displayed visual cues and performance on a monitor using virtual reality systems	Participants were walking on a treadmill in virtual reality. In real time, they adapted displacements of their virtual feet with changes in a virtual environment (Mirelman et al., 2016) or they adapted displacement of a peak of continual graphical performance such as the minimum toe clearance peak with presented targets (Begg et al., 2014a, Tirosh et al., 2013).	Participants were used for training step and foot-obstacle adaptation (Mirelman et al., 2016) or the minimum toe clearance height adaptation (Tirosh et al., 2013, Begg et al., 2014a).

Most of these instruments presented visual cues in the form of ground-projects targets (visually guided stepping) and required participants to respond to changes in location and colour of one visual cue at a time (Figure 2.7). Thus, these instruments did not give any specific feedback about performed tasks.

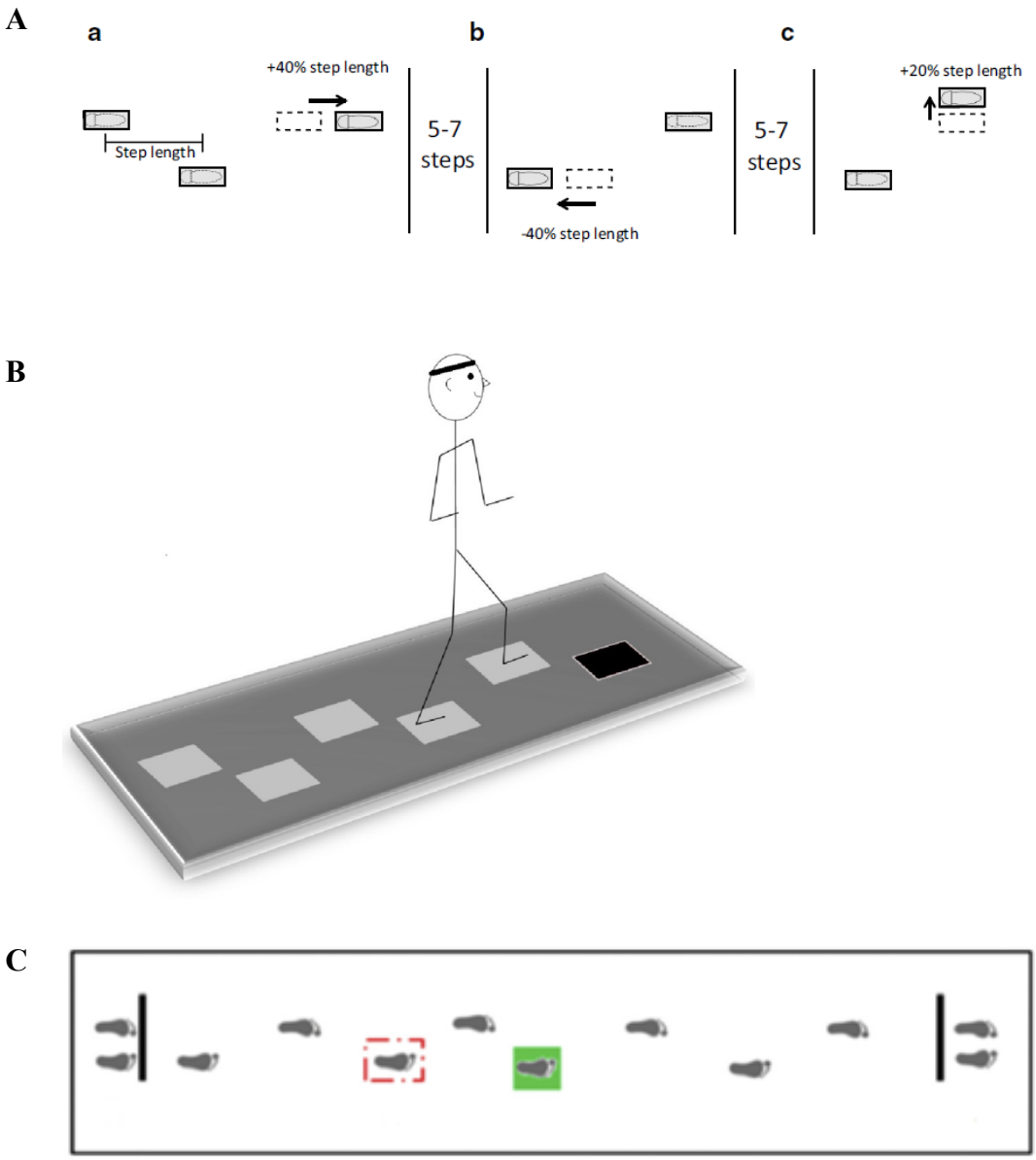


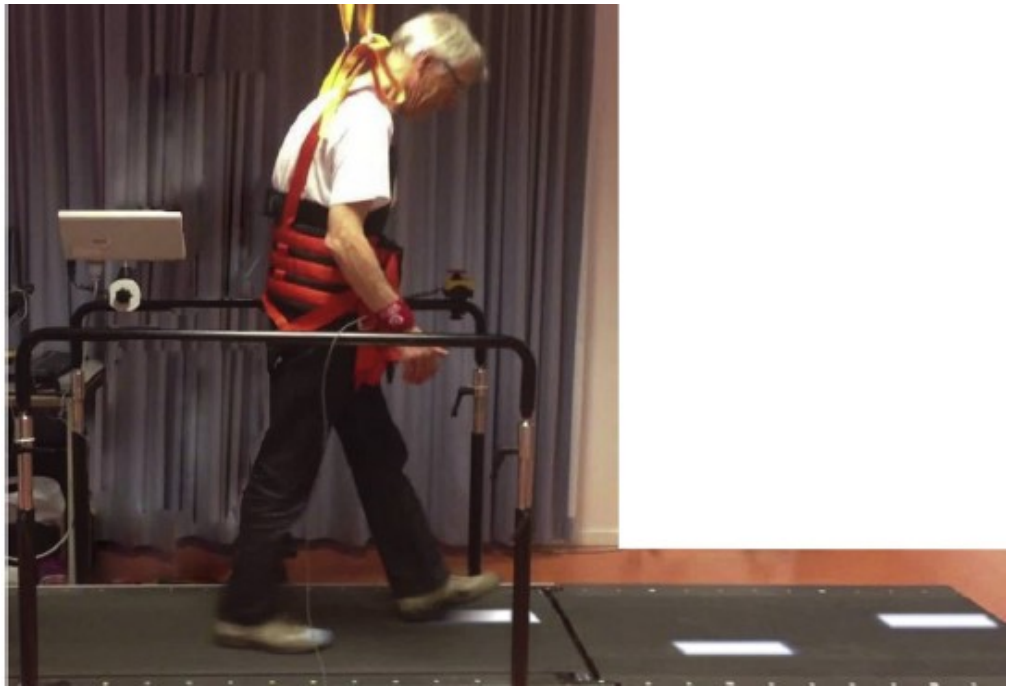
Figure 2.7. Different methods used to project a goal-task during overground walking for evaluating step adaptation. A visual cue is suddenly shifted in a direction (forward, backward, or sideways) every 5-7 steps (A) (Mazaheri et al., 2015), is projected with a different colour (B) (Potocanac et al., 2015), or is suddenly apparent two steps ahead (Caetano et al., 2016).

Training cognition and motor aspects, which are interdependent for safe walking, can reduce falls in different populations (Begg et al., 2014b, Mirelman et al., 2011). Some training programmes give feedback directly by comparison of foot displacement and position of targets on the ground (Dingwell and Davis, 1996), whereas training

programmes with a computer-simulated non-immersive virtual reality display visual feedback virtually on a monitor installed in front of a treadmill (Begg et al., 2014a, Mirelman et al., 2016). Giving visual feedback using non-immersive virtual reality programmes provides more natural head positions while walking compared with the instruments that project overground visual cues (Figure 2.8).

The present thesis used a new design of the latest overground visual cues system (Caetano et al., 2016) and a novel computer-simulated non-immersive virtual reality that evaluates and trains sagittal foot displacement adaptation during treadmill walking (Shideler et al., 2017). Two systems for quantifying foot displacements were compared for future application of the computer-simulated non-immersive virtual reality. Chapter 3 will illustrate these systems.

A



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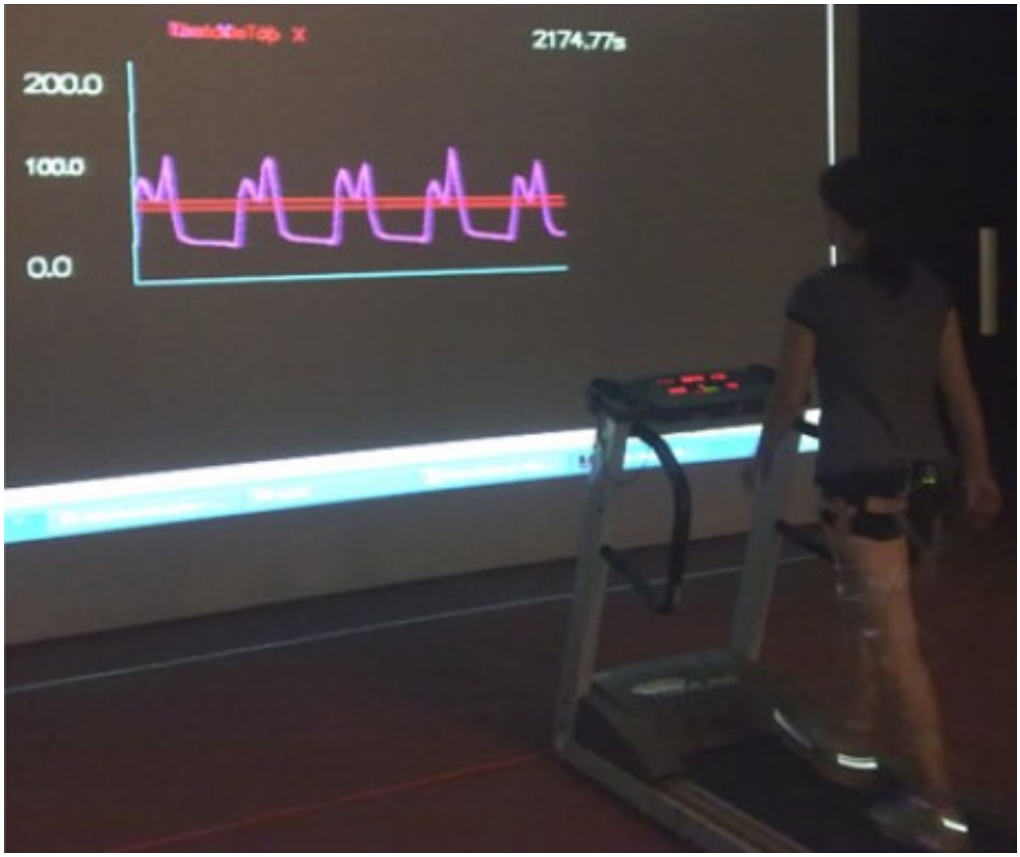


Figure 2.8. The ground-projected step target system and computer-stimulated non-immersive virtual reality. Mazaheri et al. (2015) used ground-projected step target system (A), but Begg et al. (2014) used a computer-stimulated non-immersive virtual reality (B)**.**

2.7. Aims and hypotheses

The following aims and hypotheses were addressed to answer Research Question 1: What are the effects of ageing and diabetes on gait adaptability?

Aim 1.1 To determine the effects of ageing and diabetes (Group I vs. Group II vs. Group III) on walking speed and gait spatiotemporal parameters (stance, swing, and DS and SL) in baseline.

> **Hypothesis 1.1.** Walking speed and gait spatiotemporal parameters (stance, swing, DS, and SL) would show no difference between groups in baseline.

Aim 1.2. To determine the effects of ageing and diabetes (Group I vs. Group II vs. Group III) and walking condition (baseline vs. walkthrough vs. step shortening/step lengthening/obstacle crossing) on gait adaptability spatiotemporal parameters (step velocity, stance, swing, DS, and SL) in OGA tests.

> **Hypothesis 1.2.1.** Gait adaptability spatiotemporal parameters (step velocity, stance, swing, DS, and SL) would show no difference between groups in OGA tests.

> **Hypothesis 1.2.2.** Gait adaptability spatiotemporal parameters (step velocity, stance, swing, DS, and SL) would show no difference between conditions (baseline vs. walkthrough vs. step shortening/step lengthening/obstacle crossing) within groups in OGA tests.

Aim 1.3. To determine the effects of ageing and diabetes (Group I vs. Group II vs. Group III) on foot displacement errors.

Hypothesis 1.3.1. Foot displacement error means would show no difference between groups.

Hypothesis 1.3.2. Foot displacement error means would show no difference within groups.

The following aim was addressed to answer Research Question 2: What are the effects of biofeedback on sagittal foot displacement adaptation?

Aim 2. To determine ageing and diabetes (Group I vs. Group II vs. Group III) and condition (without biofeedback vs. with biofeedback) on sagittal foot displacement adaptation.

Hypothesis 2.1. In each group, foot displacement error means would show no difference without or with biofeedback.

Hypothesis 2.2. Between groups, foot displacement error means would show no difference either without biofeedback or with biofeedback.

The following aim was addressed to answer Research Question 3: Are the treadmill and overground gait adaptability tests in agreement to quantify precise foot displacement adaptation?

Aim 3. To determine whether foot displacement error means quantified in TGA tests were in agreement with those quantified in OGA tests.

Hypothesis 3. At least 95% of points in scatter plots would fall between the limits of agreement. Points represent differences between errors measured in the TGA and OGA tests versus average errors measured in the TGA and OGA tests in response to each goal-task.

Chapter 3: Methods

All computational analyses described in this chapter were performed on experimentally recorded data in response to goal tasks during walking. The goals of this thesis were to address the research questions by acquiring high-quality data. Although markers on shoes and the pelvis were adequate to run the system, a full body marker model was used to provide additional data for further future analysis involving musculoskeletal modelling.. This chapter describes the experimental and computational pipeline (Figure 3.1). The collected experimental data in the OGA and TGA tests were processed to calculate parameters.

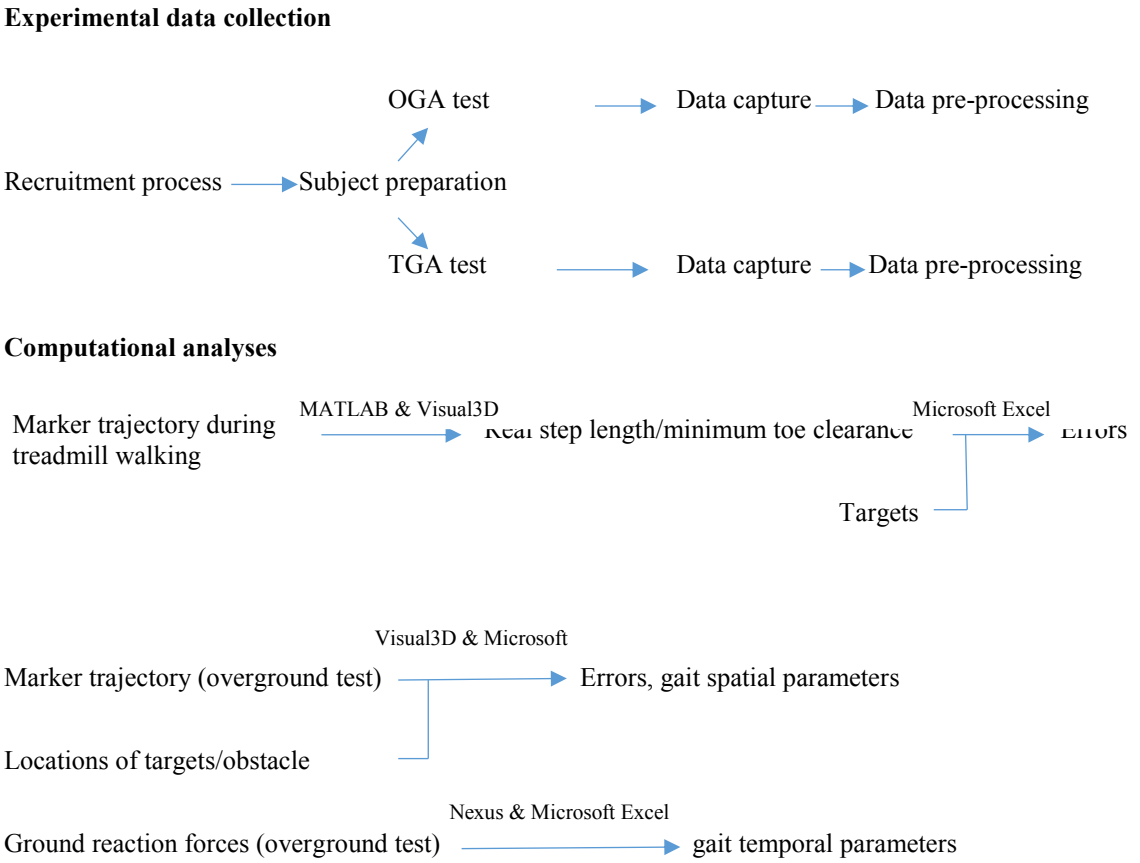


Figure 3.1. Experimental and computational pipeline used throughout the thesis. OGA: overground gait adaptability, TGA: Treadmill gait adaptability.

3.1. Experimental data collection

A priori power calculation was conducted based on a study that investigated obstacle avoidance and step adjustment (Liu et al. 2010) using G*Power (Faul et al. 2009). It was estimated that 16 participants in each group of young adults (Group I), healthy older adults (Group II) and older adults with diabetes (group III) would be required to detect an effect size of 0.66 for obstacle avoidance, with 95% power and a significant level of 0.05 using ANOVA. Therefore, Marker derived kinematics and ground reaction forces were collected from 48 participants.. They walked at their self-selected speed overground and

on a motorised treadmill, and completed baseline and adaptability tests. Data were collected at the Institute for Health and Sport, Victoria University, Melbourne, Australia. The Human Research Ethics Committee at Victoria University approved the ethics application form. All participants signed consent forms before completing the tests.

3.1.1. Recruitment process

Group I included young students who were recruited through advertisement (Appendix B) on University notice boards. Potential participants were filtered according to a specific inclusion criterion as follows:

- Aged between 18 and 40 years
- Free from any musculoskeletal injury at the time of testing
- Free from uncorrected vision issues
- Able to understand written and spoken English

Group II included healthy older adults who were recruited through advertisements on University notice boards and senior clubs within 10 km of Victoria University, Footscray. Healthy older participants were members of Yarraville Life Activities Club, and PROBUS, or saw the advertisement on the university campus. Inclusion criterions were:

- Aged between 65 and 85 years
- Free from any musculoskeletal injury at the time of testing
- Walking without any walking aids and assistive devices
- Free from cardiopulmonary conditions
- Free from neurological conditions such as stroke and Parkinson's disease
- Free from uncorrected vision issues

- No history of falling within a year before participating in the study
- Free from cognition issues
- Free from ageing-related neuropathy
- Able to understand written and spoken English

Group III included older adults with diabetes mellitus who did not have diabetes when they were young. None of them were recruited through the advertisement. They were members of diabetes support groups in the western suburbs of Melbourne and were informed about the project through their diabetes support groups.

In addition to the above mentioned inclusion criteria for healthy older adults, inclusion criteria for Group III were:

- Diagnosed with diabetes mellitus longer than five years before participating in the study
- Free from diabetes-related neuropathy
- Free from effects of diabetes on the vestibular system

Eligible participants then received a copy of the information for participants (Appendix C) through email.

3.1.2. Participant preparation

Anthropometric data were recorded for each participant (e.g. age, height, body mass) (Table 3.1). The dominant leg was determined by asking each participant to kick a ball (Kearns et al., 2001). The following tests were completed for each older adult:

(i) Visual acuity: The vision of each participant was examined using an eye chart.

(ii) Cognitive function: The mini-mental state examination (MMSE) (Cockrell and Folstein, 2002) was completed by older participants (Appendix D). MMSE included some questions to quantify cognitive function. Participants received scores for their responses to the questions.

(iii) Neuropathy: The Michigan neuropathy screening instrument (MNSI) which was used comprises two sections: questionnaire and clinical examination (Lunetta et al., 1998) (Appendix E). Participants received a score based on their responses to each question and clinical examination. The MNSI was used to examine neuropathy in older people with diabetes. Healthy older people also completed MNSI to exclude participants with ageing-related neuropathy.

Exclusion criteria were: vision acuity less than 20/40; cognition scores less than 27; or neuropathy scores of 3 and over. Eligible participants completed informed consent forms (Appendix F) approved by the Victoria University Ethics Committee. Table 3.1 shows individual participant characteristics.

Table 3.1. Participant characteristics.

Participant	Group	Gender	Age (years)	Mass (kg)	Height (m)	Dominant leg	Diabetes duration (years)
1	I	M	34	82.1	185	Right	-
2	I	F	23	70.2	182	Right	-
3	I	M	26	80.3	176	Right	-
4	I	F	24	71.7	178	Right	-
5	I	M	24	87.5	177	Right	-
6	I	F	22	92.4	173	Left	-
7	I	F	23	69.2	170	Right	-
8	I	M	24	75.0	185	Right	-
9	I	F	21	62.5	175	Right	-
10	I	M	27	66.3	168	Right	-
11	I	F	22	70.5	168	Right	-
12	I	M	25	84.2	172	Right	-
13	I	M	32	65.4	170	Right	-
14	I	F	28	70.8	172	Right	-
15	I	F	23	68.3	165	Right	-
16	I	M	39	93.3	184	Right	-
17	II	F	68	87.1	159	Right	-
18	II	F	70	75.7	176	Right	-
19	II	M	86	60.1	163	Right	-
20	II	F	70	82.3	158	Left	-
21	II	F	66	68.6	161	Right	-
22	II	F	70	75.1	156	Right	-
23	II	M	65	75.3	180	Right	-
24	II	F	65	72.7	157	Right	-
25	II	F	66	63.3	165	Right	-
26	II	M	66	97.1	193	Right	-
27	II	M	67	65.5	170	Right	-
28	II	F	65	70.3	173	Right	-
29	II	M	68	78.2	178	Right	-
30	II	F	65	79.3	162	Right	-
31	III	F	65	90.1	174	Right	16
32	III	F	73	86.3	157	Right	15
33	III	M	76	70.1	187	Right	29
34	III	F	67	61.3	165	Right	8
35	III	F	72	78.1	155	Right	10
36	III	F	75	63.3	172	Right	26
37	III	F	68	63.8	175	Right	27
38	III	M	66	62.9	154	Right	20
39	III	M	79	90.6	184	Right	5
40	III	M	65	77.6	156	Right	5
41	III	M	68	85.3	157	Right	5
42	III	M	66	79.5	172	Right	8
43	III	M	65	87.8	169	Right	7

Group I, young adults; Group II, healthy older adults; Group III, older adults with diabetes mellitus.

Kinematic data during overground and treadmill walking were acquired using a three-dimensional motion analysis system (VICON, Oxford, UK) with 14 cameras.

Experimental data were intended to be used by other students, so a full marker setup was used; however, to answer the research questions in this thesis two markers on each shoe were appropriate. Participants were provided with a pair of runners to eliminate effects of differing footwear. A modified full body marker was used (Leardini et al., 2011, Cappozzo et al., 1995) with small reflective markers (14 mm) mounted over specific locations on head, trunk, pelvis, legs, and arms (Figure 3.2, Table 3.2).

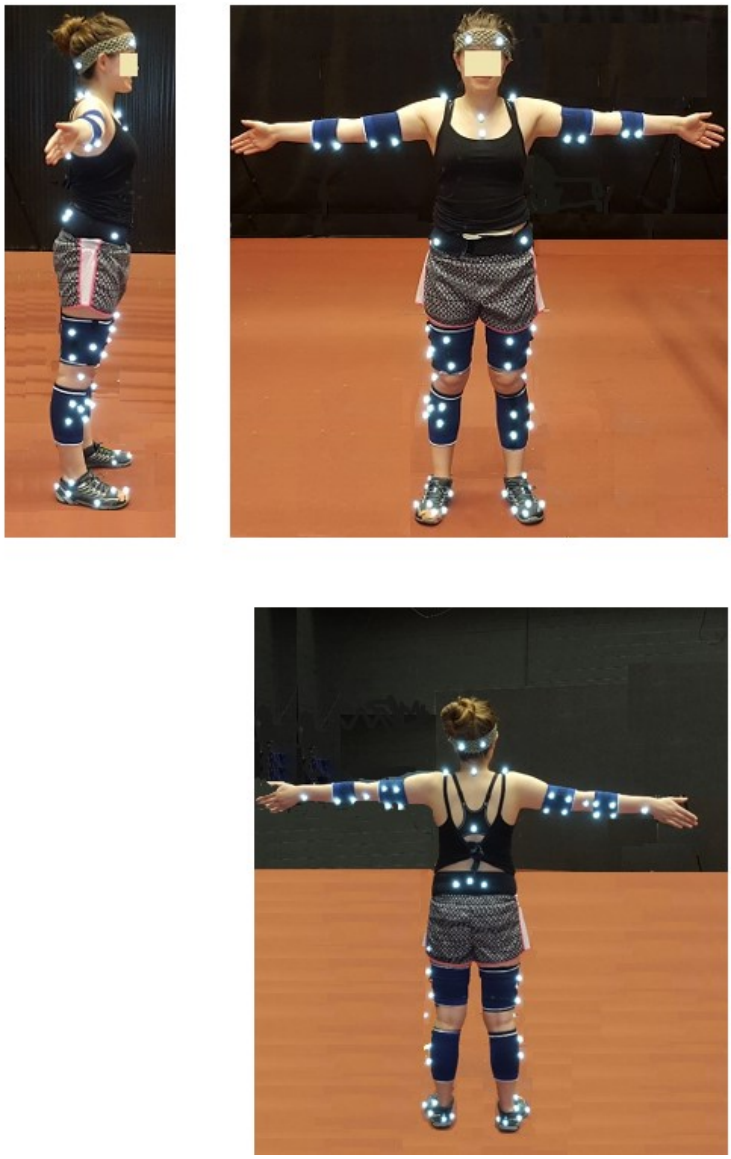


Figure 3.2. The experimental marker set used to collect kinematic data in overground and treadmill tests.

Table 3.2. Markers' locations on the body.

Marker	Location
Head	
H1	Back of the head on the left side
H2	Middle of the forehead on the left
H3	Middle of the forehead on the right
H4	Back of the head on the right side
Trunk	
RSH*	Tip of the right shoulder
LSH*	Tip of the right shoulder
T1	7th cervical vertebra
T2	7th thoracic vertebra
T3	Middle of the sternum
T4	Manubrium of the thoracic cage
Arm	
R/LUA1	Proximal posterior lateral aspect of the right/left upper arm
R/LUA2	Distal posterior lateral aspect of the right/left upper arm
R/LUA3	Distal posterior medial aspect of the right/left upper arm
R/LUA4	Proximal posterior medial aspect of the right/left upper arm
R/LLA1	Proximal posterior lateral aspect of the right/left lower arm
R/LLA2	Distal posterior lateral aspect of the right/left lower arm
R/LLA3	Distal posterior medial aspect of the right/left lower arm
R/LLA4	Proximal posterior medial aspect of the right/left lower arm
R/LMWRIST*	Right ulnar styloid process
R/LLWRIST*	Right later aspect of the styloid process of the radius
Pelvis	
RASI	Right anterior superior iliac spine
LASI	Left anterior superior iliac spine
RPSI	Right posterior superior iliac spine
LPSI	Left posterior superior iliac spine
SAC	Left and right posterior superior iliac spines
Thigh	
R/LTH1	Proximal lateral aspect of the right/left thigh
R/LHT2	Distal lateral aspect of the right thigh
R/LHT3	Distal anterior aspect of the right thigh
R/LHT4	Proximal anterior aspect of the right thigh
R/LLKNEE*	Lateral epicondyle of the right femur
R/LMKNEE*	Medial epicondyle of the right femur
Shank	
R/LSA1	Proximal lateral aspect of the right/left shank
R/LSA2	Distal lateral aspect of the right/left shank
R/LSA3	Distal anterior aspect of the right/left shank
R/LSA4	Proximal anterior aspect of the right/left shank
R/LLMAL*	Right/left lateral malleolus
R/LMMAL*	Right/left medial malleolus
Foot	
R/LFT1	Distal posterior aspect of bisection of the right/left heel on the right shoe
R/LFT2	Between RLMAL and the sole of the right/left shoe
R/LFT3	Lateral aspect of the 5th metatarsal head on the right/left shoe
R/LFT_LTB	1st toe on the right/left shoe
R/LFT5	Medial aspect of the 1st metatarsal head on the right/left shoe
R/LFT6	Between RMMAL and the sole of the right/left shoe
R/LFT7	Dorsal of the right runner on the 3rd tarsal head on the right/left shoe

* Calibration markers were removed.

Tracking markers were attached to sleeves, running shoes, head band, and a belt. Calibration markers and trunk markers were attached to the skin.

3.1.3. Laboratory setup

Ground reaction force data were captured using four A.M.T.I forces plates (Advanced Mechanical Technology Inc., Watertown, Mass., USA) during overground walking (Figure 3.3) and two A.M.T.I force plates embedded into a motorised treadmill (Figure 3.4).

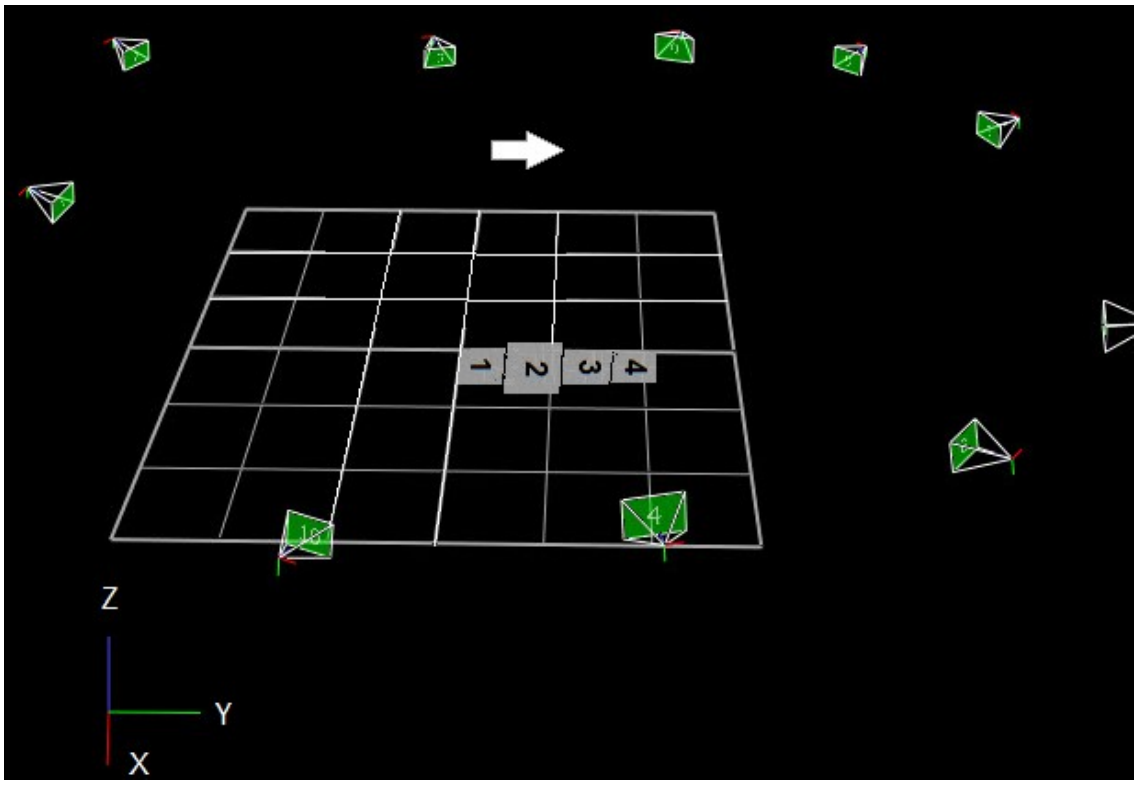


Figure 3.3. Laboratory setup in overground gait adaptability tests. The white arrow shows direction of walking.

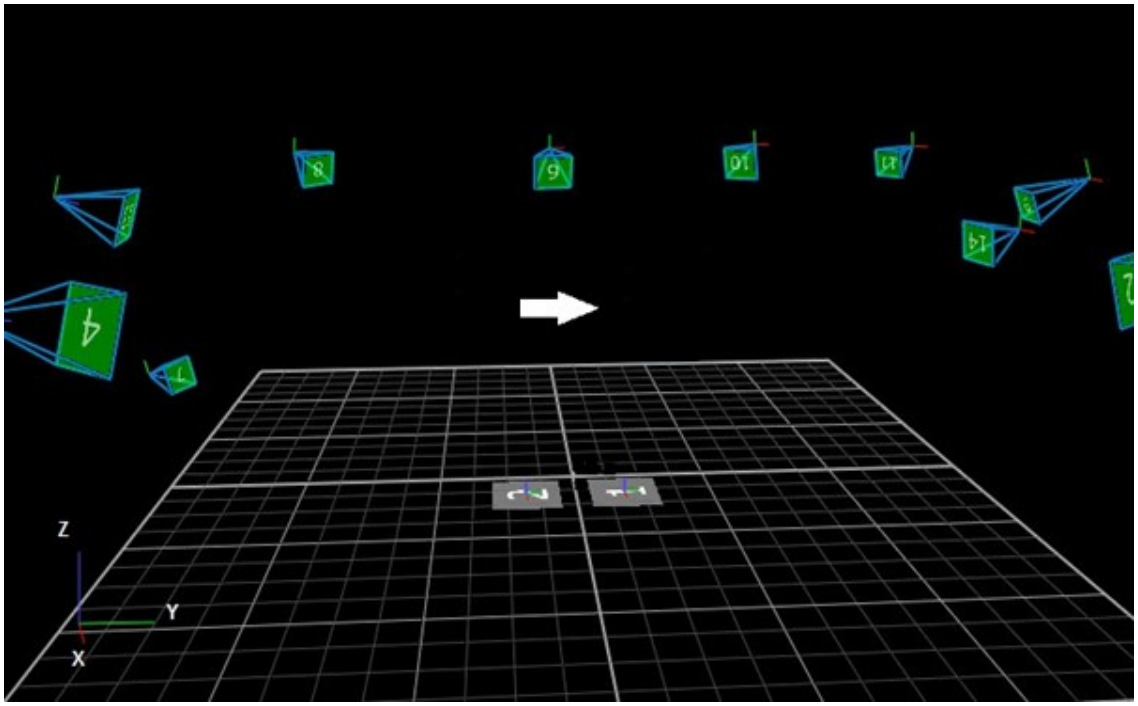


Figure 3.4. Laboratory setup in treadmill gait adaptability tests.

Instrument used in overground gait adaptability tests to address Research Question1

The instrument used in overground tests included two stepping targets and one obstacle. Each walking test presented either a target or obstacle, or neither of these. Stepping targets were projected using laser beams and the obstacle was a fine cord located on the ground and lifted 5 cm by two servomotors (Figure 3.5). Laser beam projectors presented two beams across the walkway: one for a short stepping target and the other for a long stepping length target (Figure 3.6).

Step length targets were scaled to a participant's baseline SL; however, the obstacle had 5 cm height for all participants. Targets were 0.6 × baseline SL and 1.4 × baseline SL (Hoogkamer et al., 2015).

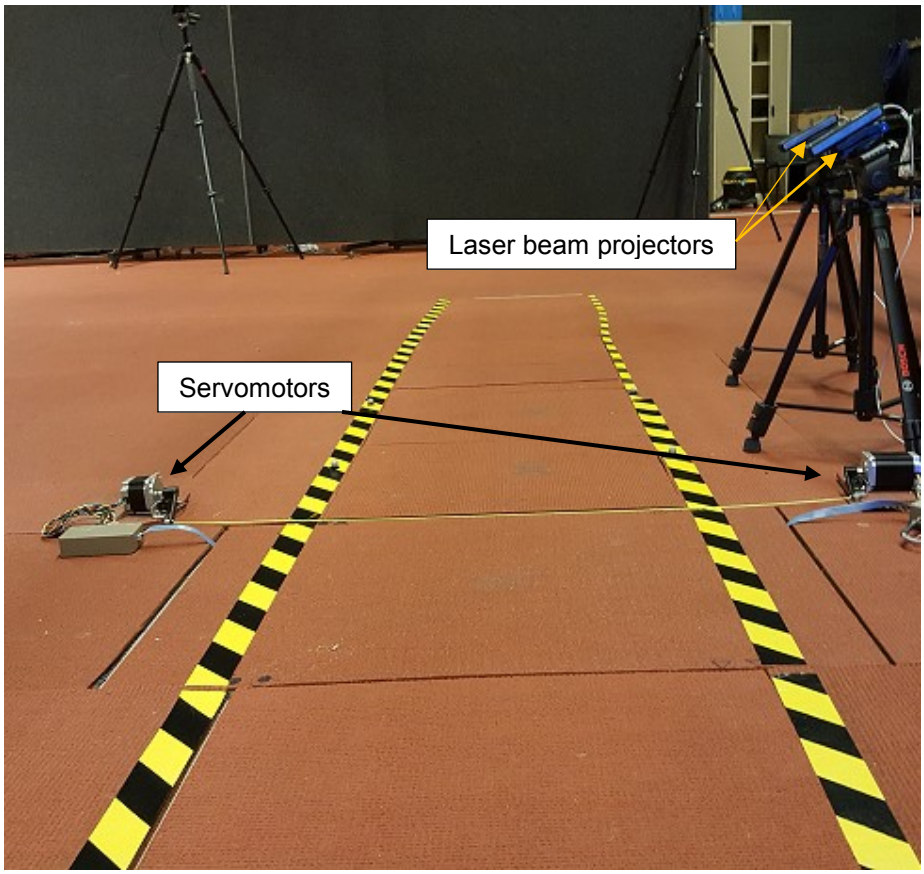


Figure 3.5. The set-up for overground gait adaptability tests. Two laser beam projectors and two servomotors were used to present two stepping targets and an obstacle during overground walking.

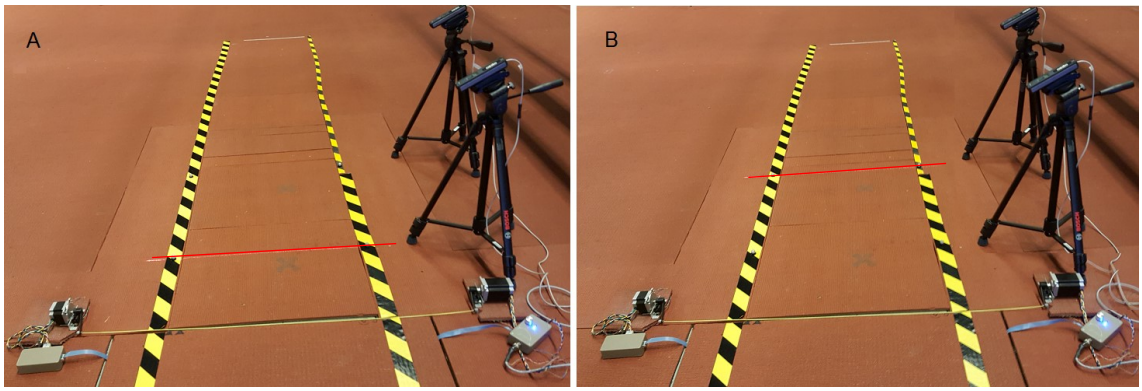


Figure 3.6. Projected stepping targets overground. A laser beam (red line) for a short stepping target (A) and a laser beam (red line) for a long stepping target (B).

The construction of the obstacle was unique. Two synchronised servomotors presented a obstacle of 5 cm in height when they were triggered (Figure 3.7). The cord was made from nylon that was attached to the servomotors by two magnets (Figure 3.8).

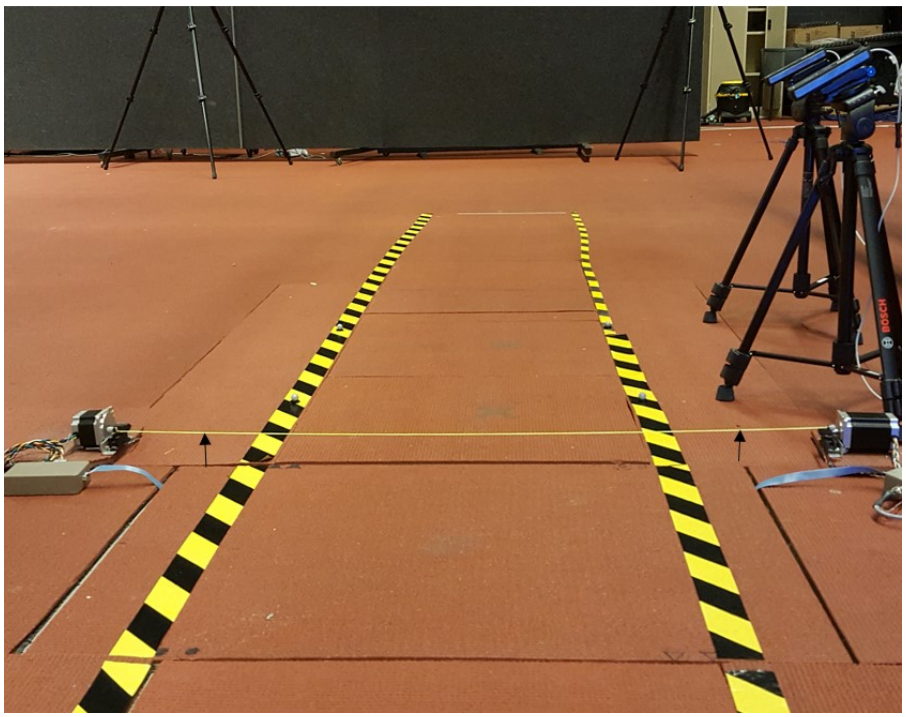


Figure 3.7. A 5-cm height was used in the overground gait adaptability tests. The yellow nylon cord was held at a 5-cm height when the synchronised servomotors were turned on.

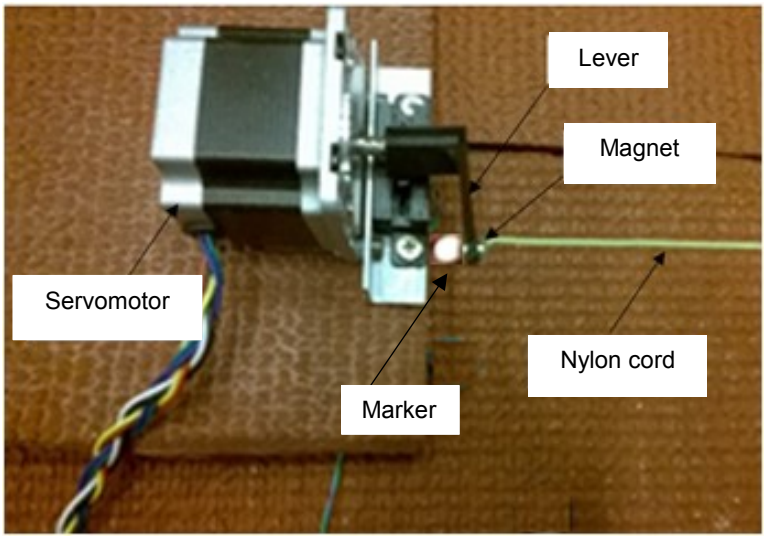


Figure 3.8. The obstacle mechanism in overground gait adaptability tests. A nylon cord was attached on each end to a servomotor's lever by a small magnet.

The nylon cord placed on the ground between the first force platform and the second force platform when it was not activated. Two markers were attached at the ends of both levers to ensure that the cameras recorded the correct location of the cord. The slight touch of a foot during crossing the obstacle immediately caused the cord attached to the servomotors to become detached. Thus, the forward swing of the leg was not obstructed if the participants could not avoid the obstacle.

A condition in which neither a stepping target nor obstacle was presented, was also added to make it more difficult for participants to predict the presenting conditions. Hence, one of four conditions was presented in each trial when one foot hit the first force plate.

Participants responded to what was presented. For example, when nothing was presented, they walked straight through. If a stepping target or the obstacle was presented, they adjusted the location of their next foot landing related to the presented stepping target or obstacle, and responded in the following step by adapting the toe marker with the stepping target or crossing the obstacle, respectively.

Instrument used in treadmill gait adaptability tests to address Research Question 2

An instrument (Mehdikhani et al., 2019) was developed and used to present virtual targets and real-time SL/MTC during treadmill walking on a monitor. A customised MATLAB program presented the real-time SL/MTC height of each participant, and their subject-specific SL/MTC height targets during walking on a large, monitor installed in front of the treadmill at eye-level. The MATLAB program was connected to the Visual3D RT Server (C-Motion, Germantown, USA), which streamed real-time 3D marker data into the MALAB program (Figure 3.9).

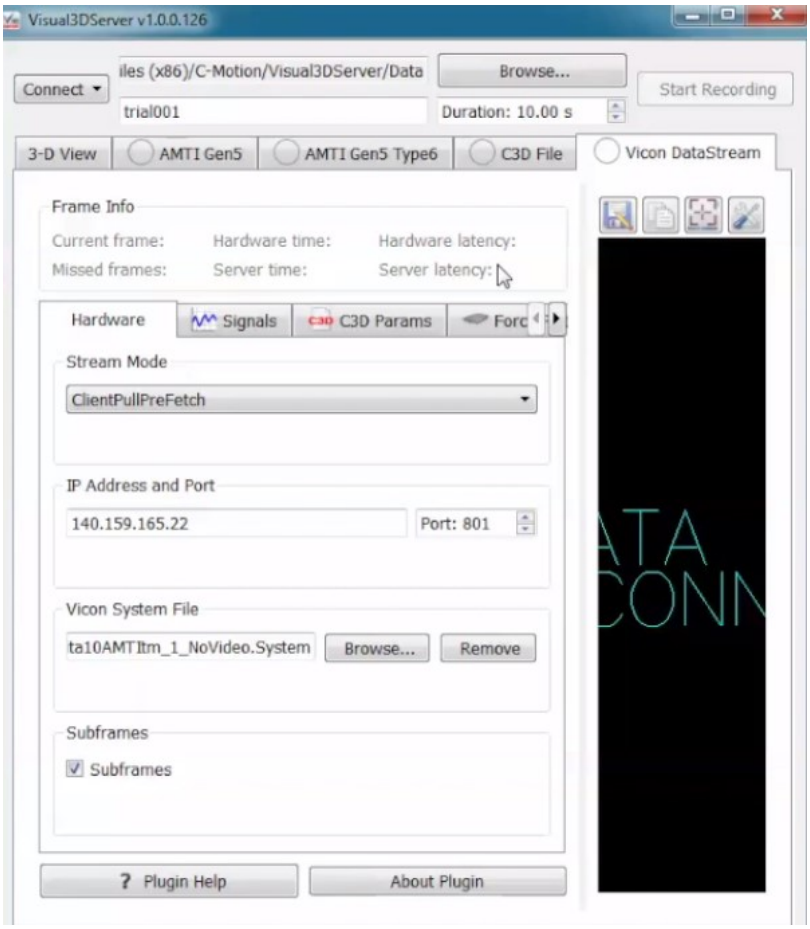


Figure 3.9. Visual3D Server connected to the VICON motion capture system.

The laptop used to run the MATLAB program was connected to the monitor. Participants walked on the treadmill while viewing a continuous real-time graph based on the differences between toe markers in the AP direction on the monitor (Figure 3.10).

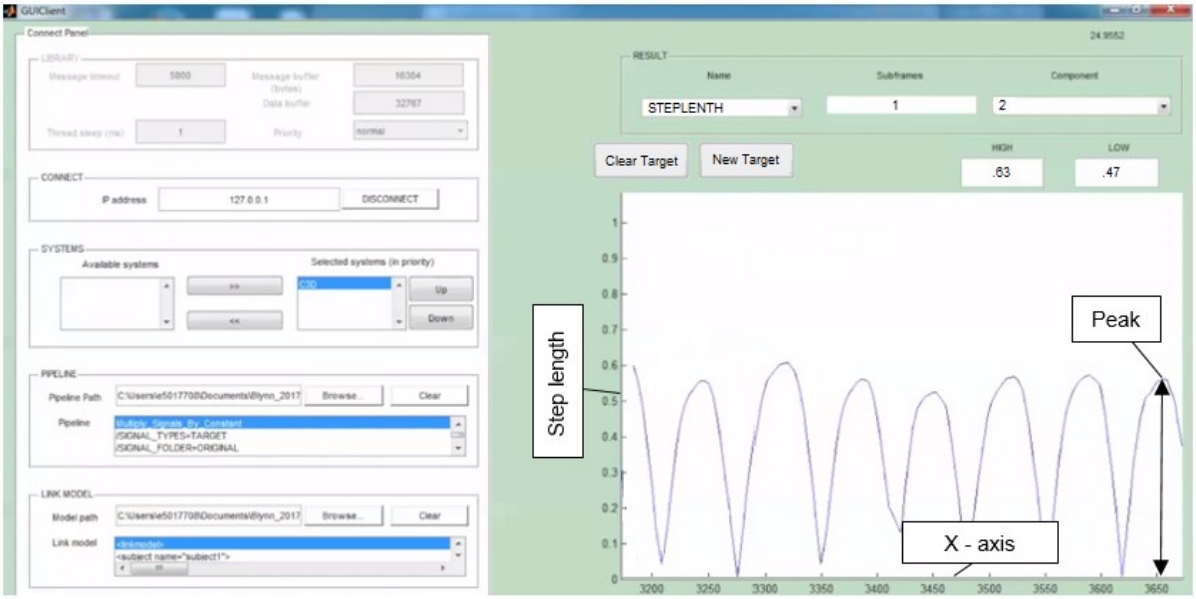


Figure 3.10. The customized MATLAB program presented the real-time step length for each step on the monitor. The peak of each graph was the maximum distance between two toe markers placed on the great toes in the anterior-posterior direction. An imaginary line for each graph presents the step length for one step (see arrow).

As shown in Figure 3.11, to give feedback on foot displacement in the AP direction, a line was discretely presented above or below the peak SL graphs. To do so, values of step length targets (10% shorter and longer than baseline SL) were added.

When a New Target was clicked, a red line appeared below the peaks on the SL graph when shortening steps, or above the peaks of the graphs when lengthening steps. For example, the red line in Figure 3.11 placed above peaks of step lengths. Therefore, the task was to walk with longer steps in a way that each peak was matched with the red line until the line was removed. Participants used biofeedback (the difference between the red line and the peak of each SL graph) to adjust lengths of steps during the time the biofeedback was presented. The red line was presented and removed every 10 steps. Therefore, participants walked normally when the data collection started by presenting

the red line (biofeedback). After 10 trials, the red line was removed for 10 steps, and again it was presented for 10 steps and so on until data for 120 continual steps (60 steps without biofeedback and 60 steps with biofeedback) were collected.

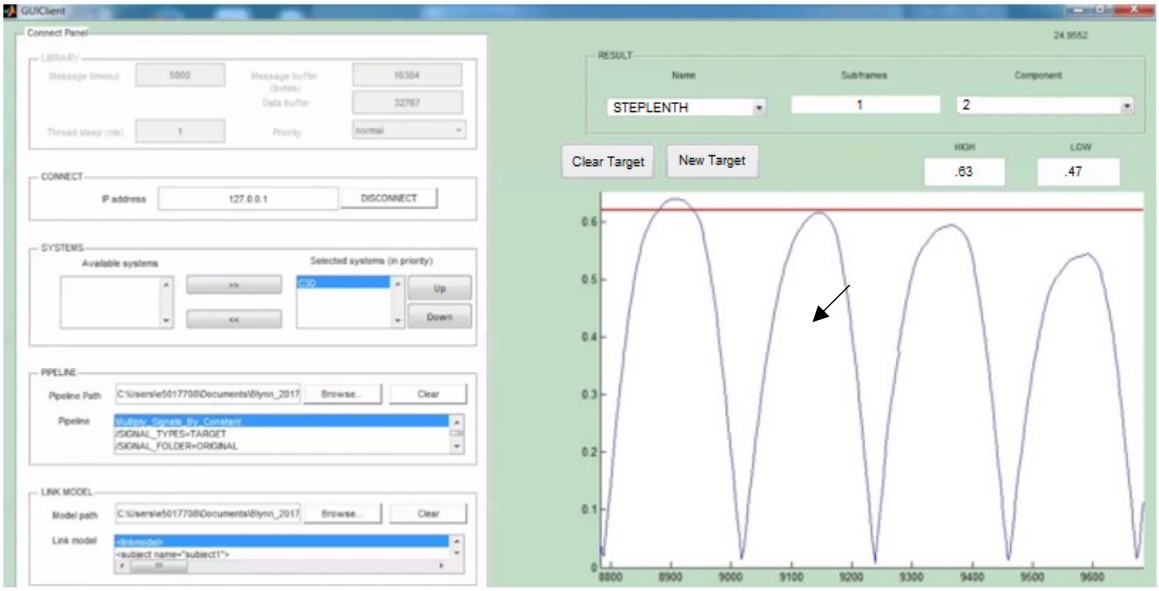


Figure 3.11. The customized MATLAB program presented long and short step length targets. Here, the participant managed to use biofeedback and match one step length with the presented step length target (see arrow).

To present feedback on foot displacement adaptation in the vertical direction, Right/Left MTC and Component 3 were selected from the lists (Figure 3.12). Based on the pilot study results (Appendix A), high and higher MTC targets were 2.5 cm and 3.5 cm higher than baseline MTC, which were added to High and Low in the software. When the red line appeared, participants increased their foot-ground height to match the MTC peak with the red line on the monitor in each step as long as the line was presented (Figure 3.13).

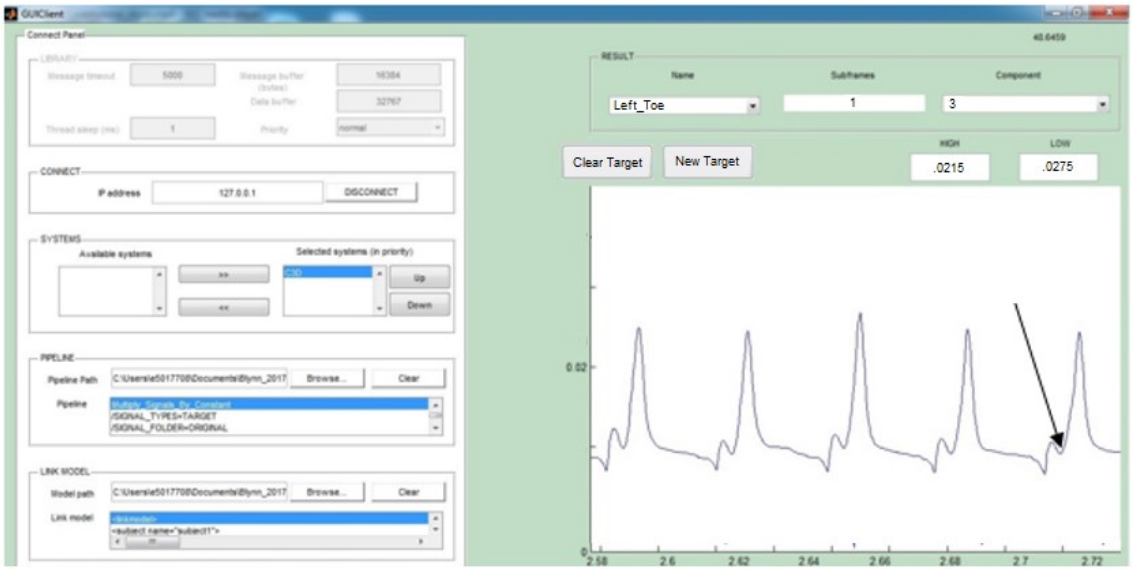


Figure 3.12. The customized MATLAB program presented the minimum toe clearance (see arrow).

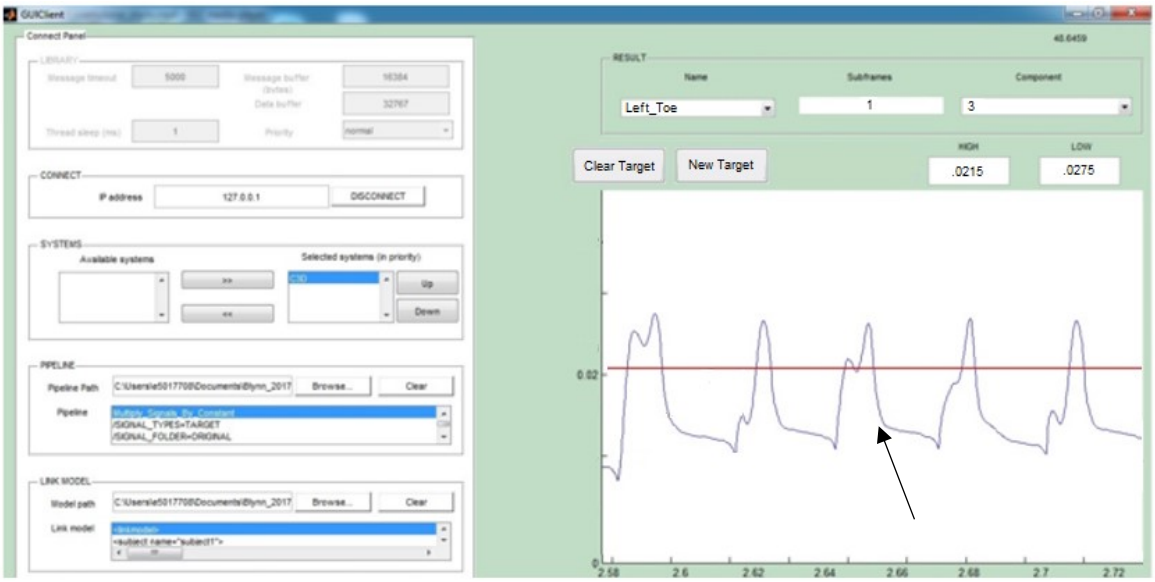


Figure 3.13. The customized MATLAB program quantifies the minimum toe clearance (MTC) error. The arrow shows where the participant was able to use visual feedback to match the MTC peak with the red line (arrow) in their third attempt.

In total, participants responded to the task for 10 steps and then walked for 10 steps without seeing any red line on the monitor until 24 blocks of 10 steps (12 blocks with

biofeedback and 12 blocks without biofeedback) were recorded. Each participant attended the Biomechanics Laboratory twice (different days) to complete either treadmill or overground gait adaptability tests (randomly allocated). Each day, the data collection session took 100-120 minutes.

3.1.4. Data collection protocol

Walking data were collected during overground and treadmill walking tests for each participant. Each participant attended the Biomechanics Laboratory twice, each time taking from 100-120 minutes to complete OGA and TGA tests.

Overground gait adaptability tests, Research Question 1

A static trial was recorded to define the marker locations in standing pose. Participants were asked to stand still on a force plate with arms abducted to about 90^{o} (see Section 3.1.2, Figure 3.2). Collected data from this static trial was used to determine the precise measurement of leg length (distance between markers on the anterior superior iliac spine and the medial malleolus) for normalising step lengths. VICON Nexus (Version V2.10.3, VICON, Oxford, UK) was used to record marker trajectories and ground reaction forces simultaneously when participants walked at self-selected walking speed in baseline and foot displacement adaptation tests. Adequate recovery time was provided between trials to avoid the effects of fatigue.

Baseline test: Participants walked on a smooth level surface at their preferred speed for 8 meters, 6 times, starting with either the right or the left foot.

Foot displacement adaptation tests: For each participant, the locations of stepping targets and the obstacle were adjusted according to baseline SL, as presented in Figure 3.14.

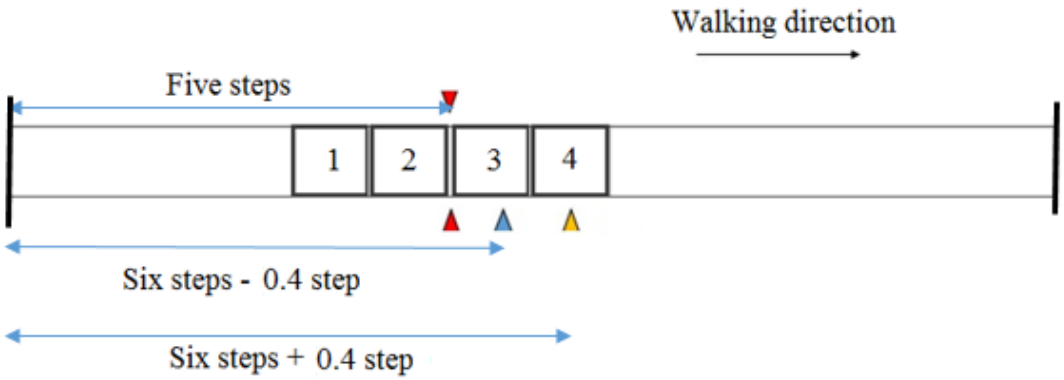


Figure 3.14. Overhead view of the experimental setup in the foot displacement adaptation test in the walkthrough condition. Red triangles present locations of the potential obstacle. The blue and yellow triangles present the locations of two laser beam projectors for potential short and long stepping targets, respectively. Two black lines present the starting and finishing points.

The starting point for each participant was adjusted in a way that participants placed their 6th step on the force platform no.4 in Figure 3.14. So all participants were required to respond to the presented task in the 3rd stride (steps 5 and 6). Participants saw a laser beam or the obstacle, adjusted the landing of the foot in the fifth step prior to the cord, and completed the step adaptation in the sixth step (Figure 3.15).

Participants did not receive any instruction on how to match the toe marker on the leading foot with presented laser beams and how to cross the obstacle. For example, participants were free to shorten or lengthen previous steps before responding to a SL target or the obstacle.

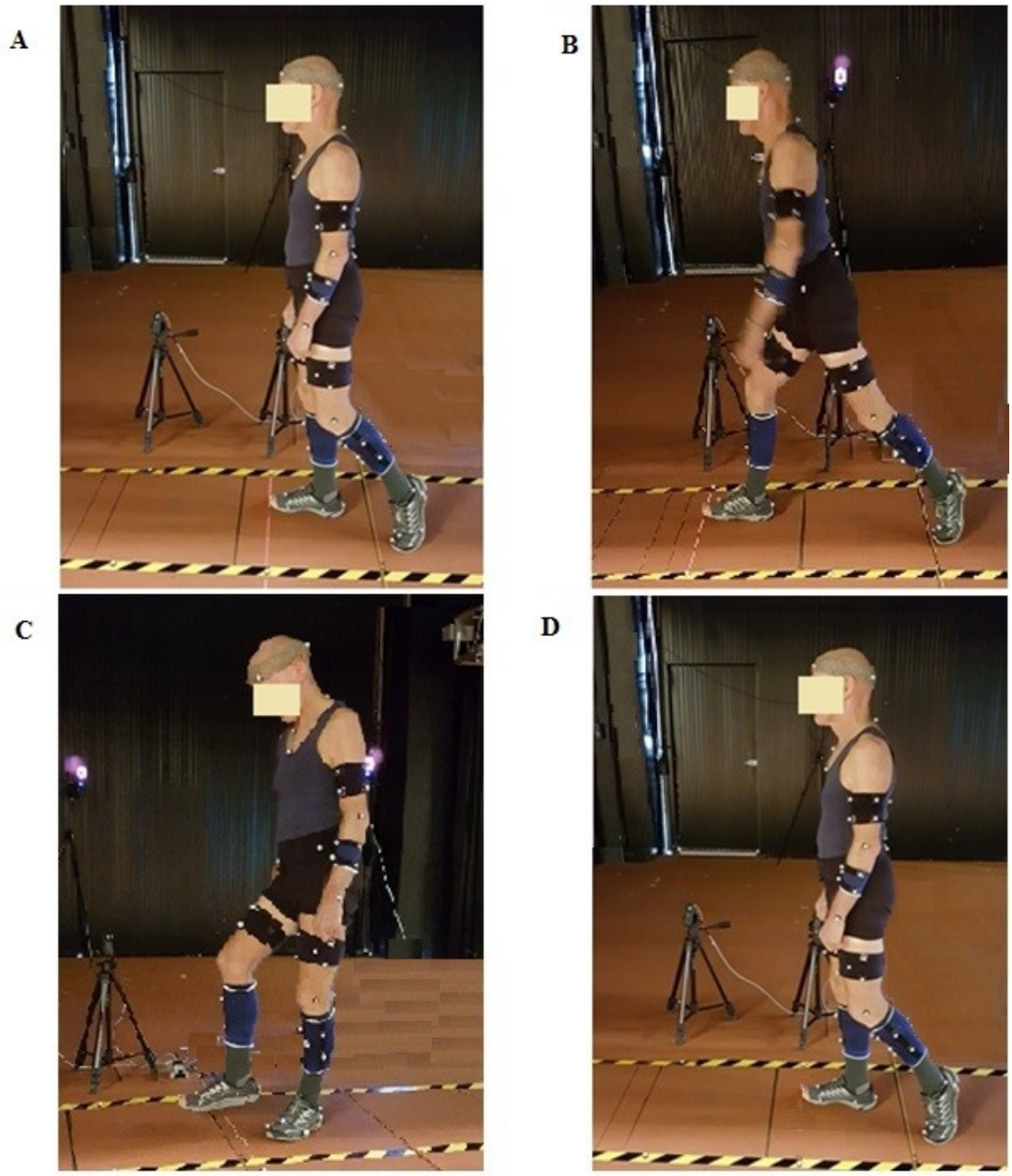


Figure 3.15. The foot displacement adaptation tests with four conditions during overground walking. Shortening one step (A), lengthening one step (B), crossing the obstacle (C), and walking through (D).

Participants practiced all conditions at least once and then completed 40 trials under four random conditions: shorten one step (A), lengthen one step (B), cross the obstacle (C), and walk through (D) when none of these conditions were presented (Figure 3.15).

Participants were unaware which condition would be presented because a random order for both conditions and leg presentation was used (Table 3.3).

Table 3.3. Order of conditions in the foot displacement adaptation tests**.**

Condition	The leg placed behind the starting point	File name
Shortening one step	Right	1
Lengthening one step	Left	2
Lengthening one step	Right	3
Shortening one step	Right	4
Walking through	Right	5
Lengthening one step	Left	6
Crossing the obstacle	Right	7
Walking through	Left	8
Crossing the obstacle	Left	9
Shortening one step	Right	10
Crossing the obstacle	Right	11
Walking through	Left	12
Shortening one step	Left	13
Shortening one step	Left	14
Crossing the obstacle	Right	15
Walking through	Left	16
Lengthening one step	Right	17
Lengthening one step	Left	18
Walking through	Right	19
Crossing the obstacle	Left	20
Walking through	Left	21
Lengthening one step	Right	22
Crossing the obstacle	Left	23
Lengthening one step	Right	24
Crossing the obstacle	Right	25
Lengthening one step	Left	26
Walking through	Left	27
Shortening one step	Left	28
Shortening one step	Right	29
Crossing the obstacle	Left	30
Walking through	Right	31
Shortening one step	Left	32
Walking through	Right	33
Crossing the obstacle	Left	34
Shortening one step	Right	35
Crossing the obstacle	Right	36
Shortening one step	Left	37
Lengthening one step	Left	38
Lengthening one step	Right	39
Walking through	Right	40

Experimental trials were deemed successful if:

- Participants took five steps and responded to a condition in their sixth steps.
- Participants correctly responded to presented conditions. For example, participants lifted their feet to cross the obstacle when it was triggered, or participants shortened their target steps to match their toe markers when the laser line projected short stepping targets.
- Participants responded to conditions and continued walking until they passed the finishing point.

Unsuccessful trials were marked and repeated at the end of the 40 random trials. Touching the obstacle was considered a successful trial if participants attempted to cross the obstacle; however, the frequency of touching the obstacle was recorded for each participant.

Treadmill gait adaptability tests, Research Question 2

Before data collection, a familiarisation session was completed for each participant. As shown in Figure 3.16, participants wore a safety harness and walked for 30 seconds at different speeds between 2 km/h and 4.5 km/h. Every 30s the speed increased by 0.5 km/h to become familiar with treadmill walking.

Preferred treadmill walking speed was determined for each participant. Treadmill speed was increased when participants were walking until they felt uncomfortable and asked not to increase their speed. Each time the speed was increased, participants walked for 30 seconds and then indicated whether they were comfortable or if they wanted to walk faster (Nagano et al., 2013). If not comfortable, the speed was recorded and reduced to about 2

km/h before being increased again. In the TGA tests, the preferred speed was the average of six uncomfortable fast and slow speeds, as calculated for each individual participant.

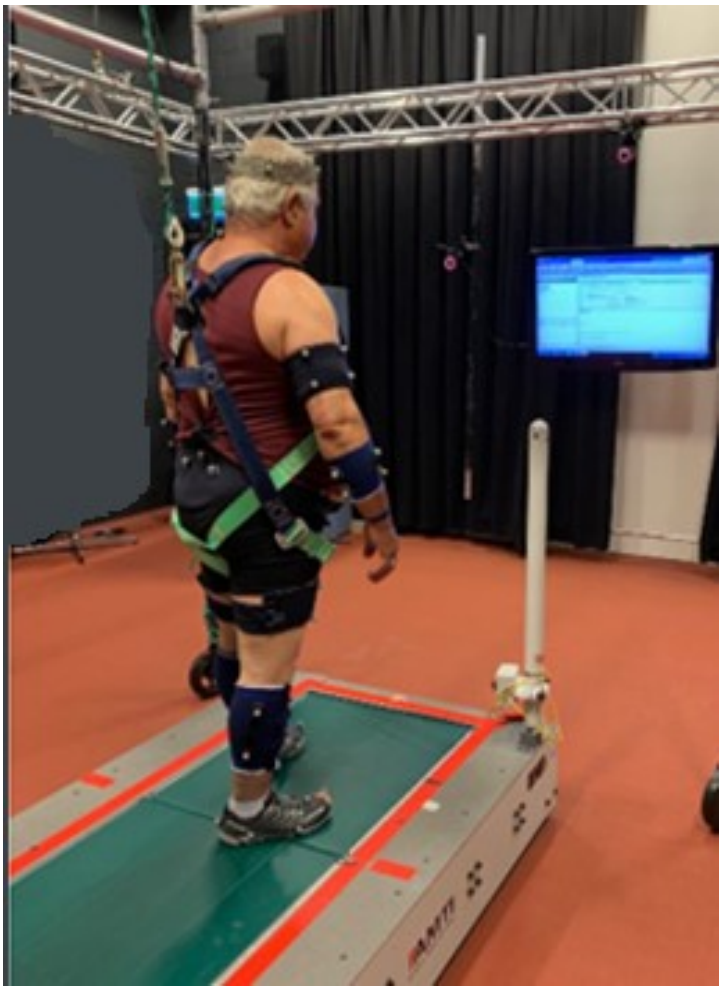


Figure 3.16. Participant wearing a safety harness to complete treadmill gait adaptability tests.

A marker model was attached to label markers of a static trial. Then, dynamic trials were recorded. In baseline and adaptability tests, both VICON Nexus and the customised MATLAB program collected data simultaneously. Data from the labelled markers streamed via the Visual3D Server software into the customised MATLAB program were

used to plot the graphs of step length and MTC height in real-time for each participant during treadmill walking.

Dynamic trials were collected during:

- **Baseline walking**: Participants were asked to walk at their preferred speed, without seeing anything on the monitor. The Visual3D Server software was connected to Nexus software to send a real-time trajectory of markers to customized MATLAB software (Version 2015b, MathWorks, USA) for three 30-second walking trials to determined step length and MTC means to scale targets for each participant individually.
- **Foot displacement adaptation with and without feedback**: The monitor was turned on and graphs and interest points were explained to participants. Participants walked at preferred speed and responded to long and short stepping targets during 120 steps with and without biofeedback. They then completed trials for quantifying the effects of biofeedback on foot displacement adaptation in the vertical direction, and responded to high and higher MTC targets. Each target was presented three times and removed three times. To minimise the cognitive interference and adaptation of the sensorimotor processes, the method of presentation of biofeedback was designed in a way to introduce an unknown planning error (Gaveau et al., 2014) by unpredictably presenting a target for every 10 steps and removing it for the following 10 steps. This planning error affects the amplitude or direction of the movement (Vindras et al., 2005). However, the awareness of a change in the location of the target does not impact on the correction of responses during online correction (Castiello et al., 1991). The order

of conditions in the foot displacement adaptation in the AP direction were: long (10 steps), no target (10 steps), short (10 steps), no target (10 steps), short (10 steps), no target (10 steps), long (10 steps), no target (10 steps), short (10 steps), no target (10 steps), long (10 steps) and no target (10 steps). The order of conditions in the foot displacement adaptation in the vertical direction was: higher (10 steps), no target (10 steps), high (10 steps), no target (10 steps), high (10 steps), no target (10 steps), higher (10 steps), no target (10 steps), high (10 steps), no target (10 steps), higher (10 steps) and no target (10 steps). .

Tasks were firstly explained to participants who then practised each condition at least once and then completed the tests.

3.2. Computational analyses

3.2.1. Overground gait adaptability tests, Research Question 1

Gait spatiotemporal parameters

Nexus Vicon software was used to analyse one gait cycle in each trial in both baseline and adaptability conditions. Ground reaction forces and trajectories of toe and heel markers determined spatiotemporal parameters of a gait cycle: step velocity, stance time, swing time, DS time, and SL.

Because height can affect SL (Murray et al., 1964), step lengths were normalised. Leg length for each participant was used to normalise step lengths. Velocity (m/s) was calculated by dividing SL to the step time (SL/step time). Stance time, swing time and DS time were reported as percentages of a gait cycle.

Foot displacement measures in response to step length targets

Visual3D was used to calculate foot displacement errors. When the leading foot was in contact with the ground, absolute and constant errors in response to SSL/LSL targets were:

$$Y_{\text{toe marker}} - Y_{\text{short/long SL target}} = \text{Short/long absolute error}$$

$$\left| Y_{\text{toe marker}} - Y_{\text{short/long SL target}} \right| = \text{Short/long constant error}$$

If the toe marker passed the laser beam, it implied an overshooting response; a step length greater than the presented step length target indicates the error as positive (Figure 3.17).

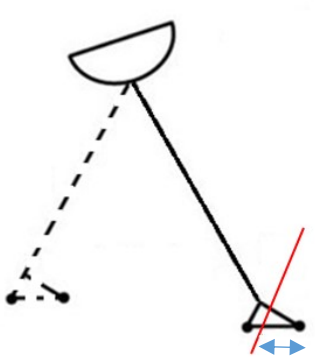


Figure 3.17. Step length error was the horizontal distance between the toe marker and the projected line. The arrow shows the step length error in response to the projection of a laser beam, an overshooting response.

Foot displacement measures in response to the triggered obstacle

When the leading foot toe marker was vertically above the obstacle (Figure 3.18), the toe-obstacle clearance height was the distance between the toe marker on the shoe of the leading limb and the cord (obstacle). In the target step, the toe-obstacle horizontal distance

was the horizontal distance between the toe marker and the cord of the obstacle. The heel-obstacle distance was the horizontal distance between the heel marker of the crossing foot and the cord of the obstacle in the target step.

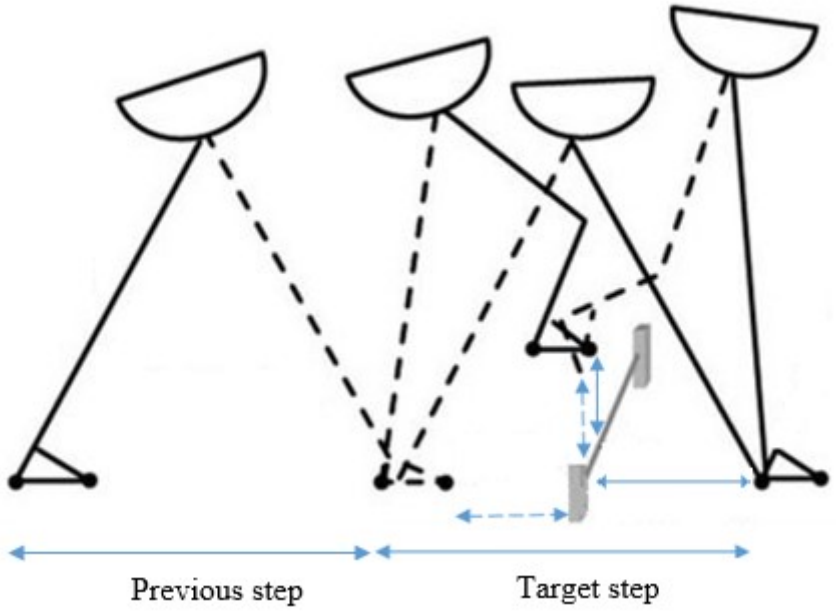


Figure 3.18. The toe-obstacle vertical distance when avoiding the obstacle. The vertical arrow shows the toe-obstacle clearance height of the leading foot.

3.2.2. Treadmill gait adaptability tests, Research Question 2

When baseline walking for each participant was recorded, a customised MATLAB program used the offline data of three 30-s trials to calculate mean step length, left MTC, and right MTC in baseline. They were used to determine four subject-specific targets.

When the session ended, labelled data were extracted as C3D files to calculate constant and absolute errors for step length and MTC adaptation in response to the sudden appearance of targets (without biofeedback) and during online correction (with

biofeedback). Constant errors showed the deviation from presented targets whereas absolute errors only presented the distant from targets without presenting directions.

Short and long step length constant and absolute errors

To quantify the step length errors for 60 steps (6 blocks of 10 steps) when step length targets were presented, the three dimensional marker coordinates of left and right toe markers were saved separately as two American Standard Code for Information Interchange (ASCII) files, using the Visual3D software. MATLAB was used to reproduce the graph of the step lengths and presented targets in TGA tests. The step length absolute error for each step was the positive difference between the step length and the target (Figure 3.19).

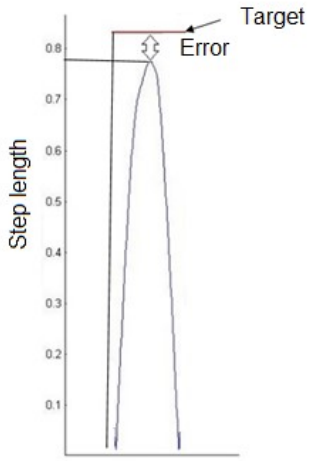


Figure 3.19. Step length error. The step length error was the distance between the peak and the line (the difference between adapted step length and the step length target).

High and higher minimum toe clearance constant and absolute errors

The three dimensional marker coordinates of left and right toe markers in the trials in which MTC targets were presented were saved separately as two ASCII files. Similar to

Figure 3.16, MATLAB plotted the trajectory of the toe marker in the vertical direction during each step when an MTC target appeared and when the MTC target was removed. The MTC absolute error was the difference between the adjusted MTC height in response to an MTC target, without considering whether the adjusted MTC height was higher or lower that the presented MTC target (Figure 3.20).

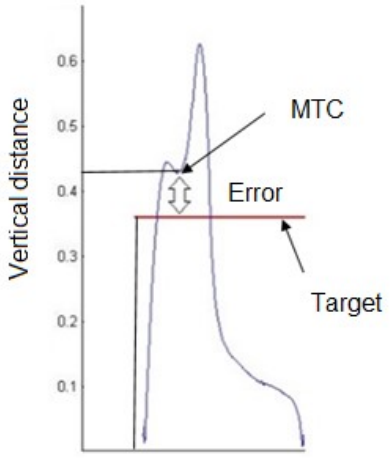


Figure 3.20. The minimum toe clearance (MTC) error. The MTC error was the difference between the MTC peak and the red line (target).

Reactive responses (the first step of each block) when targets suddenly appeared, and online correction responses (steps 2-10) when participants used feedback to adapt step lengths and MTC heights, were separately averaged.

3.2.3. Agreement of overground and treadmill gait adaptability tests, Research Question 3

A graphical method (Altman and Bland, 2003) was used to investigate the validity of the third hypothesis. The Bland-Altman plot, or difference plot, was used to find systematic biases. It compared averaged measures determined by two separate tests: OGA and TGA. A scatter graph was used to present differences between means in response to the sudden appearance of targets quantified by each test against the averages of two means.

3.3. Statistical analyses

All statistical analyses were performed in SPSS (Version 25 for Windows, SPSS Science, Chicago, USA) at a significance level of $\alpha = 0.05$. The descriptive statistics described each group in baseline and adaptability tests during TGA and OGA tests. Variables were described or illustrated by measures of central tendency and measures of variability including mean and standard deviation (SD) using tables and Microsoft Excel bar-plots.The issue of outliers was reduced during data processing. For example, if error of step length adaptation in was trial was far away from the mean, the trial was removed. Because dependent variable was the average of some trials, the chance of having a participant with an outlier became low.

Tests of normality (Shapiro-Wilk) determined whether parametric tests could be used. If they were greater than 0.05, parametric ANOVA tests were used. Homogeneity of variance and the violation of assumptions were checked using Levene's test and the assumption of sphericity.One-way non-parametric ANOVA was used to compare height and body mass between groups because dependent variables were distributed normally. Analyses of variance (ANOVA) repeated measured tests were used to

investigate group (young vs. healthy older vs. older diabetes) and condition (baseline vs. walkthrough vs. short/long/obstacle) effects, (three factors and two conditions) on dependent gait parameters (velocity, stance time, swing time, double support time, step length).The violation of the conducted tests were considered to report p-values. Multiple comparisons (Bonferroni post hoc tests) were used to investigate which groups were different when the main effect of groups was significant.

Absolute and constant errors for step lengths and MTC heights were not normally distributed (Appendix G), so nonparametric tests (Kruskal-Wallis H and Mann-Whitney U) were used to compare errors between groups and between conditions.

To investigate whether hypotheses related to the Question 3 were true, scatter plots of the difference between paired measurements (TGA – OGA) against their mean value ([TGA + OGA] / 2) were prepared for step length and MTC absolute errors in response to each target for all participants. Because the TGA tests quantified MTC absolute error means and the OGA test quantified the toe-obstacle clearance heights, the mean vertical distance mean between the MTC peak of the leading foot and the obstacle was measured for each person. This measure was used to investigate the agreement between two tests for qualifying the MTC absolute errors. If 95% of points in scatter plots lay between the limits of agreement lines (mean difference ± 1.96 SD), the TGA and OGA tests for calculating SSL, LSL absolute error means, and high and higher MTC absolute error means would be considered interchangeable.

Chapter 4: Experimental Results

Section 4.2 presents the comparison of overground gait in the baseline between groups. Section 4.3 presents the comparison of overground gait adaptability and foot displacement adaptation between groups. Section 4.4 presents the effects of biofeedback on foot displacement adaptation during treadmill walking. Finally, Section 4.5 presents the interchangeability of adaptability tests during OGA and TGA tests.

4.1. Participants

Forty-eight volunteers (16 young adults (Group I) , 16 healthy older adults (Group II) and 16 older adults with diabetes (Group II)) attended the Biomechanics laboratory at Victoria University. All older participants with diabetes had diabetes, with a mean glycated haemoglobin A1c (GHbA1c) of 7.6 % ± 1.8 %. None of the participants had fallen in the year before their participation, and all of them achieved a score of 27 and over when the MMSE was completed for them.

Two older adults with diabetes were excluded because they had peripheral neuropathy. The rest of the older participants achieved MNSI scores between zero and 1.5, so neuropathy was not confirmed. Two healthy older participants and one older participant with diabetes were excluded because they were uncomfortable walking on the treadmill.

In total, 16 young adults, 14 healthy older adults, and 13 older adults with diabetes completed tests during overground and treadmill walking. Among them only one young adult and one healthy older adult kicked the ball with their left legs. Table 4.1 presents the characteristics of participants in each group.

Table 4.1. Descriptive statistics of each group pf participants. Mean ± standard deviation of age, body mass and height in young adults (Group I), healthy older adults (Group II), and older adults with diabetes (Group III) are presented.

	Group I (n = 16)	Group II (n = 14)	Group III (n = 13)
Participants	9 males, 7 females	8 males, 6 females	7 males, 6 females
Age (years)	26.06 ± 4.97	68.36 ± 5.43 [b]	69.62 ± 4.81 [b]
Body mass (kg)	75.61 ± 9.05	75.04 ± 9.75	76.67 ± 11.14
Height (cm)	175.00 ± 6.40	167.93 ± 10.84 [b]	167.23 ± 9.97 [b]

[b] Significantly taller and older than Group I ($p < 0.05$).

Characteristics of groups were not significantly different between groups, except for age and height. Body mass means were not different between groups (H (2) = 0.121, p = 0.941). However, mean ages and height were different between groups (H (2) = 29.937, $p < 0.001$, effect size (The eta squared (η^2)) = 0.95 and H (2) = 6.117, $p < 0.047$, η^2 = 0.143).

Multiple comparisons revealed that Group I and Group II, and Group I and Group III were different in age ($p < 0.001$ and $p < 0.0009$) and height ($p = 0.034$ and $p = 0.039$). Group II and III were not different in age, body mass and height.

4.2. Baseline gait

Table 4.2 presents measured gait parameters of participants in each group while walking for 8 metres on an unobstructed level surface at preferred speed (Table 4.2).

Table 4.2. Descriptive statistics of gait parameters in each group while walking at preferred speed in baseline. Mean ± standard deviation of step velocity, stance time, swing time, double support time and step length are presented.

	Group I (n = 16)	Group II (n = 14)	Group III (n = 13)
Step velocity (m/s)	1.29 ± 0.13	1.25 ± 0.10	1.19 ± 0.12
Stance (% of gait cycle)	60.71 ± 1.32	60.87 ± 1.62	61.83 ± 1.81
Swing (% of gait cycle)	39.05 ± 1.20	38.10 ± 2.15	38.43 ± 2.79
Double support (% of gait cycle)	10.81 ± 1.10	10.83 ± 1.12	11.80 ± 1.66
Step length (% of leg length)	78.03 ± 2.99	77.37 ± 6.24	73.21 ± 6.99

Mean spatiotemporal parameters of gait were not different between groups while walking at preferred speed. Mean velocities of steps were not significantly different (H (2) = 5.786, $p = 0.055$, $\eta^2 = 0.110$). Mean step lengths were not significantly different between groups (H (2) = 4.986, $p = 0.083$, $\eta^2 = 0.131$). Mean stance, DS and swing times were not different between groups (stance time, $F_{2,40} = 2.016$, $p = 0.146$, $\eta^2 = 0.092$; swing time, $F_{2,40} = 0.808$, $p = 0.453$, $\eta^2 = 0.039$; DS time, $F_{2,40} = 2.566$, $p = 0.089$, $\eta^2 = 0.114$).

4.3. Gait adaptability

This section presents gait spatiotemporal parameters in each condition of the OGA test in the three groups of participants (see Table 4.3). Because Shapiro-Wilk tests of normality for variables were greater than 0.05, parametric

Step shortening condition

Conditions were significantly different in: target step velocity ($F_{1.65,\ 65.98} = 50.668$, $p < 0.001$); previous step velocity ($F_{1.658,\ 66.337} = 19.173$, $p < 0.001$); leading leg stance time ($F_{2,80} = 14.513$, $p < 0.001$); swing time ($F_{2,80} = 6.422$, $p = 0.003$); DS time ($F_{2,80} = 6.974$, $p = 0.002$); target step length ($F_{1.696,\ 67.850} = 109.682$, $p < 0.001$); and previous step length ($F_{2,80} = 63.767$, $p < 0.001$).

ANOVAs showed that groups differed significantly in target step velocity ($F_{2,\ 40} = 10.649$, $p < 0.001$, $\eta^2 = 0.392$), previous step velocity ($F_{2,40} = 7.114$, $p = 0.002$, $\eta^2 = 0.277$), trailing leg stance time ($F_{2,40} = 6.850$, $p = 0.003$, $\eta^2 = 0.163$), leading leg stance time ($F_{2,40} = 6.953$, $p = 0.003$, $\eta^2 = 0.172$), DS time ($F_{2,40} = 9.293$, $p < 0.001$, $\eta^2 = 0.314$), previous step length ($F_{2,40} = 3.001$, $p = 0.047$, $\eta^2 = 0.133$), and target step length ($F_{2,40} = 4.512$, $p = 0.017$, $\eta^2 = 0.189$).

Post hoc tests between groups showed that compared to young adults and healthy older adults, older adults with diabetes had significantly reduced target step velocity ($p < 0.001$ and $p = 0.004$), reduced previous step velocity ($p = 0.002$ and $p = 0.047$), increased trailing leg stance time ($p = 0.004$ and $p = 0.016$), increased leading leg stance time ($p = 0.003$ and $p = 0.018$), and increased DS time ($p = 0.001$ and $p = 0.003$).

Previous and target step lengths in older adults with diabetes differed from those in young adults ($p = 0.048$ and $p = 0.015$). Older adults with diabetes reduced the velocity of the previous step and the target step significantly (1.05 m/s and 0.88 m/s) and increased the stance time (63.65% and 61.15% of gait cycle), compared with healthy older adults (1.19 m/s and 1.14 m/s; 62.10% and 59.41% of gait cycle) and young adults (1.24 m/s and 1.14 m/s; 62.02% and 59.01% of gait cycle) when the task was to shorten the target step. They also increased the DS time (12.34% of gait cycle), compared with healthy older adults (10% of gait cycle) and young adults (10.22% of gait cycle).

Table 4.3. The comparison of gait and gait adaptability between groups. Mean ± standard deviation (SD) of spatiotemporal parameters in Group I, young adults (n =16); Group II, healthy older adults (n = 14); and Group III, older adults with diabetes (n = 13) in baseline walking and adaptability test conditions.

Variable	Group	Baseline	Walk-through	Step shortening		Step lengthening		Obstacle crossing	
				Previous step	Target step	Previous step	Target step	Previous step	Target step
Velocity (m/s)	I	1.29 ± 0.13	1.27 0.13	1.25 ± 0.12	1.14 ± 0.10 [c,d]	1.28 ± 0.16	1.43 ± 0.15 [c,d]	1.23 ± 0.05 [d]	1.27 ± 0.18
	II	1.25 ± 0.10	1.21 ± 0.11	1.19 ± 0.14 [d]	1.14 ± 0.18 [c,d]	1.24 ± 0.18 [d]	1.47 ± 0.26 [c,d]	1.21 ± 0.26 [d]	1.28 ± 0.2 [d]
	III	1.22 ± 0.12	1.08 ± 0.45 [b,d]	1.05 ± 0.14 [a,b,c,d]	0.88 ± 0.17 [a,b,c,d]	1.11 ± 0.15 [a,b,d]	1.25 ± 0.22 [a,b,d]	1.04 ± 0.28 [a,b,d]	1.01 ± 0.19 [a,b,d]
Stance (%gait)	I	60.71 ± 1.32	60.78 ± 1.04	62.02 ± 1.78	59.01 ± 1.27 [d]	59.22 ± 1.54	59.01 ± 1.27 [d]	60.12 ± 1.72	59.90 ± 1.41 [c,d]
	II	60.87 ± 1.61	61.32 ± 1.37	62.10 ± 1.69	59.41 ± 1.84 [d]	59.75 ± 2.31	59.41 ± 2.84 [d]	60.39 ± 1.73	60.02 ± 1.60 [c,d]
	III	61.83 ± 1.81	64.0 ± 1.2 [a,b,d]	63.65 ± 3.57 [a,b,d]	61.15 ± 3.16 [a,b,d]	60.88 ± 2.11 [c]	61.15 ± 3.16 [b,c]	61.42 ± 1.91 [c]	62.22 ± 2.39 [a,b,c]
Swing (% gait)	I	39.05 ± 1.2	39.43 ± 1.98	-	39.29 ± 1.91	-	43.24 ± 1.90 [c,d]	-	41.04 ± 1.34 [c,d]
	II	38.10 ± 1.15	38.98 ± 1.80	-	40.20 ± 1.89	-	43.62 ± 2.64 [c,d]	-	41.16 ± 1.17 [c,d]
	III	38.43 ± 1.79	37.28 ± 2.02 [b]	-	40.11 ± 3.16 [c,d]	-	41.39 ± 2.06 [b,c,d]	-	41.94 ± 1.95 [c,d]
Double Support (% gait)	I	10.81 ± 1.1	10.42 ± 1.17	-	10.22 ± 0.88	-	9.51 ± 0.79 [c,d]	-	10.03 ± 1.13
	II	10.83 ± 1.09	11.17 ± 1.46	-	10.00 ± 1.43	-	9.87 ± 1.66 [c,d]	-	10.04 ± 1.69 [c]
	III	11.80 ± 1.66	12.03 ± 2.72 [a,b,d]	-	12.34 ± 2.24 [a,b]	-	12.26 ± 1.70 [a,b]	-	11.61 ± 3.79 [a,b]
Step Length (% leg length)	I	78.03 ± 2.99	75.28 ± 4.57 [d]	70.89 ± 6.53 [c,d]	59.78 ± 6.9 [c,d]	78.61 ± 5.77 [c]	98.03 ± 5.77 [c,d]	72.73 ± 4.16 [d]	78.22 ± 5.62 [c]
	II	77.37 ± 6.33	73.49 ± 6.67 [d]	66.39 ± 4.48 [c,d]	60.97 ± 4.72 [c,d]	77.92 ± 7.01 [c]	97.2 ± 6.11 [c,d]	69.66 ± 11.62 [d]	80.87 ± 8.54 [c,d]
	III	73.21 ± 6.99 [b]	70.26 ± 5.28 [b,d]	67.39 ± 4.09 [b,c,d]	66.88 ± 7.29 [a,b, c,d]	79.09 ± 7.19 [c]	95.77 ± 7.2 [c,d]	69.19 ± 12.27 [d]	78.21 ± 9.64 [c,d]

[a] Significantly different from the healthy older group ($p < 0.05$).

[b] Significantly different from the young group ($p < 0.05$).

[c] Significantly different from the walk-though condition ($p < 0.05$).

[d] Significantly different from the baseline condition ($p < 0.05$).

Step lengthening condition

Conditions were significantly different in: target step velocity ($F_{1.562,\ 62.479} = 44.078$, $p < 0.001$), trailing leg stance time ($F_{2,80} = 21.245$, $p < 0.001$), leading leg stance time ($F_{1.35,54.112} = 4.395$, $p = 0.027$), swing time ($F_{1.592,63.787} = 66.114$, $p < 0.001$), DS ($F_{2,80} = 22.306$, $p < 0.001$), previous step length ($F_{1.868,74.735} = 16.537$, $p < 0.001$), and target step length ($F_{1.442,57.672} = 452.610$, $p < 0.001$).

ANOVAs showed significant differences between groups in previous step velocity ($F_{2,\ 40} = 5.849$, $p = 0.006$, $\eta^2 = 0.181$), target step velocity ($F_{2,\ 40} = 5.593$, $p = 0.007$, $\eta^2 = 0.182$), trailing leg stance time ($F_{2,40} = 59.089$, $p = 0.004$, $\eta^2 = 0.125$), swing time ($F_{2,40} = 3.886$, $p = 0.029$, $\eta^2 = 0.052$), and DS time ($F_{2,40} = 8.268$, $p = 0.001$, $\eta^2 = 0.228$). Post hoc tests between groups showed that compared to young adults and healthy older adults, older adults with diabetes had significantly reduced target step velocity ($p = 0.005$ and $p = 0.036$), reduced previous step velocity ($p = 0.002$ and $p = 0.007$), increased trailing leg stance ($p = 0.004$ and $p = 0.036$), and increased DS time ($p = 0.001$ and $p = 0.012$).

Obstacle crossing condition

Conditions were significantly different in: previous step velocity ($F_{1.734,\ 69.365} = 17.104$, $p < 0.001$); target step velocity ($F_{1.603,\ 64.127} = 6.669$, $p = 0.004$); trailing leg stance ($F_{2,\ 80} = 9.408$, $p < 0.001$); leading leg stance ($F_{1.686,\ 67.452} = 12.221$, $p < 0.001$); swing time ($F_{2,\ 80} = 39.644$, $p < 0.001$); DS time ($F_{2,\ 80} = 15.014$, $p < 0.001$); target step length ($F_{1.892,\ 75.687} = 42.007$, $p < 0.001$), and previous step length ($F_{1.204,\ 48.152} = 8.178$, $p = 0.004$).

The main effects of groups were significant in previous step velocity ($F_{2,\ 40} = 6.276$, $p = 0.004$, $\eta^2 = 0.220$), target step velocity ($F_{2,\ 40} = 7.630$, $p = 0.002$, $\eta^2 = 0.305$), trailing leg

stance ($F_{2, 40}$ = 8.772, p = 0.001, η^2 = 0.262), and DS time ($F_{2, 40}$ = 7.128, p = 0.002, η^2 = 0.191). Previous step velocity in older adults with diabetes was different from other groups (p = 0.004 and p = 0.009, η^2 = 0.220).

Target step velocity was also different compared with young adults (p = 0.002) and healthy older adults (p = 0.013). Trailing leg stance time and DS time in older adults with diabetes were different from those in young adults (p = 0.001 and p = 0.003) and healthy older adults (p = 0.007 and p = 0.015).

Foot displacement adaptation

Seven trials in which long step length targets were presented and four trials in which the obstacle was presented were repeated. Two healthy older adults and five older adults with diabetes added one step between and changed their leg in response to a LSL target (Figure 4.1). One healthy older adult and three older adults with diabetes also used the same strategy to cross the obstacle. These trials were considered failed trials, and were repeated at the end of the session.

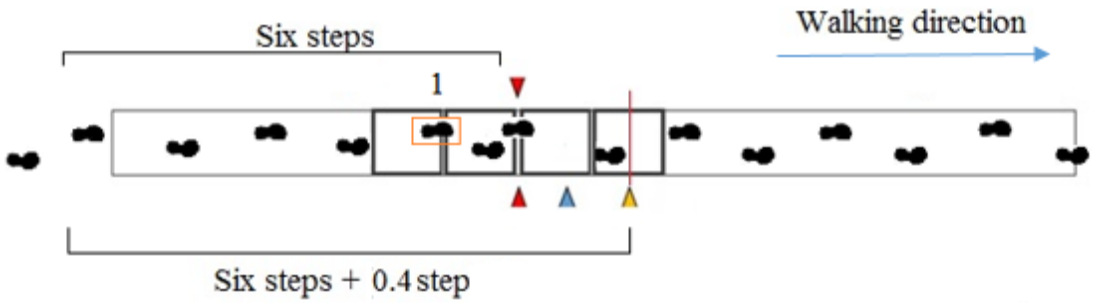


Figure 4.1. A typical older participant who had a failed trial in response to a long step length target. When the participant walked for five steps (1), the long step target (the red line) appeared. The participant shortened the next step and added an extra step before lengthening their step to match the toe marker on the tip of the shoe with the laser line.

In response to SSL targets, 75% of young adults, 67.2% of healthy older adults, and 63.8% of older adults with diabetes overshot steps. Their adapted shorter steps were longer than presented SSL targets, so the constant errors were positive. On the contrary, when LSL targets were presented, 43.8% of young adults, 21.4% of healthy older adults, and 15.4% of older adults with diabetes overshot their steps. The majority of older adults with and without diabetes could not increase their step lengths to match them with projected LSL targets, so their adapted steps were shorter than step targets.

The majority of participants increased their target step lengths when they saw that the obstacle was triggered. In the crossing step called the target step, all participants except for one young adult, four healthy older adults, and three older adults with diabetes shortened their step lengths compared with previous steps and crossed the obstacle.

Some participants touched the obstacle without any requirement to use recovery strategies (Eng et al., 1994) because the fine cord of the obstacle was detached. Participants did not notice that they had touched the obstacle until they passed the line at the end of the walkway. One young adult, two healthy older adults, and six older adults with diabetes touched the obstacle once. One healthy older adult and four older adults with diabetes touched the obstacle twice. Although participants touched the obstacle a few times, data of all trials for the obstacle crossing condition were considered unless participants added one extra step before crossing the obstacle.

Mean absolute errors in response to SSL and LSL targets were different between groups ($H(2) = 25.053$, $p = 0.000004$, $\eta^2 = 0.602$ and $H(2) = 24.134$, $p = 0.000006$, $\eta^2 = 0.589$). As Table 4.4 shows, absolute errors were different between Group I and Group III (U =

3, p = 0.000009 and U = 5, p = 0.000014) and Group II and Group III (U = 7.5, p = 0.000009 and U = 6, p = 0.000014).

Table 4.4. Sagittal foot displacement adaptation absolute errors in three groups of participants. Group I, young adults (n =16); Group II, healthy older adults (n = 14); and Group III, older adults with diabetes (n = 13). Mean ± standard deviation of absolute errors are presented.

	Step length error (cm)		Toe-obstacle vertical distance (cm)		Toe-obstacle horizontal distance (cm)	
	Shortening	Lengthening	Leading	Trailing	Leading	Trailing
Group I	2.34 ± 0.45	1.58 ± 0.56	7.95 ± 2.86	10.53 ± 4.29	13.45 ± 8.31	14.12 ± 6.52
Group II	2.58 ± 0.52	1.75 ± 0.54	7.79 ± 2.01	12.39 ± 5.18	15.26 ± 7.19	13.34 ± 7.12
Group III	4.69 ± 1.41 [a,b]	3.95 ± 1.44 [a,b]	5.46 ± 3.39 [a,b]	12.86 ± 6.33	15.87 ± 8.27	12.21 ± 8.34

[a] Significantly different from the healthy older group ($p < 0.05$).

[b] Significantly different from the young group ($p < 0.05$).

Between groups, only toe-obstacle vertical distances (Table 4.4) were significantly different (H (2) = 10.035, $p = 0.007$, $\eta^2 = 0.195$). The mean toe-obstacle vertical distance of Group III was significantly shorter than those of both Group I (U = 44, $p = 0.007$) and Group II (U = 33, $p = 0.004$).

4.4. Biofeedback effects on foot displacement adaptation during treadmill walking

All groups of participants walked with similar velocities and step lengths on the treadmill. Group effect was $F_{2, 40} = 2.653$, $p = 0.083$, $\eta^2 = 0.117$ for walking velocities, and $F_{2, 40} = 1.964$, $p = 0.154$, $\eta^2 = 0.089$ for step lengths.

Walking velocities and step lengths were 1.08 m/s (SD = 0.12 m/s) and 67.56% of leg length (SD = 6.74% of leg length) in young adults, 1.08 m/s (SD = 0.16 m/s) and 65.56%

of leg length (SD = 9.54% of leg length) in healthy older adults, and 0.98 m/s (SD = 0.11 m/s) and 60.9 % of leg length (SD = 11.45 % of leg length) in older adults with diabetes.

In baseline walking, step length targets were not different between groups. The mean baseline SL standard deviations were not significantly different between groups ($F_{2, 40}$ = 0.026, p = 0.974, η^2 = 0.001). For each group, step length targets were on average 7.2 cm shorter and longer than each person's mean baseline SL. Because baseline SL means and SL standard deviations were not different between groups, step length targets were not different between groups.

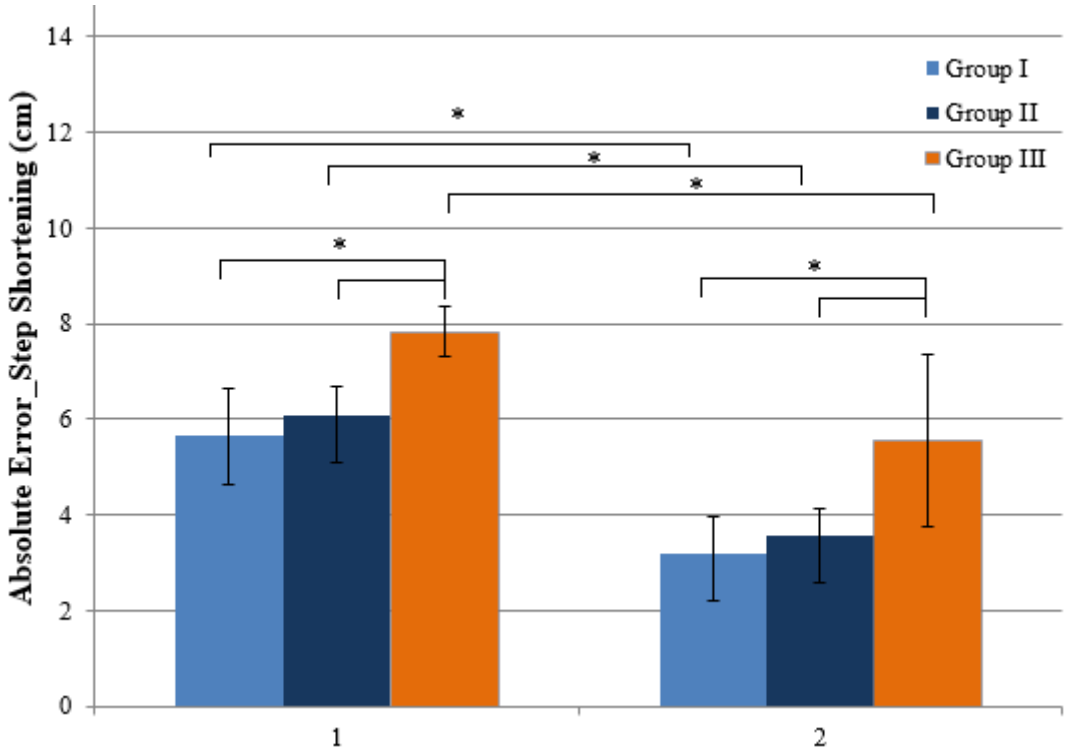
In adaptability tests when a SSL or LSL target (step shortening/lengthening) was presented unexpectedly, errors of the first responses were larger because biofeedback was unavailable, but errors reduced in the following attempts when participants could use biofeedback (visual information about their SL adaptation), with average errors of above zero in all participants in the three groups.

Step shortening

As Figure 4.2 shows, all three groups used visual feedback during online correction of step shortening to reduce the differences between real step lengths and desired short step lengths in reactive responses when targets were suddenly presented on the monitor, as shown in Figure 4.2 (Z = -3.516, p = 0.0004 in Group I, Z = -3.296, p = 0.001 in Group II, and Z = -2.970, p = 0.003 in Group III).

Groups' SL absolute errors were different significantly in reactive responses (H (2) = 11.457, p = 0.003) and planned responses with biofeedback (H (2) = 15.467, p = 0.0004). Absolute errors of step adaptation during step shortening were different between Group I

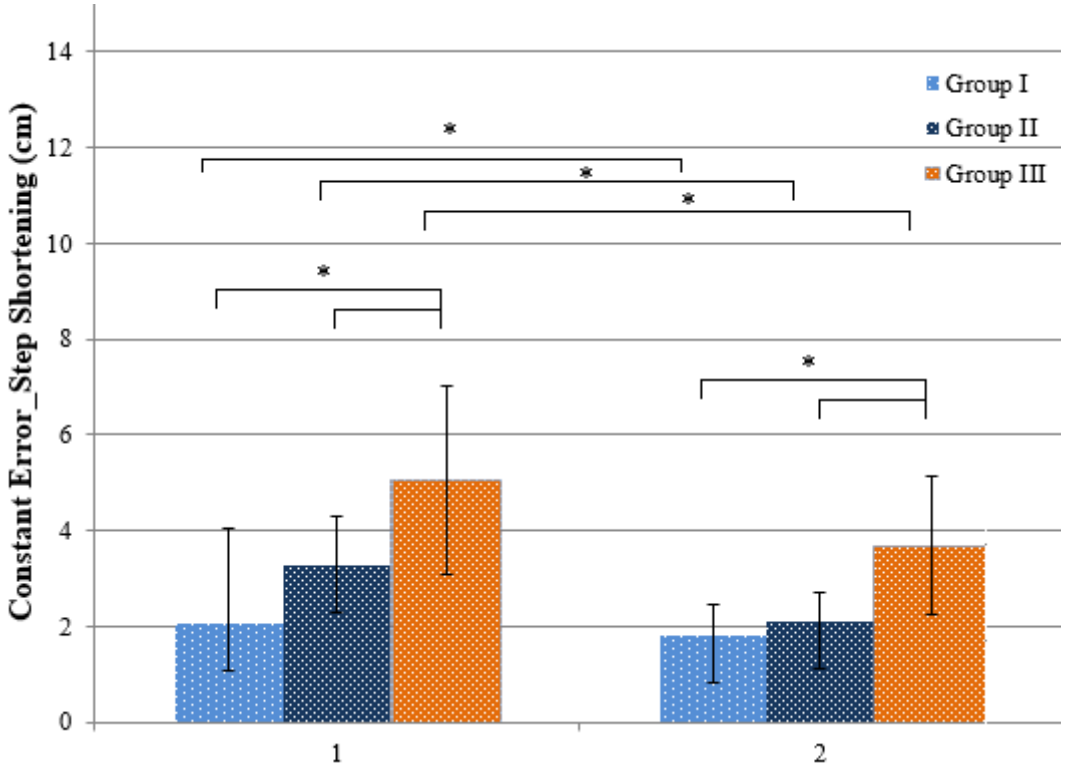
and Group III (U = 37, p = 0.003 and U = 21.5, p = 0.0003) and Group II and Group III (U = 31, p = 0.003 and U = 31, p = 0.004).



* Significant difference (p < 0.05)

Figure 4.2. Effects of biofeedback on step length absolute errors (mean and standard deviation) during walking with shorter steps. Group I (young adults), Group II (healthy older adults), and Group III (older adults with diabetes) were without biofeedback (1) and with biofeedback (2). Participants in each group significantly reduced absolute errors when they received biofeedback.

Constant errors for step adaptation during step shortening with biofeedback also showed a reduction compared with the sudden response to the task without biofeedback (Figure 4.3). The mean constant errors reduced from 2.06 to 1.81 cm in Group I (Z = -3.361, p = 0.001), from 3.28 cm to 2.10 cm in Group II (Z = -3.180, p = 0.001), and from 5.05 cm to 3.69 cm in Group III (Z = -2.411, p = 0.016).



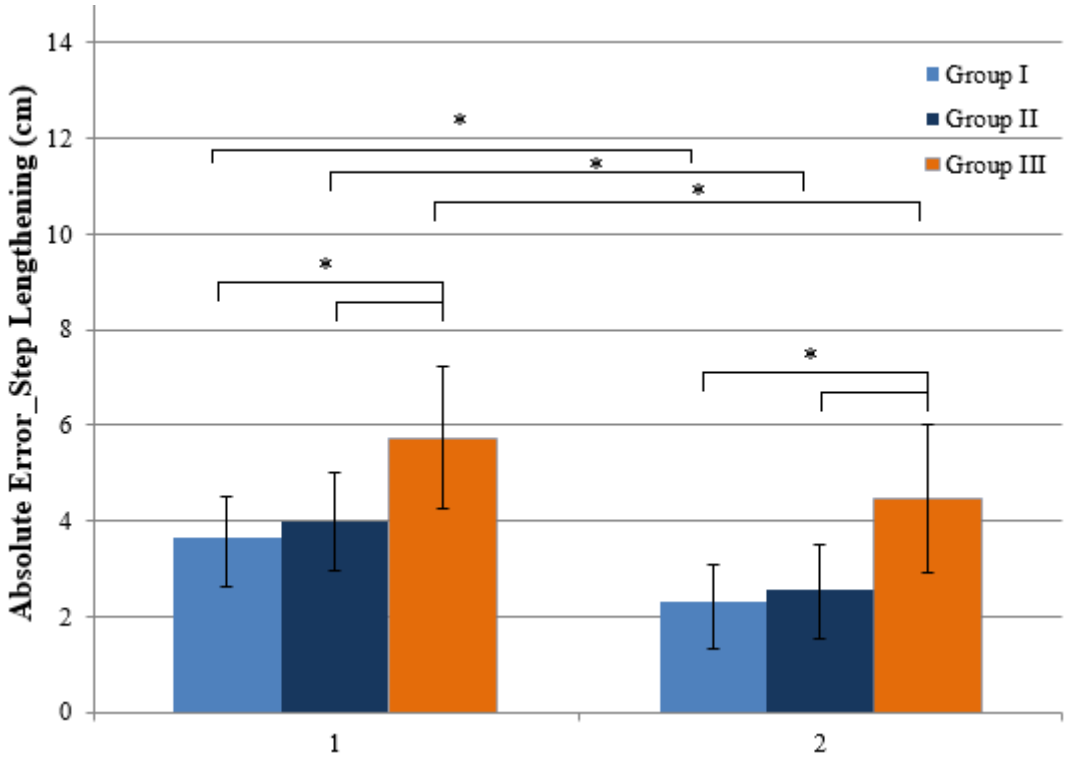
* Significant difference ($p < 0.05$)

Figure 4.3. Effects of biofeedback on step length constant errors (mean and standard deviation) during walking with shorter steps. Group I (young adults), Group II (healthy older adults), and Group III (older adults with diabetes) were without biofeedback (1) and with biofeedback (2). Participants in each group significantly reduced absolute errors when they received biofeedback.

Groups' step shortening constant errors were different without and with biofeedback (H (2) = 11.061, $p = 0.004$, $\eta^2 = 0.373$ and H (2) = 12.931, $p = 0.002$, $\eta^2 = 0.459$). Constant errors were in Group III showed significantly higher than those in Group I (U = 37, $p = 0.003$ and U = 21.5, $p = 0.0002$) and Group II (U = 32, $p = 0.004$ and U = 31, $p = 0.004$).

Step lengthening

Figure 4.4 presents step length absolute errors in each group in reactive responses to the presentation of long step length targets (1) and corrected reactive responses using biofeedback (2).



* Significant difference ($p < 0.05$)

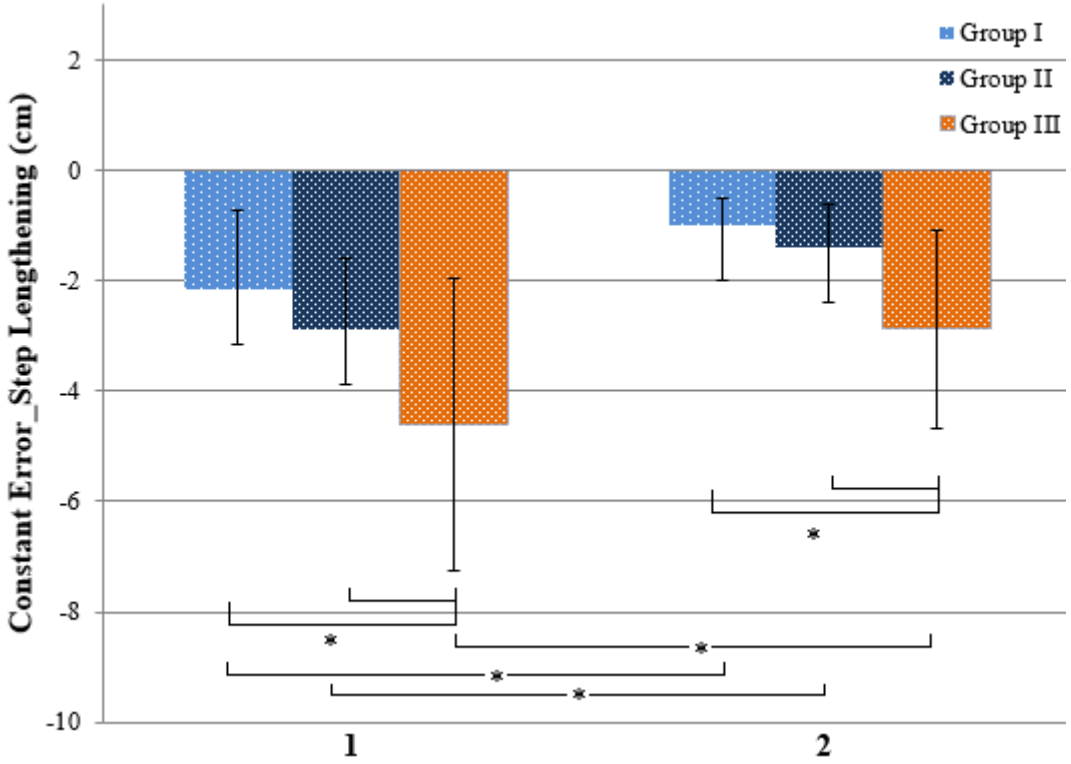
Figure 4.4. Effects of biofeedback on step length absolute errors (mean and standard deviation) during walking with longer steps. Group I (young adults), Group II (healthy older adults), and Group III (older adults with diabetes) were without biofeedback (1) and with biofeedback (2). Participants in each group significantly reduced absolute errors when they received biofeedback.

In response to biofeedback, participants in all groups adjusted their step lengths more accurately when presented step length targets stayed on, resulting in reduced mean

absolute errors (Group I, Z = -3.517, p = 0.0004; Group II, Z = -2.480, p = 0.013; and Group III, Z = -2.411, p = 0.016).

Groups' absolute errors for step adaptation during step lengthening were different with and without biofeedback (H (2) = 13.201, p = 0.001, η^2 = 0.400 and H (2) = 18.749, p = 0.00009, η^2 = 0.443). Mean errors were different between Group I and Group III (U = 30, p = 0.001 and U = 13, p = 0.0006) and Group II and Group III (U = 36, p = 0.008 and U = 20, p = 0.001).

As Figure 4.5 shows, step length constant errors also reduced during online correction of reactive responses.



* Significant difference (p < 0.05)

Figure 4.5. Effects of biofeedback on step length constant errors (mean and standard deviation) during walking with longer steps. Group I (young adults), Group II (healthy older adults), and Group III (older adults with diabetes) were without biofeedback (1) and

with biofeedback (2). Participants in each group significantly reduced constant errors when they received biofeedback.

All groups reduced the constant errors when they received biofeedback. The mean error significantly reduced in Group I ($Z = -2.793$, $p = 0.005$), Group II ($Z = -2.166$, $p = 0.03$) and Group III ($Z = -2.970$, $p = 0.003$).

Groups' mean constant errors were different between groups in the step lengthening conditions with and without biofeedback ($H(2) = 15.778$, $p = 0.003$, $\eta^2 = 0.247$ and $H(2) = 14.998$, $p = 0.001$, $\eta^2 = 0.284$). In both with and without biofeedback conditions, mean constant errors in Group III were higher than those in Group I ($U = 28$, $p = 0.001$ and $U = 27$, $p = 0.001$) and Group II ($U = 26$, $p = 0.002$ and $U = 30$, $p = 0.003$).

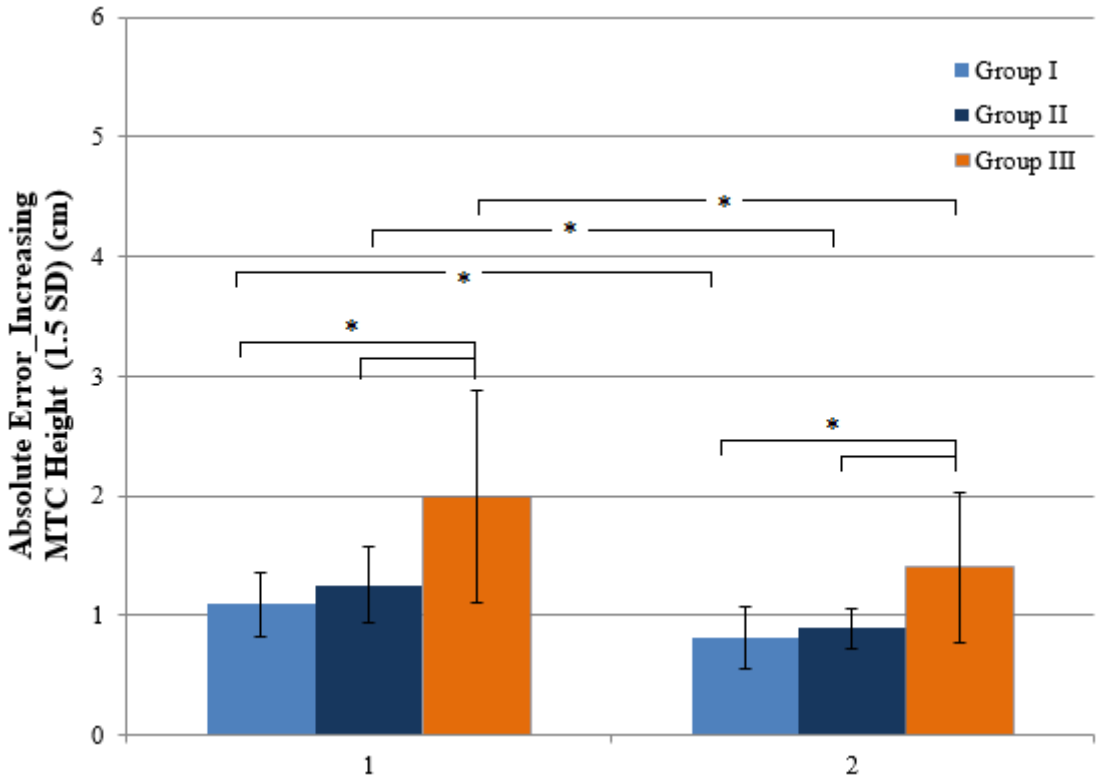
Biofeedback effects on the accuracy of vertical foot displacement

The increases of baseline MTC heights to determine high and higher MTC targets in MTC adaptation tests showed no significant differences between groups ($F_{2, 40} = 0.061$, $p = 0.941$, $\eta^2 = 0.056$). The means and standard deviations were 2.36 cm and 0.44 cm in Group I, 2.30 cm and 0.48 cm in Group II and 2.32 cm and 0.48 cm in Group III. As a result, participants had to adapt their MTC heights with high and higher MTC targets that were on average 3.5 cm and 5.8 cm higher than baseline MTC heights.

Walking with high MTC targets

As Figure 4.6 shows, all groups used biofeedback and improved the adaptation of their MTC heights with displayed MTC targets ($Z = -2.870$, $p = 0.004$ in Group I, $Z = -3.297$, $p = 0.001$ in Group II and $Z = -2.760$, $p = 0.006$ in Group III).

Groups were different in absolute error MTC height means in response to high MTC targets with and without biofeedback (H (2) = 15.599, $p = 0.0004$, $\eta^2 = 0.350$ and H (2) = 11.081, $p = 0.004$, $\eta^2 = 0.309$). Mean errors were different between Group I and Group III (U = 22, $p = 0.0003$ and U = 37.5, $p = 0.004$) and Group II and Group III (U = 31, $p = 0.004$ and U = 42, $p = 0.017$).

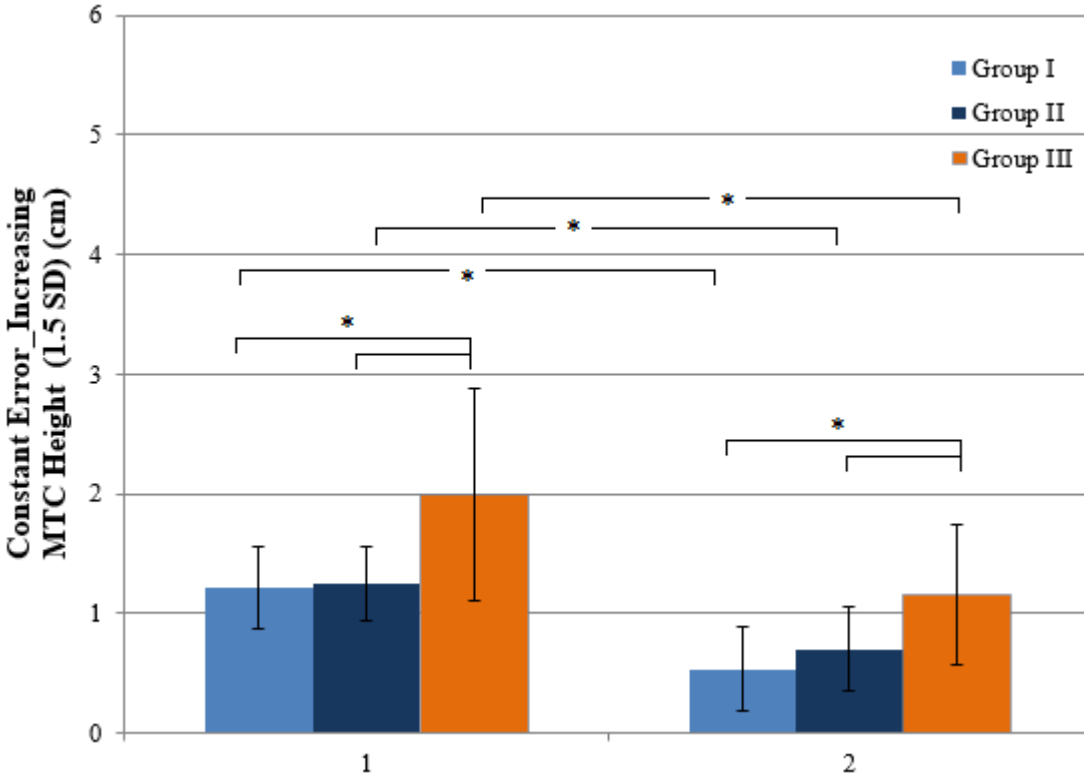


* Significant difference ($p < 0.05$)

Figure 4.6. Effects of biofeedback on minimum toe clearance (MTC) absolute errors (mean and standard deviation) during walking with high MTC targets. Group I (young adults), Group II (healthy older adults), and Group III (older adults with diabetes) were without biofeedback (1) and with biofeedback (2). Participants in each group significantly reduced absolute errors when they received biofeedback.

Constant errors for MTCs in all groups in response to the sudden appearance of high MTC targets significantly reduced during online correction (Z = -3.516, $p = 0.0004$ in Group I, Z = -3.297, $p = 0.001$ in Group II and Z = -3.110, $p = 0.002$ in Group III) (see Figure 4.7).

Groups were significantly different in constant MTC errors (H (2) = 11.187, p = 0.004, η^2 = 0.295 and H (2) = 11.813, p = 0.003, η^2 = 0.269). In both with and without biofeedback conditions, mean constant errors in Group III were significantly higher than those in Group I (U = 38, p = 0.004 and U = 30, p = 0.001) and Group II (U = 31, p = 0.004 and U = 51.5, p = 0.047).



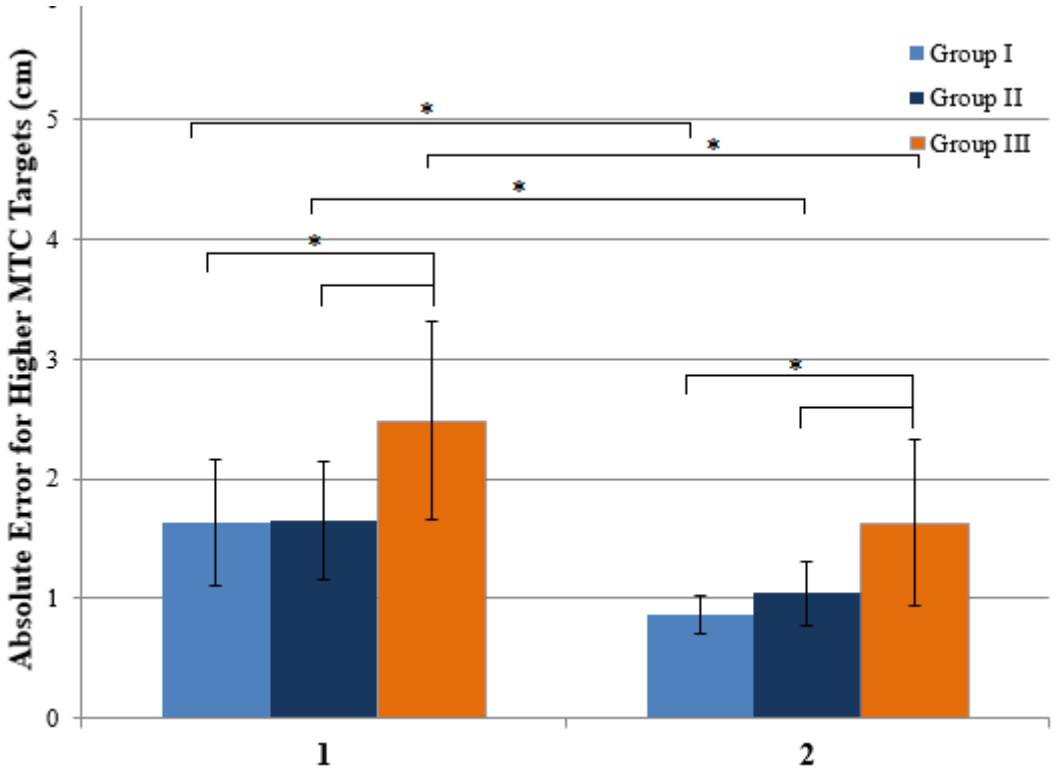
* Significant difference (p < 0.05)

Figure 4.7. Effects of biofeedback on minimum toe clearance (MTC) constant errors (mean and standard deviation) during walking with high MTC targets. Group I (young adults), Group II (healthy older adults), Group III (older adults with diabetes) were without biofeedback (1) and with biofeedback (2). Participants in each group significantly reduced constant errors when they received biofeedback.

Walking with higher MTC targets

Participants in all groups used biofeedback and significantly reduced the absolute MTC errors during online correction as Figure 4.8 shows (Z = -3.517, p = 0.0004 in Group I, Z

= -3.234, $p = 0.001$ in Group II, Z = -2.691, $p = 0.007$ in Group III). Mean absolute errors were also different between groups with and without biofeedback (H (2) = 10.846, $p = 0.004$ and H (2) = 12.563, $p = 0.002$), with significant differences between Group I and Group III (U = 35, $p = 0.002$ and U = 33, $p = 0.002$) and Group II and Group III (U = 35.5, $p = 0.007$ and U = 41.5, $p = 0.016$).

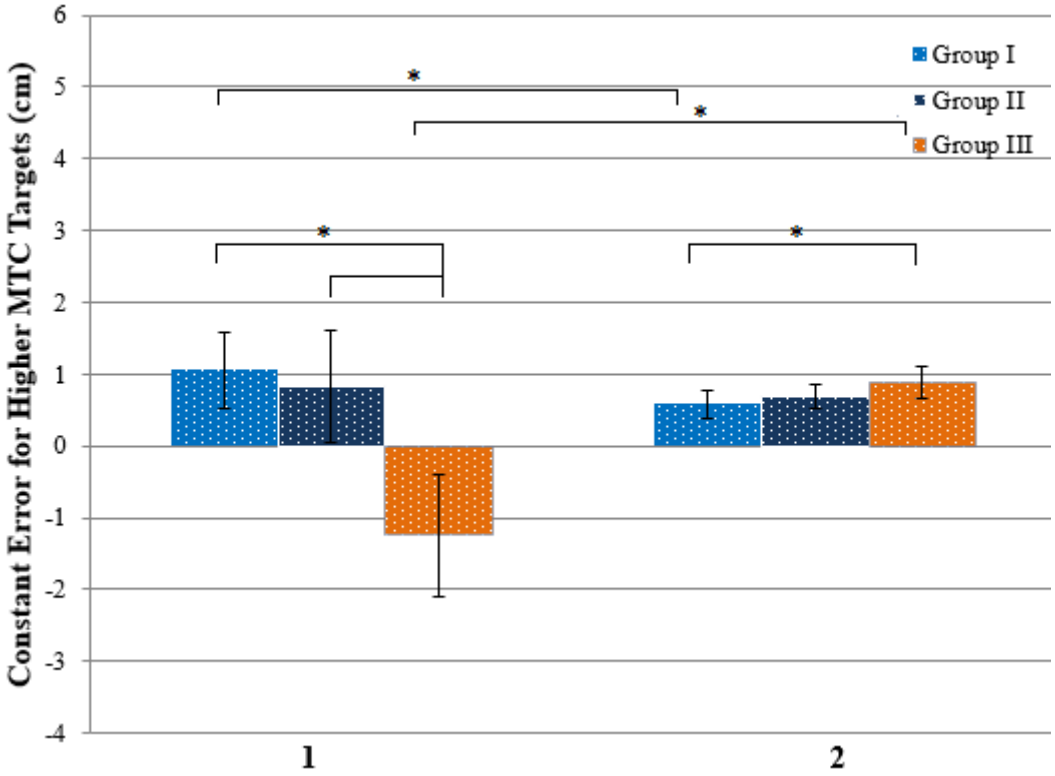


* Significant difference ($p < 0.05$)

Figure 4.8. Effects of biofeedback on minimum toe clearance (MTC) absolute errors (mean and standard deviation) during walking with higher MTC targets. Group I (young adults), Group II (healthy older adults), and Group III (older adults with diabetes) were without biofeedback (1) and with biofeedback (2). Participants in each group significantly reduced absolute errors when they received biofeedback.

In the without biofeedback condition, the mean error in Group III was negative. However, mean errors in Group I and Group II were positive, as shown in Figure 4.9.

In the with biofeedback condition, participants in Group III had positive errors like participants in Group I and Group II (Figure 4.9, with biofeedback (2)).



* Significant difference ($p < 0.05$)

Figure 4.9. Effects of biofeedback on minimum toe clearance (MTC) constant errors (mean and standard deviation) during walking with higher MTC targets. Group I (young adults), Group II (healthy older adults), and Group III (older adults with diabetes) were without biofeedback (1) and with biofeedback (2). Participants in each group significantly reduced absolute errors when they received biofeedback.

Participants in Group I and Group II used biofeedback and reduced mean constant errors ($Z = -2.585$, $p = 0.01$ in Group I, $Z = -0.659$, $p = 0.510$ in Group II, $Z = -3.181$, $p = 0.001$ in Group III). Groups' mean constant errors were different with and without biofeedback ($H(2) = 11.187$, $p = 0.004$ and $H(2) = 11.813$, $p = 0.003$), with significant differences between Group I and Group III ($U = 38$, $p = 0.004$ and $U = 30$, $p = 0.001$) and Group II

and Group III, only when biofeedback was unavailable (U = 31, $p = 0.004$). The group effect was significant ($F_{2, 40} = 31.212$, $p < 0.001$). Constant errors were different between groups I and III ($p < 0.001$), and II and III ($p < 0.001$).

4.5. Agreement between overground and treadmill gait adaptability tests

In each participant, absolute error means for SL were in agreement between the OGA and TGA tests when responding to the sudden appearance of SSL targets. Figure 4.10 shows that 96% of points were placed between the limits of agreement. The mean for adapted short step length biases (the differences between errors measured by two tests) was 2.37 cm (SD = 1.91 cm).

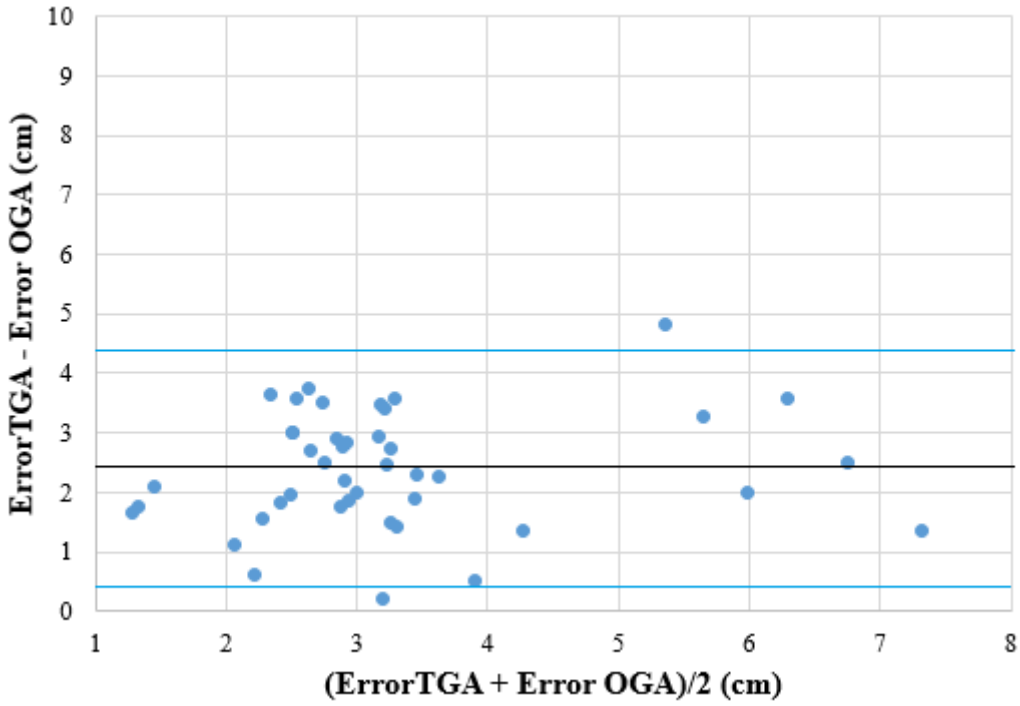


Figure 4.10. A scatter plot to investigate agreement between treadmill (TGA) and overground (OGA) tests in response to a sudden shortening of step length. The horizontal black line shows the mean of the differences (bias) between the two tests, and the blue horizontal lines show the upper and lower limits of agreement (bias ± 1.96 × SD).

Step length absolute error means in response to the sudden appearance of LSL targets in the OGA and TGA tests were in agreement, because 98% of points (see Figure 4.11) were placed between the limits of agreement. The mean for adapted long step length biases was 2.44 cm (SD = 1.35 cm).

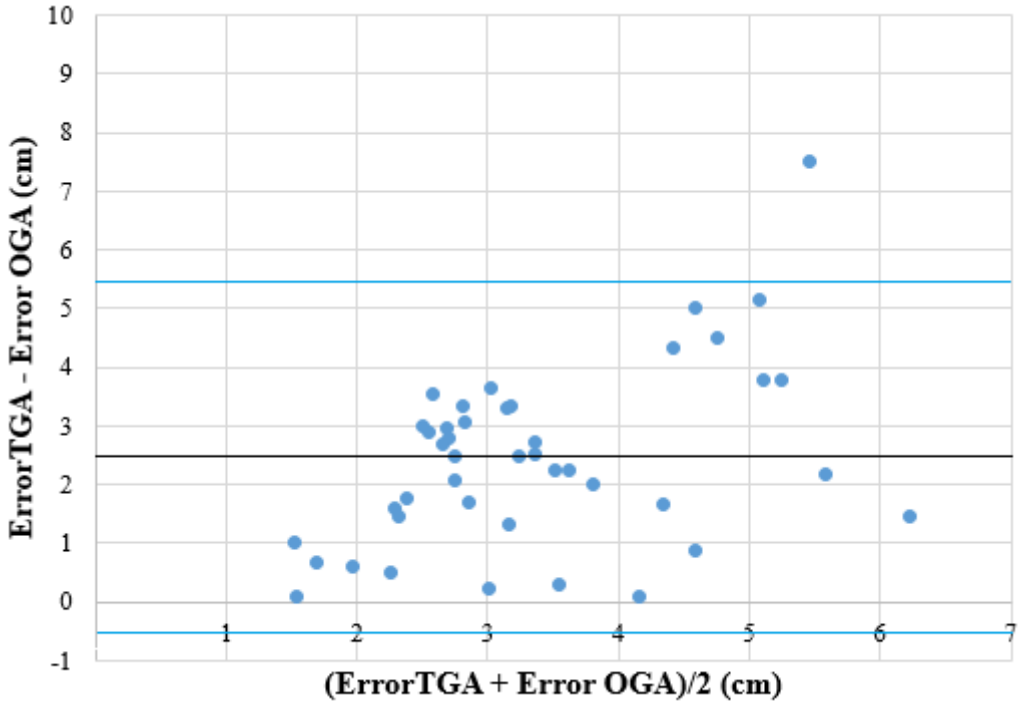


Figure 4.11. A scatter plot to investigate agreement between treadmill (TGA) and overground (OGA) tests in response to a sudden lengthening of step length. The horizontal black line shows the mean of the differences (bias) between the two tests, and the blue horizontal lines show the upper and lower limits of agreement (bias ± 1.96 × SD).

MTC absolute error means in response to high MTC targets during treadmill walking and the distances between MTCs and the obstacle in the obstacle crossing condition were in agreement, because only one point was out of the limits of the agreement (Figure 4.12). The mean for adapted MTC biases was 2.57 cm (SD = 1.76 cm).

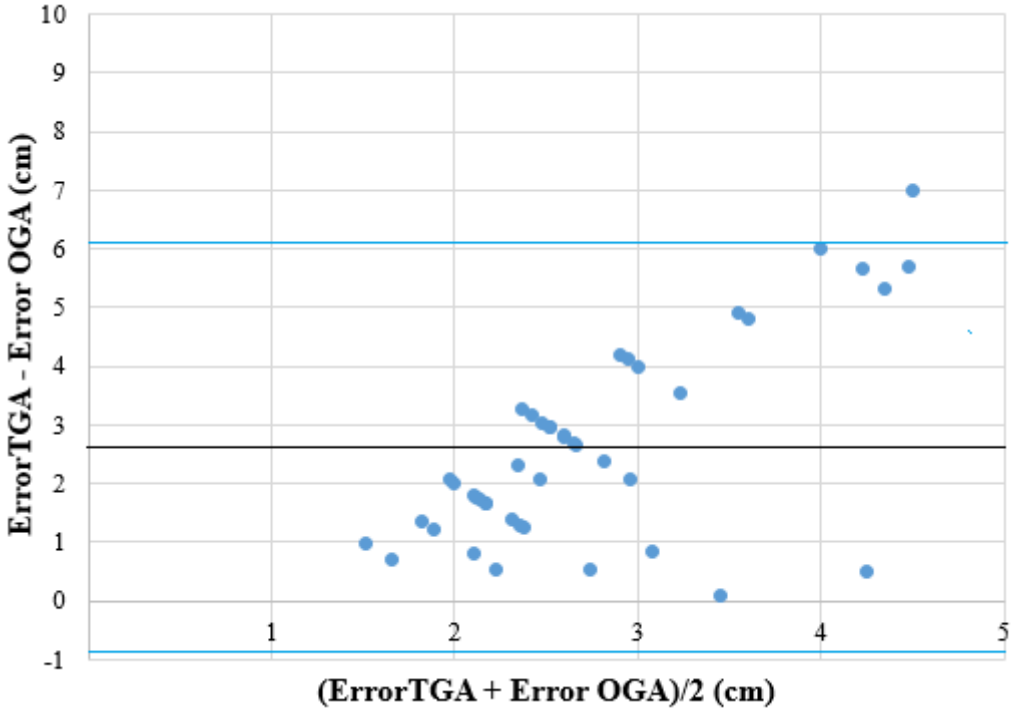


Figure 4.12. A scatter plot to investigate agreement between treadmill (TGA) and overground (OGA) tests in response to a sudden appearance of high MTC targets and the obstacle. The horizontal black line shows the mean of the differences (bias) between the two tests, and the blue horizontal lines show the upper and lower limits of agreement (bias ± 1.96 × SD).

MTC absolute error means in response to higher MTC targets during treadmill walking and the distances between MTCs and the obstacle during overground walking were not in agreement because less than 95% points were between the limits of the agreement (Figure 4.13). The mean for adapted MTC biases was 1.95 cm (SD = 1.35 cm).

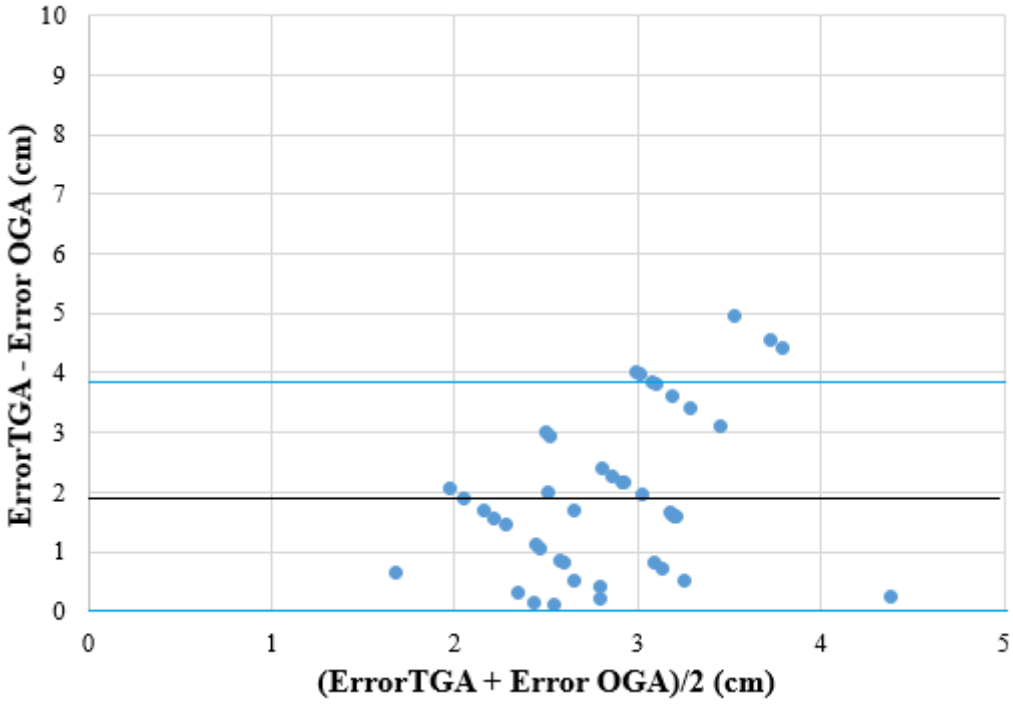


Figure 4.13. A scatter plot to investigate agreement between the treadmill (TGA) and overground (OGA) tests in response to higher MTC targets and the obstacle. The horizontal black line shows the mean of the differences (bias) between the two tests, and the orange horizontal lines show the upper and lower limits of agreement (bias ± 1.96 × SD).

4.6. Results summary

It was found out that walking speed and gait spatiotemporal parameters were not different between groups in baseline while walking on an unobstructed smooth surface. However, it was revealed that gait adaptability spatiotemporal parameters, step length errors and toe-obstacle distances were different between groups while responding to the tasks were unexpectedly presented during overground walking (Table 4.5).

Table 4.5. Results summary of the effects of age and diabetes on gait adaptability.

Variable	Condition	Step	Age effects	Diabetes effects	The combination of age and diabetes effects
Velocity	Step shortening	Previous	No	Yes	Yes
		Target	No	Yes	Yes
	Step lengthening	Previous	No	Yes	Yes
		Target	No	Yes	Yes
	Obstacle crossing	Previous	No	Yes	Yes
		Target	No	Yes	Yes
Stance	Step shortening	Previous	No	Yes	Yes
		Target	No	Yes	Yes
	Step lengthening	Previous	No	No	No
		Target	No	No	Yes
	Obstacle crossing	Previous	No	No	No
		Target	No	Yes	Yes
Swing	Step shortening	Previous	-	-	-
		Target	No	No	No
	Step lengthening	Previous	-	-	-
		Target	No	No	Yes
	Obstacle crossing	Previous	-	-	-
		Target	No	No	No
Double Support	Step shortening	Previous	-	-	-
		Target	No	Yes	Yes
	Step lengthening	Previous	-	-	-
		Target	No	Yes	Yes
	Obstacle crossing	Previous	-	-	-
		Target	No	Yes	Yes

Table 4.5. Results summary of the effects of age and diabetes on gait adaptability (Cont.).

Variable	Condition	Step	Age effects	Diabetes effects	The combination of age and diabetes effects
Step Length	Step shortening	Previous	No	No	Yes
		Target	No	Yes	Yes
	Step lengthening	Previous	No	No	No
		Target	No	No	No
	Obstacle crossing	Previous	No	No	No
		Target	No	No	No
Step length error	Step shortening	Previous	-	-	-
		Target	No	Yes	Yes
	Step lengthening	Previous	-	-	-
		Target	No	Yes	Yes
	Toe-obstacle vertical Distance	Previous	No	No	No
		Target	No	Yes	Yes
	Toe-obstacle horizontal distance	Previous	No	No	No
		Target	No	No	No

All participants walker slower on the treadmill with shorter step lengths compared with overground walking. When a task was unexpectedly presented on the monitor during treadmill walking, older adults with diabetes showed reduced step length and minimum toe clearance adjustments compared with young and older adults. All groups were able to using visual feedback and reduced their step length and minimum toe clearance errors during online correction compared with the time that a task was unexpectedly presented.

The overground and treadmill tests were exchangeable to quantify step length and minimum toe clearance errors in response to all targets except for higher MTC targets.

Chapter 5: Discussion

This study describes new gait adaptability tests in which a goal-task (adapting SL and MTC height with targets) was presented to disrupt the gait to investigate gait adaptability and sagittal foot displacement adjustments in older adults with diabetes. Participants were given biofeedback (i.e. how far they were from a presented target) during online correction to accurately adjust foot trajectory during treadmill walking. Both OGA and TGA tests in response to the sudden appearance of step length targets and high MTC targets/obstacles were found to have exchangeable errors for future application of the TGA test for quantifying foot placement errors.

This chapter discusses the findings of Chapter 4, and reviews the limitations of this research study and provides suggestions for future research.

5.1. Baseline gait

The findings supported Hypothesis 1.1 (walking speed and gait spatiotemporal parameter means including stance, swing, DS, and SL would show no difference between groups in baseline), revealing that spatiotemporal gait parameters were not different between groups in unchallenged baseline condition when participants walked for a short time on an unobstructed smooth surface. In line with previous research (Santhiranayagam et al., 2015, Hausdorff et al., 2004, Dingwell et al., 2017, Nagano et al., 2013, Ko et al., 2011), ageing and diabetes did not affect gait characteristics during walking at a preferred speed. In a study reporting gait impairments (Hausdorff et al., 2004), experimental groups were different in characteristics such as history of falls, cognition, and the distribution of gender or walking speed (Ko et al., 2011), which has been suggested to be a strong

predictor of the level of health (Studenski et al., 2011), with older adults who walk slower being less healthy. Even though groups were different in these characteristics, they might not show any significant differences because locomotion is unchallenged, so afferent information is not be required (Dingwell et al., 1999) to update efferent copies of the locomotion in the central nerve system.

In the present study, the older participants with diabetes who did not have apparent evidence of neuropathy demonstrated no altered gait patterns. Furthermore, age did not reduce the speed of walking in older adults. Older women walked more slowly than men, with the age- related speed decline after age 63 accelerating more in men (16.1% per decade) than in women (12.4% per decade) (Himann et al., 1988, Bendall et al., 1989). In the current thesis, older adults with and without diabetes were matched for age, body mass, gender, and height. Normalisation of SL and spatial gait parameters might reduce the effects of differences between heights, which may have influenced outcomes in previous research.

Both older groups included participants who did not have a history of falls or impaired cognition. Older adults were active, healthy, and free from any diagnosed pathological condition that could affect their normal gait except diabetes mellitus; however, half of them were in the early stages of diabetes or looked after themselves very well so they did not develop the full range of diabetic complications.

5.2. Gait adaptability

The results did not support Hypotheses 1.2.1 (gait adaptability spatiotemporal parameter means (step velocity, stance, swing, DS, and SL) would show no difference between groups in OGA tests) and 1.2.2 (Gait adaptability spatiotemporal parameter means (step

velocity, stance, swing, DS, and SL) would show no difference between conditions (Baseline vs. Walkthrough vs. Step shortening/Step lengthening/Obstacle crossing) in OGA tests), revealing significant differences between groups and conditions in adaptability tests. Gait patterns in older adults with diabetes differed from those in other groups when four conditions were presented in the adaptability tests randomly and unexpectedly.

Walk-through condition

In adaptability test conditions, the older adults with diabetes walked carefully even when the task was to walk through without adapting sagittal foot displacement trajectories. In the walk-through conditions, participants reduced their step velocities and walked with shorter steps, longer DS and stance times, which might be interpreted as being ready to suddenly respond to a condition. The walk-through condition might increase the load of attention and as such affect gait compared with the baseline condition, in which participants knew that none of the conditions would be presented while walking. Participants were ready to suddenly change their ongoing gait and find an alternative foot landing position to meet the presented tasks. This may assist in providing extra time for processing an appropriate response to a presented task (step shortening, step lengthening, or obstacle crossing).

Step-shortening condition

Having conservative gait patterns in the walk-through condition of adaptability tests, the older adults with diabetes adapted gait parameters in the previous step in response to the step shortening condition. They reduced velocity, increased the DS time, reduced the previous step length, and then shortened their target step. However, the other groups only

reduced the velocity and length of the previous step. The observed adjustments in temporal gait parameters are in line with the requirements of responding to the task, showing that the step shortening condition perturbed baseline gait parameters. The earlier changes in the previous step in older diabetics were more pronounced, revealing that they were more affected in the adaptability tests.

Although the older adults with diabetes adapted to the previous step, they could not adapt target steps with presented short step lengths accurately. As a small perturbation of foot displacement, shorter or longer step length, applies a modulation of the initial movement, participants were allowed very rapid and automatic (involuntary) adjustments without introducing large increases in the gait cycle with fast muscle activations (Gaveau et al., 2014). Our results of perturbing step lengths (i.e., presenting SSL targets) did not confirm differences in errors between healthy older and young adults, in disagreement with previous research reports (Caetano et al., 2016) in which some of the participants had experienced one or two falls a year before their participation. In this study, fall history might have increased the error of step shortening, which was an exclusion criterion in our research project. Our older participants were physically quite fit and walked as fast as young adults with a similar SL during overground walking. They also had no history of falls.

The errors made by older adults with diabetes in meeting step targets differed from those in healthy older and young adults. In response to SSL targets, they reduced the target step velocity and increased the stance and DS times to be able to shorten target steps according to the presented step lengths. Although they shortened target steps more than they did in baseline and the walk-through condition, target step accuracies were lower than those in

young and healthy older groups. The adaptation of temporal parameters in order to increase response times was inadequate for step shortening of target steps. Diabetes-related effects on balance (Mehdikhani et al., 2014) and gradual changes in the ankle-foot complex (Cheing et al., 2013) may explain their inability to suddenly reduce step lengths accurately while walking.

Step-lengthening condition

Although older adults with diabetes increased the step length of both previous and target steps in the step lengthening condition, they made the largest errors in the step lengthening condition compared with other groups. The healthy older and young groups increased the target step lengths, whereas the older diabetic group increased the swing time of the target step and increased the step lengths of the previous and target steps. To process appropriate responses, older adults with diabetes reduced the velocity of the previous step and increased the stance time of the leading foot. The older with and without diabetes reduced the previous step velocity when they detected the task. Healthy older adults increased the target step velocity to respond to the task; however, older adults with diabetes were unable to increase the velocity of the target step as much as the healthy older adults were. Thus, they had to increase the stance time of the trailing foot to match the toe marker with a laser beam. The older adults with diabetes adapted spatiotemporal parameters in both previous and target steps without being able to lengthen steps accurately.

Obstacle-crossing condition

Although the older adults with diabetes walked more conservatively and had similar crossing speed, they had a larger number of failed trials. This could be related to the complexity of tasks (four conditions) and the presentation of the obstacle across the

walkway, five times for each limb. In previous research (Liu et al., 2010), diabetic participants with a crossing speed as fast as participants without diabetes could fixate their gaze on the obstacle prior to its release to successfully avoid obstacles that were cylindrical and placed in the middle of the walkway. There was also no age difference between the healthy older group and the older diabetic group. The average age was 69 years. It has been suggested that the decline in obstacle avoidance occurs at about 70 years of age (Weerdesteyn et al., 2005a), so it is likely that the effects of diabetes were responsible for the increased rate of failed trials in older adults with diabetes in the present study.

When the obstacle crossing and step lengthening conditions were presented, older adults with diabetes tended to add one short step, and adjust the displacement of the trailing foot in the previous step before responding to the condition in the target step, as shown in Figure 4.1. The addition of one short step has been reported in previous research (Caetano et al., 2016, Chen et al., 1994a, Weerdesteyn et al., 2005a) as a strategy to compensate for age-related reduced lower limb strength (Pijnappels et al., 2008, Ng et al., 2014), increased muscle activation variability (Kang and Dingwell, 2009), difficulties in step lengthening (Mazaheri et al., 2015), impaired depth perception (Menant et al., 2010, Moslemi Haghighi et al., 2015), and reduced balance (Morrison et al., 2012). The extra short step may increase the timing of handling a response to the presented task including adjusting the displacement of the trailing foot on the plane of locomotion to control the trajectory of the leading foot in the sagittal plane. In the current research study, all trials in which participants added one short step were removed to reduce the effect of this compensatory strategy on the results.

Older adults with diabetes showed reduced abilities to respond to the obstacle crossing condition. They reduced step velocities and increased the DS and stance times in response to the obstacle crossing condition. However, these modulations of gait parameters which are consistent with previous research (Fernando et al., 2013) did not stop them touching the obstacle.

Experiments in which virtual targets, either changes in the colour of one of the visually guided cues during treadmill walking (Potocanac et al., 2015) or the appearance of a two dimensional light during overground walking (Caetano et al., 2016), confirmed that ageing reduces the rate of obstacle avoidance success. Although participants increased the swing time of their target steps, older adults might need to increase the swing time more than do young and healthy adults to be able to increase the rate of success, as suggested in previous research (Weerdesteyn et al., 2005b). In this thesis, a real obstacle was used in combination with other conditions in which the prediction of triggering the obstacle was difficult. Participants had to cross the obstacle and could not use any other strategy, so older adults with diabetes touched the obstacle more frequently and had reduced toe-obstacle vertical distances. Trials in which participants touched the obstacle were not repeated to avoid the learning effects that previous research reported (Potocanac et al., 2015). Reduced step velocities in both previous and target steps and increased DS time in older adults with diabetes increased the time of responding to goal tasks compared with other groups, as reported in previous research (Grewal et al., 2012). However, they were unable to accurately respond to the tasks so they touched the obstacle more frequently with a reduced toe-obstacle vertical distance.

When older adults with diabetes were faced with an increased demand of two tasks, they prioritised the walking performance task, which increased cognitive demands causing the toe-obstacle vertical distances to be reduced and failure to modify the ongoing movement of the swing leg, caused them to touch the obstacle. The young and healthy older participants in this study shortened stance time to compensate the interrupted swing phase, whereas older adults with diabetes reduced velocities of target steps which increased the chance of touching the obstacle. The older diabetes group, like other groups, reduced the previous step length when the obstacle crossing condition was presented, and increased the target SL while crossing the obstacle. The older participants with diabetes had to repeat a few trials because they increased the lengths of their previous steps and reduced the lengths of their target steps. Shortening the target step is a more conservative reaction because it increases the horizontal distance between the toe of the trailing foot and obstacle and decreases the distance between the heel of the leading foot and the obstacle in the target step (Chen et al., 1994a). Increasing the SL in the target step has been suggested to be a way to reduce the cost of obstacle avoidance (Weerdesteyn et al., 2005b), a possibility to allow more time to process and perform foot trajectory adjustments related to an obstacle. It is less risky for a self-initiated fall (Chen et al., 1994a), given step lengthening is physically demanding and more difficult to execute. These findings were in contradiction with previous research in older adults (Caetano et al., 2016) because such research did not remove trials in which participants used compensatory strategies to respond to the presented tasks.

Impaired gait adaptability with increased errors of step length adaptation and reduced toe-obstacle vertical distances are consistent with previous research of age- or diabetes-related effects on SL adaptation and poor obstacle avoidance in older adults with diabetes

(Caetano et al., 2016, Liu et al., 2010, Mehdikhani et al., 2020). When the older diabetic participants tried not to use compensatory strategies such as an addition of a short step, they touched the obstacle more often because of decreased toe-obstacle vertical distances. They were unable to lift their feet and increase heights of foot displacements in the vertical direction. Here, the inability to avoid obstacles by step lengthening in the target step may indicate impaired gait adaptability, which might originate in reduced physical fitness (Cheing et al., 2013, Leenders et al., 2013), prioritising postural balance (Bloem et al., 2001), increased ankle dorsiflexion of the trailing foot during the stance phase and compromised distal joint position senses in older adults as a result of long-term diabetes (Liu et al., 2010).

Foot displacement errors

The older adults with diabetes overshot their steps in the step shortening condition and undershot steps in the step lengthening. The majority of older diabetes participants (63.8%) overshot steps in the step shortening condition; however, the minority of older diabetes participants (15.4%) overshot steps in the step lengthening conditions. In response to the step shortening condition, the knee flexion moment required to overcome the inertia of the shank of the leading leg in the swing phase to let the toe marker be matched with laser beams presenting SSL targets. In this situation, because the knee is already in extension, step shortening requires a knee flexion moment that might be inefficient and led to increased errors of step shortening in older adults with diabetes. In line with previous research (Mazaheri et al., 2015), absolute errors of the three groups were greater in the step shortening condition than in the step lengthening condition, when targets unexpectedly appeared. Step lengthening was easier during walking because

maintaining balance was easier during step lengthening with the least change. The direction of walking and the direction of the movement of the foot was the same. However, the interaction of ageing and diabetes reduces the development of hip flexion torque in older adults with diabetes (Balducci et al., 2014). People with diabetes start knee extension late in the gait cycle (Ko et al., 2011), so they have a reduced hip range of motion, altogether reducing the ability to suddenly lengthening step.

Inability to suddenly react to targets and adapt steps resulted in larger stepping errors in the older diabetic group compared with other groups. These were also associated with impaired adaptable gait patterns in older adults with diabetes because they were unable to adapt temporal parameters efficiently in order to accurately match their real step lengths with desired step lengths. The ability to adjust foot landing accurately is required for fall prevention (Geerse et al., 2019).

Similar to other groups, the older diabetes group increased their target step lengths while crossing the obstacle; however, they had reduced toe-obstacle clearance and touched the obstacle more frequently. Previous research reported that both shortening and lengthening the target step were used to avoid obstacles (Weerdesteyn et al., 2003, Chen et al., 1994a). The available response time, stability, and ability for forward propulsion determine whether step lengthening or step shortening of the target step is used (Patla et al., 1999). The older adults with diabetes had two options to successfully cross the obstacle: adding a short step and increasing the vertical height of the leading foot. Because the participants were asked to repeat the trials in which they added one short step, the lack of ability to increase the height of the leading foot was more apparent. Increasing the target step's length revealed that the available time to respond to the obstacle was appropriate.

Only toe-obstacle vertical distances of leading feet were reduced in the older adults with diabetes, which are tripping predictors (Chen et al., 1991). The height and depth of the obstacle might be the reason that toe-obstacle and heel-obstacle horizontal distances were no significantly different between groups, which are inconsistent with previous research (Chen et al., 1991, Lowrey et al., 2007). Reduced toe-obstacle clearance of the leading foot in older adults with diabetes, in line with previous research (Liu et al., 2010), and increased variability of the MTC (Begg et al., 2007, Mills et al., 2008) might be connected with touching the obstacle more often. Even slowing down in response to the sudden appearance of the obstacle did not improve the limb end-point control (i.e., foot) during the swing phase. The effects of the interaction of age and diabetes, decline in balance control, muscle strength, reduced range of motion of the hip, and proprioception (Morrison et al., 2012, Ko et al., 2011, Schwartz et al., 2008) may reduce the accuracy of responses to the obstacle.

Older adults with diabetes tend to prioritise stability rather than respond to tasks. They could not control the trajectory of the foot, so they had increased errors of step length and reduced toe-obstacle vertical distances (Bonnet et al., 2009, Young and Dingwell, 2012). A greater standard deviation of the mean stance time in the older adults with diabetes compared with baseline walking and other groups may indicate instability in the sudden responses to goal tasks. All responses were reactive because participants could see neither the height nor an overground projected laser beam when they stood at the starting point. Participants had to react to a task that was unexpectedly presented and adapt their sagittal foot trajectories accurately. Therefore, gait adaptability tests in this thesis in line with previous tests (Geerse et al., 2019, Mazaheri et al., 2015, Caetano et al., 2016) revealed that older adults with diabetes might have an inability to accurately adjust foot

displacement in the AP and vertical directions, which can increase the risk of colliding with external hazards during walking.

An explanation for increased errors of SL adaptation and reduced toe-obstacle vertical distance in older adults with diabetes is the impaired efferent copies of tasks. If it is assumed that the internal model is similar between groups, but the ankle proprioceptive threshold is necessary for the swing limb positioning to be increased in older adults with diabetes (Richardson et al., 2014), an up weighting of visual compared to proprioceptive feedback will occur in sensorimotor integration and lead to the inaccurate foot displacement adjustments needed for safe walking. In this situation, ankle proprioceptive thresholds might have been altered before neuropathy was clinically detected.

The measurements of foot landing accuracy were predictive of falls in previous research (Mirelman et al., 2012, Herman et al., 2010) and can be improved by gait rehabilitation programs that improve the control of the foot adjustments during the swing phase for increase foot landing accuracy in the AP and vertical directions (Begg et al., 2014b, Mehdikhani et al., 2019, Mirelman et al., 2016).

5.3. Biofeedback effects

All participants walked slower in baseline treadmill walking with shorter step lengths compared with baseline overground walking, which was in line with previous research (Nagano et al., 2013) and without significant differences between groups. The decline in speed is negligible in healthy older adults less than age 70 (Alexander, 1996).

When the goal task was to walk at a preferred treadmill speed, age-related changes did not affect the speed and SL of older adults with diabetes compared with other groups in the present study. However, previous studies reported that the level of health determined

speed (Studenski et al., 2011, Dingwell et al., 2017). The participants in the present study were healthy, active people with no history of falls and gait dysfunction, so they walked normally on the treadmill in baseline.

When an unexpected target was seen on the monitor, the response was reactive and performed without any requirement to afferent (sensory) feedback (Bard et al., 1999). Thus, the motor system can modify the trajectory of the foot using central feedback loops comparing the goal for a SL or an MTC height and the existent efferent copy (i.e., a forward internal model of the relevant response). However, when the target stayed for providing feedback, a feedforward correction model was used to match the step length/MTC with the presented target.

Older adults with diabetes can benefit from gait adaptability training with biofeedback. According to the results of Aim 1, the older participants with diabetes were found to have impaired gait adaptability in response to unpredictable goal-oriented tasks (involuntary responses). In line with the findings in Section 4.2, the experimental results did not support Hypothesis 2.2. Therefore, older adults with diabetes could not accurately adjust sagittal foot trajectories without biofeedback. The findings of biofeedback effects on sagittal foot displacement adaptation in Section 4.3 did not support Hypothesis 2.1. So the older diabetes adults could use feedback about their performance and adapt their sagittal foot trajectories. Background visual information could influence responses and make them more accurate.

Methods of determining preferred treadmill speed and magnitudes of targets were consistent in the present study. Participants walked with preferred velocities that were determined in baseline treadmill walking. These velocities were not significantly different

between groups. Therefore, differences between groups did not originate in the difference between treadmill speeds and the method of determining target magnitudes.

Matching step lengths and MTC heights with presented targets led to adaptation of gait parameters; however, older diabetic adults showed the largest changes in gait parameters and as such, increased errors. Compared with the healthy older and young adults, older diabetic adults showed a reduced accuracy of sagittal foot trajectory adaptation while walking on a treadmill and responding to targets that suddenly and unexpectedly appeared on the monitor. The older adults with diabetes adjusted the displacement of their trailing foot and responded to presented tasks. However, their reactive responses were not as accurate as participants in other groups because it was more difficult for them to modulate and adapt foot trajectory with a target that shifted forward, backward, or upward when step initiation was made (Mazaheri et al., 2014, Kim and Brunt, 2013).

Older adults with diabetes had some deficits during online correction compared with other groups, presumably, due to some diabetes-related changes, not age-related changes, in the central nervous system: greater physiological noise (Dingwell et al., 2017). The frontal and parietal cortex lesions in stroke patients (Mutha et al., 2014), lateralised right hemisphere (Mars et al., 2007) and reduced abilities to control the development of force in response to tasks (Kim and Brunt, 2013) were proposed to reduce the accuracy of step adjustments.

The frontal cortex executes response inhabitation which is important to avoid falling by stopping ongoing commands and modulating them based on sensory information (Aron, 2011, Bari and Robbins, 2013). However, the underlying mechanisms were mostly

studied in people with Parkinson's disease and stroke patients, and they may not apply in older adults with diabetes.

During online correction when targets stayed for a few steps, participants made voluntary adjustments of their performance using biofeedback. During this time, decision-making and reprogramming of movement were involved (Gaveau et al., 2014). The forward model enabled the CNS to predict the consequences of motor commands by modulating feedback loops. In line with previous research (Tseng et al., 2009), the participants reduced their foot displacement errors in response to predictable targets during the online correction. However, a greater standard deviation during walking with biofeedback may indicate an increased risk of falling (Begg et al., 2007). The interaction of diabetes and ageing impair the central nervous system that is responsible for processing and integrating information (Seidler et al., 2010, Rosso et al., 2013, Biessels et al., 2002). Impaired integrated activity in the brain and spinal neural networks during walking increases the latencies of evoked potentials and as such reduces conduction velocity in peripheral nerves (Di Mario et al., 1995), so older adults with diabetes may need more time to respond compared with people without diabetes. However, because tasks were presented with the same available times for all participants, errors of step adaptation increased in older adults with diabetes. Older adults with diabetes used a forward model (Desmurget and Grafton, 2000) allowing biofeedback (the comparison between the current foot displacement state derived from the output of the internal forward model with a presented target) to increase the accuracy of the internal forward model in the following step.

The older participants with diabetes used biofeedback and reduced errors of foot trajectory adaptation. In line with previous research of foot displacement adaptation in

older adults and stroke patients (Spedden et al., 2019, Caetano et al., 2016, Begg et al., 2014b, Young and Hollands, 2012, Chapman and Hollands, 2010), this study found that detailed, meaningful information about goal-task performance in each step improved the accuracy of foot displacement in older adults with diabetes. Similar to other groups, the older diabetic group compared their real-time SL and MTC height with presented targets and reduced errors in next trials. The error may have been derived from the delayed comparison between the expected performance and the real performance during the fast online correction (Desmurget and Grafton, 2000) or from the instantaneous comparison between the goal and the expected sensory feedback during inter-sensory and visuomotor adaptation (Magescas et al., 2009). Reduced errors showed that corticospinal involvement in the control of gait increased when visual information of foot displacement was presented (Spedden et al., 2019). Visual biofeedback excites the motor cortex and corticospinal pathway during online correction compared with walking without visual feedback, which increases firing of corticospinal neurons (Schubert et al., 1999, Drew et al., 1996) and stronger electroencephalography activity during step shortening compared with step lengthening which is more difficult (Wagner et al., 2016).

The cerebellum is responsible for online movement correction since cerebellar cortical and nuclear pathways are related to vision and proprioception guide movement and encode kinematics of movement (Prevosto et al., 2010, Casabona et al., 2010, Ebner et al., 2011). The cerebellum has been also reported to be involved in predictable and unpredictable perturbations while walking. The role of the cerebellum is not critical for modifying and executing a previously acquired adaptive strategy based on an off-line modification of the motor system for predictable perturbation, but its role is crucial during unpredictable perturbation to develop the continual modification of a motor task in

response to sensory cues (Shimansky et al., 2004). Errors during both prediction and feedback are caused by the activity of Purkinje cells in providing the signals for generating the sensory prediction errors used to update a forward internal model (Popa et al., 2012). Greater errors during online correction in the older diabetes group compared with the control groups without diabetes might indicate learning deficits. Learning deficits in diabetic rats were found to be associated with changes in the hippocampus and dependent on diabetes duration and severity (Gispen and Biessels, 2000). However, the older adults with diabetes were able to reduce their errors during online correction compared with their responses without biofeedback, so learning impairment was negligible in the participants. Practice and exposure to biofeedback as reported in previous research (Potocanac and Duysens, 2017, Potocanac et al., 2015) was found to improve performance.

As discussed above, the forward internal model is likely to have an important role in the adaptation of movement in the early stages when the initial response is inaccurate. During the online correction of ongoing movement, multisensory visual and proprioceptive feedback was used to correct ongoing movements in cortical areas. Although the possibility of fast subcortical loops during online correction of movements with visual feedback in a split-brain patient has been shown (Day and Brown, 2001), it is unknown whether these loops exist in healthy human subjects. The involvement of cortical areas in online visuomotor guidance has been investigated only in animal models (Galletti et al., 2003, Buneo and Andersen, 2006, Georgopoulos, 1998).

5.4. Agreement between quantified errors in the overground and treadmill gait adaptability tests

The findings in Section 4.4 supported Hypothesis 3 (at least 95% of points in scatter plots would fall between the limits of agreement) for shortening and lengthening step lengths and increasing MTC heights in response to SSL and LSL targets and high MTC targets/obstacles during TGA and OGA tests.

Measured absolute errors for step shortening and lengthening using OGA tests were in agreement with those in TGA tests. All participants except two in the step shortening condition and one in the step lengthening condition had exchangeable errors in both tests. All participants had greater errors in the TGA test compared with those in the OGA test. During treadmill walking, each participant tried to stay in the centre of the treadmill by walking with the speed of the treadmill determined in the baseline. A sudden reaction to a SSL/LSL target in the next step meant walking slower or faster than the speed of the treadmill, so an increase in errors in the first attempt partially satisfied the response to presented tasks while considering the risk of touching back and front edges of the treadmill. Considering the principle of posture-first (Bloem et al., 2001) and prioritising walking over other concurrent tasks may explain increased errors in the TGA tests.

Only errors of MTC heights quantified in the TGA test in response to high MTC targets and errors of MTC heights in the OGA test were in agreement while crossing the 5-cm height. The height of the obstacle and high MTC targets were comparable. Participants were able to increase their MTC heights during treadmill walking at preferred speeds in response to the sudden presentation of high MTC targets.

Responding to higher MTC targets during treadmill walking produced larger errors than they did in the OGA tests. The errors were not exchangeable in over 15% of participants. Virtual higher MTC targets might increase reactive responses because the magnitudes of targets were not reasonable during treadmill walking.

5.5. Project summary and conclusion

This project was the first study to investigate overground gait adaptability, overground and treadmill sagittal foot displacement adaptation, the effectiveness of biofeedback for improving sagittal foot displacement adaptation, and exchangeability of overground and treadmill gait adaptability tests for quantifying foot displacement adaptation in older adults with diabetes. In order to distinguish the effects of ageing and diabetes, three groups of participants completed novel tests including the combined paradigms of stepping and obstacle avoidance with both virtual and real targets. The results of tests showed that older adults with diabetes had impaired gait adaptability and reduced accuracy of foot displacement adaptation during overground walking. However, they could use biofeedback to improve the accuracy of foot displacement adaptation during online correction. Assessing the agreement between TGA and OGA tests for quantifying foot displacement accuracy revealed that the tests were exchangeable.

In conclusion, the study has demonstrated the interaction of ageing and diabetes led to impaired gait adaptability and to more conservative gait patterns in older adults with diabetes. This impaired gait adaptability may place them at an increased risk of falls while reacting to unexpected challenges during walking.

Training with feedforward biofeedback such as novel biofeedback tools (Mehdikhani et al., 2019) can assist this population to reinforce their feedforward internal models which

control foot trajectories in response to a sudden change in the environment. Feedforward control allows the CNS to compute necessary motor output by updating the efferent copy of goal tasks and reducing the occurrence of touching obstacles during walking. The older adults with diabetes were able to control the sagittal trajectories of their feet by using biofeedback during online correction of their reactive responses. The use of feedback to enhance long term learning will provide opportunities for biofeedback training of the sagittal foot landing that influences fall risks during daily walking.

The advantages of the TGA tests over OGA tests are prominent when a large space, longer time associated with preparation and data collection are not available. Using only runners and a pelvis belt (Appendix H) with markers attached to them, it is possible to quantify the accuracy of sagittal foot placement adjustments to identify people suitable for gait training programs to help reduce their risk of falling. The inclusion of TGA tests in fall risk assessment (Geerse et al., 2019) can be used to enter eligible older adults into fall prevention programs (e.g., gait adaptability training with targeted biofeedback).

5.6. Limitations

This study has several limitations. Only immediate effects of the training program on gait adaptability were investigated in a group of young adults (Appendices A and H). Due to the limited time of the PhD study, the effects of the training program on gait adaptability were not investigated in this study.

The effects of diabetic neuropathy on gait adaptability were not investigated. Despite several attempts to recruit participants with diabetes-related neuropathy through Diabetes Victoria and diabetes support groups, only two older volunteers with diabetic neuropathy

completed the overground and treadmill gait adaptability tests by the official end of study time. Therefore, the effects of diabetic neuropathy on gait adaptability are limited.

The choice of parameters for investigation was limited to the sagittal plane. Foot displacement adaptation was not investigated in the mediolateral direction since the overground gait adaptability tests could not present targets for assessing the mediolateral foot displacement adjustments. Furthermore, biofeedback tools were limited to quantifying step length and MTC adaptation. It is possible to adjust target presentation in the mediolateral direction during biofeedback training (Mehdikhani et al., 2019); however, the tools for assessing foot adjustments need to be further developed to include the mediolateral errors.

The direct comparisons between step adaptation in the vertical direction quantified by two overground and treadmill gait adaptability tests (in response to the sudden appearance of targets) were impossible. In the TGA test, participants were asked to match their MTC with two presented MTC targets on a monitor whereas in the OGA test, participants were required to cross the physical obstacle (5 cm). Therefore, to compare the MTC errors, further data processing was conducted.

Some older participants only completed either the OGA or TGA test. Because the results of both tests were required for the results to be included to address Research Question 3, these participants were excluded from the study in the statistical analysis, this limited the sample size for investigating this aspect of the study.

Finally, the method of sampling might limit the generalisation of the findings. For example, almost all participants were recruited from the western suburbs of Melbourne because they lived close to Victoria University. Further studies are required to provide

insights into the gait adaptability of older adults with diabetes with participants selected more randomly across the population.

5.7. Future research

The findings of this study revealed that diabetes per se impairs gait adaptability in older adults. Future research should include older participants, both with and without diabetic neuropathy, to investigate how neuropathy affects gait adaptability.

Additional information describing gait kinetics and kinematics will help to draw more descriptions of impaired gait adaptability. Comparing the centre of mass, the centre of pressure, joint angles and joint moments can help explain strategies used in response to the sudden appearance of targets or obstacles. Another potential future direction is to use musculoskeletal modelling to investigate internal muscle forces during gait adaptation.

This thesis did not explore the effects of the training program, as described in Appendix A, on gait adaptability. The findings of the immediate effects of the training program in young adults suggest great potential. It will be interesting to explore in future research the use of the proposed TGA and OGA tests and training program to improve gait adaptability in older adults with and without diabetes.

The protocol for long-term training with targeted biofeedback as presented in Appendix C may be applied to investigate accuracies of foot displacement adjustments both during overground and treadmill gait adaptability tests.

The current research excluded older adults with a history of falls. To investigate the relationship between falls and impaired gait adaptability, the inclusion of older individuals with a history of falls is suggested in future research.

Finally, impaired gait adaptability has been found to be an issue in several pathological populations. The biofeedback tools developed in this thesis can be useful to improve gait adaptability for falls prevention in high-risk populations, such as people with stroke or Parkinson's disease.

Appendix A. Supplementary publications

Peer-reviewed journal article

Mehdikhani, M., Taylor, S., Shideler, B.L., Ogrin, R., Begg, R. (2020). Age effects on step adaptation during treadmill walking with continuous step length biofeedback, Gait & Posture, 80: 174-177.

Conference proceedings

Mehdikhani, M., Taylor, S., Shideler, B.L., Ogrin, R., Begg, R. (2019). A flexible real-time biofeedback tool that trains gait adaptability, The XXVII Conference of the International Society of Biomechanics and the American Society of Biomechanics, July 31 – August 4, Calgary, Canada.

HDR Student Conferences in iHeS

Mehdikhani, M., Taylor, S., Ogrin, R., Begg, R. (2020). Gait adaptability in Older Adults with Diabetes Mellitus during overground walking, The HDR Student Conference, The Institute for Health and Sport, Melbourne, Australia.

Mehdikhani, M., Taylor, S., Shideler, B.L., Ogrin, R., Begg, R. (2019). Aging and diabetes' effects on gait adaptability during treadmill walking, The HDR Student Conference, The Institute for Health and Sport, Melbourne, Australia.

Online publication

Martin, S. (Mehdikhani. M). (2020). Threats and opportunities for a research student during COVID-19 restrictions, Biomechanics in a COVID world: Student reflections and stories from 2020, International Society of Biomechanics.

Gait & Posture 80 (2020) 174–177

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Age effects on step adaptation during treadmill walking with continuous step length biofeedback

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Abstract

Background: The inability to adjust step length can lead to falls in older people when navigating everyday terrain. Precisely targeted forward placement of the leading foot, constituting step length adjustment, is required for adaptive gait control, but this ability may reduce with ageing. The objective of this study was to investigate ageing effects on step length adaptation using real-time biofeedback.

Research question: Does ageing affect the ability to adapt step length to match a target using real-time biofeedback?

Methods: Fifteen older adults (67±3 years; 8 females) and 27 young adults (24±4 years; 13 females) completed a step length adaptation test while walking at preferred speed on a treadmill. The test involved walking while viewing a monitor at the front of the treadmill that showed a real-time signal of absolute left-right foot displacement. The task was to match the local maxima of the signal (i.e. step length) to two target conditions, at 10 % longer or 10 % shorter than mean baseline step length. When the target was displayed, it remained unchanged for a set of 10 consecutive step attempts. Three sets of 10 attempts

for each target condition were allocated in random order, for a total of 30 step attempts per target. Average absolute error and average error (bias) of step length accuracy was computed for each target condition and compared between groups.

Results: : The step adaptation test identified that older adults had greater mean absolute error for both short and long step targets and showed a step length-dependent bias significantly different to the young.

Significance: Real-time foot position feedback could be a useful tool to train and evaluate step adaptation in older people.

Keywords: Ageing, step adaptation, gait biofeedback, treadmill walking.

1. Introduction

Adaptation and accuracy of step-to-step foot placement while walking is important to minimize falls risk factors in older people [1–5]. Methods that evaluate and train this gait ability in older people show success for reducing falls [6,7]. Currently, there are two types of visual-cued instruments that have been developed to evaluate and train goal-directed stepping tasks during walking, shown in Table 1.

Table 1. Existing types of visual-cued instruments that assess goal-directed stepping tasks.

Type	Description	Evidence
Visual cues in the form of ground-projected targets	Participants required to pay attention and navigate approaching target silhouettes along the surface of a walkway [2,8], or upon an instrumented treadmill surface [3]. No online biofeedback of accurate step placement.	Ageing effects the ability to accurately adapt foot placement for a discrete stepping task [2,3].
Visual cues in the form of non-immersive virtual-reality targets	Advantages with respect to ground-projected targets: Targets displayed on an eye-level monitor. Enables more upright walking and a more comprehensive biofeedback experience. Provides flexibility to control task conditions and visual biofeedback information to augment specific capacities of the performer.	Advantages with respect to ground-projected targets: Targets displayed on an eye-level monitor. Enables more upright walking and a more comprehensive biofeedback experience. Provides flexibility to control task conditions and visual biofeedback information to augment specific capacities of the performer.

This paper provides evidence for the use of non-immersive virtual-reality targets to measure goal-directed stepping task ability in older people. We developed a software tool that receives motion-capture data to display real-time information on relative foot position using a simple 14 mm retroreflective marker on the distal shoe. In this study, participants walked on a treadmill and received real-time step length feedback of the error between their step length and the target while the software tool quantified step length errors during continuous online corrections. The purpose of this study was to evaluate the potential merit of a new biofeedback instrument designed to measure step adaptation ability in older adults to answer the research question: Does ageing affect the ability to adapt step length to match a target using real-time biofeedback? We expected that older adults would show greater step placement error compared to a younger control group.

2. Methods

2.1. Participants

Healthy young adults (20–35 years) and older adults (60+ years) were invited to participate (Table 1). Participants were excluded if they self-reported chronic disease, cognitive impairment, history of a fall within the last year before the study, visual impairment, or musculoskeletal deficits. The study was approved by the Victoria University human research ethics committee.

2.2. Experimental set-up and procedure

Participants' gait was tested on a motorized treadmill while wearing a safety harness. Kinematic data were collected and streamed in real-time using three-dimensional motion capture (VICON, Oxford, UK). All participants wore minimal mid-sole shoes (Merrell Bare Access 4) with a tracking marker attached at the distal 1st toe. VICON cameras collected three dimensional trajectories of the toe markers. Biofeedback information in real-time was displayed using customized MATLAB software (MathWorks, Natick MA, USA), which accessed toe marker input data from VICON Nexus via Visual3D-Server software (C-Motion, Germantown MD, USA).

Participants first completed a treadmill warm-up and acclimation session (5−10 min), including determining the preferred walking speed (PWS) on a treadmill [9]. Briefly, the PWS was obtained from an average of three maximum and minimum speeds deemed by the participant to be faster and slower than their preferred treadmill walking speed. Participants then walked at their PWS for 10 min in a baseline condition without feedback. The MATLAB codes computed an updated mean step length, used to determine

the two target conditions for the subsequent stepping tests: a 10 % shorter and a 10 % longer step length. In the adaptation test, real-time step length was displayed as a line graph on a display monitor mounted at eye level in front of the treadmill (Fig. 1).

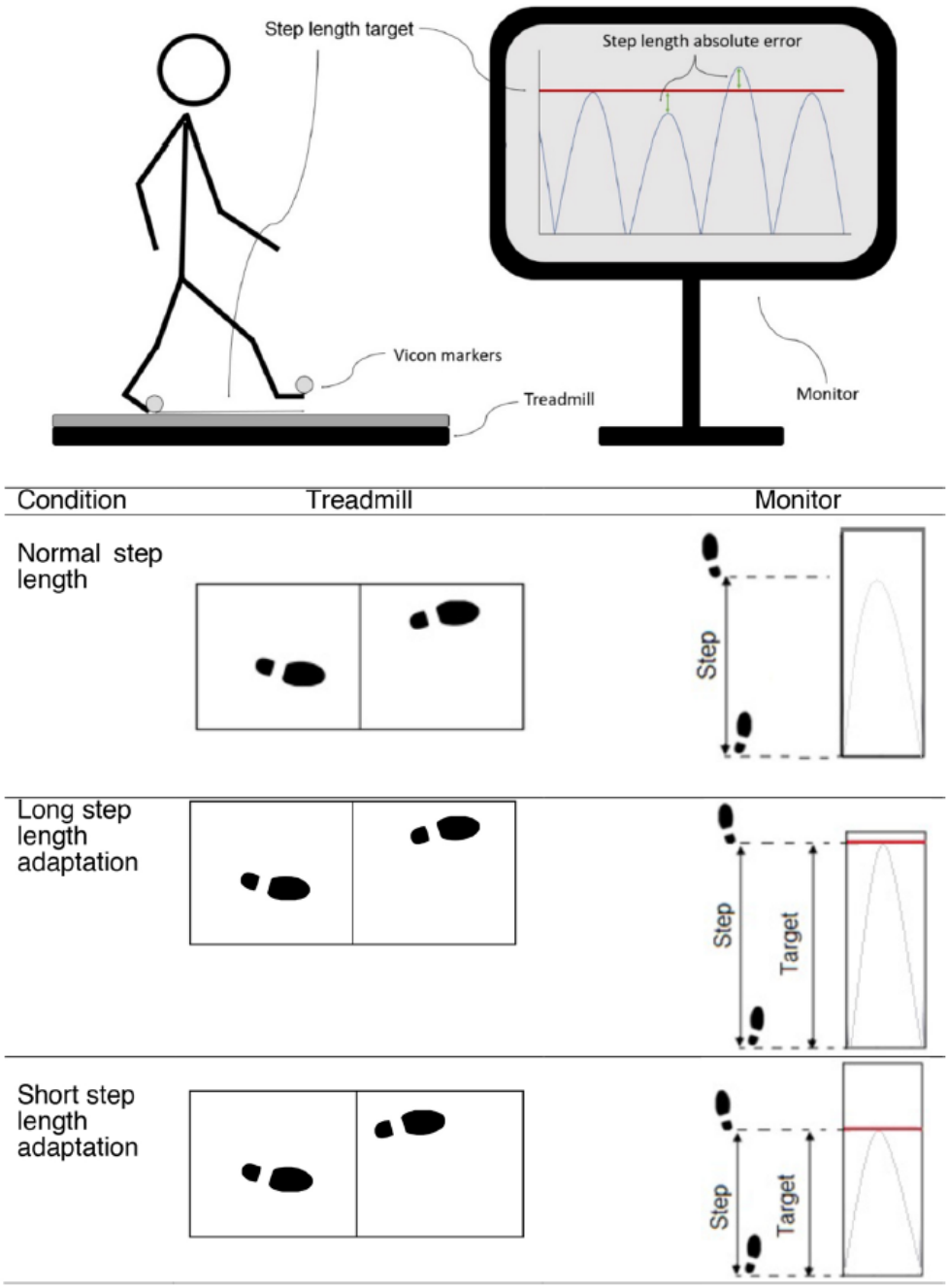

Fig. 1. Step length adaptation displayed on a monitor while treadmill walking under three conditions: normal step length (no target was presented on the monitor), long step length adaptation (a long step length target was presented on the monitor) and short step length adaptation (a short step length target was presented on the monitor).

The real-time step length was determined by local maxima of the absolute anterior displacement signal derived from online difference between the two toe marker positions. Participants were asked to match the peak maxima of the graph (step length) with a horizontal line of the target condition (Fig. 1). The target line was first presented on the display when the trailing foot was entering terminal stance, and remained visible for 10 consecutive steps. The target was then withdrawn for the following 10 steps. This alternating cycle was repeated three times per target condition and in random order. Participants thus used real-time visual information to minimize the distance between their executed step and the target (Fig. 1). Participants were unaware of the timing, duration, number of trials, and values of targets.

2.3. Data processing and statistical analysis

For each of the 60 step attempts, the error and absolute error were computed as the difference between step length attempt (Si) and the assigned target magnitude (T), specifically (Si – T) and |Si – T|, respectively (Fig. 1). Average error and average absolute error were used to represent the participant's score per target condition. Baseline measures, absolute error and average error were compared between groups per target condition using independent sample t-tests and Mann Whitney U tests (SPSS, IBM 24, Chicago, IL, USA).

3. Results

Preferred walking speed and step length were not significantly different between groups in baseline walking (Table A1 in Appendix A). Older participants had higher mean absolute error for short and long step length ($p = .001$), compared to young participants (Fig. 2A, Table A1 in Appendix A). Older adults also showed an error bias dependent on

whether the step length target was longer or shorter than their adapted step length (Fig. 2B), with error significantly greater for older than for young participants in response to both the short and long step target see Table A1 in Appendix A). Median error for older participants was greater in response to both the short and long step target, compared to young participants (Table A1 in Appendix A).

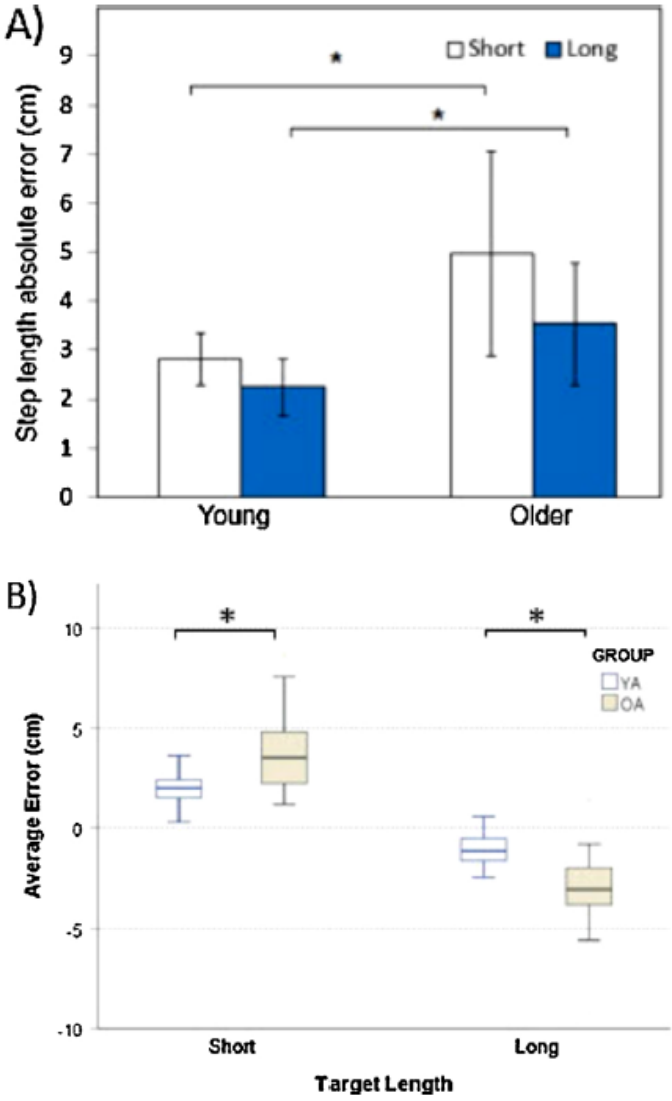

Fig. 2. A) Mean absolute error for group and target condition (error bars±1× SD), and B) box plot showing median and inter-quartile range for mean average error. Average error indicates bias towards the participants adapted step length. Accurate step is indicated by 0 cm for both dependent variables of absolute error and average error. For each participant, mean step length error/absolute error was the average of 30 attempts at the short step length target and the average of 30 attempts at the long step length target.

4. Discussion

We investigated ageing effects on step accuracy in response to virtual step length targets presented during treadmill walking at preferred speeds using a custom software tool. As hypothesized, compared to younger participants, older participants had larger step length adjustment error and greater error that was biased to the adapted step length, when matching their step length with a presented step length target. Our results are consistent with other research findings in several areas: relatively larger absolute error for short step lengths compared to large step lengths [3,10], the step length error is biased towards the baseline mean [3], and ageing is associated with reduced step length accuracy for both short and long step length targets [2,3].

To the best of our knowledge, this experiment demonstrated the first step-control intervention to incorporate a spatial-temporal gait measure into real-time visual biofeedback of the performance. In previous studies, real-time signals of vertical toe position [11], knee and ankle angles [12] and lower-limb EMG activity [13] have been used to modify and analyse gait of healthy, older adults in this manner. Using real-time foot position signal to analyse stepping accuracy reduces the need for the additional hardware used to measure stepping accuracy in previous studies, such as force plates [5] and virtual targets generated using a light-beam projector [3,8].

The present study differs from previous studies by directly quantifying step accuracy in the AP direction and allowing participants to use visual information precisely to control their performance for consecutive steps in real-time. A future application of the biofeedback instrument is to investigate the targeted toe position during leg swing, and evaluate adaptability of foot-ground clearance. This approach may provide a more

comprehensive assessment of ageing and pathology effects on step placement adjustments that will increase our understanding of the mechanisms of falls in older populations, and thereby allowing the development of strategies to minimize falls [14].

Appendix A

Table A1. Descriptive statistics of subject characteristics (mean ± standard deviation, median ± inter-quartile range) and the measurement error from the step-to-step length adaptability task.

		Young	Older		
		n = 27	n = 15		p value
Participant characteristics					
Age (years)		25 ± 4	70 ± 6		
Number of women		13	8		
Body mass index		24.0 ± 2.7	26.2 ± 3.4		p = .74
Preferred walking speed (m/s)		1.13 ± 0.10	1.06 ± 0.18		p = .35
Baseline step length (cm)		58.7 ± 8.3	58.0 ± 10.9		p = .47
Error measurements per step length comparing young and older participant groups					
Error measurement per step length		**Target Condition**			
Mean Absolute Error (cm)	Short Step mean ± SD	2.82 ± 0.53	4.97 ± 2.08	t_{40} = -3.9	p = .001
	Long Step mean ± SD	2.24 ± 0.58	3.54 ± 1.24	t_{40} = -3.8	p = .001
Mean Average Error (cm)	Short Step median ± IQR	1.98 ± 0.92	3.66 ± 3.61	U = 90.5	p = .003
	Long Step median ± IQR	−1.17 ± 1.18	−3.1 ± 2.18	U = 58.0	p = .0001

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A flexible real-time biofeedback tool that trains gait adaptability

Mahboobeh Mehdikhani, Simon Taylor, Rezaul Begg, Blynn Shideler, Rajna Ogrin

Summary

This study demonstrates a biofeedback method for training adaptable gait. The method involved a customised MATLAB program and a 3D motion capture system. 22 young adults were selected to test the effect of gait adaptability training on error-reduction of two target-oriented stepping tests while walking: (1) vertical foot height error and (2) fore-aft foot placement error. Kinematic trajectories of lower limb and target state were displayed in various forms in real-time by TV monitor at front of treadmill. Participants were allocated to either a control or experimental group. The gait adaptability training program was administered to the experimental group, while a 'placebo' training program was delivered to the control group. All participants performed the same adaptability test at pre- and post-training. The training program involved generic but targeted stepping tasks that were different to the test trial tasks. The experimental group showed improved foot placement adaptability following biofeedback training.

Introduction

Ability to adapt gait patterns that match changing environment situations is important for safe navigation. Our embodied locomotor system is equipped for these task demands, but gait can become less adaptable with ageing and pathology. Research has shown that target-oriented biofeedback has a greater beneficial effect on gait compared to walking without biofeedback [1,2]. While biofeedback programs that provide visual projection of stepping performance have demonstrated positive outcomes for gait and posture [1,3],

there is still need for further development to achieve maximal effect from an efficient training regime. Therefore, a training method should include a variety of task and target-oriented options that challenge the various sub-tasks of the gait cycle, and enable growth of abundant repertoires of locomotor solutions. The aim of this project was to evaluate a new gait adaptability training method, which incorporated a variety of walking tasks and adjustable targets that probe different locomotor control system resources. It was hypothesised that targeted biofeedback training would produce short-term improvement for a given gait adaptability performance test.

Methods

Young healthy adult participants were each fitted with customised footwear and a pelvic belt that had cluster of retro-reflective markers. A 3D motion capture system (Vicon Pty) collected the foot and limb position data and streamed to Visual3D Server (C-Motion Pty). A customised MATLAB program was developed to present a graphical display of task performance in real-time. From a warm-up period of treadmill walking, mean and standard deviation statistics of gait kinematics served as input for the tests and training program. Second, the pre-training gait adaptability test was scored by the accumulated error from four 1-min tests (task by limb, 2×2), where a target was kept constant for a block of 5-steps before random switch to a new state. Each limb was scored by a separate test that involved four blocks of targeted stepping. Third, the experimental group performed a training regime of six 3-min trials (2-min rest periods). Four 2-minute trials consisted of 'random stepping stones' where foot position targets continually varied in fore-aft and medio-lateral position. Two 2-minute trials targeted left and right hip height of the swing limb at mid swing. Two trials targeted left and right swing limb length at

mid-swing. The control group were presented with same six trials but asked to explore without target criteria. The gait adaptability tests were then repeated at completion of training. An independent t-test evaluated the post-test difference between the groups.

Results and Discussion

Post-test gait adaptability performance show the experimental group reduced their foot position error ($p < .08$ and $p < .09$) (Figure 1). Both groups scored similarly at pre-test ($p=.46$).

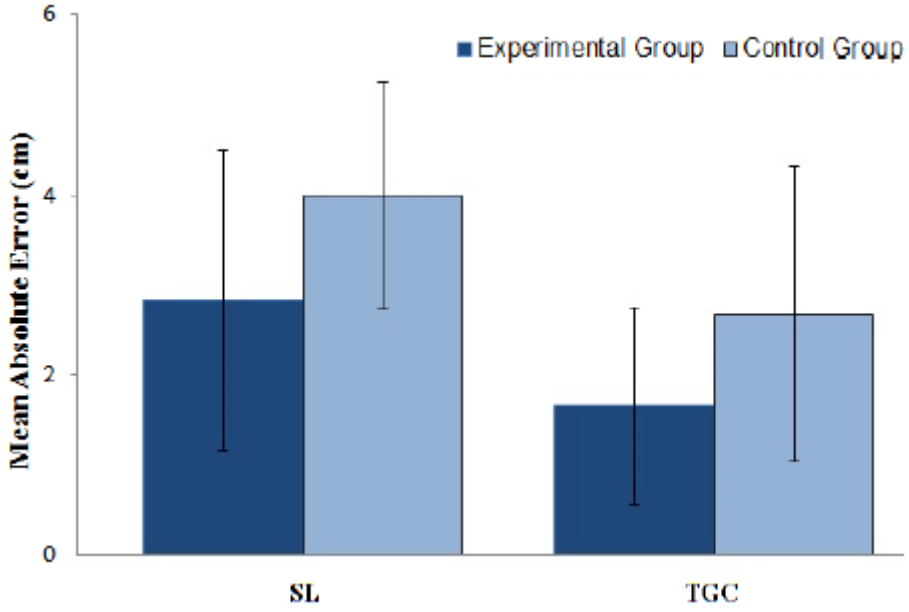

Figure 1: Comparison of mean absolute errors for step length (SL) and toe-ground clearance (TGC) after training between two groups.

Conclusions

The biofeedback method had an acute effect on gait adaptability. Future work will evaluate the effect of this training regime on adaptability tests prescribed for over ground walking tasks in ageing and gait pathology populations.

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Aging and diabetes' effects on walking adaptability during treadmill walking

Mahboobeh Mehdikhani, Simon Taylor, Blynn Shideler, Rajna Ogrin, Rezaul Begg

Falls are due to trips or misplaced steps, which suggests a reduced walking adaptability. This study investigated aging and diabetes' effects on walking adaptability.

Twenty-seven young adults (26 ± 5 years), 16 healthy older people (69 ± 6 years), 19 older people with diabetes (70 ± 7 years) participated. Walking adaptability was assessed in response to targets displayed on a head-level monitor during treadmill walking at a preferred speed using a novel biofeedback program. Four targets randomly appeared 3 times, stayed during 10 consecutive steps, and disappeared during 10 following steps.

There was a significant difference between walking adaptability means in the walking direction (F (3, 58) = 7.132, p = .000 and F (3, 58) = 4.252, p = .009) and the vertical direction (F (3, 58) = 8.288, p = .000 and F (3, 58) = 5.173, p = .003 for the dominant limb and F (3, 58) = 6.584, p = .001 and F (3, 58) = 7.363, p = .000 for the non-dominant limb). Both aging and diabetes impaired walking adaptability, potentially contributing to this population being at higher risk of tripping and falling. This information may be used to develop strategies to reduce trips/falls.

Gait adaptability in Older Adults with Diabetes Mellitus during overground walking

Mahboobeh Mehdikhani[1], Simon Taylor, Rajna Ogrin, Rezaul Begg

Background: In response to changes in environment, gait parameters are adapted (adaptable gait) to accurately place feet related to external hazard (obstacles). Inability to accurately place foot in the AP and vertical directions related to hazards may explain why older adults with diabetes are likely to fall more frequently than healthy older adults.

Aim: To investigate the effects of diabetes mellitus in older adults on gait in responses to step length targets and obstacle avoidance during walking.

Methods: Forty three subjects were involved: 16 young adults (YA) aged 18-40 years, 14 healthy older (HO), and 13 older with d system and Advanced Mechanical Technology collected marker trajectories and ground reaction older diabetes (OD) adults aged 65 year and older. Vicon three-dimensional motion analysis was used to measure spatiotemporal parameters. A novel instrument presented four conditions randomly, two steps ahead: step shortening, step lengthening, obstacle avoiding, walking through. Gait parameters and foot placement accuracy (the difference between the toe and target/obstacle) in each condition were quantified. Repeated measured ANOVA tests were used to test the main effects of group at a significance level of 0.05.

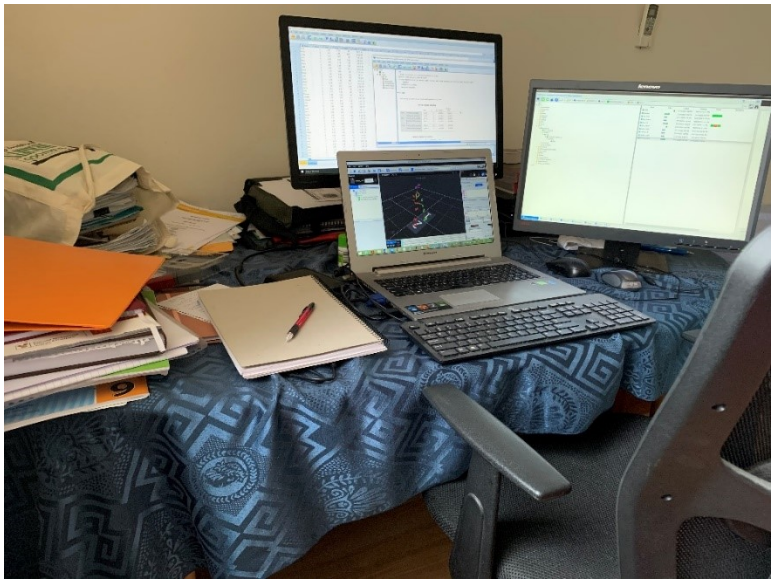
Results: Gait (velocity, stance time, and DS time) and foot placement accuracy were significantly different between YA and OD, and between HO and OD during step shortening, step lengthening, and obstacle avoiding. Diabetes impaired gait adaptability

and reduced the accuracy of foot placement in the OD participants. This may reduce a safe navigation in OD when they suddenly need to adjust the placement of their feet.

Threats and Opportunities for a Research Student in COVID-19 Restrictions

Suzanne Martin

In March 2020, COVID-19 cases in Australia were increasing rapidly. At Victoria University (VU), Melbourne, I shared an office with other research students to complete data processing and analysis of my PhD thesis entitled Gait Adaptability in Older Adults with Diabetes. Every day we were being bombarded with information on how to prepare for the pandemic. I tried to avoid any surface and used a tissue whenever I had to touch doorknobs. One day, one of my colleagues who had coughed a few times said: ‘’Excuse me, guys. Just letting you know that I do not have COVID; it is just a seasonal allergy.’’ Feeling sorry that she felt the need to explain the reason for her cough and thinking it better to work from home, I collected my things, backed up my files, and left my office. Only a few days later VU issued a directive that research students were no longer permitted to work from their offices.



Time was running out to submit my thesis by September. I had to keep working, but how? Changing my bedroom into an office, I gained permission to take my office computer home and connect it to the Biomechanics Laboratory via VPN to use special software.

Every morning I made sure my daughter was connected to her classroom virtually. Then, attempting to concentrate on my own work I could hear her calling: ‘’Mum,…, mum, can you come?’’ The internet was disconnected, or whatever other reasons. Then it was recess, lunchtime, or needing assistance with her school work. The only time I could concentrate on my work was between 4 and 6 pm and after 9.30 pm when she was in bed. I worked until after midnight, wearing myself out.

Once in bed, my mind was racing and I couldn’t sleep. ‘’What would be in store for us? What would I do if one of my family overseas caught COVID? When would the lockdown finish? What would I need to buy tomorrow?” Although my alarm was set for 7 am, most mornings I was awake by 6 am. I stayed in bed, feeling blue, until I heard my daughter rising at about 7.10 am. ‘’Come on Sue, it is time to get her breakfast; otherwise, she’ll be late,’’ I said to myself.

After surveying all research students about how the pandemic was affecting their research and professional development, the university organised support through an online program called Elevens, the COVID-19 Student Hardship Fund, Doctoral Industrial Placement (DIP), and Small Research Grant. With dwindling savings and realising I couldn’t meet the deadline for submitting my thesis, I applied for a DIP scholarship and one-semester extension with my supervisor’s support. I worked remotely for a physiotherapy centre as a data scientist. At the end of my first meeting in the centre with a face mask and sanitised hands, my project manager stepped well back, removed his mask, and said: “I look like this.” Soon after this I was informed that my application for a small grant was successful. This enabled me to strengthen my PhD project and submit an abstract for the Research Student Virtual Conference at Victoria University.

Looking back, I appreciate that students' health, safety, wellbeing, and quality of study had remained the priority of Victoria University. With the support of the university, I changed threats into opportunities and maintain quality research while remaining resilient and keeping myself, family and peers safe.

Appendix B. Flyer

Volunteers wanted for research assessing walking abilities

You may be eligible to participate if

You are ≥ 60 years, able to walk without any walking aid and able to attend the Biomechanics laboratory at Victoria University Footscray Park Campus only for ONE hour.

What will you be asked to do

Walk at your preferred walking speed to evaluate your gait adaptability at Victoria University (Footscray Park Campus). This will involve recording your motions while you walk. This project is being conducted by a research student at Victoria University College of Sport and Exercise Science.

If you are interested and want to get more information, please contact

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[redacted]

Appendix C. Information to participants

You are invited to participate

You are invited to participate in a research project entitled "Gait Adaptability and Biofeedback in Older Adults with Diabetes". This project is being conducted by a PhD research student, Ms. Mahboobeh Mehdikhani, at Victoria University under the supervision of Prof Rezaul Begg and Dr Simon Taylor from College of Sport and Exercise Science, Institute for Health and Sport.

Project explanation

The project aims to investigate the effect of diabetic peripheral neuropathy on adaptable gait; measured by the ability to respond with a matched gait pattern to a varying target criterion. To evaluate the effect of diabetes and neuropathy on gait adaptability we will perform three studies. Study-1 will test three age-matched groups: healthy older adults, older adults with diabetes, and older adults with diabetes neuropathy. Gait adaptability will be evaluated using test protocols that require the participant's stepping pattern to match a target criterion while they walk on treadmill and overground. The treadmill test will display biomechanical details of the participant's gait pattern on a screen at the front of the treadmill and through this biofeedback display they will aim to match their pattern with a target signal. For the overground test, participants will be required to adjust their gait pattern in relation to one of three real targets situated midway along an 8m walking path. The difference between the conditions of these two tests will be in the form of a virtual target verses a real target. However, the dependent variables of gait adaptability will relate to the same gait features. In addition to evaluating a pathology effect on gait

adaptability within each test condition, this study will investigate the test validity of the treadmill protocol against the overground protocol through statistical correlation analyses of the dependent variables. Study-2 will evaluate biofeedback gait training on gait adaptability performance using study-1 test protocols at post intervention. The diabetic neuropathy patients from study-1 will undergo an intervention program that involves ten separate training sessions of treadmill walking under biofeedback conditions that train gait adaptability. Study-3 will examine the mechanics of those participants that were ranked highest and lowest for gait adaptability improvement in study-2.

What will I be asked to do?

Study-1 includes two separate overground and treadmill walking tests. Overground walking test includes gait adaptation parameters in response targets that are displayed in random sequence on a walkway. Stepping targets will be in form of a projected light on the walkway. The projected light will be triggered remotely by the tester. The foot clearance obstacle will be 5 cm. The obstacle is designed to detached if subjected to minimal force caused by potential contact. You will walk along an 8 m level walkway that will form approximately 12-16 steps. Two targets and one obstacle will be positioned midway along the walkway. A baseline test condition will measure the average footfall position and determine the precise location of the targets and obstacle. You will respond to a signal presented two steps ahead. The signal will indicate one of four actions that you will be required to execute: (i) step longer than average; (ii) step shorter than average; (iii) step over an obstacle; or in absence of a signal (iv) continue to walk through uninterrupted. Therefore, each block of trials will require you to react according to a binary criterion (dual task) – that is, the negotiation signal will either be present or absent.

You will perform 10 trials per random block condition. Treadmill walking test includes gait adaptation parameters when you will respond to virtual targets displayed on a TV screen located in front of the treadmill. A baseline test of 10 minutes will be recorded.

Study-2 includes 10 training sessions. Each training session will take 45 minutes in total: familiarisation for 2-3 minutes, 4×2-min stepping stones with 1-minute rest after each 2-minute training, 5-minute rest, 2×2-min left and right support limb with 1-minute rest after each 2-minute training, a 5-minute rest, and 2×2-min left and right swing limb with 1-minute rest after each 2-min training. When 10 training sessions are completed, you will complete gait adaptability tests overground and on a treadmill, similar to Study 1.

What will I gain from participating?

We are confident that through participating in this project, you will gain a better understanding of your walking abilities. You will receive a brief 1-page report that describes your step timing and distance measures of gait. We cannot guarantee that you will receive any direct benefit from participation in the study, but there is evidence that gait training with biofeedback can improve gait patterns.

How will the information I give be used?

The effectiveness of training an adaptable gait pattern in diabetic peripheral neuropathy has potential to improve clinical outcomes and prevent falls. The scientific information generated through this project will be shared by publishing the findings in the international research community of gait, balance and posture control. Your personal data will be securely maintained at VU and is strictly confidential. All of the information gathered in this study is highly confidential between yourself and the VU project

investigators (Supervised by Prof Rezaul Begg). Data will be coded and stored under secure conditions. Only group data will be reported and presented via written publications and potential conference presentations. During testing we might ask your permission to take photos or video footage of the experimental set up (marker placement etc) which may be used in research presentations or scientific publications. This will only be done with your prior permission (see consent form), with all images made anonymous to maintain your privacy.

What are the potential risks of participating in this project?

The physical risks associated with this study are the standard risk of injury to walking on a treadmill and the allergic reaction to adhesives. There is a low risk of experiencing muscle and skeletal tiredness and fatigue as a result of extended walking periods at your comfortable walking speed. There is a small risk of stumbling from the treadmill deck during testing; however, you will be wearing a body-weight-support safety harness that will prevent any fall. To obtain motion details of your lower torso, body-mounted markers must be attached to the skin of your waist and they need to be exposed to infrared light emitted from the motion cameras during testing. It is also possible that you may become disoriented whilst walking on the treadmill and observing a display at the front of the treadmill: the visual flow of surrounding objects provides a different sensory experience compared to normal overground walking. It is also possible that you might feel some anxiousness under these walking conditions.

How will this project be conducted?

This project will be conducted in the Victoria University Biomechanics Lab using biomechanics measurement equipment and procedures. You will respond to questions

and minor tests that will screen for identifying health issues that can potentially put you at physical risk during testing, or affect the quality of the data. After we explain the testing procedures, and when you feel that you fully understand the requirements of the research protocol, you will be asked to sign an informed consent document. All data will be collected at Victoria University Biomechanics Lab, Footscray Park Campus. All data will be kept confidential.

Who is conducting the study?

Any queries about your participation in this project may be directed to [redacted] (Ph.D. candidate) by phone ([redacted]) or email ([redacted]) or, you may contact the chief investigator [redacted] by phone ([redacted]) or email ([redacted]).

If you require counselling you can contact our psychological counsellor, [redacted].

If you have any queries or complaints about the way you have been treated, you may contact:

Research Ethics and Biosafety Manager

Victoria University Human Research Ethics Committee

Victoria University

[redacted]

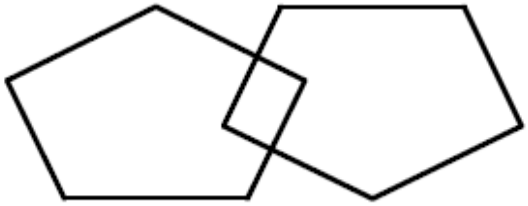
[redacted]

Appendix D. Mini-Mental State Examination

Name:

Instructions: Score one point for each correct response within each question or activity.

Maximum Score	**Score**	**Question**
5		"What is the year? Season? Date? Day? Month?"
5		"Where are we now? State? County? Town/city? Hospital? Floor?"
3		The examiner names three unrelated objects (shoes, sky, newspaper) clearly and slowly, then the instructor asks the patient to name all three of them.
5		"I would like you to count backward from 100 by sevens." (93, 86, 79, 72, 65, …) Alternative: "Spell WORLD backward." (D-L-R-O-W)
3		"Earlier I told you the names of three things. Can you tell me what those were?"
2		Show the patient two simple objects, such as a wristwatch and a pencil, and ask the patient to name them.
1		"Repeat the phrase: 'no ifs, and, or buts.'"
3		"Take the paper in your right hand, fold it in half, and put it on the floor." (The examiner gives the patient a piece of blank paper.)

1		"Please read this and do what it says." (Written instruction is "Close your eyes.")
1		"Write a sentence about anything." (This sentence must contain a noun and a verb.)
1		"Please copy this picture." (The examiner gives the patient a blank piece of paper and asks him/her to draw the symbol below. All 10 angles must be present and two must intersect.) 
30		Total

Interpretation:

- No cognitive impairment (25-30)
- Mild cognitive impairment (20-25)
- Moderate cognitive impairment (10-20)
- Severe cognitive impairment (0-10)

Appendix E. Michigan Neuropathy Screening Instrument

A. History (To be completed by the person with diabetes)

Please take a few minutes to answer the following questions about the feeling in your legs and feet. Check yes or no based on how you usually feel. Thank you.

1	Are your legs and/or feet numb?	☐ Yes	☐ No
2	Do you ever have any burning pain in your legs and/or feet?	☐ Yes	☐ No
3	Are your feet too sensitive to touch?	☐ Yes	☐ No
4	Do you get muscle cramps in your legs and/or feet?	☐ Yes	☐ No
5	Do you ever have any prickling feelings in your legs or feet?	☐ Yes	☐ No
6	Does it hurt when the bed covers touch your skin?	☐ Yes	☐ No
7	When you get into the tub or shower, are you able to tell the hot water from the cold water?	☐ Yes	☐ No
8	Have you ever had an open sore on your foot?	☐ Yes	☐ No
9	Has your doctor ever told you that you have diabetic neuropathy?	☐ Yes	☐ No
10	Do you feel weak all over most of the time?	☐ Yes	☐ No
11	Are your symptoms worse at night?	☐ Yes	☐ No
12	Do your legs hurt when you walk?	☐ Yes	☐ No
13	Are you able to sense your feet when you walk?	☐ Yes	☐ No
14	Is the skin on your feet so dry that it cracks open?	☐ Yes	☐ No
15	Have you ever had an amputation?	☐ Yes	☐ No

Total:

B. Physical Assessment (To be completed by health professional)

1. Appearance of Feet

	Right		Left	
a. Normal	☐ 0 Yes	☐ 1 No	Normal ☐ 0 Yes	☐ 1 No
b. If no, check all that apply:			If no, check all that apply:	
Deformities	☐		Deformities ☐	
Dry skin, callus	☐		Dry skin, callus ☐	
Infection	☐		Infection ☐	
Fissure	☐		Fissure ☐	
Other	☐		Other ☐	
specify:			specify:	

	Right		Left	
	Absent	Present	Absent	Present
2.Ulceration	☐	☐	☐	☐

	Right			Left		
	Present	Present/ Reinforcement	Absent	Present	Present/ Reinforcement	Absent
3.Ankle Reflexes	☐ 0	☐ 0.5	☐ 1	☐ 0	☐ 0.5	☐ 1

	Right			Left		
	Present	Decreased	Absent	Present	Decreased	Absent
4.Vibration perception at great toe	☐ 0	☐ 0.5	☐ 1	☐ 0	☐ 0.5	☐ 1

	Right			Left		
	Normal	Reduced	Absent	Normal	Reduced	Absent
5.Monofilament	☐ 0	☐ 0.5	☐ 1	☐ 0	☐ 0.5	☐ 1

Signature: Total score: /10 points

Appendix F. Consent Form

I, ----------, certify that I am voluntarily giving my consent to participate in the study: Gait Adaptability and Biofeedback in Older Adults with Diabetes being conducted at Victoria University by [redacted]

I certify that the objectives of the study, together with any risks and safeguards associated with the procedures listed hereunder to be carried out in the research, have been fully explained to me by [redacted] and that I freely consent to participation involving the below-mentioned procedures:

- Undergo a health assessment
- Attend two visits to the VU biomechanics lab for two walking tests
- Attend two visits to the VU biomechanics lab for two walking tests
- Attend training sessions if I am eligible
- Allow video recording of my body motion using reflective markers attached to the skin

☐ I agree that photographs collected from this study can be used for purposes related to the public presentation of the research (e.g. scientific conferences and science journals).

I certify that I have had the opportunity to have any questions answered and that I understand that I can withdraw from this study at any time and that this withdrawal will not jeopardize me in any way.

I have been informed that the information I provide will be kept confidential.

Signed:

Date:

Any queries about your participation in this project may be directed to the researcher

[redacted]
Phone: [redacted]
Email: [redacted]

[redacted]
Phone: [redacted]
Email: [redacted]

[redacted]
Phone: [redacted]
Email: [redacted]

If you have any queries or complaints about the way you have been treated, you may contact the Ethics Secretary, Victoria University Human Research Ethics Committee, Office for Research, Victoria University, [redacted].

Shapiro-Wilk tests used to investigate the normal distribution of gait spatiotemporal parameters.

Variable	Group	Baseline	Walk-through	Step shortening		Step lengthening		Obstacle crossing	
				Previous step	Target step	Previous step	Target step	Previous step	Target step
		Sig.	Sig.	Sig.	Sig.	Sig.	Sig.	Sig.	Sig.
Velocity (m/s)	I	0.981	0.634	0.636	0.138	0.578	0.432	0.485	0.767
	II	0.069	0.135	0.135	0.595	0.330	0.667	0.080	0.516
	III	0.231	0.897	0.625	0.828	0.536	0.817	0.528	0.271
Stance (% gait)	I	0.137	0.517	0.586	0.754	0.962	0.181	0.755	0.445
	II	0.706	0.937	0.940	0.426	0.091	0.992	0.848	0.543
	III	0.714	0.937	0.004	0.340	0.980	0.793	0.240	0.861
Swing (% gait)	I	0.718	0.216	-	0.633	-	0.566	-	0.761
	II	0.281	0.603	-	0.374	-	0.346	-	0.881
	III	0.384	0.276	-	0.974	-	0.295	-	0.749
Double Support (% gait)	I	0.338	0.430	-	0.283	-	0.236	-	0.098
	II	0.084	0.495	-	0.298	-	0.402	-	0.809
	III	0.355	0.947	-	0.577	-	0.100	-	0.739
Step Length (% leg length)	I	0.377	0.946	0.380	0.887	0.551	0.220	0.157	0.597
	II	0.568	0.304	0.107	0.502	0.608	0.696	0.237	0.197
	III	0.430	0.076	0.827	0.519	0.265	0.503	0.983	0.095

Appendix G. Gait adaptability training with targeted biofeedback

For the training purpose only runners and a pelvic belt with retro-reflective markers were used (Figure 1). In Nexus Vicon, a model of the pelvis and feet was scaled (Figure 2).

Figure 1. Runners and a pelvic belt with markers.

Nexus Vicon was connected to Visual3D Server software. Marker trajectory data collected by cameras in Nexus Vicon were streamed into customised MATLAB programs using Visual3D Server. The MATLAB programs displayed two real-time line graphs and stepping stone.

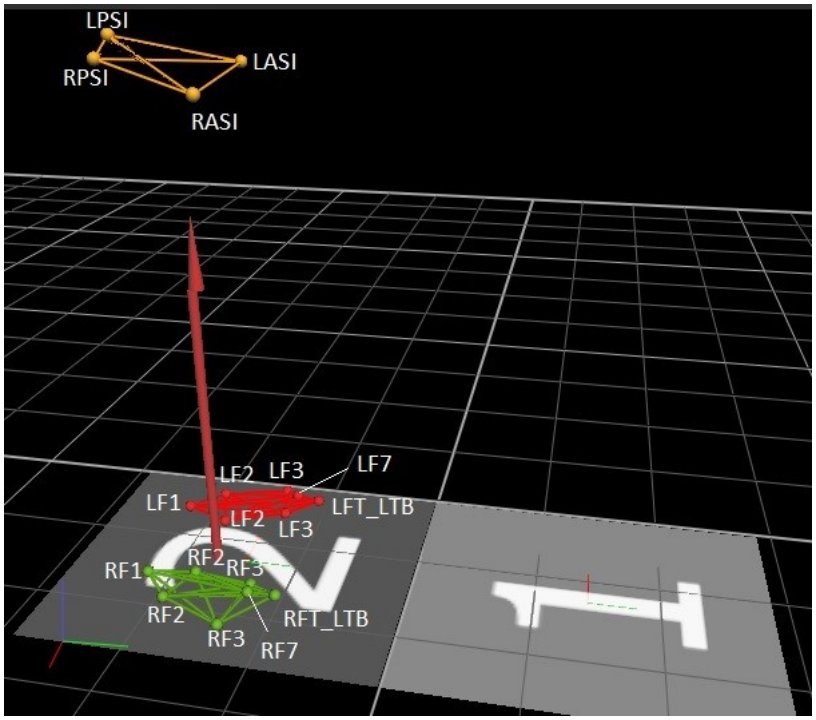


Figure 2. Model of the pelvis and feet in Nexus Vicon.

From a warm-up period of treadmill walking when the monitor was off, mean and standard deviation statistics of gait kinematics served as input for training programs were calculated using MATLAB programs. Targets (lines) appeared above the point of interest to encourage participants to match the point of the interest with the line during each step by lifting one side of pelvis and the foot higher in the Support Limb and the Swing limb respectively.

The Swing Limb graph displayed real-time distances between markers on the first toe a limb (RFT_LTB/RFT_RTB) and Anterior Superior Iliac Spine on the same limb (RASI/LASI) and the Support Limb graph displays real-time distances between the first toe on one limb (RFT_LTB/RFT_RTB) and the Anterior Superior Iliac Spine on the contralateral limb (LASI/ RASI).

The Stepping Stone program displayed foot displacement in the transverse plane during walking (Figure 3). The triangles presented the boundaries of feet during walking.

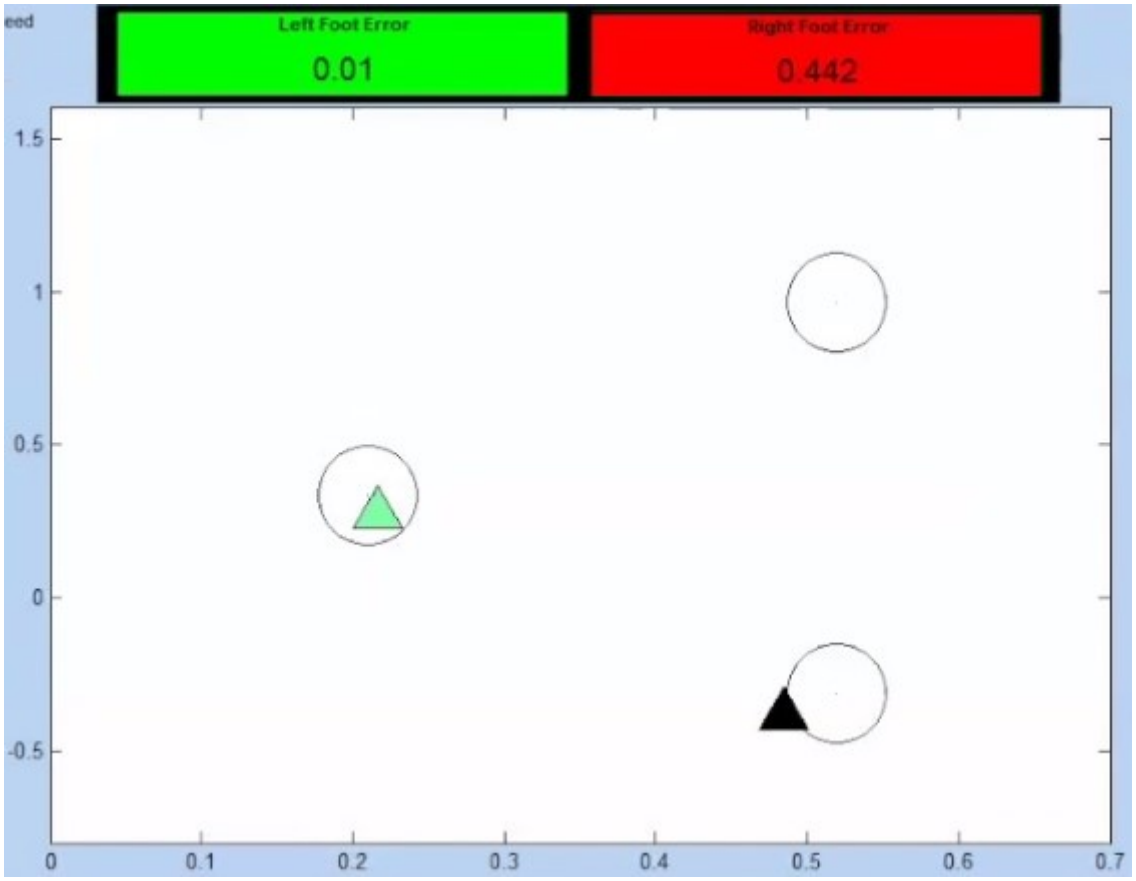


Figure 3. The Stepping Stone program for training foot displacement in the transverse plane.

Green, yellow and red lights were used in the Swing Limb program and the Support Limb program to indicate if the task was satisfied (green), almost satisfied (yellow) or not satisfied (red). In the Stepping Stones, if the foot was placed inside the target (circle), green would be shown as shown in Figure 3.

For training of foot displacement adjustments without biofeedback, the Swing Limb and Support Limb graphs and the Stepping Stones were displayed on the monitor; however, targets in graphs (lines) and in Stepping Stones (circles) were not presented.

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