MATHEMATICAL MODEL FOR THE ESTIMATE OF THE CENTER OF MASS KINEMATICS IN SPRINT START

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INTRODUCTION

In track and field sprint races, the final performance is highly dependent on an effective starting procedure. The primary goal of athletes is to perform this motor action with the shortest delay time between the start signal and the beginning of the movement, trying to achieve the highest horizontal velocity of the body center of mass (BCM) during the acceleration phase. Concurrently, for the athlete the task of regulating dynamic balance to remain upon his feet during the early start phases, represents an extremely complex motor control problem. Considering the high forces the sprinter exerts on the ground and the narrow width of the base of support during the single support phase, frontal plane balance is especially difficult and requires precise control of the BCM position and velocity.

Two approaches have been used wer the years for the assessment of BCM kinematics and his derivatives during sprint start:

1) measurement of ground reaction force data and the use of integration procedures for the estimation of BCM velocity and position (Mero et al., 1983; Cappozzo et al., 1989). One of the drawbacks of this method is that the analysis is limited to the time of the foot-force transducer contact. Furthermore, there is a complete lack of information on the kinematics of body segments and articular joints;

2) measurement of positional data, taken from two-dimensional film analysis, and the use of this information in combination with magnitudes and locations of segmental masses in which the body was modelled for BCM kinematics computing in the sagittal plane (Desipres. 1973; Mero et al., 1983, Mero. 1988; Jacobs et al., 1992). Though this approach has the merit of allowing the monitoring of BCM kinematics during a wider range of time, it does not permit the analysis of the lateral BCM trajectory. Moreover, in previous models the external landmarks position is intended to compute directly the joint centers of rotation. This 'assumption, in conjunction with the use of two-dimensional technique, may results in large errors in the linear and angular kinematics due to external-internal rotation of the limbs and to projection errors, which depend on the alignment of the segments with the film plane.

The present work provides a three-dimensional model for the calculation of BCM and body segments kinematics, by measuring the 3-D coordinates of a reduced number of external markers (technical markers) and computing the position of the correlated internal markers (internal reference points) (method 1). The procedure adopted to validate the model and the preliminary results obtained by junior sprinters are reported to illustrate the potentialities for practical applications. Knowing the high accuracy of force plate measurement system (Harman, 1991), in this study GRF data (method 2) was only used to validate the proposed model.

METHODOLOGY

Subjects and apparatus. Two male junior sprinters (S1 and S2), ages of 16 and 15, participated in this experiment. In the first experimental session, devised to collect data to validate the model, they performed 12 squat long jumps using the arm swing and 12 standing sprint starts. In both cases the athletes commenced their movement with both feet on the force plate. In a following experimental session each subject performed starts under four different starting position: normal, normal elongated, inverse and inverse elongated. Out of each set position five trials were performed, for a total of 20 sprint starts.

The 3-D coordinates of each body marker were measured by the ELITE System, an automatic motion analyzer, developed by Femgno and Pedotti (1985). Four TV cameras (paired off on the two sides of the subjects) configuration was chosen to allow a double side - three dimensional analysis, with a 100 Hz sampling rate. The accuracy of the system set up was verified by moving a 400 mm rigid bar with two markers fixed on its extremities. The errors expressed as standard deviation of the measured distance were: 0.699 mm on the left side and 0.825 mm on the right side.

Ground reaction force (GRF) signals were measured by means of a Kistler 928 1B piezoelectric force plate, with a sampling frequency of 1000 Hz. The software of ELITE System was utilized for the GRF acquisition, synchronized with kinematic data.

Filtering of 3-D markers coordinates and their derivatives computing were performed by using an algorithms based on an auto-regressive model, fitted to the signal, to evaluate the filter bandwidth and the extrapolation-of the data (D'Amico and Femgno, 1990).

Mathematical model. The model here proposed allows the estimation of the position of internal points of interest, like the joint rotation centers, from the position of marked external landmarks. The model has been designed in order to match feasibility with accuracy providing the kinematics of 16 rigid segments by means of the coordinates of only 18 external anatomical landmarks. Due to the inevitable simplifications introduced, the use of the model is constrained to movements in which large rotations of the body segments are negligible around their longitudinal axes, like in vertical jumping exercises, in running and in sprint starts. The 16 segments, include feet, lower legs, thighs, pelvis, abdomen, thorax, head-neck, upper arms, forearms and hand segments. Anthropometric estimates of each segment's mass and center of mass position were calculated using the tables provided by and Zatsiorsky and Seluyanov (1985).

During a preliminary tests it was found that camera views of shoulders' reference point were obscured for unacceptable periods; therefore, it was necessary to fix the markers with an offset, measured in a preliminary phase, to the real shoulder reference points.

RESULTS AND DISCUSSION

The validation procedure showed a high level of agreement which support the validity of the proposed model. The shape and magnitude of the BCM kinematics obtained with the model and by means of GRF were rather identical, although some minor discrepancies exist, in particular when the medio-lateral components were considered. In all trials of all subjects individual correlation was 0.97. 0.96 and 0.90 respectively for the vertical, the horizontal and the medio-lateral component. In table I, the mean and standard deviation of the percentage error between the instantaneous BCM velocities, computed with method 1 and 2 at the take off, are presented.

	Antero-posterior	Vertical	Medio-lateral
Standing Start	2.56 (1.71)	5.26 (3.53)	9.1 (5.5)
Squat Long Jump	5.74 (0.74)	2.24 (2.10)	12.5 (7.3)

Table I. Average error per cent and standard deviation between the velocity components computed by means of the model and by GRF integration.

The interpretation of the observed differences is difficult, there are at least four possible explanations of this phenomenon:

1) no correct estimation of segment masses and their center of mass location. Being the subjects of this work young people, errors may be amplified by the so called "age effect" which could affect the mass distribution calculations resulting in differences between predicted values and real ones;

2) modelling errors due to the assumptions and simplifications incorporated in the model. The more likely source of errors concerns the locations of the mass center of the upper and lower trunk and of their relative estimated joint rotation center. Systematic errors may have also resulted from the rotation of the head and neck about transverse axis. Since these four segments accounted for approximately 50% of total body mass is possible that substantial errors resulted from this source;

3) force platform characteristics. Although the method 2 seems less prone to error since it is only based on measurements of GRF and the total body mass of subjects, it may be affected by disturbing factors such as crosstalk among transducers and low frequency errors included in force signals. In particular low frequency errors, even if little, were considerably grown up by integral procedure to obtain velocity and displacement;

4) skin artefacts affecting markers' position. Although the use of reconstructed internal markers might partially reduce the influence of this phenomenon, it influences the computed kinematics of the BCM.

As expected, the velocity's time-curves were fluctuating in nature and gradually increased stride to stride with the medio-lateral component comparatively less in magnitude. The examination of the plots combined with the quantitative information allows an individual analysis and comparison of the techniques for each subject. This may be easily seen from table II and III, where the comparison of same parameters between S1 and S2, and data out of four different kinds of start position for S1 are presented. The values reported are referred to the instant of the first stance phase toe-off. S1 showed a higher instantaneous BCM horizontal velocity and less vertical and lateral velocity components, indicating a more effective starting procedure compared with S2. This was confirmed by the comparison of the interval time the athletes need from the actual movement start until the instant considered (0.598 vs. 0.700).

	Antero-posterior	Medio-lateral	dt
Subject 1	4.372 (0.092)	0.203 (0.022)	598 (14.5)
Subject 2	3.614 (0.077)	0.361 (0.034)	700 (21.2)

Table II. Velocity components at the end of the first step

	Antero-posterior	Medio-lateral	dt
Usual	4.372	0.203	589
Usual elongated	3.827	0.220	655
Inverse	4.226	0.328	1621
Inverse elongated	4.102	0.395	638

Table III. Velocity components depending on starting modality at the end of the first step

As it can be seen, the normal starting position (i.e. the position the athlete usually used in competition) seems to be the most effective for S1. In it fact provides the highest horizontal velocity with the lowest medio-lateral component. However, this result might be considered with caution and does not negate the possibility that practice

or training over an extended period of time would improve the athlete's performance in one of the other starting position.

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