The Relationship between Joint Ranges of Motion and Joint Kinetics during Sitting Pivot Wheelchair Transfers

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Introduction

A full-time wheelchair user will perform on average 14 to 18 transfers per day (Finley, McQuade, & Rodgers, 2005). During the performance of transfers, a wheelchair user often applies excessive loads on their arms, leading to upper extremity pain and injuries (Finley & Rodgers, 2004). During transfers, the loading on the upper extremity joints is greater than any other wheelchair related activity (Gagnon, Nadeau, Noreau, Dehail, & Piotte, 2008). Excessive forces acting at the shoulder during transfers can lead to the development of shoulder impingement, posterior instability, capsulitis, and tendinitis (Campbell & Koris, 1996) (Finley & Rodgers, 2004). Additionally, high superior forces generated during transfer are believed to contribute to pain and secondary impairments at the elbow (Koontz, Kankipati, Lin, Cooper, & Boninger, 2011). Moreover, the extreme wrist extension angles and forces generated during transfers may increase the pressure of carpal tunnel and exacerbate carpal tunnel syndrome (Keir, Wells, Ranney, & Lavery, 1997) (Sie, Waters, Adkins, & Gellman, 1992).

It is largely believed that injuries occur because of large joint reaction forces and moments during transfers (Finley & Rodgers, 2004) (Koontz et al., 2011) (C. Y. Tsai, Hogaboom, Boninger, & Koontz, 2014). However, it is difficult to directly measure the joint reaction force and the moment during the transfer without significant invasive procedures. Joint forces and moments are therefore indirectly estimated using inverse dynamics analyses, which use the anatomical movements (kinematics) to back calculate joint reaction forces and moments (kinetics). Using these analyses, its possible to explore the kinematics that may lead to injury-inducing kinetics. Extreme combinations of shoulder flexion, internal rotation, and abduction are known to create high internal joint forces and are difficult to avoid during transfers (Gagnon et al., 2008). Systematically varying leading hand placement and trunk position during the transfer has been shown to generate different joint forces and moments at the upper limb joints (Kankipati, Boninger, Gagnon, Cooper, & Koontz, 2015). Using certain transfer skills (e.g. placing both feet on the floor, using head-hips maneuver to pivot the body, using proper handgrip techniques) has been shown to reduce loading across the wrists, elbows and shoulders (C.Y. Tsai et al., 2016). It's not clear from these studies however which specific kinematic variables (e.g. trunk and upper body joint motions) are responsible for increasing joint kinetics. Identifying the kinds of motions that minimize joint forces or moments will help guide training interventions and help patients prevent pain and injury. The aim of this study was to investigate the relationship between the kinetics and the kinematics of the upper extremities during level bench and toilet transfers. We hypothesized that there would be specific kinematic variables at each joint that could be used to predict the joint force and moments acting on the upper extremities.

Methods

Participants.

The study was approved by the Department of Veterans Affairs Institutional Review Board. All testing occurred at the Human Engineering Research Laboratories in Pittsburgh, PA. The inclusion criteria were that the subjects were: 1) over 18 years old, 2) one year after injury or diagnosis 3) using a wheelchair for majority of mobility (40 hours per week), 4) unable to stand up without support. Individuals with pressure ulcers within the past year or history of angina or seizures were excluded.

Testing Protocol.

After subjects provided informed consent, their anthropometric measures were collected, such as upper arm length and circumference, to determine the center of mass and moment of inertia for each segment. Subjects were asked to naturally position themselves next to a bench and a commode, which was set at a height level to their own wheelchair seat, on a custom-built transfer station (Figure 1). The transfer station contains three force plates (Bertec Corporation, Columbus, OH) which were underneath the wheelchair, level bench, and the subject's feet, respectively. Two 6-component load cells (Model MC5 from AMTI, Watertown, MA; Model Omega 160 from ATI, Apex, NC) were attached to two steel beams used to simulate an armrest and grab bar (Figure 1). The position of the grab bar was adjusted based on the subjects' preferences. Reflective markers were placed on subjects' heads, trunks, and upper extremities to build local coordinate systems for each segment.

Subjects were asked to perform up to five trials of level-height bench transfers and five trials of commode transfers to and from their own wheelchairs. The order of the transfer surfaces was randomized. The subjects were provided an opportunity to adjust their wheelchair position and familiarize themselves with the setup prior to data collection. Subjects had time to rest in between trials and additional rest was provided as needed. They were asked to use their own approaches to transfer so their transfer movement pattern and techniques would be as natural as possible. Subjects were asked to use the wheelchair grab bar on the trailing side (Figure 1) when they transferred to and from the bench so the reaction forces at the hand could be accurately recorded on the wheelchair side. On the bench or the commode side, subjects were free to place their hand on either the transfer surface or the grab bar. Reflective markers were placed on subjects' heads, trunks, and upper extremities to build local coordinate systems for each segment. Marker trajectories were collected at 100 Hz using a ten-camera three-dimensional motion capture system (Vicon, Centennial, CO.) Kinetic data from all the force plates and load cells were collected at 1000 Hz and later downsampled to match the marker data sampling frequency.

Data analysis

The kinematic variables (body motions) and kinetic variables (joint forces and moments) were computed using MATLAB (Mathworks, Inc., Natick, MA, USA). A zero-lag low-pass 4th order Butterworth filter with cutoff frequency of 7 and 5 Hz was used to filter the kinetic and kinematic data, respectively (Koontz et al., 2011). A transfer was determined to begin when a vertical reaction force was detected by the load cell on the wheelchair side grab bar (Figure 1). The transfer ended before a landing spike was detected by the force plate underneath the bench. Hanavan's model was used to calculate center of mass and moment of inertia using the subjects' segment lengths and circumferences (Hanavan, 1964). Three-component forces and moments measured by the load cells and the force plates (Figure 1), the marker data of the trunk and upper extremities, and the inertial properties of each body segment were inputs into an inverse dynamic model (Cooper, Boninger, Shimada, & Lawrence, 1999). Each segment was assumed as a rigid body and linked together by ball and socket joints. The kinetic variables included maximum resultant forces and moments at the shoulders, elbows, and wrists on the leading (left arm) and trailing (right arm) sides. Each kinetic variable was normalized by body mass (in kilogram) (Gagnon et al., 2008) (Desroches, Gagnon, Nadeau, & Popovic, 2013).

Statistics

Descriptive statistics (means and standard deviations) were calculated for each variable. All of the kinetic and kinematic data were examined for normality using the Shapiro-Wilk test. Bivariate Pearson correlation tests were performed between each kinematic and kinetic variable. The kinematic variables were entered as predictor variables into a multiple linear regression model with the kinetic variables as the outcome variables. Separate models were created for each arm, surface and kinetic variable. Histograms and Q- Q plots were used to check the assumption of no outliers on both predictors and outcome variables. A scatterplot of the standardized residuals against the predicted value was used to test the assumption of linearity. The assumption of multicollinearity for the predictors was also tested using variance inflation factors (VIFs). The assumptions of homoscedasticity and independence for multiple linear regression was checked using the Breusch-Pagan test and Durbin-Watson test, respectively. Stepwise linear regression was selected for the modeling process. The level of significance was set at $p < .05$. All the statistical analyses were performed in SPSS 25 (SPSS Inc., Chicago, IL).

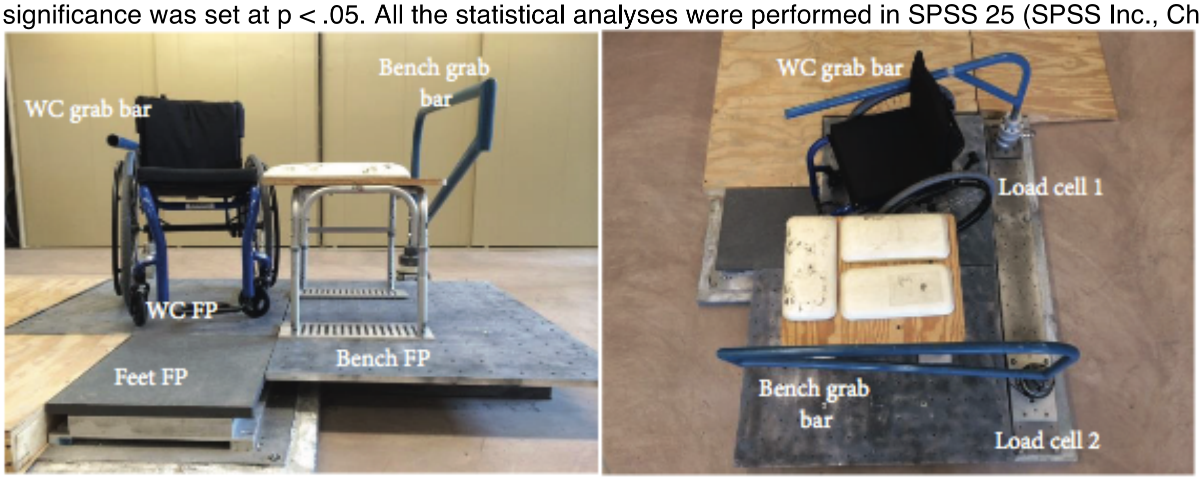

Figure 1: Front (a) and top (b) views of the transfer station. WC: wheelchair; FP: force plate.

Results:

Table 1: Subjects Demographics (n = 24)	Mean ± standard deviation
Age (years)	38.30 ± 11.07
Height (m)	1.67 ± 0.23
Weight (Kg)	67.14 ± 19.18

Average duration of using a wheelchair (years)	13.15 ± 8.13

Twenty-one men and three women participated in the study (Table 1). Eighteen subjects had a spinal cord injury (SCI); 14 subjects reported a complete SCI and four subjects an incomplete SCI (three with American Spinal Injury Association (ASIA) Grade B, one with ASIA Grade C). Three subjects had tetrapleiga (C4 to C6), nine had high paraplegia (T2 to T7), and six had low paraplegia (T8 to L3). The remaining five participants had bilateral tibial and fibular fractures with nerve damage (n=1), double above knee amputation (n=1), muscular dystrophy (n=1), osteogenesis imperfecta (n=1), and myelopathy (n=1). One subject did not complete the commode trial.

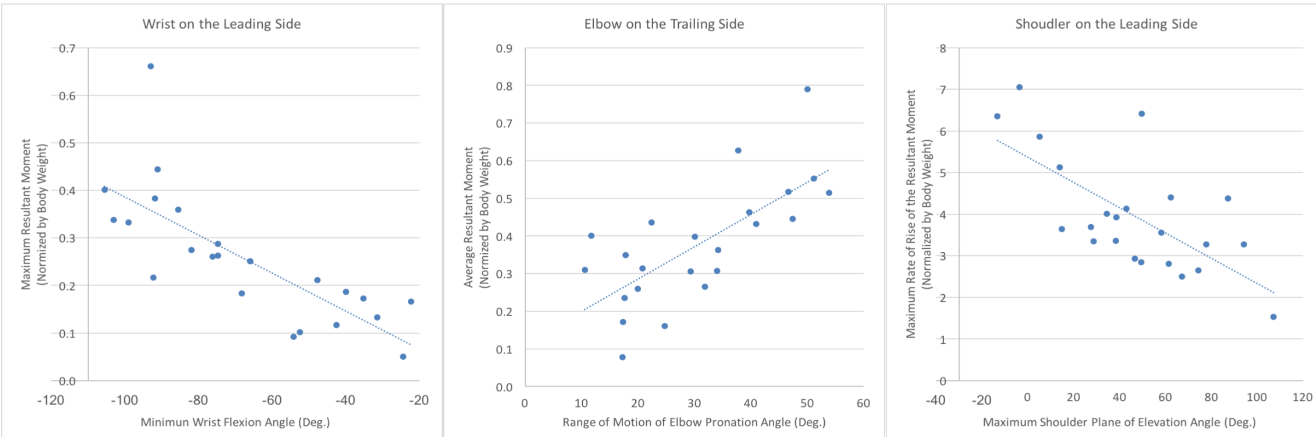

Figure 2: Examples of the correlation between kinematic and kinetic variables on upper extremities joints during the level-height bench and the commode transfer. The representative scatterplots show that a) greater leading side wrist extension was associated with larger maximum resultant moments at the wrist (R^2 = .58) b) larger trailing side forearm pronation angles was associated with higher resultant moments at the elbow (R^2 = .51); c) leading arms that were positioned more anterior of the trunk were associated with lower rates of resultant moment loading at the shoulder during the transfer (R^2 = .47).

Table 2: Correlational test results for the subjects' leading arm side during the level-height bench transfer. RF, resultant force; RM, resultant moment; RF/RM rate, rate of rise of the resultant force/moment.

Level-Height Bench Transfer Leading Side				
Joint	Kinematic Variables	Kinetic Variables		
Shoulder	Plane of Elevation	RF rate at shoulder -.524*		
		RM rate at shoulder -.683**		
Elbow	Flexion/Extension	RM rate at shoulder -.544**	RM at Elbow -.554**	
Wrist	Flexion/Extension	RM at shoulder .607**		RM at Wrist -.764**
		RM rate at shoulder .578**		
	Ulnar/Radial Deviation	RM at shoulder .536**		RM at Wrist -.561**
		RM rate at shoulder -.557**		
	Pronation/Supination			RM at Wrist -.686**

Table 3: Regression model results for the bench and commode transfers for the resultant moments at the shoulder and elbow joints.

Surface	Side	kinetic (DV)	Mean	SD	sig.	R Square	Kinematic (IV)	Standardized Coefficients Beta
Bench	Leading	MAX RM Shoulder	1.06	0.25	<0.001	0.839	Leading Shoulder Plane of Elevation_ROM	-0.564
							Leading Shoulder Rotation_MIN	-0.419
							Leading Wrist Flexion_MAX	0.484
							Leading Wrist Ulnar Deviation_ROM	-0.349
							Trailing Shoulder Elevation_ROM	0.496
							Trailing Wrist Ulnar Deviation_ROM	0.281
	Trailing	AVG RM Elbow	0.38	0.16	<0.001	0.693	Trailing Shoulder Plane of Elevation_ROM	0.676
							Trailing Elbow Pronation_ROM	-0.49
Commode	Leading	AVG RM Shoulder	0.54	0.18	<0.001	0.925	Leading Elbow Flexion_ROM	-0.319
							Leading Elbow Pronation_ROM	0.532
							Leading Wrist Flexion_MIN	0.503
							Leading Wrist Pronation_MAX	-0.509
							Trailing Shoulder Plane of Elevation_ROM	0.299
							Trailing Shoulder Rotation_MAX	-0.439
							Trailing Wrist Ulnar Deviation_ROM	0.346
	Trailing	AVG RM Elbow	0.34	0.15	<0.001	0.805	Trailing Elbow Flexion_MAX	0.684
							Trailing Wrist Ulnar Deviation_MAX	-0.642
							Leading Shoulder Plane of Elevation_MAX	0.358
							Leading Shoulder Elevation_ROM	0.219

The regression models showed that several kinematic variables were predictive factors for the maximum and average joint resultant forces and moments at the shoulder, the elbow, and the wrist on both leading and trailing arm. A subset of the results are shown in Table 2. The R^2 values for all the models were between .521 to .925. The p-values of the models were all below .005 (Table 3). These results indicate that multiple motions across different joints predict the kinetic outcomes at a single joint. Also the motions of both the leading and trailing sides predict the joint kinetics acting on a single side (leading or trailing arm). In addition, we found that a single joint motion can be a predictor for the joint kinetics at several joints. For example, wrist extension on the leading side was a predictor for the average resultant moment and the rate of rise of the resultant moment at the shoulder, and the resultant moment at the wrist on both leading side and trailing side. We also found that increasing some motions (e.g. wrist pronation) can decrease wrist kinetics but may increase the shoulder kinetics.

Discussion

Our study is one of the first to examine the detailed relationship between the joint motions and joint kinetics for sitting pivot transfers. In this study, we found that the joint kinetics at any one joint are determined by multiple motions that occur at different joints and arms. Therefore, to reduce the forces and moments at the leading shoulder requires paying attention to certain motions that are happening at the leading and trailing arms at each joint. For example, during the commode transfer the average shoulder resultant moment at leading arm is influenced by not only the flexion angles of the leading elbow and wrist, but also shoulder plane of elevation and the wrist ulnar deviation on the trailing side (Table 3).

Our results therefore highlight some of the motions that could be modified to reduce joint forces and moments during transfers and reduce the risk of injury. For example, our models show that one of the factors to reduce the resultant moment at the shoulder on the leading side is to increase the shoulder plane of elevation angle. Increasing the shoulder plane of elevation can be accomplished by positioning the leading hand in front of (anterior) to the body. Wheelchair users who place their leading arm at the front edge of the bench would be expected to generate lower joint loading than if they were to place their arm closer to their body. Another example is that decreasing trailing arm shoulder elevation reduces the resultant moment at the leading-side shoulder (positive correlation) and increases the resultant moment at the trailing-side shoulder (negative correlation). To reduce the angle of shoulder elevation, a person would need to keep their upper arm close to his/her body. Thus, subjects who keep their right arm (trailing side) closer to their trunk could reduce the left (leading side) shoulder resultant moment but then the resultant moments on the right (trailing side) shoulder will increase. Greater loading should be borne by the trailing arm during the initial lift phase of the transfer as it is responsible for initiating the transfer and elevating the body from the wheelchair. On the other hand, during the landing phase the subject relies on his/her leading arm to stabilize the body (e.g. loading is transferred from trailing to leading). Therefore, subjects who place his/her arms in the proper position at the onset of the transfer could prevent overloading on the leading arm during the lifting phase and avoid the excessive loading on the trailing arm during the landing phase. The technique can help the subject to stabilize their body during the lifting and landing phase and keep the transfer smooth and well controlled.

We also learned that level-height bench transfer and commode transfer are fundamentally different in their kinetic and kinematic relationships. The results of the correlation and the regression models are different from the level-height bench transfers for both the leading and trailing arms. For the commode trials, the initial positioning of the wheelchair and hand placement is different between the two surfaces. During the lifting phase of the transfer, the subject spent more time to lift their body between the wheelchair and the landing target perhaps because the toilet seat was more awkward in size and shape compared to the bench and lower that the wheelchair seat for most users. The commode setting forced the subject to complete the transfer with a more limited degree of freedom in body kinematics. Despite these differences, significant relationships between the motions and joint kinetics were found which provide a potential guide for the motions that would need to be altered to reduce the joint kinetics during commode transfers.

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