INTRODUCTION

The physical demands placed on the upper extremity during manual wheelchair propulsion result in a high prevalence of pain and injury [1]. Seat position is an adjustable parameter that directly influences propulsion mechanics and upper extremity demand, so identifying a position that minimizes demand holds great promise for reducing the risk of pain and injury.

Current clinical guidelines [2] recommend that the seat be positioned as far posterior as possible without compromising user stability and at a height such that the angle between the upper arm and forearm is between 100° and 120° when the hand is placed on the handrim at top dead-center (TDC). These guidelines were primarily based on experimentally identified trends in indirect measures of upper extremity demand such as cadence, push angle, handrim forces and EMG across a relatively small number of seat positions.

Forward dynamics simulations provide an alternative approach to systematically examine the influence of wheelchair seat position on direct measures of upper extremity demand. When integrated with a musculoskeletal model, they can be used to systematically examine quantities that contribute to upper extremity demand across a wide range of seat positions. The purpose of this study was to use forward dynamics simulations of wheelchair propulsion to investigate how seat position influences muscle stresses, muscle-produced joint moments and metabolic cost.

METHODS

The 3D musculoskeletal model used in this study was developed using SIMM/Dynamics Pipeline (Musculographics, Inc.) and based on the work of Rankin et al. [3]. The model had five rotational degrees-of-freedom (DOFs), representing the articulations of the shoulder, elbow and forearm. Twenty-six Hill-type musculoskeletal actuators represented the major upper extremity muscles crossing these joints.

The motion of the hand was prescribed to follow a path on a standard circular handrim during the push phase. The contact angle, release angle and push frequency were dependent on seat position and were calculated using a set of equations developed by Richter [4], assuming an initial position push frequency of 1Hz. Push time was set to 40% of the cycle time. For all seat positions, the average power output to the handrim was set at 10W.

From the initial position (hub-shoulder angle: 0°, TDC elbow angle: 110°), the seat position was systematically varied independently in the superior/inferior and the anterior/posterior directions throughout the range of achievable seat positions, which resulted in 53 investigated positions.

A forward dynamics simulation was generated for each seat position using a simulated annealing optimization algorithm to identify the muscle excitation patterns that minimized the time rate of change in handrim force and muscle-produced joint moments while producing realistic wheelchair propulsion mechanics.

The level of antagonistic muscle-produced joint moments was quantified by the co-contraction moment (τ_{cc}), defined as the difference between the average total magnitude of moments (τ_{total}) and the average net moment (τ_{net}):

\[
\tau_{total} = \frac{1}{n_{dof} \cdot t_c} \int_0^{t_c} \left( \sum_{j=1}^{n_{dof}} \sum_{i=1}^{n_{mus}} |\tau_{ij}(t)| \right) dt \tag{1}
\]

\[
\tau_{net} = \frac{1}{n_{dof} \cdot t_c} \int_0^{t_c} \left( \sum_{j=1}^{n_{dof}} \sum_{i=1}^{n_{mus}} \tau_{ij}(t) \right) dt \tag{2}
\]
\[ \tau_{cc} = \tau_{total} - \tau_{net} \] (3)

where \( \tau_{i,j} \) is the moment that the \( i^{th} \) muscle applies about the \( j^{th} \) DOF and \( t_c \) is the cycle time.

Co-contraction moment, muscle stress and metabolic cost (normalized by the total work done on the handrim throughout the cycle) were calculated and compared across seat positions. Metabolic cost was calculated using a previously described model [5].

RESULTS AND DISCUSSION

The average co-contraction moment across all DOFs (Fig. 1) was minimized (4.6 Nm) at a seat position with hub-shoulder angle of -6.4° and TDC elbow angle of 122.2°. Seat positions with hub-shoulder angles between -10° and -2.5° and TDC elbow angles between 110° and 130° had an average co-contraction moment of less than 5 Nm.

The normalized metabolic cost was minimized (8.5) at a seat position with hub-shoulder angle of -11.6° and TDC elbow angle of 106.6°. Seat positions with hub-shoulder angles between -15° and 10° and TDC elbow angle between 100° and 120° had an associated normalized metabolic cost less than 10.

Additional analysis found that further posterior seat positions placed some of the major power-producing muscles during the push phase (e.g., anterior deltoid and pectoralis major) under non-optimal operating conditions (i.e. less favorable regions of the intrinsic muscle force-length relationship). Also, a decomposition of the co-contraction moment into individual DOF components found that the muscles crossing the glenohumeral joint predominantly influenced the amount of co-contraction.

All examined upper extremity demand measures were near minimum values at positions with hub-shoulder angles between -10° and -2.5° and TDC elbow angles between 110° and 120°. These results agree with current clinical guidelines given that further posterior positions would likely compromise user stability in a standard manual wheelchair. However, even if a novel wheelchair design could achieve further posterior seat positions while maintaining user stability, our results suggest that there is a limit to how far posterior to position the seat before upper extremity demand begins to increase.

REFERENCES


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