A method for tracking centre of mass displacement during non-seated cycling using an inertial sensor

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Abstract

Instantaneous crank power does not equal total joint power if a rider's centre of mass (CoM) gains and loses mechanical energy. Thus, estimating CoM motion and the associated energy changes can provide valuable information about cycling performance. To date, an accurate and precise method for tracking CoM motion during outdoor cycling has not been validated. **Purpose:** To assess the suitability of an inertial measurement unit (IMU) for tracking CoM motion during non-seated cycling by comparing vertical displacement derived from an inertial sensor mounted to the lower back of the rider to an attached marker cluster and to a kinematic estimate of vertical CoM displacement from a full-body musculoskeletal model (Model). **Methods:** IMU and motion capture data were collected synchronously for 10 seconds while participants ($n = 7$) cycled on an ergometer in a non-seated posture at three power outputs and two cadences. A limits of agreement analysis, corrected for repeated measures, was performed on the range of vertical displacement between the IMU and the two other measures. A total of 303 crank cycles were analysed. **Results:** The IMU measured vertical displacement of the marker cluster with high accuracy (1.6 mm) and precision (3.5 mm) but substantially overestimated the kinematic estimate of rider CoM displacement. **Conclusion:** We interpret these findings as evidence that a single IMU placed on the lower back is unsuitable for tracking rider CoM displacement during non-seated cycling if the linearly increasing overestimation is unaccounted for.

Keywords: inertial measurement unit, motion analysis, joint power, bicycling

1. Introduction

During cycling, changes in mechanical energy of the rider's centre of mass (CoM) cause a discrepancy between instantaneous crank power and total joint power (van Ingen Schenau & Cavanagh, 1990). Research shows that a rider's CoM gains and loses mechanical energy at rates of $4.5 \text{ W} \cdot \text{kg}^{-1}$ during non-seated cycling (Wilkinson et al., 2020a). These changes in CoM mechanical energy increased peak crank force and peak crank

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power. Thus, tracking a cyclist's CoM position provides valuable information about rider-bicycle interactions.

Optical motion capture systems provide a gold-standard measurement of a rider's CoM position during ergometer, treadmill, and over-ground cycling. But, there are limitations to each of these experimental setups. Ergometers allow the collection of many cycles per trial, but they constrain bicycle dynamics. This constraint alters the preferred movement pattern of the rider and may affect performance (Wilkinson et al., 2020b). Cycling on a treadmill provides ecological validity, but problems arise for some tasks such as sprinting. One problem is the danger of not matching belt velocity with wheel velocity if the rider accelerates. It is possible to capture over-ground cycling, but the calibrated volume of the camera system will limit the number of captured cycles. Thus, a method for tracking a rider's CoM motion when motion capture is not feasible would broaden the scenarios for examining rider-bicycle interactions.

Examining rider-bicycle interactions outside the laboratory is important because bicycle dynamics alter the preferred movement pattern of cyclists (Wilkinson et al., 2020b). For example, changes in CoM energy during non-seated cycling are greater when riders use a preferred amount of bicycle lean compared to when self-restricting lean (Wilkinson et al., 2020b). Research also shows that changes in CoM energy during non-seated ergometer cycling increase in response to increasing crank power and decreasing cadence (Wilkinson et al., 2020a). Thus, tracking a rider's CoM motion outside the laboratory would confirm whether these results extrapolate to over-ground cycling.

Full-body kinematic analyses provide accurate estimates of CoM displacement and velocity during non-seated cycling (Wilkinson et al., 2020a) and many other forms of locomotion (Gard et al., 2004; Hamner & Delp, 2013). These analyses use segment positions and inertial parameters to compute CoM position and changes in total mechanical energy (potential + kinetic) (Eng & Winter, 1993). Assuming constant mass, potential energy changes depend on vertical displacement, while kinetic energy changes depend on velocity in three dimensions. However, because the CoM location of standing humans is anterior to the lumbo-sacral joint, there is evidence that a single sacral marker provides a convenient and reliable approximation of vertical CoM displacement during walking (Gard et al., 2004; Thirunarayan et al., 1996; Saini et al., 1998). Pioneering research on non-seated cycling used a similar simplified method to estimate the pattern of vertical CoM displacement during non-seated cycling (Soden & Adeyefa, 1979). Their results showed good agreement with more recent full-body kinematic analyses (Wilkinson et al., 2020b). Thus, tracking a single sacral marker may provide a suitable estimate of CoM position and mechanical energy during non-seated cycling.

Previous research has assessed the validity of using inertial measurement unit (IMU) data to provide an accurate measure of displacement and orientation of a point representing the CoM of a thoroughbred horse during locomotion (Pfau et al., 2005). When compared to optical motion capture, the sensor error for tracking the trunk movement of the horse in each axis was less than 5% during walking and 7% during a trot or canter. Here, we assess the validity of an IMU mounted near the sacrum for measuring vertical CoM displacement and associated energy changes of cyclists riding in a non-seated posture. Our assessment compares the derived vertical displacement of the IMU to an attached marker cluster; tracked with gold-standard optical motion capture technology, and to a full-body kinematic estimate of vertical CoM displacement.

53 2. Methods

54 Seven recreational cyclists participated in this study (5 men and 2 women, age: 24 ± 13
55 yrs, height: 1.75 ± 0.07 m, mass: 71 ± 11 kg). Each participant gave written informed
56 consent before participating in the study according to the procedures approved by the
57 Human Ethics Committee of The University of Queensland.

58 2.1. Experimental design

59 Participants performed all trials on the same cycling ergometer (Excalibur Sport,
60 Lode BV, Groningen, The Netherlands). The saddle height and handlebar position were
61 matched to each participant's accustomed cycling position. Participants wore the same
62 model of cycling shoes in their preferred shoe size (SH-R070, Shimano, Osaka, Japan)
63 that clipped into the pedals (SH-R540, Shimano, Osaka, Japan).

64 At the beginning of the session, participants warmed-up by cycling for 5 minutes at
65 100 W at their preferred cadence. Participants then performed five maximal 5-s sprints
66 in a seated posture, each separated by 3 minutes of rest. These sprints determined
67 each participant's instantaneous maximal power output ($P_{max.i}$), which was used to
68 individualise power output in the experimental trials. The experimental trials consisted of
69 six 10-s bouts of non-seated cycling at three different power outputs (10%, 30%, and 50%
70 of $P_{max.i}$) and two different cadences (70 rpm and 120 rpm), each separated by 3 minutes
71 of rest. Each participant completed the experimental trials in a randomised order. For all
72 experimental trials, the ergometer was set to Hyperbolic mode, which ensured that the
73 power output remained constant independent of cadence. Thus, participants maintained
74 the target cadence using feedback from the visual display on the ergometer for 10 seconds.

75 2.2. Optical motion capture

76 Reflective markers and lightweight marker clusters were secured to the skin using
77 a combination of double-sided tape and self-adhesive bandage at previously described
78 locations suitable for measuring full-body kinematics (Wilkinson et al., 2020c,a) (marker
79 locations are shown in Figure 1). The three-dimensional position of each marker was
80 collected for 10 seconds at 200 Hz using an eight-camera, opto-electric motion capture
81 system (Oqus, Qualisys AB, Gothenburg, Sweden). Motion capture data was processed
82 using Qualisys Track Manager software (2019.1, Qualisys AB, Gothenburg, Sweden)
83 before being exported to Matlab (R2020a, MathWorks Inc., Natick, MA).

84 2.3. IMU

85 An IMU (BlueThunder Sensor, iMeasureU, Auckland, New Zealand) attached to a
86 rigid cluster of reflective markers was secured to the rider's skin at the intersection of
87 Tuffier's line and the midline of the lumbar spine ($\sim L4$ spinous process) using double-
88 sided tape (See Figure 1). The markers were attached such that the plane created by the
89 markers corresponded to the XY plane of the IMU. A self-adhesive bandage was wrapped
90 around the cluster and torso of the rider to limit soft-tissue artefact. The IMU Research
91 Application (iMeasureU, Auckland, New Zealand) was used to collect IMU data at 100
92 Hz via Bluetooth to an iPad (Apple, California, United States). The IMU contained a tri-
93 axial accelerometer ($\pm 16g$), triaxial gyroscope ($\pm 2000^\circ s^{-1}$), and triaxial magnetometer
94 ($\pm 1200\mu T$) with micro-electro-mechanical systems (MEMS) technology connected to a

small circuit board. Thus, the sensor logged acceleration, angular velocity, and magnetic flux data in the three orthogonal planes. IMU and motion capture data were synced by tapping the sensor before the start of the trial with a motion capture calibration wand. The collision between the sensor and wand marker provided a synchronised spike in their respective resultant accelerations.

The approach used in this study for calculating orientation and linear displacement of the IMU have been described previously (Pfau et al., 2005). Based on previous research we expected the inertial sensor to accurately track the attached marker cluster (Madgwick et al., 2011; Pfau et al., 2005), hence only one model of IMU was tested. However, the IMU used here did not output orientation, and hence we used a freely available sensor fusion algorithm to determine sensor orientation (Madgwick et al., 2011). We rotated the IMU data about the vertical axis so that the heading direction matched the motion capture system in the static trial, thereby aligning the orientation of the two systems. During the cycling trials, the quaternion orientation was used to rotate the linear acceleration and angular velocity of the IMU from its local-coordinate system to the global-coordinate system defined by gravity and due north. Acceleration due to gravity was removed from the vertical component of the acceleration signal. The linear accelerations were double integrated to calculate velocity and displacement. A short window (one cycle) was implemented for drift correction under the assumption that there would only be small cycle-to-cycle variations in movement pattern because participants were cycling at a constant power output and cadence. At each time point, the position of the rigid cluster was approximated by creating a virtual marker in the centre of the cluster; calculated as the mean position of the cluster markers. The notation used in this study has been provided in Table 1.

[Figure 1 about here.]

2.4. Musculoskeletal model

Kinematic analysis was performed using a previously developed full-body musculoskeletal model (Rajagopal et al., 2016) within OpenSim software (Delp et al., 2007). A static trial was collected with each participant standing in a standard-anatomical posture to scale the model’s segment lengths and inertial parameters to each participant. The scaled model and motion capture data were used to run inverse kinematics within OpenSim, which determined the whole-body CoM location at each frame. A detailed description of the scaling and inverse kinematic process can be found elsewhere (Wilkinson et al., 2020c).

2.5. Statistical analysis

Sensor performance was quantified for each experimental condition as the root-mean-square (RMS) error in each Euler parameter describing the yaw, pitch, and roll components of the IMU angular velocity compared to the attached marker cluster body. For each trial, the mean RMS error was calculated from the RMS error in each frame over the 10 seconds of data.

Agreement in the range of vertical displacement between the IMU and the two other measures was assessed within each condition by applying a Limits of Agreement (LoA) approach (Bland & Altman, 1999). LoA analyses were corrected for repeated measures and encompassed accuracy (Bias), precision (standard deviation), average error ($\frac{bias}{range}$),

and maximum error ($\frac{\text{standard deviation}}{\text{range}}$). Different LoA calculations were carried out depending on whether a significant linear trend was identified in the distribution of differences between the IMU and the two other measures (IMU – measure) as a function of the mean range of vertical displacement ($\frac{IMU+measure}{2}$). If a linear trend was identified, bias was calculated as the equation of the linear model, rather than a constant value (Bland & Altman, 2007). Sphericity of the data was checked using Mauchly’s test and non-parametric analyses were used when necessary.

A simple linear regression was performed across all experimental conditions to identify the linear trend between the range of vertical IMU displacement to the two other measures. All statistical analyses were performed in Matlab.

[Table 1 about here.]

[Figure 2 about here.]

3. Results

To avoid any bias between the two cadence conditions at each power output, the number of cycles analysed from each participant was matched between all conditions. This meant that eight crank cycles were processed for each of the seven participants in each of the six conditions. Due to data dropout, 12 cycles were discarded, while a further 21 cycles were identified as outliers (> 3 scaled median absolute deviations from the median) (Leys et al., 2013) and removed from the analysis. Thus, a total of 303 cycles were analysed (~ 7 cycles per condition per participant). The distribution of differences between the IMU and the two other measures violated Mauchly’s test for sphericity, thus non-parametric analyses were performed.

3.1. Sensor performance

The results of the sensor performance analysis are summarised in Table 2. Group mean ($\pm$ SD) RMS errors between the IMU and marker cluster body are reported for each condition. The data shown in Figure 2 illustrates how well the IMU was able to track the orientation and motion of the marker cluster.

3.2. IMU vs optical motion capture

Linear regression results and Bland-Altman plots are presented in Figure 4A and B, respectively. Across all conditions, there was high agreement between the IMU and marker cluster range of vertical displacement (See Table 3). Differences between the IMU and marker cluster’s range of vertical displacement showed no significant linear trend. On average, the IMU overestimated the cluster results by 1.6 ± 3.5 mm (accuracy $\pm$ precision), which equated to an average error of $1.8 \pm 3.9\%$. Figure 3 illustrates how well the IMU was able to track the marker cluster’s vertical displacement.

3.3. IMU vs musculoskeletal model

Linear regression results and Bland-Altman plots are presented in Figure 4C and D, respectively. Across all conditions, agreement between the IMU and marker cluster range of vertical displacement decreased at higher ranges of vertical displacement (See Table 3). Differences between the IMU and musculoskeletal model's range of vertical displacement showed a significant linear trend. Across all conditions, the IMU overestimated the model's results, especially at higher ranges of vertical displacement. Figure 3 illustrates the discrepancy between the IMU's vertical displacement and the model's vertical CoM displacement.

[Figure 3 about here.]

[Table 2 about here.]

[Figure 4 about here.]

[Table 3 about here.]

4. Discussion

We found that the IMU overestimated the kinematic estimate of vertical CoM displacement and that this overestimation increased linearly as the measured range of vertical CoM displacement increased. The IMU identified the trend that the range of a rider's vertical CoM displacement during non-seated cycling increases in response to increasing power output and at lower cadence, however, it underestimated the rate of this increase compared to full-body kinematics. Thus, the IMU is likely to be more accurate at lower power outputs when less CoM displacement occurs. However, the non-seated posture is typically used for producing high-power outputs during climbing and sprinting.

We interpret our findings as evidence that a single IMU can track the vertical displacement of an attached marker cluster with high accuracy and precision, but unacceptable discrepancies exist between the vertical displacement of the IMU and the kinematic estimate of rider CoM displacement. Discrepancies between the single IMU and rider CoM increased with increasing power output and decreasing cadence, but these increases were systematic and may be able to be accounted for using linear regression.

The overestimation of the kinematic estimate of rider CoM displacement suggests that a rider's lower back goes through a larger range of vertical displacement during each crank cycle than their CoM. This makes intuitive sense as the motion of body segments other than the torso have a significant influence on the whole-body CoM (Kautz & Neptune, 2002). Based on our findings, it appears that the difference between lower back and CoM displacement increases at higher power outputs. This suggests that negative work done by the arms (Turpin et al., 2016; Wilkinson et al., 2020a) likely acts on the torso to decouple lower back movement from the torso CoM. Thus, a single IMU on the lower back cannot account for the entirety of changes in CoM position due to motion of the legs and rotation of the torso.

We cannot rule out that a portion of the disagreement between the IMU and model was due to the processing of the IMU data, however, the accuracy between the IMU and cluster suggests that processing errors were minimal. It is possible that small offsets

215 were present due to the manual synchronisation process between the IMU and motion
216 capture data, which could be avoided in future studies by using hardware-synchronised
217 systems. Another possible source of error was the misalignment between the IMU and
218 marker cluster. The IMU was attached using a silicone putty which prevented any relative
219 movement but does not guarantee that the alignment to the plane created by the marker
220 clusters was exact. A misalignment between the IMU and marker cluster would cause
221 a constant offset in the orientation data, however, small errors in orientation are known
222 to have a minimal influence on linear displacements in a global-coordinate system (Pfau
223 et al., 2005).

224 It is possible that multiple IMUs may provide a better estimate of a rider's CoM
225 displacement, however, any additional weight or restriction of movement cause by such a
226 system would likely prevent its widespread use. It is also possible that another location
227 on the body may provide superior tracking of the CoM compared to the lower back. In
228 general, the location of the CoM during non-seated cycling is in front of the pelvis and
229 outside the body (See Figure 1). Thus, an IMU placed on the front of the pelvis would
230 be closer to the CoM, however, this location would still not account for the influence of
231 leg motion on CoM movement. Furthermore, we suspect that this location may be more
232 likely to suffer from greater soft-tissue movement not related to CoM movement (Riddick
233 & Kuo, 2016). A single IMU placed on the lower back may still be suitable for other
234 applications in the analysis of cycling. For example, a single IMU is lightweight and low-
235 cost, meaning it could provide an accessible solution for identifying when a cyclist changes
236 their posture from seated to non-seated. Our data suggest that the precision of the IMU
237 would be adequate for providing within-condition comparisons. Given the accuracy and
238 precision of IMUs for measuring the orientation and displacement of attached objects, a
239 single IMU attached to the frame of a bicycle would be suitable for measuring its angular
240 displacement and velocity.

241 5. Conclusion

242 We found that during non-seated cycling, a single IMU placed on the lower back
243 overestimates a rider's vertical CoM displacement. As the range of CoM displacement
244 increases, the overestimation error increases linearly. We interpret these findings as
245 evidence that a single IMU placed on the lower back is unsuitable for tracking rider CoM
246 displacement during non-seated cycling if the linearly increasing overestimation is not
247 accounted for. Further development of a single-IMU method for measuring a rider's CoM
248 displacement and associated mechanical energy fluctuations during over-ground cycling
249 remains a future goal.

Conflict of interest statement

No conflict of interest to disclose.

Acknowledgements

R.D.W. was supported by an Australian Government Research Training Program
Scholarship.

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Figure 1: A single inertial sensor was used to estimate rider CoM displacement during non-seated cycling. A. An IMU attached to a rigid marker cluster was secured to the skin at the intersection of Tuffier's line (dashed) and lumbar spine midline ($\sim$ L4 spinous process). B. Sagittal plane view of a scaled musculoskeletal model cycling in a non-seated posture. The kinematic estimate of the rider's CoM position is represented by the green sphere.

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Parameter	Definitions and symbols
Local- and global-coordinate system	right-handed cartesian
Global x, y, and z axis definitions	+ x to the right, + y straight up, + z toward viewer
Euler angles	angular position about local x , y , and z axes defined as bank, heading, and attitude, respectively
Symbols	ψ = bank, θ = heading, ϕ = attitude
Angular velocity	about local x , y , and z axes defined as roll (ψ_ϵ), yaw (θ_ϵ), and pitch (ϕ_ϵ), respectively
Euler angle rotation order	Heading, attitude, then bank
Angle units	radians and radians per second

Table 1: Notation, reference coordinates, and definitions used to describe the orientation and motion of the IMU and marker cluster body in this study.

Conditions	RMS error		
	Yaw (θ_ϵ)	Pitch (ϕ_ϵ)	Roll (ψ_ϵ)
10% $P_{max.i}$			
70 rpm	0.16 ± 0.02	0.14 ± 0.02	0.11 ± 0.03
120 rpm	0.22 ± 0.03	0.17 ± 0.03	0.18 ± 0.08
30% $P_{max.i}$			
70 rpm	0.19 ± 0.05	0.15 ± 0.04	0.15 ± 0.08
120 rpm	0.19 ± 0.05	0.18 ± 0.08	0.16 ± 0.02
50% $P_{max.i}$			
70 rpm	0.21 ± 0.02	0.15 ± 0.02	0.12 ± 0.01
120 rpm	0.24 ± 0.06	0.19 ± 0.05	0.17 ± 0.05

Table 2: The sensor performance was excellent across all conditions. Group mean ($\pm$ standard deviation) RMS error in each Euler parameter describing the yaw, pitch, and roll components of the IMU angular velocity ($rad \cdot s^{-1}$) compared to the rigid marker cluster body during non-seated cycling at three different power outputs (10%, 30%, and 50% $P_{max.i}$) and two different cadences (70 rpm and 120 rpm).

		Agreement analysis				Descriptive statistics		
		LloA	UloA	Accuracy (Bias)	Precision (MAD)	Range	Ave. err.	Max. err.
IMU	vs	-8.3	11.6	1.6	3.5	88.3	1.8%	$\pm 3.9\%$
Cluster								
IMU	vs	$0.37x -$	$0.37x -$	$0.37x +$	5.6	—	—	—
Model		13.1	19.0	3.0				

Table 3: There was good agreement between the IMU and cluster, but not between the IMU and model. A fixed limits of agreement analysis was used to compare the range of vertical displacement derived from an IMU to a rigid marker cluster. Positive values indicate an overestimation of the marker cluster’s vertical displacement by the IMU (IMU – Cluster). All measures are in millimetres. A variable limits of agreement analysis was used to compare the range of vertical displacement derived from an IMU to the musculoskeletal model. All measures are in millimetres.