# Analysis Method of Breaststroke Movements Based on Biomechanical Multibody Dynamics

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## Abstract

Biomechanical analysis for competitive swimming can facilitate the evaluation of technical movements and the improvement of athletic performance; however, no mature open-source biodynamic analysis method for swimming techniques is currently available. This study aimed to develop a biomechanical analysis scheme for breaststroke that integrates inverse kinematics, fluid force, and inverse dynamics analyses, enabling quantitative characterization of joint moments and muscle force-generation patterns that are difficult to measure in vivo during swimming. Underwater motion data were collected from one collegiate athlete (male, BMI 24.0 kg/m2) using infrared motion capture technology. Body-segment simplification and fluid load calculation were performed based on Swumsuit software to obtain the fluid forces acting on each body segment during swimming. These fluid forces were then input into the OpenSim full-body musculoskeletal dynamic model, and static optimization was used to analyze joint moments and muscle group activation characteristics during breaststroke. The analyzed fluid forces, joint moments, and muscle activations were consistent with those reported for similar movements in the literature. The proposed simulation analysis method enables quantitative analysis of fluid resistance and muscle force-generation patterns during breaststroke, providing a biodynamic analysis scheme for improving training modes and enhancing competitive performance in elite swimmers.

## Full Text

# Analysis Method of Breaststroke Movements Based on Biomechanical Multibody Dynamics

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**Abstract:** Biomechanical analysis oriented toward competitive swimming helps evaluate technical movements and improve competitive performance. However, there is as yet no mature open-source biomechanical analysis method for swimming techniques. This study aims to construct a biomechanical analysis scheme for breaststroke that combines inverse kinematics, fluid-force analysis, and inverse dynamics, so as to quantitatively describe joint torques and muscle force-generation patterns that are difficult to measure in vivo during swimming. Infrared motion capture technology was used to collect underwater movements of one college-student athlete (male, BMI 24.0 kg/m^2). Based on Swumsuit software, body-segment simplification and fluid-load calculation were performed to obtain the fluid forces acting on each body segment during the athlete' s swimming process. These fluid forces were input into the OpenSim whole-body musculoskeletal-system dynamic model. Based on static optimization, joint torques and muscle force-generation characteristics during breaststroke were analyzed. The obtained fluid forces, joint torques, and muscle activations were consistent with similar movements reported in the literature. The proposed simulation analysis method realizes quantitative analysis of fluid resistance and muscle force-generation patterns during breaststroke, providing a biomechanical analysis scheme for improving training patterns and competitive performance in high-level swimmers.

**Keywords:** multibody dynamics; musculoskeletal system; inverse kinematics; fluid force; OpenSim; muscle activation

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# Analysis method of breaststroke movements based on biomechanical multibody dynamics

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**Abstract:** Biomechanical analyses of competitive swimming are beneficial for estimating technical motions and improving the competitive level. However, so far, there have been no mature and open-source biodynamics methods of analyzing swimming technical motions. This study aims to establish a biomechanics method, combining inverse kinematics, fluid force and inverse kinetics analysis, for analyzing breaststroke sports biomechanics. By this means, the joint torques and muscle force generation modes that are difficult to measure during swimming are quantitatively described. Infrared motion capture was used to collect underwater kinematics of a college student athlete (male, BMI = 24.0 kg/m^2). Fluid forces on the athlete' s

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body segments are calculated using Swumsuit software, which performs segmental simplification and fluid load calculation. The obtained fluid force is then used as a force input in a full-body musculoskeletal model in OpenSim, and the characteristics of joint torques and muscle activation patterns during breaststroke are obtained based on static optimization. The obtained fluid forces, joint torques, and muscle activation patterns are similar to those reported in the literature. The proposed simulation method realizes quantitative analyses of muscle activation and fluid drag forces during breaststroke, providing a biomechanical analysis scheme for improving the training patterns and enhancing the competitive performance of elite swimmers.

**Key words:** multibody dynamics; musculoskeletal system; inverse dynamics; fluid force; OpenSim; muscle activation

Human movement involves interactions among the musculoskeletal system and the external environment; that is, a series of movements is completed under the dynamic equilibrium of internal and external forces. The human body can be simplified as a rigid-flexible coupled multibody system composed of skeletal rigid bodies, joints, and compliant muscles. Information such as joint forces[^1] or deeper-level muscle force-generating patterns[^2] can be used to analyze human movement, thereby providing quantitative analytical methods for competitive-sport needs such as optimizing movement techniques and avoiding sports injuries.

Through experimental measurement methods, quantitative analysis of the swimming process can be achieved. In competitive swimming, cameras are commonly used to collect images for comparative technical analysis[^3], but this method is readily affected by factors such as droplets on the water surface and underwater bubbles. Sensors such as inertial measurement units[^4] and pressure sensors[^5] can collect complete swimming postures, and mechanical mechanisms can also be further designed to measure swimming resistance in real time[^6]. To avoid interference caused by wearing equipment during swimming, researchers have used particle-image velocimetry to visualize the flow field[^7], further enabling refined quantitative analysis of fluid dynamics in swimming. In addition, multimodal biomechanical measurement methods—such as video acquisition, electromyography (EMG), and force platforms[^8]—can further enable quantitative evaluation of the kinematics, neuromuscular control, and dynamics of the swimming process. However, the above experimental methods in sports biomechanics have shortcomings such as high measurement cost, long calibration time, and movement obstruction; relying on them alone makes it difficult to obtain information such as muscle force-generating patterns and joint torques during swimming.

At present, numerical simulation methods for competitive-swimming biomechanics mainly include computational fluid dynamics simulations[^9] and musculoskeletal-system multibody dynamics analysis[^8]. Among them, musculoskeletal-system multibody dynamics model simulations can obtain biomechanical information such as joint and muscle forces based on kinetic measurements. In recent years, they have been widely applied in the field of human biomechanics research, including analysis of movement characteristics in specific sports actions[^10] and interpretation of force-generating patterns in movement postures[^11]. On the basis of surface electromyography acquisition, joint-torque calculation based on skeletal models helps clarify the load-transmission mechanisms of different swimming movements[^12]. By establishing musculoskeletal dynamics models, one can further understand the driving mechanisms of arm stroking in water on the human body[^13]. However, inverse dynamics analysis requires obtaining the external-force conditions acting on the musculoskeletal system, and it remains difficult at present to achieve rapid and accurate measurement of the distribution of fluid resistance on the human body during swimming.

To obtain the forces exerted by the fluid on the human body during swimming, researchers have conducted numerical simulations of the interaction between the flow field and the human body based on methods such as moving meshes and smoothed particle assumptions[^14–15]. However, the harsh underwater environment and the large-scale motion of the human musculoskeletal system, which causes problems such as distortion of the fluid mesh, make it difficult to apply fluid–structure coupling numerical simulation to swimming-motion analysis[^16]. In addition, excessive computation time likewise makes this technology difficult to meet the need for rapid analysis and feedback in competitive-swimming training scenarios[^17]. The above problems can be addressed through systematic multibody dynamics analysis and the development of new fluid-force calculation models: Shinohara[^18] performed dynamic analysis of a mechanical hand based on fluid resistance under a steady flow field; Nakashima et al.[^19] developed Swumsuit (swimming human model with synthetic user interface tools), a numerical simulation tool for the swimming process, in which the human body segments are regarded as elliptical cones to achieve rapid solution of fluid forces. On this basis, Quental et al.[^20–21] proposed an inverse dynamics analysis framework using MATLAB and Swumsuit, and analyzed collected freestyle swimming data; Bao et al.[^22] further investigated swimming behavior in open water based on Swumsuit. However, the above studies did not consider the muscle force-generating patterns during swimming and lacked guidance for swimmers' muscle control. With the aid of fluid-dynamics calculation in Swumsuit, Yamakawa et al.[^23] combined surface electromyography and joint torques to solve the dynamics of the leg-kicking motion, but that study obtained muscle force-generating patterns only through electromyographic signals and could not consider the solution of deep-muscle activation modes.

Figure 1: Athlete marker attachment protocol and full-body musculoskeletal model

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To realize biomechanical analysis of competitive swimming, this study uses the open-source software OpenSim[^24] as a biomechanical analysis tool. This software is widely applied in biomechanical analysis of the human musculoskeletal system and can obtain biomechanical quantities such as muscle force-generating patterns and joint torques during human movement; however, it cannot directly calculate the fluid forces during swimming. Therefore, this study combines OpenSim with Swumsuit fluid-dynamics analysis, and uses a frogstroke motion collected by an underwater infrared motion-capture system···

swimming kinematics information as input, quantitative analysis and calculation of biomechanical information such as fluid interaction forces, joint moments, and muscle activation during the breaststroke movement process were achieved, thereby providing reference data for the quantitative analysis of fluid resistance and muscle force patterns.

# 1 Experiment

## 1.1 Participant information

In this experiment, one outstanding male collegiate swimmer was recruited at Beijing Sport University (height 185 cm, body mass 82 kg, sport level: National Master Athlete in breaststroke).

## 1.2 Experimental and data-processing protocol

First, reflective markers were attached at the side of the swimming pool. The attachment protocol is shown in Fig. 1, and the corresponding anatomical locations of the markers are detailed in Appendix A1. After the experiment began, the participant first stood naturally in the water with both hands raised laterally and both feet in a natural stance; the joint positions of the swimmer were spatially calibrated by means of the infrared markers. Subsequently, the participant started from 30 m away from the starting platform and advanced toward the starting platform using the breaststroke technique. Kinematic data were synchronously collected using 10 underwater infrared high-speed cameras (Qualisys U700+, Sweden, 100 Hz) and 14 above-water infrared high-speed cameras (Qualisys A12, Sweden, 100 Hz).

(a) Marker attachment positions (front view) (b) Marker attachment positions (rear view)

(b) Musculoskeletal model (front view) (d) Musculoskeletal model (rear view)

Fig. 1 Athlete marker attachment protocol and full-body musculoskeletal model

Using Qualisys Track Manager, the reflective markers on the body surface were identified, and gaps in the data were filled using a rigid-body assumption. The time range analyzed in this study included one complete cycle of the breaststroke movement, and the data from this phase were exported in Trc format for subsequent biomechanical analysis.

# 2 Modeling and solution

## 2.1 Musculoskeletal system modeling and inverse-kinematics solution

This study performed biomechanical analysis based on the open-source full-body musculoskeletal model Gait2392[25] in OpenSim. Using the main anatomical joint landmarks of the human body (upper-limb joints such as the shoulder, elbow, and wrist; lower-limb joints such as the anterior superior iliac spine, knee, and ankle) and the remaining auxiliary landmarks (the anterior aspects of the thigh and shank) as references, the scaling ratios of the skeletal rigid bodies were determined according to the positions and distances between infrared markers and the athlete's anthropometric data. A personalized full-body musculoskeletal model for the swimmer was then reconstructed, as shown in Fig. 1. In the musculoskeletal model, the human body consists of 15 body segments, including the pelvis, trunk, head, bilateral upper arms, forearms, hands, thighs, shanks, and feet. Fourteen joints constrain the motion between 20 rigid bodies: the neck, lumbar spine, shoulder, hip, and ankle joints use ball-and-socket models; the elbow and wrist joints use universal-joint models; and the knee joint uses a rotational–translational model. There are 37 degrees of freedom in total, including the three-dimensional coordinates and posture of the pelvis center of mass. When the joint angles are 0°, the musculoskeletal model is in a natural standing posture, with both arms hanging naturally, and the palms and soles facing anteriorly.

The muscles in the musculoskeletal model were established based on the Hill model and include three parts: contractile elements, parallel elastic elements, and series elastic elements. By defining nodes, muscle paths are specified and attached to the skeletal model, thereby enabling muscles to wrap around joints. The muscle characteristic curves (active force–length curve, passive force–length curve, force–velocity curve, and tendon force–length curve) are related to the muscle' s maximum voluntary contraction (MVC), and the MVC is adjusted according to the results of scaling in the T-pose[26], thereby achieving personalized modeling of the participant' s musculoskeletal model.

According to the three-dimensional coordinates of the markers acquired in the above collection environment for one breaststroke cycle, the trajectories of the infrared markers were solved based on the musculoskeletal dynamics model. A weighted least-squares method was used to drive the weighted sum of squared terms between the experimental markers and the corresponding anatomical markers on the model to a minimum, namely

$$\min_q J(q) = \left[ \sum_i w_i \left\| x_i^{\exp} - x_i(q) \right\|^2 \right] \qquad (1)$$

where: $q$ is the vector of generalized coordinates to be solved; $w_i$ is the marker weight in inverse kinematics; $x_i^{\exp}$ is the marker position during the experimental acquisition process; and $x_i(q)$ is the corresponding marker position in the musculoskeletal model (determined by the generalized joint coordinates). Higher weights are assigned to markers corresponding to the main anatomical locations of the human body on the musculoskeletal model (upper-limb joints such as the shoulder, elbow, and wrist; lower-limb joints such as the knee and ankle; and the highest point of the pelvis), to ensure the musculoskeletal

The model fits the actual frog-kick motion as closely as possible. The weight settings are detailed in Appendix A2.

## 2.2 Solution of Unsteady Fluid Forces

To consider the influence of the fluid on human joint motion, as well as the influence of changes in the flow field on human joint loads and muscle force-generation patterns, the open-source software Swumsuit (version 8.0.4), developed by Nakashima et al.

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of Tokyo Institute of Technology, Japan, was used. The SWUM (swimming human model) method was adopted to analyze the unsteady fluid forces acting on the human body during swimming.

In the SWUM method, the human body is regarded as being formed by the interconnection of 21 body segments: the left and right upper arms, forearms, hands, thighs, shanks, feet, and the head, neck, shoulders, upper chest, lower chest, upper waist, lower waist, upper hips, and lower hips. The relative motion between adjacent body segments is described by the relative joint motion angles. As shown in Fig. 2(a), a single body segment is simplified as an elliptical frustum. The two transverse and longitudinal lengths at the top and root of the measured subject' s body segment are taken as the major and minor axes of the elliptical frustum, thereby realizing individualized modeling of the subject. The body-segment dimensions are detailed in Appendix A3.

The formula for calculating the mass of a single body segment and the final human body mass is

$$m_i = \rho_i^* \rho_w V_i,$$

$$m = \sum_i m_i \qquad (2)$$

where: $m_i$ is the mass of the $i$-th body segment; $\rho_i^*$, $\rho_w$, and $V_i$ are respectively the density of the body segment relative to water, the density of water, and the volume of the body segment. The density of water and the density of the body segment relative to water are obtained through experimental measurement, and the body-segment volume is obtained by calculating the volume of the elliptical frustum. By calculating the mass of each body segment, the final human body mass $m$ can be obtained.

As shown in Fig. 2(b), this is a simplified schematic diagram of the human body. Taking the human trunk as an example, assume that the trunk is a rigid body and that point $C$ is the centroid of the body segment. According to Newton' s second law, the human-body dynamic equation is given by

$$m_{\text{trunk}} \ddot{x}_C = F_a + F_t + F_n \qquad (3)$$

where: $m_{\text{trunk}}$ denotes the mass of this body segment; $F_a$, $F_t$, and $F_n$ are the three components of the fluid force acting on this body segment; and $x_C$ is the centroid coordinate of this rigid body. Similarly, taking the trunk as an example, the Euler equations of this rigid body in the principal-axis coordinate system of rotational inertia are given as

$$\begin{cases} I_{x'}\dot{\omega}_{x'} - (I_{y'} - I_{z'})\omega_{y'}\omega_{z'} + \dot{I}_{x'}\omega_{x'} = M_{Cx'} \\ I_{y'}\dot{\omega}_{y'} - (I_{z'} - I_{x'})\omega_{z'}\omega_{x'} - \dot{I}_{y'}\omega_{y'} = M_{Cy'} \\ I_{z'}\dot{\omega}_{z'} - (I_{x'} - I_{y'})\omega_{x'}\omega_{y'} - \dot{I}_{z'}\omega_{z'} = M_{Cz'} \end{cases} \qquad (4)$$

where: $I_{x'}$ is the eigenvalue of the rotational inertia matrix along the first principal direction; $\omega_{x'}$ and $M_{Cx'}$ represent the components of angular velocity and moment along the first principal direction of the rotational inertia matrix. By analogy, the Euler equations in the three principal directions can be obtained.

(a) Human body-segment modeling in swimming posture

(b) Force analysis of the whole human body

**Fig. 2 Simplification of human body segments in Swumsuit**

The above body segments are taken out separately, and their forces are analyzed. By D' Alembert' s principle, force-equilibrium equations including inertial force $F_a$, buoyancy $F_b$, and normal drag force $F_t$ and axial drag force $F_n$ are constructed, thereby realizing the modeling and solution of the fluid forces acting on the human body. The infinitesimal force analysis is shown in Fig. 3.

(a) Analysis of inertial force and normal and axial drag forces

(b) Buoyancy analysis

**Fig. 3 Force analysis of infinitesimal in Swumsuit**

The elliptical-frustum body segment is cut into elliptical-frustum infinitesimal elements of length d$x$, which may approximately be regarded as planar ellipses. Let the directions of the major and minor axes of the elliptical infinitesimal element be $e_1$ and $e_2$, respectively; the thickness is d$x$, and the lengths of the major and minor semi-axes are set as $r_1$ and $r_2$, respectively. The inertial force acting on the infinitesimal element is related to the in-plane acceleration. Let it be $a_n$; then the inertial force can be assumed as

$$F_a = -C_a E_f \mathrm{d}x\pi\left[(a_n \cdot e_1)e_1 + (a_n \cdot e_2)e_2\right] \qquad (5)$$

where: $C_a$ is the fluid-force parameter corrected for three-dimensional flow; for two-

of the ideal fluid for micro-flows; this parameter is assumed to be 1 [19]; $E_f$ is the submergence parameter of the elliptical surface element, representing the ratio of the area below the water surface to the area of the body segment, i.e., the proportion occupied by the submerged part. The normal component $F_n$ of the fluid drag force is related to the normal velocity of the elliptical surface element. Let this velocity be $v_n$, whose direction is perpendicular to the long axis of the body segment. This force is set as

$$F_n = -C_n E_f dx\left[r_2\left(\frac{r_2}{r_1}\right)^p \|v_n\|(v_n \cdot e_1)e_1 + r_1\left(\frac{r_1}{r_2}\right)^p \|v_n\|(v_n \cdot e_2)e_2\right] \qquad (6)$$

where $C_n$ is the normal drag coefficient. In this equation, the velocity components are decomposed along the long-axis and short-axis directions; the long-axis and short-axis components of $F_n$ are proportional to the squares of the velocity components in the corresponding directions; $p$ represents the influence of the roundness of the body-segment ellipse on the fluid force.

The tangential component $F_t$ of the fluid force is related to the velocity $v_t$ along the long-axis direction, and this force is set as

$$F_t = -\frac{1}{2}C_t E_{fc} dx\|v_t\|v_t \qquad (7)$$

where $C_t$ is the tangential drag coefficient, and $c$ is the perimeter of the elliptical surface element. Buoyancy characterizes the compressive action of the fluid on the portion of the body segment submerged in water. To obtain the buoyancy acting on the body segment, the lateral surface of the elliptical frustum body

Fig. 4 Demonstration of simplification of fluid force (Taking the shank as an example)

Figure 2: Fig. 4 Demonstration of simplification of fluid force (Taking the shank as an example)

segment is divided into square micro-elements, and the compressive force on each micro-element is integrated over the surface of the elliptical frustum. The buoyancy on a surface micro-element of the elliptical frustum is

$$F_b = -z_q dsge_n \qquad (8)$$

where $z_q$ is the immersion parameter, used to determine whether the micro-element is below the water surface. When this parameter is less than 0, it indicates that the micro-element is below the water surface; $z_q = 0$ indicates that the micro-element is at the water-surface position; $g$ is the gravitational acceleration; $ds$ is the side length of the micro-element; and $e_n$ represents the outward normal direction of the surface micro-element. Finally, by integrating all the above fluid forces computed longitudinally for each elliptical plate, the fluid force acting on each truncated elliptical cone is obtained.

$$F_n = -C_n E_f \int \left[ r_2 \left( \frac{r_2}{r_1} \right)^p \|v_n\| (v_n \cdot e_1) e_1 + r_1 \left( \frac{r_1}{r_2} \right)^p \|v_n\| (v_n \cdot e_2) e_2 \right] dx \qquad (9)$$

$$F_a = -C_a E_f \int \pi \left[ (a_n \cdot e_1) e_1 + (a_n \cdot e_2) e_2 \right] dx \qquad (10)$$

$$F_t = -\frac{1}{2} C_t E_{fc} \int \|v_t\| v_{tdx} \qquad (11)$$

$$F_b = -z_q \iint g e_n \, ds \qquad (12)$$

## 2.3 Inverse dynamics solution

The above unsteady fluid-force loads are regarded as load densities acting on each body segment of the human model. Through force-system simplification, the fluid forces are simplified to the centroid positions of each body segment (upper arm, forearm, hand, trunk, thigh, shank, and foot). The moments generated during the simplification process then act on the corresponding body segment. As shown in Fig. 4, the pose of the musculoskeletal model is determined on the basis of the joint-angle data from kinematics.

Fig. 4 Demonstration of simplification of fluid force
(Taking the shank as an example)

In the inverse dynamics module of OpenSim, the above fluid-force results are applied to the musculoskeletal model as external-force conditions. Combined with the Euler-angle joint-angle data obtained from inverse kinematics, the equations of motion for all degrees of freedom of the whole-body musculoskeletal model are solved, taking into account external-force factors such as unsteady fluid forces and gravity, as well as internal-force factors such as human joint moments, thereby obtaining the joint moments in the musculoskeletal model.

$$M(q)\ddot{q} + C(q, \dot{q}) + G(q) = \tau \qquad (13)$$

where $M(q)$ is the system mass matrix; $C(q, \dot{q})$ is the vector of Coriolis and centrifugal forces; $G(q)$ is the vector of gravity; and $\tau$ is the generalized force acting on the skeletal rigid body.

## 2.4 Static optimization

In the static optimization module of OpenSim [27], the joint-angle data from inverse kinematics and the unsteady fluid-force loads applied to each body segment are used as inputs, and the joint moments in the model are solved using the known motion described above (Fig. 5). Static optimization focuses on the spatial position, velocity, acceleration, fluid force, and other conditions at each frame of the motion process; it avoids coupling multiple motion postures in the equations of motion, thereby improving computational efficiency. The activation level of each muscle group and the muscle force are obtained by analysis.

$$\sum_{m=1}^{n} \left[a_m f(F_m^0, l_m, v_m)\right] \cdot r_{m,j} = \tau_j \qquad (14)$$

$$J = \sum_{m=1}^{n} (a_m)^p \qquad (15)$$

where Eq. (14) is used to describe the relationship between muscle force and generalized force under physiological conditions; $n$ is the number of muscles in the model; $a_m$ is the activation level of muscle $m$ at a discrete time step $(0 \le a_m \le 1)$; $F_m^0$ is its maximum isometric force; $l_m$ is its length; $v_m$ is its shortening velocity; $r_{m,j}$ is its moment arm about the $j$-th joint axis; and $\tau_j$ is the generalized force acting about the $j$-th joint axis. Equation (15) gives the two-norm objective for minimizing muscle activation levels. The computational results of static optimization are the muscle activations of individual muscles, which are used to describe the degree of muscle-fiber recruitment under neural stimulation conditions and reflect the stimulation intensity of the nerve on a single muscle. If the muscle is in

Fig. 5 Mechanical analysis framework of the musculoskeletal system in breaststroke

Figure 3: Fig. 5 Mechanical analysis framework of the musculoskeletal system in breaststroke

muscle produces maximum force and is fully activated, the muscle activation is 1; if the muscle is in a naturally relaxed, resting state, the muscle activation is 0.

Fig. 5 Mechanical analysis framework of the musculoskeletal system in breaststroke

# 3 Results and Discussion

A breaststroke cycle can be divided into four parts: leg kick, end of leg kick, arm stroke, and glide, as shown in Fig. 6. Through inverse-dynamics solution, the changes in the flexion–extension angles of the left and right hip and knee joints as the breaststroke movement proceeds are obtained, as shown in Fig. 7. In the lower-limb kick, the start of leg retraction, the end of leg retraction with the knee joint flexed 90°, and the maximum leg retraction, with the knee joint flexed 90°, are selected as reference positions. The knee-joint angle reaches the maximum leg-retraction position and the kicking action begins at 25% of the cycle. Based on the kinetic information at the reference positions, the joint and muscle force-generation patterns are further analyzed. The time is nondimensionalized according to the cycle duration:

$$t^* = \frac{t}{t_0} \qquad (16)$$

where $t^*$ denotes nondimensional time; $t$ denotes actual time; and $t_0$ is the length of the cycle.

By solving the unsteady fluid forces with Swumsuit, the three-dimensional components of the fluid forces acting on different parts of the swimmer are obtained:

$$F_x = F_{\text{fluid}} \cdot e_x \qquad (17)$$

$$F_y = F_{\text{fluid}} \cdot e_y \qquad (18)$$

where the lateral force $F_y$ is along the $y$-direction and distinguishes the forces of adduction and abduction, enabling evaluation of the symmetry of the swimmer' s force production; the propulsive force $F_x$ is along the $x$-direction and quantifies the propulsive efficiency of the body segment, enabling evaluation of the effectiveness of the breaststroke action, as shown in Fig. 8.

Fig. 6 Definition of different maneuvers within a breaststroke cycle

Figure 4: Fig. 6 Definition of different maneuvers within a breaststroke cycle

Fig. 7 joint-angle curves

Figure 5: Fig. 7 joint-angle curves

(a) Leg kick (b) End of leg kick (c) Arm stroke (d) Glide

Time

Fig. 6 Definition of different maneuvers within a breaststroke cycle

(a) Knee-joint flexion–extension angle (flexion negative, extension positive)

(b) Hip-joint flexion angle (flexion positive, extension negative)

(c) Hip-joint abduction/adduction angle (adduction positive, abduction negative)

(d) Ankle-joint flexion/extension angle (flexion positive, extension negative)

**Fig. 7** Lower limb joint angle during the breaststroke maneuver

(a) Upper arm

(b) Forearm

(c) Thigh

(d) Shank

**Fig. 8** Comparison of unsteady fluid forces on arms and legs during breaststroke

Through comparison, it is found that the propulsive force acting on the upper limbs is smaller than that on the lower limbs. Among the upper limbs, the propulsive force is the vector sum of the fluid-force projections in the forward direction on the hand, forearm, and upper-arm body segments; therefore, the lower limbs are the main dynamic source for forward propulsion during swimming. However, the force on the forearm exhibits left–right asymmetry. Further calculations of the shoulder- and elbow-joint moments also show asymmetry, which is caused by asymmetric general factors such as the dominant side and developmental differences.

universally present [28]; therefore, it can be regarded as the manifestation of the athlete' s individualized characteristics. The maximum left–right differences in the propulsive forces of the thigh and shank were 4.7 and 9.3 N, respectively,

Fig. 7 joint-angle curves

Figure 6: Fig. 7 joint-angle curves

Hip joint adduction (+)

Figure 7: Hip joint adduction (+)

Ankle joint flexion (+)

Figure 8: Ankle joint flexion (+)

and the maximum left–right differences in the propulsive forces of the upper arm and forearm were 1.8 and 5.3 N, respectively. In addition, this phenomenon can also be observed by comparing the lateral force of the shank. Viewed from the anterior–posterior direction, the fluid force generated by the shank during the kicking process is the largest, followed by that of the thigh. The peak unsteady fluid forces borne by the thigh and shank were 10.9 and 31.2 N, respectively, and the peak unsteady fluid forces borne by the upper arm and forearm were 17.7 and 20.3 N, respectively.

After considering the simplified unsteady fluid forces described above, the joint moments of the hip, knee, and ankle joints of the lower limb were solved, as shown in Fig. 9. The peak extension moments of the hip and knee joints of the lower limb were 64.4 Nm and 62.4 Nm, respectively, and the peak ankle-joint flexion moment was 7.1 Nm.

(a) Knee-joint flexion–extension moment (flexion negative, extension positive)

(b) Hip-joint flexion–extension moment (flexion positive, extension negative)

(c) Hip-joint adduction–abduction moment (adduction positive, abduction negative)

(d) Ankle-joint flexion–extension moment (flexion positive, extension negative)

**Fig. 9** Lower limb joint moment during the breaststroke maneuver

Further analysis was performed on the simulation results of joint and body-segment loads: the hand propulsive force was compared with the breaststroke hand propulsive force results of Tsunokawa et al. [29], and the shoulder-joint flexion moment was compared with the research results of Schüler et al. [30]. As shown in Figs. 10–11, the peak values and the overall trends of the curves were generally consistent with existing studies.

From the activation timing of the lower-limb muscle groups obtained by static optimization, taking an activation level of 0.1 as the threshold for determining whether a muscle group is activated [31], the biceps femoris was activated first,

Propulsive force; lateral force curves for upper arm and forearm

Figure 9: Propulsive force; lateral force curves for upper arm and forearm

Propulsive force; lateral force curves for thigh and shank

Figure 10: Propulsive force; lateral force curves for thigh and shank

Fig. 9 Lower limb joint moment during the breaststroke maneuver

Figure 11: Fig. 9 Lower limb joint moment during the breaststroke maneuver

followed by the rectus femoris, then the tensor fasciae latae, and finally the gluteus maximus. The gluteus maximus here is mainly composed of three muscle bundles. As shown in Fig. 12, the differences in activation of different anatomical regions of the simulated gluteus maximus during the movement, and the activation of the left and right gluteus maximus are shown in Fig. 13. The activation time of the gluteus maximus during the leg-recovery and kicking actions (0.21 s) agreed with related studies (0.23 s) [32], preliminarily verifying that the computational results given by static optimization have a certain degree of rationality.

**Fig. 10** Compared hand fluid forces estimated in this study and Tsunokawa et al. [29]

Compared with computational methods such as muscle control and neuromusculoskeletal tracking, the computational results of static optimization are close to those of the two methods [33], but have···

faster solution speed

34

, and therefore it has advantages in guiding the direction of training. In swimming, the leg muscle groups of breaststroke swimmers are more developed than those of other athletes, and this study did not separately conduct MVC tests on the subjects. Therefore, the MVC of the gluteus maximus in the musculoskeletal model may be smaller than the actual MVC of the gluteus maximus, resulting in muscle activation greater than 1. Similar situations have also appeared in previous studies

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: the main reason is that the muscle activation time is relatively short, differing from the longer activation time in MVC tests; this can be further optimized by analyzing the difference between the two.

**Fig. 11** Compared shoulder joint flexion torque estimated in this study and

Fig. 10 Compared hand fluid forces estimated in this study and Tsunokawa et al.

Figure 12: Fig. 10 Compared hand fluid forces estimated in this study and Tsunokawa et al.

Fig. 11

Figure 13: Fig. 11

Fig. 12

Figure 14: Fig. 12

Schüler et al

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**Fig. 12** Composition of the gluteus maximus in the musculoskeletal model

The fluid-load calculation method in this study relies on Swumsuit to simplify the distribution of fluid loads. Traditional fluid-force calculation methods, such as dynamic mesh and smoothed particle methods

$14 - 15$

, have problems such as fluid mesh distortion

16

; moreover, the convergence of the smoothed particle method depends on parameters such as smoothing length and the number of neighboring particles, and the choice of these parameters has a relatively large influence on the results

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. By simplifying the human body segments and modeling the fluid force vectors, this study improves the operability and computational efficiency of the computational method in the field of human movement science, while reducing the demand for various experimental acquisition devices, especially equipment for collecting flow-field distributions. In follow-up work, markerless capture technology may be used to further reduce the influence of acquisition equipment on athletes' technical movements.

This study has important potential value in the field of swimming sports science. On the one hand, the joint torque peaks and unsteady fluid forces rapidly solved in this study effectively support the need for scientific talent selection in competitive swimming based on body morphology, and can quantitatively analyze joint flexibility and strength qualities

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. On the other hand, existing studies have revealed the mechanisms by which neuromuscular training improves swimming performance

38

Fig. 13

Figure 15: Fig. 13

. This study can quantitatively evaluate muscle force-generation patterns during swimming, providing biomechanical models and quantitative data support for optimizing the timing and amplitude of force generation in kicking and sculling, thereby improving propulsion efficiency and extending the technical parameters of competitive swimming to quantification over the whole cycle. Furthermore, the high-intensity movements of competitive swimming readily lead to joint injuries

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, and are related to injuries of soft tissues such as muscles and tendons. The method proposed in this study helps quantify information such as the activity range of each joint, joint torque, and muscle-force thresholds in swimming movements, and can effectively prevent chronic injuries in swimming

40

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**Fig. 13** Muscle activation of the gluteus maximus during breaststroke movements

This study still has certain limitations: to solve the problem of rapidly computing fluid forces, this study introduced Swumsuit as a tool for calculating external forces in the musculoskeletal model. Because it does not require offline computational fluid dynamics simulation, the computation time is far lower than that required by CFD solutions. However, according to previous research results, deviations exist between the fluid-force values solved by Swumsuit and experimental test values: in various sculling movements, the speed error is 1%-7.5%, and the force error is 10%

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. Moreover, when an athlete is affected by other athletes or by the pool wall, Swumsuit may also have computational errors. On the other hand, for the calculation results of fluid forces, the unsteady fluid forces are simplified as the centroid and corresponding torque of each body segment; the influence of this on the solution of joint torque and muscle activation requires further analysis. At the same time, because the muscles in the musculoskeletal model Gait2392 are established only in the lower limbs, muscles above the hip joint are absent from the model, causing the solution of gluteus maximus muscle activation to ignore the influence of other synergistic muscle groups; musculoskeletal dynamics analysis currently

ignored the constraint of the torso and upper-arm muscles; for the joint internal/external rotation angles, which are crucial in swimming, temporary constraints have likewise been imposed. Subsequent work can further carry out biomechanical analysis by establishing a more complete whole-body musculoskeletal model[42]. Analysis of frog-kick motion still requires recruiting more participants, so as to account for computational deviations caused by different influencing factors, such as athletes' body types and swimming speeds; surface EMG signal acquisition should also be considered to further verify the rationality of the muscle-activation analysis results. For analysis of the force-generation mode of the musculoskeletal system under fluid forces, the aim is to solve a dynamics problem of a rigid–flexible coupled system in a flow field. Both model construction and numerical solution are relatively complex; it is worth further exploring reliable modeling and efficient computational methods for coupled problems[43], thereby achieving efficient and stable numerical solutions. Finally, this study analyzed only one swimming posture—the frog kick. Future research will include more participants to analyze the cooperative mechanisms of limb movement and muscle force generation during swimming, and will discuss the effects of complex loading conditions, such as push-off from the wall[44] and flow-field variation[45], on the mechanical response of swimming.

## 4 Conclusion

This study proposed a computational framework integrating inverse kinematics, unsteady fluid forces, and inverse dynamics analysis, and obtained biomechanical information such as time-varying fluid loads, muscle forces, and muscle activation during the frog kick. During the frog kick, the participant's propulsive force was mainly generated by the leg-kicking process; the biomechanical analysis results are consistent with existing research on swimming biomechanics. This study preliminarily achieved quantitative analysis of the patterns of fluid resistance and muscle force generation in each body segment during the frog kick. By interpreting the movement process through in-depth biomechanical results, the efficiency of dynamic computation meets the needs of rapid analysis in competitive sports, providing a biomechanical theoretical basis and quantitative data support for improving the training modes and competitive performance of high-level swimmers.

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# Appendix A

**Table A1 Swimming test marker placement plan**

**Tab. A1 Swimming test posting plan**

| Marker name | Location | Marker name | Location |
|---|---|---|---|
| L_HEAD | Left side point of the head | L_IAS | Left anterior superior iliac spine |
| R_HEAD | Right side point of the head | L_IPS | Left posterior superior iliac spine |
| SGL | Anterior point of the head | L_IS | Highest point of the left iliac crest |
| CV7 | Spinous process of the seventh cervical vertebra | R_IAS | Right anterior superior iliac spine |
| L_SIA | Left inferior angle of the scapula | R_IPS | Right posterior superior iliac spine |
| R_SIA | Right inferior angle of the scapula | R_IS | Highest point of the right iliac crest |

| Marker name | Location | Marker name | Location |
|---|---|---|---|
| JN | Suprasternal notch | L_TH | Left thigh point |
| TV10 | Spinous process of the tenth thoracic vertebra | L_FLE | Lateral point of the left knee joint |
| L_SAE | Left shoulder peak point | L_FME | Medial point of the left knee joint |
| L_HUM1, L_HUM2, L_HUM3 | Left upper-arm tracking points | R_TH | Right thigh point |
| L_HLE | Lateral point of the left elbow joint | R_FLE | Lateral point of the right knee joint |
| L_HME | Medial point of the left elbow joint | R_FME | Medial point of the right knee joint |
| L_RSP | Lateral point of the left wrist joint | L_SK | Left shank point |
| L_USP | Medial point of the left wrist joint | L_FAL | Lateral point of the left ankle joint |
| L_HM2 | Left finger point | L_TAM | Medial point of the left ankle joint |
| R_SAE | Right shoulder peak point | R_SK | Right shank point |
| R_HUM1, R_HUM2, R_HUM3 | Right upper-arm tracking points | R_FAL | Lateral point of the right ankle joint |
| R_HLE | Lateral point of the right elbow joint | R_TAM | Medial point of the right ankle joint |
| R_HME | Medial point of the right elbow joint | L_FT | Left toe point |
| R_RSP | Lateral point of the right wrist joint | L_FCC | Left heel point |
| R_USP | Medial point of the right wrist joint | R_FT | Right toe point |

| | | | |
|---|---|---|---|
| R_HM2 | Right finger point | R_FCC | Right heel point |

Note: In this test, the calibration posture was with the palms facing upward.

**Table A2 Inverse kinematics analysis marker weight list**

**Tab. A2 Inverse kinematics analysis marker weight list**

| Marker name | Location | Weight | Marker name | Location | Weight |
|---|---|---|---|---|---|
| L_HEAD | Left side point of the head | 5.0 | L_SAE | Left shoulder peak point | 5.0 |
| R_HEAD | Right side point of the head | 5.0 | L_HUM1, L_HUM2, L_HUM3 | Left upper-arm tracking points | 1.0 |
| SGL | Anterior point of the head | 1.0 | L_HLE | Lateral point of the left elbow joint | 5.0 |
| CV7 | Spinous process of the seventh cervical vertebra | 1.0 | L_HME | Medial point of the left elbow joint | 5.0 |
| L_SIA | Left inferior angle of the scapula | 1.0 | L_RSP | Lateral point of the left wrist joint | 5.0 |
| R_SIA | Right inferior angle of the scapula | 1.0 | L_USP | Lateral point of the left wrist joint | 5.0 |
| JN | Suprasternal notch | 1.0 | L_HM2 | Left finger point | 1.0 |

| Marker name | Location | Weight | Marker name | Location | Weight |
|---|---|---|---|---|---|
| TV10 | Spinous process of the tenth thoracic vertebra | 1.0 | R_SAE | Right shoulder peak point | 5.0 |

Table A2 (continued)

| Marker name | Position | Weight | Marker name | Position | Weight |
|---|---|---|---|---|---|
| R_HUM1, R_HUM2, R_HUM3 | Right upper-arm tracking points | 1.0 | L_FME | Medial point of left knee joint | 5.0 |
| R_HLE | Lateral point of right elbow joint | 5.0 | R_TH | Right thigh point | 1.0 |
| R_HME | Medial point of right elbow joint | 5.0 | R_FLE | Lateral point of right knee joint | 5.0 |
| R_RSP | Lateral point of right wrist joint | 5.0 | R_FME | Medial point of right knee joint | 5.0 |
| R_USP | Lateral point of right wrist joint | 5.0 | L_SK | Left shank point | 1.0 |
| R_HM2 | Right finger point | 1.0 | L_FAL | Lateral point of left ankle joint | 5.0 |

| Marker name | Position | Weight | Marker name | Position | Weight |
|---|---|---|---|---|---|
| L_IAS | Left anterior superior iliac spine | 1.0 | L_TAM | Medial point of left ankle joint | 5.0 |
| L_IPS | Left posterior superior iliac spine | 1.0 | R_SK | Right shank point | 1.0 |
| L_IS | Highest point of left pelvis | 5.0 | R_FAL | Lateral point of right ankle joint | 5.0 |
| R_IAS | Right anterior superior iliac spine | 1.0 | R_TAM | Medial point of right ankle joint | 5.0 |
| R_IPS | Right posterior superior iliac spine | 1.0 | L_FT | Left toe point | 1.0 |
| R_IS | Highest point of right pelvis | 5.0 | L_FCC | Left heel point | 1.0 |
| L_TH | Left thigh point | 1.0 | R_FT | Right toe point | 1.0 |
| L_FLE | Lateral point of left knee joint | 5.0 | R_FCC | Right heel point | 1.0 |

**Table A3 Sizes of body segments**

Tab. A3 Sizes of body segments

| Body segment name | Lower-base minor axis/m | Lower-base major axis/m | Upper-base minor axis/m | Upper-base major axis/m | Height/m |
|---|---|---|---|---|---|
| Lower waist | 0.13 | 0.17 | 0.12 | 0.16 | 0.07 |
| Upper waist | 0.12 | 0.16 | 0.14 | 0.17 | 0.15 |
| Lower chest | 0.14 | 0.17 | 0.16 | 0.18 | 0.13 |
| Upper chest | 0.17 | 0.18 | 0.12 | 0.02 | 0.06 |
| Shoulder | 0.12 | 0.02 | 0.08 | 0.08 | 0.06 |
| Neck | 0.08 | 0.08 | 0.11 | 0.10 | 0.15 |
| Upper hip | 0.13 | 0.17 | 0.13 | 0.21 | 0.11 |
| Lower hip | 0.13 | 0.21 | 0.10 | 0.20 | 0.15 |
| Right thigh | 0.10 | 0.10 | 0.07 | 0.07 | 0.38 |
| Left thigh | 0.10 | 0.10 | 0.07 | 0.07 | 0.38 |
| Right shank | 0.07 | 0.07 | 0.04 | 0.04 | 0.36 |
| Left shank | 0.07 | 0.07 | 0.04 | 0.04 | 0.36 |
| Right foot | 0.08 | 0.04 | 0.02 | 0.06 | 0.26 |
| Left foot | 0.08 | 0.04 | 0.02 | 0.06 | 0.26 |
| Right upper arm | 0.05 | 0.05 | 0.04 | 0.03 | 0.35 |
| Left upper arm | 0.05 | 0.05 | 0.04 | 0.03 | 0.35 |
| Left forearm | 0.04 | 0.04 | 0.02 | 0.03 | 0.30 |
| Right hand | 0.02 | 0.06 | 0.01 | 0.03 | 0.21 |

**Table A4 Fluid-dynamics calculation parameters**

Tab. A4 Coefficients in fluid calculation

| Parameter symbol | Parameter meaning | Parameter symbol | Parameter meaning |
|---|---|---|---|
| $C_{\mathrm{a}}$ | Three-dimensional fluid added-mass correction coefficient | $p$ | Influence coefficient of elliptical flatness |
| $C_{\mathrm{n}}$ | Normal drag coefficient | $E_{\mathrm{f}}$ | Downstream parameter |
| $C_{\mathrm{t}}$ | Tangential drag coefficient | $Z_{\mathrm{q}}$ | Immersion parameter |

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*Note: Figure translations are in progress. See original paper for figures.*