Human Gait in Virtual Reality:

Analyses & Changes of Gait during Locomotion in Immersive Virtual Environments

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To Mom and Dad...

DECLARATION

I hereby declare, on oath, that I have written the present dissertation by my own and have not used other than the acknowledged resources and aids.

Hiermit erkläre ich an Eides statt, dass ich die vorliegende Dissertationsschrift selbst verfasst und keine anderen als die angegebenen Quellen und Hilfsmittel benutzt habe.

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ABSTRACT

Bipedal walking is generally considered to be the most natural and common locomotion technique for humans in the physical world, and the most presence-enhancing form of locomotion in virtual reality (VR). However, there are significant differences in the way people walk in VR compared to their walking behaviour in the real world.

Understanding real walking in virtual environments (VEs) is important for immersive experiences, allowing users to move through VEs in the most intuitive natural way. Previous studies have shown that basic implementations of real walking in virtual spaces, in which head-tracked movements are mapped isometrically to a VE, are not estimated as entirely natural. Instead, users estimate a virtual walking velocity as more natural when it is slightly increased compared to the user's physical locomotion. Indeed, these findings have been reported in most cases only for young persons, in particular, students, whereas older adults are clearly underrepresented in such studies. However, it appears reasonable to assume that more and more people at different ages will have access to VR, and might use this technology in application scenarios such as physiotherapy, rehabilitation or training. In this context, due to its high personal and economical impact, gait disturbances have become a main focus of interest in Parkinson's disease research. Therapeutic options of medication or deep brain surgery are limited, therefore physical training strategies have evolved to be a focus of interest to improve gait and freezing of gait.

This dissertation focuses on three aspects: (i) analyses and evaluation of the differences of gait parameters between a real and virtual environments, and furthermore investigation of how walking in VEs during (non-)isometric mappings varies across generations, i.e., healthy younger and older adults, with the goal to understand the perceptual and motor differences. (ii) Analyses of changes in velocity over time while walking within a VE and the real world with and without additional cognitive task. In particular, we performed a controlled user study to investigate locomotion adaptation over time due to prolonged exposure to these conditions. (iii) Study and treatment of gait asymmetry. In particular, we developed different virtual walking manipulation techniques to overcome the pathological spatial asymmetry of Parkinson's disease patients with respect to step length.

Zusammenfassung

In der realen Welt wird das Laufen auf zwei Beinen als die natürlichste und verbreitetste Fortbewegung angesehen. In der virtuellen Realität (VR) gilt sie als die Fortbewegungsart, die das Gefühl von Presence am meisten verstärkt. Jedoch gibt es große Unterschiede zwischen der Art, wie Menschen in VR laufen, und ihrem Laufverhalten in der wirklichen Welt.

Für immersive VR-Experiences ist es wichtig, reales Laufen in virtuellen Umgebungen (Virtual Environments, VEs) zu verstehen, um den Nutzern eine möglichst natürliche Fortbewegung durch die VE zu ermöglichen. Vorherige Studien haben gezeigt, dass einfache Implementierungen von realem Laufen in virtuellen Räumen, bei denen mittels Kopf-Tracking gemessene Bewegungen isometrisch auf eine VE gemappt werden, sich nicht komplett natürlich anfühlen. Stattdessen schätzen Nutzer die virtuelle Laufgeschwindigkeit als natürlicher ein, wenn sie im Vergleich zu der physischen Fortbewegung des Nutzers leicht erhöht ist. Diese Beobachtungen wurden jedoch größtenteils nur bei jungen Menschen gemacht, insbesondere bei Studierenden, wohingegen ältere Erwachsene in solchen Studien deutlich unterrepräsentiert sind. Man kann jedoch annehmen, dass mehr und mehr Menschen unterschiedlichen Alters Zugang zu VR erhalten und diese Technologie auch im Rahmen von Anwendungsszenarien wie Physiotherapie, Rehabilitation und Training nutzen werden. In diesem Zusammenhang sind Gangstörungen aufgrund der enormen persönlichen und wirtschaftlichen Auswirkungen in den Fokus der Parkinson-Forschung gerückt. Die Möglichkeiten einer Therapie durch Medikamente oder tiefe Hirnchirurgie sind beschränkt, weshalb Strategien für körperliches Training, mit denen das Gangbild und das Einfrieren des Gangs verbessert werden sollen, zum Forschungsschwerpunkt geworden sind.

Diese Dissertation konzentriert sich auf drei Aspekte: (i) Analyse und Bewertung der Gangparameter, die sich zwischen realen und virtuellen Umgebungen unterscheiden, sowie die Frage, wie sich Laufen in VEs aufgrund (nicht-)isometrischer Mappings zwischen den Generationen - also jüngeren und älteren Erwachsenen - unterscheidet. Ziel ist hier, die perzeptuellen und motorischen Unterschiede zu verstehen. (ii) Analyse von allmählichen Veränderungen der Geschwindigkeit beim Laufen innerhalb einer VR und in der realen

Welt mit und ohne kognitive Aufgabe. Insbesondere wurde eine kontrollierte Nutzerstudie durchgeführt, um allmähliche Anpassungen der Fortbewegung zu untersuchen, wenn die Nutzer diesen Bedingungen länger ausgesetzt sind. (iii) Untersuchung und Behandlung der Gangasymmetrie. Insbesondere wurden unterschiedliche Techniken entwickelt, den Gang virtuell zu manipulieren, um die pathologische räumliche Asymmetrie von Parkinson-Patienten in Bezug auf die Schrittlänge auszugleichen.

TABLE OF CONTENTS

INTRODUCTION

In recent years, a new wave of interest in virtual reality (VR) has evolved as a consequence of enormous advancements in the field of wearable technology, which made VR more accessible to the masses. It started with the availability of affordable, low-cost tracked head-mounted displays (HMDs) such as Oculus Rift, HTC Vive, and Samsung Gear VR. This technology is not only affordable, but also offers a higher resolution, better field of view, lower latency, lower weight, and better ergonomics compared to the high-end solutions available just a couple years ago.

The advances of the technology are based on research and developments efforts of more than 50 years in the field of VR. Stanley Weinbaum described the idea of a pair of goggles that allow the wearer to be transported through a virtual world where even smell, taste and touch can be manipulated (see Figure 1.1a) in a short story called *Pygmalion's Spectacles* [265]. In the 1960s, Morton Heilig patented the first prototype HMD named the so-called *Telesphere Mask* [83], which featured stereoscopic 3D and wide vision with stereo sound, but it did not involve any motion tracking (see Figure 1.1b). In 1965, Ivan Sutherland published *The Ultimate Display* [246], envisioning about the ultimate goal of VR, providing a virtual experience able to recreate the "Wonderland into which Alice walked".

1.1 Challenge of Walking in VR

Currently, in VR it is still difficult to support a fully natural walking experience to VR users. Hence, natural walking remains one of the unresolved challenges facing researchers and developers aspiring to provide users with access to large-scale "wonderlands" as envisioned by Sutherland [246]. Physical walking is considered as the most intuitive way of navigation, and also found to be more presence-enhancing compared to other navigation techniques [255]. Furthermore, it is proven to be superior over other techniques across users' navigational
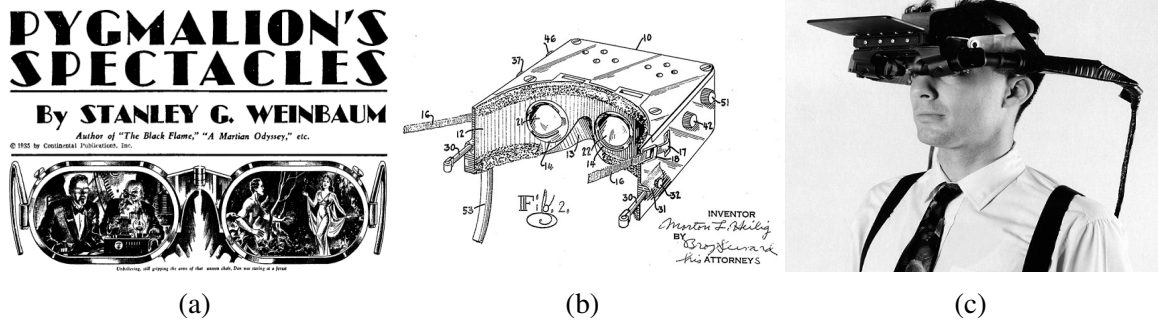
Fig. 1.1 Images from first visioning and real HMDs: (a) Pygmalion's Spectacles [265], (b) Telesphere Mask [83] and (c) Ivan Sutherland's "Sword of Damocles" (copyright by Harvard University).

tasks [208], cognitive map buildings [213], and cognitive demands [140]. However, there are limitations to this technique and sometimes it is impossible to use real walking in immersive virtual environments (VEs) [269]. The main limitation is the size of the tracked area. If the VE is larger than the physical tracked area, the user may eventually walk outside the physical space while trying to reach far spaces in the VE. A variety of locomotion techniques have been proposed (see Chapter 2) to cope with such physical restrictions, including walking in place [232, 229], redirected walking [195] or omnidirectional treadmill [35].

1.2   Research Questions

The initial motivation when I started my Ph.D. was aimed to allow users learn motions within a VE with a various scenarios. These are otherwise difficult to experience in the real world, for learning motion, often required in many different application domains such as sports, healthcare and clinical therapies. Motion profiles in these domains are often practiced with minimal interaction with the environment. That is, the user concentrates more on the overall form of the motion. Thus, the motion constraints we need to consider only come from the relative positions of the body segments. This will enhance learning of complex motion performed by users, implemented by using motion capture-based VR to provide a self-representation, which could be seen through an HMD. Particularly, this generally means that principles of motor learning can be well applied in VR training by providing goal-oriented, repetitive and varied practice that is adjusted to the abilities of the user [129].

Consequently, effective VR systems often require several characteristics, such as interaction, locomotion, audio, visual and task design [16]. Each of these components is important for a good VR experience. Specifically, locomotion is one of the most common and impor-

tant tasks in 3D VEs [16]. Locomotion interfaces based on real walking in VEs support a veridical model of reality, and have been proven to be beneficial for many applications, such as training, rehabilitation, or entertainment [239]. However, recent research has increasingly focused on the use of VR in rehabilitation, including to enhance walking [153, 151, 38, 37]. Rehabilitation interventions in VEs can manipulate practice conditions to engage motivation, motor control, cognitive processes and sensory feedback-based learning mechanisms [132]. Although task-specific training is often used in clinics, it may not be possible to configure the clinical setting to recreate the environmental contexts and challenges that patients might experience in the real world [26, 27, 134, 249].

Researchers and practitioners often observe that simple implementations of virtual walking, in which head-tracked movements or gaits are mapped isometrically to changes of the virtual camera by means of a one-to-one mapping [238]. While it seems easy to implement, previous experiments found that such isometric mappings are often not perceived as entirely natural or realistic by users [237]. Instead, users estimate a virtual walking velocity as more natural when it is slightly increased compared to the user's physical body movement. It has been found that perceptions in VR differ significantly from what we would perceive in similar situations in the real world [101, 102, 135, 198]. These perceptual-action differences have to be evaluated and analyzed, causes need to be understood and eventually fixed in order to support applications that require spatial impressions that match those in the real world, such as for motor rehabilitation applications, which are particularly concerned with a patient's gait [119, 65].

However, even within large-scale tracking spaces there are perceptual and kinematic constraints that differ from those experienced during real world walking behaviors [154, 156]. Therefore, when evaluating immersive VEs that allow for locomotion, and especially when developing a realistic walking interface to be used with VR, it is important to recognize the complex biomechanics associated with human locomotion. The human body uses various sources of sensory information by dividing them into three general categories [261]: external (vision, audition, somatosensory), internal (vestibular, kinesthetic) and efferent (efference copy, attention) to coordinate a rhythmic activation patterns of the muscles, of the limbs and of the trunk, and maintaining dynamic stability during locomotion [181]. Since the perception of the VE space could depend on some physiological information, it is important to investigate the differences between gait parameters (such as velocity, step length, step width, etc.) across different age groups in an attempt to describe walking in the real world versus within VEs. The behavior of most people is different when walking in a VE than in the real world, whereas the question remains as to whether people walking within VEs show lower stability than during walking in the real world, and in how far differences between

younger and older adults can be found presuming that these differences are exist. This inspired the following two research questions in this dissertation:

- Q1: *Which gait parameters within a VE are different from those performed in the real world?*

- Q2:*Which gait parameters between younger and older adults are different while walking in VR?*

Previous studies have shown a significant decrease of some gait parameters in the VE, in particular, velocity and step length [111, 107, 110, 154, 156, 89, 90]. However, those studies have only considered short periods of walking. In contrast, many VR applications involve extended exposures to the VE and often include additional cognitive tasks such as way-finding. Hence, it remains an open question whether a user's velocity during VR walking will further slowdown over time or if people will eventually speed-up and adapt their velocity to the VE and move with the same speed as in the real world. This led to the following general research question:

- Q3: *Is it possible to reduce gait differences while walking in the real and virtual world?*

The emergence of VR-based treatments as a rehabilitation technology has provided important insights for developing potential movement therapies for patients with neurological conditions, such as Parkinson's disease (PD). Specifically, freezing of gait (FOG) is a common symptom of PD, characterized by a progressive shortening of step length, and eventually the total inability to initiate and maintain walking. Moreover, it has been hypothesized that FOG may be a manifestation of asymmetrical and uncoordinated bilateral motor performance of gait [60, 188]. Although the mechanisms underlying FOG are poorly understood [79], FOG is associated with decreased gait rhythmicity, increased gait asymmetry, and bilateral dyscoordination of left-right stepping. [189, 275, 190]. However, such hypothesis-driven VR tasks might be useful for rehabilitation interventions in VEs to improve the spatial symmetry in PD by equalizing step length, the equalization of gait asymmetry has been applied as a motor learning principle. In particular, different virtual walking manipulation techniques were performed to define the method with the best possible equalization of the pathological gait asymmetry in view of step length. In order to carry out our investigation we set out the following research question:

- Q4: *Can VR be utilized to manipulate gait characteristics to achieve gait symmetry in Parkinson's disease patients?*

1.3   Publications

The main contributions of this dissertation have been published in peer-reviewed international journals and conferences. The following list of publication on which this dissertation is based:

Journal Articles

- **Janeh O.**, Langbehn, E., Steinicke, F., Bruder, G., Gulberti, A. and Poetter-Nerger, M.: Walking in virtual reality: Effects of manipulated visual self-motion on walking biomechanics. ACM Transactions on Applied Perception (TAP) 2017, 14(2), (pages 15).

- **Janeh O.**, Bruder, G., Steinicke, F., Gulberti, A. and Poetter-Nerger, M.: Analyses of gait parameters of younger and older adults during (non-) isometric virtual walking. IEEE Transactions on Visualization and Computer Graphics (TVCG) 2018, 24(10), 2663-2674.

- **Janeh O.**, Fründt, O., Schönwald, B., Gulberti, A., Buhmann, C., Gerloff, C., Steinicke, F. and Pötter-Nerger, M.: Gait Training in Virtual Reality: Short-Term Effects of Different Virtual Manipulation Techniques in Parkinson's Disease. Cells 2019, 8(5), (pages 15).

Conference Papers

- **Janeh O.**, Katzakis, N., Tong, J. and Steinicke, F.: Infinity Walk in VR: Effects of Cognitive Load on Velocity during Continuous Long-Distance Walking. In ACM Symposium on Applied Perception (SAP) 2019, (pages 9).

Other

- **Janeh O.**, Langbehn, E., Steinicke, F., Bruder, G., Gulberti, A. and Poetter-Nerger, M.: Biomechanical analysis of (non-) isometric virtual walking of older adults. In IEEE Virtual Reality (VR) 2017, 217-218.

Co-authorship

The following publications are not part of this dissertation, but I contributed to parts of the implementation, writing of paper sections, or supervision:

- Salah, B., **Janeh O.**, Bruckmann, T., Noche, B.: Improving the performance of a new storage and retrieval machine based on a parallel manipulator using fmea analysis. IFAC-PapersOnLine 2015, 48(3), 1658-1663.

- Salah, B., **Janeh O.**, Noche, B., Bruckmann, T., Darmoul, S.: Design and simulation based validation of the control architecture of a stacker crane based on an innovative wire-driven robot. Robotics and Computer-Integrated Manufacturing 2017, 44, 117-128.

- Zhang, J., **Janeh O.**, Katzakis, N., Krupke, D., Steinicke, F.: Evaluation of Proxemics in Dynamic Interaction with a Mixed Reality Avatar Robot. International Conference on Artificial Reality and Telexistence and Eurographics Symposium on Virtual Environments (ICAT-EGVE) 2019.

- Freiwald, J. P., Ariza, O., **Janeh O.**, Katzakis, N., Steinicke, F.: Walking by Cycling: A Novel In-Place Locomotion User Interface for Seated Virtual Reality Experiences (**submitted**)

1.4   Thesis Overview

The remainder of this dissertation is structured as follows. Chapter 2 presents commonly used virtual walking techniques for VR. Chapter 3 gives an overview about the fundamentals of human gait, and the terminology in biomechanics of walking. Chapter 4 introduces a quantitative analyses of gait parameters on younger adults during (non-)isometric virtual walking, whereas Chapter 5 provides a quantitative analyses of gait parameters for older adults. In order to reliably evaluate locomotion performance, Chapter 6 presents a controlled user study to analyze changes in velocity over time while walking overground within a VE and the real world with and without a cognitive task. Chapter 7 focuses on evaluating different virtual walking modulations in patients with Parkinson's disease and provide an insight into future rehabilitative VR training for gait symmetry. Chapter 8 discusses the contributions of the presented papers in this dissertation. Finally, Chapter 9 summarizes the main results of this work, putting the findings into a broader context for future research.

VIRTUAL WALKING TECHNIQUES

As in the real world, most immersive virtual environments are usually suitable to be explored by walking. However, allowing VR users unconstrained walking requires huge tracked free-space areas. In particular, in a VE the space may have infinite size and the user should be able to walk and explore that space freely. However, in real physical spaces users have constrained space. If the virtual space and the real space have similar sizes, a one to one mapping can be used for navigation, but if the virtual space is larger than the real space, the users may eventually walk outside the real tracking space. This interrupts the tracking and may breaks in presence and lead to reduce user experience [277]. To overcome this limitation, some techniques have been developed to enable users to explore larger VEs with real walking. In this Chapter, we summarize some of these most fundamental approaches.

2.1 Walking in Place

In *walking-in-place* (WIP) interfaces, users perform stepping-like movements without forward motion of the body, but a virtual forward motion is induced instead. In this technique, users make body gestures similar to real world walking, without actually moving with respect to the physical environment. This way, users can walk virtually and explore a larger virtual environment. Important advantages of WIP technique include: cost effectiveness [57], naturalness [255], stronger feeling of presence and easier to learn compared to other approaches [232], and proprioceptive feedback similar to real walking [231]. However, since displacement in the real world is prevented with WIP technique, vestibular feedback as in real walking is not possible. One of the first scientific implementations of the walking in place technique was published by [232, 229]. In that work, head movements were analyzed while performing WIP gesture, and virtual walking was triggered by the movement of the head. The latency was substantial; the system required four steps in place to start the virtual

(a)   (b)   (c)
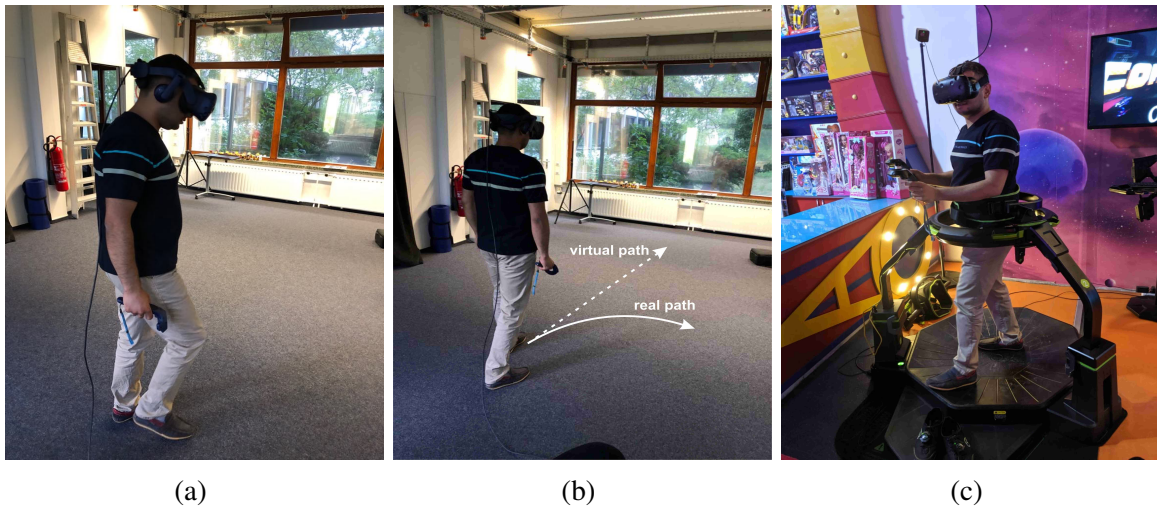
Fig. 2.1 Images for virtual locomotion interfaces: (a) Walking in place, (b) Redirected walking and (c) Omnidirectional treadmill.

walking, since false-positive steps (moving viewpoint when the user is not walking in place) were considered more confusing than a late start. Similarly, the system looked for no steps for two cycles to stop the virtual walking. Since then, different aspects of the walking in place technique have been examined, such as step detections, start and stop latency [57], and smooth motion [270].

Wendt et al. [266] proposed system used a biomechanical state machine to control the virtual walking, and found more consistent output speeds compared to a study by Feasel et al. [57]. A similar study by Kim et al. [121] have proposed a technique that triggers WIP technique using the inertial sensors embedded within two smart phones attached to the user's ankles in order to track leg movement in real time. Usually, most WIP techniques rely on the same gesture for input, a so-called *stepping gesture*, similar to soldiers marching in place. Nilsson et al. [172] performed a study comparing this gesture to two alternative gestural inputs: a gesture where the user alternately bends each knee, thus moving the lower leg backwards, and a gesture where the user in turn taps each heel against the ground without breaking contact with the toes. Furthermore, the perceived required physical effort for the tapping gesture was closer to real walking. In another study, some of those authors examined two more input gestures, i.e., hip movement and arm swinging [173]. The results showed that arm swinging was perceived as natural as the original WIP technique. Moreover, Langbehn et al. [127] have proposed WIP technique that involves a novel way of scaling the speed derived from the steps in place, i.e., the user is able to increase the speed by leaning the torso forward.

2.2   Redirected Walking

Redirected Walking (RDW) enables users to explore a virtual world that is considerably larger than the tracked real world [237]. The idea is that users walk on different paths in the real world, which may vary from the paths they perceive in the VE [22, 258, 171]. For instance, using curvature gains the user effectively starts walking in small circles in the physical space while having the illusion of being able to walk straight in the VE [195]. It is found that, when only visual input is supplied, people can successfully estimate the amount of change in direction but not the path they followed [130]. This makes it possible to manipulate the visual flow to keep the users in the tracking area without being able to notice the manipulations if a physical space of at least $45m^2$ is available [237]. These experiments have been replicated with different settings and extended several times [20, 62, 124]. For instance, Grechkin et al. [75] found that an area of approximately $25m^2$ can be sufficient for unlimited straight walking in a VE.

However, with RDW techniques large-scale VEs can be explored within a smaller tracking area. There are some variations of RDW techniques, and different taxonomies have been proposed. Steinicke et al. [237] proposed a classification based on the types of gains applied: translation, rotation or curvature. Suma et al. [240] proposed a different classification based on the geometric flexibility, the detectability of the technique and the continuity. In this taxonomy, the repositioning and reorientation techniques can either be overt or subtle according to the detectability, and either continuous or discrete according to the gain application. Bruder et al. [20] examined the limits of the gains for individuals using an electric wheelchair controlled by a joystick. The possible range for the gain values was found to be larger for such redirected driving. Recent work by Zhang et al. [278] has examined motion detection thresholds in a large VE for the purposes of improving a 360° camera telepresence robot by real walking. They found that participants could not discriminate between real and telepresence movements (i.e., translation and rotation) when translation gains are down-scaled by 6% and up-scaled by 10%, and rotation gains are about 12% less or 9% more than the actual physical rotations. This indicates that observers in this particular setup were indeed sensitive to motion discrepancies.

Redirection algorithms can also be altered to involve passive haptic feedback objects [236, 238]. A proxy object in the real environment representing virtual objects with similar size, shape and surface structure can support passive haptic feedback to the users. Although more difficult to utilize, such passive haptic feedback improve the VR experience significantly [100]. Other RDW techniques use a visuo-haptic interaction to modify the human spatial perception, such as [241, 145], to provide a sensation of walking in unlimited VR

space in spite of walking in a limited real space. In these systems, since the users actually move their bodies in space, both motor commands and proprioceptive as well as vestibular information from the body movements can be utilized. Another technique for exploring architectural 3D models scales the virtual room to fit into the real room, so that users can feel the real walls when they reach to the virtual walls [23]. In this study, an intense redirection was used to force users go through a virtual door in a virtual wall, so that they did not collide with the real walls.

Recently, novel RDW techniques consider perceptual masking effects like saccades, blinks, and other perceptual suppressions. In this context, Sun et al. [242] enhance redirected interaction by detecting saccades and amplifying redirection during the events without introducing virtual scene warping. Another work by Langbehn et al. [128] conducts perceptual experiments to measure translation and rotation thresholds during eye blinks to facilitate RDW.

2.3 Locomotion Devices

Treadmills are allowing navigation of large-scale VEs via walking movements made within a limited space. Seminal work in this field was reported by the *Walkthrough* project [19], which supported unidirectional movement, and the user could rotate by using a steering bar similar to a bicycle. It allows for walking in one direction, but severely restricting the possibilities for navigation through VEs [234]. Three generations of locomotion devices were developed for the U.S. Army's Dismounted Infantry Training Program [35]. The Uniport (Figure 2.2a) was the first device built for lower body locomotion and exertion, which did not feel natural and did not allow for making sidesteps. The second, Treadport (Figure 2.2b) is based on a standard unidirectional treadmill with the user being monitored and constrained from behind via a mechanical attachment to the user's waist. It was better compared to the first generation in which allowed for more natural locomotion, but was still limited to one direction of movement. The third generation system was the omnidirectional treadmill (Figure 2.2c) that enables locomotion in any direction of travel. The system consisted of 2D rotary motors that moved the treadmill belts to keep the user in the same place. The study showed that accurate user tracking and precise control over the speed of the belts were critical for usability of the system. Otherwise users experienced uncomfortable sudden movements. A similar system was developed in later studies and compared a 3DOF motion platform with controller-based locomotion [35, 103]. In more recent studies, an improved omnidirectional treadmill so-called *CyberWalk* was compared with real walking [223, 235], which allows for natural walking in any direction through arbitrarily large-scale VEs. The CyberWalk needed
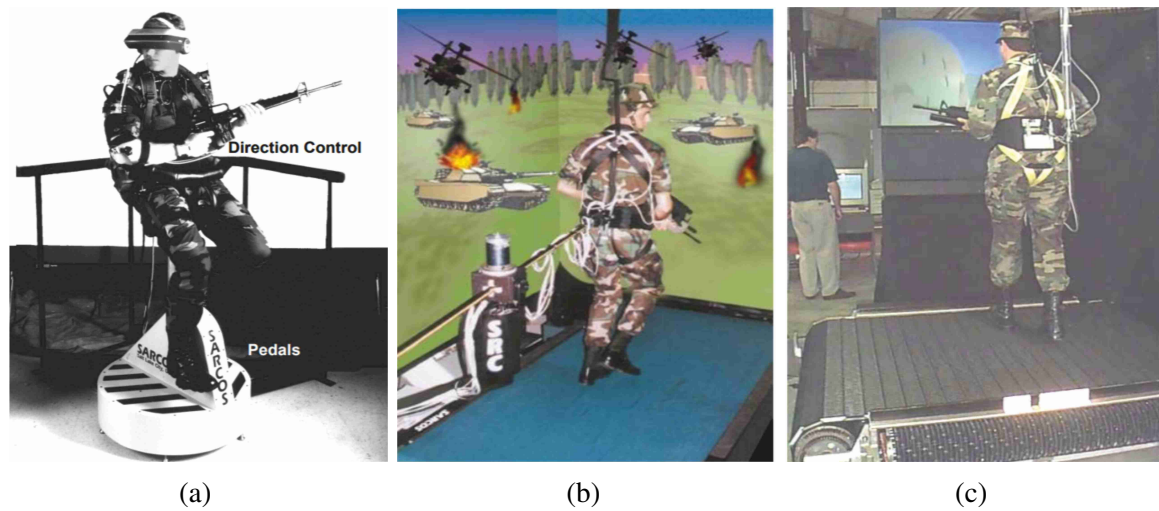
Fig. 2.2 Images for treadmill locomotion interfaces (adapted from [35]): (a) Soldier on UniPort, (b) Soldier on TreadPort and (c) Soldier on the Omnidirectional treadmill.

to ideally be large enough to accommodate a gradual accelerations on the motion platform to keep the user at its center. Although the system was found to be effective in locomotion in VEs, it is extremely expensive to maintain and difficult to adjust in the real space [64].

Furthermore, there are some atypical approaches to locomotion in this category. One of these studies was so-called *Cybersphere* [59]. The authors used a large sphere in which the user could walk, run, jump or crawl freely in any direction to explore an infinite VE. Another similar product, which was commercialized, is called *VirtuSphere* [147]. The VirtuSphere was designed to work with HMDs that enables users to walk in all directions by placing them inside a large, rotatable, hollow sphere. Due to the sphere having its own large mass, it will not stop, start, or change directions with a high degree of responsiveness, and users must essentially re-calibrate their movements to adjust for the movement of the surface under their feet. Another interesting approach to locomotion in VEs called *String Walker* [104]. In this approach, each foot was attached to four motor pulleys with strings. Once a forward motion was detected, the strings pulled the user to the center. This information was gathered with a touch sensor placed on each foot. It detected stance phase and swing phase of walking. The tension was only applied when the foot was on the ground. The motor-pulley mechanisms are mounted on a turntable driven by a motor when the walker changes direction of walking, the turntable is activated to follow the direction of the walker.

(a)	(b)	(c)
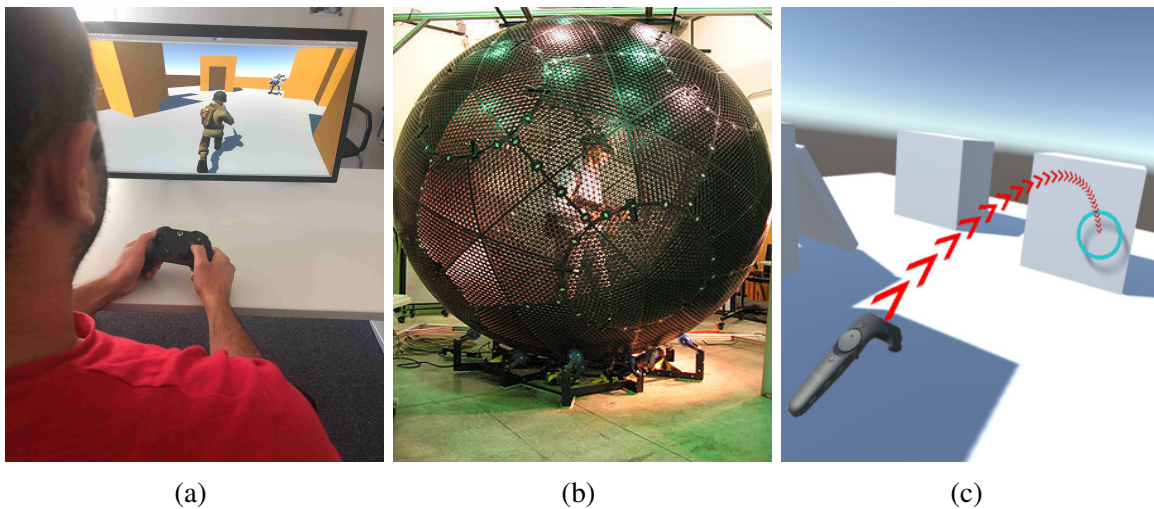
Fig. 2.3 Images for controller-based walking techniques: (a) joystick-based locomotion, (b) Virtusphere and (c) Teleportation.

2.4 Controller-based Virtual Walking

Manual devices such as joysticks, keyboards and VR controllers are widely available, which allow to perform walking inside the VE by involve user's hands and arms [36, 138]. Such joystick-based walking was compared with real walking using different display types (CAVE vs. HMD) [76]. In this study, users performed perceptual-motor coordination tasks with different locomotion techniques. The results show that different velocity controls of each locomotion technique affect the timing and success rate of actions. In real walking, the speed can be controlled easily whereas with a joystick an almost constant speed is provided. Another study by Peck et al. [184, 183] compared joysticks with other locomotion techniques in a virtual maze environment. They found that participants, who used joystick-based walking performed significantly worse than participants who used RDW or WIP. Furthermore, joystick- and keyboard-like devices were inferior for controlling spatial orientation compared to RDW techniques [210]. Riecke et al. [202] compared real walking and joystick locomotion with an additional alternative of real rotation with joystick-based walking. They found that combining real rotation with joystick-based walking produce similar task performance scores as real walking. The results show that large tracked areas are not required for reasonable navigation performance in VR. On the other hand, Nabiyouni et al. [162] compared joystick to a real walking and VirtuSphere (Figure 2.3 (b)); joystick received better results than VirtuSphere in terms of fatigue, ease of learning, ease of walking and precision. The authors concluded that well designed low fidelity locomotion techniques such as joysticks often give better results compared to designs with moderate interaction fidelity like VirtuSphere.

Alternative locomotion techniques have been developed using VR controllers such as teleportation [18]. With teleportation the user's virtual viewpoint is moved while the user itself stays at the same position and orientation in the physical space. Bolte et al. [13] developed the so-called *jumper metaphor* that uses the head direction to select the destination and a physical jump of the user to trigger the teleportation. Another work by Bozgeyikli et al. [18], utilizes gesture-based interaction to point to where the user wants to go, and the main motion takes place through teleportation. In their work, teleportation was compared to WIP and joystick regarding usability. Results show that teleportation is subjectively preferred as a user friendly locomotion technique. However, an extended version of this teleportation technique for which it was possible to set a certain target direction into which the user should face after the teleportation, showed a decrease of the user experience. Bolte et al. [13] compared teleportation to real walking and to the jumper metaphor. The result shows that teleportation and jumper metaphor are more effective techniques than real walking. Furthermore, in a CAVE setup, Freitag et al. [61] compared teleportation to joystick and real walking with portals that were used to reorient the user in the tracking space. Teleportation was faster than real walking, but led to an increased loss of orientation compared to joystick. They could not find any differences between teleportation and real walking concerning motion sickness.

In light of the above, most users are familiar with devices based locomotion and do not need extensive training to use them. However, using these devices lack realism in interaction and reduce the user's sense of presence in immersive applications.

FUNDAMENTALS OF HUMAN GAIT

Human gait refers to the repetitive locomotion pattern of how a person walks. Normally, gait is a very efficient biomechanical process, requiring relatively little energy [137]. Although, the process appears automatic and easy, gait is actually a complex and high-level motor function [137]. In order to analyze and evaluate how a person walks, it is necessary to isolate the shortest, unique, repeatable task during gait. This task is called the (bipedal) gait cycle that requires movements from the right and left sides of the body. In normal gait, the average duration of a gait cycle will be very similar for the left and right sides. In pathological (i.e., abnormal) gait, there may be a pronounced difference between the two sides, leading to arrhythmic gait patterns [254]. In this chapter, we summarize the key concepts of human gait and the terminology used to describe it.

3.1 Phases of Gait Cycle

A gait cycle begins when the heel of one foot touches the ground and ends after the leg and body have advanced through space and time and the heel of that same foot hits the ground again. Realizing aspects of the gait cycle such as phasic, time, spatial and pressure measures, which can be measured and utilized to determine the quality of a person's gait. The cycle includes a period when the leg is in contact with the ground, which is followed by a period when it is advancing through space. Because of the dynamic and continuous nature of walking, the gait cycle is described as occurring between 0% and 100% (Figure 3.1). It can be distinguished into two primary phases: (i) the stance and (ii) swing phases, which alternate for right and left lower limbs.

- *Stance phase* describes the portion of the gait cycle when the foot is in contact with the ground, which makes up to 60% of the gait cycle. Within a stance phase, the double support represents approximately 20% [3, 99], and single support represents
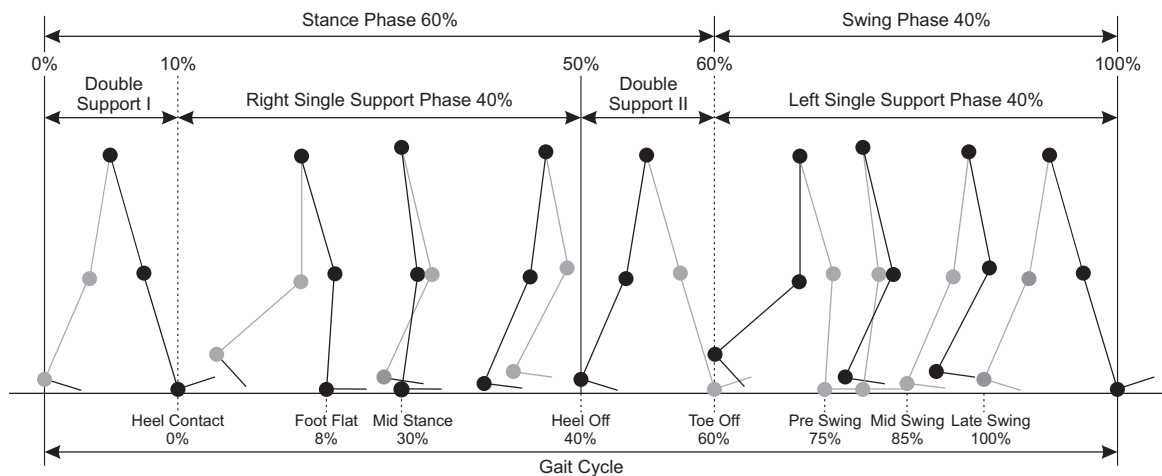
Fig. 3.1 Phases of gait cycle and their proportions as percentages of gait cycle [167].

approximately 40% of the gait cycle [12]. Therefore, when a foot is in a swing phase the other foot should be in a single support phase. When a foot is in a stance phase, it goes through a double support phase 10% of the initial stance phase, a single support phase 40%, and another double support phase 10% of the end of stance.

$$Stance\ Phase = 2 \times Double\ Support + Single\ Support \tag{3.1}$$

- *Double support* denotes the amount of time that a participant spends with both feet on the ground during one gait cycle.

- *Single support* describes the time elapsed between the last contact of the current footfall to the first contact of the next footfall of the same foot. It is equivalent to the swing time.

- *Swing phase* is the portion of the gait cycle when one foot is in the air. It is equivalent to the single support time of the opposite foot.

The phases of swing and stance are further divided into eight events during the gait cycle [185]; five of which occur in the stance phase, when the foot is on the ground, and three in the swing phase, when the foot is moving forward through the air (Figure 3.1).

1. *Heel contact* is a short period which begins the moment the heel or another part of the foot contacts the ground (at 0% of the gait cycle).

2. *Foot flat* the period that the entire plantar aspect of the foot is on the ground (at 8% of the gait cycle).

3. *Mid stance* is the point where the body weight passes directly over the supporting lower extremity (at 30% of the gait cycle).

4. *Heel off* describes the instant the heel leaves the ground (at 40% of the gait cycle).

5. *Toe off* describes the instant the toe leaves the ground (at 60% of the gait cycle).

6. *Pre swing* describes the period from toe off to mid swing (at 75% of the gait cycle).

7. *Mid swing* is the period when the foot of the swing leg passes next to the foot of the stance leg (at 85% of the gait cycle). This corresponds to the mid stance phase of the opposite lower extremity.

8. *Late swing* the period ranging from mid swing until heel contact (at 100% of the gait cycle).

3.2 Timing of Gait Cycle

The gait cycle is defined as the time interval between two successive occurrences of one of the repetitive events of walking [268], involving steps and strides (Figure 3.2).

- *Stride time* is the time elapsed between the first contact of a foot and the first following contact of the same foot, which is equivalent to the time taken to make two successive steps.

- *Step time* is the time elapsed between heel contact of one foot and heel contact of the opposite foot.

- *Gait cycle time* is the elapsed time between the heel contacts of two consecutive footfalls of the same foot.

Figure 3.2 shows the left heel contact occurs while the right foot is still on the ground and there is a period of double support between heel contact on the left and toe off on the right. During the swing phase on the right side, only the left foot is on the ground, giving a period of left single support, which ends with heel contact by the right foot. There is then another period of double support, until toe off on the left side. Right single support corresponds to the left swing phase and the cycle ends with the next heel contact on the left. In each double support phase, one foot is forward (i.e., leading foot), having just landed on the ground, and the other one is backward (i.e., behind foot), being just about to leave the ground.
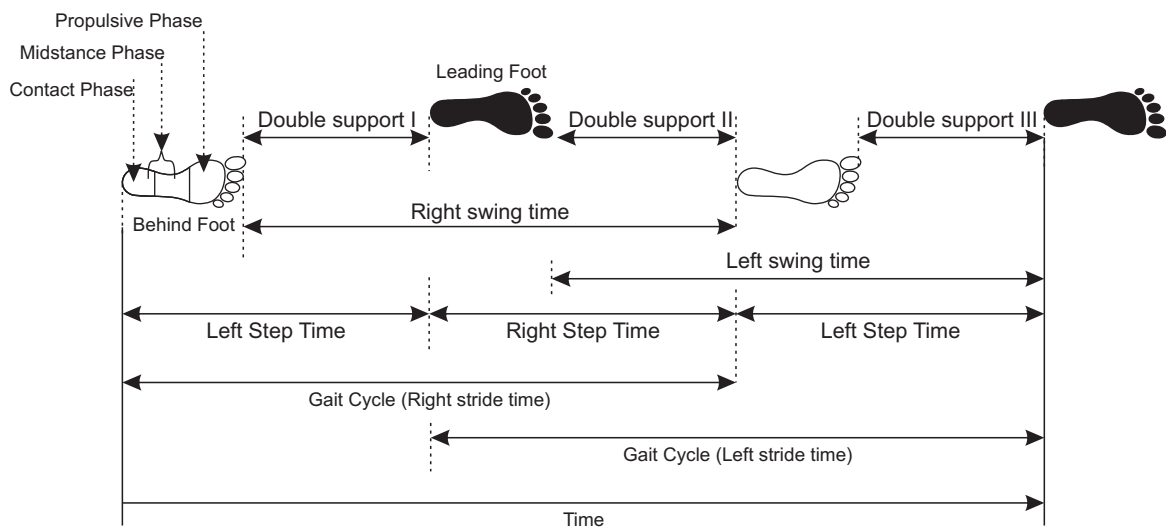
Fig. 3.2 Timing of the gait cycle, starting with right initial contact [268].

- *Swing time* is the time elapsed between the last contact of the current footfall to the first contact of the next footfall on the same foot (i.e., initiated with toe off and ends with heel contact).

- *Double Support time* is the two periods when both feet are on the floor, are called initial double support and terminal double support during one gait cycle. Initial double support occurs from heel contact of one footfall to toe-off of the opposite footfall.

Gaits can also be categorized according to how the foot contacts the ground during the stance phase, specifically which part of the foot first contacts the ground (see Figure 3.2).

1. *Contact phase* begins with heel strike and continues until about 22% of the stance phase. Forefoot loading terminates contact phase.

2. *Midstance phase* It begins with the first sign of forefoot loading. The end of midstance is heel-lift of the support limb. This occurs at about 50% of the stance.

3. *Propulsive phase* is the final 50% of the stance phase. It begins heel lift until toe off.

3.3   Distances of Gait Cycle

While Figure 3.2 emphasise the temporal aspects of human gait, Figure 3.3 illustrates how a set of footfalls can provide useful distance parameters. However, the gait cycle starts when one foot makes contact with the floor, foot contact, and continuing until the next occasion when that foot makes contact with the ground.
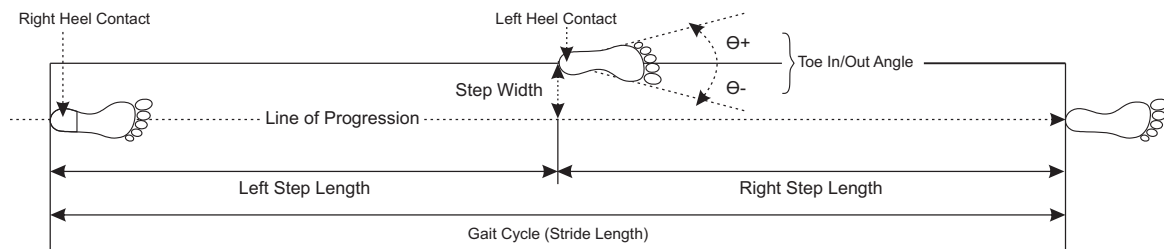
Fig. 3.3 Distance dimensions of the gait cycle, starting with the right foot placement [167].

- *Stride length* is the distance between two successive placements of the same foot. It consists of two step lengths, left and right, each of which is the distance by which the named foot moves forward in front of the other one. It is measured on the line of progression between the heel points of two consecutive footprints of the same foot (left to left, right to right). Given that during a stride the left foot moves in front of the right by right step length and the right then moves in front of the left by left step length then it can be seen that stride length must be the sum of left and right step lengths.

- *Step length* is the distance that one part of the foot travels in front of the same part of the other foot during each step. It is measured from the heel center of the current footprint to the heel center of the previous footprint on the opposite foot.

- *Base of support* or *Step width* is the lateral distance from heel center of one footprint to the line of progression formed by two footprints of the opposite foot. It is usually measured at the midpoint of the back of the heel but sometimes below the center of the ankle joint.

- *Toe in/out* is the angle between the line of progression and the midline of the footprint. In Figure 3.3, theta is the angle between the mid-line of the right footprint and the line of progression. It is zero if the geometric midline of the footprint is parallel to the line of progression; positive, toe-out, when the mid-line of the footprint is outside the line of progression and negative, toe-in, when inside the line of progression.

In pathological gait, it is common for the two step lengths to be different [96], which is known as *gait asymmetry* (see Figure 3.4). However, asymmetry begins to occur with aging, and could be indicative of neurodegenerative diseases [260, 188], such as Parkinson's disease. There are many methods to measure gait asymmetry [273]. The simplest one is to calculate a ratio between the left and right mean values, i.e., $\frac{L}{R}$. The value of 1 represents perfect symmetry; if the left mean is larger, the ratio will be greater than 1; if the right mean is larger, the ratio would be less than 1. Furthermore, gait asymmetry can also be expressed as a percent difference between the left and right mean values [217]:

(a) symmetrical gait

(b) asymmetrical gait
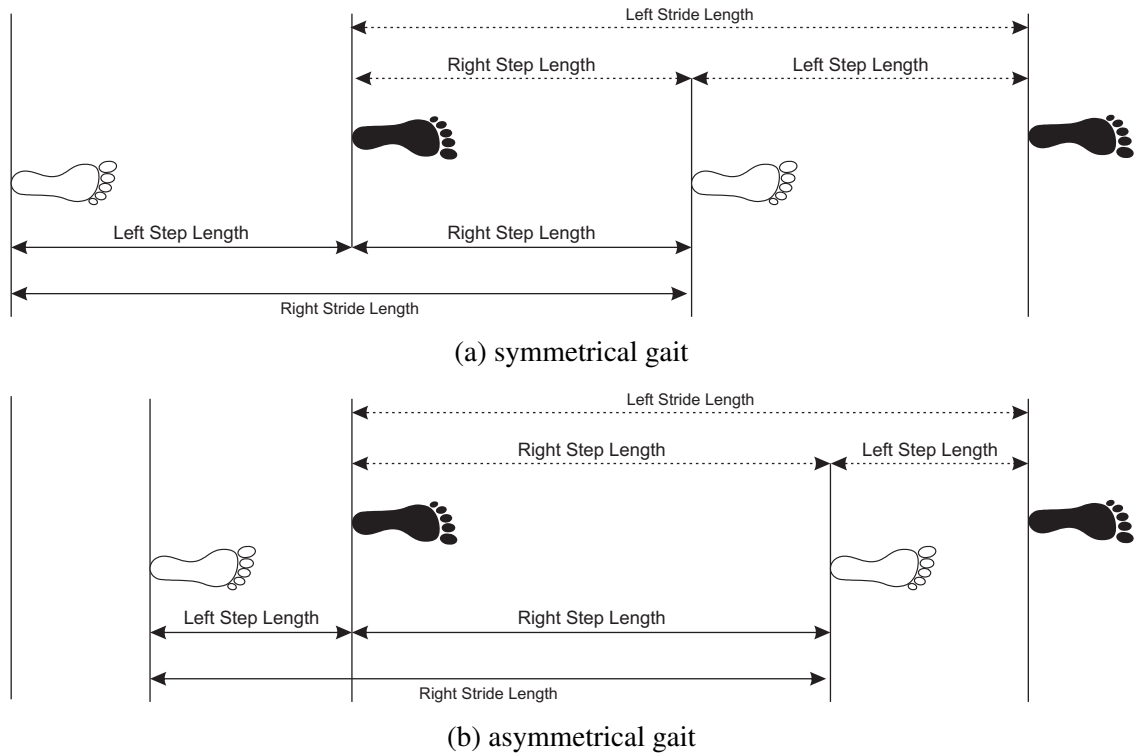
Fig. 3.4 Illustration of step and stride lengths for (a) symmetrical gait, and (b) asymmetrical gait.

$$\left(\frac{(L-R)}{\frac{(L+R)}{2}} \right) \times 100 \qquad\qquad (3.2)$$

where L and R are respectively measurements performed on the left and the right respectively, such as step length or step frequency.

3.4   Walking Pace

- *Cadence* is the number of steps or strides per unit of time, the usual units being steps per minute. However, mathematically, the cadence, which is a frequency (number of events per second), is calculated as the inverse of the cycle time. It is then assigned a factor of 60 to obtain a number in cycles per minute or a factor of 120 for the number in steps per minute. The spontaneous cadence is usually between 98-138 steps/min for women and 91-135 steps/min for men aged 18-49. Women compensate for the shortest step length with a greater frequency of steps. In general, an increased cadence occurs regardless of sex for adults who are shorter than average stature.

- *Velocity* is the distance walked divided by a unit of time in a certain direction. The average velocity is the product of the cadence and the stride length, providing appropriate units are used. The cadence, in steps per minute, corresponds to half-strides per 60 seconds or full strides per 120 seconds. The velocity can thus be calculated from cadence and step length using the formula:

$$Velocity(m/s) = \frac{Step\,Length(m) \times Cadence(steps/min)}{60} \qquad (3.3)$$

The walking velocity thus depends on the two step lengths (i.e., stride), which in turn depend to a large extent on the duration of the swing phase on each side. The step length is the amount by which the foot can be moved forwards during the swing phase, so that a short swing phase on one side will generally decrease the step length on that side. If the foot catches on the ground, this may terminate the swing phase and thereby further decrease both step length and walking velocity.

In pathological gait, the step length is often shortened. When pathological gait affects one foot more than the other, a person will usually try to spend a shorter time on the abnormal foot and correspondingly longer on the normal one. Shortening the stance phase on the abnormal foot means bringing the normal foot to the ground sooner, thereby shortening both the duration of the swing phase and the step length on that side. Thus, a short step length on one side generally means problems with single support on the other side. When making comparisons between persons, particularly with short stature, it is useful to allow for differences in size. This is done by dividing a measurement by some aspect of body size, such as height or leg length, a procedure generally known as *normalization*. It is thus fairly common to see measures such as step extremity ratio, which is step length divided by leg length [244]. Macellari et al. [136] performed a detailed study of the relationships between gender, body size, walking velocity, gait timing and foot placement.

- *Functional Ambulation Performance (FAP) score* is derived by subtracting points from a maximum score of 100 for walking at a self-selected velocity [78]. The lowest possible score is 40, with a range from 40 to 100. Hence, a higher score is better in overall walking performance, and is calculated according to the following equation (see Figure 3.5a):

$$FAP\,score = 100 - (A + B + C) \qquad (3.4)$$

where *A* denotes the average between right and left dynamic base of support during ambulation, *B* is the degree of asymmetry of the participant's gait expressed as the ratio between left and right step lengths divided by participant's leg length, and *C* denotes the

Predetermined gait parameters

step time (ST): left = .64, right = .69
step extremity ratio: left = .55, right = .52
normalized velocity: left = .86, right = .75

Normalization by Leg Length (LL)
Velocity (V) Step Length (SL)

determine deviations from normal range

| dynamic base of support (wide or narrow) | left/right SL/LL asymmetry > .03 | V/LL: (1.1-1.5) SL/LL: (.69-.86) ST: (.5-.63) |

| dynamic step width | differential step extremity ratio | left/right step functions |

| **A/8** | **B/8** | **C/44** |

(a)   (b)

Fig. 3.5 Details of FAP score (a) flowchart of FAP score calculation and (b) calculation for deductions in left and right step functions categories. Person's parameters (example data given in the flowchart) are then plotted for right and left legs and compared with normal range.

relationship of step length/leg length ratios, step times, and velocities normalized for leg lengths (see Figure 3.5b).

These points deducted in the different parts of the FAP score (see Equation 3.4) are determined by the distance between the participant's gait parameters and ranges of predefined values considered as normal for gait at the self-selected velocity [73], e.g., up to eight points are deducted if the dynamic step width is abnormally wide or narrow. Further points can be deducted from a maximum score of 100 (i.e., from 0 to 8 points for right-left asymmetry and from 0 to 22 points for right/left step functions).

- *Step extremity ratio* is defined as the step length divided by the leg length of the same leg.

- *Normalized velocity* is obtained after dividing the velocity by the average leg length and expressed in leg length per second (LL/sec). The average leg length is computed as follow:

$$Leg\ Length = (left + right)/2 \qquad (3.5)$$

(a) Left footfall   (b) Right footfall
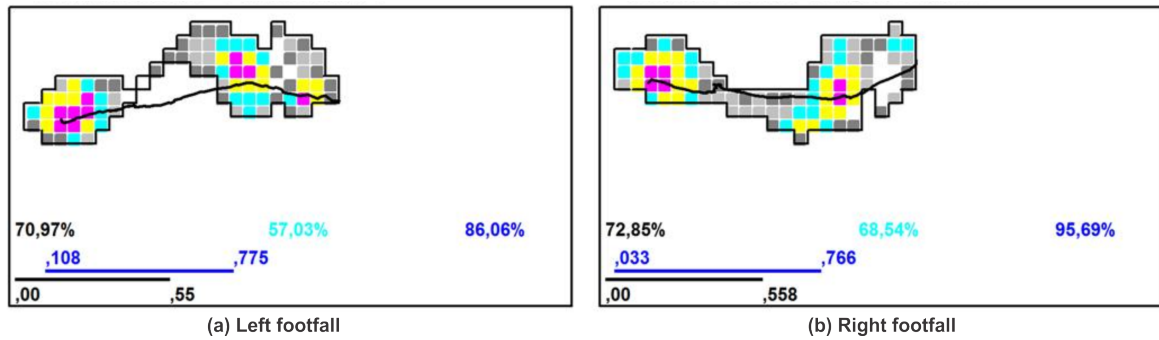
Fig. 3.6 Average pressure map determines the mean of the values measured with a single sensor.

3.5   Pressure Distribution

Pressure distribution refers to the total force applied by the foot to the ground The pressure exerted by a foot activates the sensors as the subject walks over the GAITRite[1] walkway system. The walkway detects the geometry of a step in a two dimensional space and also senses the vertical component of pressure exerted by the subject [166]. The pressure values presented by the GAITRite software are normalized and expressed as a percentage of the maximum pressure [166]. These values are then divided into seven switching levels (see Figure 3.6), i.e., dark gray(1)=lowest, light gray(2), cyan(3), yellow(4), magenta(5), red(6), blue(7)=highest. The switching levels are thus relative pressure values and cannot be converted into normal pressure units (e.g., Pa or mmHg).

The GAITRite software does not present the pressure values for each sensor separately. The footprint is divided into twelve trapezoids, six located in the medial side of the footprint and six in the lateral side. Pressure parameters presented by the software (sectional integrated pressure over time, peak time, area and peak pressure) are calculated for these trapezoids.

$$P_{average} = \frac{1}{n} \sum_{i=1}^{n} P_i \tag{3.6}$$

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[1]GAITRite system provides a simple way to obtain valid, reliable, and objective phasic, time, spatial and pressure measures of gait, in real-time both in clinical and research settings (see Appendix A)).

3.6   Gait in Younger Adults

Normal adults perform the task of walking without significant active thought or effort, which is a process learned and eventually mastered in childhood [82]. The most detailed study by Sutherland et al. [245] has been performed on the development of gait in children in which the gait of small children differs from that of adults are as follows:

- walking step width is wider.

- step length and velocity are lower and the cycle time shorter.

- heel contact being made by the flat foot.

- very little knee flexion during stance phase.

- leg is externally rotated during the swing phase.

- absence of alternate arm swinging.

These differences in gait mature at different rates, implying that changes to reach or maintain competences continue throughout the life span. By looking at the phases of motor development, gait is one of the skills achieved during the so called fundamental movement phase [67], which reaches maturity around age (6–7 years). From 7 to 14 year old and up, children start learning specialized movement that they can apply to specific sports and activities, which ends with the lifelong utilization stage around age of 14 [68]. The cycle time, step length and velocity continue to change with growth, reaching normal adult gait pattern around the age of 15 [268].

The so-called natural gait allows a person to walk forward by alternating propulsive and retropulsive motions of the lower extremities [185], but can also be adapted for lateral movement [150], which performed with the following steps [30]:

- *Step (1)* One leg is lifted off of the ground;

- *Step (2)* With the leg in contact with the ground, the body is pushed forward;

- *Step (3)* The lifted leg is swung forward until it is in front of the body;

- *Step (4)* The walker falls forward to allow the lifted leg to contact the ground;

- *Step (5)* Steps 1–4 are repeated for the other leg;

- *Step (6)* Steps 1–5 are repeated to continue walking.

3.7   Gait in Older Adults

Many investigations have been performed to investigate changes in gait which occur with advancing age, especially by Murray et al. [161], who examined the effects of age on free-speed and fast walking of men up to the age of 87. Another study by Murray [160] examined the gait of women up to age 70, but it did not provide as much information on the effects of age, but generally confirmed the observations made on males. In view of biological aspects of walking, gait performance is determined by continuous, ongoing postural adjustments by several types of control mechanisms. Stereotypical patterns of synergic muscle group activation [40] need to be scaled appropriately by peripheral sensory feedback [40] and centrally generated, anticipatory motor programs [95]. It is proposed that postural alignment requires three different processes [94]:

- Sensory organization and weighting of the orientation senses such as somatosensory proprioceptive, visual and vestibular information

- Motor adjustment processes involved with executing coordinated and properly scaled neuromuscular responses

- Background tone of muscles through which balance changes are compensated.

The process of sensory organization seems to be hierarchically organized at different levels, these systems should be coherent, any conflicting orientation inputs must be quickly suppressed in favor of those congruent with the internal reference, otherwise postural and gait performance worsens [143, 149].

The gait of the older adults is subject to two influences [268]: the effects of age itself and the effects of pathological conditions, such as osteoarthritis and parkinsonism, which become more common with advancing age. The gait of the older adults appears to be simply a slowed down version of the gait of younger adults. Murray et al. [161] were cautious to point out that the walking performance of older men are not comparable to the pathological gait. Generally, the beginning of age-related changes in gait takes place in the decade from 60-70 years of age. There is a decreased stride length, a variable but typically increased cadence and an increase in the step width. Many other changes can also be observed, such as a relative increase in the duration of the stance phase as a percentage of the gait cycle, but most of them are secondary to the changes in stride length, cycle time and step width. The velocity is almost always reduced in older people.

Furthermore, the differences between the gait of the young and the elderly are described by Murray et al. [161], which suggested that the purpose of gait changes in the elderly is

characterized by a cautious attitude of walking, which is essentially an exaggeration of the gait changes which normally occur with age. For instance, decreasing the step length and increasing the step width make it easier to maintain balance while walking. Increasing the cycle time leads to a reduction in the percentage of the gait cycle for which there is only single support, since the increase in cycle length is largely achieved by lengthening the stance phase and hence the double support time. A comprehensive review of the changes in gait with advancing age was given by Prince et al. [193].

3.8   Control of Gait

Bipedal locomotion is accomplished through a complex and coordinated pattern of nerve signals, sent to the muscles (see Figure 3.7), which in turn move the joints, the limbs and the remainder of the body [268, 51]. The control of locomotion involves the use of afferent information from a variety of sources in the visual, auditory, vestibular and proprioceptive systems [41, 248], which create a consistent multisensory representation of a person's self-motion, i.e., acceleration, velocity and walking direction. Modifying the sensory information during the movement can come from either proprioceptive information or efference copies of the motor command during the preparation for motor output [194]. Efference copies are those neural representations of motor outputs that predict reafferent sensory feedback and modulate the response of the corresponding sensory modalities. Also, accessing a copy of the efferent command allows the brain to prepare for the consequences of an intended motion before it has occurred [80].

The voluntary control of movement and high-level modulation of gait patterns is originated at the supraspinal level (see Figure 3.7, top). The latter regulates both the central pattern generator (CPG) and reflex mechanisms [41]. Also at the supraspinal level, information from vestibular and visual systems are incorporated, which are crucial for the maintenance of balance, orientation, and control of precise movement [41]. Efferent stimulation is transmitted through motor neurons to individual muscle groups, which are recruited to effect the movement. Afferent feedback, including that from proprioceptors of the muscles, joints, tendons and skin, is used to directly modulate motor commands via reflex arcs, thus contributing to the efficiency of gait under normal conditions and stabilize posture during unexpected perturbations [253, 126, 28, 170].
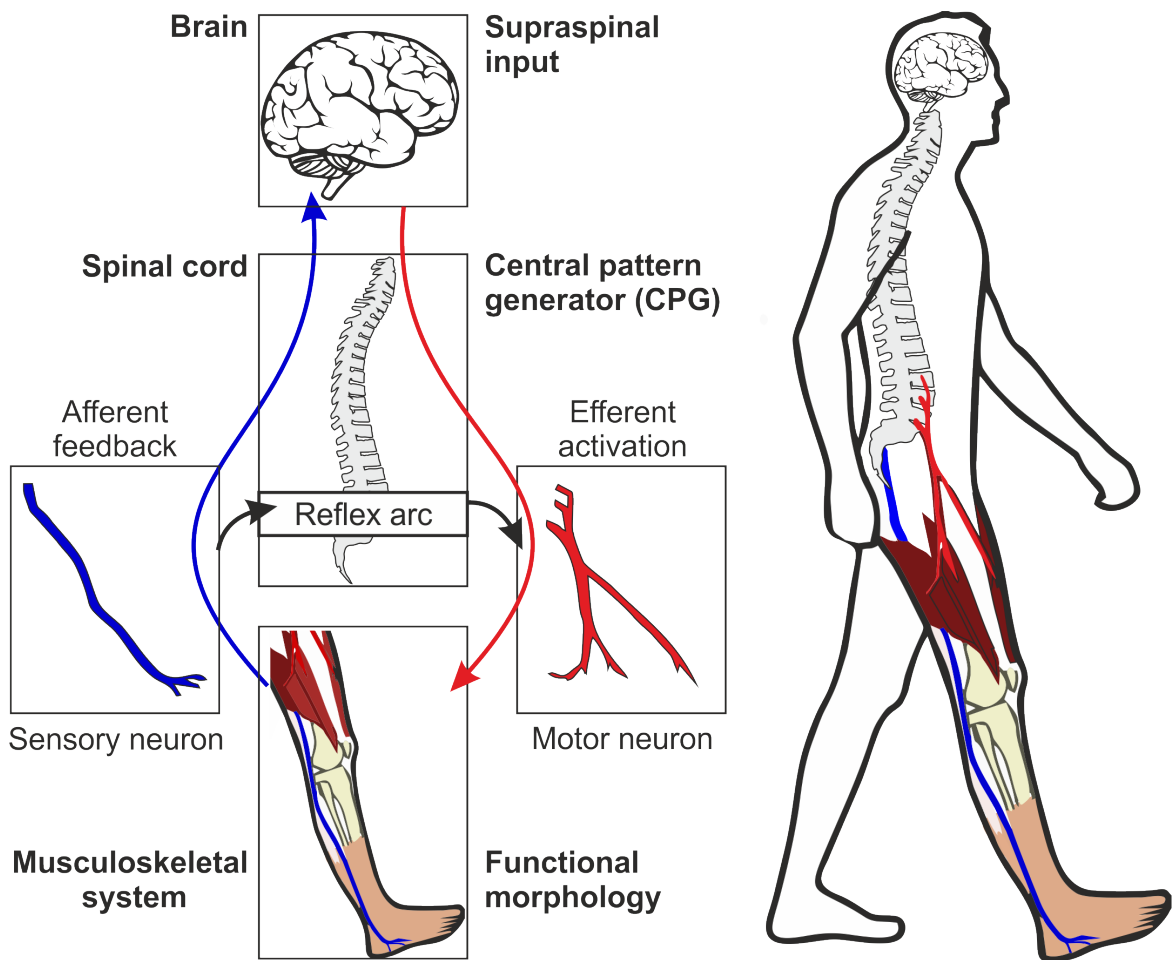
Fig. 3.7 Nominal sensory-motor control loop for human locomotion (adapted from [253]).

VIRTUAL WALKING IN YOUNGER ADULTS

In this chapter, we investigate of the effects of (non-)isometric mappings between physical movements and virtual motions in the VE on walking velocity and biomechanics of the gait cycle described in Chapter 3. Therefore, we performed an experiment in which we measured and analyzed parameters of the biomechanics of walking under conditions with isometric as well as non-isometric mappings with focus on younger adults. The results show significant differences in most gait parameters when walking in the VE in the isometric mapping condition compared to the corresponding parameters in the real world. For non-isometric mappings we found an increased divergence of gait parameters depending on the velocity of visual self-motion feedback. The results revealed a symmetrical effect of gait detriments for up- or down-scaled virtual velocities.

4.1 Introduction

Bipedal walking is often considered the most basic and natural form of locomotion human can perform. Thus, realizing real walking in VEs is essential to support a veridical model of reality (i.e., as in the real world) in a wide range of application domains, such as VR training, rehabilitation, simulation, or entertainment. To gain similar advantages from virtual walking as we gain from walking in the real world, such as navigation performance [211], spatial awareness [212] or presence [255], it is important that virtual walking matches natural walking as close as possible. Discrepancies between visual feedback and the vestibular-proprioceptive system, such as occurring while using walking-in-place [255, 250], treadmills [224, 15] or VirtuSpheres [139], have been hypothesized to cause detriments in walking performance [239].

As explained in Chapter 1 and 2, implementing real walking in VEs typically requires tracking the head movements of a user to change the virtual camera and generate self-

motion feedback, e.g., by means of an *isometric mapping* (sometimes called a one-to-one mapping [238]). In this approach, users may wear position and orientation tracked HMDs that allow the virtual movement to match their physical movement with visual motion cues. A tracked change of the user's position is defined by the vector $T = P_{cur} - P_{pre}$, where P_{cur} is the current position and P_{pre} is the previous position. T is applied one-to-one to the virtual camera, i.e., the virtual camera is moved by $|T|$ units in the corresponding direction in the VE coordinate system. While it seems easy to implement, previous experiments found that such isometric mappings are often not estimated as entirely natural or realistic by users. Williams et al. [272] introduced *translation gains*, to describe the ratio between a virtual translation and the corresponding translation of a user in the real world, i.e., $g_t := \frac{T\text{virtual}}{T\text{real}}$. Translation gains $g_t \in \mathbb{R}$ provide a way to formalize *non-isometric mappings*, in which a translation T in the real world can be mapped to a scaled translation $g_t \cdot T$ in the VE. This is particularly useful if the user wants to explore VEs whose size significantly differs from the size of the tracked space. For instance, if $g_t = 1$ the virtual scene remains stable considering the head's position change. In the case $g_t > 1$ the displacement in the virtual scene is greater than in the lab space, whereas a gain $g_t < 1$ causes a smaller displacement in the virtual scene compared to the displacement in the lab space.

In a psychophysical experiment using a two-alternative forced-choice task [58], a study by Steinicke et al. [237] analyzed at which point of subjective equality (PSE) users estimated virtual translations to match their physical movements. They found that virtual translations had to be slightly increased by 7% over a user's physical movement in order for them to estimate them as identical [237]. Other studies reported similar requirements to up-scale virtual walking velocities over physical movements, although they differed in magnitude, such as up-scaling by 53% [5] or by 36% [46]. Most researchers tried to explain these effects by limitations of the current VR hardware technologies or the subjective state of the users, who often walk more slowly and carefully in VEs than they would in the real world, e.g., due to fear of colliding with unseen walls [5, 46, 237]. In a previous study, it was shown that simple gait parameters like walking velocity, stride length, stride frequency were affected in both healthy younger and older adults when visual flow was manipulated [32]. However, this study was limited to a simple optic flow stimulation in the VE, and did not consider important gait parameters such as step width or double support.

However, while users might estimate a slightly increased virtual walking velocity as more natural, in general we have to consider that these results from perceptual matching tasks and self-adjustments not necessarily lead to changes in the biomechanics of walking that more closely match those of walking in the real world. Thus, it might be detrimental for practical applications of real walking user interfaces. Unfortunately, so far the existing body

of literature does not provide a consistent understanding of the effects of such isometric or non-isometric walking conditions on gait detriments in VEs [154, 159, 239].

In this chapter, we presented an experiment in which investigate the effects of isometric and non-isometric walking with an HMD on gait parameters. The main contribution is analysis of biomechanical walking parameters indicating a closest match between virtual and real walking for an isometric mapping, while non-isometric mappings resulted in often symmetrically-shaped detriments both for up- and down-scaled virtual velocity.

4.2   Related Literature

The study of locomotion and perception are the focus of many research groups analyzing walking in both real and virtual environments [239, 240, 159]. Since the perception of and interaction with virtual worlds may be influenced not only by visual information but also physiological information from the inner body senses, it is important to investigate contingencies that exist among the sensory and motor information that signal self-motion [50] and differences between biomechanical parameters while walking in the real world versus within virtual spaces [154]. In particular, [154] reported that walking parameters may affect a user's perception of virtual space. Differences in biomechanics of walking in virtual worlds have naturally been suggested as a potential factor contributing to the fact that distances in VEs are often overestimated or underestimated in comparison to the real world [198, 135, 101, 102]. Furthermore, Riecke and Wiener [203] found that users have larger difficulties orienting themselves in virtual worlds than in the real world, which might be affected by physiological movement signals and walking parameters as well.

Moreover, several studies investigated the potential to increase the naturalness of virtual walking velocities in VEs [49, 48]. Banton et al. [5] reported that the visually perceived velocity appears too slow compared to the physical walking velocity; and presented experiments investigating the underestimation of visual flow velocities during treadmill walking. They reported that the visual flow velocity had to be increased by about 50% in a VE to appear normal. Notably, the perceived velocity of real walking is influenced by the application of virtual velocities, which produces a discrepancy between the real and virtual velocity [237]. Similarly, users tend to underestimate travelled distances in VEs [63]. Experiments by Steinicke et al. [237] showed that users estimate the virtual distance as smaller than the physical perceived distance against the applied velocity gains. A study performed by [48] suggested that discrimination of visual velocities near walking velocity is enhanced

by the act of walking. Discrimination of slow visual velocities had a negative effect during walking, whereas discrimination of faster visual velocities was improved.

Many researchers [50, 156, 239] have investigated physiological and biomechanical aspects of walking across different samples in an attempt to describe real and virtual environments while walking within a HMD and the real world. Mohler et al. [154] reported that gait parameters (such as velocity, stride length, head angle, etc.) within a VE are different than those in the real world. Furthermore, [225, 156] found that visual information is associated with the control of locomotor behavior. In particular, they found that gait velocity of self-motion is influenced by visual flow. Hollman et al. [89] examined the effect of VEs on gait and found that walking in VEs increases variability in the ground reaction forces (such as weight acceptance peak and push-off peak) through a single footfall, which reflects compensatory efforts to control the body's center of mass over the base of support during locomotion and, therefore, represent gait instability induced by visual stimulation in VEs. However, the behavior of most people is different when walking in a VE than in the real world, whereas the question remains as to whether people walking within VEs show lower stability than during walking in the real world. We believe that it is important to investigate such aspects of human locomotion while walking in the real and virtual world.

4.3   Experiment

In this section, we present the experiment in which we have examined how walking in VEs differs from walking in the real world in terms of the biomechanics and velocity of walking. We tested different isometric and non-isometric walking conditions using the method of translation gains [237] along a straightforward movement path, and we compared the results to a baseline condition while walking in the real world. Prior the experiment, we received approval for the experimental procedure, material and methods by our institutional review board.

4.3.1   Participants

A total of 19 participants (6 female and 13 male, ages 18-38, M=26.8, heights 160-194cm, M=175.8cm, weights 52-87kg, M=69kg) completed the experiment. The participants were students or members of the department of computer science or the department of neurophysiology. Students obtained class credit for their participation. All of our participants had normal or corrected-to-normal vision. During the experiment, six participants wore glasses and two participants wore contact lenses. None of our participants reported a disorder

of equilibrium or vision disorders, such as color blindness or astigmatism. Participants wore an HMD for approximately 30 minutes during the experiment. Nine participants have had prior experience with HMDs before. We measured the leg length of our participants before the experiment (82-98cm, M=89.4cm). We used the leg length of each participant to calculate the FAP [165], which represents a quantification of participants' gait based on a selection of spatio-temporal parameters obtained at a self-selected velocity [74, 78]. The selected parameters are standard velocity normalized to leg length, step and leg length ratio, step time, right–left asymmetry of step length, and dynamic base of support. Participants were naive to the experimental conditions, wearing their normal clothes and performing barefoot walking across a walkway. They were allowed to take breaks at any time between experiment trials in order to minimize effects of exhaustion or lack of concentration. The total time per participant, including pre-questionnaires, instructions, experiment, breaks, post-questionnaires, and debriefing, was one hour.

4.3.2 Materials

We performed the experiment in a laboratory room of 9m × 4m meters in size (see Figure 4.1a). During the experiment, the room was darkened in order to reduce the participant's perception of the real world while immersed in the VE. The virtual space was rendered using Unity3D; a cross-platform game engine with a custom-enabled VR communications and rendering library. The visual stimulus was a 3D laboratory model. The start (green line) and target (red lines) were placed on the floor in front of the participant to indicate the walking distance in the virtual world (see Figure 4.1b). The participants have been instructed to walk from the start line to the target (i.e., stopping between the two target lines). In the VE, forward movement was accompanied by motion of the head-tracked without corresponding movements of the participant's self-representation. For rendering, system control and logging, we used an Intel computer with 3.4GHz Core i7 processor, 16GB of main memory and Nvidia GeForce 780Ti SLI graphics cards.

The participants wore an Oculus Rift DK2 HMD for the stimulus presentation, which provides a resolution of 960 × 1080 pixels per eye with a refresh rate of 60Hz and an approximate 110° diagonal field of view (FOV). We attached an active infrared marker to the HMD and tracked it using an optical tracking system (WorldViz precision position tracking PPT-X4) at an update rate of 60Hz with sub-millimeter precision and accuracy for position data in the laboratory. During the experiment, we provided comfort to the head from the weight of the cables by having an assistant manage the cables for each participant, making their presence transparent. In the real-world conditions without HMD, we attached an active

<div align="center">(a)   (b)</div>
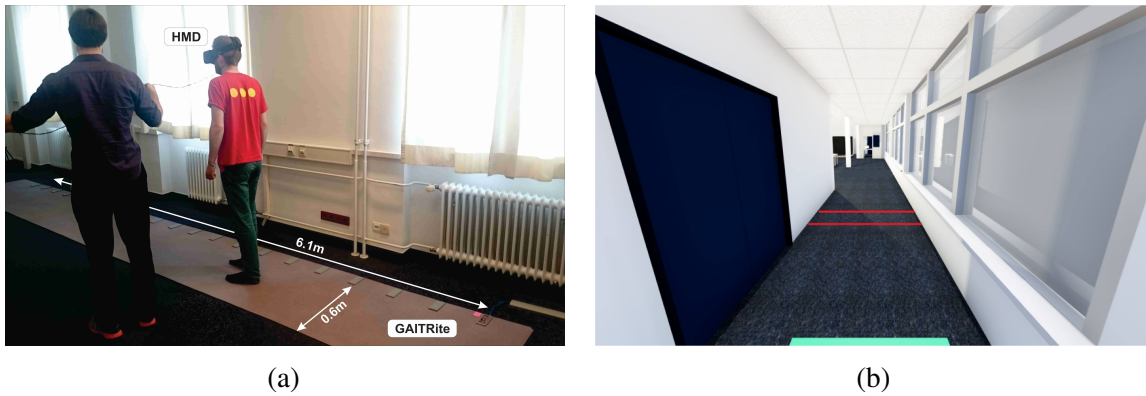
Fig. 4.1 Experiment setup: (a) A participant walks in the real workspace with an HMD over the GAITRite walking surface, and (b) participant's view on the HMD during isometric condition $g_t = 1$.

infrared marker to a baseball cap that was worn on the head of each participant and tracked it using PPT.

While walking, temporal (timing) and spatial (2D geometric indicators of the participant's feet) gait parameters were measured using a GAITRite electronic walkway system [166]. The electronic walkway system provides an active walking area of 0.6m × 6.1m with a scanning frequency of 60Hz. The pressure exerted by the feet onto the walkway activated the sensors during walking. The sensors provided measurements using (x,y) coordinates with distance recorded in centimeters and time in seconds up to an accuracy of 6dp. The walking distance was 6 meters in all conditions.

4.3.3   Methods

Participant filled out an informed consent form and received detailed instructions on how to perform the walking tasks. Furthermore, they filled out the simulator sickness questionnaire (SSQ) [118] immediately before and after the experiment, consisting of 16 symptoms that are rated by the participant in terms of severity. These symptoms include, but are not limited to headache, nausea, burping, sweating, fatigue, and vertigo. Participants rated each symptom on a Likert-type scale [133], including the options none, slight, moderate, and severe. Furthermore, the Slater-Usoh-Steed presence questionnaire (SUS PQ) [227] was filled out after the experiment as well as a demographics questionnaire.

We used a within-subjects design in which we tested eight walking conditions consisting of one real-world condition and seven translation gain conditions (cf. Section 4.1) while wearing the HMD. We tested translation gains in the following range: $g_t \in \{\frac{1}{4}, \frac{1}{2}, \frac{3}{4}, 1, \frac{5}{4}, \frac{3}{2}, \frac{7}{4}\}$, i.e., visual flow presented at lower $g_t < 1$, matched $g_t = 1$ or higher speed $g_t > 1$ (see

(a) $g_t = .25$ (b) $g_t = .5$ (c) $g_t = .75$

(d) $g_t = 1.25$ (e) $g_t = 1.5$ (f) $g_t = 1.75$
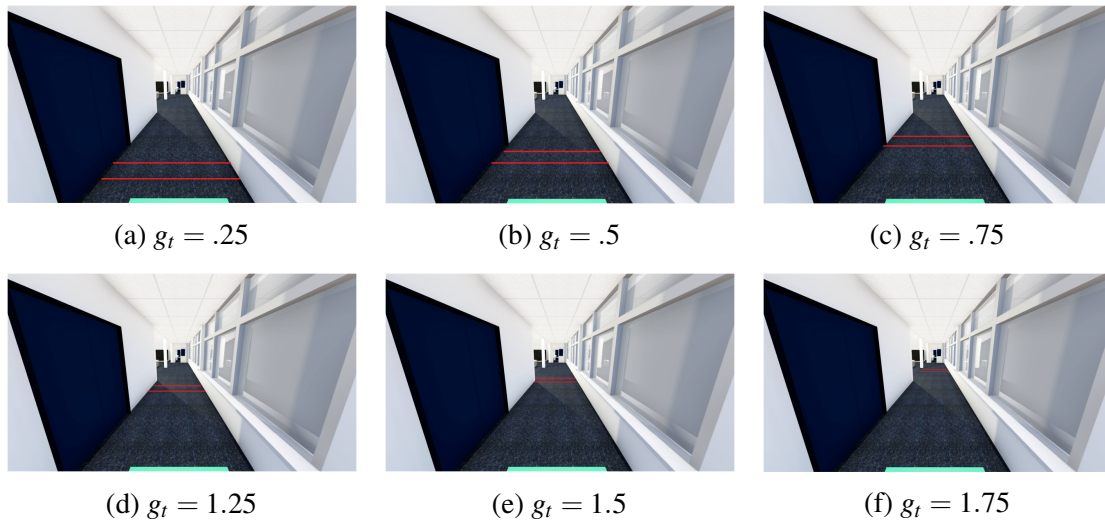
Fig. 4.2 Illustration of non-isometric conditions.

Figure 4.2). Each condition was repeated twice during the experiment. The order in which the conditions were tested was randomized. In total, each participant completed 16 trials.

The task was to first assume the start position by standing in orthostatic pose at the start line; the body is held in an upright position and supported only by the feet. Then, participants were instructed to walk in their normal pace along the walkway of the GAITRite system while coming to a halt between the location of the target lines (see Figure 4.1). This was done with or without the HMD. After each trial, the participant had to walk back to the starting point.

4.3.4 Data Collection

Several spatio-temporal gait parameters are analyzed through the GAITRite walkway system. As illustrated in Chapter 3, these parameters are: (velocity, cadence, step length, base of support, FAP score, toe in/out, single support and double support). Furthermore, we collected data from the head-tracked HMD. In particular, head pitch angles were analyzed using the orientation data of the Oculus Rift DK2 HMD. While the yaw and roll angles of the participant's head were expected to be near-constant due to the task of straightforward walking, we were interested in whether differences in gait would be accompanied by a tendency of the participants to pitch their head further down towards the floor. Such differences in pitch angles correspond to atypical head postures during walking, which indicate an additional difference from natural walking.

4.4   Results

Figures 4.3 to 4.7 show the differences between the real and virtual walking conditions for the different dependent variables in the experiment. The x-axes show the translation gains g_t and the y-axes show the measured values pooled over the participants. The vertical bars show the standard error of the mean. We summarize the results of our experiment in the following sections (see also Table 4.1):

4.4.1   Comparison Between Real and Virtual Walking

In order to compare the effects of the immersive and non-immersive walking conditions we first considered only the data in the conditions while wearing the HMD in which the virtual walking velocity matched the real-world walking velocity, i.e., with gains $g_t = 1$.

The results were normally distributed according to a Shapiro-Wilk test at the 5% level. We performed paired t-tests at the 5% significance level. We found a significant effect of immersion on walking velocity $t(18) = 3.26$, $p = .004$, step count $t(18) = -2.62$, $p = .018$, step length $t(18) = 3.83$, $p = .001$, base of support $t(18) = -2.14$, $p = .046$, FAP score $t(18) = 3.63$, $p = .002$, toe in/out $t(18) = -3.96$, $p = .001$ and double support $t(18) = -3.32$, $p = .004$. We found no significant effect of immersion on single support $t(18) = .51$, $p = .620$. The results show that most gait parameters while immersed with the HMD differed from walking in the real world.

4.4.2   Comparison Between Translation Gains

We analyzed the results for the different translation gains in the immersive conditions with a repeated-measures ANOVA and Tukey multiple comparisons at the 5% significance level with Bonferroni correction. The results were normally distributed according to a Shapiro-Wilk test at the 5% level. Degrees of freedom were corrected using [77] estimates of sphericity in case Mauchly's test indicated that the assumption of sphericity had been violated [146]. Partial eta squared (η_p^2) values provide an estimation of the effect size [117, 33]. The ranges of η_p^2 are between 0 and 1 (i.e., small $\approx .02$, medium $\approx .13$ and large $\approx .26$).

We found a significant main effect of translation gains on walking velocity, $F(2.18, 39.16) = 3.35$, $p = .042$, $\eta_p^2 = .157$, base of support, $F(4.52, 81.28) = 4.64$, $p = .001$, $\eta_p^2 = .205$, FAP score, $F(2.35, 42.34) = 4.76$, $p = .010$, $\eta_p^2 = .209$, toe in/out, $F(3.28, 59.05) = 3.17$, $p = .027$, $\eta_p^2 = .150$, double support, $F(3.76, 67.61) = 3.61$, $p = .011$, $\eta_p^2 = .167$, and head pitch angle, $F(2.68, 48.22) = 37.27$, $p < .001$, $\eta_p^2 = .674$.

Post-hoc tests showed significant differences for walking velocity only between transla-tion gains $g_t = 1$ and $g_t = 1.5$ ($p = .021$) as well as between $g_t = 1.25$ and $g_t = 1.5$ ($p = .020$), for base of support only between translation gains $g_t = .25$ and $g_t = 1.25$ ($p = .015$) as well as between $g_t = .5$ and $g_t = 1.25$ ($p = .041$), for FAP score only between translation gains $g_t = .25$ and $g_t = .50$ ($p = .015$), between $g_t = 1$ and $g_t = 1.5$ ($p < .001$), between $g_t = 1$ and $g_t = 1.75$ ($p = .005$) as well as between $g_t = 1.25$ and $g_t = 1.5$ ($p = .004$), for toe in/out only between translation gains $g_t = .5$ and $g_t = 1.75$ ($p = .003$) as well as between $g_t = 1.25$ and $g_t = 1.75$ ($p = .037$). Post-hoc tests for double support showed no significant differences but a trend between translation gains $g_t = .25$ and $g_t = .5$ ($p = .055$). Post-hoc tests for head pitch angle showed significant differences between all translation gains (all $p < .05$).

We found no significant main effect but a trend of translation gains on step count, $F(2.37, 42.57) = 2.86$, $p = .06$, $\eta_p^2 = .137$. We also found no significant main effect of translation gains on step length, $F(2.65, 47.63) = 2.20$, $p = .108$, $\eta_p^2 = .109$, or on single support, $F(3.95, 71.02) = .508$, $p = .727$, $\eta_p^2 = .027$.

4.4.3 Questionnaires

We measured a mean SSQ score of $M = 9.2$ ($SD = 16.6$) before the experiment, and a mean SSQ score of $M = 10.2$ ($SD = 12.2$) after the experiment. The scores indicate overall low simulator sickness symptoms for walking with an HMD, and we found no significant increase of symptoms over the time of the experiment $t(18) = -.33$, $p = .745$. The mean SUS PQ score for the sense of feeling present in the VE was $M = 4.2$ ($SD = .58$), which indicates a high sense of presence [227]. Additionally, participants judged their fear to collide with the walls of the room or other physical obstacles while immersed with the HMD during the experiment as comparably low (rating scale, 0=no fear, 4=high fear, $M = 1.1, SD = 1.3$).

4.5 Discussion

Our results show that participants walked differently within the real and virtual environment in terms of nearly all tested gait parameters, and most gait differences increased when large discrepancies of virtual velocity were introduced with non-isometric mappings.

Concerning walking velocity, our results (shown in Figure 4.3a) show a significant decrease of walking velocity by 6% while wearing the HMD compared to the real world, which is similar to results obtained in a study performed by [154], who reported a decrease in walking velocity by 14%. Moreover, we found that walking velocity was further decreased the more translation gains differed from an isometric mapping with $g_t = 1$, which indicates
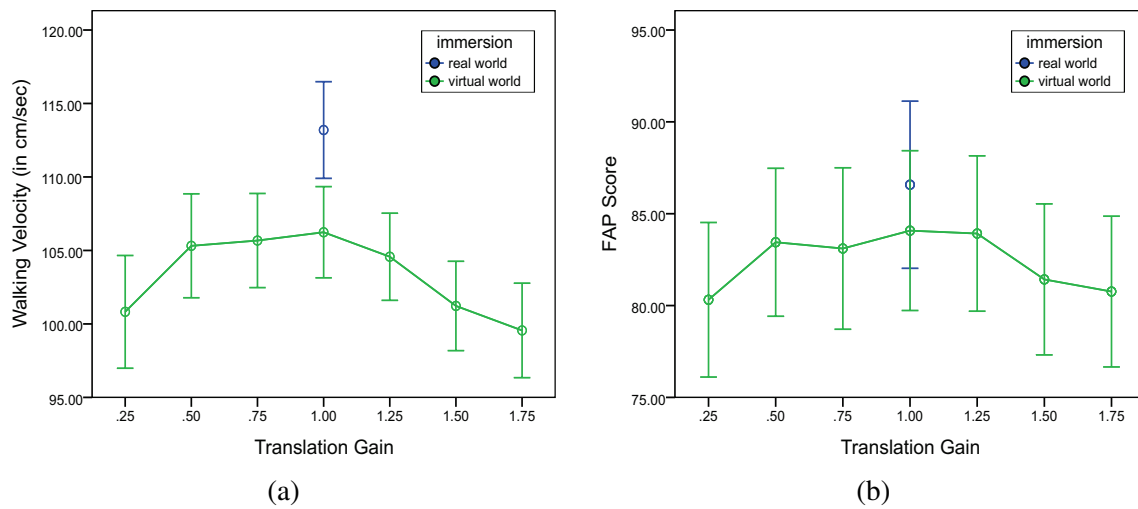
Fig. 4.3 Results for applied translation gains on the horizontal axis and pooled for (a) walking velocity and (b) FAP score on the vertical axis. The error bars show the standard error of the mean.

an almost symmetrical effect of translation gains. This result is highly interesting, since it is important to note that the effect of translation gains was actually different from what would be expected based on a previous study with slightly different experimental design performed by [156]. This study would have predicted that participants would increase their walking velocity when the virtual velocity was decreased with a gain $g_t < 1$, and not a decrease in walking velocity. Informal comments by our participants suggest that many of them felt that decreased virtual walking velocities lead to less stable walking along the path, thus inducing them to slow down even more. This result is also matched by the head pitch angles shown in Figure 4.7, which indicate that the participants looked down towards the floor when the virtual velocity was reduced, which may indicate that they received less visual flow information during walking, and thus had to orient more by the pathway shown on the floor.

Furthermore, as shown in Figure 4.3b we found that the FAP score significantly decreased in the VE compared to the real world with respect to an isometric mapping with $g_t = 1$, which indicates that the number of points subtracted increased. Moreover, the FAP scores were reduced when non-isometric mappings were applied with translation gains $g_t \neq 1$, which indicates that selected parameters increasingly differed from normal gait. The points deducted in the different parts of the FAP score (see Equation 3.4) is determined by the distance between the participant's gait parameters and ranges of predefined values considered as normal for gait at the self-selected velocity [73], e.g., up to eight points are deducted if the dynamic base of support is abnormally wide or narrow. Further points can be deducted from a maximum score of 100 (i.e., from 0 to 8 points for right-left asymmetry and from 0 to

(a)   (b)
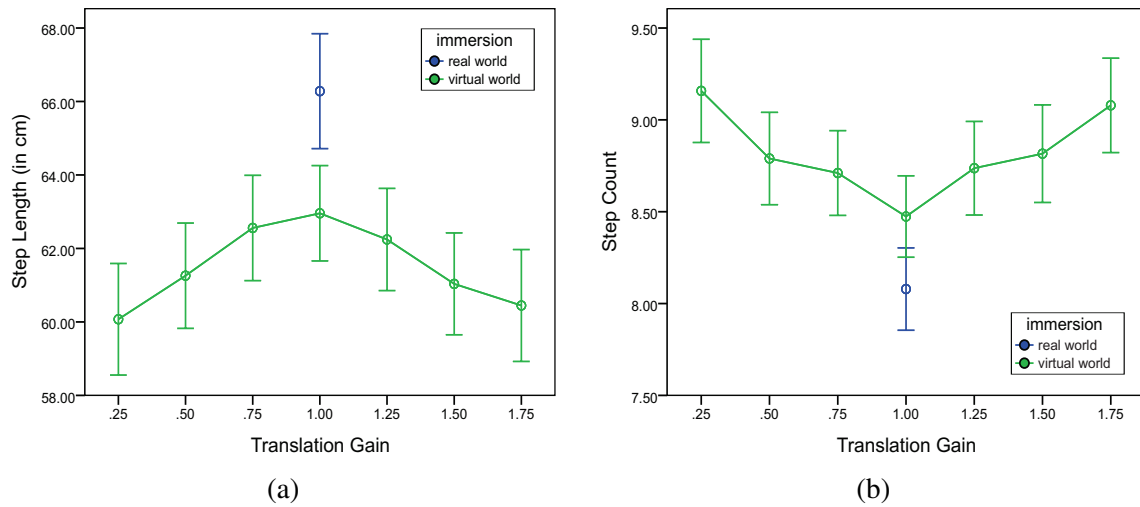
Fig. 4.4 Results for applied translation gains on the horizontal axis and pooled for (a) step length and (b) step count on the vertical axis. The error bars show the standard error of the mean.

(a)   (b)
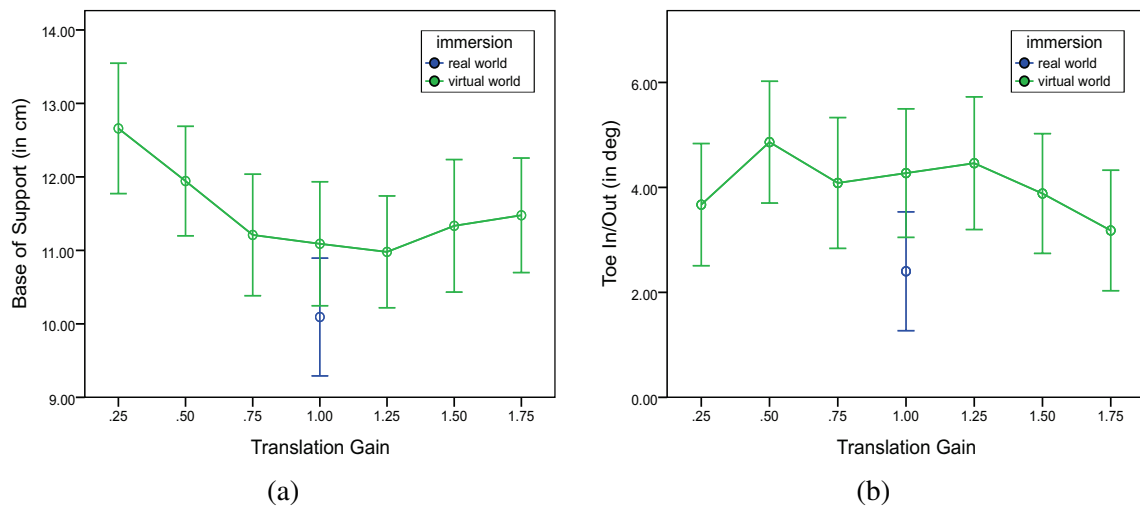
Fig. 4.5 Results for applied translation gains on the horizontal axis and pooled for (a) base of support and (b) toe in/out on the vertical axis. The error bars show the standard error of the mean.

22 points for right-left step functions). Regarding the deductible points intervening in FAP score calculation, we observed that a greater amount of points were deducted for dynamic base of support and asymmetry of step length. The deductions for functions of right and left steps were nonexistent or close of zero.

Figure 4.4a shows that participants had a shorter step length in the VE than in the real world for $g_t = 1$. Moreover, the shortened step length in the VE positively correlated with

(a)   (b)
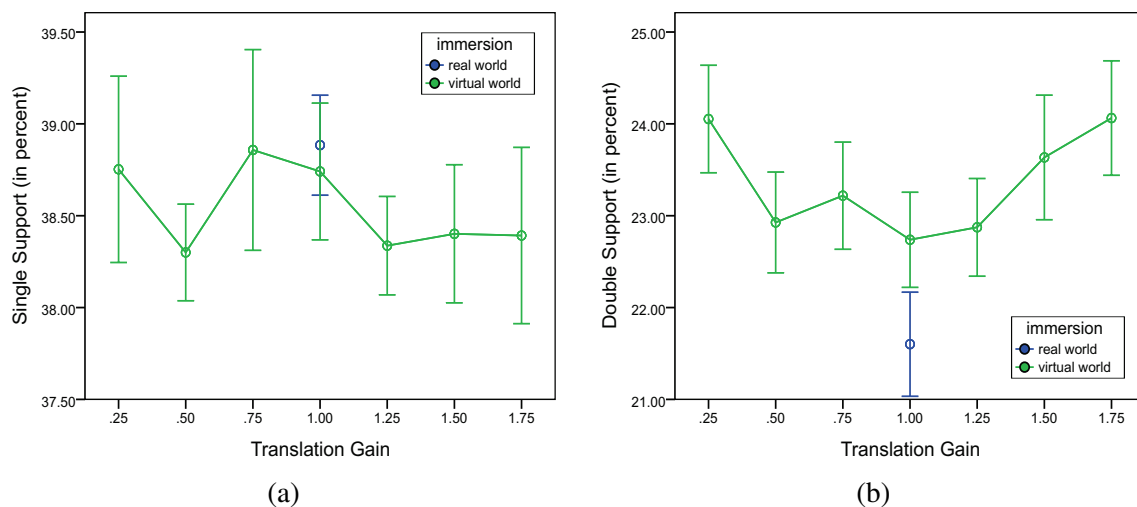
Fig. 4.6 Results for applied translation gains on the horizontal axis and pooled for (a) single support and (b) double support on the vertical axis. The error bars show the standard error of the mean.
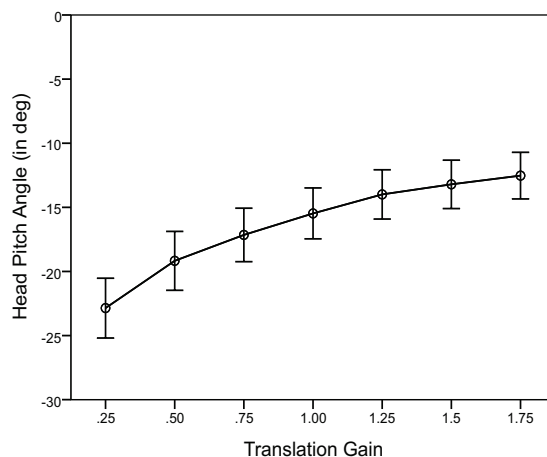
Fig. 4.7 Results for applied translation gains on the horizontal axis and pooled for head pitch angle on the vertical axis. The error bars show the standard error of the mean. A pitch angle of 0 indicates a level head, whereas negative angles indicate that the user was looking down towards the floor.

an increase of the number of steps taken (see Figure 4.4b). These results also reflect a similar response to visual stimulation during the application of translation gains regardless of whether translation gains $g_t < 1$ or $g_t > 1$ were applied. These results correlate with an instability of the participants' gait.

Figure 4.5a shows the differences of base of support in the VE compared to the real world. The differences indicate that participants walked with a widened base of support in the VE for $g_t = 1$. A wide base of support has long been believed to be a hallmark of

unsteady gait [174]. This suggests that the participants tended to spread their feet apart, thereby increasing their bases of support while walking with an HMD in the VE. The base of support was wider with slower translation gains $g_t = .25$, and was narrower with faster translation gains $g_t = 1.75$. Thus, the effects of visual information at lower translation gains induced a wider base of support and would be more likely to increase stability in the VE. In this scope, we also found that the toe in/out (foot angle) in Figure 4.5b increased, which indicates that participants walked with toes pointing further out within the VE. In contrast, we did not find any negative indicator of toe in/out within the VE, or in the real world.

Figure 4.6b shows a significantly prolonged double support period in the VE for $g_t = 1$ as well as a non-significant trend towards a shortened single support period (see Figure 4.6a). Also, the participants tended to prolong their double support period during the application of translation gains $g_t < 1$ and $g_t > 1$ in which the body weight is supported by both legs. These findings indicate that the durations of stance phases are longer than swing phases throughout the virtual walking, which also justifies why the participants walked slower in the VE than in the real world.

4.6   General Discussion

Overall, most measured biomechanical parameters of gait were affected both by immersion and the application of translation gains during bipedal locomotion in the VE. The results underline the importance to investigate the differences between gait parameters while walking in the real world and within a VE. Specifically, our results show that virtual walking had much lower stability than walking in the real world, which might be explained by hardware factors such as the weight of the HMD, which causes a participant to walk differently in the VE [271]. In particular, Willemsen et al. [271] have found evidence that issues related to the ergonomics of HMD systems may account for some of the apparent compression observed, an explanation for the larger portion of the effect remains wanting. Or, factors such as the availability of visual flow information when translation gains are applied while walking, e. g., visual flow, provides cues about the travelled distance. Within a VE, these cues are consistent and hence provide veridical information to the user about her motion. Although human subjects can use these cues to discriminate travelled distances [63], it has been shown that perception in the virtual world varies significantly from perception in the real world.

Furthermore, walking in the VE was found to correlate with decreased walking velocity, decreased FAP score, decreased step length, increased step count, widening of the base of support, positive toe-out, shortened single support and with prolonged double support. And, last but not least, we found that participants had a tendency to look down towards the floor

when translation gains $g_t < 1$ were applied, but no such effect was visible for gains $g_t > 1$. These difference might be caused by the limited FOV of current HMDs. Jones et al. [113] found evidence that a small FOV could produce improved distance judgments when real world visual flow was added to the lower part of the periphery. However, the reason for this improvement remains unclear. A follow-up study by Jones et al. [114] attempted to determine if the peripheral visual flow was causing observers to recalibrate their gait, enabling them to move more accurately in the VE. They found that gait did seem to be a partial contributing factor, but that it was insufficient to explain all the observed improvements. They speculated that the peripheral stimulation may have served as an additional reference to the location of the ground plane relative to the observers' eye position. Another potential explanation for the misinterpretation could be based on incorrect depth cues provided to the human eye when looking through an HMD, such as the accommodation convergence conflict [88]. [264] stated that inappropriate depth cues in typical HMDs may therefore contribute to distortions in perceived space.

4.7   Conclusion

In this chapter, an experiment presented in which we evaluated the differences of gait parameters between a real and virtual environment. In the virtual world conditions, we analyzed effects of non-isometric virtual walking with translation gains on gait parameters. The results of the experiment showed significant differences of walking biomechanics in the virtual and real world, and we also found a significant effect of translation gains on most gait parameters. The results provide novel insights into effects on gait that can be expected when practitioners in the field of VR follow the suggestions in previous literature to slightly increase virtual walking velocities in order for them to be estimated as more natural by users. Our findings seem to disagree with these suggestions and indicate that a closest match of gait in the real and virtual world can be found for an isometric mapping, whereas non-isometric mappings correlated with often symmetrically-shaped detriments both for up- and down-scaled virtual walking velocity.

In the next chapter, this experiment was repeated with healthy older adults, in order to compare differences in biomechanics while walking in a VE and the real world.

Table 4.1 The results of statistical analysis

	Paired sample t-tests		Repeated-measures ANOVAs with post-hoc tests			
	$t(18)$	p	g_t	F	p	η_p^2
Velocity (cm/sec)	3.26	.004		$(2.18, 39.16) = 3.35$	.042	.157
			$g_t\ (1, 1.5)$		.021	
			$g_t\ (1.25, 1.5)$		.020	
Step width (cm)	−2.14	.046		$(4.52, 81.28) = 4.64$	.001	.205
			$g_t\ (.25, 1.25)$		.015	
			$g_t\ (.5, 1.25)$		.041	
FAP score	3.63	.002		$(2.35, 42.34) = 4.76$	.010	.209
			$g_t\ (.25, .5)$		.015	
			$g_t\ (1, 1.5)$		<.001	
			$g_t\ (1, 1.75)$		.005	
			$g_t\ (1.25, 1.5)$		.004	
Toe in/out (deg)	−3.96	.001		$(3.28, 59.05) = 3.17$	.027	.150
			$g_t\ (.5, 1.75)$		.003	
			$g_t\ (1.25, 1.75)$		.037	
Double support (%)	−3.32	.004		$(3.76, 67.61) = 3.61$	.011	.167
			$g_t\ (.25, .5)$		.055	
Step count	−2.62	.018		$(2.37, 42.57) = 2.86$	.060	.137
Step length (cm)	3.83	.001		$(2.65, 47.63) = 2.20$	.108	.109
Single support (%)	.51	.620		$(3.95, 71.02) = .508$	.727	.027
Head pitch angle (degree)				$(2.68, 48.22) = 37.27$	<.001	.674
			$g_t\ (.25, .5, .75, 1, 1.25, 1.5, 1.75)$		<.050	

VIRTUAL WALKING IN OLDER ADULTS

To better understand how people at different ages walk and perceive locomotion in VR, an experiment performed to investigate the effects of (non-)isometric mappings between physical movements and virtual motions in the VE on the walking biomechanics across generations, i.e., younger and older adults. Three primary domains (pace, base of support and phase) of spatio-temporal parameters were identified to evaluate gait performance. The results show that the older adults walked very similar in the real and VE in the pace and phasic domains, which differs from results found in younger adults. In contrast, the results indicate differences in terms of base of support domain parameters for both groups while walking within a VE and the real world. For non-isometric mappings, we found in both younger and older adults an increased divergence of gait parameters in all domains correlating with the up- or down-scaled velocity of visual self-motion feedback. The results provide important insights into the design of future VR applications for older adults in domains ranging from medicine and psychology to rehabilitation.

5.1 Introduction

VR technologies are an effective way to simulate virtual worlds that are used in many application domains requiring a high degree of immersion and interactivity. In a VE, the user interacts with a multisensory computer-generated environment, which can be explored in real time. In this context, walking as a means to explore the VE is an essential part of a VR experience. Walking is often considered the most basic and natural form of locomotion humans can perform. Thus, realizing real walking in VEs is essential to support a veridical model of reality in a wide range of application domains such as training, rehabilitation, or entertainment. In previous studies, walking in VEs by means of real walking was analyzed in terms of navigation performance [209, 211], gait performance [89] or presence [255] with

a focus on younger adults. In Chapter 4, we presented an experiment that focuses on younger adults, which evaluated the differences of biomechanical walking parameters between a real and virtual environments.

Many studies demonstrated the potential of VR technology for older adults, and presented opportunities and benefits for several application domains ranging from medicine and psychology to rehabilitation [153, 205, 222, 247]. However, most of today's VR systems and applications are mainly used by younger people, whereas older adults are often not considered in applications or scientific experiments using VR technology. Recently, VR has received enormous attention by the general public, and the technology is getting widely used and accessible. Therefore, it appears reasonable to assume that more and more people at different ages will have access to VR, and use VR in the context of applications domains such as rehabilitation or physiotherapy. Hence, we believe that it is important to investigate walking in VEs involving older adults, with the goal to understand the perceptual and motor differences, but also to gain similar advantages from virtual walking as from walking in the real world for the older generation.

In the scope of this chapter, we analyzed differences in several gait parameters while walking within a VE and the real world, and furthermore investigate differences between these parameters in younger and older adults. We believe that it is important to understand how walking in VEs varies across generations. For instance, it has previously been found that walking in a VE for younger adults will result in decreases in walking velocity, increases in step width, and increases in double support compared to the real world. We assume that older adults will show less stability during walking within a VE compared to the real world, in particular, in comparison to younger ones, as a result of various factors such as increased fear of falling, slower sensory-motor coupling or less visual accuracy.

There are several technical and perceptual challenges for real walking in VEs [239]. In particular, illusions related to visual flow may change the user's perception of his self-motion in the VE independently of his actual self-motion in the VE [24]. This may be used to tune virtual locomotion cues in order to provide natural perception of self-motion in VEs. As described above, younger or older adults might estimate a slightly increased virtual walking velocity as more natural, but we have to consider that this does not necessarily lead to the situation in which the biomechanics of walking in the VE match those of the corresponding behavior in the real world. Unfortunately, the existing body of literature does not provide a consistent understanding of the effects of isometric or non-isometric walking conditions on gait detriments in VEs, or any information on the effects of age on these parameters [154, 159, 239].

In this chapter, we presented an experiment in which we investigate the effects of isometric and non-isometric walking with an HMD on gait parameters of younger and older adults. Our main contributions are:

- Analysis of gait parameters indicating a closest match between virtual and real walking for an isometric mapping, while non-isometric mappings resulted in often symmetrically-shaped detriments both for up- and down-scaled virtual velocity.

- Evaluation of the effects of manipulated visual self-motion on walking biomechanics between younger and older adults, using three domains of spatio-temporal gait parameters that may facilitate understanding of gait performance while walking within a VE and the real world.

5.2   Related Literature

Many experiments examined whether there are age-related changes in gait patterns. Two studies [10, 252] have investigated performance of gait stability with walking experience in the real world, and indicate that older adults walk as stable as or less stable than young adults. It is interesting to investigate whether similar differences can be found when younger or older people walk through VEs. In particular, in this context Chou et al. [32] reported that walking parameters may affect a user's perception of virtual space. Hollman et al. [90] have found that walking in a treadmill VE may induce changes in biomechanical parameters, which reflect gait instability. Discrepancies between perception in real and virtual environments have naturally been suggested as a potential factor contributing to the fact that distances in VEs are often over- or underestimated [101, 102, 135, 198]. However, interaction with the VE by walking with visual feedback has recently been shown to drastically improve perceived distance to within 90-100% of actual distance with an appropriate interaction [200, 155, 263, 201]. Furthermore, Ruddle and Lessels [211] found that real walking in a VE provides near-perfect performance on a navigational search task, whereas joystick directed travel resulted in less than 50% of trials performed perfectly.

While several previous studies examined gait in a VE [90, 154, 159, 239], most of those focused on young adults. According to Chou et al. [32], older adults show a comparable ability to integrate visual flow information into a VE for assessment of walking velocity and heading direction. Furthermore, Schubert et al. [220] could not find significant differences in locomotion between younger and older adults due to changes in visual information, especially when visual flow speed decreases, walking velocity and stride length increase; decreasing visual flow speed shows opposite effects. A study by Berard et al. examined

whether advanced age could impact on the directing of locomotion in response to changes in visual flow speed in the VE, and found that older adults were impaired to use visual flow cues to direct their locomotion [8]. Whether older adults are more dependent on visual flow information during locomotion compared to younger adults is still open to further investigation. However, the behavior of most people is different when walking in a VE than in the real world, whereas the question remains as to whether people walking within VEs show lower stability than during walking in the real world, and in how far differences between younger and older adults can be found presuming that these differences are exist.

5.3   Experiment

In this section, we describe the experiment in which we have examined how walking in VEs differs from walking in the real world in terms of biomechanics for younger and older adults. Since it has shown in Chapter 4 that altering the visual speed changes gait parameters in younger adults, we tested different isometric and non-isometric walking conditions using the method of translation gains (cf. Section 4.1) along a straightforward movement path for both younger and older adults. We compared the results to a baseline condition, which was walking in the real world, and with those results from younger and older adults performing the same task. Prior to the experiment, we received approval for the experimental procedure, material, and methods from our institutional review board.

5.3.1   Participants

A total of 42 healthy participants completed the experiment consist of two groups: 21 younger adults (4 female and 17 male, ages $18 - 34$ years, $M = 23.67$ $_{SD=4.04}$, heights $164 - 196$ cm, $M = 180.04$ cm $_{SD=8.84\,cm}$) and 21 older adults (12 female and 9 male, ages $45 - 83$ years, $M = 55$ $_{SD=10.08}$, heights $158 - 192$ cm, $M = 174.2$ cm $_{SD=9.98\,cm}$). The younger participants were students or members of the department of computer science or the department of neurophysiology. Students obtained class credit for their participation. The older participants were relatives of patients of the department of neurology or members of the department of neurophysiology and pathophysiology. All of our participants had normal or corrected-to-normal vision. During the experiment, 3 of the younger and 13 of the older participants wore glasses. None of our participants reported a disorder of equilibrium, vision disorders, or other abnormalities (e.g., arthritis, Parkinson's). Participants wore an HMD for approximately 30 minutes during the experiment, Six younger participants had prior experience with HMDs, whereas none of the older participants had prior experience with HMDs. We measured

the leg length of our participants before the experiment: younger adults ($92 - 111$ cm, $M = 98.57$ cm $_{SD = 5.52\,cm}$), and older adults ($90 - 115$ cm, $M = 100.83$ cm $_{SD = 6.54\,cm}$). We used the leg length of each participant to calculate a functional ambulation performance (FAP) [165], which represents a quantification of participants' gait based on a selection of spatio-temporal parameters obtained at a self-selected velocity [74, 78]. The selected parameters are standard velocity normalized to leg length, step and leg length ratio, step time, right/left asymmetry of step length, and dynamic step width. Participants were naive to the experimental conditions, wearing their normal clothes and performing barefoot walking across a walkway. They were allowed to take breaks at any time between experimental trials in order to minimize effects of exhaustion or lack of concentration. The total time per participant, including pre-questionnaires, instructions, experiment, breaks, post-questionnaires, and debriefing, was about one hour.

5.3.2 Materials

We performed the experiment in a laboratory room of 9m × 4m in size; see Figure 5.1a. During the experiment, the room was darkened in order to reduce the participant's perception of the real world while immersed in the VE. The VE was rendered using Unity3D; a cross-platform game engine with a custom-enabled VR communications and rendering library. As illustrated in Figure 5.1b, the VE showed a virtual pathway of 15m × 2.5m. The start (green line) and target (red lines) were placed on the floor in front of the participant to indicate the walking distance in the virtual world. The participants had been instructed to walk from the start line to the target (i.e., stopping between the two target lines). For rendering, system control and logging, we used an Intel computer with 3.4GHz Core i7 processor, 16GB of main memory and Nvidia GeForce 780Ti SLI graphics cards.

The participants wore an HTC Vive HMD for the stimulus presentation, which provides a resolution of 1080×1200 pixels per eye with a refresh rate of 90Hz and an approximately $110°$ diagonal FOV. The HMD uses more than 70 sensors including a MEMS gyroscope, accelerometer and laser position sensors. We tracked sensors on the HMD using a Lighthouse tracking system (2 base stations emitting pulsed IR lasers) that tracked the user's head movement with sub-millimeter precision in the laboratory.

While walking, temporal (timing) and spatial (2D geometric indicators of the participant's feet) gait parameters were measured using a GAITRite electronic walkway system [166]. The GAITRite consists of a walkway with overall dimensions of 90cm × 7m × 3.2mm on which the participant walks. A computer system controls the GAITRite and analyzes the data. The GAITRite walkway system provides an active walking area of 60cm × 6.1m with a scanning frequency of 60Hz. In addition to the active 6.1m long walkway, there are initial

(a)   (b)

Fig. 5.1 Experiment setup: (a) A participant walks in the real workspace with an HMD over the GAITRite walking surface, and (b) participant's view on the HMD during isometric condition $g_t = 1$.

20cm and final 70cm inactive sections to allow for walk on/off areas of the participant (i.e., where the start and target lines were placed). The pressure exerted by the feet onto the walkway activated the sensors during walking. The sensors provided measurements using (x,y) coordinates with distance recorded in centimeters and time in seconds up to an accuracy of 6dpi. The walking distance was 6 meters in all conditions.

5.3.3 Design

A mixed factorial design was used, with two levels of age group (younger, older) as the between-subjects factor and seven levels of translation gains (cf. Section 4.1) as the within-subjects factor. The tested translation gains were in the following range: $g_t \in \{\frac{1}{4}, \frac{1}{2}, \frac{3}{4}, 1, \frac{5}{4}, \frac{3}{2}, \frac{7}{4}\}$, i.e., visual flow presented at lower $g_t < 1$, matched $g_t = 1$ or higher speed $g_t > 1$ (see Figure 4.2). Hence, the experiment consisted of eight walking conditions, i.e., one real-world condition and seven translation gain conditions while participants wore the HMD. Each condition was repeated twice and the order of the tested translation gain conditions was randomized. Hence, each participant completed 16 walking trials. The experiment was conducted in two blocks, the first block with 21 older participants, and the second block with 21 younger participants.

5.3.4 Procedure

Prior to the walking tasks, participants filled out an informed consent form and received detailed instructions on how to perform the task. In addition, they filled out the simulator sickness questionnaire (SSQ) [118] immediately before and after the experiment, consisting

of 16 symptoms that are rated by the participant in terms of severity. These symptoms include, but are not limited to headache, nausea, sweating, fatigue, vertigo, and burping. Participants rated these symptoms on a Likert-type scale [133] from none, slight, moderate, to severe. After the experiment, they filled out a demographics questionnaire as well as the Slater-Usoh-Steed presence questionnaire (SUS PQ) [227].

The task was to first assume the start position by standing in an orthostatic pose at the start line. Then, participants were instructed to walk at a normal pace along the walkway of the GAITRite system while coming to a halt between the location of the target lines (see Figure 5.1). This was done with and without the HMD. After each trial, the participant had to walk back to the starting point with their eyes open in the real world. During the experiment, an experimental assistant managed the cables of the HMD for each participant and ensure that participants could walk safely.

5.3.5 Gait Data

Several spatio-temporal gait parameters (i.e., time and distance variables of the gait cycle) are analyzed through the GAITRite walkway system. The mean of consecutive gait cycles measured during steady-state walking was $M = 3.77_{SD=0.5}$ for younger adults, and $M = 4.42_{SD=0.6}$ for older adults. As illustrated in Chapter 3, these parameters are: (velocity, cadence, step length, step width, FAP score, toe in/out, double support, single support, stance phase and swing phase).

5.4 Results

Figures 5.2 to 5.4 show the differences between the real and virtual walking conditions for the different dependent variables in the experiment. The x-axes show the translation gains g_t and the y-axes show the measured values pooled over the participants. The vertical bars show the standard error of the mean. We summarize the results of our experiment in the following sections (see also Table 5.1) :

5.4.1 Real and Virtual Walking Comparisons

In order to compare the effects of walking in a VE with walking in the real-world conditions we first considered only the data in the conditions while wearing the HMD in which the virtual walking velocity matched the real-world walking velocity, i.e., with gains $g_t = 1$.

The results were normally distributed according to a Shapiro-Wilk test at the 5% level. We performed paired t-tests at the 5% significance level. We found a significant effect

between real world and VE in younger adults group on walking velocity $t(20) = 3.35$, $p = .003$, FAP score $t(20) = 3.039$, $p = .006$, step length $t(20) = 4.840$, $p = .001$, step width $t(20) = -3.056$, $p = .006$, toe in/out $t(20) = -4.064$, $p = .001$, double support $t(20) = -3.278$, $p = .004$, single support $t(20) = 4.020$, $p = .001$ and stance phase $t(20) = -4.135$, $p = .001$. The only dependent variable showing no significant effect of immersion was cadence $t(20) = .563$, $p = .580$.

On the contrary, in older adults we did not find any significant effect between real world and VE on walking velocity $t(20) = -.780$, $p = .445$, FAP score $t(20) = 1.088$, $p = .290$, step length $t(20) = .309$, $p = .760$, double support $t(20) = -.906$, $p = .376$, single support $t(20) = .258$, $p = .799$ and stance phase $t(20) = -.321$, $p = .752$. The only dependent variables showing a significant effect between real world and VE were cadence $t(20) = -2.166$, $p = .043$, step width $t(20) = -3.861$, $p < .001$ and toe in/out $t(20) = -2.534$, $p = .020$.

The results show that most gait parameters of the conditions with HMD significantly differ from walking in the real world for the younger adult participants. In contrast for older adults the results show that most gait parameters of the VE conditions do not significantly differ from walking in the real world.

5.4.2 Translation Gains Comparisons

We analyzed the results for the different translation gains in the immersive conditions with a mixed ANOVA with age group (younger, older) as the between subject factor and translation gains (i.e., seven levels) as the within subject factor. The results were normally distributed according to a Shapiro-Wilk test at the 5% level. Degrees of freedom were corrected using Greenhouse-Geisser estimates of sphericity in case Mauchly's test indicated that the assumption of sphericity had been violated. For all analyses, post hoc Bonferroni corrections for multiple comparisons at the 5% significance level were used to explore significant effects across all analyses

We found a significant main effect of translation gains on walking velocity ($F_{3.63,145.21} = 25.82$, $p < .001$ $\eta_p^2 = .392$), cadence ($F_{3.23,129.14} = 22.21$, $p < .001$, $\eta_p^2 = .356$), FAP score ($F_{4.25,169.79} = 6.06$, $p < .001$, $\eta_p^2 = .132$), step length ($F_{3.51,140.36} = 13.63$, $p < .001$, $\eta_p^2 = .254$), step width ($F_{4.19,167.49} = 5.20$, $p < .001$, $\eta_p^2 = .115$), toe in/out ($F_{5.21,208.55} = 7.05$, $p < .001$, $\eta_p^2 = .150$). No significant main effect found of translation gains on double support ($F_{3.82,152.76} = 1.36$, $p = .251$, $\eta_p^2 = .033$), single support ($F_{2.92,116.86} = 1.60$, $p = .193$, $\eta_p^2 = .039$), stance phase ($F_{3.16,126.26} = 1.66$, $p = .177$, $\eta_p^2 = .040$).

The main age group effects were significant only for FAP score ($F_{1,40} = 7.36$, $p = .010$, $\eta_p^2 = .155$), double support ($F_{1,40} = 5.84$, $p = .020$, $\eta_p^2 = .127$), single support ($F_{1,40} = 4.92$, $p = .032$, $\eta_p^2 = .110$), stance phase ($F_{1,40} = 5.79$, $p = .021$, $\eta_p^2 = .126$). No significant

main effect found of group on walking velocity ($F_{1,40} = .02$, $p = .874$, $\eta_p^2 = .001$), cadence ($F_{1,40} = .50$, $p = .482$, $\eta_p^2 = .012$), step length ($F_{1,40} = .53$, $p = .471$, $\eta_p^2 = .013$), step width ($F_{1,40} = .48$, $p = .492$, $\eta_p^2 = .012$), toe in/out ($F_{1,40} = .009$, $p = .923$, $\eta_p^2 = .001$).

Post-hoc tests (Table 5.1) showed significant differences for most gait parameters between translation gains within younger and older groups. The results revealed no significant interaction effects between age group and translation gains for all gait parameters.

5.4.3   Questionnaires

We measured a mean SSQ score of younger adults (*M*=11.75, *SD*=14.45) and older adults (*M*=9.61, *SD*=12.2) before the experiment, and a mean SSQ score of younger adults (*M*=13.17, *SD*=21.4) and older adults (*M*=12.64, *SD*=18.26) after the experiment. We analyzed the SSQ questionnaire scores with a non-parametric Wilcoxon Signed Rank Test at the 5% significance level. The SSQ scores indicate overall low simulator sickness symptoms for walking with an HMD, and we found no significant increase of symptoms over the time of the experiment; younger adults (*Z*=-1.2, *p*=.23) and older adults (*Z*=-.629, *p*=.529). The mean SUS PQ score for the sense of feeling present in the VE was for younger adults (*M*=27.47, *SD*=5.97) and older adults (*M*=24.61, *SD*=6.32) which indicates a high sense of presence [227]. We analyzed the SUS PQ and SSQ questionnaires with a non-parametric Mann-Whitney U Test but found no significant difference between younger and older adults in SUS PQ scores (*U*=164.5, *p*=.157). Additionally, participants judged their fear to collide with the walls of the room or other physical obstacles while immersed with the HMD during the experiment as comparably low (rating scale, 0=no fear, 5=high fear, younger adults (*M*=0.3, *SD*=1.53) and older adults (*M*=0.8, *SD*=1.43).

5.5   Discussion

The aim of this study was to investigate whether older adults show differences in gait performance while walking within a VE and the real world compared with younger adults. Based on gait analysis, three domains of gait performance were identified. A pace domain was characterized by walking velocity, step length, and cadence. A base of support domain was characterized by step width, toe in/out (i.e., foot angle), and FAP score. A phase domain was characterized by temporal divisions of the gait cycle.

5.5.1   Gait Pace

Parameters in the pace domain are characterized by walking velocity, step length and cadence. Walking velocity and step length for example, are the most often reported reference values for gait performance [92].

Figure 5.2a shows that there was no significant difference for older adults between walking velocity while wearing the HMD compared to the real world for translation gain $g_t = 1$. These results are different from results found in younger adults, which show a significant decrease of walking velocity by 6.5% while wearing the HMD compared to the real world, which is similar to results obtained in a previous study performed by Janeh et al. [111], who reported a decrease in walking velocity by 6%. The main effects of translation gain were significant for walking velocity, indicating that both groups increased their walking velocity when the virtual velocity was decreased with a translation gain $g_t < 1$ and a decreased in walking velocity with a gain $g_t > 1$, which indicates an almost asymmetrical effect of applied translation gains between $g_t = .5$ and $g_t = 1.75$. Informal comments by our participants suggest that many of them felt that decreased virtual walking velocity with a gain $g_t = .25$ lead to less stable walking along the path, thus inducing them to slow down even more. It is important to note that the effect of translation gains was actually similar to a previous study by [156] performed on young adults with a slightly different experimental design.

Furthermore, Figure 5.2b shows that older adults had comparable step length in the VE and in the real world for $g_t = 1$, whereas younger adults had a significantly shortened step length in the VE than in the real world by 5.7%. The results also reflect a similar response for both groups to visual stimulation during the application of translation gains regardless of whether translation gains $g_t < 1$ or $g_t > 1$ was applied. The performance of older and younger adults was significantly different in the real condition for walking velocity $t(20) = 3.436$, $p = .003$, and for step length $t(20) = 4.436$, $p < .001$. Older adults walked at a slower velocity and with shorter steps by $\approx 7\%$ than younger adults. However, as walking velocity and step length can be modified by cognitive influences [267] or muscle activity [115], it is also likely that the walking performance observed in older people are partly due to a reluctance rather than an inability to walk more quickly.

Although both groups exhibited comparable cadence within the real world condition, older adults had a significantly higher step rate in the VE compared to the younger adults. Figure 5.2c shows that older adults had a significant difference in cadence by 2.3% while walking within a VE compared to the real world, which implies that the older group took an extra steps in the VE compared to the real condition. This increment in cadence may be attributed to a compensation for the shorter step lengths taken in a relatively shorter period of time. No significant difference for cadence was found in younger adults. In addition, the

(a)

(b)

(c)
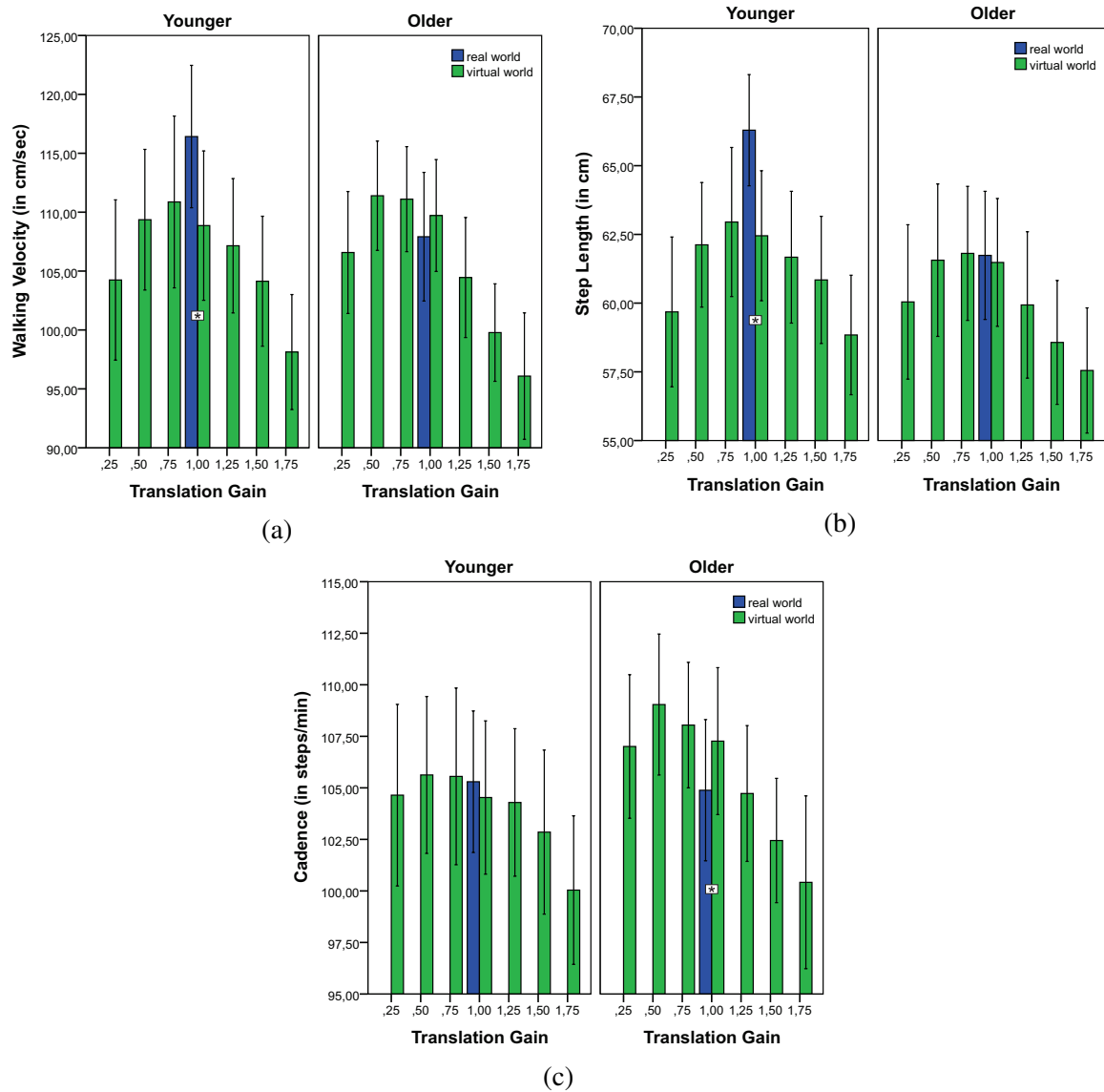
Fig. 5.2 The pace domain results for applied translation gains on the horizontal axis and pooled for (a) walking velocity, (b) step length and (c) cadence on the vertical axis. The error bars show the standard error.

linear main effects of translation gain for cadence were significant, showing that both groups walked at a higher step rate when the translation gain $g_t < 1$ was applied.

5.5.2 Base of Support

Figure 5.3a shows a significant increase of step width by 14.4% in older adults and 12.8% in younger adults while walking in the VE compared to the real world, which is similar to results obtained in a study performed by [90], who reported an increase in step width by 23%

in younger adults. This suggests that both groups tended to spread their feet apart, thereby increasing their step width while walking with an HMD in the VE and a translation gain of $g_t = 1$. Moreover, step width was significantly wider with slower translation gains $g_t = .25$, and was narrower with translation gains $g_t > 1$. Thus, the effects of visual information at lower translation gains induced a wider base of support and would be more likely to increase stability in the VE. In this scope, we also found that the toe in/out in Figure 5.3b increased positively, which indicates that participants walked with toes pointing further out within the VE. (i.e., the feet are outside the line of progression as shown in Figure 3.3) In contrast, we did not find any negative angles of toe in/out within the VE, or in the real world.

Furthermore, as shown in Figure 5.3c we found that the FAP score significantly decreased in the VE compared to the real world in younger adults with respect to an isometric mapping with $g_t = 1$, which indicates that the number of points subtracted increased. No significant difference was found in older adults, which justifies their comparable velocity and step length in the real world and the VE. Regarding the deductible points intervening in the FAP score calculation, we observed that a greater amount of points for older adults were deducted for dynamic step width and functions of right and left steps. The deductions for asymmetry of step length were nonexistent or close to zero. For younger adults the points were deducted mostly for dynamic step width.

Moreover, the FAP scores were reduced for both groups when non-isometric mappings were applied with translation gains $g_t \neq 1$, which indicates that selected parameters increasingly differed from normal gait. Additionally, the main age group effects were significant for FAP score, indicating that older adults walked at lower FAP score 80.5 than younger adults. This also suggests that older adults showed worse performance with different modulations in walking with response to the availability of translation gains compared with younger adults. This finding can be attributed to the fact that impairments occurred in the individual components of the FAP score, i.e., dynamic step width and functions of right and left steps led to lower the scores for the older group.

5.5.3 Gait Phases

A phase domain represents temporal divisions of the gait cycle. Figure 5.4 shows a non-significant difference between gait phases when older adults walk in the real world compared to walking through a VE. In older adults, double support occupied $\approx 24.9\%$ of the gait cycle, single support/swing phase $\approx 37.2\%$ and stance phase $\approx 62.8\%$. These findings indicate that the durations of gait cycle phases differ slightly from norms established by [45], i.e., double support 24%, single support/swing phase 38% and stance phase 62%, but likely represents the fact that older participants in our study walked with convergent gait pace in the real

(a)

(b)

(c)
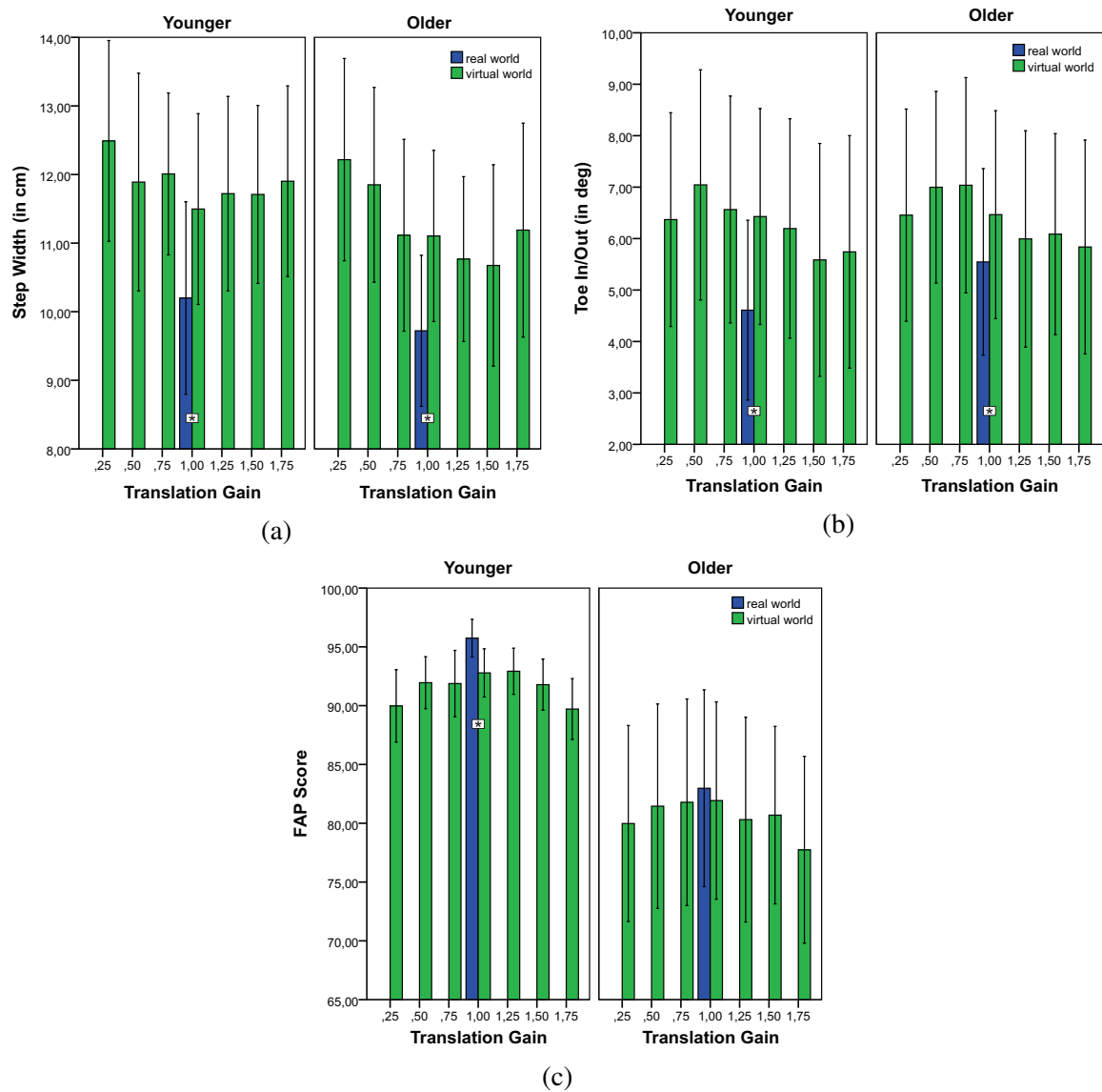
Fig. 5.3 The base of support domain results for applied translation gains on the horizontal axis and pooled for (a) step width, (b) toe in/out and (c) FAP score on the vertical axis. The error bars show the standard error.

world and the virtual environment. In contrast, we can see in Figure 5.4 that younger adults show significant differences between all gait phases while walking in the VE and real world. Double support, in which the body weight is supported by both legs, was prolonged by 5% during VE walking. Single support decreased by 2% and stance phase increased by 1.3%, which justifies why younger adults walked with slower gait speed and shorter steps in the VE compared to the real world.

Moreover, no significant main effects of translation gains were found on all phasic parameters, showing that both group tended to walk during the application of translation gains $g_t < 1$ and $g_t > 1$ with non-significant discrepancies in gait pattern throughout the virtual walking. Post hoc analyses revealed significant differences in single support and stance phase only between translation gains $g_t = .25$ and $g_t = .5$. The effects of age group were statistically significant for all gait phases, indicating that the effects of translation gains were influenced by age. Older adults walked with longer double support 25.6%, less single support 37.4% and a longer stance phase 62.6% than the younger adults.

5.6   General Discussion

Generally, it is interesting to note that our findings underline the importance of investigating the differences between gait parameters across generations while walking in the real world and within a VE. Specifically, our results show that the older adults exhibited comparable gait stability in most parameters within the pace and phasic domains during walking with and without the HMD.

The finding of unchanged walking parameters of older adults in the real and VE was an interesting observation. Older adults are known to have a more unstable gait pattern with about 50% of them will suffer recurrent falls while walking [144]. Increased gait variability during walking characterizes gait instability in older adults, which is further worsened with increasing cognitive demands such as dual tasking [91]. One would suggest further deterioration of walking capability by use of VR instead of constant gait performance in real and virtual conditions. In older adults, virtual environments have been found to impose a cognitive load that demands attention, response selection, and the processing of rich visual stimuli involving several perceptual processes [151].

Hypotheses about the unchanged gait performances in older adults in real and the VE remain speculative. It could be suggested that older adults already walk more slowly with decreased step length in the real world, so that the relative change between conditions is not obvious, i.e., *floor effect*. Another explanation could be that the hierarchical process of sensory organization is already altered in older compared to younger adults with another emphasis of the different orientational senses on the motor organization (e.g., less visual or proprioceptive weighting in the older adults). Besides, the older group could use the VR as an *external locus of control* [207], fixing their gaze to the screen and thus walking more stable and faster. Another explanation could be that older adults are more attentive and motivated when using VR technology [11], which invigorates their walking and prevents gait deterioration.
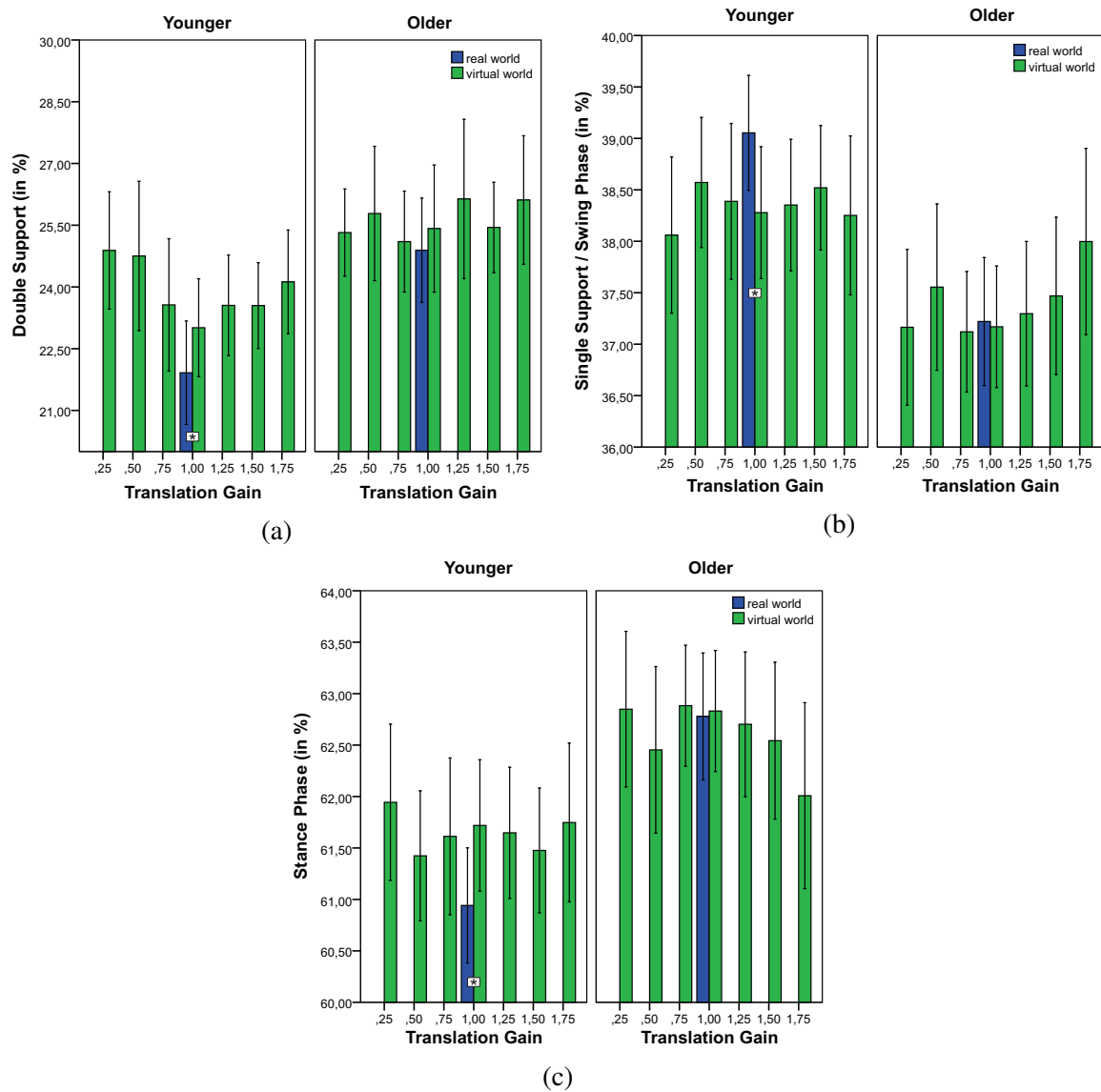
Fig. 5.4 The phase domain results for applied translation gains on the horizontal axis and pooled for (a) double support, (b) single support/swing phase and (c) stance phase on the vertical axis. The error bars show the standard error.

Our findings further suggest that various aspects of gait were found in the base of support domain to maintain stability, which has been a hallmark of unsteady gait [174]. Also, walking in the VE was found to correlate with widening step width and positive toe-out. While the walking behavior of older adults was similar to young adults in terms of step width and toe in/out, there seems to be little agreement on the idea that an increased step width and foot angle represent a compensatory strategy [39] also correlated to fear of falling [29].

Regarding age group differences in the phase domain as shown in Figure 5.4, the older group had a 4.6% decrease in the single support phase of the gait cycle compared to the younger group, which directly reflects a decreased step time. Also, a 13.2% increase in the double support phase indicates that older adults spent longer periods with both feet in contact with the ground while walking with and without the HMD.

Confirming previous results [111], younger adults walked significantly different within the real and virtual environments in terms of almost all gait parameters within the tested domains, and most gait differences increased when large discrepancies of virtual velocity were introduced with non-isometric mappings. This might be explained by hardware factors such as the weight, limited FOV and latency of the HMD, which cause a participant to walk differently in the VE. In particular, small FOV and latency could have a greater impacts on younger adults who may be dependent more on the availability of visual information to guide their walking behavior compared to older adults. Although the HTC Vive's end-to-end system latency is low, about 22ms measured between physical movement of the headset and the corresponding update of the Vive's display [169], but still can provides delay from the user's physical movement until the response becomes available on the headset's screen. Another potential explanation for the misinterpretation could be based on incorrect depth and motion cues provided to the human eye, when looking through an HMD, which introduces accommodation-convergence conflicts [88], in particular, in combination with age-related accommodation loss. Watt et al. [264] stated that inappropriate depth cues in typical HMDs may therefore contribute to distortions in perceived space. Or, non-visual factors (e.g., fear of falling or mobility-related anxiety) might be have a greater impacts on older adults compared to younger ones [153].

5.7   Conclusion

In summary, we evaluated the differences of biomechanical walking parameters of younger and older adults between walking in the real world and in the VE. In the VE conditions, we analyzed effects of non-isometric virtual walking with translation gains on gait parameters. Interestingly unlike younger group, the results of the older group showed similar results for the most walking biomechanics in the virtual and real world, and we could not find any significant effects of translation gains on most gait parameters. The results of older adults indicate similar gait patterns in the most parameters within pace and phasic domains in the real and virtual world with an isometric mapping. In contrast, the base of support domain indicated a significant increment in step width and toe in/out angle while walking within the VE. During the application of non-isometric mappings, we found that most gait

parameters within the domains correlated with often symmetrically-shaped detriments both for up- and down-scaled virtual walking velocity. Hence, we neither advocate increasing nor decreasing the virtual walking velocity of the user, but rather suggest maintaining an isometric one-to-one relation whenever possible to minimize gait detriments and the risk of falling.

Table 5.1 The results of statistical analysis

	Younger adults			Older adults			Mixed ANOVA with post-hoc			
	M_{SD}	$t(20)$	p	M_{SD}	$t(20)$	p	Source	F	p	η_p^2
Velocity (cm/sec)	$116.42_{(13.28)}$ r	3.35	$.003$	$107.92_{(12.01)}$ r	$-.78$	$.445$	gain	$(3.63, 145.21) = 25.82$	$<.001$	$.392$
	$108.86_{(13.94)}$ v			$109.72_{(10.44)}$ v			group	$(1,40) = .02$	$.874$	$.001$
							post-hoc	$g_t(.25)_{.5,.75,1.75}$ $g_t(.5,.75,1)_{1.25,1.5,1.75}$ $g_t(1.25)_{1.5,1.75}$ $g_t(1.5)_{1.75}$	$<.02$	
Cadence (steps/min)	$105.29_{(7.54)}$ r	$.563$	$.58$	$104.88_{(7.52)}$ r	-2.166	$.043$	gain	$(3.23, 129.14) = 22.16$	$<.001$	$.356$
	$104.53_{(8.17)}$ v			$107.26_{(7.83)}$ v			group	$(1,40) = .50$	$.482$	$.012$
							post-hoc	$g_t(.25)_{1.5,1.75}$ $g_t(.5,.75)_{1.25,1.5,1.75}$ $g_t(1,1.25)_{1.5,1.75}$ $g_t(1.5)_{1.75}$	$<.01$	
FAP score	$95.73_{(3.51)}$ r	3.039	$.006$	$82.97_{(18.38)}$ r	1.088	$.290$	gain	$(4.25, 169.79) = 6.06$	$<.001$	$.132$
	$92.78_{(4.49)}$ v			$81.92_{(18.44)}$ v			group	$(1,40) = 7.36$	$.010$	$.155$
							post-hoc	$g_t(.5,.75,1,1.25,1.5)_{1.75}$	$<.01$	
Step length (cm)	$66.29_{(4.44)}$ r	4.84	$<.001$	$61.73_{(5.12)}$ r	$.309$	$.76$	gain	$(3.51, 140.36) = 13.63$	$<.001$	$.254$
	$62.45_{(5.19)}$ v			$61.47_{(5.11)}$ v			group	$(1,40) = .53$	$.471$	$.013$
							post-hoc	$g_t(.25)_{.5,.75}$ $g_t(.5,.75,1,1.25)_{1.5,1.75}$ $g_t(1.5)_{1.75}$	$<.02$	
Step width (cm)	$10.19_{(3.08)}$ r	-3.056	$.006$	$9.72_{(2.41)}$ r	-3.861	$.001$	gain	$(4.19, 167.49) = 5.20$	$<.001$	$.115$
	$11.49_{(3.05)}$ v			$11.10_{(2.74)}$ v			group	$(1,40) = .48$	$.492$	$.012$

Measure	Value	t	p	Value	t	p	Effect	Test	$< .03$	
Toe in/out (deg)	$4.61_{(3.84)}^{r}$	-4.064	.001	$5.54_{(3.98)}^{r}$	-2.534	.020	gain	$g_t(.25)_{(5.21,208.55)} = 7.05$	$<.001$	.150
	$6.42_{(4.61)}^{V}$			$6.46_{(4.44)}^{V}$			group	$(1,40) = .009$	.923	.001
							post-hoc	$g_t(.5)_{1.25,1.5,1.75}$ $g_t(.75)_{1.5,1.75}$	$<.008$	
Double support (%)	$21.91_{(2.75)}^{r}$	-3.278	.004	$24.89_{(2.78)}^{r}$	$-.906$	.376	gain	$g_t(.25)_{(3.82,152.76)} = 1.36$	.251	.033
	$23_{(2.61)}^{V}$			$25.41_{(3.39)}^{V}$			group	$(1,40) = 5.84$	.020	.127
Single support (%)	$39.06_{(1.22)}^{r}$	4.02	.001	$37.22_{(1.36)}^{r}$	$.258$	.799	gain	$g_t(.25)_{(2.92,116.86)} = 1.60$	.193	.039
	$38.29_{(1.41)}^{V}$			$37.18_{(1.29)}^{V}$			group	$(1,40) = 4.92$	.032	.110
							post-hoc	$g_t(.25)_{.5}$	$<.002$	
Stance phase (%)	$60.94_{(1.22)}^{r}$	-4.135	.001	$62.77_{(1.5)}^{r}$	$-.321$	.752	gain	$g_t(.25)_{(3.16,126.26)} = 1.66$	.177	.040
	$61.71_{(1.40)}^{V}$			$62.83_{(1.29)}^{V}$			group	$(1,40) = 5.79$	.021	.126
							post-hoc	$g_t(.25)_{.5}$	$<.002$	

CONTINUOUS GAIT IN VIRTUAL REALITY

In this chapter, we presented a study to compare the effects of cognitive task on velocity during long-distance walking in VR compared to walking in the real world. Therefore, we used an exact virtual replica model of the users' real surrounding. To reliably evaluate locomotion performance, we analyzed walking velocity during long-distance walking. This was achieved by 60 consecutive cycles using a left/right figure-8 protocol[1], which avoids the limitations of treadmill and non-consecutive walking protocols (i.e., start-stop). The results show a significant decrease of velocity in the VE compared to the real world even after 60 consecutive cycles with and without the cognitive task.

6.1 Introduction

In everyday life, humans move in the most natural form of locomotion (i.e. walking) to explore the real environment. Likewise, walking in a VE is important for many applications of VR such as architecture, training, or entertainment. To achieve this goal, an obvious implementation of virtual walking is to simply track the user's actual real-world motion and map them one-to-one to the virtual camera. However, one problem is that the user must be tracked within the available space of the room, which prevents the user from walking longer distances. Previous research primarily focused on treadmill virtual walking in terms of cognitive aspects of navigation [213, 214], gait performance [89, 90, 156]. Little is known regarding the tendency for locomotion performance during overground walking (i.e., natural ground surface) and over time [141]. Treadmill walking is supposed to be biomechanically identical to normal walking, but it alters users' perception of motion due to missing vestibular feedback and therefore may alter gait parameters as compared to overground walking. [50]

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[1]The "figure-8" cycle is also used in Infinity Walk, which is a therapeutic method for progressively developing coordination [243].

found that perceived speed in a complex environment decreased by about 10% on a treadmill compared to wide-area walking. In contrast, the experiments described in [5] report that the visually perceived velocity appears too slow compared to the physical walking velocity and presented experiments investigating the underestimation of visual flow velocities during treadmill walking. However, so far it remains unclear if the differences between a user's velocity during VR walking will further slowdown over time or if people will eventually speed-up and adapt their velocity to the VE and move with the same speed as in the real world. In this context, recent advancements in VR technology with the availability of affordable, low-cost tracked HMDs have allowed access to unexplored paradigms [222, 247], which provides a safe environment for analyzing how humans react and adapt to the VE during continuous walking.

In previous work described in [111, 107, 154], authors evaluated the differences of spatiotemporal parameters between a VE and the real world while using non-consecutive protocols (i.e., start-stop) for a few laps of walking on the pathway. This non-consecutive walking might not only be different from continuous walking, but its evaluation might also fail to identify the key characteristics of variability during locomotion tasks [180, 123]. Therefore, in order to reliably evaluate locomotion performance, we propose the collection of walking velocity for at least 60 consecutive cycles using left/right figure-8 protocol as illustrated in Figure 6.2a, which avoids the limitations of treadmill and non-consecutive walking protocols that lead to erroneous results [123]. By using a continuous overground walking protocol, researchers have shown that it is a valid and reliable method for investigating gait changes overtime with age [179]. Recently, two studies by [141, 251] have investigated the visuomotor adaptation of gait over time due to prolonged walking in a VE with healthy younger adults. In particular, they showed that differences in stride length, step width and stride variability were reduced significantly over time while walking in the VE. However, it seems necessary to provide participants with enough time to become familiar with the VE in order to reduce differences between real and virtual walking. In terms of gait training, it might enrich motor learning and rehabilitation outcomes by offering life-like environments making it possible to practice activities of daily living in a personalized, safe and accessible manner.

Moreover, we measured velocity under single and dual-task paradigms [1] to examine the influence of cognitive demands on gait parameters while walking in the real and the VE. In dual-task paradigms, participants were asked to perform two tasks simultaneously (i.e., walking while completing a cognitive task). We employed a dual-task paradigm typically to investigate the effects of concurrent cognitive task on locomotion performance, and evaluate the cognitive demands of locomotion task within the VE. The dual-task paradigm has not been widely used in VR, but it can be applied to understand influences of cognitive tasks

<table>
<tr><td>(a)</td><td>(b)</td></tr>
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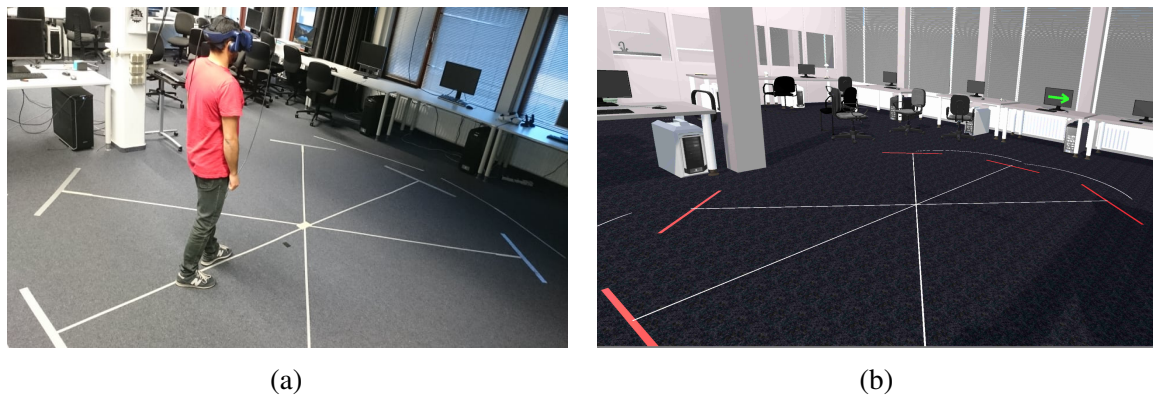
Fig. 6.1 Schematic representation of the experiment: (a) A participant walks in the real workspace with an HMD on the walkway of the figure-8 protocol, and (b) the VE, as viewed in the HMD.

on traveling or locomotion gait and balance [274]. Locomotion through VEs requires that the users have the ability to move about while accomplishing tasks [17]. These aspects may cause that users resort to employ strategies requiring additional cognitive resources within the VE, which compete for resources that are utilized for successful completion of the locomotion task.

The study aims at analyzing locomotion performance within a VE and the real world during overground walking with and without the cognitive task. In particular, we investigated the locomotion adaptation of gait parameters over time due to prolonged exposure to these conditions. Hypothesis are described as follow:

- Continuous walking leads to reduced differences in the gait velocity over time while walking within a VE and the real world.

- In contrast, distraction (i.e., performing a dual task) during continuous walking would slow adaptation within a VE and the real world.

6.2   Related Literature

Currently, the literature in longitudinal studies does not provide a consistent insight into the effects of overground walking on locomotion performance over time and the amount of cognitive demands that are induced by prolonged exposure to the VE. A VE with a treadmill-based walking interface has recently been used in several research studies [251, 219, 176]. Authors of these studies have investigated biomechanical aspects of walking over time in an attempt to analyze how humans react and adapt to the VE during treadmill walking. In

particular, [251] investigated the tendency for visuomotor adaptation in walking balance control using prolonged exposure to the visual flow perturbations. In addition, [219, 176] showed changes in spatiotemporal gait parameters and their variability as an effect of prolonged exposure to VR while walking on a treadmill. Recently, [120] explored clinical utilization of a VE over long periods of time for neuro-rehabilitation to treat individuals with Parkinson's disease. They concluded that using HMDs for longer bouts of walking in the VE did not induce simulator sickness, which could be a useful for future rehabilitation-based applications with this population. Importantly, it is unclear whether treadmill walking induces similar motor responses of overground walking [93]. Therefore, even after long periods of treadmill walking in a VE, gait does not completely approximate overground walking [219, 257].

Research on overground walking in a VE is relatively new, and little is known about how walking overground with a HMD matches gait parameters in the real world [154, 111, 107]. These aforementioned studies involved only a short period of walking as they were designed to investigate responses to particular manipulation in a VE. Therefore, time spent in the VE in these studies was much shorter, approximately total 2–3 min based on the distance walked and reported spatiotemporal gait parameters. The authors showed that healthy young adults walked in the VE with reduced velocity, increased step width and longer step times. In contrast, Janeh et al. reported that the gait parameters of healthy older adults [110, 107] and individuals with Parkinson's disease [108] (such as velocity, step length, double support, etc.) within a VE were not different than those in the real world. Thus, analyzing gait changes within a VE while walking overground for longer periods of time in healthy younger and older adults have not been systematically explored. A recent study suggests that participants can modify their overground walking pattern over time to accommodate the VE [141]; it provided an evidence for visuomotor adaptations during overground walking while wearing an HMD. In particular, results showed that participants adapted to the VE over time by increasing stride length and reducing stride width, stride length variability, step time variability, and step time. Despite that, participants still took shorter strides with wider stride width while walking in the VE.

Because most environments (i.e., real or virtual) are characterized by background noise, obstacles, and distracting visual/auditory stimuli, while walking, humans are required to perform cognitive and locomotion task simultaneously [6]; (i.e., walking while talking, texting on a mobile, or thinking about one's shopping list. Consequently, Nadkarni et al. [163] have shown that cognitive tasks which activate working memory and spatial attention can have an effect on human locomotion. In particular, they found that changes in gait, including speed, stride length, and double support time, were affected by cognitive tasks. Moreover, Oh and

LaPointe [175] have recently demonstrated the impact of cognitive load on gait parameters in a dual-task walking paradigm. In experiments by Marsh et al. [140] performed to investigate the cognitive costs of various locomotion interfaces. It was shown that locomotion techniques required spatial working memory resources. Unnatural locomotion techniques have been shown to affect performance in cognitive tasks negatively [276]. Their results suggest that locomotion with a less natural interface in a VE can increase spatial working memory demands, and locomotion with a smaller FOV can increase general attentional demands. A more recent study by Bruder et al. [21] investigated the cognitive costs that are induced by redirected walking based on curvature gains of different magnitudes on the walking user. They used a dual-task method to determine the mutual influence between a locomotion task using redirected walking and a concurrent task, which draws from either verbal or spatial cognitive resources. The results showed that using gain values outside these limit values increased the cognitive load and decreased the locomotion task performance.

For this paper we considered the role of cognitive demands as it relates to locomotion performance. Therefore, evaluation of locomotion performance under dual-task conditions may reveal subtle gait detriments that would otherwise go unnoticed. The primary purpose of this study was to examine whether convergence in velocity, occur over time while walking in the VE compared with the real world. A secondary purpose of the study was to compare errors in the cognitive task during dual-task walking in both real and virtual worlds to quantify the relationships between cognitive performance and locomotion performance.

6.3 Experiment

6.3.1 Participants

A total of 18 participants (4 female and 14 male, ages 19–37 years, $M = 25.16_{SD=4.65}$) completed the experiment. The participants were students, who obtained class credits, or professionals at the local department of computer science. All of our participants had normal or corrected-to-normal vision (i.e., 6 participants wore glasses). None of our participants reported a disorder of equilibrium. No other vision disorders have been reported by our participants. Participants wore an HMD for approximately 90 minutes during the experiment, 17 participants had some experience with HMDs before. The experience of the participants with 3D stereoscopic displays (cinema, games etc.) in a range of 1 (no experience) to 5 (much experience) was $M = 4.05_{SD=1.01}$. The body height of the participants varied between 1.64–1.9 m, $M = 1.81\,\text{m}_{SD=0.08m}$.

6.3.2 Apparatus

We performed the experiment in a laboratory room of 15m × 7.5m in size; see Figure 7.1 (a). During the experiment, the room was darkened in order to reduce the participant's perception of the real world while immersed in the VE. The VE was rendered using Unity3D; a cross-platform game engine with a custom-enabled VR communications and rendering library. As illustrated in Figure 7.1 (b), the VE showed a 3D laboratory, which is an exact virtual replica of the real laboratory. For rendering, system control and logging, we used an Intel computer with 3.4GHz Core i7 processor, 16GB of main memory and Nvidia GeForce GTX 1080 graphics card. The participants wore an HTC Vive Pro HMD for the visual stimulus presentation, which provides a resolution of 1440×1600 pixels per eye with a refresh rate of 90Hz and an approximately $110°$ diagonal FOV. We tracked sensors on the HMD using a Lighthouse tracking system (2 base stations emitting pulsed IR lasers) that tracked the user's head movement with sub-millimeter precision in the laboratory. We used the VIVE Deluxe Audio Strap with integrated over-ear headphones to render cognitive task, which provides an auditive feedback in the form of a letter sounds. Participants performed the cognitive task via button presses on a gamepad remote controller. Before the experiment, all participants provided written informed consent form to participate in this study, which was approved by the local ethics committee.

6.3.3 Method

We used a 4×6 within-subject factorial design, with 4 blocks of locomotion conditions (i.e., walking alone in the real world and the VE, walking while completing a cognitive task in the real world and the VE), and 6 blocks of cycle conditions (i.e., consecutive walking with left/right figure-8 cycles $\in \{1, 2, \ldots, 59, 60\}$). The order of the tested locomotion conditions was counter-balanced. In total, the participants completed $4 \times 60 = 240$ cycles (i.e., ≈ 2.1km within a duration of ≈ 33.6min). Participants completed 5 training cycles before each block. They were allowed to abort the experiment at any time and to take breaks at any time between blocks.

In addition, they filled out the simulator sickness questionnaire (SSQ) [118] immediately before and after the experiment, further the Slater-Usoh-Steed (SUS) presence questionnaire [256], and a demographic questionnaire. The total time per participant, including pre-questionnaires, instructions, experiment, breaks, post-questionnaires, and debriefing, was about two hours.
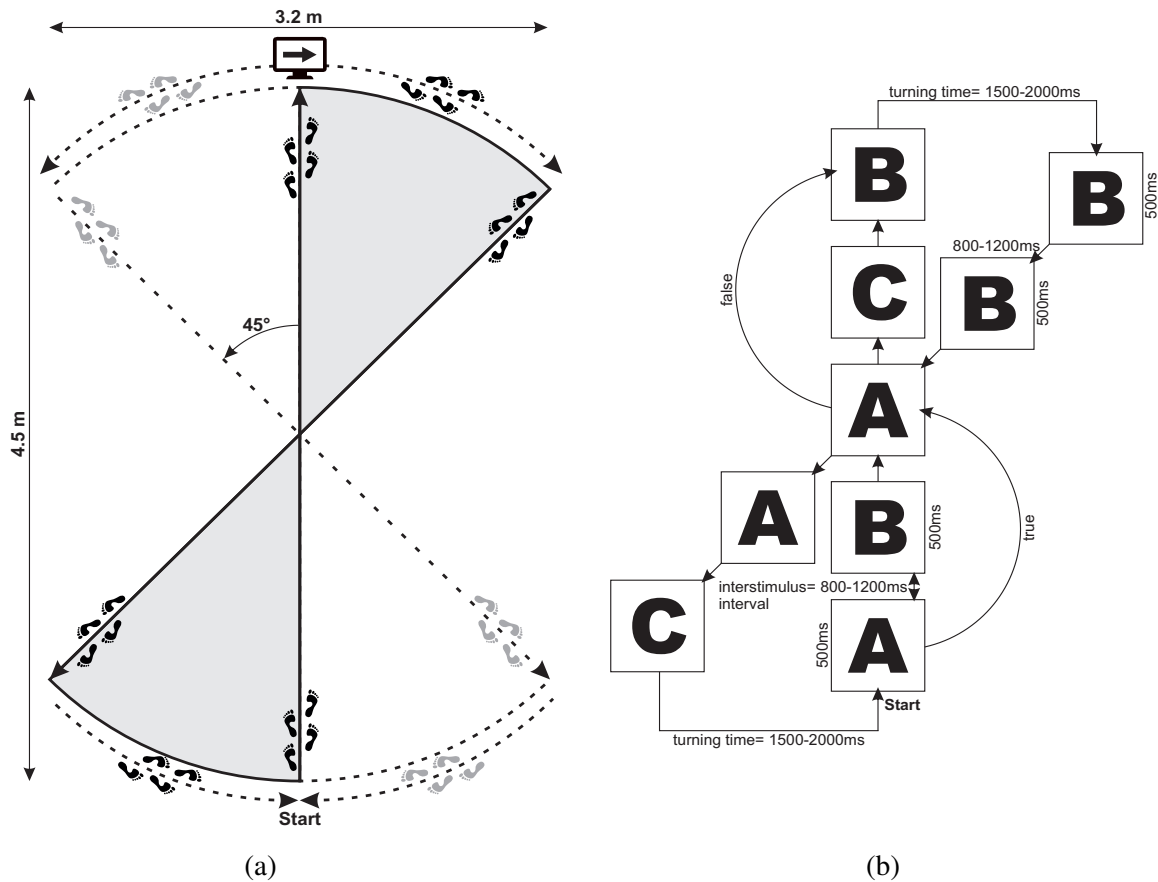
Fig. 6.2 Schematic representation of the left/right figure-8 protocol: (a) An Illustration of the typical pathway length during right figure-8 cycle (i.e. shaded area), and (b) illustration of the cognitive two-back task: auditory working memory sequences during right figure-8 cycle.

Locomotion Task

The task started by standing at the start line. Participants were then instructed to walk continuously at a normal pace in a path that described a left/right "8" placed on the floor in front of the participant to indicate the walking distance in the real and the VE. The two straight sections of the so-called "figure-8" cycle (as suggested by König et al. [123]) were each 4.5m in length and 1m wide Figure 6.2a. We collected gait data from the tracked HMD during walking of the 4.5m straight sections. The curved sections were not included in the assessment to allow for acceleration/deceleration while turning. The total distance walked during figure-8 cycle was 9m over a duration of $\approx$ 8.4s, after which the two sections of the figure-8 cycle ended, and participants were guided to turn left/right by the presentation of an arrow on a computer monitor mounted on the table in front of the first section of the figure-8 cycle. The next figure-8 cycle started once participants reached the start position. In each

condition, the participant had to walk 60 consecutive left/right figure-8 cycles about 540m over a duration of $\approx$ 8.4min with the HMD in the real and virtual worlds. This was done with and without the cognitive task.

During the experiment, an experimental assistant managed the cables of the HMD for each participant and ensured that participants could walk safely. In the experiment, mean values of walking velocity were computed along 60 consecutive figure-8 cycles.

Cognitive Task

As illustrated in Figure 6.2b, the auditory working memory task was a letter *two-back task* [122]. In every cycle, participants were shown a continuous stream of letters that were presented on HMD's headphones in both VE and the real world conditions. Participants were instructed to respond by pressing the button on the gamepad remote if a presented letter was the same as the one that came up two stimuli back in the sequence (true condition in Figure 6.2b). This task has a high auditory working memory load since it requires continuous on-line monitoring and maintenance of the presented letter until two consecutive letters presented. If (and only if) the stimulus matched the one that came up two stimuli before it, participants had to press a button on the gamepad remote controller. This task did not require large shifts of spatial attention or memory as the letters were presented continuously in the headphones of the HMD. The presentation duration for every stimulus on the cognitive paradigms was ≈ 500 ms with a pseudo-randomized interstimulus interval of $800-1200$ ms, thereby allowing for 10 stimuli for every trial with 6 recorded responses as shown in Figure 6.2b. Participants were instructed to perform the cognitive task to the best of their ability while walking the straight sections of figure-8 cycle in the locomotion dual-task conditions. In order to evaluate the effect of cognitive task on walking velocity within a VE and the real world. The decrements in walking velocity during the cognitive task were computed as a percentage using the following formula:

$$\Delta Velocity = \frac{Single\ Task - Dual\ Task}{Single\ Task} \times 100 \qquad (6.1)$$

6.4 Results

The results were normally distributed (Shapiro-Wilk test at the 5% level). We analyzed the results with a repeated-measures ANOVA and Tukey multiple comparisons at the 5% significance level with Bonferroni correction. Degrees of freedom were corrected using
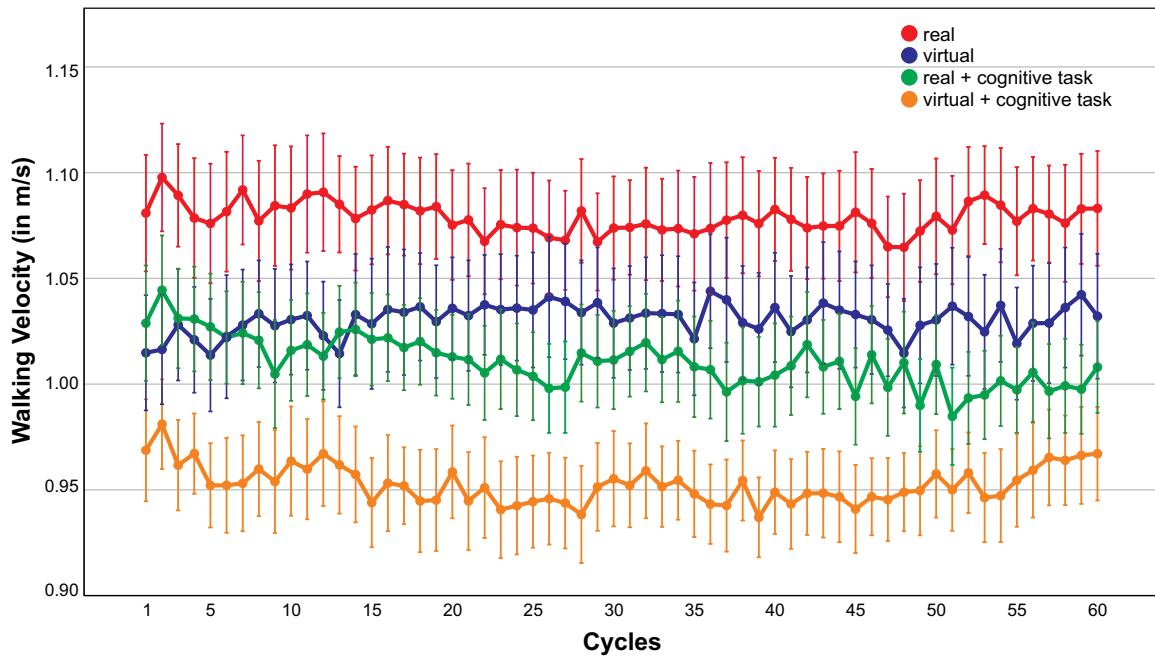
Fig. 6.3 Pooled results of the experiment with figure-8 cycles on the horizontal axis for walking velocity on the vertical axis.

Greenhouse-Geisser estimates of sphericity when Mauchly's test indicated that the assumption of sphericity had been violated.

6.4.1 Locomotion Performance

Figure 6.3 shows the pooled results for the figure-8 cycles plotted against the performance in the locomotion with and without the cognitive task for both real and virtual worlds. The vertical bars show the standard error of the mean. The *x*-axes show the pooled figure-8 cycles, the *y*-axes show the mean of walking velocity.

We found a significant main effect of locomotion conditions for walking velocity ($F_{2.38,33.39} = 23.06$, $p < .001$, $\eta_p^2 = .622$). Post-hoc tests showed significant differences for walking velocity between locomotion conditions. While walking in the VE with no task, participants walked significantly with slower velocity ($p = .04$) compared to the real world walking. During walking in the VE with the cognitive task, participants walked significantly with slower velocity ($p = .02$). The results revealed no significant interaction effects between (*cycle × locomotion*) conditions for walking velocity.

Furthermore, in order to ensure that counterbalancing had worked and does not affect the study outcome. We performed a repeated-measures ANOVA across all participants in each locomotion condition given first followed by second, third and fourth as an independent

variable on walking velocity as dependent variable. We found no significant main effects of locomotion conditions with given order on walking velocity ($F_{2.41,40.99} = 1.29$, $p = .2$, $\eta_p^2 = .071$). The group effects are thus symmetric.

6.4.2 Cognitive Performance

No main effects found of cycles conditions along a continuous walking velocity ($F_{4.11,57.59} = .91$, $p = .4$, $\eta_p^2 = .061$) and percentage of the correct responses ($F_{2.77,44.25} = .48$, $p = .6$, $\eta_p^2 = .029$) and therefore pooled the data. Figure 6.4 shows the pooled results for the tested locomotion conditions plotted against the performance in the locomotion and cognitive tasks. The colored lines show the results for the verbal task, spatial task, or condition without cognitive task. The x-axes show the pooled real/virtual locomotion conditions, the y-axes show the walking velocity in Figure 6.4a, and the percentage of correct responses in the cognitive tasks in Figure 6.4b. The vertical bars show the standard error of the mean.

Walking velocity was compared for the real/virtual locomotion with and without the cognitive task with a paired t-test. We found a significant decrease of 6.5% velocity during dual-task walking in the real world ($p < .001$) compared to the single-task walking. Also while walking in the VE, we found a significant decrease of 8% velocity during dual-task walking ($p < .001$) compared to the single-task walking. Additionally, the differences in accuracy of serial subtraction were analyzed using a paired t test, between velocity of walking alone and and with cognitive task within the VE and the real world. We found a significant velocity decrease by 13.3% in the VE ($p < .001$) compared to the real world.

Moreover, we compared the percentage of correct responses for the real/virtual locomotion with the cognitive task with a paired t-test. We found no significant main effect of the cognitive task on the percentage of correct responses between the real and virtual walking ($p = .3$). Participants made comparably high task performance with a continuous walking both for the real $M = 80.99\%\ _{SD=11.39}$ and the virtual worlds $M = 82.99\%\ _{SD=8.48}$.

6.4.3 Questionnaires

We measured a mean SSQ score $M = 7.27\ _{SD=22.44}$ before the experiment, and a mean SSQ score of $M = 11.29\ _{SD=26.6}$ after the experiment. We analyzed the SSQ questionnaire scores with a non-parametric Wilcoxon Signed Rank Test at the 5% significance level. The SSQ scores indicate overall high simulator sickness symptoms for extensive continuous walking with an HMD, and we found a significant increase of symptoms over the time of the experiment; $Z = -2.38$, $p = .01$. The mean SUS PQ score for the sense of feeling present in the VE was $M = 5.26\ _{SD=.52}$ which indicates a high sense of presence [256]. Additionally,
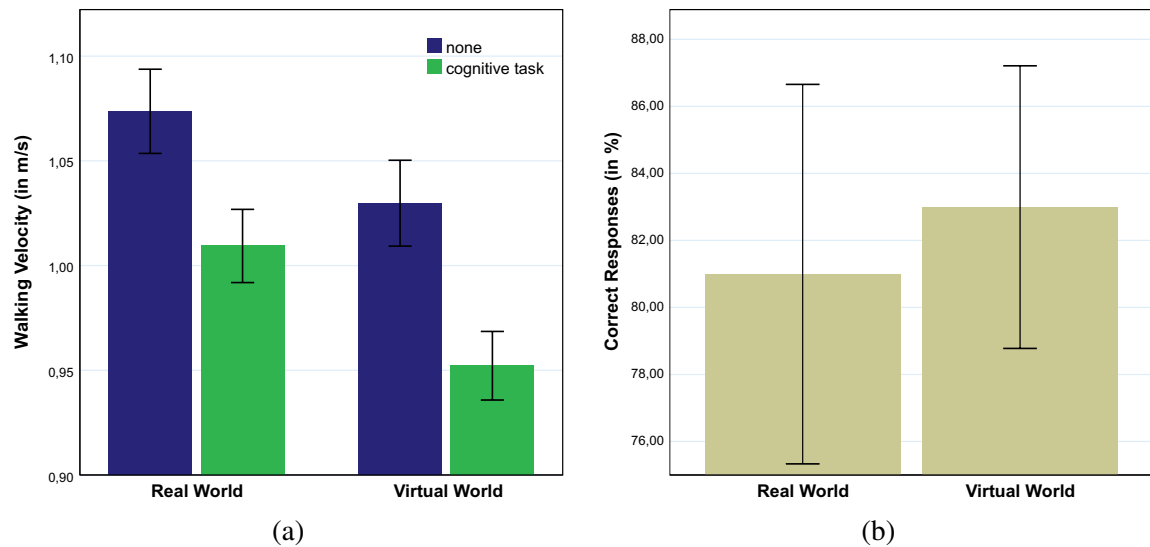
Fig. 6.4 Pooled results of the experiment with real/virtual worlds on the horizontal axis for (a) walking velocity, (b) percentage of correct responses of the cognitive task on the vertical axis.

participants judged their fear to collide with the walls of the room or other physical obstacles while immersed with the HMD during the experiment as comparably low (rating scale, 0=no fear, 5=high fear); $M = 1.68_{SD=.76}$.

6.5 Discussion

Our results suggest participants gait between the real and virtual environment was different in terms of velocity. These differences persisted even after long periods of walking. These results are consistent with those reported by [111, 107, 154], whose younger adults tended to walk slower within a VE during a short periods of walking. Interestingly, we found that velocity (Figure 6.3) was reduced significantly while walking continuously in the VE after 60 cycles compared to the real world, which indicates an almost linear effect of consecutive cycles over time. However, our results differed from those of [141, 251]. These authors reported that over time, participants adapt to the VE and provide comparable results to motor adaptation in real environments [4].

Moreover, walking velocity (Figure 6.3) was significantly decreased while performing a cognitive task in the VE and the real world. This suggests that that cognitive demands have an impact on gait and that attention processes are involved in walking. Earlier works [91, 6, 178] reported a reduction in walking velocity, cadence and stride length together with

an increase in stride time when cognitive demands were modulated. These modulations may serve as a way to maintain gait stability in conditions of higher cognitive demand [2].

These are important findings for prolonged use of VR technology, which could help developers and researchers improve the naturalness of walking in VEs [159], or to propose evaluation metrics where participants practice over a prolonged period. Our findings further suggest that gait detriments in locomotion performance within VEs highly depend on many potential contributing factors, which span the whole range from technical, perceptual, cognitive to simulator sickness aspects (see Chapter 8).

6.6   Conclusion

The objective of this study was to analyze changes in velocity over time while walking overground within a VE and the real world with and without the cognitive task. We ran a controlled user study to investigate locomotion adaptation over time due to prolonged exposure to these conditions. Results show a significant decrease of velocity along 60 consecutive left/right figure-8 cycles in the VE compared to the real world. Moreover, walking velocity was further decreased within the VE compared to the real world while performing a cognitively challenging task. Furthermore, these findings bring forth several issues that potentially have interest for the scientific and clinical community and may have positive implications for gait training [86]. The experiment was based upon the evaluation of younger adults, and therefore may be relevant to the examination of, or interventions for, gait disorders in older adults.

GAIT TRAINING IN PARKINSON'S DISEASE

In PD, a strong relationship between gait asymmetry and FOG has been found. The purpose of this pilot study was to find a VR based gait manipulation strategy to improve gait symmetry by equalizing step length. Natural gait was compared with walking conditions during VR based gait modulation tasks that aimed at equalizing gait symmetry using visual or proprioceptive signals. Compared to natural gait, VR manipulation tasks significantly increased step width and swing time variability for both body sides. Within the VR conditions, only the task with proprioceptive-visual dissociation by artificial backward shifting of the virtual foot improved significantly spatial asymmetry with comparable step lengths of both sides. The hypothesis-driven VR tasks represent an efficient tool to manipulate gait features as gait symmetry in PD potentially preventing FOG. This pilot study offers promising VR based approaches for rehabilitative training strategies to achieve gait symmetry and prevent FOG.

7.1 Introduction

Gait disturbances represent the main symptom impacting every-day self-dependence and quality of life in PD patients. Gait disturbances and FOG promote frequent falls occurring in up to 87% of PD patients [84, 215, 216], which result in hospitalization, immobilization, and a loss of self-confidence. The Parkinsonian gait disturbance is characterized by continuous features [69] as reduced speed, shorter step length, increased stride-to-stride variability, reduced automaticity, and increased gait asymmetry [199, 81] as well as episodic phenomena such as freezing episodes, festination, and starting arrests [69]. The freezing episodes are closely associated with continuous gait characteristics particular to the degree of gait asymmetry [60, 188]. With regard to limited therapeutic options of medication or deep brain surgery, physical training strategies have evolved to be a focus of interest to improve gait and the freezing of gait.

VR has emerged as an efficient tool in physical rehabilitation [192, 43] in PD. VR offers the opportunity to simulate immersive, controllable, changeable environments with the option to create individualized, specific training programs. To achieve a more natural experience and highly immersive VR, simulations are often generated using a HMD. This system represents a low-cost, easy accessible, portable pedestrian simulator system, which can be used in different settings, in the lab, at home, or during physiotherapy. VR-based interventions attempt to promote neuroplasticity and motor learning [106, 177, 66]. *motor learning strategies* (MLS) consider specific motor learning principles, which are patient and task-specific. VR offers the opportunity to facilitate the incorporation of motor learning principles such as real-time multisensory feedback, task variation, objective progression, and task-oriented repetitive training [192]. VR has been shown to improve balance and gait especially in PD [192, 43, 151], particularly when combined with conventional rehabilitation. However, to date, many of the VR studies lacked a clear rationale for intervention programs and did not utilize motor learning principles. Particularly VR training strategies developed by theory-driven protocols may assist motor learning implementation for optimized VR-based treatments. Implementation of patient-tailored motor learning strategies into the design and planning of VR interventions may enhance the efficiency and improve the therapeutic outcome [192].

Recently, novel insights revealed the heterogeneity of the freezing of gait in PD with distinct freezing phenotypes as asymmetric-motor, anxious, and sensory-attention phenotypes [53]. Particularly, the strong relationship between gait asymmetry and freezing of gait in PD patients [60, 188] represents a therapeutic clue. Reduced episodes of FOG might be attained by establishing enhanced lower limb gait symmetry. Dopaminergic medication seems to promote improved gait symmetry in PD patients [188]. Deep brain stimulation (DBS) of the subthalamic nucleus improves gait symmetry and freezing of gait by reducing the amplitude of the better side [55], which can be used for trouble-shooting in PD patients with FOG and DBS. Specific physiotherapeutic approaches designed to achieve gait symmetry [199] or using treadmill training [60] or split belt-locomotion [56] effectively reduced gait asymmetry and FOG.

The rational of the current study is to apply the equalization of gait asymmetry as the motor learning principle by use of VR techniques with real-time multisensory feedback to increase efficacy of the training strategy. Specifically, we aim to exploit and optimize the VR environment by variation of different conditions to define the method with the best possible equalization of the pathological gait asymmetry in view of step length. This would lead to a theory-driven, individualized therapeutic approach, which might be used in future studies for long-term gait symmetry training to reduce FOG and falls in PD patients.

7.2 Materials and Methods

The study was conducted in accordance with the Declaration of Helsinki [71] and approved by the local Ethics Committee of the Medical Council in Hamburg and the local Ethics Commission of the Department of Informatics in Hamburg (reference number PV5281). All participants gave written informed consent.

7.2.1 Subjects, clinical data and questionnaires

A total of 16 male patients with idiopathic PD participated in the study. They were recruited using an announcement from the outpatient clinic/Parkinson day clinic of the Department of Neurology of the University Medical Center Hamburg-Eppendorf. Inclusion criteria were diagnosis of idiopathic PD, according to the UK PD Society Brain Bank criteria [97] and the criteria of the German Society of Neurology (Deutsche Gesellschaft für Neurologie, DGN), Hoehn & Yahr stage 2–3 [87], no DBS or medication pump, clinically relevant gait disorder with freezing, but without the tendency to fall, walking independently without a walking aid, and normal vision, if necessary, with a vision aid. Other inclusion criteria include no severe dementia (Montreal Cognitive Assessment MoCA [164] > 21, this cut-off has also been used in [85]) as dementia can be associated with gait disturbances [157], Giladi's freezing of gait questionnaire [70] with a score > 6 (this cut-off has been used in [206]), no severe polyneuropathy (pallaesthesia > 4/8), and no spinal stenosis or severe musculoskeletal disorders that impair sensorimotor function or gait.

All patients were tested in the *on* medication state, which was defined as about 1 h after the intake of their regular medication. Prior to the main experiment, patients were characterized by a short interview, demographic questions, and clinical scores and questionnaires:

1. Unified Parkinson Disease Rating Scale of the Movement Disorder Society (MDS-UPDRS) part III [72] as a general motor score.

2. German version of the freezing of gait questionnaire by Giladi et al. [70], Vogler et al. [259] as a subjective assessment of FOG in PD.

3. Ziegler's freezing of gait course [279] as an objective assessment of FOG in PD.

4. Short, 7-item version of the Berg balance scale [31] as an objective measure of balance as a parameter of gait stability.

5. German version of the Montreal cognitive assessment (MoCA) [164] as a measure of cognitive function in PD.

Table 7.1 Patient characteristics, clinical data, and questionnaires

Patient Characteristics and Clinical Data	Mean ± SD	(min–max)
Age	67.6 years ± 7	(49–77)
Gender	15 male	–
Handedness	1 ambidextrous, 14 right	–
Hoehn & Yahr Scale	2–3	–
Levodopa (Minutes After Intake)	58.3 minutes ± 19.8	(40–120)
Leg Length	Left: 93.8cm ± 3.7	(88–102cm)
	Right: 93.7cm ± 4.0	(88–102cm)
	Short: 93.7 cm ± 4.0	(89–102 cm)
	Long: 93.8 cm ± 3.8	(88–102 cm)
Onset of PD Symptoms	11.5 years ± 4.9	(2–19)
LPD Diagnosis	9.5 years ± 4.9	(1–17)
Onset of Gait Disturbance	5.5 years ± 4.4	(1–17)
MDS-UPDRS part III	25.5 ± 7.2	(12–37)
Giladi's FOG	27.5 ± 10.6	(12–47)
Berg Balance	24.7 ± 1.8	(19–26)
Ziegler's FOG	7.2 ± 5.9	(0–17)
MoCA	27.5 ± 2.0	(23–31)
Pre-SSQ	16.45 ± 16.59	(3.74–52.36)
Post-SSQ	15.21 ± 17.04	(3.74–56.1)
SUS	3.5 ± 0.8	(1–5.83)
PDQ-39	25.31 ± 12.83	(5.76–45.21)

6. German version of the Parkinson's Disease Questionnaire (PDQ-39) by Jenkinson et al. [112], Berger et al. [9] as a measure of quality of life in PD.

7. Simulator Sickness Questionnaire (SSQ) [118] before and after the experiment as a measure of cybersickness symptoms.

8. Slater, Usoh, and Steed Questionnaire (SUS) [230] as a measure of presence in the virtual environment.

One patient withdrew from the study due to exhaustion while 15 male patients with idiopathic PD completed the experiment. Table 1 gives an overview of patient characteristics as well as results of clinical scores and questionnaires.

7.2.2   Apparatus

Gait parameters were measured using a GAITRite electronic walkway system [165]. The GAITRite consists of a walkway with overall dimensions of 90cm × 7m × 3.2mm. The

virtual space was rendered using Unity3D and showed an outdoor scene with a long grass pathway (see Figure 7.1b). We included a virtual mat in the VE that exactly matched the real GAITRite system walkway. The virtual item correlates to the start (green line) and target (red line) lines were placed on the floor in front of the participant to indicate the walking distance in the virtual world as well as the real world. We used an Intel computer for rendering with 2.7 GHz Core i7-6820HK processor, 16GB of main memory and Nvidia GeForce GTX 1070 graphics cards. The participants wore an HTC Vive HMD for the stimulus presentation, which provides a resolution of 1080×1200 pixels per eye with a refresh rate of 90Hz and an approximately $110°$ diagonal FOV. The HMD uses more than 70 sensors including a MEMS gyroscope, accelerometer and laser position sensors. We tracked sensors on the HMD using a Lighthouse tracking system (2 base stations emitting pulsed IR lasers) that tracked the user's head movement with sub-millimeter precision in the laboratory. In order to provide a realistic VR scenario while walking, we attached the Vive controllers to the participant's leg to present virtual feet in VR resembling the participant's real feet. We tracked sensors on the HMD (i.e., that tracked the user's head movement) and the Vive controllers (i.e., that tracked the user's leg movement) using a Lighthouse tracking system with sub-millimeter precision in the laboratory.

7.2.3   Experimental Procedure

We performed the experiment in a laboratory room of $9m \times 4m$ in size (see Figure 7.1a), which was shielded from light and noise. After providing the patients with the full equipment, participants were asked to first assume the start position by standing in an orthostatic pose at the start line. Then, participants were instructed to walk at their self-selected pace along the walkway of the GAITRite system while stopping at the location of the target line (see Figure 7.1b). After each trial, the participant had to walk back to the starting point alongside the GAITRite walkway. During the experiment, we provided comfort to the head from the weight of the cables by having an assistant manage the cables for each participant. The total time per participant, including questionnaires, interview, instructions, experiment, individual breaks, and debriefing was about 1.5 to 2 hours.

Seven different walking conditions were performed with three repetitions each (trials 1, 2, and 3) and a duration of about 5 to 6 min. After a short familiarization phase to become used to the setup of each trial, patients were asked to walk at their self-selected pace and start using the dominant leg, which was labeled as the leg that was predominantly used in the pull test to stabilize the stance. Gait analysis was divided into two parts:
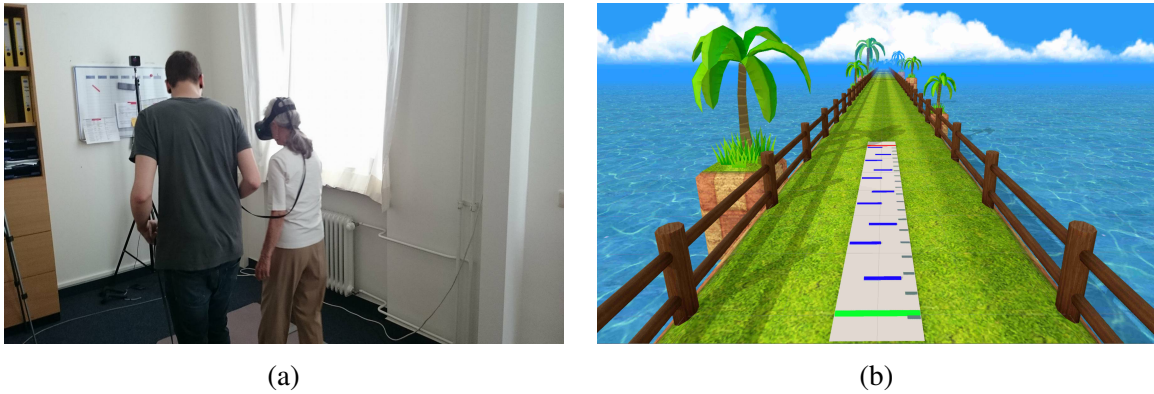
Fig. 7.1 Experimental setup: (a) a participant walks in the real workspace with an HMD over the GAITRite walking surface. (b) participant's view of the virtual environment on the HMD.

1. Conditions without gait asymmetry equalization as a motor learning strategy (non-MLS):

   (a) Natural walk: Walking naturally on the GAITRite without HMD. The HMD was reversed and positioned on the participants' head to ensure the same weight and posture during each condition.

   (b) Diving glasses: Walking naturally on the GAITRite with diving glasses. The diving glasses had a similar weight and FOV compared to the HMD.

   (c) Natural virtual walk: Walking naturally on the GAITRite with HMD presenting the virtual environment without visual targets (see Figure 7.2a).

2. Conditions with gait asymmetry equalization as a motor learning strategy (MLS):

   (a) Symmetrical without virtual feet: walking on the GAITRite with HMD. Virtual lines are presented each with a distance that corresponds to the individuals' step length of the longer side (see Figure 7.2b); which step length of shorter leg expressed as follow.

   $$SLs_{new} = SLs_{old} + (SLl - SLs) \tag{7.1}$$

   Participants were asked to step on the lines, but walk at the natural speed. To evaluate if the visual target signals might lead to greater gait symmetry.

   (b) Symmetrical with virtual feet: walking on the GAITRite with HMD. Virtual lines are presented each with a distance that corresponds to the individuals' step length

of the longer side (see Figure 7.2c); step length of shorter leg (see equation 7.1). Participants are asked to step on the lines with the middle of their feet, but walk as normal as possible while remaining in the middle of the pad. To evaluate if multiple visual signals (i.e., target and proprioceptive signals) might influence gait symmetry.

(c) Asymmetrical with virtual feet: walking on the GAITRite with HMD. Virtual lines are presented with asymmetrical distances (see Figure 7.2d); step length of shorter leg is exaggerated as the following.

$$SLs_{new} = SLs_{old} + 2(SLI - SLs) \qquad (7.2)$$

Participants are asked to step on the lines with the middle of their feet, but walk as normal as possible while remaining in the middle of the pad. To evaluate if an exaggeration of step lengths of the shorter leg is needed to achieve gait symmetry.

(d) Manipulated virtual feet: walking on the GAITRite with HMD. Virtual lines are presented each with a distance that corresponds to the individuals' step length of the longer side. However, the foot on the shorter side is visually shifted backwards (see Figure 7.2e); shift of the manipulated virtual foot expressed as follow:

$$M_{shift} = SLI - SLs \qquad (7.3)$$

To evaluate if a visual shifting of the proprioceptive signal leads to a greater gait symmetry.

In the end, a natural walk was repeated to ensure gait stability over time and explore possible after-effects of short-term application of VR gait manipulation tasks.

7.2.4 Gait Data

For all seven conditions mentioned above, spatio-temporal gait parameters were analyzed through the GAITRite walkway system. The mean value of all the three trials per condition was used to calculate the individual gait parameters. For the non-MLS and MLS conditions, we focused on gait variability and gait asymmetry. The following gait parameters were used: Step length (cm), velocity (cm/s), cadence (steps/min), the gait asymmetry index (G_iAsym) corresponding to the ratio of longer step length/short step length, i.e., $G_iAsym := \frac{long}{short}$ (adapted from [56, 182]), stride velocity (cm/s), step time (s), step width (cm), double support time (s), and swing time (s). Furthermore, we used the leg length of each participant
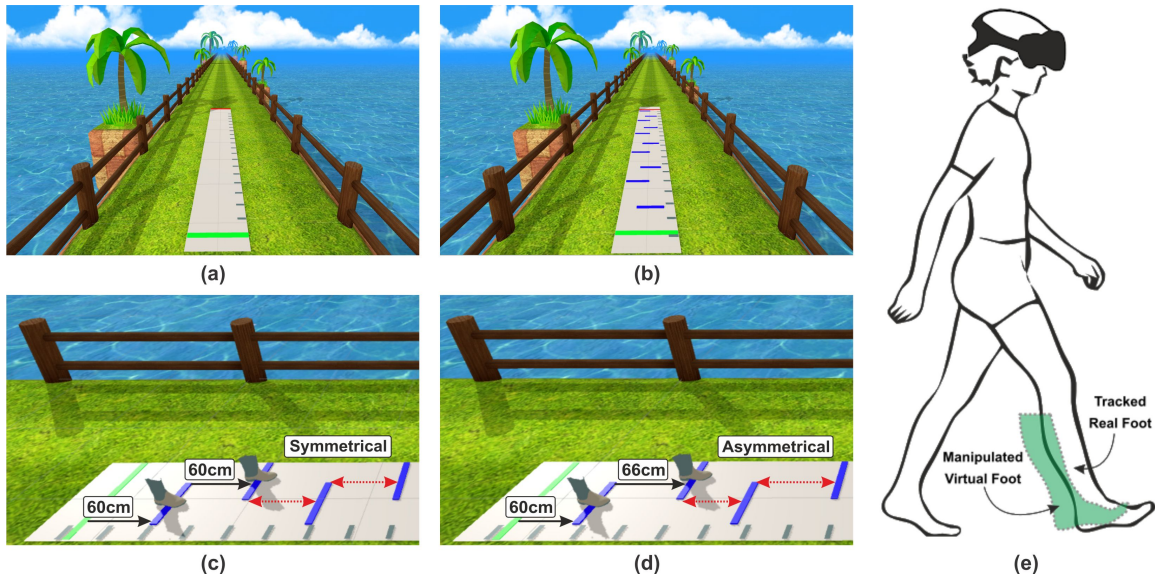
Fig. 7.2 Illustration of experimental conditions.

(see Table 7.1) to calculate FAP score. As a measure of gait variability, we used the coefficient of variation (CV) of the previously mentioned gait parameters that was already introduced in previous studies [25, 7].

$$CV = \frac{SD}{Mean} \times 100 \tag{7.4}$$

7.2.5   Data Analysis

We confirmed the assumptions of the ANOVA for the experimental data. The results were normally distributed according to a Shapiro-Wilk test at the 5% level. Degrees of freedom were corrected using Greenhouse and Geisser estimates of sphericity in case the assumption of sphericity had been violated. Data analyses were performed using the IBM SPSS software version 25.0.

For exploratory reasons, we first performed an overall ANOVA to look for gait velocity differences between each of the three trials (trials 1, 2, and 3) during all nine conditions to check if the self-selected pace of the patients differed between the trials or conditions.

Afterward, for non-lateralized gait parameters (velocity, cadence, gait asymmetry, and FAP score), we performed repeated measures one-way ANOVAs to evaluate the effects of the different gait manipulation methods on these parameters, i.e., three factors in the non-MLS conditions (baseline, diving glasses, and real virtual), five factors in the MLS conditions (baseline, symmetrical without feet, symmetrical with feet, asymmetrical with feet, visual-proprioceptive dissociation). For lateralized gait parameters, we analyzed the

results with two-way repeated-measures ANOVA (gait modulation method × body side). For each condition, the term *body side* represented the leg with longer or shorter step length during the baseline condition. For the testing of after-effects, the natural gait condition at the beginning and at the end of the experimental session were compared. Post-hoc effects were calculated at the 5% significance level. We have also provided η_p^2 effect sizes to supplement the interpretation of the results. The interpretation of η_p^2 suggests that effect sizes greater than .01 but less than .06 be considered small, greater than .06 but less than .14 be considered medium, and greater than .14 be considered large [34].

The SSQ questionnaire scores before and after the experiment were compared using the non-parametric Wilcoxon Signed Rank Test at the 5% significance level. Spearman's correlation coefficient was used to correlate clinical data (age, Ziegler score, Berg balance score, MoCA, and MDS-UPDRS III, SUS) with the gait parameters mentioned above in the baseline condition. Thus, 17 parameters were used for correlations, which needed the Bonferroni correction for multiple testing.

7.3 Results

7.3.1 General Aspects

In six patients, the left side was the most affected body side (side with symptom onset and ongoing symptom predominance) whereas the other eight patients had pronounced symptoms on their right body side. In only five out of 15 patients (= 33.3%), the clinically most affected body side represented the leg with the shorter step length. None of our PD patients represented FOG during the experiment even though all patients claimed freezing in the questionnaires, which is a common phenomenon in experimental settings [142]. Nine patients used their left leg and six patients used their right leg as the dominant leg in the pull test (i.e., starter leg). Gait parameters did not differ between conditions in which PD patients started walking with the dominant or non-dominant leg. Although patients were asked to walk at their self-selected pace and performed a training session before each recorded condition, gait velocity during the first trial of each condition was significantly slower compared to the second and third trial, which might be explained by a prolonged adaptation to the changing conditions.

7.3.2 Objective Gait Parameters

Objective gait parameters of PD patients in the different non-MLS conditions (baseline, diving glasses, real virtual) and MLS conditions (symmetrical without feet, symmetrical with feet, asymmetrical with feet, and manipulated) are shown in Figure 7.2.

Non-MLS Conditions

The two-factorial ANOVA of lateralized gait parameters with the intra-subject factors gait manipulation method (baseline, diving mask, and real virtual) × body side (short side, long side) revealed a significant main effect of the factor gait manipulation method on step time ($F_{1.76,24.68} = 4.93$, $p = .01$, $\eta_p^2 = .261$) , step width ($F_{1.62,22.7} = 13.48$, $p < .001$, $\eta_p^2 = .491$) , swing time ($F_{1.63,22.92} = 9.15$, $p = .002$, $\eta_p^2 = .395$) , and step time variability ($F_{1.9,26.61} = 3.97$, $p = .03$, $\eta_p^2 = .221$). The factor body side was significant for step length ($F_{1,14} = 42.62$, $p < .001$, $\eta_p^2 = .753$) step time ($F_{1,14} = 7.64$, $p = .01$, $\eta_p^2 = .353$) , and swing time variability ($F_{1,14} = 10.49$, $p = .006$, $\eta_p^2 = .428$). One-way ANOVA revealed a significant main effect of gait modulation on cadence ($F_{1.8,25.91} = 7.39$, $p = .003$, $\eta_p^2 = .346$). Post-hoc tests for both ANOVAs with significant differences of gait parameters within the different gait conditions are shown in 7.3a. In post-hoc tests, there was no significant difference between the baseline condition and diving glass condition with regard to all gait parameters. These findings indicate that reducing the FOV to the size we used in the experiment by use of HMD did not impact gait (see Figure 7.3a). Furthermore, while walking with an HMD in the real-virtual condition, the step length was comparable to the baseline condition, but other gait parameters changed significantly. PD participants walked with a decreased step time, an increased step time variability, a decreased swing time, a widening step width, and increased cadence, indicating a rather insecure gait pattern in VR.

MLS Conditions

The two-factorial ANOVA with the intra-subject factors gait manipulation method (baseline, symmetrical without feet, symmetrical with feet, asymmetrical with feet, manipulated) × body side (short side/long side) revealed a significant effect of gait manipulation method on step length ($F_{2.21,30.99} = 8.85$, $p = .001$, $\eta_p^2 = .387$), step time ($F_{2.65,37.17} = 3.67$, $p = .02$, $\eta_p^2 = .208$), swing time ($F_{2.43,34.1} = 3.39$, $p = .03$, $\eta_p^2 = .195$), step width ($F_{3.05,42.74} = 10.8$, $p = .001$, $\eta_p^2 = .436$), step time variability ($F_{2.97,41.57} = 11.51$, $p = .001$, $\eta_p^2 = .451$), and swing time variability ($F_{1.95,27.43} = 6.18$, $p = .006$, $\eta_p^2 = .306$). The factor body side revealed significant differences only for step length ($F_{1,14} = 6.63$, $p = .02$, $\eta_p^2 = .322$). One-way ANOVA revealed a significant main effect of gait modulations only on cadence ($F_{3,42.07}$

$= 5.44$, $p = .003$, $\eta_p^2 = .280$). Furthermore, a repeated measure ANOVA determined that the G_iAsym mean differed statistically and significantly between gait manipulation methods ($F_{2.82, 39.54} = 3.17$, $p = .03$, $\eta_p^2 = .185$). Post-hoc tests of gait parameters within the different gait manipulation conditions are shown in 7.3b. There were no significant interaction effects between the gait manipulation method$\times$body side for all gait parameters.

Post-hoc tests revealed that the different gait manipulation strategies have been associated with an increase in the step width, step time variability, and swing time variability for both sides compared to the baseline condition reflecting an insecure gait. G_iAsym significantly decreased only during a manipulated virtual foot condition with visual-proprioceptive dissociation. Step length of the short side approached the step length of the long side, which reflects a lower degree of gait asymmetry in the manipulated condition compared to the baseline condition ($F_{2.82, 39.54} = 3.17$, $p = .03$, $\eta_p^2 = .185$). In summary, the manipulated condition with visual-proprioceptive dissociation was the most effective method to reduce gait asymmetry and to adjust step lengths of both legs.

After-effects

Comparison of natural gait at the beginning and at the end of the experiment revealed no significant differences for any gait features. 2-factorial ANOVA revealed no significant effects for the factor gait modulation method or body side for all gait parameters.

7.3.3   Simulator Sickness and Presence

SSQ scores of PD patients before and after the experiment are given in Table 1. They indicate low simulator sickness symptoms for walking with an HMD, and we found no significant increase of symptoms over the time of the experiment (Z=-1.02, p=.3). The mean SUS Questionnaire scores for the sense of feeling present in the virtual environment was (M=3.5, SD=.8), which indicates a moderate sense of presence. Additionally, participants judged their fear to collide with the walls of the room or other physical obstacles while immersed with the HMD during the experiment as comparably low (i.e., along a 0–5-point rating scale, 0 = no fear, 5 = high fear, M=.38 SD=.47).

7.3.4   Correlations

Clinical characteristics of PD patients as age, cognition (MoCA), motor score (MDS-UPDRS III), gait and freezing related scores (Berg balance score, Ziegler's FOG score), and the stimulation sickness scores (SUS, pre-/post-SSQ) were not significantly correlated with any of the gait parameters after the Bonferroni correction for multiple testing.

7.4   Discussion

The main purpose of this project was to develop a theory-driven, individualized therapeutic rehabilitative approach by defining optimized motor learning strategies implemented in VR for the best possible equalization of the pathological gait asymmetry in PD. To our knowledge, this is the first study with the specific VR-based approach to implement gait symmetry as the motor learning principle in PD.

The present study revealed three main findings, which are of relevance in the design of future rehabilitative VR training strategies. First, during a natural walk in the baseline condition, PD patients revealed the typical, clinically relevant gait asymmetry with a significant difference between step lengths of both legs. The leg with the shorter step length was only in 33.3% of patients the clinically the most affected one contradicting the clinical impression that the range of motion decreased more on the clinically most affected side. This finding implies that, in future gait symmetry training, the first step should be to define the worst and best side by measuring step lengths in quantitative gait analyses. Second, walking in VR induced an increase of step width, cadence, and gait variability, which indicates an insecure gait pattern during the VR conditions. This is not due to a simple reduction of the FOV, since walking with diving glasses did not impact the gait pattern. Similarly, in a previous study, we found alterations of nearly all tested gait parameters in healthy subjects using HMD [111, 107, 110, 109]. Third, we found the optimized equalization of gait asymmetry in the VR condition with dissociation of the visual and proprioceptive inputs (i.e., manipulated condition). PD patients overcame the spatial asymmetry and exhibited a comparable step length by enlarging the step length of the short side, an adapted step time, and a swing time variability of both sides in this manipulated condition. Thus, our findings suggest that virtual walking with visual-proprioceptive dissociation might have important implications for the restoration of gait symmetry in PD patients with FOG [199, 60, 56].

Virtual reality provides a unique platform to study the complex interactions between an individual's movement and the environment. VR has tremendous potential to advance both our understanding and treatment of gait impairments. It has been applied successfully in PD patients in previous studies for motor and cognitive assessment, motor trainings, and rehabilitation [192, 43]. In a currently published review, evidence for a positive effect of the VR exercise on certain gait parameters as step and stride length has been found with overall effects on gait, balance, and quality of life that were comparable to that of physiotherapy [43]. It has been found to be a safe method without relevant adverse effects [120], which is easily accessible for private use as well. In addition, the use of VR increases motivation and enjoyment that may assist motor learning by ensuring continuous training.

In view of the current hypothesis of gait asymmetry in the pathophysiology of FOG and falls, there have been rehabilitative approaches focused on the improvement of gait asymmetry using split-belt treadmill training [56], conventional physiotherapy [199], or treadmill training with visual cues on a screen and acoustic feedback [60]. Those training strategies on gait symmetry revealed beneficial effects. However, they were partly unspecific [60] or induced even intermittent, short-term deterioration with the better side-down strategy [56]. The advantage of this specific manipulation of gait symmetry in the current study is the exploitation of immersive VR techniques such as proprioceptive-visual dissociation, which is not available outside the VR. It remains to be elucidated how specific and effective this training strategy is in future rehabilitative training sessions.

The superiority of visual-proprioceptive dissociation in VR compared to other VR conditions was quite astonishing. Most of the current rehabilitative training strategies such as the Lee Silverman Voice Treatment (LVST BIG) therapy[1] focus on the improvement of self-awareness and recalibration of movements [52]. The hypothesis of the current rehabilitative motor learning strategies is that basal ganglia sensorimotor processing deficits result in disturbed amplitude scaling and bradykinesia in PD [186]. Partially, PD patients might overcome this sensorimotor deficit by focusing on movement execution counteracting the under-scaling of the amplitude. However, when the conscious focus on movement execution stops, PD patients fall back into the automated gait pattern with restricted, under-scaled movements. LSVT-BIG therapy aims at the conscious recalibration of the under-scaled movement amplitudes with continuous, graded implementation of the newly gained movement pattern into increasingly complex everyday activities to transport the recalibrated movement template into automated behaviour [52]. In our specific, effective VR paradigm, we deceived the patient's self-awareness of the movement by visual-proprioceptive dissociation instead of reinforcement of self-awareness.

Recently, utilization of error–driven motor learning has been implemented in gait training strategies [197, 196]. In post-stroke patients, gait asymmetry due to hemiparesis was artificially exaggerated on the split-belt treadmill by augmentation of the velocity differences of both sides [197, 196]. The hypothesis was that error augmentation is necessary to drive the nervous system to make corrections. Error augmentation induced an adjusted gait pattern with short-term and long-term effects after repeated training sessions in stroke patients [196]. In our specific, effective VR paradigm, we deceived the patient's self-awareness of the movement by visual backward shifting of the shorter leg resulting in error augmentation. This might be the reason for the effectiveness of that particular VR manipulation method. Due to current positive long-term findings of error augmentation in stroke patients after

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[1]LSVT BIG focuses on intensive exercising of high-amplitude movements.

repeated training sessions [196], we are confident that the VR-based multi-sensory error augmentation might also be effective in the long-term in PD patients.

In the manipulated VR condition, the manipulated foot induces a conflict between the visual and proprioceptive signals about foot position. Such a conflict can be used in different ways to estimate the way in which visual and proprioceptive signals are synchronized. The present study shows that our PD subjects mapped their real foot movements onto corresponding movements of the manipulated virtual foot in the virtual world. Therefore, the consistency can be derived from the responses to the adjustments between proprioceptive and visual information. PD patients initially showed notable asymmetry that was gradually adjusted toward symmetry during the manipulated foot condition.

In another proprioceptive study, the rubber hand illusion was assessed in PD [42]. In this experimental paradigm, synchronous stroking of a rubber hand and the subject's hidden real hand resulted in the illusory experience of feeling the rubber hand and proprioceptive mis-localization of the real hand toward the rubber hand (i.e., proprioceptive drift). PD patients predicted larger proprioceptive drift as compared to healthy controls. The amount of the proprioceptive drift was correlated with disease duration and interpreted as deficient multisensory integration in cortico-basal ganglia-thalamic circuitry in PD [42]. Nevertheless, although PD seems to affect illusory perceptions of body ownership, it was found that dopaminergic treatment appears to increase suggestibility generally rather than having a specific effect on own-body illusions [42]. As in our study, all patients were in the medication on state and results demonstrated sufficient gait symmetry in the manipulated condition. We assume that suggestibility in the sense of mapping the visually perceived foot onto the own body was present in our PD sample. To our knowledge, there is no data on identical tests performed in healthy subjects, but similar setups suggest that the manipulation of visual-proprioceptive integration in virtual environments lead to gait adaption in healthy subjects as well [191].

Study limitations were a small PD patient sample size with predominantly male participants and the lack of control groups such as PD patients without freezing or age-matched healthy controls. Furthermore, information on the body mass index should consistently be recorded since obesity can also alter gait patterns [148]. Since we currently only assessed short-term effects, we need to determine the effects of long-term training in the future. Current investigations are underway in our clinic to examine whether training using the manipulated condition over a few weeks can lead to long-term improvement of step length symmetry following virtual walking. Although we used a very realistic setup with over-ground walking and spatial consistency of the real and virtual world, an even more realistic

and dynamic setup might be helpful in the future such as the moving environment or using wireless or transparent glasses.

7.5   Conclusion

This study presents the results of different VR-based gait manipulation methods in PD patients with FOG with the main purpose to find a method equalizing step length asymmetry in PD patients. We found a significant step length difference between both legs in PD with FOG. The virtual dissociation of visual and proprioceptive signals was most promising in accomplishing this goal and might therefore be a sufficient rehabilitative technique to achieve gait symmetry and, hypothetically, to prevent FOG. Future studies are needed to further investigate the long-term training effects of this specific visual-proprioceptive VR manipulation technique.
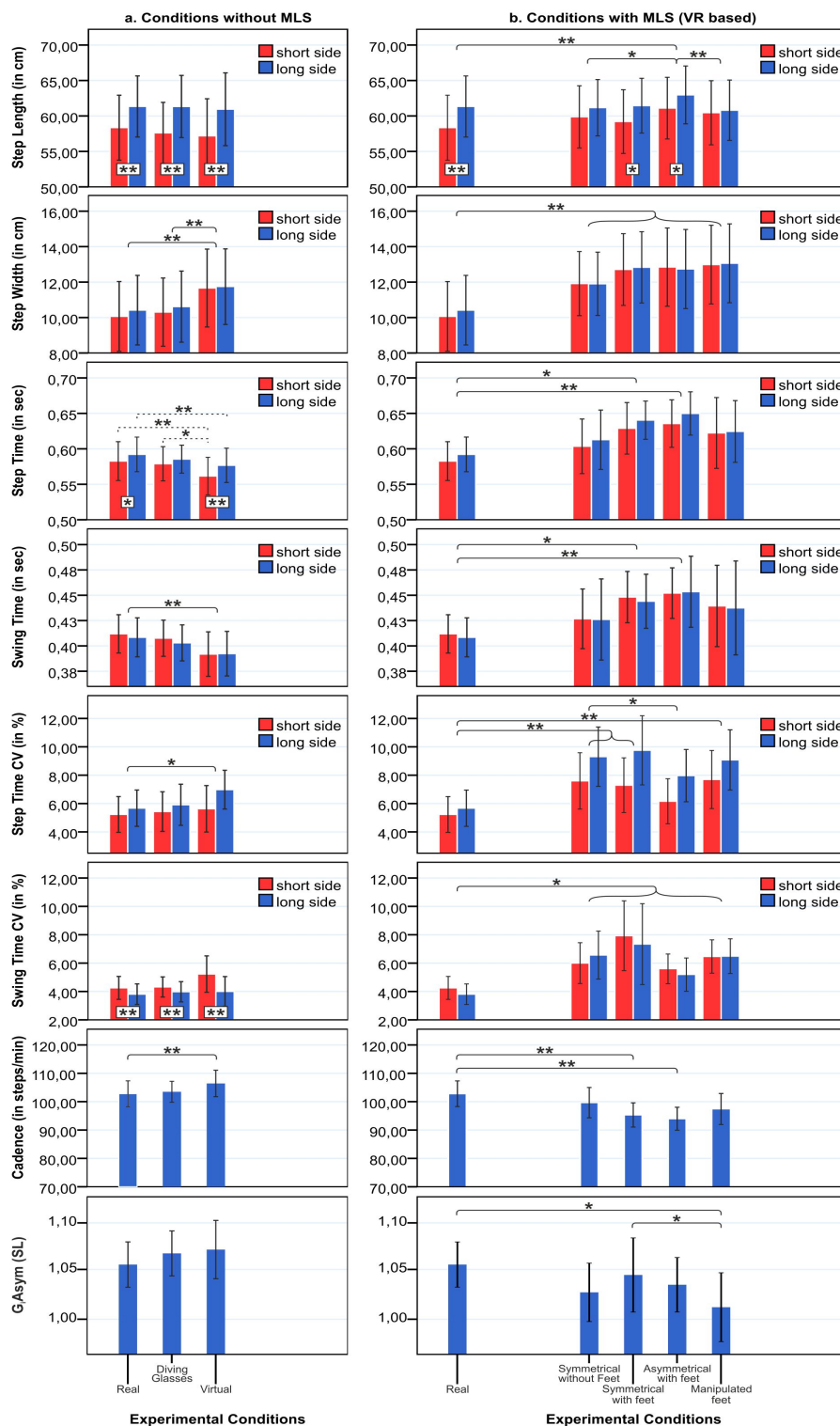
Fig. 7.3 General gait parameters for (a) non-MLS conditions, and (b) MLS conditions are given as mean values ± standard error of mean. Significant differences are marked as follows: (*) = p < 0.05 and (**) = p< 0.01.

DISCUSSION OF CONTRIBUTIONS

In this chapter, we summarize and discuss the contributions relevant to each of the four general research questions presented in Chapter 1. We endeavored to address relevant factors impacting the general validity and applicability of the findings, and suggest directions for potential future research. The main contribution of this dissertation lies in the four papers presented in Chapter 4, 5, 6 and 7. Each paper includes an empirical study induced by the general research questions. The research questions are reprinted here for the sake of clarity.

- Q1: *Which gait parameters within a VE are different from those performed in the real world?*

- Q2:*Which gait parameters between younger and older adults are different while walking in VR?*

- Q3: *Is it possible to reduce gait differences while walking in the real and virtual world?*

- Q4: *Can VR be utilized to manipulate gait characteristics to achieve gait symmetry in Parkinson's disease patients?*

8.1   Overview

8.1.1   Gait Differences in Younger Adults

Q1 served as a primary source of the motivation for Chapter 4 of this dissertation that led to the following findings:

The results underline the importance to investigate the differences between gait parameters while walking in the real world and within a VE. Furthermore, walking in the VE was found to correlate with decreased walking velocity, decreased FAP score, decreased step length,

increased step count, widening of the base of support, positive toe-out, shortened single support and with prolonged double support. And, last but not least, we found that younger adults had a tendency to look down towards the floor when translation gains $g_t < 1$ were applied, but no such effect was visible for gains $g_t > 1$. Considering that the previous literature indicated that users wearing an HMD subjectively estimate a slightly increased virtual walking velocity as more natural than walking using an isometric mapping, our results suggest that this advantage of translation gains $g_t > 1$ is not visible in the gait parameters that we measured in this experiment. Conversely, independently of whether gains $g_t < 1$ or $g_t > 1$ were applied, we found that most parameters of the biomechanics of walking deviated from what is considered normal walking.

8.1.2 Gait Differences in Younger & Older Adults

Q2 served as a primary source of the motivation for Chapter 5 of this dissertation that led to the following findings:

Generally, it is interesting to note that our findings underline the importance of investigating the differences between gait parameters of younger and older adults with the widespread use of VR technology. In particular, our results show that the older adults exhibited comparable gait stability in most gait parameters during walking in the real and virtual world with an isometric mapping, which differs from results found in younger adults. Therefore, older adults were not dependent on the presence of visual information during walking compared to younger adults, which induces comparable gait performance in older adults while walking in the VE compared to the real world. During the application of non-isometric mappings, we found that older adults modulated their walking speeds within the VE asymmetrically with manipulations of visual flow velocity in much the same manner as younger adults. For instance, increasing visual flow with translation gains $g_t > 1$ resulted in a reduced walking velocity and step length. However, such manipulations might be useful when it comes to applications in the area of rehabilitation or physiotherapy.

8.1.3 Reducing Gait Differences

Q3 was addressed throughout a controlled user study presented in Chapter 6, which yielded the following findings:

We found that participants gait while walking within a VE and the real world was different in terms of walking velocity. These differences persisted even after long periods of walking. Moreover, walking velocity was significantly decreased while performing a cognitive task in the VE and the real world. This suggests that cognitive demands have an impact on gait and

that attention processes are involved in walking. The findings appear to disagree with previous literature that promote adaptation [141, 251], which suggests that staying and interacting within a VE for a longer time lead to reduction of differences in gait. This disagreement with prior literature is most likely due the limitations of current VR technology, such as computers, tracking systems, and HMDs. Results of the this study provide a foundation for future longitudinal studies exploring approaches such as left/right figure-8 protocol to simulate real world challenges while walking long distances in a controlled VE.

8.1.4   Manipulating Gait Features

Q4 was addressed to define a virtual manipulation technique with the best possible equalization of the pathological gait asymmetry in Parkinson's disease patients. The study presented in Chapter 7 and revealed three main findings, which are of relevance in the design of future rehabilitative VR training strategies:

1. The leg with the shorter step length was only in 33.3% of patients clinically the most affected one contradicting the clinical impression that the range of motion decreased more on the clinically most affected side.

2. Walking in VR induced an increase of step width, cadence, and gait variability, which indicates an insecure gait pattern during the VR conditions. Similarly, in our previous studies, walking in the VE was found to correlate with widening step width and increasing cadence in healthy younger and older adults.

3. PD patients overcame the spatial asymmetry and exhibited a comparable step length by enlarging the step length of the short side, an adapted step time, and a swing time variability of both sides during the manipulation of visual-proprioceptive signals.

8.2   Factors Influencing Gait in VR

8.2.1   Technical-related Factors

Although VR technology is advancing, there are still major technical limitations of current HMDs such as resolution, FOV and latency, which can cause a participant to walk differently in the VE. Slater et al. [228] found that visual realism (i.e., geometric and illumination) induces greater participant presence within the VE. Interestingly, Jones et al. [114] investigated that higher peripheral visual flow (i.e., relatively high FOV) was causing participants to recalibrate their gait, enabling them to move more accurately in the VE. In contrast, the

effect of potential latency in the HMD can create negative effects on gait characteristics [218]. Although the HTC Vive Pro's end-to-end system latency is stable and low [131], but still can provides delay from the user's physical movement until the response becomes available on the HMD's screen.

8.2.2   Perceptual-related Factors

Discrepancies between perception in real and virtual environments have naturally been suggested as a potential factor contributing to the fact that distances in VEs are often over- or underestimated [101, 102, 135, 198]. In particular, they have found that people underestimate egocentric distances in the VE as compared to real-world performance. Consequently, the studies using shorter distances found an influence, while studies using longer distances did not [198]. Many researchers tried to explain these effects through VR hardware technology limitations or the subjective state of the users, who often walk more slowly and carefully in VEs than they would in the real world, e.g., due to fear of colliding with unseen walls [5, 47]. A common potential explanation for the misinterpretation could be based on incorrect depth and motion cues provided to the human eye, when looking through an HMD, which introduces accommodation-convergence conflicts [88]. Watt et al. [264] stated that inappropriate depth cues in typical HMDs may therefore contribute to distortions in perceived space. Some studies have suggested that this effect is an issue of perception-action recalibration [125], while others suggest that walking through the virtual environment with continuous visual feedback is necessary to cause rescaling of the perceived space [116]. Rieser et al. [204] demonstrated that adapting to a new perception-action coupling had an influence on subsequent dynamic updating of space, as indicated by visually directed walking.

8.2.3   Cognitive-related Factors

It is unknown to what extent the nature of navigation in VEs interferes with other cognitive processes. Waller et al. [262] have assumed that difficulty of navigation in VE (i.e., an exact replica of real environment) caused by a lack of environment fidelity and/or movement fidelity, when compared with the equivalent real world. Additionally, VEs have been found to impose a cognitive load that demands attention, response selection, and the processing of rich visual stimuli involving several perceptual processes [152]. Peterson et al. [187] found that the use of VR increases cognitive load. Due to cognitive-motor interference that reflects interactions between cognition and gait, performance in either or both tasks declines when cognitive load increases, suggesting that gait control relies on higher cognitive systems [2]. In the scope of the present study, despite the cognitive performance in the VE and the real

world was comparable, we found a significant decrements in velocity within the VE by 13.3% compared to the real world (see Figure 6.4a). This indicates that participants resort to employ strategies requiring additional cognitive resources within the VE, which contend for resources that are utilized for successful completion of the locomotion task.

8.2.4   Simulator Sickness Factors

Due to the significant differences found through the SSQ evaluations administrated at the beginning and the end of the experiment, it is possible that participants' gait was influenced by the simulator sickness symptoms (i.e., increased fatigue, sweating, nausea, dizziness, difficulty focusing and eyestrain) that introduced by prolonged walking paths in the VE. Jaeger and Mourant [105], Ibánez et al. [98] have shown no clear correlation between walking and simulator sickness in the VE. However, there are also studies which do not support this claim [14, 221]. Although previous studies [44, 226] suggest that simulator sickness can be prevented when users are allowed a sufficient amount of time to adapt the VE. However, there are also studies which do not support this claim [233, 168], therefore further research on this issue is necessary.

CONCLUSION AND FUTURE WORK

All research described in this dissertation is designed to target the set of research questions initially set out in Chapter 1, and later presented in Chapter 4, 5, 6 and 7. Therefore, in this chapter we conclude with an overall summary in the light of the work carried out, and finally we discuss potential future work that could be carried out to extend this research.

9.1 Summary

In this dissertation, age-related differences were observed in the gait pattern during walking within a VE and the real world. This underlines how important it is to study different users groups with the widespread use of VR technology. For individuals in good health, the gait of older adults differs from the walking pattern of younger adults for some gait parameters. Unlike younger adults, older adults demonstrate similar gait patterns in most parameters within real and virtual world with an isometric mapping. During the application of non-isometric mappings, we found that most gait parameters in both younger and older subjects correlated with often symmetrically-shaped detriments both for up- and down-scaled virtual walking velocity. Hence, we neither advocate increasing nor decreasing the virtual walking velocity of the user, but rather suggest maintaining an isometric one-to-one relation whenever possible to minimize gait detriments and the risk of falling.

In contrast, our previous experiments on younger adults have shown a significant decrease of gait parameters in the VE, in particular, velocity and step length. However, those studies have only considered short periods of walking. Therefore, in this work another controlled study has shown that the gait of younger participants in the real and virtual environments was different in terms of velocity; and these differences persisted even after long periods of walking. Moreover, velocity was further decreased along a continuous walking within the VE compared to the real world while performing a cognitively challenging task.

In addition, gait training protocols in VR have been shown as more motivating and entertaining for people with neurological dysfunction, thereby inducing more time of practice and higher number of repetitions, which are considered to be important factors in the rehabilitation of individuals with Parkinson's disease. In this study, we developed a gait manipulation strategy to improve gait symmetry by equalizing step length. Therefore, we assessed 15 PD patients with FOG while walking in the real world and through a VE comparing natural gait with walking conditions during "VR-based" gait modulation tasks. The results showed that PD patients overcame the spatial asymmetry and exhibited a comparable step length by enlarging the step length of the short side, an adapted step time, and a swing time variability of both sides during the manipulation of visual-proprioceptive signals. Thus, our findings suggest that virtual walking with manipulated feet may have important implications for the restoration of gait symmetry in PD patients, who are affected by step length asymmetry. The improvements that were observed in this study can induce sudden changes in the walking pattern, which may offer promising and simple solution for training gait asymmetry adaptation and quick adjustments in PD patients. Our findings may have important implications for rehabilitation; suggesting that it may be important to also address the feasibility of adding VR intervention during walking to enhance the effects of implicit training, particularly in PD patients with FOG .

9.2 Outlook

Virtual spaces have proven useful over the years to create manipulations that cannot be implemented in real life. Therefore, further exploration of the effects of visual flow manipulations is warranted to determine the specific combination of gait parameters that could eventually be most beneficial in the rehabilitation of individuals with sensorimotor issues. However, the application of (non-)isometric mappings might open up new vistas for rehabilitation or physiotherapy, for instance, in the context of reducing the risk of fall for older adults with cognitive or motor disabilities.

Moreover, to determine the role of virtual gait training in rehabilitation, further research is needed to investigate the effects of long-term training. Current investigations are underway to examine whether training over weeks can lead to long-term learning of improved step length symmetry following virtual walking and to investigate the impact of feedback about symmetry during and after virtual modulations walking. The overall aim is to transfer our previous finding of effective gait asymmetry improvement by visual-proprioceptive drift in VR to a rehabilitative therapeutically approach to equalize step length, reduce freezing

episodes and falls, which has fundamental impact on quality of life and economical aspects in PD patients.

The transfer of motor learning principles attained during the training sessions to everyday life is key for the development of new rehabilitation strategies. Particularly in PD, the transfer of exercised movements in the training session into the daily routine is often unsuccessful and constitutes a serious limitation of rehabilitative approaches [158]. It remains to be assessed whether the use of proprioceptive-visual dissociation in VR is a rehabilitative training strategy that overcomes these potential drawbacks.

9.3 Further Research and Possible Extensions

The research described in this dissertation has potential to stimulate further work. For instance, one might study the biomechanics of virtual walking with further visual information about self-motion by incorporating a tracked virtual model of the participant's body. Considering that participants in our experiments tended to look down towards the floor when the virtual walking velocity was reduced, providing such a visual self-representation might help users to acquire a more stable gait profile in VEs. Furthermore, we plan to develop different techniques, multimodal illusions and feedback, which support users in adapting their biomechanics while walking in the VE to the corresponding biomechanics in the real world.

In principle, we would like to see further studies to explore strategies for measuring an accurate perception of distances in VEs, which do not exactly match a user's actual occupied environment. Since, many researchers have found that the estimation of the travel distance of a simulated movement shows characteristic errors, sometimes overestimating and sometimes underestimating the true travel distance [101, 102, 135, 198]. Another open issue is the effect of adaptation to motion manipulations in VEs. Such adaptive properties of the perceptual system also open up possibilities for manipulation, which have not been investigated yet in this context. Adaptation requires that the user stays and acts within a VE for a longer time. This transforms the user's perception of the VE such that he learns to interact with the VE in a particular way. The potential of these learning effects remains to be explored, which over a longer time period of time that the gait parameters may adapt and become more stable.

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GAITRITE® SYSTEM

General Description

The GAITRite is an electronic walkway utilized to measure the temporal (timing), spatial (two dimension geometric position) and relative pressure level values (switching levels) parameters of its pressure activated sensors for the events occurring during bipedal and quadruped locomotion [166]. It measures cadence, step length, velocity and other gait parameters. Assess step-to-step symmetry and variability to determine dynamic balance, predict fall risk, and quantify motor and cognitive dual tasking. Inferred parameters are easily obtained by applying common physics and math formulas to the directly measured relationships between spatial and temporal events. This data offers clinicians an efficient, easy to use, yet powerful tool to document functional outcomes. Through a graphical user interface (GUI), clinicians can see the pressure points affected in a patient's foot. The software then turns data collected into analysis on foot placement patterns and overall gait patterns.

Table A.1 GAITRite Walkway Technical Specifications

Parameters	Walkway Specification
Overall Dimensions	90cm × 7m × 3.2mm
Active area	60cm × 6.1m
Sample rate	60, 80, 100, 120, 180, 240Hz
Number of pressure sensors	2.304 in (48 × 48) grid pattern
Spatial accuracy	±1.27cm
Temporal accuracy	±1 sample
Switching level accuracy	±.5 switching level

QUESTIONNAIRES

Hoehn and Yahr Scale

The Hoehn and Yahr Scale is used to measure how Parkinson's symptoms progress and the level of disability. Originally published in 1967 in the journal Neurology by Melvin Yahr and Margaret Hoehn [87], it included stages 1 to 5. Since then, stage 0 has been added and stages 1.5 and 2.5 have been proposed and are widely used.

Unified Parkinson's Disease Rating Scale (UPDRS III)

The UPDRS combines elements of several scales to produce a comprehensive and flexible tool to monitor the course of Parkinson's and the degree of disability. The scale was introduced in 1987 [54] and has since been updated by specialists from the Movement Disorder Society (MDS) to include new assessments of non-motor symptoms.

Freezing Of Gait Questionnaire (FOG-Q)

The German version of (FOG-Q) reliably detects freezing of gait (FOG) in patients with Parkinson's disease (PD) without dementia Vogler et al. [259]. It consists of six questions. Questions 1, 2, 4, 5, and 6 refer to the patient's experiences, related to FOG, of the previous week. For question 3 the patient is asked about his unique experience of FOG in different situations, which is not limited in time. Each question has a 5-point scale, where 0 means an absence of symptoms and 4 represents the worst stage. Consequently, the total score on the FOG-Q ranges from 0 to 24 points. The higher the score is, the more the FOG is pronounced. The time needed to administer the questionnaire is approximately 5–10 minutes.

Parkinson's Disease Questionnaire (PDQ-39)

The PDQ-39 assesses how often people affected by Parkinson's experience difficulties across 8 dimensions of daily living including relationships, social situations and communication. It also assesses the impact of Parkinson's on specific dimensions of functioning and well-being.

Berg Balance Scale

The Berg balance scale is used to objectively determine a patient's ability (or inability) to safely balance during a series of predetermined tasks. It is a 14 item list with each item consisting of a five-point ordinal scale ranging from 0 to 4, with 0 indicating the lowest level of function and 4 the highest level of function and takes approximately 20 minutes to complete. It does not include the assessment of gait.

Montreal Cognitive Assessment scale (MoCa)

The MoCA was designed as a rapid screening instrument for mild cognitive dysfunction. It assesses different cognitive domains: attention and concentration, executive functions, memory, language, visuoconstructional skills, conceptual thinking, calculations, and orientation. Time to administer the MoCA is approximately 10 minutes. The total possible score is 30 points; a score of 26 or above is considered normal.

Presence Questionnaires (PQ-SUS)

Slater et al. [230] identified that both external and internal factors contribute to the sense of presence in VEs. The external factors were quality and resolution of displays, consistency of environment, interactivity, realistic self-representation, and simple connection between actors and effects. Internal factors were based on primary presentation system (visual, auditory or kinesthetic) and perceptual position (egogenic or exogenic). The SUS questionnaire consisted of six items rated on a seven point rating scale.

Simulator Sickness Questionnaire (SSQ)

The SSQ [118] is currently the standard for measuring simulator sickness in VR. The questionnaire asks participants to score 16 symptoms on a four point scale (0-3). A factor analysis

revealed that these symptoms can be placed into three general categories: Oculomotor, Disorientation, and Nausea. Weights are assigned to each of the categories and summed together to obtain a single score. Although the score is not intended to predict illness, it does provide a description of overall simulator sickness scores for a given simulation or simulator environment.