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The Development and validation of an intertial sensor for measuring cycling kinematics: a preliminary study

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Abstract: A biomechanical variable of interest to cyclists and cycling coaches is postural stability. A cyclist's position on a bicycle can be easily measured in a laboratory environment using motion capture software, but is difficult to measure in the field. The focus of this paper was to identify the legitimacy of a sacrum mounted triaxial accelerometer to identify temporal acceleration magnitudes of the centre of mass (CoM) whilst cycling against a motion analysis system. To provide validation of the sensor, data was collected at the torso as cyclists pedaled at varied cadences against a motion analysis system. The effects of cycling cadence and changes to torso angle via changes to hand position revealed that wearable technology (accelerometers) provide legitimacy in the assessment of torso accelerations during cycling. The minimal variation and change in agreement between the two systems during cycling indicates the adherence method of the accelerometer was suitable.

Keywords: Wearables; Accelerometer; Triathlon; Torso Centre of Mass; Cycling.

1. Introduction

Use of wearable technologies (wearables) such as accelerometers has increased in recent years. Wearables have been cited as an advancement in fitness tracking as they provide more precise data than classic self-assessment methods (Meyer and Hein 2013). Such technology is attractive because of the potential to measure movement unobtrusively, in the ambulatory environment and at a comparative cost compared to laboratory-based equipment (Lee et al. 2019a).

This research provides a basis for wearable technology, particularly accelerometers, to be applied qualitatively during movement recognition of cycling. At present the most accepted method of measuring kinematics such as cycling during different body positions and cadences (revolutions per minute, rev.min^{-1}) is to use three-dimensional motion capture (3D MoCap) (Mayagoitia et al. 2002). The need for new methods of monitoring is due to the cost of 3D MoCap systems and the restrictive nature of the systems that are typically confided to laboratory environments (Luinge and Veltink 2005) and require participants to remain in close proximity and frequently tethered to equipment (Lee et al. 2019b). Although commonly referred to as the gold standard, 3D MoCap has limited use outside of laboratories. The current generation of body-worn accelerometers are attractive due to the potential to measure human movement unobtrusively and in an environment that more closely simulates real-world requirements (i.e. outdoors) (Miller et al. 2013). Evaluating the agreement between accelerometers and an accepted gold standard measure of kinematics would determine whether the method is valid for measuring movement during cycling.

An accelerometer is an electromechanical device that can measure acceleration forces. These forces may be static, like the constant force of gravity, or dynamic, such as those caused by moving or vibrating the accelerometer during motion. While previous research has validated temporal gait kinematics in running (Lee et al. 2010) and armstroke classification in swimming (Lee et al. 2008) there is a research gap in the present literature for accelerometry validation, with limited information regarding cycling movement validations specific to temporal torso kinematics during cycling (Evans et al. 2020).

Cycling is a series of repetitive motions often performed over a prolonged duration. Anthropometric dimensions of the cyclist and component selection (Bini and Carpes, 2014) are critical for mechanical fitting the athlete to the bicycle (bike sizing), as these factors may impact the athlete's ability to perform the coordinated task of pedaling. Poorly distributed or excessive rocking of the torso in the saddle can contribute to discomfort and degrade performance. In contrast, when the upper body is kept stable it serves as a brace for the power-producing limbs of the lower leg (Fleming et al. 1998). Relative to cycling, an increase in workload produces higher accelerations of the torso (Hewitt, 2005) and the resulting need of stabilisation of the upper body (Costes et al. 2015).

As a comparative amount of torso stability is needed in order to balance the bicycle (McDaniel et al. 2005), a forward shift of the torso centre of mass (CoM) means that cyclist's body is less supported by the saddle which would require more stabilisation from the torso. In this instance, the torso CoM refers to the gross motion of the body. As the torso undergoes movement, the magnitude of torso acceleration, as observed at the spinous process, will be a function of its local coordinates (i.e., x, y, z) acceleration components. In this regard a postural change will be apparent in the local acceleration components. In this paper, torso accelerations of each local component were compared for each participant (i.e., participant motion only) to examine the longitudinal (x), mediolateral (y) and anteroposterior (z) changes in two commonly used saddle positions, namely a drops and aerodynamic position. Knowledge concerning temporal accelerations of the torso could provide valuable information on the movement pattern for performance and possibly injury prevention. However, the timing of these acceleration magnitude outputs should ideally have a high agreement with a gold standard measurement method (i.e., 3D MoCap) to accurately portray movement as it occurs in real time. Therefore, the scope of this preliminary study is to fill the gap in the current literature by determining the agreement between a sacrum mounted triaxial accelerometer and 3D MoCap to measure postural movements of the torso.

Thus, the aim of this research was to investigate the validity relative to measurement of temporal torso kinematics using a triaxial accelerometer compared to a 3D MoCap during cycling at varied cadence. Four random yet progressively increased cadence conditions were chosen. Additionally, temporal magnitudes of acceleration were compared as cyclists changed from a drops to aerodynamic for timing comparisons between the triaxial accelerometer and 3D MoCap. Validating accelerometers for this purpose provides evidence for the development of an accurate, quantitative and practical means of assessing postural changes,

2. Methods

2.1 Participants

This preliminary observational study consisted of four participants (three males, one female, 33 ±2.5 years, 175 ± 4.6 cm, 68 ±6.2 kg, 6.3 ±0.3 weekly training hours).

Informed consent was obtained, then triathlon history and physical activity readiness questionnaires (PAR-Q) completed before commencement of the test. The criterion included a minimum of 12 months prior experience of competing and training in cycling events, be in a well-trained state and be injury free with no adhesive tape allergies. Ethical clearance was granted by the Charles Darwin University Ethics Committee (HREC19028). Participants were asked to refrain from vigorous exercise 24 hours prior to testing and instructed to preserve their normal diet prior to testing. Participants were tested on the same day using their own bicycles to eliminate the effects of an unfamiliar bike.

Protocol

Bicycles were secured on an ergometer (Cyclus2, Leipzig, Germany) with participants asked to perform a self-selected warm up for the initial 5 minutes at an Individual Preferred Cadence (IPC). Cadence was measured in rev.min^{-1} and was visible to participants on the Cyclus2 display screen in order for participants to self regulate exertion. The warm up was immediately followed by 15 minutes of continuous cycle that consisted of 4 different cadence conditions, each of which lasted 3 minutes. The cadence conditions consisted of: (1) IPC rev.min^{-1}, (2) 55–60 rev.min^{-1}; (3) 75–80 rev.min^{-1}; and (4) 95–100 rev.min^{-1}, all of which were randomly ordered from minutes 5–20 to simulate the varying cadences of normal cycling. Cadences were selected as representative of typical for this population (Chapman et al. 2007). Hand position was altered at minutes 5-8 whereby participants adopted a drops position, defined as torso low, wrists straight and elbows slightly bent on drop bars. From minutes 8-17 participants cycled in an aerodynamic position, defined as forearms extended resting on aero bars with torso positioned greater to the horizontal), before reverting to a drops position for minutes 17-20. Body position was defined as the location of the cyclist relative to the pedal axle of the bicycle which was determined by the angle of the bicycle seat tube and a vertical line (perpendicular to the ground) passing through the pedal axle (Bini et al. 2014). Participants kept within the selected cadence by monitoring the feedback screen attached to the ergometer.

Clipless pedals were used with a yellow Shimano SPD-SL pedal cleat with approximately 6° floatation in the mediolateral direction. All participants used cleated shoes in the fore-aft location, defined as the centre of the pedal axle in line with the metatarsophalangeal joint. Individual pedal strokes (top dead centre to the subsequent ipsilateral top centre) were determined visually and deemed measurable when the components coincided with an event in the plotted traces of one of the orthogonal planes (e.g., a positive or negative spike in the acceleration data). Ratings of perceived exertion (RPE) were verbally obtained at each 3 minute epoch prior to the proceeding change of cadence.

Cycling events and corresponding cadence changes were identified in raw MoCap and accelerometer data to ensure no loss or timing shift. Longitudinal acceleration was used to identify a change in posture that represented acceleration of the torso and was identified at the point where the acceleration magnitude began increasing towards its large impact peak. Mediolateral acceleration was used to identity pedal strokes. For each cadence condition, temporal torso acceleration magnitude and MoCap was initially averaged to 60 seconds in order to obtain a true reflection of steady cycling. Accelerometer data were calibrated to produce a gravitational (*g*) scale output. In relating accelerometer outputs to the customary 3D physical environment and Cartesian coordinates, acceleration is defined as the rate of change of velocity, or, equivalently, as the second derivative of position. It is thus a vector quantity with SI units measured in metres per second/per second ($m.s^{-2}$). Therefore, accelerations in the present work were subsequently scaled into ($m.s^{-2}$).

Intensity was managed to a maximum RPE of 13-14, where 13 is generally defined as 'somewhat hard' (Borg 1998). The Borg scale has been used previously to regulate exercise intensity during cycle ergometry (Buckley et al. 2000). Verbal encouragement was given to participants throughout the protocol. Time was recorded using a Sportline 240 Econosport manual stopwatch (Yonkers, New York City).

Measurements

Three-dimensional motion was measured by an anatomical coordinate system defined by two 14 mm reflective markers positioned over the second thoracic and first sacral vertebra. A single tri-axial accelerometer (52 mm x 30 mm x 12 mm, mass 23 g; resolution 16-bit, full-scale range 16 *g*, sampling at 100 Hz: SABEL Labs, Darwin, Australia) was calibrated as described elsewhere (Lai et al. 2004).

The accelerometer was fixated between the L5 and S1 spinous process (James et al. 2011) using double sided elastic adhesive tape (Medtronic Australasia Pty Ltd, Macquarie, NSW) to reduce unwanted movement. A single accelerometer placed on the low back provides a simple and effective method to examine movement near the body CoM (Kosbar et al. 2014) and is the closest external point to the CoM (Winter et al. 2016) (Figure 1).

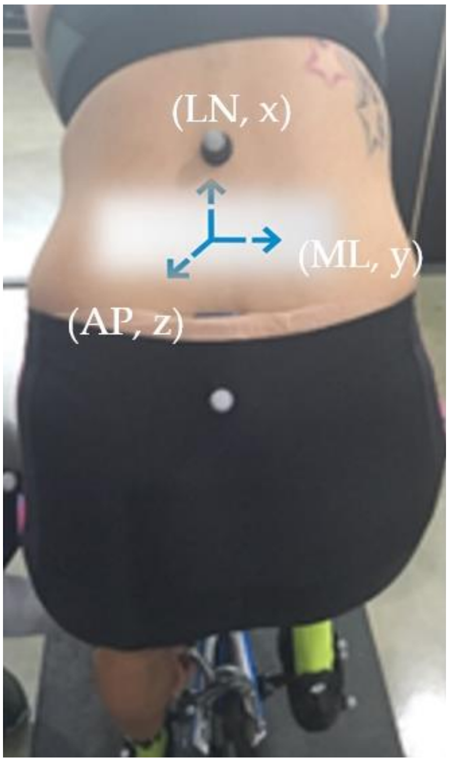


Figure 1: Location of anatomical markers and accelerometer used for all participants. Where LN is longitudinal, ML is mediolateral and AP is anteroposterior.

The accelerometer was positioned to capture acceleration data in three orthogonal planes where longitudinal (LN), mediolateral (ML), and anteroposterior (AP) aligned with X, Y and Z respectively. A postural change will result in a change of displacement and hence give a change in acceleration magnitude. Therefore, the accelerometer provides information about torso motion based on this postural change which enables detection of repetitive movement during cycling. The MoCap measures displacement change and can be converted to an acceleration magnitude for comparison with the accelerometer output.

Data was collected by the sensor and wirelessly transferred to a computer for analysis. The timing points of identifiable acceleration timing peaks in accelerometer X axis (longitudinal direction) and 3D MoCap raw data were manually identified at the commencement and conclusion of each cadence condition and epoch. The reliability of this method may be considered subjective due to the manual process of selecting timing data points, which may cause repeatability differences between researchers (Gleadhill et al. 2016a). This may be area for future research to test the reliability of this method. However, due to the strong correlation of results, any agreement error between methods may be due to errors in manual picking and not relate to the actual differences between sensors and 3D MoCap. Three dimensional MoCap data was collected with an integrated Cycling 3DMA System and software (San Sebastián, Spain) to track the movement of the two reflective markers (Figure 2).

(**a**) (**b**)

Figure 2: (a) Cycling 3DMA (STT Systems) with mounted cycle; (**b**) Participant cycling during protocol

All markers were positioned by the second author to avoid intertester variability. The output data was collected, processed and displayed to filter at a frame rate of 100 Hz/FPS. Reflective markers were not removed between cadence changes to maximize reproducibility of kinematic data. The extracted timing points were transferred to Excel 2007 as the initial stage was to plot the raw data and analyse torso acceleration magnitude characteristics relative to timing points in the longitudinal direction before exploring mediolateral and anteroposterior data.

Statistical analysis was carried out using Analyse-it (Leeds, United Kingdom, version 4.92). Raw MoCap data was converted to acceleration by taking the double derivative of displacement with respect to time. using double derivative calculations.

A Will Hopkins Typical Error of the Estimate validation was implemented to determine timing agreement between both validation methods (Hopkins 2000a). The typical error in raw units was divided by the ± SD of the values of the criterion (MoCAP) predicted by the practical (accelerometer) but evaluated via the correlation coefficient, to allow estimation of confidence limits. Pearson's correlation coefficient (r) is reported. Root Mean Square Error (RMSE) is described as the standard deviation of the residuals (prediction errors). The classification scheme published by Fitz-Gibbon and Morris (1987) was used to interpret the coefficient of determination (r^2). According to these authors, r^2 values <0.4 are interpreted as weak relationships; 0.4~0.6 moderate and >0.6 are strong. The Will Hopkins modified Cohen's scale was used to determine error and bias: <0.1, trivial; 0.1–0.3, small; 0.3–0.6, moderate; 0.6–1.0, large; 1.0–2.0, very large; >2.0, extremely large.

3. Results

The results demonstrate that the accelerometer had high agreement for accurately detecting temporal accelerations of the torso when cycling (Table 1).

The Typical Error of the Estimate (TEE) showed the standardised error to be trivial, demonstrating a small mean bias. The total error (all cadence conditions) illustrates a small positive variability between accelerometer and 3D MoCap. This was supported with correlation and small confidence limits, demonstrating the strength of the proposed method for monitoring temporal kinematics of the torso. The modified Cohen's scale was large for all conditions across the three accelerometric directions (>1.0-2.0).

In order to determine angular changes to longitudinal torso acceleration, a tilt sensing measurement calculation was performed. Data was then scaled into m.s^{-2} where 9.8 m.s^{-2} equates to 1g (Equation 1).

$$\Theta = \sin^{-1}\left(\frac{\mathrm{LN}}{1g}\right) \tag{1}$$

Where Θ represents the angle with respect to the ground, LN is the longitudinal axis and 1g is 9.8 m.s^{-2}

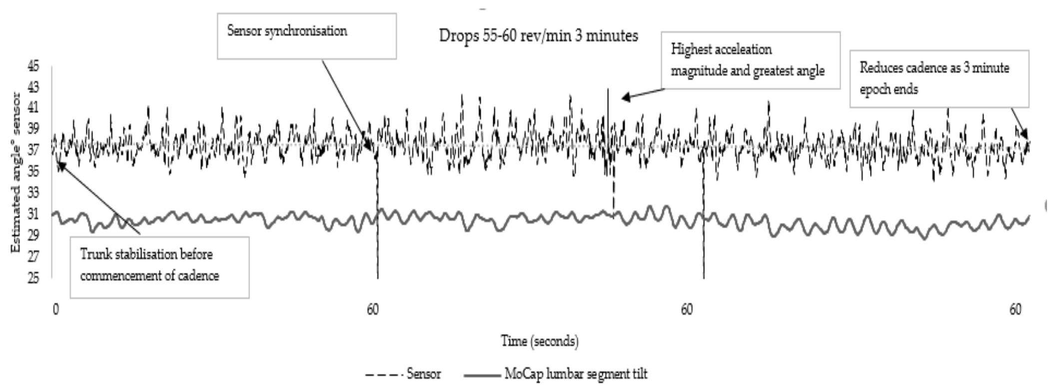
Raw MoCap data pertaining to angular rotations (measured in degrees) that referenced lumbar-segment tilt were then applied to sensor data to obtain angular equivalents and calculated according to equation 1. Changes in both drops and the aerodynamic position were then identified in both methods (Figure 3).

The comparison at the anteroposterior direction showed least error and lowest coefficient of variation at all cadences despite the lowest coefficient of determination compared to other directions (Figure 4).

	Typical error of the estimate (TEE)					Correlation		
Axis (100Hz)	**TEE**	**Upper CL**	**Lower**	**Residuals**	***r***	**Upper**	**Lower**	**Cohen**
Raw longitudinal acceleration (x) in m.s^{-2}								
5–8min (drops)	0.06	0.01	0.30	1.53	0.99	0.96	1.00	>1
8–11min	0.09	0.02	0.52	1.78	0.99	0.89	1.00	> 1
11–14min	0.17	0.03	1.09	2.26	0.98	0.68	1.00	>1
14–17min	0.14	0.03	0.81	2.21	0.99	0.78	1.00	>1
17–20min (aero)	0.14	0.03	0.87	1.90	0.99	0.77	1.00	> 1
All cycles	41.75%	39.22%	53.35%	15.23%	-	20.32%		>1
Raw mediolateral acceleration (y) in m.s^{-2}								
5–8min (drops)	0.18	0.02	0.54	0.18	0.99	0.88	1.00	>1
8–11min	0.12	0.02	0.65	0.54	0.99	0.84	1.00	> 1
11–14min	0.04	0.01	0.22	0.06	0.99	0.98	1.00	>1
14–17min	0.22	0.04	1.64	0.22	0.97	0.52	1.00	>1
17–20min (aero)	0.22	0.07	0.79	0.20	0.97	0.78	1.00	> 1
All cycles	0.16	0.03	0.77	0.17	0.98	0.80	1.00	> 1
CV	43.97%	66.73%	61.82%	139.57	-	19.30%		
Raw anteroposterior acceleration (z) in m.s^{-2}								
5–8min (drops)	0.30	0.06	3.46	0.54	0.95	0.28	1.00	>1
8–11min	0.31	0.06	3.95	0.31	0.95	0.25	1.00	> 1
11-14min	0.35	0.07	7.79	0.86	0.94	0.13	1.00	>1
14–17min	0.23	0.07	0.82	1.04	0.96	0.78	1.00	>1
17–20min (aero)	0.28	0.32	1.00	0.40	0.96	0.32	1.00	> 1
All cycles	0.04	0.11	2.83	0.31	-	0.25	1.00	> 1
CV	13.37%	88.02%	74.31%	44.01%	-	63.41%		

Table 1. Agreement and correlation effects of sensor against MoCap criterion measure

(a)



(b)

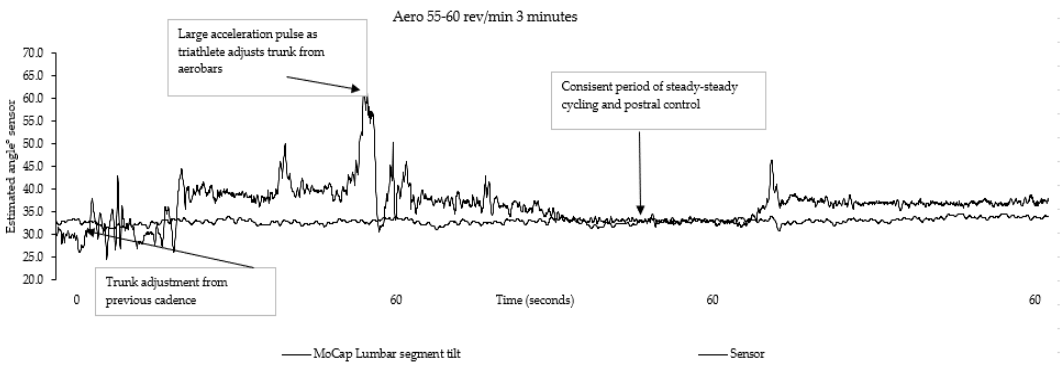
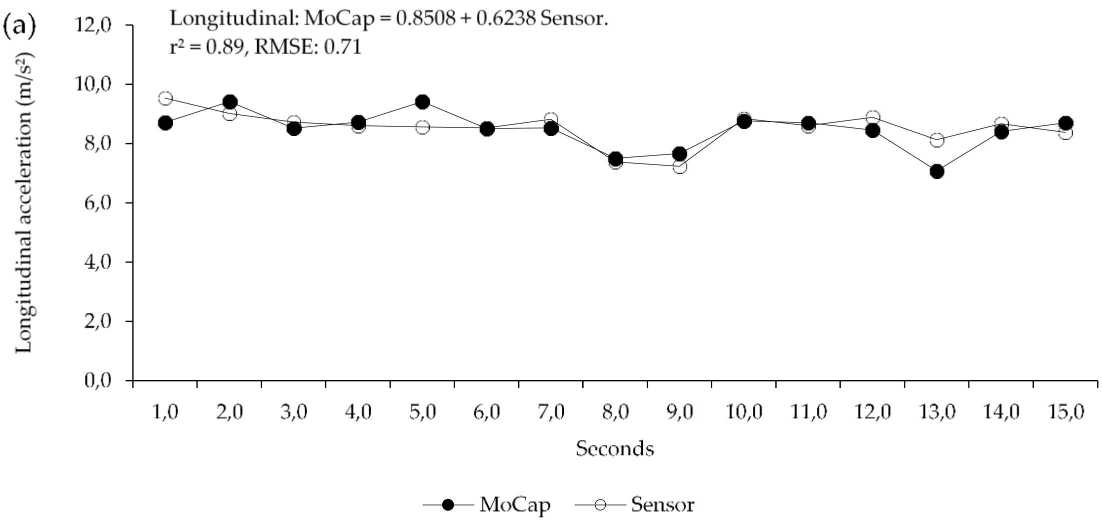
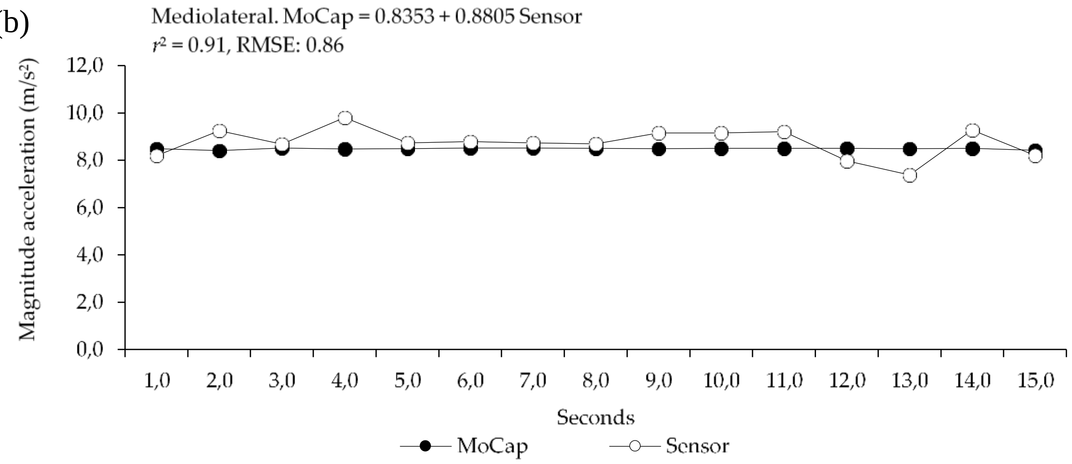


Figure 3: Comparison of torso angles in two hand positions using a triaxial accelerometer (sensor) to the criterion measure of an infrared 3D MoCap for 3-minute cadence condition of 55-60 rev/min¹ (a) drops position; (b) aerodynamic position.

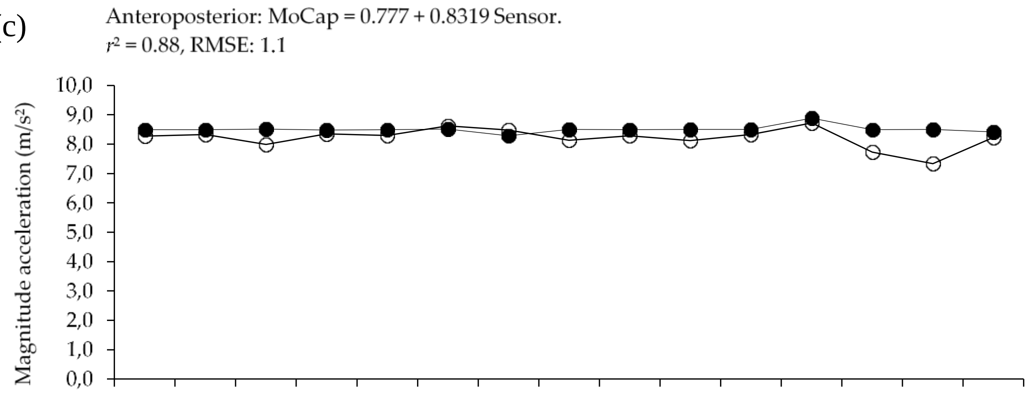
(a)



(b)



(c)



4. Discussion

The aim of this research was to report the temporal validity of using a triaxial accelerometer to determine whether an accelerometer measured torso kinematics during cycling at varied cadence at the same time as a 3D MoCap system. Temporal validity was also determined in two different styles of cycling position.

This validation of accelerometry is a contribution to the continuation of wearable software to support human assessment and dynamic movement. The significance of this validation translates to numerous applications, where overall timing and the time series of events occur are measured, such as gait symmetry (Lee et al. 2010) and resistance exercises (Gleadhill et al. 2016b). Despite these advancements, no known validation has been completed to measure the agreement of accuracy of these measures relative to temporal accelerations of the torso in cycling as well as changes to cycling position. The significance of this research is that results support the practical use of accelerometry in dynamic and cyclic human movement applications. The methods used in this research were selected due to their ease of use and the reviewed success of these procedures to quantity the variables they are designed to measure (Hopkins 2000b).

Cycling involves mostly motion of the lower limbs, isolated from any movement occurring up the chain. However, movement of the torso is relevant for the cyclist in achieving a position that offers improved aerodynamic efficiency whilst limiting excessive saddle movements. For example, greater torso flexion was observed in triathletes (Bini et al. 2014). In this instance, there are similarities with our study as variations in torso position corresponded with an increase in longitudinal acceleration magnitude due to a change in cycling position (Figure 3). This demonstrates that accelerometry is capable of detecting changes in hand position and associated changes in magnitude. This could be of benefit when forming a training intervention given excessive torso acceleration could indicate poor core strength and a weakness to maintain a precise cycling position.

Mediolateral and longitudinal acceleration data demonstrated the greatest accuracy of torso kinematics with high levels of agreement and correlations between data (Table 1). While it would be assumed that all orthogonal axis of data should compare similarly in an agreement, this was not seen in this study (Figure 4). Anteroposterior acceleration displayed less agreement when compared to MoCap. The greater variability may be attributable to the accelerometer measuring positional displacement changes of participants simultaneously moving in multiple directions (i.e., both up and down whilst adjusting mediolateral position) combined with the manual process undertaken to identify commencement and conclusion of cadence conditions. Furthermore, anteroposterior acceleration in cycling is in the forward direction which has a lower magnitude than the mediolateral torso peaks and longitudinal increases when changing cycling position, therefore, directional acceleration measured by the sensor may be blending into other accelerations. The blending of forward and rotation acceleration components may explain why the direction of movement measured by sensors provides a variation to data collected by the motion analysis system. Despite these limitations, the proportion of variance determination was strong between both systems which is indicative of significant correlation. Specific to cycling, previous studies have found that during strenuous pedalling the anteroposterior direction seems to be sensitive for stability decreases (Wiest et al. 2011). A measure of cycling stability according to musculoskeletal state was the centre of pressure sway velocity which has been found to correlate strongly with acceleration of the CoM (Masani et al. 2014). Whilst cyclists in the current study did not perform strenuous activity or a cycling to exhaustion protocol, the significance of these results are relevant given the recreational level of the athletes and that a low level of core muscle strength could have caused supplementary movement of the upper body and thus greater magnitudes in temporal torso kinematics. This means that the variation observed in torso acceleration magnitudes could relate to core stability and each individual's functional movement ability level. The results of our study support the previous accounts of beneficial effect of the core stability training on the cycling specific stability (Asplund et al. 2010).

Accelerometer data was analysed for participants cycling at familiar cadences in two different yet accustomed cycling positions. The timing of different torso acceleration magnitudes in cycling may be an indicator of technique. An accelerometer monitoring each body segment may provide comprehensive dynamic information about which body segments are moving and how they are moving at different times throughout the crank cycle. The implication of these findings is that past research was supported and future research can be designed to monitor upper body technique with accelerometers, with strong evidence to support this analytical method. It is assumed that other outputs including measurements of bicycle kinematics (roll and steer) and rider movements (joint angles and rider lean) (Cain 2016; Xu et al. 2015) can be used in future research to analyse temporal torso kinematics due to changes in the pedalling cycle in different surroundings.

These past studies support that accelerometery can accurately monitor upper body postural changes via torso acceleration with a tri axial accelerometer, further warranting the validation and assessment of these outputs, specific to changes in cadence and body position. Ideally, different saddle positions (e.g., standing, seated) and terrain (uphill, downhill) with full body accelerometer mark up could be compared to comprehensively validate all possible sensor outputs to measure and classify differences in technique and common pedalling variances. This was outside the scope of the current study and is an area for future research.

5. Practical Applications.

A triaxial accelerometer is capable of measuring temporal kinematics of torso CoM acceleration magnitudes during a varied cadence cycling protocol as well as cycling position. The strength of using an accelerometer is that it can appear to monitor movement patterns with as much confidence as 3D MoCap for timing measures, whilst remaining highly practical with possibilities to provide feedback in real time. Additionally, accelerometers are easily configurable to any cyclist; does not impede bicycle or cyclist motion; provides estimates of meaningful kinematic variables; and is not confided to a laboratory of capture volume. The important outcome and strength of this study is that it filled a gap in the literature for validating sensor timing measures in cycling, and provided evidence and interpretations to support using accelerometers to monitor differences in torso positions. This research provided support for past sensor and accelerometer applications, a foundation for future applications, and future research recommendations to expand or repeat this research. Therefore, accelerometers have the potential for practical applications to lead to significant benefits for possible injury prevention and performance in cycling.

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Conflicts of Interest: The authors declare no conflict of interest.

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