Assessment by 3D modelling of the swimmer’s lower limbs muscles contribution during a grab start

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Abstract

The aim of the study was to assess a method that, through the estimation the lower limb muscle activity of a swimmer during the initial phase of the grab start, would allow optimizing the performance. The method uses, as input, only kinematic data. The musculo-skeletal system was modelled in the 3-D space using BRG.LifeMOD™, subject-specific morphological parameters and kinematic data obtained by a high speed video camera and passive markers located over points approximating the relevant joints. The model was validated by comparing the predicted speed and power values with experimental reaction forces data collected in situ. Our analysis showed the important role played by the Quadriceps, Hamstring, Tibialis Anterior and Soleus during the grab start.

Keywords: high level swimming, grab start, lower limbs muscles, kinematics, kinetics, 3D modelling.

1. Introduction

Regardless of style, the study of swimmers’ performance involves the identification and analysis of three technical phases: start, turn and stroke. In short distance races (50m and 100m), the start represents a particularly important factor. For instance, at the 2004 Athens Olympic Games, the time that separated the eight men’s 50m freestyle finalists was 0.44 s, representing only 2% of the winner’s total race time. The analysis of the temporal distribution has shown that the start phase accounted for 30 % for the 50 meters and 15 % for the 100 meters freestyle events of the total time. Thus, performance differences may be caused by different start phase durations.

A good start must allow the swimmer to plunge as far as he can in the shortest period of time and with an orientation favourable to the reduction of drag and to the adoption of an effective stroke.

In the aim to determine the important parameters of the start, lots of studies have been carried out. Many authors tried to identify the discriminating parameters between the track start and the grab start [1, 2, 3, 4, 5, 6, 7]. The results obtained did not show any real difference between them.

Concerning the grab start, different studies using kinetic and kinematic approaches have been published. They provide only some information concerning the relationships between the swimmers’ movements and their performance. For [8] and [9], the vertical and horizontal acceleration components of the swimmer’s centre of mass in the takeoff phase are good indicators of the effectiveness of the start. Another study has focused on the modelling of body movements during the start phases in order to compare the kinematics between a grab start and a track start [10].

The aim of the present study was to determine the variables that describe the mechanics of the grab start and that may be used to optimize performance. The reaction force, and thus, the force impulse generated by the swimmer musculature, were estimated in a high level swimmer using kinematic information. The contribution of the main lower limb muscles to the force impulse was determined by using a model of the musculo-skeletal system. Ground reaction forces, normally not measurable during competitions, were measured for model validation purposes only.
2. Methods

One national level male swimmer was asked to perform a grab start. His height and body mass were 179 cm and 72.2 kg, respectively. The swimmer was equipped with 41 passive markers according to the Plug-in-Gait protocol marker placement based on models of [11] and [12]. The movement studied could be described as follows. The swimmer stood on the platform with his feet shoulder-width apart. He clutched the edge of the platform with both his toes and with his hands. The centre of mass of the body was placed on the vertical axis passing through the anterior portion of the feet. The head remained flexed and motionless until the starting signal. At the starting signal, the trunk leaned forward first and after the legs followed. The arms were stretched forward, piloting the total body movement during this phase. The movement was recorded using three high speed video cameras (Photron Fastcam PCI at 125 frames per second). Simultaneously, the ground reaction was measured using a six component force-plate (AMTI OR6-7-2000) mounted on the starting platform with sampling frequency of 1000 samples per second. Before the capture, a calibration cage, equipped with fourteen control points, was placed on the starting platform. The global system of reference (X, Y, Z) was defined in the calibration cage and had the Y-X plane parallel with the starting platform (leaning by 6 degrees with respect to the horizontal plane). The video cameras were calibrated (Fig. 1) and clock synchronized (0.008 s accuracy).

![Fig.1. Position of the three cameras around the swimming pool.](image_url)

The image coordinates of the calibration object control points were digitalized using the Snap 3.4 software. The three-dimensional global position of the markers was reconstructed using the direct linear transformation (DLT) method [13] taking into account the reconstruction error [14].

MSC ADAMS – BRG.LifeMOD was used for the reconstruction of a model of the human musculo-skeletal system. This is viewed as a mechanical system made of members (segments) and joints under the control of
muscular forces. The geometric and inertial properties of the body segments and the location of the points of origin and insertion of muscles were drawn from anthropometric databases. The description of the mechanical properties of the physical environment (starting platform) permitted the inclusion of the contact reaction forces.

A two step process is used for the musculoskeletal simulation. In the first step, the muscles are created on the human body as training elements (passive elements). In the inverse-dynamics analysis, the body is in movement using the motion capture data fed to the model and muscle shortening/lengthening patterns are recorded. Then to perform the forward-dynamics analysis, these patterns serve as actuators to drive the motion.

The above-mentioned analysis entailed a number of assumptions. The forces exchanged between the hands and the starting platform were neglected. In fact, it has been shown that the role of the upper limbs in the force impulse generation is more than 30 times weaker compared to that of the lower limbs [15]. In addition, since, for the sake of determinacy, only the resultant reaction force acting on both feet could be estimated through inverse dynamics, the reaction forces acting on the individual feet was determined by dividing the resultant force by two. This entailed the assumption of sagittal symmetry of the movement.

The musculoskeletal model used to generate the solution was made up of 19 segments, 18 joints and 118 muscles. In this study, only the results concerning the lower limb model are presented. Each hip was modelled as a ball and socket joint (three rotations) and each knee [16] and ankle joint as a revolute hinge joint (one rotation in the sagittal plane).

These joints are regarded as constraints between two adjacent body segments. For this mechanical system, embedding ns segments, the expression is:

\[ \phi(q) = 0 \]

where the generalized coordinates

\[ q = \begin{bmatrix} q_1 \mid q_2 \mid q_3 \mid q_4 \mid q_5 \mid q_6 \end{bmatrix}^T = \begin{bmatrix} q_1, q_2, \ldots, q_n \end{bmatrix}^T \]

with n=6 ns which describes the position and the orientation of each segment as a function of time.

The segments correspond in the generalized coordinates to:

\[ q_i = \begin{bmatrix} p \mid \varepsilon \end{bmatrix} \]

(where \( p \) \((i, \ldots, 3)\) is the position of a rigid segment defined by 3 Cartesian coordinates \( x, y, z \), and \( \varepsilon \) the orientation of the rigid segment defined by 3 Euler angles that correspond to the 3-1-3 sequence rotation: \( X=1, Y=2, Z=3 \).

The motion is represented as a time dependent constraint equation. Because of the non-linear nature of the constraint equations, a Newton-Raphson iterative method was used.

To model muscles, LifeMOD uses forces that replicate the desired body motion, while staying within each muscle physiological limits. For the calculation of the muscle forces, the physiological cross sectional area (pCSA) was used. It was defined as the volume of the muscle divided by its muscle length or its fibre length with or without accounting for pennation. The formula used is that presented by Alexander:

\[ pCSA = \begin{cases} \frac{m}{\rho l'} , & \text{for fusiform muscles} \\ \frac{m}{2 \rho t} \sin(2\alpha) , & \text{for unipennate muscles} \end{cases} \]

\( m, \rho \) and \( l \) are respectively muscle mass, density (1.05 gm/cm\(^2\) is a typical value) and length. \( t \) is the layer thickness of muscle pennation and \( \alpha \) is the pennation angle.

The muscle geometry data in LifeMOD were derived from [17]. The muscle geometry data (pCSA) were used by different authors. They used planimetry to measure areas and the sum of the products of area and slice thickness for estimating volumes.

The upper limit of the muscle force (\( F_{\text{max}} \)) is generated by multiplying pCSA of each muscle by a maximum tissue stress (\( M_{\text{stress}} \)) value derived from [18].
The formulation of the active muscle force ($F_1$) used in LifeMOD is:

$$F_1 = \begin{cases} 
F_{\text{max}} & : \text{if } F_1 \geq F_{\text{max}} \\
0 & : \text{if } F_1 < F_{\text{max}} \\
P_{\text{gain}} (L_{\text{desired}} - L_{\text{actual}}) + D_{\text{gain}} (\dot{L}_{\text{desired}} - \dot{L}_{\text{actual}}) & : \text{if } L_{\text{desired}} \geq L_{\text{actual}}
\end{cases}$$

(5)

where: $F_{\text{max}} = pCSA \times M_{\text{stress}}$ ($M_{\text{stress}}$: the maximum tissue stress)

$P_{\text{gain}}$ is the corrector to minimize the error

$D_{\text{gain}}$ is the multiplier of the derivative of errors

$L_{\text{desired}}$ is the muscle shortening / lengthening pattern recorded in the first step. Then this pattern is used as actuators.

$L_{\text{actual}}$ is the instantaneous shortening / lengthening pattern of the muscle in the final step.

$L_{\text{desired}}$ is the derivative of $L_{\text{desired}}$

$L_{\text{actual}}$ is the derivative of $L_{\text{actual}}$

$F_{\text{filter}}$ is a specified filter we can apply. A coherent value is 100%.

In our study, we focused on 16 muscles in each lower limb. These were: Gluteus Maximus 1 and 2, Gluteus Medius 1 and 2, Adductor Magnus, Semitendinosus, Vastus Medialis, Vastus Lateralis, Biceps Femoris 1 and 2, Rectus Femoris, Iliacus, Gastrocnemius 1 and 2, Soleus and Tibialis Anterior.

Contacts between each foot and the starting platform were characterized by two three-dimensional elements: six ellipsoids for one foot and a rectangular box for the floor. (Fig. 2)

![Fig. 2. Position of ellipsoids on the foot for the ground contact](image)

The contact model for one ellipsoid contains a normal contact force ($F_n$) and a horizontal force ($F_f$). (Fig. 3)

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The normal (vertical) contact force on the foot $F_n$ uses the non-linear spring and damper function:

$$F_n = \begin{cases} 
0 & y \geq y_1 \\
\max(0, k(y_1 - y)^e - \text{step}(y, y_1 - d, c_{\text{max}}, y_1, 0)y) & y < y_1 
\end{cases}$$

where $y_1 - y$ is the depth of penetration, $\dot{y}$ is the penetration rate. In the first term, $k$ specifies the stiffness of the surface and $e$, the exponent of the force deformation characteristic (for a stiffening spring characteristic, $e > 1.0$ and for a softening spring characteristic, $0 < e < 1.0$). The second term is the damping coefficient. This function varies linearly between 0 and $c_{\text{max}}$ (maximum damping coefficient). The penetration varies between 0 and $d_{\text{max}}$ (boundary penetration). The damping coefficient is 0 for a penetration = 0 and $c_{\text{max}}$ for a penetration $= d_{\text{max}}$.

The horizontal force ($F_f$) is modelled using the Coulomb's friction force:

$$F_f = \mu N$$

where $\mu$ is the coefficient of friction, $N$ is the normal force exerted between the surfaces, and $F_f$ is the force exerted by friction, in a direction opposing the relative motion. In this study, we took the assumption that the dynamic friction was negligible because the feet and the platform were not slipping relative to each other.

The contact properties consist of:
- stiffness (the higher the value, the more rigid the bodies in contact are),
- damping (a good rule of thumb is that this value is about 1 % of the stiffness),
- penetration depth (0.01 mm is a correct value because if $y_1 - y = 0$, no penetration occurs and the force is zero).

Table 1 summarizes the contact model parameters used.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>Value used in this study</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stiffness</td>
<td>$k$</td>
<td>$2.0^e$ N/m</td>
</tr>
<tr>
<td>Force exponent</td>
<td>$e$</td>
<td>2</td>
</tr>
<tr>
<td>Max Damping Penetration</td>
<td>$d_{\text{max}}$</td>
<td>0.01 mm</td>
</tr>
<tr>
<td>Friction coefficient</td>
<td>$\mu$</td>
<td>0.9</td>
</tr>
</tbody>
</table>
Using the reconstructed 3-D position of the markers, the angles between the subject's segment longitudinal axes and the horizontal plane were estimated (Fig. 4).

![Image](a), (b), (c)

Fig. 4 (a). Time = 0.304 s: The swimmer's body model in the initial moment of the impulse phase. (b). Time = 0.680 s: in the maximal efforts of the impulse phase. (c) Time = 0.784 s in the final grab start phase.

The subject's inertial parameters were estimated using the measured values for the height and the body mass and the anthropometric tables from [19]. The joint muscular moments at ankle, knee, and hip, in addition to the resultant force and couple exchanged between feet and platform were estimated using the Newton-Euler equations (inverse dynamics) as defined in [20].

A particular attention was given to the numerical differentiation process and to the required filtering of the raw position data. We used a method of the residual analysis of the difference between filtered and unfiltered signals [20]. The residual at any cutoff frequency choice is calculated as follows for a signal of N sample in time:

\[
R(f_c) = \sqrt{\frac{1}{N} \sum_{i=1}^{N} (X_i - \hat{X}_i)^2}
\]

Where \(X_i\) = raw data at ith sample

\(\hat{X}_i\) = filtered data at the ith sample

The intercept \(a\) (Fig. 5) on the ordinate (at 0 Hz) is the rms value of the noise (\(\hat{X}_i\) for a 0-Hz filter is the mean of the noise over the N samples).

![Image](Residual)

Fig. 5. Residual between a filtered and an unfiltered signal as a function of the filter cutoff frequency

To have a compromise between the signal distortion and the noise passed through filter, we decided to project a line horizontally from \(a\) to intersect the residual line at \(b\). The line vertically from \(a\) is the cutoff frequency chosen (\(f_c^1\)).

So, fourth order zero-phase lag Butterworth filter was used with a cut-off frequency set at 6 Hz [20, 21, 22].
Using the above-described optimization approach, the muscular forces time histories were estimated.

3. Results

The grab start movement may be divided into two phases: (a) first, the lower limbs of the swimmer flex and the line of gravity moves forward; (b) second, the extension of the lower limbs occurs and a force impulse is generated that is followed by the take off. (Fig. 6)

Fig. 6. Femur-Tibia and Tibia-Foot inter segmental angle during the grab start.

In the first phase (unbalance), there was no hip flexion but only a knee flexion and an ankle dorsal flexion. The Hamstrings (Biceps Femoris 1 and 2, Semitendinosus), which generated a hip extension moment and a knee flexion moment, acted the knee flexion in our analysis, but had very little effect on the hip unbalance-phase (Fig. 7).
Fig. 7. a) Muscular forces of the lower limbs during the grab start. b) Contributions of the different muscles of the lower limbs in percent of the capability of the hamstrings during this movement. In the negative values, there are the muscles which generate the unbalance and in the positive values, the muscles which generate the impulse.

The Tibialis Anterior generated the ankle dorsal flexion. The capabilities of the Hamstrings and Tibialis Anterior muscles to induce angular movement of the knee and ankle dorsal flexion during this phase were higher than the other muscles. Indeed, in this phase, the Adductor Magnus and the Gluteus Medius contributed substantially to the hip adduction and abduction. There is equilibrium between the capabilities of the two muscles. The Gluteus Maximus (1 and 2) participated to the hip extension. There was a weak contribution of this muscle to the maintenance of the pelvis attitude during this first phase.

In the second phase (impulse), there was an extension of the lower limbs: hip and knee extension and ankle dorsal flexion. The Quadriceps (Rectus Femoris, Vastus Lateralis, Vastus Medialis and Vastus Intermedius), which generated a hip flexion and a knee extension, had an impact on the knee extension in our results. The Soleus participated to the ankle plantar flexion. The capability of these two muscles (Quadriceps and Soleus) in this phase is quasi-identical because of a total extension of the knee and the ankle to create impulse. The Gastrocnemius (1 and 2) contributed to ankle plantar flexion and knee flexion. In our model, the function of the gastrocnemius is to help the Soleus in the ankle plantar flexion.
In order to assess the accuracy with which kinetic quantities were estimated using movement data, the measured ground reaction X component was compared with its estimated counterpart (Fig. 8).

![Graph showing force vs time](image)

Fig. 8. Comparison between the measured reaction forces (force plate) and the predicted reaction forces (LifeMOD) for the horizontal axis (X-axis). (a) Time = 0.304 s: the initial moment of the impulse phase. (b) Time = 0.680 s: the maximal effort of the impulse phase. (c) Time = 0.784 s the final grab start phase.

Relevant resultant force impulses during the second phase were also compared (Table 2).

<table>
<thead>
<tr>
<th>Time</th>
<th>Impulse (N.s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>from 0s to 0.784s</td>
<td>283</td>
</tr>
<tr>
<td>from 0.304s to 0.680s</td>
<td>197</td>
</tr>
<tr>
<td>from 0.680s to 0.784s</td>
<td>61</td>
</tr>
</tbody>
</table>

### Table 2.

Impulse Values in accordance with the starting phase

4. Discussion

This study, using all the studied kinematic variables of a swimmer during the impulse phase of the grab start, allowed a classification of muscle recruitment and coordination. The estimation of the kinetic variables was validated by comparing those that could also be measured using a force plate with the relevant measures.

We compared only the measured and predicted the horizontal component of the reaction force (X axis) because the grab start (impulse movement) is mainly executing in this component. In fact, the maximal value of the vertical component (= 400 N) is less than half of the maximal horizontal component.

Furthermore, we have determined the force impulse (both using experimental and estimated data). For the time interval 0.304s to 0.680s (corresponding to the phase of unbalance forward), we have obtained an important
discrepancy (30.26%) between the impulses. This corresponds to the fact that the LifeMOD model does not simulate the feet displacements which counterbalance the body unbalance, contrary to the real swimmer who constantly controls his balance using feet movements.

For the time interval 0.680s to 0.784s (corresponding to the “push phase”), the gap between the impulses of the two functions was quite reasonable (6.8%). It means that for this most important grab start phase, the predicted time-varying force is in good agreement with the experimental data.

The predicted maximal force amplitude (about 1050 N) was found to be higher than the experimental one (about 950 N). This is due to the frictional contact parameters in LifeMOD corresponding to a dry skin and a body stabilization.

The time delay between the predicted results and the measured ones can be explained by the fact that the swimmer takes support mainly on the edge of the force plate and has an unbalance phase due to the equilibrium in the horizontal axis during push. Moreover, the foot in LifeMOD 2005 is composed with only one segment and not with all real foot joints. Furthermore, noise is mainly introduced by the digitizing process. The high speed cameras have a resolution of 512 x 480 pixels. If we calculate the accuracy of the cameras during this experimentation, we have 8 \(10^{-9}\) m/pixel. According to [23], measurement error on 3D marker coordinates propagates unpredictably to the estimation of body segment kinematics.

Concerning the muscular contributions, during the first phase of the grab start (flexion of the lower limbs) the activity of the Hamstrings and Tibialis Anterior muscles was higher than the Adductor Magnus and the one of Gluteus Medius higher than the Gluteus Maximus.

In the second phase (extension of the lower limbs), the Quadriceps and the Soleus have quasi-similar values. The Gastrocnemius (lower contribution) helps the Soleus in the ankle plantar flexion.

Even if, the grab start presents a counter movement before the propulsion phase, our results are in good agreement with other relevant studies dealing with jump. [24], found for the propulsion phase of maximum height jumping from the squat that the uniaxial extensors developed most of the propulsive energy. Moreover, he found that hamstrings were excited first and gastrocnemius last. [25], described for the propulsion phase that the acceleration of the body in vertical direction is produced primarily by the gastrocnemius and the soleus As tibialis anterior counteracts the plantar motion of the foot and does not seem to be activated during the propulsion phase.

The originality of our study was to estimate the lower limb muscle activity of a swimmer during the initial phase of the grab start while including only kinematics data in our analysis.

The proposed approach has of course some limits. We have not taken into account the EMG in this study for a further data validation. This is due to the swimming pool experimental toughness. In addition, we have only studied one subject with the aim of validating our model. It is clear that to identify the swimmers’ performance parameters, we should do our study with more that one subject and that’s why the next stage of our research will be to enlarge sufficiently the number of subjects.

References

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