Clinical Biomechanics 28 (2013) 378–385

Contents lists available at SciVerse ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

A theoretical analysis of the influence of wheelchair seat position on upper extremity demand

Jonathan S. Slowik, Richard R. Neptune *

Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX, USA

ARTICLE INFO

Article history:
Received 10 October 2012
Accepted 13 March 2013

Keywords:
Forward dynamics simulation
Musculoskeletal model
Biomechanics
Wheelchair propulsion
Muscle stress
Co-contraction
Metabolic cost

ABSTRACT

Background: The high physical demands placed on the upper extremity during manual wheelchair propulsion can lead to pain and overuse injuries that further reduce user independence and quality of life. Seat position is an adjustable parameter that can influence the mechanical loads placed on the upper extremity. The purpose of this study was to use a musculoskeletal model and forward dynamics simulations of wheelchair propulsion to identify the optimal seat position that minimizes various measures of upper extremity demand including muscle stress, co-contraction and metabolic cost.

Methods: Forward dynamics simulations of wheelchair propulsion were generated across a range of feasible seat positions by minimizing the change in handrim forces and muscle-produced joint moments. Resulting muscle stress, co-contraction and metabolic cost were examined to determine the optimal seat position that minimized these values.

Findings: Muscle stress and metabolic cost were near minimal values at superior/inferior positions corresponding to top-dead-center elbow angles between 110 and 120° while at an anterior/posterior position with a hub-shoulder angle between −10 and −2.5°. This coincided with a reduction in the level of muscle co-contraction, primarily at the glenohumeral joint.

Interpretation: Deviations from this position lead to increased co-contraction to maintain a stable, smooth propulsive stroke, which consequentially increases upper extremity demand. These results agree with previous clinical guidelines for positioning the seat to reduce upper extremity overuse injuries and pain for wheelchair users.

© 2013 Elsevier Ltd. All rights reserved.

1. Introduction

There are approximately 3.3 million wheelchair users in the United States (CDC, 2009), with the vast majority (>90%) of users relying on manual wheelchair propulsion as their primary method of mobility (Kaye et al., 2000). Upper extremity pain and injuries that frequently occur in wheelchair users can be extremely debilitating and lead to a decrease in independence and quality of life (e.g., Gutierrez et al., 2007). The high incidence of pain and injury is correlated with the high physical demand placed on the upper extremity during wheelchair propulsion (e.g., Curtis et al., 1999). In addition to generating the mechanical power required to propel the wheelchair, the upper extremity muscles must also help maintain joint stability (e.g., Requejo et al., 2008). These stability requirements, along with the kinematic constraints of the push phase, require significant intermuscular coordination and co-contraction (e.g., Rankin et al., 2010, 2011, 2012; van der Helm and Veeger, 1996). Although co-contraction has many beneficial purposes (e.g., helping to stabilize a joint), it can also have detrimental effects (e.g., elevated joint loading and muscle fatigue). Notably, the glenohumeral joint has relatively few stabilizing structures (Veeger and van der Helm, 2007), requiring the muscles responsible for stabilizing the joint to be highly active and have an elevated risk of injury (e.g., Mulroy et al., 2004; Veeger et al., 2002).

Seat position is an easily adjustable parameter that directly influences propulsion mechanics (e.g., Boninger et al., 2000; Gorce and Louis, 2012; Kotajarvi et al., 2004; Richter, 2001) and upper extremity demand (e.g., Gutierrez et al., 2005; Mulroy et al., 2005; Paralyzed Veterans of America Consortium for Spinal Cord Medicine (PVACSCM), 2005; Requejo et al., 2008). Thus, identifying the optimal seat position that minimizes upper extremity demand holds great promise for reducing the risk of pain and injury. A number of studies have examined the influence of seat position on propulsion mechanics and found relationships with specific biomechanical measures such as cadence (e.g., Boninger et al., 2000; Gorce and Louis, 2012; Kotajarvi et al., 2004; Masse et al., 1992; Richter, 2001), handrim forces (e.g., Boninger et al., 2000; Kotajarvi et al., 2004; van der Woude et al., 2009), joint ranges of motion (e.g., Gorce and Louis, 2012; Wei et al., 2003) and electromyography (EMG) activity (e.g., Gutierrez et al., 2005; Louis and Gorce, 2010; Masse et al., 1992). High levels of these measures have been identified as risk factors for upper extremity pain and injury (e.g., Gorce and Louis, 2012; Gutierrez et al., 2005; PVACSCM, 2005).

* Corresponding author at: Department of Mechanical Engineering, The University of Texas at Austin, 204 E. Dean Keeton Street, Stop C2200, Austin, TX 78712, USA.
E-mail address: rneptune@mail.utexas.edu (R.R. Neptune).

0268-0033/$ – see front matter © 2013 Elsevier Ltd. All rights reserved.
http://dx.doi.org/10.1016/j.clinbiomech.2013.03.004

Recent clinical guidelines based on these relationships suggest that seat position should be adjusted as far posterior as possible without compromising wheelchair stability (PVACSCM, 2005). The guidelines also recommend superior/inferior positions that correspond to an elbow angle between 100° and 120° when the hand is at the top-dead-center (TDC) position on the handrim (full extension is 180°). However, recent studies have found with such low, posterior seat positions, the joint ranges of motion and muscle activity levels may be increased (Gorce and Louis, 2012; Louis and Gorce, 2010), which may adversely affect upper extremity demand.

One challenge in assessing the influence of seat position on upper extremity demand is the difficulty in directly measuring demand-related quantities such as muscle stress or co-contraction that may elevate joint loading. Because these measures are difficult to obtain experimentally, indirect measures are frequently used (e.g., Erdemir et al., 2007). For example, inverse dynamics techniques are often used to determine joint moments, but identifying individual muscle force and stress values is challenging due to muscle redundancy and co-contraction (Erdemir et al., 2007; Zajac et al., 2002). In addition, systematically investigating the influence of seat position on upper extremity demand using experimental techniques is difficult and time-consuming (e.g., to assess the influence of seat position on metabolic cost). As a result, most studies have investigated a limited number of seat positions (e.g., two or three).

Forward dynamics simulations provide an alternative approach to systematically examine the influence of wheelchair seat position on direct measures of upper extremity demand. Forward dynamics techniques have been successfully used to analyze various human movement tasks such as pedaling (e.g., Raasch et al., 1997; Rankin and Neptune, 2008), walking (e.g., Anderson and Pandy, 2001; Neptune et al., 2004) and running (e.g., Miller et al., 2012; Sasaki and Neptune, 2006). More recently, simulations have been applied to analyzing wheelchair propulsion to identify individual muscle contributions to push and recovery mechanics (e.g., Rankin et al., 2010, 2011, 2012). Simulations can also be used to systematically vary system parameters and analyze their influence on specific biomechanical measures. Such studies have recently been used to optimize the design of bicycle configurations (e.g., Rankin and Neptune, 2010) and lower limb prostheses (e.g., Fey et al., 2012).

The purpose of this study was to use forward dynamics simulations of wheelchair propulsion to investigate how seat position influences upper extremity demand including muscle stress, co-contraction and metabolic cost. Understanding these relationships can help guide clinicians in determining the optimal wheelchair configuration to reduce upper extremity demand, and ultimately overuse injuries and pain in wheelchair users.

2. Methods

2.1. Musculoskeletal model

The musculoskeletal model used in this study was based on a previously described upper extremity model representing a 50th percentile able-bodied male (Holzbaur et al., 2005; Rankin et al., 2010, 2011) and will be summarized here (Fig. 1). The model was developed using SIMM (Musculographics, Inc., Santa Rosa, CA, USA) and consisted of a trunk and right side upper arm, forearm and hand segments. There were three degrees-of-freedom (DoFs) at the shoulder, defined as plane of elevation, elevation angle and internal–external rotation. The motion at the shoulder also included a scapulohumeral rhythm based on regression equations derived from cadaver data (Holzbaur et al., 2005). The model had two additional DoFs, elbow flexion–extension and forearm pronation–supination. The wrist was fixed in the standard anatomical position.

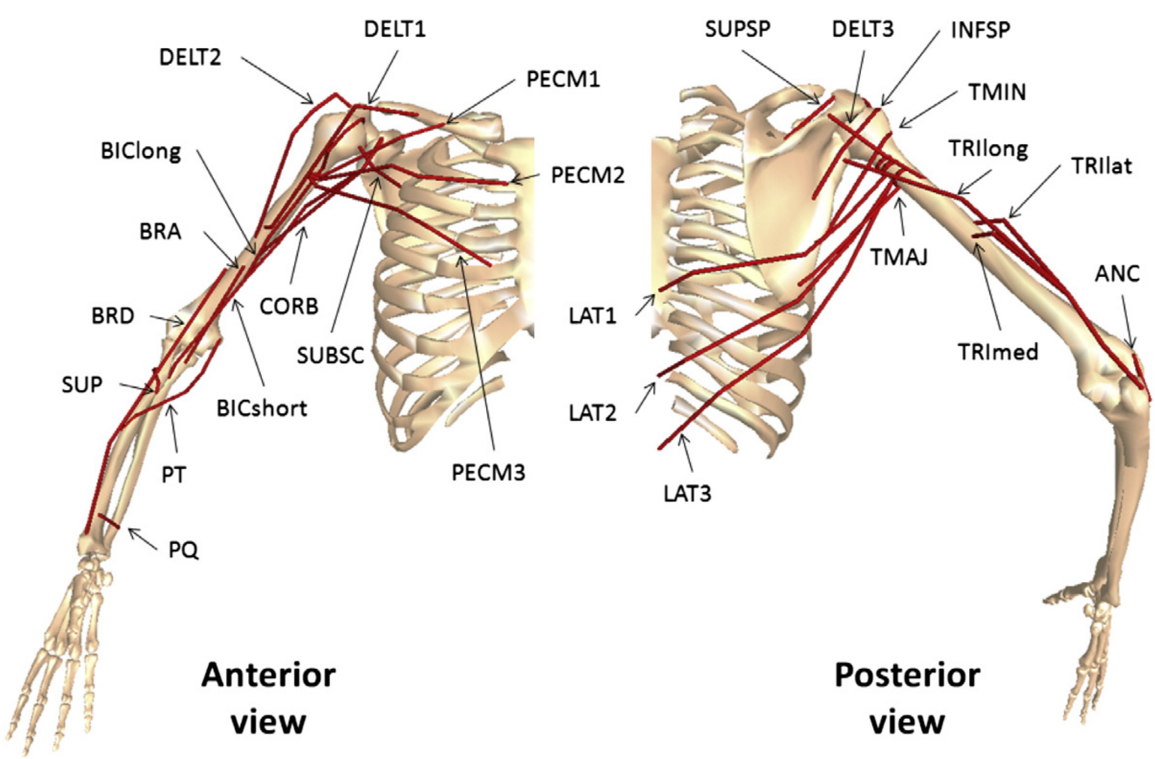

Fig. 1. 3D musculoskeletal model used in the wheelchair propulsion simulations. The model had 5 degrees-of-freedom: plane of elevation, elevation angle, internal–external rotation, elbow flexion–extension and forearm pronation–supination. Twenty-six Hill-type musculotendon actuators represented the major upper extremity muscles crossing the shoulder and elbow joints. These actuators were: DELT1 (anterior deltoid), DELT2 (middle deltoid), DELT3 (posterior deltoid), PECM1 (pectoralis major, clavicular head), PECM2 (pectoralis major, sternocostal head portion 1 — sternum), PECM3 (pectoralis major, sternocostal head portion 2 — ribs), CORB (coracobrachialis), TMAJ (teres major), LAT1 (latissimus dorsi, thoracic portion), LAT2 (latissimus dorsi, lumbar portion), LAT3 (latissimus dorsi, iliac portion), SUBSC (subscapularis), INFSP (infraspinatus), TMIN (teres minor), SUPSP (supraspinatus), BRD (brachioradialis), BRA (brachialis), BICshort (biceps brachii, short head), BIClong (biceps brachii, long head), ANC (anconeus), TRIlat (triceps brachii, lateral head), TRImed (triceps brachii, medial head), TRIlong (triceps brachii, long head), SUP (supinator), PQ (pronator quadratus) and PT (pronator teres).

Download English Version:

https://daneshyari.com/en/article/6204938

Download Persian Version:

https://daneshyari.com/article/6204938

Daneshyari.com