Phase determination during normal running using kinematic data

A. Hreljac[1] N. Stergiou[2]

[1]Department of Kinesiology & Health Science, California State University, Sacramento, USA
[2]University of Nebraska at Omaha, Nebraska, USA

Abstract—*Algorithms to predict heelstrike and toe-off times during normal running at subject-selected speeds, using only kinematic data, are presented. To assess the accuracy of these algorithms, results are compared with synchronised force platform recordings from ten subjects performing ten trials each. Using a single 180 Hz camera, positioned in the sagittal plane, the average RMS error in predicting heelstrike times is 4.5 ms, whereas the average RMS error in predicting toe-off times is 6.9 ms. Average true errors (negative for an early prediction) are +2.4 ms for heelstrike and +2.8 ms for toe-off, indicating that systematic errors have not occured. The average RMS error in predicting contact time is 7.5 ms, and the average true error in predicting contact time is 0.5 ms. Estimations of event times using these simple algorithms compare favourably with other techniques requiring specialised equipment. It is concluded that the proposed algorithms provide an easy and reliable method of determining event times during normal running at a subject selected pace using only kinematic data and can be implemented with any kinematic data-collection system.*

Keywords—*Running, Heelstrike, Toe-off, Contact time*

Med. Biol. Eng. Comput., 2000, **38**, 503–506

1 Introduction

AN ESSENTIAL aspect of most gait analyses is the accurate estimation of event times such as heelstrike and toe-off. This information is necessary to subdivide a stride into stance and swing periods, regardless of the type of datum being collected, and is often required to make meaningful comparisons between subjects and studies possible.

In experiments in which a force platform is utilised, the times of a single heelstrike and toe-off event can be determined accurately, but, if the temporal components of one or more complete strides are required to be measured, it is necessary to use alternative methods of determining phase durations, unless a laboratory is equipped with large or multiple force platforms. In experiments conducted outside a laboratory setting or on a treadmill, the accurate measurement of temporal components is generally not possible without specialised equipment.

One alternative to the force platform commonly used for determining the onset of stance and swing phases during gait involves the placing of pressure-sensitive foot switches on the shoe or foot (LIGGINS and BOWKER, 1991; MINNS, 1982; ROSS and ASHMAN, 1987). Relatively accurate determination of heelstrike and toe-off times can be obtained during walking with these simple devices, provided the foot switches are properly positioned and a predetermined offset time is taken into account (HAUSDORFF *et al.*, 1995).

Correspondence should be addressed to Dr A. Hreljac; email: ahreljac@hhs4.hhs.csus.edu

First received 28 April 2000 and in final form 26 June 2000

MBEC online number: 20003505

© IFMBE: 2000

Other specialised techniques that have been utilised to determine these timing parameters during gait include the use of an instrumented walkway (CROUSE *et al.*, 1987; GIFFORD and HUGHES, 1983), the mounting of a rubber tube, connected with a pressure transducer, on the foot or shoe (NILSSON *et al.*, 1985) and the use of a photocell contact mat (VIITASALO *et al.*, 1997). Although reasonably accurate, these techniques require equipment that is not typically available to most researchers.

In situations in which researchers are only interested in, or limited to, kinematic data collection, relatively few options exist for the determination of phase timing. In these situations, researchers may be required to rely upon visual inspection of video records to determine the times of heelstrike and toe-off (e.g. MANN and HERMAN, 1985; VILENSKY and GEHLSEN, 1984). The accuracy of this time-consuming process is limited by the sampling frequency and the quality of the video recording. The problem of phase determination is further exacerbated when opto-electric systems are utilised for data collection, as video records are not obtained with these systems.

Given the fact that kinematic patterns of walking are relatively consistent from stride to stride and between speed conditions (WINTER, 1987), researchers have been able accurately to determine temporal components of the walking stride of horses (PEHAM *et al.*, 1999) and humans (HRELJAC and MARSHALL, 2000) using only kinematic data over a range of speeds.

Kinematic patterns of running are generally consistent from stride to stride, but these patterns have been shown to vary with speed (MANN and HAGY, 1980; MANN *et al.*, 1986), suggesting that a kinematically based model of predicting phase times during running should be speed dependent. A speed condition that is utilised in much running-related research is a self-selected pace (e. g. SHIAVI *et al.*, 1981; STERGIOU *et al.*, 1999).

The purpose of the present investigation was to evaluate the accuracy of algorithms designed to predict heelstrike and toe-off times during normal (heelstrike) running at subject-selected speeds, using only kinematic data. The accuracy of the event time predictions was evaluated by comparing results with those determined from force platform recordings.

2 Methods

Ten young (23.5 ± 2.6 years), healthy, physically active subjects (four males, six females), wearing their own running footwear, ran at self-selected speeds down a 25 m runway, over a floor-mounted force platform, upon which subjects landed with their right foot. All subjects exhibited a heelstrike pattern at the test speed.

Ten successful trials, 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 (right) foot, were completed by each subject. The motion of four reflective markers placed on the knee joint centre, lateral malleolus, calcaneus and head of the fifth metatarsal of the landing leg (Fig. 1) were recorded in the sagittal plane with a single video camera (180 Hz) for at least ten frames prior to heelstrike and after toe-off of each trial.

Two-dimensional (2D) kinematic data were synchronised with ground reaction force (GRF) data (900 Hz). The raw 2D co-ordinate data were smoothed using a fourth-order, zero-lag Butterworth filter, with optimum cutoff frequencies uniquely chosen for each co-ordinate of each marker using the residual method (WELLS and WINTER, 1980). Segmental angles of the leg (knee to ankle) and foot (heel to toe) were calculated from the smoothed co-ordinate data. Derivatives of segment angles were calculated using finite difference equations. Counterclockwise rotations of a segment were considered to be in the positive direction (Fig. 1).

The times of heelstrike (HS) and toe-off (TO) were first determined from force platform (FP) recordings, regarded as true representations of contact timing events. In this FP method, HS was considered to occur during the sample in which the vertical (y) component of the GRF rose above a threshold level of 10 N, whereas TO was considered to occur during the sample in which the y-component of the GRF fell below the 10 N threshold. True contact time T was calculated from these values. Predictive algorithms, based upon calculated derivatives of segment angular motion were then applied for estimation of HS and TO times. The accuracy of the predictive algorithms was assessed by comparing results with those obtained from FP recordings.

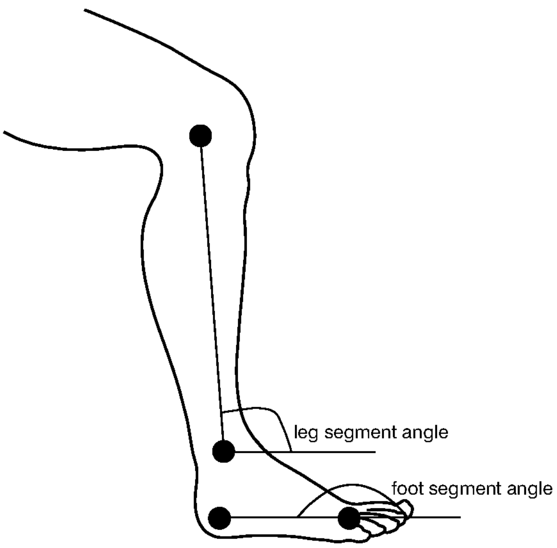

Fig. 1 *Lateral view of right leg, showing marker locations and segment angles*

The minimum foot angular acceleration α_{foot} was used as the criterion to estimate the time of HS t_{HS}. In this minimum α_{foot} algorithm, t_{HS} was predicted to occur at the time of a minimum (maximum in the clockwise direction) of the foot segment angular acceleration. As with all maxima and minima of curves, the actual minimum value of α_{foot} occurred when the derivative curve (jerk) was equal to zero. As the true minimum of α_{foot} generally occurred between discrete data frames, a linear interpolation equation (eqn 1) was used to estimate the actual time that α_{foot} occurred.

$$t_{HS} = t_1 + \left(\frac{J(t_1)}{J(t_1) - J(t_2)}\right) t_{int} \tag{1}$$

where t_1 is the time of the last negative value of the foot segment angular jerk prior to the jerk curve crossing zero, occurring at either the data frame of minimum α_{foot} or the frame prior to minimum α_{foot}; t_2 is the time of the first positive value of the foot angular jerk after the jerk curve crosses zero, occurring at either the frame of minimum α_{foot} or the frame following minimum α_{foot}; $J(t_1)$ is the value of the foot segmental jerk at frame t_1; $J(t_2)$ is the value of jerk at frame t_2; and t_{int} is the time interval between frames (5.56 ms for 180 Hz data collection).

The criterion algorithm used to predict the time of toe-off t_{TO} was a local minimum in the leg segment angular acceleration α_{leg}. As with the algorithm used to predict t_{HS}, the minimum of α_{leg} was assumed to occur at the point where the leg segment angular jerk curve was equal to zero. A linear interpolation equation similar to eqn 1 was used to estimate the fraction of a frame in which t_{TO} occurred. Predicted contact time T was calculated as the time period between predicted t_{HS} and t_{TO}.

Errors in predicting t_{HS}, t_{TO}, and T were calculated in two ways. Directional errors were determined by calculating the true error (TE), defined as the arithmetical difference between predicted event times and actual event times. A negative TE in t_{HS} or t_{TO} indicated that the predicted event time preceded the actual event time. A negative TE in T indicated that contact time was underestimated. Root mean square (RMS) errors were indicative of the magnitude of error, regardless of the direction. After calculating true and RMS errors for each trial, average true and RMS errors and maximum RMS errors were determined for each subject.

3 Results

Average values of true and RMS errors in predicting each of the event times are shown for subjects, individually and collectively, in Table 1. The maximum RMS error of any single trial in predicting each of the event times is also shown in Table 1. The average TE in predicting t_{HS} was 2.5 ms, and the average RMS error was 4.5 ms. The maximum error in predicting t_{HS} was 14.5 ms. The average TE in predicting t_{TO} was 2.8 ms, and the average RMS error was 6.9 ms. The maximum error in predicting t_{TO} was 18.8 ms. The average TE in predicting contact time was 0.5 ms, with an average RMS error of 7.5 ms and a maximum error of 28.7 ms.

Fig. 2 illustrates a representative curve of α_{foot} for a time period from 50 ms prior to HS until the time of TO. The estimation of t_{HS} occurs when this curve reaches a minimum value, as illustrated. Fig. 3 shows a representative curve of α_{leg} for a time period from HS to 50 ms after TO. The estimation of t_{TO} occurs at a local minimum of this curve, as illustrated.

Table 1 Average TEs and RMS errors (in ms) in estimating heelstrike (HS), toe-off (TO) and contact times (T) for individual subjects. For each subject, n = 10

Subject	HS		TO		T	
	TE	RMS	TE	RMS	TE	RMS
1	0.5	2.6	−0.9	2.8	−1.3	4.1
2	3.8	4.1	12.3	12.3	8.5	8.7
3	12.3	12.3	−5.1	9.2	−17.4	18.6
4	−1.4	3.9	3.5	10.9	4.9	12.5
5	1.6	3.0	0.9	5.7	−1.1	3.8
6	−2.4	2.9	−2.5	6.6	−0.1	7.8
7	4.6	4.6	11.8	11.8	7.4	7.4
8	0.7	2.5	2.8	2.8	2.0	3.5
9	2.1	5.0	3.3	4.4	1.9	5.1
10	2.2	3.9	1.7	2.1	−0.2	3.3
Average	2.4	4.5	2.8	6.9	0.5	7.5
Maximum		14.5		18.8		28.7

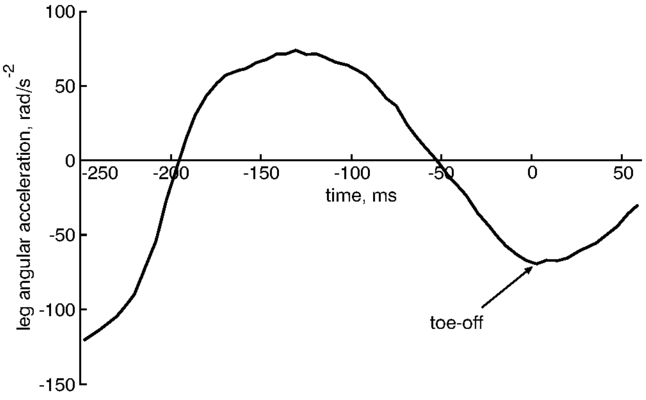

Fig. 3 *Representative curve of leg angular acceleration α_{leg} against time from heelstrike to 50 ms after toe-off. Toe-off occurs at time of local minimum of α_{leg}*

4 Discussion

The results of this study verified that the proposed algorithms provide accurate information regarding heelstrike, toe-off and contact times during normal heelstrike running at a subject-selected pace. The small value of the true errors in estimating each of the event times (<3.0 ms) demonstrates that errors are generally random, although, for some subjects, errors did appear to be directional (Table 1), indicated by the average TE equalling the average RMS error. Errors in the prediction of t_{HS} (4.5 ms) were smaller than errors in predicting t_{TO} (6.9 ms), which could be partly owing to the fact that the minimum of the α_{foot} curve (Fig. 1) is a more distinct peak than the local minimum in the α_{leg} curve (Fig. 2), thereby producing less uncertainty in the estimated time at which this minimum occurs. Another possible reason for errors in the prediction of t_{HS} being smaller than errors in the prediction of t_{TO} involves the setting of a 10 N vertical force threshold to determine when contact was made. As there was a rapid rise in the vertical force reading at heelstrike, setting a 10 N threshold would not have affected the determination of t_{HS}. Because the dropoff of the vertical force reading at toe-off was relatively gradual, setting a 10 N threshold could have had an effect on the estimation of the t_{TO} during some trials.

The algorithms presented in this study compare favourably with other techniques for determining gait event times that utilise more complex instrumentation. In a study that estimated event times at three different running speeds using a photocell-mat method (VIITASALO *et al.*, 1997), errors in estimating the time of HS ranged from 3.3 to 47.1 ms, with all predicted times following the true time of HS, whereas errors in estimating TO times ranged from 11.0 to 37.5 ms, with all estimations preceding the true time of TO. Even after improving the accuracy of the contact-mat method by implementing various correcting regression equations, the errors in estimating event times using this fairly complex system were greater than the errors found in the present study, in which no equipment apart from a single camera is required. Using a simple foot switch technique and adjusting by a predetermined offset time, a group of researchers (HAUSDORFF *et al.*, 1995) determined heelstrike times within ± 10 ms and toe-off times within ± 22 ms. These values are only slightly greater than the errors found in the present study, although subjects do not require wiring using the present technique.

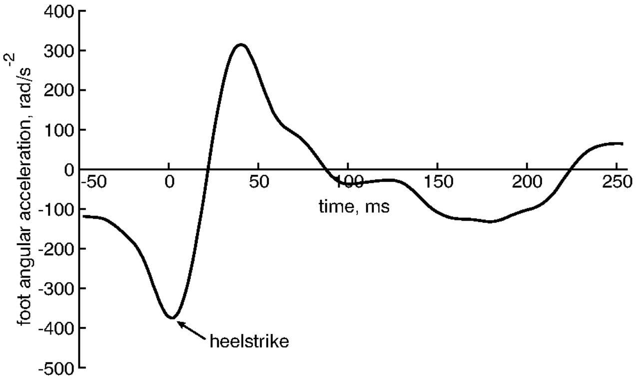

Fig. 2 *Representative curve of foot angular acceleration α_{foot} against time from 50 ms before heelstrike to toe-off. Heelstrike occurs at time of minimum value of α_{foot}*

Researchers (STANHOPE *et al.*, 1990) who used a kinematic model based upon ankle position data in conjunction with force platform records to predict the event times of subsequent walking strides reported that errors in predicted event times were greater than 20 ms in over 20% of the cases. In the present study, the maximum error in predicting either heelstrike or toe-off time was less than 20 ms.

NILSSON *et al.* (1985) presented a technique for predicting heelstrike and toe-off times during walking that required a specially designed contact device consisting of, 'a monolithic pressure transducer ... attached to one end of a flexible silicone rubber tube ...' fastened to a subject's foot or shoe. These authors reported RMS errors of 3.9 ms and 4.2 ms in estimations of HS time during walking at two different speeds, and errors of 2.5 ms and 6.2 ms in the estimation of TO at the same speeds. Comparable errors during running were found in the present study using no special instrumentation. Event times found by NILSSON *et al.* (1985) always lagged behind force platform responses, whereas event times calculated with the algorithms in the present study showed no systematic errors.

In a recent study in which kinematic data were used to predict event times during walking (HRELJAC and MARSHALL, 2000), predictions of heelstrike times were within ± 4.7 ms, and predictions of toe-off times were within ± 5.6 ms, using only a 60 Hz data collection system. Slightly greater errors were found in the present study, even though a 180 Hz data collection system was utilised. It appears that a greater variability exists in the kinematic patterns of running than of walking.

The algorithms presented provide an easy and accurate method to calculate event times during kinematic data collection of heelstrike running at subject selected speeds. As the implementation of these algorithms requires no special equipment, they can be utilised in any setting in which kinematic data are normally collected, including on a treadmill and outdoors. Any number of consecutive stride events can be measured using these algorithms. The resulting errors in estimating gait event times compare favourably with those of other techniques requiring specialised equipment, whereas the present method can be implemented solely with any 2D or 3D kinematic data collection system.

Acknowledgment—The authors would like to thank Shane Scholten, Estelle Abdala, Kyle Patterson, Gerson Brandalesi and Michael Garrett for their assistance with the data collection and digitising process in this project.

References

CROUSE, J., WALLS, J. C., and MARBLE, A. E. (1987): 'Measurement of the temporal and spatial parameters of gait using a microcomputer based system', *J. Biomed. Eng.*, **9**, pp. 64–68

GIFFORD, G., and HUGHES, J. (1983): 'A gait analysis system in clinical practice', *J. Biomed. Eng.*, **5**, pp. 297–301

HAUSDORFF, J. M., LADIN, Z., and WEI, J. Y. (1995): 'Footswitch system for measurement of the temporal parameters of gait', *J. Biomech.*, **28**, pp. 347–352

HRELJAC, A., and MARSHALL, R. N. (2000): 'Algorithms to determine event timing during normal walking using kinematic data', *J. Biomech.*, **33**, pp. 783–786

LIGGINS, A. B., and BOWKER, P. (1991): 'A simple low cost footswitch', *J. Biomed. Eng.*, **13**, pp. 87–88

MANN, R. A., and HAGY, J. (1980): 'Biomechanics of walking, running, and sprinting', *Am. J. Sports Med.*, **8**, pp. 345–350

MANN, R., and HERMAN, J. (1985): 'Kinematic analysis of Olympic sprint performance: Men's 200 meters', *Int. J. Sport Biomech.*, **1**, pp. 151–162

MANN, R. A., MORAN, G. T., and DOUGHERTY, S. E. (1986): 'Comparative electromyography of the lower extremity in jogging, running, and sprinting', *Am. J. Sports Med.*, **14**, pp. 501–510

MINNS, R. J. (1982): 'A conductive rubber footswitch design for gait analysis', *J. Biomed. Eng.*, **4**, pp. 328–330

NILSSON, J., STOKES, V. P., and THORSTENSSON, A. (1985): 'A new method to measure foot contact', *J. Biomech.*, **18**, pp. 625–627

PEHAM, C., SCHEIDL, M., and LICKA, T. (1999): 'Limb locomotion-speed distribution analysis as a new method for stance phase detection', *J. Biomech.*, **32**, pp. 1119–1124

ROSS, J. D., and ASHMAN, R. B. (1987): 'A thin foot switch', *J. Biomech.*, **20**, pp. 733–734

SHIAVI, R., CHAMPION, S., FREEMAN, F., and GRIFFIN, P. (1981): 'Variability of electromyographic patterns for level-surface walking through a range of self-selected speeds', *Bull. Prosth. Res.*, **18**, pp. 5–14

STANHOPE, S. J., KEPPLE, T. M., MCGUIRE, D. A., and ROMAN, N. L. (1990): 'Kinematic-based technique for event time determination during gait', *Med. Biol. Eng. Comput.*, **28**, pp. 355–360

STERGIOU, N., BATES, B. T., and JAMES, S. L. (1999): 'Asynchrony between subtalar and knee joint function during running', *Med. Sci. Sports Exerc.*, **31**, pp. 1645–1655

VIITASALO, J. T., LUHTANEN, P., MONONEN, H. V., NORVAPALO, K., PAAVOLAINEN, L., and SALONEN, M. (1997): 'Photocell contact mat: a new instrument to measure contact and flight times in running', *J. Appl. Biomech.*, **13**, pp. 254–266

VILENSKY, J. A., and GEHLSEN, G. (1984): 'Temporal gait parameters in humans and quadrupeds: How do they change with speed?', *J. Hum. Mov. Stud.*, **10**, pp. 175–188

WELLS, R. P., and WINTER, D. A. (1980): 'Assessment of signal and noise in the kinematics of normal, pathological, and sporting gaits'. Proc. Special Conf. Canadian Society for Biomechanics, University of Western Ontario, London, pp. 92–93

WINTER, D. A. (1987): 'The biomechanics and motor control of human gait', (University of Waterloo Press, Waterloo)

Authors' biographies

ALAN HRELJAC is currently an Assistant Professor of Biomechanics at California State University, Sacramento. He received his BSc degree in Physics from the University of Waterloo prior to studying Biomechanics at San Diego State University, where he received his MSc degree. His doctoral studies in Biomechanics were completed at Arizona State University. He completed a post-doctoral fellowship at the University of Auckland. His primary areas of research have been in the realm of gait mechanics, with special interests in gait transitions and running injuries.

NICK STERGIOU received his BSc degree in Physical Education from Aristotle University of Thessaloniki before going to the USA to study Exercise Science at the University of Nebraska at Omaha, where he earned his MSc degree. Subsequently, he received his PhD in the field of Biomechanics from the University of Oregon. His primary research interests are related to lower-extremity joint functions during gait. He is currently an Assistant Professor of Biomechanics at the University of Omaha at Nebraska.