Algorithms to determine event timing during normal walking using kinematic data

Alan Hreljac\textsuperscript{a,}*, Robert N. Marshall\textsuperscript{b}

\textsuperscript{a}Department of Kinesiology and Health Science, California State University, Sacramento, 6000 J Street, Sacramento, CA 95819-6073, USA
\textsuperscript{b}Sport and Exercise Science Department, University of Auckland, Auckland, New Zealand

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Abstract

Algorithms to predict heelstrike and toeoff times during normal walking using only kinematic data are presented. The accuracy of these methods was compared with the results obtained using synchronized force platform recordings of two subjects walking at a variety of speeds for a total of 12 trials. Using a 60 Hz data collection system, the absolute value errors (AVE) in predicting heelstrike averaged 4.7 ms, while the AVE in predicting toeoff times averaged 5.6 ms. True average errors (negative for an early prediction) were +1.2 ms for both heelstrike and toeoff, indicating that no systematic errors occurred. It was concluded that the proposed algorithms provide an easy and reliable method of determining event times during walking when kinematic data are collected, with a considerable improvement in resolution over visual inspection of video records, and could be utilized in conjunction with any 2-D or 3-D kinematic data collection system.

Keywords: Walking; Heelstrike; Toeoff; Contact time

1. Introduction

In gait analysis, a stride is generally subdivided into a stance and a swing phase. The accurate identification of these phases requires precise knowledge of heelstrike and toeoff times. When force plate data are collected during the analysis, these event times could be estimated accurately by determining the sample at which the vertical force rises above zero (or some threshold level) for heelstrike, and falls below that level again for toeoff. This method is excellent for recording a single heelstrike and toeoff event during overground gait in a laboratory situation, but it would be necessary for a laboratory to be equipped with two force platforms (or a large force platform) to determine the temporal components of a complete stride. Outside of a laboratory setting or on a treadmill, the use of a force platform to measure these parameters is generally not possible without specialized equipment. Pressure sensitive switches, placed inside or outside the shoe, at the heel and toe, have often been used as an alternative to determine the onset of the stance and swing phases (Abernethy et al., 1995; Cavanagh et al., 1983; Grillner et al., 1979). Assuming that the foot switches are properly positioned, and synchronized with other data collection procedures, accurate data regarding heelstrike and toeoff timing could be obtained. Other techniques that have been utilized to determine the time of events during gait include mounting a rubber tube instrumented with a pressure transducer to the foot or shoe (Nilsson et al., 1985), and the use of a photocell contact mat (Viitasalo et al., 1997). Although fairly accurate, these methods require special equipment which is not readily available to most researchers.

When researchers are interested in collecting kinematic data only, or when the necessary instrumentation is not available to determine the timing of heelstrike and toeoff, it is possible to rely upon visual inspection of video or film data to determine the timing of events (e.g. Chengzhi and Zongcheng, 1985; Mann and Hagy, 1980). The maximum resolution that could be expected with this method is limited by the sampling rate. The inherent inaccuracy of the visual inspection method due to imperfections (such as blur) in the film or video records, along with the processing time required to perform these inspections for a large number of trials makes this method difficult to implement, and sometimes impractical.
problem of event recognition becomes more difficult to solve when employing kinematic data collection systems (such as optoelectric systems) which do not produce video or film records. The purpose of this investigation was to present new algorithms for determining heelstrike and toecoff times during normal walking using only available kinematic data, and to evaluate the accuracy of the proposed algorithms by comparing results to the “gold standard” of force platform recordings.

2. Methods

Two healthy subjects (one male, one female), wearing their own athletic shoes, walked down a 10 m walkway over a floor mounted force platform six times each at a variety of self-selected speeds, ranging from slow to fast while the motion of two reflective markers, placed on the heel and toe (fifth metatarsal) of the left shoe (dominant side for one subject, non-dominant for the other) were recorded simultaneously by a system of four Motion Analysis Falcon cameras (60 Hz). Walking speed was neither monitored nor controlled. A successful trial was considered to be one in which the subject did not make any noticeable alterations in stride length during the trial (i.e. no targeting), and contacted the force platform with the entire landing (left) foot. Ground reaction force (GRF) data were collected at a frequency of 600 Hz. Synchronized data collection was initiated when subjects were approximately 1 m in front of the force platform to accommodate the calculation of the force threshold, and to produce endpoint data frames required to improve the accuracy of subsequent data smoothing procedures. Subjects reported no history of neuromuscular pathology, and were free of musculoskeletal injuries at the time of testing. Subjects signed informed consent forms, which reiterated the basic procedures and intent of the study, as well as warning of any potential risks involved as a result of participation.

A standard direct linear transformation (DLT) method was used to reconstruct three-dimensional coordinate data of the two markers. The coordinate system was oriented so that the positive x-direction was in the direction of forward movement, the positive z-direction was vertically upward, and the positive y-direction was to the subject’s left. Raw 3-D coordinate data were smoothed using a fourth-order, zero lag, Butterworth filter, with optimal cutoff frequencies uniquely chosen for each coordinate of each marker using the residual method of Wells and Winter (1980). Derivatives were calculated using finite difference equations.

Two methods were used to determine the time of heelstrike (HS) and toecoff (TO) during each trial. Contact time (T) was calculated from these values. In the force platform (FP) method, considered to be the “true” representation of the contact timing events, HS was considered to occur during the sample at which the vertical (z) component of the GRF rose above 10 N, while TO was considered to occur during the sample at which the z-component of GRF fell below the 10 N threshold. The second method of determining HS and TO times utilized two algorithms based upon the derivatives of the marker positions. In the HZACC algorithm, the time of HS (tHS) was estimated to occur at the time of a local maximum in the horizontal component of acceleration (ax) of the heel marker. The actual maximum value of ax generally occurred between discrete data frames, a linear interpolation equation (Eq. (1)) was used to estimate the actual time (including fraction of a data frame) that the maximum acceleration (zero jerk) occurred.

\[ t_{\text{HS}} = t_1 + \left( \frac{J(t_1)}{J(t_1) - J(t_2)} \right) t_{\text{INT}}, \]  

where \( t_1 \) is the time of the last positive value of the vertical component of jerk prior to the jerk curve crossing zero, occurring at either the data frame of maximum ax or the frame prior to maximum ax, \( t_2 \) is the time of the first negative value of jerk after the jerk curve crosses zero, occurring at either the frame of maximum ax or the frame following maximum ax, \( J(t_1) \) is the value of jerk (vertical component) at frame \( t_1 \), \( J(t_2) \) is the value of jerk at frame \( t_2 \), and \( t_{\text{INT}} \) is the time interval between frames, which is 16.7 ms for 60 Hz data collection.

The TXACC algorithm made use of a similar equation to calculate the time of TO (tTO), estimated to occur at the local maximum of the horizontal component of acceleration (ax) of the toe marker. A similar linear interpolation method (Eq. (2)) was used to determine the time of maximum ax (value of the horizontal component of jerk equal to zero), and thus the time of TO is

\[ t_{\text{TO}} = t_1 + \left( \frac{J(t_1)}{J(t_1) - J(t_2)} \right) t_{\text{INT}}, \]  

where jerk values refer to the horizontal component of the toe, and times relate to events close to \( t_{\text{TO}} \).

After determining \( t_{\text{HS}} \) and \( t_{\text{TO}} \) using these algorithms, contact time (T) for each trial was calculated. Based upon the assumption that the FP method produced actual representations of event times, the errors in estimating \( t_{\text{HS}}, t_{\text{TO}}, \) and \( T \) were determined. True error (TE) was defined as the arithmetic difference between predicted event times (using the algorithms) and actual event times determined by the FP method. A negative value of TE indicated that the predicted event time preceded the actual time. Absolute value errors (AVE), calculated as the absolute value of TEs for individual trials, were indicative of the magnitude of error that each algorithm produced, regardless of the direction. Simple two-tailed
3. Results

No significant differences in $t_{HS}$, $t_{TO}$, or $T$ were found between the algorithm methods and the FP method ($p > 0.15$). The average AVE in estimating $t_{HS}$ was 4.7 ms, with a maximum error of 13.9 ms (Table 1). The average AVE in estimating $t_{TO}$ was 5.6 ms (maximum error of 11.5 ms). The average AVE in estimating $T$ was 5.8 ms (maximum error of 14.7 ms). The average TE for predicting both $t_{HS}$ and $t_{TO}$ was 1.2 ms, with an average TE of 0.0 ms in the prediction of $T$.

The vertical component of the heel marker displacement, acceleration, and jerk are shown for a representative trial to illustrate $t_{HS}$ determination (Fig. 1). Similar curves for the horizontal component of the toe marker could be used to illustrate how $t_{TO}$ was determined.

4. Discussion

The results of this study verified that the proposed algorithms provide accurate information regarding heelstrike, toeoff, and contact times for walking over a range of speeds. The validation of the algorithms has been limited to normal walking. With average errors in estimating event times ranging between 4.7 and 5.8 ms, the expected results using these algorithms together with a 60 Hz data collection system are comparable to what may be expected with visual inspection using a 200 Hz system. There did not appear to be any change in the accuracy of the algorithms as speed varied, although no statistical comparisons were made between speeds (Table 1).

The algorithms were tested using a four camera, three-dimensional kinematic data collection system, but could be implemented with any properly positioned single camera, two-dimensional system since they are based upon the vertical component of the heel marker and the horizontal component of the toe marker, respectively. Since the implementation of these algorithms requires no special equipment, they may be utilized in any setting in which kinematic data would normally be collected, including on a treadmill and outdoors. If the method were to be used in conjunction with another data collection system (such as an EMG system), however, it would be necessary to assure that a means of synchronizing the systems was available.

The algorithms presented in this study compare favorably to other methods of determining gait event times using more complex instrumentation. Nilsson et al. (1985) reported errors of 3.9 and 4.2 ms in estimations of HS time during walking at two different speeds, and errors of 2.5 and 6.2 ms in the estimation of TO at the same speeds, while using a specially designed contact device which consisted of “a monolithic pressure transducer ... attached to one end of a flexible silicone rubber tube ...” which was then fastened to a subject’s foot or
shoe. Comparable errors were found in the present study using no special instrumentation. Event times found by Nilsson et al. (1985) always lagged behind force platform responses, while event times calculated with the algorithms in the present study showed no systematic errors. Results using a photocell mat method (Viitasalo et al., 1997) were presented only during running with errors ranging from 3.3 to 47.1 ms (late) for the estimation of HS, and from 11.0 to 37.5 ms (early) in the estimation of TO, with calibration corrections presented which improved the accuracy considerably. Although validated only for running, the errors in estimating event times using this fairly complex system were greater than the errors found in the present study in which equipment beyond a single camera is required. Stanhope et al. (1990) used a kinematic model based upon ankle position data, in conjunction with force platform records to predict the event times for several subsequent walking strides. Using a 50 Hz data collection system, these authors reported that errors in predicted event times were greater than 20 ms in over 20% of the cases. In the present study, which incorporated a 60 Hz data collection system, the maximum error in predicting any event time was less than 15 ms.

The validity of the HZACC algorithm is based upon the fact that the heel marker reaches a local maximum in $a_z$ as the heel contacts the ground. Since the maximum $a_z$ of the heel marker always precedes the minimum vertical displacement of the heel (by approximately 2–10% of stride time, depending upon walking speed), this event could be used as a starting point to locate the local maximum in $a_z$ that represents heelstrike. As is the case when calculating any derivative of kinematic data, results are critically dependent upon marker placement and data smoothing. In the trial that produced the worst results in this study, there were several missing data frames for the heel prior to the calculated time of heelstrike, leading to minor endpoint smoothing errors, thus contributing to the results being less accurate for this trial.

The accuracy in predicting toeoff time using the TXACC algorithm demonstrates that the toe marker (at the fifth metatarsal) accelerates maximally in the horizontal direction as the toe leaves the ground. The toeoff always follows the minimum vertical displacement of the toe marker by approximately 10–15% of stride time, allowing this event to be used as a starting point in determining the time of the local maximum in the horizontal component of the toe marker acceleration.

The general form of the algorithms presented in this study could be used in a variety of situations to improve the resolution in the estimation of any event timing that may be obtained from kinematic data. As an example, if the time of a maximum (or minimum) joint angle is important to a researcher, a similar algorithm as was used in this study could improve the resolution beyond that of a single frame of data by determining the fraction of a frame at which the angular velocity curve was equal to zero (true time of maximum or minimum).

The algorithms presented provide an easy and accurate method to calculate event times during kinematic data collection of normal walking. Any number of consecutive stride events could be measured using these algorithms, which would be particularly useful when subjects are walking on a treadmill. The resulting errors in estimating heelstrike and toeoff times compare favorably to other methods which require specialized equipment, while the present method could be implemented with any two- or three-dimensional kinematic data collection system. The resolution obtained in estimating event times with these algorithms is considerably greater than the resolution of the data collection system used.

References


