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INSTRUMENTATION AND BIOMECHANICAL MODEL FOR KINEMATIC AND KINETIC ANALYSIS OF UPPER LIMBS DURING GAIT WITH CRUTCHES

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Abstract: The goal of this study was to develop a three-dimensional kinematic and kinetic model of the right upper extremity and a Lofstrand crutch in order to analyze joint displacements and loads during crutch-assisted gait. A Lofstrand crutch was instrumented with a six-component load cell to measure forces and moments at the crutch tip. The crutch and the right upper extremity of a subject were instrumented with markers to obtain kinematic data. A biomechanical model based on rigid bodies was implemented in biomechanical analysis software. To demonstrate the functionality of the model, a pilot test was conducted on one healthy individual during Lofstrand crutch-assisted gait. The shoulder extended during the support phase and flexed in the swing phase, the elbow flexed during the swing, and the wrist remained in extension throughout the cycle. In the shoulder and elbow joints, the predominant reaction forces were upward, whereas the internal force moments were flexion and extension, respectively. This tool will be useful when it comes to identifying risk factors for joint pathology associated with pattern gait, aid design or crutch overuse.

Keywords: biomechanics; biomechanical model; upper extremity; crutch-assisted gait.

Introduction

Restoring the capacity to walk in patients with incomplete spinal cord injury (SCI) is one of the main goals in rehabilitation programs. In the United States, it is estimated that there are currently 236,000 to 327,000 persons living with the consequences of SCI.[1] Furthermore, the estimated annual incidence of 40 per million population means that every year there are an additional 8,000-11,000 persons with SCI:[1] indeed, the percentage of the population with incomplete SCI increased from 44% in 1973 to 54% in 1990.[2,3] Nevertheless, with the use of proper bracing and walking aids, many such individuals become functional walkers.[4]

Shoulder pain is one of the most usual complaints among persons with SCI, due to the increased mechanical demands placed on the upper extremities for the purpose of achieving different types of displacements. Musculoskeletal conditions caused by excessive use of the upper extremities -rotator cuff overload in particular- are the most common cause of shoulder pain among subjects with spinal cord injury. In tasks involving upper-extremity weight bearing, such as wheelchair propulsion or transfers, the application of repetitive forces to the shoulder joint can lead to rotator cuff impingements or injuries[5]. During these weight-bearing actions, a force is transmitted via the forearm to the glenohumeral joint, elevating the head of the humerus. Failure to control this movement can lead to impingement of the tendons of the shoulder rotator cuff or other structures found in the subacromial space. In the case of manual wheelchair propulsion, these demands have been widely studied, in terms of both incidence and biomechanical characteristics. For instance, Sie et al.[6] and Dalyan et al.[7] estimated that: 59% of patients with spinal cord injury complained of some type of pain in the upper extremities; 30% presented with important pain which required medication, or experienced limitation or pain in the performance of two or more activities of daily living; 66% had symptoms related with carpal tunnel syndrome; 36% reported shoulder pain; 16% reported elbow pain; 13% reported wrist pain; and 11% reported hand pain.

In the case of persons with incomplete SCI who can walk, a high incidence of shoulder pain has also been reported, with prevalence figures of 39% at ages

31-45 years and 61% at ages over 45 years.[8] As in the case of manual wheelchair propulsion or transfers, weight bearing on the upper extremities during walking with aids has been implicated as a causative factor in developing shoulder pain.[6] The use of canes, crutches or wheelchairs entails related risks, such as degenerative arthritis of the hand and wrist, carpal tunnel syndrome, or ulnar neuropathy at the wrist. These repetitive actions exerting a load on the palm of the hand cause radial deviation and extension of the wrist, pressure on the carpal tunnel, and prolonged or repetitive contractions of the muscles of the hand and wrist. A number of studies have shown that actions associated with the use of crutches can cause an increase in carpal-tunnel pressure which could, in turn, lead to ischemia of the median nerve.[9] It is important to identify the factors that predispose persons to such injuries. Shoulder joint loads during assisted walking are not as well documented as those experienced in wheelchair propulsion. Biomechanical gait analysis with crutches yields pertinent information. Hence, there is a case for conducting biomechanical gait analysis of the upper extremities of persons who require crutches, canes or walkers to move about, and thereby ascertain their kinematic and dynamic gait patterns, i.e., the movements, forces and moments undergone by their upper limb joints. A comprehensive biomechanical model is required to identify potential mechanical sources of shoulder injury.[4] An inverse relationship has been reported between assistive device axial load and lower extremity strength.[10] Efforts to relate assistive device loads to demands placed on the upper limbs during ambulation have been limited. Previous endeavors have gone some way to examine upper extremity dynamics during Lofstrand crutch-assisted gait.[4,11,12] Requejo et al.[4] presented a system involving sensors positioned around the crutch handle. A more recent experiment was performed with a four-sensor crutch system for the purpose of directly measuring crutch-cuff kinetics and fully quantifying the dynamics of the wrist, in addition to those of the elbow and shoulder.[11]

The goal of this study was to develop a 3D kinematic and kinetic model of the right upper extremity when walking with a Lofstrand crutch, using a six-component strain-gage load cell located at the crutch tip to measure forces and moments of force, and a system of active markers to capture movement.

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Unlike previous experiments, which envisage one rigid body below the hand-handle interface, this model divides each part of the crutch into rigid bodies.[4,11] To verify the application of this biomechanical model, the kinematic and kinetic patterns registered by a single subject during Lofstrand crutch-assisted gait are described and discussed below.

Materials and methods

Instrumentation

For study purposes, a Guardian-model crutch (Sunrise Medical, Fresno, CA, USA) was instrumented with a six-degree-of-freedom MCW-6-1000 load cell (AMTI, Watertown, MA, USA) at the distal end. This sensor incorporates strain gages capable of measuring the components of the forces and moments received by the sensor (Fx, Fy, Fz, Mx, My, Mz), with respect to its effective origin and coordinate system. The manufacturer calibrated the sensor in-house, indicating the calibration constants of each variable, along with the location of the effective origin and orientation of the axes of the coordinate system. The signals generated by the dynamometer sensor were transmitted via a 10m-long cable to the pertinent amplifier, and after being amplified, to the motion-capture device, where they were recorded in synchronization with the positions of the markers.

Load-cell accuracy and precision were determined by simultaneously recording force data from the forceplate (Kistler, Winterthur, Switzerland) and crutch during crutch-assisted gait, with the crutch tip contacting the forceplate. Root-mean-square (RMS) error (i.e., the difference between the forceplate and the resultant three-component crutch force) was used to calculate sensor accuracy, and standard deviation was computed to assess precision.[4,13]

Tracking marker trajectories were recorded by two scanners, placed at either side of the walkway and fitted to an active-marker motion-capture system (Charnwood Dynamics Limited, Leicestershire, England).

Kinematic model

To implement the model of the right upper extremity, segments of the trunk, arm, forearm and wrist were defined.[4] Active markers were placed on different bony landmarks in order to compute segment movements and joint-center positions during the readings (Figure 1).

The center of the shoulder joint was deemed to be at the glenohumeral joint center (GHJC). This joint's position was calculated by rotating the plane formed by the markers of the greater tubercule of the right humerus (RTB), medial epicondyle (LE) and medial epicondyle of the humerus (ME), through an angle of 30º in an anticlockwise direction about the line formed by the RTB and LE markers. A straight line was created belonging to the rotated plane and parallel to the line formed by the LE and ME markers, which passed through the RTB marker. The GHJC was located on this latter line at a distance from the RTB equal to the mean distance between the LE and ME.

The elbow joint center (EJC) was located at the midpoint between the LE and ME markers; the wrist joint center (WRJC) was located at the midpoint between the markers of the ulnar styloid process (USP) and radial styloid process (RSP); and the third metacarpal joint center (3JC) was located at the point at which the third carpometacarpal (M3) marker projected onto the plane formed by the RSP, USP and lateral fifth metacarpal (LM5) markers.

In line with International Society of Biomechanics (ISB) guidelines:[14] the X axis was defined as anteroposterior, with the anterior direction being deemed positive; the Y axis was defined as longitudinal, with the upward direction being deemed positive; and the Z axis was defined as mediolateral, with the lateral direction being deemed positive.

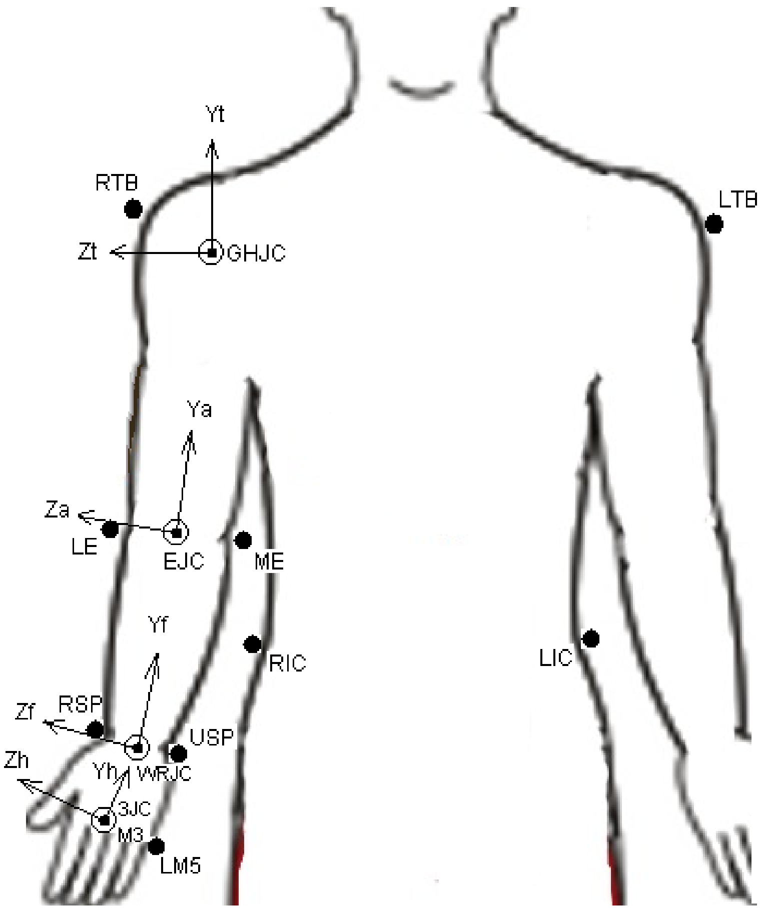
The Yt axis of the trunk segment was parallel to the line formed by the midpoint between the right iliac crest (RIC) and left iliac crest (LIC) markers, and the midpoint between the RTB marker and the marker of the greater tubercule of the left humerus (LTB); the Zt axis was perpendicular to the Yt axis, pointing toward the RTB marker; and the Xt axis was perpendicular to the Yt and Zt axes described above.

The Ya axis of the arm segment passed through the EJC and GHJC, the Za axis was perpendicular to the Ya axis, pointing toward the LE marker, and the Xa axis was perpendicular to the Ya and Za axes. The Yf axis of the forearm passed through the WRJC and EJC, the Zf axis was perpendicular to the Yf axis, pointing toward the RSP marker, and the Xf axis was perpendicular to the Yh and Zh axes of the forearm. The Yh axis of the hand passed through the 3JC and WRJC, the Zh axis was perpendicular to the Yh axis, pointing toward the LM5 marker, and the Xh axis was perpendicular to the Yh and Zh axes of the hand.

As a general rule, rotations about the Z, Y and X axes were assumed to be flexoextension, abduction/adduction, and internal/external rotation movements, respectively. To describe the angular movements between coordinate systems, Cardan displacements were calculated following the ZXY sequence; and to describe joint movement, the angular displacement of the distal segment coordinate system was shown relative to that of the proximal segment coordinate system. In view of the shoulder's anatomical complexity, in which a number of joints are involved, its joint movement was regarded as the displacement of the arm coordinate system with respect to the trunk coordinate system.

Three active markers were used to obtain crutch kinematics: two were placed on the anterior surface of the crutch shaft, distally (DC) and proximally just below the handle (PC), and a third was extended on a wand, 25 mm from the anterior end of the handle (HC). Four segments were considered, i.e., handle, shaft, load cell, and crutch tip (Figure 2). To define each of these segments, some landmarks, calculated on the basis of the position of the markers and the dimensions of the crutch, had to be identified. These landmarks corresponded to the ends of each of the above segments. The Y axis of each segment of the crutch was assumed to coincide with its longitudinal axis; the X axis was perpendicular to the Y axis, pointing toward the DC marker; and the Z axis of each segment was perpendicular to the Y and X axes described above.

Figure 1. Right upper extremity marker set and segment coordinate systems:[4] RTB, right greater tubercule; LTB, left greater tubercule; RIC, right iliac crest; LIC, left iliac crest; LE, lateral epicondyle; ME, medial epicondyle; RSP, radial styloid process; USP, ulnar styloid process; M3, third metacarpal; LM5, fifth metacarpal; HC, crutch handle marker; PC, proximal crutch marker; DC, distal crutch marker.



Kinetic model

Each segment was assumed to be a rigid body with its mass uniformly distributed,[4] with the trunk segment being deemed to be an elliptical cylinder, the arm and forearm segments being deemed to be cylinders, the hand segment being deemed to be a sphere, and each segment of the crutch being deemed to be a cylinder. Inverse dynamic Newton-Euler methodology (Figure 3) was used to calculate the joint-kinetic values between two consecutive rigid bodies, for both the upper extremity and crutch.[15] The joint-reaction forces, i.e., the forces exerted by the lower segment on the upper segment of any given joint, were calculated and referenced to the upper segment coordinate system. The internal moments of each joint were calculated with respect to the proximal segment coordinate system, following the right-hand rule (Figure 1). The kinetic data of the wrist were not considered valid, due to a lack of data on load values between forearm and crutch cuff.

The point of application of the forces of the hand on the crutch handle was assumed to be at the projection of the 3JC on the line joining the HC to the proximal end of the shaft. The forces and the moments measured by the load cell were deemed to be applied on the load-cell segment, at the sensor's effective origin, and with respect to its coordinate system.

The biomechanical model was implemented in biomechanical analysis software (C-Motion, Inc., Germantown, MB, USA) in order to calculate kinematic and kinetic joint data on the basis of the tracking marker trajectories, and data on the forces and moments measured by the load cell.

Figure 2. Crutch model. The crutch model was divided into handle, shaft, load cell, and tip segments, defined on the basis of the handle marker (HC), proximal marker (PC), and distal marker (DC)

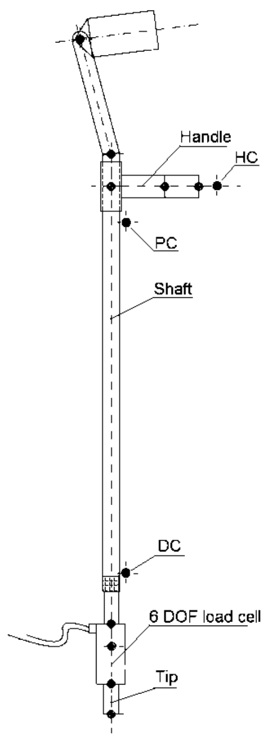


Figure 3. Proximal joint force and moment of any segment in the Global Coordinate System. mi = mass of segment i, ai = acceleration of segment i, n = number of distal segments connected in chain, q = number of external forces, Fj = applied external forces, p = number of external couples, τk applied external couples, Pj = vector from the application of the external force to the proximal joint, Ri = distance from the center of mass of each distal segment i to the proximal joint, Mi = inertial moment due to segment i.

$$\vec{F}_{proximal} = \sum_{i=1}^{n} m_i(\vec{a}_i + \vec{g}) + \sum_{i=1}^{q} \vec{F}_j$$

$$\vec{M}_{proximal} = \sum_{i=1}^{n} \left\{ \vec{M}_i + \vec{R}_i \times \left[m_i(\vec{a}_i + \vec{g}) \right] \right\} + \sum_{j=1}^{q} \left(\vec{P}_j \times \vec{F}_j \right) + \sum_{k=1}^{p} \vec{\tau}_k$$

Pilot test

To verify the application of this biomechanical model, a pilot study was conducted with the aid of a healthy subject. A healthy female, age 24 years, weight 68kg and height 1.66m was instrumented. The height of the instrumented crutch was adjusted so that the handle was at the level of the ulnar styloid apophysis, with the subject standing upright, and her arms hanging by her side and flexed 15°.[16,17] In order to perform the pilot test, the subject was required to walk at free speed along an 8 meters-long walkway, holding the crutch with the right hand and loading it simultaneously with left lower limb support. The subject was required to lean the trunk slightly to the right side when loading the crutch. The subject performed 10 trials to become familiar with the instrumentation. Then, she was asked to walk along the walkway 8 more trials. Five gait cycles were selected to be analyzed based on a good signal acquisition and a correct movement performance by the subject[12,13,18] .Informed consent was obtained from the subject as stipulated following project approval from the Clinical Research Ethics Committee and in accordance with the Declaration of the World Medical Association.

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Data-collection, -processing, and -analysis

Kinematic data were recorded at a frequency of 200Hz. and load cell signals at a frequency of 1000 Hz. The kinematic signals were interpolated using a polynomial least squares fitting procedure, and a second-order bidirectional Butterworth filter and cut-off frequency of 6 Hz was then applied.[12,19] In the case of the force and moment signals, a second-order, bidirectional, low-pass Butterworth filter and cut-off frequency of 7 Hz was applied.[20]

To obtain the results, the subject completed the walkway 5 times. From each test run, a cycle corresponding to that performed in the center of the walkway was selected. The cycle so selected was referred to as the "crutch cycle", which was in turn defined as all the data recorded between two consecutive initial contacts between crutch tip and ground. The data reported here thus refer to 5 crutch cycles. In each cycle, the instants at which the crutch left the ground were also defined, in order to differentiate the crutch's support phase from its swing phase. The initial contacts and lifting of the shaft tip were manually detected by visualization of the vertical force recorded by the sensor.

Once the cycles targeted by this study had been selected: the kinematic and kinetic joint data were normalized with respect to 100% of the crutch-cycle duration; the mean and standard deviation of the normalized variables of the 5 cycles were calculated for each moment of the cycle; and the spatial-temporal, kinematic and kinetic variables were obtained.

The spatial-temporal variables were calculated on the basis of the data on the position and time point of the landmark located at the distal end of the crutch-tip segment. The following parameter definitions were used: Cycle Length, the anteroposterior distance between two consecutive initial contacts between crutch tip and ground; Cycle Time, the time elapsed between the two initial contacts defined above; Stance Time, the time during which there is contact between crutch and ground; Swing Time, the time during which the crutch is not in contact with the ground; Speed, Cycle Length divided by Cycle Time; Cycles/minute, the number of cycles performed per minute calculated on the basis of Cycle Time; and % Stance

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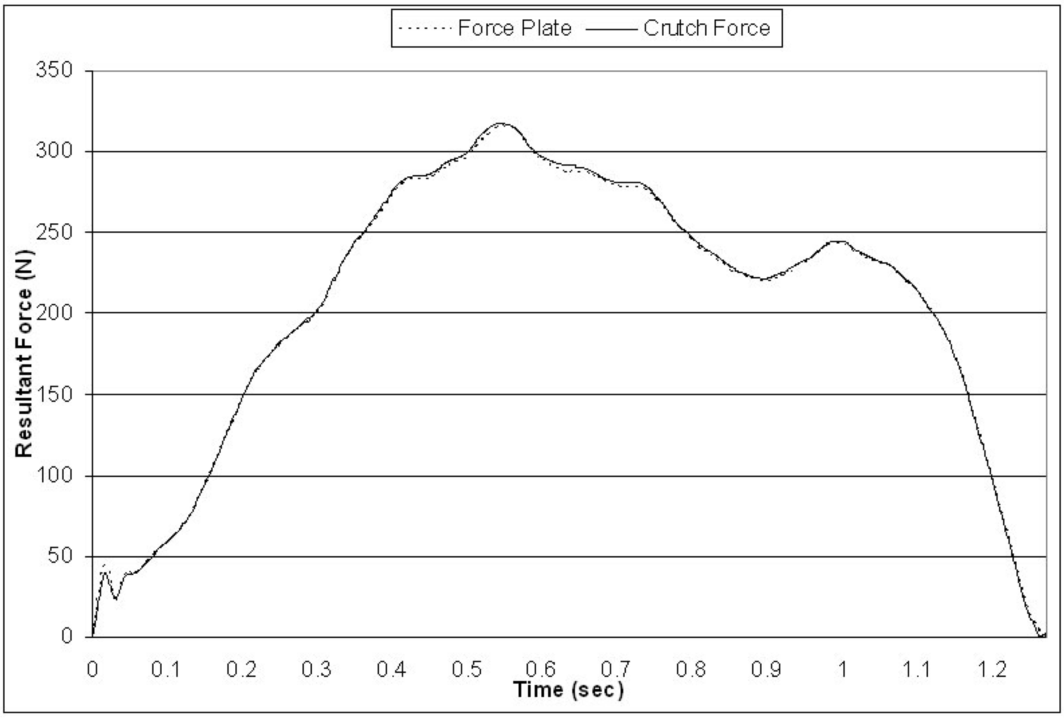
Phase, the percentage duration of the cycle during which there is contact between crutch and ground. In the case of the kinematic variables, the joint displacements of the shoulder, elbow and wrist were analyzed; and in the case of the kinetic variables, the forces and moments of force of the shoulder and elbow were obtained.

Results

Accuracy

The accuracy test yielded an RMS error of 2.14N, which represented 0.67% with respect to the maximum value of the resultant crutch force obtained (317.60N). In the precision test, the standard deviation of the error was 1.40N (Figure 4).

Figure 4. Resultant crutch (solid line) and forceplate (dashed line) forces simultaneously recorded.



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Spatial-temporal parameters

The mean velocity reached for the 5 cycles recorded was 0.35 (+/-0.03) m/s. The mean length of the right crutch cycle was 0.95 (+/-0.05) m. The remaining spatial-temporal parameters are shown in Table 1 below.

Table 1. Spatial-temporal crutch cycle data.

Parameter	Mean	Std. Dev.
Cycle Length (m)	0.95	0.05
Cycle Time (s)	2.7	0.2
Stance Time (s)	1.8	0.2
Swing Time (s)	0.9	0.0
Speed (m/s)	0.35	0.03
Cycles/minute	22.0	1.3
Stance Phase (%)	67.0	2.8

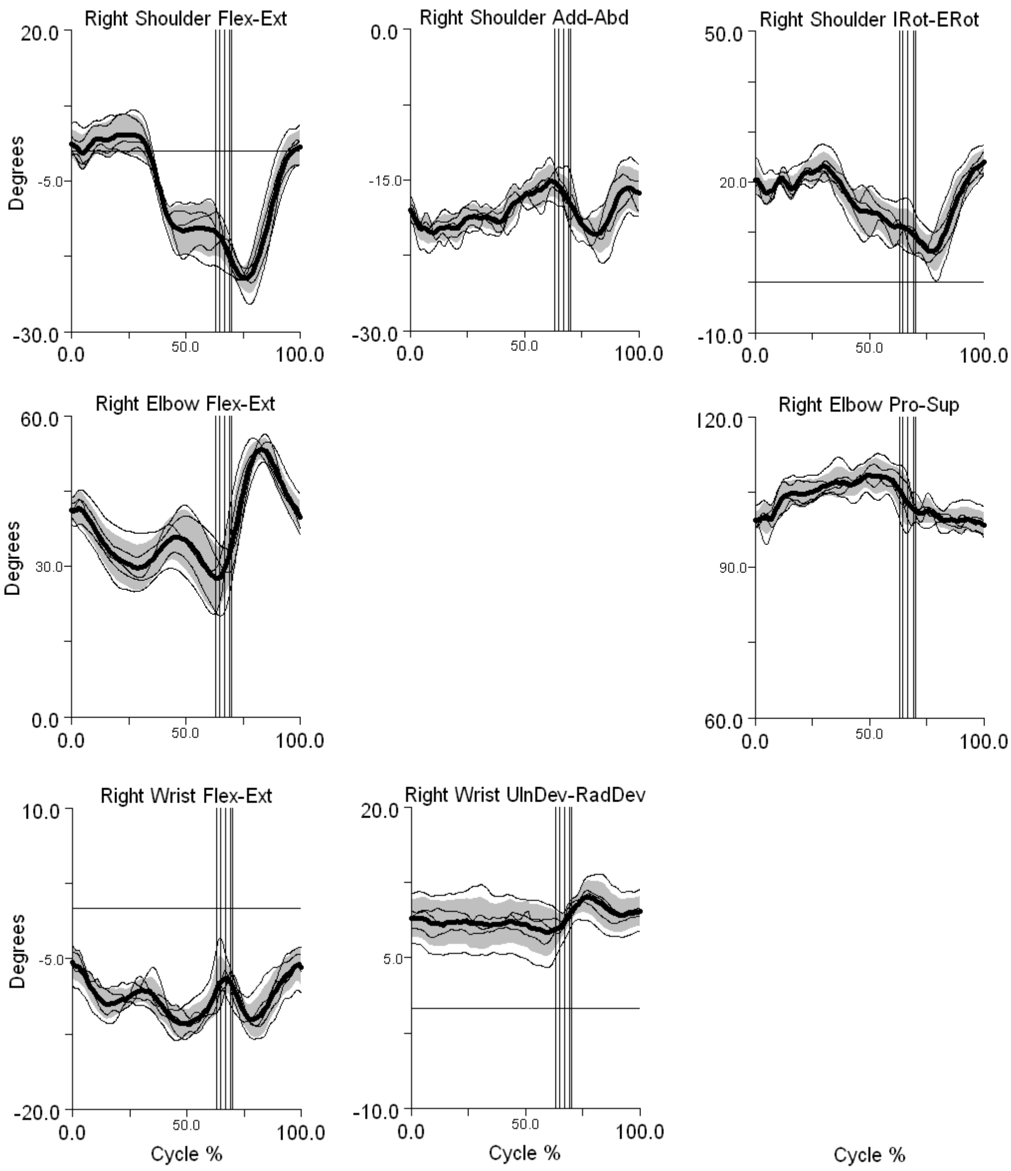
Kinematics of the upper extremity

Insofar as the shoulder was concerned, hardly any mobility was observed in the joint during the first half of the support phase (Figure 5). Once the loading response phase had passed, the shoulder extended. Then it flexed during the swing phase. In the case of the trunk, the humerus remained in abduction and internal rotation throughout the cycle.

The elbow remained flexed throughout the cycle, attaining a peak value in the swing phase (Figure 5). There was little variation in pronation across the cycle.

The wrist remained in extension throughout the cycle (Figure 4). During the support phase, it experienced slight consecutive flexion and extension movements, with flexion becoming more rapid and displaying a greater range of movement at the end of the phase. The wrist was observed to remain in ulnar deviation across the cycle.

Figure 5. Graphs depicting joint displacement: thin lines represent the 5 recorded graphs, solid lines the mean, and grey bands the mean +/- standard deviation of the 5 readings



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Joint kinetics in the upper extremity

Within the shoulder joint complex, the reaction force in the glenohumeral joint was observed to be mainly vertically upward, posterior and medial in the support phase (Figure 6). During this same phase, there was an internal flexor moment in this joint (Figure 7), which continued to be flexor in the swing phase, albeit with the minimum value required to advance the crutch in order for its tip to make the next point of contact with the ground.

As with the shoulder, the reaction force in the elbow joint was essentially vertically upward and posterior to the joint (Figure 6). As was to be expected, whereas the internal moment in the elbow was extensor during the support phase, it was mainly flexor in the swing phase (Figure 7).

Figure 6. Graphs depicting joint forces: thin lines represent the 5 recorded graphs, thick lines the mean, and grey bands the mean +/- standard deviation of the 5 readings.

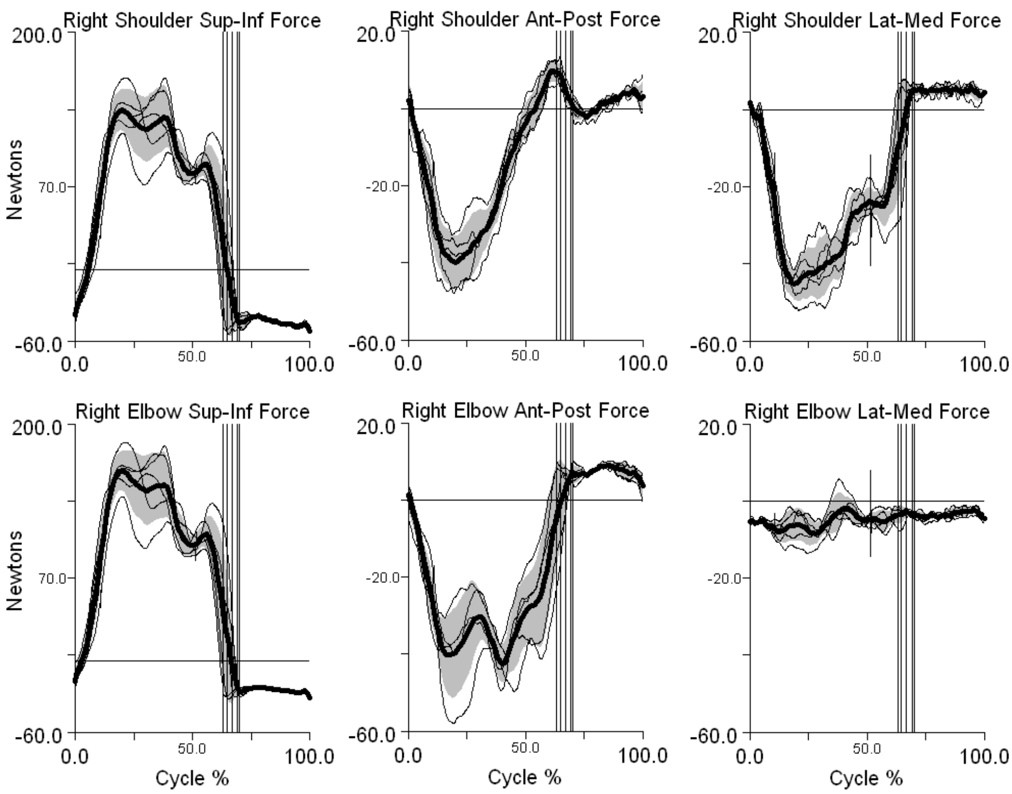
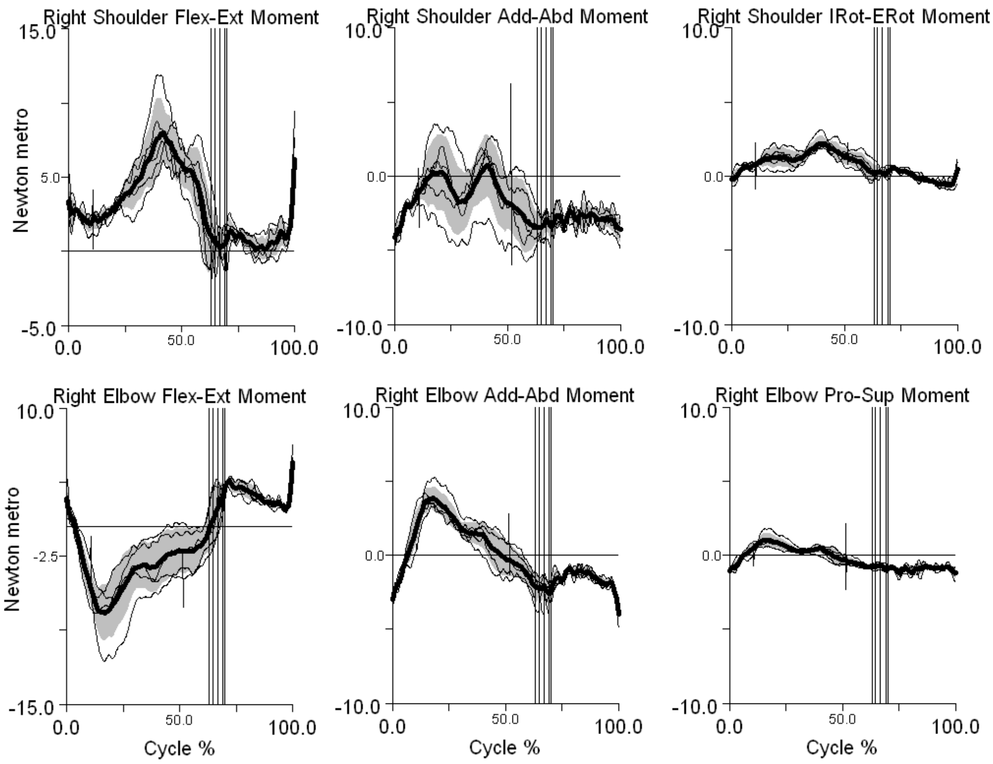


Figure 7. Graphs depicting joint moments: thin lines represent the 5 recorded graphs, solid lines the mean, and grey bands the mean +/- standard deviation of the 5 readings.



Discussion

The paper describes a biomechanical model, as well as the instrumentation and methodology, for the study of upper extremity joint kinematics and kinetics during crutch-assisted gait. Compared to previous works,[4,11,12,22,25] here we have presented a three-dimensional kinematic and kinetic model that incorporates a full description of forces application points and the rigid body geometry, and meets ISB joint coordinate system guidelines[15]. Studies following the kinetic model presented in this work, and that adhere the ISB standards will be reproducible and comparable. The functioning of the model was demonstrated by conducting a pilot test with a healthy subject.

The inclusion of a force sensor placed at the tip of the crutch enabled accurate estimation of the major net joint forces and moments of the upper extremities. The data obtained by comparing the resultant crutch and forceplate forces confirmed the accuracy of the crutch instrumentation, as previously reported.[4,13]

For the purposes of this study an active-marker motion capture system was used, whereas other authors have used systems based on passive markers.[11,12,21,22] While the placement of the crutch markers, namely, one at the anterior end of the handle and two on the shaft, coincided with that used in previous experiments,[4,12] other earlier studies used two markers placed on the anterior and posterior positions of the sensor, which was itself located at the distal end of the crutch, and a further two markers on the shaft, one in the middle and another at the height of the handle.[18,23] In this study, the force sensor was situated on the distal part of the crutch, in line with previous models.[18,23,24] Other authors have included a load cell just below the handle to reduce inertial loading effects.[4,11,12] The kinetic model proposed in this study assumes the crutch to be an element formed by 3 rigid bodies, i.e., handle, shaft and force sensor, whereas others regard it as a single rigid body lying between the handle and the center of the sensor.[4,11,18]

This study reports the results registered in respect of a healthy subject who performed a three-point gait pattern, using the crutch with the upper right extremity. Other authors have studied subjects with different medical conditions, using reciprocal four-point or swing-through gait patterns, with two crutches, a walker or even a cane.[4,11,12,18,20-22,25]

With regard to joint kinematics, after the moment corresponding to the loading response phase, the shoulder extends to enable the trunk to advance vis-à-vis the crutch. Thereafter, the shoulder flexes in the swing to move the crutch tip forward, so as to ensure that the following initial contact is made at approximately the height of the foot of the contralateral extremity.

During approximately 40% of the cycle, the elbow has to flex to avoid raising the body's center of gravity when the tip of the shaft passes at the height of the body. The elbow then extends to ensure that the crutch maintains contact with the ground and continues transferring the weight of the upper part of the body to the ground until the crutch is lifted. In the swing phase, the elbow must flex to raise the shaft tip and prevent it from coming into contact with the ground, and then extends once again to prepare for the following initial contact. As regards the adduction-abduction movement of the elbow, this is a passive rotation and is not reported.[15] The elbow's peak flexion in the swing phase proved of greater intensity in this than in a previous study.[4] Its value depends on the anthropometric characteristics of the user, cycle duration and crutch length.

Overall, the kinetic data obtained in this study were of the same order of magnitude as those reported by other similar studies.[4,11,18,22,23] During the support phase, the vertical reaction forces in the shoulder and elbow are similar in form but the magnitude of the forces observed in the elbow is superior to that of the shoulder, a finding in line with that of other authors.[4,11] During the swing phase, in contrast, this situation becomes inverted, with the vertically downward force being greater in the shoulder than in the elbow.

This effect could be due to the fact that, in the support phase, the weight of the lower extremity is transmitted to the ground, and the shoulder has to bear part of the trunk's weight which is loaded onto the crutch, whilst the elbow during the support phase has to bear the loading of the trunk plus the weight of the arm. In the swing phase, in contrast, the shoulder has to control the full weight of the upper extremity plus the crutch, whereas the elbow has to control the same weight, less that of the arm. As observed in previous studies, during the support phase there is a predominance of forces exerted in a superior, posterior and lateral direction in the shoulder.[4,12]

It is noteworthy that, for around 40% of the cycle, in the second half of the support phase there is an increase in the value of the vertically upward force and flexor moment in the shoulder while this joint rotates toward extension. This situation might entail an increase in the load borne by the shoulder and a moment of risk of joint overuse.

The predominance of the flexor moment observed in the shoulder during the support phase coincides with that found by other authors who have examined reciprocal gait.[11,18,25] Whereas studies of swing-through gait showed a predominance of internal adductor moments in the shoulder during the support phase,[25] in reciprocal gait internal abductor moment values were detected in the shoulder during the support phase.[4] This difference between adductor and abductor moments in the shoulder in swing-through and reciprocal gait was subsequently confirmed.[18] In this study, which examined reciprocal gait with a crutch, the internal moments in the anteroposterior axis of the shoulder joint were in line with those described above, tending to values of small magnitude but with a trend toward abduction.

As reported by previous studies,[4,12] the elbow shows a predominance of superior and posterior forces during the support phase. In the loading response phase, there is a peak force directed vertically upward in the elbow, together with a posterior peak force and an increase in the extensor moment at this level which, as in the case of the shoulder, coincides with a displacement toward extension. This circumstance may also amount to the joint being compromised. This study observed an extensor moment of the elbow during the support phase, a finding in line with other studies that have likewise addressed reciprocal gait.[4,11,18]

Some limitations must be mentioned, however. First, there was no information about the load existing between the crutch cuff and the forearm, with the result that no kinetic wrist data were reported. Similarly, the study was performed on a single upper extremity. Accordingly, the model should be extended to include both of the upper limbs and two crutches, and the instrumentation should be modified to ascertain the kinetics of the carpus. Once these conditions have been met, the respective gait patterns of subjects with different medical conditions could then be analyzed

Conclusions

The procedure described in this work provides the basis for the kinematic and kinetic analysis of the upper extremity joints during crutch-assisted gait. The present model is described in detail, including all force application points and rigid body geometries, and meets ISB guidelines, warranting precise comparison with future analyses. The results of a pilot test demonstrates the functionality of the trial configuration and its potential application for the clinical practice. The results of the present wil be used to initiate a study of different gait patterns using crutches (reciprocal versus swing trough) in order to evaluate which offer a higher risk to develop overuse upper limb joint pathology.

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