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Effect of multifocal lens glasses on the stepping patterns of novice wearers

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ABSTRACT

Multifocal lens glasses (MfLs) negatively affect vision, increase falling risk and contribute to gait changes during stepping. Previous studies on the effects of MfLs on gait have focused on experienced wearers. Thus, the initial response of first-time wearers, who may face significant challenges in adapting to these glasses, is not well understood. This study aimed to quantify the effects of MfLs on novice wearers during stepping up and down. Additionally, young adults were compared against a middle-aged adults to determine the validity of convenience sampling in testing novice response to MfLs. Fifteen young adults (18–34 y.o.) and seven middle-aged adults (46–56 y.o.) were recruited to perform stepping trials while wearing progressive MfLs and blank single lens glasses. Participants stepped up and down from a 75 mm and 150 mm step in randomized order. Step placement, minimum toe clearance, lower body kinematics and stepping time were measured during step up. Step placement, minimum heel clearance, vertical forces and stepping time were measured during step down. MfLs significantly increased toe clearance in the lead and trailing legs, hip flexion, knee flexion and stepping time during step up and increased vertical forces and stepping time during step down. Step placement and hip angle explained 17% of the toe clearance variability. Changes during step up suggest a more conservative adaptation while increased forces during step down suggest a reduced level of control. No age group effects were observed, which supports the use of convenience sampling for evaluating the novice response to MfLs.

1. Introduction

Falls are a major health problem with high societal costs. Falling is prevalent in occupational settings costing 13 billion dollars annually [1]. Multifocal lens glasses (MfLs) increase falling risk by 70–130% in elderly adults [2,3]. These studies only included elderly adults despite MfL's prevalence among middle-aged adults who also have an increased fall risk. Presbyopia, which is typically treated with MfLs [4], affects >85% of middle aged adults [5] and adults aged 45–54 fall at 88% higher rate than adults aged 25–34 [6]. While the MfL diopter assists in near distance viewing, it causes significant vision impairments for distant vision in the lower visual field [7]. Unlined progressive lenses distort ground-level objects due to the shape of the lens as it transitions between the distance focal length to the shorter focal length diopter. Furthermore, MfLs reduce contrast sensitivity and distort depth perception in older [2,8] and middle-aged [7] populations, which are needed for precisely locating steps. This visual impairment may explain why MfL wearers experience more trips, falls outside of the home and falls around steps than non-MfL wearers [2]. Understanding the effects of MfL-induced distortions and impairments on gait patterns of middle-aged adults is critical to developing strategies that reduce falls in this group.

Visual feedback is critical to maintaining a safe and stable gait pattern. Visual information allows for detection of obstacles or steps and is used to alter foot placement and stepping strategies [9]. When ascending or descending stairs, visual gaze is typically fixated on step surfaces and edges located approximately 3–4 steps away [10]. MfLs affect foot movement patterns, step placement, stepping time and kinematics of stepping. Specifically, participants alter gait patterns when stepping up by increasing the distance [11,12] and variability [11] between their toe and the step edge, increasing toe clearance of the stepping foot [11,13] and reducing magnitude [11,13] while increasing variability [11] of the distance between their heel and the step edge after step up. Strong MfL diopters also cause an increase in tripping on steps [13]. Additionally, MfLs are thought to negatively affect landing control during step down by altering ground reaction forces, center of mass dynamics, lower body kinematics and step time [14,15].

Studies examining the effects of MfLs on gait have primarily focused on elderly and very experienced MfL wearers [11–14,16,17].

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A paucity of information is available about the gait adaptations during stepping when MfLs are first used. Increased fall incidences have been linked with new prescription [18], suggesting that fall risk may increase during the initial adaptation period.

The purpose of this study is to gain better insight of the effects of MfL glasses on stepping gait patterns for novice MfL-wearers. Specifically, this study:

1. Examined the initial changes to gait caused by MfLs in novice wearers by considering foot kinematics, spatiotemporal variables, lower-body kinematic variables and kinetic variables during step up and step down tasks.
2. Determined if young adult novice wearers can be used as a valid representation of middle-aged novice wearers for testing the initial effects of MfLs on stepping.

2. Methods

2.1. Participants and experimental protocol

Fifteen young adults between the ages of 18 and 28 years (5 male and 10 female; aged 24.3 $\pm$ 3.65 years, 1.68 $\pm$ 0.083 m tall, 66.4 $\pm$ 16.5 kg) and seven middle-aged adults between the ages of 46 and 56 years (3 male and 4 female; aged 51.9 $\pm$ 3.98 years, 1.63 $\pm$ 0.13 m tall, 71.4 $\pm$ 13.4 kg) were recruited to participate. Participants were only included if they had never used MfLs previously (i.e., were novice MfL wearers) and did not have any clinically significant musculoskeletal or neurological condition that would affect their walking. Because participants were required to wear different MfL and single lens (SL) glasses during the study, only participants who wore contact lens or no corrective lenses at all were included (i.e., subjects requiring glasses for corrected distance viewing were excluded). The University of Wisconsin-Milwaukee Institutional Review Board approved the study and participants consented prior to being enrolled.

Participants were fit with 29 markers and a safety harness. Relevant marker locations for the anatomical coordinate systems included bilateral anterior superior iliac spine, medial and lateral femoral epicondyles, medial and lateral malleoli, heel and the front of the toe as well as a sacral marker. Three additional markers were placed on each of the thigh and shank segments to create a measurement coordinate system that was tracked during the dynamic trials [19]. Data were collected using a motion capture system (10 Motion Analysis Raptor cameras, Santa Rosa, CA) and two force platforms (AMTI, Watertown, MA). Participants completed three trials of four different stepping conditions (75 mm step up, 150 mm step up, 75 mm step down and 150 mm step down) (Fig. 1) as well as six walking trials without any step for each eyewear type (4 stepping conditions $\times$ 3 trials per stepping condition + 6 walking trials = 18 trials per type of glasses). Participants wore SLs with no magnification and MfLs (18 trials per type of glasses $\times$ 2 eyewear types = 36 trials per participant). Progressive lens (i.e., no lined bifocals) MfLs transitioned from no power in the upper region to a diopter add region of +2.75 in the lower quarter and central third ($\sim$15 mm wide). All eyewear was from Zenni®. Participants wore glasses high up on the nose and were not allowed time to adapt to new glasses before the start of data collection. The step contained a level platform (2 m $\times$ 2 m) and an adjacent ramp (2 m $\times$ 2 m). The riser of the step was unpainted wood while the landing of the step and the surrounding floor were blue vinyl tile illuminated at 460–480 lux (Fig. 1). The step edge was 2–3 cm after the force platform during step up and 2–3 cm before the force platform during step down. Participants took at least two strides before and after stepping up or down to avoid gait initiation or termination from affecting their step. Participants' starting position was adjusted until participants cleanly hit the first force platform during level walking trials. After this point, the starting location was not adjusted during session. Participants were instructed to walk at a comfortable pace to the opposite side of the room. The five unique tasks were intended to force participants to use visual feedback for navigating the steps as opposed to proprioception similar to other studies examining the effects of MfLs on stepping [11]. In addition to randomizing the order of trials within a single glasses condition, the order in which eyewear were presented was also randomized. Walking trials were collected but not analyzed in this study.

2.2. Data processing and statistical analyses

Variables that were considered during step up included step placement before and after the step, toe clearance of the first stepping foot ("lead foot") and the second stepping foot ("trailing foot"), stepping time and lower-body kinematics during stepping. Toe clearance was calculated as the minimum perpendicular distance between the front of the toe marker and a line connecting two markers placed on the step edge during step-up. The variables considered during step down included step placement before ("toe-to-step distance") and after the step ("heel-to-step distance"), heel clearance during step down, stepping time and peak normal

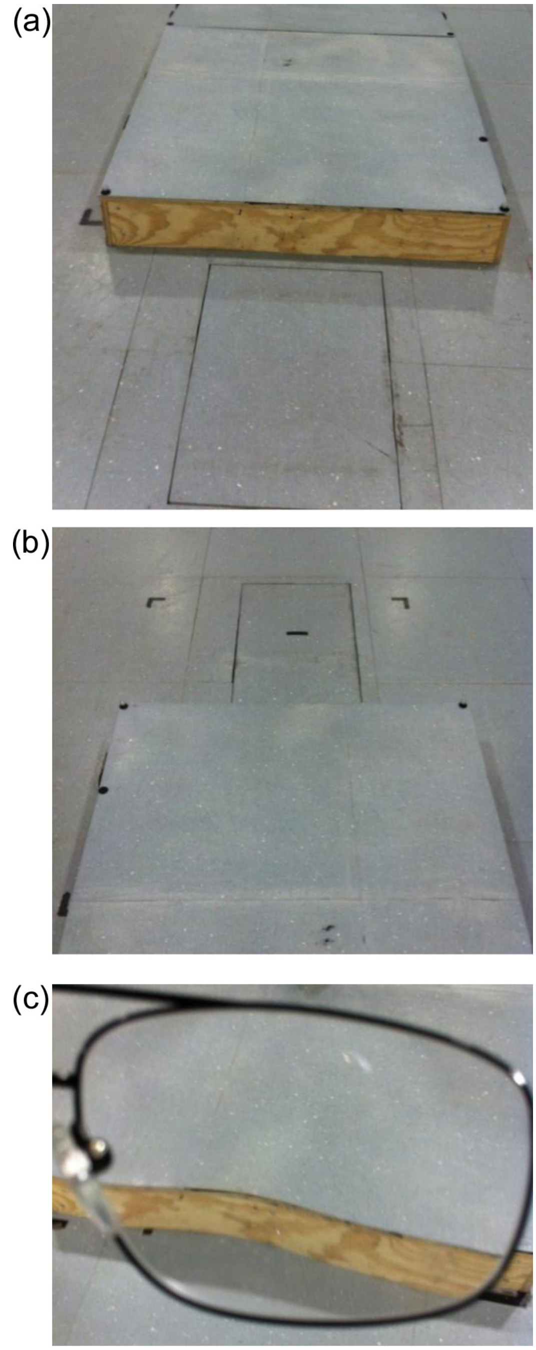

Fig. 1. (A) Participants' point of view approaching step up condition, (B) participants point of view approaching step down and (C) distortion of step edge caused by progressive lens.

ground reaction forces. Step placement variables were analyzed to determine if participants altered foot placement in response to MfLs, while ankle, knee and hip angles were analyzed to determine if participants altered their stepping patterns. Stepping time was evaluated to determine if participants quickened or slowed their stepping process due to the MfLs. Toe clearance during step up and heel clearance during step down were evaluated as foot clearance has been suggested to be related with trip risk [20]. Peak normal force was calculated as a measure of stepping control.

The timing of foot contact was identified using force platform data, when available, or heel marker data similar to [21]. Lower-body kinematics were calculated by developing a seven segment model including the feet, lower legs, thighs and pelvis. Joint angles were calculated in 3D according to the rotation order suggested in [22–24], where zero reference angles represent the state of the joint in standard anatomical position. Peak angles in ankle dorsiflexion, knee flexion and hip flexion were used to characterize the stepping gait adaptations. Peak normal force was normalized to body weight and only clean hits with one foot were included.