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Short communication

Biomechanical model for evaluation of pediatric upper extremity joint dynamics during wheelchair mobility

Alyssa J. Schnorenberg [a,b,c,*], Brooke A. Slavens [a,b,d], Mei Wang [b,c,e], Lawrence C. Vogel [d], Peter A. Smith [b,c,d], Gerald F. Harris [b,c,d,e]

[a] Department of Occupational Science & Technology, University of Wisconsin-Milwaukee, Milwaukee, WI, USA
[b] Orthopaedic and Rehabilitation Engineering Center (OREC), Marquette University/Medical College of Wisconsin, Milwaukee, WI, USA
[c] Department of Biomedical Engineering, Marquette University, Milwaukee, WI, USA
[d] Shriners Hospitals for Children, Chicago, IL, USA
[e] Department of Orthopaedic Surgery, Medical College of Wisconsin, Milwaukee, WI, USA

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ABSTRACT

Pediatric manual wheelchair users (MWU) require high joint demands on their upper extremity (UE) during wheelchair mobility, leading them to be at risk of developing pain and pathology. Studies have examined UE biomechanics during wheelchair mobility in the adult population; however, current methods for evaluating UE joint dynamics of pediatric MWU are limited. An inverse dynamics model is proposed to characterize three-dimensional UE joint kinematics and kinetics during pediatric wheelchair mobility using a SmartWheel instrumented handrim system. The bilateral model comprises thorax, clavicle, scapula, upper arm, forearm, and hand segments and includes the sternoclavicular, acromioclavicular, glenohumeral, elbow and wrist joints. A single 17 year-old male with a C7 spinal cord injury (SCI) was evaluated while propelling his wheelchair across a 15-meter walkway. The subject exhibited wrist extension angles up to 60°, large elbow ranges of motion and peak glenohumeral joint forces up to 10% body weight. Statistically significant asymmetry of the wrist, elbow, glenohumeral and acromioclavicular joints was detected by the model. As demonstrated, the custom bilateral UE pediatric model may provide considerable quantitative insight into UE joint dynamics to improve wheelchair prescription, training, rehabilitation and long-term care of children with orthopedic disabilities. Further research is warranted to evaluate pediatric wheelchair mobility in a larger population of children with SCI to investigate correlations to pain, function and transitional changes to adulthood.

1. Introduction

A 2012 Americans with Disabilities Report states that about 3.7 million people in the United States of America (USA) use a wheelchair, with about 124,000 wheelchair users under the age of 21 and 67,000 under the age of 15 (Brault, 2012). Additionally, 90% of wheelchair users in the USA are manual wheelchair users (MWU) and among children under the age of 18, the wheelchair is the most used assistive mobility device (Kaye et al., 2000).

It has been reported that adults experience approximately a 4.2–6.6 N-m mean net shoulder moment, with peak moments between 6.7 and 10.3 N-m (Van Drongelen et al., 2011). Specifically, maximum shoulder flexion moments have been reported at 5.7 N m and 7.9 N-m during wheelchair propulsion at speeds of 3 km/h and 4 km/h, respectively (Gil-Agudo et al., 2010a). As the upper extremities (UEs) are not constructed for high magnitude or high frequency force demands, overuse injuries such as carpal tunnel syndrome, shoulder impingement and UE pain are commonly developed (Crane, 2007; Veeger et al., 1998) and were reported in 50% of MWU with spinal cord injury (SCI) (Boninger et al., 2005). These secondary problems have been associated with increased loading at extremes of joint excursions (Corfman et al., 2003). Due to increased life expectancy and continual wheelchair usage, these injuries may reduce or severely limit independent function and quality of life (Crane, 2007).

These issues may be of even greater concern in the pediatric population of long-term wheelchair users. Due to pediatric growth and maturation, simple scaling of adult body segment parameters is not appropriate (Jensen, 1989). While studies have examined wheelchair mobility in the adult population (Mercer et al., 2006; Gil-Agudo et al., 2010b), current biomechanical models for pediatric wheelchair mobility are limited by non-standard kinematic models (Bednarczyk and Sanderson, 1995; Wu et al., 2005).

* Corresponding author at: University of Wisconsin-Milwaukee, Department of Occupational Science and Technology, Enderis Hall 983, 2400 East Hartford Avenue, Milwaukee, WI 53201, USA. Tel.: +1 414 229 4137.
E-mail address: paulaj@uwm.edu (A.J. Schnorenberg).

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We propose a pediatric UE model for biomechanical analysis of wheelchair mobility in children and adolescents with orthopedic disabilities, such as spinal cord injury. Previous models from our group lay the groundwork for pediatric assistive mobility modeling (Slavens et al., 2009; Konop et al., 2009).

Studies have shown that assuming symmetry between left-hand and right-hand sides, when assessing wheelchair propulsion, may result in errors (Boninger et al., 2002). While the clinical implications of asymmetry are still unclear, asymmetrical propulsion may be a contributing factor to the development of pain and pathology (Hurd et al., 2008). Characterizing bilateral UE joint dynamics during wheelchair propulsion will provide insight to motion and loading patterns. Investigation of bilateral peak joint dynamics may help identify potentially injurious risk factors. Better knowledge of UE joint dynamics during wheelchair mobility may improve our understanding of the onset and propagation of UE pathologies. Additionally, this work may be applied to identify biomechanical risk factors of pain and pathology for long-term wheelchair users. This may guide clinicians and engineers to improvements in wheelchair prescription, design, training and transitional and long-term care. Thereby, pain and pathology onset may be slowed or prevented, and quality of life restored.

2. Methods

2.1. Kinematic model

The bilateral UE model comprises 11 segments, including the thorax, clavicles, scapulae, upper arms, forearms and hands. The joints of interest are three degree-of-freedom wrist, glenohumeral (GH) and acromioclavicular (AC) joints; and two-degree-of-freedom sternoclavicular (SC) and elbow joints. To define the segments, twenty-seven passive reflective markers are placed on the subject (Fig. 1). Additionally, the wheelchair was modeled as one rigid body segment using four markers. Joint axes were embedded at the joint centers, which are calculated using subject specific anthropometric data. A *Z–X–Y* Euler sequence is used to determine the joint angles of the distal segment with respect to the proximal segment for the thorax and the GH, elbow and wrist joints. A *Y–X–Z* Euler sequence is used for AC joint (scapula with respect to the clavicle) and SC joint (clavicle with respect to the sternum) angle determination. All rotation sequences follow ISB recommendations (Wu et al., 2005). Matlab (MathWorks, Inc., Natick, MA) was used for model development.

2.2. Model features

Previous, validated UE models for the evaluation of pediatric assisted mobility, created by our group (Slavens et al., 2011a, 2011b, 2010, 2009; Konop et al., 2009) served as the foundation for the development of this model. The model incorporated ISB recommendations for segment design (Wu et al., 2005) as well as custom features specific to pediatric wheelchair mobility. To avoid possible marker contact with the wheelchair during propulsion, a single marker was placed on the olecranon, a method previously validated (Hingtgen et al., 2006). The marker set used to describe the thorax was refined to closely reflect the pediatric model described by Nguyen and Baker (2004) to reduce the influence of shoulder girdle movement on thoracic kinematic measurements. To provide a comprehensive description of shoulder girdle dynamics, scapula and clavicle segments were included in this model. ISB recommended modeling methods were closely followed (Wu et al., 2005), except that, in order to accommodate a child's smaller size, a single marker on the suprasternal notch was used in place of dual SC joint markers for clavicle segment definition, after research showed a minimal effect on the results (Van der Helm and Pronk, 1995). A regression method was applied for determining GH joint center locations that used the positions of five markers on the scapula (Meskers et al., 1998). This method was chosen because of its ease of implementation with this model, its general acceptance and its increased accuracy over more simplistic methods, such as using shoulder circumference (Campbell et al., 2009). A marker tracking method was utilized for the scapula markers to reduce the effects of skin motion artifact as well as possible marker-wheelchair interaction (Šenk and Chèze, 2010). Body segment parameters were calculated by equations specifically developed for the pediatric population (Jensen, 1989; Yeadon and Morlock, 1989).

2.3. Kinematic model: joint centers

The positions of the joint centers were the origins for each segment's local coordinate system, except for the thorax, scapulae and forearms, whose origins were the suprasternal notch (IJ), acromial angles (AA) and ulnar styloids (ULN), respectively (Wu et al., 2005). All joints were assumed to have fixed centers of rotation.

2.4. Kinematic model: segment coordinate systems

Segment coordinate systems (SCS) were determined for each segment. The joint angles were determined by the relative motion between two adjacent SCS, distal relative to proximal. The SCS follow the right-hand rule with the *Z*-axis as the flexion/extension axis; the *X*-axis as the abduction/adduction axis; and the *Y*-axis as the internal/external rotation axis. Equations presented in Table 1 define the right-hand side.

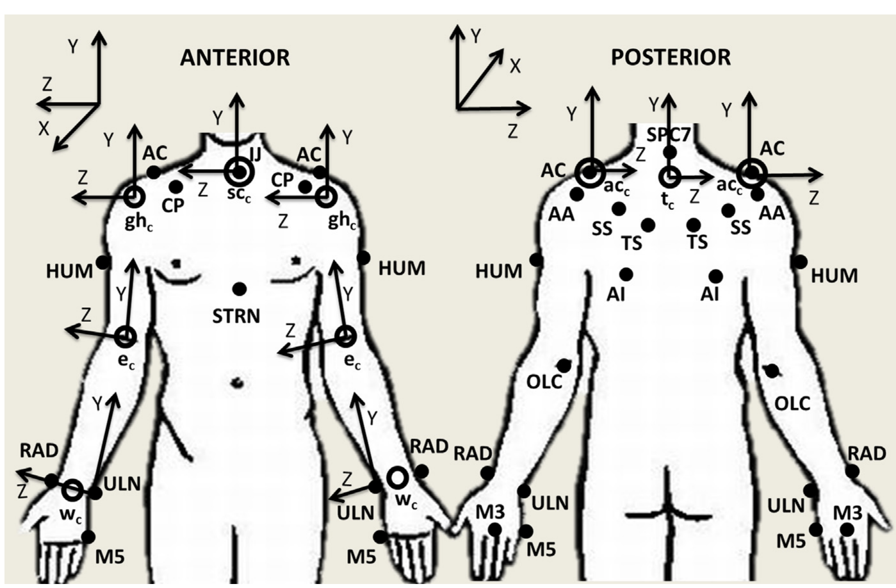

Fig. 1. Upper extremity model marker set: suprasternal notch (IJ), xiphoid process (STRN), spinous process at C7 (C7), acromioclavicular joint (AC), inferior angle (AI), trigonum spine (TS), scapular spine (SS), acromial angle (AA), coracoid process (CP), humerus (HUM), olecranon (OLC), radial styloid (RAD), ulnar styloid (ULN), third and fifth metacarpals (M3 and M5). Joint centers: wrist joint center (w_c), elbow joint center (e_c), glenohumeral joint center (gh_c), acromioclavicular joint center (ac_c), sternoclavicular joint center (sc_c) and thorax center (t_c) are represented by the open circles. Following ISB recommendations, definition of axes of rotation follow right-hand rule and the *Z*-axis points laterally towards the subject's right side, the *X*-axis points anteriorly, and the *Y*-axis points superiorly (Wu et al., 2005).