2000

Phase determination during normal running using kinematic data

Alan Hreljac
California State University - Sacramento

Nikolaos Stergiou
University of Nebraska at Omaha, nstergiou@unomaha.edu

Phase Determination During Normal Running Using Kinematic Data

Alan Hreljac[1] and Nick Stergiou[2]

[1]California State University, Sacramento

[2]University of Nebraska at Omaha

Corresponding author: Alan Hreljac, Ph.D.

Department of Kinesiology and Health Science

California State University, Sacramento

6000 J Street

Sacramento, CA 95819-6073

Telephone: (916)278-5411

Fax: (916)278-7664

E-Mail: ahreljac@hhs4.hhs.csus.edu

Abstract

Algorithms to predict heelstrike and toeoff times during normal running at subject selected speeds using only kinematic data are presented. To assess the accuracy of these algorithms, results were compared to synchronized force platform recordings of 10 subjects performing 10 trials each. Using a single 180 Hz camera, positioned in the sagittal plane, the average RMS error in predicting heelstrike times was 4.5 ms, while the average RMS error in predicting toeoff times was 6.9 ms. Average true errors (negative for an early prediction) were +2.4 ms for heelstrike and +2.8 ms for toeoff, indicating that systematic errors did not occur. Average RMS error in predicting contact time was 7.5 ms, while the average true error in contact time was 0.5 ms. Estimations of event times using these simple algorithms compare favorably to other techniques requiring specialized equipment. It was 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 could be implemented with any kinematic data collection system.

Keywords: running, heelstrike, toeoff, contact time

Introduction

An essential aspect of most gait analyses is the accurate estimation of event times such as heelstrike and toeoff. This information is necessary to subdivide a stride into stance and swing periods regardless of the type of data being collected, and is often required to make meaningful comparisons between subjects and studies. In experiments in which a force platform is utilized, times of a single heelstrike and toeoff event could be determined accurately, but if 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 of a laboratory setting or on a treadmill, the accurate measurement of temporal components is generally not possible without specialized equipment.

One commonly used alternative to the force platform for determining the onset of stance and swing phases during gait is placing 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 toeoff times could be obtained during walking with these simple devices provided that the foot switches are properly positioned, and a predetermined offset time is taken into account (HAUSDORFF et al., 1995). Other specialized techniques that have been utilized to determine these timing parameters during gait include the use of an instrumented walkway (CROUSE et al., 1987; GIFFORD and HUGHES, 1983), mounting of 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 reasonably accurate, these techniques require equipment which 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 toeoff (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 optoelectric systems are utilized for data collection since video records are not obtained with these systems.

Utilizing the fact that kinematic patterns of walking are relatively consistent from stride to stride, and between speed conditions (WINTER, 1987), researchers have been able to accurately 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 which is utilized in numerous 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 toeoff times during normal (heelstrike) running at subject selected speeds, using only kinematic data. The accuracy of the event time predictions were evaluated by comparing results to those determined from force platform recordings.

Methods

Ten young (23.5 ± 2.6 y), healthy, physically active subjects (4 males, 6 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 center, lateral malleolus, calcaneus, and head of fifth metatarsal of the landing leg (Figure 1) were recorded in the sagittal plane with a single video camera (180 Hz) for at least 10 frames prior to heelstrike and after toeoff of each trial. Two-dimensional kinematic data were synchronized with ground reaction force (GRF) data (900 Hz). The raw 2-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 (WELLS and WINTER, 1980). Segmental angles of the leg (knee to ankle) and foot (heel to toe) were calculated from the smoothed coordinate data. Derivatives of segment angles were calculated using finite difference equations. Counterclockwise rotations of a segment were considered to be in the positive direction (Figure 1).

Insert Figure 1 about here

The times of heelstrike (HS) and toeoff (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 at which the vertical (y) component of the GRF rose above a threshold level of 10 N, while TO was considered to occur during the sample at 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 to estimate HS and TO times. Accuracy of the predictive algorithms were assessed by comparing results to those obtained from FP recordings.

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. Since the true minimum of α_{foot} generally occurred between discrete data frames, a linear interpolation equation (Eq. 1) was used to estimate the actual time that α_{foot} occurred.

$$t_{HS} = t_1 + \left(\frac{J(t_1)}{J(t_1) - J(t_2)} \right) t_{int} \qquad \text{(Eq. 1)}$$

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 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 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 toeoff, 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 Eq. 1 was utilized 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 arithmetic 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 true error 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.

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, while 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, while 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.

Insert Table 1 about here

Figure 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. Figure 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.

Insert Figures 2 and 3 about here

Discussion

The results of this study verified that the proposed algorithms provide accurate information regarding heelstrike, toeoff, 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 equaling the average RMS error. Errors in the prediction of t_{HS} (4.5 ms) were less than errors in predicting t_{TO} (6.9 ms) which could be partly due 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 less than errors in the prediction of t_{TO} involves the setting of a 10 N vertical force threshold to determine when contact was made. Since 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 toeoff 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 favorably to other techniques of determining gait event times which utilized 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, while 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 beyond a single camera is required. Using a simple foot switch technique, along with adjusting by a predetermined offset time, a group of researchers (HAUSDORFF et al., 1995) determined heelstrike times within ± 10 ms and toeoff 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.

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 toeoff time was less than 20 ms. Nilsson et al. (1985) presented a technique of predicting heelstrike and toeoff times during walking which 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. (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.

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 toeoff times 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 utilized. It appears that a greater variability exists in the kinematic patterns of running than 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. Since the implementation of these algorithms requires no special equipment, they may be utilized in any setting in which kinematic data are normally collected, including on a treadmill and outdoors. Any number of consecutive stride events could be measured using these algorithms. The resulting errors in estimating gait event times compare favorably to other techniques requiring specialized equipment, while the present method could be implemented solely with any 2-D or 3-D kinematic data collection system.

Acknowledgements

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 digitizing 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**, 64-68.

GIFFORD, G. and HUGHES, J. (1983). A gait analysis system in clinical practice. *J. Biomed. Eng.*, **5**, 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**, 347-352.

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

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

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

MANN, R. and HERMAN, J. (1985). Kinematic analysis of Olympic sprint performance: Men's 200 meters. *Int. J. Sport Biomech.*, **1**, 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**, 501-510.

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

NILSSON, J., STOKES, V. P. and THORSTENSSON, A. (1985). A new method to measure foot contact. *J. Biomech.,* **18**, 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**, 1119-1124.

ROSS, J. D. and ASHMAN, R. B. (1987). A thin foot switch. *J. Biomech.,* **20**, 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**, 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**, 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**, 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**, 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**, 175-188.

WELLS, R. P. and WINTER, D. A. (1980). Assessment of signal and noise in the kinematics of normal, pathological, and sporting gaits. In *Proceedings of the Special Conference of the Canadian Society for Biomechanics*. University of Western Ontario, London, pp. 92-93.

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

Table 1. Average true errors (TE) and RMS errors in estimating heelstrike (HS), toeoff (TO), and contact times (T) for individual subjects. All values in units of ms. For each subject, n=10.

	HS		TO		T	
Subject	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

Figure Captions

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

Fig. 2. Representative curve of foot angular acceleration (α_{foot}) vs. time from 50 ms before heelstrike to toeoff. Heelstrike occurs at the time of the minimum value of α_{foot}.

Fig. 3. Representative curve of leg angular acceleration (α_{leg}) vs. time from heelstrike to 50 ms after toeoff. Toeoff occurs at the time of the local minimum of α_{leg}.

Figure 1

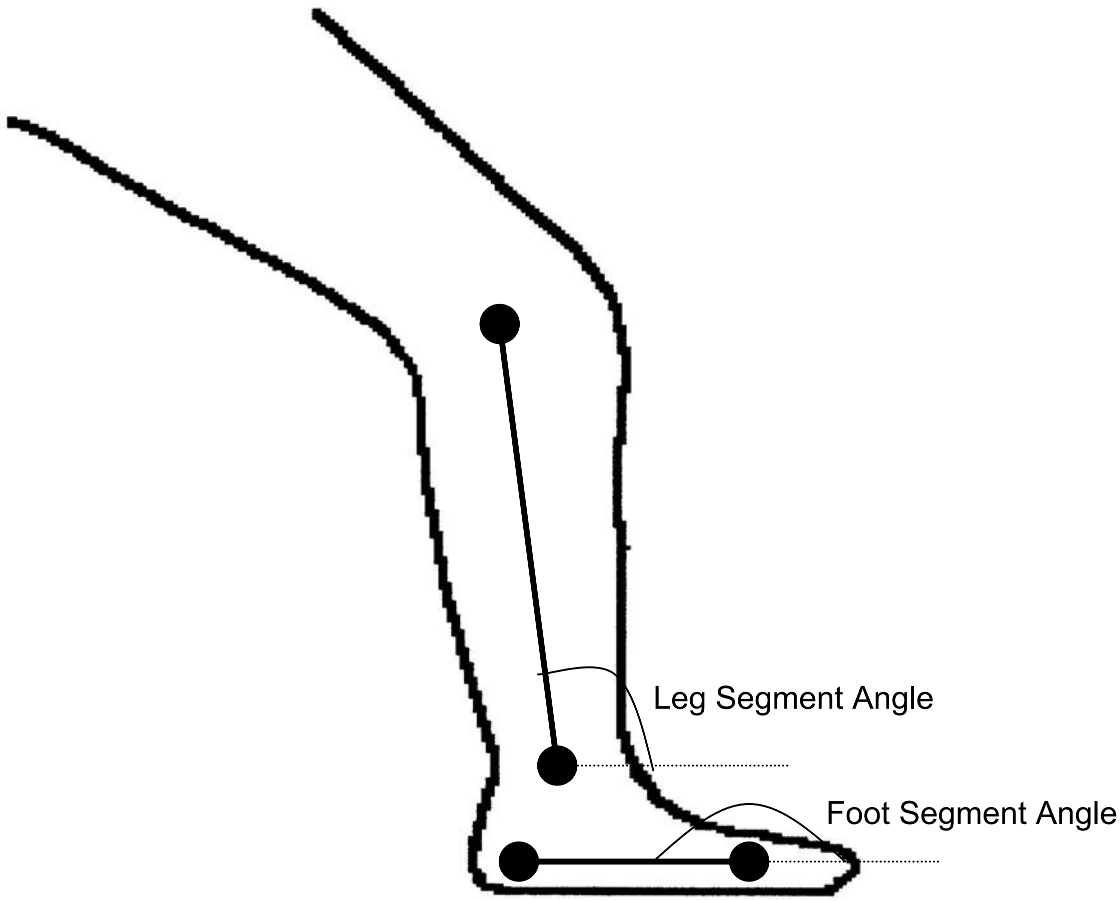


Figure 2

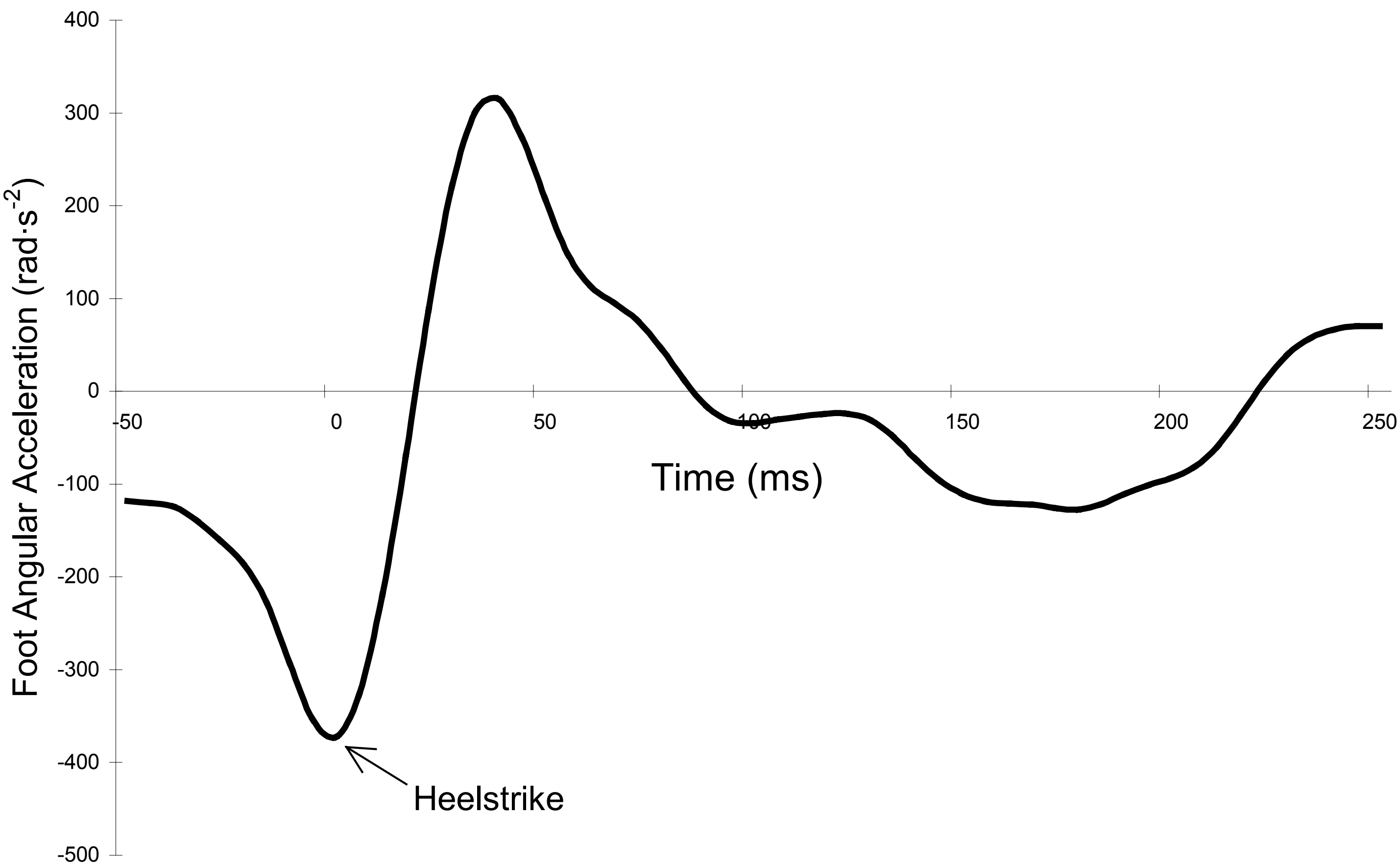
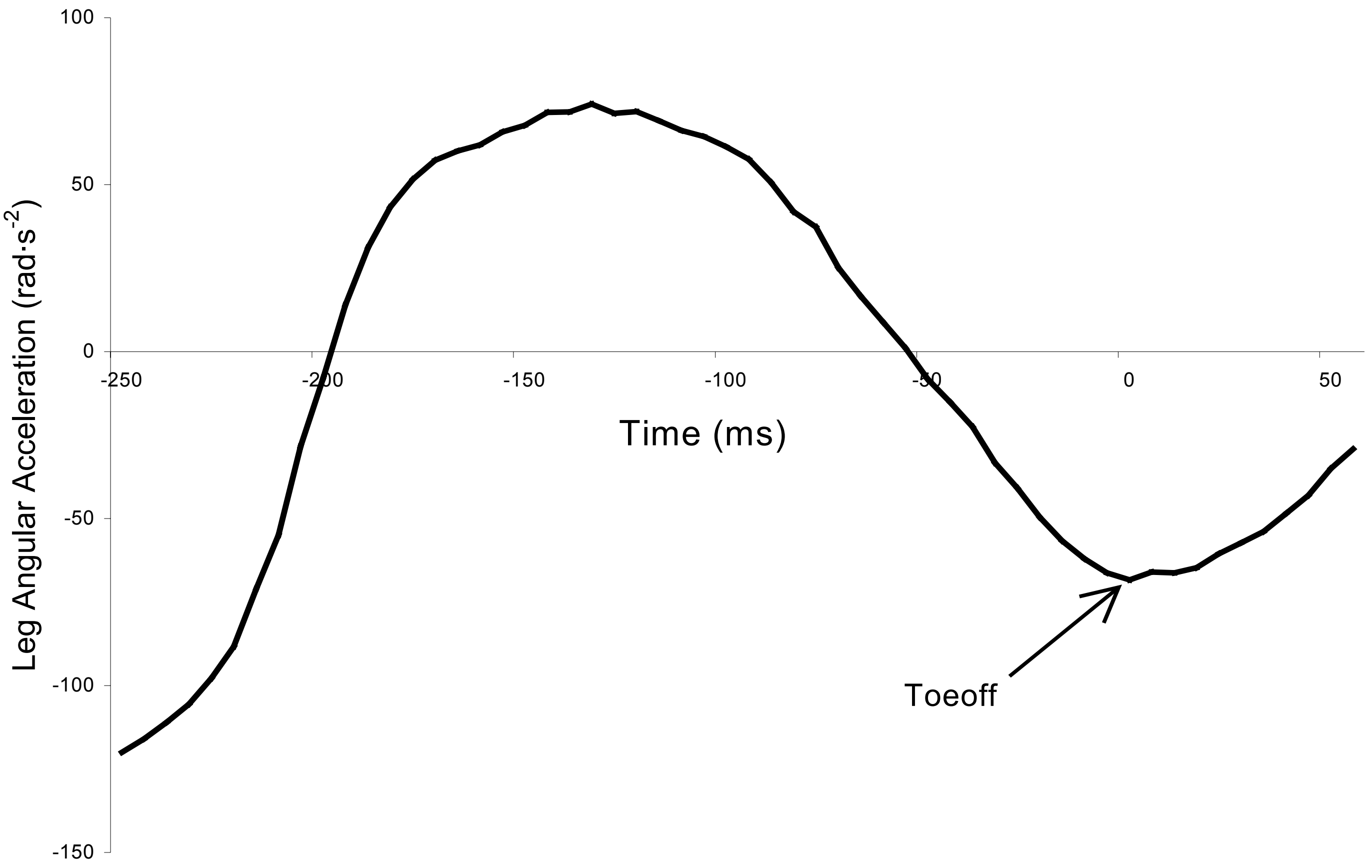

400
300
200
100
0
-100
-200
-300
-400
-500
Foot Angular Acceleration (rad·s-2)
-50
0
50
100
150
200
250
Time (ms)
Heelstrike

Figure 3



100
50
0
-50
-100
-150
-250
-200
-150
-100
-50
0
50
Leg Angular Acceleration (rad·s^{-2})
Time (ms)
Toeoff