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Shoulder joint kinetics and dynamics during underwater forward arm elevation

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ABSTRACT

Aquatic exercises are widely implemented into rehabilitation programs. However, both evaluating their mechanical demands on the musculoskeletal system and designing protocols to provide progressive loading are difficult tasks. This study reports for the first time shoulder joint kinetics and dynamics during underwater forward arm elevation performed at speeds ranging from 22.5 to 90°/s. Net joint moments projected onto anatomical axes of rotation, joint power, and joint work were calculated in 18 participants through a novel approach coupling numerical fluid flow simulations and inverse dynamics. Joint dynamics was revealed from the 3D angle between the joint moment and angular velocity vectors, identifying three main functions—propulsion, stabilization, and resistance. Speeds <30°/s necessitated little to no power at all, whereas peaks about 0.20 $W \cdot kg^{-1}$ were seen at 90°/s. As speed increased, peak moments were up to 61 × higher at 90 than at 22.5°/s, (1.82 ± 0.12%BW·AL vs 0.03 ± 0.01%BW·AL, $P < 0.038$). This was done at the expense of a substantial decrease in the joint moment contribution to joint stability though, which goes against the intuition that greater stabilization is required to protect the shoulder from increasing loads. Slow arm elevations (<30°/s) are advantageous for joint mobility gain at low mechanical solicitation, whereas the intensity at 90°/s is high enough to stimulate muscular endurance improvements. Simple predictive equations of shoulder mechanical loading are provided. They allow for easy design of progressive protocols, either for the postoperative shoulder or the conditioning of athlete targeting very specific intensity regions.

1. Introduction

Aquatic therapy is a widespread modality used in early stages of shoulder rehabilitation to accelerate recovery and transition to dry-land exercises. Shoulder loading should be tailored so as not to exceed the biomechanical limits of healing tissues, which might cause pain and tendon repair failure, while sufficient solicitation should be guaranteed to gradually restore joint function (Galatz et al., 2009). Planning continua of exercises for progressive aquatic rehabilitation however remains challenging because of the difficulty in measuring water resistance (Colado et al., 2008; Pöyhönen et al., 2001). This is why aquatic exercise intensity is most often inferred from electromyographical (EMG) recordings of shoulder muscles. Current clinical guidelines recommend arm elevation in water up to 90°/s (Thigpen et al., 2016), a speed below which muscle activation levels were consistently lower in water than on land, supposedly marking the positive effect of buoyancy (Kelly et al., 2000). More recently though, aquatic exercises were reevaluated to no longer be assistive at a speed of 45°/s (Castillo-Lozano et al., 2014), leaving large uncertainty in aquatic exercise prescription.

Monitoring exercise intensity through comparison of in-water and on-land EMG is questionable. Muscle activity inadequately estimates joint load, and its relationship with muscle force is highly nonlinear, complicating the interpretation of task mechanical requirements (Escamilla et al., 2009). Moreover, EMG amplitude inevitably decreases during body immersion (Pöyhönen and Avela, 2002), such that a same level of muscle activity in both media might not correspond at all to the same mechanical demand. Alternatively, some authors adopted simplified experimental approaches centered on hydrodynamics (Pöyhönen et al., 2001, 2002; Tsourlou et al., 2006). Nonetheless, although they offer a more direct evaluation of exercise intensity, they ignore impor-

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tant flow phenomena and were showed to inaccurately reflect force production by the arm (Lauder and Dabnichki, 2005).

In contrast to these studies, the actual mechanical demands of aquatic exercises have been successfully estimated through a novel, noninvasive approach integrating numerical fluid flow simulations with inverse dynamics (Lauer et al., 2016). Unprecedented insight was gained into the load on the musculoskeletal system and the way it adapts to accommodate varying mechanical demands from the environment (Lauer et al., 2017). Importantly, this approach constitutes the basis from which to examine joint function. For instance, by analyzing the angle formed by the vectors of joint moment and angular velocity during wheelchair locomotion, Desroches et al. (2010) could identify whether the shoulder was predominantly driven, stabilized, or absorbing energy to resist motion. A parallel was drawn between increased risk of overuse and the simultaneous need for shoulder stabilization and force production, shedding light onto shoulder pathomechanics.

Although an in-depth evaluation of shoulder kinetics and dynamics could inform aquatic rehabilitation protocol design, such analysis has never been carried out. The aim of this study was therefore twofold: to compute the mechanical demands of forward arm elevations performed on land and in water; and to elucidate shoulder mechanical roles at speeds ranging from 22.5 to 90°/s. Accordingly, we formulated two hypotheses. First, judging from previous EMG-based estimations, we predicted that shoulder mechanical work would become higher in water than on land at a speed between 45 and 90°/s. We anticipated, however, that calculation of shoulder kinetics would provide a narrower estimation band than EMG. Second, because of its ball-and-socket nature and the need to provide active stabilization, we hypothesized that the shoulder would be mainly stabilized, more so at high speeds to protect the shoulder against increasing load.

2. Material and methods

2.1. Participants and numerical procedure

Eighteen volunteers (7 women: 30.8 ± 9.6 years, 1.63 ± 0.06 m, 58.1 ± 9.3 kg, 21.8 ± 3.2 BMI; 11 men: 33.1 ± 9.0 years, 1.80 ± 0.09 m, 76.5 ± 13.2 kg, 23.6 ± 2.7 BMI) provided written informed consent to participate in the study. Ethical approval was granted by the University of Porto review board. Their upper bodies were scanned with a Mephisto 3D scanner (4DDynamics, Antwerp, Belgium). Virtual geometries were edited and converted into computer-aided design models prior to import into ANSYS® Fluent® Release 14.5 computational fluid dynamics software (ANSYS, Inc., Canonsburg, PA, USA). Seven anatomical landmarks (C7, T8, suprasternal notch, xiphoid process, right glenohumeral joint center, medial and lateral epicondyles of the humerus) were located in order to construct the thorax and upper arm coordinate systems (Wu et al., 2005). The glenohumeral joint center was determined from the least squares approach proposed by Gamage and Lasenby (2002), since this method yields the most accurate and reliable results when compared to the actual anatomical joint center obtained through medical imaging (Lempereur et al., 2013).

Forward arm elevations were numerically simulated in Fluent. Virtual upper body models were positioned vertically, immersed in a single phase numerical domain, with the arms in the glenohumeral neutral ("full can") position. They were smoothly animated at 22.5, 30, 45, and 90°/s with triangle waves of period 16, 12, 8, and 4 s and ranging between 0 and π, thus covering full (0–180°) shoulder range of motion. Our dynamic mesh algorithm preserved mesh quality and ensured simulation convergence. The surface of the virtual models was meshed with ∼40,000 mm-scale triangular facets onto which Fluent flow solver evaluated instantaneous pressure and shear stress. The resultant hydrodynamic force was then obtained through integration over the upper limb surface.

2.2. Net shoulder moments, power and mechanical work computation

Weight $\mathbf{F}_\mathrm{w}$, buoyancy $\mathbf{F}_\mathrm{b}$, and hydrodynamic force $\mathbf{F}_{\mathrm{h},i}$ acting onto element i, as well as their respective points of application $\mathbf{r}_\mathrm{com}$, $\mathbf{r}_\mathrm{cob}$, $\mathbf{r}_i$ in the local coordinate system, were substituted in the following equation to solve for the net shoulder joint moment $\mathbf{M}_\mathrm{s}$:

$$\mathbf{M}_\mathrm{S} = I\boldsymbol{\alpha} - \underbrace{\mathbf{r}_\mathrm{COM} \times \mathbf{F}_\mathrm{W}}_{\mathbf{M}_\mathrm{W}} - \underbrace{\mathbf{r}_\mathrm{COB} \times \mathbf{F}_\mathrm{B}}_{\mathbf{M}_\mathrm{B}} - \underbrace{\sum_{i=1}^{n} \mathbf{r}_i \times \mathbf{F}_{\mathrm{H},i}}_{\mathbf{M}_\mathrm{H}}, \tag{1}$$

with I, the moments of inertia of the upper limb and $\boldsymbol{\alpha}$, its angular acceleration. Upper limb volume (hence, Archimedes' thrust and buoyancy) and center of buoyancy location were obtained from the virtual model; upper limb mass, center of mass location and moments of inertia were estimated from scaling equations based on subject anthropometry (Dumas et al., 2007). Net shoulder moments were projected on a non-orthogonal joint coordinate system so that they correspond to the load that muscles and ligaments must resist about each individual axis (Kristianslund et al., 2014), and normalized to body weight times arm length (%BW·AL; Hof, 1996).

Instantaneous shoulder joint power was calculated as the dot product of the net shoulder moment and shoulder angular velocity vectors, and normalized to body mass. Power time series were individually integrated with respect to time over discrete periods of positive power, yielding the positive mechanical work W^+_water done by the shoulder musculature during an elevation of the arm. W^+_water was recomputed once drag and buoyancy moments subtracted to assess the positive mechanical work W^+_land done if the same movement were to be performed on land. The mechanical load in the water was expressed as a fraction of the load on land as follows: $W_\mathrm{rel} = 100 \times \frac{W^+_\mathrm{water}}{W^+_\mathrm{land}}$. A value of 100% therefore indicates that an equal amount of work should theoretically be apportioned in both physical environments. Values below and above this threshold reflect load reduction and amplification, respectively.

2.3. Interpretation of mechanical power in 3D

Negative, null, or positive power is associated with eccentric, isometric and concentric muscle actions (Robertson and Winter, 1980). However, unlike joint moment, joint power cannot simply be decomposed into three components along the three axes of a coordinate system, as power is a scalar quantity (Dumas and Chèze, 2008). Consequently, mechanical power does not readily inform about joint dynamics, nor does it indicate the proportion of the joint moment contributing to the movement (Samson et al., 2009). For ease of interpretation, Dumas and Chèze (2008) proposed the computation of the 3D angle $\theta_{\mathbf{M}\boldsymbol{\omega}}$ between the joint moment $\mathbf{M}$ and the joint angular velocity $\boldsymbol{\omega}$ according to:

$$\theta_{\mathbf{M}\boldsymbol{\omega}} = \tan^{-1}\left(\frac{\|\mathbf{M} \times \boldsymbol{\omega}\|}{\mathbf{M} \cdot \boldsymbol{\omega}}\right). \tag{2}$$

Eq. (2) returns an angle in the range [0–180°]. Recalling that joint power P equals:

$$P = \|\mathbf{M}\|\|\boldsymbol{\omega}\| \cos\theta_{\mathbf{M}\boldsymbol{\omega}}, \tag{3}$$

Three angular intervals of interest were identified. When $\theta_{\mathbf{M}\boldsymbol{\omega}}$ is in the interval 0–60° (i.e., $\cos\theta_{\mathbf{M}\boldsymbol{\omega}}$>0.5), it follows from Eq. (3) that more than 50% of the joint moment contributes to positive joint