The relationship between whole-body external loading and body-worn accelerometry during team sports movements

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Abstract

Purpose: The aim of this study was to investigate the relationship between whole-body accelerations and body-worn accelerometry during team sports movements. ***Methods:*** Twenty male team sport players performed forward running, and anticipated 45° and 90° side-cuts at approach speeds of 2, 3, 4 and 5 $m \cdot s^{-1}$. Whole-body Centre of Mass (CoM) accelerations were determined from ground reaction forces collected from one foot-ground-contact and segmental accelerations were measured from a commercial GPS/accelerometer unit on the upper trunk. Three higher specification accelerometers were also positioned on the GPS unit, the dorsal aspect of the pelvis, and the shaft of the tibia. Associations between mechanical load variables (peak acceleration, loading rate and impulse) calculated from both CoM accelerations and segmental accelerations were explored using regression analysis. In addition one-dimensional Statistical Parametric Mapping (SPM) was used to explore the relationships between peak segmental accelerations and CoM acceleration profiles during the whole foot-ground-contact. ***Results:*** A weak relationship was observed for the investigated mechanical load variables regardless of accelerometer location and task (R^2 values across accelerometer locations and tasks: peak acceleration 0.08-0.55, loading rate 0.27-0.59 and impulse 0.02-0.59). Segmental accelerations generally overestimated whole-body mechanical load. SPM analysis showed that peak segmental accelerations were mostly related to CoM accelerations during the first 40-50% of contact phase. ***Conclusions:*** Whilst body-worn accelerometry correlates to whole-body loading in team sports movements and can reveal useful estimates concerning loading, these correlations are not strong. Body-worn acclerometry should therefore be used with caution to monitor whole-body mechanical loading in the field.

Keywords: CoM acceleration, GPS/accelerometry, training load, mechanical load, peak acceleration

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Introduction:

Team sports players experience high external forces on the body, in particular during the large number of accelerations and decelerations they perform [1]. As a consequence, soft tissues (bones, cartilage, muscles, tendons and ligaments) are put under considerable mechanical load. The accumulation of this mechanical load over time can result in structural adaptations that are beneficial (repair, regeneration, and strengthening of the tissue) and/or detrimental (leading to overuse or acute injury). A subtle balance of mechanical load that depends on the frequency, duration and intensity of the external forces acting on the body is required to have beneficial adaptation yet avoid soft tissue injury [2]. Quantifying the external forces acting on the body during team sport movements in the field could therefore help researchers and practitioners to better monitor and understand the mechanical load experienced by players in training and matches.

Accelerometers embedded in Global positioning Systems (GPS) devices are commonly used in professional team sport to monitor the players' energetic demands, e.g. from the distance players cover and the speed they run at or to estimate the external forces acting on the players' body [3,4]. The GPS/accelerometer devices are worn on the dorsal part of the upper trunk within an elastic vest and allow the registration of acceleration of the (upper) trunk segment. It has previously been demonstrated that the accelerations registered from these GPS embedded accelerometers overestimate the peak external forces acting on the players' body during running and changes in direction [5], or in landing and jumping tasks [6]. However the relationship between trunk acceleration from GPS accelerometers and whole-body mechanical loading during team sports movements it is still largely unexplored.

The estimation of external forces acting on the body from trunk accelerometry is based on Newton's second law of motion ($F_{whole\text{-}body} = m_{whole\text{-}body}\ a_{whole\text{-}body}$) and the assumption that body-worn accelerometers are able to measure whole-body acceleration. However because the GPS accelerometers measures trunk accelerations the external forces measured are actually the external forces acting on the trunk ($F_{trunk} = m_{trunk}\ a_{trunk}$). If however segmental accelerations from the trunk accelerometer are related to the whole-body acceleration it could be feasible to estimate the external forces experienced by players in the field. Whole-body accelerations, biomechanically expressed as Centre of Mass (CoM) accelerations, do however depend on the complex inter-segmental dynamics of the body. Since the position of the CoM relative to individual segments varies depending on the player's movements it remains questionable whether trunk mounted accelerometers and body-worn accelerometry in general are able to measure the multi-segment dynamics during those movements that are typically performed in team sports.

The relationship between segmental acceleration from body-worn accelerometry and CoM accelerations seems to be affected by the location of the accelerometer. Accelerometers located at the hip have for example demonstrated an acceptable association with the external forces acting on the whole body, biomechanically expressed as the ground reaction forces (GRF), during daily life activities [7,8]. In addition accelerometers located at the hip and tibia have shown a strong association with GRF in vertical jumping [9,10]. Furthermore, higher accumulated accelerometer-based loading values have recently been observed from a GPS accelerometer located at the hip compared to the trunk for a 90 minute football simulation [11,12] but it remains uncertain which segmental accelerations would better relate to whole-body mechanical loading during typical team sports movements.

Altogether, the influence of accelerometer location on the relationship between measured accelerations and CoM accelerations during team sports movements such as running and changes in direction is still largely unexplored. The aim of this study was therefore to investigate the association between whole-body mechanical loading and accelerations measured from an accelerometer that is attached to an individual body segment. This was done by investigating whether accelerations from the body-worn accelerometers are related to variables that represent whole-body loading, and whether peak accelerations are related to specific features of the CoM accelerations during the time when the player is in contact with the ground.

Methods:

Twenty recreational male team sports athletes volunteered to participate in this study (age 22 ± 4 years, height 178 ± 8 cm, mass 76 ± 11 kg). No participants had a history of severe lower limb injuries (e.g. ACL injuries or ankle sprains). The study was approved by the Institutional Ethics Committee and written consent was obtained from all participants.

The participants completed four forward running trials (Run), four anticipated 45° (Cut45) and four 90° side cutting trials (Cut90) at approach speeds of 2, 3, 4 and 5 $m \cdot s^{-1}$ (± 5%) in a randomised condition order. Approach speed was measured with photocell timing gates (Brower Timing System, Utah, USA) that were positioned 2 m apart and 2 m from the centre of a force platform. The participants were instructed to hit the force platform with their dominant leg (defined as their preferred kicking leg) during the Run trials and to perform the cutting step with their dominant leg on the force platform. An individual number of practice trials were incorporated in the warm up routine until the participants were familiar with the different tasks and approach speeds (typically around 4 ± 2 practise trials for each conditions).

Segmental acceleration data were collected from four body mounted accelerometers: 1) a trunk mounted tri-axial accelerometer (KXP94, Kionex, Inc., Ithaca, NY, USA) embedded within a commercial GPS device (MinimaxX S4, Catapult Innovations, Scoresby, Australia). This accelerometer had a sampling frequency of 100 Hz and an output range of ± 13 g. The GPS device was positioned on the dorsal part of the upper trunk between the scapulae within a small pocket of a tight fitted elastic vest according to the manufactures recommendations; 2) A tri-axial wireless laboratory accelerometer (518, DTS accelerometer, Noraxon Inc., Scottsdale, USA) with an effective sampling frequency of 1000 Hz, an output range of 24 g, a total weight of 5.7 grams and 19 x 14.2 x 6.3 mm in dimension was tightly fixated to the posterior side of the GPS device using double sided tape. Pilot work showed a difference of approximately 0.34 g in peak acceleration between a laboratory accelerometer fixated to the posterior side of the GPS device compared to the anterior side. The posterior location was therefore used for all measurements; 3) A tri-axial wireless accelerometer (same specifications as accelerometer 2) was located inside the shorts worn by the participants (level with the 5^{th} lumbar vertebra) during the session with double sided tape. An elastic belt was strapped around the participant's waist and accelerometer to minimise the movement of the accelerometer relative to pelvis; 4) A tri-axial wireless accelerometer (same specifications as accelerometer 2) was fixed to a lightweight fibre glass plate shaped to the shaft of the tibia with double sided tape and with elastic velcro straps tightly strapped to the front of the tibia shaft with which the subject performed the pivot/cutting step.

The accelerometers' static validity were tested pre and post every test session by rotating the box through 6 degrees of freedom to detect a ± 1g acceleration due to gravity. The average resultant acceleration were calculated over a 10 second time period for each of the sensing axes and the overall averages were calculated from the average values of the sensing axis. A one sample t-test was used to test if the average resultant acceleration obtained from each accelerometer were significant different ($\alpha \leq 0.01$) from 1g pre or post every test session. Neither of the accelerometers showed a significant difference from 1g pre or post for any of the test sessions.

Ground reaction forces (GRF) were collected from a 0.9 x 0.6 m^2 Kistler force plateform (9287C, Kistler Instruments Ltd., Winterthur, Switzerland) embedded in the floor sampling at 3000 Hz. The GRF data were synchronised with the accelerometer data from the three laboratory tri-axial accelerometers through an analogue board and recorded simultaneously in Qualisys Track Manager (Qualisys AB, Gothenburg, Sweden). The Trunk accelerometer was gently tapped three times before each trial creating three clear spikes in the acceleration traces which were used to synchronise the Catapult acceleration data with the other acceleration data (accuracy of ± 10 ms).

All acceleration and GRF data were exported to Matlab (Version R2014a, The MathWorks, Inc., Natick, MA, USA) where the whole-body CoM acceleration was determined by dividing the GRF data by the participants' body mass and subtracting the gravitational acceleration from the vertical GRF data). The GRF data were filtered with a 6th order lowpass filter with a cut off frequency of 20 Hz, while a similar lowpass filter with a cut off frequency of 60 Hz, 60 Hz and 90 Hz were applied to the Trunk, Pelvis and Tibia acceleration data, respectively. The raw Catapult acceleration data were not filtered, as the accelerometer data from the commercial GPS embedded accelerometers according to the authors' knowledge is left unfiltered when used in the field. Resultant accelerations were calculated from the individual axes for the accelerometry and CoM acceleration data. The foot-ground-contacts on the force platform were determined from the vertical GRF, where touch down and take off events were created when the vertical GRF crossed a 20 N threshold. The following variables were calculated from the accelerometry and CoM acceleration data for each trial: peak resultant acceleration (Peak Acc); the average loading rate (Loading Rate) defined as the average gradient of the resultant acceleration data from touch down to Peak Acc within the first 140 ms of the stance phase; the impulse (Impulse) calculated as the integral of the resultant acceleration over time.

A linear regression analysis was used to explore the within task relationship between Peak Acc, Loading Rate, Impulse of the CoM acceleration and accelerometry from the different accelerometers. In addition a linear multiple regression using the three laboratory accelerometers was used to explore if accelerometry from multiple accelerometers would improve the relationship with the variables obtained from the CoM acceleration. The linear regression analyses were performed using SPSS (Version 22, SPSS Inc., Chicago, IL, USA).

One-dimensional Statistical Parametric Mapping (SPM)[13] was used to explore the within task relationship between Peak Acc from the different accelerometer locations and CoM acceleration across the entire stance phase for the Run, Cut45 and Cut90 tasks respectively. The SPM analysis is an n-dimensional statistical approach of the traditionally 0-dimensional linear regression and one-sample t test approach performed in SPSS [13]. SPM analysis makes it possible to explore the relationship without having to impose the temporal focus bias [14], that may occur in the 0-dimensional linear regression approach described above, because of the between task variation in the GRF pattern. The SPM analysis will reveal the periods of the stance phase where Peak Acc from the individual accelerometers is significantly related to the CoM acceleration.

$$\text{CoM acceleration}(t) = (\beta_1(t) \times \text{Peak Acc from accelerometry}) + \alpha_1(t) + \varepsilon(t)$$

The slopes of the regression line between Peak Acc from the Catapult, Trunk, Pelvis and Tibia accelerometer ($\beta 1$, $\beta 2$, $\beta 3$ and $\beta 4$, respectively) and the CoM acceleration were computed at each time node (t) of the stance phase resulting in beta (β) trajectories (third row Figure 2). These β trajectories were computed for each participant and were subsequently submitted to a population level one-sample t test, yielding statistical curves (SPM{t}) for each of the four accelerometers describing the strength and slope of the relationship between Peak Acc and CoM acceleration (fourth row Figure 2). The significance of each SPM{t} was then determined topologically using random field theory [15], with an alpha level at 0.0125, for each of the three tasks Run, Cut45 and Cut90, respectively.

Results:

The segmental acceleration data overestimated the CoM acceleration (Figure 1) and whole-body mechanical loading variables regardless of task (Table 1). In general, the Catapult and Trunk accelerations were the closest to the CoM acceleration, followed by Pelvis and Tibia accelerations regardless of task and variable of interest. The loading variables increased with an increase in approach speed regardless of task and accelerometer location.

Weak to moderate within task relationships were observed between the segmental acceleration data and CoM acceleration data (Table 2). The Catapult and Trunk accelerometry data most strongly predicted whole-body Peak Acc and Impulse whereas Pelvis and Tibia accelerometry data were the strongest predictor of Loading Rate regardless of task. The addition of multiple accelerometers only showed minor improvements of the relationship with the CoM acceleration loading variables.

The SPM analysis for the Run and Cut45 task generally showed that peak segmental accelerations, regardless of accelerometer location, were significantly positive related to the CoM accelerations during the 10-75% of the stance phase with the strongest relationship from the 10-50% of the stance phase (Figure 2 and 3). While a significantly negative relationship were observed for all accelerometers from the 75-95% of the stance phase between peak segmental acceleration and CoM acceleration for the Run task before take off where the CoM acceleration were low (Figure 2). For the Cut90 task, Peak Acc and CoM acceleration was in general positive significantly related to the CoM acceleration in the initial part of the weight acceptance phase (10-25% stance phase), apart from the peak Tibia acceleration which also demonstrated a positive significant relationship from 70-80% of the stance phase (Figure 4).

Discussion:

The aim of the study was to investigate the association between whole-body mechanical loading and segmental accelerations measured from body-worn accelerometers. The segmental acceleration data consistently overestimated the whole-body mechanical loading variables investigated in this study regardless of task and a weak relationship was observed between segmental acceleration and CoM acceleration. Furthermore this study showed that peak segmental acceleration data is primarily related to whole-body mechanical loading in the 10-50% of foot-ground-contact.

Body-worn accelerometry only measures the acceleration of the segment it is attached to and therefore according to our results it is inadequate to measure the acceleration of the whole body due to the complex multi-segment motion during team sports movements. Furthermore, this linear relationship has previously been questioned, because the relationship between lower limb segmental acceleration and whole-body loading is influenced by the kinematics of the lower limbs at initial foot-ground-contact [16]. The difference between acceleration of individual segments and the acceleration of the whole body can explain the consistent overestimation of peak whole-body loading from body-worn accelerometers observed in this study. These results are in line with the weak relationship previously observed between peak resultant accelerations from a GPS embedded trunk mounted accelerometer and resultant peak GRF during running and change of directions at similar intensities [5].

The peak segmental accelerations measured with the Catapult and Trunk accelerometers were the closest to the peak CoM acceleration. This may be explained by the attenuation of the acceleration signal as it travels up through the body [17]. In addition, the trunk segment represents the largest proportion of the whole-body mass (49.7%) compared to the pelvis (14.2%) and tibia (4.7%) segments [18] which may explain why the segmental acceleration of the trunk best represented the acceleration of the whole body in the current study. The trunk segment's higher mass may also explain why the two trunk mounted accelerometers demonstrated a higher relationship with the impulse of the CoM acceleration as the impulse represent the acceleration measured over time. This indicates that the current practice of positioning GPS-embedded accelerometers on the trunk may be the best location to represent the accumulated whole-body mechanical loading to which team sport players are exposed in the field.

The results from this study showed that tibial segmental accelerations were not a good indicator of whole-body mechanical loading. However tibial segmental accelerations could potentially provide valuable information about the impact forces the lower extremities are exposed to during initial foot-ground-contact. Studies on overuse injuries in running have for instance showed that runners with previous stress fracture history were exposed to high initial peak ground reaction forces and higher loading rate than runners with no previous stress fracture history [19]. The potential of using tibia mounted accelerometer to monitor initial loading rate in team sport is supported by the results of this study as the tibia mounted accelerometer demonstrated a higher relationship with whole-body loading rate than the trunk mounted accelerometer. Consideration should therefore be given to accelerometer location in team sports based on the mechanical variable/s of interest.

The GPS-embedded Catapult accelerometer consistently measured lower accelerations than the Trunk laboratory accelerometer, and the Peak Acc was slightly delayed in the Catapult data (see Figure 1). The difference in sampling frequencies (Catapult: 100 Hz, laboratory accelerometer: 1000 Hz) may explain the systematic difference between the two trunk mounted accelerometers. The commercial GPS embedded accelerometers' ability to measure peak acceleration during high frequency movements has previously been questioned when compared to laboratory accelerometers with a higher sampling frequency [20,21]. Increasing the sampling frequency of the commercial GPS embedded accelerometers may improve their ability to represent the true accelerations experienced in team sports.

The Statistical Parametric Mapping analysis enabled us to investigate the relationship between peak segmental accelerations from body-worn accelerometry and CoM acceleration across the stance phase. This analysis showed that peak segmental accelerations, regardless of accelerometer location, were strongest related to CoM acceleration from the 10-50% of the stance phase. Peak segmental accelerations, which previously have been used to investigate whole-body mechanical loading in daily life activities [7,8] or as in this and previously studies to validate whole-body loading from body-worn accelerometry [5,6], can therefore describe only part of the loading the body's soft tissues is exposed to during foot-ground-contact. Trying to use peak segmental accelerations to understand whole-body mechanical loading during foot-ground-contact in team sport movements could therefore be misleading. Additional information other than peak segmental accelerations is needed to better represent the whole-body mechanical loading across the stance phase in dynamic sports movements.

Our results indicated that the relationship between peak segmental acceleration and whole-body loading is task dependent. The difference observed between the two change-in-direction tasks may be explained by the difference in the segmental and CoM acceleration patterns during the stance phase with a clear initial peak after touch down in the Cut90 task (Figure 4) compared to the later occurrence of peak CoM acceleration in the Cut45 task (Figure 3). Furthermore the CoM accelerations of the Cut45 task indicated that approach speed changed the shape of CoM acceleration pattern while the accelerometer trace remained consistent (Figure 3) and thereby affect the relationship with the peak segmental acceleration.

Limitations within this study include the attachment of the individual accelerometers which may have resulted in errors in the accelerometry signal due to the movement of the accelerometer relative to the segment. The attachment methods and locations were chosen with a combination of ideal an applied approach in mind for potential use in team sports. Fixing the accelerometer directly to the skin may have improved the accuracy of the accelerometer data but this is currently less feasible in an everyday field context. In addition, lower filtering cut-off frequency of the accelerometry data may have improved the relationship with the CoM accelerations, as previously demonstrated for GPS-embedded accelerometers [5,6]. However it was beyond the scope of this study to determine the optimal cut off frequency as this will most likely will be dependent on task and task intensity making it difficult to apply optimal filter settings in the field. Importantly though, improving the relationship with specific cut off frequencies does not change the fundamental issue with the use of body-worn accelerometry to estimate CoM acceleration as it only measures the accelerations of the segment it is attached to and not the accelerations of the whole-body.

The assumption of a simple linear relationship, based on Newton's second law of motion, where segmental accelerations is measured from body-worn accelerometers is not sufficient to determine the linked multi-segment dynamics of the whole body during team sports movements in the field. For instance when this linear assumption is used to investigate the relationship between GPS accelerometry data and risk of soft tissue injuries [22,23]. To better estimate whole-body acceleration, the multibody dynamics of a complex system, such as the human body, must be accounted for. Future studies should not assume that a linear approach is sufficient to estimate the mechanical external force acting on players in the field but investigate the application of multi-segment models for this purpose [24].

Practical Applications

Although a linear relationship exists between body-worn accelerometry (e.g. GPS accelerometers) and whole-body accelerations the assumption of a simple linear relationship, based on Newton's second law of motion, should be used with caution. Practitioners should therefore be careful when attempts are made to monitor, summarise and evaluate the mechanical load the players are exposed to from body-worn accelerometry or associated to soft tissue injury risk. New methods need to be developed to use body-worn accelerometry to more accurately explain whole-body mechanical loading in dynamic team sports.

Conclusion

Whilst a weak to moderate correlation was observed between segmental accelerations from body-worn accelerometry and can reveal useful estimations of whole-body mechanical loading in team sports movements, particularly in the first 10-50% of foot-ground-contact, the linear relationship is weak regardless of accelerometer location and task. Body-worn accelerometry only measures the acceleration of the segment it is attached to and is inadequate to measure the acceleration of the whole body due to the complex multi-segment motion during team sports movements. Practitioners should consider the weak to moderate linear relationship between body-worn accelerometry and whole-body mechanical loading when interpreting the accelerometry data in this context.

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Table 1: Peak Acc, Loading Rate and Impulse for all tasks (Run, Cut45, Cut90) at all approach speeds (2-5 m·s^{-1}) for CoM accelerations and the four different accelerometers mounted on the body. The values presented are means (±SD) and n = 80 trials in total for each task. [a] One of the participants was not able to perform the four Cut90 trials with an approach speed at 5 m·s^{-1} (n = 76 for this task).

	COM M ± SD	Catapult M ± SD	Trunk M ± SD	Pelvis M ± SD	Tibia M ± SD
Peak Acc (g)					
Run 2 m·s^{-1}	**1.32 ± 0.30**	2.82 ± 0.60	3.78 ± 1.13	4.56 ± 1.70	8.02 ± 2.77
Run 3 m·s^{-1}	**1.56 ± 0.33**	3.33 ± 0.69	4.52 ± 1.22	5.38 ± 1.57	10.47 ± 3.65
Run 4 m·s^{-1}	**1.80 ± 0.30**	2.79 ± 0.80	5.09 ± 1.32	6.38 ± 1.72	14.25 ± 3.78
Run 5 m·s^{-1}	**1.85 ± 0.41**	2.82 ± 0.89	5.34 ± 1.75	7.39 ± 2.48	20.36 ± 5.39
Cut45 2 m·s^{-1}	**1.40 ± 0.34**	2.81 ± 0.63	3.73 ± 1.19	4.90 ± 2.11	8.69 ± 3.54
Cut45 3 m·s^{-1}	**1.72 ± 0.38**	3.41 ± 0.79	4.52 ± 1.30	6.06 ± 1.89	11.62 ± 3.86
Cut45 4 m·s^{-1}	**2.04 ± 0.42**	2.92 ± 1.09	5.40 ± 1.56	8.62 ± 3.20	16.83 ± 5.19
Cut45 5 m·s^{-1}	**2.25 ± 0.49**	3.10 ± 0.95	5.78 ± 1.65	11.36 ± 4.89	18.95 ± 5.99
Cut90 2 m·s^{-1}	**1.49 ± 0.37**	3.10 ± 0.82	3.99 ± 1.38	5.52 ± 2.40	9.92 ± 4.15
Cut90 3 m·s^{-1}	**1.90 ± 0.50**	3.89 ± 0.96	5.01 ± 1.49	8.73 ± 4.71	14.37 ± 6.27
Cut90 4 m·s^{-1}	**2.08 ± 0.51**	2.86 ± 1.03	5.08 ± 1.36	10.33 ± 4.28	16.95 ± 6.26
[a] Cut90 5 m·s^{-1}	**2.28 ± 0.51**	3.05 ± 1.04	5.35 ± 1.56	12.53 ± 5.45	19.85 ± 5.72
Loading Rate (g·s^{-1})					
Run 2 m·s^{-1}	**18.6 ± 4.6**	31.7 ± 9.8	56.2 ± 24.2	83.6 ± 38.2	233.1 ± 111.8
Run 3 m·s^{-1}	**22.7 ± 5.5**	38.3 ± 10.9	70.7 ± 27.1	116.9 ± 45.0	318.8 ± 166.8
Run 4 m·s^{-1}	**30.8 ± 11.1**	34.6 ± 16.6	83.4 ± 28.6	146.4 ± 52.1	463.6 ± 176.5
Run 5 m·s^{-1}	**44.8 ± 18.4**	51.9 ± 16.8	93.1 ± 34.2	191.9 ± 73.7	731.4 ± 249.9
Cut45 2 m·s^{-1}	**15.4 ± 3.8**	30.8 ± 10.5	54.9 ± 26.7	87.8 ± 53.3	261.8 ± 141.3
Cut45 3 m·s^{-1}	**19.8 ± 6.5**	38.4 ± 12.1	67.3 ± 28.1	126.2 ± 57.9	355.3 ± 128.3
Cut45 4 m·s^{-1}	**36.9 ± 20.2**	45.8 ± 18.2	86.2 ± 36.4	202.8 ± 96.8	565.7 ± 234.3
Cut45 5 m·s^{-1}	**52.7 ± 26.2**	63.6 ± 19.3	97.1 ± 36.2	266.1 ± 145.7	690.7 ± 315.3
Cut90 2 m·s^{-1}	**18.3 ± 10.7**	33.7 ± 13.1	55.0 ± 28.5	92.4 ± 53.6	301.2 ± 180.1
Cut90 3 m·s^{-1}	**32.8 ± 20.9**	42.8 ± 17.1	69.5 ± 26.6	154.8 ± 88.6	446.1 ± 224.2
Cut90 4 m·s^{-1}	**44.1 ± 23.6**	52.8 ± 13.0	71.9 ± 24.6	199.4 ± 105.6	567.9 ± 268.5
[a] Cut90 5 m·s^{-1}	**56.3 ± 21.4**	65.3 ± 15.8	76.9 ± 28.6	247.2 ± 126.6	701.0 ± 237.9
Impulse (g·s)					
Run 2 m·s^{-1}	**0.25 ± 0.04**	0.42 ± 0.03	0.43 ± 0.04	0.51 ± 0.04	0.75 ± 0.09
Run 3 m·s^{-1}	**0.24 ± 0.04**	0.40 ± 0.03	0.41 ± 0.04	0.50 ± 0.05	0.84 ± 0.09
Run 4 m·s^{-1}	**0.24 ± 0.04**	0.38 ± 0.04	0.39 ± 0.05	0.51 ± 0.05	1.00 ± 0.11
Run 5 m·s^{-1}	**0.21 ± 0.04**	0.33 ± 0.05	0.34 ± 0.05	0.47 ± 0.08	1.14 ± 0.14
Cut45 2 m·s^{-1}	**0.28 ± 0.05**	0.47 ± 0.04	0.47 ± 0.04	0.55 ± 0.05	0.81 ± 0.11
Cut45 3 m·s^{-1}	**0.30 ± 0.04**	0.46 ± 0.05	0.47 ± 0.05	0.57 ± 0.06	0.93 ± 0.12
Cut45 4 m·s^{-1}	**0.31 ± 0.04**	0.46 ± 0.06	0.49 ± 0.07	0.63 ± 0.08	1.12 ± 0.15
Cut45 5 m·s^{-1}	**0.29 ± 0.04**	0.41 ± 0.07	0.46 ± 0.07	0.62 ± 0.13	1.25 ± 0.21
Cut90 2 m·s^{-1}	**0.35 ± 0.06**	0.55 ± 0.08	0.56 ± 0.08	0.64 ± 0.09	0.92 ± 0.14
Cut90 3 m·s^{-1}	**0.38 ± 0.05**	0.58 ± 0.08	0.60 ± 0.09	0.72 ± 0.13	1.09 ± 0.20
Cut90 4 m·s^{-1}	**0.41 ± 0.06**	0.58 ± 0.09	0.68 ± 0.10	0.80 ± 0.13	1.28 ± 0.27
[a] Cut90 5 m·s^{-1}	**0.38 ± 0.06**	0.54 ± 0.09	0.66 ± 0.10	0.81 ± 0.12	1.44 ± 0.25

Table 2: Within task linear regression values (R^2) for Peak Acc, Loading Rate and Impulse between the CoM acceleration and acceleration data from the individual accelerometers and multiple laboratory accelerometers. [a] One of the participants was not able to perform the four Cut90 trials with an approach speed at 5 $m \cdot s^{-1}$.

	N	Catapult	Trunk	Pelvis	Tibia	Trunk & Hip	Trunk & Shank	Trunk, Hip & Shank
Peak Acc (g)								
Run	320	0.26	0.20	0.08	0.26	0.21	0.31	0.31
Cut45	320	0.42	0.32	0.35	0.50	0.42	0.52	0.54
Cut90	316 [a]	0.55	0.46	0.48	0.34	0.60	0.53	0.61
Loading Rate ($g \cdot s^{-1}$)								
Run	320	0.27	0.41	0.29	0.45	0.47	0.56	0.56
Cut45	320	0.38	0.34	0.59	0.45	0.59	0.49	0.62
Cut90	316 [a]	0.36	0.32	0.59	0.43	0.62	0.49	0.64
Impulse ($g \cdot s$)								
Run	320	0.26	0.25	0.13	0.02	0.26	0.26	0.26
Cut45	320	0.26	0.25	0.17	0.10	0.27	0.29	0.29
Cut90	316 [a]	0.59	0.57	0.44	0.27	0.57	0.57	0.57

Figures

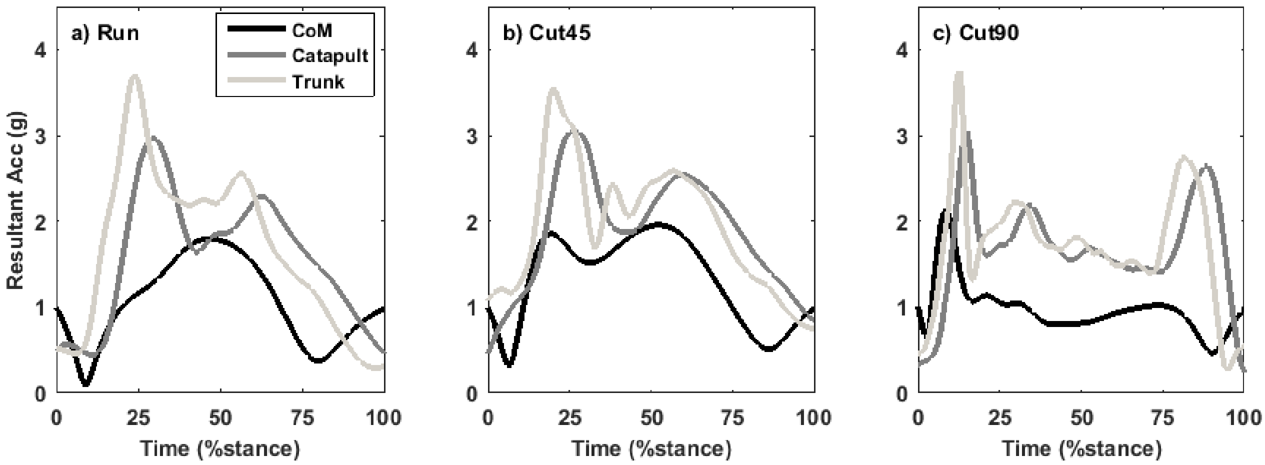

Figure 1: Representative examples of the resultant CoM acceleration and resultant acceleration from the Catapult and Trunk accelerometer for the Run, Cut45 and Cut90 each with an approach speed of 5 $m \cdot s^{-1}$. All curves are normalised over the stance phase (%).

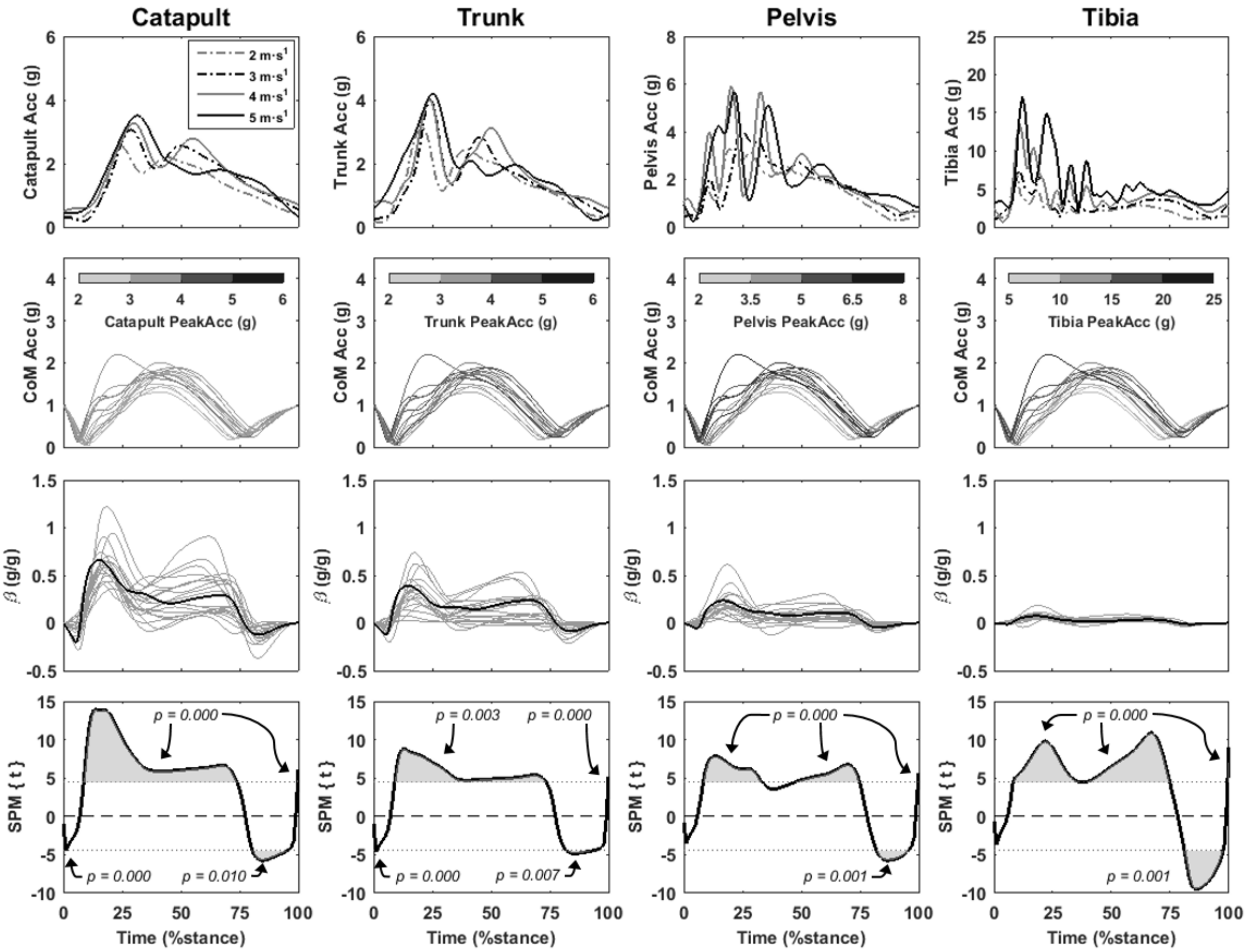

Figure 2: SPM1D regression analysis of the Run task for the four body-worn accelerometers, all curves are normalised over the stance phase (%). The top row shows a representative acceleration from the four approach speeds and accelerometer locations for one subject. The second row shows the CoM acceleration, coloured according to the peak acceleration from the same participant for all trials. The third row shows the β curves from all participants. The specific β curve generated from the data in the second row is shown in black. The bottom row shows the statistical relationship (SPM{t}) between Peak Acc and CoM acceleration across the entire stance phase. Shaded areas indicate a significant relationship (p<0.0125) between Peak Acc from the accelerometer and CoM acceleration.

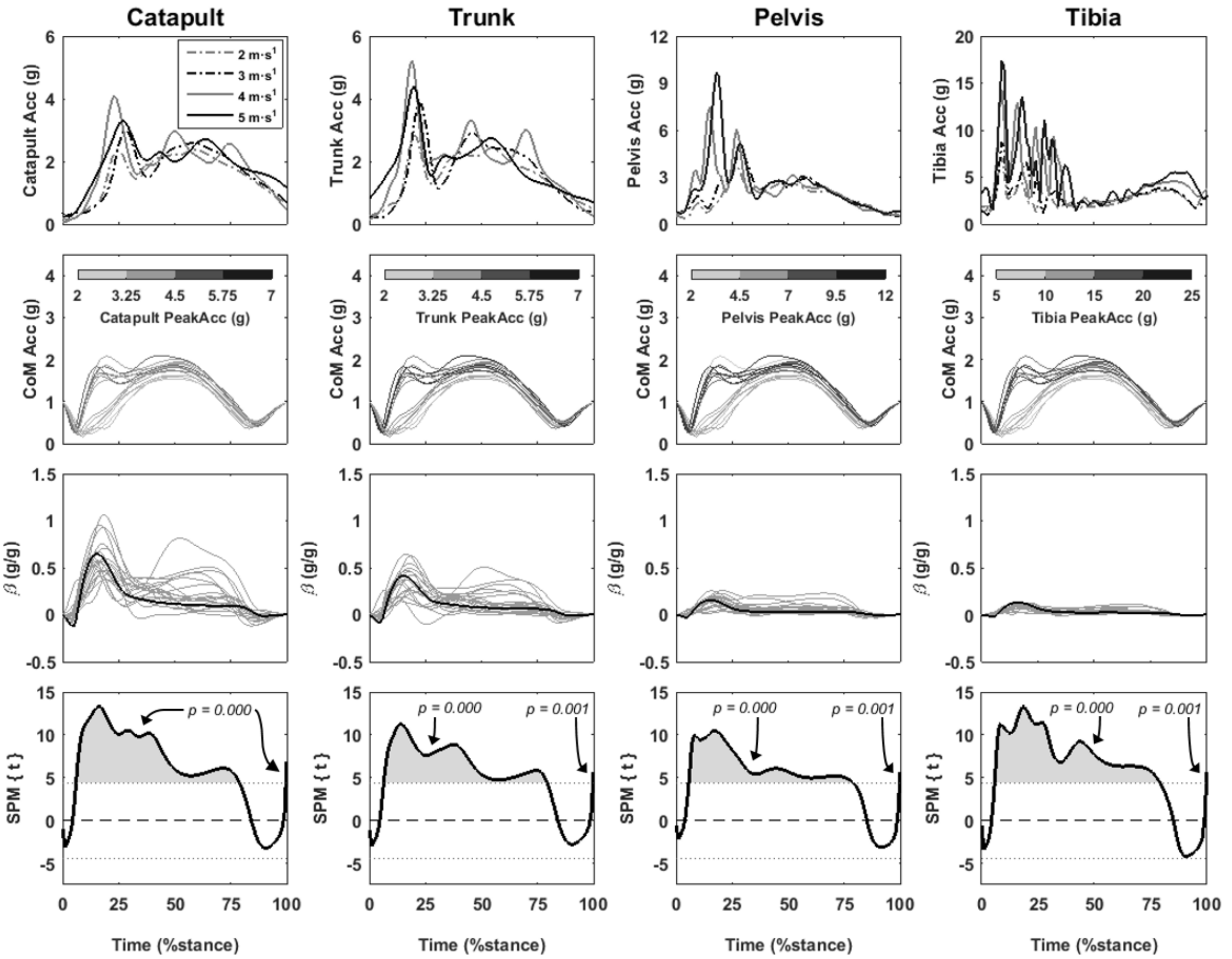

Figure 3: SPM1D regression analysis of the Cut45 task for the four body-worn accelerometers, all curves are normalised over the stance phase (%). See Figure 2 for a detailed explanation of the data displayed in the individual rows.

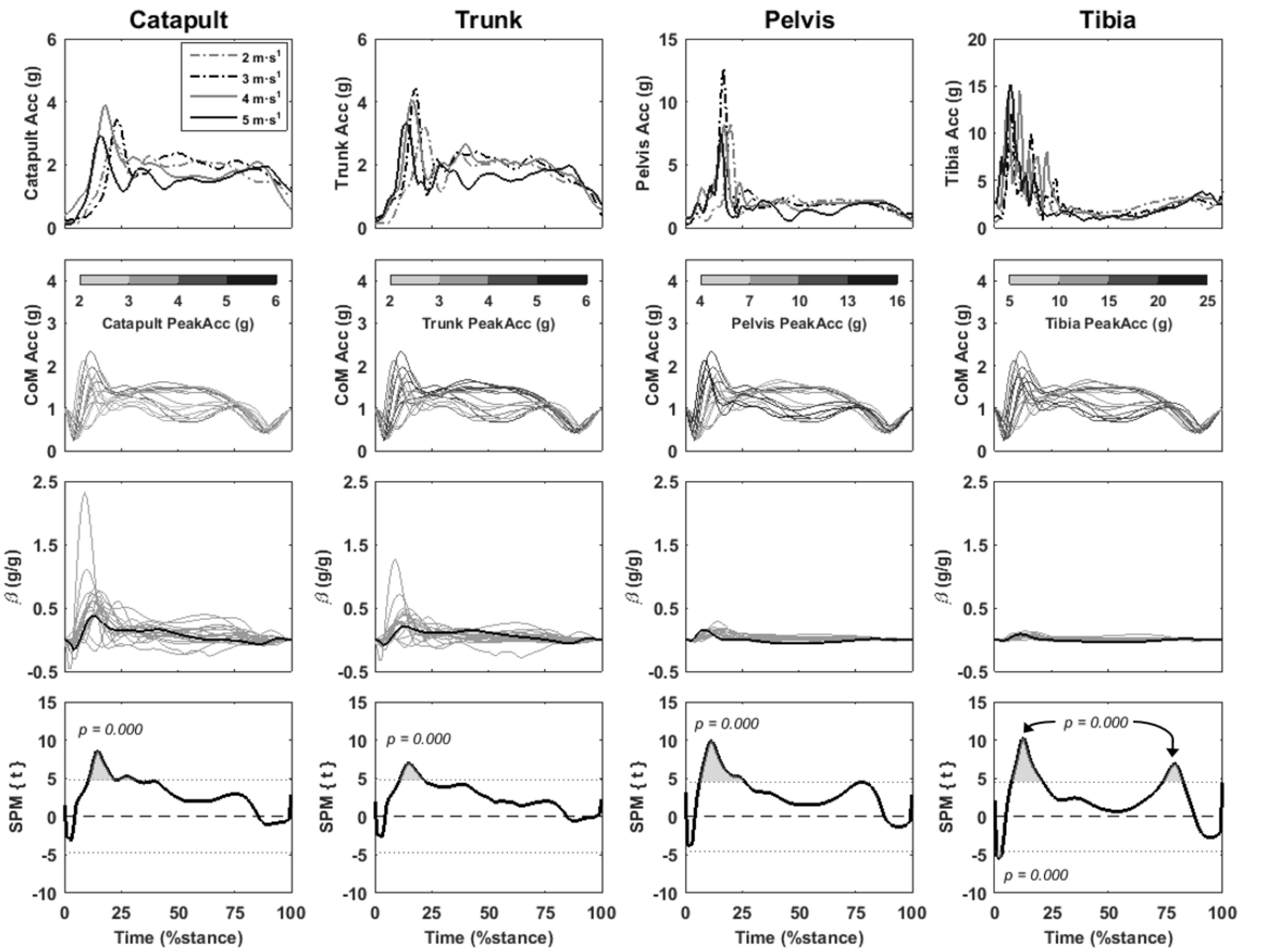

Figure 4: SPM1D regression analysis of the Cut90 task for the four body-worn accelerometers, all curves are normalised over the stance phase (%). See Figure 2 for a detailed explanation of the data displayed in the individual rows.