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The effects of a positional feedback device on rollator walker use: a validation study

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ABSTRACT

Background: According to clinical guidelines, rolling walker users should walk with their feet between the posterior wheels of the walker; however approximately 50% of users do not. Objective: To describe the development and effects of a custom device designed to attach to a walker and provide visual feedback to encourage improved user position. Methods: Fourteen older adults participated in this study to validate the effects of this device when a 10% decrease in the users' habitual distance away from the walker was encouraged via feedback. Users' relative distances were recorded using a non-contact distance sensor within the device, while kinematics were measured using commercial wearable wireless inertial sensors. Results: Individuals were able to ambulate on average 20% closer or more to their walker when prescribed the visual feedback. This was primarily achieved through a reduction in shoulder flexion. Trunk and cervical postures were less generalizable as only small and variable changes were observed. Conclusions: These findings suggest that the device has promise, as individuals attended to the device and walked in a position closer to that recommended by clinical guidelines. The device did not appear to improve posture. Future work is needed to determine long-term effects.

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KEYWORDS
kinematics of walker users; mobility aids; older adults; rolling walkers; stability and posture; visual feedback

Introduction

Mobility aid use has increased significantly over the last few decades, with an estimated 24% of older adults in the United States now using some form of a mobility aid (Gell, Wallace, Lacroix, Mroz, & Patel, 2015). Over 4 million of these individuals use walkers (Gell et al., 2015). In addition to improving mobility and helping to maintain balance, the use of a walker has other documented clinical benefits including compensation for lower-extremity weakness; increased confidence and feelings of safety; physiological benefits; and increased levels of activity and independence (Bateni & Maki, 2005). However, adverse effects and potential disadvantages associated with walker use are also commonly highlighted in the literature including: increased attentional and neuromotor demands; destabilizing biomechanical effects during device advancement; and interference with reactive limb movements during balance recovery which could be influenced by walker fit and positioning (Bateni & Maki, 2005; Stevens, Thomas, Teh, & Greenspan, 2009).

Rolling walkers deserve particular attention, but remain understudied. These walkers are very common, with an estimated 25% of residents of retirement communities and over 40% of assisted living residents using them (Liu, 2009). Unfortunately, the majority of rolling walkers are often purchased without the input of a medical professional (Liu, 2009). Further, rollator walker users are at higher risk of more severe injury and hospitalization if they fall while using their walkers when compared to fallers of a similar age not using a device (Van Riel et al., 2014). However, typical walker users may also be frailer and have more comorbidities compared to non-device users which may increase the risk of injury during a fall and was not controlled for in Van Riel's study.

Published clinical recommendations provide guidance on using rolling walkers (O'Sullivan & Schmitz, 2000; Pierson, 2002). When ambulating, rolling walker users should have their hands positioned so that the walker height is at the wrist joint line with both elbows bent to 20° to 30°. Their feet should be positioned between the posterior wheels of the rolling walker. These guidelines are intended to enable individuals to use the walker more comfortably and more safely, putting them in a favorable position where wrist and back pain are minimized and the user is in greater control of the walker for maneuvering and balance recovery if needed. Despite the common acceptance of these clinical guidelines, they should be approached cautiously since there is little empirical data to support them. Given the complex interaction between the individual, the device and the environment, optimal fitting and positioning that maximizes safety and comfort are likely to be different for each individual. For example, an individual with a thoracic spine compression fracture may be encouraged to walk closer to the walker with a more upright posture to reduce anterior vertebral body compressive forces while a person with spinal stenosis may be instructed to walk further from the walker with

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a more flexed posture to reduce narrowing of the vertebral foramen. Additional factors such as limb strength, postural control, anthropometrics, flooring, and environmental conditions may also influence walker fitting and positioning recommendations.

Research suggests that approximately 80% of rolling walker users have never received demonstration or education on how to use their walker, likely explaining the high prevalence of walker use that deviates from the general clinical recommendations (Liu, 2009). Liu (2009) studied rolling walker use and identified that the most common form of misuse (55% of users) was having the walker set to the incorrect height. Two additional notable issues emerged that were present both in those who had the walker appropriately adjusted to their height and those who did not: 1. A forward leaning posture during standing and ambulation and 2. Pushing the walker too far forward during ambulation such that the feet were not in the clinically recommended position.

Correcting a forward leaning posture during standing is harder to address because over half of those individuals who exhibited a forward leaning posture were unable to correct their posture due to kyphosis or spinal stenosis (Liu, 2009). This suggests that targeting improvement in the position of the feet may be more appropriate. Liu (2009) observed that while most individuals were able to position their feet appropriately during stance, over half of the users were not able to maintain the position during ambulation. This included 27% of rolling walker users who initiated walker movement by pushing the walker too far forward and then stayed too far behind and another 23% of rolling walker users who moved the rolling walker faster than their feet at certain times during ambulation. Often this caused or increased the forward leaning position. Therefore, figuring out how to promote better adherence to the clinical guidelines, particularly in the positioning of the feet during ambulation, could be beneficial.

Several previous studies have used walker mounted laser or camera (RGB-D, IR) systems to measure foot and body position to estimate center of gravity (CoG) and measure select gait and balance characteristics of walker users (Ballesteros, Peula, Martinez, & Urdiales, 2018; Hirata, Komatsuda, & Kosuge, 2008; Joly, Dune, Gorce, & Rives, 2013). Efforts to instrument walkers have also included the development of a walker that initiated active braking in response to changes in CoG relative to the walker that may indicate a loss of balance (Hirata et al., 2008). While users did remain in a better position relative to the walker, the delay in estimating fall risk yielded unanticipated braking and was validated with only three non-walker users. Despite these advancements in walker technology, no study has yet attempted to provide real-time feedback to promote immediate changes in walker use that is achieved through action of the user. Additionally, the systems that have been proposed may be complex, cost prohibitive and/or require custom software and instrumented walkers that are not commercially available.

To address this need, we designed and investigated an intervention in the form of a simple and potentially affordable device that can be mounted to most commercially available walkers and provide real-time visual feedback on the user's relative position with respect to the walker. The primary objective of this research was to determine the ability of individuals to attend to the device and decrease their relative distance from their walker during both simple walking and obstacle negotiation tasks. The secondary objective of this study was to determine the common methods of biomechanical adaptations that were used to adjust the walking position and whether upright posture during ambulation was changed.

Methods

Device development

The design goals of the system were to provide a position sensing device with visual feedback with the following attributes: (1) attach to and be easily removed from the crossbar of a variety of commercially available rollator walkers; (2) have capabilities for non-contact measurement of the user's location up to the maximum distance the user may be away from the walker front crossbar (~120 cm); (3) actuate a form of feedback when the user was outside of the maximum specified distance; (4) record instances when the user was outside of the maximum specified distance; and (5) be safe for use by allowing the user to maintain line of sight while walking.

The custom device (Figure 1) consisted of a mechatronic system with the following main components: (1) an Arduino Uno R3 microprocessor for storing the program to provide the visual feedback cues when the user was outside the specified distance range; (2) a microSD shield and 16GB microSD card for logging the user's relative distance measurements; (3) an ultrasonic proximity sensor for non-contact distance measurements; (4) a Real Time Clock for recording a time stamp of the distance measurements (recognizing this may not be necessary if the Arduino Uno R3 internal clock were used instead); (5) a variable light emitting diode (RGB LED) to produce a range of colors for visual feedback; and (6) a 3.7v 1200 mAh Li-Ion battery for powering the device.

Various forms of attachments were trialed to attach the device to the crossbar. The final version of the clamp was a modified mount sold for action cameras, which was found to adjust well to fit crossbars of different diameters and allowed for easy multi-axis rotation of the device for positioning purposes, with the ability to lock into place once correctly positioned. An ultrasonic sensor (Parallax PING) was selected as the non-contact distance measurement over infrared sensors due to the fact that it was more robust in several lighting conditions and provided both the accuracy (± 0.3 cm) and range (2 cm – 3 m) needed, with a field of view of ±30 degrees at 60 cm (Karmali, Tomlinson, & Goyal, n.d.; Parallax Technical Specification Sheet for Ping Sensor, n.d.). Figure 1 shows a representation of the sensor's field of view when the device is attached to the walker for use.

Several different forms of feedback were considered based on effectiveness, feasibility, and intrusiveness. The main forms of feedback considered were visual, auditory, and haptic (Giggins, Persson, & Caulfield, 2013). Visual feedback in the form of a RGB LED was ultimately selected due to its reported effectiveness, feasibility of application, and lack of obtrusive nature as compared to the alternative options. Auditory feedback was not selected due to the high rate of hearing impairment in older adults (Walling & Dickson, 2012) and haptic

Figure 1. Representation of the sensor's field of view when the device is attached to the walker for use as shown with an anthropometric model generated by licensed commercial software (Zygote Media Group, American Fork, Utah) (left) and Close-up of the unattached device as would be facing the user (right).

feedback was not chosen based on concerns that the user would not be able to feel the vibration during ambulation on rough ground or would find this sensation uncomfortable or distracting. Given the location of the device on the crossbar, the visual display resided in the general field of vision, allowing the feedback to be noticeable while not blocking primary lines of sight. Through a simple closed loop feedback at 4 Hz sampling, the device illuminated the RGB LED a red color to indicate to the user when the ultrasonic sensor detected the user outside the prescribed range, or a green color to indicate to the user when the user was within the range. Providing the user visual feedback on their distance every 0.25 s was found to be reasonable based on user feedback during the prototyping stage. Distance data was stored along with time stamps to the microSD card. All was packaged within an electrical enclosure box for safety. The system was then tested with five rollator walker users to confirm the reliability of the visual feedback system prior to the validation study.

Device validation study

Recruitment

A total of 14 older adults participated in this device validation study. All participants were independent or assisted living residents of an area retirement community and used a rollator walker regularly for ambulation. Exclusion criteria included a score of 26 or less on the Mini-Mental Status Exam (Bassuk & Murphy, 2003), a gait speed less than 0.40 m/s during a 10 Meter Walk (10-MW), significant spinal deformity causing a fixed flexed posture, inability to walk for 30 feet with a walker without rest or other assistance, a level of pain greater than 2/10 during ambulation, and severe macular degeneration that could prevent them from observing the device LED. The local Institutional Review Board approved the data collection protocol, and all participants provided written informed consent prior to participation.

The 14 participants all required the use of a rollator walker but no additional physical assistance or support for ambulation. The average gait speed was 0.64 m/s among participants, which is within the gait speed norms for older adults in clinical settings as reported by Peel et al. (Peel, Kuys, & Klein, 2013). Participant characteristics are summarized in Table 1.

Data collection

During the testing session, participants were first fitted with an Xsens Awinda Biomech full suite of 17 inertial sensors (XSENS, Culver City, California) in order to wirelessly record full body kinematics during trials at their residence. This approach allowed for users to perform the study outside of the laboratory environment using a system that has been validated for joint angle measurements when compared to the optimal motion capture system to within a root-mean-square error (RMSE) below ~8° for sagittal plane flexion/extension (Marreiros & Karatsidis, Angelos, 2017).

All 14 participants used the same 4-wheel rolling walker with the position monitor attached to the crossbar to ensure

Table 1. Participant characteristics (n = 14).

Characteristic	Values
Age (y)	87.54 ± 4.90, (77–94)
Sex: female/male	9/5
Height (cm)	163.67 ± 12.62, (149.9–198.0)
Weight (kg)	82.69 ± 14.46, (61.7–112.1)
BMI	30.97 ± 5.08, (24.2–42.1)
MMSE (max = 30)	29.71 ± 0.82, (27–30)
Gait Speed (m/s)	0.64 ± 0.10, (0.45–0.82)

Note. Values are mean ± SD, (range: min – max) unless otherwise noted
Abbreviations: BMI, body mass index; MMSE, Mini-Mental State Exam

consistency in walker design, wheel size, and overall walker condition. The walker was individually adjusted for each study participant by a licensed physical therapist on the research team who followed the accepted walker fitting guidelines defined by O'Sullivan and Schmitz, aligning the handles of the walker with each respective wrist joint center with hands at the side of the body (O'Sullivan & Schmitz, 2000). Once the walker was properly fitted, the participant was instructed to stand with their hands on the handles and their feet positioned within the walker, placing their 5th metatarsals heads in line with the center axle of the rear wheels. The distance from the front of the ultrasonic sensor to a location at their waist, approximated between the anterior superior iliac spines, was measured and recorded. This value was known as their "forward limit", and was considered the optimal distance from the walker as well as the closest distance to the walker that was desired for the participants to be positioned based on guidelines for walker fitting and positioning (Irimia, 2010).

Participants completed three baseline 10-MW trials at a self-selected comfortable pace without receiving any verbal corrections on their posture or visual feedback from the position monitor. During this time the Xsens system monitored their body kinematics, and the position monitor recorded their distance from the walker. From the three baseline walking trials, their average habitual walking distance from the walker was calculated. The target distance was then set at 90% of their habitual distance for the intervention trials. While it would be optimal for a target distance to be set individually based on a variety of clinical factors by a qualified health-care professional, we chose the target of moving all participants 10% closer to the walker for consistency in this first study. This 10% decrease in habitual distance was chosen as a distance that was thought to be an achievable by most participants and was also small enough so as not to bring any user closer to their walker than their previously established forward limit distance, as being too close to the walker can also be problematic (Kirby, Little, & MacLeod, 1999).

Participants then completed one trial of a standardized obstacle course, 32 m in total, without position feedback, walking at a self-selected comfortable pace. The obstacle course was designed to simulate an environment more typical of daily ambulation, requiring greater awareness and attention while navigating over thresholds, through cones, around furniture, and performing sit-to-stand and stand-to-sit transitions. Figure 2 shows the set-up of the obstacle course used in this study.

After a short rest, the position monitor feedback light was then turned on, and the purpose of the light was explained consistently to each participant by the researcher. The researcher demonstrated and described that correct walker use is usually achieved by a combination of moving closer to the walker and standing up straighter. Participants were told that any time the light was red, then they were outside of the pre-determined target distance and should make appropriate corrections to turn the light green. They then completed three trials of the 10-MW and single trial of the obstacle course with feedback.

Data processing and statistical analysis

The position data retrieved from the device's microSD card was processed in MATLAB R2015A (MATLAB, Natick, Massachusetts). Preprocessing of the data from the position monitor included only downsampling to 4 Hz. Joint angles for elbow flexion, shoulder flexion, hip flexion, and flexion of the T1-C7 and atlanto-occipital articulations (C1-Head) were determined by the Xsens MVN Studio software and analyzed. Elbow and shoulder flexion were used to assess changes in the relationship between the upper extremities and the walker, hip flexion was used to assess postural changes at the hip, and T1-C7 and C1-Head joints were used to assess changes in neck and head position, respectively. Together, these joint angles were used to help identify the movement strategies that were used to achieve a closer position to the walker.

Forward trunk inclination was calculated during each walking trial as an indicator of upright posture. Forward trunk inclination was calculated by creating a vector from the pelvis sensor to the neck sensor in the sagittal plane and determining the angle of that vector with respect to vertical. Trunk inclination has been shown to be valuable for assessing posture and is associated with functional gait performance in older adults (Hirose, Ishida, Nagano, Takahashi, & Yamamoto, 2004).

All statistical analyses were performed using IBM SPSS 23 statistical analysis software (IBM Corp., Armonk, NJ). Paired samples t-tests were used to assess differences in the mean values between the no feedback and feedback conditions with the significance level set at $\alpha < 0.05$ for all comparisons. Effect sizes (Cohen's d) were also calculated for each variable to determine the magnitude of change. Additionally, post-hoc statistical power for each comparison was determined using G*Power software (University of Düsseldorf, Düsseldorf, Germany).

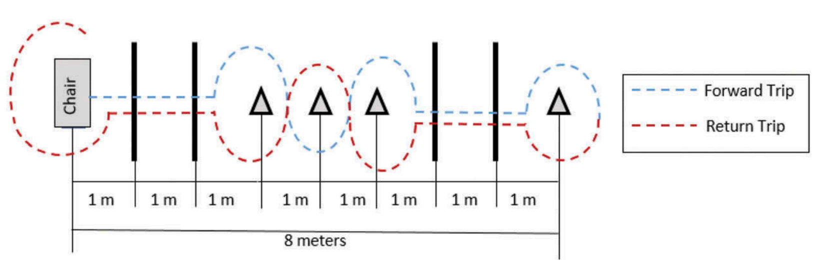

Figure 2. Obstacle Course Layout with Dimensions and Path for a Single Lap. Participants first stood from a chair, then crossed over two rubber thresholds placed on the floor (denoted as dark vertical lines), wove through a series of three small cones (denoted as gray triangles), crossed two additional thresholds, rounded a final cone, and returned through the obstacle course, went behind the chair, and returned to a seated position. Individuals then stood again and completed another complete lap of the course.

Results

Walker position

Table 2 shows the mean habitual distance that each participant stood from their walker during the 10-MW without feedback as well as the target distance for each participant that was required to achieve a position 10% closer to the walker. Additionally, the percentage of time that was spent within the target distance during both the 10-MW and obstacle course when feedback was provided is reported. The mean distance from the walker during the 10-MW and obstacle course with and without feedback is shown in Table 3. With the feedback, participants walked significantly closer to the walker during the 10-MW ($P < .001$) and obstacle course ($P < .001$).

Kinematics and posture

Kinematic analysis was performed using position and joint angle data from the Xsens system. One subject was excluded from this analysis due to the impacts of a past stroke. Although stroke was not an exclusion criterion and this participant met all criteria for the study, impacted motor and postural control led to inconsistent kinematic values found upon data review. Joint angle and inclination values are shown in Table 4. During the 10-MW with feedback provided there was a significant decrease in shoulder flexion ($P = .001$) and significant increases in elbow flexion ($P = .013$), T1-C7 flexion ($P = .017$) and C1-Head Flexion ($P = .032$). During the obstacle course with feedback provided there was a significant decrease in shoulder flexion ($P = .01$), hip flexion ($P = .023$) and trunk inclination ($P = .016$) and significant increases in elbow flexion ($P = .013$).

Gait speed and obstacle course completion time

There were no significant differences ($P = .31$) in gait speed during the 10-MW trials between the no feedback (0.56 ± 0.12 m/s) and feedback (0.59 ± 0.13 m/s) conditions. There was also no significant difference ($P = .079$) in obstacle course completion time between the no feedback (84.4 ± 22.7 secs) and feedback (94.3 ± 33.4 secs) conditions.

Discussion

This study was designed to assess the effects of a newly developed position feedback device that provides visual feedback about the distance between the walker and the user during ambulation and that can be fit to a variety of common walkers. It is the first study to evaluate a device designed for this purpose that we are aware of.

Walker position

The primary aim of this research was to determine the ability of the device to encourage participants to ambulate 10% closer to their walker under different conditions. To accomplish this, participants were provided with visual feedback (green light) when they were within the target distance. During a 10-MW, participants were able to stay within their target distance approximately 90% of the time with feedback. During an obstacle course that was designed to simulate the navigation of common household objects, participants were able to stay within the target distance 67% of the time. These findings demonstrate that the device did achieve its intended purpose but was less effective when used in complex environments that require greater attentional demands (e.g. visual scanning and obstacle avoidance) and physical demands (e.g. increased propulsion over thresholds, turning,

Table 2. Posture monitor distance and threshold data.

Subject	Mean Habitual Distance* (cm)	Target Distance** (cm)	Difference (cm)	10-MW % of time ≤ target distance with feedback	Obstacle Course % of time ≤ target distance with feedback
1	63.91	58	5.91	95.58	67.65
2	60.07	54	6.07	93.86	81.75
3	45.74	42	3.74	86.01	39.74
4	51.53	46	5.53	87.19	64.84
5	57.05	51	6.05	75.15	35.99
6	54.03	49	5.03	95.58	96.80
7	46.85	42	4.85	99.61	94.09
8	53.97	50	3.97	83.33	29.32
9	47.73	43	4.73	88.18	62.50
10	67.16	61	6.16	87.87	42.67
11	60.60	54	6.6	92.11	79.89
12	49.39	45	4.39	89.50	94.33
13	54.84	49	5.84	86.41	82.49
14	51.26	46	5.26	92.21	11.83
Mean ± SD	54.58 ± 6.27	49.29 ± 5.65	5.30 ± 0.87	89.47 ± 5.89	67.3 5 ± 20.98

Abbreviations: 10-MW, 10-Meter Walk
*Habitual Distance = mean distance from walker with no feedback determined during 10-MW
**Target Distance = (Habitual Distance) – (Habitual Distance x 0.10)

Table 3. Distance from walker crossbar during baseline and feedback trials.

Variable	No Feedback	Feedback	*P* value	Effect size (d)	Power
10-Meter Walk					
Average Distance (cm)	54.58 ± 6.27	43.16 ± 5.54	< 0.001*	1.92	1.00
Obstacle Course					
Average Distance (cm)	63.45 ± 8.28	48.40 ± 6.99	< 0.001*	1.95	1.00

Table 4. Joint kinematics during the 10-meter walk and obstacle course.

Variable	No Feedback	Feedback	P	Effect size (d)	Power
10 Meter Walk					
Elbow Flexion†	24.18 ± 8.90°	27.37 ± 9.30°	0.013*	0.78	0.74
Shoulder Flexion†	9.03 ± 10.06°	−1.12 ± 7.95°	0.001*	1.15	0.96
Hip Flexion†	2.65 ± 7.93°	−1.28 ± 14.88°	0.255	0.33	0.19
T1-C7 Flexion	18.85 ± 3.31°	20.46 ± 3.60°	0.017*	0.76	0.71
C1-Head Flexion	−1.16 ± 6.02°	1.57 ± .6.57°	0.032*	0.674	0.60
Trunk Inclination	6.45 ± 5.20°	7.29 ± 4.97°	0.141	0.438	0.30
Obstacle Course					
Elbow Flexion†	25.39 ± 7.51°	29.96 ± 8.40°	0.013*	0.806	0.76
Shoulder Flexion†	19.57 ± 8.43°	4.94 ± 9.59°	0.000*	2.822	1.00
Hip Flexion†	10.12 ± 12.37°	3.81 ± 16.48°	0.023*	0.721	0.66
T1-C7 Flexion	19.52 ± 4.04°	19.85 ± 3.42°	0.330	0.282	0.15
C1-Head Flexion	0.00 ± 7.34°	0.35 ± 6.21°	0.588	0.154	0.08
Trunk Inclination	10.21 ± 5.64°	8.69 ± 4.37°	0.016*	0.776	0.72

†Elbow, shoulder, and hip joint angles calculated for the right side of the body
*Statistically significant at $p \leq .05$

acceleration and deceleration). This finding was expected and is consistent with previous research which found lower response efficiencies during activities that require greater attentional demand (Verhoeff, Horlings, Janssen, Bridenbaugh, & Allum, 2009). It can be assumed that this phenomenon would be especially true when the feedback is communicated visually and when greater visual observance of the environment is necessary for task completion. That said, these findings support that even under these greater attentional and physical demands, individuals were still able to attend to the feedback provided by the device the majority of the time.

In addition, it was found that the individuals, on average actually shifted 20% closer to their walker with the feedback during the 10-MW (11.4 cm, $P < .001$) and 24% closer during the obstacle course (15.0 cm, $P < .001$). This meant that individuals were even closer to the position recommended by clinical guidelines than the target had been set at. This was a promising finding and indicates that future work is now needed to determine how to best shift individuals all the way to the ultimately desired position.

While achieving a closer position to that recommended by guidelines may have benefits, we recognize that there could be a trade-off if the attentional demands of using the device are too great and cause excessive distraction when used. While this possible effect was not explicitly addressed in this study, we did look for indications that the attentional demands required by attending to the device were interfering with ambulation. All individuals demonstrated an ability to navigate safely and without collisions or task difficulty while attending to the device. There were also no significant differences in walking speed during the 10-MW and obstacle course completion times between the trials where no feedback was provided and when it was. These preliminary indications are encouraging and may indicate that, for the population studied, individuals were able to attend to the device without significant detriment to their functional performance. It is unknown, however, what the effects of the increased attentional demands may be in more novel environments and/or in other populations who may find the increased cognitive load more taxing.

Kinematics and body position

A secondary aim of this study was to analyze the biomechanical and postural changes that were used to achieve the closer position to the walker that was encouraged by the feedback device. The analysis of the specific joint angles showed that a decrease in shoulder flexion was the primary method used by the participants to bring the walker closer to them. This reduction in shoulder flexion is logical due to anatomical linkages and its relationship to humeral orientation when the hand is holding onto an object such as a walker.

In addition to upper extremity joint motions, changes in trunk segment motions and posture were assessed. This was done to evaluate if there were any changes in upright posture when the participants moved closer to their walkers. It was speculated that the participants might improve their posture as they moved closer to the walker by standing up straighter. This was assessed by measuring forward trunk inclination during each of the trials. During the 10-MW there was no change in trunk inclination; however, during the obstacle course, there was a significant decrease in forward trunk inclination during the feedback trials. While this decrease was statistically significant, the actual change was only ~1.5°, which may be of questionable clinical relevance. Therefore, it does not seem that trunk posture was meaningfully improved through the use of the device alone. In retrospect, this finding may have been anticipated given previous research showing that up to 50% of walker users may have difficulty changing trunk posture due to fixed kyphotic deformities and spinal stenosis which was not controlled for in the present study (Liu, 2009).

Because the device uses visual feedback and is placed on the front crossbar of the walker in low position to prevent visual obstruction, there was a concern that users may need to look down at the device which could lead to unwanted increases in cervical and/or head flexion. This was assessed by measuring flexion at both the lower (T1-C7) and upper (C1-Head) cervical regions. When using the feedback, there were statistically significant increases in flexion at both the T1-C7 and C1-Head segments during the 10-MW; however, the actual changes were relatively small (<3°) in comparison to the total available range of motion of these joints (Kuhlman, 1993). During the obstacle course trials, where more attention and visual scanning of the environment may be required, there were no differences in cervical or head flexion between conditions. This may be because individuals did not look down at the device as often as they did during the 10-MW, choosing to attend to the environment over the light at times. If so, this may also explain why individuals were outside of the target range more often during the obstacle course than they were during the 10-MW (67.3% of the time in range, versus 89.5% of the time). Another reason that participants may have spent less time in the target range during the obstacle course was the multiple turns and thresholds that they were required to negotiate. When turning or pushing the walker over the thresholds participants may have pushed the walker further in front of them to improve the leverage and mechanical advantage needed for these tasks. Even so, it was encouraging that individuals were able to attend to the device for the majority of the time and without increased cervical or head flexion required.

In summary, changes in trunk inclination and cervical flexion were generally small and demonstrate that the feedback device had minimal negative or positive impacts on posture. Therefore, while the primary goal of improving relative position was effectively addressed by the current device, in order to improve upright posture, additional or different types of feedback would need to be explored.

Limitations

While this study suggested that the developed device has potential, there are also a number of limitations that should be considered. From a device perspective, it is acknowledged that with loose clothing the relative distance measurement would be closer than the true position. However, given the simplicity of this technique, it does lend itself well for providing an effective way of estimating relative user position. If looking to improve such sensing, emphasis may move to sensing and monitoring foot position which would have less of an error given that loose clothing (e.g. gowns) likely do not extend to near the floor. However, doing so would be more involved in order to accurately estimate the relative user position.

From a participant perspective, this study involved a small sample of motivated individuals with normal cognitive function and therefore may not be an adequate representation of typical walker users. It is also realized that the attentional demands required to attend to the device were not explicitly examined in this study. This means that, especially for individuals who may have increased difficulty with dual tasking (e.g. Parkinson's disease, dementia), the benefits gained by using the device to obtain a more optimal positon may be negated if the increased attentional demands are detrimental to the safety and stability of the user. This is an important limitation that future work will need to address before determining who may best benefit from such a device.

It is also acknowledged that in this study, the device was programmed so that individuals were encouraged to make a change that moved them only 10% closer to the position recommended by clinical guidelines. While most individuals shifted even more than that, ultimately the goal would be for individuals to get to and maintain the recommended position. Future work is needed to determine whether this distance would be able to be successfully reached by the majority of users of the device and how improvements in position, even if incremental, relate to improvements in stability, safety, and other functional outcomes. It should also be acknowledged that there is little data to support the current clinical recommendations for rollator walker fitting or positioning, therefore, attempting to achieve this recommended position may not be relevant or even appropriate for many individuals. Until more research is conducted, clinicians will need to rely on their clinical expertise and an understanding of the individual needs of the patient to provide the best fit and positioning recommendations. The device could then be used as a tool to help the patient achieve and maintain this individually determined position.

Another important study limitation is that this study only examined the immediate effects of feedback over short walking distances and time frames. Future research should assess use over longer distances and time frames as well as the frequency of feedback that may optimize carry-over effects when feedback is reduced or discontinued. Longer term use will require device refinement and considerations such as maximizing battery life. Exploration of different forms (tactile and auditory) or combinations of feedback for individuals for which visual feedback is not appropriate would also be of future interest.

Conclusions

In this study, a positon feedback device was developed and validated by assessing walker position and body kinematics during a 10-MW and obstacle course with and without feedback from the device. It was found that participants were able to attend and respond to the visual biofeedback cues provided by the device. The position monitor helped individuals achieve and maintain a closer distance to their walker in both simple and complex walking conditions. Individuals achieved this change predominantly by decreasing their amount of shoulder flexion. There were only small changes in trunk inclination when feedback was provided, suggesting that while the device addresses the common misuse of pushing the walker too far in front of the user, in its current form it does not seem to elicit a more upright and favorable position, another common issue. Changes to neck posture were also small, indicating that individuals did not need to take on unfavorable positions in order to view and attend to the device. In conclusion, this study validates the ability of the newly designed position monitor to encourage an individual to ambulate closer to the position encouraged by clinical guidelines and motivates future work to refine and fully recognize the potential of such a device.

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