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SIMPLIFIED MARKER SETS FOR THE CALCULATION OF CENTRE OF MASS LOCATION DURING BEND SPRINTING

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Simplified marker sets for the calculation of whole body centre of mass (CoM) location and associated variables (velocity, touchdown distance and turn of CoM) used in the analysis of bend sprinting performance were examined. CoM related variables were compared between a whole-body (13 segment), lower limb and trunk and lower limb model. Both simplified models showed strong agreement with whole-body CoM (Intraclass correlation: 0.873 - 0.998). The lower limb and trunk model (LLT) was the most accurate representation of whole body calculations, with acceptably low differences in all variables examined. Therefore, the LLT model is recommended for future use.

KEYWORDS: velocity, touchdown distance, three-dimensional kinematics, 200m.

INTRODUCTION: Analysis of the position, trajectory and velocity of the centre of mass (CoM) is common within bend sprinting research. In addition, key variables such as touchdown distance and turn of CoM during ground contact require the location of CoM to be known (Churchill, Salo & Trewartha, 2015a; Churchill, Trewartha, Bezodis & Salo, 2015b). Kinematic adaptations that occur during bend sprinting are multi-dimensional (Alt, Heinrich, Funken & Potthast, 2015; Churchill et al., 2015a; Churchill et al., 2015b). Therefore, evaluation of bend sprinting necessitates the use of three-dimensional (3D) data collection methods. With the use of a full body marker set, 3D methods such as optoelectronic motion capture allow calculation of CoM position. Segmental methods and associated inertial parameters (derived from anthropometric studies e.g. Demspter & Gaughran, 1967; de Leva, 1995; Zatsiorksy & Seluyanov, 1983) are used to estimate the whole-body CoM location based on the distribution of mass within the multi-segment model. However, there are difficulties with optoelectronic data collections that have complex lower limb marker sets (for example a multi-segment foot model) with a required minimum of three markers for each segment. The ability to achieve an appropriate frame rate, resolution and capture volume to aid marker detection is both determined and restricted due to the number of cameras and laboratory space available. Additional factors such as the time-consuming application of markers, increased interference with the athlete, thus a potential decrease in ecological validity, make collection of only lower limb data preferable to using a full body marker set. Furthermore, kinematic analysis of the arms is rarely performed due to the implication that the arms do not have a significant impact on sprint performance (Mann & Herman, 1985). Therefore, the ability to calculate the required variables (including CoM location) using a simplified marker set would be desirable. The purpose of this study was to ascertain to what extent the simplification of a kinematic marker set is appropriate for the calculation of CoM during bend sprinting.

METHODS: Following ethical approval from the Sheffield Hallam Local Research Ethics Committee, seven sprinters (five males; mean age 20 ± 1.11 years; body mass 70.37 ± 4.88 kg; stature 1.79 ± 0.48 m and two females; mean age 22 ± 3.50 years; body mass 58.85 ± 1.35 kg; stature 1.66 ± 0.40 cm) volunteered for the study. All athletes had competitive experience of bend sprinting (200 and/or 400 m) with a mean 200 m personal best time of 22.90 ± 0.85 s (range from 22.00 to 24.10 s). Data were collected on a flat standard indoor track surface with a reconstructed bend, replicating lane 1 (radius 36.5 m) of a standard 400 m running track (IAAF, 2008). Participants performed five trials at maximal effort for 60 m. Approximately eight minutes were allowed between trials to allow recovery and avoid the onset of fatigue (Churchill et al., 2015b).

Kinematic data were collected using a 12-camera optoelectronic motion capture system (10 x model Raptor and 2 x model Eagle, Motion Analysis Corporation, Santa Rosa, CA, USA) operating at 240 Hz. A right-handed global coordinate system was defined using a rigid L-frame with four markers of known locations. A three marker wand of 500 mm was used within the calibration volume to scale the individual camera views. The calibration volume (7 m long, 2 m wide and 2 m high) was located tangentially to the apex of the curve to record data at the 40 - 47 m section of the 60 m sprints where athletes are likely to be at maximum speed (Krzysztof & Mero, 2013). A full body marker set (59 markers) was used to model 13 segments (head, upper arms, lower arms, torso, pelvis, thighs, shanks and feet). Retro-reflective spherical markers (12.7 mm) were placed on the following anatomical landmarks; head, acromio-clavicular joint, upper arm, lateral humeral epicondyle, medial humeral epicondyle, forearm, ulnar styloid, radial styloid, C7, T10, jugular notch, xiphoid process, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), greater trochanter, medial femoral epicondyle, lateral femoral epicondyle, thigh, tibia, medial malleolus, lateral malleolus, posterior calcaneus, medial calcaneus, lateral calcaneus, 1st and 5th metatarsal base, 1st, 2nd and 5th metatarsal head and head of the 2nd toe. A static trial was performed before the protocol began to determine neutral position of the joints.

Cortex (version 5.3, Motion Analysis Corporation, Santa Rosa, CA, USA) was used to track and export raw 3D coordinate data. Automatic gap filling was performed using a cubic spline where gaps were <10 frames. Raw marker positions were filtered at 18 Hz using a low-pass Butterworth filter. Segments, local coordinate systems and joint centres were defined and constructed in Visual 3D (version 4, C-Motion, Rockville, MD, USA). Body segment parameters were estimated from de Leva (1996) and adjusted to allow the addition of 0.2kg to each foot which represents the mass of a spiked shoe (Hunter, Marshall & McNair, 2004). Two further, simplified models were defined following methods of Vanrenterghem et al., (2010). Firstly, a CoM representation was calculated with lower limb and trunk segments (LLT) including pelvis, thighs, shanks, feet and thorax (total markers: 42). Following this, the trunk segment was then excluded and CoM calculated based upon lower limbs (LL) segmental position only (total markers: 38). CoM trajectories across two steps (one left, one right) were calculated with each of the three models and the mean calculated across all participants. Take-off and touchdown events were identified using the position of the marker coordinates for the 2nd toe and 5th metatarsal in the static trial as a threshold in accordance with Bezodis, Thomson, Gittoes and Kerwin (2007).

Horizontal velocity for each CoM representation was calculated in Matlab (v2015b, Mathworks, Natick, USA) using a finite difference algorithm (Winter, 2005). Velocity data were analysed for the duration of a left and right step and normalised to 101 data points to represent 100% of the gait cycle. Left and right steps were defined by the foot that initiated the step. Touchdown distance (the horizontal displacement between the CoM and second metatarsophalangeal joint) was calculated using an instantaneous progression vector for the CoM (calculated from the horizontal position of the CoM one frame before the instant of interest to the horizontal position of the CoM one frame after the instant of interest, then divided by its norm to create a unit vector) and a horizontal vector from the CoM to the 2nd metatarsal head of the touchdown limb. A scalar projection of this vector onto the instantaneous progression vector gave touchdown distance (Churchill et al., 2015b). For all trials the middle step was used to calculate Turn of CoM, this resulted in 16 trials for the right step and 19 for the left step. Turn of CoM (the amount the athlete turns 'into' the bend during ground contact, Churchill et al., (2015a) was calculated using the angle between CoM progression vectors during the flight phase before and after the ground contact of interest. Bland-Altman plots with 95% limits of agreement were used to evaluate the agreement between models for all variables. Intraclass correlations (ICC), model (3,k) were used to further assess agreement according to the criteria set by Vincent (1994) where >0.9 indicates excellent agreement.

RESULTS AND DISCUSSION: Mean velocity for all participants was 8.14 ± 0.74 m/s (right step) and 8.12 ± 0.73 m/s (left step). Deviations from whole-body CoM across mean velocity profiles for all participants during the left and right step are shown in Figure 1. Visual inspection of these curves suggests that representation of the CoM with both LLT and LL provide a close (0.00 – 0.10 m/s) approximation when compared to the whole-body CoM trajectory, with LLT providing a more accurate representation throughout the entire step. It should be noted that the deviations are greater at certain points which may impact on the calculation of certain variables, particularly during the flight phase.

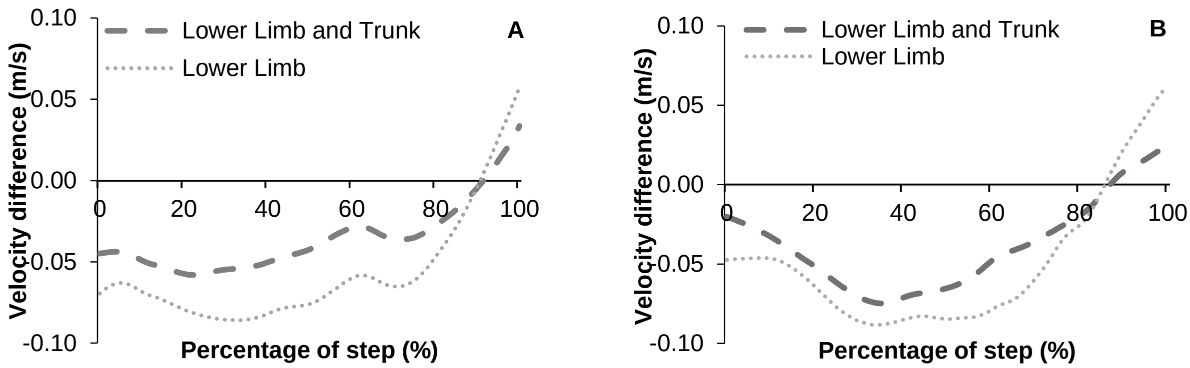

Figure 1: Comparison of the difference between whole-body CoM and CoM representations, zero indicates no deviation from whole-body CoM during mean velocity profiles for (A) the right step and (B) the left step.

Results from Bland-Altman plots and ICC analysis are shown in Table 1. Observations from Bland-Altman plots and the small mean difference found suggest no evidence of systematic bias in the CoM representations with LLT and LL when compared to whole-body CoM. The mean difference for mean step velocity in the LLT condition was 0.0207 ± 0.0643 m for the right step and 0.0037 ± 0.0808 m for the left step. This increased to 0.0524 ± 0.0860 m (right step) and 0.0375 ± 0.0661 m (left step) during the LL condition. The wide (≥ 0.09 m/s) limits of agreement could be related to the high standard deviation shown between trials which is likely a consequence of the differing maximal velocities achieved by participants. Furthermore, the ICC results demonstrate excellent agreement between methods ranging from 0.997-0.998 and providing support for the use of a LLT model to measure mean velocity. Mean difference for touchdown distance in the LLT condition was 0.0040 ± 0.0111 m (right step) and 0.0066 ± 0.0092 m (left step). These values indicate the method provides near equivalent results, supported by narrow limits of agreement and all trials falling within these limits. The mean difference was also low for the LL condition (Table 1, 0.010 ± 0.0151 m (right step) and 0.0010 ± 0.0153 m (left step)). The limits of agreement for the LLT condition are similar to the 1.11 limits of agreement ratio for touchdown distance reported as part of a reliability assessment of sprint variables (Hunter, Marshall & McNair, 2004b). The mean difference was <1° across all conditions for the Turn of CoM. ICC results again demonstrated excellent agreement between whole body and LLT methods (0.995-0.998). Both models showed very good agreement with the whole-body condition with the LLT condition being marginally better.

Table 1: Mean difference, limits of agreement and ICC values for the left and right step during bend sprinting

	Mean step velocity (m/s)		**Touchdown distance (m)**		**Turn of CoM (°)**	
	Right	*Left*	*Right*	*Left*	*Right*	*Left*
Whole body Mean ± standard deviation	**8.14** ± 0.74	**8.12** ± 0.73	**0.28** ± 0.13	**0.43** ± 0.13	**3.59** ± 1.83	**4.60** ± 1.13
LLT Mean ± standard deviation	**8.16** ± 0.75	**8.13** ± 0.74	**0.28** ± 0.13	**0.42** ± 0.13	**3.55** ± 1.95	**4.66** ± 1.16
LL Mean ± standard deviation	**8.18** ± 0.76	**8.14** ± 0.74	**0.28** ± 0.14	**0.43** ± 0.12	**3.13** ± 1.78	**5.47** ± 1.34
Mean difference (Whole body – LLT)	0.0207	0.0037	0.0040	0.0066	0.0381	0.0402
Mean difference (Whole body – LL)	0.0524	0.0375	0.010	0.0010	0.4659	0.8696
Upper Limits of Agreement (Whole body – LLT)	0.14	0.16	0.02	0.03	0.11	0.33
Lower Limits of Agreement (Whole body – LLT)	0.11	0.15	-0.01	-0.02	-0.04	-0.25
Upper Limits of Agreement (Whole body – LL)	0.12	0.09	0.04	0.09	1.95	1.63
Lower Limits of Agreement (Whole body – LL)	0.22	0.16	-0.02	-0.09	-1.02	0.11
ICC (Whole body – LLT)	0.998	0.997	0.998	0.995	0.998	0.995
ICC (Whole body – LL)	0.996	0.998	0.995	0.900	0.942	0.873

These results are in line with the conclusions of Vanrenterghem et al., (2010) who found a better agreement in the LLT model compared to LL in the analysis of lateral cutting movements. This suggests that whilst both CoM representations provide a close approximation of whole-body CoM, LLT is a more accurate solution. Since the use of LLT requires only a small number (four) of additional markers, its use should be favoured over LL. The LLT marker set is comprised of 17 fewer markers than the whole-body marker set and thus its use can decrease the amount of time required in the application of markers and the data cleansing process whilst also increasing the ecological validity of the protocol through reducing the amount of intrusion the athlete experiences.

CONCLUSION: The purpose of this study was to establish whether the CoM and associated variables required for the analysis of performance during bend sprinting could be estimated from a simplified marker set. The LLT model was deemed to provide good agreement with whole-body CoM. It is thought the benefits associated with utilising this simplified model outweigh the small differences reported. Therefore, use of an LLT marker set is suggested as an appropriate alternative to a full body marker set in kinematic studies of bend sprinting.

REFERENCES:

Alt, T., Heinrich, K., Funken, J. & Potthast, W. (2015). Lower extremity kinematics of athletics curve sprinting. *Journal of Sports Sciences, 33:6,* 552-560, DOI 0.1080/02640414.2014.960881

Bezodis, I., Thomson, A., Gittoes, M., & Kerwin, D. (2007). Identification of instants of touchdown and take-off in sprint running using an automatic motion analysis system. *Proceedings of the XXVth Symposium of the International Society of Biomechanics in Sports,* Ouro Preto, Brazil.

Churchill, S., Salo, A. & Trewartha, G. (2015a). The effect of the bend on technique and performance during maximal effort sprinting. *Sports Biomechanics, 14:1,* 106-121, DOI: 10.1080/14763141.2015.1024717

Churchill, S., Trewartha, G., Bezodis, I. & Salo, A. (2015b). Force production during maximal effort bend sprinting: theory vs reality. *Scandinavian Journal of Medicine and Science in Sports,* DOI: 10.1111/sms.12559

de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics, 29*,1223-1230.

Dempster, W. & Gaughran, G. (1967). Properties of body segments based on size and weight. *American Journal of Anatomy, 18(7414)*, 33-54.

Hunter, J. P., Marshall, R. N. & McNair, P. J. (2004). Interaction of step length and step rate during sprint running. *Medicine and Science in Sports and Exercise, 36*, 261-271.

Hunter, J. P., Marshall, R. N. & McNair, P. J. (2004). Reliability of biomechanical variables in sprint running. *Medicine and Science in Sports and Exercise, 36:5,* 850-861.

IAAF (2008). IAAF Track and Field Facilities Manual 2008 Edition - Marking Plan 400m Standard Track. http://www.iaaf.org/about-iaaf/documents/technical#manuals-guidelines Last accessed 6th January 2017

Krzysztof, M., & Mero, A. (2013). A kinematics analysis of three best 100 m performances ever. *Journal of Human Kinetics, 36*(1), 149-160.

Vanrenterghem, J., Gormley, D., Robinson, M. & Lees (2010). Solutions for representing the whole-body centre of mass in side cutting manouevres based on data that is typically available for lower limb kinematics. *Gait & Posture, 31,* 517-521.

Vincent J. Statistics in kinesiology. Champaign (IL): Human Kinetics Books, 1994

Winter, D. (2005) Biomechanics and motor control of human movement, Third Edition, New Jersey: Hoboken.

Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., Whittle, M., D'Lima, DD., Cristofolini, L., Witte, H., Schmid, O, & Stokes, I. (2002). ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion - part I: ankle, hip and spine. *Journal of Biomechanics, 34:4,* 543-548.

Zatsiorsky, V. & Seluyanov, V. (1983). The mass and inertia characteristics of the main segments of the human body. *Biomechanics VIII-B,* Human Kinetics, 1152-9.