LOWER LIMB MUSCLES MAY DEVELOP HIGHER FORCES WHILE WALKING IN SHALLOW WATER THAN ON LAND

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Abstract: *While walking in water at chest deep and comfortable speed we observe a reduction in the peak plantar flexor torque at the ankle and in the peak extensor torque at the knee compared to comfortable walking on land. However, the peak flexor torque at knee and the peak extensor torque at the hip are similar between environments, which suggests that forces generated at the muscles crossing hip and knee joints are not always lower in water. To verify this hypothesis we used a computational model of the human lower limb and trunk together with experimental data of 10 volunteers walking in both environments, and applied an inverse dynamic-based static optimization algorithm to estimate the forces in the main flexors and extensors hip, knee and ankle muscles during comfortable walking on land and in water. The results confirmed our hypothesis, and showed that the hamstrings and gluteus maximus may generate higher forces during middle support in water than on land, while rectus femoris, iliacus and psoas may generate similar forces in both environments during swing. These results support the idea that walking in water is effective for muscle strengthening.*

Keywords: *Static Optimization, OpenSim, Hydrotherapy.*

Introduction

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 [1, 2]. Such changes in gait biomechanics result in a diminished peak plantar flexor torque at the ankle and peak extensor torque at the knee, but similar peak flexor torque at knee and peak extensor torque at the hip during walking in water compared to land [2]. This may indicates that forces generated by the muscles crossing hip and knee joints are not always lower in water in comparison to land. The estimation of muscle forces can help to comprehend the origin of the differences in the net joint torques between environments and to better quantify the mechanical load of walking in shallow water.

Muscle forces can be estimated non-invasively associating computational models of the human musculoskeletal system and optimization methods [3]. Several approaches using dynamic and static optimization have been implemented successfully to estimate muscle forces during walking on land [3, 4]. However, the solution of the same problem in water requires the calculation of drag forces, increasing the complexity of the problem and making dynamic optimization approaches computationally expensive. A cheaper alternative is to apply inverse dynamics-based static optimization algorithms that account for the force-length-velocity properties of each muscle. This approach has been shown to provide realistic results as those obtained with dynamic optimization for walking on land [4]; in addition, it may facilitate the inclusion of drag forces in the problem, hence these forces can be estimated using the kinematic and subject's anthropometric data [2]. We used this approach to estimate muscle forces during comfortable walking in water and on land, to verify our hypothesis that muscle forces generated by the hip extensors and knee flexors are not always lower in water compared to land.

Material and methods

We used the Static Optimization tool available in the OpenSim 3.1 [5] to estimate force and activation level in the tibialis anterior and posterior (TA, TP), soleus (SOL), gastrocnemius medialis and lateralis (Gmed, Glat), rectus femoris (RF), vastus lateralis (VL), medialis (VM) and intermedius (VI), long and short head of the biceps femoris (BFlh, BFsh), semitendinosus (SMT) and semimembranosus (SMM), gluteus maximus (Gmax), iliacus (ILI) and psoas (PSO), from a two-dimensional (2D) simulation of adults walking in water at chest level with comfortable speed and on land at comfortable speed.

Experimental data – The data set used in the simulations came from a 2D gait analysis of a stride (two consecutive right heel strikes), performed by 10 adults (6 female, 4 male; 24±3 years; 168±7 cm; 63±8 kg). They consisted of the vertical and anterior-posterior components of: all markers used to locate the modeled body segments in space ($\{\vec{x_i}\}_{exp}$), ground reaction force ($\vec{F}_{GR}$) and center of pressure ($\vec{x}_{CoP}$), 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 and buoyancy forces ($\vec{D}$,$\vec{B}$). The experimental setup and data collection procedures are reported in more detail elsewhere [1, 2].

The musculoskeletal model – We modeled right lower limb and torso movements using a 7-degrees-of-freedom, 43-Hill-type-muscles model that included the ankle, knee, hip and low back joints, and represented 38 human muscles. The model was obtained by modifying an existing 23-degree-of-freedom and 92-Hill-type-muscles model, whose lower extremity joints were defined as in [6] and the back joint and anthropometry as in [7]. Modifications included removing left lower limb and right muscles that did not contribute for flexion or extension, constraining hip and back to move in the sagittal plane, not considering motion at the subtalar and metatarsophalangeal joints and move ankle flexion-extension axis to a normal position in relation to the sagittal plane, to account for our 2D approach.

Muscle force and activation level estimates – A set of virtual markers was allocated at the 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 each volunteer from static trial data and body size measurements. For each trial, the Inverse Kinematic problem [5] was solved to obtain the set of generalized coordinates, $\{q_k\}$, that characterize the movement, in both water and on land. Point Kinematic tool was employed to obtain the segments proximal and distal joint trajectories ($\vec{x}_{JCj}$), in order to calculate $\vec{D}$ and its respective torque around the proximal joint ($\vec{\tau}_D$), as well as the center of volume ($\vec{x}_{CV}$) in each submersed segment *j*. To solve the muscle force-sharing problem we considered the constraints given by the active muscle force-length-velocity surface (eq.1, $f(F_m^0, l_m, v_m)$ defined as in [8]) in eq. 2 and the cost function given by eq. 3:

$$F_m = a_m f(F_m^0, l_m, v_m) \qquad (1)$$

$$\tau_k = \sum_{m=1}^{43} F_m r_{mk} \qquad (2)$$

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

where 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, fiber length and contraction velocity [4,8]. External forces prescribed in the problem were: $\vec{F}_{GR}$ on the right foot, at $\vec{x}_{CoP}$;$\vec{D}$ and corresponding $\vec{\tau}_D$, at the proximal joint and $\vec{B}$ at $\vec{x}_{CV}$ of each immersed segment (figure 1) .

Hydrodynamic forces estimates – To estimate $\vec{D}_j$, $\vec{\tau}_{D\,j}$ and $\vec{B}_j$ the body segments volume and frontal area were calculated by considering them truncated cones, with dimensions related to the individual body-segment size. We assumed the predominance of pressure drag and estimated $\vec{D}_j$ and $\vec{\tau}_{D\,j}$ by applying the stripe theory as described in details in [2]. We considered $\vec{B}_j$ constant in time and we calculated them from each segment volume, adopting for water density a value of 1000 kg/m^3. We also assumed $\vec{x}_{CV\,j}$coincident with the segment center of mass.

Data analysis – The a_m and F_m time series were normalized in time by the stride period (0-100% in steps of 1%); we also normalized a_m by their mean activation during the stride, and F_m by the individual body weight (BW). These cycles were averaged across trials to obtain the mean cycle for each participant and the same process was repeated to obtain the mean cycle among participants. The simulated a_m for BFsh, BFlh, VL, TA and Gmed were compared to the electromyography (EMG) data of 10 adults [1]. The mean and maximal force during the 1st and 2nd half of support phase (ST1 and ST2) and during swing (SW) were calculated and averaged across trials. The mean difference in these variables between environments was calculated among individuals. Paired Student's t test or the Wilcoxon signed-rank test, when sample was not normally distributed, were applied to verify the effect of environment on muscle forces (α=5%)

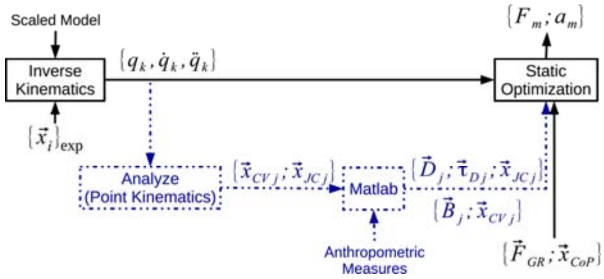

Figure 1: Muscle force estimation using OpenSim. In blue are the procedures necessary to account for drag and buoyancy forces during walking in water.

Results

The individuals analyzed here walked with the spatial-temporal characteristics and joint kinematic and kinetic described in [2], and our simulations reproduced those results within one standard deviation. In addition, the simulated activation level for the selected muscles agreed well with EMG data reported by Barela et al. (2006) [1], except for BFsh, during the transition from support to swing (figure 2).

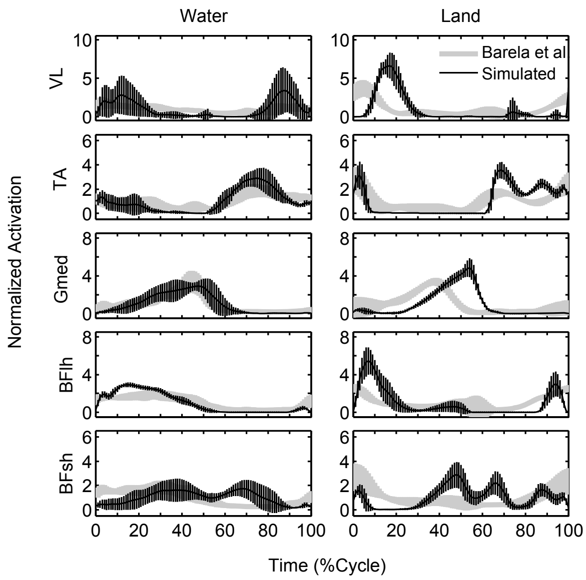

Figure 2: Simulated muscle activation level while walking in water and on land compared to the electromyography data of 10 adults reported by Barela et al. (2006) [1].

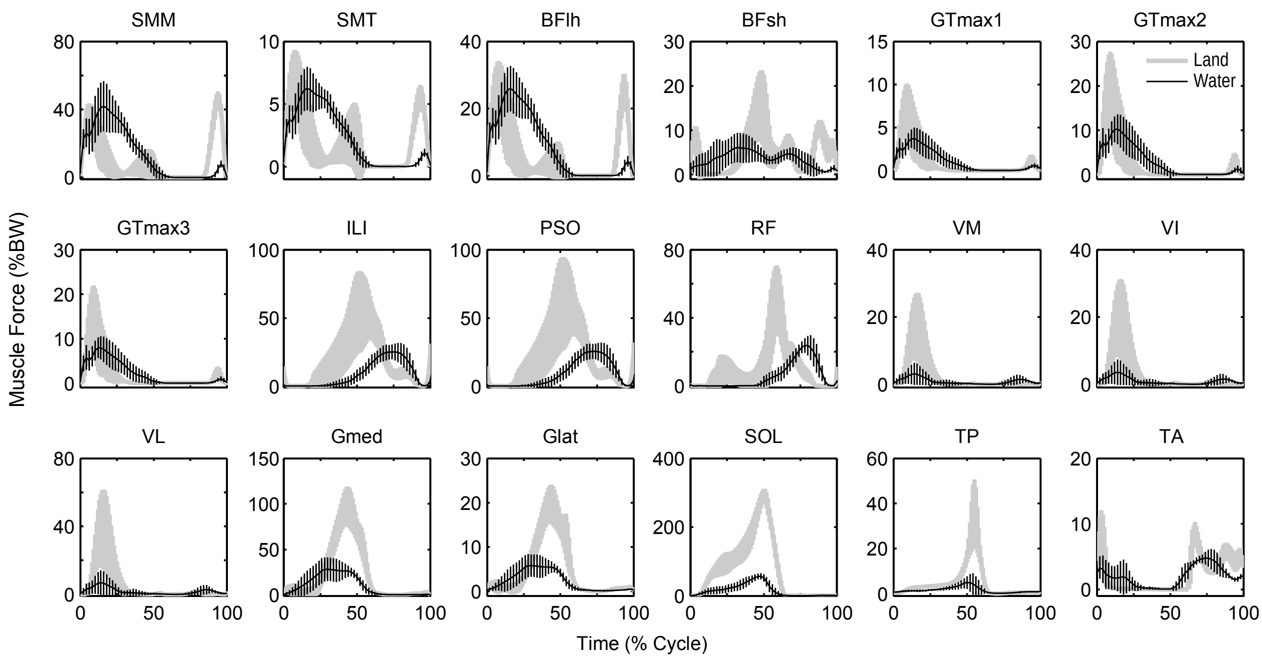

Figure 3: Mean and standard deviation across subjects for the muscle force developed in the main hip, knee and ankle flexors and extensors while walking in water and on land.

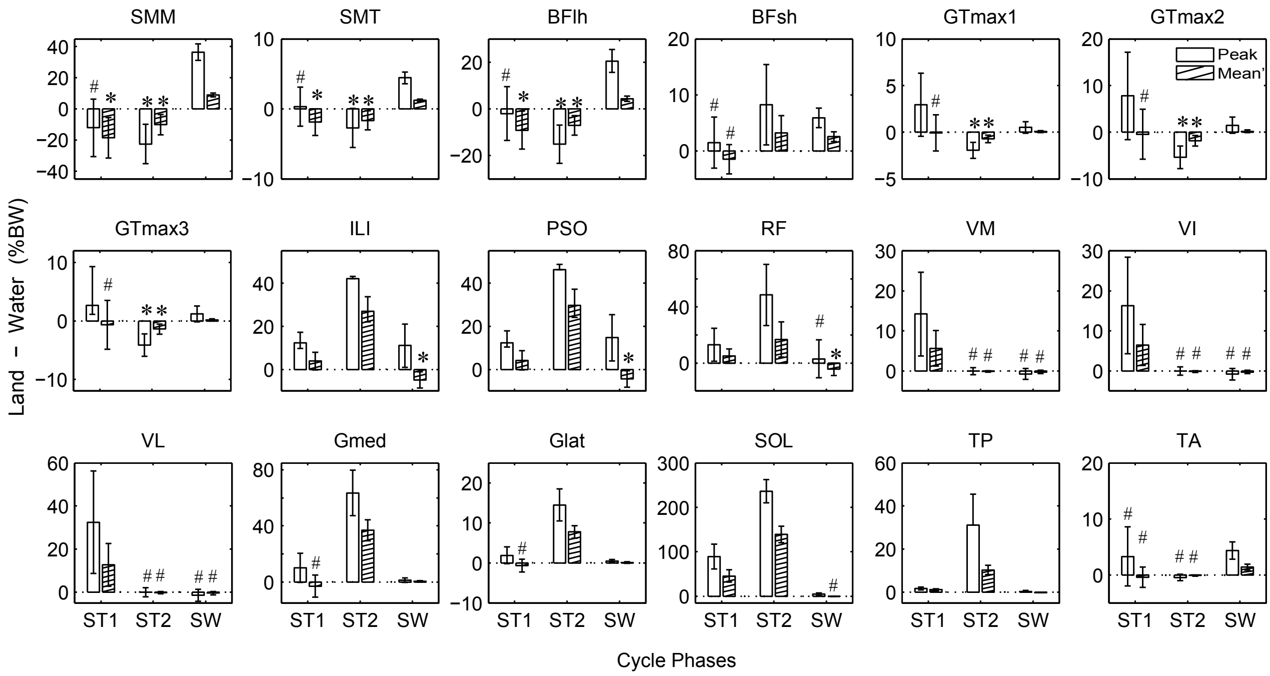

Figure 4: Difference between peak and mean force developed by the main hip, knee and ankle flexors and extensors muscles while walking on land and in water during 1st half of stance (ST1), 2nd half of stance (ST2) and swing (SW). The * indicates when values for water were greater than on land (negative difference, $p<0.05$) and # indicates no difference between environments ($p >0.05$), otherwise values for land were greater than in water (positive difference, $p<0.05$).

Concerning muscle forces, our results showed that hip, knee and ankle joints were actuated differently by their main flexors and extensors muscles, in both environments, as shown in figure 3. The peak and mean forces developed by SMT, SMM and BFlh during ST1 and ST2 of walking in the aquatic environment were comparable to or greater than those observed on land. The peak and mean forces developed by GTmax (modeled by three independent bundles) were greater in water during ST2 (the net difference for the three bundles were -11±3% and -4±1%, for the peak and mean respectively). During ST1, we also observed similar mean forces between environments for GTmax, BFsh, Gmed, Gmax and TA. During SW, we observed greater mean forces in water for ILI, PSO and RF, and similar peak forces for the RF (see figure 4).

Discussion

Our goal was to estimate the necessary lower limb muscle forces to walk in water with comfortable speed, at chest deep, and compare them to the forces observed during comfortable walking on land, in order to verify our hypothesis that muscle forces generated by the hip extensors and knee flexors are not always lower in water compared to land. Our results confirmed this hypothesis once the gluteus maximus and the hamstrings generated peak and mean forces comparable to or greater than those observed on land during the stance phase of walking in water (figure 4). In addition, we also observed that the mean force developed in the ILI, PSO, hip flexors, and RF, hip flexor and knee extensor, were greater during swing phase of walking in water than on land.

The prolonged action of hip extensors, during stance, and flexors, during swing, and the increased mean forces observed in these muscles despite the decreased gait speed while walking in water (figures 3 and 4), are explained by the propulsive role that hip flexors and extensors play in walking [9]. The continuous need for propelling the body forward against the drag forces may demand more of the hip extensors during the middle stance, as well as the hip flexors during middle swing. Gmed and Glat may also contribute to forward propulsion during early stance in water, since mean force generated by these muscles are similar in both environments, despite the decreased need to support body weight due to the action of buoyancy. The similar peak force and the greater mean force developed by RF during swing phase of walking in water, is also due to its action around the knee, since knee extension at middle swing may be generated actively by contraction of extensors, due to the combined action of drag, breaking the forward motion, and buoyancy, inducing floating, at the shank.

The fact that the forces generated by hip and knee muscles are not always lower in water than on land, support the idea that walking in the aquatic environment is effective for muscle strengthening. However, there should be a special concern about the fact that muscle forces around the hip can be greater in water than on land, since increased muscle forces can also increase joint contact loads. Other authors have shown that when increasing walking speed inside the water there is an increase in the net hip extensor and knee flexor torque during stance, which suggest that increasing walking speed may require greater forces at the hamstring and gluteus maximus [10].

Although the 2D approach employed in this study to estimate muscle forces is certainly a limitation, we think it is justified given the lack of information about the loads muscles and joints are subjected while walking inside water, and the technical difficult of performing gait analysis in the aquatic environment. In addition, our results for muscle activation agreed well with EMG activity recorded by other authors [1] and muscle forces magnitudes were within the range observed in other studies [10].

Conclusion

We were able to estimate muscle forces during comfortable walking in shallow water, at chest deep, and on land using an inverse dynamics-based static optimization. Our results showed that forces developed by hip and knee muscles are not always lower in water compared to land, which support the idea that walking in the aquatic environment is effective for muscle strengthening. This result also points that walking speed and immersion depth has to be carefully selected when applying this exercise in hip joint rehabilitation, since increased hip muscle forces may increase hip joint load.

Acknowledgments

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