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Upper Extremity Biomechanical Model of Crutch-Assisted Gait in Children

Neha Bhagchandani[1, 2], Brooke Slavens [1-3], Mei Wang[1, 2] and Gerald Harris [1-3]
[1]Orthopaedic and Rehabilitation Engineering Center, (OREC), [2]Department of Biomedical Engineering, Marquette University, Milwaukee, WI and [3]Shriners Hospitals for Children, Chicago, IL

Abstract— **A 3D biomechanical model with a novel instrumented Lofstrand crutch system is presented. The novel Lofstrand crutch system consists of two six-axis load cells incorporated in the crutch to study the reaction forces occurring at the crutch handle and the cuff. The goal of this study is to quantify the effect of the cuff forces with the help of this improved crutch system. The kinematic model developed is verified based on previous studies. The kinetic model, consisting of the forces, is derived using the kinematic data, anthropometric data and the reaction forces generated from the load cells. The kinetic data is also in accordance with previous studies. Thus, the novel crutch system has been verified for evaluating the force loading on shoulder, elbow and wrist. This model would be further implemented on children suffering from Osteogenesis Imperfecta (OI), which would help in evaluating injury prevention criteria for long-term crutch users.**

Keywords— Instrumented Crutches, Gait, Upper Extremity, Osteogenesis Imperfecta

I. INTRODUCTION

While biomechanical analysis has been used extensively to study unassisted motion during gait, little has been accomplished to characterize upper extremity dynamics during assisted gait. Motion analysis is a noninvasive and painless technique that allows evaluation of multiplanar motion during functional activity [1]. Improved motion analysis technology and modeling software has allowed more rapid development of complex models such as those needed to study the upper extremities (UE) during assisted gait. We have applied current instrumentation and modeling concepts to study Lofstrand crutch-assisted gait dynamics. Evaluating the UE dynamics of crutch users may ultimately help to prevent injuries due to excessive loading or inappropriate gait patterns. These evaluations may also assist in pre-treatment planning and post treatment rehabilitation.

An earlier study of UE dynamics during Lofstrand crutch-assisted gait in children with myelomeningocele (MM) demonstrated a limitation in the instrumentation to fully define the complete kinetics of the UE joints during ambulation [2]-[5]. The early model was unable to include the kinetic interactions between the crutch cuff and the forearm. Another earlier study analyzed Lofstrand crutch-assisted gait of a single adult subject with spinal cord injury (SCI) [6]. Their system consisted of one six-axis load cell and a three-axis strain gauge which measured only moments and failed to quantify the forces at the cuff. The goal of the current study was to develop a novel crutch system to evaluate the complete set of kinematics and kinetics to study the UE joint dynamics during Lofstrand crutch-assisted gait.

II. METHODOLOGY

A. Kinematic Model

The UEs are modeled using seven rigid body segments which are the thorax, upper arms, forearms, and hands. The UE segments are connected by a 3 degree of freedom (dof) shoulder joint, a 2 dof elbow joint and a 2 dof wrist joint. Each crutch is modeled using two rigid body segments defined by the cuff and handle of the crutch. Eighteen reflective markers are used to define these segments (Fig 1). These markers are placed on bony anatomical landmarks to reduce skin and soft tissue motion between bones and markers. In order to determine crutch kinematics five reflective markers are placed on each crutch.

Joint centers were calculated using the markers and subject specific anthropometric data. Joint coordinate segment axes are based on ISB standard recommendations [7]. The thorax model is based on the study done by Nyugen et al, for analyzing thorax kinematics in children with MM [8]. The shoulder joint is determined by locating the humeral head [9]. The midpoint of the lateral and the medial epicondyles is used to calculate the elbow joint center. Similarly, for the wrist joint center the midpoint of the radial and the ulnar joint was evaluated. Z-X-Y Euler rotation sequence is used to define rotations of the segments. The rotations of the distal coordinate system are defined with respect to the proximal coordinate system, while the crutch and the thorax are referenced to the lab coordinate system.

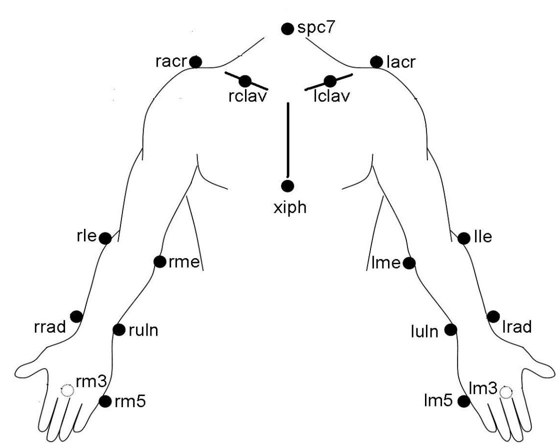

Fig. 1. Reflective marker placement for defining the UE segments.

B. Kinetic Model

The crutch and the UE forces are evaluated by the kinetic model. The kinetic model is developed using the inverse dynamics Newton-Euler approach [10]. The reaction forces and the moments from the instrumented crutches are used to evaluate the 3D forces and moments occurring at the crutch, wrist, elbow and shoulder. In order to evaluate the kinetics of the UEs, inputs such as joint velocities and accelerations are needed. These are computed by means of Euler angles obtained from kinematics for the wrist, elbow, shoulder and trunk. The centers of mass of the segments are determined by the marker positions and inertial properties of human body segments [11]. The crutch is assumed to be composed of cylindrical shells and solid cylinder for the load cells. This assumption is used to define the center of mass and the moment of inertia of the crutches. Segmental masses, segmental acceleration of the center of the mass and the distal forces are analyzed through inverse dynamics to evaluate the forces at the proximal segments. For evaluating the forces, the Newton-Euler force equation was used for the individual joint:

$$\overline{F_i} - \overline{F_{i+1}} - m\overline{g} - m\overline{a} = 0 \quad (1)$$

where forces $\overline{F_i}$ and $\overline{F_{i+1}}$ are the distal and proximal forces acting on a segment *i* where *i* represents the distal joint and *i+1* represents the proximal joint [3],[10]. Mass of the particular segment is given by *m*. $\overline{g}$ is the acceleration due to gravity and $\overline{a}$ is the acceleration of the i^{th} segment.

C. Instrumentation

Crutches were instrumented with FS-6 load cells (AMTI, Watertown, MA) to measure the applied reaction forces. These load cells measure forces in the x, y and z directions. Each crutch consists of two load cells placed above and below the handle (Fig. 2).

The load cells are made from high-strength aluminum alloy. Strain gage bridges are incorporated in the load cells to evaluate the forces. Thin co-axial cables are used to multiplex crutch data for data acquisition. The load cells have high sensitivity, high stiffness, low cross talk and long term stability. The analog data from the load cells are amplified using AMTI MSA-6 high gain amplifiers. The weight of the load cells is 0.10 kg each and the weight of the crutch used for this study is 0.43 kg. Thus, the addition of these load cells to the crutches would not deter the natural gait pattern. All force data were tested and calibrated.

Fig. 2. Instrumented crutches.

D. Subject

This pilot study was designed to test and verify the novel instrumented crutch system. A normal subject, 24 year old female, participated in the study. Her height and weight were 1.7 m and 54.4 kg, respectively. Since the subject had no previous experience using crutches, prior practice sessions were performed for acclimatization to using Lofstrand crutches.

E. Data Collection, Processing and Analysis

The subject was asked to walk with the bilateral instrumented Lofstrand crutches on a 6-meter walkway for 6 trials. A Vicon motion analysis system, with 15 infrared cameras, was used to capture 3D motion of the reflective markers placed on the bony landmarks of the subject and the crutches. The motion data was sampled at the rate of 120 Hz. Vicon workstation was used for processing the motion data and generating 3D coordinates of the markers, which are then analyzed using Vicon BodyBuilder. The motion data was filtered using a Woltering filter. Data was averaged over 6 trials. Right foot heel strike to heel strike is defined as a 100 % gait cycle with data being processed every 1 % of the gait cycle. Matlab (The MathWorks Inc. Natick, MA) was used for further data analysis.

Cadence, walking speed, stride length and stance duration were calculated for each subject. Motion for the thorax, shoulder, elbow and the wrist were evaluated for the sagittal, coronal and the transverse planes. Maximum motion was seen in the sagittal plane which was used for calculating the range of motion. Forces at the shoulder, elbow, wrist and crutch were also determined for all three planes. Mean forces and peak forces were analyzed as % body weight (BW). Peak forces are maximum forces seen over the complete gait cycle in a particular plane.

III. RESULTS

A. Temporal-Distance Parameters

The average right and left temporal parameters are displayed below (Table 1). These parameters are averaged over 6 trials.

TABLE 1
TEMPORAL-DISTANCE PARAMETERS

Temporal-Distance Parameter	Reciprocal gait pattern
Cadence (Steps/min)	37
Walking Speed (m/s)	0.31
Stride Length (m)	1.02
Stance Duration (%)	55

B. Kinematics

Kinematics of the thorax, shoulder, elbow and wrist were further analyzed in the most significant sagittal plane. The mean and the standard deviation of the joint angles are shown below (Fig. 3). The kinematics of the right UE is reported, which are similar to the left UE.

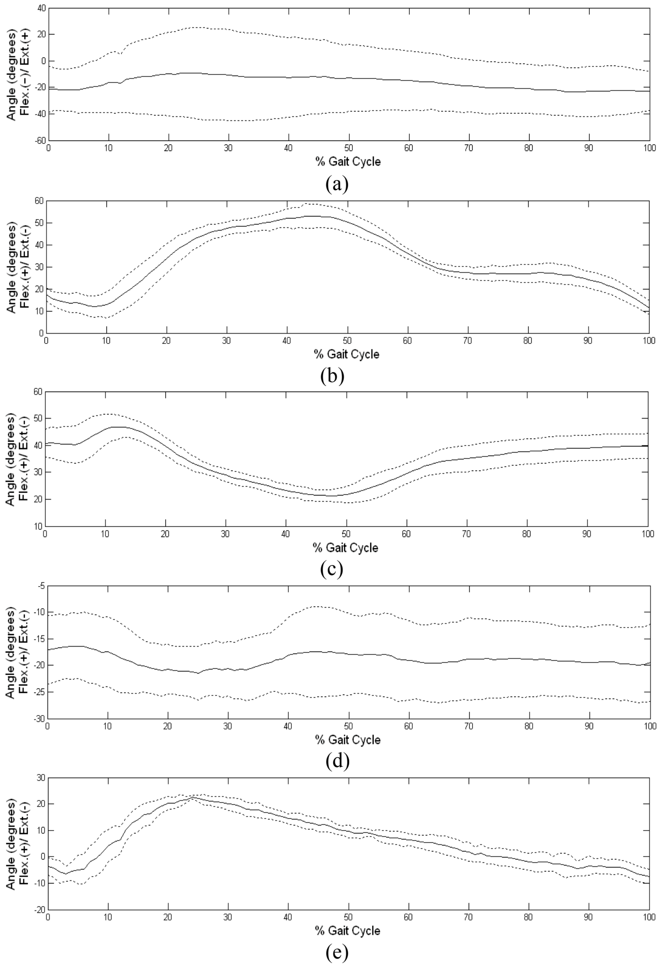

Fig. 3. Mean (± std. dev.) kinematics of the (a) thorax, (b) shoulder, (c) elbow (d) wrist and (e) crutch.

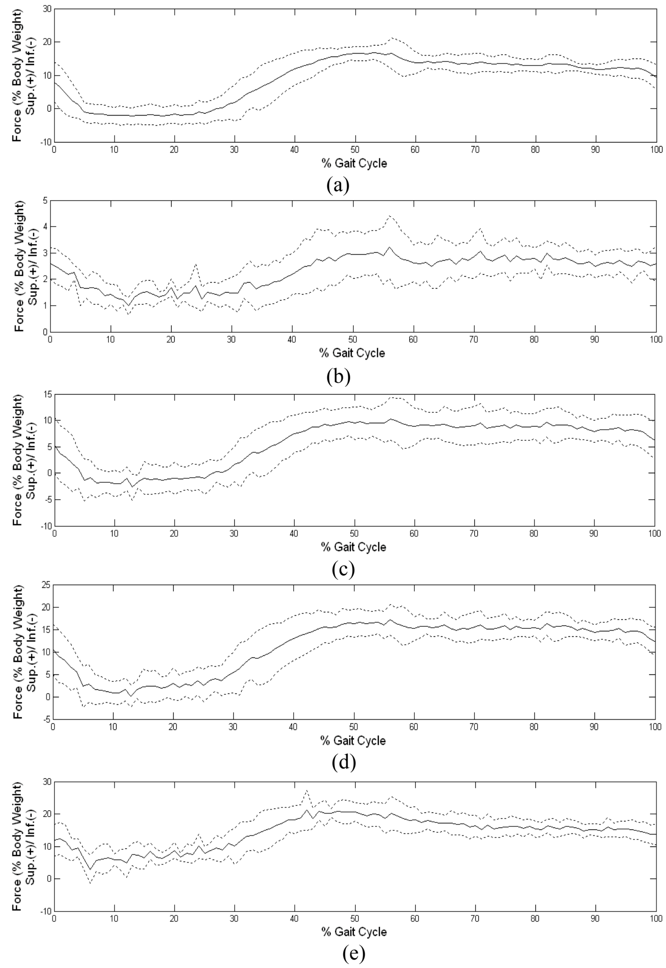

Fig. 5. Mean (± std. dev.) reaction forces seen at the (a)lower sensor, (b)upper sensor, (c)wrist, (d)elbow and (e)shoulder.

The thorax remained in flexion during the entire gait cycle but showed high variation. During crutch gait the shoulder joint demonstrated extension during 0-10%, flexion for 10-45% and remained in extension for the rest of the gait cycle. The elbow joints showed flexion for 0-10%, extension for 10-45% and remained in flexion for rest of the gait cycle. On the other hand, the wrist joint remained extended during the complete gait cycle. Also, the wrist joint motion showed high variation across the complete gait cycle. The average right and left range of motion for the thorax, shoulder, elbow and wrist as seen in the sagittal plane are displayed (Fig. 4).

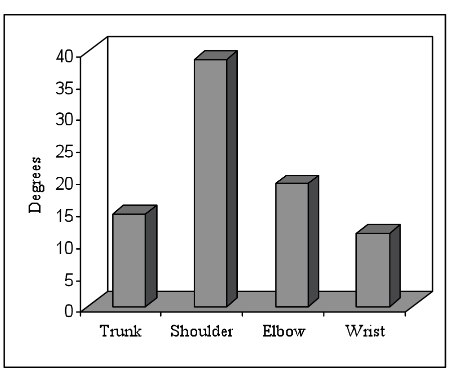

Fig. 4. Range of motion.

The joint reaction forces observed on the right upper extremity during crutch-assisted gait are displayed. The mean and the standard deviation of the forces were evaluated for 6 different trials (Fig. 5). Crutch loading was observed from 30-100% of the gait cycle. Forces established on the shoulder, elbow and wrist after 30% of gait cycle were maintained for the rest of the gait cycle. Peak forces evaluated on planes are reported (Table 2). Maximum peak forces were observed at the shoulder joint for which the resultant force was 33.57 % BW.

TABLE 2
PEAK FORCES OBSERVED AT THE SHOULDER, ELBOW, WRIST, UPPER AND LOWER CRUTCHES

Plane of force occurrence	Lower Crutch (% BW)	Upper Crutch (% BW)	Wrist (% BW)	Elbow (% BW)	Shoulder (% BW)
Ant.(+)/Post.(-)	-3.1	2.9	-15.9	-22.3	-14.2
Sup.(+)/Inf.(-)	16.8	3.2	10.3	17.1	21.1
Lat.(+)/Med.(-)	7.4	7.5	-5.8	-8.4	21.9

VI. DISCUSSION

The kinematic model was developed based on previous studies [2]-[8]. The kinematics of the UE showed similar morphologies with previous studies [2],[3],[5]. ISB recommendations were implemented for the coordinate system design of the upper arm and the forearm [7]. This kinematic model has been verified; hence it can be used for future studies to investigate upper extremity dynamics of children with OI.

We developed a novel crutch system to completely define the upper extremity kinetics. Previous studies which failed to evaluate kinetic involvement of the cuff during crutch-assisted gait were surpassed with the help of incorporating two six-axis load cells on each crutch. A previous study demonstrated the limitation of calculating the net wrist and the elbow forces due to the placement of the load cells at the tip of crutch causing increased inertial loading [2],[3]. This limitation was overcome by the incorporating the six-axis load cells just below the handle, as suggested by the study done by Requejo et al. [6].

This model, so far, has been validated for evaluating the kinematics and forces accurately. It was observed that the crutch loading occurred for 70% of the gait cycle. For long-term crutch usage, load on the UE could cause potential damage. Maximum forces were observed at the shoulder joint. Thus, excessive loading of the shoulder joint in case of long-term crutch users may cause additional shoulder pathologies. Inclusion of the involvement of the cuff forces would further helped evaluate the moment contribution at the cuff. This model serves as a basis for developing a complete kinetic model for evaluating the forces and the moments in the upper extremities.

A biomechanical model was developed to study and verify the novel instrumented crutch system. This model will aid in accurate assessment of UE loading, which may further help in preventing injuries in long-term crutch users. This 3D biomechanical model would thus, help us in evaluating complete kinematics and kinetics in children suffering from OI.

V. CONCLUSION

A 3D biomechanical model of the UE was developed and will be applied to children with OI. Quantitative assessment of UE joint dynamics may ultimately assist in activity modification, treatment planning, medical/surgical intervention and crutch prescription for OI patients.

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