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Technical note

Two simple methods for determining gait events during treadmill and overground walking using kinematic data

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Abstract

The determination of gait events such as heel strike and toe-off provide the basis for defining stance and swing phases of gait cycles. Two algorithms for determining event times for treadmill and overground walking based solely on kinematic data are presented. Kinematic data from treadmill walking trials lasting 20–45 s were collected from three subject populations (healthy young, $n = 7$; multiple sclerosis, $n = 7$; stroke, $n = 4$). Overground walking trials consisted of approximately eight successful passes over two force plates for a healthy subject population ($n = 5$). Time of heel strike and toe-off were determined using the two new computational techniques and compared to events detected using vertical ground reaction force (GRF) as a gold standard. The two algorithms determined 94% of the treadmill events from healthy subjects within one frame (0.0167 s) of the GRF events. In the impaired populations, 89% of treadmill events were within two frames (0.0334 s) of the GRF events. For overground trials, 98% of events were within two frames. Automatic event detection from the two kinematic-based algorithms will aid researchers by accurately determining gait events during the analysis of treadmill and overground walking.

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1. Introduction

Accurate and efficient detection of gait events is essential for the analysis of human gait. Determination of heel strike (HS) and toe-off (TO) allows walking trials to be broken up into gait cycles consisting of a stance and swing phase. This permits easy comparison of joint angles, forces and moments across multiple strides and walking trials. Analysis of gait data will often examine gait variables with reference to one or more of these gait events or phases, such as knee flexion at heel strike or knee moment at 50% of stance phase. It is critical that these events are detected accurately and consistently throughout a trial.

Researchers have used multiple experimental methods to determine gait events. Pressure sensitive foot switches have been employed to detect when a load is placed on or removed from the foot, and corresponding HS and TO can be determined [1,2]. This requires the use of additional equipment and or the modification of the subject's footwear. The technique also has limited use in subject populations with abnormal gait, especially in populations in which the subject lacks sufficient foot clearance during the swing phase. Event detection by these techniques is also dependent on appropriate and accurate placement of the sensors. Novel techniques, such as the use of miniature gyroscopes have also been used to determine the timing of gait events [3,4]. Other techniques require a force sensitive walkway, or a walkway containing multiple force plates [5,6]. This equipment can be quite expensive and is not practical for many labs.

Computational methods of event detection often rely on data from reflective marker systems where the position of

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the heel or toe marker is tracked through multiple frames. Algorithms based on the velocities and accelerations of these markers allow researchers to determine when HS and TO occur with relatively successful results [7–9]. However, these algorithms are subject to problems when walking speed or joint kinematics substantially deviate from normative values.

While many of these studies have examined overground gait, few studies investigated event detection on a treadmill. The use of a treadmill allows for the collection of large amounts of data and gait cycles in a small volume. Because of the multitude of strides that can be collected in a single treadmill trial, auto-identification of gait events becomes extremely important. It saves clinicians a significant amount of time when processing the gait data and provides an objective and reliable approach to determine the events.

In this paper we will introduce and discuss two computational methods of determining treadmill and overground gait events from kinematic data. Both methods are relatively simple algorithms that are based on the positional changes of markers on the foot and do not involve the use of force plates or other equipment. The overall objective of this study was to evaluate the ability of the two novel algorithms to predict gait events by comparing the computationally predicted events to those detected using vertical ground reaction force (GRF). In order to evaluate the robustness of the algorithms, data was collected from a variety of subject populations including young healthy subjects, subjects with multiple sclerosis (MS) and subjects who had suffered a stroke.

2. Methods

The data used in this study were collected from subjects participating in three research projects. All subjects signed informed consent forms that had been approved by the institution's Human Subjects Review Board. A 20–45 s walking trial on the treadmill was used from the subjects in each of the adult populations (healthy young, $n = 7$; MS, $n = 7$; stroke, $n = 4$). Overground trials consisted of approximately eight successful trials of walking where the heel contacted the center of the force plate and was tested only in healthy subjects ($n = 5$). Kinematic marker data for heel and toe markers was collected at 60 Hz using a six camera motion analysis system (Motion Analysis Corporation, Santa Rosa, CA, USA). The subjects all walked in the direction of the lab's positive X -axis. Event detection was performed with a custom written program using LabView 7.1.

2.1. Coordinate-based treadmill algorithm

The first algorithm developed for determining events on the treadmill is based on position of a foot marker. Because of the moving walking surface, there is a characteristic

sinusoidal curve when the X coordinate of a foot marker is graphed versus time. The peaks correlate to the time at which the foot comes into contact with the treadmill belt and the valleys coincide with the initiation of swing phase, or toe-off. When the position of the foot marker is graphed relative to the sacral marker, this sinusoidal curve is also seen and oscillates about the origin. The mean value of this curve is used to set a threshold beyond which the peaks (HS) and valleys (TO) can be detected.

All peaks and valleys (i.e., HS and TO events) were found by

$$t_{\text{HS}} = (X_{\text{heel}} - X_{\text{sacrum}})_{\text{max}}, \quad t_{\text{TO}} = (X_{\text{toe}} - X_{\text{sacrum}})_{\text{min}}$$

which represents the maximal displacement of the heel and toe from the sacrum marker.

2.2. Velocity-based treadmill algorithm

On a treadmill the foot is placed on a constantly moving surface, which results in predictable changes in the position of foot markers at both heel strike and toe-off. In our lab coordinate system, the X coordinate of a heel marker changes from moving in a positive X direction during swing to a negative X direction at each heel strike. Therefore, the X component of the velocity vector at this instant changes from positive to negative as well. This frame at which the foot begins moving backward on the treadmill is labeled heel strike. As the swing phase begins, the X component of the velocity vector for the toe or heel markers changes from negative to positive. This indicates the point in time that the foot begins moving in the direction of walking and is labeled toe-off. These easily discernable changes in the direction of velocity allows for the creation of a simple and effective velocity-based algorithm to determine gait events on a treadmill.

2.3. Application to overground trials

In order to incorporate the treadmill event detection algorithms into conventional overground event calculations, the coordinate systems of each subject are modified. With treadmill walking there is little displacement of the trunk as the lower extremities move relative to a semi-fixed point near the center of mass. In order to recreate a similar situation in overground walking, the X coordinate of the sacral marker at each frame was subtracted from the X coordinate of each marker at each corresponding frame. This created the effect of markers moving relative to a fixed point on the body as opposed to the markers on the body moving relative to a fixed origin in the lab coordinate system. The result is a subject who appears to be walking in place, as if their feet were placed on a dynamic surface that moved at the speed of the subject's forward velocity. Joint angles remain the same, while the coordinates of each marker relative to the room's coordinate system are

Table 1

Percentages of computationally determined events offset by the frame error for the velocity- and coordinate-based algorithms in the healthy subject population

Frame error	RHS		RTO		LHS		LTO	
	Velocity	Coordinate	Velocity	Coordinate	Velocity	Coordinate	Velocity	Coordinate
-3	0.00	0.00	0.52	0.52	0.00	0.00	1.05	1.05
-2	0.52	16.75	1.05	4.71	0.00	10.99	0.52	3.66
-1	35.08	71.20	13.61	22.51	32.46	73.30	10.47	10.99
0	61.78	11.52	46.07	60.73	63.35	12.04	19.90	56.02
1	2.62	0.52	38.74	11.52	2.09	2.09	64.40	28.27
2	0.00	0.00	0.00	0.00	1.57	1.57	3.66	0.00
3	0.00	0.00	0.00	0.00	0.52	0.00	0.00	0.00

Ninety-four percent of the events fall within one frame of the force plate data (subject $n = 7$ and gait cycles per side $n = 191$). RHS: right heel strike; LHS: left heel strike; RTO: right toe-off; LTO: left toe-off.

changed. The coordinate- and velocity-based algorithms (described above) were applied to detect gait events from overground data.

2.4. Events from force plate data

In order to determine the validity of these algorithms, they were compared to a gold standard. Consistent with previous literature, the gold standard in this study was event detection based on force plate data [6,7,9]. The treadmill used to collect the data had dual integrated force plates capable of capturing X, Y and Z force components (Bertec Corporation, Columbus, OH, USA). For overground trials, subjects walked on a platform perpendicular to the treadmill, which allowed the force plates to collect force data during the overground trials as well. Analog force data was captured at 600 Hz. Because the treadmill motors and moving walking surface introduced a significant amount of noise into the data during treadmill trials, the force data was filtered with a zero phase shift, fourth order Butterworth filter with a cutoff frequency of 20 Hz. Heel strike was calculated at the first frame at which the vertical GRF was greater than 20 N for a width of at least 40 samples. Similarly, toe-off was determined at the first frame that was less than 20 N for a width of 40 frames. This threshold is higher than what has been previously reported because of the noise that the moving belts on the treadmill introduced into the vertical GRF. After filtering and event calculation, the frame number of each event was divided by a factor of 10 in order to match the video frame rate.

Each event determined from the two computational methods was compared to the corresponding event determined by GRF. The difference in the number of frames was then recorded. A positive value corresponded to an event that was predicted after the GRF determined event and a negative value indicates that the computational method found the event prior to the GRF event. For example, a frame error of -2 corresponds to an event that was detected by the computational method two frames prior to a GRF event. A value of zero indicates that the computational and force plate determination were the same.

3. Results

3.1. Healthy young subjects—treadmill

Table 1 shows the distribution of the actual error using each computational method for each event expressed as a percent of total trials studied. A total of 191 gait cycles for each side were used in the calculation of the percentages. The maximum error found in the healthy subject's event detection was three frames, which corresponds to a time difference of .050 s. However, this error occurred in only 0.4% of all velocity and coordinate determined events. Ninety-four percent of all events were within one frame (0.0167 s) of the GRF event. Average frame offset for the coordinate-based algorithm was -1.04 and -0.90 for the respective right and left heel strikes. The average frame offset for toe-off using the same algorithm was closer to zero with -0.022 and 0.7 for the right and left foot. The velocity-based algorithm had an average offset of -0.34 and -0.28 frames for right and left heel strike and -0.21 and 0.57 frames for right- and left toe-off (Fig. 1).

3.2. Multiple sclerosis subjects—treadmill

In order to test the algorithms in populations where gait abnormalities may affect event detection, subjects with MS were included in this investigation. Trials lasting 45 s from seven subjects with mild to moderate MS provided a total of 247 gait cycles on each leg. The results show that 75% of the events from the coordinate- and velocity-based algorithms fall within one frame from the GRF events. The velocity-based algorithm performed slightly better than the coordinate-based algorithm with 89% of the events in the ± 1 frame range compared to 67% of the coordinate-based events. In determining heel strikes, the coordinate-based algorithm calculated the heel strike an average of 2.22 and 1.95 frames after the GRF events for the right and left heel strikes, respectively (Fig. 1). Average toe-off was placed at 0.77 and 0.55 frames before the respective right and left toe-off determined from the force plates. The average frame offset for heel strike from the velocity algorithm was -0.83 and -0.24 for the right and left, respectively. Right and left offset for toe-off were found to be 0.15 and 0.47 frames. The

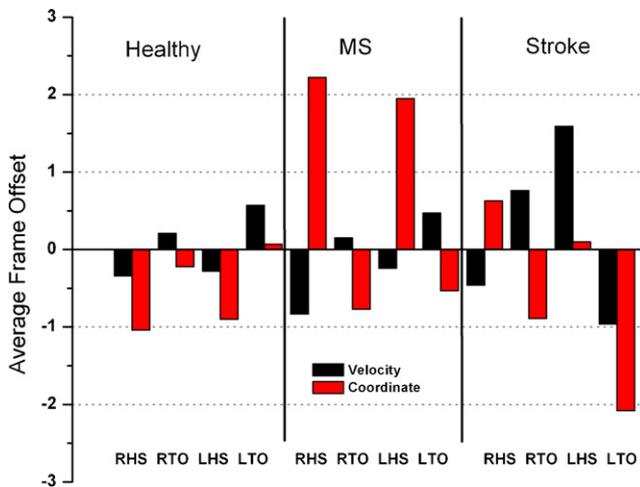


Fig. 1. Average frame offsets of the velocity- and coordinate-based algorithms for each of the subject populations.

maximum offset was found to be -6 frames (0.100 s) for the velocity algorithm and eight frames (0.134 s) for the coordinate algorithm.

3.3. Stroke subjects—treadmill

Data from four subjects who had suffered a stroke were also included in this investigation. The results from this group are based on 20 s walking trials for a total of 55 gait cycles on each side. 75.5% of the velocity-based events were within one frame from the GRF events compared to 53.6% of the coordinate-based events. The average offset for velocity-based events was -0.46 and 1.59 frames for right and left heel strikes and 0.76 and -0.96 frames for right and left toe-off. For the coordinate-based algorithm, the average offset was 0.63 and 0.10 frames for right and left heel strike and slightly larger at -0.89 and -2.08 frames for right and left toe-off, respectively (Fig. 1). The maximum offset was four frames for both the velocity- and coordinate-based algorithms which corresponds to a time difference of 0.067 s.

3.4. Healthy young subjects—overground

A total of 102 gait events were used for comparison of overground gait events. The computational methods determined all of the gait events within a range of -3 to $+2$ frames when compared to GRF determined events. The maximal offset was three frames which corresponds to .05 s. This only occurred less than 2% of all predicted events, whereas 82% of all predicted events were within one frame of the GRF derived events (85% coordinate algorithm and 80% velocity algorithm). Ninety-eight percent of all events were within two frames of the GRF determined events (100% coordinate algorithm and 96% velocity algorithm).

Events from the left- and right-side were combined for analysis since no difference was observed between sides when compared using *t*-test ($p > 0.125$ for all left and right events). On average, the coordinate algorithm detected heel

strike at 0.125 frames (0.0021 s) before the force plate and detected the average toe-off at 0.74 frames (.0123 s) after the force plate determined event. The trend of the velocity algorithm was the opposite. Determining events with this algorithm placed heel strike at 1.41 frames (0.023 s) after force plate derived events and placed toe-off at 0.50 frames (0.0083 s) before the force plate determined toe-off.

4. Discussion

The results from this study support the theory that velocity- and coordinate-based algorithms are capable of detecting critical gait events on a treadmill and overground. While event detection for the healthy unimpaired subjects correlated closest with the gold standard, the algorithms appear to be valid for use in populations where altered gait patterns are present. The average sample error with healthy subjects is similar to what has been reported by other investigators studying overground walking [2,10]. It is hard to compare our results of event detection in impaired populations to previous literature because nearly all of the previous reported event detection techniques were evaluated solely in healthy subjects without significant gait abnormalities. While we found maximal sample error in impaired subjects ranged up to eight frames (0.134 s), it is important to note the extremely low frequency with which these large errors occurred. Out of 494 combined left- and right-side gait cycles on the treadmill (1976 total predicted events), an error of that magnitude occurred only once. Many of the previous studies have examined overground walking only in healthy subjects which provides a few strides of gait data per subject. The large amount of gait cycles used in this study and small percentage of significant error in event prediction demonstrates the ability of the two algorithms to calculate gait events even in the presence of altered gait patterns on the treadmill.

The errors for overground event detection are equivalent to, or smaller than errors reported by other authors. Mickelborough et al. [10] determined that accuracy of the kinematic algorithm used in their study is to be within approximately 0.03 s. Hreljac and Marshall [7] performed a study using a kinematic-based algorithm for determining events and found offset times similar to what we found. However, they had a small sample size of two subjects and found average error of heel strike and toe-off to be 0.0047 and 0.0056 s at a variety of speeds. Hansen et al. [6] found the average error to be 0.0083 s for heel strike and 0.0167 s for toe-off which is slightly higher than the error we observed with the coordinate-based algorithm. Their algorithm required both kinematic data and the center of pressure from a force plate. Other studies on event determination have been performed that show similar average offset times, but rely on the use of pressure sensitive walkways, shoe inserts or forceplates [2,8]. The algorithms used in this study require only kinematic data and predict 82% of overground events within one frame and 98% of overground events within two frames of GRF derived events.

These methods eliminate the need for force plates or pressure sensitive walkways and require only a motion analysis system to collect kinematic data.

Because walking patterns differ dramatically in impaired populations, it is difficult to develop an algorithm that works in a uniform fashion for gait patterns that are not homogeneous. Some of the error recorded with the stroke population may be attributed to a shuffling gait pattern and insufficient foot clearance during swing. Determination of heel strike in this situation may require a refined definition of heel strike. The definition offered by Perry [11] defines heel strike (initial contact) as the “moment when the foot just touches the floor”. While this definition is appropriate for normal gait, it does not describe the typical heel strike in many impaired populations. During a scuff in late swing phase, there may be measurable forces because the foot has come in contact with the ground, but the maximal forward position of the foot has not been achieved. Perry [11] also describes heel strike (initial contact) as the time at which “the joint postures... determine the limb’s loading response pattern”. Determining heel strike during a scuff does not describe the time at which the lower extremity prepares for weight acceptance. In the impaired populations, there are significant idiosyncrasies with the conventional definition of heel strike. We feel that heel strike determined by the coordinate-based algorithm may provide the most valid representation of heel strike in these subject populations.

There are similar difficulties with toe-off detection. Foot movement in the positive X direction may not be coincident with zero GRF due to weakness or other problems associated with the impairment. This lends itself to some error that may be found in using the force plates as a gold standard in investigating the event detection in these populations. Consistent with this theory, our results for the stroke group and MS group using the coordinate algorithm follow the trend of a late detection for heel strike and early detection of toe-off. We believe that the algorithms demonstrated accurate detection of events in this population, even though this may not have been fully exemplified by the results. The subsequent use of visual inspection may provide more accurate accounts of gait events and decrease some of the error associated with using the GRF as a gold standard in the impaired population.

Another source of error in using the force plate as a gold standard is the relatively large threshold required for the vertical GRF. While this larger threshold was unavoidable because of the noise introduced by the moving belts, it is likely a source of error in the results of the healthy subject group. In this population, the coordinate algorithm tended to predict the majority of heel strikes one frame prior to the GRF predicted heel strike. This likely reflects the delay in GRF event prediction secondary to a threshold of 20 N. If the threshold was lowered, it is likely that the GRF events in healthy subjects would have been determined closer to the coordinate- and velocity-based detection of heel strike.

The use of these kinematic algorithms to calculate gait events is beneficial to investigators because of the need for

limited equipment. These algorithms can be used in any situation where kinematic data is being collected. It does not require the use of pressure sensitive foot switches or expensive force plates. Gait events will be detected even if the subject lacks sufficient foot clearance during swing or does not bear full body weight on the support limb in stance phase.

The new computational methods of event detection provide researchers with an accurate means of determining event times from kinematic data. Although the frame offsets for computationally determined events may seem high in the impaired populations, we feel that this may be representative of the problems with using the GRF as a gold standard. As the use of a treadmill becomes more prudent in motion analysis labs for clinical, research or training purposes, the implementation of these algorithms will provide useful information and efficient, automatic event detection. The application in overground data collections will also allow for automatic event detection and decrease both human error and time of analysis that often is associated with visual inspection of gait events.

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