

Estimating running spatial and temporal parameters using an inertial sensor

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Abstract An inertial measurement unit (IMU) is widely considered to be an economical alternative to capture human motion in daily activities. Use of an IMU for clinical study, rehabilitation, and in the design of orthoses and prostheses has increased tremendously. However, its use in defining running gait is limited. This study presents a practical method to estimate running spatial and temporal parameters using an inertial sensor by placing it on a shoe. A combination of a zero-crossing method and thresholding is used to identify foot-strike and foot-off based on foot acceleration during running. Stride time, ground contact time and flight time can then be identified. An off-phase segmentation algorithm is applied to estimate stride length and running speed. These two parameters are commonly used to evaluate running efficiency and to differentiate elite runners. This study found that an IMU can estimate foot-strike and foot-off with average absolute time differences of 2.60–6.04 and 2.61–16.28 ms, respectively. Stride time was estimated with error between -4.04 and 0.33 ms. Stride length and running speed were estimated with maximum average errors of 45.97 mm and 0.41 km/h.

Keywords Inertial sensor · Gait event detection · Running gait

1 Introduction

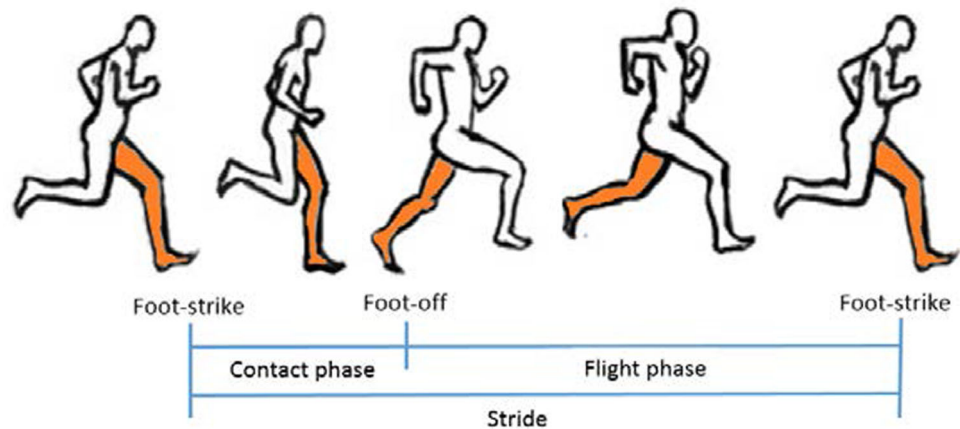
Spatial and temporal characteristics, such as stride time, contact time, flight time, stride frequency, and stride length (distance traveled per stride), are key parameters in evaluating running efficiency [1, 2] and commonly used to differentiate elite runners. For example, an endurance runner has a longer stride, shorter ground contact time and fewer changes in speed during ground contact [3]. Force plates and optical motion capture systems are widely used to determine these parameters [4–6]. Foot-strike time (FS) and foot-off time (FO) are first identified using a ground reaction force profile. Using these events, measurements are then divided into several strides and phases. Stride is defined as the period when one foot strikes the ground, lifts off the ground, and ends when the same foot hits the ground again. A stride can be divided into two phases: contact phase and flight phase (Fig. 1). Contact phase starts when the foot strikes the ground and ends when the toe lifts off the ground. Flight phase starts when toe lifts off the ground and ends when the same foot touches the ground. 3D kinematic data captured by motion capture systems are segmented on stride-to-stride basis to estimate stride length, stride frequency and running speed. Even though these technologies are accepted in running gait analysis, they have several drawbacks. They are expensive, require a controlled environment and have a relatively small capture volume.

The wearable inertial measurement unit (IMU) has been widely considered to be an economical alternative. While there is an increasing interest in the application of IMU for clinical studies and rehabilitation, its use in defining running gait is still limited. Among them are the works reported in [7], [8] and [9]. Yang et al. [7] placed IMU on the shank to estimate running speed. Their estimation

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Fig. 1 Running gait phases



method achieved root mean square error of 4.10% with maximum average error of 0.11 m/s. Bergamini et al. [8] placed the sensor on the lower back trunk. Their method used distinctive features exhibited by the second derivative of the angular velocity to determine FS and FO. It yielded an average absolute difference of 5 ms between IMU and the force plate and optical motion capture system with 95% of the differences were less than 25 ms.

Placing an IMU at the trunk or shank may not be the best option for the runners, as it requires them to wear semi-elastic belt to mount and hold the sensor at the desired location. This can lead to discomfort and often restrains muscle movements. At some instances, if the belt is not secured properly, it may slide down the trunk or shank due to the running impact and the elliptical shape of these body parts. This may cause misalignment of the sensing axes and lead to less accurate results. Moreover, some runners prefer to run without anything attached to them during training or competition. Hence, there is a need for a more practical location for the sensor to define spatial and temporal characteristics of the running gait. Mounting an IMU at a new location may require a different set of approaches. Change in location may generate different waveforms with different distinctive features. This can be observed in the acceleration of the trunk and shank reported in [7], [8] and [9], respectively. Integrating an IMU on the shoe (above the third metatarsal) can be the solution. Runners do not have to wear additional accessory that may restrain their motions during running. It also minimizes effort in setting up the inertial sensor.

This paper proposes a new approach to define running gait. It presents a new gait event identification method, which uses acceleration of the foot to determine FS and FO. Temporal parameters, e.g., contact time and flight time can then be derived from this information. This study also aims to demonstrate that spatial parameters, such as running speed and stride length, can be estimated accurately using the method presented in [7]. An optical motion

capture system is used here as the benchmark to validate the proposed method.

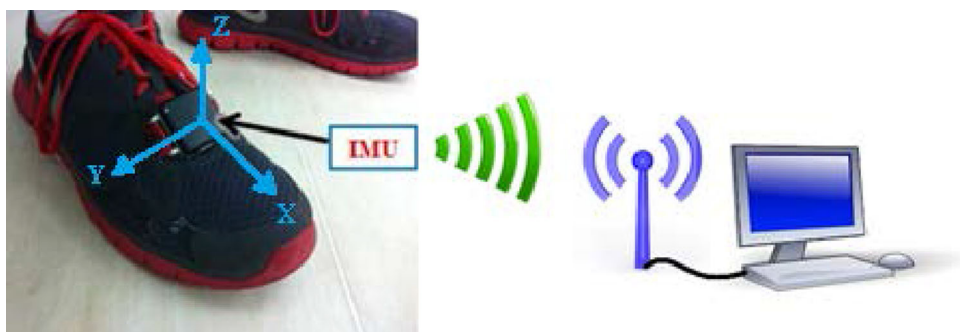
2 Method

Ten healthy male subjects (age: 25.5 ± 3.8 years/old; height: 174.4 ± 19.5 cm; weight: 65.5 ± 15.2 kg) participated this study. Subjects were briefed on the purpose and method before obtaining their consent. In this study, an IMU (Opal, APDM Inc.) was placed on the right shoe (Fig. 2) and looped together using shoe string to minimize measurement errors due to vibratory motion when the foot hits the ground. This sensor can measure acceleration ± 6 g m/s^2 and angular velocity ± 2000 deg/s at sampling rate of 128 Hz [10]. It weighs 550 g and has dimension of $36.50 \times 36.10 \times 13.40$ mm. Participants were instructed to run on a treadmill for two sessions. Each session lasted for 10 min. During the first session, they were requested to walk at a speed of 1.3 km/h for the first minute and at 4 km/h for the subsequent minute, then started running for 3 min at a speed of 8 km/h and for another 3 min at 9 km/h. Lastly, these participants were allowed to cool down by walking at a speed of 2 km/h for 1 min and at 1.3 km/h for the last minute. After this session, they were allowed to rest for 3 min before proceeding to the next session. Second session was similar to the first session. Instead of running at 8 and 9 km/h, they ran at 10 and 11 km/h. Subjects gave their consent to participate in the experiment in accordance with the university policy. This study was approved by university research ethics committee. Twenty continuous strides at each running speed were investigated here.

2.1 Estimation of temporal parameters

This study found that acceleration measured by the accelerometer along its X-axis, a_x can identify FS and FO.

Fig. 2 Placement of wireless Inertial Measurement Unit (IMU) on shoe and its inertial sensing axes



In every stride, two prominent local minima are present in a_x , Fig. 3. The first local minimum corresponds to FS and the latter corresponds to FO. A heuristic-based algorithm, similar to [11], was developed to detect these events. First, it finds the derivative of a_x and identifies the point in which the derivative crosses zero. Subsequently, the point in which the derivative crosses zero is identified as the potential gait event. Next, a threshold of -8 m/s^2 is applied. If a_x is less than this threshold then it can be considered as FS or FO. A comparison is then made between two subsequent minima: The local minimum with greater amplitude is identified as FS and the other is regarded as FO.

Once FS and FO are identified, stride time, T_S , duration of contact phase, T_C , and flight phase, T_F , can be calculated, where n is the number of stride.

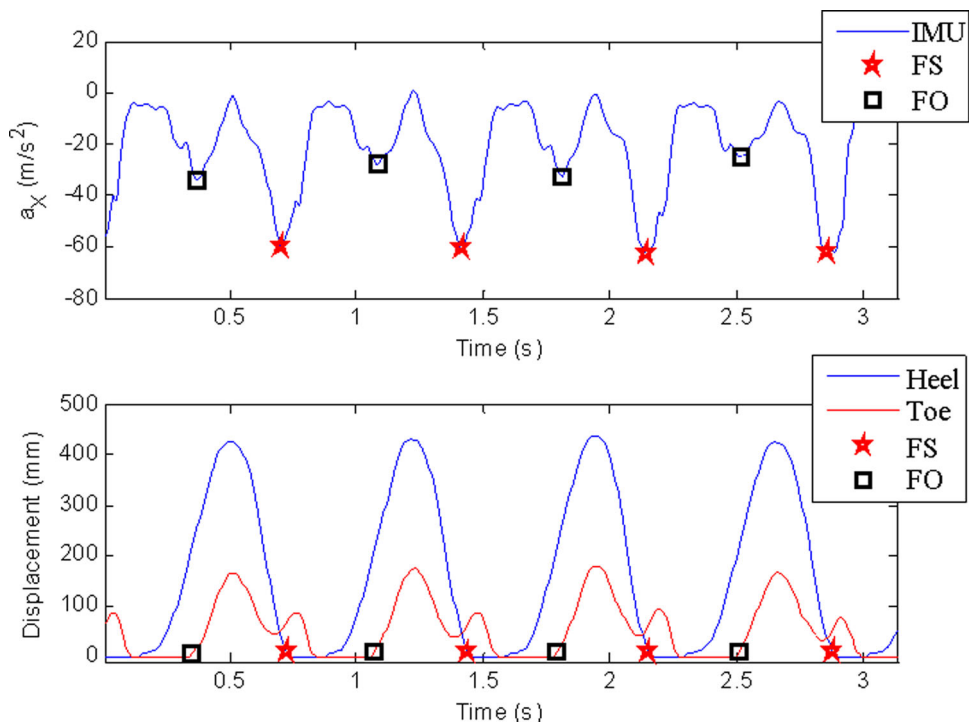
$$T_S = \text{FS}(n) - \text{FS}(n - 1) \tag{1}$$

$$T_C = \text{FO}(n) - \text{FS}(n) \tag{2}$$

$$T_F = \text{FS}(n) - \text{FO}(n - 1) = T_S - T_C \tag{3}$$

An optical motion capture system (Qualisys, Qualisys AB) was used to validate the temporal parameters defined by the IMU. Reflective markers were placed on the heel and third metatarsal of the foot to detect FS and FO, respectively. A 10 mm threshold was applied to the vertical trajectories of these markers. FS is identified to be the time when heel marker approaches the threshold. FO is the time when marker at the third metatarsal marker lifts off 10 mm above the ground. The sampling rate for the motion capture system was set at 128 Hz. Equations (1–3) were used to estimate T_S , T_C , and T_F using events identified by the system.

Fig. 3 Signatures of foot-strike time (FS) and foot-off time (FO) in acceleration measured by IMU (top figure) and in the displacements of heel and toe markers (bottom figure)



2.2 Estimation of spatial parameters

This study adopted an off-phase segmentation algorithm proposed in [7]. Instead of using angular rate of the shank, foot angular rate, ω , was used to derive the stride length and running speed. Angular rate of the foot and shank exhibits a similar waveform and thus can be used to estimate spatial parameters of the running gait, as reported in our earlier study [12]. The angle cycle starts at vertical event ($\theta = 0$)—when the shank is in its vertical position. This is also a point where ω experiences a transition from negative value to positive (Fig. 4a). This cycle ends at the next vertical event. The starting point and ending point of this cycle are used as the integration boundary to determine instantaneous angle, $\theta(t)$:

$$\theta(t) = \int_0^t \omega(t)dt + \theta(0) \tag{4}$$

where $\omega(t)$ is the measured angular velocity and $\theta(0)$ is the initial angle at the beginning of the integration. Coordinate

system transformation is then performed to calculate horizontal and vertical acceleration, $a_x(t)$ and $a_y(t)$:

$$\begin{bmatrix} a_x(t) \\ a_y(t) \end{bmatrix} = \begin{bmatrix} \sin \theta(t) & \cos \theta(t) \\ \cos \theta(t) & -\sin \theta(t) \end{bmatrix} \begin{bmatrix} a_n(t) \\ a_t(t) \end{bmatrix} - \begin{bmatrix} 0 \\ g \end{bmatrix} \tag{5}$$

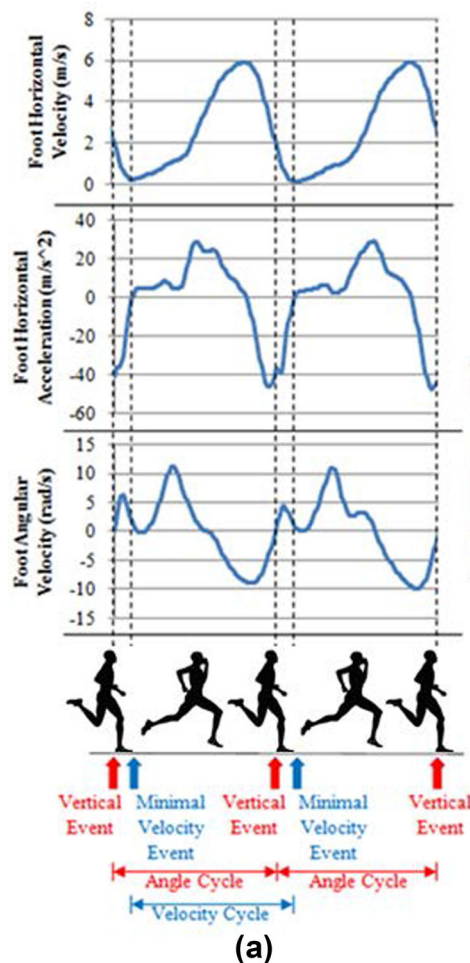
where $a_n(t)$ is the normal acceleration, which is equal to the acceleration measured in Z-direction of the IMU sensing axis (Fig. 4b), $a_t(t)$ is the tangential acceleration, which is the acceleration measured in X-direction of the IMU sensing axis, and g is the gravitational acceleration. The calculated $a_x(t)$ and $a_y(t)$ are integrated to obtain instantaneous horizontal and vertical velocities, $v_x(t)$ and $v_y(t)$:

$$v_x(t) = \int_0^t a_x(t)dt + v_x(0) \tag{6}$$

$$v_y(t) = \int_0^t a_y(t)dt + v_y(0) \tag{7}$$

where $v_x(0)$ and $v_y(0)$ represent the initial horizontal and vertical velocities, respectively. Next, the velocity cycle is

Fig. 4 a Characteristics of foot horizontal velocity, horizontal acceleration and angular velocity during running. **b** The IMUs configuration



(b)

defined to determine instantaneous horizontal and vertical velocities. This cycle starts at a minimal velocity event, where $a_x(t)$ experiences transition from negative slope to positive slope, and ends at the next event.

Assuming the shank and foot is a single rigid body that approximately rotates about the ankle joint during the stance phase; the initial velocity can be estimated at the beginning of the velocity cycle, as presented:

$$v_t(0) = \omega(0) \cdot L \quad (8)$$

$$\begin{bmatrix} v_x(0) \\ v_y(0) \end{bmatrix} = \begin{bmatrix} \cos \theta(0) \\ -\sin \theta(0) \end{bmatrix} \cdot v_t(0) \quad (9)$$

where $v_t(0)$ represents the tangential velocity at the beginning of the velocity cycle and L represents the distance between the sensor and the ankle joint.

The effect of sensor bias is then reduced by determining the end velocity of each cycle, $v_{xe}(T)$ and $v_{ye}(T)$:

$$v_t(T) = \omega(T) \cdot L \quad (10)$$

$$\begin{bmatrix} v_{xe}(T) \\ v_{ye}(T) \end{bmatrix} = \begin{bmatrix} \cos \theta(T) \\ -\sin \theta(T) \end{bmatrix} \cdot v_t(T) \quad (11)$$

$$v_{xc}(t) = v_x(t) + \frac{v_{xe}(T) - v_x(T)}{T} t \quad (12)$$

$$v_{yc}(t) = v_y(t) + \frac{v_{ye}(T) - v_y(T)}{T} t \quad (13)$$

where T is the period of the corresponding velocity cycle and $v_{xc}(t)$ and $v_{yc}(t)$ are the instantaneous horizontal and vertical velocities after bias compensation. The horizontal $s_x(T)$ or the stride length is calculated by integrating $v_{xc}(t)$ over the entire velocity cycle:

$$s_x(t) = \int_0^t v_{xc}(t) dt + s_x(0) \quad (14)$$

where $s_x(0)$ is the initial horizontal displacement at the beginning of the velocity cycle, which is always zero. Finally, the estimated running speed in each stride cycle, $V_e(T)$, is defined as follows:

$$V_e(T) = \frac{s_x(T)}{T} \quad (15)$$

2.3 Validation

The root mean square error (RMSE) and estimation error were calculated to evaluate the accuracy of the proposed method. Temporal parameters measured by the motion capture system were set as the reference. For each treadmill speed, the mean estimation error and its standard deviation were calculated by averaging across all subjects. A T test was conducted to examine the significance difference of

T_S , T_C and T_F estimated by the IMU and the motion capture system.

Similar approaches were carried out to validate the spatial parameters. Differences in the stride length (ΔS) were calculated by comparing the horizontal displacement estimated by the IMU (S_X) and horizontal distance traveled by the third metatarsal marker in each stride. The difference in running speed (ΔV) was determined by the difference between running speed estimated by IMU and the treadmill running speed.

3 Results

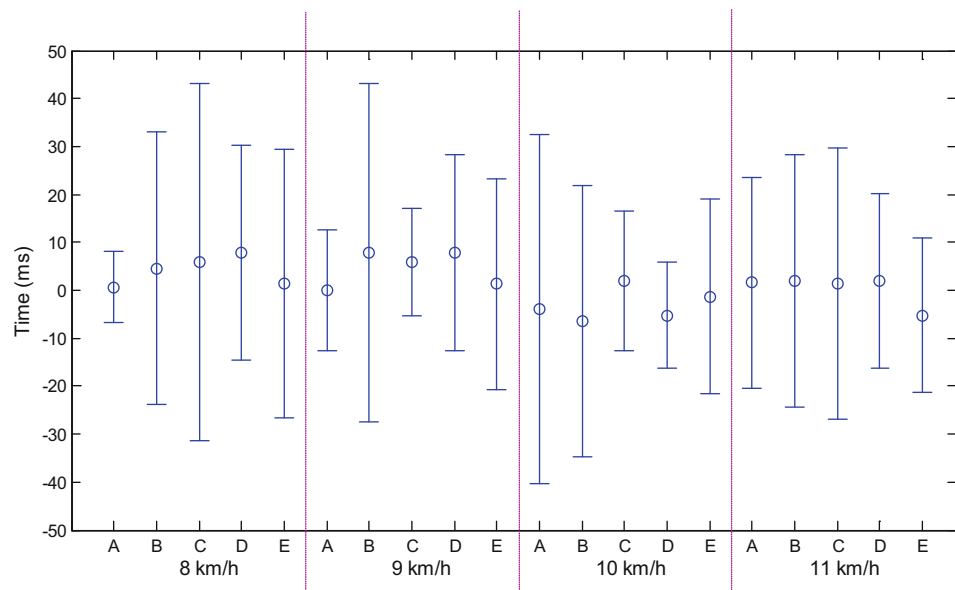
The temporal estimation results gathered from all subjects at each running speed are presented in Table 1. Several negative values were found in the timing of FS and FO. These values imply that the IMU detected FS and FO earlier than the optical system. The FSs were detected earlier at 8 and 9 km/h, whereas FOs were detected earlier at 10 km/h. Nevertheless, these errors were comparable to findings reported using joint angles [6] and velocity and acceleration profile of the foot during running [13]. They were also comparable to results estimated using IMU placed on the lower back [8].

The duration of the running stride was estimated (Fig. 5) with the mean differences ranging between -0.98 and 4.04 ms and RMSE ranging from 17.34 to 24.81 ms. No statistical difference was found ($p = 0.92$) between the measures from both systems (IMU and optical system). These results support the findings presented in [14].

Table 1 Temporal differences between IMU and optical system

		8 km/h	9 km/h	10 km/h	11 km/h
ΔFS (ms)	Mean	- 2.60	- 6.04	4.34	3.04
	S. D.	12.78	14.12	17.89	14.08
	RMSE	4.66	5.31	8.28	4.66
ΔFO (ms)	Mean	3.26	16.28	- 4.34	2.61
	S. D.	20.90	16.70	15.03	19.54
	RMSE	9.01	11.06	7.57	10.99
ΔT_S (ms)	Mean	- 4.04	- 3.15	- 0.98	0.33
	S. D.	24.00	22.71	25.60	22.12
	RMSE	17.62	17.34	24.81	21.37
ΔT_C (ms)	Mean	- 6.09	- 8.20	- 8.07	- 8.10
	S. D.	6.19	5.29	3.08	2.52
	RMSE	7.76	9.17	9.11	9.95
ΔT_F (ms)	Mean	6.09	8.20	8.07	8.10
	S. D.	6.19	5.29	3.08	2.52
	RMSE	7.76	9.17	10.11	9.95

Fig. 5 Mean and standard deviation of temporal difference of stride time (ΔT_s) of five subjects (subject A, B, C, D and E) at different running speeds



The duration of the contact phase estimated by the IMU was shorter than the ones measured using the optical system. Hence, negative values were found in ΔT_C in each running speed. No statistical difference was found in T_C ($p = 0.75$).

Table 2 summarizes the spatial differences at each running speed. Distance estimated by the IMU had differences ranging from 32.33 to 45.97 cm. These differences were only a fraction of the stride length which could range between 146.3 and 165.4 cm depending on subject's leg length and running speed. Mean differences between the running speed and treadmill speed were between 0.08 and 0.41 km/h. These results demonstrated that the off-phase segmentation method can estimate running speed even when the sensor was placed on a shoe rather than on the shank.

4 Discussion

This study demonstrates that the running gait spatial and temporal parameters can be estimated with minimal errors. These errors can be attributed to several factors. Among

Table 2 Spatial differences in stride length and running speed between IMU and optical system

		8 km/h	9 km/h	10 km/h	11 km/h
ΔS (cm)	Mean	32.33	14.07	43.99	45.97
	S. D.	48.16	46.03	56.74	62.64
	RMSE	62.43	70.16	63.84	59.17
ΔV (km/h)	Mean	0.16	0.08	0.24	0.41
	S. D.	0.24	0.23	0.29	0.32
	RMSE	0.31	0.36	0.53	0.63

them is the assumption that knee-to-ankle joint distance, L , is the rotation arm during the stance phase and is used to calculate the initial and end velocities [7]. However, during running, the heel-lift takes place before foot-off, and hence the shank segment does not rotate about the ankle joint. The actual rotation arm is slightly longer than L . Minor misalignment of the sensor-sensing axes and manually measuring L can contribute to these errors. Lastly, shoe movement during contact phase may introduce errors in the measured acceleration due to contact and friction between the shoe and the ground. Nevertheless, to the best of our knowledge, this is the first study that reports the use of the inertial sensor to estimate stride length during running. With these results, it is hoped that it can be a baseline for the future study to improve its accuracy, potentially by introducing an additional device, such as GPS (Global Positioning System). A study by Beato et al. [15] reported that the 10 Hz GPS they used to measure shuttling distance has an accuracy between 22 and 41 cm. Hence, there is a possibility of combining these two devices to create a more accurate measurement that can estimate running spatial parameters. The differences in the estimated temporal parameters are in agreement with existing studies. Using an optical system and force plate, Osis et al. [6] reported that all predicted foot-strike were within 20 ms, with a maximum error of 28 ms. Fellin et al. [5] reported error from 22.4 to 24.6 ms for FS, and from 4.9 to 5.4 ms for FO.

Integrating an inertial sensor on a shoe provides several advantages than at other locations. First, it does not inhibit athlete's movement during running. The athlete does not have to wear additional accessories, e.g., elastic band to hold the sensor on the body. Second, fixing the sensor on the shoe can minimize sensor-sensing axis alignment errors

in both intra- and inter-subjects experiments, thus reducing errors in the estimated running gait parameters. Yang et al. [7] reported that a 10° misalignment would lead to 8% reduction of the estimated running speed. Lastly, it can be mounted on the shoe by the wearer without technical guidance or advice. This minimizes the hassle to mount this device during training routines and competitions.

Algorithms used in this study have several merits. Identification of gait events using the zero-crossing method and thresholding minimizes computational time and resources to identify FS and FO. Due to its simplicity, it can be easily embedded into online and offline data processing—allowing runners to monitor their gait characteristics during running. A threshold of -8 m/s^2 was selected based on a preliminary study, in which this threshold was found to be able to identify gait events for a wide range of walking, jogging and running speeds (4–12 km/h) and in all subjects. On the other hand, a 10 mm threshold applied to the markers' trajectories to determine FS and FO was selected with several intentions. First, the distance of the heel marker is approximately 20 mm from the ground. Second, the distance of the toe marker is approximately 50 mm above the ground. Hence, this threshold is deemed to be sufficient to ensure that the gait events are detected. An off-phase segmentation algorithm was proven to be sufficiently robust in defining spatial parameters. It successfully determined the walking speed [16] and running speed [7] using shank acceleration and angular velocity. It also estimated running speed using measurements collected on runner's shoe as demonstrated here and in [12].

Several limitations should be noted. Treadmill running is a norm in gait analysis. It provides a standardized and reproducible environment where speed can be easily controlled and the required calibration volume for the optical system is considerably reduced. However, there are biomechanical differences between overground and treadmill running. These differences can be attributed to several factors such as treadmill familiarity, change of visual and auditory surroundings, mechanical property of the running surface, and air resistance [17]. These factors result in a subtle change to the lower limb kinematic patterns [18]. Nevertheless, treadmill running can be considered as a representative expression of overground running [17–19]. Limited running speed is the other limitation of this study. In uncontrolled overground running, participants may run slower than 8 km/h and faster than 11 km/h. Even though, the proposed method may do well in determining the spatial and temporal parameters, qualitative and quantitative examinations should be carried out to further investigate these parameters beyond the tested running speed.

5 Conclusion

Mounting a wireless IMU on a shoe can be a practical approach in defining running gait characteristics. The experimental study demonstrated that information derived from an IMU can be used to estimate the timing of FS and FO, the duration of stride, stance phase and swing phase, stride length and running speed. The average temporal differences between the IMU and force plate are between -2.60 and 3.04 ms for FS, and -4.34 and 16.28 ms for FO. No statistical difference was found in the duration of stride, stance phase and swing phase. The stride length and running speed were estimated with average errors between 14.07 and 45.97 cm and between 0.08 and 0.41 m/s, respectively.

Compliance with ethical standards

Conflict of interest The authors declare that they have no conflict of interest.

References

- Gindre C, Lussiana T, Hebert-Losier K, Morin J (2015) Reliability and validity of Myotest for measuring running stride kinematics. *J Sports Sci*. doi:10.1080/02640414.2015.1068436
- Garcia Perez JA et al (2013) Effect of overground vs treadmill running on plantar pressure: influence of fatigue. *Gait Posture* 28:929–933
- Nummela A, Keränen T, Mikkelsen LO (2007) Factors related to top running speed and economy. *Int J Sports Med* 28(8):655–661
- Hreljac A, Stergiou N (2000) Phase determination during normal running using kinematic data. *Med Biol Eng Compu* 38:503–506
- Fellin RE, Rose WC, Royer TD, Davis IS (2010) Comparison of methods for kinematic identification of foot-strike and toe-off during overground and treadmill walking. *J Sci Med Sports* 13:646–650
- Osis ST, Hettinga BA, Leitch J, Ferber R (2014) Predicting timing of foot strike during running, independent of striking technique, using principal component analysis of joint angles. *J Biomech* 47:2786–2789
- Yang S, Mohr C, Li Q (2011) Ambulatory running speed estimation using inertial sensor. *Gait Posture* 34(4):462–466
- Bergamini E, Picerno P, Pillet H, Natta F, Thoreux P (2012) Estimation of temporal parameters during sprint running using a trunk mounted inertial measurement unit. *J Biomech* 45:1123–1126
- Wixted AJ, Billing DC, James DA (2010) Validation of trunk mounted inertial sensors for analysing running biomechanics under field conditions, using synchronously collected foot contact data. *Sports Eng* 12(4):207–212
- APDM, Inc. (2013) Motion Studio User's Guide
- Gouwanda D, Gopalai AA (2015) A robust real-time gait event detection using wireless gyroscope and its application on normal and altered gaits. *Med Eng Phys* 37(2):219–225
- Chew D, Gouwanda D, Gopalai AA (2015) Investigating running gait using a shoe-integrated wireless inertial sensor. In: Proceedings of 2015 IEEE region 10 conference - TENCON 2015, 1–4 Nov 2015, Macao, China, pp 1–6

13. Leitch J, Stebbins J, Paolini G, Zavatsky AB (2011) Identifying gait events without a force plate during running: a comparison of methods. *Gait Posture* 33(1):130–132
14. Lee JB, Mellifont RB, Burkett BJ (2010) The use of a single inertial sensor to identify stride, step, and stance durations of running gait. *J Sci Med Sport* 13:270–273
15. Beato M, Bartolini D, Ghia G, Zamparo P (2016) Accuracy of a 10 Hz GPS Unit in Measuring Shuttle Velocity Performed at Different Speeds and Distances (5–20 M). *J Hum Kinet* 54(1):15–22
16. Li Q, Young M, Naing V, Donelan JM (2010) Walking speed estimation using a shank-mounted inertial measurement unit. *J Biomech* 43(8):1640–1643
17. Schache AG, Blanch PD, Rath DA, Wrigley TV, Starr R, Bennell KL (2001) A comparison of overground and treadmill running for measuring three-dimensional kinematics of the lumbo-pelvic-hip complex. *Clin Biomech* 16:667–680
18. Dixon SJ, Collop AC, Batt ME (2000) Surface effects on ground reaction forces and lower extremity kinematics during running. *Med Sci Sports Exerc* 32:1919–1926
19. Riley PO, Dicharry J, Franz J, Croce UD, Wilder RP, Kerrigan DC (2008) A kinematics and kinetic comparison of overground and treadmill running. *Med Sci Sports Exerc* 40(6):1093–1100