

## A PRELIMINARY COMPUTATIONAL STUDY OF OPTIMAL HAND TRAJECTORIES FOR MANUAL WHEELCHAIR PROPULSION

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**Abstract.** Manual wheelchairs are widely used by people with different physical disabilities. However, handrim propelled wheelchair locomotion is inefficient and biomechanically unfavorable, as evidenced by several studies in the literature. In this context, this study investigates circular hand trajectories within the physiological workspace of the upper extremity that are biomechanically more favorable for propulsion. We used an OpenSim musculoskeletal model of the upper extremity and a mechanical work-based performance criterion. So far, suitable locations of 0.1 m radius circumferences have been determined.

**Keywords:** Manual wheelchairs. Wheelchair propulsion. Biomechanical models. Computer simulation. Optimal trajectories.

### 1. INTRODUCTION

In the world, about 1 billion people have some form of disability (ONU, 2011). Besides that, about 80% of the people with disability lives in developing countries and 50% of the people with disability cannot pay for health. In Brazil, there are about 62.6 million people with disability, 8.8 million of whom have motor disabilities (IBGE, 2010).

Considering the large number of wheelchair users and the economic conditions of most of them, the economically viable option is the manual wheelchair. Despite economic viability, the conventional propulsion type, through rims connected directly to the wheels, is considered inefficient and leads to a high incidence of upper limb injuries (Woude *et al.*, 2001b; Boninger *et al.*, 2002; Curtis *et al.*, 1999; Rice *et al.*, 2017). Also according to Sabbik *et al.* (2004), approximately 68% of wheelchair users face some type of upper-extremity pain, generally on the shoulder.

In order to mitigate some of these negative effects, the scientific community has been proposing alternative forms of propulsion for manual wheelchairs (van den Woude *et al.*, 2001a; Choromański *et al.*, 2013; Dysterheft *et al.*, 2017). However, studies that objectively analyze these strategies are still scarce. In this context, this project aims at objectively investigating optimal hand trajectories for manual wheelchair propulsion within the upper extremity physiological workspace that reduce muscle demand.

### 2. METHODOLOGY

#### 2.1. Musculoskeletal model

The adopted musculoskeletal model, Fig. 1, is composed by two rigid bodies, the arms and the forearms, considering bilateral symmetry, contained in the sagittal plane, and its motion is described by two generalized coordinates,  $\alpha$  and  $\beta$ , representing shoulder and elbow flexion angles, respectively. To ensure anatomical movements and avoid upper extremity injury (Cooper *et al.*, 1999),  $\alpha$  ranges from  $-30^\circ$  to  $90^\circ$  and  $\beta$  from  $0^\circ$  to  $130^\circ$ . Considering the ranges of  $\alpha$  and  $\beta$ , the workspace of the hands can be defined as in Fig. 2.

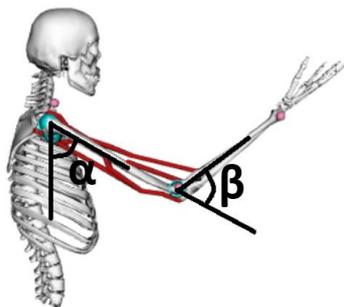


Figure 1. Musculoskeletal model with the representation of the two generalized coordinates.

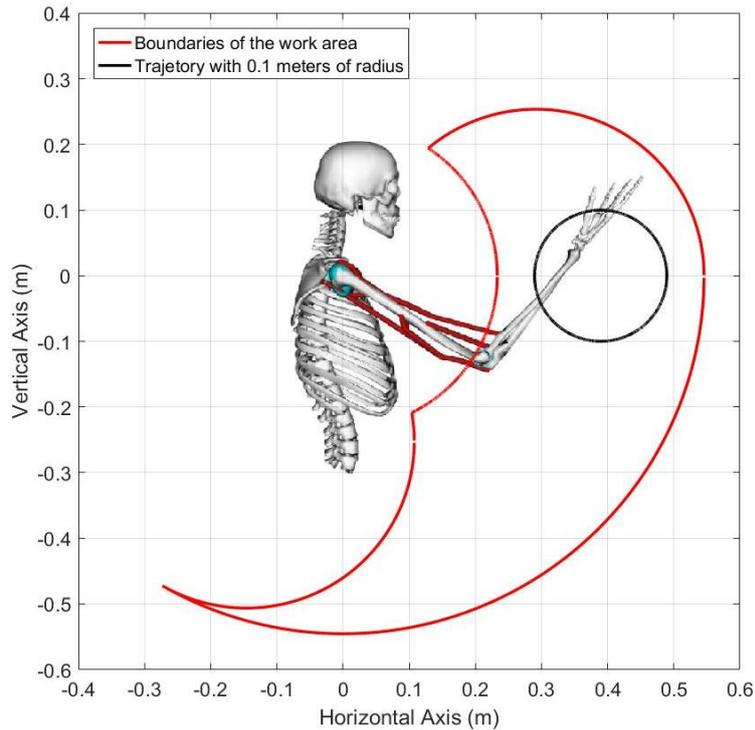


Figure 2. Representation of the model's workspace limits (in red) together with an example of circular hand trajectory (in black).

Six Hill-type muscles were used in the model. The muscles are the long head triceps, the lateral head triceps, the medial head triceps, the long head biceps, the short head biceps and the brachialis, whose parameters are extracted from the OpenSim (Delp *et al.*, 2007) model arm26.

In this preliminary study, only closed, circular, 0.1 m-radius trajectories, as in Fig. 2, are investigated. Furthermore, the motion is assumed as quasi-static so that inertial forces and the effect of the muscles' force-velocity relationship is not considered. In the future, this study will investigate the effect of different velocities and radii, as well as noncircular trajectories.

## 2.2. Performance criterion and formulation

The trajectories were evaluated by the maximum mechanical work  $\tau$  the model is able to produce over a complete closed, circular trajectory, as

$$\tau = \oint \vec{F}_{tangential_{max}} \cdot d\vec{s}, \quad (1)$$

where  $\vec{F}_{tangential_{max}}$  is the maximal tangential force the model is able to exert and  $d\vec{s}$  is an infinitesimal displacement along the trajectory.

In order to evaluate Eq. (1), the trajectory is discretized and  $\vec{F}_{tangential_{max}}$  is computed in each one of the resulting discrete positions. In the following, we will describe how the magnitude of the maximal tangential force is computed from the maximal muscle activations.

The magnitude of the maximal tangential force  $F_{tangential_{max}}$  can be computed from the two components of the contact forces  $F_x$  (horizontal) and  $F_y$  (vertical) according to Fig. 3 as

$$F_{tangential_{max}} = [\cos\theta \quad \sin\theta] \begin{bmatrix} F_x \\ F_y \end{bmatrix}, \quad (2)$$

where,  $\theta$  is the angle between  $\vec{F}_{tangential_{max}}$  and the horizontal.

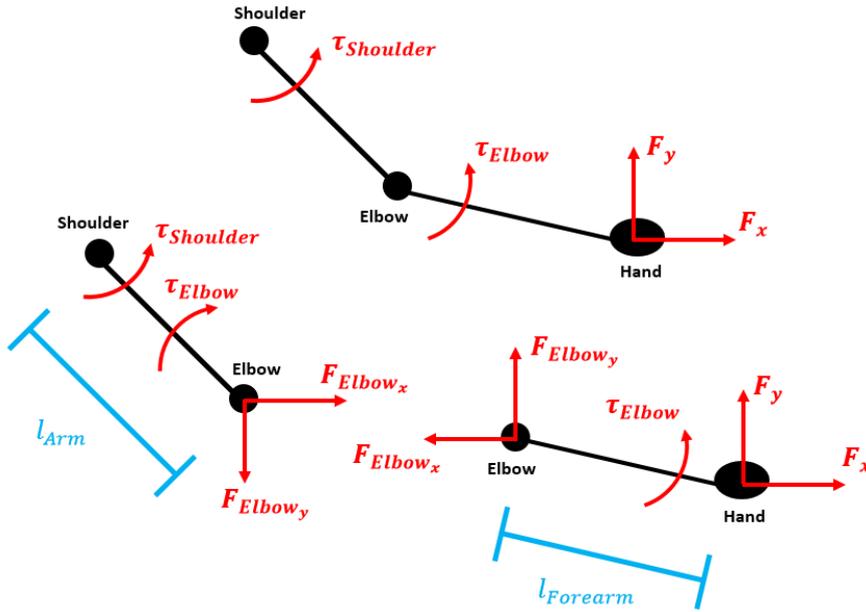


Figure 3. Free body diagram of the arm and the forearm.

The two components of the contact force, in turn, are related to the joint moments at the shoulder  $\tau_{Shoulder}$  and elbow  $\tau_{Elbow}$  as

$$\begin{bmatrix} F_x \\ F_y \end{bmatrix} = A \begin{bmatrix} \tau_{Shoulder} \\ \tau_{Elbow} \end{bmatrix}, \quad (3)$$

where matrix A is determined by applying static equilibrium conditions to the segments of the model, Fig. 3, resulting in

$$A = \begin{bmatrix} -l_{Forearm} \cos(\alpha + \beta) & -l_{Forearm} \sin(\alpha + \beta) \\ -l_{Arm} \cos(\alpha) & -l_{Arm} \sin(\alpha) \end{bmatrix}^{-1} \begin{bmatrix} 0 & 1 \\ 1 & -1 \end{bmatrix}. \quad (4)$$

The joint moments are computed by multiplying the force exerted by each muscle by the corresponding moments arms with respect to each of the generalized coordinates (Nigg and Herzog, 2006), as

$$\begin{bmatrix} \tau_{Shoulder} \\ \tau_{Elbow} \end{bmatrix} = \begin{bmatrix} b_{1Shoulder} & b_{2Shoulder} & b_{3Shoulder} & b_{4Shoulder} & b_{5Shoulder} & b_{6Shoulder} \\ b_{1Elbow} & b_{2Elbow} & b_{3Elbow} & b_{4Elbow} & b_{5Elbow} & b_{6Elbow} \end{bmatrix} \begin{bmatrix} F_1 \\ F_2 \\ F_3 \\ F_4 \\ F_5 \\ F_6 \end{bmatrix} \quad (5)$$

where  $F_i$  is the force applied by the  $i^{th}$  muscle. The corresponding moments arms with respect to the shoulder and elbow,  $b_{iShoulder}$  and  $b_{iElbow}$ , respectively, are extracted from the OpenSim model. However, each muscle force depends on the maximum isometric force that the muscle can generate  $F_{IsometricMax,i}$ , the force-length  $f_{l,i}$  and force-velocity  $f_{v,i}$  relationships and on the muscle activation  $a_i$  as

$$F_i = F_{IsometricMax,i} \cdot a_i \cdot f_{v,i} \cdot f_{l,i}, \quad (6)$$

neglecting the muscles' pennation angles. In the present preliminary study, the force-length and the force-velocity relationships are not considered so that they are both set to one.

Therefore, the maximal tangential force can be computed according to

$$F_{tangential_{max}} = [\cos\theta \quad \sin\theta] A B F_{IsoMax} F_v F_l \begin{bmatrix} a_1 \\ a_2 \\ a_3 \\ a_4 \\ a_5 \\ a_6 \end{bmatrix} \quad (7)$$

where the matrices are constructed according to Eqs. (1-6). The computation of  $F_{tangential}$  requires setting the muscle activations either to one (for agonists muscles) or zero (for antagonist) muscle, so as to maximize tangential force for a given upper extremity configuration on the trajectory.

Once, maximum tangential force is determined, is it possible to evaluate numerically the integral in Eq. (1) to estimate the maximal work over a complete close-loop trajectory.

### 3. RESULTS

The simulations were performed using MATLAB and OpenSim (Lee and Umberger, 2016). The interaction between MATLAB and OpenSim have been developed and studied (Mansouri and Reinbolt, 2012) in order to facilitate the exchange of information between the two programs. A finite number of possible 0.1 m-radius, circular trajectories within the workspace were investigated, Fig. 4 by varying the position of the circumferences' centers, as illustrated in Fig. 4.

The circular trajectories were discretized using 1801 points resulting from angular steps of  $\frac{2\pi}{1800}$  radians. The circumference centers were placed 0.02 meter apart in the x and y-axis directions and within the boundaries for the centers of 0.1 m-radius circumferences, resulting in 152 different trajectories.

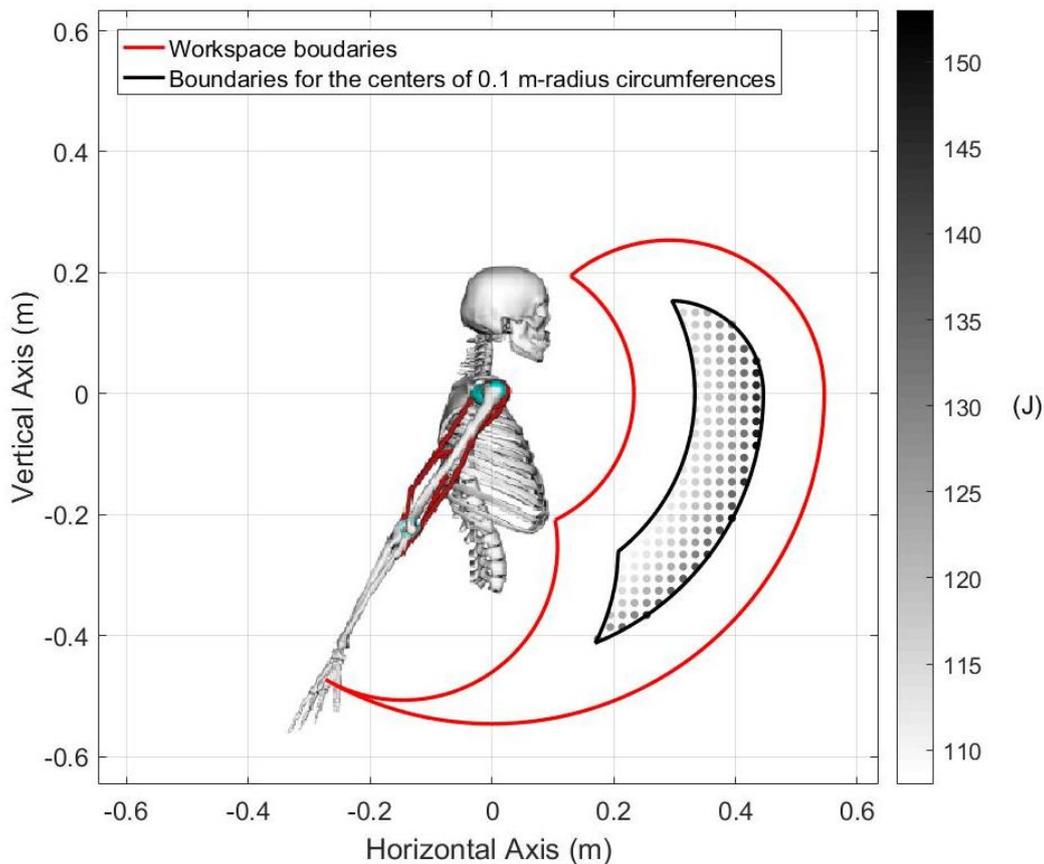


Figure 4. Graphical representation of the simulations results. The dots in gray-scale represent the centers of the respective circular trajectories. The darker the dot the larger is the maximum mechanical work computed for the respective circular trajectory.

Figure 4 shows the centers of the respective 0.1 m-radius trajectories in gray scale. A darker point indicates the center of a circular trajectory for which a larger mechanical work can be produced and vice-versa.

#### 4. DISCUSSION AND CONCLUSION

So far, our preliminary results show that trajectories that can produce larger maximal mechanical work are located farther away from the body. This would indicate that placing circular trajectories for manual wheelchair propulsion away from the user while still within its workspace could lead to less muscle activation and effort, i.e. to biomechanically more favorable configurations.

However, it is important to consider that this preliminary study has not yet considered the force-length and force-velocity relationships or the effects of inertial and gravitational forces, which are important biomechanical factors affecting performance and could substantially change these results. Therefore, further investigations on the effects of muscle intrinsic properties as well as of inertial and gravitational forces are needed to determine suitable trajectories for manual wheelchair propulsion. Future investigations should also focus on different radii and trajectory shapes.

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#### RESPONSIBILITY FOR INFORMATION

The authors are the only responsible for the information in this project.