INVESTIGATION OF TRANSFER TECHNIQUE BIOMECHANICS AMONG PERSONS WITH TETRAPLEGIA AND PARAPLEGIA

by

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Due to lower limb paralysis people with SCI rely heavily on their upper extremities (UEs) for performing activities of daily living (ADL) of which wheelchair transfers lists as one of the ADL's that are particularly taxing on the UEs. Preservation of UE function is extremely important to maintain independence and quality of life amongst people with Spinal Cord Injury. Although the Paralyzed Veterans of America developed Clinical Practice Guidelines to preserve UE function, limited research/recommendations are available for optimal transfer strategies that reduce loading on UE joints. Specific aims of this dissertation were to 1) Describe biomechanical strategies for preferred methods of transferring amongst persons with paraplegia and tetraplegia (Chapter 2), 2) Determine how taught transfer techniques and self-selected transferring reduce mechanical loading at the UE (Chapter 3), 3) Investigate how upper limb strength and balance impact loading at the UE joints during transfers (Chapter 4). 20 participants took part in the study, only 18 (17 male and 1 female) could execute the taught techniques. The group consisted of 12 persons with paraplegia (6 with complete & 6 with incomplete injury) and 6 with tetraplegia (all with incomplete injury). A custom transfer measurement system was used to capture kinetic and kinematic measures of the UEs and feet while participants performed wheelchair transfers. Participants performed self-selected transfers & three transfer techniques that varied on leading hand placement and trunk flexion. Functional measures recorded included:

Strength, balance, anthropometrics and pain scores. Comparison of mechanical loading between the group with paraplegia and tetraplegia shed light on potential risk of injuries that may occur for each individual group. Taught technique comparison pointed towards a tradeoff, among force and moment components, at the leading and trailing arms, influenced by leading hand placement compared to amount of trunk flexion. Primarily a modified trunk upright technique was employed, trunk remained upright and hand placed close to the body, as preferred method of transferring. Impact of functional measures on transfer kinetics showed that transfer strategy may override functional capacity. Findings of this study will assist in refining clinical practice guidelines on safe level sitting pivot transfer strategies.

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PREFACE

The work on this dissertation would not have been completed without the help of a team of intelligent and wonderfully supportive individuals. I would like to first thank my doctoral advisor Dr. Koontz who provided me with academic and professional guidance and for providing me the opportunity to be a part of the wonderful HERL team. A special thanks to the rest of my committee members: Dr. Boninger, Dr. Cooper and Dr. Gagnon for their support and guidance in defining my dissertation work. I would like to acknowledge Dr. Rubinstein who provided me with the statistical support not just for the dissertation but through the course of the PhD program.

I would like to thank the HERL staff who have provided technical, administrative and moral support. I am truly grateful to the clinical co-coordinators, in particular Michelle Oyster who made sure that I kept on track with the execution of the study. I would like to thank the study participants, without whom this study could not have come to fruition. They were extremely patient inspite of the long testing periods, which were sometimes marred with technical issues.

I made a lot of friends living in Pittsburgh particularly from working at HERL. To the past and present Herlites I owe a debt of gratitude for making Pittsburgh my home away from home. I would like to thank Yen-Sheng Lin who was my partner in crime while testing and provided tremendous support. To my exceptionally supportive family and my loving husband Prashant

Singh, who in no uncertain terms always made it clear that the safety net of being a student was a temporary phase and always motivated me to keep on track. I dedicate this dissertation to the three people who have played the most influential roles in my life: my maternal grandfather, my mom and dad.

1.0 INTRODUCTION

People with spinal cord injury rely on their upper extremities (UEs) extensively to maintain their functional independence. The activities of daily living that require repetitive use of their UEs are propulsion, wheelchair transfers, pressure relief. Any pain /injury to their arms will severely affect their quality of life (1). Manual wheelchair propulsion has been extensively researched. The most favorable propulsion patterns to preserve shoulder function have been identified and supported by quantitative data (kinetics, kinematics and electromyography). Limited literature/research however has been conducted in the field of wheelchair transfers and only preferred methods of transferring have been explored. ***Biomechanical evaluation of optimal transfer movement strategies is non-existent***.

Transfers and weight reliefs were identified to be common causes of shoulder pain complaints amongst individuals with SCI (2, 3). Dalyan et al. (4) conducted a survey based study examining symptoms of pain amongst 130 individuals with SCI (paraplegia (68) & tetraplegia (62)) one year post injury. They found that 58.5% of the individuals reported UE pain of which 71% had shoulder pain, 35% elbow pain, 53% wrist pain and 43% hand pain. In addition they also found that 65% of the participants reported UE pain that hindered transfer performance.

Wheelchair transfers can be classified based on three factors namely height (level and non-level), approach (lateral, front or back approach) and strategy (rotatory and translatory movement patterns). The Clinical Practice Guidelines on preservation of UE strength (5) recommends performing level transfers whenever possible. The lateral transfer is the most common type of transfer since it is quick and requires less strength, and is essential for maintaining an independent lifestyle (6). Allison et al (7) identified two predominant transfer movement strategies but these strategies have not been biomechanically verified.

We have developed three transfer techniques based on the strategies suggested by Allison et al (7)and ergonomic principles. This study will verify the techniques biomechanically and examine if one of them could be potentially implemented as a technique to prevent UE pain/injury in the long run. The findings of this study will aid in improving clinical practice guidelines for wheelchair transfers.

1.1 TRANSFER MOVEMENT STRATEGIES

People with SCI perform an average of 15 to 20 wheelchair transfers per day(8) (9), (10). Transferring results in majority of the body weight being borne by the UEs which leads to high stresses at the shoulder, elbow and wrist. The combination of high stress and large number of transfers carried out per day predispose the UEs to secondary injuries over time. The Consortium for Spinal Cord Injury published clinical practice guidelines on the preservation of UL function however limited recommendations on wheelchair transfers are present (5).

Therapists chiefly use clinical judgment while training people with SCI on how to perform a wheelchair transfer. Functional level plays a vital role in determining which transfer training strategy can be taught to a particular individual. Compensatory strategies are employed to enable successful transferring which include muscle substitution, momentum and Head-Hips relationship (moving the buttocks by moving the head in the opposite direction) (6). Allison *et al.* (7) described two general movement strategies used when performing lateral transfers: *rotational strategy* (head moves in an opposite direction to the pelvis) and *translational strategy* (head and pelvis move simultaneously in the same direction). When viewed from the sagittal plane, individuals performing the *rotational strategy* leaned forward during the transfer and those using the *translational strategy* kept their trunk more upright during the transfer. The *rotational strategy* is analogous to what clinical practice refers to as the 'head-hips' relation. It is often taught to patients with weak triceps and/or those with high levels of trunk involvement. They are trained to use the concept of momentum by moving their UEs to enable motion of their pelvis while transferring from one surface to another.

Wheelchair transfer related research has focused predominantly on the paraplegic population. Research involving pressure reliefs, posterior transfers and assisted transfers (i.e. transfer using an assistive device) have been explored in the tetraplegic population (11-13); however limited research has been carried out in the field of independent wheelchair transfers within the same population. Harvey et al (11), who explored pressure reliefs in a group of people with tetraplegia, reported that the participants began the lift with their trunk forward flexed. Individuals with levels of injury at C5-C6, rely on the anterior deltoids, sternal pectoralis and biceps brachii to carry out elbow extension to '*lock*' it prior to weight bearing (11, 12). They found that the compensatory movements that enable persons with C5 - C6 motor complete injuries to lift their body causes large shoulder flexor moments, lower elbow extension moments and higher wrist flexion moments. We predict to see similar results in our study.

1.2 TRANSFER BIOMECHANICS

Ergonomic literature deems that a classic impingement position involves the arm being forward flexed, internally rotated and abducted (14). Significant amounts of vertical stress borne at the shoulder in combination with the humeral elevation during UE weight bearing activities compress the rotator cuff (15). It is very difficult to avoid an impingement position while performing a wheelchair transfer. Studies have found that using a forward-flexed trunk position during transfers and pressure relief engages sternal pectoralis major and latissimus dorsi muscles (15, 16). This muscle substitution may help transfer the body weight between the leading arm and trailing arm with less loading of the glenohumeral joint thereby reducing the risk of rotator cuff impingement (15-17).

Nawoscenzki et al. (18)studied 3D scapular and humeral movement, during a weight relief raise and a wheelchair transfer, in a group of unimpaired asymptomatic individuals. They found similar patterns between the weight relief raise and transfer i.e. increased upward rotated and anteriorly tipped scapula with reduced humeral external rotation. Additional differences were found while comparing the trailing and leading limb; the leading limb had increased scapular anterior tipping and internal rotation along with reduced upward rotation and humeral external rotation. These movement patterns result in reduction of the subacromial space contributing to deformation of the rotator cuff tendons.

Finley et al. (9)studied scapular kinematics in 23 participants with paraplegia of which UE impingement syndrome was present in 10 participants and remaining 13 were asymptomatic. The results were found to be similar to the study conducted by Nawocenski (18) in the asymptomatic group. People with UE impingement syndrome were found to however have reduced thoracic flexion, increased scapular upward rotation and reduced humeral internal rotation compared to the asymptomatic group. This movement pattern adopted by WC users who already have the presence of UE pain could be a potential preservation technique to reduce impingement of the greater tuberosity under the acromion thereby allowing the individuals to continue with functional activity. This leads us to believe that individuals who are further out from injury may have optimized their preferred method of transferring thereby reducing shoulder joint loading and preserving functional capability in order to continue performing ADL's.

Hand placement i.e. placing hand further away from the body or closer to the body, while performing transfers has not been explored in previous research. Glenohumeral joint instability has been found to increase with increasing shoulder flexion and abduction(14). Placing the leading hand further away from the body increases the moment arm, which results in generation of a higher angular momentum. Although the increased momentum assists in transfer performance there are drawbacks namely increased shoulder moments and poor positioning of the arm i.e. forward flexed and abducted.

1.3 ROLE OF LOWER EXTREMITIES DURING WHEELCHAIR TRANSFERS

Role of LE's during wheelchair transfers has been highlighted albeit minimally investigated. Although motor function is absent in the lower limbs, when positioned appropriately they can weight bear and also assist in stabilizing the trunk, which makes it easier to pivot on the arms (5, 6). Tanimoto et al (19) recorded hand and feet kinetics in a group of individuals with SCI (2 with tetraplegia and 11 with paraplegia) while performing level transfers. They also found that while studying similar protocol in healthy subjects they found that the vertical reaction forces at the feet increased with increase in trunk inclination. Gagnon et al (20) explored the role of the feet during level and non-level transfers in a group of 12 participants with paraplegia. Both studies found that the feet bore about 25% BW during the lift phase of the transfer.

1.4 FUNCTIONAL MEASURES AND WHEELCHAIR TRANSFERS

Muscle strength has been identified as a key element to be considered in the preservation of upper extremity function (21). A recent study investigated the relationship between shoulder and elbow strength and transfer ability (22). No association was found between upper limb strength and how high and how low one could transfer or how far away from the target surface one could be and still safely transfer. Thus, this study raised questions as to how critical strength is to performing transfers and whether other factors such as trunk balance, upper limb pain, skill/technique or anthropometry (e.g. weight, height, etc.) have greater impact on transfer performance.

Very limited research is available exploring the relationship between biomechanical variables during transfer activity with sitting balance and UE strength. Amongst people with SCI, sitting balance has been found to be highly correlated to functional performance of daily tasks such as transferring (23). Enhanced sitting balance allows for increased movement control (24) and potentially may result in reduced mechanical loads at the shoulder. Previous literature has revealed a correlation between high forces with injury/pain at the UE joints (5).

Van Drongelen (25) investigated relationship between upper extremity pain with respect to UE strength, lesion level and motor score (obtained using Functional Independence Measure), during and one year post rehabilitation. He found the functional outcome and UE strength to be inversely related to UE pain. From his results he concluded that higher muscle strength would lead to a reduction in UE pain. What we predict to find in our study is that sitting balance and UE muscle strength will be sensitive predictors of UE forces and moments while performing a wheelchair transfer.

1.5 SIGNIFICANCE

High reliance on upper extremities for daily functioning predisposes people with SCI to experience UE pain/injury. UE pain is usually complemented with reduced functioning and a lower quality of life. Among the ADL's transfers and wheelchair propulsion have been identified to be the leading cause of shoulder pain (26, 27). Prior research has found high prevalence of UE joint pain [shoulder: 30-60%, elbow: 5- 16% and wrist: 40-66%] (5, 10). It has been suggested that damage to the UEs is functionally equal to an SCI of higher neurological level (28). Transfer strategies that are used by people with SCI are adapted based on level of SCI, UE strength and functioning. This study is important since there has been no research that has been conducted in identifying optimal strategies in performing wheelchair transfers in a controlled and consistent manner (controlling hand placement, Trunk range of motion etc.).

2.0 UPPER LIMB KINETIC COMPARISON OF SITTING PIVOT TRANSFERS AMONG PEOPLE WITH PARAPLEGIA AND TETRAPLEGIA

2.1 INTRODUCTION

Every year there are 12,000 new cases of Spinal Cord Injury, with the neurological impairment level at discharge being: 40% incomplete tetraplegia, 16.3 % complete tetraplegia, 21.7% incomplete paraplegia and 22.1% complete paraplegia (29). Those persons with SCI, who are full time manual wheelchair (MWC) users, perform an average of 15 to 20 wheelchair transfers per day to complete basic activities of daily living (8, 9, 30). Maximum dependence and repetitive stress on the upper extremities (UEs) leads to high incidence of shoulder, elbow and wrist pain (8, 28, 31). People with tetraplegia reportedly experience higher intensity and prevalence of shoulder pain as compared to those with paraplegia (25, 28, 32).

Wheelchair transfer related research has focused predominantly on the population with paraplegia. Research involving pressure reliefs, posterior transfers and assisted transfers (i.e. transfer using an assistive device) have been explored in the population with tetraplegia (11-13); however limited research has been carried out in the field of independent wheelchair transfers within the same population. Harvey et al (11), who explored pressure reliefs in a group of people with tetraplegia, reported that the participants began the lift with their trunk forward flexed and could bear some weight through flexed elbows. Individuals with levels of injury at C5-C6, rely on the anterior deltoids, sternal pectoralis and biceps brachii to carry out elbow extension to '*lock*' it prior to weight bearing (11, 12). They found that the compensatory movements that enable persons with C5 - C6 motor complete injuries, to lift their body is associated with the generation of large shoulder flexor, shoulder adduction and wrist flexion moments.

Gagnon et al (17) analyzed posterior transfers amongst two groups of individuals with SCI: 6 with level of injury (LOI), between C7 to T6 and 5 persons LOI ranging between T11 to L2. They found that the overall muscle activity particularly for the anterior deltoids and pectoralis major, were greater for persons with higher LOI compared to those with lower LOI. They also observed that the individuals with higher LOIs performed the transfer maneuver with the trunk forward flexed.

Therapists chiefly use clinical judgment while training people with SCI on how to perform a wheelchair transfer. Lesion level plays a vital role in determining which transfer training strategy can be taught to a particular individual. Key muscles such as the triceps brachii, lattismus dorsi and the sternal part of the pectoralis major are often compromised with higher LOI's (12). Additionally higher LOI is associated with poorer trunk stability due to paralysis of thoracohumeral muscles (32, 33). Compensatory strategies are employed to enable successful transferring which include muscle substitution, momentum and Head-Hips relationship (moving the buttocks by moving the head in the opposite direction) (6). The Clinical Practice Guidelines (CPG) on the preservation of upper limb function (5) lists generic recommendations on how to perform transfers. However these recommendations are not sensitive to the characteristics of the transfer strategies adopted by people with tetraplegia and paraplegia. A complete systematic kinetic analysis of the UE joints for preferred method of level sitting pivot transfers (SPTs) will provide a clearer understanding of the transfer strategies adopted by both groups. The analysis will also assist in determining the necessity of tailored transfer training protocols /interventions.

The aim of this study was to compare trunk, shoulder and elbow joint ranges of motion; forces at the shoulder, elbow and hand; and moments at the shoulder, elbow and wrist, between people with paraplegia and tetraplegia as they perform SPTs. We hypothesize that similar to the results by Harvey et al (11), higher shoulder flexion and adduction moments, lower elbow extension moments, and higher wrist flexion moments will be observed in the group with tetraplegia (TG)compared to the group with paraplegia (PG). We hypothesize that participants with paraplegia will adopt a translatory method of transferring while the group with tetraplegia will use a rotatory manner of transferring. Gagnon et al (20) evaluated tri-axial component forces at the hand in a group of MWC users with thoracic level SCI for level and non level transfers. They found a simultaneous increase in vertical and *horizontal* forces to be associated with the momentum produced as a combination of a higher forward flexed trunk velocity and mass (trunk and head), while performing an SPT. Similarly we hypothesize that the non vertical forces (i.e. medio-lateral and anterior-posterior) will be higher for the TG compared to the PG.

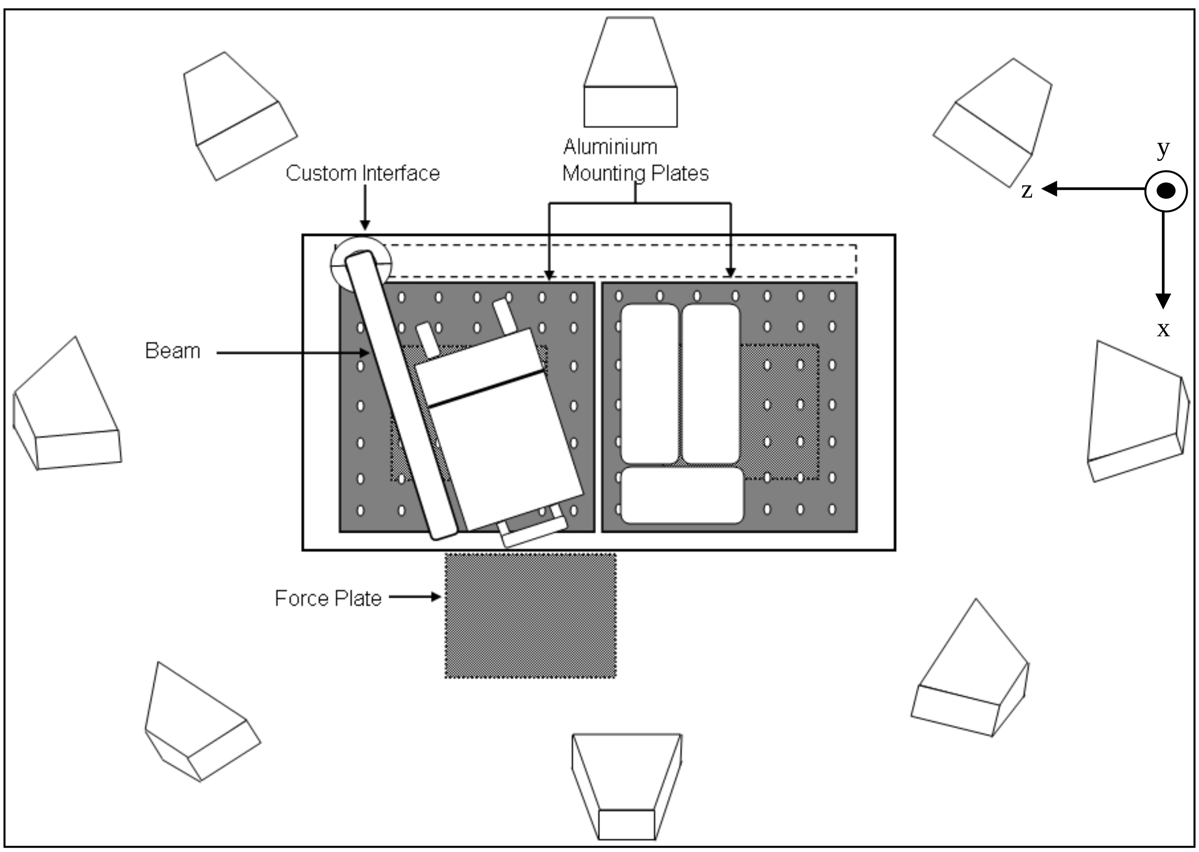
2.2 METHODS

2.2.1 Subjects

This study received ethical approval from the Department of Veterans Affairs Institutional Review Board. After reading and providing informed consent, twenty subjects (19 male, 1 female), volunteered to participate in this study. The inclusion criteria were: spinal cord injury

C4 level or below that occurred over one year prior to the start of the study, able to independently transfer to/from a manual wheelchair without human assistance or assistive devices, over 18 years of age, and free from upper extremity pain that influenced their ability to transfer.

Figure 1. Wheelchair (left) and tub bench (right) shown secured to the aluminum mounting plates of the base frame. The custom interface consists of the load cell and a beam that can be positioned anywhere along base frame.Sixteen 3D motion cameras surround the base frame.



2.2.2 Data collection

Strength Testing

Isokinetic strength measurements, at a torque arm speed of 60 deg/sec, were recorded using an instrumented dynamometer (Biodex Medical System, New York, USA). The measurements were recorded in a randomized fashion for both arms for the following test maneuvers: shoulder flexion/extension in the sagittal plane, shoulder abduction/adduction in the frontal plane, shoulder internal/external rotation in the transverse plane, elbow flexion/extension and wrist flexion/extension (tested range of motion shown in Table 1).

Table 1. Isokinetic testing of UE joint at 60 deg/sec

Joint	Movement	Tested Range of Motion (Degrees)
Shoulder	Extension/Flexion	-30 to 50
	Adduction/ Abduction	10 to 70
	Internal/External Rotation	0 to 45
Elbow	Extension/Flexion	0 to 90
Wrist	Extension/Flexion	-45 to 45

Two practice repetitions for each movement tested were completed prior to data collection. In order to ensure the maximal force production of the tested upper extremity participants were secured into the chair with three padded belts: two diagonally across their chest and one across their lap. Five repetitions were recorded for each maneuver and participants were allowed to rest for five minutes to avoid fatigue from becoming a confounding factor. A rest period of 30 to 60 minutes was taken after the strength testing before transfer biomechanics were recorded. From strength testing, peak isokinetic torques were calculated using customized software (MATLAB 2011b, MathWorks; Natick, Massachusetts) and were averaged over five repetitions for each of the movements recorded at the shoulder, elbow and wrist. The peak torques were normalized by body weight for each participant and reported in % Meter (%m). Additionally influence of asymmetric strength differences between groups were investigated by computing peak torque ratios i.e. right/left.

2.2.3 Experimental protocol

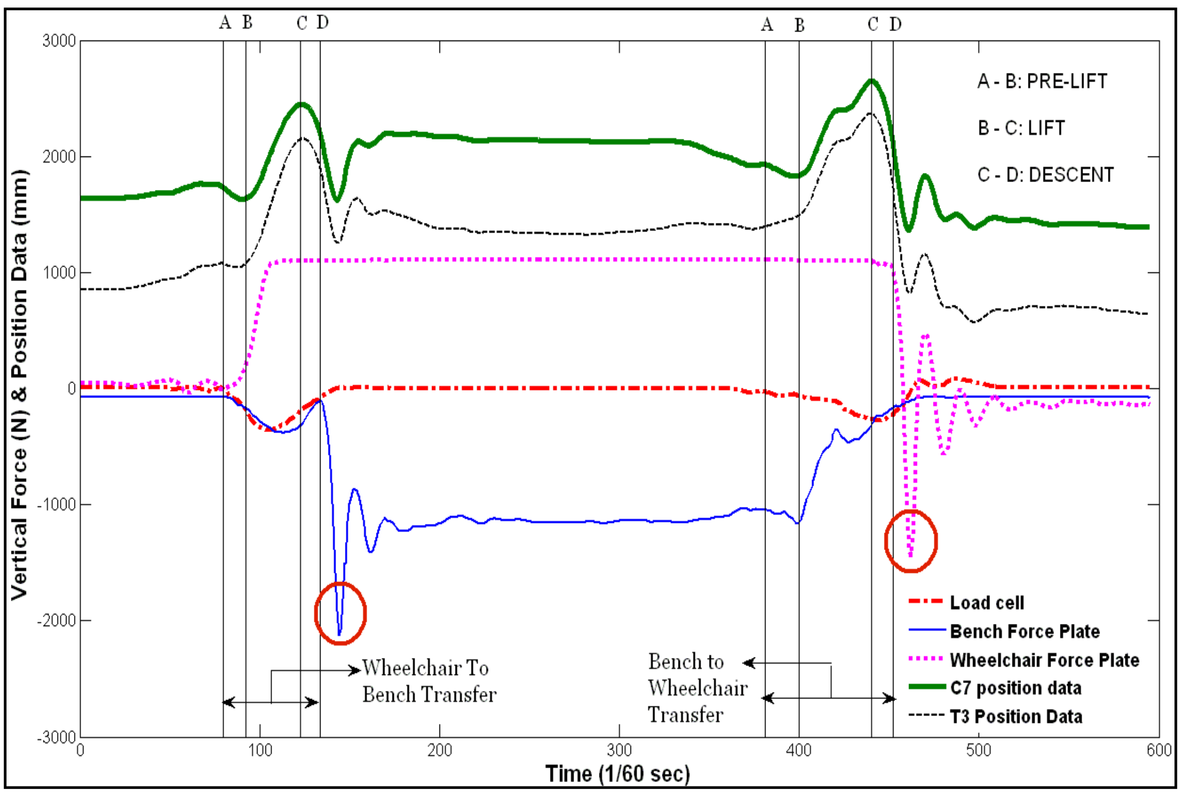
Participants used their personal wheelchairs to transfer to and from a bench. For all transfers the wheelchair was positioned and secured at a comfortable angle from an adjustable height tub bench as shown in Figure 1. The bench was adjusted to be level with the wheelchair seating surface. The platform contains three force plates (Bertec Corporation, Columbus, OH), one beneath the wheelchair, one beneath the tub bench and one located below the feet (34). The wheelchair and bench were secured to the platform. A steel beam attached to a 6-component load cell (Model MC5 from AMTI, Watertown, MA) was positioned to simulate a wheelchair armrest. Reflective markers were placed on the subjects C7, T3 and T8 vertebrae, right and left acromion processes, 3rd metacarpalphalangeal joints, radial and ulnar styloid processes, and lateral epicondyles. The coordinates of the markers were recorded based on a global reference frame using a sixteen camera three-dimensional (3D) motion capture system (Vicon Peak, Lake Forest, CA). Several anthropometric measurements were recorded such as: axiliary arm, wrist, fist and elbow circumference, upper arm and forearm length.

Based on the experimental set up, all transfers began with the left arm leading and moving the body from the wheelchair to the bench. Subjects were instructed to perform a sitting-pivot transfer (SPT) as they normally would to an adjacent level tub bench. They were instructed to place their left hand (e.g. leading hand) anywhere on the tub bench and their right hand (e.g. trailing hand) on the steel beam (height of wheelchair arm rest). Subjects practiced transferring before recording began. Each transfer technique was performed three times and kinetic and kinematic data were recorded at 360 Hz and 60 Hz respectively, for the length of the transfer. The point of external force application on the hand was assumed to be at the base of the third metacarpal(35).

2.2.4 Data analysis

Kinetic, kinematic, and anthropometric data were entered into an inverse dynamic model to calculate the 3D net shoulder and elbow joint forces and moments. Kinetic data were down sampled to 60 Hz, to align with the kinematic data. Both kinematic and kinetic data were filtered with a 4th order zero-lag Butterworth filter (cut off frequency of 5 Hz and 7 Hz respectively). The inverse dynamic model used was based on the general rigid-link segment model using a Newton-Euler method and a variable degree of freedom body co-ordinate system (35). The local coordinate systems were approximated on the recommendations by the International Society of Biomechanics (ISB) (36). Customized software (MATLAB 2009b, MathWorks; Natick, Massachusetts) was used to compute the trunk movement using a Cardan angle sequence (ZXY, along a anterior/posterior axis [x], superior/inferior axis [y], and medial/lateral axis [z] acting to flex/extend [z], lateral rotation [x], and axial rotation [y] with respect to the laboratory coordinate system). Upper extremity joint movements were computed using an Euler angle sequence: shoulder movement (YXY along plane of elevation [y], amount of elevation [x], and internal/external rotation [y] and elbow movement (ZXY, flexion/extension [z])). All upper extremity joint coordinate systems acted with respect to the trunk coordinate system.The vertical reaction force from the force plate under the bench and the grab bar were used to determine the start (absolute value of vertical force > 0 at the grab bar and bench) and the end (determined prior to the generation of a large spike in the vertical force at the bench) of the transfer. The SPT was delineated into three phases: prelift, lift and descent (Figure 2). The vertical forces from the force plate under the tub bench, wheelchair and the grab bar (wheelchair side) were superimposed onto one plot to analyze the phases of transfer. Trunk motion, represented by the C7 and T3 markers, were added to the force data to assist with the delineation of phases of transfer (37).

Figure 2. Representative self selected transfer recordings. Position and kinetic data were combined to delineate the sitting pivot transfer into three phases: pre-lift, lift and descent.



2.2.5 Kinetic and kinematic outcome measures

Analysis was conducted for transfers from the wheelchair to tub bench. Kinematic Variables: maximum and minimum angles of trunk flexion/extension; lateral and axial rotation; shoulder flexion/extension; abduction/adduction, and internal/external rotation and elbow flexion/extension were identified for each transfer trial.

Kinetic variables: For each trial 3D component and net resultant forces were calculated for the shoulder, elbow and hand. Flexor/adductor moments at the shoulder, flexor/extensor moments at the elbow and flexor moments at the wrist were calculated for both leading and trailing arms. The maximum and minimum of the aforementioned variables were identified for the three phases of the transfer. Variables were computed using Matlab (Mathworks, Inc., Natwick, MA) and averaged over the three trials. All mean force and moment values were normalized against body weight for each participant and reported a % Body Weight (%BW) and % Meter (%m) respectively.

2.2.6 Statistical analysis

Means and standard deviations were computed for each group i.e. Group with paraplegia (PG) and Group with Tetraplegia (TG). Independent t-tests were performed to determine group differences for the demographic variables: age, height, weight and years since injury. Based on the normality of the strength measures a Student's t-test or a Mann-Whitney *U* test was used to investigate peak torques for each arm and peak torque ratios(right/left) across the UE joints between the two groups (PG /TG).

A 3 way mixed model Repeated Measures ANOVA was used: 2 (Group: TG/PG) X 2(Arm: Leading/Trailing) X 3 (Phase: Prelift/Lift/Descent), to compare UE joint kinematics, UE joint moments and reaction forces for both groups. If the 3 way interaction was significant then the 3 way model was collapsed into two 2- Way Mixed Models (38):

- 2 (Group) X 2 (Arm) for each phase of the transfer

- 2 (Group) X 3 (Phase) for each arm.

In cases where in the 3-way interaction was not significant, analysis of the 2-way interactions: Group X Arm averaged across phase and Group X Phase averaged across arm was performed. For all models the main and interaction effects were investigated using simple pair wise comparisons and a post hoc Bonferroni correction. Level of significance was set to 0.05. Due to the small sample size we have considered p values less than 0.05 to be statistically significant and p values ranging between 0.1- 0.05, to indicate '*significant trends*'. The statistical tests were performed using SPSS statistical software (SPSS Inc., Chicago, IL). Effect size (ES) was computed for group differences averaged across both arm & phase, and group differences for each arm averaged across phases.

2.3 RESULTS

2.3.1 Demographics

Table 2 provides the demographic information of the subjects, by group. No significant differences were found between the groups in regards to the demographic variables of age, height, weight and years since injury.

Table 2. Mean (standard deviation) of subject demographics for each of the groups of interest.

	Group with paraplegia (PG) (N = 13)	Group with Tetraplegia (TG) (N = 7)
Age (Yrs)	37.77 (12.63)	38.85 (7.15)
Height (m)	1.79 (0.11)	1.55 (0.69)
Weight (Kg)	73.99 (20.87)	81.95 (16.81)
Years Since Injury (Yrs)	15.92 (8.34)	12.43 (8.16)
Lesion Level	T3/T4 - L5	C7-C5
Injury: Complete/Incomplete	7 complete/6 incomplete	7 incomplete

2.3.2 Group comparison

2.3.2.1 Upper extremity strength

Three participants could not complete the entire protocol and belonged to the TG. Of the three participants, one of them owing to left arm weakness had missing data for all strength measures at the shoulder, elbow and wrist of the left arm. Of the two remaining participants, one had missing data for right shoulder flexion/extension and abduction/adduction. Additionally both had missing data for left wrist flexion extension strength measurements. The PG had greater right shoulder external rotation peak torque compared to the TG (Figure 3, $p = 0.028$). Differences in strength measures were more apparent at the elbow and wrist. The isokinetic peak torques associated with right elbow F/E, right wrist F/E and left elbow extension was significantly higher for the PG compared to the TG (Figure 4, p values < 0.047). Wrist flexion and extension torque tended to be higher for the PG compared to the TG ($p = 0.05$ and $p = 0.064$).

The groups were similar in terms of strength asymmetry (right arm vs. left arm) with the exception of elbow flexion torque ratio. The PG (1.27 ± 0.44) had significantly higher asymmetric strength compared to the TG (0.75 ± 0.32), with respect to elbow flexion ($p = 0.027$). This finding indicates that the PG had stronger elbow flexors in their right arm compared to the left arm which was significantly different compared to the TG who had weaker elbow flexor strength in their right versus left.

Figure 3. Isokinetic peak torques at the shoulder for three contractions: Flexion/Extension, Abduction/Adduction and Internal/External Rotation.

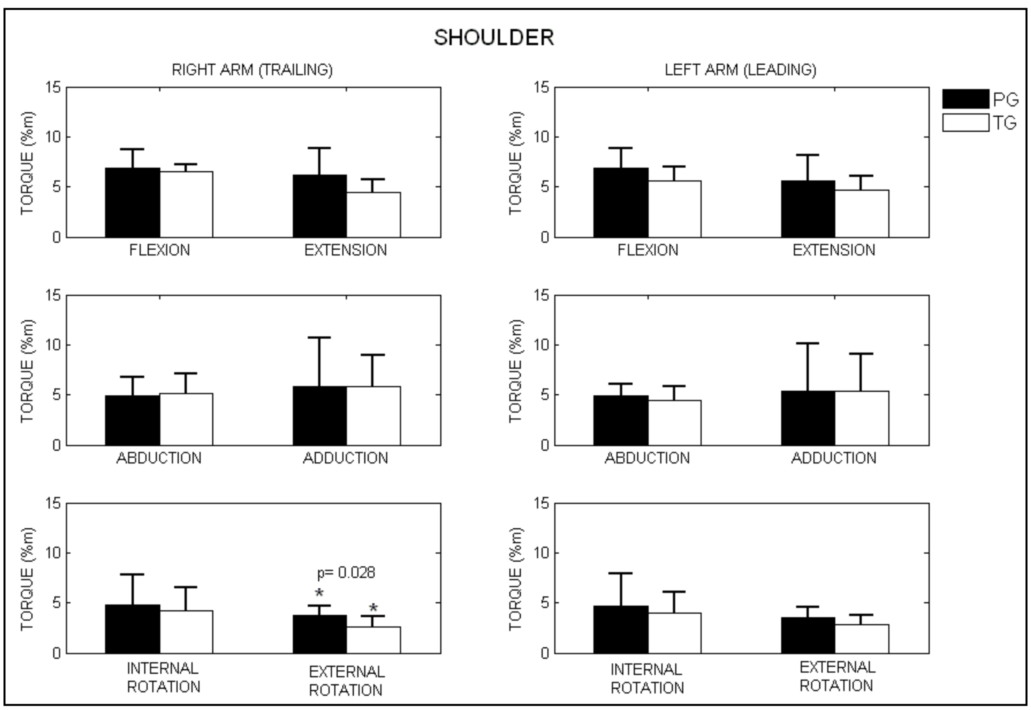


Figure 4. Isokinetic peak torques recorded at the elbow and wrist. *: p<0.05 and &: 0.05<p<0.10.

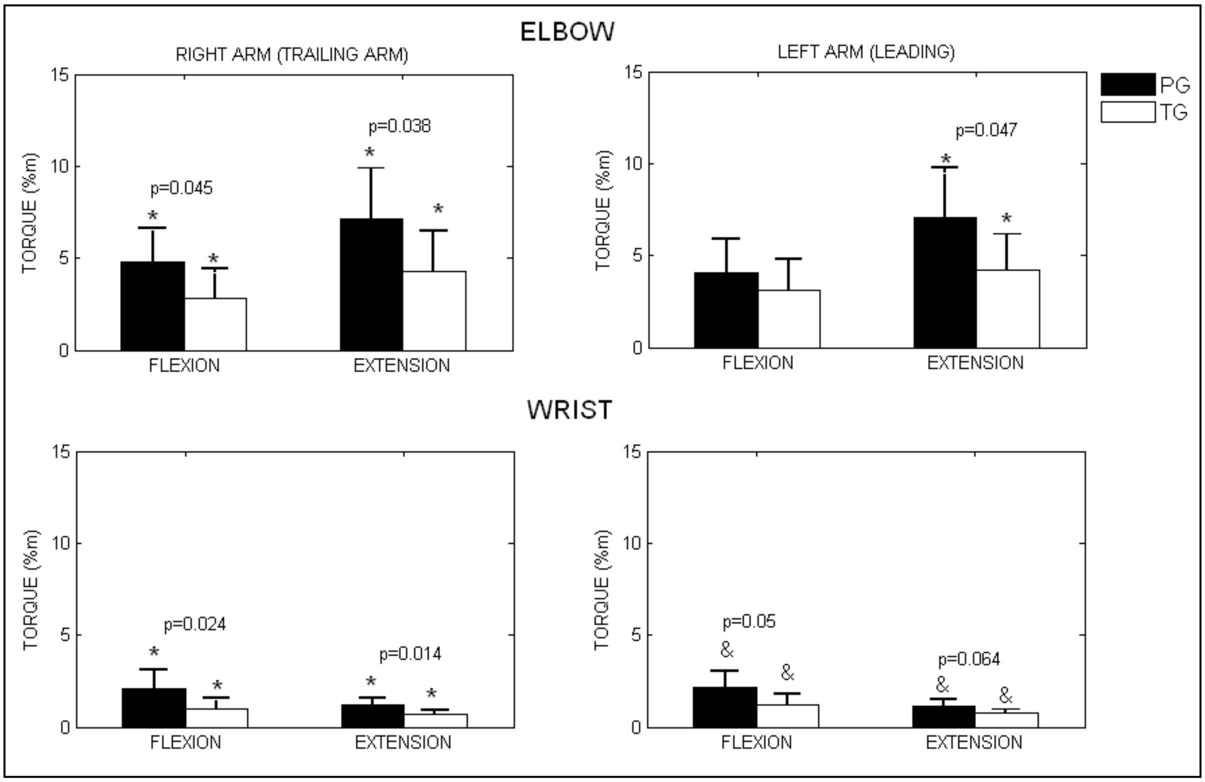


Table 3. Individual group mean (standard deviations) for the primary kinematic outcome variables during the lift phase.

Joint	Movement	PG Trailing Arm	PG Leading Arm	TG Trailing Arm	TG Leading Arm
SHOULDER	Flexion +; Extension -				
	Max	23.05 (33.64)	32.06 (15.83)	27.67 (10.53)	40.28 (14.34)
	Min	17.46 (35.92)	-8.14 (10.65)	10.48 (11.45)	-16.71 (11.01)
	Internal Rotation +; External Rotation -				
	Max	13.59 (36.29)	-18.07 (20.29)	13.27 (13.53)	-14.88 (26.64)
	Min	32.16 (35.63)	42.24 (19.80)	31.67 (11.94)	54.19 (23.71)
	Adduction +; Abduction -				
	Max	-26.49 (6.33)	-31.58 (9.42)	-27.35 (5.78)	-32.81 (14.12)
	Min	37.44 (6.86)	59.27 (13.56)	38.75 (7.28)	54.72 (18.45)
ELBOW	Flexion +; Extension -				
	Max	31.14 (52.42)	36.52 (17.87)	66.30 (48.72)	36.55 (17.42)
	Min	9.90 (54.74)	26.23 (15.96)	49.55 (58.10)	19.74 (17.31)

TRUNK		PG	TG
	Flexion	46.32 (11.32)	43.34 (7.59)
	Axial Rotation: Left +; Right -		
	Max	-11.96 (9.97)	-17.57 (11.88)
	Min	-21.09 (9.79)	-28.92 (13.14)

2.3.2.2 Kinematic and kinetic differences

No significant kinematic differences were found between the two groups: PG and TG (Table 3). Peak resultant forces were not significantly different between groups (Figure 5) across the UE joints and hand. We found that the trailing shoulder bore 30 ± 9.56 %BW over the pre-lift, 40 ± 12.97 %BW during the lift and 27 ± 10.34 %BW during the descent phases. At the leading shoulder we found 20 ± 12.5 %BW being borne during the pre-lift, 35± 9.55 %BW over the lift and 42 ± 11.86 %BW during the descent phases. However significant group differences were found in the component forces and moments and are described below for each joint.

Figure 5. Peak resultant forces at the shoulder, elbow and hand across transfer phases for each arm

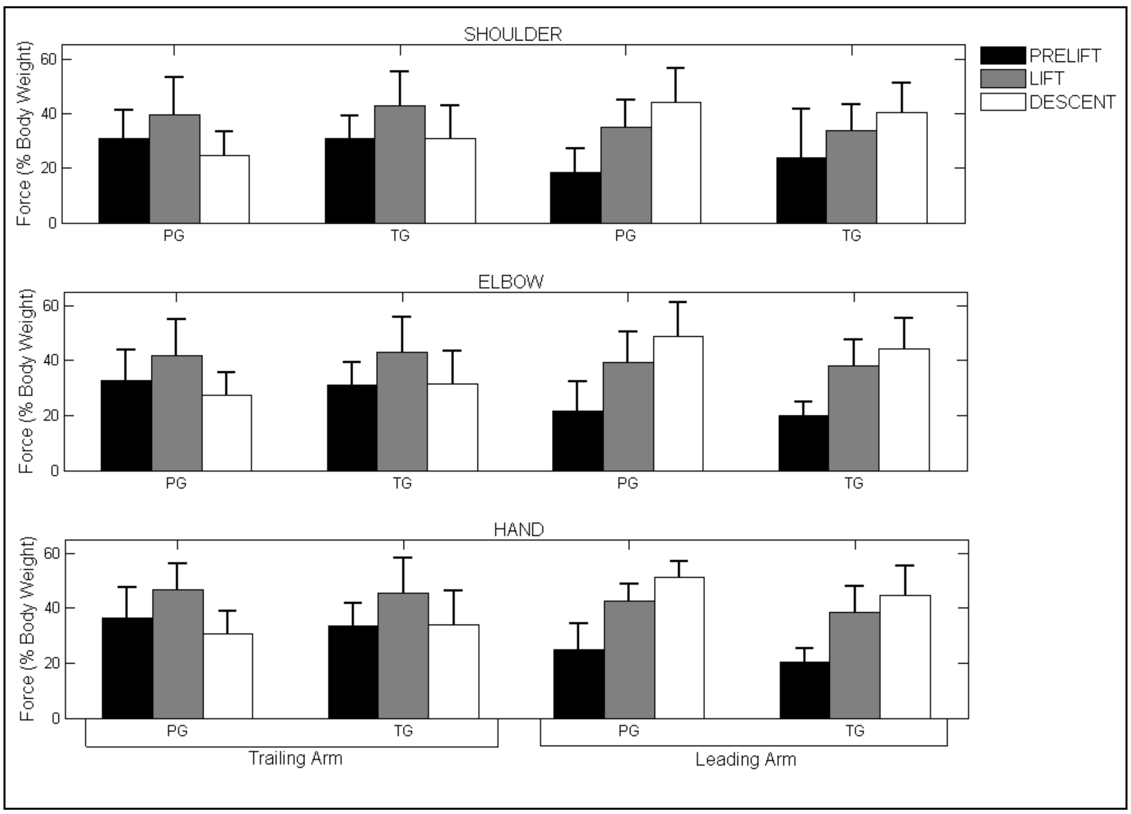


Figure 6. Peak Moments across the upper extremity joints for each phase of the transfer, for both arms across groups.

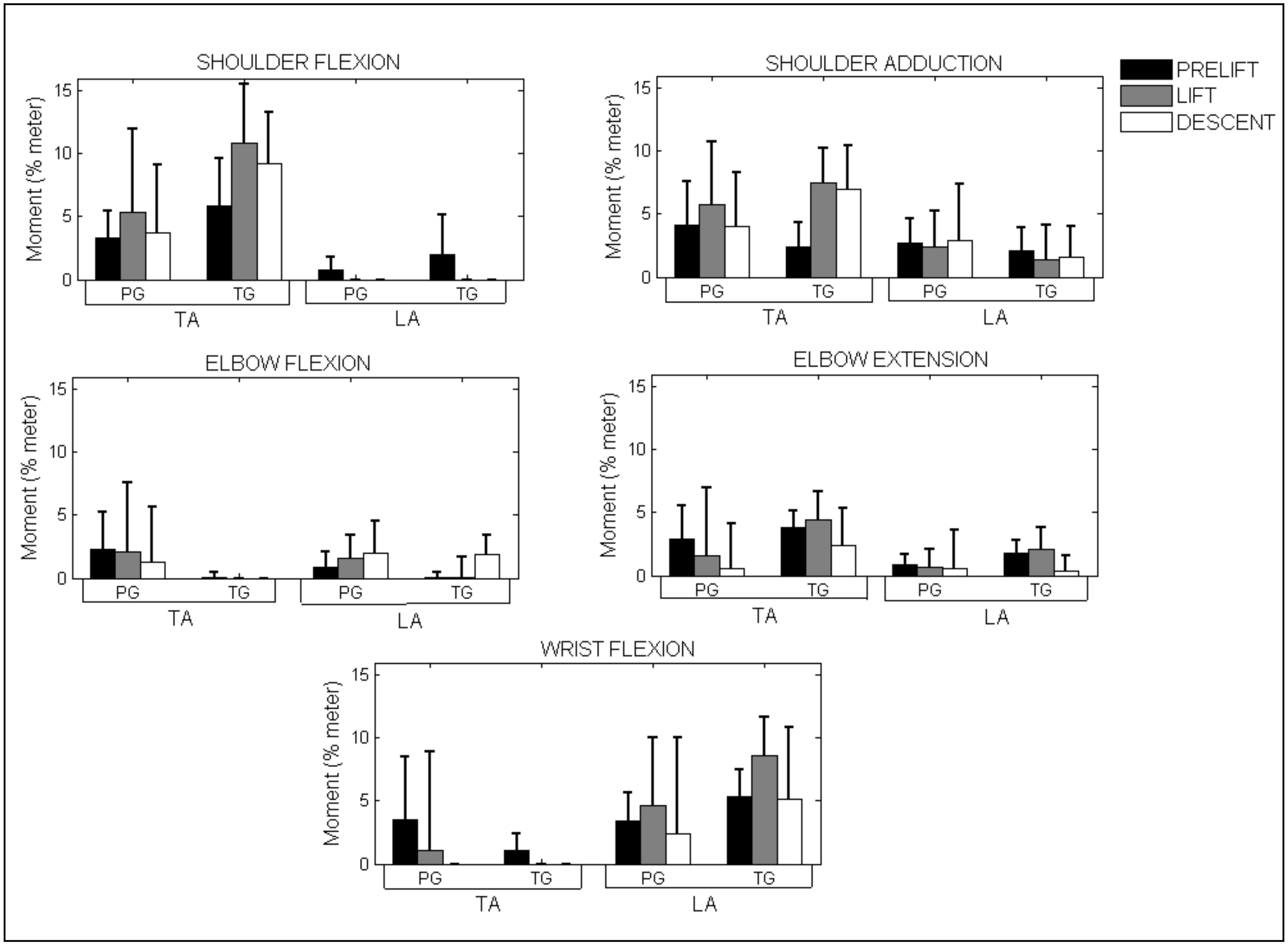


Figure 7. Peak 3D reaction forces at the shoulder across the phases of the transfer for each arm. [Note: Bonferroni adjusted p-values are shown. * indicate significant comparison at p < 0.05; + - indicates significant 'trends']

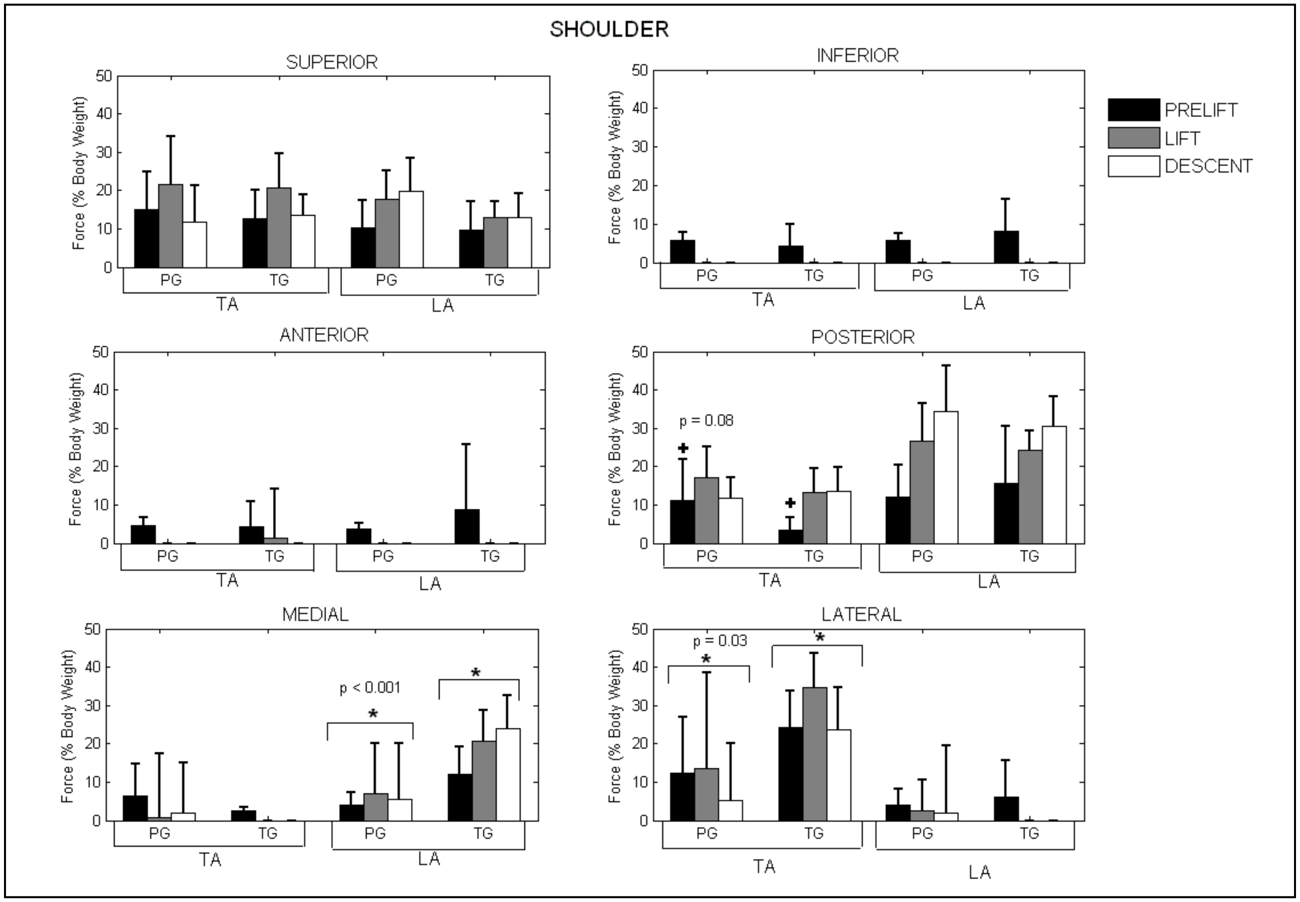


Figure 8. Peak 3D reaction forces at the elbow across transfer phases for each arm.

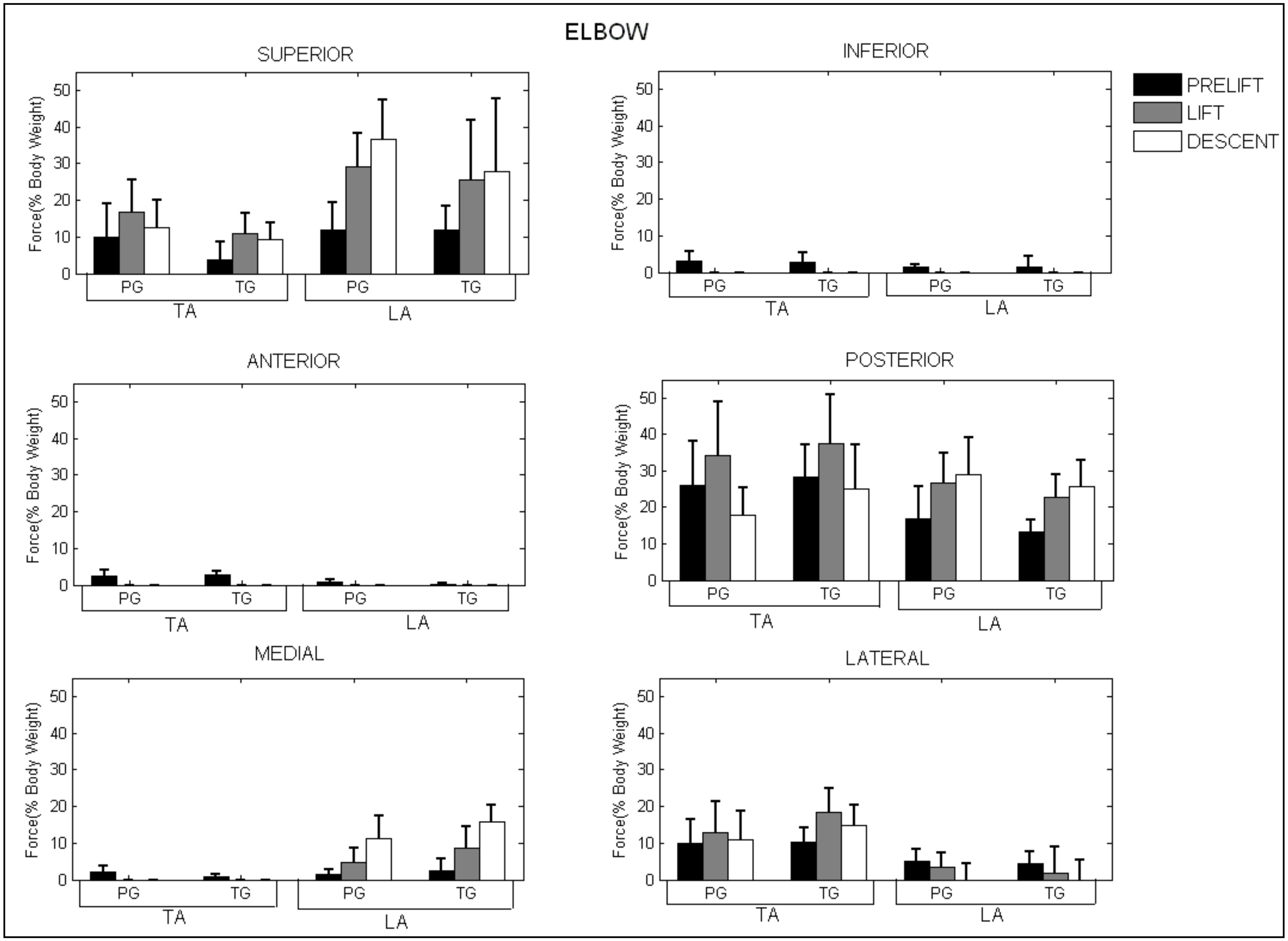


Figure 9. Peak 3D reaction forces at the hand across transfer phases for each arm.

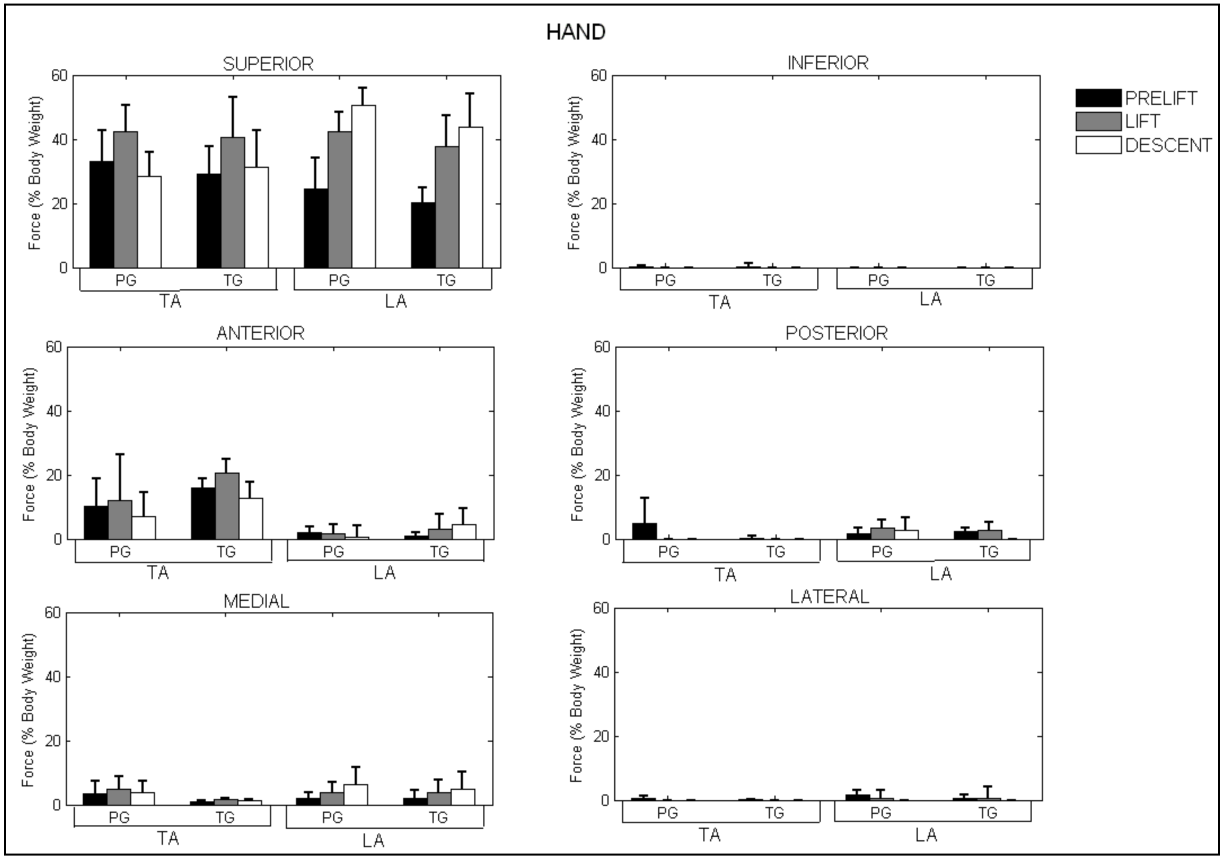


Table 4. Mean (SE) peak moments at the UE joints and forces at the shoulder, elbow and hand during the SPT. Effect Size computed for significant and non- significant results of main effect of group averaged across arm and phases. Group

differences of the variables not influenced by arm and transfer phases are tabulated. [NOTE: Symbols: ¥ - significance p<0.05; * - trends 0.1 ≤ p ≤ 0.05; NA – indicates the variables that could not be included due to the influence of arm and phase on group difference.]

		Main Effect of Group		
MOMENT (%m)		PG	TG	ES
Shoulder	Flexion +			
	Max	1.58 (0.65) ¥	3.91 (0.89) ¥	0.99
Elbow	Flexion +; Extension -			
	Max	1.66 (0.48) ¥	-0.13 (0.66) ¥	1.03
	Min	-1.2 (0.45) *	-2.48 (0.62) *	0.78
Wrist	Flexion +			
	Max	2.10(0.88)	1.83(1.19)	0.09
FORCE (% BW)				
Shoulder	Anterior +; Posterior -			
	Max	NA	NA	NA
	Min	-18.84(1.71)	-16.72(2.33)	0.34
Elbow	Anterior +; Posterior -			
	Max	9.21(1.12)	9.53(1.52)	0.08
	Min	NA	NA	NA
	Medial +; Lateral -			
	Max	1.13 (0.66)	1.34(0.90)	0.09
	Min	-6.5(0.94)	-7.03(1.28)	0.16
Hand	Anterior +; Posterior -			
	Max	5.54(1.26)*	9.66(1.72)*	0.91
	Min	-1.18(1.13)	2.01(1.54)	0.78
	Medial +; Lateral -			
	Max	3.92(0.67)	2.44(0.92)	0.61
	Min	-0.44(0.45)	-0.17(0.61)	0.17

Table 5. Mean (SE) peak moments at the UE joints and forces at the shoulder, elbow and hand, in the trailing and leading arm, during the SPT. Group differences within each arm averaged across phases, for the variables not influenced by transfer phases are tabulated.[NOTE: Symbols: ¥ - significance p<0.05; * - trends 0.05≤p ≤ 0.1].

		Simple Main Effect of Arm by Group					
		Trailing Arm			Leading Arm		
MOMENT (% m)		**PG**	**TG**	**ES**	**PG**	**TG**	**ES**
Shoulder	Flexion +						
	Max	4.13(1.18)	8.61(1.60)	1.06	-.976(0.54)	-0.79(0.73)	0.10
Elbow	Flexion +; Extension -						
	Max	1.87 (0.96)	-0.94(1.30)	0.81	1.45(0.39)	0.67(0.53)	0.56
	Min	-1.69(0.87)	-3.56(1.18)	0.60	0.7(0.41)	1.41(0.56)	0.48
Wrist	Flexion +						
	Max	0.73(1.43)	-2.69(1.95)	0.66	3.47(1.05)	6.34(1.43)	0.76
FORCE (% BW)							
Shoulder	Anterior +; Posterior -						
	Max	-2.28(1.3)	-1.12(1.77)	0.25	-10.8(1.72)	7.04(2.35)	2.87
	Min	-13.32(1.81)	-10.00(2.46)	0.51	-24.37(2.09)	-23.43(2.85)	0.12
	Medial +; Lateral -						
	Max	3.07(2.64) ¥	-7.23(3.59) ¥	1.08	5.51(2.32) ¥	18.92(3.17) ¥	1.60
	Min	-10.30(4.19) ¥	-27.56(5.71) ¥	1.14	-2.82(2.30)	-5.75(3.13)	0.35
Elbow	Anterior +; Posterior -						
	Max	-7.24(1.2)	9.01(1.63)	3.76	11.18(1.23)	10.06(1.68)	0.25
	Min	-26.07(2.89)	-30.21(3.93)	0.40	-24.2(1.94)	-20.56(2.64)	0.52
	Medial +; Lateral -						
	Max	-3.65(1.23)	-6.24(1.68)	0.58	5.92(1.00)	8.92(1.36)	0.83
	Min	-11.20(1.64)	-14.52(2.24)	0.56	-1.80(1.05)	0.45(1.43)	0.59
Hand	Anterior +; Posterior -						
	Max	9.69(2.29)	16.42(3.13)	0.81	1.39(0.68)	2.89(0.93)	0.61
	Min	0.22(2.14)	5.40(2.91)	0.67	-2.59(0.59)	-1.38(0.81)	0.57
	Medial +; Lateral -						
	Max	3.95(0.83) *	1.21(1.14) *	0.91	3.88(0.9)	3.67(1.23)	0.06
	Min	1.02(0.41)	0.44(0.56)	0.39	-1.45(0.69)	-0.097(0.94)	0.54

2.3.2.3 Shoulder

Shoulder flexion was significantly higher for the TG (3.91 ± 0.90) compared to PG (1.58 ± 0.65), averaged across arms and transfer phases ($p = 0.048$, ES = 0.99; Figure 6 & Table 4). Posterior forces at the trailing shoulder tended to be higher for the PG compared to the TG during the prelift phase ($p = 0.081$, Figure 7). In agreement with our hypothesis the TG tended to have larger forces in the lateral and medial directions at the shoulder. Medial forces were significantly higher for TG compared to PG in the leading arm averaged across phases ($p = 0.003$, ES = 1.60; Table 5). The TG had primarily lateral forces acting at the trailing shoulder, which was significantly higher than the PG averaged across phases ($p =0.025$, ES = 1.14; Figure 7).

2.3.2.4 Elbow

The PG had significantly higher elbow flexion moment compared to the TG, averaged across both arms and phases of the transfer ($p = 0.042$, ES = 1.03; Table 5). Higher elbow extension moment was observed in the TG (2.48 ± 0.62) than the PG (1.20 ± 0.45), averaged across the arms and phases of the transfer ($p = 0.11$, ES = 0.78; Figure 7). Superior and posterior forces were the two largest component forces acting at the elbow for both groups, Figure 5. Effect sizes revealed that posterior forces were higher for the PG compared to the TG (ES = 0.52, Table 5) and a large effect size (ES = 0.83, Table 5) was associated with the medial forces tending to be higher for the TG compared to the PG in the leading arm.

2.3.2.5 Wrist and hand

Wrist flexion moment did not significantly vary between the two groups (Figure 6). However large effect size (ES = 0.76) was associated with wrist flexion being larger for the TG compared to the PG, in the leading arm (Table 5). The forces at the hand were predominantly in the superior direction for both TG and PG. Anterior forces were the second largest component reactive forces. These forces tended to be higher for the TG compared to PG averaged across arms and phases of the transfer ($p = 0.069$, ES= 0.91;Figure 9). An additional trend relates to the trailing hand medial forces being higher for the PG compared to the TG ($p = 0.068$, ES = 0.91; Table 5).

2.4 DISCUSSION

This study is one of the first to our knowledge to perform a kinetic analysis of all three upper extremity joints: shoulder, elbow and wrist, for SPTs across SCI groups. Our kinetic results are in general agreement with previous studies (11, 20). Similar to their results we found that the largest component forces at the shoulder while performing SPTs were the superior and posterior forces for both groups. Similar to Gagnon et al (39), we found the elbow flexion and extension moments to be of smaller amplitudes in comparison to the shoulder flexion moments for both groups.

2.4.1 Group comparison

2.4.1.1 UE strength

Bernard et al (40) examined isokinetic peak torques and the influence of lesion level in a group of wheelchair athletes with paraplegia. They found that the group with lower level lesions had significantly higher external rotation strength compared to those with higher lesion levels. Our results were similar in nature where we found the PG had significantly higher external rotation strength compared to the TG. The level of injury and the absence of innervated triceps describe the significant strength differences observed between the groups (PG>TG) at the elbow and wrist(6). Our experimental set up constrained all transfers to start with the left arm leading. Based on discovering the PG having higher elbow flexor strength and the TG having lower elbow flexor strength in their right arm compared to their left respectively in addition to TG may explain some of differences in transfer strategies and joint kinetics between the two groups.

2.4.1.2 Kinematic and kinetic differences

We found primarily kinetic differences between the groups in contrast to kinematic differences. The post hoc power observed for the statistical tests on the kinematic variables ranged from 10% - 60%. The large variance in the data and the low power indicates a larger sample size would have been required to detect statistical significance.

Individuals with paraplegia have better core strength, when compared with persons with tetraplegia, which allows them to better support themselves in a posturally stable position (41). Individuals with tetraplegia have to work a lot more to overcome the effects of gravity to maintain both seated and dynamic postural stability, compared to the PG (21). Therefore the reaction forces imposed at the shoulder are greater for the TG compared to the PG as is evident by the elevated shoulder flexion moments and anterior forces at the shoulder, elbow and hand, in contrast to the large posterior forces at the shoulder for the PG. The large lateral forces at the trailing shoulder and medial forces at the leading shoulder in the TG compared to the PG are indicative of the higher rotational demands at the shoulder, to maintain dynamic postural stability when executing the SPT (10).

Similar to previous literature superior forces were the principle forces acting at the hand (20). It has been suggested that the increase in the horizontal forces at the hand as the SPT progresses may serve to prevent forward fall of the trunk and assist with the lift (20). The TG displayed higher anterior forces in comparison to the PG, not only at the hand but also at the elbow and shoulder, as indicated by the large effect sizes. The presence of large anterior forces at the hand could indicate that the TG work harder to prevent a forward fall when performing an SPT as compared to the PG.

In accordance with our hypothesis, larger shoulder & wrist flexion moments were found in the TG compared to the PG. Contrary to our hypothesis; larger elbow extension moments in the TG were found compared to the PG. This alternative finding could indicate that compared to the PG group, TG relied more on the shoulder flexors (e.g. anterior deltoids) as a compensatory way to stabilize and passively assist with elbow extension (23). This can be further supported by the strength differences found between the groups, with increased elbow and wrist flexor/extensor strength for the PG in comparison to the TG. As a closed chain system is formed with the hand/wrist fixed on the surface, finding that wrist flexion moments also increased in this group further supports compensatory elbow stabilization (28). We based our hypothesis related to elbow extension moments on a study by Harvey et al (9), wherein they found the execution of lift in a weight relief maneuver involved generation of active shoulder flexion moment and passive wrist flexion moments to 'lock' the flexed elbow. However with the more pronounced dynamic component of an SPT (lifting of the body and sideways weight shift) the reliance on shoulder and wrist flexors to generate the closed chain passive extension at the elbow throughout the movement is more critical.

2.4.2 Joint comparison

Superior forces were largest at the hands compared to the shoulder and elbow as observed in previous literature(10). As the forces translate upwards the posterior forces dominate over the superior forces followed by the lateral forces at the shoulder. Shoulder flexion/adduction and wrist flexion moments were the most dominant moments followed by elbow extension moments similar to Gagnon et al(39). The advantage of applying inverse dynamics to compute forces and moments at each of the joints assists in identifying differing characteristics of movement patterns between the two groups.

2.4.3 Phase & arm comparison

Overall we found that the leading arm tended to bear more forces compared to the trailing arm whereas moments generated at the UE joints of the trailing arm were higher than those generated at the joints of the leading arm. Interpretation of the roles of the arms are still unclear and comparisons to other studies are confounded based on the experimental setup being used to record the data (10). Based on our observation the arms weight bear in an alternative manner as the person progresses through the SPT. This is highlighted more notably through the delineation of the SPT into phases. At the trailing arm the most significant amount of weight bearing occurs during the lift phase (10). However for the leading arm the weight bearing occurs during the lift and descent phases and in some cases the weight bearing during the descent phase exceeds the lift phase. This can be explained based on the leading arms role in assisting with weight shift that occurs just prior to landing and assists with dynamic postural stabilization (42). The findings support the clinical recommendation, from the clinical practice guidelines on preservation of upper limb function, of varying the leading arm whenever possible (5).

2.4.4 Clinical implications

Finding significantly higher posterior forces acting at the shoulder in PG compared to the TG may make this group more susceptible to problems such as posterior shoulder instability, capsulitis and tendonitis (10, 43, 44). Mercer et al (45) investigated the relationship between shoulder kinetics and shoulder pathology in MWC users during WC propulsion, and found that individuals who presented large lateral forces, were more likely to have coracoacromial (CA) ligament thickening. This condition contributes to narrowing of the supraspinatus outlet which has been recognized as a cause of impingement syndromes and rotator cuff diseases (46, 47).

At the elbow attention must be given to not only superior forces but also the shear forces acting at the elbow which may prove detrimental for both groups if elbow instability is a pre-existing condition (39, 48). The increase in anterior deltoid activity used to stabilize the elbow joint during depression style weight reliefs likely increases the chances of shoulder impingement in the TG (28). The differences in the kinetics at the upper extremity joints highlight the different injury mechanisms that can occur in each group.

These study findings may be useful for tailoring strength training/exercise programs to the individual based on their LOI. As for the TG group exercise and strength training programs focusing on strengthening the anterior deltoids sternal pectoralis major are crucial due to the dependence on those muscles to enable a successful lift. Additionally overall for both groups the strength training of the larger thoracohumeral muscles in non weight bearing positions will aid in shoulder pain management (21).

2.4.5 Limitations

That all subjects with tetraplegia were incomplete may have made it more difficult to find differences between the two groups as the paraplegic group included both incomplete and complete injuries. An ASIA motor score would have provided a more complete understanding of sensory-motor functioning of each group. Also we examined the simplest of transfer scenarios that being a level height transfer to plain surface. Transfers to higher/lower heights and different types of surfaces may have shown greater differences between groups. Despite these limitations we still found statistically significant differences in SPT performance. In order to allow for enhanced clinical interpretation the effect size for group differences were estimated. We based parts of our discussion and conclusions on results that were not significant but had a high effect size (>0.80) indicative of a meaningful difference that may be statistically detectable with a larger sample size. Direct comparison to previous literature is difficult due to the differences in experimental setup with particular regard to the fact that participants in our study transferred from their own chair.

2.5 CONCLUSION

The results of the study shed light on potential risk of injuries that may occur for each individual group. Results suggest alternative strength training programs be considered for preservation of UE function for each group separately. Future studies involving additional variables such as trunk kinematics and upper extremity kinematics will assist in understanding the differences in strategies used between the two groups. Muscle activity recordings from dominant muscles involved in performance of SPTs would provide a greater understanding of the moment data presented, and should be considered for future studies.

3.0 UPPER LIMB KINETIC ANALYSIS OF SITTING PIVOT WHEELCHAIR TRANSFER TECHNIQUES IN INDIVIDUALS WITH SPINAL CORD INJURY

3.1 INTRODUCTION

Due to lower limb paralysis, people with spinal cord injury (SCI), commonly have upper limb pain, due to their high reliance on their arms to perform activities of daily living such as wheelchair propulsion, pressure relief and transfers (49). Being able to transfer independently is a key factor to achieving an optimal level of independence. Therefore any loss of upper limb function will severely affect overall functional mobility and independence. Research has found that on average a person with SCI performs 15-20 transfers per day (10). Based on the frequency of transfers performed and the magnitudes of upper extremity (UE) loading, transfers have been found to be a large contributor to development of UE pain. Studies on UE pain amongst people with SCI have found high prevalence of shoulder pain (30-60%), elbow (22-45%) and wrist (40-66%) (5). Research has indicated that the onset of upper extremity pain/damage is functionally and economically equivalent to an SCI of higher neurological level (28). Therefore preservation of upper extremity function is of utmost importance for an individual to maintain a good quality of life and community participation.

Earlier studies have described general movement strategies, upper limb kinematics, and muscle activity for long-sitting transfers which are transfers where the legs are extended out in front of the body (Allison *et al.* (7) and Gagnon et al.(17)). Allison et al.(7) described two general movement strategies used when performing long sitting transfers: *rotational strategy* (head moves in an opposite direction to the pelvis) and *translational strategy* (head and pelvis move simultaneously in the same direction). When viewed from the sagittal plane, individuals performing the *rotational strategy* leaned forward during the transfer and those using the *translational strategy* kept their trunk more upright during the transfer. The *rotational strategy* is analogous to what clinical practice refers to as the 'head-hips' relation. It is often taught to patients with weak triceps and/or those with high levels of trunk involvement and can be applied to other types of transfers as well.

Sitting-pivot type transfers are most common among individuals with SCI who are unable to stand unsupported (6). The execution of an SPT requires the individual to primarily lift and move their body with their UEs. Individuals usually scoot forward in their chair prior to beginning the transfer. This is followed by hand placement: one hand is placed on a target surface (leading hand) while the other hand is placed on a stable surface (part of the wheelchair or a grab bar) close to the individual (trailing hand). From this safe starting position the individual begins the transfer by a simultaneous flexion and rotation of the trunk and head first in the forward direction and then sideways while lifting their body of the wheelchair followed by a pivoting motion to the target surface (10). The transfer is complete when the person lands on the target surface and regains seated postural stability. In theory, if the head-hips relation is used more momentum can be generated to facilitate moving the body and research on a related task (pressure relief pushups) suggests that the forward-flexed trunk position is ideal for engaging sternal pectoralis major and latissimus dorsi muscles (17). This muscle substitution may help transfer the body weight between the leading arm (arm reaching to new surface) and trailing arm (arm behind during move to new location) with less loading of the glenohumeral joint thereby reducing the risk of rotator cuff impingement (13, 17).

Poor transfer techniques have been identified as a risk factor associated with loss of function of the UEs (50). Although some general recommendations exist regarding optimal setup (e.g. performing level height transfers when possible), hand, arm and foot placement during transfers there is little evidence to support all of them (5, 10, 21). In a recent study we investigated the effects on upper limb biomechanics for several transfer techniques which varied on two factors: hand placement and trunk flexion, in an unimpaired population (51). The three techniques that were compared included two techniques which used the head-hips relation where the trunk was forward flexed and the leading arm abducted and away from the body (HH-A); again with the leading arm close to the body and internally rotated (HH-I), while the third technique mimicked the translational strategy where the trunk remained upright and the leading arm abducted and away from the body (TU). We compared how trunk flexion and hand placement affected the load distribution at the shoulder, elbow and hand through the different transfer techniques executed by the participants. We found that the head hips transferring techniques reduced superiorly directed forces across all three UE locations, in comparison to the trunk upright style of transferring. Although there was a reduction in superior forces with the head hips transferring techniques, the moments at the shoulder (flexion/extension and external rotation) were higher in comparison to the TU strategy. Although technique differences were found no conclusive information could be drawn that could be generalized to a SCI population. Therefore the purpose of this study was to evaluate the same techniques in a cohort of manual wheelchair users with SCI. The objectives of the study were:

1. To compare UE joint (shoulder, elbow and wrist and hand) kinetics for the above mentioned techniques: HH-A, HH-I and TU
2. To compare UE joint (shoulder, elbow and wrist and hand) kinetics for each of the taught transfer techniques to the participant's own preferred method of transferring.
3. Classify the participants self-selected technique based on data obtained for the taught techniques.

We hypothesized that the transfer techniques using either of the head hips techniques will result in reduced peak superiorly directed forces at the shoulder and elbow compared to the Trunk upright transfer technique (analogous to translatory strategy).

Role of LE's during wheelchair transfers has been highlighted albeit minimally investigated. Although motor function is absent in the lower limbs, when positioned appropriately they can weight bear and also assist in stabilizing the trunk, which makes it easier to pivot on the arms (5, 6). Tanimoto et al (19) recorded hand and feet kinetics in a group of individuals with SCI (2 with tetraplegia and 11 with paraplegia) while performing level transfers. They found that the vertical reaction forces at the feet increased with increase in trunk inclination. Gagnon et al (20) explored the role of the feet during level and non-level transfers in a group of 12 participants with paraplegia. Both studies found that the feet bore about 25% BW during the lift phase of the transfer. *We believe that the overall forces will be lower at the shoulder due to greater off loading through the lower extremities (LE's) during the head-hips transfers as compared to the trunk upright transfer strategy.*

Finley et al. (9) studied scapular kinematics in 23 participants with paraplegia of which UE impingement syndrome was present in 10 participants and 13 were asymptomatic. The results were similar to the study conducted by Nawocenski (18) in the asymptomatic group.

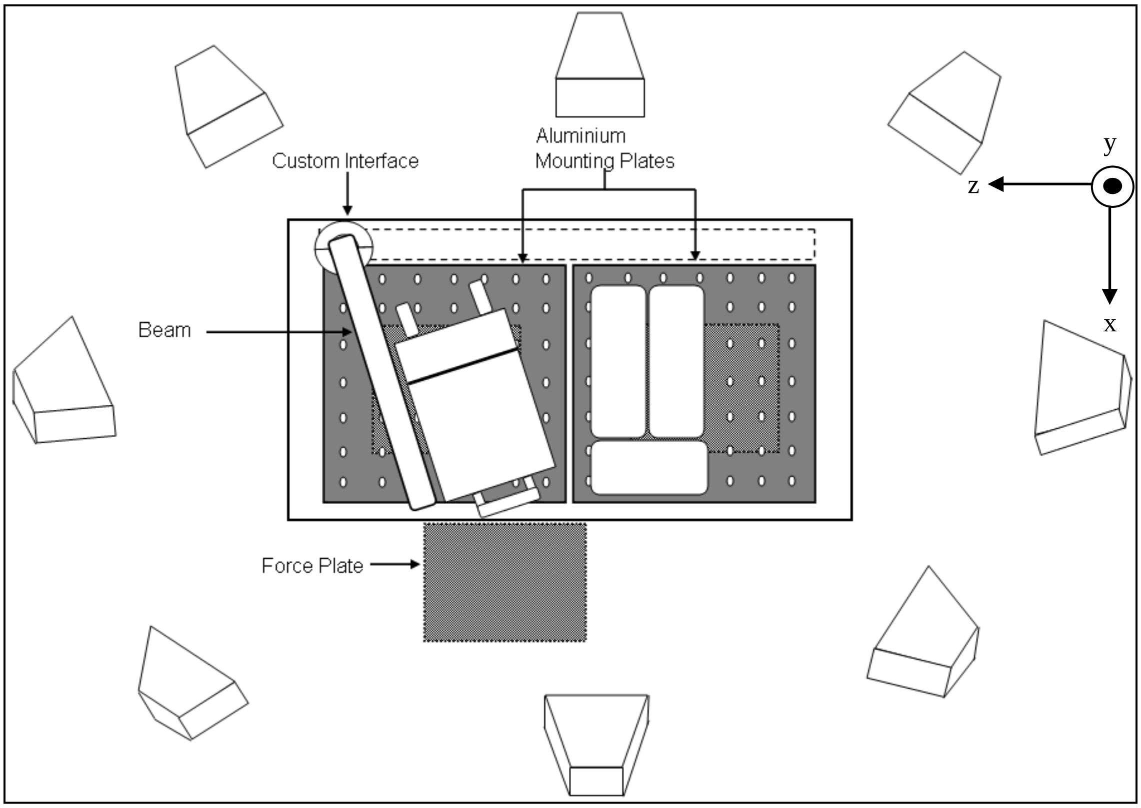
People with UE impingement syndrome were found to however have reduced thoracic flexion, increased scapular upward rotation and reduced humeral internal rotation compared to the asymptomatic group. This movement pattern adopted by WC users who already have the presence of UE pain could be a potential preservation technique to reduce impingement of the greater tuberosity under the acromion thereby allowing the individuals to continue with functional activity. Individuals who are further out from injury may have optimized their preferred method of transferring thereby reducing shoulder joint loading and preserving functional capability in order to continue performing ADL's. *This leads us to believe that persons for whom pain does not interfere with independent transfers will transfer using sub-optimal self-selected techniques and that using one of the head-hips taught techniques will show reduced loading compared to the self-selected technique*.

3.2 METHODS

3.2.1 Subjects

This study received ethical approval from the Department of Veterans Affairs Institutional Review Board. After reading and providing informed consent, twenty subjects (19 male, 1 female), volunteered to participate in this study. The inclusion criteria were: spinal cord injury C4 level or below that occurred over one year prior to the start of the study, able to independently transfer to/from a manual wheelchair without human assistance or assistive devices, over 18 years of age, and free from upper extremity pain that influenced their ability to transfer.

Figure 10. Wheelchair (left) and tub bench (right) shown secured to the aluminum mounting plates of the base frame. The custom interface consists of the load cell and a beam that can be positioned anywhere along base frame.

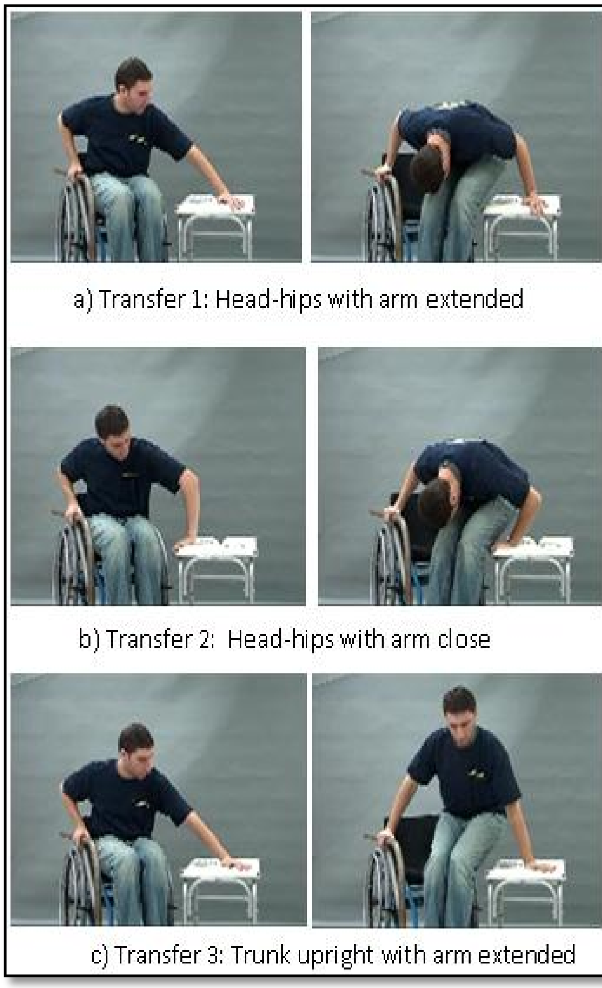


3.2.2 Experimental protocol

Participants used their personal wheelchairs to transfer to and from a bench. For all transfers the wheelchair was positioned and secured at a comfortable angle from an adjustable height tub bench as shown in Figure 10. The bench was adjusted to be level with the wheelchair seating surface. The platform contains three force plates (Bertec Corporation, Columbus, OH), one beneath the wheelchair, one beneath the tub bench and one located below the feet (34). The wheelchair and bench were secured to the platform. A steel beam attached to a 6-component load cell (Model MC5 from AMTI, Watertown, MA) was positioned to simulate a wheelchair armrest. Reflective markers were placed on the subjects C7, T3 and T8 vertebrae, sternum, xiphoid, right and left acromion processes, 3rd metacarpalphalangeal joints, radial and ulnar styloid processes, and lateral epicondyles. The coordinates of the markers were recorded based on a global reference frame using a sixteen camera three-dimensional motion capture system (Vicon Peak, Lake Forest, CA). Several anthropometric measurements were recorded such as: axiliary arm, wrist, fist and elbow circumference, upper arm and forearm length.

All transfers began with the left arm leading and moving the body from the wheelchair to the bench. For the first transfer, subjects were instructed to perform a SPT as they normally would from their wheelchair to the adjacent level tub bench. For this transfer, they could place their left hand anywhere on the bench and right hand on the steel beam (height of wheelchair arm rest). The other three transfers were performed in random order. Prior to performing each of the transfer techniques, subjects were shown an instructional video on how to complete the transfer. For the HH-A transfer, subjects were instructed to place left hand on the far circular target on the bench, right hand on the target on the force beam, and transfer leaning their trunk forward as far as possible while moving their buttocks toward a large target on the bench while moving their head in the opposite direction (Figure 11.a). The HH-I transfer required the same instructions except that the left hand was placed on a near circular target on the bench, with the left arm internally rotated (Figure 11.b). For the Trunk Upright (TU) transfer subjects were instructed to place their left hand on the far target of the bench, right hand on the target on the force beam, and transfer with their trunk upright while moving their buttocks toward the large target on the bench; while moving their head in the same direction (Figure 11.c). Subjects practiced each technique until they were confident they could perform the transfer based on instructions provided. Each transfer technique was performed three times and kinetic and kinematic data were recorded synchronously at 360 Hz and 60 Hz respectively, for the length of the transfer.

Figure 11. Still photos from the transfer instruction video for the two types of head-hips relation transfer and translational transfer.



a) Transfer 1: Head-hips with arm extended

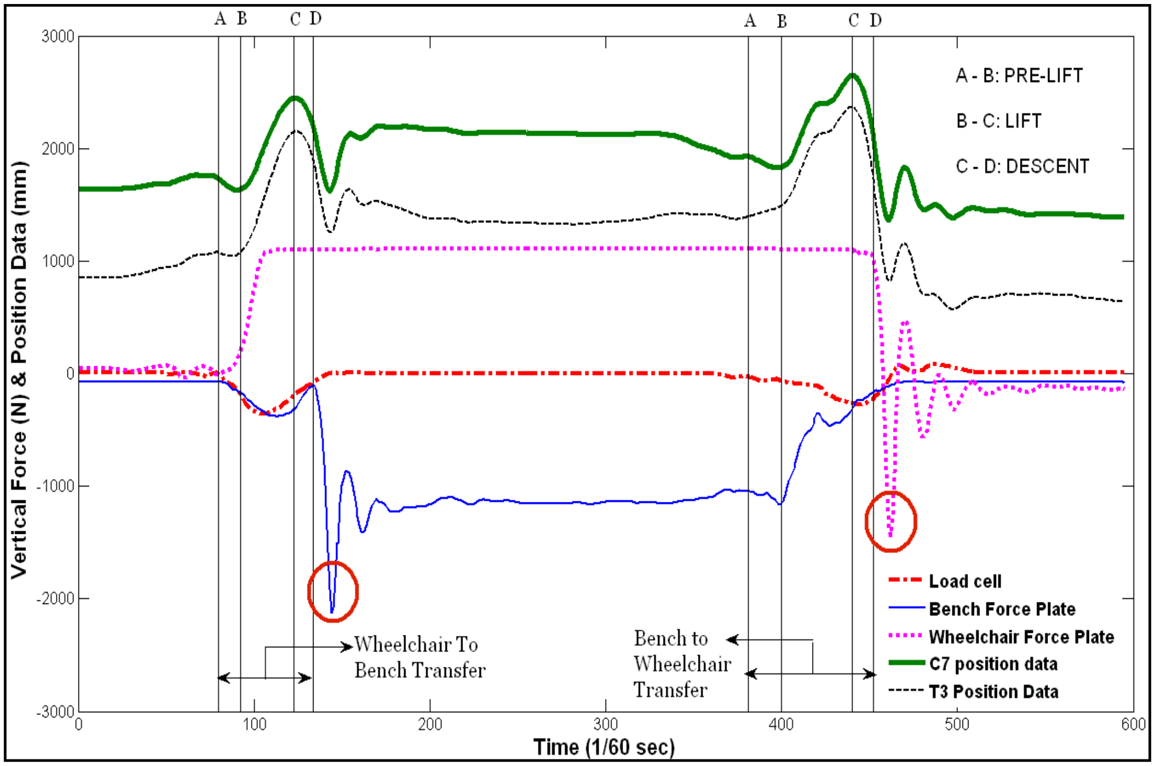
b) Transfer 2: Head-hips with arm close

c) Transfer 3: Trunk upright with arm extended

3.2.3 Data analysis

Kinetic, kinematic, and anthropometric data were entered into an inverse dynamic model to calculate the 3D net shoulder and elbow joint forces and moments. Kinetic data were down sampled to 60 Hz, to align with the kinematic data. Both kinematic and kinetic data were filtered with a 4th order zero-lag Butterworth filter (cut off frequency of 5 Hz and 7Hz respectively). The inverse dynamic model used was based on the general rigid-link segment model using a Newton-Euler method and a variable degree of freedom body co-ordinate system (35). The local coordinate systems were approximated on the recommendations by the International Society of Biomechanics (ISB) (36). Trunk movement was computed using a Cardan angle sequence (ZXY, along a anterior/posterior axis [*x*], superior/inferior axis [*y*], and medial/lateral axis [*z*] acting to flex/extend [*z*], lateral rotation [*x*], and axial rotation [*y*] with respect to the laboratory coordinate system) and shoulder movement using a Euler angle sequence (YXY along plane of elevation [*y*], amount of elevation [*x*], and internal/external rotation [*y*] for the shoulder coordinate system acting with respect to the trunk coordinate system). The vertical reaction force from the force plate under the bench and the grab bar were used to determine the start (absolute value of vertical force > 0 at the grab bar and bench) and the end (determined prior to the generation of a large spike in the vertical force at the bench) of the transfer. The SPT was delineated into three phases: prelift, lift and descent (Figure 12). The vertical forces from the force plate under the tub bench, wheelchair and the grab bar (wheelchair side) were superimposed onto one plot to analyze the phases of transfer. Trunk motion, represented by the C7 and T3 markers, were added to the force data to assist with the delineation of phases of transfer (37).

Figure 12. Representative self selected transfer recordings. Position and kinetic data were combined to delineate the sitting pivot transfer into three phases: pre-lift, lift and descent.



3.2.4 Outcome measures

Analysis was conducted for transfers from the wheelchair to tub bench. For each trial 3D component and net resultant force were calculated for the shoulder, elbow and hand for both arms. Flexor/extensor, abduction/adduction and internal /external rotation moments at the shoulder, flexor/extensor moments at the elbow and wrist were calculated for both leading and trailing arms. Net resultant feet forces were collected on a subset of the sample (n=11), as an upgrade to add another force plate to the transfer system was made after starting the study. Additionally trunk linear and angular velocities for the flexion and axial rotation were also computed, with the centre of mass defined for the upper trunk using the right/left acromion and the xiphoid markers (52). The maximum and minimum of the aforementioned variables were identified for the lift phase of the transfer. Maximum values of trunk flexion, leading shoulder abduction angles and distance of the third metacarpal on the leading hand from global zero were identified for the prelift phase of the transfer and used to test for instruction adherence. Transfer duration was also determined from the beginning of the prelift to the end of descent phase. Variables were computed using Matlab (Mathworks, Inc., Natwick, MA) and averaged over the three trials for each transfer technique.

Categorization of the self selected technique was based on two variables: leading hand position (i.e. distance from global zero) and the amount of trunk flexion. The leading hand position was classified into two positions namely:

1. Near: Leading hand distance for the self selected technique was close to the mean within one standard deviation of the leading hand position recorded for the HH-I technique
2. Far: Leading hand distance for the self selected technique was close to the mean within one standard deviation of the leading hand position recorded for the HH-A or TU techniques.

The second variable that assisted with categorization of the self-selected technique was trunk flexion, which was classified as:

1. Head-Hips: Trunk flexion angle when transferring with the self-selected technique was close to the mean within one standard deviation of either of the HH techniques.
2. Trunk Upright: Trunk flexion angle when transferring with the self-selected technique was close to the mean or within one standard deviation of the TU technique.

3.2.5 Statistical analysis

Group means and standard deviations were computed. One way repeated measures ANOVA for the resultant feet forces, trunk flexion, trunk linear/angular velocities, leading shoulder abduction, leading hand position and transfer duration were carried out. Differences between transfer techniques and between the leading and trailing arm biomechanical variables were analyzed using a Two- Way within subjects ANOVA. Main and interaction effects were investigated using simple pair wise comparisons and a post hoc Bonferroni correction. Differences across the self-selected transfer techniques after classification were analyzed using Mann Whitney U tests due to the small sample size, on group characteristics (age, height, weight and years since injury) and net resultant forces at the upper limb locations for both the trailing and leading arm. The statistical tests were performed using SPSS statistical software (SPSS Inc., Chicago, IL), with the level of significance set to 0.05.

3.3 RESULTS

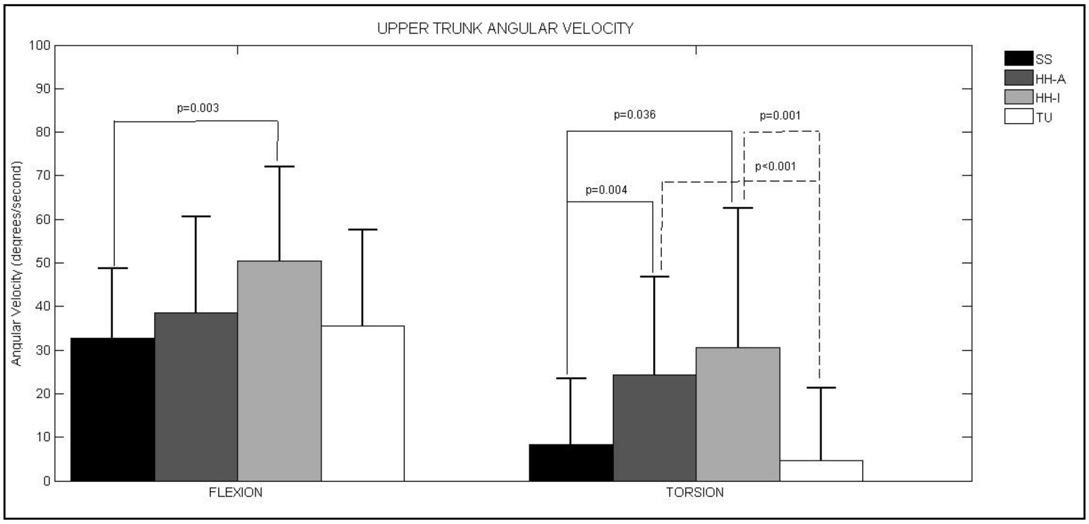
3.3.1 Demographics

Of the 20 participants who took part in the study, 18 (17 male and 1 female) could complete all SPT techniques. Data was missing for one out of the three taught techniques for the two persons excluded from the analysis. One person could not physically complete the taught technique, while the second person owing to technical difficulties with the set up, the data could not be collected on one of the taught techniques. The group consisted of 12 persons with paraplegia (6 with complete & 6 with incomplete injury) and 6 with tetraplegia (all with incomplete injury). The group mean (standard deviation) of age, height, weight and years since injury were 36.83 (10.5) years, 1.70 (0.4), 76.21 (20.0) and 13.72 (7.6) years.

3.3.2 Instruction adherence

The peak trunk flexion angles were significantly higher for the Head hips techniques (HH-A: 51.48 ± 14.41 & HH-I: 48.74 ± 12.85) compared to the TU (35.48 ± 13.57) technique ($p=0.001$ and $p <0.001$ respectively). The leading hand was positioned significantly further away for the HH-A (1.74 ± 0.08 m) and TU (1.73 ± 0.08 m) techniques compared to the HH-I (1.56 ± 0.08 m) technique (p-values< 0.001). Leading shoulder abduction angles were not significantly different across taught techniques: HH-A (71.59 ± 13.26), HH-I (70.14 ± 15.22) and TU (66.96 ± 3.47). These findings affirm that the taught techniques varied on namely: trunk flexion and hand placement.

Figure 13. Trunk Angular velocity in the forward flexion and torsion directions across techniques



3.3.3 Technique comparison

Statistical differences in the shoulder, elbow, wrist and hand kinetics for the taught techniques and the preferred versus the taught techniques have been summarized in Table 10 and Table 11 respectively and are described in more detail below.

3.3.3.1 Trunk linear and angular velocities

Upper trunk flexion angular velocity was significantly higher for the HH-I technique compared to the SS technique ($p = 0.003$,Figure 13). Upper trunk torsion velocity was significantly higher for the head-hips techniques compared to the TU transferring technique (p values <0.001, Figure 13). Additionally the upper trunk torsion velocity was significantly higher for the head-hips techniques compared to the SS style of transferring (HH-A: $p = 0.004$ and HH-I: $p = 0.036$ respectively, Figure 13).

3.3.3.2 Feet kinetics

No significant differences were observed for net resultant feet force (% Body Weight) across the 4 techniques- SS: 24.59 ± 15.94, HH-A: 27.77±15.77, HH-I: 24.99 ± 14.51, and TU: 25.77± 14.45.

3.3.3.3 Temporal characteristics

The transfer durations were significantly longer for the HH-A (2.57 ± 1.69 seconds) technique compared to the TU (2.13 ± 1.44 seconds) technique of transferring, $p < 0.001$. No significant differences in transfer durations were found for the HH-I (2.26 ± 1.34 seconds) and the SS techniques (2.29 ± 2.14 seconds) compared to the remaining taught techniques.

Table 6. Mean (± standard error) of the peak forces at the shoulder, elbow, wrist and hand averaged across both arms. The variables found to have significant interaction effect between technique and arm are denoted by '¥'. Simple main effects of these variables are shown in Table 2 and 3.

Moment (Nm)		**TECHNIQUE DIFFERENCES AVERAGED ACROSS BOTH ARMS**			
		SS	**HHA**	**HH-I**	**TU**
Shoulder	Internal rotation	33.86 (4.82)	37.10(4.45)*	24.69 (4.75)$^{* +}$	41.21 (6.03)$^{+}$
Wrist	Flexion	18.72 (7.35)*	35.78 (9.91)	5.36 (7.77)$^{+}$	44.10 (9.55)$^{* +}$
Force (N)					
Shoulder	Superior	147.44 (14.23)*	98.68 (13.58)*	114.84 (16.44)	133.09 (13.21)
	Posterior ¥	155.27 (12.52)	162.40 (13.90)	185.05 (20.82)*	147.65 (12.63)*
Elbow	Superior	165.93 (13.80)*	128.27 (11.23)$^{* +}$	135.90 (12.24)	167.47 (12.50)$^{+}$
	Lateral ¥	65.09 (8.11)	75.67 (9.90)*	52.00 (8.04)$^{* +}$	77.22 (10.53)$^{+}$
Hand	Medial ¥	29.52 (4.89)$^{* \$}$	47.97 (7.15)$^{* +}$	31.65 (4.00)$^{+ \#}$	46.11 (5.23)$^{\# \$}$

Note: Symbols *, +, $ and # indicate p < 0.05

Table 7. Means (± 1 standard deviation) the peak moment across the upper extremity joints.

Moment (Nm)		**Simple Main Effect of Technique by Arm**									
		Trailing Arm					Leading Arm				
Joint	Component	SS	HH-A	HH-I	TU	p-value	SS	HH-A	HH-I	TU	p-value
SHOULDER	Abduction+, Adduction -										
	Max	-9.55 (35.75)	-16.47 (37.28)	-20.00(23.52)	-4.23(46.32)	NA	5.38 (21.66)	-0.71 (20.70)	10.09 (12.62)	-2.62 (26.05)	NA
	Min	-47.94^{+} (35.20)	-80.38$^{+ \#}$ (45.55)	-71.27* (25.72)	-48.04$^{* \#}$ (32.84)	+: 0.044 #: 0.010 *: 0.021	-14.33^{+} (23.21)	-27.20 (27.26)	-12.04 (10.33)	-23.21^{+} (23.99)	+: 0.001

Table 8. Means (± 1 standard deviation) of the peak forces at the hand, elbow and shoulder.

FORCE (N)		Simple Main Effect of Technique by Arm									
		Trailing Arm					Leading Arm				
Joint	Component	SS	HH-A	HH-I	TU	p-value	SS	HH-A	HH-I	TU	p-value
Shoulder	Medial +, Lateral-										
	Max	-36.89 (120.00)	-27.50 (131.86)	-15.94 (129.89)	-31.66 (125.41)	NA	99.59 (92.21)	115.96 (75.36)	103.36 (113.42)	112.10 (81.86)	NA
	Min	- 164.64 (166.35)	-172.59^{+} (189.13)	-144.61^{*+} (172.40)	-167.39* (186.55)	*: 0.046 +: 0.03	9.68 (61.32)	14.13 (56.75)	12.84 (84.11)	15.91 (46.76)	NA
Elbow	Anterior +; Posterior -										
	Max	-124.56 (70.17)	-133.81 (65.50)	-132.83 (63.64)	-104.05 (51.65)	NA	-104.05 (51.65)	-109.46 (62.54)	-108.63 (48.00)	-78.31 (50.19)	NA
	Min	-271.27 (101.40)	-294.01 (74.64)	-271.38 (72.47)	-294.43 (98.37)	NA	-192.19 (63.35)	-206.30* (61.59)	-204.87^{+} (72.47)	-167.82^{*+} (56.69)	*: <0.001 +: 0.008
	Medial +; Lateral -										
	Max	-52.30 (49.68)	-42.90 (81.81)	-43.42 (54.57)	-44.31 (59.61)	NA	51.41 (40.31)	68.37 (51.64)	75.92 (35.45)	48.66 (63.69)	NA
	Min	-111.13 (63.53)	-126.81 (80.24)	-105.80 (65.39)	-114.66 (81.45)		-19.05 (37.81)	-24.53 (49.07)	1.80* (31.68)	-39.79* (51.66)	*: 0.013
Hand	Superior +, Inferior -										
	Max	315.26$^{*\#}$ (65.87)	325.68^{+} (65.77)	306.69$^{*+\text{€}}$ (67.86)	333.38$^{\text{€}\#}$ (72.26)	*:0.015 +: 0.015 €: <0.001 #: 0.011	305.53$^{*\$}$ (66.07)	280.02^{*+} (66.68)	306.88$^{+\#}$ (67.07)	273.63$^{\$\#}$ (77.35)	*:0.018 +: 0.001 $: 0.039 #: 0.001
	Min	176.88 (60.97)	179.64 (71.91)	174.14 (65.14)	170.12 (72.63)	NA	164.70 (64.62)	145.00 (73.70)	178.44 (76.50)	138.70 (77.67)	NA
	Medial +; Lateral -										
	Max	25.29 (26.05)	33.30 (38.52)	23.83 (28.61)	32.01 (32.11)		33.74$^{\$\#}$ (25.07)	62.63$^{*\$}$ (33.39)	39.46^{+*} (21.56)	60.20$^{+\#}$ (33.41)	*:0.014 +: 0.004 $: <0.001 #: 0.001
	Min	13.93 (22.39)	18.64 (28.50)	12.50 (23.20)	19.55 (29.19)	NA	1.72 (19.97)	17.02 (30.61)	10.69 (14.09)	14.09 (29.49)	NA

Table 9. Means (± 1 standard deviation) of the peak resultant forces at the hand, elbow and shoulder

Resultant Force (N)	**Simple Main Effect of Technique by Arm**									
	Trailing Arm					Leading Arm				
	SS	HH-A	HH-I	TU	p-value	SS	HH-A	HH-I	TU	p-value
SHOULDER	308.90 $^{+}$ (94.99)	329.96* (80.84)	307.11* (76.11)	330.69 $^{+}$ (95.62)	+:0.014 *: 0 011	262.45 (78.06)	250.00 (64.10)	276.9 * (80.17)	241.9 * (78.71)	*:0.003
ELBOW	320.3 $^{+}$ (90.89)	344.95 $^{\#}$ (71.16)	319.91 * $^{\#}$ (71.41)	351.42 * $^{+}$ (76.38)	+: 0.036 *:<0.001 #:0.005	293.97 (87.24)	279.39 $^{€}$ (67.27)	309.68 $^{€}$ Ω (81.69)	273.93 ^{Ω} (79.02)	€:0.021 Ω:0.004
HAND	347.46* ^{υ} (71.22)	362.18 $^{\#}$ (72.93)	337.47 * $^{\# +}$ (73.82)	368.85 $^{+}$ υ (78.61)	*: 0.02 +:<0.001 #:0.006 υ:0.006	309.88 (68.06)	288.61 $^{+}$ (68.82)	311.07 $^{\# +}$ (67.96)	283.17 $^{\#}$ (79.58)	+:0.005 #:0.004

Figure 14. Mean (±standard errors) of the peak flexion/extension, abduction/adduction and internal/external rotation moments at the shoulder for all transfer techniques.

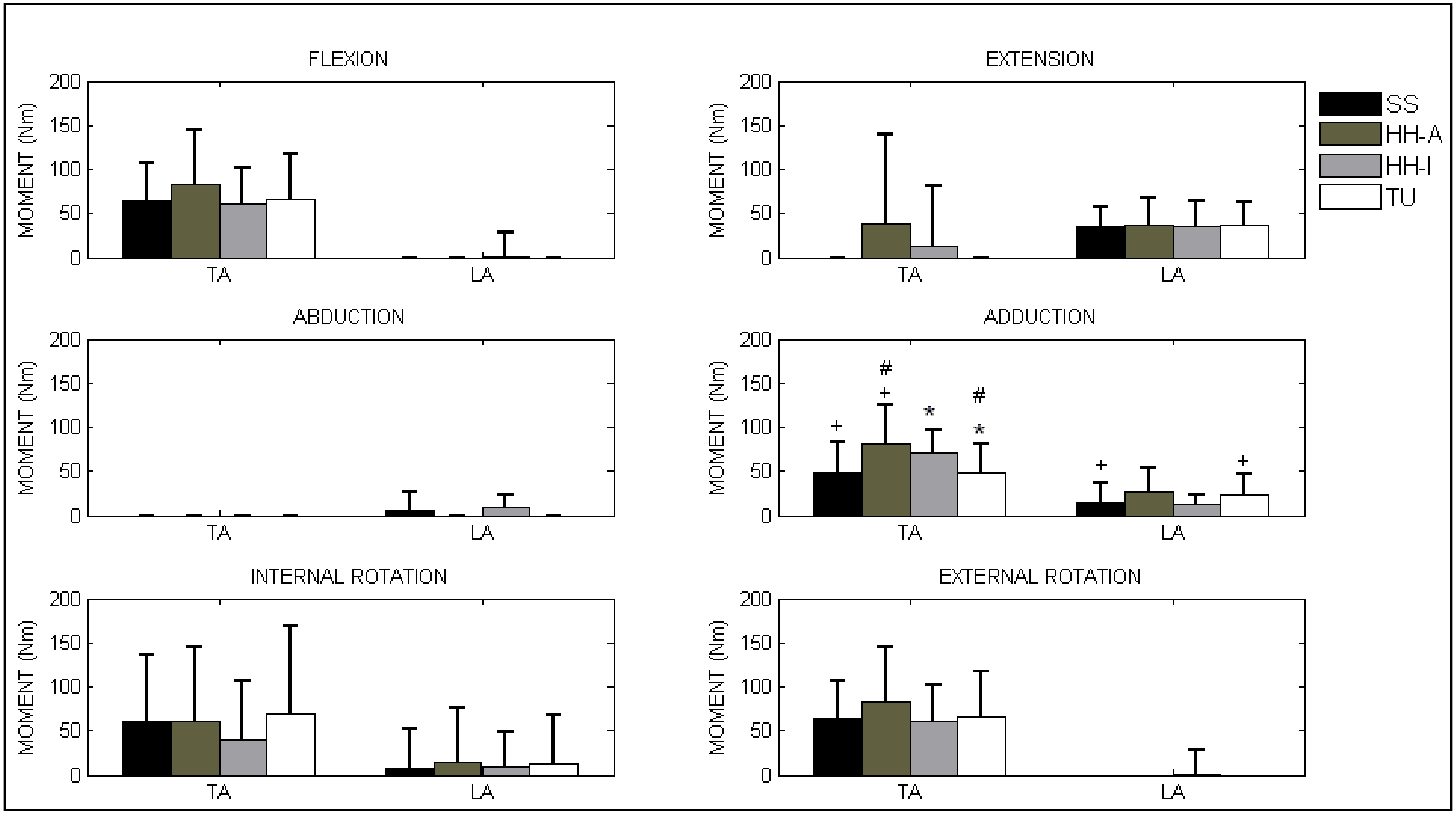


Figure 15. Mean (±standard errors) of the peak flexion and extension moments across the wrist and elbow for all transfer techniques.

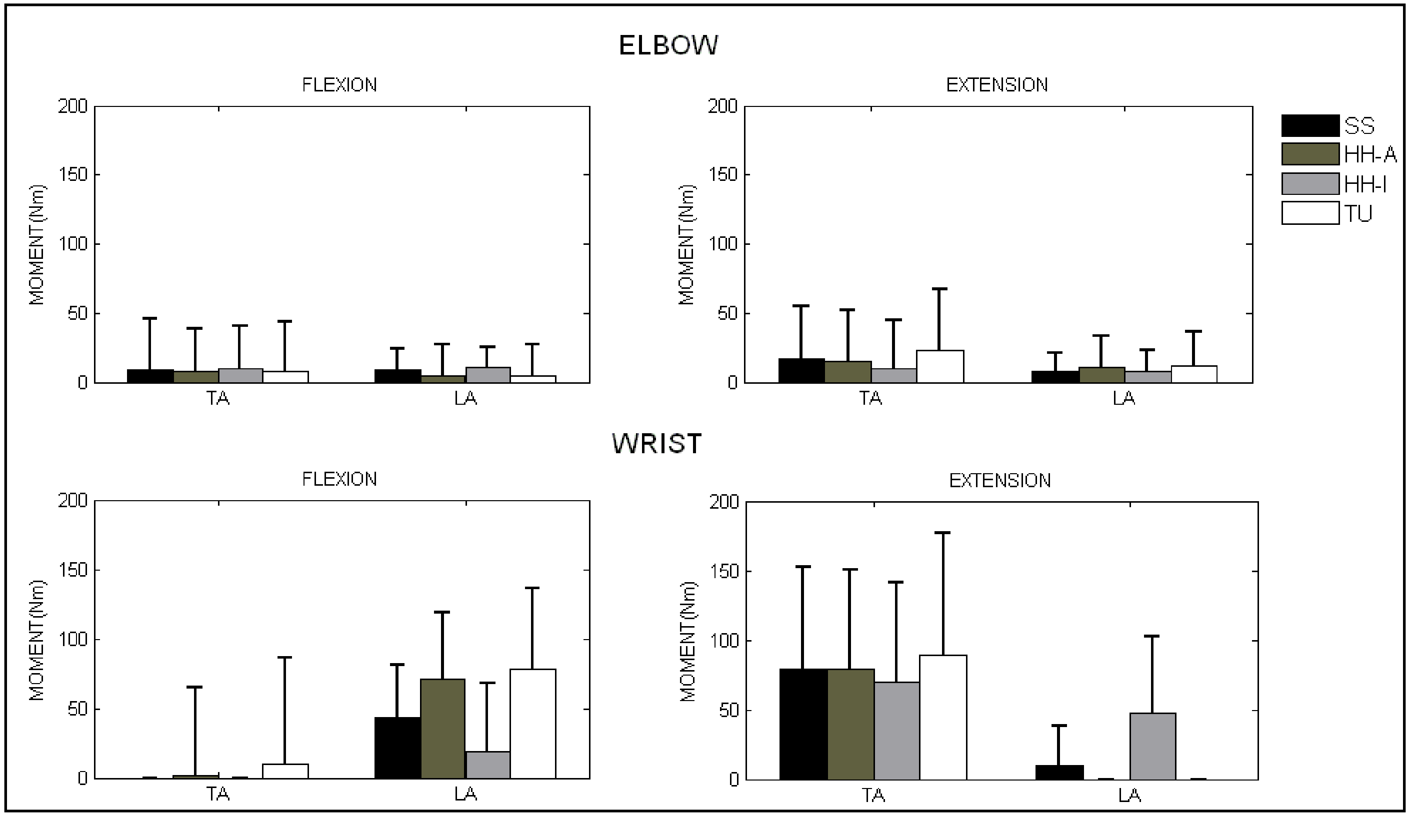


Figure 16. Mean (±standard errors) of the peak 3D component forces at the shoulder across the four transfer techniques.

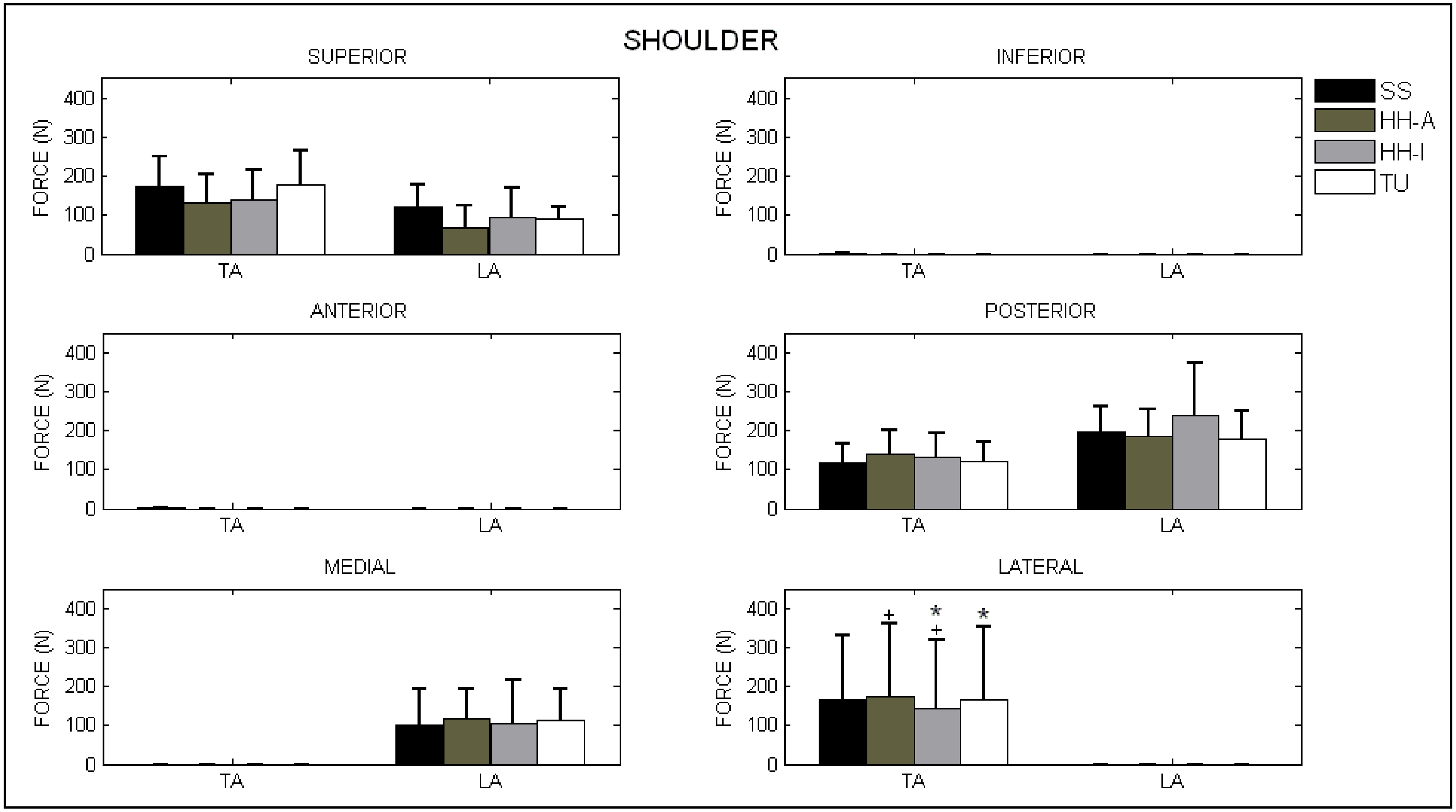


Figure 17. Mean (±standard errors) of the peak 3D component forces at the elbow across the four transfer techniques.

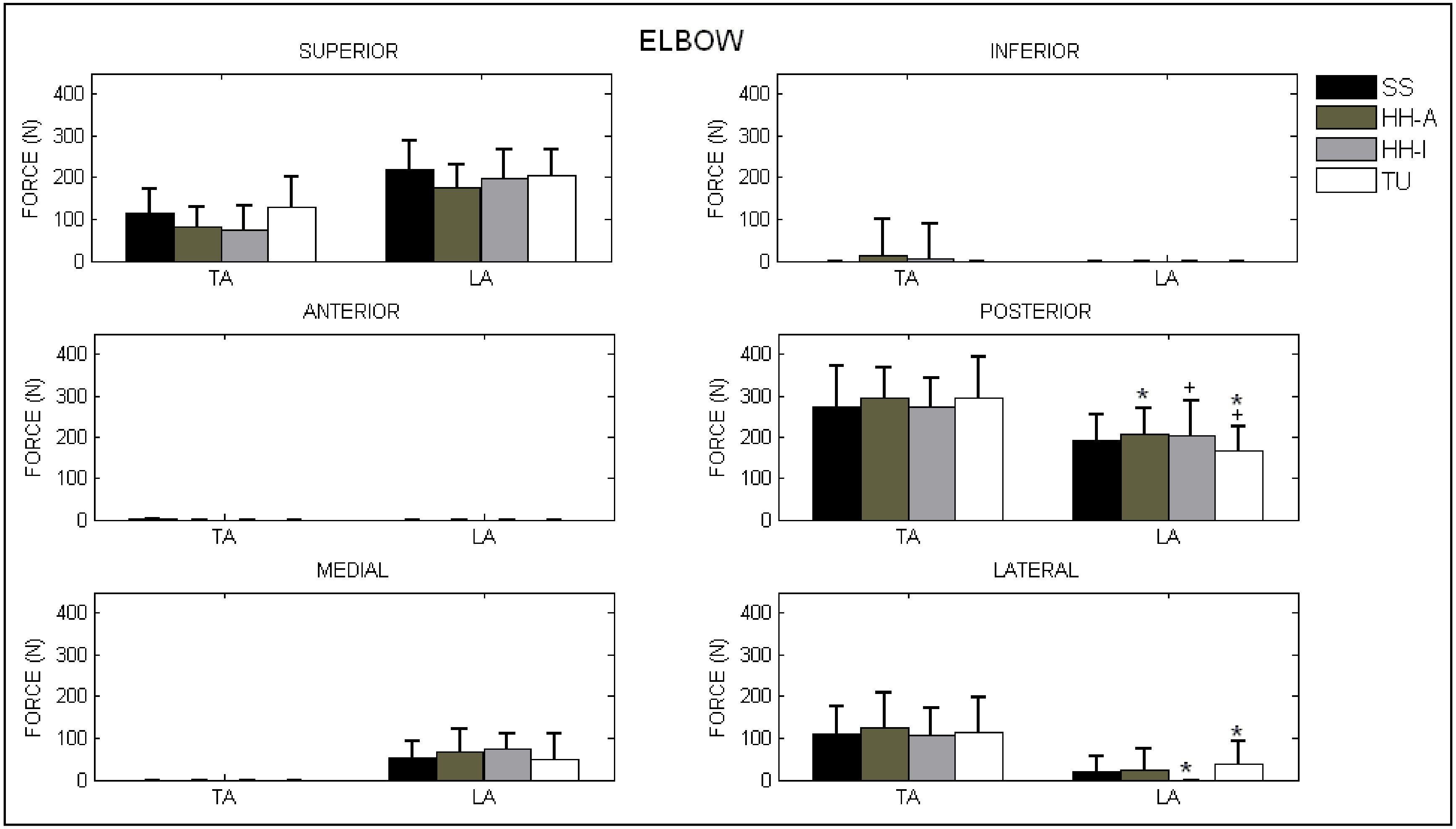


Figure 18. Mean (±standard errors) of the peak 3D component forces at the hand across the four transfer techniques

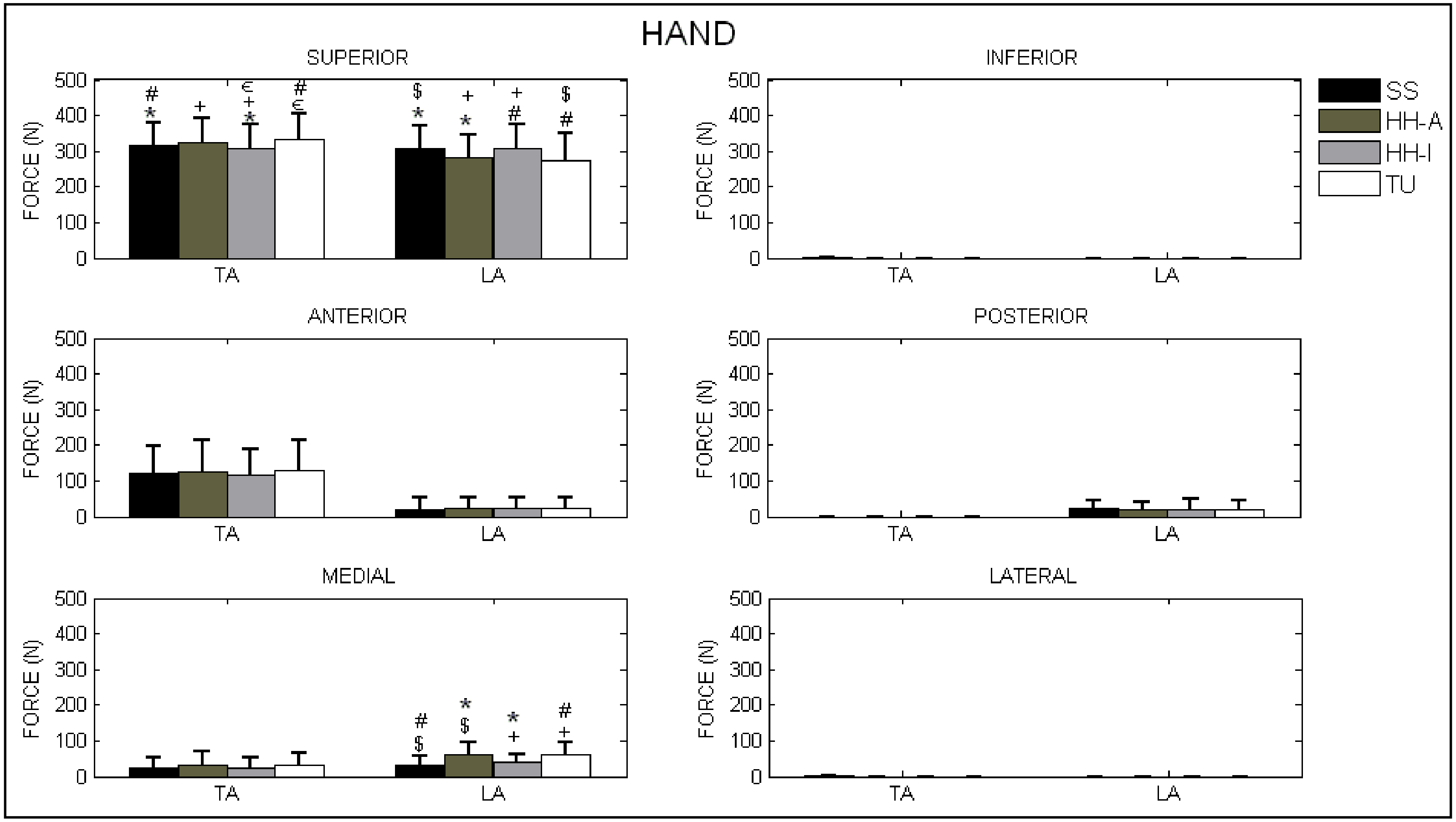


Figure 19. Resultant forces across the hand, elbow and shoulder.

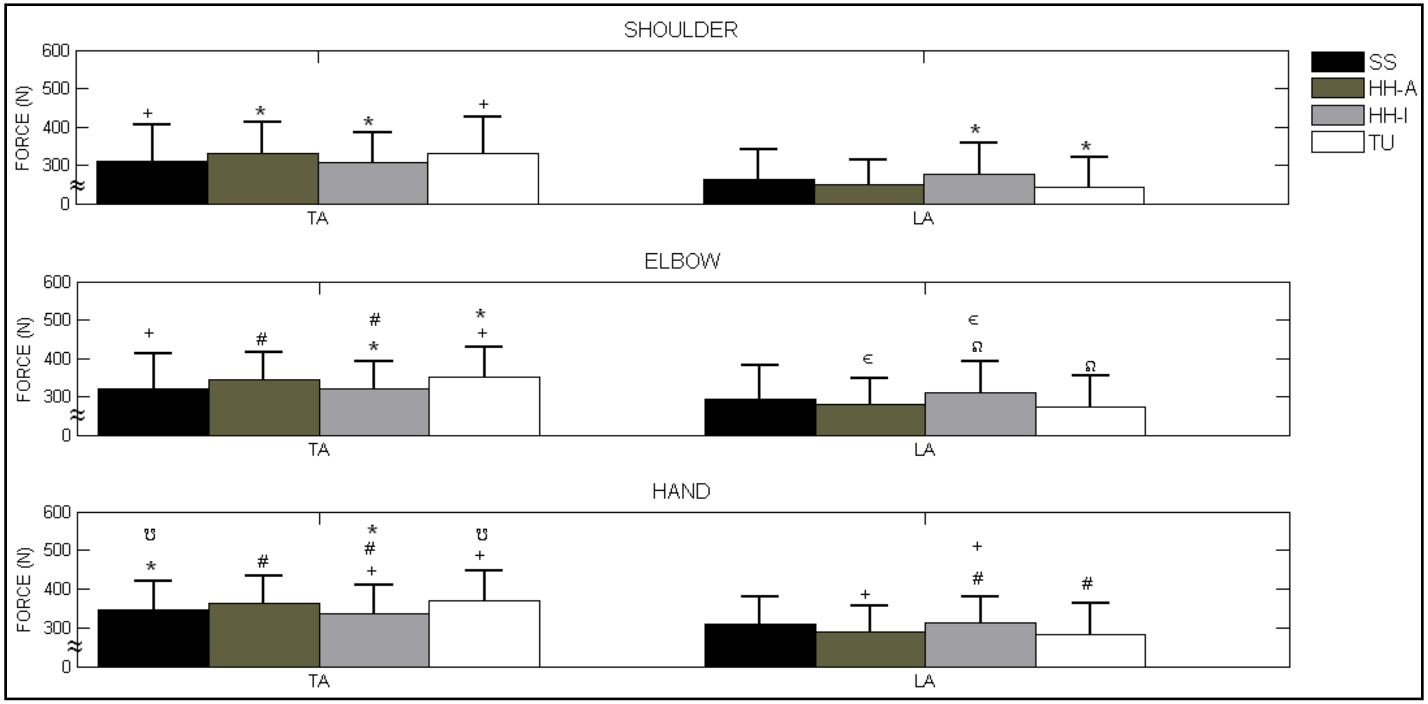


Table 10. Significant comparisons of UE joint reaction force and moment across the taught techniques

SUMMARY TABLE COMPARISON OF TAUGHT TECHNIQUES		
Technique Differences Averaged across arms		
Shoulder Internal rotation moment	HH-I < HH-A	
	HH-I < TU	
Shoulder Posterior Force	TU < HH-I	
Elbow Superior force	HH-A < TU	
Wrist Flexion moment	HH-I < TU	
Technique Differences within each arm		
	Trailing Arm	**Leading Arm**
Shoulder Adduction Moment	TU < HHA	NS
	TU < HH-I	
Shoulder Lateral Force	TU > HH-I	NS
	HH-A > HH-I	
Shoulder Resultant Force	HH-I < HH-A	TU < HH-I
Elbow Posterior Force	NS	TU < HH-I
		TU < HH-A
Elbow Lateral Force	NS	HH-I < TU
Elbow Resultant Force	HH-I < TU	TU < HH-I
	HH-I < HH-A	HH-A < HH-I
Hand Superior Force	HH-I < TU	HH-A < HH-I
	HH-I < HH-A	TU < HH-I
Hand Medial Force	NS	HH-I < TU
		HH-I < HH-A
Hand Resultant Force	HH-I < TU	HH-A < HH-I
	HH-I < HH-A	TU < HH-I

Table 11. Significant comparisons of the UE joint force and moments for the preferred method of transferring versus the taught techniques

SUMMARY TABLE SELF-SELECTED VS TAUGHT TECHNIQUES		
Technique Differences Averaged across arms		
Shoulder superior force	SS > HH-A	
Elbow superior force	SS > HH-A	
Wrist Flexion Moment	SS < TU	
Technique Differences within each arm		
	Trailing Arm	**Leading Arm**
Shoulder resultant force	SS < TU	NS
Elbow resultant force	SS < TU	NS
Hand superior force	SS > HH-I	SS > HH-A
	SS < TU	SS > TU
Hand medial force	NS	SS < HH-A
		SS < TU
Hand resultant force	SS < TU	NS
Shoulder adduction moment	SS < HH-A	SS < TU

3.3.3.4 Shoulder

In general across all techniques higher flexion moments acted at the trailing shoulder whereas higher extension moments acted at the leading shoulder (Figure 3). Similarly we found higher internal/external rotation and adduction moments in the trailing arm compared to the leading arm irrespective of technique (Figure 14). Reaction forces at the shoulder were mostly directed in the superior and posterior direction in both arms whereas forces were directly medially in the lead arm and laterally in the trailing arm (Figure 5).

Taught Technique Comparison

Shoulder internal rotation moment was significantly lower for the HH-I technique compared to the HH-A ($p = 0.002$) and TU ($p = 0.009$), averaged across both arms (Table 6 & Table 10). Shoulder adduction moments in the trailing arm were significantly lower when transferring with the TU technique compared to the HH-A ($p = 0.010$) and the HH-I technique (Figure 14, Table 2 & Table 10; $p = 0.021$). Posterior forces were significantly lower when transferring with the TU technique compared to the HH-I technique (Table 6 & Table 10; $p = 0.042$) averaged across arms. Lateral forces were lower at the trailing shoulder when transferring with the HH-I technique compared to the other taught techniques (HH-A: $p = 0.03$ and TU: $p = 0.004$; Figure 16, Table 8 & Table 10). Resultant forces were significantly lower at the trailing arm, for the HH-I technique compared to the HH-A ($p = 0.014$) technique (Figure 19, Table 8 & Table 10). At the leading shoulder the resultant forces were significantly lower for the TU compared to the HH-I ($p=0.003$; Figure 19, Table 9 & Table 10).

Self-Selected Vs Taught Techniques

Shoulder adduction moments in the trailing arm were significantly lower when transferring with the SS technique compared to the HH-A (p=0.044, Table 7 & Table 11). Significantly lower adduction moments were generated at the leading shoulder when transferring with the SS technique compared to the TU transfer technique (p = 0.001; Figure 14, Table 7 & Table 11). Superior forces at the shoulder were significantly lower when transferring with the HH-A technique compared to the SS (p = 0.009; Table 6 & Table 11) averaged across both arms. Resultant forces were significantly lower at the trailing arm, for the SS technique compared to the TU technique of transferring (p = 0.011; Figure 19 , Table 8 & Table 11).

3.3.3.5 Elbow

In general peak elbow flexion and extension moments were similar across all techniques (Figure 15). Superior and posterior forces were the larger component forces at the elbow. Additionally medial forces were primarily present at the leading elbow in comparison to lateral forces of similar magnitudes at the trailing elbow (Figure 17).

Taught Technique Comparison

Superior forces at the elbow, averaged across both arms, were significantly lower for the HH-A compared to the TU (p = 0.002; Table 6 & Table 10). Transferring with the TU technique resulted in significantly lower posterior forces at the leading elbow, compared to both head hips techniques (p values< 0.008; Figure 17, Table 8 &Table 10). Lateral forces at the leading elbow were significantly lower for HH-I compared to the TU (p = 0.013; Table 8 & Table 10). At the trailing elbow net resultant forces were significantly lower for the HH-I in comparison to the TU and the HH-A techniques of transferring (p-values< 0.036; Figure 19, Table 9 & Table 10). In contrast at the leading elbow the TU and HH-A transferring techniques resulted in significantly lower resultant forces compared to the HH-I technique (p-values < 0.021; Figure 19, Table 9 & Table 10).

Self-Selected Vs Taught Techniques

Superior forces at the elbow, averaged across both arms, were significantly lower for the HH-A compared to the SS (p = 0.007; Table 6 & Table 11). At the trailing elbow net resultant forces were significantly lower for the SS in comparison to the TU (p-values = 0.036; Figure 19, Table 9 & Table 11).

3.3.3.6 Wrist & hand

Flexion moments were more predominant in the leading wrist compared to extension moments, which were predominant in the trailing wrist (Figure 15). Superior forces constituted the principal component forces present at both hands across all techniques. Small magnitude posterior forces, similar across all techniques were present only at the leading hand (Figure 18).

Taught Technique Comparison

Wrist flexion, averaged across both arms, was significantly lower for the HH-I (p = 0.003) technique compared to the TU (Table 6& Table 10). As shown in Figure 18, superior forces were significantly lower at the trailing hand for the HH-I compared to the remaining taught techniques (p-values<0.015; Table 8 &Table 10). Superior forces at the leading hand were lower for the HH-A and TU techniques compared to HHI (p = 0.001 for both). Significantly lower medial forces at the leading hand were observed for HH-I compared to HH-A (p-values < 0.001; Figure 18, Table 8 & Table 10) and TU (p values <0.014; Figure 18, Table 8 & Table 10). The HH-I resulted in significantly lower net resultant forces at the trailing hand compared to the remaining two taught techniques (p-values < 0.02; Figure 19, Table 9 & Table 10). The resultant forces at the leading hand were significantly lower for the HH-A and TU in comparison to HH-I technique of transferring (p values< 0.005; Figure 19, Table 9 & Table 10).

Self-Selected Vs Taught Techniques

Wrist flexion moment, averaged across both arms, was significantly lower for the SS technique compared to the TU (p = 0.048, Table 6 & Table 11). As shown in Figure 18, superior forces were significantly lower at the trailing hand for the HH-I compared to the SS (p-value= 0.015, Table 8 & Table 11) technique of transferring. The preferred method of transferring resulted in significantly lower superior forces for the SS compared to TU (p = 0.011; Table 8 & Table 11) at the trailing hand. At the leading hand superior forces were lower for the HH-A and TU techniques compared to the SS technique (p = 0.018 and p=0.018; Table 8 & Table 11). Significantly lower medial forces at the leading hand were observed for SS compared to HH-A (p-values < 0.001; Figure 18, Table 8 & Table 11) and TU (p values <0.014; Figure 18 Table 8 & Table 11). At the trailing arm, the net resultant force was significantly lower for the SS technique than the TU (p = 0 .006; Figure 19, Table 9 & Table 11).

3.3.4 Classification of self-selected technique based on taught techniques

The leading hand position and trunk flexion angles for each participant have been super imposed on the mean and 1 standard deviation of the taught techniques for the same variables, shown in Figure 20 and Figure 21. The leading hand position was classified based on the description in 3.2.4. Owing to the overlap of the standard deviations of the mean and standard deviations of trunk flexion angles for the taught techniques the cut off criterion used was:

- Trunk upright: Trunk flexion <= 35°
- Head-Hips: Trunk flexion > 35°

Figure 20. Group mean and standard deviation of the leading hand distance for the taught techniques with the individual participant leading hand distance for the self-selected technique superimposed.

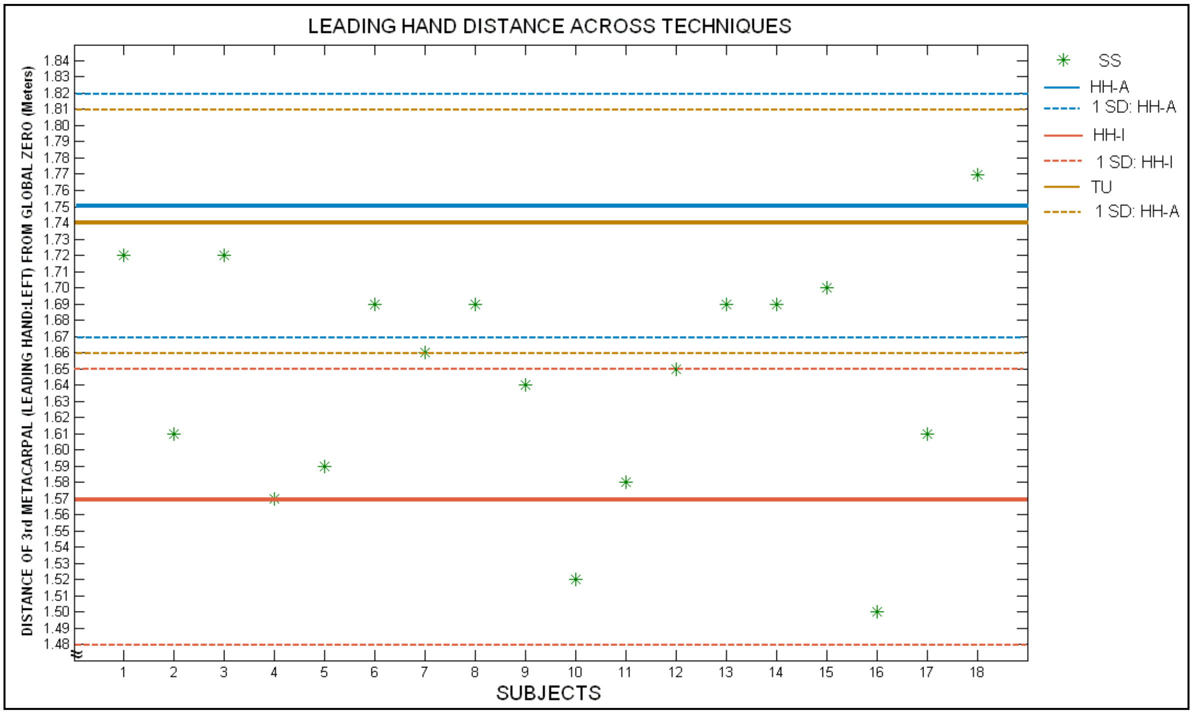


Figure 21. Group mean and standard deviation of the trunk flexion angle for the taught techniques with the individual participant trunk flexion angles for the self-selected technique superimposed.

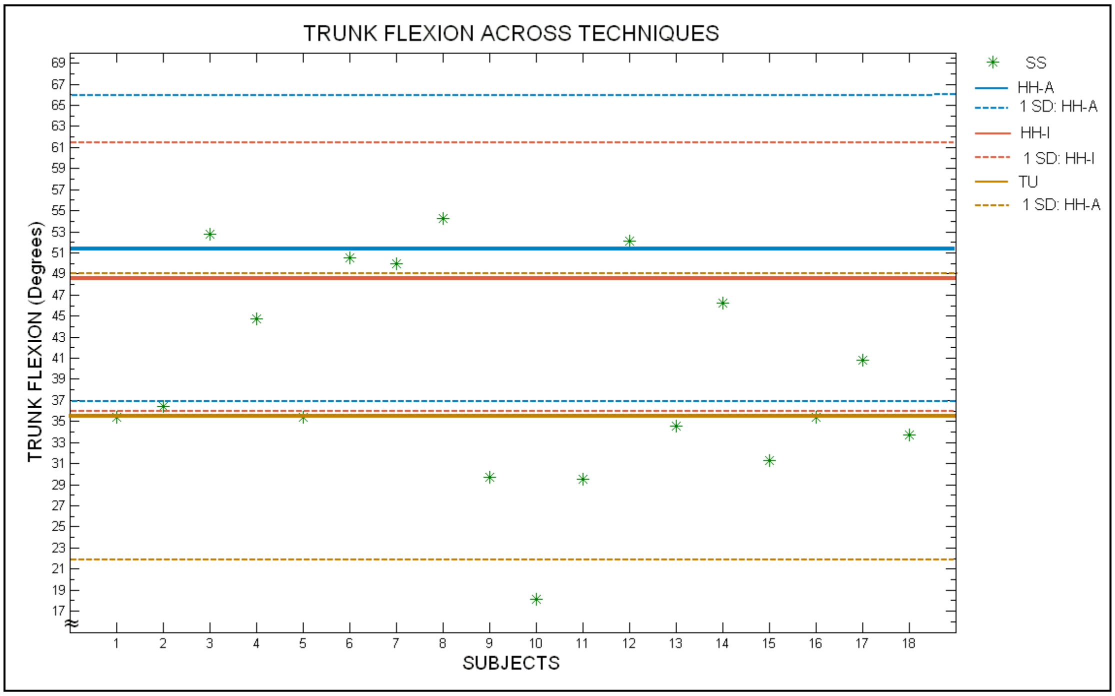


Table 12. Classification of Self-Selected technique based on variables that the taught techniques vary on namely: trunk flexion and leading hand position. * indicates observing a trunk upright posture with the leading hand close to the body.

SUBJECT ID	Target	Trunk Flexion	Transfer Technique
1	Far	<= 35°	TU
3	Far	> 35°	HH-A
6	Far	> 35°	HH-A
7	Far	> 35°	HH-A
8	Far	> 35°	HH-A
13	Far	<= 35°	TU
14	Far	> 35	HH-A
15	Far	<= 35°	TU
18	Far	<= 35°	TU
2	Near	<= 35°	TU*
4	Near	> 35	HH-I
5	Near	<= 35°	TU*
9	Near	<= 35°	TU*
10	Near	<= 35°	TU*
11	Near	<= 35°	TU*
12	Near	> 35	HH-I
16	Near	<= 35°	TU*
17	Near	> 35	HH-I

Based on a simultaneous consideration of the two categorizing variables and the definitions of the taught techniques, twelve subjects fit into one of the three taught techniques with the exception of 6 subjects who transferred with their trunk upright and with their hand in close to the body (referred to in the table as TU*). Due to small subject numbers, subjects in the two head-hips categories and subjects in the two TU categories were combined to analyze the effects of the two trunk postures on the differences in biomechanics and demographics. Similarly, we compared all subjects using a 'near' leading hand placement to all subjects using a 'far' hand position. It was found that those participants who placed their hand further away from their body were significantly younger (far: 31 ± 8.5) in comparison to those who placed their hand closer to their body (near: 42± 9.4; p =0.03).

3.4 DISCUSSION

This study has three purposes which are to compare the upper extremity joint loading for taught transfer techniques: HH-A, HH-I and TU; secondly to compare these taught techniques to the preferred method of transferring: SS; and thirdly to classify the SS technique into the taught techniques, amongst a cohort of manual wheelchair users with SCI. This study is a follow-up to recently published data, which studied the same taught techniques amongst a group of unimpaired subjects (n=14) (51) . This study found more subtle, but significant differences in techniques and enhances the understanding of the impact of paralysis on technique execution and subsequent UE joint loading.

3.4.1 Taught technique comparison

3.4.1.1 Trunk kinematics and temporal characteristics

The forward flexed trunk combined with the pivoting action of the SPT account for the significantly larger right axial rotation velocity head-hips techniques in comparison to the SS and the TU transfer techniques. From the classification of the SS technique we found that the participants predominantly employed a TU technique or modified version of the TU technique, which explains the significantly higher trunk flexion velocity for the HH-I technique compared to the SS technique. The transfer duration of the HH-A technique was significantly longer in comparison to the TU technique, which can be explained, based on the nature of the two techniques; the HH-A technique required longer duration to prepare and generate the momentum to transfer in comparison to the TU technique, which is a quicker lateral shift.

3.4.1.2 Shoulder

Performing HH techniques entails forward flexion of the trunk causing a shift of the center of mass forward (anterior with respect to the spine) and the shoulders translate anteriorly (17, 53). The larger distance between the pivot hand (leading arm) and the body's centre of mass increases the moment arm resulting in an increase in moments about the shoulder (54). Based on the significantly large shoulder adduction moments generated at the trailing shoulder when transferring with the leading arm further away from the body (HH-A and TU), we believe that the transfer required greater activation of the shoulder adductors (latissimus dorsi, pectoralis major and subscapularis). This can be recognized as a method of stabilizing the upper body when the leading arm is abducted, during the lift phase of the SPT. With the HH-A and TU techniques the larger distance between the leading and trailing hands could be forcing the arms into a closed chain internal rotation, as indicated by the significantly higher internal rotation moments for the aforementioned techniques as compared to the HH-I technique. However the axial rotation velocity was higher for the head-hips techniques in comparison to the TU technique. Therefore the larger internal rotation moments occurring at the shoulders may possibly be more influenced by the leading hand being further away from the body rather than trunk flexion.

As hypothesized shoulder superior forces tended to be larger, although not significantly, for the TU in comparison to the head- hips style of transferring. Based on the style of transferring keeping the trunk upright results in the shoulder joint bearing larger vertically directed forces in comparison to when the trunk is flexed. Significantly higher posteriorly directed force components for the HH-I technique compared to the HH-A and TU transfer may be a cause of concern in the long run if this technique was to be adopted due to the risk of development of posterior instability and tendinitis(10). The large lateral forces associated with the HH-A and TU techniques in comparison to that observed when transferring with the HH-I technique at the trailing arm, may be described based on the rotational demands made on the shoulder muscles while to maintain stabilization of the trailing limb.

3.4.1.3 Elbow

Previous literature has underscored the importance of elbow extensor strength in transfer strategy selection (55). Prior research has reported the combined effect of increased superior and posterior force is a predecessor of impingement syndrome (9, 56). An increase in superior forces at the elbow when using the TU technique indicates a potential increase in the risk for secondary impairments, due to ulnar nerve compressive neuropathy at the joint (57). Similar to Gagnon et al. (39), where the shoulder and elbow joint forces and moments were examined for both weight relief and lateral level transfers, we found that the posterior peak force was a dominating component force during SPTs which was larger in the trailing arm. Unlike our previous results (51), we find that when the trunk was upright, as hypothesized we found an increase in superiorly directed force, however at the leading elbow the posteriorly directed forces actually decreased for the TU transfer. Interestingly the HH-I resulted in lower net resultant forces at the leading elbow however the same technique produced largest net resultant force at the leading elbow. Therefore one must consider existing UE function/condition when adopting/choosing a transfer technique. We see that trunk flexion is beneficial in reducing the vertical forces however hand placement influences the loading in the non-vertical forces at the elbow. Therefore careful selection of choosing which arm will play the role of leading or trailing must be considered.

3.4.1.4 Wrist and hand

Primarily extension moments were seen at the trailing arm and flexion moments were seen at the leading arm, which can be attributed to the gripping style of the force beam. We found that the TU technique generated the largest flexion moments at the leading arm and extension moments at the trailing arm. Large excursions of the wrist are detrimental due to the inherently unstable bony anatomy associated with the wrist; which also increases the risk of peripheral neuropathy of the median nerve (58-61). SPT's have been found to be associated with larger wrist range of motion compared to wheelchair propulsion (61, 62). Incidence of CTS has been found to be associated to duration of SCI(60). People with CTS have elevated pressure in the carpel tunnel when the wrist is in a full active flexion or active extension (8, 60). Therefore if the individual is already predisposed to median nerve neuropathy then the TU style of transferring would prove to be detrimental.

3.4.2 Overall summary of technique comparison across joints

There was an imbalance in the superior forces and net resultant forces, being higher at the trailing arm and lower at the leading arm when transferring with the HH-A and TU techniques in comparison to the HH-I technique across all joints. This imbalance draws attention to leading hand placement being a key factor to balancing the distribution of forces across the upper limb joints. There were no statistical differences in vertical force at the shoulder for the trained techniques. Although statistical differences existed at the hand and elbow for the trained techniques, this was accompanied with larger non-vertical forces at the joints. The HH-I technique also resulted in lower medial and lateral forces in comparison to the HH-A and TU which highlights the reduced rotational demands on the muscles when the leading arm is closer to the body which likely helps with shifting bodyweight between the two arms while maintaining postural stability (10).

3.4.3 Comparison of taught techniques to self-selected technique

Results showed that most statistical differences were between the SS and HHA and SS and TU, which would suggest that the biomechanics of SS for the group on average most closely resembled that of the HH-I technique. This can be supported based on the classification of the SS technique with respect to the taught techniques. We classified 10 participants preferred method of transferring to be a hybrid method where in they employed moderate trunk flexion and 9 out of the 10 persons placed their hand closer to their body. We did find that the upper trunk flexion velocity was significantly higher for the HH-I technique compared to the preferred method of transferring which can be described based on the momentum generated as a result of the greater forward flexed trunk associated with the head-hips style of transferring. The only statistically significant difference found between these two techniques was for trailing hand superior force being 8.6 N greater for SS than for HH-I which while a statistically significant the clinical relevance of such a small magnitude difference may be minimal. The factor that was similar between the two techniques was that the hand placement was close to the body in comparison to the remaining two taught techniques (HH-A and TU), which required the hand to be placed further away from the body. As described above the HH-I technique enabled for a greater balance in joint loading across both arms, which was a similar pattern, observed for the TU * (classified SS transfer) (Figure 8).

Similar to previous literature we too found that the superior forces were the largest component forces acting at the hand (10, 20). The large vertical component forces combined with horizontal forces have been shown to be associated with high mechanical demands placed on the shoulder flexors and adductors and elbow extensors during the lift phase of the SPT (63). The SS technique compared to HHA and TU resulted in smaller magnitudes in all statistically significant variables with the exception of superior forces at all three joints. Observation of the results leads us to believe that the movement pattern adopted by the participants for the preferred style of transferring is directed towards optimizing (minimizing) *moments and non-vertical forces* - versus the vertical forces acting at the shoulder. This also points towards the trade-off that occurs between maintaining balance and the effort required to perform a successful SPT.

We found transferring with the SS technique, resulted in net resultant forces at the trailing shoulder, elbow and hand were significantly lower while not being significantly higher at the leading arm, compared to the taught techniques. The inclusion criteria controlled for the enrollment for persons with SCI who could perform an independent transfer free from any inhibiting pain. However it is a possibility that cohort of MWC users who were a high functioning group and had an average duration of 14 years since injury, developed compensatory mechanisms that reduced mechanical loading at the UEs.

An important observation regarding the SS technique in comparison to the taught techniques has to do with the comparatively even load distribution between the trailing and leading arm across all three joint which is in stark contrast to what was observed when transferring with the taught techniques and in particular HH-A and TU.

3.4.4 Feet

Similar to previous research (19, 20), majority of the weight bearing during the lift phase of the transfer occurs at the hand compared to the feet across all the transfer techniques. We found that the LE's bear close to 25% of body weight (BW) while transferring with the SS technique which corroborates with previous literature (10, 19, 20). This study not only analyzed participants preferred method of transferring but also taught techniques. Contrary to our hypothesis we did not find the head-hips techniques to have significantly higher LE weight bearing in comparison to the TU style of transferring. Although not significant the HH-A technique tended to have the largest LE weight bearing amongst the techniques. The results of our study not only highlights but also reiterates the importance of the LE's during transfers in accordance with the recommendations in the clinical practice guidelines for preservation of UE function(5).

3.4.5 Joint comparison

Regardless of technique, during the lift phase of the transfer, maximum weight bearing occurred in the vertical direction at the hand with much less force seen in the other component directions. However at the shoulder and elbow the vertical forces followed by posterior constituted the principal component forces. This is likely due to the extended position of the humerus with a flexed trunk. We also observed that at the elbow and shoulder medial forces were present primarily at the leading arm compared to the lateral forces present at the trailing arm across all techniques. This can be explained based on the pivoting action entailed in the SPT. Similar to Gagnon et al.(39), we found elbow extension/flexion moments to be smaller compared to shoulder flexion/extension. Our study is the first to examine wrist moments with regards to

SPTs. Magnitudes of the peak vertical forces were larger at the hand compared to the shoulder and elbow. The results indicate that there is a high demand on the forearm muscles to keep the wrist in a stable position during the transfer, like that observed during the performance of weight relief maneuvers (11). This finding underscores the propensity for transfers to accelerate the development of wrist pain and injuries particularly since the wrist is commonly placed in an extreme position of extension during the weight-bearing portion of the transfer (19). This was observed in our results as well wherein the wrist extension moments were larger than the shoulder and elbow extension moments.

3.4.6 Limitations

An important factor to consider with the comparison of the taught techniques is that fact that results are based on short term transfer training of the individual. The participants in our study were a more seasoned cohort of MWC users, with the average years since injury being 14 years. A constraint of the experimental set-up is that all transfers began with the left arm leading. Individual SPT direction preference was studied with respect to demands at the UEs, the results showed that muscle demands across the UEs did not vary between preferred versus a non-preferred direction when the participants transferred (54). Most MWC users face a variety of environmental situations wherein they have to be adept in transferring in either direction.

3.5 CONCLUSION

Our results point towards there being a tradeoff among force and moment components amongst the techniques and between leading and trailing arms which seems to depend mainly on hand placement more than trunk flexion. A technique that minimized vertical forces at the joints maximized non-vertical forces and moments. Keeping the leading arm close to the body during the head-hips motion appears to balance the UE joint loading across arms, which was similar to that found for the preferred method of transfer. Therefore the similarities between biomechanics of the SS and the HH-I technique may be primarily accounted to the leading hand placement being closer to the body. Classification of the SS technique into the taught techniques identified a technique we did not teach our subjects to use, transferring with an upright trunk posture with the hand close to body. Further study is needed to examine differences between the four classifications of self-selected SPTs. Longitudinal studies are also needed to understand which is better long-term; a technique that minimizes vertical forces (e.g. HH-A) or techniques that minimize non-vertical forces and moments (HH-I).

4.0 IMPACT OF SUBJECT CHARACTERISTICS AND FUNCTIONAL MEASURES ON UPPER EXTREMITY TRANSFER KINETICS

4.1 INTRODUCTION

Transfers have been identified as one of the key activities that lead to development of shoulder pain and injury among persons with spinal cord injury (SCI) (26, 30, 64). Those persons with SCI, who are full time manual wheelchair (MWC) users, perform up to 35 wheelchair transfers per day to complete basic activities of daily living (8, 9, 30). Maximum dependence and repetitive stress on the upper extremities (UEs) leads to high incidence of shoulder, elbow and wrist pain (8, 28, 31). People with tetraplegia reportedly experience higher intensity and prevalence of shoulder pain as compared to those with paraplegia (25, 28, 32).

Being able to perform independent sitting pivot transfers (SPTs) is a function of physical capabilities, subject characteristics and transfer technique (65) . Upper limb muscle strength has been identified as key to maintaining the highest level of independence in daily activities and preservation of upper extremity function (21). The primary muscles that are involved in elevating the trunk during weight-bearing activities like pressure relief pushups and transfers are the large thoracohumeral muscles, pectoralis major and the lattismus dorsi (10, 21). The muscle activity reaches peak intensity during the lift phase of the SPT. Very limited research is available exploring the relationship between biomechanical variables during transfer activity and UE strength.

A recent study investigated the relationship between shoulder and elbow strength and transfer ability (22). No association was found between upper limb strength and how high and how low one could transfer or how far away from the target surface one could be and still safely transfer. Thus, this study raised questions as to how critical strength is to performing transfers and whether other factors such as trunk balance, upper limb pain, skill/technique or anthropometry (e.g. weight, height, etc.) have greater impact on transfer performance.

Sitting balance has been highly correlated to functional performance of daily tasks such as transferring (23). Several studies have implicated the extensive trunk impairment resulting from high-level thoracic and cervical spinal cord injuries as a significant risk factor for developing shoulder pain and injuries (41, 66). Enhanced sitting balance allows for increased movement control (24) which may potentially result in reduced mechanical loads at the shoulder during transfers.

Bergstrom et al (67) investigated the influence of anthropometric variables on transfer ability amongst a group of persons with tetraplegia below C6 level. Amongst the 23 anthropometric variables measured they found nine variables that were significant in the model. The most influential of the significant predictors were sitting height, body weight and triangular base lift, which were positively correlated to the inability to transfer independently. Arm to torso ratios were also examined although were not entered into the model due to low statistical tolerance. No joint kinetics were collected in this study so the effect of anthropometry on joint loading is unknown.

The objective of this paper was to investigate how upper limb strength, sitting balance, anthropometry and pain influence UE joint kinetics during level independent wheelchair transfers. We hypothesized that having greater upper limb strength, low UE pain, better balance control and longer arms relative to the torso will be associated with less mechanical loading at the shoulders during transfers. The information gathered in this study will be helpful for developing individualized therapeutic interventions that enhance transfer performance while minimizing the risk of shoulder injuries (10, 50, 68, 69).

4.2 METHODS

4.2.1 Subjects

This study received ethical approval from the Department of Veterans Affairs Institutional Review Board. After reading and providing informed consent, twenty subjects (19 male, 1 female), volunteered to participate in this study. The inclusion criteria were: spinal cord injury C4 level or below that occurred over one year prior to the start of the study, able to independently transfer to/from a manual wheelchair without human assistance or assistive devices, over 18 years of age, and free from upper extremity pain that influenced their ability to transfer.

4.2.2 Data collection

4.2.2.1 Functional measures

Balance

Seated balance was tested using the Modified Functional Reach Test (MFRT) (24). As part of the balance protocol, participants were asked to transfer to a mat table, which was provided with a back support to rest in between trials. Once the subject was positioned, a yardstick was attached horizontally to the wall along the subject's shoulder at the level of the acromion. Subjects were asked to reach as far forward as possible maintaining a 90 degree shoulder flexion angle with palm down and hand flat. The opposite arm was not permitted to bear weight or hold on during the reach maneuver. The linear distance that the third metacarpalphalangeal joint moved was recorded as the 'reach' distance in centimeters. Subjects were permitted to practice before collecting the data. Three reach trials were collected.

Pain

Upper- limb pain was recorded using the Wheelchair Users Shoulder Pain Index (WUSPI)(70). The WUSPI is a visual analog scale (VAS) that targets activity limitation resulting from shoulder pain. It covers various activities (15 items in total), which include transfers, wheelchair mobility, self care and general activities. The individual item scores in the WUSPI were summed to give a total score. Because many of the subjects did not perform one or more activities measured in WUSPI items, we calculated a performance corrected shoulder pain score PC-WUSPI by dividing the raw total WUSPI score by the number of activities performed and multiplying by 15 (32).

Strength Testing

Isokinetic strength measurements, at a torque arm speed of 60 deg/sec, were recorded using an instrumented dynamometer (Biodex Medical System, New York, USA). The measurements were recorded in a randomized fashion for both arms for the following test maneuvers: shoulder flexion/extension in the sagittal plane, shoulder abduction/adduction in the frontal plane, shoulder internal/external rotation in the transverse plane, elbow flexion/extension and wrist flexion/extension (tested range of motion shown in Table 13).

Table 13. Isokinetic testing of UE joint at 60 deg/sec

Joint	Movement	Tested Range of Motion (Degrees)
Shoulder	Extension/Flexion	-30 to 50
	Adduction/ Abduction	10 to 70
	Internal/External Rotation	0 to 45
Elbow	Extension/Flexion	0 to 90
Wrist	Extension/Flexion	-45 to 45

Two practice repetitions for each movement tested were completed prior to data collection. In order to ensure the maximal force production of the tested upper extremity participants were secured into the chair with three padded belts: two diagonally across their chest and one across their lap. Five repetitions were recorded for each maneuver and participants were allowed to rest for five minutes to avoid fatigue from becoming a confounding factor. A rest period of 30- 60 minutes was taken after the strength testing before transfer biomechanics were recorded. From strength testing, peak Isokinetic torques were calculated using customized software (MATLAB 2011b, MathWorks; Natick, Massachusetts) and were averaged over 5 repetitions for each the of movements recorded at the shoulder, elbow and wrist. The peak torques were normalized by body weight for each participant and reported % Meter (%m).

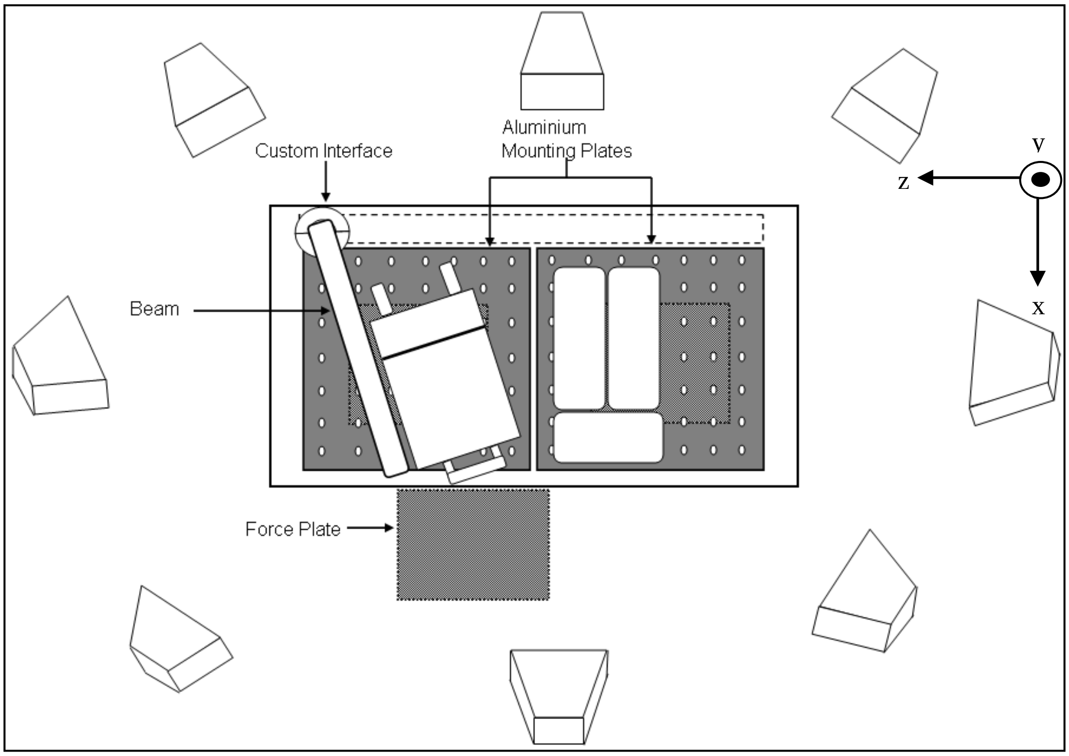
Anthropometric measure

Arm length was recorded as the distance between the acromion and the ulnar tuberosity. The Trunk length was recorded as the distance between the acromion and the greater trochanter. The Arm to torso ratio was computed with the aforementioned measurements.

4.2.3 Experimental protocol

Participants used their personal wheelchairs to transfer to and from a bench. For all transfers the wheelchair was positioned and secured at a comfortable angle from an adjustable height tub bench as shown in (Figure 22). The bench was adjusted to be level with the wheelchair seating surface. The platform contains three force plates (Bertec Corporation, Columbus, OH), one beneath the wheelchair, one beneath the tub bench and one located below the feet (34). The wheelchair and bench were secured to the platform. A steel beam attached to a 6-component load cell (Model MC5 from AMTI, Watertown, MA) was positioned to simulate a wheelchair armrest. Reflective markers were placed on the subjects C7, T3 and T8 vertebrae, right and left acromion processes, 3rd metacarpalphalangeal joints, radial and ulnar styloid processes, and lateral epicondyles. The coordinates of the markers were recorded based on a global reference frame using a sixteen camera three-dimensional motion capture system (Vicon Peak, Lake Forest, CA). Several anthropometric measurements were recorded such as: axiliary arm, wrist, fist and elbow circumference, upper arm and forearm length.

Figure 22. Wheelchair (left) and tub bench (right) shown secured to the aluminum mounting plates of the base frame. The custom interface consists of the load cell and a beam that can be positioned anywhere along base frame.



All transfers began with the left arm leading and moving the body from the wheelchair to the bench. For the first transfer, subjects were instructed to perform a lateral transfer as they normally would from their wheelchair to the adjacent level tub bench. For this transfer, they could place their left hand anywhere on the bench and right hand on the steel beam (height of wheelchair arm rest). Each transfer technique was performed three times and kinetic and kinematic data were recorded synchronously at 360 Hz and 60 Hz respectively, for the length of the transfer.

4.2.3.1 Data analysis

Kinetic, kinematic, and anthropometric data were entered into an inverse dynamic model to calculate the 3D net shoulder and elbow joint forces and moments. Kinetic data were down sampled to 60 Hz, to align with the kinematic data. Both kinematic and kinetic data were filtered with a 4th order zero-lag Butterworth filter (cut off frequency of 5 Hz and 7Hz respectively). The inverse dynamic model used was based on the general rigid-link segment model using a Newton-Euler method and a variable degree of freedom body co-ordinate system (35). The vertical reaction force from the force plate under the bench and the grab bar were used to determine the start (absolute value of vertical force > 0 at the grab bar and bench) and the end (determined prior to the generation of a large spike in the vertical force at the bench) of the transfer. The SPT was delineated into three phases: prelift, lift and descent. The vertical forces from the force plate under the tub bench, wheelchair and the grab bar (wheelchair side) were superimposed onto one plot to analyze the phases of transfer. Trunk motion, represented by the C7 and T3 markers, were added to the force data to assist with the delineation of phases of transfer (37).

4.2.3.2 Kinetic outcome measures

Analysis was conducted for transfers from the wheelchair to tub bench. For each trial 3D component and net resultant force were calculated for the shoulder, elbow and hand. Flexor/extensor, abduction/adduction and internal /external rotation moments at the shoulder, flexor/extensor moments at the elbow and wrist were calculated for both leading and trailing arms. The maximum and minimum of the variables were identified for the lift phase of the transfer. Variables were computed using Matlab (Mathworks, Inc., Natwick, MA) and averaged over the three trials. All mean force and moment values were normalized against body weight for each participant and reported a % Body Weight (%BW) and % Meter (%m) respectively.

4.2.4 Statistical analysis

Group means and standard deviations were computed. Descriptive analyses were completed for the data. Shapiro-Wilk test was used to determine normality. A Mann Whitney *U* test was used to investigate differences in shoulder pain and balance between the group with paraplegia (PG) and group with tetraplegia (TG). Because the level of SCI has been significantly associated with upper limb isokinetic strength (71),the sample was split into three groups based on level of injury: group with high paraplegia (above T7), group with low paraplegia (at T7 or below) and group with tetraplegia. A Kruskal-Wallis test was used to test UE strength differences between the PG and TG groups. Individuals with incomplete SCI were analyzed using a Mann Whitney *U* Test. Spearman Rho correlations were performed on all weight normalized strength and kinetic measures for all groups combined. Additionally spearman rho correlations were run between the non-weight normalized kinetic measures and demographic variables: age, height, weight, years since injury, arm/torso ratio and functional measures: balance and pain. For the correlations a pairwise exclusion analysis was used to handle variables with missing data (i.e. analysis by analysis in which the participant data is excluded only for the calculation involving the variable which has missing data). All analyses were performed using SPSS software, Version 19.0 (SPSS Inc., Chicago, IL) and significance level of 0.05.

4.3 RESULTS

4.3.1 Participants

Of the 20 participants who took part in the study, 18 (17 male and 1 female) could complete all techniques. The group consisted of 12 persons with paraplegia (6 with complete & 6 with incomplete injury) and 6 with tetraplegia (all with incomplete injury). The group mean (standard deviation) of age, height, weight and years since injury were 36.83 (10.5) years, 1.70 (0.4), 76.21 (20.0) and 13.72 (7.6) years.

4.3.1.1 Balance

Of the 20 participants MFRT (cm) was recorded for 14 participants in the study. The mean MFRT scores for the PG (N =9) and the TG (N =7) were 18.74 (6.19) and 6.35 (5.15) centimeters respectively. Balance was found to be significantly higher for the PG compared to the TG ($p = 0.039$). Five of the participants in the TG and one person in the PG could not perform a functional reach without support.

4.3.1.2 Pain

Pain measures were recorded on 16 participants. A performance corrected WUSPI (PC-WUSPI) score was computed to reflect shoulder pain amongst the groups. There were no significant differences between levels of pain between groups ($p = 0.55$). Although pain scores were in the lower range for both groups, we found the PC-WUSPI scores were 52% higher for the TG (N = 7, 8.92 ± 10.12) compared to the PG (N =9, 5.22 ± 8.33). When looking at the pain scores related solely to transfer tasks, we found that the pain scores were 62% higher for the TG (2.71 ± 3.12) compared to the PG (1.43 ± 3.49).

4.3.1.3 Upper extremity strength

Isokinetic peak torques recorded at the shoulder, elbow and wrist are shown in Figure 25. The lowest strength was recorded at the wrist in comparison to the shoulder and elbow. Strength measures were recorded for sixteen participants. Three participants who could not complete the entire protocol belonged to the TG. Of the three participants, one of them owing to left arm weakness had missing data for all strength measures at the shoulder, elbow and wrist of the left arm. Of the two remaining participants, one had missing data for right shoulder flexion/extension and abduction/adduction. Additionally both had missing data for left wrist flexion extension strength measurements. The strength values were similar across the left and right arms across the entire sample.

Upper Extremity strength amongst stratified sample

a. Groups stratified based on level of injury (LOI):

   i. Group- Low paraplegia (LPG): LOI below T7

   ii. Group- High paraplegia (HPG): LOI above T7 and below C1

   iii. Group- Tetraplegia (TG): Cervical level lesion

No significant strength differences were found with the stratified samples (Table 14 & Table 15). The sample size was small for the three groups with the LPG having 5 participants, HPG 4 participants and the TG 7 participants.

Table 14. Group mean and (standard deviations) of the weight normalized peak torque values for the PG with incomplete injury versus the group with tetraplegia (all participants in this group had incomplete injuries).

Strength Variables (%m)	GROUPS	
	PG - Incomplete Injury N = 5	TG Incomplete injury N=7
Left Elbow Extension	6.45 (2.96)	4.24 (1.95)
Left Elbow Flexion	3.66 (1.47)	3.12(1.66)
Right Elbow Extension	7.08 (3.17)	4.28 (2.19)
Right Elbow Flexion	5.35 (1.89)	2.80 (1.64)
Left Wrist Extension	2.31 (.91)	1.19 (.59)
Left Wrist Flexion	1.20 (0.31)	0.74 (.23)
Right Wrist Extension	.23 (1.00)	0.98 (0.61)
Right Wrist Flexion	1.22 (0.49)	0.70 (0.25)
Left Shoulder Extension	4.75 (1.61)	4.62 (1.44)
Left Shoulder Flexion	6.18 (2.07)	5.64 (1.35)
Right Shoulder Extension	6.43 (2.71)	4.43 (1.29)
Right Shoulder Flexion	6.87 (1.67)	6.54 (.63)
Left Shoulder IR	4.01 (3.73)	3.98 (2.04)
Left Shoulder ER	3.17 (.96)	2.77(1.01)
Right Shoulder IR	4.13 (3.53)	4.16 (2.33)
Right Shoulder ER	3.46 (0.56)	2.57 (1.01)
Left Shoulder Adduction	5.17 (6.05)	5.31 (3.70)
Left Shoulder Abduction	4.55 (1.57)	4.40 (1.44)
Right Shoulder Adduction	5.74 (6.67)	5.83 (3.08)
Right Shoulder Abduction	5.04 (2.02)	5.15 (1.93)

Table 15. Weight normalized peak torque values for the three groups stratified on the level of injury (LOI) : Low Paraplegia, High Paraplegia and Tetraplegia. The group means, standard deviation (SD) and the median are shown for the tested movements for both arms.

STRATIFIED GROUPS BASED ON LOI	Group: Low Paraplegia (LOI below T7, N = 5)			Group: High Paraplegia (LOI above T7, N = 4)			Group: Tetraplegia (LOI Cervical level, N = 7)		
Strength Variables (%m)	Mean	Median	SD	Mean	Median	SD	Mean	Median	SD
Left Elbow Extension	5.56	5.24	1.46	8.92	8.71	2.63	4.23	4.86	1.95
Left Elbow Flexion	3.66	3.11	1.47	4.50	4.10	2.43	3.12	3.66	1.66
Right Elbow Extension	6.26	6.22	2.59	8.30	8.44	2.77	4.27	4.09	2.18
Right Elbow Flexion	4.53	4.42	.62	5.04	4.81	2.88	2.80	3.08	1.64
Left Wrist Extension	2.14	1.68	.94	2.19	2.42	.69	1.18	1.23	.59
Left Wrist Flexion	1.13	1.24	.25	1.10	1.22	.48	.74	.69	.23
Right Wrist Extension	1.92	1.70	.98	2.29	2.21	1.16	.98	1.00	.61
Right Wrist Flexion	1.21	1.22	.48	1.15	1.13	.30	.70	.85	.24
Left Shoulder Extension	5.08	5.07	1.43	6.26	4.95	3.70	4.62	4.71	1.44
Left Shoulder Flexion	6.79	6.89	2.09	6.88	6.18	2.26	5.64	6.13	1.35
Right Shoulder Extension	5.93	5.17	2.45	6.49	6.54	3.18	4.43	4.42	1.29
Right Shoulder Flexion	6.70	7.27	1.55	7.16	7.09	2.25	6.54	6.28	.63
Left Shoulder IR	4.89	4.19	3.52	4.39	3.43	3.26	3.98	4.56	2.03
Left Shoulder ER	3.73	3.33	1.16	3.30	3.20	.90	2.77	2.82	1.01
Right Shoulder IR	4.97	4.13	3.42	4.63	4.11	2.73	4.16	4.48	2.33
Right Shoulder ER	3.72	3.42	.84	3.89	3.68	1.14	2.57	2.90	1.07
Left Shoulder Adduction	5.33	2.39	5.93	5.50	5.85	3.21	5.31	6.72	3.70
Left Shoulder Abduction	4.34	4.36	1.44	5.47	5.70	.63	4.40	4.98	1.44
Right Shoulder Adduction	5.97	3.85	6.55	5.58	5.38	2.59	5.83	7.23	3.08
Right Shoulder Abduction	4.73	5.71	1.61	5.22	5.45	2.17	5.15	4.92	1.93

4.3.2 UE kinetics, UE strength, functional measures and subject demographic correlations

4.3.2.1 Demographic variables

Body weight was positively correlated to several non-weight normalized transfer kinetic variables shown in Table 16. For this reason all other correlation test were run using weight-normalized kinetic data. Superior (r= -0.48, p= 0.031, n = 20) and resultant (r= -0.47, p= 0.036, n = 20) forces, at the trailing shoulder, were found to be negatively correlated to arm/torso ratio. Additionally the medial forces at the trailing hand were found to be negatively correlated to height (r= -0.54, p= 0.015, n = 20).

Table 16. Significant correlations of body weight to upper extremity kinetics recorded during the lift phase of sitting pivot transfer.

Correlations between Body Weight and Transfer kinetics	Component	Correlation	p-value
Trailing shoulder	Posterior force	0.46	0.044
	Shoulder adduction	0.54	0.014
Leading shoulder	Posterior force	0.53	0.019
	Superior Force	0.54	0.013
	Resultant force	0.68	0.001
Trailing Elbow	Resultant Force	0.53	0.015
Leading Elbow	Superior Force	0.55	0.012
	Posterior Force	0.59	0.006
	Resultant Force	0.69	0.001
Trailing Hand	Superior force	0.65	0.002
Leading hand	Superior force	0.70	0.001

Table 17. Significant correlation results between shoulder, elbow and hand forces and strength variables.

Arm	Force (%BW, N= 20)		Upper Extremity Movements (%m)		Sample size	Coefficient of Determination (r)	p-value
LEADING SHOULDER	Superior	15.95 (6.93)	Right shoulder internal rotation	4.53 (2.63)	16	0.52	0.037
	Medial	11.92 (13.15)	Left shoulder abduction	4.67 (1.30)	15	0.58	0.025
TRAILING ELBOW	Posterior	35.37 (14.11)	Left shoulder internal rotation	4.40 (2.74)	15	0.54	0.037
LEADING ELBOW	Posterior	25.25 (7.70)	Left shoulder internal rotation	4.40 (2.74)	15	0.61	0.016
			Right shoulder internal rotation	4.53 (2.63)	16	0.69	0.007
			Right shoulder external rotation	3.26 (1.15)	16	0.56	0.024
			Left shoulder adduction	5.37 (4.15)	15	0.56	0.029
			Left shoulder abduction	4.67 (1.30)	15	0.71	0.003
			Left elbow extension	5.93 (2.75)	15	0.71	0.003
			Right elbow extension	5.90 (2.84)	16	0.62	0.0011
	Resultant	38.77 (10.49)	Left shoulder internal rotation	4.40 (2.74)	15	0.53	0.044
			Right shoulder internal rotation	4.53 (2.63)	16	0.50	0.047
TRAILING HAND	Anterior	15.00 (12.27)	Left elbow extension	5.93 (2.75)	15	0.58	0.023
			Right elbow extension	5.90 (2.84)	16	0.54	0.031
	Resultant	46.14 (10.53)	Right shoulder abduction	5.03 (1.77)	15	0.54	0.037
LEADING HAND	Anterior	2.17 (3.42)	Left wrist flexion	1.86 (0.85)	13	0.58	0.039
	Medial	3.81 (3.42)	Left wrist flexion	1.86 (0.85)	13	0.54	0.046

Table 18. Significant correlation results between upper extremity joint moments and strength variables.

Arm	Moment (%m)		Upper extremity movement (%m)		Sample size	Correlation of Determination (r)	p value
TRAILING ARM	Elbow Extension	2.59(4.82)	Right shoulder extension	5.48 (2.31)	16	0.70	0.004
			Right elbow extension	5.90 (2.84)	16	0.50	0.047
			Right wrist flexion	0.97 (0.41)	16	0.54	0.029
LEADING ARM	Shoulder Adduction	2.04 (2.80)	Left wrist flexion	1.86 (0.85)	13	0.71	0.006
	Elbow Flexion	1.03 (1.91)	Right shoulder extension	5.48 (2.31)	16	0.67	0.006
			Right shoulder external rotation	3.26 (1.15)	16	0.55	0.026
			Right elbow extension	5.90 (2.84)	16	0.67	0.004
			Right wrist flexion	0.97 (0.41)	16	0.59	0.017
	Wrist Extension	1.02 (4.03)	Right shoulder internal rotation	4.53 (2.63)	16	0.53	0.034
			Right shoulder external Rotation	3.26 (1.15)	16	0.78	p<0.001
			Left shoulder abduction	4.67 (1.30)	15	0.61	0.015
			Left elbow extension	5.93 (2.75)	15	0.71	0.003
			Right elbow extension	5.90 (2.84)	16	0.72	0.002
			Right wrist flexion	0.97 (0.41)	16	0.51	0.043

4.3.2.2 Functional measures

No significant associations were found between UE kinetics and shoulder pain and balance.

4.3.2.3 UE strength

Shoulder

Medial forces at the leading shoulder were significantly correlated to the left shoulder abduction strength (Table 17). Lead shoulder adduction moments were found to be positively correlated to the left wrist flexion strength variables (Table 18).

Elbow

Elbow posterior and resultant forces were found to be significantly correlated to the shoulder internal/external rotation, adduction/abduction and elbow extension strength variables (Table 17). Elbow flexion/extension moments were significantly correlated to the right arm shoulder extension/external rotation, elbow extension and wrist flexion strength variables (Table 18).

Hand

Anterior forces were found to be significantly correlated to the elbow extension (left/right) on the trailing side and left wrist flexion on the leading arm (Table 17). Additionally resultant forces were significantly correlated to the right shoulder abduction strength variables. Medial forces at the leading hand were significantly correlated to the strength of the left wrist flexion variable (Table 17).

4.4 DISCUSSION

The purpose of this study was to examine the relationship between UE joint loading and anthropometry, functional measures and UE strength.

4.4.1 Demographics and anthropometrics

We found the net resultant force and the large vertical force at the trailing shoulder were significantly lower for individuals with longer arms and shorter torso. Although not a strong correlation, the results indicate that this anthropometric attribute may provide an increased mechanical advantage thereby potentially alleviating the demand at the shoulder by easing the lifting process (67). This finding in part may explain why women, who tend to have smaller torso to arm ratios, report higher percentages of upper limb pain due to transfers (8).

4.4.2 Pain

Our study sample consisted of high functioning MWC users with SCI. Functioning amongst persons with SCI has been found to be inversely related to pain (25). The self reported pain measurements were considerably low for both groups and may explain why we did not find a link between pain and joint kinetics during transfers. The observation that TG had higher pain scores compared to the PG. corroborates with previous literature that found people with tetraplegia reportedly experience higher intensity and prevalence of shoulder pain as compared to those with paraplegia (25, 28, 32).

4.4.3 Balance

Similar to previous literature we found that balance is sensitive to level of injury (72). Chen et al (72) identified that sitting balance is necessary to perform functional activities. Although five of the seven subjects with TG had no unsupported seated balance, it did not influence their ability to successfully perform an SPT. In a recent study, the relationship between transferring capability and physical function was examined for SPTs that varied on three factors: height, gap and transferring with a side guard (73). They found balance measured using the same MFRT scale to be a significant predictor of transferring capability i.e. greater stability equated to higher transfer capability. Non- level transfers have been shown to have a higher demand of shoulder muscle activation in comparison to level SPTs (74). Our study only analyzed level SPTs, which could explain why we didn't find a significant association between sitting balance and transfer kinetics.

4.4.4 Strength

Contrary to our belief the overall association between strength and transfer kinetics were all positive in nature. This indicates that the stronger the participant the greater the mechanical loading observed at the UE joints, particularly in regards to the vertical and medial forces at the shoulder, non-vertical forces at the elbow and hand, flexion/extension moments at the elbow and extension moments at the wrist. Ambrosio et al (75) examined the relationship between isokinetic shoulder strength and manual wheelchair biomechanics. They found that an increase in strength does not necessarily indicate the use of an optimal propulsion strategy. And that training is often necessary to adopt a propulsion technique that reduces forces borne by the upper limbs.

The amount and quality of transfer skill training varies widely across rehab facilities and clinicians. For individuals who are not skilled in performing 'best' transfer techniques it is possible that being stronger affords greater freedom in choice of hand and trunk positions and trajectory of body movements and that sub-optimal strategies are chosen get from point A to B resulting in increased loading. An assessment of transfer skill (e.g. Transfer Assessment Instrument (76)) should be included in future studies to evaluate the association between quality of transfer and the resulting biomechanics. The data seem to suggest that either weaker individuals have been trained better how to transfer or that weaknesses in upper limb muscles may force adopting a technique that not only enables these individuals to be successful with the transfer but also minimizes mechanical loading at the joints. Future work should analyze the transfer techniques of those with upper limb weakness similar to that described in Chapter 2 to determine what aspects of their technique (e.g. hand placement, feet and trunk positioning) may be lending to the reduction in the joint forces/moments during transfer.

During transfers the primary muscles groups engaged are the shoulder flexors, abductors/adductors, elbow extensors, and wrist flexors (16, 77). However no significant correlations were identified with the respective strength variables: shoulder flexion strength, elbow flexion strength, and wrist extension. The shoulder flexors in particular is a predominant muscle group involved in transfers(21). Based on the nature of SPTs a certain level of shoulder strength is required to facilitate a successful lift and enable a person to transfer successfully. However there could be a threshold effect wherein having shoulder strength above the threshold, does not affect the capability of performing a transfer. The strength variables that were correlated were those that might assist less with lifting and more with dynamic stabilization of the body as it rotates and pivots from one surface to the other. Therefore the results point toward training to assist with control and execution of a smooth successful transfer.

Shoulder internal rotation strength was found to be significantly associated with majority of the non-vertical forces and the flexion/extension moments across the UE joints. An interesting observation of the strength correlations are the contralateral findings of left strength variables affecting right side kinetics and vice versa. The findings suggest how the strength of the contralateral side is utilized in assisting with both upper limb stabilization and dynamic postural stability as the weight is shifted during the transfer. This may be especially true for those individuals who have higher trunk impairment level. The stabilization rationale can be further supported when we observe, particularly for the association with the non-vertical forces (anterior/posterior and medial), agonist and antagonist muscle strengths being correlated to the same transfer kinetic variable.

Contrary to prior research, a secondary analysis yielded no significant differences between upper limb strength and level of SCI. This may have happened because the sample size was too small making it difficult to detect statistical differences. Or it may have occurred because the demands of the protocol (e.g. able to perform 30+ transfers in a short time frame) excluded subjects who were physically less able, weaker, lower stamina, etc.

Extension moments were found to be the largest and were seen at the trailing wrist, while flexion moments were observed at the leading wrist. The generated moments can be due to the gripping style of the force beam on the trailing side while on the leading side the hand was placed flat on the tub bench. Most common correlates to the transfer moments were chiefly elbow extension and wrist flexion strength. For individuals with higher level spinal cord injuries (cervical and high thoracic injury level), the elbow extension moment during transfers usually occurs through the generation of large shoulder and wrist (passive stretching) flexion moments (refer to Chapter 1 results). Again this ‘technique’ to generate passive extension enables them to be able to be successful with transfers while subsequently lowering forces and moments at the joints. The moment and upper limb strength correlations infer that individuals with greater elbow and wrist function are transferring in sub-optimal ways that generate larger moments at all three joints.

4.4.5 Limitations

Strength measurement using the Biodex dynamometer has been proven to be a reliable measurement technique (78, 79). However a study by Souza et al (80) that compared UE strength measures between people with SCI and unimpaired controls found that proximal trunk strength can aid in generating upper limb forces. Therefore it’s possible we observed an under-estimation of strength for subjects with higher levels of SCI. However a secondary analysis using data collected in this study found no relationship between strength and level of SCI thus strengthening the finding that having strength above a threshold necessary to complete a successful transfer may have a negative impact on mechanical loading at the joints in the absence of training. The preliminary descriptive results and the small sample size prevented us from exploring the predictive relationship of how functional measures and strength influence transfer biomechanics.

4.5 CONCLUSION

Transfer strategy may play a stronger role when considering UE joint loading as compared to the influence of functional capacity. Future investigations must examine the impact of early interventions to understand safe transfer strategies and ultimately preserve UE function.

5.0 CONCLUSION

The goal of this dissertation was to get a better understanding of the upper extremity mechanical loading while performing sitting pivot transfers (SPTs) amongst persons with spinal cord injury (SCI), as they performed their preferred method of transferring and taught techniques. The taught techniques varied on two factors- hand placement and trunk flexion.

The experimental protocol employed in this study went through a few modifications once data collection had begun. The experimental set up was upgraded to collect feet forces after collecting hence we have feet forces for 11 out of the 20 participants. The functional measures were included in the protocol, after the first 4 participants had been tested, thus balance, strength and pain scores were collected on 16 participants. Additionally 19 of the 20 participants could transfer successfully with the taught techniques. Of the 19 participants, one participants' data for the HH-I technique could not be used due to technical difficulties while recording the trials. The participants' data could not be used due to the presence of multiple ghost markers that could not be rectified during post processing.

Due to our inclusion criteria and the objective of the study being, investigation of weight bearing at the UEs during SPTs, we did have to excluded a number of participants from participating in the study, who employed active weight bearing on their LEs (i.e. stand-pivot transfer). We initially had participants performing 10 transfers/ techniques but found that to be very exhaustive for the participant, who also completed the strength testing protocol prior to the transfer testing. Therefore to make testing more efficient we reduced the number of transfers to 5 per technique.

Our first study was to acquire knowledge on how UE joint biomechanics compared between persons with paraplegia and tetraplegia. Although kinetic differences (statistical differences and trends) were detected, the study was underpowered to detect kinematic differences. The sample being analyzed presented a complex scenario due to the TG participants having incomplete injuries. Persons with incomplete injuries are associated with huge variability in functioning. Therefore although we did not have ASIA scores for the participants we did have UE strength measures that were analyzed between the two groups that could be used to help interpret the kinetic differences obtained. Overall we had a high functioning group and the strength differences were primarily associated with the elbow and wrist flexor/extensor strength variables In accordance with our primary hypotheses we found that TG group generated significantly larger shoulder flexion and wrist flexion moments to facilitate the SPT. This not only provides a better understanding of the execution of the SPT but also highlights the difference demands on the UE by the groups. The TG tended to have lower strength for wrist and elbow flexion/extension movements. Therefore the passive wrist flexion and the closed kinematic chain with the hand fixed on a surface allow for the passive extension and stabilization of the elbow, which was represented by our results. Additionally the medio-lateral forces at the shoulder tended to be higher for the TG compared to the PG. The non-vertical forces, medio-lateral, associated with stabilization tended to be higher for the TG compared to the PG. The results of the study shed light on potential risk of injuries that may occur for each individual group based on the different loading at the UEs.

The second study focused on conducting a thorough kinetic comparison of the taught techniques as well as investigating the preferred method of transferring compared to the taught techniques. The third aim of the study was to classify the preferred method of transferring to one of the taught techniques based on the factors that the taught techniques varied on. Comparison of the taught techniques presented an imbalance in the superior forces and net resultant forces, being higher at the trailing arm and lower at the leading arm when transferring with the HH-A and TU techniques in comparison to the HH-I technique across all joints. This imbalance draws attention to leading hand placement being a key factor to balancing the distribution of forces across the upper limb joints. Transferring with the HH-I resulted in the reduction of non-vertical forces as well, meaning less demand on the UEs to maintain postural stability. From the comparison of the taught techniques we found that there was a trade off based on hand placement. Placing the leading hand further away from the body led to reduced loading at the leading arm and higher loading at the trailing arm, which was vice versa when placing the hand close to the body. Therefore based on UE weakness/impairment choice of the leading arm is of the utmost importance, in light of the trade off that occurs based on leading hand placement. For example if there is upper arm weakness/impairment in the right arm, based on the results, choosing the left arm as the leading arm and placing the leading hand close to the body will allow for less loading on the trailing arm.

The biomechanics of the self-selected transferring technique resembled the HH-I style of transferring. However compared to the taught techniques the SS transfer technique resulted in a more even distribution of mechanical loading across the limbs. In order to understand the transfer strategy applied by the participants as they performed the SS transfer and how it related to the taught techniques we analyzed the two variables that defined each of the taught techniques: leading hand placement and trunk flexion. We observed that 10 out of the 18 participants used a TU (4) or a TU* (modified version of TU: Trunk upright and hand close to the body; transfer strategy. The remaining 8 participants used the head-hips methods of transferring with 5 placing their hand further away from their body and 3 placing their leading hand close to the body. Based on our results we found that transferring with the SS technique resulted in significantly lower net resultant forces at all three upper limb locations, at the trailing arm, compared to the TU technique. From our classification we can conclude that these differences can be explained based on the greater influence of the leading hand placement compared to the amount of trunk flexion.

Groups defined by the classification of the SS technique into the taught technique were further examined to determine if there was any statistical difference in terms of net resultant forces at the UE joint or demographics influenced by hand placement or trunk flexion. We did not find any differences in the net resultant forces. However we found that amongst our participants those who chose to place their hand further away from their body were of a lower age compared to those who chose to place their hand closer to their body. This may shed light to the compensatory mechanisms that are employed to preserve UE function with time. Evaluation of the taught techniques, were based on short term transfer training therefore future studies should look at conducting longitudinal studies to understand the long term benefits of the transfer.

The third study looked at the how functional measures and strength related to transfer kinetics. Strength variables were positively correlated to the transfer kinetics variables suggesting that increased strength does not necessarily translate to an optimal strategy being used. We found no significant correlations with the respective strength variables: shoulder flexion strength, elbow flexion strength, and wrist extension. Based on the nature of SPTs a certain level of shoulder strength is required to facilitate a successful lift and enable a person to transfer successfully. However there could be a threshold effect wherein having shoulder strength above the threshold, does not affect the capability of performing a transfer. Interestingly the positive nature of the correlations suggest that the stronger the participant the greater the mechanical loading observed at the UE joints, particularly in regards to the vertical and medial forces at the shoulder, non-vertical forces at the elbow and hand, flexion/extension moments at the elbow and extension moments at the wrist. The data seem to suggest that either weaker individuals have been trained better how to transfer or that weaknesses in upper limb muscles may force adopting a technique that not only enables these individuals to be successful with the transfer but also minimizes mechanical loading at the joints. Future work should analyze the transfer techniques of those with upper limb weakness similar to that described in Chapter 2 to determine what aspects of their technique (e.g. hand placement, feet and trunk positioning) may be lending to the reduction in the joint forces/moments during transfer. An assessment of transfer skill should be included in future studies to evaluate the association between quality of transfer and the resulting biomechanics.

Additionally we examined one of the simplest scenarios of SPT's i.e. level transfers. It would be beneficial to see how the mechanical loading varies when the taught techniques are executed for non-level transfers. Another setting that requires further investigation are overhead reach transfers.

APPENDIX A

INVERSE DYNAMICS PROGRAM

```
%References used in this program:

%Hanavan, EP.  A Mathematical Model of the Human Body.  Wright-Patterson Air
Force Base. Pub:AMRL-TR-64-102, 1964.

%Winter, DA. Biomechanics and Motor Control of Human Movement, Second Edition.
Wiley-Interscience, New York, 1990.

%Cooper RA, Boninger ML, Shimada SD, Lawrence BM. (1999) Glenohumeral Joint
Kinematics and Kinetics for
%Three Coordinate System Representations During Wheelchair Propulsion. Am J
Phys Med Rehab. 78(5):435-446.

%Wu G, van der Helm FCT, Veeger HEJ, Makhsous M, Van Roy P, Anglin C,
%Nagels J, Karduna AR, McQuade K, Wang X, Werner FW, Bucholz B. (2005) ISB
%recommendation on definitions of joint coordinate systems of various
%joints for the reporting of human joint motion-PartII: shoulder, elbow,
%wrist, and hand. Journal of Biomechanics. 38: 981-992.
close all
clear all
%% Load Subject data
subject_num = 53;
ID= num2str(subject_num);
trial_num = [3];
% trial_num = [2];
for j = 1:length(trial_num)

    transfer_trial = num2str(trial_num(j));
    technique= 'U';
    % %
%         filename = ['AnalyzedKinematicData_TX', ID, 'V',technique,
'L',transfer_trial];
%         filename_kinetic = ['AnalyzedKineticData_TX', ID, 'V',technique,
'L',transfer_trial];
    %     filename_test = ['AnalyzedkinematicData_TX', ID, 'V',technique,
'R',transfer_trial];
%         filename_kinetic = ['AnalyzedKineticData_TX', ID, 'V',technique,
'R',transfer_trial];
    %     load(filename_kinetic)

    filename = ['AnalyzedData_TX', ID, 'V',technique, 'L',transfer_trial];
```

```
%      filename = ['AnalyzedKinematicData_TX', ID, 'V',technique,
'L',transfer_trial];
load (filename)
% load (filename_kinetic)
    filename_position = ['MarkerData_TX',ID, 'V', technique, 'L',
transfer_trial];
%     filename_position = ['MarkerData_TX',ID, 'V', technique, 'R',
transfer_trial];
%     filename = ['AnalyzedData_TX', ID, 'V','L',technique,transfer_trial];
% TX24
%      load (filename)
%     filename_position = ['MarkerData_TX',ID,
'V','L',technique,transfer_trial];%TX24

    load (filename_position,
'C7_global','T3_global','T8_global','RSHO_global','RMEP_global',
'RLEP_global', 'RRS_global','RUT_global','R3MP_global',......
        'LSHO_global','LMEP_global', 'LLEP_global',
'LRS_global','LUT_global','L3MP_global');


    %% PHASE DEFINITION
    %
    kinematic_trial = ['TX' ID 'V' technique 'L' transfer_trial];
%     load (kinematic_trial, 'x', 'kinetics')
%     kinematic_trial = ['TX21VLL2'];
    load (kinematic_trial, 'x')

    bench_force = [Fx_FP_Bench Fy_FP_Bench Fz_FP_Bench ];
        bench_force = padarray(bench_force,[1 0],'post');
    wc_force = [Fx_FP_WC Fy_FP_WC Fz_FP_WC];
    LC_force = [Fx_LC Fy_LC Fz_LC];
%     bench_force = bench_force(1:length(LC_force),:);
%     wc_force = bench_force(1:length(LC_force),:);

    kinetics = [bench_force wc_force LC_force];
%     kinetics = [bench_force wc_force LC_force feet_force];

%% filtering feet forces & resampling forces

            [b,a]=butter(4,10/300);
%
%             test = Kinetics(1,4).FP_Feet;to be used for TX53
%             feetFP = Kinetics.FP_Feet;
            %feet_force = feetFP (:,1:3);
            feet_force= Kinetics(1,4).FP_Feet;
             feet_force = padarray(feet_force,[663 0],'post');
            filt_feet= filtfilt(b,a,feet_force);
            feet_FP = resample(filt_feet(:,1:3),60,360);
% % Johnny's pgm forces to Herl coordinate system- Subjects TX21 - TX33
% Bench Force
% x = x
% y = y
% z = -z

% LC Force
% x = -x
% y = -y
% z = -z

% % Johnny's pgm forces to Herl coordinate system- Subjects TX40- TX53
% Bench Force
% x = x
```

```
% y = y
% z = z

% LC Force
% x = x
% y = y
% z = z

    bench_FP = resample(bench_force,60,360);
    bench_FP(:,3)= -bench_FP(:,3);
    WC_FP = resample(wc_force,60,360);
%     loadcell= resample(-LC_force,60,360);
      loadcell= resample(LC_force,60,360);
    kinetics_processed.bench_FP =bench_FP;
    kinetics_processed.WC_FP =WC_FP ;
    kinetics_processed.loadcell =loadcell ;
    kinetics_processed.feet_FP= feet_FP;
%     plot_variables = [bench_FP(1:length(C7_global),2)
WC_FP(1:length(C7_global),2) loadcell(1:length(C7_global),2) C7_global(:,3)
T3_global(:,3)];


    %     end

    %    if length(feet_force)~=length(bench_force)
    %        diff_length = length(bench_force)
    %        feet_force=zeros (feet_force)

    %% OLD CODE
    %----------------Load subject data----------------------------------%
    %User input
    % newID= input('Enter the subject 4 digit ID: ', 's');
    % condition = input('Enter transfer type (i.e. L, U, A, I, or T) : ',
's');
    % side = input('Enter non-dominant side: ', 's');
    % trial = input('Enter trial number: ', 's');
    %
    % %Kinematic data
    % kin=[newID,'V',condition,side,trial,'s'];
    % kin=load(kin);
    % %kinetic data
    % forcedata=[newID,'fm','V',condition,side,trial,'c'];
    % load(forcedata);
    %
    % FMr_raw=loadcell;
    % FMl_raw=forceplate1;
    %  [r,c]=size(kin);
    % %Calculate baseline to subtract from forces
    % fmrbaseline=mean(FMr_raw(1:10,:));
    % fmlbaseline=mean(FMl_raw(1:10,:));
    % %remove baseline to get actual force data
    % for f=1:r
    % FMr(f,:)=FMr_raw(f,:)-fmrbaseline;
    % FMl(f,:)=FMl_raw(f,:)-fmlbaseline;
    % end

    %% Anthropometric
    %anthropometric data (first row= height(in), second row=weight(lbs), rest
of measurements are in meters)
    %rows 3-8 are for the right side: axillary arm circ, elbow circ, wrist
    %circ, fist circ, upper arm length, forearm length
    %rows 9-14 are for the left side: axillary arm circ, elbow circ, wrist
    %circ, fist circ, upper arm length, forearm length
```

```
    anth=['anthroTX',ID, '.txt'];
    anthro=load(anth);


    %----------------define anthropometric variables (used for both sides) ---
--------------%
    heightinch=anthro(1); %height in inches
    heightm=heightinch*0.0254; %height in meters
    weightlbs=anthro(2); %weight in pounds
    weightN=weightlbs*4.448222; %weight in Newtons
    pindex=heightinch/(weightlbs^(1/3)); %ponderal index (Winter pg. 53)
    bodydenkgl=0.69 + (0.0297*pindex); %body density in kg/l
    bodyden=bodydenkgl/.001; %body density in kg/m^3
    swua=0.5*(0.08*weightlbs-2.9); %segment weight of upper arm in lbs
(Hanavan)
    swfa=0.5*(0.04*weightlbs-0.5); %segment weight of forearm in lbs (Hanavan)
    swha=0.5*(0.01*weightlbs-0.7); %segment weight of hand in lbs (Hanavan)
    handdens=1.16/.001; %hand density in kg/m^3 from Winter
    fadens=1.13/.001; %forearm in kg/m^3 density
    uadens=1.07/.001; %upper arm in kg/m^3 density

%% Converting from Vicon Coordinate system to HERL coordinate system

    kin = [C7_global T3_global T8_global RSHO_global RMEP_global RLEP_global
RRS_global RUT_global R3MP_global,.....
    LSHO_global LMEP_global LLEP_global LRS_global LUT_global L3MP_global];
    kintemp = kin;

    for i=1:size(kintemp,2)
        if i>15
        break;
        else
        kin(:,1+3*(i-1)) = -kintemp(:,2+3*(i-1));
        kin(:,2+3*(i-1)) = kintemp(:,3+3*(i-1));
        kin(:,3+3*(i-1)) = -kintemp(:,1+3*(i-1));
        end
    end

    [r,c]=size(kin);
    %----------get rid of NaNs left over from interp-------------------%
    a=isnan(kin);
    for t=1:length(kin)
        for c=1:c
            if a(t,c)==1;
                kin(t,c)=0;
            end
        end
    end
    kin=(kin/1000); %convert from mm to meters
    kin=kin+1; %shifts data by 1 meter so that all coordinates are positive
    [kinrows,kincolumns]=size(kin);
       processed_kinematics.c7 = kin (:,1:3);
    processed_kinematics.T3= kin (:,4:6);
    processed_kinematics.T8 = kin (:,7:9);
    processed_kinematics.RSHO = kin (:,10:12);
    processed_kinematics.RMEP= kin (:,13:15);
    processed_kinematics.RLEP = kin (:,16:18);
    processed_kinematics.RRS = kin (:,19:21);
    processed_kinematics.RUT = kin (:,22:24);
    processed_kinematics.R3MP = kin (:,25:27);
    processed_kinematics.LSHO = kin (:,28:30);
    processed_kinematics.LMEP= kin (:,31:33);
```

```
    processed_kinematics.LLEP = kin (:,34:36);
    processed_kinematics.LRS = kin (:,37:39);
    processed_kinematics.LUT = kin (:,40:42);
    processed_kinematics.L3MP = kin (:,43:45);

    plot_variables.bench = bench_FP;
    plot_variables.wc= WC_FP;
    plot_variables.lc= loadcell;
    plot_variables.c7= kin(:,1:3);
    plot_variables.t3= kin(:,4:6);

    %% Old Code

    % %----------------Filter kinematic data----------------------------%
    % [kinrows,kincolumns]=size(kin);
    % [b,a]=butter(2,7/30); %defines 4th order Butterworth filter with 7Hz
cutoff frequency ; 30 = F/2
    % for i=1:kincolumns
    %     filteredkin(:,i)=filtfilt(b,a,kin(:,i)); %runs filter
    % end
    %
    % kin=(filteredkin/1000); %convert from mm to meters
    % kin=kin+1; %shifts data by 1 meter so that all coordinates are positive


    %% Variable Definition
    n=1;
   FMr =loadcell;
   FMl = bench_FP;
% FMl = bench_force;
    for n=1:2

        %-------------Define variable names for right and left side-----------
----------%
        if n==1 %right side
            FM=FMr;
            forces=FM(:,1:3); %sampled force data
%             plot(forces(:,2))

            thirdmp=kin(:,25:27); %third MP                 % we can just
rename them!!!!
            radsty=kin(:,19:21); %radial styloid
            ulnsty=kin(:,22:24); %ulnar styloid
            wristcen=0.5*(radsty+ulnsty); %wrist center
            rmep=kin(:,13:15); %medial epicondyle for transfers
            latep=kin(:,16:18); %lateral epicondyle
            acro=kin(:,10:12); %acromion
            t3=kin(:,4:6);%t3
            t8=kin(:,7:9);%t8
            axilc=anthro(3); %axillary arm circumference
            elbc=anthro(4); %elbow circumference
            wrc=anthro(5); %wrist circumference
            fistc=anthro(6); %fist circumference
            ualen=anthro(7); %upper arm length
            falen=anthro(8); %forearm length
        else %left side
            FM=FMl;
            forces=FM(:,1:3); %sampled force data
%             plot(forces(:,2))

            thirdmp=kin(:,43:45); %third MP
            radsty=kin(:,37:39); %radial styloid
            ulnsty=kin(:,40:42);%ulnar styloid
```

```
            wristcen=0.5*(radsty+ulnsty); %wrist center
            lmep=kin(:,31:33); %Left medial epicondyle
            latep=kin(:,34:36); %lateral epicondyle
            acro=kin(:,28:30); %acromion
            t3=kin(:,4:6);%t3
            t8=kin(:,7:9);%t8
            axilc=anthro(9);%axillary arm circumference
            elbc=anthro(10); %elbow circumference
            wrc=anthro(11); %wrist circumference
            fistc=anthro(12); %fist circumference
            ualen=anthro(13); %upper arm length
            falen=anthro(14); %forearm length

        end %end of if loop to set FM file to FMr or FMl

        %---------------Calculate mass moment of inertia / center of mass-----
-----------------------%

        g=9.81; %gravity m\s^2
        dt=1/60; %sampling interval

        %upper arm
        uapr=axilc/(2*pi); %upper arm proximal radius (shoulder)
        uadr=elbc/(2*pi); %upper arm distal radius (elbow)
        uavol=(pi*ualen/3*(uapr^2+uapr*uadr+uadr^2)); %segment volume in m^3
(modeled as elliptical cylinder (Hanavan))
        uamass=uadens*uavol;  %upper arm mass in kg (density in kg/m^3)
        uamu=uadr/uapr; %radius ratio constant "mu" defined by Hanavan
        uasigma=1+uamu+uamu^2; %constant "sigma" defined by Hanavan
        uaAA=(9/(20*pi))*((1+uamu+uamu^2+uamu^3+uamu^4)/(uasigma^2));
%constant AA defined by Hanavan
        uaBB=(3/80)*((1+4*uamu+10*uamu^2+4*uamu^3+uamu^4)/(uasigma^2));
%constant BB defined by Hanavan

        %check to make sure y is longitudinal and x,z are perpendicular to
        %longitudinal
        uaIxx=uamass*((uaAA*(uamass/(uadens*ualen)))+uaBB*(ualen^2)); %moment
of inertia perpendicular to longitudinal axis(kg*m^2)
        uaIzz=uaIxx; %moment of inertia perpendicular to longitudinal
axis(kg*m^2)
        uaIyy=(3/10)*uamass*((uapr^5-uadr^5)/(uapr^3-uadr^3));%moment of
inertia about the longitudinal axis of the upper arm (kg*m^2)
        uaIxy=0;
        uaIxz=0;
        uaIyz=0;
        uaI=[uaIxx uaIxy uaIxz; uaIxy uaIyy uaIyz; uaIxz uaIyz uaIzz]; %matrix
of upper arm mass moments of inertia

uacmratio=((uapr^2+2*uapr*uadr+3*uadr^2))/(4*(uapr^2+uapr*uadr+uadr^2));
%upper arm center of mass ratio (center of mass/length)with respect to
proximal end (Hanavan)
        uacm=uacmratio*(latep-acro)+acro; %3-D coordinates of upper arm center
of mass

        %forearm
        fapr=elbc/(2*pi); %forearm proximal radius (elbow)
        fadr=wrc/(2*pi); %forearm distal radius (wrist)
        favol=(pi*falen/3*(fapr^2+fapr*fadr+fadr^2)); %segment volume in m^3
(modeled as elliptical cylinder (Hanavan))
        famass=fadens*favol;  %forearm mass in kg (density in kg/m^3)
        famu=uadr/uapr; %radius ratio constant "mu" defined by Hanavan
        fasigma=1+uamu+uamu^2; %constant "sigma" defined by Hanavan
        faAA=(9/(20*pi))*((1+famu+famu^2+famu^3+famu^4)/(fasigma^2));
```

```
%constant AA defined by Hanavan
        faBB=(3/80)*((1+4*famu+10*famu^2+4*famu^3+famu^4)/(fasigma^2));
%constant BB defined by Hanavan
        faIyy=famass*((faAA*(famass/(fadens*falen)))+faBB*(falen^2)); %moment
of inertia perpendicular to longitudinal axis(kg*m^2)
        faIzz=faIyy; %moment of inertia perpendicular to longitudinal
axis(kg*m^2)
        faIxx=(3/10)*famass*((fapr^5-fadr^5)/(fapr^3-fadr^3));%moment of
inertia about the longitudinal axis of the forearm (kg*m^2)
        faIxy=0;
        faIxz=0;
        faIyz=0;
        faI=[faIxx faIxy faIxz; faIxy faIyy faIyz; faIxz faIyz faIzz]; %matrix
of forearm mass moments of inertia

facmratio=((fapr^2+2*fapr*fadr+3*fadr^2))/(4*(fapr^2+fapr*fadr+fadr^2));
%upper arm center of mass ratio (center of mass/length) with respect to
proximal end (Hanavan)
        facm=facmratio*(wristcen-latep)+latep; %3-D coordinates of forearm
center of mass

        %hand
        handrad=fistc/(2*pi); %hand radius
        handvol=(4/3)*pi*handrad^3; %hand volume in m^3
        handmass=handdens*handvol; %hand mass in kg
        handIany=(2/5)*handmass*handrad^2; %hand mass moment of inertia about
any axis (kg*m^2)
        handI=[handIany 0 0; 0 handIany 0; 0 0 handIany];
        handcmratio=0.5; %center of mass ratio for the hand (sphere) (Hanavan)
        handcm=handcmratio*(thirdmp-wristcen)+wristcen; %3-D coordinates of
hand center of mass

        %Save all segment masses into a matrix
        %1x3 matrix
        massall=[handmass famass uamass];

        %Save all center of mass locations in a matrix
        %kinrows(1200)x9 matrix
        cmall=[handcm facm uacm];

        %------------------------------Calculate absolute limb angular
positions----------------------------%

        %Upper Arm
        upperarmvector=latep-acro; %vector along the long axis of the upper
arm
        uazyangle=atan2(upperarmvector(:,2),upperarmvector(:,3)); %absolute
upper arm angle in ZY plane
        uaxzangle=atan2(upperarmvector(:,3),upperarmvector(:,1)); %absolute
upper arm angle in XZ plane
        uaxyangle=atan2(upperarmvector(:,2),upperarmvector(:,1)); %absolute
upper arm angle in XY plane

        %Forearm
        forearmvector=wristcen-latep; %vector along the long axis of the
forearm
        fazyangle=atan2(forearmvector(:,2),forearmvector(:,3)); %absolute
forearm angle in ZY plane
        faxzangle=atan2(forearmvector(:,3),forearmvector(:,1)); %absolute
forearm angle in XZ plane
        faxyangle=atan2(forearmvector(:,2),forearmvector(:,1)); %absolute
forearm angle in XY plane
```

```
        %Hand
        handvector=thirdmp-wristcen; %vector along the long axis of the hand
        handzyangle=atan2(handvector(:,2),handvector(:,3)); %absolute hand
angle in ZY plane
        handxzangle=atan2(handvector(:,3),handvector(:,1)); %absolute hand
angle in XZ plane
        handxyangle=atan2(handvector(:,2),handvector(:,1)); %absolute hand
angle in XY plane

        %-------------------------Calculate angular velocities and
accelerations------------------------------%
        %Velcities and accelerations calculated according to 3 point centered
different method (Winter)

        %store absolute angles in a single matrix
        %kinrows(1200)x9 matrix
        angles=[uazyangle uaxzangle uaxyangle fazyangle faxzangle faxyangle
handzyangle handxzangle handxyangle];

        %check to make sure all angles are in proper quadrant
        for row=1:kinrows
            for col=1:9
                if angles(row,col) <= -pi
                    angles(row,col)=(angles(row,col)+2*pi);
                elseif angles(row,col) > pi
                    angles(row,col)=(angles(row,col)-2*pi);
                end
            end
        end


        %calculate velocities
        count1=2;
        for count1=2:(kinrows-1)
            velocities(count1,1:9)=(angles(count1+1,:)-angles(count1-
1,:))/(2*dt);
            count1=count1+1;
        end
        %correct # of rows
        velocities(1,1:9)=velocities(2,1:9);
        velocities(kinrows,1:9)=velocities((kinrows-1),1:9);

        %calculate accelerations
        index1=2;
        for index1=2:(kinrows-2)
            accelerations(index1,1:9)=(velocities(index1+1,:)-
velocities(index1-1,:))/(2*dt);
            index1=index1+1;
        end
        %correct # of rows
        accelerations(1,1:9)= accelerations(2,1:9);
        accelerations((kinrows-1),1:9)= accelerations((kinrows-2),1:9);
        accelerations(kinrows,1:9)= accelerations((kinrows-2),1:9);

        %previously used to check data
        %kept here in case troubleshooting needs to be done in the future
        %     if n==1
        %         angaccr=accelerations;
        %         angvelr=velocities;
        %         angr=angles;
        %     else
        %         angaccl=accelerations;
        %         angvell=velocities;
```

```
        %         angl=angles;
        %     end

        %--------------------------Calculate linear velocities and
accelerations--------------------------------%
        %Velcities and accelerations calculated according to 3 point centered
different method (Winter)

        %Calculate linear velocities and accelerations for center of mass of
each segment

        %linear velocities of center of mass
        count2=2;
        for count2=2:(kinrows-1)
            cmvel(count2,1:9)=(cmall(count2+1,:)-cmall(count2-1,:))/(2*dt);
            count2=count2+1;
        end
        %correct # of rows
        cmvel(1,1:9)=cmvel(2,1:9);
        cmvel(kinrows,1:9)=cmvel((kinrows-1),1:9);

        %linear accelerations of center of mass;
        index2=2;
        for index2=2:(kinrows-2)
            cmaccel(index2,1:9)=(cmvel(index2+1,:)-cmvel(index2-1,:))/(2*dt);
            index2=index2+1;
        end
        %correct # of rows
        cmaccel(1,1:9)=cmaccel(2,1:9);
        cmaccel((kinrows-1),1:9)=cmaccel((kinrows-2),1:9);
        cmaccel(kinrows,1:9)=cmaccel((kinrows-2),1:9);

        %---------------------Calculate Net Joint Reaction Forces and Moments-
------------------------------%
        %Reference is Cooper et al. Glenohumeral Joint Kinematics and
Kinetics.....Am J Phys Med Rehab 1999.
        %All variable names in reference to Cooper et al.

        %Define blank arrays to be filled (defined) later

        %Hand matrices
        PHI_rD_hand=zeros(6,1,kinrows); %kinrows=#data points in kinematic
file
        M_hand=zeros(6,1,kinrows);
        Mg_hand=zeros(6,1,kinrows);
        omega_hand=zeros(6,6,kinrows);
        T_hand=zeros(3,3,kinrows);
        Ip_hand=zeros(3,3,kinrows);
        I_hand=zeros(6,6,kinrows);
        w_hand=zeros(6,1,kinrows);
        omegaIw_hand=zeros(6,1,kinrows);
        a_hand=zeros(6,1,kinrows);
        Ia_hand=zeros(6,1,kinrows);
        rP_hand=zeros(6,1,kinrows);

        %Forearm matrices
        PHI_rD_fa=zeros(6,1,kinrows); %kinrows=#data points in kinematic file
        M_fa=zeros(6,1,kinrows);
        Mg_fa=zeros(6,1,kinrows);
        omega_fa=zeros(6,6,kinrows);
        T_fa=zeros(3,3,kinrows);
        Ip_fa=zeros(3,3,kinrows);
        I_fa=zeros(6,6,kinrows);
```

```
        w_fa=zeros(6,1,kinrows);
        omegaIw_fa=zeros(6,1,kinrows);
        a_fa=zeros(6,1,kinrows);
        Ia_fa=zeros(6,1,kinrows);
        rP_fa=zeros(6,1,kinrows);

        %Upper arm matrices
        PHI_rD_ua=zeros(6,1,kinrows); %kinrows=#data points in kinematic file
        M_ua=zeros(6,1,kinrows);
        Mg_ua=zeros(6,1,kinrows);
        omega_ua=zeros(6,6,kinrows);
        T_ua=zeros(3,3,kinrows);
        Ip_ua=zeros(3,3,kinrows);
        I_ua=zeros(6,6,kinrows);
        w_ua=zeros(6,1,kinrows);
        omegaIw_ua=zeros(6,1,kinrows);
        a_ua=zeros(6,1,kinrows);
        Ia_ua=zeros(6,1,kinrows);
        rP_ua=zeros(6,1,kinrows);

        %Phi Matrix (distances between proximal and distal landmarks with -1
on diagonals) EQN. 20
        PHI_hand=zeros(6,6,kinrows);
        PHI_fa=zeros(6,6,kinrows);
        PHI_ua=zeros(6,6,kinrows);
        for i=1:6
            PHI_hand(i,i,1:kinrows)=-1; %put -1 along diagonal
            PHI_fa(i,i,1:kinrows)=-1; %put -1 along diagonal
            PHI_ua(i,i,1:kinrows)=-1; %put -1 along diagonal
        end

        %Hand segment
        rD_hand=zeros(kinrows,6);
        %Assume hand has a point contact with the pushrim at the third mp
        %Therefore SW forces are input to the third mp, but there is no moment
arm between the pushrim and the thirdmp, so the input moments are zero
        for t=1:kinrows
            %if step(t,1) > 0, %will only input SW forces when hand is on the
rim, determined by step function
            rD_hand(t,1:3)=-(forces(t,1:3)); %reaction forces at hand are the
negative of the forces applied to the pushrim
%             plot(forces(:,2))
            %end %
        end
%         figure
%         plot(rD_hand(:,2))
        rD_hand=rD_hand';

        for t=1:kinrows
            %fill in Phi_hand matrix with distances between third mp and wrist
center
            %Signs in PHI matrix are different from Cooper et al. because his
            %paper assumes distances rather than directional vectors
            PHI_hand(4,2,t)=-(thirdmp(t,3)-wristcen(t,3)); %negative of vector
from prox to dist. in z direction EQN.20 (-Zdp)
            PHI_hand(5,1,t)=(thirdmp(t,3)-wristcen(t,3)); %vector from prox to
dist. in z direction EQN.20 (Zdp)
            PHI_hand(4,3,t)=-((thirdmp(t,2)-wristcen(t,2))); %negative of
vector from prox to dist. in y direction EQN.20 (-Ydp)
            PHI_hand(6,1,t)=((thirdmp(t,2)-wristcen(t,2))); %vector from prox
to dist. in y direction EQN.20 (Ydp)
            PHI_hand(6,2,t)=-(thirdmp(t,1)-wristcen(t,1)); %negative of vector
from prox to dist. in x direction EQN.20 (-Xdp)
            PHI_hand(5,3,t)=(thirdmp(t,1)-wristcen(t,1)); %vector from prox to
```

```
dist. in x direction EQN.20 (Xdp)


            %EQN. 21 PHI matrix times the reaction forces and moments at the
distal end of the segment
            PHI_rD_hand(:,:,t)=PHI_hand(:,:,t)*rD_hand(1:6,t);

            %EQN. 20 Define M matrix for hand (mass and moment arm vector)
            M_hand(2,1,t)=handmass;
            M_hand(4,1,t)=handmass*-1*(handcm(t,3)-wristcen(t,3));%hand mass
times distance in z direction b/w wrist center and hand center of mass
            %negative corrects for direction of moment
            M_hand(6,1,t)=handmass*(handcm(t,1)-wristcen(t,1));%hand mass
times distance in x direction b/w wrist center and hand center of mass

            %EQN. 21 Calculate M*g matrix
            Mg_hand(:,1,t)=M_hand(:,1,t)*g; %M matrix times gravity

            %EQN. 20 Calculate Capital Omega matrix
            omega_hand(4,5,t)=-(velocities(t,9)); %negative angular velocity @
z axis
            omega_hand(5,4,t)=(velocities(t,9)); %angular velocity @ z axis
            omega_hand(4,6,t)=(velocities(t,8)); %angular velocity @ y axis
            omega_hand(6,4,t)=-(velocities(t,8)); %negative angular velocity @
y axis
            omega_hand(5,6,t)=-(velocities(t,7)); %negative angular velocity @
x axis
            omega_hand(6,5,t)=(velocities(t,7)); %angular velocity @ x axis

            %EQN.18 Set up transformation matrix to convert inertias about
            %segment axes to inertias about global x,y,z axes
            %angles(7)=psi_hand; angles(8)=theta_hand; angles(9)=phi_hand
            T_hand(1,1,t)=cos(angles(t,9))*cos(angles(t,8));
            T_hand(1,2,t)=sin(angles(t,9))*cos(angles(t,8));
            T_hand(1,3,t)=-sin(angles(t,8));
            T_hand(2,1,t)=-
sin(angles(t,9))*cos(angles(t,7))+cos(angles(t,9))*sin(angles(t,8))*sin(angles
(t,7));

T_hand(2,2,t)=cos(angles(t,9))*cos(angles(t,7))+sin(angles(t,9))*sin(angles(t,
8))*sin(angles(t,7));
            T_hand(2,3,t)=cos(angles(t,8))*sin(angles(t,7));

T_hand(3,1,t)=sin(angles(t,9))*sin(angles(t,7))+cos(angles(t,9))*sin(angles(t,
8))*cos(angles(t,7));
            T_hand(3,2,t)=-
cos(angles(t,9))*sin(angles(t,7))+cos(angles(t,7))*sin(angles(t,8))*cos(angles
(t,7));
            T_hand(3,3,t)=cos(angles(t,8))*cos(angles(t,7));

            %EQN.18 Calculate inertias about global x,y,z
            Ip_hand(:,:,t)=T_hand(:,:,t)*handI*T_hand(:,:,t)';

            %All inertia characteristics of the hand (angular velocity and
            %acceleration) will not be included in the calculated because they
            %have a very small contribution and are susceptible to noise)

            %EQN. 20 Set up angular velocity vector(lowercase omega-- will
call "w")

%w_hand(:,:,t)=[0;0;0;velocities(t,7);velocities(t,8);velocities(t,9)];
```

```
            %EQN. 21 Calculate product of angular velocity matrices
(omega*I*w)

%omegaIw_hand(:,:,t)=omega_hand(:,:,t)*I_hand(:,:,t)*w_hand(:,:,t);

            %EQN. 20 Define acceleration vector(linear [of center of mass] and
angular accelerations)

%a_hand(:,:,t)=[cmaccel(t,1);cmaccel(t,2);cmaccel(t,3);accelerations(t,7);acce
lerations(t,8);accelerations(t,9)];

            %EQN. 21 Calculate matrix that combines inertial properties and
linear accelerations
            %Ia_hand(:,:,t)=I_hand(:,:,t)*a_hand(:,:,t);

            %EQN. 21 Calculate reaction force at wrist center in global
coordinate system
            rP_hand(:,:,t)=PHI_rD_hand(:,:,t)+Mg_hand(:,:,t);

            %         plot3(rP_hand(1:3,1,t)) %checking the forces




            if n==1
                fxr_hand(t,1)= rD_hand(1,t);
                fyr_hand(t,1)= rD_hand(2,t);
                fzr_hand(t,1)= rD_hand(3,t);


resultant_force_R3mp(t,1)=sqrt(rD_hand(1,t)^2+rD_hand(2,t)^2+rD_hand(3,t)^2);

            else
                fxl_hand(t,1)=rD_hand(1,t);
                fyl_hand(t,1)=rD_hand(2,t);
                fzl_hand(t,1)=rD_hand(3,t);


resultant_force_L3mp(t,1)=sqrt(rD_hand(1,t)^2+rD_hand(2,t)^2+rD_hand(3,t)^2);

                %
            end


            % previously  use for checking results
            %left in for future troubleshooting
            %         if n==1
            %             fxr_wrist(1,t)=-rP_hand(1,1,t);
            %             fyr_wrist(1,t)=-rP_hand(2,1,t);
            %             fzr_wrist(1,t)=-rP_hand(3,1,t);
            %             mxr_wrist(1,t)=-rP_hand(4,1,t);
            %             myr_wrist(1,t)=-rP_hand(5,1,t);
            %             mzr_wrist(1,t)=-rP_hand(6,1,t);
            %
resultant_force_wrist(1,t)=sqrt(rP_hand(1,1,t)^2+rP_hand(2,1,t)^2+rP_hand(3,1,
t)^2);
            %
resultant_force_3mp(1,t)=sqrt(rD_hand(1,t)^2+rD_hand(2,t)^2+rD_hand(3,t)^2);
            %
resultant_moment_wrist(1,t)=sqrt(rP_hand(4,1,t)^2+rP_hand(5,1,t)^2+rP_hand(6,1
,t)^2);
            %         else
            %             fxl_wrist(1,t)=-rP_hand(1,1,t);
```

```
            %               fyl_wrist(1,t)=-rP_hand(2,1,t);
            %               fzl_wrist(1,t)=-rP_hand(3,1,t);
            %               mxl_wrist(1,t)=-rP_hand(4,1,t);
            %               myl_wrist(1,t)=-rP_hand(5,1,t);
            %               mzl_wrist(1,t)=-rP_hand(6,1,t);
            %           end
        end

        %Forearm segment
        rD_fa=-rP_hand; %reaction forces at hand are the negative of the
forces applied to the wrist (negative applied in PHI matrix below)

        for t=1:kinrows
            %fill in Phi_fa matrix with distances between wrist center and
lateral epicondyle
            %Signs in PHI matrix are different from Cooper et al. because his
            %paper assumes distances rather than directional vectors
            PHI_fa(4,2,t)=-(wristcen(t,3)-latep(t,3)); %negative of vector
from prox to dist. in z direction EQN.20 (-Zdp)
            PHI_fa(5,1,t)=(wristcen(t,3)-latep(t,3)); %vector from prox to
dist. in z direction EQN.20 (Zdp)
            PHI_fa(4,3,t)=-((wristcen(t,2)-latep(t,2))); %negative of vector
from prox to dist. in y direction EQN.20 (-Ydp)
            PHI_fa(6,1,t)=((wristcen(t,2)-latep(t,2))); %vector from prox to
dist. in y direction EQN.20 (Ydp)
            PHI_fa(6,2,t)=-(wristcen(t,1)-latep(t,1)); %negative of vector
from prox to dist. in x direction EQN.20 (-Xdp)
            PHI_fa(5,3,t)=(wristcen(t,1)-latep(t,1)); %vector from prox to
dist. in x direction EQN.20 (Xdp)

            %EQN. 21 PHI matrix times the reaction forces and moments at the
distal end of the segment
            PHI_rD_fa(:,:,t)=PHI_fa(:,:,t)*rD_fa(1:6,t);

            %EQN. 20 Define M matrix for forearm (mass and moment arm vector)
            M_fa(2,1,t)=famass;
            M_fa(4,1,t)=famass*-1*(facm(t,3)-latep(t,3));%forearm mass times
distance in z direction b/w latep and forearm center of mass
            %negative corrects for direction of moment
            M_fa(6,1,t)=famass*(facm(t,1)-latep(t,1));%forearm mass times
distance in x direction b/w latep and forearm center of mass

            %EQN. 21 Calculate M*g matrix
            Mg_fa(:,1,t)=M_fa(:,1,t)*g; %M matrix times gravity

            %EQN. 20 Calculate Capital Omega matrix
            omega_fa(4,5,t)=-(velocities(t,6)); %negative angular velocity @ z
axis
            omega_fa(5,4,t)=(velocities(t,6)); %angular velocity @ z axis
            omega_fa(4,6,t)=(velocities(t,5)); %angular velocity @ y axis
            omega_fa(6,4,t)=-(velocities(t,5)); %negative angular velocity @ y
axis
            omega_fa(5,6,t)=-(velocities(t,4)); %negative angular velocity @ x
axis
            omega_fa(6,5,t)=(velocities(t,4)); %angular velocity @ x axis

            %EQN.18 Set up transformation matrix to convert inertias about
            %segment axes to inertias about global x,y,z axes
            %angles(4)=psi_fa; angles(5)=theta_fa; angles(6)=phi_fa
            T_fa(1,1,t)=cos(angles(t,6))*cos(angles(t,5));
            T_fa(1,2,t)=sin(angles(t,6))*cos(angles(t,5));
            T_fa(1,3,t)=-sin(angles(t,5));
            T_fa(2,1,t)=-
```

```
sin(angles(t,6))*cos(angles(t,4))+cos(angles(t,6))*sin(angles(t,5))*sin(angles
(t,4));

T_fa(2,2,t)=cos(angles(t,6))*cos(angles(t,4))+sin(angles(t,6))*sin(angles(t,5)
)*sin(angles(t,4));
            T_fa(2,3,t)=cos(angles(t,5))*sin(angles(t,4));

T_fa(3,1,t)=sin(angles(t,6))*sin(angles(t,4))+cos(angles(t,6))*sin(angles(t,5)
)*cos(angles(t,4));
            T_fa(3,2,t)=-
cos(angles(t,6))*sin(angles(t,4))+cos(angles(t,4))*sin(angles(t,5))*cos(angles
(t,4));
            T_fa(3,3,t)=cos(angles(t,5))*cos(angles(t,4));

            %EQN.18 Calculate inertias about global x,y,z
            Ip_fa(:,:,t)=T_fa(:,:,t)*faI*T_fa(:,:,t)';

            %EQN.20 Set up I matrix that contains mass and inertia information
            I_fa(1,1,t)=famass;
            I_fa(2,2,t)=famass;
            I_fa(3,3,t)=famass;
            I_fa(4:6,4:6,t)=Ip_fa(:,:,t);

            %EQN. 20 Set up angular velocity vector(lowercase omega-- will
call "w")

w_fa(:,:,t)=[0;0;0;velocities(t,4);velocities(t,5);velocities(t,6)];

            %EQN. 21 Calculate product of angular velocity matrices
(omega*I*w)
            omegaIw_fa(:,:,t)=omega_fa(:,:,t)*I_fa(:,:,t)*w_fa(:,:,t);

            %EQN. 20 Define acceleration vector(linear [of center of mass] and
angular accelerations)

%a_fa(:,:,t)=[cmaccel(t,4);cmaccel(t,5);cmaccel(t,6);accelerations(t,4);accele
rations(t,5);accelerations(t,6)];

a_fa(:,:,t)=[cmaccel(t,4);cmaccel(t,5);cmaccel(t,6);0;0;accelerations(t,6)];
            %xz and yz plane angular accelerations ignored because they are
            %prone to quadrant changes when the arm is vertical.
contributions
            %are negligable in these two planes

            %EQN. 21 Calculate matrix that combines inertial properties and
linear accelerations
            Ia_fa(:,:,t)=I_fa(:,:,t)*a_fa(:,:,t);

            %EQN. 21 Calculate reaction force at elbow center in global
coordinate system
            rP_fa(:,:,t)=PHI_rD_fa(:,:,t)-Ia_fa(:,:,t)-
omegaIw_fa(:,:,t)+Mg_fa(:,:,t);

            %         plot3(rP_fa(1:3,1,t)) %checking the forces

            % previously  use for checking results
            %left in for future troubleshooting
            %         if n==1
            %             fxr_elbow(1,t)=-rP_fa(1,1,t);
            %             fyr_elbow(1,t)=-rP_fa(2,1,t);
            %             fzr_elbow(1,t)=-rP_fa(3,1,t);
            %             mxr_elbow(1,t)=-rP_fa(4,1,t);
            %             myr_elbow(1,t)=-rP_fa(5,1,t);
```

```
            %              mzr_elbow(1,t)=-rP_fa(6,1,t);
            %
resultant_force_elbow(1,t)=sqrt(rP_fa(1,1,t)^2+rP_fa(2,1,t)^2+rP_fa(3,1,t)^2);
            %
resultant_moment_elbow(1,t)=sqrt(rP_fa(4,1,t)^2+rP_fa(5,1,t)^2+rP_fa(6,1,t)^2)
;
            %          else
            %              fxl_elbow(1,t)=-rP_fa(1,1,t);
            %              fyl_elbow(1,t)=-rP_fa(2,1,t);
            %              fzl_elbow(1,t)=-rP_fa(3,1,t);
            %              mxl_elbow(1,t)=-rP_fa(4,1,t);
            %              myl_elbow(1,t)=-rP_fa(5,1,t);
            %              mzl_elbow(1,t)=-rP_fa(6,1,t);
            %          end

        end

        %Upper arm segment
        rD_ua=-rP_fa; %reaction forces at shoulder are the negative of the
forces applied to the elbow (negative applied in PHI matrix below)

        for t=1:kinrows
            %fill in Phi_ua matrix with distances between lateral epicondyle
and acromion
            %Signs in PHI matrix are different from Cooper et al. because his
            %paper assumes distances rather than directional vectors
            PHI_ua(4,2,t)=-(latep(t,3)-acro(t,3)); %negative of vector from
prox to dist. in z direction EQN.20 (-Zdp)
            PHI_ua(5,1,t)=(latep(t,3)-acro(t,3)); %vector from prox to dist.
in z direction EQN.20 (Zdp)
            PHI_ua(4,3,t)=-((latep(t,2)-acro(t,2))); %negative of vector from
prox to dist. in y direction EQN.20 (-Ydp)
            PHI_ua(6,1,t)=((latep(t,2)-acro(t,2))); %vector from prox to dist.
in y direction EQN.20 (Ydp)
            PHI_ua(6,2,t)=-(latep(t,1)-acro(t,1)); %negative of vector from
prox to dist. in x direction EQN.20 (-Xdp)
            PHI_ua(5,3,t)=(latep(t,1)-acro(t,1)); %vector from prox to dist.
in x direction EQN.20 (Xdp)

            %EQN. 21 PHI matrix times the reaction forces and moments at the
distal end of the segment
            PHI_rD_ua(:,:,t)=PHI_ua(:,:,t)*rD_ua(1:6,t);

            %EQN. 20 Define M matrix for upperarm (mass and moment arm vector)
            M_ua(2,1,t)=uamass;
            M_ua(4,1,t)=uamass*(1-uacmratio)*PHI_ua(4,2,t);
            M_ua(6,1,t)=uamass*(1-uacmratio)*PHI_ua(6,2,t);

            M_ua(2,1,t)=uamass;
            M_ua(4,1,t)=uamass*-1*(uacm(t,3)-acro(t,3));%upperarm mass times
distance in z direction b/w acromion and upperam center of mass
            %negative corrects for direction of moment
            M_ua(6,1,t)=uamass*(uacm(t,1)-acro(t,1));%upperarm mass times
distance in x direction b/w acromion and upperam center of mass


            %EQN. 21 Calculate M*g matrix
            Mg_ua(:,1,t)=M_ua(:,1,t)*g; %M matrix times gravity

            %EQN. 20 Calculate Capital Omega matrix
            omega_ua(4,5,t)=-(velocities(t,3)); %negative angular velocity @ z
axis
            omega_ua(5,4,t)=(velocities(t,3)); %angular velocity @ z axis
```

```
            omega_ua(4,6,t)=(velocities(t,2)); %angular velocity @ y axis
            omega_ua(6,4,t)=-(velocities(t,2)); %negative angular velocity @ y
axis
            omega_ua(5,6,t)=-(velocities(t,1)); %negative angular velocity @ x
axis
            omega_ua(6,5,t)=(velocities(t,1)); %angular velocity @ x axis

            %EQN.18 Set up transformation matrix to convert inertias about
            %segment axes to inertias about global x,y,z axes
            %angles(1)=psi_ua; angles(2)=theta_ua; angles(3)=phi_ua
            T_ua(1,1,t)=cos(angles(t,3))*cos(angles(t,2));
            T_ua(1,2,t)=sin(angles(t,3))*cos(angles(t,2));
            T_ua(1,3,t)=-sin(angles(t,2));
            T_ua(2,1,t)=-
sin(angles(t,3))*cos(angles(t,1))+cos(angles(t,3))*sin(angles(t,2))*sin(angles
(t,1));

T_ua(2,2,t)=cos(angles(t,3))*cos(angles(t,1))+sin(angles(t,3))*sin(angles(t,2)
)*sin(angles(t,1));
            T_ua(2,3,t)=cos(angles(t,2))*sin(angles(t,1));

T_ua(3,1,t)=sin(angles(t,3))*sin(angles(t,1))+cos(angles(t,3))*sin(angles(t,2)
)*cos(angles(t,1));
            T_ua(3,2,t)=-
cos(angles(t,3))*sin(angles(t,1))+cos(angles(t,1))*sin(angles(t,1))*cos(angles
(t,1));
            T_ua(3,3,t)=cos(angles(t,2))*cos(angles(t,1));

            %EQN.18 Calculate inertias about global x,y,z
            Ip_ua(:,:,t)=T_ua(:,:,t)*uaI*T_ua(:,:,t)';

            %EQN.20 Set up I matrix that contains mass and inertia information
            I_ua(1,1,t)=uamass;
            I_ua(2,2,t)=uamass;
            I_ua(3,3,t)=uamass;
            I_ua(4:6,4:6,t)=Ip_ua(:,:,t);

            %EQN. 20 Set up angular velocity vector(lowercase omega-- will
call "w")

w_ua(:,:,t)=[0;0;0;velocities(t,1);velocities(t,2);velocities(t,3)];

            %EQN. 21 Calculate product of angular velocity matrices
(omega*I*w)
            omegaIw_ua(:,:,t)=omega_ua(:,:,t)*I_ua(:,:,t)*w_ua(:,:,t);

            %EQN. 20 Define acceleration vector(linear [of center of mass] and
angular accelerations)

%a_ua(:,:,t)=[cmaccel(t,7);cmaccel(t,8);cmaccel(t,9);accelerations(t,1);accele
rations(t,2);accelerations(t,3)];

a_ua(:,:,t)=[cmaccel(t,7);cmaccel(t,8);cmaccel(t,9);0;0;accelerations(t,3)];
            %xz and yz plane angular accelerations ignored because they are
            %prone to quadrant changes when the arm is vertical.
contributions
            %are negligable in these two planes

            %EQN. 21 Calculate matrix that combines inertial properties and
linear accelerations
            Ia_ua(:,:,t)=I_ua(:,:,t)*a_ua(:,:,t);

            %EQN. 21 Calculate reaction force at shoulder center in global
```

```
coordinate system
            rP_ua(:,:,t)=PHI_rD_ua(:,:,t)-Ia_ua(:,:,t)-
omegaIw_ua(:,:,t)+Mg_ua(:,:,t);

            %         plot3(rP_ua(1:3,1,t)) %checking the forces

            % previously  use for checking results
            %left in for future troubleshooting
            %         if n==1
            %             fxr_shoulder(1,t)=-rP_ua(1,1,t);
            %             fyr_shoulder(1,t)=-rP_ua(2,1,t);
            %             fzr_shoulder(1,t)=-rP_ua(3,1,t);
            %             mxr_shoulder(1,t)=-rP_ua(4,1,t);
            %             myr_shoulder(1,t)=-rP_ua(5,1,t);
            %             mzr_shoulder(1,t)=-rP_ua(6,1,t);
            %
resultant_force_shoulder(1,t)=sqrt(rP_ua(1,1,t)^2+rP_ua(2,1,t)^2+rP_ua(3,1,t)^
2);
            %
resultant_moment_shoulder(1,t)=sqrt(rP_ua(4,1,t)^2+rP_ua(5,1,t)^2+rP_ua(6,1,t)
^2);
            %         else
            %             fxl_shoulder(1,t)=-rP_ua(1,1,t);
            %             fyl_shoulder(1,t)=-rP_ua(2,1,t);
            %             fzl_shoulder(1,t)=-rP_ua(3,1,t);
            %             mxl_shoulder(1,t)=-rP_ua(4,1,t);
            %             myl_shoulder(1,t)=-rP_ua(5,1,t);
            %             mzl_shoulder(1,t)=-rP_ua(6,1,t);
            %         end
            %
            %         %static check of shoulder Fy forces
            %         check_sho_fy(t,1)=-
forces(t,2)+massall(1,1)*9.8+massall(1,2)*9.8+massall(1,3)*9.8;


            %----------------------- Calculate Local Coordinate Systems for
Segments--------------------%

            %------------------------Hand local coordinate system------------
---------%

            %temporary k axis of hand (use to calculate i)
            if n==1 %vector points to right for both sides in standard
anatomical position
                v1_hand(t,1:3)=radsty(t,1:3)-ulnsty(t,1:3); %vector 1, not
normalized
                k_hand_temp(t,1:3)= v1_hand(t,1:3)/norm(v1_hand(t,1:3));
%normalized vector 1 (temporary k vector)
            else %vector points to right for both sides in standard anatomical
position
                v1_hand(t,1:3)=ulnsty(t,1:3)-radsty(t,1:3); %vector 1, not
normalized
                k_hand_temp(t,1:3)= v1_hand(t,1:3)/norm(v1_hand(t,1:3));
%normalized vector 1 (temporary k vector)
            end

            %j axis of the hand
            v2_hand(t,1:3)=wristcen(t,1:3)-thirdmp(t,1:3); %vector 2, not
normalized
            j_hand(t,1:3)= v2_hand(t,1:3)/norm(v2_hand(t,1:3)); %normalized
vector 2 (j vector)

            %i axis of the hand
```

```
            v3_hand(t,1:3)=cross(j_hand(t,1:3),k_hand_temp(t,1:3));%vector 3,
not normalized
            i_hand(t,1:3)=v3_hand(t,1:3)/norm(v3_hand(t,1:3)); %normalized
vector 2 (k vector)

            %k axis of the hand
            v4_hand(t,1:3)=cross(i_hand(t,1:3),j_hand(t,1:3));%vector 4, not
normalized
            k_hand(t,1:3)=v4_hand(t,1:3)/norm(v4_hand(t,1:3)); %normalized
vector 2 (i vector)

            %rotation matrix for hand
            rot_hand(1,1:3,t)=i_hand(t,1:3); %first row is i unit vector
            rot_hand(2,1:3,t)=j_hand(t,1:3); %second row is j unit vector
            rot_hand(3,1:3,t)=k_hand(t,1:3); %third row is k unit vector

            %------------------Forearm local coordinate system-----------------
--------%

            %temporary k axis of forearm (use to calculate i)
            if n==1 %vector points to right for both sides in standard
anatomical position
                v1_fa(t,1:3)=radsty(t,1:3)-ulnsty(t,1:3); %vector 1, not
normalized
                k_fa_temp(t,1:3)= v1_fa(t,1:3)/norm(v1_fa(t,1:3)); %normalized
vector 1 (temporary k vector)
            else %vector points to right for both sides in standard anatomical
position
                v1_fa(t,1:3)=ulnsty(t,1:3)-radsty(t,1:3); %vector 1, not
normalized
                k_fa_temp(t,1:3)= v1_fa(t,1:3)/norm(v1_fa(t,1:3)); %normalized
vector 1 (temporary k vector)
            end

            %j axis of the forearm
            v2_fa(t,1:3)=latep(t,1:3)-ulnsty(t,1:3); %vector 2, not normalized
            j_fa(t,1:3)= v2_fa(t,1:3)/norm(v2_fa(t,1:3)); %normalized vector 2
(j vector)

            %i axis of the forearm
            v3_fa(t,1:3)=cross(j_fa(t,1:3),k_fa_temp(t,1:3));%vector 3, not
normalized
            i_fa(t,1:3)=v3_fa(t,1:3)/norm(v3_fa(t,1:3)); %normalized vector 2
(i vector)

            %k axis of the forearm
            v4_fa(t,1:3)=cross(i_fa(t,1:3),j_fa(t,1:3));%vector 4, not
normalized
            k_fa(t,1:3)=v4_fa(t,1:3)/norm(v4_fa(t,1:3)); %normalized vector 2
(k vector)

            %rotation matrix for forearm
            rot_fa(1,1:3,t)=i_fa(t,1:3); %first row is i unit vector
            rot_fa(2,1:3,t)=j_fa(t,1:3); %second row is j unit vector
            rot_fa(3,1:3,t)=k_fa(t,1:3); %third row is k unit vector


            %-------------------Humerus local coordinate system---------------
-------%
            %Reference is Cooper et al. Glenohumeral Joint Kinematics and
Kinetics.....Am J Phys Med Rehab 1999.
            %EQN. 1-2,5
```

```
            %temporary i axis of upper arm (use to calculate k)
            v1_ua(t,1:3)=ulnsty(t,1:3)-latep(t,1:3); %vector 1, not normalized
            i_ua_temp(t,1:3)= v1_ua(t,1:3)/norm(v1_ua(t,1:3)); %normalized
vector 1 (temporary i vector)

            %j axis of the upper arm (called j_s in cooper's paper)
            v2_ua(t,1:3)=acro(t,1:3)-latep(t,1:3); %vector 2, not normalized
            j_ua(t,1:3)= v2_ua(t,1:3)/norm(v2_ua(t,1:3)); %normalized vector 2
(j vector)

            %k axis of the upper arm (called k_s in cooper's paper)
            v3_ua(t,1:3)=cross(i_ua_temp(t,1:3),j_ua(t,1:3));%vector 3, not
normalized
            k_ua(t,1:3)=v3_ua(t,1:3)/norm(v3_ua(t,1:3)); %normalized vector 2
(k vector)

            %i axis of the upper arm (called i_s in cooper's paper)
            v4_ua(t,1:3)=cross(j_ua(t,1:3),k_ua(t,1:3));%vector 4, not
normalized
            i_ua(t,1:3)=v4_ua(t,1:3)/norm(v4_ua(t,1:3)); %normalized vector 2
(i vector)

            %rotation matrix for upper arm
            rot_ua(1,1:3,t)=i_ua(t,1:3); %first row is i unit vector
            rot_ua(2,1:3,t)=j_ua(t,1:3); %second row is j unit vector
            rot_ua(3,1:3,t)=k_ua(t,1:3); %third row is k unit vector

            %-----------Trunk local coordinate system-----------------%
            %Cooper used a triad on the chest to create coordinate system
            %I updated the coordinate system to follow the same convention,
but
            %avoided using the chest triad

            shocen(t,1)=(kin(t,14)+kin(t,41))/2;
            shocen(t,2)=(kin(t,15)+kin(t,42))/2;
            shocen(t,3)=(kin(t,16)+kin(t,43))/2;

            j_Gtrnn(t,1:3)=t3(t,1:3)-t8(t,1:3); %vector j, not normalized
            j_tr(t,1:3)= j_Gtrnn(t,1:3)/norm(j_Gtrnn(t,1:3)); %normalized j
vector

            %intermediate axis of the trunk points anteriorly
            i_Gtrint(t,1:3)=shocen(t,1:3)-t3(t,1:3);

            %k axis of trunk (points to the right in setpo)
            k_Gtrnn(t,1:3)=cross(i_Gtrint(t,1:3),j_tr(t,1:3)); %vector k, not
normalized
            k_tr(t,1:3)= k_Gtrnn(t,1:3)/norm(k_Gtrnn(t,1:3));

            %i axis of the trunk
            i_Gtrnn(t,1:3)=cross(j_tr(t,1:3),k_tr(t,1:3));
            i_tr(t,1:3)=i_Gtrnn(t,1:3)/norm(i_Gtrnn(t,1:3)); %normalized i
vector

            %rotation matrix for trunk
            rot_tr(1,1:3,t)=i_tr(t,1:3); %first row is i unit vector
            rot_tr(2,1:3,t)=j_tr(t,1:3); %second row is j unit vector
            rot_tr(3,1:3,t)=k_tr(t,1:3); %third row is k unit vector

            %          figure
            %          plot(rot_tr(1))
```

```
            %          hold on
            %          plot(rot_tr(2),'r')
            %          hold on
            %          plot(rot_tr(3),'g')

            %----------Calculate reaction forces/moments in anatomical
coordinate systems---------%
            %forces at the wrist
            f_wrist(1:3,1,t)=rot_fa(:,:,t)*-rP_hand(1:3,1,t); %local
forces=T*global forces

            %moments at the wrist
            m_wrist(1:3,1,t)=rot_fa(:,:,t)*-rP_hand(4:6,1,t); %local
moments=T*global moments

            %reformat variables for plotting
            if n==1
                fm_rwrist(t,1)=f_wrist(1,1,t);
                fm_rwrist(t,2)=f_wrist(2,1,t);
                fm_rwrist(t,3)=f_wrist(3,1,t);
                fm_rwrist(t,4)=m_wrist(1,1,t);
                fm_rwrist(t,5)=m_wrist(2,1,t);
                fm_rwrist(t,6)=m_wrist(3,1,t);

                rf_rwrist(t,1)=sqrt(fm_rwrist(t,1)^2+ fm_rwrist(t,2)^2 +
fm_rwrist(t,3)^2);
                rm_rwrist(t,1)=sqrt(fm_rwrist(t,4)^2+ fm_rwrist(t,5)^2 +
fm_rwrist(t,6)^2);
            else
                fm_lwrist(t,1)=f_wrist(1,1,t);
                fm_lwrist(t,2)=f_wrist(2,1,t);
                fm_lwrist(t,3)=f_wrist(3,1,t);
                fm_lwrist(t,4)=m_wrist(1,1,t);
                fm_lwrist(t,5)=m_wrist(2,1,t);
                fm_lwrist(t,6)=m_wrist(3,1,t);
                rf_lwrist(t,1)=sqrt(fm_lwrist(t,1)^2+ fm_lwrist(t,2)^2 +
fm_lwrist(t,3)^2);
                rm_lwrist(t,1)=sqrt(fm_lwrist(t,4)^2+ fm_lwrist(t,5)^2 +
fm_lwrist(t,6)^2);
            end

            %forces at the elbow
            f_elbow(1:3,1,t)=rot_ua(:,:,t)*-rP_fa(1:3,1,t); %local
forces=T*global forces

            %moments at the wrist
            m_elbow(1:3,1,t)=rot_ua(:,:,t)*-rP_fa(4:6,1,t); %local
moments=T*global moments

            %reformat variables for plotting
            if n==1
                fm_relbow(t,1)=f_elbow(1,1,t);
                fm_relbow(t,2)=f_elbow(2,1,t);
                fm_relbow(t,3)=f_elbow(3,1,t);
                fm_relbow(t,4)=m_elbow(1,1,t);
                fm_relbow(t,5)=m_elbow(2,1,t);
                fm_relbow(t,6)=m_elbow(3,1,t);
                rf_relbow(t,1)=sqrt(fm_relbow(t,1)^2+ fm_relbow(t,2)^2 +
fm_relbow(t,3)^2);
                rm_relbow(t,1)=sqrt(fm_relbow(t,4)^2+ fm_relbow(t,5)^2 +
fm_relbow(t,6)^2);
            else
                fm_lelbow(t,1)=f_elbow(1,1,t);
```

```
                fm_lelbow(t,2)=f_elbow(2,1,t);
                fm_lelbow(t,3)=f_elbow(3,1,t);
                fm_lelbow(t,4)=m_elbow(1,1,t);
                fm_lelbow(t,5)=m_elbow(2,1,t);
                fm_lelbow(t,6)=m_elbow(3,1,t);
                rf_lelbow(t,1)=sqrt(fm_lelbow(t,1)^2+ fm_lelbow(t,2)^2 +
fm_lelbow(t,3)^2);
                rm_lelbow(t,1)=sqrt(fm_lelbow(t,4)^2+ fm_lelbow(t,5)^2 +
fm_lelbow(t,6)^2);
            end


            %forces at the shoulder
            %EQN. 27 from Cooper et al.
            f_shoulder(1:3,1,t)=rot_tr(:,:,t)*-rP_ua(1:3,1,t); %local
forces=T*global forces

            %moments at the shoulder
            %EQN. 28 from Cooper et al.
            m_shoulder(1:3,1,t)=rot_tr(:,:,t)*-rP_ua(4:6,1,t); %local
moments=T*global moments

            %reformat variables for plotting
            if n==1
                fm_rsho(t,1)=f_shoulder(1,1,t);
                fm_rsho(t,2)=f_shoulder(2,1,t);
                fm_rsho(t,3)=f_shoulder(3,1,t);
                fm_rsho(t,4)=m_shoulder(1,1,t);
                fm_rsho(t,5)=m_shoulder(2,1,t);
                fm_rsho(t,6)=m_shoulder(3,1,t);
                rf_rsho(t,1)=sqrt(fm_rsho(t,1)^2+ fm_rsho(t,2)^2 +
fm_rsho(t,3)^2);
                rm_rsho(t,1)=sqrt(fm_rsho(t,4)^2+ fm_rsho(t,5)^2 +
fm_rsho(t,6)^2);
                %stepr=step; %save step function
            else
                fm_lsho(t,1)=f_shoulder(1,1,t);
                fm_lsho(t,2)=f_shoulder(2,1,t);
                fm_lsho(t,3)=f_shoulder(3,1,t);
                fm_lsho(t,4)=m_shoulder(1,1,t);
                fm_lsho(t,5)=m_shoulder(2,1,t);
                fm_lsho(t,6)=m_shoulder(3,1,t);
                rf_lsho(t,1)=sqrt(fm_lsho(t,1)^2+ fm_lsho(t,2)^2 +
fm_lsho(t,3)^2);
                rm_lsho(t,1)=sqrt(fm_lsho(t,4)^2+ fm_lsho(t,5)^2 +
fm_lsho(t,6)^2);
                %              if subject_num > 33
                %
                %              rf_feet(t,:)=sqrt(feet_FP(t,1)^2+ feet_FP(t,2)^2
+ feet_FP(t,3)^2);
                %              end
                %stepl=step; %save step function
            end

            %---------------------------------Calculate Euler Angles-----------
-------------------------------%

            %Reference for all Euler Angle calculations is Wu G. et al... ISG
            %recommendation on definitions of joint coordinate systems of
            %various joints (see top for complete citation)


            %-------------Relating trunk position to the global coordinate
```

```
system------------------%

            %need to take transpose of rotation matrices so that LCS axes are
in columns instead of rows
            rot_tr(1:3,1:3,t)=rot_tr(1:3,1:3,t)';

            %Assume a yx'z'' rotation for Euler Angle Calculations
            %This is updated from 1996 standards
            %Code for Euler Angle calculations using 1996 standards was
adapted from rotyxz.m function on ISB webpage
            %alpha is the first rotation (about y)
            %+ alpha = torsion to the left
            %- alpha = torsion to the right
            %beta is the second rotation (about x)
            %+ beta = lateral bending to the right
            %- beta = lateral bending to the left
            %gamma is the third rotation (about z)
            %+ gamma = extension
            %- gamma = flexion


            beta1_tr(t)= asin(-rot_tr(2,3,t)); %calculate x' rotation first

            salpha_tr(t) = rot_tr(1,3,t)/cos(beta1_tr(t)); %sin alpha
            calpha_tr(t) = rot_tr(3,3,t)/cos(beta1_tr(t)); %cos alpha
            alpha1_tr(t) = atan2(salpha_tr(t),calpha_tr(t)); %alpha one

            sgamma_tr(t) = rot_tr(2,1,t)/cos(beta1_tr(t)); %sin gamma
            cgamma_tr(t) = rot_tr(2,2,t)/cos(beta1_tr(t)); %cos gamma
            gamma1_tr(t) = atan2(sgamma_tr(t),cgamma_tr(t)); %gamma one

            if beta1_tr(t)>=0
                beta2_tr(t)=pi-beta1_tr(t); %beta two
            else
                beta2_tr(t)=-pi-beta1_tr(t); %beta two
            end

            salpha2_tr(t) = rot_tr(1,3,t)/cos(beta2_tr(t)); %sin alpha two
            calpha2_tr(t) = rot_tr(3,3,t)/cos(beta2_tr(t)); %cos alpha two
            alpha2_tr(t) = atan2(salpha2_tr(t),calpha2_tr(t)); %alpha two

            sgamma2_tr(t) = rot_tr(2,1,t)/cos(beta2_tr(t)); %sin gamma two
            cgamma2_tr(t) = rot_tr(2,2,t)/cos(beta2_tr(t)); %cos gamma two
            gamma2_tr(t) = atan2(sgamma2_tr(t),cgamma2_tr(t)); %gamma two


            if -pi/2 <= beta1_tr(t) & beta1_tr(t) <= pi/2 %loop sets values of
all angles, based on the x rotation
                %convert to degrees
                alpha_tr(t)=alpha1_tr(t)*(180/pi);
                beta_tr(t)=beta1_tr(t)*(180/pi);
                gamma_tr(t)=gamma1_tr(t)*(180/pi);
            else
                %convert to degrees
                alpha_tr(t)=alpha2_tr(t)*(180/pi);
                beta_tr(t)=beta2_tr(t)*(180/pi);
                gamma_tr(t)=gamma2_tr(t)*(180/pi);
            end


            %------------Relating humeral motion to the trunk (shoulder
angles)--------%
```

```
            %need to take transpose of rotation matrices so that LCS axes are
in columns instead of rows
            %trunk rotation matrix already transposed
            rot_ua(1:3,1:3,t)=rot_ua(1:3,1:3,t)';

            %calculate the inverse of the trunk rotation matrix
            rot_tr_inv(1:3,1:3,t)=(rot_tr(1:3,1:3,t))';

            %find rotation matrix from trunk to humerus
            rot_tr_ua(1:3,1:3,t)=rot_tr_inv(1:3,1:3,t)*rot_ua(1:3,1:3,t);

            %Assume a yx'y'' rotation for Euler Angle Calculations
            %This is updated from 1996 standards
            %Code for Euler Angle calculations using 1996 standards was
adapted from rotyzy.m function on ISB webpage
            %There is no code on the ISB webpage for yxy rotations, so these
changes were made by JLM
            %All output is relative to the defined local coordinate system
            %Before averaging, the sign of either the right or left side will
have to be flipped
            %alpha is the first rotation (about y)
            %Right side
            %+ alpha = plane of elevation in front of horizontal line
connecting the two acromions
            %- alpha = plane of elevation behind horizontal line connecting
the two acromions
            %Left side
            %+ alpha = plane of elevation behind horizontal line connecting
the two acromions
            %- alpha = plane of elevation in front of horizontal line
connecting the two acromions
            %beta is the second rotation (about x)
            %Right side
            %+ beta = negative elevation (or adduction)
            %- beta = positive elevation (or abduction)
            %Left side
            %+ beta = positive elevation (or abduction)
            %- beta = negative elevation (or adduction)
            %gamma is the third rotation (about y)
            %Right side
            %+ gamma = internal rotation
            %- gamma = external rotation
            %Left side
            %+ gamma = external rotation
            %- gamma = internal rotation

            beta1(t)= acos(rot_tr_ua(2,2,t)); %calculate x' rotation first
            if beta1(t)==0 %if there is no x rotation, then the first and
third rotations will be the same
                alpha(t)=acos(rot_tr_ua(1,1,t)); %assign all rotation to be
about the first y axis
                beta(t)=beta1(t); %x rotation is still zero
                gamma(t)=0.0; %assign y'' rotation equal to zero since all
rotation was about y
            end


            salpha(t) = rot_tr_ua(1,2,t)/sin(beta1(t)); %sin alpha
            calpha(t) = rot_tr_ua(3,2,t)/sin(beta1(t)); %cos alpha
            alpha1(t) = atan2(salpha(t),calpha(t)); %alpha one

            sgamma(t) = rot_tr_ua(2,1,t)/sin(beta1(t)); %sin gamma
```

```
            cgamma(t) = -rot_tr_ua(2,3,t)/sin(beta1(t)); %cos gamma
            gamma1(t) = atan2(sgamma(t),cgamma(t)); %gamma one

            beta2(t)=-beta1(t); %beta two

            salpha2(t) = rot_tr_ua(1,2,t)/sin(beta2(t)); %sin alpha two
            calpha2(t) = rot_tr_ua(3,2,t)/sin(beta2(t)); %cos alpha two
            alpha2(t) = atan2(salpha2(t),calpha2(t)); %alpha two

            sgamma2(t) = rot_tr_ua(2,1,t)/sin(beta2(t)); %sin gamma two
            cgamma2(t) = -rot_tr_ua(2,3,t)/sin(beta2(t)); %cos gamma two
            gamma2(t) = atan2(sgamma2(t),cgamma2(t)); %gamma two


            if n==1

                %beta should always be negative on the right side
                if beta1(t) <= 0 & beta1(t) >= -pi %loop sets values of all
angles, based on the x rotation
                    alpha(t)=alpha1(t);
                    beta(t)=beta1(t);
                    gamma(t)=gamma1(t);
                else
                    alpha(t)=alpha2(t);
                    beta(t)=beta2(t);
                    gamma(t)=gamma2(t);
                end

                %convert to degrees
                alpha_rsho=alpha*(180/pi);
                beta_rsho=beta*(180/pi);
                gamma_rsho=gamma*(180/pi);

            else

                %beta should always be positive on rightside
                if beta1(t) >= 0 & beta1(t) <= pi %loop sets values of all
angles, based on the x rotation
                    alpha(t)=alpha1(t);
                    beta(t)=beta1(t);
                    gamma(t)=gamma1(t);
                else
                    alpha(t)=alpha2(t);
                    beta(t)=beta2(t);
                    gamma(t)=gamma2(t);
                end

                %convert to degrees
                alpha_lsho=alpha*(180/pi);
                beta_lsho=beta*(180/pi);
                gamma_lsho=gamma*(180/pi);
            end

            %------------Relating forearm motion to the upper arm (elbow
angles)--------%

            %need to take transpose of rotation matrices so that LCS axes are
in columns instead of rows
            %rot_ua already transposed
            rot_fa(1:3,1:3,t)=rot_fa(1:3,1:3,t)';

            %calculate the inverse of the upper arm rotation matrix
```

```
            rot_ua_inv(1:3,1:3,t)=(rot_ua(1:3,1:3,t))';

            %find rotation matrix from upper arm to forearm on the
            rot_ua_fa(1:3,1:3,t)=rot_ua_inv(1:3,1:3,t)*rot_fa(1:3,1:3,t);


            rot_tr_fa(1:3,1:3,t)=
rot_tr_inv(1:3,1:3,t)*(rot_ua_inv(1:3,1:3,t)*rot_fa(1:3,1:3,t));


            %Assume a zx'y'' rotation for Euler Angle Calculations
            %This is updated from 1996 standards
            %Code for Euler Angle calculations uwas adapted from rotzxy.m
function on ISB webpage
            %All output is relative to the defined local coordinate system
            %Before averaging, the sign of either the right or left side will
have to be flipped
            %alpha is the first rotation (about z)
            %Right and left side are the same
            %+ alpha = flexion
            %- alpha = extension
            %beta is the second rotation (about x) (usually ~=0)
            %Right side
            %+ beta = adduction
            %- beta = abduction
            %Left side
            %+ beta = abduction
            %- beta = adduction
            %gamma is the third rotation (about y)
            %Right side
            %+ gamma = internal rotation
            %- gamma = external rotation
            %Left side
            %+ gamma = external rotation
            %- gamma = internal rotation

            beta1_elb(t)= asin(rot_ua_fa(3,2,t)); %calculate x' rotation first

            sgamma_elb(t) = -rot_ua_fa(3,1,t)/cos(beta1_elb(t)); %sin gamma
            cgamma_elb(t) = rot_ua_fa(3,3,t)/cos(beta1_elb(t)); %cos gamma
            gamma1_elb(t) = atan2(sgamma_elb(t),cgamma_elb(t)); %gamma one

            salpha_elb(t) = -rot_ua_fa(1,2,t)/cos(beta1_elb(t)); %sin alpha
            calpha_elb(t) = rot_ua_fa(2,2,t)/cos(beta1_elb(t)); %cos alpha
            alpha1_elb(t) = atan2(salpha_elb(t),calpha_elb(t)); %alpha one

            if beta1_elb(t)>=0
                beta2_elb(t) = pi-beta1_elb(t);
            else
                beta2_elb(t)= -pi-beta1_elb(t);
            end

            %the next 2 if loops check to see if beta is unstable at 180/-180
            %degrees and sets beta to zero if it is unstable
            %the elbow should have very little ROM about the x axis
            if beta2_elb(t) == pi
                beta2_elb(t)=0;
            end
            if beta2_elb(t)== -pi
                beta2_elb(t)=0;
            end
```

```
            sgamma2_elb(t) = -rot_ua_fa(3,1,t)/cos(beta2_elb(t)); %sin gamma
two
            cgamma2_elb(t) = rot_ua_fa(3,3,t)/cos(beta2_elb(t)); %cos gamma
two
            gamma2_elb(t) = atan2(sgamma2_elb(t),cgamma2_elb(t)); %gamma two

            salpha2_elb(t) = -rot_ua_fa(1,2,t)/cos(beta2_elb(t)); %sin alpha
two
            calpha2_elb(t) = rot_ua_fa(2,2,t)/cos(beta2_elb(t)); %cos alpha
two
            alpha2_elb(t) = atan2(salpha2_elb(t),calpha2_elb(t)); %alpha two


            if n==1

                if -pi/2 <= beta1_elb(t) & beta1_elb(t) <= pi/2 %loop sets
values of all angles, based on the x rotation
                    alpha_elb(t)=alpha1_elb(t);
                    beta_elb(t)=beta1_elb(t);
                    gamma_elb(t)=gamma1_elb(t);
                else
                    alpha_elb(t)=alpha2_elb(t);
                    beta_elb(t)=beta2_elb(t);
                    gamma_elb(t)=gamma2_elb(t);
                end

                if gamma_elb(t)<0 %correct for switch at 180 degrees due to
atan2
                    gamma_elb(t)=gamma_elb(t)+2*pi;
                end

                %convert to degrees
                alpha_relb=alpha_elb*(180/pi);
                beta_relb=beta_elb*(180/pi);
                gamma_relb=gamma_elb*(180/pi);


            else

                if -pi/2 <= beta1_elb(t) & beta1_elb(t) <= pi/2 %loop sets
values of all angles, based on the x rotation
                    alpha_elb(t)=alpha1_elb(t);
                    beta_elb(t)=beta1_elb(t);
                    gamma_elb(t)=gamma1_elb(t);
                else
                    alpha_elb(t)=alpha2_elb(t);
                    beta_elb(t)=beta2_elb(t);
                    gamma_elb(t)=gamma2_elb(t);
                end

                if gamma_elb(t)>0 %correct for switch at -180 degrees due to
atan2
                    gamma_elb(t)=gamma_elb(t)-2*pi;
                end

                %convert to degrees
                alpha_lelb=alpha_elb*(180/pi);
                beta_lelb=beta_elb*(180/pi);
                gamma_lelb=gamma_elb*(180/pi);
            end

            %------------------Relating hand motion to the forearm (wrist
```

```
angles)--------%

            %need to take transpose of rotation matrices so that LCS axes are
in columns instead of rows
            %rot_fa already transposed
            rot_hand(1:3,1:3,t)=rot_hand(1:3,1:3,t)';

            %calculate the inverse of the forearm rotation matrix
            rot_fa_inv(1:3,1:3,t)=(rot_fa(1:3,1:3,t))';

            %find rotation matrix from upper arm to forearm on the
            rot_fa_hand(1:3,1:3,t)=rot_fa_inv(1:3,1:3,t)*rot_hand(1:3,1:3,t);


rot_tr_hand(1:3,1:3,t)=rot_tr_inv(1:3,1:3,t)*(rot_ua_inv(1:3,1:3,t)*(rot_fa_ha
nd(1:3,1:3,t)));



            %Assume a zy'x'' rotation for Euler Angle Calculations
            %This is updated from 1996 standards
            %Code for Euler Angle calculations uwas adapted from rotzyx.m
function on ISB webpage
            %All output is relative to the defined local coordinate system
            %Before averaging, the sign of either the right or left side will
have to be flipped
            %alpha is the first rotation (about z)
            %Right and Left side are the same
            %+ alpha = flexion
            %- alpha = extension
            %beta is the second rotation (about y)
            %Right side
            %+ beta = internal rotation
            %- beta = external rotation
            %Left side
            %+ beta = external rotation
            %- beta = internal rotation
            %gamma is the third rotation (about x)
            %Right side
            %+ gamma = ulnar deviation
            %- gamma = radial deviation
            %Left side
            %+ gamma = radial deviation
            %- gamma = ulnar deviation

            beta1_wr(t)= asin(-rot_fa_hand(3,1,t)); %calculate y' rotation
first

            sgamma_wr(t) = rot_fa_hand(3,2,t)/cos(beta1_wr(t)); %sin gamma
            cgamma_wr(t) = rot_fa_hand(3,3,t)/cos(beta1_wr(t)); %cos gamma
            gamma1_wr(t) = atan2(sgamma_wr(t),cgamma_wr(t)); %gamma one

            salpha_wr(t) = rot_fa_hand(2,1,t)/cos(beta1_wr(t)); %sin alpha
            calpha_wr(t) = rot_fa_hand(1,1,t)/cos(beta1_wr(t)); %cos alpha
            alpha1_wr(t) = atan2(salpha_wr(t),calpha_wr(t)); %alpha one

            if beta1_wr(t)>=0
                beta2_wr(t) = pi-beta1_wr(t);
            else
                beta2_wr(t)= -pi-beta1_wr(t);
            end
```

```
            sgamma2_wr(t) = rot_fa_hand(3,2,t)/cos(beta2_wr(t)); %sin gamma
two
            cgamma2_wr(t) = rot_fa_hand(3,3,t)/cos(beta2_wr(t)); %cos gamma
two
            gamma2_wr(t) = atan2(sgamma2_wr(t),cgamma2_wr(t)); %gamma two

            salpha2_wr(t) = rot_fa_hand(2,1,t)/cos(beta2_wr(t)); %sin alpha
two
            calpha2_wr(t) = rot_fa_hand(1,1,t)/cos(beta2_wr(t)); %cos alpha
two
            alpha2_wr(t) = atan2(salpha2_wr(t),calpha2_wr(t)); %alpha two


            %         if -pi/2 <= beta1_wr(t) & beta1_wr(t) <= pi/2 %loop sets
values of all angles, based on the x rotation
            %             alpha_wr(t)=alpha1_wr(t);
            %             beta_wr(t)=beta1_wr(t);
            %             gamma_wr(t)=gamma1_wr(t);
            %         else
            %             alpha_wr(t)=alpha2_wr(t);
            %             beta_wr(t)=beta2_wr(t);
            %             gamma_wr(t)=gamma2_wr(t);
            %         end


            if n==1 %convert to degrees
                alpha_rwr=alpha1_wr*(180/pi);
                beta_rwr=beta1_wr*(180/pi);
                gamma_rwr=gamma1_wr*(180/pi);
            else
                alpha_lwr=alpha1_wr*(180/pi);
                beta_lwr=beta1_wr*(180/pi);
                gamma_lwr=gamma1_wr*(180/pi);
            end
    %% Checking the co-ordinate system

%             Ftr2ua(:,t) = rot_tr(:,:,t)*-rP_ua(1:3,1,t);
%             Ftr2fa(:,t) = rot_tr(:,:,t)*-rP_fa(1:3,:,t);
%             Ftr2hand(:,t) = rot_tr(:,:,t)*-rP_hand(1:3,:,t);
%
%             plotcoord(eye(3), [0 0 0], 30);
%             plotcoord(rot_tr_ua(:,:,t), shocen(t,:)*100, 45);
%             plotcoord(rot_tr_fa(:,:,t), latep(t,:)*100, 30);
%             plotcoord(rot_tr_hand(:,:,t), wristcen(t,:)*100, 15);
%             line([0 shocen(t,1)*100], [0 shocen(t,2)*100], [0
shocen(t,3)*100], 'Marker','o','LineStyle','-');
%             line([latep(t,1)*100 shocen(t,1)*100], [latep(t,2)*100
shocen(t,2)*100], [latep(t,3)*100 shocen(t,3)*100],
'Marker','o','LineStyle','-');
%             line([latep(t,1)*100 wristcen(t,1)*100], [latep(t,2)*100
wristcen(t,2)*100], [latep(t,3)*100 wristcen(t,3)*100],
'Marker','o','LineStyle','-');
%
%             line([shocen(t,1)*100
shocen(t,1)*100+Ftr2ua(1,t)],[shocen(t,2)*100
shocen(t,2)*100+Ftr2ua(2,t)],[shocen(t,3)*100 shocen(t,3)*100+Ftr2ua(3,t)],
'Marker','*','LineStyle','--', 'color', 'm');
%             line([latep(t,1)*100 latep(t,1)*100+Ftr2fa(1,t)],[latep(t,2)*100
latep(t,2)*100+Ftr2fa(2,t)],[latep(t,3)*100 latep(t,3)*100+Ftr2fa(3,t)],
'Marker','*','LineStyle','--', 'color', 'm');
%             line([wristcen(t,1)*100
wristcen(t,1)*100+Ftr2hand(1,t)],[wristcen(t,2)*100
wristcen(t,2)*100+Ftr2hand(2,t)],[wristcen(t,3)*100
```

```
wristcen(t,3)*100+Ftr2hand(3,t)], 'Marker','*','LineStyle','--', 'color',
'm');


%
%              hold on;
%              grid on;
%              pause;
%              axis vis3d ;
        end

        %     n=n+1; %analyze left side

    end %end of for n=1:2 loop (right and left side)
    %% Phase detection using Fy of Bench, WC, LC adn Y coordinate of C7 and T3
%
%     figure
%     plot(FMr(:,2),'r')
%     hold on
%     plot(FMl(:,2))
%     title (kinematic_trial)
%     % sfig_filename= [newID,'_',condition, trial];
%     saveas (gcf,kinematic_trial,'bmp')
%     % saveas (gcf,fig_filename,'fig')
%
%     % hold on
%     % plot(FM_wc(:,2),'g')
%     % hold on
%     % plot((kin(:,6)*10000-20000),'c')
%     % hold on
%     % plot((kin(:,9)*10000-20000),'K')
%     close
    % [x,y]= ginput(8);
    % pause;
    % fig_filename= [newID,'_',condition, trial];
    % saveas (gcf,fig_filename,'bmp')
    % saveas (gcf,fig_filename,'fig')

    % figure
    % plot(FMr_raw(100:length(FMr_raw),2),'r')
    % hold on
    % plot(FMl_raw(100:length(FMr_raw),2))
    % hold on
    % plot(FM_wc(100:length(FMr_raw),2),'g')
    % hold on
    % plot((kin(100:length(FMr_raw),6)*10000-20000),'c')
    % hold on
    % plot((kin(100:length(FMr_raw),9)*10000-20000),'K')
    % pause
    % [x,y]= ginput(8);
    % pause;
    % fig_filename= [newID,'_',condition, trial];
    % saveas (gcf,fig_filename,'bmp')
    % saveas (gcf,fig_filename,'fig')


    %% Transfer Time

    % CB Transfer--Prelift:x(1)-x(2),Lift: x(2)-x(3),descent: x(3)-x(4)
    % BC Transfer--Prelift:x(5)-x(6),Lift: x(6)-x(7),descent: x(7)-x(8)

    time_CB=(x(4)-x(1))/60;
```

```
time_BC=(x(8)-x(5))/60;

rest_time = (x(5)-x(4))/60;

transfer_time = [time_CB time_BC rest_time];

%% Kinetic variables
% CB transfer
for n = 1:7

    %Shoulder
    rspeak_Fx1(:,n)=max(fm_rsho((x(n):x(n+1)),1));
    rspeak_Fy1(:,n)=max(fm_rsho((x(n):x(n+1)),2));
    rspeak_Fz1(:,n)=max(fm_rsho((x(n):x(n+1)),3));

    lspeak_Fx1(:,n)=max(fm_lsho((x(n):x(n+1)),1));
    lspeak_Fy1(:,n)=max(fm_lsho((x(n):x(n+1)),2));
    lspeak_Fz1(:,n)=max(fm_lsho((x(n):x(n+1)),3));

    rsmin_Fx1(:,n)=min(fm_rsho((x(n):x(n+1)),1));
    rsmin_Fy1(:,n)=min(fm_rsho((x(n):x(n+1)),2));
    rsmin_Fz1(:,n)=min(fm_rsho((x(n):x(n+1)),3));

    lsmin_Fx1(:,n)=min(fm_lsho((x(n):x(n+1)),1));
    lsmin_Fy1(:,n)=min(fm_lsho((x(n):x(n+1)),2));
    lsmin_Fz1(:,n)=min(fm_lsho((x(n):x(n+1)),3));


    rsavg_Fx1(:,n)=mean(fm_rsho((x(n):x(n+1)),1));
    rsavg_Fy1(:,n)=mean(fm_rsho((x(n):x(n+1)),2));
    rsavg_Fz1(:,n)=mean(fm_rsho((x(n):x(n+1)),3));

    lsavg_Fx1(:,n)=mean(fm_lsho((x(n):x(n+1)),1));
    lsavg_Fy1(:,n)=mean(fm_lsho((x(n):x(n+1)),2));
    lsavg_Fz1(:,n)=mean(fm_lsho((x(n):x(n+1)),3));

    rspeak_Mx1(:,n)=max(fm_rsho((x(n):x(n+1)),4));
    rspeak_My1(:,n)=max(fm_rsho((x(n):x(n+1)),5));
    rspeak_Mz1(:,n)=max(fm_rsho((x(n):x(n+1)),6));

    lspeak_Mx1(:,n)=max(fm_lsho((x(n):x(n+1)),4));
    lspeak_My1(:,n)=max(fm_lsho((x(n):x(n+1)),5));
    lspeak_Mz1(:,n)=max(fm_lsho((x(n):x(n+1)),6));


    rsmin_Mx1(:,n)=min(fm_rsho((x(n):x(n+1)),4));
    rsmin_My1(:,n)=min(fm_rsho((x(n):x(n+1)),5));
    rsmin_Mz1(:,n)=min(fm_rsho((x(n):x(n+1)),6));

    lsmin_Mx1(:,n)=min(fm_lsho((x(n):x(n+1)),4));
    lsmin_My1(:,n)=min(fm_lsho((x(n):x(n+1)),5));
    lsmin_Mz1(:,n)=min(fm_lsho((x(n):x(n+1)),6));


    rsavg_Mx1(:,n)=mean(fm_rsho((x(n):x(n+1)),4));
    rsavg_My1(:,n)=mean(fm_rsho((x(n):x(n+1)),5));
    rsavg_Mz1(:,n)=mean(fm_rsho((x(n):x(n+1)),6));

    lsavg_Mx1(:,n)=mean(fm_lsho((x(n):x(n+1)),4));
    lsavg_My1(:,n)=mean(fm_lsho((x(n):x(n+1)),5));
```

```
lsavg_Mz1(:,n)=mean(fm_lsho((x(n):x(n+1)),6));

rspeak_RF(:,n)=max(rf_rsho((x(n):x(n+1)),1));
rsmin_RF(:,n)=min(rf_rsho((x(n):x(n+1)),1));
rsavg_RF(:,n)=mean(rf_rsho((x(n):x(n+1)),1));
lspeak_RF(:,n)=max(rf_lsho((x(n):x(n+1)),1));
lsmin_RF(:,n)=min(rf_lsho((x(n):x(n+1)),1));
lsavg_RF(:,n)=mean(rf_lsho((x(n):x(n+1)),1));

rspeak_RM(:,n)=max(rm_rsho((x(n):x(n+1)),1));
rsmin_RM(:,n)=min(rm_rsho((x(n):x(n+1)),1));
rsavg_RM(:,n)=mean(rm_rsho((x(n):x(n+1)),1));
lspeak_RM(:,n)=max(rf_lsho((x(n):x(n+1)),1));
lsmin_RM(:,n)=max(rf_lsho((x(n):x(n+1)),1));
lsavg_RM(:,n)=mean(rm_lsho((x(n):x(n+1)),1));


%Elbow
repeak_Fx1(:,n)=max(fm_relbow((x(n):x(n+1)),1));
repeak_Fy1(:,n)=max(fm_relbow((x(n):x(n+1)),2));
repeak_Fz1(:,n)=max(fm_relbow((x(n):x(n+1)),3));

lepeak_Fx1(:,n)=max(fm_lelbow((x(n):x(n+1)),1));
lepeak_Fy1(:,n)=max(fm_lelbow((x(n):x(n+1)),2));
lepeak_Fz1(:,n)=max(fm_lelbow((x(n):x(n+1)),3));


remin_Fx1(:,n)=min(fm_relbow((x(n):x(n+1)),1));
remin_Fy1(:,n)=min(fm_relbow((x(n):x(n+1)),2));
remin_Fz1(:,n)=min(fm_relbow((x(n):x(n+1)),3));

lemin_Fx1(:,n)=min(fm_lelbow((x(n):x(n+1)),1));
lemin_Fy1(:,n)=min(fm_lelbow((x(n):x(n+1)),2));
lemin_Fz1(:,n)=min(fm_lelbow((x(n):x(n+1)),3));

reavg_Fx1(:,n)=mean(fm_relbow((x(n):x(n+1)),1));
reavg_Fy1(:,n)=mean(fm_relbow((x(n):x(n+1)),2));
reavg_Fz1(:,n)=mean(fm_relbow((x(n):x(n+1)),3));

leavg_Fx1(:,n)=mean(fm_lelbow((x(n):x(n+1)),1));
leavg_Fy1(:,n)=mean(fm_lelbow((x(n):x(n+1)),2));
leavg_Fz1(:,n)=mean(fm_lelbow((x(n):x(n+1)),3));

repeak_Mx1(:,n)=max(fm_relbow((x(n):x(n+1)),4));
repeak_My1(:,n)=max(fm_relbow((x(n):x(n+1)),5));
repeak_Mz1(:,n)=max(fm_relbow((x(n):x(n+1)),6));

lepeak_Mx1(:,n)=max(fm_lelbow((x(n):x(n+1)),4));
lepeak_My1(:,n)=max(fm_lelbow((x(n):x(n+1)),5));
lepeak_Mz1(:,n)=max(fm_lelbow((x(n):x(n+1)),6));

remin_Mx1(:,n)=min(fm_relbow((x(n):x(n+1)),4));
remin_My1(:,n)=min(fm_relbow((x(n):x(n+1)),5));
remin_Mz1(:,n)=min(fm_relbow((x(n):x(n+1)),6));

lemin_Mx1(:,n)=min(fm_lelbow((x(n):x(n+1)),4));
lemin_My1(:,n)=min(fm_lelbow((x(n):x(n+1)),5));
lemin_Mz1(:,n)=min(fm_lelbow((x(n):x(n+1)),6));

reavg_Mx1(:,n)=mean(fm_relbow((x(n):x(n+1)),4));
reavg_My1(:,n)=mean(fm_relbow((x(n):x(n+1)),5));
```

```
reavg_Mz1(:,n)=mean(fm_relbow((x(n):x(n+1)),6));

leavg_Mx1(:,n)=mean(fm_lelbow((x(n):x(n+1)),4));
leavg_My1(:,n)=mean(fm_lelbow((x(n):x(n+1)),5));
leavg_Mz1(:,n)=mean(fm_lelbow((x(n):x(n+1)),6));

repeak_RF(:,n)=max(rf_relbow((x(n):x(n+1)),1));
remin_RF(:,n)=min(rf_relbow((x(n):x(n+1)),1));
reavg_RF(:,n)=mean(rf_relbow((x(n):x(n+1)),1));
lepeak_RF(:,n)=max(rf_lelbow((x(n):x(n+1)),1));
lemin_RF(:,n)=min(rf_lelbow((x(n):x(n+1)),1));
leavg_RF(:,n)=mean(rf_lelbow((x(n):x(n+1)),1));

repeak_RM(:,n)=max(rm_relbow((x(n):x(n+1)),1));
remin_RM(:,n)=min(rm_relbow((x(n):x(n+1)),1));
reavg_RM(:,n)=mean(rm_relbow((x(n):x(n+1)),1));
lepeak_RM(:,n)=max(rm_lelbow((x(n):x(n+1)),1));
lemin_RM(:,n)=min(rm_lelbow((x(n):x(n+1)),1));
leavg_RM(:,n)=mean(rm_lelbow((x(n):x(n+1)),1));

%Wrist

rwpeak_Fx1(:,n)=max(fm_rwrist((x(n):x(n+1)),1));
rwpeak_Fy1(:,n)=max(fm_rwrist((x(n):x(n+1)),2));
rwpeak_Fz1(:,n)=max(fm_rwrist((x(n):x(n+1)),3));

lwpeak_Fx1(:,n)=max(fm_lwrist((x(n):x(n+1)),1));
lwpeak_Fy1(:,n)=max(fm_lwrist((x(n):x(n+1)),2));
lwpeak_Fz1(:,n)=max(fm_lwrist((x(n):x(n+1)),3));

rwmin_Fx1(:,n)=min(fm_rwrist((x(n):x(n+1)),1));
rwmin_Fy1(:,n)=min(fm_rwrist((x(n):x(n+1)),2));
rwmin_Fz1(:,n)=min(fm_rwrist((x(n):x(n+1)),3));

lwmin_Fx1(:,n)=min(fm_lwrist((x(n):x(n+1)),1));
lwmin_Fy1(:,n)=min(fm_lwrist((x(n):x(n+1)),2));
lwmin_Fz1(:,n)=min(fm_lwrist((x(n):x(n+1)),3));


rwavg_Fx1(:,n)=mean(fm_rwrist((x(n):x(n+1)),1));
rwavg_Fy1(:,n)=mean(fm_rwrist((x(n):x(n+1)),2));
rwavg_Fz1(:,n)=mean(fm_rwrist((x(n):x(n+1)),3));

lwavg_Fx1(:,n)=mean(fm_lwrist((x(n):x(n+1)),1));
lwavg_Fy1(:,n)=mean(fm_lwrist((x(n):x(n+1)),2));
lwavg_Fz1(:,n)=mean(fm_lwrist((x(n):x(n+1)),3));

rwpeak_Mx1(:,n)=max(fm_rwrist((x(n):x(n+1)),1));
rwpeak_My1(:,n)=max(fm_rwrist((x(n):x(n+1)),2));
rwpeak_Mz1(:,n)=max(fm_rwrist((x(n):x(n+1)),3));

lwpeak_Mx1(:,n)=max(fm_lwrist((x(n):x(n+1)),1));
lwpeak_My1(:,n)=max(fm_lwrist((x(n):x(n+1)),2));
lwpeak_Mz1(:,n)=max(fm_lwrist((x(n):x(n+1)),3));

rwmin_Mx1(:,n)=min(fm_rwrist((x(n):x(n+1)),1));
rwmin_My1(:,n)=min(fm_rwrist((x(n):x(n+1)),2));
rwmin_Mz1(:,n)=min(fm_rwrist((x(n):x(n+1)),3));

lwmin_Mx1(:,n)=min(fm_lwrist((x(n):x(n+1)),1));
lwmin_My1(:,n)=min(fm_lwrist((x(n):x(n+1)),2));
```

```
lwmin_Mz1(:,n)=min(fm_lwrist((x(n):x(n+1)),3));

rwavg_Mx1(:,n)=mean(fm_rwrist((x(n):x(n+1)),1));
rwavg_My1(:,n)=mean(fm_rwrist((x(n):x(n+1)),2));
rwavg_Mz1(:,n)=mean(fm_rwrist((x(n):x(n+1)),3));

lwavg_Mx1(:,n)=mean(fm_lwrist((x(n):x(n+1)),1));
lwavg_My1(:,n)=mean(fm_lwrist((x(n):x(n+1)),2));
lwavg_Mz1(:,n)=mean(fm_lwrist((x(n):x(n+1)),3));

rwpeak_RF(:,n)=max(rf_rwrist((x(n):x(n+1)),1));
rwmin_RF(:,n)=min(rf_rwrist((x(n):x(n+1)),1));
rwavg_RF(:,n)=mean(rf_rwrist((x(n):x(n+1)),1));
lwpeak_RF(:,n)=max(rf_lwrist((x(n):x(n+1)),1));
lwmin_RF(:,n)=max(rf_lwrist((x(n):x(n+1)),1));
lwavg_RF(:,n)=mean(rf_lwrist((x(n):x(n+1)),1));

rwpeak_RM(:,n)=max(rm_rwrist((x(n):x(n+1)),1));
rwmin_RM(:,n)=min(rm_rwrist((x(n):x(n+1)),1));
rwavg_RM(:,n)=mean(rm_rwrist((x(n):x(n+1)),1));
lwpeak_RM(:,n)=max(rm_lwrist((x(n):x(n+1)),1));
lwmin_RM(:,n)=min(rm_lwrist((x(n):x(n+1)),1));
lwavg_RM(:,n)=mean(rm_lwrist((x(n):x(n+1)),1));

%Hand

rhpeak_Fx1(:,n) = max(fxr_hand(x(n):x(n+1),1));
rhpeak_Fy1(:,n) = max(fyr_hand(x(n):x(n+1),1));
rhpeak_Fz1(:,n) = max(fzr_hand(x(n):x(n+1),1));

rhmin_Fx1(:,n) = min(fxr_hand(x(n):x(n+1),1));
rhmin_Fy1(:,n) = min(fyr_hand(x(n):x(n+1),1));
rhmin_Fz1(:,n) = min(fzr_hand(x(n):x(n+1),1));

rhavg_Fx1(:,n) = mean(fxr_hand(x(n):x(n+1),1));
rhavg_Fy1(:,n) = mean(fyr_hand(x(n):x(n+1),1));
rhavg_Fz1(:,n) = mean(fzr_hand(x(n):x(n+1),1));

lhpeak_Fx1(:,n) = max(fxl_hand(x(n):x(n+1),1));
lhpeak_Fy1(:,n) = max(fyl_hand(x(n):x(n+1),1));
lhpeak_Fz1(:,n) = max(fzl_hand(x(n):x(n+1),1));

lhmin_Fx1(:,n) = min(fxl_hand(x(n):x(n+1),1));
lhmin_Fy1(:,n) = min(fyl_hand(x(n):x(n+1),1));
lhmin_Fz1(:,n) = min(fzl_hand(x(n):x(n+1),1));

lhavg_Fx1(:,n) = mean(fxl_hand(x(n):x(n+1),1));
lhavg_Fy1(:,n) = mean(fyl_hand(x(n):x(n+1),1));
lhavg_Fz1(:,n) = mean(fzl_hand(x(n):x(n+1),1));

lhpeak_RF(:,n) = max(resultant_force_L3mp(x(n):x(n+1),1));
rhpeak_RF(:,n) = max(resultant_force_R3mp(x(n):x(n+1),1));
lhmin_RF(:,n) = min(resultant_force_L3mp(x(n):x(n+1),1));
rhmin_RF(:,n) = min(resultant_force_R3mp(x(n):x(n+1),1));
lhavg_RF(:,n) = mean(resultant_force_L3mp(x(n):x(n+1),1));
rhavg_RF(:,n) = mean(resultant_force_R3mp(x(n):x(n+1),1));


    for p= 1:length(feet_FP)

        rf_feet(p,:)=sqrt(feet_FP(p,1)^2+ feet_FP(p,2)^2 +
```

```
feet_FP(p,3)^2);
            end
            %FEET

            feet_peak(:,n) = max(feet_FP(x(n):x(n+1),:));
            feet_min(:,n) = min(feet_FP(x(n):x(n+1),:));
            feet_avg(:,n)  =  max(feet_FP(x(n):x(n+1),:));
            feetpeak_RF(:,n)= max(rf_feet(x(n):x(n+1),:));
            feetmin_RF(:,n)= min(rf_feet(x(n):x(n+1),:));
            feetavg_RF(:,n)= mean(rf_feet(x(n):x(n+1),:));

    end


    ShoF=   [rspeak_Fx1' rspeak_Fy1' rspeak_Fz1' rsmin_Fx1' rsmin_Fy1'
rsmin_Fz1' rsavg_Fx1' rsavg_Fy1' rsavg_Fz1' lspeak_Fx1' lspeak_Fy1'
lspeak_Fz1' lsmin_Fx1' lsmin_Fy1' lsmin_Fz1' lsavg_Fx1' lsavg_Fy1' lsavg_Fz1'
rspeak_RF' rsmin_RF' rsavg_RF' lspeak_RF' lsmin_RF' lsavg_RF'];
    %
    ShoM=   [rspeak_Mx1' rspeak_My1' rspeak_Mz1' rsmin_Mx1' rsmin_My1'
rsmin_Mz1' rsavg_Mx1' rsavg_My1' rsavg_Mz1' lspeak_Mx1' lspeak_My1'
lspeak_Mz1' lsmin_Mx1' lsmin_My1' lsmin_Mz1' lsavg_Mx1' lsavg_My1' lsavg_Mz1'
rspeak_RM' rsmin_RM' rsavg_RM' lspeak_RM' lsmin_RM' lsavg_RM'];

    ElbowF= [repeak_Fx1' repeak_Fy1' repeak_Fz1' remin_Fx1' remin_Fy1'
remin_Fz1' reavg_Fx1' reavg_Fy1' reavg_Fz1' lepeak_Fx1' lepeak_Fy1'
lepeak_Fz1' lemin_Fx1' lemin_Fy1' lemin_Fz1' leavg_Fx1' leavg_Fy1' leavg_Fz1'
repeak_RF' remin_RF' reavg_RF' lepeak_RF' lemin_RF' leavg_RF'];
    %
    ElbowM= [repeak_Mx1' repeak_My1' repeak_Mz1' remin_Mx1' remin_My1'
remin_Mz1' reavg_Mx1' reavg_My1' reavg_Mz1' lepeak_Mx1' lepeak_My1'
lepeak_Mz1' lemin_Mx1' lemin_My1' lemin_Mz1' leavg_Mx1' leavg_My1' leavg_Mz1'
repeak_RM' remin_RM' reavg_RM' lepeak_RM' lemin_RM' leavg_RM'];

    WristF= [rwpeak_Fx1' rwpeak_Fy1' rwpeak_Fz1' rwmin_Fx1' rwmin_Fy1'
rwmin_Fz1' rwavg_Fx1' rwavg_Fy1' rwavg_Fz1' lwpeak_Fx1' lwpeak_Fy1'
lwpeak_Fz1' lwmin_Fx1' lwmin_Fy1' lwmin_Fz1' lwavg_Fx1' lwavg_Fy1' lwavg_Fz1'
rwpeak_RF' rwmin_RF' rwavg_RF' lwpeak_RF' lwmin_RF' lwavg_RF'];

    WristM= [rwpeak_Mx1' rwpeak_My1' rwpeak_Mz1' rwmin_Mx1' rwmin_My1'
rwmin_Mz1' rwavg_Mx1' rwavg_My1' rwavg_Mz1' lwpeak_Mx1' lwpeak_My1'
lwpeak_Mz1' lwmin_Mx1' lwmin_My1' lwmin_Mz1' lwavg_Mx1' lwavg_My1' lwavg_Mz1'
rwpeak_RM' rwmin_RM' rwavg_RM' lwpeak_RM' lwmin_RM' lwavg_RM'];

    HandF=  [rhpeak_Fx1' rhpeak_Fy1' rhpeak_Fz1' rhmin_Fx1' rhmin_Fy1'
rhmin_Fz1' rhavg_Fx1' rhavg_Fy1' rhavg_Fz1' lhpeak_Fx1' lhpeak_Fy1'
lhpeak_Fz1' lhmin_Fx1' lhmin_Fy1' lhmin_Fz1' lhavg_Fx1' lhavg_Fy1' lhavg_Fz1'
rhpeak_RF' rhmin_RF' rhavg_RF' lhpeak_RF' lhmin_RF'  lhavg_RF'];

    FeetF = [feet_peak' feet_min' feet_avg' feetpeak_RF' feetmin_RF'
feetavg_RF'];
%     outputFM = [ShoF ShoM ElbowF ElbowM WristF WristM HandF ];
    outputFM = [ShoF ShoM ElbowF ElbowM WristF WristM HandF FeetF];
% eval(['save
S:\Students\padmaja(kankipatip)\DISSERTATION\PROCESSED_DATA\processed_kinetics
\' ,'TX' ID 'V' technique 'L' transfer_trial '_kinetic', ' x
kinetics_processed processed_kinematics transfer_time plot_variables fm_rsho
fm_lsho rf_rsho rf_lsho rm_rsho rm_lsho fm_relbow fm_lelbow rf_relbow
rf_lelbow rm_relbow rm_lelbow fm_rwrist fm_lwrist rf_rwrist rf_lwrist
rm_rwrist rm_lwrist  fxr_hand fyr_hand fzr_hand fxr_hand fyr_hand fxl_hand
fyl_hand fzl_hand resultant_force_L3mp resultant_force_R3mp  ShoF ShoM ElbowF
ElbowM WristF WristM HandF outputFM ';])
```

```
eval(['save
S:\Students\padmaja(kankipatip)\DISSERTATION\PROCESSED_DATA\processed_kinetics
\','TX' ID 'V' technique 'L' transfer_trial '_kinetic', ' x kinetics_processed
transfer_time plot_variables fm_rsho fm_lsho rf_rsho rf_lsho rm_rsho rm_lsho
fm_relbow fm_lelbow rf_relbow rf_lelbow rm_relbow rm_lelbow fm_rwrist
fm_lwrist rf_rwrist rf_lwrist rm_rwrist rm_lwrist  fxr_hand fyr_hand fzr_hand
fxr_hand fyr_hand fxl_hand fyl_hand fzl_hand resultant_force_L3mp
resultant_force_R3mp feet_FP rf_feet ShoF ShoM ElbowF ElbowM WristF WristM
HandF FeetF outputFM ';])

end
```

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