USING WEARABLE TECHNOLOGY TO DETECT CHANGES TO TRUNK POSITION AND POWER IN CYCLING

Stuart A Evans[1], Daniel James[1], David Rowlands[2], James B Lee[1]

SABEL Labs, College of Health and Human Science, Charles Darwin University, Darwin, Australia[1]
School of Engineering, Griffith University, Nathan, Queensland, Australia[2]

This pilot study aimed to quantify the change of CoM acceleration when the trunk position and the power output was varied during indoor cycling. Triathletes (n=4) performed a varied indoor cycling power protocol in their natural training environment whilst wearing a trunk (sacrum) mounted wearable device containing a triaxial accelerometer. Mediolateral acceleration increased 26% as athletes moved from the drops position, to the aerodynamic position and back to the drops position. Although longitudinal acceleration increased 8%, minimal differences in anteroposterior CoM magnitude were observed. Power output was found to have an effect on both the mediolateral and longitudinal acceleration alongside increases in RPE. The results indicate that accelerometers may be effective in monitoring changes to trunk position and power output.

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

INTRODUCTION: Cycling performance is a fine balance between biomechanical effectiveness and physiological efficiency. The use of technology in triathlon has increased with triathletes and coaches utilising a variety of devices to analyse and improve skill acquisition, training efficiency, and performance enhancement. The aerodynamic advantage from a reduced frontal projection area (FPA) when the cyclist assumes a forward crouched upper body position is well established (Capelli et al. 1993). This crouched position changes the angle of the trunk and the body's Centre of Mass (CoM) compared to the more upright position adopted by road cyclists. In this paper the term CoM will only refer to the centre of mass of the body of the cyclist.

The CoM can be defined as the unique point where the weighted position of mass is equally distributed. This is the point to which a force may be applied to a cause linear acceleration. However, the importance of trunk angle in cycling is often overlooked with the focus typically on lower limb kinematics that result in force application to the pedals (Bini et al. 2013). Nevertheless, trunk angle affects leg kinematics, associated muscle activity and limb length (Savelberg et al. 2003).This is supported by Rannama et al. (2017) who observed that the forward shift of the CoM meant that the cyclist's body is less supported by the saddle and therefore more stabilisation from the trunk muscles is needed to control the force on the axis defined from handlebars to pedals. Such conclusions suggest that manipulation of the trunk position can affect cycling economy and biomechanical efficiency. This is especially applicable to triathletes in which trunk angle is often manipulated in order to improve streamlining and to maintain postural stability.

Cycling assessment is often confined to laboratories with participants typically cycling on an ergometer. While laboratory systems provide accurate measurements, they usually only provide snippets of information, rarely allowing for continuous analysis of the prolonged cycling seen in the training environment. Advances in wearable technology (wearables) have allowed the creation of small, accurate, and low-cost devices containing inertial sensors capable of accurately sensing motion (Rowlands et al. 2013). Due to their small size, they can be worn with negligible discomfort and used to measure activity and performance. An advantage of these devices is their ability to be used in various situations as data can be wirelessly transmitted or stored in the device. Therefore, it seems logical to assess changes to the magnitude of CoM acceleration using wearable technology containing a triaxial accelerometer since the wearable does not compromise the power output of the triathlete. Thus, we designed the presented pilot study to quantify the magnitude of CoM acceleration of a group of triathletes in their natural training environment whilst investigating if changes to the trunk position and changes to the power output results in corresponding changes to the magnitude of the CoM acceleration.

METHODS: Four (3 male & 1 female) healthy and experienced triathletes from a local club (age: 31 ± 8.8yrs: body mass 77 ± 8.5kg: height 178 ± 0.5.9cm: weekly training volume: 8.7 ± 3.71hrs) participated in the project having given informed consent (approved by the University Research Ethics Committee:HREC19028). The study analysed two cycling positions commonly used in triathlon which were the drops position and the aerodynamic position. Minimal instruction was given to the participants as they settled into their preferred positions. In the first position, the participant's hands moved down onto the drops (DP) with slight elbow flexion and trunk inclination. In the second position, the triathlon racing position was adopted using the tri (aero) - bars (AO) which has been described as being, "elbows on triathlon aero bars, arms inside the projected frontal body area, torso flat, head tucked low between the shoulders" (Bassett et al., 1999). Tests were performed on an indoor cycle (Schwinn Carbon Blue, Schwinn Bicycles, Dorel Industries, Inc, Washington, USA) that consisted of a belt drive (122 cm x 109 cm x 51 cm). Strap-in pedals were used to replicate their training position.

Participants warmed up for 2 min using their customary routine. Each participant completed the following test schedule. The participants assumed a DP and cycled for 2-5 minutes at their self-selected cadence (SSC). From minutes 5-8 participants adopted an AO position whilst cycling between 0-150 Watts (W). Participants remained in the AO position from minutes 8-11 and 11-14 (152-205 W) to reflect their typical training situation. Minutes 15-17 were between 206-246 W.

At minutes 17-20 participants assumed a DP position and cycled at their SSC. Ratings of perceived exertion (RPE) were obtained at each change of power to control intensity and to limit fatigue. Consequently, intensity was managed to a maximum RPE of 13-14, where 13 generally defined as 'somewhat hard' (Borg, 1988). Power feedback was provided via the display unit on the cycle. When a change of power was required, participants adjusted the gearing in order to maintain a consistent RPE.

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 positioned at the sacrum as described by Rowlands et al. (2013). The sensor was positioned to capture acceleration data in the three orthogonal planes where longitudinal (LN), mediolateral (ML), and anteroposterior (AP) aligned with the accelerometer's X, Y and Z respectively. Data was collected by the sacrum mounted sensor and wirelessly transferred to a computer for analysis. Calibration of the accelerometer prior to the measurement was performed as described by Lai et al. (2004).

Cycling events and corresponding power changes were identified in the raw accelerometer data to ensure no loss or timing shift. The longitudinal acceleration was used to identify a change in posture of the trunk and was identified at the point where the acceleration magnitude began increasing towards its largest impact peak. The mediolateral acceleration was used to identify pedal strokes. For each power range, the average CoM acceleration magnitude for each axis (X, Y, Z) was determined over each 3 minute epochs.

A two factor ANOVA (power x epoch) was used with an alpha (α) level set at 5% ($P \leq 0.05$) to compare participant changes to CoM acceleration in three orthogonal axes and determine whether CoM acceleration variables differed due to alterations in power output. The relationship between power output and CoM acceleration were analysed using a correlation coefficient (*r*). The variance between power output was described using the coefficient of variation ((SD/mean x 100) with focus on the AO position as it is used during triathlon events.

RESULTS

The power outputs indicated that the magnitude of CoM acceleration was significantly different (all $P \leq 0.05$) when considered from a kinematic perspective. Orthogonal (X, Y, Z) acceleration against the power range protocol and RPE were statistically significant, supporting the decision to analyse CoM acceleration via a trunk mounted wearable. Longitudinal acceleration was found to increase relative to power with the largest overall magnitude observed at 206-246 W (-0.25 ± 0.42) and the final SSC (-0.25 ± 0.37). Mediolateral CoM acceleration peaked highest at 152-250 W (-5.52 ± 10.1) whilst anteroposterior acceleration was highest at 152-250 W (-13.6 ± 10.6) and the final SSC (-14.2 ± 7.1). RPE steadily increased with time, power and CoM acceleration in each axis. Longitudinal and mediolateral CoM acceleration across the 2 cycling positions (related to trunk angle) were significantly different. Minimal overall change was observed in anteroposterior magnitude across durations and power outputs. The results showed a general linear relationship between longitudinal and mediolateral CoM acceleration relative to power output with concomitant increases in CV from 152-250 W (Table 1).

Table 1: Power range protocol and magnitude of timeseries CoM acceleration for all participants in x, y and z axis in m/s²

Minutes Power Trunk	**2-5 SSC DP**		**5-8 <150 W AO**		**8-11 >152-205 W AO**		**11-14 > 152-205 W AO**		**14-17 >206-246 W AO**		**17-20 SSC DP**	
	p	*r*	*p*	*r*	*p*	*r*	*p*	*r*	*p*	*r*	*p*	*r*
LN	0.07	0.8*	0.07*	0.8*	0.03*	0.9*	0.03*	0.9*	0.01*	0.9*	0.01*	1.0*
ML	0.02*	0.9*	0.79	0.4	0.81	0.6*	0.83	0.2	0.05*	0.9*	0.01*	0.9*
AP	0.75	0.1	0.71	0.5	0.10	0.7	0.12	0.6	0.06	0.71	0.08	0.6
CV (%)	22.65		31.74		15.30		24.51		26.1		21.01	

LN (longitudinal, x); ML (mediolateral, y); AP (anteroposterior, z); W = Watts; * Significant at (α) ≤ 0.05. Data are n=4 for DP and AO positions. * correlation coefficient (*r*) indicative of relationship between variables. CV = coefficient of variation

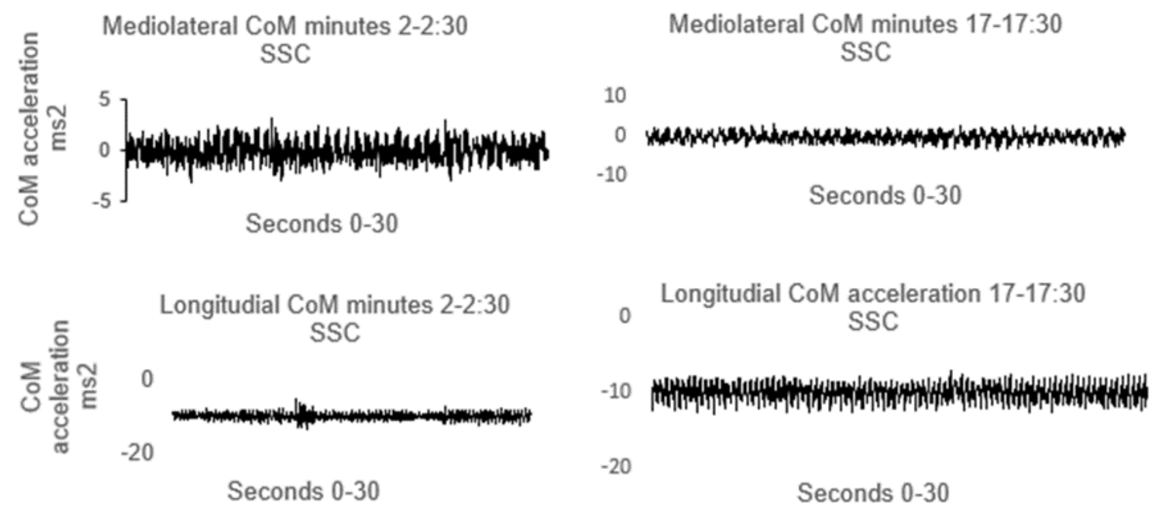

Figure 1: Representative of raw timeseries data for CoM acceleration obtained for one participant during SSC in mediolateral and longitudinal axes. Plot shows 30 second duration from minutes 2-2:30 and 17-17:30 in m/s²

DISSCUSSION: This pilot study found a greater increase in the magnitude of CoM acceleration in two orthogonal axes. Peak power was gradually increased throughout the protocol which corresponded with significant mediolateral CoM acceleration ($P \leq 0.001$) increase. This was represented by a 26% increase from onset to completion. The sensitivity of the mediolateral acceleration suggests occurrences of medio-lateral sway of the trunk. This shares similarities with prior findings. Abt et al. (2007) detected that fatiguing trunk muscles

have significant compensatory effect on cyclist movement kinematics without alterations in pedalling. Although pedalling technique was not monitored, RPE was checked at regular intervals to monitor the onset of fatigue since a change in trunk lean can occur early in the fatigue process.
The overall magnitude of CoM longitudinal acceleration increased 8%. Changes in longitudinal position when cycling are not necessarily surprising as participants may adjust position to reduce projected frontal area to adapt to the cycling conditions if cycling outside as well as for comfort. Although relatively small, the increased acceleration and strong correlation could suggest low core stability and ability to control the trunk. Rannama et al. (2017) observed that a low level of core muscle strength can cause more movement of the upper body. Specifically, cyclists had tendency of a more pronounced inclination and lateral movement of the bicycle at all intensity levels. This was supported by the findings of Costes et al. (2015) that along with an increase in power, acceleration forces directed to pelvis and upper body will increase.
Despite observable changes in two axes, the value reported in the present study for anteroposterior acceleration (32.62 ± 2.5) was modest in comparison to the mediolateral and longitudinal axes. In analysing cycling, it would be expected that movement in this direction would remain quite stable due to the supported trunk position.
Although the CoM is an imaginary point, it is possible to use this position to approximate the entire body's CoM. Even small adjustments to CoM magnitudes may contribute to improved biomechanical efficiency Our findings are comparable to Jobson et al (2008) in that body position can significantly affect power output in endurance cycling, reflected in increases to CV when power increased. The accelerometer allows for a vast amount of data to be extracted. This information may be of practical benefit for coaches, cyclists and triathletes to improve performance, monitor postural stability as well as delay the onset of fatigue.

CONCLUSION: This study identified that a wearable can detect changes to the magnitude of CoM acceleration when trunk position and power are altered. In order to obtain real-time data reflective of a typical training session, the trunk-mounted wearable used in this study inferred changes to the magnitude of CoM acceleration during a varied indoor cycling power protocol in the triathlete's natural training environment. This is particularly relevant for coaches who seek a low-cost and portable method to help athletes improve postural control whilst cycling. Results indicate that accelerometers can be applied to monitor trunk kinematics in cycling and can indicate change in power/cadence.

REFERENCES

Abt, J. P., Smoliga, J. M., Brick, M. J., Jolly, J. T., Lephart, S. M., & Fu, F. H. (2007). Relationship between cycling mechanics and core stability. *The Journal of Strength & Conditioning Research, 21*(4), 1300-1304.
Bassett, D.R., Kyle, C.R., Passfield, L., Broker, J.P., & Burke, E.R. (1999). Comparing cycling world hour records, 1967-1996: modeling with empirical data. *Med Sci Sports Exerc*. *31*(11),1665-76.
Bini, R.; Hume, P.; Croft, J.L.; & Kilding, A. (2013). Pedal force effectiveness in cycling: A review of constraints, *J Sci Cycling*, ,2, 11-24.
Borg, G. (1993). Borg's perceived exertion and pain scales. Human Kinetics, Champaign, IL, US.
Capelli, C., Rosa G, Butti F, Ferretti G, Veicsteinas A, & di Prampero, P.E. (1993). Energy cost and efficiency of riding aerodynamic bicycles. *Eur J Appl Physiol Occup Physiol,* 67,144–149.
Costes, A., Turpin, N. A., Villeger, D., Moretto, P., & Watier, B. (2015). A reduction of the saddle vertical force triggers the sit–stand transition in cycling. *Journal of Biomechanics*, *48*(12), 2998-3003.
Jobson, S. A., Nevill, A. M., George, S. R., Jeukendrup, A. E., & Passfield, L. (2008). Influence of body position when considering the ecological validity of laboratory time-trial cycling performance *J Sport Sci*, 26, 1269-1278.
Lai, A., James, D.A., Hayes, J.P., et al. (2004). Semi-automatic calibration technique using six inertial frames of reference, *SPIE*, 5274, 531–42
Rowlands, D.D., James, D.A., & Lee, J.B. (2013). Visualization of wearable sensor data during swimming for performance analysis, *Sport Tech*, *6*(3), 130-136.
Rannama, I., Pedak, K., Bazanov, B., & Port, K. (2017). Cycling specific postural stability during incremental exercise: The relationship with cyclists functional movement screen score. *Journal of Human Sport and Exercise*, *12*(1), 83-95.

Savelberg, H.; Van de Port I.G.L.; & Willems, P.J.B. (2003). Body configuration in cycling affects muscle recruitment and movement pattern, *Appl Biomech*, 19, 310-324.