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Short communication

Performance of a lateral pelvic cluster technical system in evaluating running kinematics

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ABSTRACT

Valid measurement of pelvic and hip angles during posterior load carriage gait task requires placement of pelvic markers which will not be occluded or physically displaced by the load. One solution is the use of pure lateral pelvic clusters to track the pelvis segment. However, the validity of this method has not been compared against pelvic marker systems recommended by the International Society of Biomechanics (ISB) during high impact tasks, such as running. The purpose of this study was to validate the lateral tracking pelvic clusters against the ISB pelvis during running. Six participants performed overground running at a self-selected running speed with shoes. Three dimensional motion capture and synchronised in-ground force plates were used to determine lower limb joint angles and gait events respectively. Two biomechanical models were used to derive pelvic segment and hip joint angles. The ISB pelvis used the anterior and posterior iliac spines as anatomical and tracking markers, whilst the other model used lateral pelvic clusters as tracking markers. The between participant averaged coefficient of multiple correlation suggested good to excellent agreement between the angle waveforms generated from the two marker protocols. In addition, both marker protocols had similar sensitivity in detecting three dimensional pelvic and hip joint angles during the stance phase. This study suggests that in the event posterior load carriage is involved in running gait, pelvic and hip kinematics can be measured by the use of lateral pelvic clusters.

1. Introduction

The International Society of Biomechanics (ISB) recommends modelling and tracking the pelvic segmental coordinate system using the anterior and posterior superior iliac spine markers (ASIS, PSIS) (Wu et al., 2002). In recent years, posterior pelvic clusters have been used to track the pelvis in tasks involving significant hip flexion (Borhani et al., 2013; Vogt et al., 2003) to overcome the ASIS markers being occluded or displaced (Hara et al., 2014). However, posterior pelvic clusters may not be suitable in studies involving posterior load carriage, as these markers can be displaced or occluded during motion capture (Dames and Smith, 2015). One solution is to use lateral pelvic clusters during these motor tasks (Benedetti et al., 1998; Bruno and Barden, 2015; McClelland et al., 2010).

A concern on the use of lateral pelvic clusters is the presence of significant soft tissue artefact (STA), as this region contains greater soft tissue compared to the posterior pelvis (Schwenzer et al., 2010). In contrast, markers positioned on bony prominences are less likely to be affected by STA. STA influence has been shown to be greater during running (Dumas et al., 2014) compared to walking (Peters et al., 2010). However, previous studies have only evaluated the performance of lateral pelvic markers on walking (Benedetti et al., 1998; Bruno and Barden, 2015; McClelland et al., 2010), which may not be translated into running. In addition, previous studies using lateral pelvic markers used a composite lateral and posterior and/or anterior pelvic marker system (e.g. ASIS-iliac crest markers) (Bruno and Barden, 2015; Kisho Fukuchi et al., 2010; McClelland et al., 2010). This composite pelvic tracking system may have disadvantages during load carriage tasks involving significant hip flexion range, where both ASIS and PSIS markers can be occluded. Given that there is an increasing interest in the influence of posterior load carriage in running (Brown et al., 2014; Silder et al., 2015), there is a need to first validate the use of purely lateral pelvic clusters to track the pelvis during body weight running (i.e. running with no external load).

Given the emerging interest in load carriage running biomechanics, there is a need to validate a new pelvic marker tracking protocol, which does not require markers placed on pelvic bony prominences. Hence, the primary aim of this study was to validate

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the use of purely lateral pelvic clusters against the traditional ISB pelvic marker protocol, as a means to track the pelvis during running, when deriving variables of pelvic and hip angles.

2. Methods

2.1. Participants

Six healthy runners [five male, one female; mean (SD) age=25.5 (4.0) years old; height=1.72 (0.14) m; weight=64.9 (12.5) kg; BMI=21.6 (1.1)] participated in this study. Ethics approval was attained from the Curtin University Human Research Committee and written consent was sort from participants prior to study commencement.

2.2. Marker set placement and biomechanical model

Reflective markers were adhered to the pelvic, thigh, leg, and foot segments (Diamond et al., 2014). Two biomechanical models were created, the only difference being the method of tracking the pelvic segment (Supplementary material for details). For the ISB pelvis model, two ASIS and two PSIS markers were used to define the pelvic segment coordinate system (CS), and to track its motion. The origin of the pelvis anatomical coordinate system was defined as the mid-point between the ASIS markers. The (x–y) plane of the segment coordinate system is defined as the plane passing through the right and left ASIS markers, and the mid-point of the right and left PSIS markers. The x-axis was defined from the ORIGIN towards the Right ASIS. The z-axis was orthogonal to the (x–y) plane. The y-axis was the cross product of the x-axis and z-axis. The pelvic CS was defined using Visual 3D (version 5.0, C-Motion Inc., Germantown, USA) default pelvic CS, and does not follow the ISB's pelvic CS (Supplementary material for differences). For the pelvic cluster model, ASIS and PSIS markers were used to only model the pelvis, with all six markers on two lateral clusters (each cluster on each side of the pelvis) acting as tracking markers (Fig. 1). The dimension of each cluster triad was identical for all participants (Supplementary material). All clusters positioned on the pelvis, thigh and shank were attached via double sided tape and rigid sports tape.

2.3. Experimental protocol

Trajectories and ground reaction force were captured using 18 motion capture cameras (Vicon T-series, Oxford Metrics, UK) (250 Hz), and synchronised in-ground force plates (AMTI, Watertown, MA) (1000 Hz), and stored using manufacturer supplied software (Vicon Nexus v2.1.1, Oxford Metrics, UK). All participants performed at least 10 over ground self-paced running trials (10 m run up before, and 10 m run off after the force plate). Initial contact and toe off was determined by a 20 N threshold in ground reaction force. All six participants wore their own running shoes.

2.4. Data processing

Visual 3D was used for post processing. Trajectories were filtered using a zero-lag, fourth order Butterworth (12 Hz) (Sinclair et al., 2014). Trajectories were normalised to 101 data points between initial contact and toe-off. Pelvic segment (relative to "virtual lab" axes) and hip joint angles were quantified using a ZYX and XYZ Cardan rotation sequence (Baker, 2001). A ZYX sequence was used for the pelvis as it produces pelvic rotation angles that more closely relates to clinical understanding of pelvic movement (Baker, 2001).

2.5. Statistical analysis

Overall between-protocol reliability was assessed using the coefficient of multiple correlations (CMC) (Ferrari et al., 2010), performed in Matlab (version 14a, Mathworks Inc., USA). CMC is routinely used to provide a metric that summarises the average waveform similarity between two marker protocols (Ferrari et al., 2010).

A one dimensional SPM paired t-test was performed in Python 2.7 (Canopy 1.5.2, Enthought Inc., Austin, USA) (Pataky et al., 2013). A statistical parametric map (SPM $\{t\}$) was created using the paired difference in mean angle waveforms (between the two protocols) at each normalised time point in the stance phase. Significance level was set at alpha (α) 0.0167, which was α=0.05 corrected for three comparisons per joint. Statistical inference was undertaken using random field theory (Adler and Taylor, 2007). Since biomechanical signals routinely use one dimension (e.g. time varying joint angles), and there is no prior expectation of when in a gait cycle differences in derived angles would occur between marker sets, statistical parametric mapping (SPM) enables a more robust statistical inference testing between the two protocols over an entire gait cycle (Pataky et al., 2013).

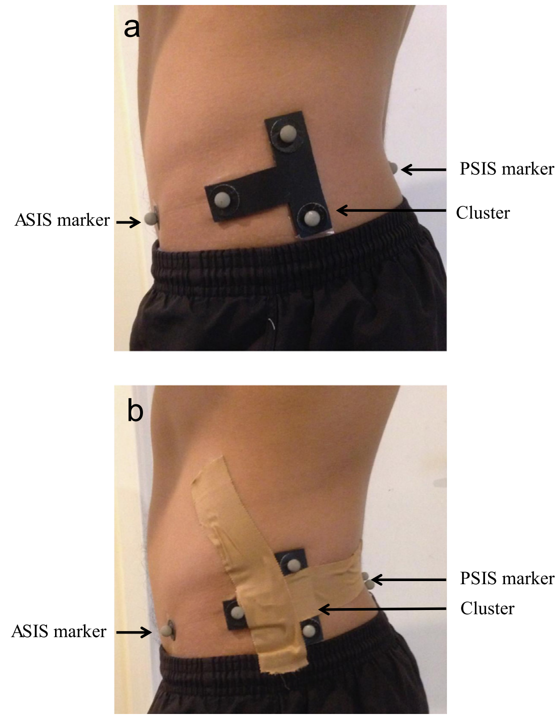

Fig. 1. Position of lateral pelvic marker clusters (lateral view).

Table 1
Between-protocol (ISB vs cluster pelvic protocols) coefficient of multiple correlations.

Participant	Pelvis			Hip		
	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse
1	0.87	0.97	1.00	0.99	0.99	1.00
2	0.98	0.98	0.97	1.00	0.99	0.99
3	0.90	0.95	0.81	0.75	0.99	0.96
4	0.75	0.92	0.87	0.99	0.83	0.96
5	0.73	0.99	0.84	0.92	0.99	0.99
6	1.00	1.00	1.00	0.91	0.99	0.98
Average	0.87	0.97	0.91	0.93	0.96	0.98

Sagittal plane: Hip=flexion/extension, Pelvis=anterior/posterior tilt; Frontal plane: Hip=abduction/adduction, Pelvis=obliquity; Transverse plane: Hip=rotation, Pelvis=rotation.

The "standard error of measurement (SEM)" (Hopkins, 2000) of discrete joint angles between the two protocols (Hopkins, 2000) was calculated at 20% and 80% of the stance phase. A post-hoc decision was made to identify the sensitivity at these two gait phases, as visual inspection suggested that the standard deviation of angle waveforms for each marker protocol were large at these two time points. SEM was calculated by dividing the standard deviation of the angle metric across gait trials by the square root of the number of gait trials (Hopkins, 2000).

3. Results

The mean (SD) running speed was 3.07 (0.19) m s^{-1}. Average CMC values varied from 0.87 for pelvic tilt to 0.98 for hip transverse rotation (Table 1). SPM paired t-test found that for the hip frontal plane, the ISB pelvis resulted in significantly greater hip adduction between 21% and 29% of the right stance phase (P=0.00313) (Fig. 2a). For pelvis frontal plane, the ISB pelvis also resulted in significantly greater pelvic obliquity compared to the cluster protocol between 14% and 26% of the right stance phase (P=0.00073) (Fig. 2b). The sensitivity for hip and pelvic angle

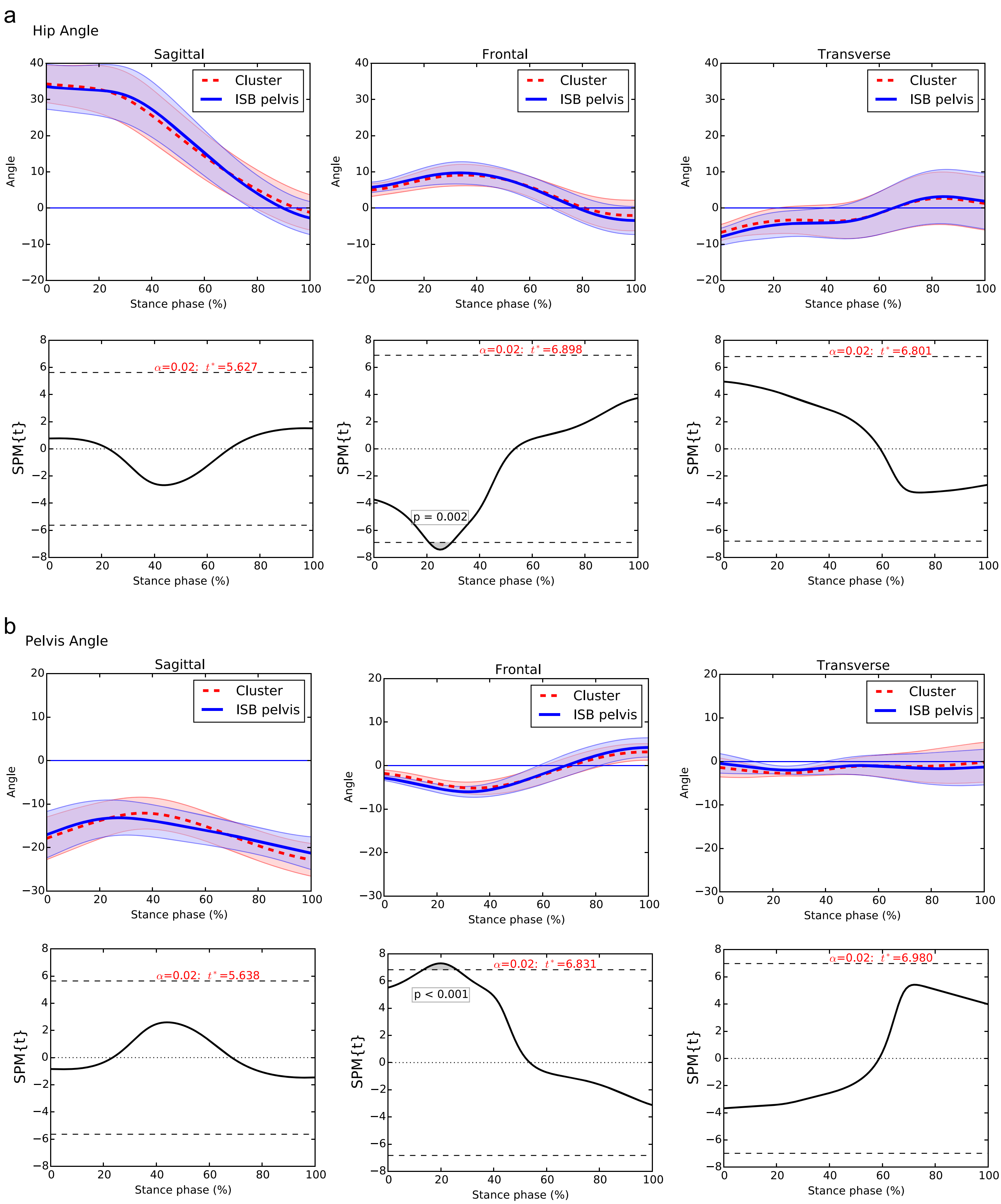

Fig. 2. (a) Hip joint angle group mean (SD) in the top row: joint angles derived from ISB pelvis and pelvic cluster protocol in the sagittal (left), frontal (middle), and transverse (right) plane. SPM paired *t*-test results in the bottom row of angles in the sagittal, frontal, and transverse plane. The horizontal dotted line represents the critical random field threshold of $t=0.0167$. (b) Pelvis segment angle group mean (SD) in the top row: joint angles derived from ISB pelvis and pelvic cluster protocol in the sagittal (left), frontal (middle), and transverse (right) plane. SPM paired *t*-test results in the bottom row of angles in the sagittal, frontal, and transverse plane. The horizontal dotted line represents the critical random field threshold of $t=0.0167$.

Table 2
Within protocol standard error of measurement (deg).

Subject (% gait cycle)	ISB model (deg)						Cluster (deg)					
	Hip			**Pelvis**			**Hip**			**Pelvis**		
	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse
1 (20%)	0.60	0.50	0.56	0.35	0.4	0.99	0.67	0.46	0.52	0.46	0.34	0.97
1 (80%)	0.56	0.21	0.20	0.38	0.33	0.73	0.61	0.18	0.20	0.45	0.30	0.71
2 (20%)	0.28	0.09	0.28	0.18	0.18	0.31	0.28	0.10	0.29	0.19	0.17	0.30
2 (80%)	0.24	0.14	0.24	0.17	0.09	0.17	0.25	0.14	0.23	0.15	0.09	0.16
3 (20%)	0.70	0.84	0.24	0.47	0.63	0.89	0.62	0.84	0.23	0.45	0.62	0.88
3 (80%)	0.66	0.35	0.23	0.30	0.35	0.64	0.68	0.36	0.17	0.42	0.38	0.67
4 (20%)	0.58	0.18	0.28	0.29	0.28	0.33	0.61	0.18	0.27	0.31	0.24	0.34
4 (80%)	0.38	0.14	0.27	0.26	0.12	0.49	0.39	0.14	0.27	0.26	0.13	0.48
5 (20%)	0.56	0.34	0.38	0.46	0.38	0.68	0.55	0.34	0.40	0.46	0.37	0.68
5 (80%)	0.47	0.20	0.29	0.37	0.42	0.48	0.50	0.19	0.29	0.40	0.38	0.51
6 (20%)	0.31	0.35	0.17	0.23	0.22	0.56	0.36	0.30	0.15	0.27	0.18	0.58
6(80%)	0.19	0.27	0.35	0.26	0.30	0.71	0.19	0.28	0.35	0.25	0.28	0.72
Average	0.46	0.30	0.29	0.31	0.31	0.58	0.48	0.29	0.28	0.34	0.29	0.58

Sagittal plane: Hip=flexion/extension, Pelvis=anterior/posterior tilt; Frontal plane: Hip=abduction/adduction, Pelvis=obliquity; Transverse plane: Hip=rotation, Pelvis=rotation.

varied from an average of 0.30° to 0.58° depending on the participant, marker protocol, joint angle and phase of gait (Table 2).

4. Discussion

Relatively few studies involving posterior load carriage in gait have reported pelvic and hip angles (Smith et al., 2006). This may be due to the difficulty in positioning markers on pelvic bony prominences during these tasks. Given the emerging interest in load carriage running research (Brown et al., 2014; Silder et al., 2015), this study demonstrated that pelvic and hip kinematics derived using purely lateral tracking pelvic clusters was comparable with that using the traditional ISB pelvis, in running.

This study reported relatively high between marker protocols CMC values, indicating that the pelvic and hip angles derived using both marker protocols correlated well. Our CMC values were higher than previously reported results, when posterior pelvic clusters were compared against the traditional ISB pelvic model, when considering participants with a BMI of $<24\ \text{kg/m}^2$ (Borhani et al., 2013). One reason could be that bilateral pelvic clusters were used, which increased the inter-marker separation distance in the medio-lateral axis, which minimises propagation of error from markers to the bone position and orientation (POSE) (Cappozzo et al., 1997). In contrast, posterior pelvic marker clusters are usually spaced closely over the sacrum (Borhani et al., 2013), which increases the error propagating to bone POSE estimation (Cappozzo et al., 1997).

CMC values were greater for hip compared to pelvic angles, due to the dependence of CMC values on joint excursion (Ferrari et al., 2010). Participant four and five had relatively lower CMC values for pelvic tilt, and participant three had relatively lower CMC value for hip flexion (Table 1). Visual inspection of these individual's waveform data suggests that there was a small temporal phase lag in angles derived from the two marker protocols, which could contribute to a reduced magnitude of CMC values. A previous investigation also reported slightly greater phase lag in pelvic tilt angles in walking between the ISB pelvis method and a composite pelvis tracking method (ASIS-iliac crest markers) (Bruno and Barden, 2015). Variation between participants could also be due to variations in manual identification of iliac bony prominences, which could differ between participants by as much as 20 mm (della Croce et al., 1999). This could influence derived joint angle magnitudes, which could in turn influence the magnitude of calculated CMC values.

Despite high CMC values, significant differences were detected at approximately loading response of stance. CMC negates within-cycle variability of biomechanical signals and does not provide a measure of statistical inference. The lateral pelvic clusters resulted in smaller hip adduction and pelvic obliquity compared to the ISB pelvis. Angles derived from the lateral pelvic clusters may be more sensitive to soft-tissue artefact from high hip abductor muscle activity, especially during the high impact phase of initial contact to loading response of running (Chumanov et al., 2012). However, no study to our knowledge has quantified the relationship between hip muscle activity and pelvic STA. Despite the significance, these differences were small (hip difference at 25%= 0.73°; pelvic difference at 20%=1.11°) (McGinley et al., 2009), and likely clinically acceptable. The sensitivity of each marker protocol was similar in magnitude, for all cardinal planes and all participants. Although the sensitivity of the lateral pelvic clusters used in this study was greater compared to previous research using a composite lateral pelvic protocol (Bruno and Barden, 2015), the significance of this finding is unclear, given that clinically meaningful changes in kinematics have not been defined.

A limitation of this study was the use of a relatively homogenous dimensioned participant group (mean BMI of 21.6). Although a previous study which used a composite lateral pelvic protocol validated it in participants with a range of BMI (23–43), only walking was investigated. Using lateral tracking pelvic clusters in an overweight population during running requires a specific validation study. In conclusion, purely lateral tracking clusters may be used as an alternate form of tracking the pelvis in load carriage running studies, where posterior pelvic markers may not be feasible.

Conflicts of interest and source of funding

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Appendix A. Supplementary material

Supplementary data associated with this article can be found in the online version at http://dx.doi.org/10.1016/j.jbiomech.2016.05.010.

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