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FEASIBILITY OF USING OPENSIM FOR ESTIMATING MUSCLE FORCES DURING WALKING IN THE AQUATIC ENVIRONMENT

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SUMMARY

To walk with comfortable speed in the aquatic environment, adults decrease the gait speed and increase the horizontal impulse applied on the ground when compared to comfortable walking on land. Such changes in gait biomechanics are accompanied by differences in the net joint torques at the ankle, knee and hip between the aquatic and terrestrial environment. To comprehend the origin of those differences and to better estimate the mechanical load of walking in shallow water, we want to estimate muscle forces during this task. For that, we conduct a study on the feasibility of using Static Optimization tool available in the OpenSim (OS) software. Using OS, we estimated muscle forces and activation levels in seven lower limb and trunk muscles. The simulated activation levels were in good agreement with experimental electromyography data, except for biceps femoris long head; the muscle forces values obtained were reasonable when compared to those of walking on land. We conclude that it is possible to use the Static Optimization tool of OS to estimate muscles forces at the flexors and extensors muscles of lower limb and trunk.

INTRODUCTION

To walk in shallow water is a low-impact activity often recommended for training and rehabilitation. The increased buoyancy and drag forces in comparison to the terrestrial environment require the individual to adopt new strategies to walk, such as decreasing the gait speed and increasing the horizontal impulse applied on the ground [4, 6]. Those changes in gait biomechanics are accompanied by differences between the mechanical loads associated with walking in water and on land: in water, there is a decrease in the net torques on the ankle and the knee joints, but there is not a significant decrease in the hip joint torque. We attribute these results to the difference in the amount of apparent weight each joint should support and also to the role of each joint in body propulsion [6]. However, the contributions of these two factors need to be better understood.

To comprehend the origin of the differences in the net joint torques between both environments and also to better estimate the loads associates with walking in water, we want to calculate muscle forces during this task. We report here the results of a preliminary study, which aimed at verifying the feasibility of using the Static Optimization tool of OpenSim software to estimate muscle forces in adults during walking in the aquatic environment.

METHODS

We used the Static Optimization tool available in the 2.2.1 OpenSim version to estimate muscle force and activation level in the tibialis anterior (TA), gastrocnemius medialis (GM), vastus lateralis (VL), long and short head of the biceps femoris (BFLH, BFSH), tensor fasciae latae (TFL) and erector spinae (EE), from a three-dimensional (3D) simulation of a young adult walking in water at chest level with comfortable speed. The 23-degree of freedom and 92-Hill-type-muscles model (Gait Model 2392), available with the software and created by the OS team based on [1, 5, 7], was adopted to model the lower limbs, pelvis and trunk movements and muscle force generation characteristics.

The data set used in the simulation came from a two-dimensional (2D) gait analysis of a stride (two consecutive right heel strikes), performed by a female (25 years, 160 cm and 53 kg). They consisted of the vertical (V) and anterior-posterior (A-P) components of: all markers used to locate the modeled body segments in space, ground reaction force (GRF) and center of pressure (CoP), acquired during five different strides performed by the participant; the marker positions during a static trial (upright position); and the participant's body-segment measures necessary to estimate the drag forces (DF). The experimental setup and data collection procedures are reported in more detail elsewhere [4, 6].

To simultaneously work with the 2D data of only one side of the body and the chosen 3D model, it was necessary to simplify the problem by neglecting all external M-L forces and by considering constant the M-L positions of markers and CoP. In adition, the left body side movement was accessed by assuming the existence of a left marker set symmetrically positioned to the right one. The positions of the missing left markers were estimated from the right markers by considering the gait symmetrical and cyclic (left side movement delayed by one step). The components of GRF and CoP of the left side were estimated likewise.

A set of virtual markers was allocated on the right and left sides of the OS neuromuscular model in the same anatomical positions they had been placed in the volunteer's body. The model was scaled to represent the anthropometric characteristics of the individual from static trial data and body size measurements. For each trial, the Inverse Kinematic (IK) problem was solved to obtain the generalized coordinates that characterize the movement. Point Kinematic tool was employed to obtain the segments proximal and distal joint trajectories [2] in order to calculate DF and its respective torque around the proximal joint at each segment [6].

To evaluate the scaling and IK solution, sagittal angular displacement (AD) and the net torque (JT) on the right hip, knee and ankle joints were compared to corresponding experimental data [6] calculated with Matlab 7.5 (Mathworks Inc., US). The differences between the results obtained with Matlab and OS were evaluated. The muscle force-sharing problem solution was obtained considering the constraints given by the muscle force-length-velocity surface (eq.2, $f(F^0_m, l_m, v_m)$) and the objective function given by eq. 3:

$$\tau_k = \sum_{m=1}^{N} \left(a_m f\left(F^0_m, l_m, v_m\right)\right) r_{m,k} \qquad (2)$$

$$J = \sum_{m=1}^{N} a_m^2 \qquad (3)$$

where: N=92; τ_k, is the net torque at joint k; $r_{m,k}$ the moment arm of muscle m around joint k; a_m, F^0_m, l_m, v_m are respectively, its activation level, maximum isometric force, length, and shortening velocity [2]. External forces prescribed in the problem were: GRF on booth feet, at the corresponding CoP; DF and corresponding torque at the proximal joint of each immersed segment; Buoyancy (calculated as in [6]), at the center of mass of each immersed segment. The a_m time series were normalized in time by the stride period, divided by the mean activation level along the stride and averaged across the 5 trials, in order to be compared to electromyography (EMG) data of 10 adults [4]. The mean and maximal force (mF and MxF) developed by each muscle during the stride was calculated and averaged across trials.

RESULTS AND DISCUSSION

The mean quadratic difference (MQE) between AD and JT calculated with Matlab and OS were respectively 7%, 2% and 12% of ankle, knee and hip AD range and 8%, 5% and 7% of ankle, knee and hip JT range. We consider those differences acceptable given all the assumptions necessary to simulate the 3D movement.

The a_m of all muscles analyzed, were in good agreement to EMG data [4] (see figure 1), except for BFLH, which presented a high activation level in swing phase when compared to experimental data. As BFLH is also a hip rotator, it could be due to the fact that hip rotation could not be correctly modeled. For those muscles whose a_m was in accordance with literature, mF and MxF were computed (table 1). It is possible to note that TA and VL MxF were lower in water than on land [5]. Those results are in accordance to the fact that peak extensor torques on the ankle and knee joints are reduced in water. Other values are around those observed for land and they seem reasonable.

As we are conducting pilot studies on 3D gait analysis in water, we chose a 3D model keeping in mind our future investigation. Despite the simplifications made to fit the 2D data on the 3D model may compromise the quality of force estimative, it was possible to obtain reasonable values for muscle force of the flexors and extensors muscles of lower limb and trunk.

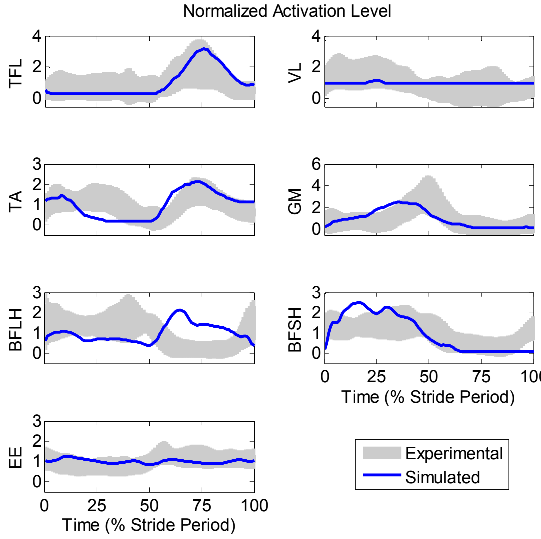

Figure 1: Comparison between the normalized activation level simulated with OS and experimental data on EMG activity (mean ± standard deviation of 10 subjects) [4].

Muscle	mF (N)	MxF (N)	Muscle	mF (N)	MxF (N)
BFSH	74±11	208±32	TA	42±7	98±34
VL	15±1	24±1	TFL	6.6±0.5	19.4±3.1
GM	78±22	252±45	EE	283±27	399±60

Table 1: Mean and maximal force (mF and MxF, respectively) developed by each muscle during a stride.

CONCLUSIONS

We conclude that it is possible to use the static Optimization tool of OpenSim to estimate the muscle forces during locomotion in aquatic environment. However, further studies with 3D kinematic data, are necessary to evaluate the quality of muscle force estimation.

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