SUBJECT-SPECIFIC MUSCULOSKELETAL MODEL TO IDENTIFY MUSCLE CONTRIBUTION TO THE ACCELERATION PHASE IN ELITE SPRINTING

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The purpose of this study was to identify which lower limb joint moments of force and lower limb muscles had the highest contribution to the forward and upward acceleration of the Body Centre of Gravity (BCG) in an elite sprinter during the first step after starting from blocks in 100 meters dash. An induced acceleration analysis approach was used to mechanically ascertain this contribution. Two models were used, one based in a simple multilink rigid body's model and a second using a musculoskeletal model developed using Opensim. Our results showed a possible synergist action between hip and ankle extensors that allow an optimal combination that resulted in a forward propulsion of the BCG, this results seem to be corroborated using an Opensim musculoskeletal model.

KEY WORDS: IAA, Musculoskeletal Model, Sprint, Modelling.

INTRODUCTION: The sprint start and subsequent acceleration phase are crucial in sprinting success. Several studies report that best times in the 100 meters are achieved by athletes who generate the highest horizontal velocities when leaving the block. Considering that the main purpose of the sprint start is to generate the greatest horizontal velocity in the shortest time interval, it is crucial, from a performance point of view, to understand how the athlete accelerates its BCG. Usually the biomechanical determinant factors of sprint performance are studied and described using a combination of conventional kinematics and kinetics variables related using statistically based methods, usually correlating biomechanical variables and performance (Bezodis et al, 2014). Nevertheless there are biomechanical and simulation based methods that allow us to estimate the possible contribution of the different athlete's musculoskeletal system actuators to the acceleration of the body by solving the equation of motion using as input the experimental inverse dynamics estimations. One of this methods is the Induced Acceleration Analysis (IAA) that could be used to estimate the CG acceleration portion generated by each joint moment, or each muscle when individual muscle forces are estimated using a musculoskeletal model (Dorn et al., 2012). The present study aims to investigate the contribution of each joint moment, and each muscle force, to the overall acceleration of the CG, immediately after sprint block start in one elite sprinter.

METHODS: After his warm-up routine, the participant, a male National Elite Sprinter (1.87m; 80Kg; 25yrs; 10.21 s yearly 100 meter dash best result), performed a series of block starts. The motion and ground reaction forces of the first step, after leaving the blocks, were captured at 200Hz using an optoelectronic system of 10 infrared cameras (Qualisys Oqus 300 and Qualisys Track Manager), synchronized with a strain-gauge force plate (FP6090-15-2000, Bertec Corporation, Columbus, USA). The best trial was selected for analysis. A multibody biomechanical model was developed using Visual 3D motion analysis software. The model composed by 7 rigid segments (HAT, bilateral thighs, shanks and feet) was built and optimized through inverse kinematics. Net joint moments of force were computed using inverse dynamics. The joint moments of force obtained for ankle, knee and hip of both lower limbs during the support phase of the first step of the sprint were used in order to evaluate the contribution of all joint moments and gravity to the horizontal and vertical acceleration of the participant's center of mass. This contribution was computed through joint moment induced

acceleration analysis (Kepple et al, 1997) based on equation 1 and considering the 7 rigid multibody biomechanics model described earlier using SD/Fast solver.

$$\mathbf{q} = M^{-1} T + M^{-1} C + M^{-1} G + M^{-1} F \qquad \text{(equation 1)}$$

Were **q** is matrix of the joints acceleration, M^{-1} is the inverse of the inertia matrix, T is the joint moments matrix, C is the Coriolis terms, G is the gravitational terms matrix and F is the external forces matrix.

The results of moment of force IAA are sensitive to the biomechanical model used and particularly to the contact model between the body and ground. In the present study two contact models were used, the foot-floor contact was initially modeled as a hinge joint, which allowed the foot to rotate over its center of pressure, about an axis aligned with the foot's medio-lateral axis (Free-foot model) and secondly the IAA was calculated with a contact model in which the foot was fixed to the floor (Fixed-foot model). The use of each model was decided based on the observed angular position the foot to floor angle. The fixed-foot model was used only when this angle was kept almost unchanged and the free-foot was used when the foot rotate in relation to the ground.

Additionally kinematic and kinetic data was exported to OpenSim (version 3.0) where a 92 muscle subject-specific musculoskeletal model was generated (Delp et al., 2007). The model was scaled geometrically based on the athletes' body markers placement but the muscle parameters were derived and scaled partially from direct measurements based on the methods proposed by Maganaris et al., (2001), Maganaris et al., (2003) and Erskine et al., (2009).. This data was obtained using a combination of ultrasound (US) during isokinetic dynamometer testing combined with MRI data from tight and shank. Muscle architectural parameters from the quadriceps and gastrocnemius muscles groups were estimated by using MRI data to measure the muscle volume and US to obtain pennation angle, and physiologic cross sectional area (PCSA) when MRI and US data was combined. Using the referred direct methods the following parameters were also obtained: optimal fiber length, maximal force at optimal fibbers length. The remaining lower limb muscles parameters were estimated using linear transformation in order to reflect the same proportional differences between muscles in the data base used on the standard Gait2392_Simbody Opensim model and the data obtained by our direct measurements. The movement data together with the correspondent GRF was used to perform residual reduction (RRA) followed by a Estimation of muscle activation and muscles forces using Computer Muscles Control (CMC) in OpenSim using the adjusted muscle parameters described earlier (Delp et al., 2007).

RESULTS: Our results for the joint moments IAA using the fixed foot model during first ¼ of support and the free foot model after shows that the hip extensor moment contributes to CG acceleration and that during the first quarter of the stance phase its contribution to the vertical CG acceleration is higher than that from the plantarflexor moment. The same was visible in the horizontal acceleration. The results from fixed foot only apply to the first quarter of stance phase, when the foot is relatively immobilized in relation to the ground. For the remaining task duration the analysis was performed with the free-foot model. Therefore, after the first quarter of the stance phase the hip moment contribution to accelerate the CG reduces, and the ankle plantar flexor becomes the main contributor to both the horizontal and vertical CG acceleration (Figure 1). The rationale of this solution is that the plantar flexor muscles during the first quarter of the contact time are acting synergistically as stable base for the actuation of the hip extensors against the ground. As a result the hip extensors accelerate the CG forward. In order to ascertain if this rationale is correct it will be crucial that the musculoskeletal models results showed a simultaneous strong activation of both hip extensors and plantar flexor in this period of the support.
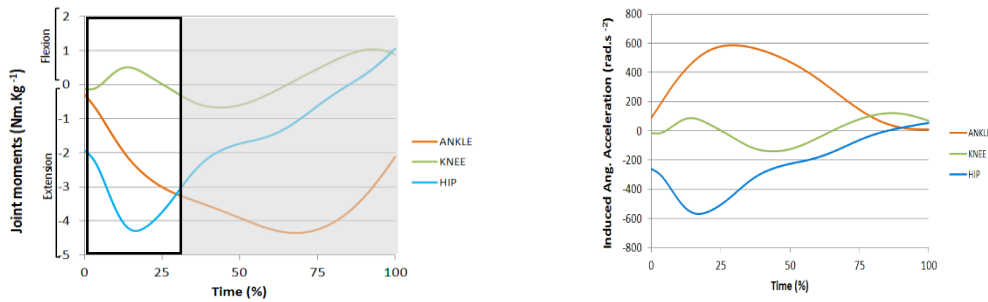
Figure 1 – Estimated joints moments of force and foot-floor "joint" angular induced acceleration. Joint moment's data is showed normalized to body mass and duration of support while induced angular acceleration is presented in absolute values.

The results from CMC support the finding obtained by estimating the joints contribution. In fact the combined action of the hip extensors produce a considerable amount of force during the first ¼ of support and the gastrocnemius act predominantly in the main push-off phase. Soleus muscle is predominant and produces a substantial amount of force during all the support being probably responsible for the fixation of the ankle and as a consequence of the foot to floor stabilization during the ¼ of the support, guarantying the transfer of the hip extensors action against the ground in a synergistic action apparently showed by joints moments of force IAA analysis. The results of CMC also report important level of activation from knee extensors being the vasti muscles predominant in the first half of the support and the rectus femoris (RF) activated later.
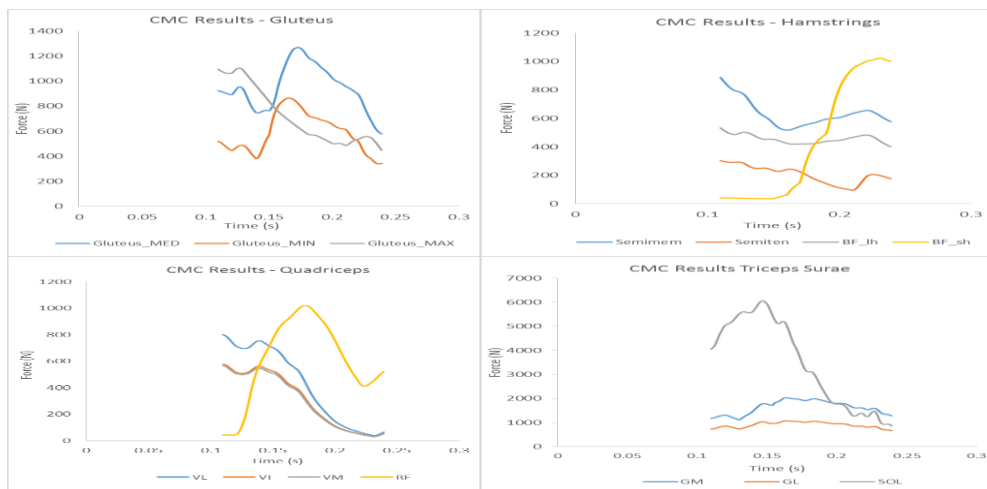

Figure 2 – Estimated muscles forces for Gluteus, Hamstrings, Quadriceps and Triceps Surae muscle groups using CMC routine in OpenSim. Data is presented in absolute values both for force estimations as well as for time.

Noticed that when using OpenSim the free foot-floor interaction was modeled with a pure rolling constraint meaning that the muscles results should explain the observed kinematics. As predicted on the joint moments of force IAA analysis the major muscle actuators are the hip extensors and the plantar flexors in particular the eccentric action of the soleus that seems to act in synergy with the hip extensors during the first stages of the support.

DISCUSSION: In the current study a synergy between the hip and the ankle joint moments was identified, where the hip action appears to be responsible for the forward movement of the pelvis at the first half of the support and the plantar flexor moment counteract this action resisting to the dorsiflexion effect of the predominant hip extensor moment. This interpretation

was corroborated by the results of the Opensim CMC that using the relevant force-activation coupling algorithms allow us to compute the muscle individual forces (fig 2). Our results show a strong participation of the triceps surae during the support and this action is able to maintain the foot to ground interface in a fixed position while the hip extensors provide the forward acceleration of the pelvis and trunk during the first half of the support. In the second half of the support the high levels of triceps muscle tension produces a strong plantar flexor moment that propels the BCG forwards when the hip extensor reduce the contribution for forward acceleration. The knee has a strong upward action allowing the rotation of the lower limb around the ankle joint in the first half of the support and actively extending the knee in the second half and as a consequence participating in the advance of the CG position accelerating it forward together with the plantar flexors. RF seems to have a strong contribution in this last action probably due to their bi-articular geometry. The relative contribution of plantar flexor and hip extensors and their synergistic action was also identified by Dorn et al., (2012) and was associated to the increase of running speed in collegiate sprinters.

CONCLUSION: The use of moment of force IAA analysis enable us to identify the main contributors to forward and upward acceleration in the acceleration phase of the sprint in one elite sprinter. Using a musculoskeletal model developed in Opensim, scaled based on individual muscle parameters obtained in vivo from the studied athlete (subject specific model), muscles actions were estimated and seem to confirm this synergist interplay between hip extensors and plantar flexors. The plantar flexors action is able to maintain the foot in a fixed position against the ground while the hip extensors propel the body forward in the first half of support and contribute to forward acceleration of BGC in the last stages of support. An obvious limitation off our work is that uses data from just one elite sprinter limiting the generalisation of our analysis.

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