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Abstract

Trips are a major cause of falls. Sagittal-plane kinematics affect clearance between the foot and obstacles, however, it is unclear which kinematic measures during obstacle-free walking are associated with avoiding a trip when encountering an obstacle. The purpose of this study was to determine kinematic factors during obstacle-free walking that are related to obstacle avoidance ability. It was expected that successful obstacle avoidance would be associated with greater peak flexion/dorsiflexion and range of motion (ROM), and differences in timing of peak flexion/dorsiflexion during swing of obstacle-free walking for the hip, knee and ankle. Three-dimensional kinematics were recorded as 35 participants (young adults age 18–45 (N = 10), older adults age 65+ without a history of falls (N = 10), older adults age 65+ who had fallen in the last six months (N = 10), and individuals who had experienced a stroke more than six months earlier (N = 5)) walked on a treadmill, under obstacle-free walking conditions with kinematic features calculated for each stride. A separate obstacle avoidance task identified trippers (multiple obstacle contact) and non-trippers. Linear discriminant analysis with sequential feature selection classified trippers and non-trippers based on kinematics during obstacle-free walking. Differences in classification performance and selected features (knee ROM and timing of peak knee flexion during swing) were evaluated between trippers and non-trippers. Non-trippers had greater knee ROM (P = .001). There was no significant difference in classification performance (P = .193). Individuals with reduced knee ROM during obstacle-free walking may have greater difficulty avoiding obstacles.

Keywords

Falls risk, Tripping, Linear discriminant analysis, Gait analysis

1. Introduction

Achieving adequate foot clearance is crucial for preventing trips, one of the greatest causes of falls (Berg et al., 1997, Blake et al., 1988, Overstall et al., 1977, Tuunainen et al., 2014, Robinovitch et al., 2013, Heijnen and Rietdyk, 2016). Foot clearance can be accounted for by the sagittal plane motion of the lower extremity joints (Winter, 1992). While individuals may employ different strategies to achieve adequate foot clearance (Little et al., 2014, Levinger et al., 2012), each strategy for avoiding an obstacle ultimately relies on the magnitude of lower extremity joint angles and the timing of the joint motion. Since changes in the sagittal plane ankle, knee and hip angles individually affect foot clearance throughout swing phase of gait (Winter, 1992, Gates et al., 2012, Schulz, 2011, Schulz et al., 2010, Moosabhoy and Gard, 2006, deficits in the magnitude of these joint motions may contribute to an inability to avoid obstacles while walking. Likewise, reduced time to avoid an obstacle, perhaps due to a shorter swing time, shorter distance in the approach between the foot and obstacle, or faster walking speed, may also result in reduced foot clearance (Chou and Draganich, 1998). This is particularly relevant if an individual's gait pattern during obstacle-free walking is not conducive to avoiding obstacles that appear suddenly, such as a previously unseen curb or a pet that darts into the walking path.

Just as the prevalence of falls increases with age (Talbot, Musiol, Witham, & Metter, 2005) or disability (Forster and Young, 1995, Wagner et al., 2009, Mackintosh et al., 2005), the risk of tripping is different among certain demographic groups. Healthy young adults reported on average at least one slip or trip per week, with only 5% of those perturbations leading to a fall (Heijnen and Rietdyk, 2016). On the other hand, older adults are more likely to trip than young adults (Garman, Franck, Nussbaum, & Madigan, 2015), and have a greater risk of contacting an obstacle while walking, likely related to longer reaction times than young adults (Chen, Ashton-Miller, Alexander, & Schultz, 1994). Stroke survivors may have motor impairments that limit foot clearance during walking (Batchelor, Mackintosh, Said, & Hill, 2012). In particular, those who have experienced a stroke have been shown to exhibit less overall limb shortening, with maximal limb shortening occurring later in the gait cycle (Little, McGuirk, & Pattern, 2014). Thus, the kinematic factors that influence obstacle avoidance ability may be related to age or disability status.

Despite the obvious consequences of inadequate foot clearance and the incidence of trips, it is unclear which factors during obstacle-free walking are specifically related to the ability to avoid obstacles that appear suddenly. These factors may manifest as specific joint kinematics during obstacle-free walking that influence foot clearance in the presence of an obstacle, and these factors may be more pronounced among certain demographic groups. By investigating the walking patterns of stroke survivors as well as older adults with and without a history of falls and young adults, the purpose of this study was to determine the sagittal plane kinematics during obstacle-free walking that are related to the ability to avoid an obstacle that appears suddenly. It was expected that the stroke survivors and older adults with a history of falls would not be able to avoid the obstacle, and participants who were able to avoid an obstacle would have different obstacle-free gait characteristics than those who were not able to avoid the obstacle. In particular, it was projected that successful obstacle avoidance would be associated with greater peak hip and knee flexion, ankle dorsiflexion, and sagittal plane range of motion during swing for the lower extremity joints. Additionally, differences in timing of peak hip and knee flexion and ankle dorsiflexion during swing were expected.

2. Methods

2.1. Participants

The study protocol was approved by the University of Wisconsin-Milwaukee Institutional Review Board, and all participants provided informed consent. Thirty-five community-dwelling participants included young adults age 18–45 (N = 10), older adults age 65+ without a history of falls (N = 10), older adults age 65+ who had fallen in the last six months (N = 10), and individuals who had experienced a stroke more than six months earlier (N = 5) (Table 1). A fall was defined as unintentionally coming to rest on the ground (Senden, Savelberg, Grimm, Heyligers, & Meijer, 2012). All participants were able to walk without an assistive device for five minutes at a time. Inclusion was limited to participants with a Mini-Mental State Examination score greater than 22 (Savin, Morton, & Whitall, 2014). The participants with chronic stroke completed the lower extremity sub-scale of the Fugl-Meyer assessment, which has a range of possible scores of 0–34 (Sanford et al., 1993, Sullivan et al., 2011.

Table 1. Participant Characteristics.

	Young Adult	**Older Adult – Non-faller**	**Older Adult – Faller**	**Stroke**
N	10	10	10	5
Age (range), yr	30.5 (22–44)	71.9 (65–87)	75.3 (66–91)	61.6 (40–83)
Height (SD), m	1.74 (0.14)	1.68 (0.08)	1.72 (0.12)	1.68 (0.10)
Weight (SD), kg	76.0 (18.1)	75.9 (16.2)	86.3 (23.0)	82.6 (13.4)
Sex	5M, 5F	3M, 7F	5M, 5F	2M, 3F
Number of Falls 6 Months (range)	0.1 (0–1)	0	1.4 (1–3)	0.4 (0–1)

Mini Mental State Exam (range)	29.6 (28–30)	29.3 (28–30)	28.6 (27–30)	27.6 (24–30)
LE Fugl-Meyer (range)	–	–	–	24.6 (17–31)
Affected Side	–	–	–	3 R, 2 L
Type of Stroke	–	–	–	5 ischemic
Time since stroke onset (range), mo	–	–	–	43.2 (10–120)

Note. SD = standard deviation; LE = lower extremity.

2.2. 3D motion capture

Each participant was provided a pair of standard laboratory shoes (Saucony Jazz, Lexington, MA) and tight-fitting shorts. Retroreflective markers used for motion capture were applied bilaterally to track the motion of the thigh, leg and foot. The tracking markers were placed on the right and left anterior and posterior superior iliac spines, a four-marker plate on the thighs and the legs, and a rigid four-marker cluster attached to the heel counter of the shoes. A standing calibration was recorded with calibration markers on the following bilateral anatomical locations: greater trochanter, lateral and medial femoral epicondyles, malleoli and first and fifth metatarsal heads. Additional calibration markers were placed on the distal end of each shoe and on the second metatarsal head. The location of these markers in the local coordinate system of the foot was used to determine their virtual positions during the movement trials to avoid the risk of the markers falling off during the subsequent obstacle avoidance task. The calibration markers were removed following a three-second standing calibration trial. During all trials, the three-dimensional positions of each marker were continuously collected at 200 Hz with a ten-camera Eagle system (Motion Analysis, Inc., Santa Rosa, CA). These data were filtered using a 4th order, zero-lag, recursive Butterworth filter with a cutoff at 10 Hz, allowing components of the signal due to walking to be retained while removing noise associated with higher frequency content from the signal (Angeloni, Riley, & Krens, 1994).

From the calibration trial, the joint center of each hip was established as 25% of the distance between the left and right greater trochanters (Weinhandl and O'Connor, 2010), and the knee and ankle joint centers were defined as the midpoint between the lateral and medial femoral epicondyles and malleoli, respectively. Right-handed local coordinate systems were defined for the pelvis, thigh, shank and foot segments as outlined by Wu et al. (2002). Three-dimensional joint angles at the hip, knee and ankle were calculated using a joint coordinate system approach (Wu et al., 2002, Grood and Suntay, 1983). Processing of the kinematic data was done using Visual 3D software (v5.00.24; C-Motion, Inc., Rockville, MD, USA).

2.3. Walking and obstacle avoidance tasks

Each participant walked at their self-selected walking pace on a treadmill (Precor, C964i, Woodinville, WA, USA). Participants practiced treadmill walking for one minute while no kinematic data were recorded. Participants then walked on the treadmill for one minute and performed a separate treadmill-based obstacle avoidance task. The order of the obstacle-free walking condition and the obstacle avoidance task was randomized, but kinematic data were recorded during both tasks. Participants wore a safety harness that provided no support during walking but prevented the participant from landing on the ground in the case of a fall. Participants were allowed to rest at any point if their perceived exertion was above very light (Borg, 1970).

The obstacle avoidance task consisted of four one-minute periods of treadmill walking where participants were presented with obstacles, which they were instructed to attempt to avoid. The obstacles were lightweight pieces of foam cut to length, width and height dimensions of 20 x 16 × 6 cm (Airex AG, Balance-pad, CH-5643 Sins, Switzerland). Similar to the process outlined by Weerdesteyn, Schillings, van Galen, and Duysens (2003), at random heel-strike events, the foam was placed by an experimenter on the belt of the treadmill in front of the foot entering stance phase so that the obstacle would have to be avoided in the subsequent swing phase. Considering typical minimal foot clearance for most elderly adults has been reported to be no more than 5 cm (Begg, Best, Dell'Oro, & Taylor, 2007), using a 6-cm obstacle required the participant to react to the object to avoid coming in contact with it. This is also within the range of obstacle heights used in previous studies of obstacle avoidance in stroke survivors (Said, Goldie, Patla, & Sparrow, 2001). The participant continued to walk on the treadmill until another obstacle was presented, for a total of six obstacles in a one-minute period, and this process was repeated four times for 24 obstacles overall. Although the participants were aware that obstacles would be present, the conditions were designed so that the participants would not know when they would encounter an obstacle. The number of steps between obstacles was randomized, as was the foot (right or left), however, within each period the obstacle was presented on the right side three times and the left side three times. To further ensure that participants were reacting to – and not anticipating – the obstacles, participants completed a concurrent visual dual task that required participants to look straight ahead and not down at their feet or the experimenter placing the obstacles on the treadmill. For the visual task, an arrow appeared on a screen positioned at eye level approximately two meters from the treadmill. The participants reported the direction the arrow was pointing, and a new arrow appeared one second after each response. Participants were not instructed to prioritize either the visual or obstacle avoidance task.

Once placed on the treadmill belt, the obstacle did not move relative to the treadmill belt unless the participant came in contact with the obstacle. As the obstacle was lightweight foam, contact by the participant's toe did not result in an actual trip, but rather the obstacle was kicked toward the experimenter who then batted the obstacle out of the way to prevent the participant from having a second exposure to the obstacle. A participant was labeled as a tripper if they came in contact with an obstacle more than once. Non-trippers were participants that came in contact with an obstacle just one or no times. The classification was determined by tracking the location of retroreflective markers attached to the obstacle, and using custom software (Matlab v8.0.0.783, Mathworks, Inc., Natick, MA, USA) to identify any changes in velocity of the markers relative to the treadmill belt speed, as well as the location of the heel and virtual distal shoe markers relative to the position of the obstacle.

2.4. Data processing

For the obstacle-free walking condition only, heel-strike and toe-off events were determined from the location of the heel and virtual second metatarsal markers using the horizontal velocity algorithm (Zeni, Richards, & Higginson, 2008). These gait events were used to segment the kinematic data into strides. Only strides from the affected leg of the stroke survivors and a randomly chosen leg of all other participants were used in the analysis. For each stride, a total of nine features were extracted: peak joint angle, time to peak joint angle, and range of motion (ROM) during swing for the sagittal plane motion at the ankle, knee and hip.

2.5. Classification model

A Linear Discriminant Analysis (LDA) was applied to the dataset of extracted features for each stride to determine the features that best discriminate between trippers and non-trippers. First, a sequential feature selection method was employed to reduce the number of features included in the LDA. Starting with an empty set, features were added until there was no improvement in prediction for a linear discriminant model that classified trippers and non-trippers, evaluated using 10-fold cross-validation. This process was repeated 10 times, and features that were selected at least once during the feature selection process were retained.

The full dataset was separated into training and testing datasets, representing approximately 90% and 10% of all strides by all participants, respectively. To prevent overfitting, at any point, all strides of a given participant were in either the training or the testing dataset, not both. The 35 participants were organized into the training and testing datasets similar to a leave-one-out approach, such that the testing dataset included all strides of one tripper with the remaining strides from randomly selected non-trippers (up to approximately 10% of all strides), and the training dataset included all strides from the participants not included in the testing dataset. Pairs of training and testing datasets were generated until each participant was included in exactly one testing dataset. This process was then repeated 10 times, for a total number of folds equal to 10 times the number of trippers. For each fold, a LDA was applied to the training dataset. The model that was created was then applied to the testing dataset, with performance evaluated by classification accuracy (Fig. 1). The performance was averaged across all of the folds. Further iterations were performed as the selected features were entered into the feature set in a stepwise manner, beginning with the feature with the highest selection rate.

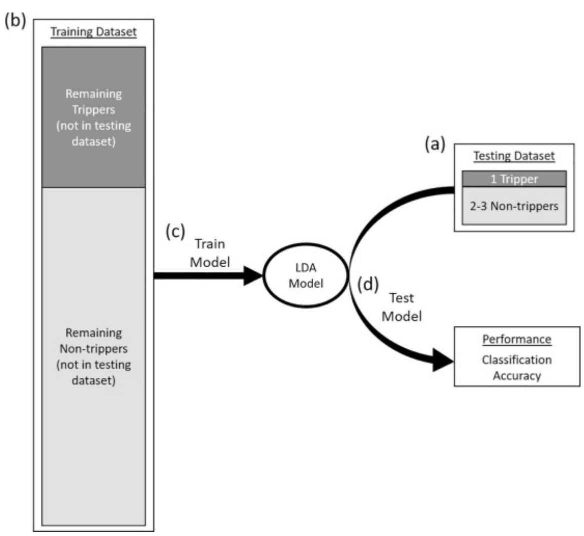

Fig. 1. For each feature set, classification accuracy was determined according to 10 times the number of trippers folds of the following process: a) A testing dataset was created from all strides of one tripper and two or three non-trippers (accounting for approximately 10% of the total number of strides across all participants); b) A training dataset was created from all strides of the remaining trippers and non-trippers not included in the testing dataset (accounting for approximately 90% of the total number of strides across all participants); c) The training dataset was used to train a LDA model to discriminate trippers and non-trippers; d) The testing dataset was applied to the LDA model, with performance evaluated as classification accuracy. This process represents one fold; additional folds were completed until all participants appeared in exactly one testing dataset, and the entire procedure was repeated 10 times.

2.6. Statistical analysis

Average model performance was compared for each feature set to identify which features resulted in the best discrimination between trippers and non-trippers. Additionally, for the best feature set, determined by classification accuracy, classification accuracy for each individual subject was recorded. Differences in classification accuracy were evaluated across groups and between trippers and non-trippers using a one-way ANOVA. Accompanying ANOVAs were used to determine group differences and differences between trippers and non-trippers for the variables included in the best feature set. For significant ($P < .05$) group differences, follow-up tests included all pairwise comparisons, with significance determined at a Bonferroni-adjusted level of $\alpha = 0.008$. The LDA was implemented using custom software (Matlab v9.1.0.441655, Mathworks, Inc., Natick, MA, USA), and all statistical analyses were done in SPSS (v24.0.0.1; SPSS, Inc., Chicago, IL).

3. Results

Over the course of the obstacle-free walking condition for all participants, a total of 1580 strides on the affected side (stroke survivors) or randomly chosen leg (all other participants) were recorded. During the obstacle avoidance task, 10 participants (3 older non-fallers, 4 older fallers, and 3 stroke survivors) were identified as trippers, and the remaining 25 as non-trippers. Of the 10 trippers, only 5 had reported at least one fall in the previous 6 months. Differences between the tripper and non-tripper waveforms for sagittal plane joint angles during obstacle-free walking can be observed visually (Fig. 2). Four features were selected at least once during the feature selection process (Table 2). Therefore, the LDA was evaluated over 100 folds (10 × 10 trippers) for each of four feature sets. As the first two features were added to the feature set, the classification accuracy improved, however, the performance declined when more than two features were included in the feature set (Table 3). The features included in the best feature set were knee ROM during swing and time to peak knee angle during swing.

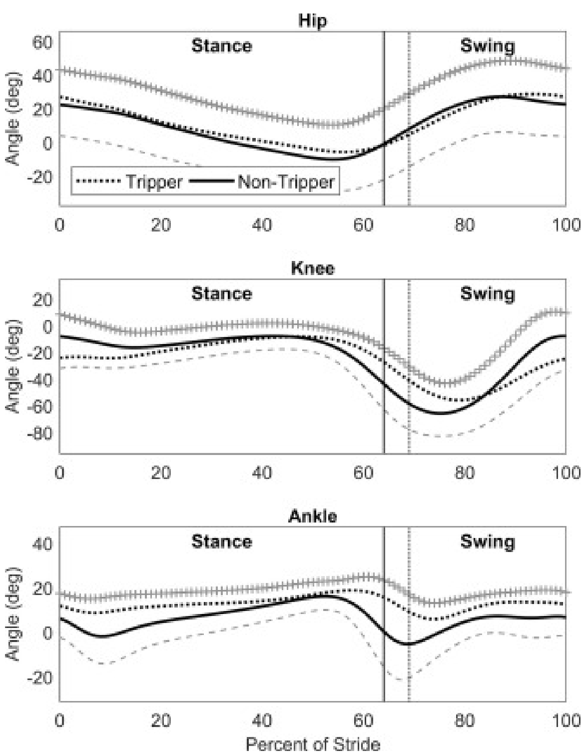

Fig. 2. Mean sagittal plane joint angles throughout the gait cycle for trippers (dashed black line) and non-trippers (solid black line). The dashed and solid vertical lines represent toe-off for the trippers and non-trippers, respectively. The tripper and non-tripper waveforms are bracketed by ±1.96 standard deviations of the overall mean (gray lines). Positive angles represent hip flexion, knee extension and ankle dorsiflexion.

Table 2. Features ranked in order of number of times they were selected in the feature selection algorithm.

Rank	Feature
1	Knee ROM swing
2	Time to peak knee swing
3	Ankle ROM swing
4	Time to peak hip swing

Table 3. Features included and model performance for each feature set. Features were added in a stepwise manner based on their rank (Table 2) during the feature selection process. The second feature set, which included the top two ranked features, had the best classification accuracy. Sensitivity represents the proportion of actual trippers correctly identified. Specificity represents the proportion of actual non-trippers correctly identified.

Features Included	Accuracy	Sensitivity	Specificity
1	0.917	0.722	0.960
1, 2	0.923	0.763	0.957
1, 2, 3	0.896	0.774	0.917
1, 2, 3, 4	0.902	0.776	0.926

3.1. Kinematic differences

When assessing the differences between groups for each of the features in the best feature set, the assumption of homogeneity of variance was violated for knee ROM during swing ($Levene_{(3,31)} = 4.322$, *P* = .012). While Levene's test was not significant for time to peak knee angle during swing ($Levene_{(3,31)} = 2.526$, *P* = .076), when comparing the group with the greatest variance to the group with the least variance for each of the variables, there was a high ratio (>4) of variances. Therefore, Brown-Forsythe F tests were run and there were significant group differences in knee ROM ($F_{(3,12.493)} = 4.360$, *P* = .026) but no significant group differences in timing of the peak knee angle during swing ($F_{(3,9.469)} = 1.610$, *P* = .252). Pairwise comparisons indicated that the young participants had significantly greater knee ROM than the stroke survivors (Fig. 3). During the comparison between trippers and non-trippers for the features in the best feature set, the assumption of homogeneity of variance was violated for time to peak knee angle during swing ($Levene_{(3,31)} = 9.039$, *P* = .005). Results of the Brown-Forsythe F tests indicated that non-trippers (M = 60.0, SD = 10.6 degrees) had greater knee range of motion than trippers (M = 35.8, SD = 15.5 degrees; $F_{(1,12.479)} = 20.327$, *P* = .001), but there were no significant differences between trippers (M = 77.7, SD = 4.7 percent stride) and non-trippers (M = 76.1, SD = 2.1 percent stride) in timing of the peak knee angle during swing ($F_{(1,10.476)} = 1.125$, *P* = .313).

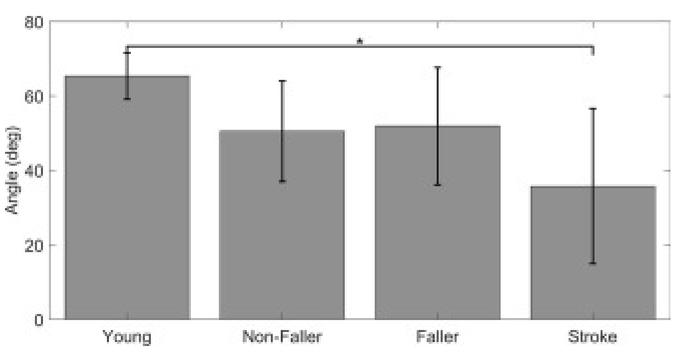

Fig. 3. Knee range of motion during swing for each group. *Significant difference between groups at the Bonferroni-adjusted α = 0.008.

3.2. Model performance

Overall, a LDA using knee ROM and timing of peak knee flexion during swing classified all strides with 92.3% accuracy. The proportion of actual trippers correctly classified (sensitivity) was 0.763, and the proportion of actual non-trippers correctly classified (specificity) was 0.957 (Table 3). The strides for all but four individuals were classified with accuracy ranging from 83.4 to 100% (Table 4). Four individuals within the older faller group had very poor classification accuracy (<50%). The assumption of homogeneity of variance was violated, particularly because the variance in the stroke survivor group was zero (all strides of all stroke survivors were correctly classified). Despite the unequal variances, a Brown-Forsythe F test could not be used to assess group differences in classification performance due to the zero variance of the stroke survivor group. Therefore, a one-way ANOVA indicated significant differences between groups ($F_{(3,31)} = 5.016$, $P = .006$), and pairwise comparisons indicated that classification accuracy for the older faller group was significantly lower than the young adult and older non-faller groups after the Bonferroni correction of $\alpha = 0.008$ (Table 4). Results of a Brown-Forsythe F test indicated there was no significant difference ($F_{(1,10.932)} = 1.925$, $P = .193$) in classification performance between trippers (M = 76.3, SD = 39.3 percent) and non-trippers (M = 94.4, SD = 20.1 percent).

Table 4. Average model performance for each individual. Participants are organized according to their group, and sorted from best classification accuracy to worst within groups.

Young			Older Non-fallers			Older Fallers			Stroke		
Participant	T/NT	Accuracy	Participant	T/NT	Accuracy	Participant	T/NT	Accuracy	Participant	T/NT	Accuracy
Y1	NT	1.000	ON1	T	1.000	OF1	T	1.000	S1	T	1.000
Y2	NT	1.000	ON2	T	1.000	OF2	NT	1.000	S2	T	1.000
Y3	NT	1.000	ON3	T	1.000	OF3	NT	1.000	S3	T	1.000
Y4	NT	1.000	ON4	NT	1.000	OF4	NT	1.000	S4	NT	1.000
Y5	NT	1.000	ON5	NT	1.000	OF5	NT	1.000	S5	NT	1.000
Y6	NT	1.000	ON6	NT	1.000	OF6	NT	0.834			
Y7	NT	1.000	ON7	NT	1.000	OF7	T	0.400			
Y8	NT	1.000	ON8	NT	0.988	OF8	T	0.228			
Y9	NT	1.000	ON9	NT	0.978	OF9	T	0.000			
Y10	NT	0.894	ON10	NT	0.913	OF10	NT	0.000			
Mean (SD)	0.989 (0.034)			0.988 (0.028)			0.646 (0.439)			1.000 (0.000)	

Note: T = tripper; NT = non-tripper.

4. Discussion

Participants were identified as trippers or non-trippers based on an obstacle avoidance task. Trippers came in contact with the obstacles multiple times, suggesting that their walking pattern was not adequate for reacting to and avoiding obstacles. The identification of trippers and non-trippers did not necessarily correspond with the participants' recorded history of falls, as falls recorded in the previous six months may have been due to causes other than trips. Due to differences in the prevalence of falls and risk of tripping among demographic groups (Heijnen and Rietdyk, 2016, Talbot et al., 2005, Garman et al., 2015, Batchelor et al., 2012), it was expected that older adults with a history of falls and stroke survivors would not be able to avoid the obstacles, yet less than 50% of these participants were trippers. Therefore, the ability to predict trippers and non-trippers cannot be accurately determined based on a history of falls or stroke.

4.1. Kinematic differences

The kinematic features during obstacle-free walking that best discriminated between trippers and non-trippers were knee ROM and timing of peak knee flexion during swing, with trippers exhibiting a significantly smaller knee ROM than non-trippers. Reduced knee ROM indicates that an individual may not be able to achieve adequate foot clearance when reacting to an obstacle. While there were no significant differences between trippers and non-trippers in timing of peak knee flexion during swing, including this variable in the model improved the ability to distinguish between trippers and non-trippers using kinematic variables. This result is consistent with the findings of Chou and Draganich (1998), who showed that greater angular velocity of the knee is required to avoid obstacle contact, as faster knee angular velocity means a shorter time to peak knee flexion, and more time to react to the obstacle.

Sagittal plane ROM and timing of peak joint angles were examined for the ankle, knee and hip because of the relationship between sagittal plane joint motion and foot clearance. Gait adaptations to accommodate varying walking surfaces (Gates, Wilken, Scott, Sinitski, & Dingwell, 2012) and perform everyday tasks while walking (Schulz, Lloyd, & Lee, 2010) include concurrent changes in joint kinematics and foot clearance height, and a physiological range of joint angles at the ankle, knee and hip could independently account for the variability observed in foot clearance (Winter, 1992). Previous studies have examined how the sagittal plane motion of the lower extremity joints influences foot clearance. Moosabhoy and Gard (2006) developed a theoretical model which suggested that ankle dorsiflexion has a greater effect on foot clearance during mid-swing than knee or hip flexion, while knee and hip flexion have the greatest effect on foot clearance at the beginning and end of swing phase of healthy gait. More recently, Little et al. (2014) found that among stroke survivors, the knee has the greatest influence on foot clearance and limb shortening at the lowest trajectory of the toe, regardless of the time during swing. Discrepancies may be attributed to the fact that different populations may use different strategies to achieve adequate foot clearance (Levinger et al., 2012). While the achievement of adequate foot clearance relies on contributions from all of the joints in the lower extremity, in this study, greater knee ROM and the timing of peak knee flexion in the stride cycle were the variables that best identified those able to avoid obstacles while walking.

4.2. Model performance

The group differences in classification accuracy show that the LDA performance was significantly better for the young and older non-faller groups than the older faller group. This is likely due to the very poor classification accuracy for four members of the older faller group: one who was a non-tripper, and the other three trippers. Of these two scenarios, the misclassification of the trippers (OF7, OF8 and OF9 in Table 4) is the most concerning, as these individuals appeared to walk like non-trippers, but were unable to avoid the obstacles. These misclassifications are represented in the low sensitivity, or proportion of actual trippers correctly classified. Therefore, for some individuals – and perhaps those with a history of falls in particular – the risk of tripping would be identified by other factors besides knee kinematics. Additionally, while knee kinematic measures were able to identify most individuals at risk for tripping, a trip does not occur every time that someone walks, and falls may occur due to events other than tripping. Therefore, it would be beneficial to detect gait characteristics such as limited ROM and risk factors for other types of falls in real-time as part of a falls-prevention program.

4.3. Limitations

Three limitations to this study are acknowledged. First, the number of participants, particularly in the stroke survivor group, was small. The stroke group was expected to be the most susceptible to tripping, and effect of the small stroke group was magnified as two of the five stroke survivors were non-trippers. The second limitation is that based on the study design, all participants were relatively high-functioning, with the ability to walk without assistance for at least five minutes. Taken together, having a small group of high-functioning participants suggests that it would be difficult to identify gait kinematics related to the risk of tripping, yet, using knee ROM and time to peak knee flexion during swing, it was possible to distinguish trippers and non-trippers with 92.3% accuracy. Nevertheless, additional factors may be helpful in determining the risk of tripping. Therefore, the third limitation is that other independent variables – such as walking speed, available ROM, kinematic variability, strength, reaction time, and balance – were not included in the model. The potential effects of some of these factors are discussed below.

Some of the kinematic differences between participants may be attributed to the effects of walking at a slower speed. Walking at a slower than normal speed has been shown to result in a decrease in peak knee flexion (Kirtley, 2006, Kwon et al., 2014), and is a common characteristic of gait for older adults or stroke survivors who fear they may be at risk for falling (Park and Yoo, 2014, Maki, 1997). In this study, participants walked at a self-selected speed, and non-trippers walked faster than trippers ($P < .001$). This is in contrast to the expectation that a faster walking speed corresponding with less reaction time to avoid the obstacles might translate into a greater risk of tripping. Nevertheless, to ensure that the kinematic differences observed between trippers and non-trippers were not simply a function of the speed difference (Lelas, Merriman, Riley, & Kerrigan, 2003), a posthoc analysis was performed using linear regression across all participants to predict knee ROM and timing of peak knee flexion during swing from speed. Speed accounted for 63.7% of the variance in knee ROM, and 12.2% of the variance in timing of peak knee flexion during swing. While the relationship between speed and knee ROM is strong, speed does not account for all of the variation in either of the variables, indicating that other factors contribute to the kinematic differences between trippers and non-trippers.

It is possible that the trippers had a physical limitation in their ability to produce knee flexion, although this was not measured statically. The stroke participants were expected to have this limitation in their affected leg (Balaban and Tok, 2014, Olney and Richards, 1996), and in fact had a lower knee ROM during swing than the young adult group. Some older adults were also expected to have limited knee ROM (Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998), as the available ROM at the hip, knee and ankle has been reported to be lower in older adults than younger adults (Chen, Ashton-Miller, Alexander, & Schultz, 1991). However, while there was a trend for a lower knee ROM during walking among both older adult groups compared to the young adults, consistent with other findings (Chen et al., 1991), these differences were not significant. It appears lower knee ROM during swing occurred for the older adults in the faller and non-faller groups that were unable to avoid the obstacle while walking. Although it was not significant by itself, an individual with a delay in the peak knee angle combined with a low swing-phase knee ROM during obstacle-free walking may not be able to react to an obstacle, which could result in a trip. Other factors such as kinematic variability, balance, strength and reaction time may also contribute to the inability to avoid an obstacle, or have manifestations in the observed kinematic differences between trippers and non-trippers. In particular, it is possible that the introduction of the dual task influenced the ability of participants to react to the obstacles and may not be representative of their real-world obstacle avoidance ability. Therefore, interventions designed to improve the ability to avoid obstacles may need to address multiple factors and not simply the kinematic differences between trippers and non-trippers. However, the potential role of these other factors was beyond the scope of this investigation.

5. Conclusion

In conclusion, individuals with reduced sagittal plane knee ROM combined with a delay in peak knee flexion in swing during obstacle-free walking may not be able to react to an obstacle that appears suddenly, which could result in a trip.

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Conflicts of interest

None.

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