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A comparison of kinematic algorithms to estimate gait events during overground running

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ABSTRACT

The gait cycle is frequently divided into two distinct phases, stance and swing, which can be accurately determined from ground reaction force data. In the absence of such data, kinematic algorithms can be used to estimate footstrike and toe-off. The performance of previously published algorithms is not consistent between studies. Furthermore, previous algorithms have not been tested at higher running speeds nor used to estimate ground contact times. Therefore the purpose of this study was to both develop a new, custom-designed, event detection algorithm and compare its performance with four previously tested algorithms at higher running speeds. Kinematic and force data were collected on twenty runners during overground running at 5.6 m/s. The five algorithms were then implemented and estimated times for footstrike, toe-off and contact time were compared to ground reaction force data. There were large differences in the performance of each algorithm. The custom-designed algorithm provided the most accurate estimation of footstrike (True Error 1.2 ± 17.1 ms) and contact time (True Error 3.5 ± 18.2 ms). Compared to the other tested algorithms, the custom-designed algorithm provided an accurate estimation of footstrike and toe-off across different footstrike patterns. The custom-designed algorithm provides a simple but effective method to accurately estimate footstrike, toe-off and contact time from kinematic data.

1. Introduction

The gait cycle is frequently divided into two distinct phases, stance and swing. Identification of the phases is reliant upon accurate detection of the gait events footstrike and toe-off, which is typically achieved using force plates. However, in some situations, force plates cannot be used [1] and therefore it is necessary to rely on kinematic algorithms to estimate events. Algorithms to detect gait events were primarily developed within walking studies [2–4], however research recognises there are significant kinematic differences between walking and running gait [5]. As inaccuracies may be evident when applying walking based algorithms to running gait data, an increasing number of algorithms have been developed specifically for running gait [1,6,7].

Previously published algorithms for detecting gait events during running are based around the timing of kinematic events, such as minimum shank and foot angular accelerations [6]. Although relatively good estimation accuracy has been reported [5,6], some algorithms fail to accurately determine the timing of footstrike [7]. Furthermore, other published algorithms have not been shown to perform as well in subsequent studies [5,6]. For example, Hreljac and Stergiou [6] determined gait events utilising two-dimensional (2D) calculations in which footstrike and toe-off coincided with the local minima of the foot and shank angular accelerations in the sagittal plane, respectively. The algorithm proved effective, with average Root Mean Square (RMS) errors for footstrike of 4.5 ms and toe-off of 6.9 ms. However, when tested by Fellin et al. [5], this same algorithm demonstrated lower accuracy, with errors of −20.8 ms for footstrike and −245.0 ms for toe-off.

Contact time is an important biomechanical variable which is often used to characterise running gait [8]. However, to date, most papers investigating kinematic algorithms for event estimation, have not reported the associated errors in ground contact times [1,4,5,7]. It is possible that small absolute errors in footstrike and toe-off estimation may accumulate, leading to relatively large errors in estimated contact time. Additional limitations of previous studies are that they have only tested event estimation algorithms at relatively slow speeds, for example 3.35 m/s [5] or have not reported speed [6,7], and some have only investigated one

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footstrike pattern, e.g. a rearfoot footstrike [5,6]. Therefore it is unknown how previously published algorithms perform at higher speeds, characteristic of competitive races and whether they work consistently across different footstrike patterns.

The purpose of the present study was three-fold: (1) to propose a new custom-designed kinematic algorithm, (2) to compare its accuracy with established algorithms for both event and contact time estimation at a relatively fast running speed and (3) to compare algorithm accuracy between different footstrike patterns. The accuracy of all algorithms was validated with force plate data.

2. Methods

2.1. Participants

Twenty elite runners (16 males and 4 females) participated in the study. The mean (±SD) age was 24.3 ± 4.4 years, mass 63.3 ± 8.5 kg and height 177.3 ± 6.8 cm. Participants reported no history of lower leg pain or pathology. Prior to participation all participants gave written informed consent and the research was approved by the University ethics committee. The study complied with the Declaration of Helsinki.

2.2. Data collection

Three-dimensional kinematic data (240 Hz) were collected using a twelve-camera motion capture system (ProReflex, Qualisys AB, Gothenburg, Sweden). Synchronised kinetic data (1200 Hz) were also collected with one 600 mm × 900 mm AMTI force plate (Advanced Medical Technologies Inc, Newton, MA, USA), which was embedded into the floor. Twenty-four 14 mm retro-reflective markers were placed on anatomical landmarks of the lower extremity in order to define the foot, shank, thigh and pelvis. Markers were placed bilaterally over the iliac crests, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), greater trochanters, medial and lateral femoral condyles, medial and lateral malleoli, calcaneous, and the first, second and fifth metatarsal heads. In addition, rigid cluster plates, each with four markers, were securely attached bilaterally on the participant's shank and thigh. Prior to testing, a static calibration trial was recorded, with the participants standing in an anatomically neutral position [9], after which the dynamic data were collected. Participants wore their own running shoes and ran along a 35 m track at 5.6 m/s (±2.5%), contacting the force plate with their right foot. Following a 5-min warm-up, five acceptable trials were collected for each participant. Trials in which the participant visibly targeted or partially contacted the force plate were discarded.

2.3. Data processing and analysis

For each of the five trials, data were trimmed visually in Qualisys Track Manager (version 2.5, Qualisys AB, Gothenburg, Sweden) to within the flight phase prior to the landing foot contact (right). Kinematic data were exported to Visual 3D (Visual 3D Inc., Rockville, MD, USA) and all raw kinematic marker trajectories were filtered with a fourth order Butterworth low pass filter with a cut-off frequency of 10 Hz. A CODA pelvis was created and hip joint centres were defined utilising adaptations to the equations developed by Bell et al. [10]. Segmental kinematics were derived using a global optimisation approach [11], described in detail in Mason et al. [12]. For each trial, true footstrike and toe-off events were identified by applying a 20 N threshold to the vertical component of the ground reaction force (vGRF). The strike Index [13] was then used to characterise footstrike patterns. This index is calculated as the position of the centre of pressure at footstrike, relative to the length of the foot [13]. A strike index of 0–33%, 34–67% and 68–100% indicate rearfoot, midfoot and rearfoot strikers, respectively. Using the strike index, 6 participants were classified as rearfoot, 8 as midfoot and 6 as forefoot footstrikers.

In order to estimate gait events based on kinematic data, five algorithms were implemented within Matlab (Mathworks, Natick, MA, USA) which used both raw marker trajectories and computed joint angles. Of these five algorithms four had been published previously and one was custom-designed. These algorithms are described in detail here:

Alton: We implemented an adapted version of the Alton et al. [2], algorithm outlined in Fellin et al. [5]. Footstrike and toe-off were defined as the minimum vertical position of the heel and second metatarsal head respectively. Due to the presence of multiple minima in the vertical position of the second metatarsal head, a secondary criteria [3] was implemented. Specifically, toe-off was defined as the minimum vertical second metatarsal head position closest to the second peak in knee extension.

Dingwell: Footstrike was defined to be the first peak in knee extension and toe-off as the second peak in knee extension [3].

Hreljac: An algorithm suggested by Hreljac and Stergiou [6] defined footstrike and toe-off as the local minima of the foot and shank angular accelerations in the sagittal plane respectively [5]. In order to account for the multiple minima in these accelerations Fellin et al. [5] trimmed data to 50 ms before footstrike. However, such an approach would not be appropriate in any practical application where the aim is to accurately determine footstrike in the absence of vGRF data. Therefore, instead of data trimming, a secondary criteria was incorporated. Specifically, footstrike was defined as the local minima of the foot angular acceleration in the sagittal plane which was closest to the first in peak knee extension.

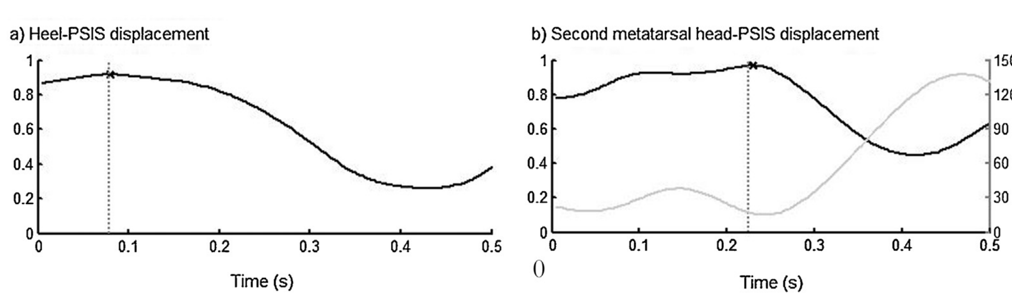

Fig. 1. Representation of the custom-designed algorithm (Smith). (a) Vertical displacement between the heel and PSIS marker is plotted against time and footstrike estimated to occur at the first maximum (denoted with an x). The dashed line shows footstrike time dervied from the vGRF. (b) Vertical displacement between the second metatarsal head and PSIS marker is plotted against time and toe-off estimated to occur at the local maxima closest to the second peak in knee extension. This estimate is marked on the graph with an x, in which the black line represents the custom-designed algorithm, whereas the grey line represents knee flexion–extension and again the dashed line represents vGRF derived toe-off.