

# **Trunk Function in Manual Wheelchair Users in Static and Dynamic Conditions**

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## **Abstract**

### **Trunk Function in Manual Wheelchair Users in Static and Dynamic Conditions**

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The relationship between functional capacity and trunk control is of significant interest to a variety of people, from wheelchair sport clinicians and professionals to rehab specialists, but has proven difficult to quantify. This project utilizes a “Volume of Action” framework to observe trunk function in both static and dynamic conditions across a functional spectrum using the functional classification system employed in wheelchair basketball.

Study 1 (chapter 2) examines the role of functional capacity in a static environment using a Volume of Action framework using a seated limits of stability test to examine functional limitations in wheelchair users across a functional spectrum. This investigation stratified wheelchair users into two groups using the functional classification system used in wheelchair basketball, and observed functional differences in the limits of stability in three planes of movement.

Study 2 (chapter 3) went one step further to observe how wheelchair users use their trunk to brake their wheelchair. It is well documented that wheelchair users actively use their trunk to

propel their wheelchair, but no known research has been conducted on how wheelchair users use their trunk to come to a complete stop. Study 2 examined 8 wheelchair users using motion capture technology to determine movements patterns leading up to and during the action of bringing their wheelchair to an abrupt stop at high speeds.

The findings of these studies provide insight into the role that functional capacity plays in the strategies that manual wheelchair users utilize to maintain balance statically and dynamically. These results provide insight for wheelchair designers and wheelchair sport athletes and professionals into the role that functional capacity plays in maintaining balance under static and dynamic conditions. Future investigations should examine the role of sport wheelchair configuration in this relationship between functional capacity and postural control. Additionally, future models should be developed to provide a more complete picture of the pelvis during wheelchair propulsion.

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## **Chapter 1**

### **Introduction**

There are currently between 1.7 and 3.6 million wheelchair users in the United States (Kaye, Kang, & Laplante, 2002) with an estimated 1.5 million of those under the age of 65 (Brault 2012), and between 245,000 and 353,000 individuals living with a spinal cord injury (National Spinal Cord Injury Statistical Center, 2018). Of this range of 1.7 to 3.6 million, 90% use a manual wheelchair. Worldwide, the World Health Organization estimates that a staggering 76 million people require a wheelchair (WHO, 2010). Coupled with the fact that adults with a disability are 53% more likely to be obese than adults without a disability (Fox et al., 2014), physical activity research and the factors that contribute to wheelchair users' ability to engage in it is important for this population.

The study of wheelchair propulsion biomechanics is important for the wheelchair using community, rehab specialists, and wheelchair manufacturers, and has provided extensive knowledge that has contributed to advancements in MWC set up (Van der Woude et al, 1989) and injury prevention (Boninger et al., 2002). For example, The advent of new technology such as the SMARTWheel in 1989 (Cooper & Cheda, 1989) has allowed researchers around the world to observe kinetic and kinematic properties at the pushrim, as well as at the joints of the upper extremities, including the trunk. However, the overwhelming majority of this research has focused on forward propulsion of the wheelchair. To our knowledge no work has been done specifically on the role of functional capacity in MWC braking. As will be demonstrated in this



review of the literature, there remains large gap in what is known about functional differences in how manual wheelchair users bring their wheelchair to a complete stop using their hands.

### **Trunk Function**

The role of the trunk in manual wheelchair propulsion is still not completely understood and can be influenced by a number of factors, including spinal cord lesion level (Schantz et al, 1999; Newsam et al., 1999), surface incline (Chow et al, 2009), and even seat height (Hughes et al, 1992). One school of thought suggests that individuals who use a trunk flexion style of propulsion put themselves at increased risk of injury due to decreased activation of key propulsion muscles during fatigue (Rodgers et al., 2000). However, there is also evidence to suggest that the trunk may serve as a source of power generation in wheelchair propulsion (Sanderson and Sommer, 1985). Vanlandewick et al. ( 2011) found that when able-bodied individuals who were put into a wheelchair to simulate positions associated with stability in wheelchair sports with each position limiting trunk range of motion differently, individuals who had their thighs at 90° relative to their trunk were able to provide significantly more power to the first push of a sprint than those who had their hips either maximally flexed or at flexed at 45°. In a study looking at the role of the trunk in ascending a curb, the degree of trunk flexion increased as curb height increased when manual wheelchair users are attempting to ascend a curb (Lalumiere et al., 2013).

Due to the variety of movement patterns in wheelchair users across the functional spectrum, a number of researchers have taken to using able-bodied subjects to study movement

patterns in wheelchair propulsion (Tordi et al., 1998; Veeger & Rozendal, 1991). This provides more uniformity in results but can often make extrapolating the conclusions to the wheelchair using community. In a study examining trunk muscle activation in able-bodied subjects during wheelchair propulsion, Yang et al. (2006) found that abdominal and back muscle groups co-contracted during the recovery and early push phase to provide stability during propulsion. Putting these results in context, individuals of varying functional capacities would be expected to have different muscular contraction patterns during the push and recovery phases of wheelchair propulsion.

These findings suggest that those with decreased trunk control have a decreased capacity to generate power during propulsion. By extension, it can be hypothesized that those with decreased trunk control will be less able to stop the forward momentum of the trunk during abrupt manual wheelchair braking. Studies looking at postural control in manual wheelchair users have shown that the degree to which the user has control of their trunk has a significant impact on their ability to maintain postural control. For example, significant accelerations, forces, and moments have been seen in experiments that simulate a wheelchair running into a curb (Fast et al., 1997). Additionally, Kamper et al. (1999), in a study examining lateral postural stability, found that the ability to perform static leaning was strongly correlated with dynamic balance, highlighting the relationship between static limits of stability and dynamic postural control. In the absence of functional ability of the trunk to control balance, wheelchair users may rely on strategies to adapt to external perturbations, such as stabilize themselves with their hands. This

may prevent the user from falling out of their wheelchair, however this can interfere with task performance or cause the wheelchair to diverge from its intended path.

Postural control can play a significant role in the user's center of mass, and by extension risk of falls, as variations in the center of mass can reasonably be attributed to variations in the upper body center of mass compared to the fixed lower body in the wheelchair. Sauret et al. (2006) found that the center of mass of the upper body was the most forward at the beginning of a propulsion phase and the most back at the end of the propulsion phase. Unknown is the role of functional capacity in postural control during braking. Considering Newton's first law of motion, an object in motion stays in motion, these data raise some interesting questions about the role of functional capacity in manual wheelchair users during different tasks, including bringing the wheelchair to a complete stop. Questions that need answers include: what strategies do wheelchair users of different functional capacities use to maintain postural control during abrupt braking? How does decreased functional capacity influence the motor planning needed to maintain balance?

An inverted pendulum model has sometimes been used to model trunk stability. An inverted pendulum has a center of mass above a pivot point and will fall without additional help. This model is frequently applied to human gait. The inverted pendulum model seeks to provide information on how humans control balance (Anderson, 1989), and in the context of seated balance is a function of the relationship between the center of mass and the distance from the joint center (Bidard et al., 2000). The model follows such that as the degree to which

wheelchair users have access to stabilizer muscles increases, the relative contribution from individual muscles needed decreases. According to Bidard et al. (2000), when the system stands in isolation the upright position is unstable and requires feedback of disturbances to the positioning to achieve stability. When linear joint stiffness is positive, static stability is achieved. Static stability requires a smaller joint stiffness than is required to resist against disturbances, and the ability to resist against disturbances is directly related to functional capacity, or access to musculature to assist in increasing that joint stiffness. Wilkenfield et al. (2006) developed a model suggested that functional electrical stimulation (FES) may help improve trunk flexion by applying muscle actuator properties to passive muscles. When normal function is impaired compensatory mechanisms are utilized to maintain erect posture, such as relying on a backrest or increased activation of muscles that are typically not heavily relied upon for postural control (Janssen-Potten et al., 2001).

## **Braking**

The ability of manual wheelchair users to proficiently execute basic wheelchair skills is critical in promoting independence in this population. Wheelchair skill proficiency has been shown to increase participation in leisure-time physical activity (Phang et al, 2012) and improve independence and integration in their community, and is also associated with higher overall life satisfaction. This is significant, given the barriers individuals face to engaging in community-based physical activity. Nearly 80% of parents of children with disabilities reported that lack of facilities was a major barrier to encouraging their children to engage in physical activity (Levinson & Reid,

1991), and lack of physical skills was cited as a major theme in barriers to physical education to children with a disability (Shields, Synnot, & Barr, 2011). Skill proficiency has also been shown to be associated with higher community participation and improved life satisfaction (Hosseini et al., 2012). Given this data, it is worth noting that currently, the Wheelchair Skills Test does not incorporate bringing the wheelchair to a complete stop by the user by using their hands on the push rim (Kirby et al., 2002). As will be discussed, braking activities are a significant aspect of manual wheelchair locomotion, so the omission of this skill from the Wheelchair Skills Test ignores an important skill needed by wheelchair users in order to be independent.

The biomechanics of wheelchair propulsion have been a topic of study since the early 1970s, and in that time much has been elucidated about wheelchair propulsion kinetics and kinematics. We have considerable information on wheelchair propulsion in different environments, during sporting events, and across levels of disability. Additionally, we have substantial evidence on the role manual wheelchair propulsion plays in the development of shoulder injuries in this population. However, very little is known about the kinetics and kinematics of manual wheelchair braking. For the purposes of this project, braking is being considered as bringing the wheelchair to a complete stop with the hands on the handrim. Work has been done on braking moments within propulsion cycles (Kwarciak et al., 2009; Richter et al., 2011; Kwarciak et al., 2012), which are considered to be “periods of nonpropulsive contact...[b]efore and after propulsive contact” (Kwarciak et al., 2009). These moments are not considered in these investigations.

Manual wheelchair locomotion has been demonstrated to consist of short bouts of mobility, including starting, stopping, and turning (Sonnenblum, Sprigle, & Lopez, 2012), and is actually consistent with the movement patterns of able-bodied adults (Orendurff et al, 2008). Additionally, adapted sports such as wheelchair basketball consist of rapid changes in direction and bouts of stopping and starting (Cavedon, Zancanaro, & Milanese, 2015). For example, Coutts (1992) found that during an average game wheelchair basketball players will spend 36% of their time engaging in braking activities. A separate analysis attempting to quantify activity during wheelchair basketball and wheelchair rugby games found that wheelchair basketball and wheelchair rugby players will stop their wheelchairs an average of  $239.78 \pm 60.61$  times during a game (Spörner et al, 2009), and in a study examining the movement patterns of children who use wheelchairs, Slavens et al., (2015) estimated that subjects completed up to 250 start and stop activities per day. These data have led to increased calls for research in all aspects of manual wheelchair locomotion in addition to forward propulsion (Morrow et al., 2010; Sonnenblum, Sprigle, & Lopez, 2012; Koontz et al, 2005), and illustrate the need for further research into the biomechanics and physiology of manual wheelchair braking, including the role of different body segments in executing this skill.

While most research on wheelchair biomechanics has focused on wheelchair propulsion, recently there have been a few attempts to examine the effects of braking. Morrow et al. (2010) found that shoulder demand during a stop condition was similar to that of level propulsion ( $0.12 \pm 0.04$  N/N), although this was measured at a self-selected speed and subjects were instructed

to simulate a stop as if they were approaching a door. Speeds in wheelchair sports such as wheelchair basketball can reach up to 4 m/s (Coutts, 1992), and Kim et al. (2002) have found that shoulder forces increase proportionally with speed. Additionally, Kim et al. (2002) found that when subjects were required to abruptly stop their wheelchair after pushing as fast as possible, forces tangential to the pushrim were 160% higher than force needed to start the wheelchair from a complete stop, and 230% higher than maintaining constant speed. In a study looking at abrupt braking forces needed to quickly turn a wheelchair to complete a figure 8 drill, Hwang et al. (2017) found that braking forces were almost the same as the body weight of the subject. Similarly, subjects who achieved higher velocities (and therefore needed greater reductions in velocity, i.e. deceleration) required higher braking forces, thereby increasing risk for injury. In the context of rapidly changing direction, the deceleration of the inside wheel (the direction of the turn) requires direct opposite mechanisms needed to propel the contralateral wheel. Indeed, when considering the magnitude of the forces needed for deceleration, the risk of injury to the shoulder complex was assessed as higher than propulsion. Slavens et al. (2015) found that the glenohumeral joint incurred the largest force of 10.6% of bodyweight during a stopping task in an analysis looking at braking in children, and high joint forces for the elbow and wrist. They determined that propulsion, starting, and stopping a wheelchair all have very different biomechanical features. Trunk movements patterns in this population (wheelchair users under the age of 21) were not significantly different in the sagittal plane, but subjects were moving at a self-selected speed, so there may be differences in movement patterns needed to stop a wheelchair moving at high velocities seen in wheelchair sports. Therefore, one of the aims of

Study 2 (Chapter 3) is to provide a preliminary analysis of the functional differences of manual wheelchair braking as a function of angular velocity and joint kinematics while bringing the wheelchair to an abrupt stop.

### **Angular Momentum**

Control of angular momentum plays an important role in preventing falls during walking (Simoneau & Krebs, 2000) as well as recovering from a trip (Pijnappels et al., 2004), but it remains to be seen if functional impairments of the trunk play a role in control of angular momentum of the trunk, as well as how that manifests in balance during manual wheelchair braking.

Changes in velocity during wheelchair locomotion (both propulsion and braking activities) result in changes in upper body angular momentum that must be arrested and controlled in order to prevent a loss of balance. The cyclical nature of wheelchair propulsion requires that wheelchair users develop muscle synergies to respond to the anterior/posterior inertial forces that must be absorbed by the upper body while pushing a wheelchair (Gagnon et al., 2009). Eccentric contraction of the trunk stabilizers are required for the successful arrest of this momentum during deceleration and gait termination in ambulatory individuals (Bastian, 1997; Dvorznak et al., 2001) and while this has not been observed in manual wheelchair users, we expect that the same principles of physics would apply. To that end, the extent to which that angular momentum can be arrested relies on the user's ability (and by extension, functional capacity) to eccentrically contract their trunk stabilizers. Users with less functional capacity would be expected to be at a disadvantage in that they have less musculature with which to



respond to changes in angular momentum (Dvorznak et al., 2001). Despite this, very little information is available on the role of angular momentum in manual wheelchair users during braking.

What research that does exist on upper body angular momentum during wheelchair braking is demonstrated in crash test dummies. Cooper et al. (1998) observed three different braking methods to examine angular kinematics of the trunk, and in each condition angular velocity of the trunk put the dummy at increased risk of falling out of the wheelchair. In this study speed played a large role in the response of the angular velocity of the trunk, with abrupt braking having the highest risk of falling out of the wheelchair. Results of this study led researchers to recommend that wheelchair users with less trunk control utilize a braking strategy that allows them to come to a more gradual stop. Dvorznak et al. (2001) compared a hybrid II test dummy to a single wheelchair user to compare trunk kinematics, and during the course of data collection found that the wheelchair user lost balance several times in response to the braking of the wheelchair. It is hypothesized that lower functional capacity will be reflected in a higher angular momentum, reflecting the higher degree of difficulty in maintaining balance for these subjects.

### **Wheelchair Configuration**

There is also evidence that wheelchair set up may play a role in the effects of manual wheelchair propulsion, including braking, by attenuating the force incurred. Hwang et al. (2017) found that brake force was significantly lower in subjects who used a wheelchair that had an axle

position that was increased in the anterior position, and peak forces were shown to be decreased (Boninger et al., 2000). This seating position has also been associated with lower electromyography activity (Masse et al., 1992) and higher mechanical efficiency (Van der Woude et al., 1989). A lower and more rearward seating position relative to the axle put the wheelchair user at lower risk for injury to the upper limbs (Kotajarvi et al., 2004; Van der Woude et al., 1989), although it does put the user at higher risk of tipping over backwards (Gaal et al., 1997).

Another strategy that wheelchair users, particularly wheelchair athletes, utilize is increasing posterior tilt to the seat base (known in wheelchair sport circles as “dump”). This is most often seen in wheelchair athletes with lower functional capacity to provide more stability. This allows the athlete to push their back against the backrest (increasing stability) and also decreases the angle between the trunk and thighs, improving balance (Cooper and De Luigi, 2014). This strategy is one component of a concept called “passive pelvic stability”, which is a key concept in wheelchair basketball classification. Typically, athletes that are classed 2.5 and below utilize this strategy due to a lack of muscle control in the lower trunk and hips to maintain pelvic positioning during trunk motion (IWBF, 2014). Further discussion on this classification system follows below.

As previously mentioned, the postural control mediated changes in center of mass are also pertinent to this particular part of the discussion. The fore and aft changes in upper body center of mass change the amount of pressure on both the rear wheels as well as the front wheels (Sauret et al., 2006), and this added pressure on the front wheels can result in an accidental fall

if the wheel comes in contact with a rock or other protruding part of the environment. Tips and falls are the most common form of wheelchair-related injury (Gaal et al., 1997) with 46% of them being in the forward direction (Kirby et al., 1994), and can result in expensive and time consuming rehabilitation. For example, if a wheelchair user fractures their wrist as a result of trying to brace themselves for an impact, they can expect to have severely limited mobility. This can be exacerbated if the wheelchair is equipped with smaller, harder front wheels (Gaal et al., 1997), which provide more efficient mobility, but put the user at higher risk of falls if the wheelchair comes into contact with uneven terrain. With this in mind, an understanding of the relationship between postural control and conditions that result in abrupt stopping of the wheelchair are a necessary and important piece of the literature. In wheelchair sports where stopping, starting, and changing direction are all integral to competition, more attention needs to be focused on the role that manual wheelchair braking plays in health and injury.

### **Classification**

With trunk function in mind, a discussion on how wheelchair users are stratified for both competition and research purposes is warranted. A common practice in research is to stratify wheelchair users based on spinal cord lesion level, with the rationale being that spinal cord lesion level should correlate to functional capacity. While this often holds true, this is not always the case. There are numerous instances where spinal cord injury lesion level did not correlate to functional capacity. For example, in one study examining postural control in individuals with spinal cord injury, the subject with the lowest injury (and presumably, the highest functional

capacity) exhibited the least postural control ability (Fast et al., 1997). In a study looking at lateral stability in spinal cord-injured subjects, Kamper et al. (1999) found that injury level was not always indicative of balance. Another example of this discrepancy in practice is seen in individuals with spina bifida meningocele (MC) and myelomeningocele (MMC). Briefly, spina bifida is a birth defect that results in the failure of the spinal cord to fully develop. This can occur at a number of places on the spinal cord, impacting functional capacity to varying degrees. This often results in an incomplete lesion of the spinal cord, and the loss of function is influenced by both the location of the lesion as well as the nerves affected by the lesion (Northup and Volcik, 2000). Finally, improvements in postural control have been seen in longitudinal studies evaluating balance in spinal-cord injured subjects. Seelen et al. (1998) found that subjects with spinal cord injury developed unique compensatory muscle activation patterns to maintain balance. This suggests a learning effect, and by extension implies that as wheelchair users spend more time learning about their new motor abilities, function may be observed to increase, particularly among those who exercise the function more.

Even if lesion level could guarantee an accurate estimation of functional capacity, this method of stratification leaves out a large population of wheelchair users. The National Spinal Cord Injury Statistical Center estimated that as of 2017 there were anywhere between 245,000 and 353,000 individuals living with a spinal cord injury (National Spinal Cord Injury Statistical Center, 2018) compared to the estimated range of 1.7 million to 3.6 million wheelchair users in the United States. In this range, 1.5 million are under the age of 65 (Kaye, Kang, & Laplante, 2002;

Brault, 2012), and 90% are manual wheelchair users (Kaye, Kang, & Laplante, 2002). If spinal cord injury lesion level is the only means of stratification in research settings, only 20% (conservatively) of the wheelchair using population is being considered for research projects.

There have been several attempts to classify wheelchair users, both within the spinal cord injury population as well as the broader wheelchair-using community. The American Spinal Injury Association (ASIA) (Maynard et al, 1997) has developed a classification system that stratifies individuals with spinal cord injury into classifications A, B, C, D, and E. By examining multiple levels of damage (sensory, motor, and neurological) at 28 key dermatomes on the body, clinicians are able to provide a classification for individuals with spinal cord injury. This system, while including muscular function through manual muscle testing, is of limited utility in evaluating the functional ability of manual wheelchair users since the evaluation of each subject does not take place in their personal wheelchair. The Functional Classification System used by the National Wheelchair Basketball Association (NWBA) and the International Wheelchair Basketball Association (IWBF) was designed and implemented to ensure equitable competition for players of all disability types and functional levels, the rationale being that if such a system was not in place, only those with the highest functional capacity would play and those with higher level injuries or lower levels of functional ability would be left out (IWBF, 2014). This system uses a classification panel of three trained classifiers to observe functional volume of action during competition. Given that wheelchair basketball players present a variety of injuries that limit more than just trunk function (upper limb amputations, limited upper limb range of motion, etc), these

deficits are taken into consideration for overall classification. The IWBF Functional Classification originated as a five point scale, of 1, 2, 3, 4, and 4.5, with higher numbers signifying higher levels of functional ability. Those players who exhibit functional ability that borders between two classes (for example, trunk rotation to one side but not the other) maybe be given 1.5, 2.5, or 3.5 classifications (table 1). A noteworthy point on the Functional Classification System used in wheelchair basketball is that players are classified in their sport wheelchairs, and wheelchair set up is often manipulated in a way that maximizes functional ability (IWBF, 2014). For example, players may utilize straps or belts around their hips or waist in order to maximize trunk stability, or manipulate the angle at which their knees sit relative to their hips in order to generate more power (Kotajarvi et al., 2004). This can make it seem as though the player has more actual function than they do, and make it difficult to assign a classification, as the Functional Classification System relies on classifying based on true functional capacity, not on how wheelchair set up and manipulation facilitates movement. Therefore, one of the goals of study 1 (Chapter 2) is further validate the Functional Classification System used in national and international wheelchair basketball by using a modified limits of stability test. This will be one of the first investigations to quantify rotation of the trunk along the Z axis.

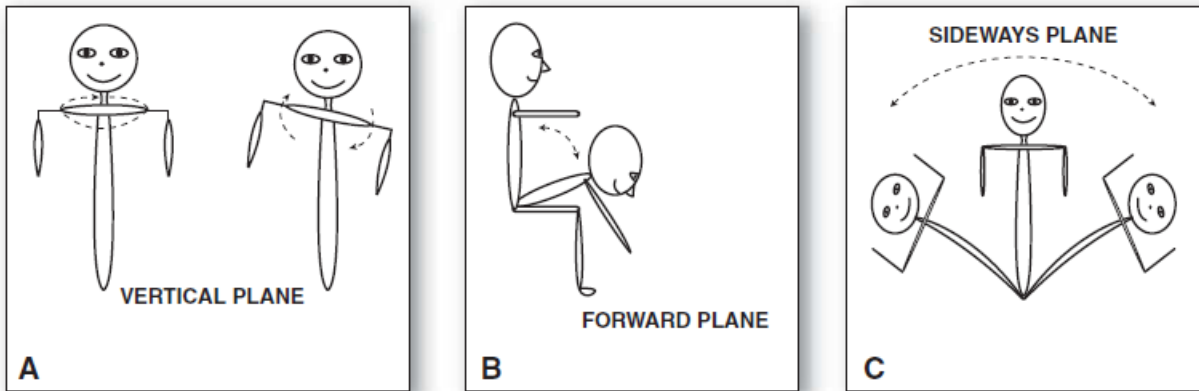
## Figure Legends

**Figure 1.** Illustration of the Volume of Action concept used in the IWBF Functional Classification system

**Table 1:** IWBF Classification Based on Function

Classification	Function
1.0	Little or no controlled trunk movement in the forward plane No active trunk rotation Balance in both forward and sideways directions is significantly impaired Players rely on their arms to return to the upright position when unbalanced
2.0	Partially controlled trunk movement in the forward plane Active upper trunk rotation but no lower trunk function No controlled sideways movement
3.0	Good trunk movement in the forward direction Good trunk rotation No controlled trunk movement sideways
4.0	Normal trunk movements, but usually due to limitations in one lower limb the player has difficulty with controlled movement to one side
4.5	Normal trunk movement in all directions Able to reach side to side with no limitations





**Figure 1.** Volume of Action, adapted from IWBF (2014).

## References

- Anderson, C. W. (1989). Learning to control an inverted pendulum using neural networks. *IEEE Control Systems Magazine*, 9(3), 31-37.
- Bastian, A. J. (1997). Mechanisms of ataxia. *Physical therapy*, 77(6), 672-675.
- Bidard, C., Rienstra, S., Veltink, P. H., Koopman, H. F. J. M., Grady, J., De Vries, J., & Huttenhuism, L. (2000). Trunk stability while standing or sitting: a static analysis. In *IFESS 2000 Conference at Aalborg University, Denmark*.
- Boninger, M. L., Baldwin, M., Cooper, R. A., Koontz, A., & Chan, L. (2000). Manual wheelchair pushrim biomechanics and axle position. *Archives of physical medicine and rehabilitation*, 81(5), 608-613.
- Boninger, M. L., Souza, A. L., Cooper, R. A., Fitzgerald, S. G., Koontz, A. M., & Fay, B. T. (2002). Propulsion patterns and pushrim biomechanics in manual wheelchair propulsion. *Archives of physical medicine and rehabilitation*, 83(5), 718-723.
- Brault M. (2012). Americans with disabilities: 2010. Household economic studies. Retrieved from <http://www.census.gov/people/disability/>
- Cavedon, V., Zancanaro, C., & Milanese, C. (2015). Physique and performance of young wheelchair basketball players in relation with classification. *PloS one*, 10(11), e0143621
- Chow, J. W., Millikan, T. A., Carlton, L. G., Chae, W. S., Lim, Y. T., & Morse, M. I. (2009). Kinematic and electromyographic analysis of wheelchair propulsion on ramps of different slopes for young men with paraplegia. *Archives of physical medicine and rehabilitation*, 90(2), 271-278.

- Cooper RA, Cheda A. Measurement of Racing Wheelchair Propulsion Torque. Proceedings 11th Annual International Conference of the IEEE/EMBS; Seattle, Washington. November 9–12.1989.
- Cooper, R. A., Dvorznak, M. J., O'Connor, T. J., Boninger, M. L., & Jones, D. K. (1998). Braking electric-powered wheelchairs: Effect of braking method, seatbelt, and legrests. *Archives of Physical Medicine and Rehabilitation*, 79(10), 1244-1249.
- Cooper, R. A., & De Luigi, A. J. (2014). Adaptive sports technology and biomechanics: wheelchairs. *PM&R*, 6, S31-S39.
- Coutts, K. D. (1992). Dynamics of wheelchair basketball. *Medicine and science in sports and exercise*, 24(2), 231-234.
- Dvorznak, M. J., Cooper, R. A., O'Connor, T. J., Boninger, M. L., & Fitzgerald, S. G. (2001). Kinematic comparison of Hybrid II test dummy to wheelchair user. *Medical engineering & physics*, 23(4), 239-247.
- Fast, A., Sosner, J., Begeman, P., Thomas, M., & Drukman, D. (1997). Forces, moments, and acceleration acting on a restrained dummy during simulation of three possible accidents involving a wheelchair negotiating a curb: Comparison Between Lap Belt and Four-Point Belt1. *American journal of physical medicine & rehabilitation*, 76(5), 370-377.
- Fox, M. H., Witten, M. H., & Lullo, C. (2014). Reducing obesity among people with disabilities. *Journal of disability policy studies*, 25(3), 175-185.

- Gaal, R. P., Rebholtz, N., Hotchkiss, R. D., & Pfaelzer, P. F. (1997). Wheelchair rider injuries: causes and consequences for wheelchair design and selection. *Journal of rehabilitation research and development*, 34(1), 58.
- Gagnon, D., Verrier, M., Masani, K., Nadeau, S., Aissaoui, R., & Popovic, M. (2009). Effects of trunk impairments on manual wheelchair propulsion among individuals with a spinal cord injury: a brief overview and future challenges. *Topics in Spinal Cord Injury Rehabilitation*, 15(2), 59-70.
- Hosseini, S. M., Oyster, M. L., Kirby, R. L., Harrington, A. L., & Boninger, M. L. (2012). Manual wheelchair skills capacity predicts quality of life and community integration in persons with spinal cord injury. *Archives of physical medicine and rehabilitation*, 93(12), 2237-2243.
- Hughes, C. J., Weimar, W. H., Sheth, P. N., & Brubaker, C. E. (1992). Biomechanics of wheelchair propulsion as a function of seat position and user-to-chair interface. *Archives of physical medicine and rehabilitation*, 73(3), 263-269.
- Hwang, S., Lin, Y. S., Hogaboom, N. S., Wang, L. H., & Koontz, A. M. (2017). Relationship between linear velocity and tangential push force while turning to change the direction of the manual wheelchair. *Biomedical Engineering/Biomedizinische Technik*, 62(4), 439-44
- IWBF Player Classification Commission. (2014). *A Guide to the IWBF Functional Classification System for Wheelchair Basketball Players*.
- Janssen-Potten, Y. J., Seelen, H. A., Drukker, J., Huson, T., & Drost, M. R. (2001). The effect of seat tilting on pelvic position, balance control, and compensatory postural muscle use in paraplegic subjects. *Archives of physical medicine and rehabilitation*, 82(10), 1393-1402.

- Kamper, D., Barin, K., Parnianpour, M., Reger, S., & Weed, H. (1999). Preliminary investigation of the lateral postural stability of spinal cord-injured individuals subjected to dynamic perturbations. *Spinal Cord*, 37(1), 40.
- Kaye, H. S., Kang, T., & Laplante, M. P. (2002). Wheelchair use in the United States. In *Disability Statistics Abstract* (Vol. 23, pp. 1-4).
- Kirby, R. L., Ackroyd-Stolarz, S. A., Brown, M. G., Kirkland, S. A., & MacLeod, D. A. (1994). Wheelchair-related accidents caused by tips and falls among noninstitutionalized users of manually propelled wheelchairs in Nova Scotia. *American Journal of Physical Medicine & Rehabilitation*, 73(5), 319-330.
- Kirby, R. L., Swuste, J., Dupuis, D. J., MacLeod, D. A., & Monroe, R. (2002). The Wheelchair Skills Test: a pilot study of a new outcome measure. *Archives of Physical Medicine and Rehabilitation*, 83(1), 10-18.
- Kim SJ, Boninger ML, Koontz AM, Cooper RA, Souza AL, Wheelchair braking biomechanics, Proceedings of RESNA 2002 Annual Conference, Minneapolis, MN, 2002.
- Koontz, A. M., Cooper, R. A., Boninger, M. L., Souza, A. L., & Fay, B. T. (2002). Shoulder kinematics and kinetics during two speeds of wheelchair propulsion. *Journal of Rehabilitation Research and Development*, 39(6), 635.
- Kotajarvi, B. R., Sabick, M. B., Kai-Nan An, Zhao, K. D., & al, e. (2004). The effect of seat position on wheelchair propulsion biomechanics. *Journal of Rehabilitation Research and Development*, 41(3), 403-14.

- Kwarciak, A. M., Sisto, S. A., Yarossi, M., Price, R., Komaroff, E., & Boninger, M. L. (2009). Redefining the manual wheelchair stroke cycle: identification and impact of nonpropulsive pushrim contact. *Archives of physical medicine and rehabilitation*, 90(1), 20-26.
- Kwarciak, A. M., Turner, J. T., Guo, L., & Richter, W. M. (2012). The effects of four different stroke patterns on manual wheelchair propulsion and upper limb muscle strain. *Disability and Rehabilitation: Assistive Technology*, 7(6), 459-463.
- Lalumiere, M., Gagnon, D. H., Hassan, J., Desroches, G., Zory, R., & Pradon, D. (2013). Ascending curbs of progressively higher height increases forward trunk flexion along with upper extremity mechanical and muscular demands in manual wheelchair users with a spinal cord injury. *Journal of Electromyography and Kinesiology*, 23(6), 1434-1445.
- Levinson, L. J., & Reid, G. (1991). Patterns of physical activity among youngsters with developmental disabilities. *CAHPER Journal*, 57(3), 24-28.
- Masse, L. C., Lamontagne, M., & O'riain, M. D. (1992). Biomechanical analysis of wheelchair propulsion for various seating positions. *Journal of rehabilitation research and development*, 29(3), 12-28.
- Maynard, F. M., Bracken, M. B., Creasey, G., Ditunno Jr, J. F., Donovan, W. H., Ducker, T. B., ... & Waters, R. L. (1997). International standards for neurological and functional classification of spinal cord injury. *Spinal cord*, 35(5), 266-274.
- Morrow, M. M., Hurd, W. J., Kaufman, K. R., & An, K. N. (2010). Shoulder demands in manual wheelchair users across a spectrum of activities. *Journal of Electromyography and Kinesiology*, 20(1), 61-67.

- National Spinal Cord Injury Statistical Center. (2018). Spinal cord injury facts and figures at a glance: 2018 SCI data sheet. Birmingham, AL: University of Alabama at Birmingham.
- Newsam, C. J., Rao, S. S., Mulroy, S. J., Gronley, J. K., Bontrager, E. L., & Perry, J. (1999). Three dimensional upper extremity motion during manual wheelchair propulsion in men with different levels of spinal cord injury. *Gait & posture*, *10*(3), 223-232.
- Northrup, H. & Volcik, K. A. (2000). Spina bifida and other neural tube defects. *Current Problems in Pediatrics*, *30*, 317-332.
- Orendurff, M. S., Schoen, J. A., Bernatz, G. C., & Segal, A. D. (2008). How humans walk: bout duration, steps per bout, and rest duration. *Journal of rehabilitation research and development*, *45*(7), 1077.
- Phang, S. H., Martin Ginis, K. A., Routhier, F., & Lemay, V. (2012). The role of self-efficacy in the wheelchair skills-physical activity relationship among manual wheelchair users with spinal cord injury. *Disability and rehabilitation*, *34*(8), 625-632.
- Pijnappels, M., Reeves, N. D., & van Dieën, J. H. (2008). Identification of elderly fallers by muscle strength measures. *European journal of applied physiology*, *102*(5), 585-592.
- Richter, W. M., Kwarciak, A. M., Guo, L., & Turner, J. T. (2011). Effects of single-variable biofeedback on wheelchair handrim biomechanics. *Archives of Physical Medicine and Rehabilitation*, *92*(4), 572-577.
- Rodgers, M. M., Keyser, R. E., Gardner, E. R., Russell, P. J., & Gorman, P. H. (2000). Influence of trunk flexion on biomechanics of wheelchair propulsion. *Journal of rehabilitation research and development*, *37*(3), 283.

- Sanderson, D. J., & Sommer Iii, H. J. (1985). Kinematic features of wheelchair propulsion. *Journal of biomechanics*, 18(6), 423-429.
- Sauret, C., de Saint Rémy, N., Vaslin, P., Cid, M., Dabonneville, M., & Kauffmann, P. (2006). Theoretical comparison of the resultant braking force applied on a manual wheelchair within a propulsion cycle on the field and on laboratory ergometers. *Model Meas Control C*, 67, 43-52.
- Schantz, P., Björkman, P., Sandberg, M., & Andersson, E. (1999). Movement and muscle activity pattern in wheelchair ambulation by persons with para-and tetraplegia. *Scandinavian journal of rehabilitation medicine*, 31(2), 67-76.
- Seelen, H. A. M., Potten, Y. J. M., Drukker, J., Reulen, J. P. H., & Pons, C. (1998). Development of new muscle synergies in postural control in spinal cord injured subjects. *Journal of Electromyography and Kinesiology*, 8(1), 23-34.
- Shields, N., Synnot, A. J., & Barr, M. (2011). Perceived barriers and facilitators to physical activity for children with disability: a systematic review. *Br J Sports Med*, bjsports-2011.
- Simoneau, G. G., & Krebs, D. E. (2000). Whole body momentum during gait: A preliminary study of non-fallers and frequent fallers. *Journal of Applied Biomechanics*.
- Slavens, B. A., Schnorenberg, A. J., Aurit, C. M., Tarima, S., Vogel, L. C., & Harris, G. F. (2015). Biomechanics of pediatric manual wheelchair mobility. *Frontiers in bioengineering and biotechnology*, 3, 137.
- Sonenblum, S. E., Sprigle, S., & Lopez, R. A. (2012). Manual wheelchair use: bouts of mobility in everyday life. *Rehabilitation research and practice*, 2012.



Sporner, M. L., Grindle, G. G., Kelleher, A., Teodorski, E. E., Cooper, R., & Cooper, R. A. (2009).

Quantification of activity during wheelchair basketball and rugby at the National Veterans

Wheelchair Games: A pilot study. *Prosthetics and orthotics international*, 33(3), 210-217.

Tordi, N., Gimenez, M., Predine, E., & Rouillon, J. D. (1998). Effects of an interval training

programme of the upper limbs on a wheelchair ergometer in able-bodied subjects.

*International journal of sports medicine*, 19(06), 408-414.

Van der Woude, L. H., Veeger, D. J., Rozendal, R. H., & Sargeant, T. J. (1989). Seat height in

handrim wheelchair propulsion. *J Rehabil Res Dev*, 26(4), 31-50.

Vanlandewijck, Y. C., Verellen, J., & Tweedy, S. (2011). Towards evidence-based classification in

wheelchair sports: Impact of seating position on wheelchair acceleration. *Journal of Sports*

*Sciences*, 29(10), 1089-1096.

Veeger, H. E., & Rozendal, R. H. (1991). Within-cycle characteristics of the wheelchair push in

sprinting on a wheelchair ergometer. *Medicine and science in sports and exercise*, 23(2),

264-271.

Wilkenfeld, A. J., Audu, M. L., & Triolo, R. J. (2006). Feasibility of functional electrical stimulation

for control of seated posture after spinal cord injury: A simulation study. *Journal of*

*Rehabilitation Research & Development*, 43(2).

World Health Organization (WHO). (2010). Fact sheet on wheelchairs. Retrieved from:

[http://www.searo.who.int/entity/disabilities\\_injury\\_rehabilitation/wheelchair\\_factsheet.pdf](http://www.searo.who.int/entity/disabilities_injury_rehabilitation/wheelchair_factsheet.pdf)

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Yang, Y. S., Koontz, A. M., Triolo, R. J., Mercer, J. L., & Boninger, M. L. (2006). Surface electromyography activity of trunk muscles during wheelchair propulsion. *Clinical Biomechanics*, 21(10), 1032-1041.

## Chapter 2

### **The effects of trunk function on volume of action in manual wheelchair users**

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## **ABSTRACT**

### **Objective**

The aims of this study were (1) quantify the Volume of Action of wheelchair basketball players across a functional spectrum (2) determine if seated limits of stability are consistent with the functional classification system used by the International Wheelchair Basketball Federation.

### **Design**

Eight manual wheelchair users completed a seated limits of stability test to determine maximum range of motion of their trunk forward, backward, right, left, and rotation of the trunk.

### **Results**

The low function group had significantly less trunk segment flexion than the high group with a mean difference (MD) = -36.50, 95% confidence interval (CI) = -52.34 to -20.17,  $p < 0.0001$ . There were no significant differences between groups for trunk segment extension angle MD = 9.28 degrees, CI: -21.12 to 39.67,  $p = 0.54$ . The low function group also had significantly less abdominal joint ROM than the high function group, MD = 8.41. 95% CI = 0.71 to 16.12 degrees. The high function group exhibited significantly more trunk segment lateral ROM than the low function group with a MD = 55.16 degrees, 95% CI: 22.51 to 87.81. The high function group exhibited significantly more trunk segment rotational ROM than the low function group with a MD = 44.47 degrees, 95% CI = 28.68 to 28.68. The high function group exhibited significantly more abdominal joint ROM than the low function group with a MD = 38.51 degrees, 95% CI = 27.31 to 49.72.

### **Conclusions**

Wheelchair basketball players in lower functional classification groups have a lower capacity to control their trunk excursion in lateral, anterior, and posterior directions, as well as rotation. These results lend support to the validity of the functional classification system used in wheelchair basketball.

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## Introduction

The ability to maintain seated balance is an important skill for wheelchair users in the execution of activities of daily living. These include stationary activities where the wheelchair is not in motion (reaching to retrieve an object, grooming, bathing) and dynamic activities such as propelling the wheelchair. For example, when pushing up a ramp a wheelchair user will lean their trunk as far forward as possible to prevent the wheelchair from falling backward, and will lean their trunk back to prevent falling forward when descending a ramp (Sisto, Druin, & Sliwinski, 2008). Additionally, falls during the execution of functional activities can be a problem for this population (Nelson et al., 2003).

When comparing groups of wheelchair users there is a need to differentiate between functional capacity, and one way of doing that is by using trunk function. This can be done by using spinal cord lesion level (Kulig et al., 2001) or stratifying subjects by paraplegia and tetraplegia (van Drongelen et al., 2005). While spinal cord injury lesion level provides a simple way of stratifying wheelchair users, not all wheelchair users have a spinal cord injury. There remains a need for a framework to examine wheelchair users based on functional ability regardless of mechanism of injury. The International Wheelchair Basketball Federation (IWBF) Functional Classification system is used to stratify wheelchair players based on the functional reach of their trunk, among other sport specific measures (IWBF, 2014). This system functions using a “Volume of Action” framework, based on the idea that the higher the classification, the greater the volume of action and by extension, more function. This system has strong

correlations with field performance tests (Vanlandewijck, Spaepen, & Lysens, 1995; Vanlandewijck et al., 2003). On average, players with higher classifications exhibit higher power output and  $VO^2$  peak (De Lira et al., 2010), and score higher on the Comprehensive Basketball Grading System (Vanlandewijck et al., 2004). However, very little work has been done to examine this system with respect to trunk range of motion.

Wheelchair basketball players are assigned a classification after being observed during sanctioned game play by a panel of three individuals called classifiers. Classifiers receive extensive training and observe trunk movement of players during game play, including during the execution of skills such as dribbling, passing, shooting, and rebounding. Players are given a classification, called points, ranging from 1.0 (minimal function) to 4.5 (maximal function) based on a consensus of the classification panel. Half points are given if "the player functionally blends characteristics, specific criteria or the volume of action of two classes" (Courbariaux, 1996). The panel provides for objectivity, in that there must be a consensus between all three classifiers before a classification is awarded (IWBF, 2014). An overview of the functional capacities for each classification can be seen in table 1 and figure 2. Generally speaking, a class I is identified by an inability to rotate their trunk along the Z axis. A class II player has the functional capacity to rotate their trunk but lacks the functional capacity to exhibit active hip flexion and extension. A class III player has the ability to execute the aforementioned movements of the trunk, but lacks the functional capacity to actively move their trunk laterally to the left or right and return to an upright position. Finally, a class IV player has the functional capacity to move their trunk

in all planes of motion (rotation, flexion/extension, and lateral movement). ½ points are given to players who exhibit some but not all of these movements (for example, a player may be classified as a 3.5 if they are able to control the movement of their trunk to one side laterally, but not the other).

There have been very few attempts at quantifying the Volume of Action of wheelchair basketball players. Santos et al., (2016) used a limits of stability test to quantify trunk balance and found that trunk excursion increased progressively with classification. However, they were unable to measure trunk function with respect to rotation, which is a major aspect of the classification process and an important skill for a wheelchair user in the execution of activities of daily living. In a study comparing people with disabilities to able-bodied individuals, Rehm (2015) determined that functional limitations play a large role in the differences seen between classifications with respect to volume of action.

A Volume of Action framework has practical applications outside the world of wheelchair sports, and the movements in the present investigation are consistent with activities of daily living for wheelchair users (picking an object up off the ground, leaning to reach into a cabinet, rotating to reach the back of a wheelchair). As such, the Volume of Action framework is applicable to a myriad of situations. In a longitudinal study examining the role of muscle synergy in postural control of subjects with spinal cord injury, Seelen et al. (1998) found that subjects developed unique muscle activation patterns to maintain balance during bimanual tasks. Specifically, increased use of the latissimus dorsi and trapezius muscle was seen as a

compensatory strategy to maintain balance. In a study examining lateral perturbations in spinal cord-injured subjects, Kamper et al. (1999) found that the ability to perform static leaning was strongly correlated to dynamic lateral balance.

Therefore, the purpose of this study was to (1) quantify the Volume of Action of wheelchair basketball players across a functional spectrum (2) determine if seated limits of stability are consistent with the functional classification system used by the National Wheelchair Basketball Association.

## **MATERIALS AND METHODS**

### **Participants**

Eight wheelchair users (4 men, 4 women) participated in this study. Mean age of wheelchair users was  $24.75 \pm 7.57$  years, and mean mass was  $65.10 \pm 14.15$  kg. Functional Classifications ranged from 1.0 to 3.5. Self-reported shoulder pain or injury within the last six months was considered an exclusion criteria. Use of a wheelchair as primary means of locomotion were required to participate in this study. Subjects were divided into two subgroups: high functional classification (HFC), in which the participant's functional classification was above 3.0 or above, and low functional classification (LFC), in which the participant's functional classification was 2.5 or below (Wilson et al., 2018). All participants provided written informed consent, and all procedures were approved by the Institutional Review Board of the University of Texas at Arlington.

### **Experimental Procedures**



Data collection took place during a single visit to the Applied Biomechanics Laboratory at the University of Texas at Arlington. Participants were asked to seat themselves on an adjustable seated platform (Per4max, Grand Prairie, TX) and place a strap around their waist for safety (figure 1). A 14 segment full-body marker set with 6 DoF joints was used to model the body. Reflective markers (14 mm) were attached bilaterally to the skin over anatomical landmarks. Acromion process (RAC, LAC), joint center of the shoulder complex (RADL, RPDL, LADL, LPDL), neck in line with C7 (RNECK, LNECK), C7, T8, T2, L1, L3, L5 vertebrae, superior most point of iliac crest in the sagittal plane (RPP, LPP), anterior superior iliac spine (RAS, LAS), posterior superior iliac spine (RPS, LPS), greater trochanters (RHP, LHP), medial and lateral epicondyles of the femur (RMK, RLK, LMK, LLK), medial and lateral epicondyle of the humerus (RMEL, RLEL, LLEL, LMEL), radial and ulnar epicondyles (RWRR, RWRU, LWRR, LWRU), second, third, and fifth metacarpals (LHR, LHM, LHU, RHR, RHM, RHU) medial and lateral malleoli (RMA, RLA, LMA, LLA), first metatarsal, base and fifth of the metatarsals. Markers were also placed on the top of the head (THEAD), forehead (AHEAD), occipital bone (PHEAD), zygomatic bone (RHEAD, LHEAD). Non-collinear markers on molded thermo-plastic shells were placed on the posterior thorax, upper arms, forearms, proximal thighs, and distal shanks. Three tracking markers were placed on the medial, lateral, and posterior heel. A Vicon T-Series motion capture system (Vicon Motion Systems Ltd., Denver, CO) with sixteen MX T40S cameras (4 MP resolution 2336 x 1728) was used to track the position of the markers at 100 Hz. A static trial was then recorded. All anatomical markers were then removed for the limits of stability trials. During data collection, it became necessary to digitally define the RAS and LAS markers. A

spring-loaded digitizing pointer (C-Motion, Germantown, MD, USA) was used to create digital markers to be used when adipose tissue occluded the physical markers, or when the markers became occluded due to changes in position. The tip of the digitizing pointer was placed on the soft tissue directly over the anterior superior iliac spine, after which the clinician depressed the digitizing pointer until it reached the underlying bone (Lerner et al., 2014). Prior to limits of stability trials, subjects had the opportunity to practice each motion they would be asked to complete (trunk flexion/extension, left and right lateral extension, and trunk rotation). In the limits of stability trials participants completed ten trials in each direction where they were required to flex and extend the trunk as far as possible without falling. Trials were disqualified if the participant lost their balance, and the trial was repeated.

### **Data Analysis**

Visual 3D (C-Motion, Germantown, MD, USA) was used to process three-dimensional kinematic data for each participant. Marker trajectories were filtered with a fourth order zero lag Butterworth low-pass filter with cutoff frequency of 6 Hz. Body segment parameters (mass, center of mass location) were obtained using de Leva (1996). Hip joint center locations were obtained using Bennett (2016). Three-dimensional joint angles were calculated using an x (flexion/extension), y (abduction/adduction), z (axial rotation) Cardan rotation sequence. The limits of stability were quantified using the orientation of the trunk relative to the laboratory reference frame (trunk segment angle) and the abdominal joint angle (angle between the pelvis local coordinate system and the trunk local coordinate system).

### **Statistical Analysis**

Separate SAS version 9.4 proc GLIMMIX linear mixed effects models were used to compare differences in each dependent variable (trunk segment angles and abdominal joint angles) with functional capacity as a fixed effect grouping factor (high, low), subjects as a random factor and trials as a covariate. Separate intercepts were fit for each subject using an unstructured variance-covariance matrix to account for the correlations between trials. A significant main effect for group was followed by post hoc analysis with Tukey correction for multiple comparisons between groups with alpha set at 0.05.

## **Results**

### *Trunk Flexion/Extension and Anterior/Posterior Limits of Stability*

Means and 95% confidence intervals for anterior/posterior limits of stability can be seen in upper left and upper right of Figure 3. The low function group had significantly less trunk segment flexion than the high group with a mean difference (MD) = -36.50, 95% confidence interval (CI) = -52.34 to -20.17,  $p < 0.0001$ , Cohen's  $d = 0.706$ . The trunk flexion angle represents the anterior limits of stability for a seated individual. The high classification group had a mean  $\pm$  SE (95% CI) anterior limit of stability of  $71.87 \pm 5.78$  degrees (60.33 to 83.43). In the low function group the mean anterior limit of stability was  $35.37 \pm 5.78$  degrees (23.82 to 46.92).

There were no significant differences between groups for trunk segment extension angle MD = 9.28 degrees, CI: -21.12 to 39.67,  $p = 0.54$ , Cohen's  $d = 0.096$ . The high function group posterior limit of stability occurred with a mean  $\pm$  SE (95% CI) trunk segment extension

angle of  $30.56 \pm 10.75$ , (9.07 to 52.05 degrees). The low functional classification group posterior limit of stability occurred with a mean  $\pm$  SE (95% CI) of  $21.29 \pm 10.75$  (-0.21 to +42.77 degrees). Trunk segment flexion-extension range of motion (ROM) was significantly different with MD =  $45.78 \pm 16.80$  degrees, CI: 12.21 to 79.34,  $p = 0.0083$ , Cohen's  $d = .431$ . Subjects in the high function group had a flexion-extension ROM of  $102.44 \pm 16.80$  degrees, CI: 78.70 to 126.17. Subjects in the low function group a flexion-extension ROM of  $56.66 \pm 11.88$  degrees, CI: 32.92 to 80.39.

Abdominal joint flexion-extension ROM was also significantly different with a MD of  $23.74 \pm 9.77$  degrees, CI: 4.21 to 43.28,  $p = 0.018$ , Cohen's  $d = .384$ . Low functioning subjects had a mean flexion – extension abdominal joint ROM of  $19.02 \pm 6.91$  degrees, CI: 5.21 to 32.83, and whereas high functioning subjects had a mean flexion – extension abdominal joint ROM of  $42.77 \pm 6.91$  degrees, CI: 28.95 to 56.58.

#### *Lateral ROM for Trunk Segment Angle and Abdominal Joint Angle*

Means and 95% confidence intervals for lateral range of motion can be seen in lower left of Figure 3. The high function group exhibited significantly more trunk segment lateral ROM than the low function group with a MD = 55.16 degrees, 95% CI: 22.51 to 87.81,  $p = 0.0013$ , Cohen's  $d = .534$ . In the high function group the mean  $\pm$  SE (95% CI) trunk segment flexion ROM was  $105.42 \pm 11.55$  degrees (82.33 to 128.50). For the low functional classification group, mean  $\pm$  SE (95% CI) trunk segment ROM was  $50.26 \pm 11.55$  degrees (27.17 to 73.35).

Furthermore, there was a significant difference in abdominal joint lateral range of motion with a MD of  $8.41 \pm 3.85$  degrees,  $p = .033$ , 95% CI: 0.71 to 16.11, Cohen's  $d = 0.345$ .

The mean  $\pm$  SE (95% CI) abdominal joint ROM for high functional was  $48.47 \pm 8.47$  degrees (31.54 to 65.39) compared to  $40.06 \pm 8.44$  degrees (23.18 to 56.94) in the low function group (Figure 4, left).

#### *Rotational ROM for Trunk Segment Angle and Abdominal Joint Angle*

Means and 95% CI for longitudinal rotation of the trunk segment about the Z axis can be seen in bottom right Figure 3. The high function group exhibited significantly more trunk segment rotational ROM than the low function group with a MD = 44.47 degrees, 95% CI: 28.68 to 28.68,  $p < 0.0001$ , Cohen's  $d = .890$ . The high functional classification group trunk segment angle had a mean  $\pm$  SE (95% CI) ROM of  $114.24 \pm 8.54$  degrees (97.18 to 131.31) compared to  $69.78 \pm 8.45$  degrees (52.89 to 86.65) of trunk segment rotation for the low classification group.

Means and 95% CI for longitudinal rotation of the abdominal joint about the Z axis can be seen on the left of Figure 4. The high function group exhibited significantly more abdominal joint rotational ROM than the low function group with a MD = 38.51 degrees, 95% CI = 27.31 to 49.72,  $p = 0.0001$ , Cohen's  $d = 1.09$ . The high functional classification group abdominal joint angle had a mean  $\pm$  SE (95% CI) ROM of  $93.62 \pm 6.35$  degrees (80.92 to 106.32) compared to  $55.10 \pm 6.29$  (42.53 to 67.68) for the low functional classification group.

#### **Discussion**

The aims of this study were (1) quantify the Volume of Action of wheelchair basketball players across a functional spectrum (2) determine if seated limits of stability are consistent with the functional classification system used by the International Wheelchair Basketball Federation. The results of this study indicate that static limits of stability are significantly

different across a functional spectrum using the functional classification system employed in wheelchair basketball. Limits of stability via lateral trunk flexion to the left and right, rotation of the trunk along the Z axis, and flexion in the sagittal plane was lower in class 2.5 and below than it was in class 3.0 and above. These results lend further support to the validity of the functional classification system used in wheelchair basketball. Importantly, this investigation was among the first to observe rotation of the trunk, which is an integral part of the functional classification system.

The Volume of Action framework is the foundation of the functional classification system and is defined as *“the limit to which a player can move voluntarily in any direction, and with control return to the upright seated position, without holding the wheelchair for support or using the upper extremities to aid the movement”* (IWBF, 2014). As important as this concept is to both the sport of wheelchair basketball and independence in wheelchair users, very little work has been done to examine or quantify it. Indeed, the concept of limits of stability is particularly appropriate in the observation of volume of action, as it requires that the subject reach the limits of their ability to maintain their posture, then actively return to the upright position. Santos et al. (2016) employed a modified limits of stability test using a Neurocom Balance Master to determine the maximum distance traveled by the center of gravity and found significant differences between classification groups for trunk flexion/extension and left and right lateral flexion. Importantly, they were unable to examine rotation of the trunk, which is the key differentiator between class I and class II individuals in this system (IWBF, 2014). The ability to rotate the trunk increases the ability of wheelchair users to execute numerous

functional and contributes to overall stability (IWBF, 2014), and as such is an accurate indicator of overall function within this functional classification system. In the present study high functional classified subjects were able to rotate the trunk about the longitudinal axis with a ROM of 114.24 degrees (95% CI of 97.18 to 131.31). In contrast, low functional classified individuals had significantly less longitudinal trunk ROM rotation (69.78 degrees, 95% CI of 52.90 to 86.66). Contrary to the results of Santos et al., we did not find a significant difference between groups for trunk extension. This may have been a result of a small sample size, as well as an absence of players with a classification of 4.0 or 4.5.

The requirement that subjects return to the upright position is a key feature of this research, and to the concept of functional capacity. (Serra-Ano et al., 2013). The ability to flex and extend the trunk is the primary identifier of class III and above, and differentiates players between class II and class III. Individuals who are capable of this action are able to generate more power in their push (Howarth et al., 2010; Sanderson and Sommer, 1985) and also more competitive in other aspects of the game, such as rebounding or retrieving a ball from the ground. Additionally, this is applicable to activities of daily living, such as functional reach or stabilizing the trunk during ascending or descending a ramp. However, the key requirement is that the movement be active trunk flexion, as opposed to passive flexion due to gravity. If a player requires the use of their hands to push themselves back up into the upright position, they are not considered to demonstrate the appropriate level of function needed for a class III or above. This study sought to mimic that requirement by requiring participants to flex their

trunk to the point at which they felt they would lose their balance and then return to an upright position.

Differences in the lateral limits of stability between the high and low function groups have real world implications when performing activities of daily living. The high function group in our study had 8.96 degrees more lateral range of motion in the trunk segment and 8.41 degrees of lateral abdominal joint range of motion when compared to the low function individuals. The high function individuals were able to laterally tilt the pelvis and laterally tilt the trunk further than the low function group. These differences in lateral limits of stability greatly impact lateral balance when bending to the side to grasp an object, placing the low function individual at great risk of fall in the lateral direction.

This study required participants to sit on a flat surface, which removed any potential passive stability that may be provided by the wheelchair during competition (IWBF, 2014). While this allowed us to examine functional capacity without having to elucidate the role of the wheelchair in providing stability, it does remove the wheelchair user from the system in which they operate on a day to day basis. Wheelchair configuration has been shown to influence the user's ability to stabilize themselves (Trudel et al., 1997; Thomas et al., 2017). Curtis et al., (1995) found that wheelchair users who used a strap to stabilize their trunk and lower body significantly increased their limits of stability when going through the motions that wheelchair basketball classifiers look for during competition. Future work should examine the role of wheelchair configuration in the manifestation of the Volume of Action framework as it applies to both wheelchair athletes and manual wheelchair users in general. Potential practical



applications may exist for this framework in examining functional differences in activities of daily living, or important injury prevention techniques such as the wheelchair push-up to prevent pressure ulcers (Van Drongelen et al., 2005).

### **Limitations**

There are some potential limitations to this study. First, this study used wheelchair basketball functional classification as a means of stratifying subjects into functional groups, but all data was collected in a laboratory setting. This is potentially significant, as classifications are given to players only after observation during competition. While there is evidence to suggest that the functional classification system reasonably predicts functional capacity (Yanci et al., 2015; Gil et al., 2015; Molik et al., 2013), it is unclear what role sport wheelchair set up plays in the demonstration of functional capacity. Second, similar to other researchers (Wilson et al., 2018; Fliess-Douer et al., 2003; Marszałek et al., 2019), we divided our subjects into two groups due to a small sample size in each of the wheelchair basketball classifications. While this is a common practice in the literature, an appropriate sample size of each different classification may have provided more clarity in the role of trunk function in the limits of stability, as well as further differentiated between each classification (class I versus class II, for example). Third, we did not have any subjects who were classified as 4.0 or 4.5. These two classifications are the highest functional classifications, and typically consist of amputees or semi-ambulatory individuals who have full trunk function but have lower limb impairments that preclude them from participating in able-bodied sports. It is expected that inclusion of individuals with these two classifications would have further increased the differences seen between the two groups.

Inclusion of these classifications and increasing the sample size such that each classification is able to be grouped together (as opposed to two separate groups of “high classification” and “low classification”) may further elucidate the role of active vs passive pelvic stabilization strategies for both sport purposes and activities of daily living. Finally, as mentioned, there was no accounting for limb deficits that may have influenced classifications. Future studies should work with classification professionals to differentiate subjects into groups based solely on trunk function in order to elucidate functional differences of the trunk in the wheelchair-using population.

### **Conclusions**

In conclusion, the results of this study indicate that seated limits of stability differ significantly between high functioning and low functioning wheelchair basketball players, and that these differences are consistent with the Volume of Action concept that is the basis of the wheelchair basketball functional classification system. Wheelchair users in higher classifications (3.0 and above) exhibited greater volume of action in all planes of movement. Taken together, these results indicate that the functional classification system used by the IWBF and NWBA objectively stratifies wheelchair users into classifications based on trunk function, and that a volume of action framework can be used to stratify wheelchair users to examine functional differences in movement.

## Figure Legends

**Figure 1:** A 14 segment full-body marker set with 6 DoF joints was used to model the body. Reflective markers (14 mm) were attached bilaterally to the skin over anatomical landmarks.

**Figure 2:** Volume of Action framework.

**Figure 3:** Mean and 95% confidence interval of trunk segment extension angle (upper left), trunk segment flexion angle (upper right), trunk segment lateral range of motion (bottom left), longitudinal axis trunk segment range of motion (lower right) by trunk functional capacity groups.

**Figure 4:** Mean and 95% confidence interval of abdominal joint lateral range of motion (left) and abdominal joint longitudinal axis range of motion (right) by trunk functional capacity groups.

**Table 1: IWBF Classification Based on Function**

Classification	Function
1.0	Little or no controlled trunk movement in the forward plane No active trunk rotation Balance in both forward and sideways directions is significantly impaired Players rely on their arms to return to the upright position when unbalanced
2.0	Partially controlled trunk movement in the forward plane Active upper trunk rotation but no lower trunk function No controlled sideways movement
3.0	Good trunk movement in the forward direction Good trunk rotation No controlled trunk movement sideways
4.0	Normal trunk movements, but usually due to limitations in one lower limb the player has difficulty with controlled movement to one side
4.5	Normal trunk movement in all directions Able to reach side to side with no limitations

<b>Table 2.</b> Trunk excursions in high (HFC) and low functional classifications (LFC) of wheelchair basketball players		
Variable (degrees)	HFC	LFC
Flexion <sup>a</sup>	71.87 ± 5.78	35.37 ± 5.78
Extension	30.56 ± 10.75	21.29 ± 10.75
Trunk segment lateral ROM	82.45 ± 11.83	73.49 ± 11.80
Abdominal joint lateral ROM <sup>a</sup>	48.47 ± 8.47	40.06 ± 8.44
Trunk segment rotation ROM <sup>a</sup>	19.20 ± 3.54	15.79 ± 3.50
Abdominal joint rotation ROM <sup>a</sup>	93.62 ± 6.35	55.10 ± 6.29
Data are presented as Least Squares Means ± SE <sup>a</sup> Significant differences (p < .05) between groups		

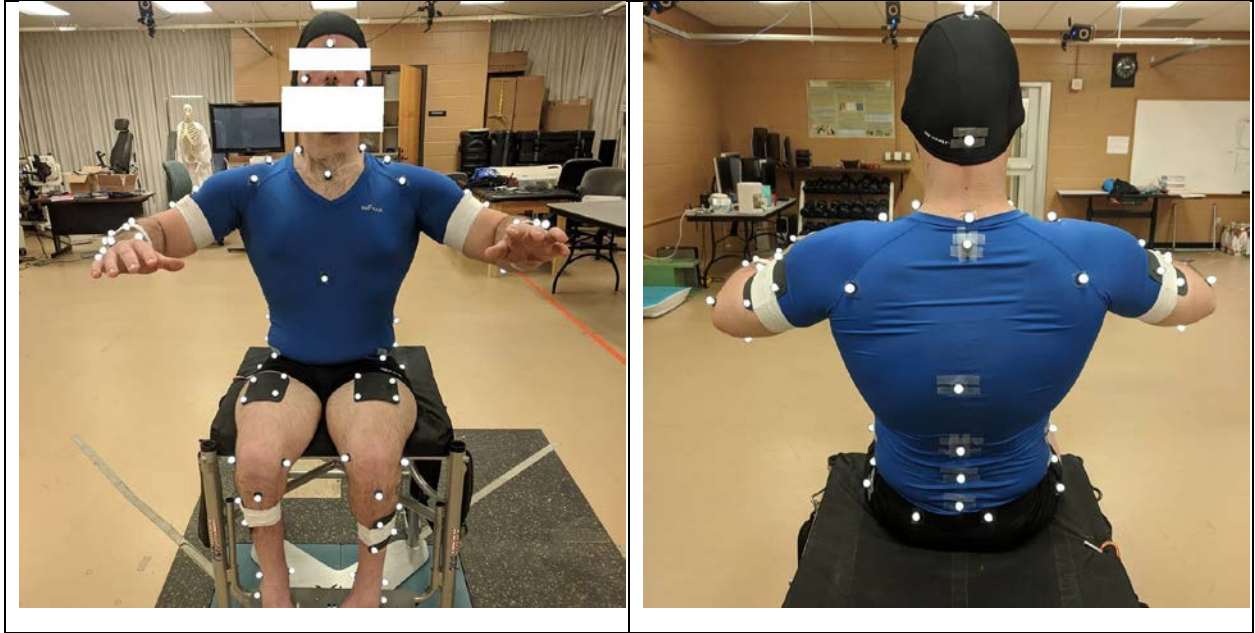
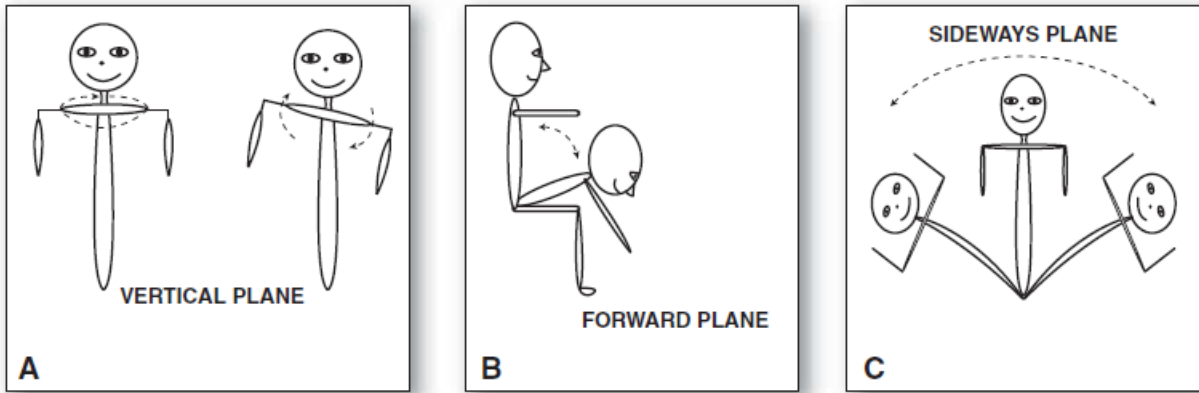
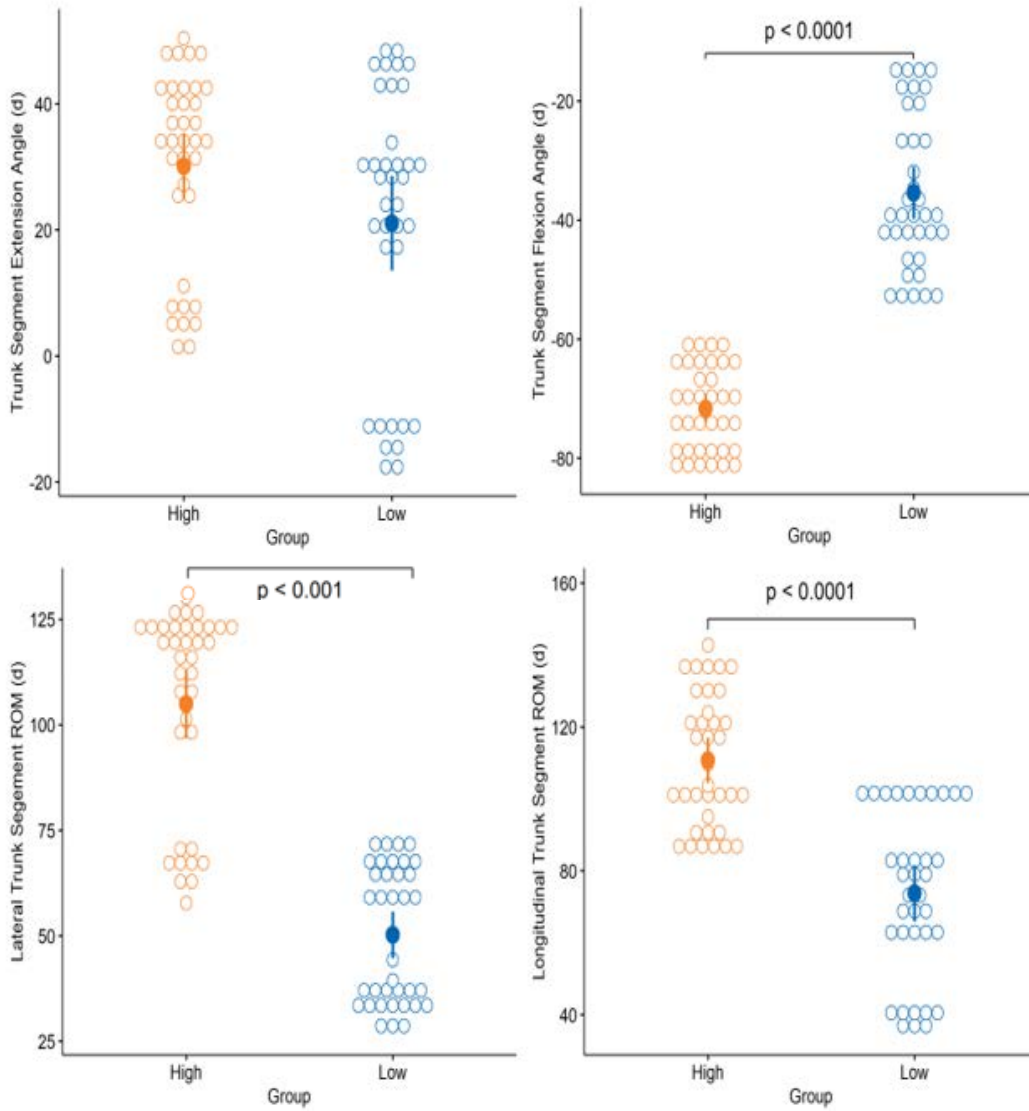


Figure 1.



**Figure 2.** Volume of Action, adapted from IWBF (2014).



**Figure 3.**



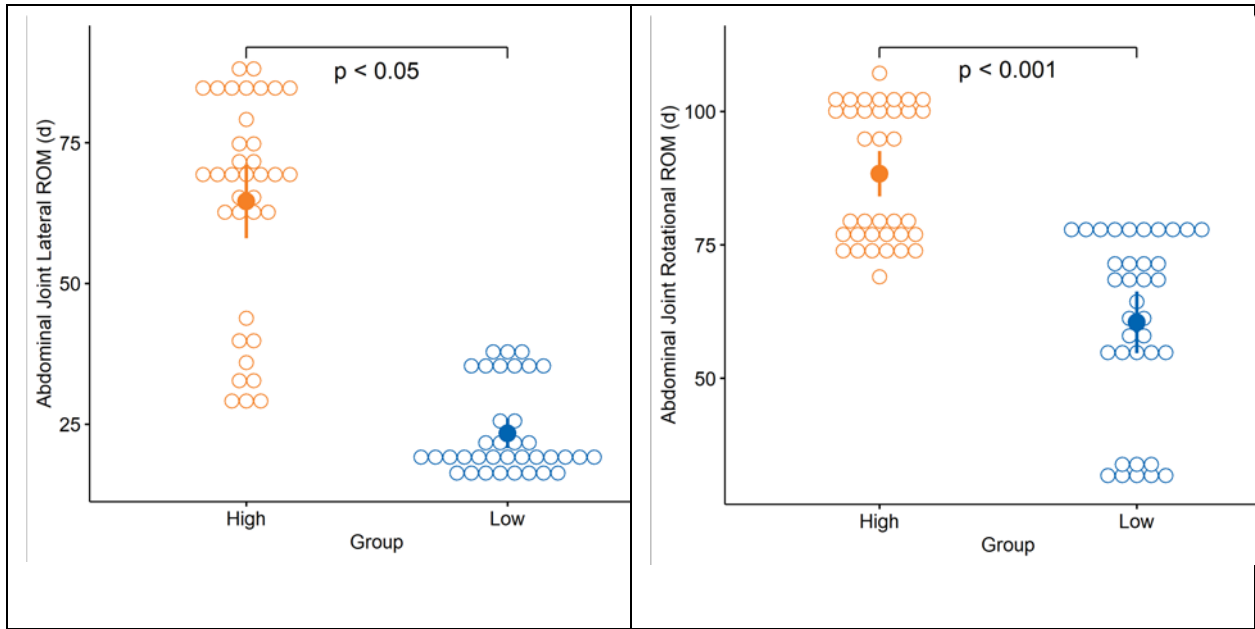


Figure 4

## References

- Bennett, H. J., Shen, G., Weinhandl, J. T., & Zhang, S. (2016). Validation of the greater trochanter method with radiographic measurements of frontal plane hip joint centers and knee mechanical axis angles and two other hip joint center methods. *Journal of Biomechanics*, 49(13), 3047-3051.
- Courbariaux, B. (1996). The classification system for wheelchair basketball players. New York: IWBF.
- Curtis, K. A., Kindlin, C. M., Reich, K. M., & White, D. E. (1995). Functional reach in wheelchair users: the effects of trunk and lower extremity stabilization. *Archives of physical medicine and rehabilitation*, 76(4), 360-367.
- de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics*. 29: 1223-1230, 1996.
- De Lira, C. A. B., Vancini, R. L., Minozzo, F. C., Sousa, B. S., Dubas, J. P., Andrade, M. S., ... & Da Silva, A. C. (2010). Relationship between aerobic and anaerobic parameters and functional classification in wheelchair basketball players. *Scandinavian Journal of Medicine & Science in Sports*, 20(4), 638-643.
- Fliess-Douer, O., Hutzler, Y., & Vanlandewijck, Y. C. (2003). Relation of functional physical impairment and goal perspectives of wheelchair basketball players. *Perceptual and motor skills*, 96(3), 755-758.

- Gil, S. M., Yanci, J., Otero, M., Olasagasti, J., Badiola, A., Bidaurrezaga-Letona, I., ... & Granados, C. (2015). The functional classification and field test performance in wheelchair basketball players. *Journal of human kinetics*, 46(1), 219-230.
- Howarth, S. J., Polgar, J. M., Dickerson, C. R., & Callaghan, J. P. (2010). Trunk muscle activity during wheelchair ramp ascent and the influence of a geared wheel on the demands of postural control. *Archives of physical medicine and rehabilitation*, 91(3), 436-442.
- IWBF Player Classification Commission. (2014). *A Guide to the IWBF Functional Classification System for Wheelchair Basketball Players*.
- Kamper, D., Barin, K., Parnianpour, M., Reger, S., & Weed, H. (1999). Preliminary investigation of the lateral postural stability of spinal cord-injured individuals subjected to dynamic perturbations. *Spinal Cord*, 37(1), 40.
- Kulig, K., Newsam, C. J., Mulroy, S. J., Rao, S., Gronley, J. K., Bontrager, E. L., & Perry, J. (2001). The effect of level of spinal cord injury on shoulder joint kinetics during manual wheelchair propulsion. *Clinical biomechanics*, 16(9), 744-751.
- Lerner, Z. F., Board, W. J., & Browning, R. C. (2014). Effects of an obesity-specific marker set on estimated muscle and joint forces in walking. *Med Sci Sports Exerc*, 46(6), 1261-1267.
- Marszałek, J., Gryko, K., Kosmol, A., Morgulec-Adamowicz, N., Mróz, A., & Molik, B. (2019). Wheelchair Basketball Competition Heart Rate Profile According to Players' Functional Classification, Tournament Level, Game Type, Game Quarter and Playing Time. *Frontiers in psychology*, 10.

- Molik, B., Laskin, J. J., Kosmol, A., Marszałek, J., Morgulec-Adamowicz, N., & Frick, T. (2013). Relationships between anaerobic performance, field tests, and functional level of elite female wheelchair basketball athletes. *Human Movement, 14*(4), 366-371.
- Nelson, A., Ahmed, S., Harrow, J., Fitzgerald, S., Sanchez-Anguiano, A., & Gavin-Dreschnack, D. (2003). Fall-related fractures in persons with spinal cord impairment: a descriptive analysis. *SCI nursing: a publication of the American Association of Spinal Cord Injury Nurses, 20*(1), 30-37.
- Rehm, J. (2015). *Measuring Trunk Stability and Range of Motion: Two Field Tests for Wheelchair Basketball Classification* (Doctoral dissertation).
- Sanderson, D. J., & Sommer III, H. J. (1985). Kinematic features of wheelchair propulsion. *Journal of biomechanics, 18*(6), 423-429.
- da Silva Santos, S., Krishnan, C., Alonso, A. C., & Greve, J. M. D. A. (2017). Trunk function correlates positively with wheelchair basketball player classification. *American journal of physical medicine & rehabilitation, 96*(2), 101-108.
- Santos, S. D. S., Alonso, A. C., & Greve, J. M. D. A. (2016). Quantitative evaluation of trunk muscle strength in wheelchair basketball players. *Motriz: Revista de Educação Física, 22*(2), 69-72.

- Seelen, H. A. M., Potten, Y. J. M., Drukker, J., Reulen, J. P. H., & Pons, C. (1998). Development of new muscle synergies in postural control in spinal cord injured subjects. *Journal of Electromyography and Kinesiology*, 8(1), 23-34.
- Serra-Ano, P., Pellicer-Chenoll, M., Garcia-Massó, X., Brizuela, G., Garcia-Lucerga, C., & González, L. M. (2013). Sitting balance and limits of stability in persons with paraplegia. *Spinal Cord*, 51(4), 267-272.
- Sisto, S. A., Druin, E., & Sliwinski, M. M. (2008). *Spinal Cord Injuries-E-Book: Management and Rehabilitation*. Elsevier Health Sciences.
- Thomas, L., Borisoff, J., & Sparrey, C. (2017). Defining the stability limits of a manual wheelchair with adjustable seat and backrest. In *Rehabilitation Engineering and Assistive Technology Society of North America Conference*. New Orleans, LA.
- Trudel, G., Kirby, R. L., Ackroyd-Stolarz, S. A., & Kirkland, S. (1997). Effects of rear-wheel camber on wheelchair stability. *Archives of physical medicine and rehabilitation*, 78(1), 78-81.
- van Drongelen, S., van der Woude, L. H., Janssen, T. W., Angenot, E. L., Chadwick, E. K., & Veeger, D. H. (2005). Glenohumeral contact forces and muscle forces evaluated in wheelchair-related activities of daily living in able-bodied subjects versus subjects with paraplegia and tetraplegia. *Archives of physical medicine and rehabilitation*, 86(7), 1434-1440.

Vanlandewijck, Y. C., Spaepen, A. J., & Lysens, R. J. (1995). Relationship between the level of physical impairment and sports performance in elite wheelchair basketball athletes.

*Adapted Physical Activity Quarterly*, 12(2), 139-150.

Vanlandewijck, Y. C., Evaggelinou, C., Daly, D. D., Van Houtte, S., Verellen, J., Aspeslagh, V., ... & Zwakhoven, B. (2003). Proportionality in wheelchair basketball classification. *Adapted Physical Activity Quarterly*, 20(4), 369-380.

*Physical Activity Quarterly*, 20(4), 369-380.

Vanlandewijck, Y. C., Evaggelinou, C., Daly, D. J., Verellen, J., Van Houtte, S., Aspeslagh, V., ... & Zwakhoven, B. (2004). The relationship between functional potential and field performance in elite female wheelchair basketball players. *Journal of Sports Sciences*, 22(7), 668-675.

in elite female wheelchair basketball players. *Journal of Sports Sciences*, 22(7), 668-675.

Wilson, J. R., Liegey-Dougall, A., Garner, D.M. (2018). Relationship between Elite Women's Wheelchair Basketball Skills Testing and Future Success in the Sport. *Sport Exerc Med Open J.* 4(1), 3-8.

Yanci, J., Granados, C., Otero, M., Badiola, A., Olasagasti, J., Bidaurrezaga-Letona, I., ... & Gil, S. M. (2015). Sprint, agility, strength and endurance capacity in wheelchair basketball players.

*Biology of Sport*, 32(1), 71.

### **Chapter 3**

**Effects of trunk functional capacity on the angular distance utilized to absorb angular momentum in wheelchair braking.**

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## **ABSTRACT**

### **Background**

Abrupt manual wheelchair braking generates upper body angular inertia which must be arrested to prevent a forward fall. The purpose of this study was to determine the effects of trunk functional capacity (low, high) on the angular distance utilized to absorb angular momentum during abrupt manual wheelchair braking.

### **Methods**

Eight wheelchair (4 low function, 4 high function) users completed 10 trials of abrupt wheelchair braking. Trunk segment angles and abdominal joint angles, and normalized upper-body angular impulses were computed for each trial. Linear mixed effects models with initial velocity as a covariate were used to determine differences between groups.

### **Results**

The low function group had a higher angular impulse than the high function group with a mean difference (MD) of 0.01 N·m·s/kg/m<sup>2</sup>, 95% confidence interval (CI) = .002 to .02, p = .017. The low function group also had a higher trunk segment extension angle, MD = -13.98 degrees, 95% CI: -26.27 to -1.69, p = .027. The low function group utilized a larger range of trunk angular distance, MD = 12.14 degrees, 95% CI: -24.48 to 0.21, p = 0.054 to bring the upper body momentum to rest during braking.

### **Conclusions**



Lower trunk functional capacity adversely affects the wheelchair user's ability to attenuate upper-body angular momentum caused by abrupt braking. Despite using a greater range of motion to absorb upper-body angular momentum, low functional capacity users had higher angular impulse and therefore greater fall risk during braking.

## **Introduction**

Manual wheelchair braking is an important skill for wheelchair users. This skill is utilized in wheelchair sports (Cavedon, Zancanaro, & Milanese, 2015; Coutts, 1992; Sporer et al., 2009) as well as casual locomotion, with wheelchair users engaging in up to 250 start/stop activities per day (Slavens et al., 2015). As critical as this skill is to wheelchair users, very little research has been conducted on this aspect of wheelchair mobility, particularly with respect to functional capacity.

Seated balance is impaired in wheelchair users across a functional spectrum (Santos et al., 2016), and this is demonstrated by the high incidence of falls in this population. In one study of 659 wheelchair users, 31% reported a total of 553 fall events, with 15% of these occurring during wheelchair propulsion (Nelson et al., 2010). The main cause of falls in this population was found to be loss of balance during functional activities (Nelson et al., 2003). Manual wheelchair braking uniquely challenges balance control systems in wheelchair users and requires a coordination between the upper limb joints and body segments to decelerate the wheelchair while resisting the forward momentum of the wheelchair/user system, as well as the momentum of the upper body once the wheelchair has come to a stop. Wheelchair users have been shown to have an impaired ability to maintain balance both in stable environments and in response to external perturbations, with differences existing across a functional spectrum. Kamper et al. (1999) used a tilting platform to simulate deceleration conditions similar to those seen in a moving vehicle to examine the relationship between limits of stability and balance in response to external perturbations in the sagittal plane, and found that cervical

spinal cord injured subjects lost balance at a smaller magnitude of disturbance than thoracic spinal cord injured subjects. These results are consistent with functional differences seen in seated balance and functional reach of wheelchair users and demonstrate the role of trunk function in maintaining balance. These difficulties in balance control are influenced both by lack of control of postural muscles as well as impairments in sensorimotor integration of the trunk and lower body (Serra-Ano et al., 2013). Given the inertial forces that wheelchair users are exposed to during propulsion, particularly in the anteroposterior direction (Gagnon et al., 2009), similar challenges could be expected when a wheelchair user uses their hands as friction brakes (Frogley, 2010).

Changes in velocity during wheelchair locomotion (both propulsion and braking activities) result in changes in upper body angular momentum that must be arrested and controlled in order to prevent a loss of balance. The trunk is the largest body segment and is the largest contributor to angular momentum during locomotion (Gillet et al., 2003), so the ability to arrest changes in angular momentum of the trunk becomes of utmost importance. In a study examining the effects of different braking methods in an electric power wheelchair on a crash test dummy, Cooper et al. (1998) observed three different braking methods to examine angular kinematics of the trunk, and in each condition angular velocity of the trunk put the dummy at increased risk of falling out of the wheelchair. They found that speed played a large role in the response of the angular velocity of the trunk, with abrupt braking having the highest risk of falling out of the wheelchair. Results of this study led researchers to recommend that wheelchair users with less trunk control utilize a braking strategy that allows them to come to a

more gradual stop. While these results have relevance to manual wheelchair users, this phenomena has not been observed in this population.

The role of trunk function and the importance of a better understanding of how wheelchair users stop their wheelchairs are interrelated. Gu et al. (1996) put it succinctly, stating “quantification requires knowledge not only of how much angular momentum must be arrested, but also what moment can be developed to do that arresting.” Functional capacity of the trunk influences both propulsion (Vanlandewijck et al, 2010) and deceleration/braking activities (Sisto, Druin, & Sliwinski, 2008), and the degree to which a wheelchair user has the ability to actively engage their trunk to push or stop their wheelchair is influenced by their limits of stability. A wheelchair user can only engage in trunk flexion to generate power during propulsion to the extent that they do not exceed their anterior limits of stability. Conversely, a wheelchair user can only engage in trunk extension to prevent themselves from falling forward to the extent that they do not exceed their posterior limit of stability. While trunk extension has been observed during ramp descent, the role of functional capacity in abrupt braking remains to be seen. Given the role of braking in wheelchair mobility, there remains a need for more information on how manual wheelchair users stop their wheelchairs.

Therefore, the purpose of this study was to determine the effects of trunk functional capacity on the angular distance utilized to absorb angular momentum during abrupt manual wheelchair braking.

## **MATERIALS AND METHODS**

### **Participants**

Eight wheelchair users were recruited to complete this study (4 men, 4 women). Mean age of wheelchair users was  $24.75 \pm 7.57$  years, and mass was  $65.10 \pm 14.15$  kg. Functional Classifications ranged from 1.0 to 3.5. Self-reported shoulder pain or injury within the last six months was considered an exclusion criteria. Use of a wheelchair as primary means of locomotion were required to participate in this study. This group was divided into two subgroups: high functional classification (HFC), in which the participant's functional classification was above 2.5, and low functional classification (LFC), in which the participant's functional classification was below 2.5. Each group had 4 subjects.

### **Experimental Procedures**

Data collection took place during a single visit to the Applied Biomechanics Laboratory at the University of Texas at Arlington. Participants used their personal wheelchair that they used for everyday use. Reflective markers (14 mm) were attached bilaterally to the skin over anatomical landmarks. Acromion process (RAC, LAC), joint center of the shoulder complex (RADL, RPDL, LADL, LPDL), neck in line with C7 (RNECK, LNECK), C7, T8, T2, L1, L3, L5 vertebrae, superior most point of iliac crest in the sagittal plane (RPP, LPP), anterior superior iliac spine (RAS, LAS), posterior superior iliac spine (RPS, LPS), greater trochanters (RHP, LHP), medial and lateral epicondyles of the femur (RMK, RLK, LMK, LLK), medial and lateral epicondyle of the humerus (RMEL, RLEL, LLEL, LMEL), radial and ulnar epicondyles (RWRR, RWRU, LWRR, LWRU), second third, and fifth metacarpals (LHR, LHM, LHU, RHR, RHM, RHU) medial and lateral malleoli (RMA, RLA, LMA, LLA), first metatarsal, base and fifth of the metatarsals. Markers were also placed on the top of the head (THEAD), forehead (AHEAD), occipital bone (PHEAD),

zygomatic bone (RHEAD, LHEAD). Non-collinear markers on molded thermo-plastic shells were placed on the posterior thorax, upper arms, forearms, proximal thighs, and distal shanks. Three tracking markers were placed on the medial, lateral, and posterior heel. All anatomical markers were then removed for wheelchair braking trials.

Prior to the collection of the braking trials, subjects had the opportunity to practice bringing their wheelchair to an abrupt stop at a predefined location within the laboratory. Participants were then asked to complete ten trials where they were instructed to push their wheelchairs as quickly as possible to a predefined spot on the floor of the laboratory, covering a distance of approximately ten meters, and then bring their wheelchair to an abrupt stop as quickly as possible. Trials were disqualified if the participant lost their balance, and the trial was repeated.

### **Data Analysis**

Visual 3D (C-Motion, Germantown, MD, USA) was used to process three-dimensional kinematic and kinetic data for each participant. Marker trajectories were filtered with a fourth order recursive Butterworth low-pass filter with cutoff frequency of 6 Hz.

During processing, it became necessary to digitally define the RAS and LAS markers. A spring-loaded digitizing pointer (C-Motion, Germantown, MD, USA) was used to create digital markers to be used when adipose tissue occluded the physical markers, or when the markers became occluded due to changes in position. The tip of the digitizing pointer was placed on the soft tissue directly over the anterior superior iliac spine, after which the clinician depressed the digitizing pointer until it reached the underlying bone (Lerner et al., 2014).

Kinematic data were low-pass filtered using fourth-order Butterworth filters with cut-off frequencies of 6 Hz. A 14-segment model including the head, torso, pelvis, upper arms, lower arms, thighs, shanks and feet was used to determine the COM location and velocity of each segment. Segment masses and inertial properties were determined using de Leva (1996). Upper-body angular momentum (head, torso, pelvis, arms and hand) about the medial/lateral axis of the upper body center of mass was determined using the following equation:

$$(1) \quad \bar{H} = \sum_{i=1}^n [(\bar{r}_i^{CM} - r_{upperbody}^{CM}) \times m_i(\bar{v}_i^{CM} - \bar{v}_{upperbody}^{CM}) + I_i \bar{\omega}_i]$$

where  $\bar{r}_i^{CM}$ ,  $\bar{v}_i^{CM}$ , and  $\bar{\omega}_i$  are the position, velocity and angular velocity vectors of the  $i$ th segment's CM,  $r_{upperbody}^{CM}$ ,  $\bar{v}_{upperbody}^{CM}$ , are the position and velocity vectors of the upper body CM,  $m_i$  and  $I_i$  are the mass and moment of inertia of the  $i$ th segment and  $n$  is the number of segments. Angular momentum ( $\bar{H}$ ) was normalized by dividing by the mass of the upper body segments (kg) and upper body height squared. The upper body height was defined as the distance from the hip joint to top of head (m). The direction of the angular momentum vector was defined by the right hand rule with positive angular momentum defining rotation of the wheelchair user toward the back of the wheelchair and negative angular momentum defining the magnitude of rotation causing or tending to cause the user to fall forward out of the wheelchair. Angular impulse was computed by integrating the negative angular momentum. The negative angular impulse quantifies the amount of momentum that must be arrested to prevent a forward fall of the user from wheelchair.

## Statistical Analysis

Separate SAS version 9.4 proc GLIMMIX linear mixed effects models were used to compare differences in each dependent variable (trunk segment angles, angular impulse and anterior limits of stability) with functional capacity as a fixed effect grouping factor (high, low), subjects as a random factor with trials and initial horizontal velocity as a covariate. Separate intercepts were fit for each subject using an unstructured variance-covariance matrix to account for the correlations between trials. A significant main effect for group was followed by post hoc analysis with Tukey correction for multiple comparisons between groups with alpha set at 0.05.

## Results

### *Initial Horizontal Velocity prior to Braking*

There were no differences between groups (high, low) in the initial horizontal velocity of the wheelchair-user prior to the initiation of abrupt wheelchair braking, mean difference MD  $\pm$  SE, MD =  $0.17 \pm 0.24$  m/s, with a 95% CI of -0.31 to 0.64,  $p = 0.49$ . The high function group had an initial horizontal velocity of  $3.02 \pm 0.17$  m/s, CI: 2.68 to 3.36 and the low function group had an initial horizontal velocity of  $2.85 \pm 0.17$  m/s, CI: 2.51 to 3.19.

### *Upper Body Kinematics*

There was a significant difference between groups for trunk segment extension,  $p = .027$ . The low functional classification group leaned their trunk back further at the onset of braking their wheelchairs than the high functional classification group ( $30.32 \pm 4.35$  degrees, 95% CI of 21.63 to 39.02, and  $16.34 \pm 4.35$  degrees, 95% CI of 7.66 to 25.03, respectively).



There were no significant differences between groups for the trunk segment flexion attained at the end of braking,  $p = 0.82$ . In the low functional group subjects had a final trunk segment flexion angle of  $-0.02 \pm 5.85$  degrees with a 95% CI of  $-11.72$  to  $11.68$ . Final trunk segment flexion angle in the high function group was  $-1.86 \pm 5.85$  degrees with a 95% CI of  $-13.56$  to  $9.83$ .

There were no significant differences between groups in the trunk segment total range of motion,  $p = 0.054$ . The high function group utilized a mean  $\pm$  SE trunk segment angular distance of  $18.21 \pm 4.36$  degrees, 95% CI of  $9.48$  to  $26.93$  to bring the upper body momentum to rest. Subjects in the low function group utilized an angular distance of  $30.34 \pm 4.36$  degrees, 95% CI of  $21.61$  to  $39.08$  to bring the upper body momentum to rest.

#### *Normalized Upper-Body Angular Momentum and Angular Impulse*

A typical trial of normalized upper body angular momentum about a medial/lateral axis of the upper body CM during abrupt wheelchair braking is shown in Figure 1. During the initial phase of braking the angular momentum is positive as a result of the wheelchair user leaning backward ( $t = 0$  to  $t = 0.5$  s). From  $t = 0.5$  to  $t = 2.0$  s upper body angular momentum is negative as the wheelchair decelerates causing negative angular momentum.

There was a significant difference in the normalized upper-body angular impulse about a medial/lateral axis of the upper-body center of mass,  $p = 0.018$ . Subjects in the low function group had greater fall inducing normalized upper-body angular impulse with  $-0.039 \pm 0.0046$  N·m·s/kg/m<sup>2</sup>, 95% CI of  $-0.03$  to  $-0.0479$ . In the high function group, the fall inducing angular impulse was with  $-0.022 \pm 0.0046$  N·m·s/kg/m<sup>2</sup>, 95% CI of  $-0.0137$  to  $-0.032$ .

## Discussion

Abrupt manual wheelchair braking generates upper body angular inertia which must be counteracted to prevent a forward fall of the user from the wheelchair. The upper limits of this counteracting torque about the hip joint center is determined by neuromuscular control of the trunk and lower limbs by the wheelchair user. During the initial phase of braking the wheelchair user leans the trunk backward effectively increasing the angular distance over which he/she can counteract the angular inertia imposed by wheelchair braking. Subjects in our low functional classification leaned their trunk backwards 13.97 degrees more than subjects in our high functional classification group. The low function group also utilized a larger angular distance of 12.14 degrees to bring the upper body momentum to rest during braking. These differences in trunk segment angular kinematics during braking directly reflect the functional capacity of the wheelchair user. High functional capacity subjects can generate greater extension torque and therefore require a shorter angular distance to absorb angular inertia.

To successfully prevent a forward fall from the wheelchair during abrupt braking the wheelchair user must attenuate forward angular momentum as the wheelchair decelerates. Figure 2 illustrates an example of successful (left) and unsuccessful braking (right) for a single subject. The velocity, normalized relative linear momentum of the upper body, and normalized angular momentum is shown for each trial. Graphs in the middle row depict the linear momentum of the upper body relative to the wheelchair velocity. In successful braking on the left, initial relative linear momentum is negative due to trunk extension, from 0.75 s to 1.5 s the upper body linear momentum continues to move forward as the chair decelerates to 0 m/s

(top). The angular momentum graph on the bottom left illustrates that the subject successfully attenuates the negative angular momentum caused by deceleration of the wheelchair, thus preventing a forward fall. In the fall condition shown on the right, the subject did not have the functional capacity to arrest the angular momentum of their trunk. Sensing the impending loss of balance at 1.2 s the subject releases the pushrims, causing chair/user velocity increase from 1.2 to 1.7 s. The subject then grabs the vertical uprights of the wheelchair to prevent a forward fall.

As mentioned, there is evidence to suggest that trunk function plays a role in the anticipatory response to changes in position to maintain balance (Chikh et al., 2018; Crommert et al., 2015; Yang et al., 2006), including activation of the abdominal and trunk extensor muscles. In this study the low function group had a higher normalized angular impulse than the higher function group, suggesting that they have a decreased ability to slow the forward inertia of the trunk upon bringing the wheelchair to a complete stop. Coupled with a smaller anterior stability limit, this puts users with less functional control of their trunk at an increased risk of falling out of their wheelchair if angular momentum exceeds the moment that can be developed to arrest it before the center of mass falls outside the functional base of support. Gait termination studies provide a unique insight into this phenomenon. During transition periods such as gait termination (or wheelchair braking) the CNS is uniquely challenged to respond to changes to the environment, and must determine factors such as appropriate braking forces, including the direction and magnitude of those forces, to maintain to maintain the center of mass over the functional base of support (Meier et al., 2001). Combining an

increased angular impulse with a decreased ability to control that center of mass (which, for these purposes consists of the upper body) puts the user at an increased risk of suffering a fall. This has been demonstrated in able-bodied subjects with decreased proprioception in type II diabetes (Meier et al., 2001), incomplete spinal cord injury (Lemay et al., 2015), and Parkinson's Disease (Oates et al., 2008). One factor that may influence this in wheelchair users is decreased proprioception in individuals with lower functional capacity. In a study examining wheelie performance in wheelchair users, Kauzlarich & Thacker (1987) determined that proprioception, vision, and vestibular function may be as important in the balance of a wheelie as they are in maintaining standing balance. It is possible that decreased proprioception of the lower body and trunk puts wheelchair users with less functional capacity at a disadvantage when it comes to arresting angular momentum and maintaining balance when their wheelchair comes to a complete stop. These results provide a valuable look into the role that functional capacity in manual wheelchair users plays in manual wheelchair braking.

Very little research has been conducted on trunk utilization and movement patterns during manual wheelchair braking. In a study examining different braking strategies of electric power wheelchairs Dvorznak et al. (2001) observed that when across three braking conditions, the center of gravity and height of the wheelchair directly influenced the tipping moment of the wheelchair and user. If this moment is great enough, this can result in the wheelchair tipping forward. Alternatively, if the center of gravity is higher, the user may fall out of the wheelchair. This says nothing of functional capacity or the user's ability to control their trunk and prevent the fall in the first place. Additionally, this research was conducted in electric power

wheelchairs, so extrapolating the results to manual wheelchair users may be of limited utility. In a study examining the response to motor vehicle braking, Kamper et al., (1999) found that SCI subjects lost balance at thresholds below those seen in standard braking patterns in motor vehicles. These subjects were not able to use their hands or arms to support themselves, and the inertia they were required to resist was the result of a motor vehicle, not wheelchair propulsion. We recommend that future studies focus further on the strategies that wheelchair users with lower functional capacity may use to mitigate the risk of falling during abrupt braking.

Additionally, inferences can be made from work that has been done regarding the role of the trunk during forward propulsion. For example, Rice et al. (2004) found that the trunk actually moved backwards at the beginning of the push phase and attributed this movement to the reactive forces of the pushrim in individuals with impaired trunk control. Taken with the results from the present study, we hypothesize that manual wheelchair users with lower functional capacity may have preemptively leaned their trunk backwards in order to try to prevent their trunk from falling forward once their hands came into resistance from the forward momentum of the handrim. This increases the braking distance, decreasing the overall force incurred on the trunk. This is consistent with work conducted by Yang et al. (2006), who demonstrated that able-bodied volunteers who propelled a manual wheelchair showed activation of the trunk muscles even before the initiation of the push cycle, stabilizing their trunk in anticipation of the reactionary forces incurred during the push itself.

The greater trunk extension angle for the lower functional classification group was an interesting discovery, particularly as it relates to the results seen for anterior/posterior limits of stability. It is possible that, lacking the functional ability to actively engage in trunk extension, lower functional classification subjects used their limited function to shift their center of mass over their back rest, using their wheelchair set up as a means of passive stability. This increased the amount of distance that their center of mass needed to travel before reaching their anterior limit of stability, thereby increasing the time that they were able to adjust before coming to a complete stop. There is evidence to suggest that backrest height influences a number of outcomes with respect to wheelchair propulsion biomechanics. Yang et al. (2012) found that wheelchair users with a lower backrest had a greater range of shoulder motion, increased stroke angle, and reduced cadence during forward propulsion, all of which decreased the risk of injury to the shoulder complex. According to the IWBF (2014), lower classification players rely on backrest height for improved stability, so future work should examine the role of backrest height in trunk kinematics during braking.

### **Limitations**

There are some limitations to this study. While we were able to note the location of the wheelchair using reflective markers, we did not place a marker on the wheel. This required that we estimate the positioning of the hand on the wheel and thus were not able to accurately determine hand placement with respect to braking phase. Additionally, we did not take note of the wheel diameters for each subject. There is evidence to suggest that wheel size may play a role in physiologic and biomechanical outcomes (Mason et al., 2012), and documenting that

aspect may have provided insight into the biomechanics of wheelchair braking. Additionally, the nature of sitting in a wheelchair makes placing markers on the hips and pelvis difficult. Active and passive stability of the hips and pelvis is an important aspect of functional classification in wheelchair sports, so understanding the behavior of these segments may provide insight into stabilization strategies during manual wheelchair braking activities. This can be done either mathematically or through methods that allow for the visualization of specific markers on bony landmarks.

### **Future Directions**

Future research should focus on the role of different wheelchair configurations in attenuating forces of manual wheelchair braking. There is evidence to suggest that manipulating the angle of the base of support may increase stability during propulsion (Hastings et al., 2003), and this may extend to braking activities as well. Additionally, future research should examine the kinetics both at the pushrim and at the joints of the upper limb during abrupt wheelchair braking. This can be done using technology such as the SmartWHEEL. Future research should also focus on the relationship between reaction time and functional capacity, as those with less functional capacity may require a greater braking distance in order to maintain postural control and avoid falling out of the wheelchair. In gait termination research using ambulatory subjects, there is evidence to suggest that subjects may use a trunk flexion strategy to avoid falling (Bishop et al., 2004), so the role of the trunk in wheelchair users with decreased functional capacity as a function of reaction time, particularly during activities of daily living that require abrupt stopping, may be of interest. The role of manual wheelchair

braking in shoulder pain is also an area that should be further explored. Yildirim & Ozengin (2010) found that individuals with less trunk control reported greater instances of shoulder pain on the Wheelchair User's Shoulder Pain Index (WUSPI). Kinetic variables influencing shoulder, elbow, and wrist pain during braking would provide valuable information on the role of this aspect of wheelchair locomotion on the risk of injury across the functional spectrum. Finally, EMG analysis on trunk stiffness to elucidate the role of specific muscle groups in maintaining trunk stiffness during braking should also be examined.

### **Conclusions**

Lower trunk functional capacity adversely affects the wheelchair user's ability to attenuate upper-body angular momentum caused by abrupt braking. Despite using a greater range of motion to absorb upper-body angular momentum, low functional capacity users had higher angular impulse and therefore greater fall risk during braking. The functional ability to control the angular momentum of the trunk to prevent it from exceeding the limits of stability is a critical skill for wheelchair users, and strategies need to be developed for those with less functional capacity. Rehab and sport specialists should take these findings into consideration when developing rehab or training protocols in order to assist in the maximization of musculature to complete activities of daily living, and braking should be more of a focus in wheelchair skills training.



## Figure Legends

**Figure 1:** A typical trial of normalized upper body angular momentum about a medial/lateral axis of the upper body CM during abrupt wheelchair braking. During the initial phase of braking the angular momentum is positive as a result of the wheelchair user leaning backward ( $t = 0$  to  $t = 0.5$  s). From  $t = 0.5$  to  $t = 2.0$  s upper body angular momentum is negative as the wheelchair decelerates causing negative angular momentum. Angular impulse was computed by integrating the negative angular momentum. The negative angular impulse quantifies the amount of momentum that must be arrested to prevent a forward fall of the user from wheelchair.

**Figure 2:** Example of successful (left) and unsuccessful braking (right) for a single subject. The velocity, normalized relative linear momentum of the upper body, and normalized angular momentum is shown for each trial. The middle row graphs depict the linear momentum of the upper body relative to the wheelchair velocity. In successful braking on the left, initial relative linear momentum is negative due to trunk extension, from 0.75 s to 1.5 s the upper body linear momentum continues to move forward as the chair decelerates to 0 m/s (top). The angular momentum graph on the bottom left illustrates that the subject successfully attenuates the negative angular momentum caused by deceleration of the wheelchair, thus preventing a forward fall. In the fall condition shown on the right at 1.2 s the subject realizes he is about to fall so he releases the pushrims causing the velocity to increase from 1.2 to 1.7 s (top right). The subject quickly grabs the vertical uprights of the wheelchair to prevent himself from falling forward out of the chair.

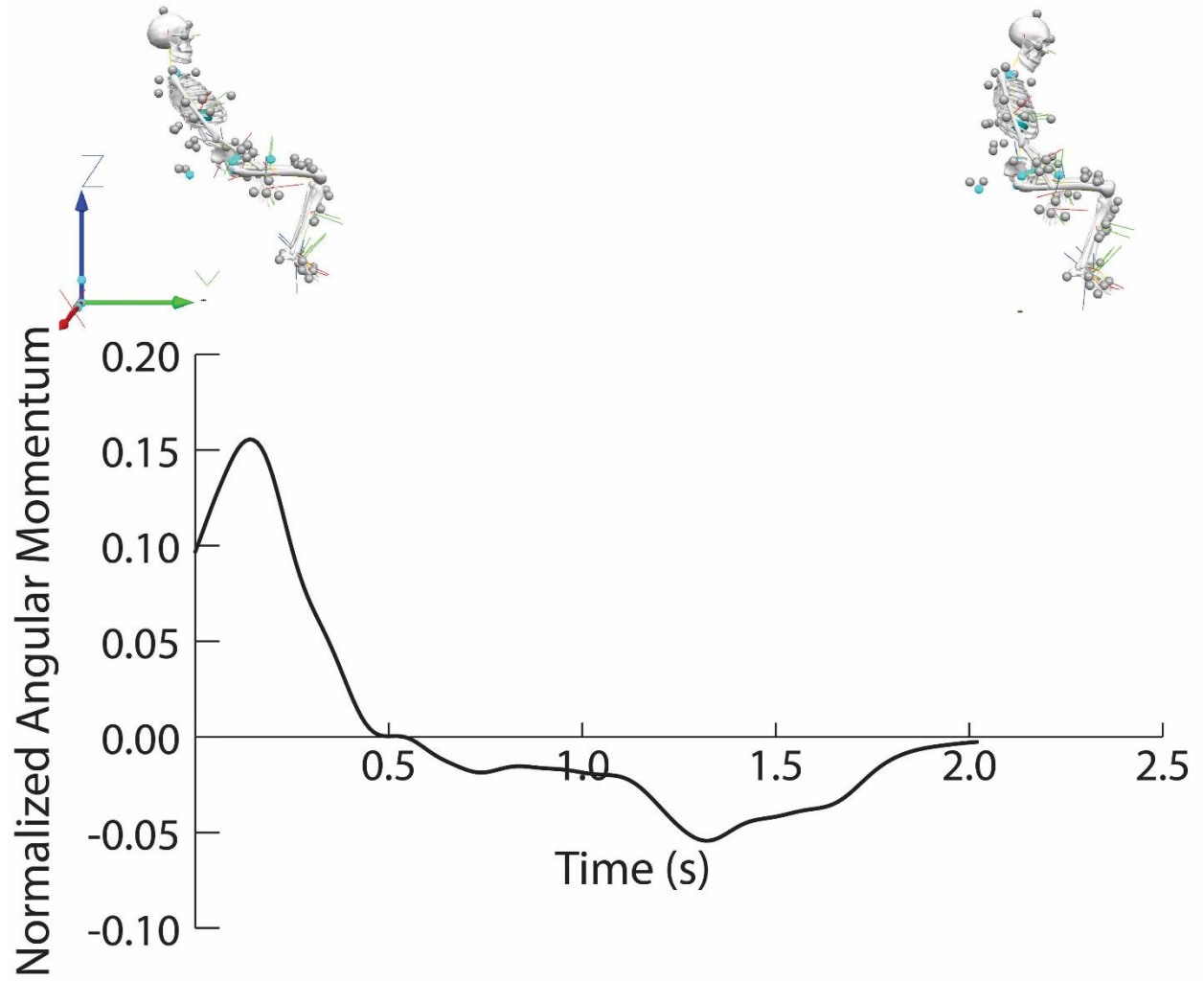


Figure 1. Angular impulse of a typical trial

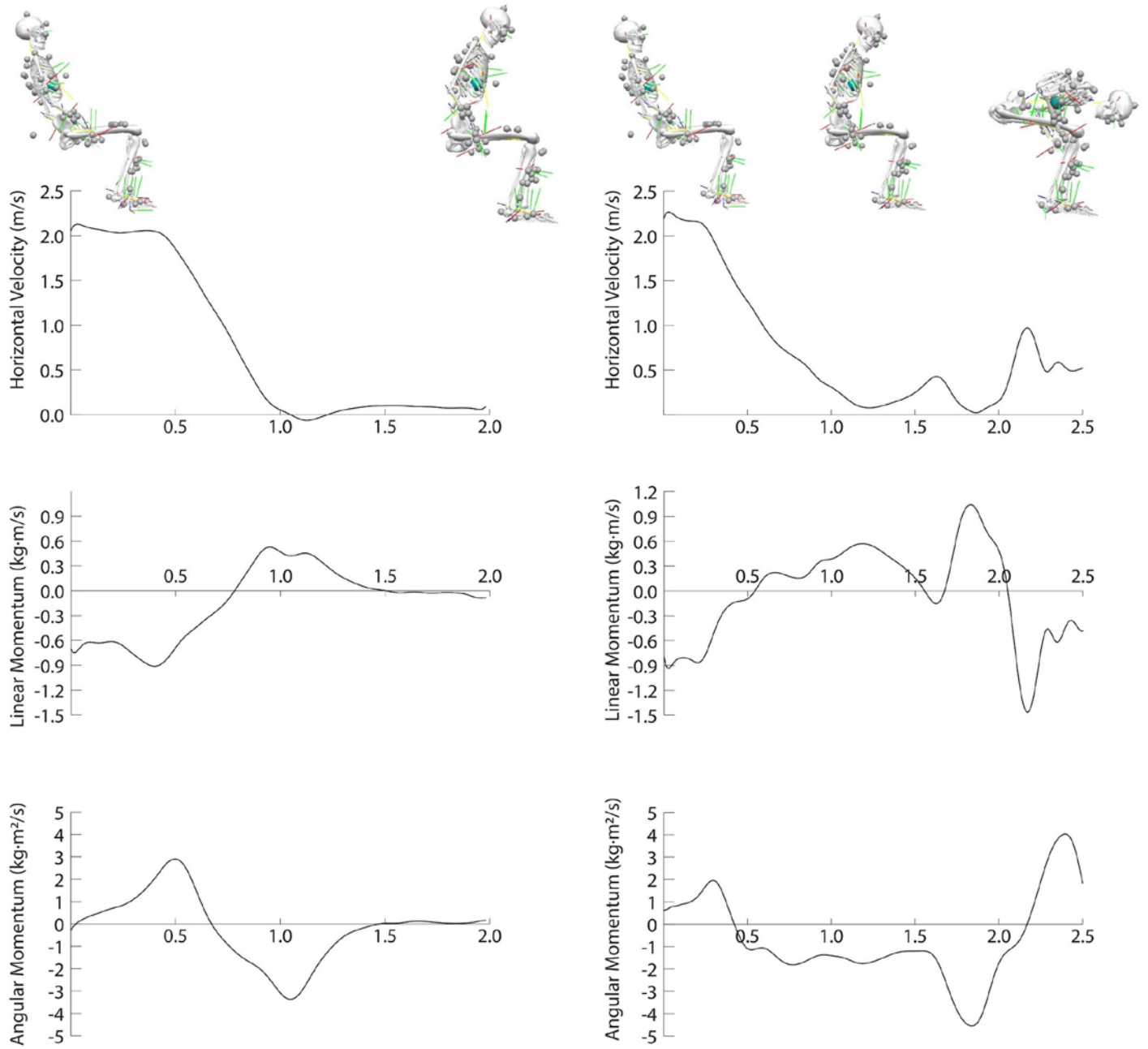


Figure 2. Subject losing balance

## References

- Bishop, M., Brunt, D., Pathare, N., & Patel, B. (2004). The effect of velocity on the strategies used during gait termination. *Gait & posture*, 20(2), 134-139.
- Cavedon, V., Zancanaro, C., & Milanese, C. (2015). Physique and performance of young wheelchair basketball players in relation with classification. *PloS one*, 10(11), e0143621.
- Chikh, S., Garnier, C., Faupin, A., Pinti, A., Boudet, S., Azaiez, F., & Watelain, E. (2018). Arm-trunk coordination in wheelchair initiation displacement: A study of anticipatory and compensatory postural adjustments during different speeds and directions of propulsion. *Journal of Electromyography and Kinesiology*, 40, 16-22.
- Cooper, R. A., Dvorznak, M. J., O'Connor, T. J., Boninger, M. L., & Jones, D. K. (1998). Braking electric-powered wheelchairs: Effect of braking method, seatbelt, and legrests. *Archives of Physical Medicine and Rehabilitation*, 79(10), 1244-1249.
- Coutts, K. D. (1992). Dynamics of wheelchair basketball. *Medicine and science in sports and exercise*, 24(2), 231-234.
- Crommert, M. E., Halvorsen, K., & Ekblom, M. M. (2015). Trunk muscle activation at the initiation and braking of bilateral shoulder flexion movements of different amplitudes. *PloS one*, 10(11), e0141777.
- de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics*. 29: 1223-1230, 1996.

Dvorznak, M. J., Cooper, R. A., O'Connor, T. J., Boninger, M. L., & Fitzgerald, S. G. (2001). Kinematic comparison of Hybrid II test dummy to wheelchair user. *Medical engineering & physics*, 23(4), 239-247.

Frogley, M. (2010). Wheelchair basketball. *Wheelchair Sport: A Complete Guide for Athletes, Coaches, and Teachers*, 119-132.

Gagnon, D., Verrier, M., Masani, K., Nadeau, S., Aissaoui, R., & Popovic, M. (2009). Effects of trunk impairments on manual wheelchair propulsion among individuals with a spinal cord injury: a brief overview and future challenges. *Topics in Spinal Cord Injury Rehabilitation*, 15(2), 59-70.

Gillet, C., Duboy, J., Barbier, F., Armand, S., Jeddi, R., Lepoutre, F. X., & Allard, P. (2003). Contribution of accelerated body masses to able-bodied gait. *American journal of physical medicine & rehabilitation*, 82(2), 101-109.

Gu, M. J., Schultz, A. B., Shepard, N. T., & Alexander, N. B. (1996). Postural control in young and elderly adults when stance is perturbed: dynamics. *Journal of biomechanics*, 29(3), 319-329.

Hastings, J. D., Fanucchi, E. R., & Burns, S. P. (2003). Wheelchair configuration and postural alignment in persons with spinal cord injury. *Archives of physical medicine and rehabilitation*, 84(4), 528-534.

IWBF Player Classification Commission. (2014). *A Guide to the IWBF Functional Classification System for Wheelchair Basketball Players*.

- Kamper, D., Barin, K., Parnianpour, M., Reger, S., & Weed, H. (1999). Preliminary investigation of the lateral postural stability of spinal cord-injured individuals subjected to dynamic perturbations. *Spinal Cord*, 37(1), 40.
- Kauzlarich, J. J., & Thacker, J. G. (1987). A theory of wheelchair wheelie performance. *Journal of rehabilitation research and development*, 24(2), 67-80.
- Kim SJ, Boninger ML, Koontz AM, Cooper RA, Souza AL, Wheelchair braking biomechanics, Proceedings of RESNA 2002 Annual Conference, Minneapolis, MN, 2002.
- Lerner, Z. F., Board, W. J., & Browning, R. C. (2014). Effects of an obesity-specific marker set on estimated muscle and joint forces in walking. *Med Sci Sports Exerc*, 46(6), 1261-1267.
- Lemay, J. F., Duclos, C., Nadeau, S., & Gagnon, D. H. (2015). Postural control during gait initiation and termination of adults with incomplete spinal cord injury. *Human movement science*, 41, 20-31.
- Mason, B. S., Van, L. D. W., Tolfrey, K., Lenton, J. P., & Goosey-Tolfrey, V. L. (2012). Effects of wheel and hand-rim size on submaximal propulsion in wheelchair athletes. *Medicine and science in sports and exercise*, 44(1), 126-134.
- Meier, M. R., Desrosiers, J., Bourassa, P., & Blaszczyk, J. (2001). Effect of type II diabetic peripheral neuropathy on gait termination in the elderly. *Diabetologia*, 44(5), 585-592.

- Nelson, A., Ahmed, S., Harrow, J., Fitzgerald, S., Sanchez-Anguiano, A., & Gavin-Dreschnack, D. (2003). Fall-related fractures in persons with spinal cord impairment: a descriptive analysis. *SCI nursing: a publication of the American Association of Spinal Cord Injury Nurses*, 20(1), 30-37
- Nelson, A. L., Groer, S., Palacios, P., Mitchell, D., Sabharwal, S., Kirby, R. L., ... & Powell-Cope, G. (2010). Wheelchair-related falls in veterans with spinal cord injury residing in the community: a prospective cohort study. *Archives of physical medicine and rehabilitation*, 91(8), 1166-1173.
- Oates, A. R., Frank, J. S., Patla, A. E., VanOoteghem, K., & Horak, F. B. (2008). Control of dynamic stability during gait termination on a slippery surface in Parkinson's disease. *Movement disorders: official journal of the Movement Disorder Society*, 23(14), 1977-1983.
- Rice, I., Koontz, A. M., Boninger, M. L., & Cooper, R. A. (2004, June). An analysis of trunk excursion in manual wheelchair users. In *Proceedings of the 27th International RESNA Conference* (pp. 20-22).
- Santos, S. D. S., Alonso, A. C., & Greve, J. M. D. A. (2016). Quantitative evaluation of trunk muscle strength in wheelchair basketball players. *Motriz: Revista de Educação Física*, 22(2), 69-72.
- Serra-Ano, P., Pellicer-Chenoll, M., Garcia-Massó, X., Brizuela, G., Garcia-Lucerga, C., & González, L. M. (2013). Sitting balance and limits of stability in persons with paraplegia. *Spinal Cord*, 51(4), 267-272.

- Sisto, S. A., Druin, E., & Sliwinski, M. M. (2008). *Spinal Cord Injuries-E-Book: Management and Rehabilitation*. Elsevier Health Sciences.
- Slavens, B. A., Schnorenberg, A. J., Aurit, C. M., Tarima, S., Vogel, L. C., & Harris, G. F. (2015). Biomechanics of pediatric manual wheelchair mobility. *Frontiers in bioengineering and biotechnology*, 3, 137.
- Spornier, M. L., Grindle, G. G., Kelleher, A., Teodorski, E. E., Cooper, R., & Cooper, R. A. (2009). Quantification of activity during wheelchair basketball and rugby at the National Veterans Wheelchair Games: A pilot study. *Prosthetics and orthotics international*, 33(3), 210-217.
- Vanlandewijck, Y., Verellen, J., & Tweedy, S. (2010). Towards evidence-based classification- the impact of impaired trunk strength on wheelchair propulsion. *Advances in Rehabilitation*, 3, 1-5.
- Yang, Y. S., Koontz, A. M., Triolo, R. J., Mercer, J. L., & Boninger, M. L. (2006). Surface electromyography activity of trunk muscles during wheelchair propulsion. *Clinical Biomechanics*, 21(10), 1032-1041.
- Yang, Y. S., Koontz, A. M., Yeh, S. J., & Chang, J. J. (2012). Effect of backrest height on wheelchair propulsion biomechanics for level and uphill conditions. *Archives of physical medicine and rehabilitation*, 93(4), 654-659.



Yildirim, N. U., Comert, E., & Ozengin, N. (2010). Shoulder pain: a comparison of wheelchair basketball players with trunk control and without trunk control. *Journal of back and musculoskeletal rehabilitation*, 23(2), 55-61.

## Chapter 4

### Summary and Future Directions

The role of trunk function in the movement patterns of manual wheelchair users is a complex issue that is as diverse as the injuries that require wheelchair use in the first place. Several factors can contribute to this, including functional capacity, wheelchair configuration, and time spent in the wheelchair. This is particularly evident in activities such as manual wheelchair braking, where the user is required to respond to changes in momentum due to outside perturbations. As previously mentioned, manual wheelchair braking is a critical skill and a major aspect of wheelchair mobility, yet little work has been conducted on this behavior. While advances have been made in our understanding of the areas related to postural control in this population, particularly with respect to functional capacity, there remains a considerable gap when it comes to functional classification in wheelchair sport. The diversity of the nature of injuries that lead to the need for wheelchair use makes it difficult to determine the role of injury in trunk function. Spinal cord injury lesion level is not always indicative of function, and the manifestation of function can improve longitudinally (Seelen et al., 1998). This suggests a motor learning effect gained either from experience exploring functional capacity, or via direct practice during rehabilitation.

Furthermore, few frameworks have been developed to observe trunk function in this population. For example, inverted pendulum models have been used to attempt to quantify seated postural control, using mathematical modeling to define single segment and two segment models and the joint stiffness required to maintain both static stability and the joint stiffness and torques required to resist against external perturbations (Bidard et al., 2000). The current investigation uses a Volume of Action framework to observe trunk function using a modified limits of stability test, and then applied those functional limits to how those manual

wheelchair users used their trunk to bring their wheelchair to a complete, abrupt stop. This framework is unique in that it allows researchers and clinicians to recruit wheelchair users from across a functional spectrum without respect to injury level, and therefore increase the subject pool for research purposes by looking at wheelchair users as a population rather than limiting participation to subjects with spinal cord injury.

Study 1 employed the concept of Volume of Action used in wheelchair basketball classification to compare those in high functional classification groups and low functional classification groups in three planes of movement. Subjects with higher functional capacity (represented by higher functional classification) were able to increase their trunk excursion to a greater extent in flexion/extension at the hip, left and right lateral flexion, and rotation of the trunk along the Z axis. Although the primary focus of this investigation was the trunk, movement of the pelvis was also different between groups, with subjects using different movement patterns in an attempt to keep their center of mass above their base of support. Previous work in this area observed movement from the perspective of the wheelchair. This investigation placed subjects in an environment that prevented them from relying on wheelchair configuration or other strategies to stabilize, allowing us to observe true functional capacity. Additionally, using the criteria of the functional classification system allowed for the observation of specific movement patterns that are used not only in wheelchair sports, but also activities of daily living common to manual wheelchair users.

An interesting observation from this investigation is the differences in pelvic stabilization strategies used between groups to facilitate movement. The high classification group had higher range of motion for thoracic rotation, primarily through the added ability to

rotate the pelvis in conjunction with the trunk. Consistent with the IWBF functional classification criteria, low classification subjects were unable to effectively use the musculature of the hips and pelvis to further rotate the upper body, limiting rotation to primarily the upper trunk. The inclusion of a marker that further delineated the “upper trunk” vs the “lower trunk” may have been able to provide more insight into the musculature recruited to facilitate these movements. This observation would be particularly useful for further examination into the functional classification system in wheelchair basketball used by the IWBF and NWBA (National Wheelchair Basketball Association). For example, in this system players who are classed 2.0 exhibit active upper trunk rotation, but no lower trunk functions. The IWBF does not clarify what constitutes “upper trunk rotation”, and this may result in variability between classifiers across classification panels. Work by Bae et al. (2013) define upper trunk rotation as “moving the shoulders forward and backward” and lower trunk rotation as “moving the knees forward and backward”, suggesting a functional recruitment of the pelvis, hip, and lower abdominal muscles in lower trunk rotation. These definitions provide context and provide a clear delineation between the upper and lower trunk and may help to provide objective criteria for future work that seeks to observe differences in these two segments. Standardized definitions on how these movements manifest within the classification system would help to provide further objectivity in functional classification systems within wheelchair sports.

The differences between upper and lower trunk may also provide some context into the concept of active vs passive pelvic stabilization (IWBF, 2014). One aspect of Study 1 provided valuable insight into how wheelchair users of different functional capacities utilized their ability

(or lack thereof) to compliment trunk movement by increasing the range of motion these body segments were able to achieve.

Results from study 2 provide a foundation for future research into manual wheelchair braking. While studies have been conducted using crash test dummies (Cooper et al., 2003; Dvorznak et al., 2001; Fast et al., 1997), the current investigation demonstrates that this type of research can be conducted safely using human subjects. Technological advances such as the SMARTWheel (Cooper & Cheda, 1989) provide a valuable opportunity to shed light on the impact on the shoulder joint of braking for both athletic endeavors and activities of daily living. The application and adaptation of inverted pendulum models for wheelchair users also have potential to provide more insight into postural control of wheelchair users.

It was surprising to find that current Wheelchair Skills Tests (Kirby et al., 2002), which seek to evaluate and train new wheelchair users as part of the rehabilitation process, contained no braking aspects of the battery of tests for wheelchair users. Competency in these skills increases self-efficacy in wheelchair users, and there is a relationship between wheelchair skills self-efficacy and participation in physical activity (Phang et al., 2012). Evidence-based training protocols on manual wheelchair braking should be incorporated into future iterations of Wheelchair Skills Tests, with special consideration given to strategies to accommodate various levels of functional capacity. Given that rates of obesity in wheelchair-using populations are higher than those of able-bodied populations (Froehlich-Grobe, 2010), there remains a need to further promote physical activity in this population, and self-efficacy in wheelchair skills, including braking, is an important piece of that puzzle.

Another potential area of interest for future investigations is the role of functional capacity in braking distance. The results from study 2 lend credence to the idea that individuals with lower functional capacity may need to coordinate their body segments differently to prepare for the wheelchair to come to a complete stop, and this may put the user at a disadvantage if the braking actions is required to be completed abruptly or unexpectedly. The amount of braking distance needed for a wheelchair user to decelerate/stop their wheelchair across a range of functional capacities would be of value to rehab and sport professionals.

There have been numerous attempts to describe the upper extremity movement during manual wheelchair propulsion (Newsam et al., 1999; Finley et al., 2004; Rao et al., 1996) Study 2 was the first investigation to provide a descriptive analysis of upper extremity kinematics during manual wheelchair braking, filling an important gap in the literature with respect to motor skills of the manual wheelchair using population. Upper extremity kinematics of wheelchair propulsion can vary significantly between groups of wheelchair users with different levels of functional ability. This can include joint angles (Crespo-Ruiz & Del Ama-Espinoza, 2011; Finley et al., 2004) and kinematic variables with respect to contact with the handrim (Crespo-Ruiz & Del Ama-Espinoza, 2011; Haydon et al., 2017). This is the first study to examine joint angle changes during the braking cycle during manual wheelchair braking. One limitation of this study is that data was collected only at the onset of braking through the end of the braking cycle. Future studies should examine kinematic patterns during the recovery phase of the propulsion cycle as it relates to the initiation of braking. There are four primary types of propulsion patterns (SLOP, DLOP, ARC) (Sisto et al., 2008), and differences in propulsion kinematics have been observed during each of the four patterns ( Koontz et al., 2009). Future

work should determine if these patterns play a role in manual wheelchair braking across the functional spectrum, as well as during various conditions such as variations in speed, or initiating a braking cycle going down a hill.

Given the amount of time that wheelchair users engaged in sport spend in braking activities (Coutts et al., 1992; Spörner et al., 2009), this study provides a foundation by which future researchers may examine the role of abrupt braking in injuries incurred during wheelchair propulsion and braking.

Another area of interest that should be expanded on is reaction time across the functional spectrum for manual wheelchair braking across a range of activities. Studies examining postural control with subjects that have trunk impairments have shown that increased reaction time is seen with decreased function (Hodges, 2001). Future research should expand on this concept to include wheelchair athletes with respect to braking.

A potential limitation to study 2 was the inability to adequately capture multiple forward propulsion cycles prior to the subject engaging in the braking activity. This would have provided valuable information on multiple levels. First, the differences in stroke patterns within wheelchair users for propulsion are well documented (Shimada et al., 1998; Koontz et al., 2009; Boninger et al., 2002), and these differences may play a role in risk of injury to the upper limbs (Boninger et al., 2005). Capturing propulsion stroke pattern may allow for questions to be asked about how these differences influence the way that wheelchair users initiate braking patterns. Rodgers et al. (2000) found that wheelchair users who utilized a trunk flexion propulsion strategy placed their hands on the handrim differently than those who had less trunk variability within the propulsion cycle. It remains to be seen whether these differences have any influence



on how wheelchair users bring their wheelchair to a complete stop. Additionally, subjects in the current investigation were instructed to stop their wheelchair at a point known to them. While this may reflect conditions such as decelerating to open a door (Morrow et al., 2010), it does not reflect the rapid starting and stopping seen in wheelchair sports, particularly in conditions where the athlete is required to abruptly stop to avoid colliding with another player.

Additionally, braking at different speeds may provide more insight into how this action affects wheelchair users during activities of daily living. Adding more complex variables to braking conditions will also allow for more complex statistical analysis to further elucidate various factors that may influence this action. Another limiting factor that resulted as a function of the data collection area was that the opportunity to adequately collect elbow joint angles during the braking cycle and the time period immediately preceding the braking cycle.

Rehabilitation specialists, wheelchair sport classifiers, wheelchair manufacturers, and other entities have an active interest in the continuation of this research. Sport classifiers should be aware of the role of wheelchair configuration in the manifestation of stability, and work to elucidate the role of stabilizing techniques such as strapping and seat inclination in the Volume of Action of the trunk.

These investigations sought to provide insight and context into the role of functional capacity in the limits of stability in manual wheelchair users, and the role that those limits play in how manual wheelchair users use their trunk to bring their wheelchair to an abrupt, complete stop without experiencing a fall. Observing this relationship using a Volume of Action framework provides valuable insight into the motor patterns of this diverse population and

gives researchers and clinicians tools that can assist in a wide array of functions, from sport classification to wheelchair design.

## References

- Bae, S. H., Lee, H. G., Kim, Y. E., Kim, G. Y., Jung, H. W., & Kim, K. Y. (2013). Effects of trunk stabilization exercises on different support surfaces on the cross-sectional area of the trunk muscles and balance ability. *Journal of physical therapy science*, 25(6), 741-745.
- Bidard, C., Rienstra, S., Veltink, P. H., Koopman, H. F. J. M., Grady, J., De Vries, J., & Huttenhuism, L. (2000). Trunk stability while standing or sitting: a static analysis. In *IFESS 2000 Conference at Aalborg University, Denmark*.
- Boninger, M. L., Souza, A. L., Cooper, R. A., Fitzgerald, S. G., Koontz, A. M., & Fay, B. T. (2002). Propulsion patterns and pushrim biomechanics in manual wheelchair propulsion. *Archives of physical medicine and rehabilitation*, 83(5), 718-723.
- Boninger, M. L., Koontz, A. M., Sisto, S. A., Dyson-Hudson, T. A., Chang, M., Price, R., & Cooper, R. A. (2005). Pushrim biomechanics and injury prevention in spinal cord injury: recommendations based on CULP-SCI investigations. *Journal of rehabilitation research and development*, 42(3), 9.
- Cooper RA, Cheda A. Measurement of Racing Wheelchair Propulsion Torque. *Proceedings 11th Annual International Conference of the IEEE/EMBS; Seattle, Washington. November 9–12.1989.*
- Cooper, R. A., Wolf, E., Fitzgerald, S. G., Boninger, M. L., Ulerich, R., & Ammer, W. A. (2003). Seat and footrest shocks and vibrations in manual wheelchairs with and without suspension. *Archives of physical medicine and rehabilitation*, 84(1), 96-102.
- Coutts, K. D. (1992). Dynamics of wheelchair basketball. *Medicine and science in sports and exercise*, 24(2), 231-234.

- Crespo-Ruiz, B. M., & Del Ama-Espinosa, A. J. (2011). Relation between kinematic analysis of wheelchair propulsion and wheelchair functional basketball classification. Adapted physical activity quarterly, 28(2), 157-172.
- Dvorznak, M. J., Cooper, R. A., O'Connor, T. J., Boninger, M. L., & Fitzgerald, S. G. (2001). Kinematic comparison of Hybrid II test dummy to wheelchair user. Medical engineering & physics, 23(4), 239-247.
- Fast, A., Sosner, J., Begeman, P., Thomas, M., & Drukman, D. (1997). Forces, moments, and acceleration acting on a restrained dummy during simulation of three possible accidents involving a wheelchair negotiating a curb: Comparison Between Lap Belt and Four-Point Belt1. American journal of physical medicine & rehabilitation, 76(5), 370-377.
- Finley, M. A., Rasch, E. K., Keyser, R. E., & Rodgers, M. M. (2004). The biomechanics of wheelchair propulsion in individuals with and without upper-limb impairment. Journal of rehabilitation research & development, 41.
- Froehlich-Grobe, K. (2010). Exploring obesity among wheelchair users: BMI vs body composition. In International Journal of Exercise Science: Conference Proceedings (Vol. 6, No. 1, p. 28).
- Haydon, D. S., Pinder, R. A., Grimshaw, P. N., & Robertson, W. S. (2017). Overground-propulsion kinematics and acceleration in elite wheelchair rugby. International journal of sports physiology and performance, 13(2), 156-162.
- IWBF Player Classification Commission. (2014). A Guide to the IWBF Functional Classification System for Wheelchair Basketball Players.

- Kirby, R. L., Swuste, J., Dupuis, D. J., MacLeod, D. A., & Monroe, R. (2002). The Wheelchair Skills Test: a pilot study of a new outcome measure. *Archives of Physical Medicine and Rehabilitation*, 83(1), 10-18.
- Koontz, A. M., Roche, B. M., Collinger, J. L., Cooper, R. A., & Boninger, M. L. (2009). Manual wheelchair propulsion patterns on natural surfaces during start-up propulsion. *Archives of physical medicine and rehabilitation*, 90(11), 1916-1923.
- Morrow, M. M., Hurd, W. J., Kaufman, K. R., & An, K. N. (2010). Shoulder demands in manual wheelchair users across a spectrum of activities. *Journal of Electromyography and Kinesiology*, 20(1), 61-67.
- Newsam, C. J., Rao, S. S., Mulroy, S. J., Gronley, J. K., Bontrager, E. L., & Perry, J. (1999). Three dimensional upper extremity motion during manual wheelchair propulsion in men with different levels of spinal cord injury. *Gait & posture*, 10(3), 223-232.
- Phang, S. H., Martin Ginis, K. A., Routhier, F., & Lemay, V. (2012). The role of self-efficacy in the wheelchair skills-physical activity relationship among manual wheelchair users with spinal cord injury. *Disability and rehabilitation*, 34(8), 625-632.
- Rao, S. S., Bontrager, E. L., Gronley, J. K., Newsam, C. J., & Perry, J. (1996). Three-dimensional kinematics of wheelchair propulsion. *IEEE Transactions on Rehabilitation Engineering*, 4(3), 152-160.
- Rodgers, M. M., Keyser, R. E., Gardner, E. R., Russell, P. J., & Gorman, P. H. (2000). Influence of trunk flexion on biomechanics of wheelchair propulsion. *Journal of rehabilitation research and development*, 37(3), 283-296.

Shimada, S. D., Robertson, R. N., Bonninger, M. L., & Cooper, R. A. (1998). Kinematic characterization of wheelchair propulsion. *Journal of rehabilitation research and development*, 35(2), 210-218.

Sisto, S. A., Druin, E., & Sliwinski, M. M. (2008). *Spinal Cord Injuries-E-Book: Management and Rehabilitation*. Elsevier Health Sciences.

Sporner, M. L., Grindle, G. G., Kelleher, A., Teodorski, E. E., Cooper, R., & Cooper, R. A. (2009). Quantification of activity during wheelchair basketball and rugby at the National Veterans Wheelchair Games: A pilot study. *Prosthetics and orthotics international*, 33(3), 210-217.