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Optimising the front foot contact phase of the cricket fast bowling action

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ABSTRACT

Cricket fast bowling is a dynamic activity in which a bowler runs up and repeatedly delivers the ball at high speeds. Experimental studies have previously linked ball release speed and several technique parameters with conflicting results. As a result, computer simulation models are increasingly being used to understand the effects of technique on performance. This study evaluates a planar 16-segment whole-body torque-driven simulation model of the front foot contact phase of fast bowling by comparing simulation output with the actual performance of an elite fast bowler. The model was customised to the bowler by determining subject-specific inertia and torque parameters. Good agreement was found between actual and simulated performances with a 4.0% RMS difference. Varying the activation timings of the torque generators resulted in an optimised simulation with a ball release speed 3.5 m/s faster than the evaluation simulation. The optimised technique used more extended front ankle and knee kinematics, increased trunk flexion and a longer delay in the onset of arm circumduction. These simulations suggest the model provides a realistic representation of the front foot contact phase of fast bowling and is suitable to investigate the limitations of other kinematic or kinetic variables on fast bowling performance.

Keywords: computer simulation; performance; ball release speed; modelling; biomechanics

INTRODUCTION

Cricket fast bowling is a dynamic activity in which a bowler runs up and repeatedly delivers the ball at high speeds. Ball release speed is considered a major contributor to fast bowling success as it reduces the time the batter has to interpret the path of the ball and make decisions over which shot to play. The fast bowling action has previously been analysed using experimental methods to explore the relationships between ball release speed and the run-up (Davis & Blanksby, 1976a; Elliot et al., 1986; Worthington et al. 2013a), the front leg kinematics (Elliot et al., 1986; Burden & Bartlett, 1990a; Portus et al., 2004; Worthington et al. 2013a), the motion of the thorax (Davis & Blanksby, 1976a; Elliot et al., 1986; ; Burden & Bartlett 1990b; Portus et al., 2004; Worthington et al. 2013a), the position of the bowling arm (Davis & Blanksby, 1976a; Elliot et al., 1986; Foster at al., 1989; Worthington et al. 2013a), and the ground reaction force (Worthington et al., 2013b). Conflicting results have been found between studies aiming to investigate the effect of individual technique parameters on ball release speed due to potential differences within the sample population, data collection techniques or statistical approaches.

Recent research has identified that a combination of technique parameters within the front foot contact phase (period between front foot contact and ball release) may be fundamental to maximising ball release speed (Worthington et al., 2013a). The study however adopted a cross-sectional research design which prevents the cause and effect of these technique parameters being determined. To determine cause and effect, a theoretical approach in which a forward-dynamics simulation model is used may be appropriate due to the motor system difficulties encountered when attempting to manipulate a single variable experimentally. Provided the model is evaluated for accuracy, this approach allows the user to manipulate one variable, and measure its effect on performance, whilst keeping all other variables constant (Yeadon & King, 2018).

Three-dimensional (3D) forward-dynamics simulation models are increasingly being adopted to analyse human movement, but their application to investigate the effect of technique on performance is restricted due to the limitations on obtaining realistic subject-specific 3D strength constraints (Yeadon & King, 2018). Ferdinands et al. (2008) has a three-dimensional whole-body forward-dynamics simulation model driven using kinetic inputs derived from inverse dynamics as part of a preliminary methodological paper and a two-segment arm model (Ferdinands et al., 2002). The application of these models to investigate kinematic effects are restricted due to the inability to optimise the model within subject-specific strength constraints. Middleton et al. (2015) also developed a three-dimensional forward-dynamics kinematic model, driven using the angular waveforms calculated from motion capture data between front foot contact and ball release. To investigate the effect of differing elbow kinematics on wrist velocity the elbow waveforms were manipulated. This method also fails to consider whether the manipulated waveforms are realistic and whether the corresponding joint torques are feasible within subject-specific strength constraints.

To overcome the limitation on determining 3D subject-specific strength parameters, a planar approach is often adopted since subject-specific strength parameters can be obtained in vivo (King et al., 2006). While previous research has shown that the kinematics and kinetics of the front foot contact phase of the fast bowling action can be accurately reproduced using a planar representation, the simulation model was angle-driven and incorporated a method to allow the hip and shoulder joint centres to be non-coincident (Felton et al., 2019). While it was concluded that this method was suitable to incorporate the non-sagittal plane rotations of the pelvis and torso in a planar representation, the method of implementing it in a torque-driven model requires evaluating to ensure model accuracy.

The weakness of many simulation models is that the level of accuracy is unknown (Yeadon & Challis, 1994). If a simulation model is to be used to investigate optimum performance, it is crucial that the model is appropriately evaluated. King et al. (2006) argue that if research is to be considered ‘scientific’ then it should be a necessity that simulation models are evaluated both kinematically and kinetically.

The aim of this paper is to develop a torque-driven simulation model of the front foot contact phase of fast bowling, assess the model accuracy and optimise fast bowling performance.

METHODS

A male fast bowler who was 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) participated in this study. All procedures were approved by Loughborough University’s Ethics Committee and written informed consent was obtained from the participant prior to the study commencing.

Data Collection

Kinematic and force data were collected at the National Cricket Performance Centre. Fifty 14 mm retroreflective markers were placed in positions on the bowler’s body in order that joint centres could be calculated (Felton et al., 2019). An additional 15 x 15 mm reflective patch was attached to the ball to enable ball release velocity and the instant of release to be determined (Worthington et al., 2013a). Eighteen MX13 Vicon cameras (OMG Plc, Oxford, UK) sampling 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 sampling at 1800 Hz (Type 9287B, Kistler AG, Switzerland). The participant bowled 12 maximal effort deliveries of a good line and length (directed towards and landing 6-8 metres in front of the target wickets), striking the force plate with their front foot (Felton et al., 2019).

Subject-specific torque and inertia measurements were also collected from the participant. Maximal voluntary torque data were obtained assuming bilateral symmetry using a Con-Trex MJ isovelocity dynamometer for flexion and extension of the ankle, knee and hip on the non-dominant side of the body (corresponding to the front leg in the bowling action) and flexion and extension of the shoulder on the bowling arm (King et al., 2006). Isometric measurements were taken at seven joint angles (five at the ankle) evenly distributed throughout the joint range of motion. Isovelocity measurements were taken at increments of 50°s^{-1} between 50°s^{-1} and joint specific maximums of: hip - 250°s^{-1}; shoulder - 275°s^{-1} (last increment 25°s^{-1}); ankle - 300°s^{-1}; and knee - 400°s^{-1}. Ninety-five anthropometric measurements of the bowler were also taken in order to calculate subject-specific inertia parameters using Yeadon's geometric inertia model (Yeadon, 1990).

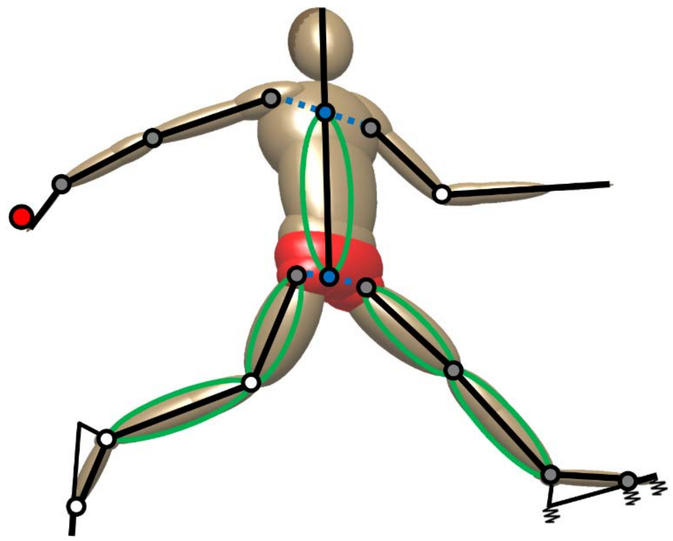

Figure 1 - 16-segment simulation model with wobbling masses within the shank, thigh and trunk segments, torque drivers at the ball, ankle and knee on the front leg, both hip and shoulder joints and the elbow and wrist on the bowling arm (grey circles), angle drivers at the ball, ankle and knee on the rear leg, as well as the non-bowling arm elbow (white circles), two massless segments (blue lines) connecting the hip and shoulder joint centres to the torso (blue circles) and spring-dampers at three points of contact on the front foot.

Simulation model

A 16-segment planar torque-driven computer simulation model (Figure 1) was constructed using AUTOLEV (Online Dynamics, 1990) for the front foot contact phase of the fast bowling action (Kane & Levinson, 1985). The 14 segments comprised: 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 included within the shank, thigh and trunk representations. Non-linear spring dampers connected the ends of the wobbling and rigid segments (Pain & Challis, 2001). Two massless segments (pelvis and shoulder girdle) with variable length and orientation connected the bilateral hip and shoulder joint centres, allowing non-coincident hip joint centres and non-coincident shoulder joint centres to represent the non-sagittal rotations of the pelvis and torso which occur during the front foot contact phase (Felton et al., 2019). In addition, to incorporate lateral side-flexion the length of the torso plus head segment was allowed to vary while adjusting the inertia parameters to take into account the change in length (Felton et al., 2019). Horizontal and vertical linear spring-dampers were used at the three points of contact (heel, metatarsophalangeal joint (MTP), and toe) to represent the foot-ground interface of the front foot (Felton et al., 2019). The ball was represented as a point mass and attached to the end of the bowling hand using a viscoelastic spring to ensure a smooth release (Felton et al., 2019).

The model was driven by torque generators consisting of contractile and series elastic components which were employed to flex and extend both shoulder and hip joints, as well as the knee, ankle and MTP joints on the front leg and the elbow and wrist joints on the bowling arm (King et al, 2006). The elbow on the non-bowling arm, and the MTP, ankle and knee joints on the rear leg were angle-driven since previous research has not identified these joints to significantly impact performance during the front foot contact phase (Ferdinands et al., 2014).

Input to the simulation model comprised the mass centre position and velocity, the trunk orientation angle and angular velocity, and the angle and angular velocity of each joint at front foot contact. Model parameters comprised the joint angle time histories of the angle-driven joints, the viscoelastic parameter values for the wobbling mass and foot-ground interface, and the torque parameters and the activation profiles for each of the torque generators.

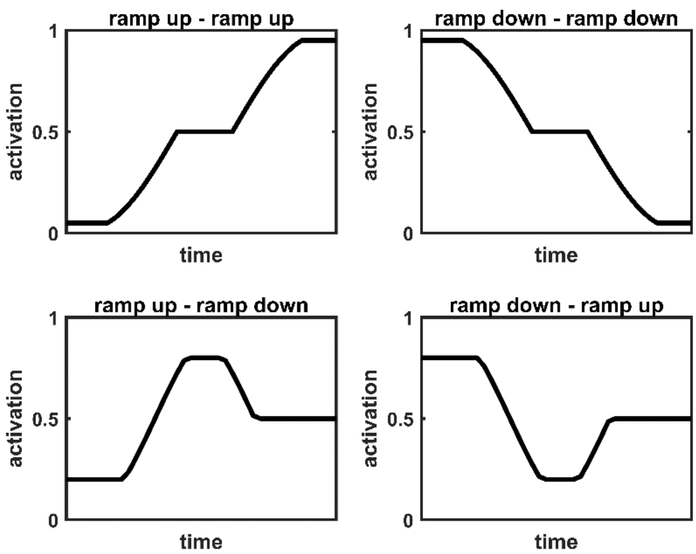

Figure 2 - Potential torque activation profiles possible in the simulation model.

The activation of each torque generator at any given time was governed by a quintic spline with zero accelerations and velocities at the end points (Yeadon & Hiley, 2000). The minimum time in which the activation could ramp up from zero to maximum was 70 ms (Yeadon et al., 2010). Torque profiles were either: ramp up - ramp up, ramp up - ramp down, ramp down - ramp up or ramp down - ramp down (Figure 2). Seven parameters were used to defined each torque profile and comprised: the time of the onset of the activation ramp, the initial activation level, time taken for first ramp to ramp up from zero to maximum (or if ramp down, from maximum to zero), the activation level after the first ramp, the time of onset of the second ramp, time taken for second ramp to ramp up from zero to maximum (or if ramp down, from maximum to zero), and the final level of activation.

The output from the model comprised the mass centre position, trunk orientation and joint configuration angles, ground reaction forces and ball release velocity.

Parameter determination

To determine the kinematic inputs required for the simulation model, the three best bowling trials (greatest ball velocity with the correct line and length and minimal marker loss) were processed within the Vicon Nexus software and joint centres calculated between front foot contact and ball release (Felton et al., 2019). 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 exceeded the length of the hand (taken from the anthropometric measurements) indicating the ball was no longer in contact with the fingers. Ball release velocity was calculated as the average resultant velocity over the first ten frames after ball release (Felton et al., 2019). The trunk orientation angle (angle of the trunk in the global coordinate system) and the joint configuration angles were determined using the projections of the joint centres onto the sagittal plane. Quintic splines (Wood & Jennings, 1979) were fit to the time histories of the joint configuration angles for input to the simulation model, with the closeness of fit at each point based on the difference between the actual data and a pseudo data set generated by averaging the two angles from the adjacent times points (Yeadon and King, 2002). The centre of mass position was also calculated using the segmental inertia parameters collected from the bowler (Yeadon, 1990).

The lengths (the distance between the projected hip joint centres, the distance between the projected shoulder joint centres, and the distance between the mid-point of the projected hip joint centres and the top of the head) and joint configuration angles of the massless segments within the simulation model were driven by expressing them as a function of the trunk orientation angle. To determine these functions, a third order Fourier series approximation was fitted to the coordinates from the three best recorded performances.

Subject-specific torque parameters were calculated from the isovelocity dynamometer data to produce a nine-parameter maximal torque function for flexion and extension of the ankle, knee, hip, and non-bowling shoulder joints. Each nine-parameter joint torque generator function allowed maximal voluntary torques to be determined within the simulation model spanning the joint angle and angular velocity ranges of the movement (King et al., 2006; Yeadon et al., 2006). The torque function for the wrist was determined using the method of King et al. (2009). Preliminary analysis of the bowling shoulder joint configuration angle during the front foot contact phase revealed the planar joint angle to be outside the flexion/extension joint range collected on dynamometer due to a combination of non-planar internal/external rotation and abduction/adduction. Instead, a torque function for the bowling shoulder was determined where the flexor and extensor joint torques were combined and represented as a constant. The value of this constant was determined by two-dimensional inverse dynamics with the maximal flexor and extensor torque values across the four bowling trials used. Similarly, as the bowling arm elbow remains at its anatomical limit (straight) throughout the front foot contact phase the maximal voluntary torque was assumed to be small and would have a negligible effect so no active torque generator was employed and the movement was restricted using a passive viscoelastic element (Felton & King, 2016). Passive elastic elements were also used at the ankle, knee, hip, and wrist to stop the joints exceeding anatomical limits (Riener & Edrich, 1999; Hiley et al., 2015). As the movement of the MTP joint was small due to the structure of the shoe and floor, no active torque generator was employed, and movement was restricted using a passive torque in the form of a linear rotational spring.

The viscoelastic parameter values consisting of the wobbling mass and foot-ground interface coefficients were determined using a 16-segment angle-driven simulation model (Felton et al., 2019). To ensure the viscoelastic parameters were appropriate penalties were utilised to prevent excess compression at the ground (Allen et al., 2012). Wobbling mass movements were limited to a maximum of 45 mm at the shank, 70 mm at the thigh, and 100 mm at the trunk (Lafortune et al., 1992; Minetti and Belli, 1994). None of the penalties were incurred for the determined viscoelastic parameters.

Model Evaluation

The torque-driven simulation model was evaluated by assessing how accurately a simulation could match the performance data for the best recorded performance (the performance with the fastest ball release speed). This simulation was found by varying a total of 119 parameters via a genetic algorithm to minimise an objective score function representing the difference between a simulation and the kinematics and kinetics of the performance (Carroll, 1996). The 119 parameters comprised: 10 parameters varying the initial kinematic conditions, 10 parameters varying the initial massless segment conditions, 7 parameters varying passive torque elements, 91 parameters varying the torque generator activation timing and one parameter varying the timing of ball release. The 10 initial kinematic and the 10 initial massless segment parameters were used to allow small amounts of variation to compensate for possible inaccuracies in the initial conditions (Hubbard & Alaways, 1989). The 7 passive torque elements comprised the stiffness of the torsional spring employed at the MTP, and the stiffness and damping coefficients governing the elbow torque profile. The 91 torque generator activation timings comprised the parameters from the profiles of all the torque-driven joints.

The objective score function value F (Equation 1) was calculated as the overall root mean square (RMS) difference between the simulation and performance for six components:

$$\mathrm{S} = \sqrt{\left(\frac{{S_1}^2 + {S_2}^2 + {S_3}^2 + {S_4}^2 + {S_5}^2 + {S_6}^2}{6}\right)} \qquad (1)$$

where S_1 is the average of the horizontal and vertical force RMS time history differences expressed as a percentage of the peak vertical force, S_2 is the sum of the mass centre horizontal and vertical velocity absolute differences at release expressed as a percentage of the resultant mass centre velocity at ball release, S_3 is the trunk orientation angle RMS time history difference in degrees, S_4 is the sum of the horizontal and vertical ball release velocity absolute differences expressed as a percentage of the resultant ball release velocity, S_5 is the difference in duration of the front foot contact phase as a percentage of the actual duration, S_6 is the overall RMS difference of the nine torque-driven joint angles in degrees.

Each component was weighted equally with 1% difference in $S_{1,2,4,5}$ considered comparable to 1º difference in $S_{3,6}$ (Yeadon and King, 2002). This approach was taken to prevent the objective function being weighted towards the kinematic or kinetic characteristics. The trunk orientation angle component (S_5) was given equal weighting verse the nine joint angles as the trunk orientation angle represents the whole-body orientation whereas the joint angles define the configuration (Yeadon and King, 2002).

Model Optimisation

Technique was optimised in order to maximise ball release speed. The 10 parameters varying the initial kinematic conditions, the 10 parameters varying the initial massless segment conditions and the 7 passive torque parameters were the same as the matched simulation and the 91 torque activation parameters were varied using a genetic algorithm. The initial position at front foot contact was identical to the matched simulation and ball release occurred when the bowling arm had passed the vertical and the predicted horizontal landing distance was equal to the matched simulation. Penalties were imposed if any of the joints exceeded anatomical limits.

RESULTS

The torque-driven simulation model was able to closely match the recorded performance (Figures 3-5) with an objective score function of 4.0% (Table 1), demonstrating sufficient complexity for subsequent optimisation of performance. The kinematic difference score (disregarding the force component) was very low, 0.9%, which suggests the simulation can accurately reproduce the kinematics of the front foot contact phase. None of the penalties regarding anatomical limits of the joints were incurred in the matching simulation.

Table 1 - Objective score function component differences between the matched simulation and performance

component	difference
S_1 – force (%)	9.59
S_2 – COM (%)	0.06
S_3 – orientation (°)	0.67
S_4 – ball velocity (%)	0.03
S_5 – time (%)	0.19
S_6 – joint angles (°)	1.81
RMS difference (%)	**3.99**

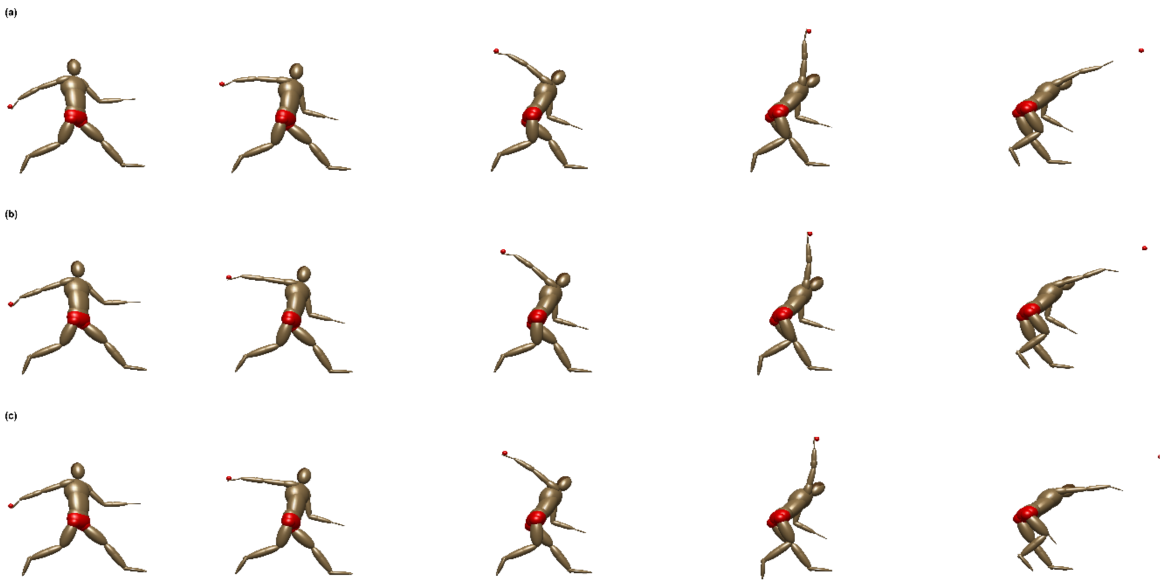

Figure 3 - (a) performance, (b) matched simulation, and (c) optimised simulation, of the front foot contact phase of fast bowling.

Optimisation of the technique during the front foot contact phase yielded an increase in ball release velocity of 9.8% compared to the matched simulation (38.8 m/s vs. 35.3 m/s). The optimisation chose a technique where the front ankle and knee remained extended compared to the matched simulation (Figures 3 & 4). The front hip also remained more extended in the optimised technique before flexing to allow more trunk flexion to occur. Similarly, shoulder extension of the bowling arm occurred later in the optimised technique, while rear leg hip flexion occurred earlier and faster. (Figures 3 & 4). The optimised technique also resulted in an increase in horizontal ground reaction force whilst the vertical ground reaction force remained similar to the matched simulation (Figure 5).

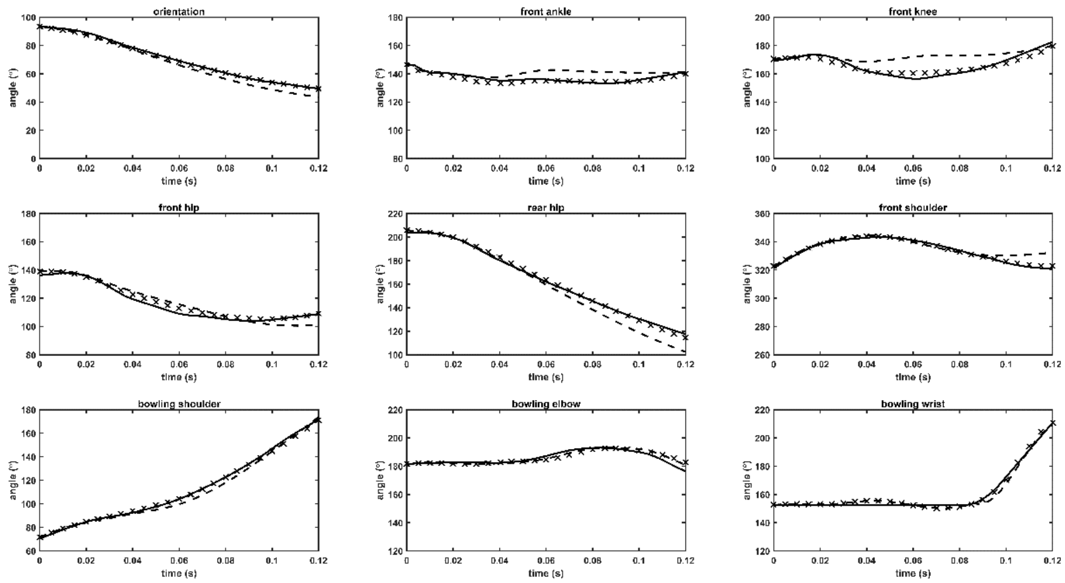

Figure 4 - Joint angle time histories of the recorded performance (solid lines) and the matched (crossed lines) and optimised (dashed lines) simulations.

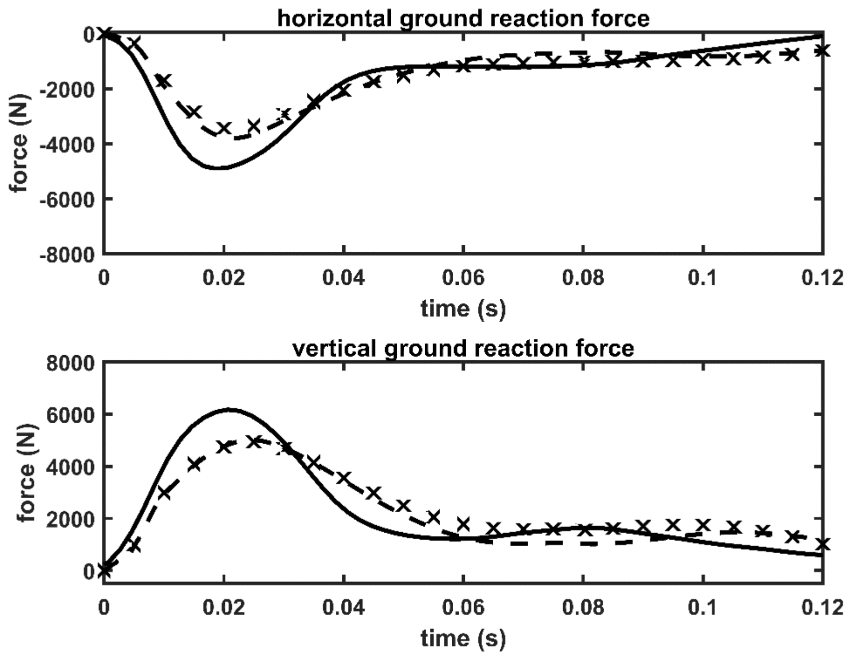

Figure 5 - Ground reaction force time histories of the recorded performance (solid lines) and the matched (crossed lines) and optimised (dashed lines) simulations.

DISCUSSION

The current study investigated the assumption that the optimal technique for ball The aim of this paper was to develop, evaluate and optimise a torque-driven simulation model of the front foot contact phase using both kinematic and kinetic variables. Close agreement between performance and simulation was achieved with an overall difference of 4.0%, resulting in a ball release speed difference of 0.03%. The average kinematic difference of 0.9% indicates that the simulation model can accurately reproduce the kinematics of the front foot contact phase of fast bowling. Additionally, the kinetic differences of 9.6% show reasonable agreement in a model which incorporates pin joints (Allen et al., 2012). Optimising the simulation model by varying the torque activation parameters led to an increase of 9.8% (3.5 m/s) in ball release speed which may be considered to be feasible and suggests that the subject-specific strength parameters are appropriate.

The biggest difference between the matched and optimised simulations can be seen in the front leg (Figures 3 & 4). The ankle and knee joints remain more extended throughout the front contact phase on the front leg in the optimised simulation compared to the matched simulation. These differences in the front leg kinematics act to increase the horizontal ground reaction force rather than the vertical ground reaction force (Figure 5) indicating that the horizontal component of the ground reaction force is more important to fast bowling performance than the vertical ground reaction force. This finding agrees with Portus et al. (2004) who found in 42 high performance male fast bowlers that the faster bowlers had higher peak horizontal ground reaction forces, and King et al. (2016) who found that the faster bowlers had greater horizontal impulses. Future optimisations should investigate the effect of changing the initial configuration and its effect on ball release speed and ground reaction forces.

The optimised simulation employs a technique where trunk flexion is initially delayed but then flexes further than in the matched simulation (Figures 3 & 4). This agrees with previous research which has found that the amount of trunk flexion during the front foot contact phase of fast bowling is positively correlated with ball release speed (Davis & Blanksby, 1976a; Elliot et al., 1986; Burden & Bartlett 1990b; Portus et al., 2004; Worthington et al. 2013a). It has been proposed that the amount of trunk flexion during the front foot contact phase is a mechanism of the bowling action rather than a result of the muscular work in the torso (Worthington et al. 2013a). The differences in technique between the matched and optimised simulations potentially support this statement. As discussed earlier, the optimised solution employs front leg kinematics which more efficiently brakes the lower half of the body, increasing angular momentum, and as a result produces more trunk flexion. It is probable that the delay in the onset of trunk flexion is due to changes in the proximal to distal sequencing within the kinetic chain when straighter front leg kinematics are used.

Another distinguishable difference is that the onset of circumduction of the bowling arm is delayed in the optimised simulation compared to the matched simulation (Figures 3 & 4). This mechanism was first identified by Tyson (1976), and later by Worthington et al. (2013a) as one of the key features which explained the variance in ball release speed across 20 elite fast bowlers. Although varying the initial position of the bowling arm was outside the scope of this paper, the optimised solution delays arm circumduction to occur after trunk flexion has started (Figure 4). It is probable that there is a relationship between the amount of trunk flexion and the delay in the bowling arm which maximises performance. For example, if the bowling arm is delayed, the trunk can flex further during the front foot contact phase, while still being able to release the ball towards the intended target. The results of this study appear to suggest that a relationship between trunk flexion and the bowling arm delay may exist, although further investigation is required to understand this relationship and its effect on performance.

The final observable difference in the optimised technique is the earlier onset of rear hip flexion compared to the matched simulation (Figures 3 & 4). Although the motion of the rear leg during the front foot contact phase has not previously been linked to ball release speed, coaching literature has previously advocated it (Ferdinands et al., 2013). Theoretically, rear hip flexion brings the distribution of mass of the rear leg segments closer to the rotation axis of the torso during the front foot contact phase. This will affect the relationship between rear leg motion and trunk flexion. It is proposed that the earlier onset of rear hip flexion in the optimised simulation reduces the moment of inertia of the body about the front hip, therefore maximising trunk flexion during the front foot contact phase, which is known to be linked to ball release speed (Davis & Blanksby, 1976a; Elliot et al., 1986; Burden & Bartlett 1990b; Portus et al., 2004; Worthington et al. 2013a). Rear hip flexion may also help maintain linear momentum and stability. Future research is required to understand the role of rear leg kinematics on fast bowling performance.

The complexity of the model was kept to a minimum by adopting a planar approach with the non-planar rotations of the pelvis and torso being incorporated using massless segments. The method of driving the lengths and orientations of the massless segments as a function of the trunk orientation angle using Fourier series approximations was considered suitable, since the optimisation algorithm was able to find a simulation with close agreement with the recorded performance. A limitation of this approach is that the non-planar rotations of the pelvis and torso cannot be perturbed as it is not possible to determine whether the new rotations would be feasible and the corresponding torques are within subject-specific strength constraints.

The optimisation of the front foot contact phase of fast bowling for the same initial conditions as the evaluation simulation produced a 3.5 m/s faster ball release speed.

This optimised simulation collaborates pre-existing experimental research on fast bowling biomechanics and predicts a ball speed, 38.8 m/s, which is comfortably within a range previously observed in elite fast bowlers (32.8 – 39.7 m/s: Worthington et al., 2013a). This suggests that the subject-specific strength estimates are suitable. If the model strength estimates were too high it might be expected that the optimised simulation would be unrealistic whereas if they were too low, then the evaluation would be poor (Yeadon & King, 2002). A limitation of the optimisation procedure is that the objective function maximises ball release speed for a single set of activation parameters. This may lead to a solution which Yeadon (2004) describes with a landscape metaphor: *"an optimisation routine may find the top of a pinnacle which stands on a narrow base high above the surrounding terrain. Even if this is the global optimum it is a summit that should not be attempted, since any small location error will land on the low terrain"*. In the future, additional work is required to consider how robust optimum solutions are to perturbations of the activation timings, as it is likely elite performers develop techniques that are relatively insensitive to perturbations.

This study has developed and evaluated a torque-driven simulation model of the fast bowling action. The close matching of the simulation with performance indicated that the model complexity was sufficient to accurately reproduce the kinematics of a recorded performance. The optimised simulation produces an increase in ball release speed which is feasible and collaborates previous experimental research which links the fastest bowlers with straighter front leg kinematics, increased trunk flexion and a delay in circumduction of the bowling arm. This indicates that the subject-specific strength parameters are appropriate and not too strong. In the future this simulation model could be used to vary other parameters such as mass centre velocity, front leg plant angle, and knee angle at foot contact and observe the effect on ball release speed. In addition, the model can be used to assess the influence of strength on optimum fast bowling technique.

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