Kinematic Modeling of the Shoulder Complex in Tetraplegia

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In comparison to other joints in the human body, the shoulder complex is particularly reliant on the coordination of active muscle forces to generate both movement and stability during activities using the upper extremities. The resultant imbalance of muscle forces across the shoulder, coupled with the increased reliance on the shoulder for functional mobility, puts the individual with tetraplegia at great risk for developing shoulder pathology. The ability to quantify the movement of the shoulder, and in particular the sequence of shoulder complex movement components within functional tasks, can provide information to better inform clinical and surgical decision making. In this article, we will discuss the impact of tetraplegia on shoulder biomechanics and function, provide an overview of general principles and current status of kinematic modeling of the shoulder complex, and describe emerging applications of quantitative motion analysis of the shoulder complex. **Key words:** *biomechanics, kinematic modeling, shoulder, tetraplegia*

The shoulder complex consists of four articulations that produce functional movement through coordinated muscle activity acting across several joints. Movements in all planes, both joint rotations and translations, are supported by the skeletal architecture in combination with muscles. The shoulder girdle is dependent on 31 muscles and muscle parts to provide functions related to both mobility and stability. Unlike lower extremity muscles, most shoulder complex muscles have large attachment sites and span multiple joints. In addition, many of the shoulder complex muscles, such as the deltoid, are comprised of muscle parts that contract independently and provide different forces across the joints. The closed chain configuration of the thorax, clavicle, and scapula along with various ligaments and the shape of the articular surfaces provide stability. Movements are typically described in Cartesian coordinates, in which three axes comprise the coordinate systems. This allows for description of six possible motions or degrees of freedom at each joint, rotation around each of the three axes, and translations along each axes.

The shoulder complex can be modeled by four segments: the thorax, scapula, clavicle, and humerus. The only skeletal articulation

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the upper extremity (UE) has with the trunk is through the sternoclavicular (SC) joint. The SC joint has three degrees of freedom, allowing for rotation around three axes. The acromioclavicular (AC) joint, the articulation between the clavicle and scapula, has three rotational degrees of freedom as well. Coupled action of muscles acting at both the AC and SC joints further constrain motion at these joints. The scapulothoracic (ST) articulation occurs through muscle contact; this is an atypical joint in that the movement constraints are primarily determined by muscle action rather than bony configuration. The ST joint has five degrees of freedom: three rotational and two gliding planar movements. The possible sixth degree of freedom that would be provided by a translational glide away from the thorax is not allowed. The glenohumeral (GH) joint, what is commonly called the shoulder joint, is the fourth articulation in the shoulder complex. This joint is typically considered to have three rotational degrees of freedom, with translation of the humeral head in the glenoid fossa ignored. The glenohumeral joint is inherently unstable and relatively unconstrained by bony architecture providing the greatest amount of mobility of any joint in the body. Passive and active stabilizers act to hold the humeral head steady. The joint geometry provides minimal stabilization, although the limited joint volume, glenoid labrum, and ligamentous restraints provide additional stabilization. The majority of the stability, and mobility, of the GH joint is contributed by compression of joint surfaces, dynamic ligamentous tension, and neuromuscular control.

The shoulder complex's most common role in functional tasks is to position the humerus to support use of the hand for manipulation in space. Although little movement occurs at the SC and AC joints, both the ST and GH joints provide considerable movement. Function and performance of activities of daily life are greatly impacted by impaired movement at the ST and GH joints. In addition, the effectiveness of the GH joint in positioning the hand in space is enhanced by the stability provided by the ST joint. When the scapula is well stabilized on the thorax, the muscles crossing and acting on the GH are optimized as movers. During large movements, simultaneous motions of the scapula on the thoracic wall and of the GH joint occur and are referred to as *scapulohumeral rhythm*. The development of this scapulohumeral rhythm has been studied in adults as well as in typical children and children with brachial plexus injuries.[1–3]

Impact of Tetraplegia on Function and Shoulder Biomechanics

The condition of tetraplegia occurs with a spinal cord injury (SCI) to the low cervical levels of the spinal cord. Tetraplegia refers to weakness or paralysis of all four extremities, often with trunk involvement as well. The degree of sensory and motor loss may vary with none (complete) or some residual sensory or motor function remaining at levels below the SCI level. Individuals with injuries at cervical level 5 (C5) may have remaining function in deltoid, biceps, brachialis, rhomboids, and serratus anterior (partial). Movements that remain within the shoulder complex are ST abduction/adduction, GH flexion, abduction and extension, elbow flexion, and forearm supination. At C6, the following muscles may also retain innervation: clavicular pectoralis, serratus anterior, latissimus dorsi, and extensor carpi longus. These muscles provide additional scapular protraction and some shoulder horizontal adduction, forearm supination, and radial wrist extension. C7 injuries result in limited grasp and release, but elbow extension and wrist and finger movements become possible. In addition, at C7, the latissimus dorsi, sternal pectoralis, and triceps are innervated. Elbow extension significantly improves functional capabilities and may provide enough UE function for manual wheelchair propulsion.

There are several critical impairments resulting from tetraplegia that impact shoulder movement and biomechanics. First, decreased strength of the shoulder's prime movers clearly impacts the ability to use active shoulder movement to position the arm and hand in space. In addition, because shoulder stability is in large part reliant on muscle activity, decreased strength in muscles will impact the coordinative interplay of muscles acting as stabilizers at the ST and GH joints. Such altered coordination imposes atypical patterns of use and forces on the shoulder complex. A normally developed shoulder architecture was not meant to support the extra stability and mobility needs of the tetraplegic shoulder. The long-term impact of such atypical use often results in chronic pain syndromes. Shoulder pain is commonly reported in the majority of individuals with tetraplegia.[4]

In individuals with tetraplegia, the shoulder now also bears additional force and weight in terms of transfers, wheelchair propulsion, or ambulation with assistive devices. These activities pose additional workload on the shoulder, a joint primarily designed to support large movements rather than generate sufficient force to perform such activities or bear weight. In individuals with C5 or lower level spinal injuries, the shoulder is used for manual wheelchair propulsion. Wheelchair propulsion becomes the most repetitive activity performed by the shoulder and often occurs at higher work rates than ergonomic guidelines suggest as limits within work settings. Pain and limited shoulder motion are quite commonly reported as long-term outcomes of tetraplegia. Decreased stability of the GH joint brought on by weakness of the rotator cuff and biceps tendon also acts to overload the passive restraints, and impingement of the rotator cuff by the coracoacromial arc may occur. The most common symptom of impingement is painful abduction and restricted range of motion.

The use of quantitative motion analysis may better document the ranges of movement used and the patterns associated with pain or poorer functional outcomes. Kinematic modeling when combined with electromyography or kinetic data can provide a fairly complete picture of how the shoulder performs and responds from a mechanical perspective to interventions designed to improve function.

Biomechanical Modeling of the Shoulder

Motion capture in combination with kinematic and inverse modeling is one of the most common techniques used to quantify shoulder complex motions. Typically markers, called *technical markers*, are placed on the surface of the skin, and their movement is the moving segment to represent true joint displacements. The International Society of Biomechanics (ISB) has recently published definitions of joint coordinate systems for reporting UE motions.[5] The humeral segment is described based on the humeral joint center of rotation. Previous studies have described the GH joint center based on visual inspection, offset values, or as a function of the trunk width.[6] The ISB recommends using either a regression analysis or instantaneous helical axis (IHA) of rotation, with the IHA being preferred. However the IHA can present problems when used in persons with pathological abnormality at the GH joint. Gamage and Lasenby[7] describe the use of a least squares fit method to calculate the center of rotation for a true ball and socket joint. The aim of this article was to describe a video-based model that was developed to more completely characterize movement in the UE; the technique provides a picture of scapular motion and involvement in UE tasks.

Rau et al.[8] have illustrated the difficulty of translating knowledge gained from lower extremity analysis to UE analysis, specifically the larger ranges of motion (e.g., shoulder versus hip), and the resulting ambiguity regarding selection of rotation sequence for angular decomposition. Rab et al.[6] subsequently described a UE model comprised of 10 segments that used anatomical offsets based on a small set of measures from a typical adult.

The model developed by Rab et al. has been used to study motion in children with typical development and with brachial plexus birth palsy.[9] The models mentioned do not provide a comprehensive analysis of the scapula segment or a description of scapulothoracic movement using a video-based marker system. For the study of scapular kinematics, electromagnetic systems seem to be preferred to video-based systems, and numerous studies have been published using such a system.[2,10] No concrete consensus has been reached when comparing the two systems or describing a video-based scapular model. However because both rely on tracking markers placed on the skin and similar underlying kinematic models, the hardware system used to detect marker movement would be expected to have minimal effect.

In kinematic modeling of the shoulder complex, data collection and processing must adequately track segment rotations of the thorax relative to global coordinate system, humerus relative to thorax coordinate system, and the ulna and radius relative to humerus. Of note is that several joints are bypassed in measuring the relative motion of the humerus relative to the thorax. As techniques improve in tracking the motion of the scapula and clavicle using skin-mounted markers, these joints will routinely be included in shoulder complex analyses. Estimation of clavicular rotations relative to the thorax is often used, as it is quite difficult to measure AC rotations. Such estimations often minimize rotation at the AC joint and ignore rotation about the long axis of the clavicle. By far one of the largest issues in modeling of the shoulder complex is the measurement of the ST joint movement. This difficulty arises as direct measurements of the scapula relative to the trunk are complicated by few bony landmarks and significant relative motion between the scapula and the overlying skin to which markers are attached. The GH joint center is often determined by anthropometric-based regression equations from the literature or from an individual basis or the use of dynamic joint centering trials at the start of movement capture.

One other UE kinematic model issue is the definition of forearm axes. Although beyond the scope of this article, the use of standardized humeral reference frames and single axis forearms results in different simulated predictions of motion relative to actual measures of forearm kinematics. The Rab model uses the long axis of the forearm for elbow flexion/extension calculations but then shifts the axis to a line that goes from the center of the elbow joint to the distal ulnar styloid to let the radius rotate around the ulna for pronation/supination. The take-home point is that a model using two bones rather than one bone for the forearm may need to be considered. Active discussion and work by several groups continue to improve our kinematic modeling of the forearm.

In a kinematic model, the different segment coordinate systems, or movement of a given segment in time, require the use of coordinate system transformations. Euler angles are most commonly used, and the order of rotation is still a topic of debate. Many have proposed using different decomposition sequences depending on the UE task to be analyzed (i.e., is the task mostly in the sagittal vs. horizontal plane). Still, such decomposed graphs can be difficult for the clinician to interpret if the motion being studied lies between clinical planes of motion. One can minimize this effect by referencing the decomposition graphs generated from a normal population performing the same task. Recently Rab's group[11] demonstrated the equivalence of the Euler rotation sequence and the Pearl method.[12] The Pearl method provides a globally based graphic presentation of arm position. A similar spherical system has been discussed by Cheng[13–15] and used by Doorenbosch[16] to illustrate shoulder motion.

Current Applications of Kinematic Shoulder-Complex Models

Several models do exist to support future UE work, but they have several clear weaknesses. These can be classified as weaknesses in determining the GH joint center, modeling the full number of degrees of freedom, and quantifying ST joint motion. Issues also remain in approaches that better account for skin movement artifact and that model forearm motion as well as hand and finger motion within functional movement tasks. In the following sections, specific approaches to shoulder modeling address the issues with GH models appropriate for clinical work, scapulothoracic tracking, a model that accommodates six degrees of freedom, and a method used to link the more complex motion analyses with a functional measure. Each of these applications represents how kinematic models are evolving, but it should be noted that their application to individuals with tetraplegia are limited and not validated in this population. Although many advances have been made in understanding the biomechanics of wheelchair propulsion,[17] these are not discussed in this article.

A. I. duPont Hospital for Children's Upper Extremity Model for Clinical Motion Analysis

We refined a model originally developed by James Richards. This model addressed the lack of a functional fit for the GH joint and for the elbow axis, with assignment of flexion/extension and abduction/adduction to the wrist and supination/pronation to the forearm. This model also recognized that a single method of calculating shoulder angles yields results that may not make clinical sense for all possible movements.

To analyze UE motion, we utilized a series of nine functional motions (bilateral motions: walking, running, ball raise; unilateral motions: hand to mouth, hand to neck, hand to back, abduction, external rotation, and supination). Three markers defined the orientation of the head: two placed on a baseball cap (head front: center bill, head top at cap button) and the marker at C7. The pelvis was determined by the markers at the ASIS and PSIS (i.e., Cleveland Clinic or Helen Hayes marker sets). The trunk coordinate system was established by defining the long axis to be from the middle of markers at C7 and a marker just inferior to the jugular notch of the sternum (cervical center) and the middle of markers on T8 and a marker at the xiphoid process (thoracic center). Crossing the longitudinal vector with a vector from the thoracic segment center to T8 defined the medial/lateral (ML) axis. The anterior/posterior (AP) direction of the trunk resulted from the cross product of the trunk longitudinal and ML vectors. A pseudo-scapular segment was constructed from the cervical center and the thoracic center and the marker at the acromion process. The motion follows the vector from the cervical center to the acromion process in the plane of all three markers. Thus, we can measure rotation in the frontal and transverse planes but not rotation about the long axis of the clavicle. The GH joint center is fixed in the scapular segment and is calculated by a spherical fit[18] of the motion of the elbow joint center (middle of humeral epicondyles). The humerus was defined by the GH joint center, the elbow joint center, and the average vector produced from the cross product of the long axis of the humerus and the forearm as the elbow flexes and extends. The long axis of the forearm was defined by the elbow joint center and the wrist joint center that was midway between the styloid processes of the ulna and the radius. Crossing the long axis of the forearm with a vector through the styloid processes defined the AP axis of the forearm. In this way, pronation/supination was defined to be about the long axis of the forearm. The hand was defined as a vector from the bisection of the styloid process to a marker at the head of the third metacarpal. This model tracked wrist movement as flexion/extension and radial/ulnar deviation.

From our experience, all methods of calculating shoulder motion were clinically inadequate. That is, the use of Euler/Cardan angles, helical rotations, and quinterions all accurately map one coordinate system to another; however, the resulting decompositions do not necessarily describe the motions of the shoulder in a clinically meaningful way (**Figure 1**). Another method using "instantaneous" Euler angles initially looked promising but was subject to accumulation of small errors. A system that describes shoulder position based the location of the elbow on a sphere that is graduated with respect to elevation (latitude) and plane of elevation (longitude) is intuitive but does not correspond to traditional clinical terms of abduction/adduction and flexion/extension, and it is difficult to define internal/external rotation. Our solution was to utilize different Euler rotation orders for different movements. The third Euler rotation was always internal/external rotation, and the first Euler rotation was assigned to flexion/extension or abduction/adduction depending on which motion was dominant.

From comparison of curves from the two decomposition sequences, we determined that the order of Euler rotations that best fit the motions we studied were flexion, abduction, and internal rotation for walking, running, and bilateral ball raise, and abduction, flexion, and internal rotation for abduction, hand to mouth, neck, and back. For both

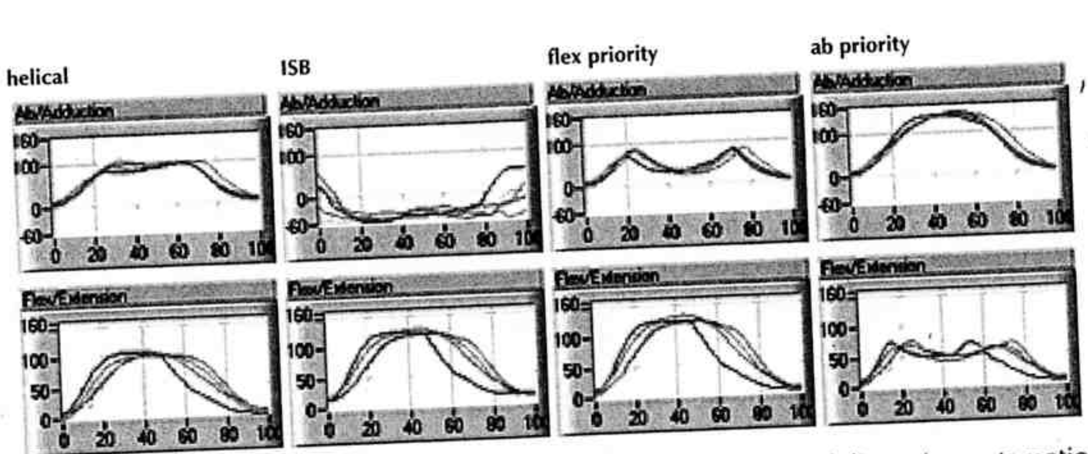

Figure 1. Relevant graphs of abduction and flexing at the shoulder. Each line represents motion from hands at side to hands up above head as far as possible and back to side. First row represents abduction/adduction component of decomposition by the method listed in the column heading of a normal person starting with the arms and hands at the sides and moving to hand overhead by straight abduction. Then returning to the hands and arms at the sides of the body. The participant was to keep the arms in the frontal plane as much as possible throughout the movement Second row represents flexion/extension component of decomposition by the method listed in the column heading of a normal person starting with the arms and hands at the sides and moving to hand overhead by straight flexion. The participant was to move the arms in a sagittal plane. ISB = International Society of Biomechanics.

supination and external rotation motions, we used an abduction priority; however using flexion as the first rotation in the sequence yields similar results.

This model has been actively used at A. I. duPont Hospital for Children to clinically study individuals with brachial plexus palsy or hemiplegic cerebral palsy (CP), and we are currently looking to quantify the mechanics of intubation techniques. To analyze the clinical data, we utilize normative data of the aforementioned functional tasks processed using the rotation order as assigned previously as the standard for comparison. If abnormal movement/posturing is noted so that the primary movement plane differs from norms, the Euler angle sequence can be changed. At present, there is no adequate method to track the scapula using skin-mounted markers. The spherical fit method[18] has been validated for the hip but not for the shoulder joint. We hope to eliminate these problems in the near future.

Richmond Six Degrees of Freedom UE Model

Many users of motion analysis systems simplify their kinematic models and allow only certain motions by restricting the degrees of freedom. However, more robust models that provide the full six degrees of freedom may better capture true anatomic joint motions leading to more exact and accurate quantification of UE movement. In this section, the Richmond UE kinematic model will be described that uses six degree of freedom modeling and Visual3D software (C-Motion, Inc., Rockville, MD). Although six degrees of freedom are modeled and measured, we typically do not report the translations in our clinical evaluations.

To measure UE movement, we follow International Standard Orthopaedic Measures[19] so that upper limb kinematics correspond with known goniometric standards. The comprehensive upper limb evaluations that we perform include a detailed musculoskeletal and neuromuscular examination and functional testing specific to each diagnosis and the clinical question to be answered along with kinematics, for the purpose of enhanced clinical decision making. For example, in the individual with tetraplegia, additional movements may be performed during kinematic data collection and videotaping including specific antigravity movements with the shoulder complex stabilized. In some cases, we use the standardized Jebsen-Taylor Hand Function Test.[20] After 10 years of performing gait analysis evaluations using the six degree of freedom Move3D software developed at the National Institutes of Health, we began using the updated modeling tool, Visual3D, for UE motion processing. To date, we have used this kinematic model for the past 4.5 years to quantify upper limb movement in nearly 100 individuals with brachial plexus birth palsy or injury, hemiplegic or quadriplegic CP, tetraplegia, arthrogryposis, and a number of other diagnoses affecting UE movement. Early correlation work has been performed.[21]

We use a head, arm, and trunk marker set with a combination of static and tracking markers. All static markers are removed after the static data capture prior to dynamic data collection. Markers on the head and trunk track three-dimensional head relative to trunk movements and global trunk movements. Tracking markers for each arm segment consist of very distal placements of T-shaped triad plates consisting of three noncollinear markers, which track movements of the humerus, forearm, and hand. Rigorous testing in our laboratory compared the two methods of standard skin-based single markers placed more proximally over the skin versus distal positioning of triad plates to measure true axial motion of the bones beneath. We found that these distally placed triad plates permit the most accurate measurement of axial rotations, when compared with more proximally placed single skin-mounted markers.

In our model, the x-axis traverses the medial/lateral direction perpendicular to the sagittal plane, with rotation around this axis referred to as flexion/extension for the shoulder, elbow, and wrist joints. Similarly, the y-axis extends in the anterior/posterior direction, perpendicular to the frontal plane, with shoulder abduction/adduction and wrist radial/ulnar deviation occurring around this axis. The z-axis is oriented vertically, perpendicular to the transverse or horizontal plane allowing for the axial rotations of shoulder external/internal rotation and forearm pronation/supination.

Head and cervical spine movements measured include flexion/extension, lateral side bend, and rotation. Head marker placements consist of a static marker placed on the tip of the nose and three tracking markers placed at midline on the forehead just superior to the brow line and immediately anterior to the tragus at the external auditory meatus of the ear on both right and left sides.

Because clinical management of upper limb dysfunction aims to restore (or improve) movements of the entire shoulder complex, and due to the difficulty separating GH from ST motion, we measure the sum total of shoulder complex movement (i.e., humerus relative to the thorax). We recognize that the scapula and humerus move together in concert, but due to problems measuring pathologic scapular motion with skin-mounted markers, the final "output" of thoracohumeral movement is used, as these motions correspond with goniometric standards and are understood by the clinicians who use the data we collect. Therefore, shoulder movements measured include thoracohumeral flexion/extension, abduction/adduction, and external/internal rotation. We define the shoulder using four static markers placed superiorly on each acromion and over the anterior, middle, and posterior deltoid on each side to define the shoulder joint center. The humeral tracking triad is attached via Coban™ to the distal humerus bridging the femoral epicondyles posteriorly as well as the olecranon just distal to the epicondyles. This triad tracks humeral segment flexion/extension, abduction/adduction, and internal/external rotation relative to the trunk.

For data collection of shoulder kinematics, we have each individual perform the five Mallet scale movements[22] in either a sitting position or standing (if able). The Mallet movements include (a) maximal active abduction, (b) external/internal rotation through the full arc of motion (with the humerus abducted 90° and with the humerus adducted to the side), (c) hand to nape of neck behind head, (d) hand to ipsilateral back pocket (buttock), and (e) hand to mouth. Five repetitions of each movement are performed unilaterally by each limb. Additional shoulder movements, such as maximal sagittal plane flexion or other functional tasks, are added to the evaluation if desired.

Elbow and forearm movements measured include elbow flexion/extension and forearm pronation/supination. To define the elbow joint axis, static markers are placed on each medial and lateral epicondyle. The forearm tracking triad is attached to the distal forearm dorsally over both ulnar and radial styloid processes. The forearm triad tracks elbow flexion/extension as well as forearm supination/pronation movements through the full 180° arc of motion. Elbow, forearm, and wrist joint kinematics are measured during various movements depending on the diagnosis of the individual as well as the specific information needed from the evaluation. For example, in an individual with C6 tetraplegia lacking antigravity elbow extension, elbow motion would be performed with the shoulder in various positions to quantify antigravity elbow extension and not necessarily gravity-assisted elbow extension (see **Figure 2**). Likewise, a similar procedure would be followed for pronation to quantify antigravity pronation abilities and not substitution using gravity as an assist.

Wrist movements measured include flexion/extension and radial/ulnar deviation. To define the wrist joint center, static markers are placed at the radiocarpal joint over the lateral aspect of the radial styloid and medial aspect of the ulnar styloid on each arm. To define the distal aspect of the hand segment, static markers are placed just proximal to the metacarpophalangeal joints of the index finger (lateral aspect) and little finger (medial aspect) on each hand. The hand tracking triad is attached to the hand segment with the plate extending distally to the metacarpophalangeal joints of the fingers. The hand triad tracks flexion/extension and radial/ulnar de-

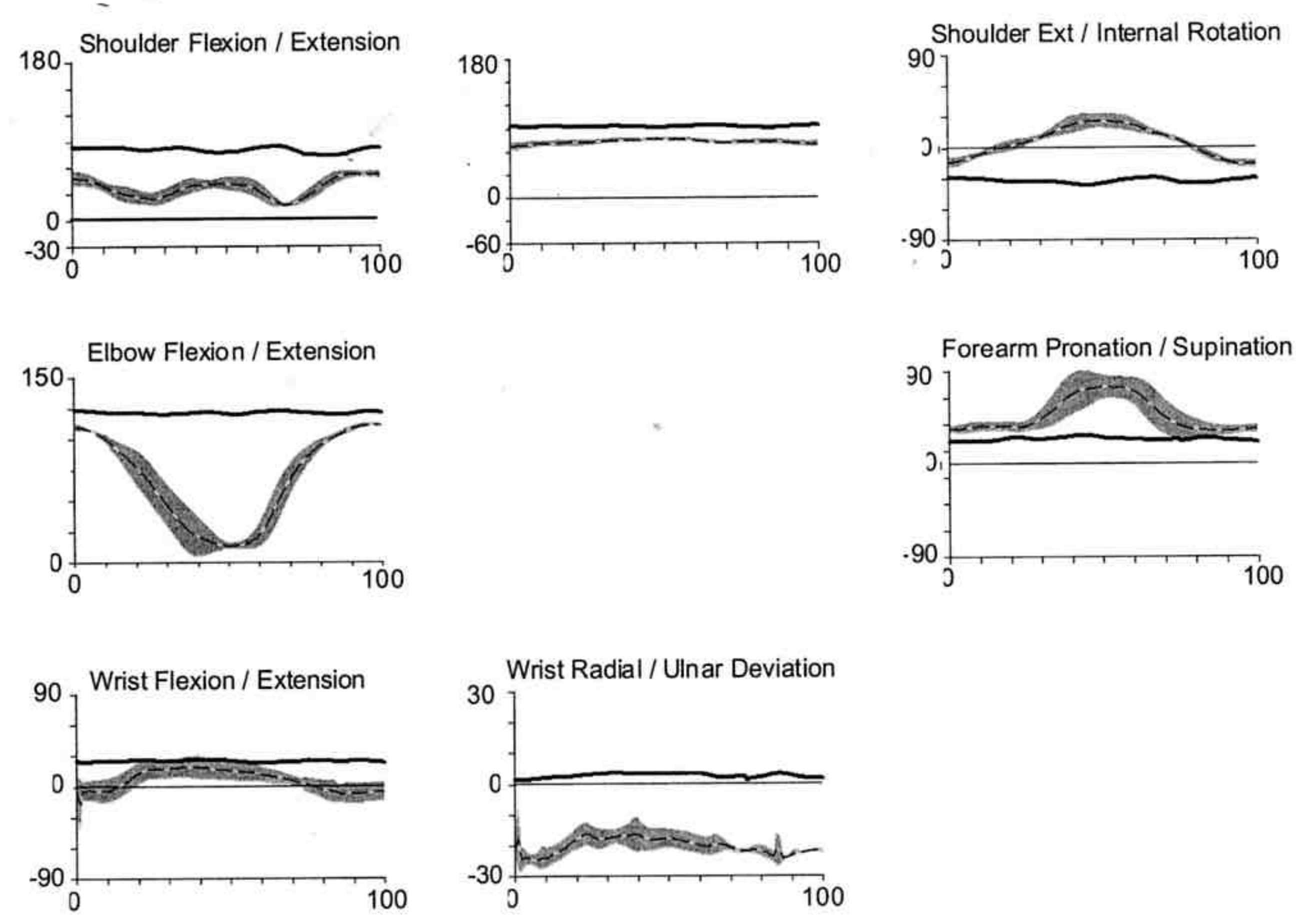

Figure 2. Upper limb kinematics in an individual with C6 tetraplegia during gravity-eliminated active elbow extension with shoulder stabilized in 90° flexion/abduction and adducted 45° from frontal plane. (Starting position 120° elbow flexion.) Grey band with dashed line (mean ± 1 SD) = right side 10 months post-op biceps to triceps tendon transfer. Black lines = left side pre-op biceps to triceps tendon transfer.

viation of the wrist joint relative to the forearm segment. When collecting wrist kinematic data in the individual with tetraplegia, one must take into account antigravity substitutions for absent muscle function depending on the level of spinal cord involvement (e.g., absent wrist flexors in C6 tetraplegia).

Global trunk movements measured include trunk flexion/extension, lateral flexion, and rotation. To define the trunk, the acromion static markers are used superiorly, while iliac crest static markers are used inferiorly. The trunk tracking triad is usually placed posteriorly just below C7 but can be placed alternatively over the superior aspect of the sternum.

Graphic displays of kinematic data are customized for each individual. For unilateral UE dysfunction, kinematic graphs display the individual's noninvolved side motion as the normal reference with the mean ± 1 *SD* band along with multiple trials of involved-side movements. For bilateral UE dysfunction, graphs can contain either multiple trials of right- versus left-sided movements or involved-side motion trials versus reference movement from a normal database. All graphs are color coded for comparison of right and left sides.

Due to the Euler angle estimate problems, we chose to use clinical angles. Use of clinical angles involves taking the flexion-only graph of the flexion-priority Euler sequence, then recalculating abduction/adduction for the abduction/adduction priority sequence or axial for the axial priority. For the axial rotations of the humerus (external/internal rotation), and forearm (pronation/supination), we create a virtual limb defined with the same static markers of the original segment, but we use the tracking targets of the proximal segment. Static captures are taken at 30° and 90° of shoulder abduction with the elbow at 90° of flexion and the forearm fully pronated. The static closest to the dynamic movement to be tested is used for that trial. For example, during the Mallet movements of shoulder external/internal rotation and hand to nape of neck, we use the 90° abduction static trial so that the humerus z-axis is parallel to the reference z-axis. Otherwise, the humerus z-axis is referenced to the trunk x-axis. This method provides the most clinically relevant description of active joint motion, corresponding to what a clinician would measure when using a goniometer. This method works for kinematic data collection during the Mallet and other functional movements in individuals with shoulder pathology such as tetraplegia, brachial plexus palsy, or CP but possibly not for high-velocity movements such as baseball pitching. Future development includes adopting a digitizer pointer method of static data collection and dynamic joint center optimization for the shoulder joint.

Reachable Workspace: Linking Biomechanics to Shoulder Function

Considering the position of the hand in space as the critical determinant of daily function, independent of propulsive needs, several measures of work space have been developed. The primary outputs of these models are three-dimensional hand position translated into a volume of space around the person. The motion of the shoulder and elbow to accomplish these ranges of reach have not been extensively analyzed.

Kukke and Triolo[23] reported increased reach with functional electrical stimulation of the lumbar trunk extensors in four persons with complete SCI. The participants were seated. Reflective markers were placed on a handlebar that the participants held bimanually. They were instructed to sweep the bar to maximum reach in seven planes (the three primary orthogonal planes [sagittal, transverse, and coronal] and four planes at 45° to the primary planes [right and left sagittal and up and down transverse]). The shell of the area covered by the handlebar sweeps was calculated using the mean acromion position as the origin of the coordinate system.

Sison-Williamson et al.[24] reproduced the Kukke and Triolo measurement with modifications in the test protocol and in the data thinning algorithms. Twenty children with SCI were tested both wearing and not wearing a thoracic-lumbar-sacral orthosis (TLSO). Participants were seated with poles placed in front and to their left and right sides. Markers were placed on the poles at floor level and at 20° increments below and above seated shoulder height. Participants were instructed to position their dominant hand in front of them at a specific level (starting at the floor as the first trial) and then sweep in a transverse plane to the right and left poles and back to the front. Each level was done in sequence as a separate trial then all trials were repeated with the nondominant hand. All testing was done both wearing and not wearing the TLSO. Hand data were combined to create a dominant and nondominant file for both test conditions and were processed according to the Kukke and Triolo format with some additional data thinning temporally and spatially to achieve more uniform weighting factors for the volume calculations. On average, reachable workspace volume decreased 28% when the TLSO was worn.

Gutierrez-Farewik et al.[25] reported volumetric range of motion for the UE in three children with spastic hemiplegia undergoing botulinum toxin injections to the biceps. The impaired arm had reduced volumetric range of motion relative to the unimpaired arm, but range of motion increased after treatment in two of the cases. The volume calculation was measured with the thorax as the origin and was based on a convex hull algorithm[26] in Matlab. This technique could be further explored for application to the tetraplegia population.

Conclusion and Future Directions

Quantitative motion analysis based on kinematic models of the shoulder complex and upper limb can provide a means to track changes in shoulder complex motion; when combined with performance of functional tasks, it can help with treatment evaluation. Clearly, many groups are actively focused on improving kinematic modeling of the shoulder complex. Several major issues are critical to advancing the clinical applicability of quantitative motion analysis of shoulder function: procedure for dynamic GH joint centering; more accurate methods to track the scapula; development of models that are not overly complex in terms of degrees of freedom, yet sufficiently sophisticated to adequately capture shoulder movement; and kinematic models and protocols that can account for multisegment movements during functional tasks. In addition, the format of reports and presentation of the kinematic data will be critical for clinical utility. The combination of kinematic and kinetic measures using inverse dynamics with calculation of individual joint forces and mechanical power will provide a more complete picture of shoulder function. The combination of kinematic analyses with a device such as the SMARTWheel (Three Rivers, Phoenix, AZ) that provides handrim kinetics will provide much needed information about wheelchair propulsion. Given the large role of muscles as both stabilizers and movers of the shoulder complex, simultaneous analysis of EMG with motion capture will also be critical. Computational techniques have also improved and support the use of forward models—models in which the musculoskeletal architecture is defined and used to predict the segment movements. Recent work has highlighted the potential of this approach in improving our understanding of UE function.[27]

The use of kinematic models to better quantify shoulder complex motion is an active field of investigation. Within the next decade, advances in quantitative motion analysis techniques will be able to support and enhance the clinical use of such approaches to improve our understanding of shoulder biomechanics in individuals with tetraplegia.

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