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Are planar simulation models affected by the assumption of coincident joint centres at the hip and shoulder?

[1]P.J. Felton, [1]M.R. Yeadon and [1]M.A. King.

[1]School of Sport, Exercise and Health Sciences, Loughborough University, Leicestershire, LE11 3TU, UK

ABSTRACT

Planar simulation models which assume coincident joint centres at the hip and shoulder are often used to investigate subject-specific maximal performances rather than 3D models due to the viability of determining subject-specific parameters. To investigate the effect of coincident joint centres on model accuracy, three variants of a 16-segment planar subject-specific angle-driven model were evaluated using an elite cricket fast bowling performance: (a) planar representation assuming coincident joint centres; (b) planar representation with non-coincident hip joint centres; (c) planar representation with non-coincident hip and shoulder joint centres. Model (c) with non-coincident hip and shoulder joint centres best matched the recorded performance with better estimates of the ground reaction force (mean RMS differences: (a) 18%; (b) 12%, (c) 11%) and ball release velocity (mean RMS differences: (a) 3.8%, (b) 3.2%, (c) 1.7%) due to a better representation of the mass centre location and link system endpoint velocity. Investigations into the subject-specific performance of maximal effort movements, where non-sagittal plane rotations of the pelvis and torso could affect model accuracy, should consider the use of non-coincident hip and shoulder joint centres within a planar model rather than using a simple planar model or a full 3D model.

Keywords: computer simulation, performance, angle-driven, modelling

INTRODUCTION

In recent years 3D forward dynamics muscle-driven models have increasingly been employed to analyse human movement. However due to the difficulty associated with obtaining accurate subject-specific muscle model parameters, most subject-specific 3D muscle-driven models typically use muscle-model parameters scaled from previous in-vitro research or determined using optimisation procedures (Rajagopal etal., 2016; Shoa et al., 2009). This has thus far limited the application of these models to investigating neuromuscular coordination, estimating internal loading of the musculoskeletal system and evaluating the risk of future injuries (Rajagopal et al., 2016; Shoa et al., 2009; Dao, 2016, Delp et al., 2007; Reinbolt et al., 2011), rather than being able to investigate the optimisation of subject-specific performance of maximal effort activities. With this in mind, forward dynamics torque-driven models are often used to investigate performance of maximal effort activities since subject-specific torque parameters can be obtained in-vivo (Wilson et al., 2006; Allen et al., 2012). These models are often limited to movements where 2D planar representations are appropriate, due to the difficulty associated in obtaining a full set of subject-specific 3D torque parameters (Yeadon and King, 2018).

One assumption that planar models typically adopt to simplify the human link system is to use coincident hip and shoulder joint centres (where the left and right joint centres share a common joint centre in the plane of the model). This assumption reduces the number of segments in the link segment model but fails to incorporate the effect of any non-sagittal plane rotations of the pelvis and torso which cause the projections of the hip and pelvis joint centres to become non-coincident in the sagittal plane. In previous planar models the use of coincident hip and shoulder joint centres has been found to be acceptable for models of take-off where the key outcome parameters are centre of mass velocity and/or angular momentum (Wilson, et al., 2006; Allen et al., 2012; Pandy et al., 1990; Cole et al., 1996; Ackermann and van den Bogert, 2010). In such running and jumping movements the amplitude of non-sagittal plane movements of the hip and shoulder are limited whereas movements such as javelin throwing, cricket bowling and overhead racket shots adopt a “side-on” position where the hip and shoulder projections approach maximum separation. It may be expected that these movements are more affected by the assumption of coincident joint centres.

Modifications have previously been made to planar simulation models to incorporate non-sagittal plane motion of the legs and segment length changes during movements on the gymnastics high bar apparatus. Hiley et al. incorporated the non-sagittal plane motion of straddling the legs prior to high bar release by modifying the inertia parameters of the legs as a function of the thigh abduction angle (Hiley et al., 2007). To represent the effect of scapular rotation Begon et al. expressed the torso length as a function of upper arm elevation angle (Begon et al., 2008).

The aim of this study was to investigate the assumption of using coincident hip and shoulder joint centres on the accuracy of simulations for reproducing the recorded kinematics and kinetics of a movement with non-sagittal plane rotations of the pelvis and torso.

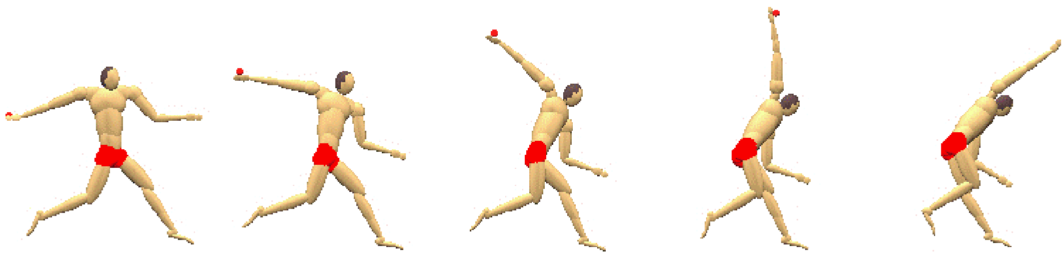

Figure 1. The front foot contact phase of the cricket fast bowling action.

METHODS

A forward dynamics simulation model of the front foot contact phase of the fast bowling action in cricket was chosen since the movement is predominately planar in all aspects other than the non-sagittal plane pelvis and torso rotations (Figure 1). Three variants of the simulation model customised to an elite fast bowler, were developed and were then evaluated by comparison with the bowler’s recorded performance.

Data collection

Kinematic and kinetic data were collected from a member of the England and Wales Cricket Board (ECB) elite fast bowling group (age: 18 years, mass: 85.0 kg, height: 1.94 m) at the National Cricket Performance Centre in accordance with the Loughborough University ethics committee guidelines. Eighteen MX13 Vicon cameras (OMG Plc, Oxford, UK) operating at 300 Hz were used for motion capture within a volume of 7 x 3 x 3 m which spanned the whole bowling action and was centred on a Kistler force platform (Type 9287B, Kistler AG, Switzerland). Fifty 14 mm retro-reflective markers were positioned over bony landmarks, in accordance with the marker set used by Worthington et al., such that joint centres could be calculated (Worthington et al., 2013). An additional 15 x 15 mm reflective patch was placed on the ball to enable ball release velocity and the instant of ball release to be determined.

The participant bowled 12 maximal effort stock deliveries, striking the force plate with his front foot.

Data processing

The four best trials (greatest ball velocity and minimal marker loss) were processed using the Vicon Nexus software. To determine the inputs for the planar simulation model, the projections of the joint centres (toe, MTP, ankle, knee, hip, shoulder, elbow, wrist and hand) onto the sagittal plane were used to determine the trunk orientation angle (angle of the trunk in the global coordinate system) and the joint configuration angles. The distance between the projected hip joint centres, the distance between the projected shoulder centres, and the distance between the mid-point of the projected hip joint centres and the top of the head were calculated. Quintic splines were fitted to the time histories of these variables for input into the simulation model using error estimates calculated as the difference between a data value and the mean of adjacent values (Wood and Jennings 1979; Yeadon and King, 2002). The centre of mass position was calculated using segmental inertia parameters determined via the inertia model of Yeadon using ninety-five anthropometric measurements of the bowler (Yeadon, 1990).

Front foot contact was identified as the first frame in which the vertical ground reaction force exceeded 25 N due to the front foot contacting the force plate. Ball release was determined to have occurred when the distance between the ball marker and the wrist joint centre was greater than the length of the hand (taken from the anthropometric measurements) indicating that the ball was no longer in contact with the hand. The coordinates of the reflective tape on the ball in the sagittal plane were used to calculate the ball release velocity as the average resultant velocity calculated over the first ten frames after release.

Simulation model

A 16-segment planar forward dynamics angle-driven computer simulation model of the front foot contact phase of fast bowling (Figure 2) was constructed using AUTOLEV (Online Dynamics, 1990), (Kane and Levinson, 1985). The 14 rigid segments represented the head plus trunk, two upper arms, two thighs, two shanks, two two-segment feet, forearm plus hand (non-bowling arm), forearm (bowling arm), hand (bowling arm) with wobbling masses (separate rigid segments with mass and inertia) included within the shank, thigh and trunk representations. Two massless segments with variable length and orientation connected the bilateral hip and shoulder joint centres to allow non-coincident hip centres and non-coincident shoulder centres. The front foot had three points of contact with the ground at the heel, ball (metatarsophalangeal joint), and toe. A cricket ball was represented at the end of the bowling arm hand as a point mass.

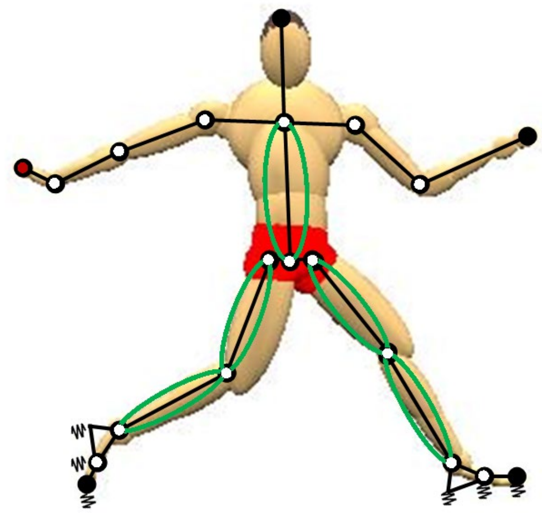

Figure 2 - 16-segment simulation model with wobbling masses within the shank, thigh and trunk segments, angle drivers at all joints (white circles) and spring-dampers at three points on each foot.

Horizontal and vertical linear spring-dampers were used to model the foot-ground interface at the three points of contact with the ground (Equations 1 and 2). The total horizontal and vertical forces (R_x and R_y) acting on the foot were the sums of the forces at these three points (R_{xi} and R_{yi}, i = 1,3), Allen et al., 2010.

$$R_{yi} = -\mathrm{s}_{1i} \cdot y_i - d_{2i} \cdot \dot{y}_i \cdot |y_i| \tag{1}$$

$$R_{xi} = (-\mathrm{s}_{3i} \cdot x_i - \mathrm{d}_{4i} \cdot \dot{x}_i) \cdot R_{yi} \tag{2}$$

$$R_y = \sum_{i=1}^{3} R_{yi} \tag{3}$$

$$R_x = \sum_{i=1}^{3} R_{xi} \tag{4}$$

where x and y are the horizontal and vertical displacements between the point on the foot and its initial point of contact with the ground, $\dot{x}$ and $\dot{y}$ are the time derivatives of x and y, s_1 and s_3 are stiffness coefficients, d_2 and d_4 are damping coefficients, and i represents the point of contact on the foot.

The vertical damping term included multiplication by the vertical displacement which prevented a force occurring due to damping prior to contact or after take-off. The horizontal spring-damper expression included multiplication by the vertical force to ensure that the force was zero prior to contact and after take-off. The horizontal and vertical stiffness and damping coefficients were allowed to vary across the three points of contact.

The wobbling masses of the torso, thigh and shank were connected to the fixed segments using non-linear viscoelastic springs (Equation 5), Pain and Challis, 2001.

$$R_{wi} = (-s_{5\mathrm{i}} \cdot |v_i|^3 - d_{6\mathrm{i}} \cdot |\dot{v}_i|) \cdot \hat{v}_i \tag{5}$$

where R_w is a force vector, v is a vector connecting the points of attachment from the wobbling mass to the fixed segment, $\dot{v}$ is the derivative of v, $\hat{v}$ is a unit vector in the direction of v, $|v|$ is the magnitude of v, s_5 and d_6 are stiffness and damping coefficients, and i represents the segment containing the wobbling mass.

The ball was attached to the distal end of the hand segment using a viscoelastic spring that allowed the ball to release from the hand smoothly (Equation 6). When the ball was in contact with the hand s_7 was set to 1000 Nm^{-1} and d_8 was set to 1000 Nsm^{-1} in order to keep the ball fixed in the hand. At ball release the stiffness and damping parameters instantaneously changed to zero to allow release to occur. Ball release occurred when the time of release in the simulation matched the time of release in the performance (approximately 110 ms).

$$R_B = (-s_7 \cdot |u| - d_8 \cdot |\dot{u}|) \cdot \hat{u} \tag{6}$$

where R_B is a force vector, u is a vector connecting the position of the points of attachment from the ball to the distal end of the hand segment, $\dot{u}$ is the derivative of u, $\hat{u}$ is a unit vector in the direction of u, and s_7 and d_8 are stiffness and damping coefficients respectively.

The simulation model was driven using the recorded kinematics. Input to the simulation model comprised the initial position and velocity of the centre of mass and the initial trunk orientation angle and angular velocity. Segmental inertia parameters derived using Yeadon's inertia model and the coefficients defining the viscoelastic elements were also input (Yeadon, 1990). The output from the simulation model comprised the trunk orientation angle, mass centre position and velocity, ground reaction force time histories, and ball release velocity.

Model variants

Three variants of the simulation model were evaluated to investigate the effect of the assumption of coincident hip and shoulder joint centres on the accuracy of the simulation model to reproduce the recorded kinematics and kinetics when non-sagittal plane pelvis and torso rotations cause the assumption to be violated (Figure 3). In these variants the complexity of the simulation model was increased to incorporate the non-sagittal plane movement of the pelvis and torso. Initially, a simple planar simulation model (simple planar model) was invoked by setting the massless segment lengths to zero and was evaluated to determine the level of agreement with recorded forces and ball release velocity (Figure 3a). In the second variant non-sagittal plane rotations of the pelvis were represented in the planar simulation model (non-coincident hip model) by allowing non-coincident hip joint centres (Figure 3b). This was achieved by driving the length and orientation of the pelvis massless segment using the time histories from the recorded performance data. In the third variant (non-coincident hip and shoulder model) non-sagittal plane rotations of the torso were also included in the planar simulation model (Figure 3c). To incorporate lateral side-flexion the length of the torso plus head segment was driven using the time history from the recorded performance data while adjusting the inertia parameters to take into account the change in length (Hiley et al., 2007). The shoulder joint centres were also allowed to be non-coincident by driving the length and orientation of the shoulder massless segment using the respective time histories from the recorded performance data.

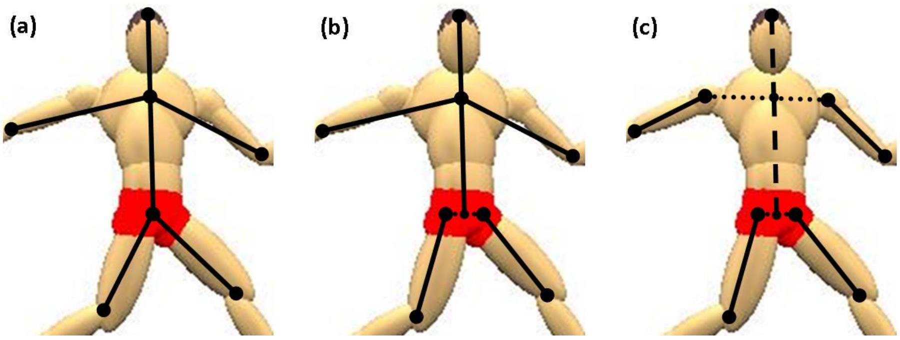

Figure 3 - Three different model variants used to investigate the assumption of coincident hip and shoulder joint centres on model accuracy: a) simple planar model; b) non-coincident hip model; c) non-coincident hip and shoulder model. Solid lines represent rigid segments; dotted lines represent massless segments; dashed line represents rigid segment with variable length.

Parameter Determination

For each model variant, a common set of viscoelastic parameters consisting of the wobbling mass and foot-ground interface coefficients were determined concurrently for three bowling trials (Wilson et al., 2006). A total of 33 parameters were varied via a simulated annealing algorithm in order to minimise an objective function representing the difference between the simulation and the kinematics and the kinetics of the recorded performances (Kirkpatrick et al., 1983). The parameters comprised: 12 stiffness and damping parameters on the front foot, six stiffness and damping coefficients for the wobbling masses, three natural vertical foot spring lengths, and 12 performance specific parameters specifying the initial horizontal and vertical centre of mass velocities, trunk orientation angular velocity and ball release time for the three trials. The performance specific parameters were allowed to vary to compensate for small inaccuracies associated with calculating velocities and the difficulties of determining ball release from performance data (Pain and Challis, 2001; Kirkpatrick et al., 1983).

The objective function value F was taken to be the average of a difference score function applied to each of the three performance trials in order to determine a common set of parameters (Wilson et al., 2006). The score function (Equation 7) was calculated as the overall RMS difference between simulation and performance for four components where 1° was considered to be equivalent to 1% difference in other measures (Yeadon and King, 2002).

$$\mathrm{F} = \frac{1}{3}\sum_{i=1}^{3}\sqrt{\left(\frac{{F_{1\mathrm{i}}}^2 + {F_{2\mathrm{i}}}^2 + {F_{3\mathrm{i}}}^2 + {F_{4\mathrm{i}}}^2}{4}\right)} \qquad (7)$$

where F_1 is the average of the horizontal and vertical force RMS differences expressed as a percentage of the peak vertical force, F_2 is the sum of the mass centre horizontal and vertical velocity absolute differences expressed as a percentage of the resultant mass centre velocity at ball release, F_3 is the trunk orientation angle RMS difference in degrees, F_4 is the sum of the horizontal and vertical ball velocity absolute differences expressed as a percentage of the resultant ball velocity at ball release, and i represents the bowling trial.

Each simulation incurred a penalty which was added to the score function value if the displacement of the front foot exceeded 6 cm vertically or 9 cm horizontally from the initial point of ground contact. These limits were chosen to include the mean vertical compression and horizontal slide of the front foot from the performance trials (3.5 cm vertically and 5 cm horizontally) as well as some extra compliance to account for compression in the system that is not currently accounted for in pin joint models (Allen et al., 2012). Penalties were also employed to ensure that the foot did not 'bounce' in and out of contact with the ground during the simulation period or continuously 'slide' throughout the simulation (Allen et al., 2012). A penalty was also added to limit wobbling mass movement to a maximum of 4.5 cm at the shank, 7 cm at the thigh and 10 cm at the trunk (Lafortune et al., 1992; Minetti and Belli, 1994). No penalties were incurred in the optimised simulations for any of the model variants.

Model Evaluation

The viscoelastic parameters for each simulation model variant were evaluated using a fourth independent bowling trial. For each model variant an optimisation was run using tight bounds on the initial centre of mass velocity, trunk orientation angular velocity and ball release time in order to minimise (via simulated annealing) the same objective function (Equation 7) and obtain the closest overall match to the fourth performance.

RESULTS

The simple planar model was unable to closely match the recorded performances with objective function values of 9.1% and 8.2% between simulation and performance for the parameter determination and model evaluation process respectively (Table 1). Including the non-sagittal plane rotations of the pelvis within the planar simulation model (non-coincident hip model) resulted in more realistic performances with differences of 6.6% and 5.7% for the parameter determination and model evaluation process respectively (Table 1). The recorded performances were best matched in the model variant that included the non-sagittal plane rotations of both the pelvis and torso (non-coincident hip and shoulder model) with differences of 5.8% and 5.3% for the parameter determination and model evaluation process respectively (Table 1).

Since the model evaluation objective function value was similar to the parameter determination objective function value and the score components were comparable, the four trials were averaged to best describe the accuracy of each model (Table 1).

Table 1. The objective function value and score components for parameter determination, model evaluation and the four trials combined for each of the model variants.

score component	parameter determination (mean ± SD of 3 trials)			model evaluation			combined (mean ± SD of 4 trials)		
	PM	HM	TM	PM	HM	TM	PM	HM	TM
F_1 (%)	18 ± 2	12 ± 1	11 ± 1	16	11	11	18 ± 2	12 ± 1	11 ± 1
F_2 (%)	0.2 ± 0.2	0.2 ± 0.1	0.1 ± 0.0	0.0	0.2	0.2	0.2 ± 0.2	0.2 ± 0.1	0.1 ± 0.1
F_3 (°)	0.9 ± 0.3	1.2 ± 0.7	0.9 ± 0.1	1.0	1.2	0.8	0.9 ± 0.3	1.2 ± 0.6	0.9 ± 0.1
F_4 (%)	4.1 ± 1.3	3.6 ± 3.1	1.9 ± 0.5	2.9	2.1	0.2	3.8 ± 1.2	3.2 ± 2.6	1.5 ± 1.0
F (%)	9.1 ± 0.8	6.6 ± 0.7	5.8 ± 0.3	8.2	5.7	5.3	8.9 ± 0.8	6.4 ± 0.7	5.7 ± 0.3

Note:
PM – simple planar model
HM – non-coincident hip model
TM – non-coincidental hip and shoulder model
F_1 - average of the horizontal and vertical force RMS differences expressed as a percentage of the peak vertical force
F_2 - Centre of mass velocity vector difference at ball release as a percentage of the resultant centre of mass velocity at ball release
F_3 – trunk orientation angle RMS difference in degrees
F_4 - ball velocity vector difference at ball release expressed as a percentage of the resultant ball velocity at release

The difference in ball release velocity (F_4) decreased as more complexity was included to model the non-sagittal plane rotations of the pelvis and torso with mean differences between simulation and performance of 3.8%, 3.2% and 1.7% for the simple planar model, the non-coincident hip model, and the non-coincident hip and shoulder model respectively (Table 1).

The model variants which employed non-coincident hip joint centres were better able to more closely match ground reaction forces with mean differences (F_1) of 12% and 11% in the non-coincident hip model and the non-coincident hip and shoulder model compared to 18% in the simple planar model (Table 1). Whilst the increase in complexity from the second to the third model variant resulted in the model more accurately reproducing the horizontal ground reaction force (non-coincident hip model: 10.5% vs non-coincident hip and shoulder model: 8.6%; Figure 4), the difference in the vertical ground reaction force showed was marginal (non-coincident hip model: 13.7% vs non-coincident hip and shoulder model: 13.6%).

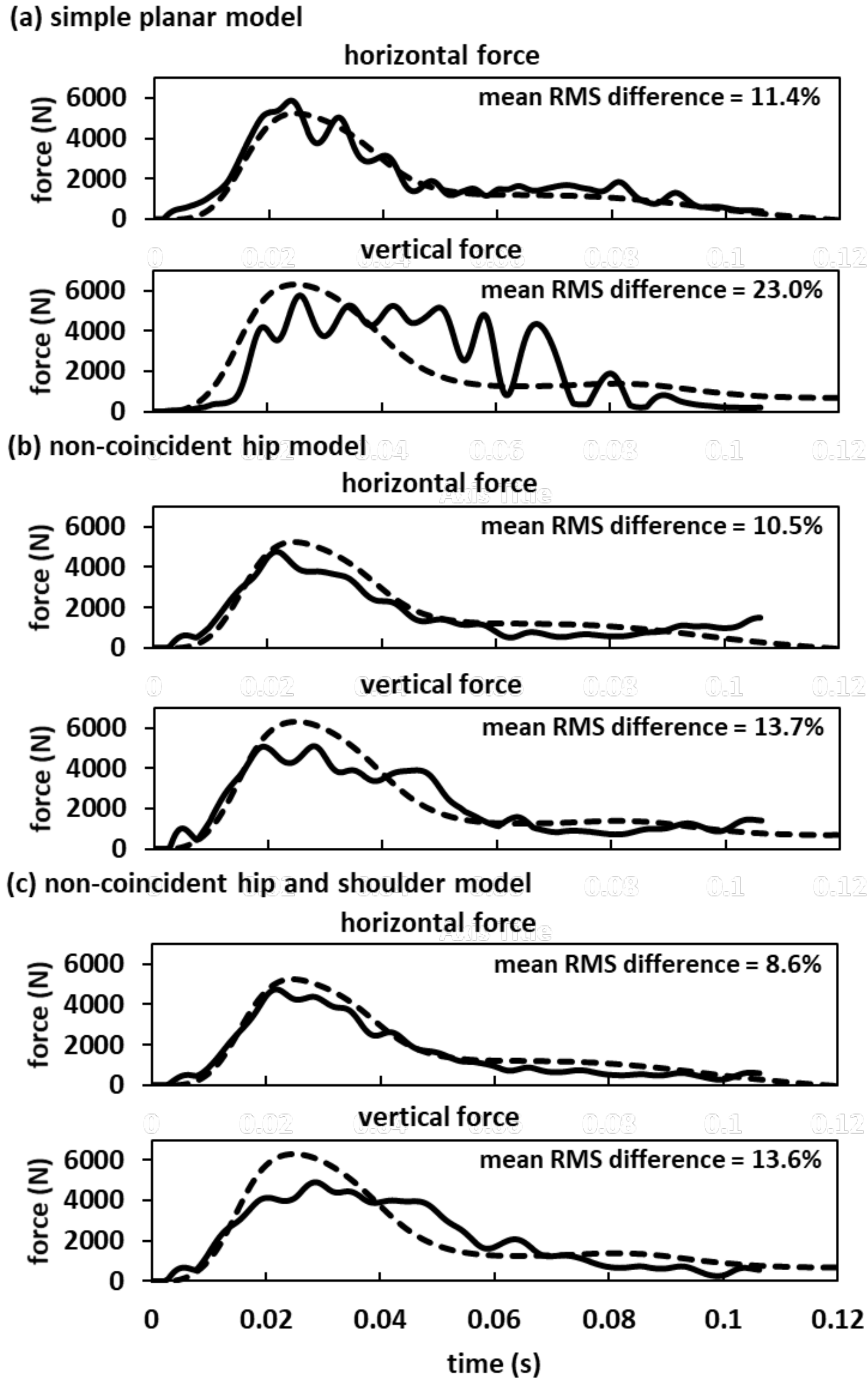

Figure 4 - Comparison of simulation (solid line) and recorded (dashed line) ground reaction forces (simulation representative of the mean RMS difference for each model variant): (a) simple planar model; (b) non-coincident hip model; (c) non-coincident hip and shoulder model.

Coefficients of the horizontal and vertical springs governing the foot-ground interface became more uniform across the three points of contact as the complexity of the simulation model increased (Table 2). No pattern in the stiffness or damping coefficients of the shank, thigh and trunk wobbling masses or the natural spring lengths was evident as the model complexity increased to include non-sagittal plane rotations of the pelvis and torso (Table 2).

Table 2. Viscoelastic parameters for the three model variants

parameter	model variant PM	HM	TM
Vertical toe stiffness, s_{11} (Nm^{-1})	94589	38548	33566
Vertical toe damping, d_{21} (Nsm^{-1})	57098	48600	10834
Vertical MTP stiffness, s_{12} (Nm^{-1})	11798	33140	33209
Vertical MTP damping, d_{22} (Nsm^{-1})	4725	99	6077
Vertical heel stiffness, s_{13} (Nm^{-1})	11778	56639	47265
Vertical heel damping, d_{23} (Nsm^{-1})	3162	5175	6286
Horizontal toe stiffness, s_{31} (m^{-1})	0.10	0.11	16.0
Horizontal toe damping, d_{41} (sm^{-1})	0.00	0.13	0.25
Horizontal MTP stiffness, s_{32} (m^{-1})	0.10	20.0	14.6
Horizontal MTP damping, d_{42} (sm^{-1})	1.57	0.24	0.11
Horizontal heel stiffness, s_{33} (m^{-1})	189	15.9	16.0
Horizontal heel damping, d_{43} (sm^{-1})	0.00	0.09	0.1
Vertical toe spring natural length, (m)	0.026	0.025	0.027
Vertical MTP spring natural length, (m)	0.060	0.060	0.058
Vertical heel spring natural length, (m)	0.051	0.061	0.059
Wobbling mass shank stiffness, s_{51} (Nm^{-3})	2348	1625	2190
Wobbling mass shank damping, d_{61} (Nsm^{-1})	1500	1045	1317
Wobbling mass thigh stiffness, s_{52} (Nm^{-3})	2207	589	2331
Wobbling mass thigh damping, d_{62} (Nsm^{-1})	549	743	744
Wobbling mass trunk stiffness, s_{53} (Nm^{-3})	5608	5485	3297
Wobbling mass trunk damping, d_{63} (Nsm^{-1})	1000	1003	1015

Note:
PM – simple planar model
HM – non-coincident hip model
TM – non-coincidental hip and shoulder model

The trunk orientation angle was closely matched with mean differences (F_3) of 0.9°, 1.2°, and 0.9° for the simple planar model, the non-coincident hip model, and the non-coincident hip and shoulder model respectively (Table 1). The centre of mass velocity at ball release was also closely matched with mean differences (F_2) between simulation and recorded performances of 0.2%, 0.1%, and 0.1% for the simple planar model, the non-coincident hip model, and the non-coincident hip and shoulder model (Table 1).

DISCUSSION

This paper has investigated the effects of the assumption of coincident hip and shoulder joint centres on the accuracy of planar simulation models reproducing the recorded kinematics and kinetics of a movement with non-sagittal plane rotations of the pelvis and torso. A simple planar representation which assumed the hip and shoulder joint centres were coincident was unable to match closely the recorded performances with substantial differences in the ground reaction force and ball release velocity (Table 1; Figure 4). Allowing the hip joint centres to be non-coincident to incorporate the non-sagittal plane rotations of the pelvis within the model improved the difference in the ground reaction force substantially and the difference in ball release velocity slightly (Table 1). When both the hip and shoulder joint centres were allowed to be non-coincident and the trunk plus head segment length was allowed to vary, the model closely matched the recorded performances with the differences in the ground reaction force and ball release velocity decreased substantially compared to the simple planar model.

The models with non-coincident joint centres lead to improved representations of the human link system compared to the simple planar model. This led to a better estimation of the centre of mass position and a closer match to the ground reaction forces (Figure 4). In the simple planar model the origins of the upper and lower extremities are attached to average positions on the torso due to the assumption that the hip and shoulder joint centres are coincident. The disadvantage of this is that the estimation of the centre of mass location relative to the front foot contact is affected since the lower extremities are closer to the torso. This shortens the moment arm between the centre of mass and centre of pressure of the ground reaction force. In order for the model to produce the same impulse and match the centre of mass velocity and trunk orientation at ball release, the ground reaction force is different and thus cannot match closely the recorded ground reaction force. The large vertical toe stiffness (Table 2) in the simple planar model reflects an attempt to move the centre of pressure towards the toe to maximise the moment arm and match the ground reaction force as best as possible.

Comparing the ground reaction force between the model with only non-coincident hip joint centres and the model with non-coincident hip and shoulder joint centres shows that increasing the complexity further improved the accuracy of the horizontal ground reaction force although the difference in the vertical ground reaction force showed no consistent changes (Figure 4). The improvement in the horizontal ground reaction force is most likely due to the origin of the upper extremities being more accurate and providing a better representation of the human link system horizontally. The fact that the vertical ground reaction force difference did not improve suggests that the difference is not caused by the assumption of coincidental shoulder joint centres. The average difference of 13.6% (Figure 4) with a vertical depression at the foot-ground interface of 3.5 cm is similar to that found by Allen et al., 2012 who concluded that pin jointed simulation models are suitable for simulating performance but may require additional complexity to incorporate compliance throughout the link system (e.g. spring-damper elements between segments) to reproduce internal and ground reaction forces accurately.

The difference in ball release velocity between simulation and recorded performance improved with the inclusion of the non-sagittal plane pelvis and torso rotations within both the simulation models when non-coincident joint centres were used compared to the simple planar model. Previous research has found ball release velocity to be linked with both bowling shoulder kinematics (Worthington et al., 2013; David and Blanksby, 1976; Elliott et al., 1986; Foster et al., 1989) and trunk kinematics (Worthington et al., 2013; Elliott et al., 1986; Davis and Blanksby, 1976). Using massless segments to represent the non-sagittal plane rotations of the pelvis and torso introduces extra degrees of freedom which allow the simulation model to better reproduce the human link system during the fast bowling action and more accurately model ball release velocity.

The centre of mass velocity was closely matched for all three model variants (Table 1). In order to match the centre of mass velocity the simple planar model was unable to accurately reproduce the ground reaction force due to the difference in the moment arm between the centre of mass and centre of pressure. This suggests that previous models which have used coincident joint centres may have also been unable to accurately match the ground reaction forces. This may result in a model producing peak forces which the human body would be unable to dissipate safely. The use of coincident joint centres in future planar simulation models should consider the effect it has on the ground reaction forces.

This study has indicated that the use of coincident hip and shoulder joint centres to represent a movement with non-sagittal plane rotations of the pelvis and torso resulted in an over-simplified representation of the human link system. To investigate subject-specific cause and effect relationships for movements with non-sagittal plane rotations of the pelvis and torso a model with increased complexity is required. Although a full 3D model might seem to offer a solution, the need for a full set of accurate subject-specific strength parameters renders such an approach non-viable for investigating maximal effort activities (Wilson et al., 2006) The results of this investigation indicate that an alternative approach using a planar simulation model, where non-sagittal plane rotations of the pelvis and torso are incorporated by allowing hip and shoulder joint centres to be non-coincident, can improve the accuracy of the human link system resulting in closer matched ground reaction forces and link system endpoint velocity. This provides a method for which non-sagittal plane rotations of the pelvis and torso can be represented in a planar simulation and provides a viable solution to enable individual optimal performance and cause and effect relationships to be investigated. The advantages of this proposed approach compared to a 3D model include a simplified optimisation problem and reduced kinematic and kinetic process being required. In the future, this method could be used within forward dynamics planar simulation models to investigate end point velocity in movements with non-sagittal plane rotations of the pelvis and torso such as in throwing, cricket bowling or overhead racket shots.

REFERENCES

Allen, S.J., King, M.A., Yeadon, M.R. (2010). Is a single or double arm technique more advantageous in triple jumping? *Journal of Biomechanics*, 43, 3156-3161.

Allen, S.J., King, M.A., Yeadon, M.R. (2012). Models incorporating pin joints are suitable for simulating performance but unsuitable for simulating internal loading. *Journal of Biomechanics,* 45,1430-1436.

Ackermann, M., van den Bogert, A.J. (2010). Optimality principles for model-based prediction of human gait. *Journal of Biomechanics*, 43, 1055-1060.

Begon, M., Wieber, P-B., Yeadon, M.R. (2008). Kinematics estimation of straddled movements on high bar from a limited number of skin markers using a chain model. *Journal of Biomechanics*, 41, 581-586.

Cole, G.K., Nigg, B.M., van den Bogert, A.J., Gerritsen, K.G.M. (1996). Lower extremity joint loading during impact in running. *Clinical Biomechanics*, 11, 181-193.

Dao, T.T. (2016). Rigid musculoskeletal models of the human body systems: a review. *Journal of Musculoskeletal Research,* 19.

Davis, K., Blanksby, B. (1976). The segmental components of fast bowling in cricket. *Australian Journal for Health, Physical Education and Recreation*, 71(suppl.), 6-8.

Davis, K., Blanksby, B. (1976). A cinematographical analysis of fast bowling in cricket. *Australian Journal for Health, Physical Education and Recreation*, 71(suppl.), 9-15.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Thelen, D. (2007). OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE transactions on biomedical engineering,* 54,1940-1950.

Elliott, B.C., Foster, D.H., Gray, S. (1986). Biomechanical and physical factors influencing fast bowling. *Australian Journal of Science and Medicine in Sport*, 18, 16-21.

Foster, D., John, D., Elliott, B., Ackland, T., Fitch, K. (1989). Back injuries to fast bowlers in cricket: A prospective study. *British Journal of Sports Medicine*, 23, 150-154.

Hiley, M.J., Yeadon, M.R., Buxton, E. (2007). Consistency of performances in the Tkatchev release and re-grasp on high bar. *Sports Biomechanics,* 6, 121-130.

Hubbard, M., Always, L.W. (1989). Rapid and accurate estimation of release conditions in the javelin throw. *Journal of Biomechanics*, 22, 583-595.

Kane, T.R, Levinson, D.A. (1985). Dynamics, theory and applications. New York, NY: McGraw Hill.

Kirkpatrick, S., Gelatt, C.D., Vecchi, M.P. (1983). Optimization by simulated annealing. *Science*, 220, 671-680.

Lafortune, M.A., Cavanagh, P.R., Sommer Iii, H.J., Kalenak, A. (1992). Three-dimensional kinematics of the human knee during walking. *Journal of Biomechanics*, 25, 347-357.

Minetti, A.E., Belli, G. (1994). A model for the estimation of visceral mass displacement in periodic movements. *Journal of Biomechanics*, 27, 97-101.

Pain, M.T.G., Challis, J.H. (2001). The role of the heel pad and shank soft tissue during impacts: a further resolution of a paradox. *Journal of Biomechanics*, 34, 327-333.

Pandy, M.G., Zajac, F.E., Sim, E., Levine, W.S. (1990). An optimal control model for maximum-height human jumping. *Journal of Biomechanics*, 23, 1185-1198.

Rajagopal, A., Dembia, C.L., DeMers, M.S., Delp, D.D., Hicks, J.L., Delp, S.L. (2016). Full-body musculoskeletal model for muscle-driven simulation of human gait. *IEEE transactions on biomedical engineering,* 63, 2068-2079.

Reinbolt, J.A., Seth, A., Delp, S.L. (2011). Simulation of human movement: applications using OpenSim. *Procedia IUTAM,* 2,186-198.

Shoa, Q., Bassett, D.N., Manal, K., Buchanan, T.S. (2009). An EMG-driven model to estimate muscle forces and joint moments in stroke patients. *Computers in Biology and Medicine,* 39, 1083-1088.

Wells, D., Dols, A., Donnelly, C.J., Alderson, J. (2015). An examination of ball release determination methods in cricket bowling and its influence on bowling legality. In proceedings of the 5th World Congress of Science & Medicine in Cricket. Sydney, Australia.

Wilson, C., King, M.A., Yeadon, M.R. (2006). Determination of subject-specific model parameters for visco-elastic elements. *Journal of Biomechanics*, 39,1883-1890.

Wood, G.A., Jennings, L.S. (1979). On the use of spline functions for data smoothing. *Journal of Biomechanics*, 12, 477-479.

Worthington, P.J., King, M.A., Ranson, C.A. (2013). Relationships between fast bowling technique and ball release speed in cricket. *Journal of Applied Biomechanics*, 29, 78-84.

Yeadon, M.R. (1990). The simulation of aerial movement—II. A mathematical inertia model of the human body. *Journal of Biomechanics*, 23, 67-74.

Yeadon, M.R., King, M.A. (2002). Evaluation of a torque driven simulation model of tumbling. *Journal of Applied Biomechanics,* 18, 195-206.

Yeadon, M.R., King, M.A. (2018). Computer simulation modelling in sport. In: Payton, C.J., Bartlett, R.M. (eds.) Biomechanical Evaluation of Movement in Sport and Exercise: BASES Guidelines. London, UK: Routledge, 176-205.