NEUROMECHANICS OF WALKING AND SLIPPING USING MUSCLE SYNERGY ANALYSES

A Dissertation

by

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ABSTRACT

Falls have numerous adverse effects on the economy and society. Slipping is the main contributor to falling. However, some individuals possess the ability to recover upon slipping while some do not. This dissertation studied the early indicators of falling in different individuals and compared them between mild slippers and severe slippers. The studied variables (sagittal angular moment (*H*), Center of Mass height, and the duration of single and double stance) all failed to show significant differences between the two severity groups before slip initiation, while the all differed after slip onset. *H* showed the earliest deviations among variables, indicating its importance in slip control. A muscle synergy (a united group of muscles that may act as a building block of the Central Nervous System in motor control) approach was chosen to study walking and slipping. Four synergies were identified for slipping, from which two were common between walking and slipping. Next, muscle synergies of slipping were compared for different severity groups. Severe slippers' slipping synergies differed from mild slippers', showing a probable difference in their motor control. Severe slippers were unable to replace their slipping limb as effectively. Upon finding post-slip differences, the walking behavior of mild and severe slippers were compared using muscle synergies. The walking muscle synergies also differed between mild and severe slippers, suggesting a difference in their gait control, causing their inclination to slip severely and fall. The results suggested severe slippers had a deficiency in decelerating their swing limb at the terminal swing phase of the gait and had higher dorsiflexion upon their heel strikes. Lastly, a muscle synergy approach was used to track the gait improvements of ADS patients upon surgical alignment. While their clinical gait variables showed improvements, they also required a higher number of synergies for walking, suggesting a more advanced gait control. Also, the entropy of muscle synergies were compared before and after surgery. Lower entropy is associated with more deterministic control. Upon the gait improvements, the entropy of the walking muscle synergies showed a significant decrease, proving it as a novel tool to track motor control enhancements in rehabilitation.

DEDICATION

To Zahra and Hasan, whose sacrificial care made this possible.

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All other work conducted for the dissertation was completed by the student independently.

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NOMENCLATURE

CNS	Central Nervous System
COM	Center of Mass
H	Angular momentum
E	Entropy
PHS	Peak Heel Speed
MH	Medial Hamstring
TA	Tibialis Anterior
RF	Rectus Femoris
MG	Medial Gastrocnemius
EO	External Oblique
GM	Gluteus Maximus
MF	Multifidus
ES	Erector Spinae
VL	Vastus Lateralis
NS	non-slipping (limb)
S	slipping (limb)
FFA	Foot-floor angle
SD	Standard Deviation
PVC	Polyvinyl Chloride
L	Leading

T	Trailing
VAF	Variance Accounted For

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1. INTRODUCTION

1.1. The true cost of falling

Falls pose a drastic negative impact on the U.S. economy and population. Injuries due to slips, trips, and falls cause the U.S. to sustain damage of $180 billion annually (Beschorner et al., 2016). Falls affect older adults notoriously. Older adults fall more frequently and often suffer from more severe consequences due to pre-existing health conditions and complications, where up to about 60% of the community-dwelling older adults fall each year. Unfortunately, up to 47% of these falls that come to medical attention are eventually fatal (Chisholm and Harruff, 2010; Layne and Pollack, 2004; T. E. Lockhart et al., 2003; Rubenstein and Josephson, 2002), and 75% of all fall-related deaths occur in persons older than 65 (Rubenstein and Josephson, 2002).

"Falls on the same level" impact the young and working population as the leading contributor to the days-away-from-work cases, causing about a third of the total non-fatal occupational injuries. Also, in 2015, injuries caused by slips, trips, and falls were the second most common cause of fatal occupational injuries in the U.S. (Bureau of Labor Statistics US Department of Labor, 2016a) causing 17% of the fatal injuries (Bureau of Labor Statistics U.S. Department of Labor, 2016a, 2016b). According to Layne and Pollack (2004), "fall on the same level," was primarily triggered by slip. Slipping, tripping, and stumbling were the main causes of 64% of all falls in the US (Courtney et al., 2001). Consequently, slipping was reported to be the main contributor to fall initiation (Courtney et al., 2001; Di Pilla, 2009; Gao and Abeysekera, 2004).

Furthermore, injuries caused by slips, trips, and falls have increased by 10% from 2013 to 2014 (Bureau of Labor Statistics US Department of Labor, 2015a), and still showing a growing trend (Bureau of Labor Statistics U.S. Department of Labor, 2015a, 2015b). Considering the prevalence and the increasing trend of fall-related injuries coupled with slipping being the main cause of falling, understanding the slip recovery process is of paramount importance in fall prevention.

1.2. The critical role of slipping

Although slips cause most of the falls, not all slips result in a fall. “Severe slips” (sometimes referred to as hazardous slips) are more likely to result in falls compared to “mild slips.” To assess the severity of a slip, several studies have tried to introduce different measures such as Peak Heel Speed (PHS) or slipping distance as the key factors in assessing slip severity. (T. Lockhart et al., 2003; Perkins, 1978; Strandberg and Lanshammar, 1981). Lockhart et al. (T. Lockhart et al., 2003) claimed that slipping distances and speeds higher than 3.91 cm and 1.44 m/s should be considered severe in younger adults.

Interestingly, being a “severe slipper” can be considered as a characteristic of an individual (T. Lockhart et al., 2003; T. E. Lockhart et al., 2003). Lockhart et al. (T. E. Lockhart et al., 2003) found that even though younger and older adults have the same potential for slip initiation, older adults slipped more severely compared to younger adults. This fact indicates that severe slipping is also highly related to the post-slip-initiation motor responses rather than motor behaviors before the slip initiation.

Consequently, the identification of severe slippers and the prevention of severe slips may have significant potential for targeting interventions and preventing falls.

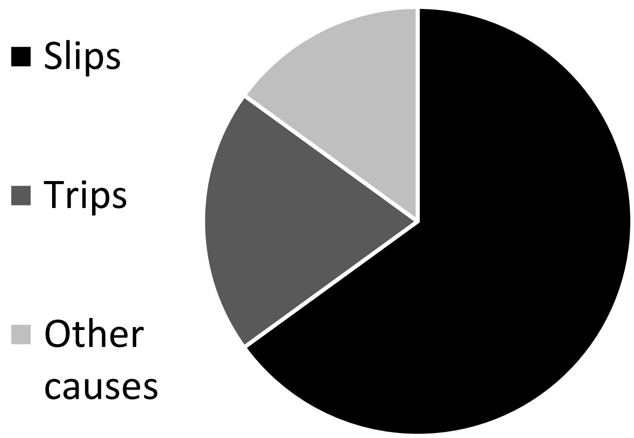

Figure 1 Main contributors to falling

While slips can be classified into two types, researchers agree that in general, maintaining balance during slipping requires fast and appropriate corrective responses (Cham and Redfern, 2001; Marigold and Patla, 2002; Tang and Woollacott, 1998). There were also several studies in which muscle patterns during slipping was investigated. Qu et al. (2012) found relations between muscle activation patterns and successful slip recoveries as well as failed slip recoveries. Studies also examined the latencies and the role of the muscle activation patterns of both lower and upper extremities on the recovery from a slip, implying intralimb and interlimb coordination strategies in maintaining balance (Marigold et al., 2003; Marigold and Patla, 2002;

Moyer et al., 2009). Also, studies focused on the stance leg after a slip found a minimum latency of 175ms in muscle activations in major leg muscles (Chambers and Cham, 2007), and a 200ms latency for restorative moments (Cham and Redfern, 2001). However, studying the interlimb coordination can be challenging using conventional EMG analysis and sometimes resulted in contradictory conclusions (Chambers and Cham, 2007; Marigold et al., 2003; Qu et al., 2012). Also, another challenge in studying interlimb coordination is that the kinematic and kinetic variables are prone to inter-subject differences such as height and weight.

1.3. A 'Muscle Synergies' approach

Studies suggest that the Central Nervous System (CNS) might control muscles using a low-dimensional organization of co-activated muscles, or muscle synergy (d'Avella et al., 2003; d'Avella and Bizzi, 2005; Overduin et al., 2012; Ting and Macpherson, 2005). In other words, muscle synergies are a group of co-active muscles recruited by a single control input, or activation coefficient (d'Avella and Bizzi, 2005; Ting and Macpherson, 2005). While the concept of muscle synergies stays consistent among different researchers, they have tried different mathematical notations and terminologies to describe them. Some studies have used a vector with constant ratios for each muscle to present muscle synergies and a constant activation pattern for each synergy (Neptune et al., 2009; Ting and Macpherson, 2005), while other studies used time-varying muscle synergies and an activation pattern with an adjustable time delay (d'Avella et al., 2003; d'Avella and Bizzi, 2005). Not only have scientists different views about mathematical notations of synergies, but also they have different opinions about the existence of such a lower-dimensional modular organization to control and describe motor-tasks (de Rugy et al., 2013; Tresch and Jarc, 2009). Thus, a significant number of studies aimed to examine the muscle synergy hypothesis. Proponents of the synergy hypothesis as a descriptive tool have shown that synergies can be efficient in explaining the variability observed in the EMG signals for a vast range of activities in different animals; such as human gait, hand posture of macaques, the posture of cats, and kicking in frogs (d'Avella et al., 2003; Neptune et al., 2009; Overduin et al., 2012; Ting and Macpherson, 2005). Muscle synergy as a neural control mechanism has been substantiated by several studies showing that electrical microstimulation on different parts of the CNS results in multi-degree-of-freedom motor behaviors and invariant postures, which may indicate the presence of a coupling between the joint movements and muscle patterns (Overduin et al., 2012; Steele et al., 2015; Zimmermann et al., 2011).

On the contrary, the opponents of this hypothesis state that the CNS is more likely to use an uncontrolled manifold or an optimal control schema to perform and control the motor-tasks (Todorov and Jordan, 2002; Valero-Cuevas et al., 2009) based on their experiments. Although the opponents raised deep questions about muscle synergies with their research, other studies have shown that the uncontrolled manifold and optimal control methods often result in extraction of structures that are highly similar to muscle synergies (Danna-Dos-Santos et al., 2009; de Rugy et al., 2013; Krishnamoorthy et al., 2004; Todorov, 2004). This fact makes both endorsers and adversaries agree that muscle synergies are at least an effective descriptive tool to explain the variations observed in different motor-tasks using a lower-dimensional organization. As a result, the intralimb and interlimb coordination happening in response to slipping incidents (Marigold et al., 2003; Moyer et al., 2009) may be better represented using muscle synergies resolving the issue with the kinematic studies and conventional EMG analysis. Muscle activation patterns during normal walking have also been described by muscle synergies. Interestingly, studies showed each of the walking muscle synergies corresponds to a known sub-function of the gait cycle (e.g., propulsion) (Clark et al., 2008; Neptune et al., 2009).

1.4. Scope of this dissertation

We aim to utilize muscle synergies to study walking and slipping. In chapter 2, we use kinematic variables to find differences between mild and severe slippers. As kinematic and kinetic variables fail in labeling mild and severe slippers, we move to chapter 3 to initiate the usage of muscle synergies to study slipping. Chapter 3 covers the similarities between muscle synergies of walking and slipping. Next, we move to study the differences in slipping muscle synergies of mild and severe slippers in chapter 4. In chapter 5, we study differences in walking of mild and severe slippers to look for earlier identifiers. At last, in chapter 6, we utilize muscle synergies to track the improvements of ADS patients to verify the potential of the muscle synergies in improving the quality of life in different patients.

2. ANGULAR MOMENTUM MAY DICTATE THE SLIP SEVERITY IN YOUNG ADULTS[1]

2.1. Introduction

Studies have argued that upon slipping, the Central Nervous System (CNS) has to react with appropriate signals to avoid falling and retain balance (Cham and Redfern, 2001). Obviously, failing to provide proper responses to slip would result in falling. To provide a safer experiment environment to study slips, scientists enforced usage of harness systems and have developed different indicators of falling instead of an actual fall. These measures mainly consisted of a load cell average force during falling, percentage of body height drop while slipping, slipping distance, and peak slipping velocity where some of them were reported to predict falls with 90-100% accuracy (T. Lockhart et al., 2003; Perkins, 1978; Strandberg and Lanshammar, 1981; Yang et al., 2013; Yang and Pai, 2014, 2011). For instance, Lockhart et al. (T. Lockhart et al., 2003) claimed that slippers can be classified into mild and severe slippers by the peak heel speed after slipping to predict their falls. Specifically, severe slips are described as slips in which the peak heel speed exceeds 1.44 m/s and severe slippers are more prone to fall (T. Lockhart et al., 2003). Conversely, mild slips are less dangerous and mild slippers can recover from slips without falling compared to their severe slipper counterparts.

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Additionally, prior studies have shown that one's risk of fall is affected by both pre-slip control (gait control) and post-slip response (slip control) (Moyer et al., 2006; Nazifi et al., 2017a, 2017b; Parijat and Lockhart, 2012). In other words, mild slippers possess different control techniques for both walking and slipping compared to severe slippers. Identification of such differences in kinematics, dynamics, and control of walking and slipping between mild and severe slippers would facilitate diagnosis of severe slippers (who naturally have a higher risk of fall). Consequently, numerous studies have tried to identify discrepancies based on individuals' fall/recovery outcome and/or slip severity. These studies targeted a wide range of variables to detect differences between fallers and non-fallers (i.e. persons who recover from slips), such as kinematic variables (e.g., foot-floor angles, slipping distances) (Honeycutt et al., 2016; Moyer et al., 2006; Nazifi et al., 2020; Yang et al., 2009), kinetic variable (torques) (Beschorner and Cham, 2008; Cham and Redfern, 2001), and neuromuscular variables (activation onsets) (Nazifi et al., 2017b; Qu et al., 2012; Sawers et al., 2016).

While numerous studies tried to find potential associations between slip severity and kinetic and kinematic variables, there are still several critical variables that have not been studied and compared between mild and severe slippers. More importantly, the causal nature of these associations is still unclear. For instance, numerous studies have studied the lower extremity kinematics and kinetics and their association to severe slipping (Beschorner and Cham, 2008; Cham and Redfern, 2002; Lockhart and Kim, 2006; Marigold and Patla, 2002; Moyer, 2006; Moyer et al., 2009, 2006; Parijat et al., 2015). Despite the important role upper body kinematics play during slip control, few studies have examined the association of the slip severity with upper extremity kinematics (elbow and shoulder joint angles) (Marigold et al., 2003). Also, while several studies have argued that COM height and its stability play a key role in prediction of a slip outcome (Yang et al., 2009; Yang and Pai, 2014), very few studies have compared the COM height based on slip severity to find potential differences. In addition to COM height, angular momentum (denoted by H from engineering literature), a quantity representing the movement of rotation of an object, is also known to be of importance in gait. Different studies have examined angular momentum manipulation for human gait (Herr and Popovic, 2008; Hinrichs, 1987; Pijnappels et al., 2004; Simoneau and Krebs, 2000; Vistamehr, 2014). Nevertheless, no studies have attempted to compute and compare H between mild and severe slippers. Specifically, since slips mostly result in backward falls (Qu et al., 2012), studying angular momentum in the sagittal plane (backward/forward falls are equivalent a rotation in the sagittal plane) is of our interest. Lastly, the duration of single and double support phase of the gait and slipping is another relevant gait parameter (Moyer et al., 2009; Tsai and Lin, 2013) that has never been compared between mild and severe slippers. We argue that comparing these variables among individuals with different slip severity may address the gap in our knowledge and find possible associations. Also, since COM height has been used as the main indicator of the falls in slip studies (Yang et al., 2009; Yang and Pai, 2014, 2011), any variable that shows a time-lag in its deviations compared to COM height, will be ruled out from having a causal relationship with falls while a time-lead over COM height deviations would increase the likelihood of causal nature of that variable to falls.

The objective of this chapter is to i) compare the shoulder and elbow joint angles, the COM height, sagittal angular momentum (H), and length of single/double support between mild and severe slippers, ii) compare the timing of the deviations relative to changes in COM height to find the potential cause of the severe slipping. We hypothesize that these measures would differ between mild and severe slippers, indicating the different motor control in kinematics and kinetics of walking and slip in both mild and severe slippers. Also, we hypothesize that some of the variables would deviate sooner than COM height drop (i.e., an indicator of falls), suggesting a potential causal relationship to severe slipping, and hence, falling.

2.2. Methods

2.2.1. Participants

Twenty healthy young adults age (11 males and 9 females, age mean ± SD: 23.6 ± 2.52) participated in this experiment at the University of Pittsburgh. Participants signed a written consent form before participation and were excluded in case of any gait disorder history/condition. The de-identified data were transferred to Texas A&M University for further analysis. Both the experiment and the data analysis were approved by the University of Pittsburgh's IRB and TAMU IRB according to their Human Research Protection (HRP) regulations and the Declaration of Helsinki.

2.2.2. Measurements, Experimental protocol, and Data Processing

Participants were asked to walk in a ten-meter pathway at their comfortable speed. They were told that the floor (vinyl-composite tile) was dry such that they were not anticipating any slips. After two or three walking trials, a slippery contaminant (75% glycerol, 25% water) was applied to the middle of the walkway (to secure at least four normal steps before slipping) to generate and collect a slip trial data (Figure 2). Participants looked away from the walkway and listened to music with headphones between each trail to minimize awareness of the contaminant. Participants donned an overhead harness for their safety throughout the trials. PVC-soled shoes in the participants' sizes were provided for all participants. During the first few walking trials, the location of the starting point was adjusted to align the participants' foot placement with the slippery surface.

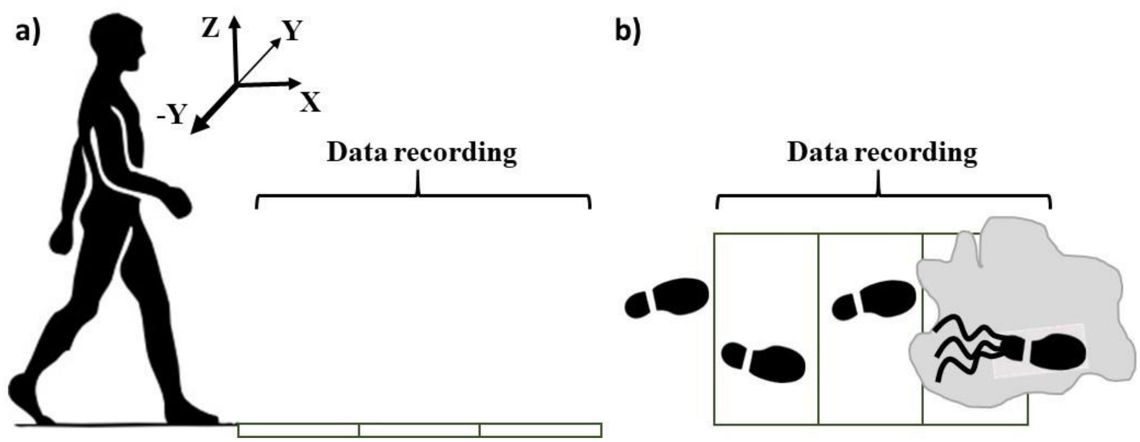

Figure 2 Experimental setup, contamination, and foot placing during the experiment. (reprint from Nazifi et al., 2020)

A set of 79 reflective markers was placed on anatomic bony body landmarks (Moyer et al., 2006) to collect the kinematics at 120 Hz (Vicon 512, Oxford, U.K.). Participants' weight and height were recorded. The markers' data were low-passed filtered (at 10 Hz) with a second-order Butterworth filter (MATLAB, MathWorks, Natick, MA) (Beschorner and Cham, 2008). Using the heel marker information after the slip trial, participants were classified into mild and severe slippers based on their PHS (T. Lockhart et al., 2003; Nazifi et al., 2017a) to investigate their inter-group differences. Next, based on heel and toe markers, the heel strike and toe-off were calculated, and the corresponding double/single support phase of the gait were measured for each individual. The filtered markers data were also used in generic code (MATLAB, R2017a MathWorks, Natick, MA) to compute limb and joint positions (for both upper and lower extremity) on both right/leading/slipping side (L) and left/trailing/non-slipping side (T). The rotations of the upper extremity joints, the head kinematics, and the hands' kinematics were not studied as they have little to no effect on the angular momentum. Using anthropometric relative joint and COM positions (Winter, 1990), the center of mass of each limb was calculated and used to measure the position and velocity of the whole body's center of mass. The center of mass was then normalized using participants' heights and presented as a height percentage. Finally, using the same segmental analysis method as COM, the angular momentum of the body was calculated by multiplying the relative velocity of each limb compared to COM to its relative distance to COM and its mass as described in :

$$H = \sum_{i=1}^{10} m_i\left(r_{com/i} \times v_{com/i}\right) + \ I_i\omega_i$$

Equation 1

where m_i is the mass of the i-th limb, and $r_{COM/i}$ and $v_{COM/i}$ are the relative distance and velocity of the i-th limb with respect to the whole-body COM and I_i and ω_i are the mass moment of inertia and absolute sagittal plane angular velocity, respectively. According to our reference frame (Figure 2), a positive angular momentum indicates a general backward rotation, whereas a negative H shows a forward rotation (Neptune and Mcgowan, 2011; Vistamehr, 2014). Moreover, H is a function of COM velocity (m/s), the relative distance of each limb to whole-body-COM (m, a function of participant's height), and mass (kg). Hence, a unitless/non-dimensional H was created by dividing the original H to one's average COM velocity, mass, and height (Herr and Popovic, 2008). This would remove subjective differences and make unitless H a more appropriate candidate to present inter-participant differences.

To eliminate the effect of different gait speeds, the gait cycle was normalized to 100 points for each participant to facilitate a point-to-point inter-participant comparison. The comparison was made between a full gait cycle (0% to 100%) for normal walking and an additional 30% of the gait cycle through slipping (100% + 30% = 130% of gait cycle time). According to existing literature, 30% of gait cycle time is enough to capture the slip response of the participants (Hur and Beschorner, 2012). Considering the slip to happen at time = 0%, the prior full gait cycle would have happened from -100% to 0%. Also, the slipping would happen starting from 0%, and the analysis continued until 30%. The upper body kinematics, the z component of the COM (COM height), and the y component of H (angular momentum in the sagittal plane) (Figure 2) were compared between the mild and severe slippers at each percentage of the gait and slipping (i.e., 130 data points). Since double stance happens later in a gait cycle, we studied this variable for a full gait cycle before slip initiation (i.e., from -100% to 0%) and a full gait cycle time length after slip initiation (i.e., from 0% to 100%, a total of 200% instead of 130%).

The data were checked for normality and homogeneity of variance (using Shapiro Wilk and Levene's test, respectively). Statistical Parametric Mapping (SPM) at significance of 0.05 was used (MATLAB, MathWorks, Natick, MA) to identify the regions of the gait cycle where the upper body kinematics, H, and COM height deviate significantly between groups. SPM is a statistical technique that can be used to examine differences observed in time-series data or spatial data. Unlike *t*-tests, SPM is not based on Gaussian theory and is based on the Random Field theory. Since our data is continuous through time, SPM can be an effective replacement for running multiple statistical tests to avoid the inflation of Type 1 error due to the multiple comparisons (Friston et al., 1995; Pataky et al., 2013). Moreover, an independent *t*-test was used to detect statistically significant differences in the single/double stance duration between mild and severe slippers at a significance of 0.05 (SPSS v21, IBM, Chicago, IL) as this variable is not considered a time-series and only presents the time of the transition from single to double stance (The variances were also checked and in case of significant difference in variance, a Welch *t*-test was used instead of an independent *t*-test).

2.3. Results

Eight of the twenty participants were found to be severe slippers due to their PHS, while the rest were mild slippers. Statistical tests showed no gender, age, or sex-related association for slip severity (p-value > 0.05) (Table. 1). The upper body kinematics were extracted (Figure 3), and the statistical comparison indicated that there were no significant inter-group differences in the upper body kinematics both before and after the slip initiation (p-value > 0.05), meaning that upper extremity differences between mild and severe slippers were modest.

Mean±SD	PHS (m/s)	Age	Mass (Kg)	Height (cm)	Sex (M/F)
Mild	0.63±0.25	24.17±2.79	68.41±11.89	171.75±8.59	5/7
Severe	1.87±0.27	22.75±1.48	70.00±11.37	175.19±7.57	6/2
p-value	<0.001	0.228	0.780	0.395	0.142

Table 1 Demographics of different severity groups and the resulting statistical analysis. No significant difference in any of the variables at the level of 0.05 (p-value > 0.05), except PHS. Pearson's Chi squared test was used for Sex while independent t-test was used for the others. (reprint from Nazifi et al., 2020)

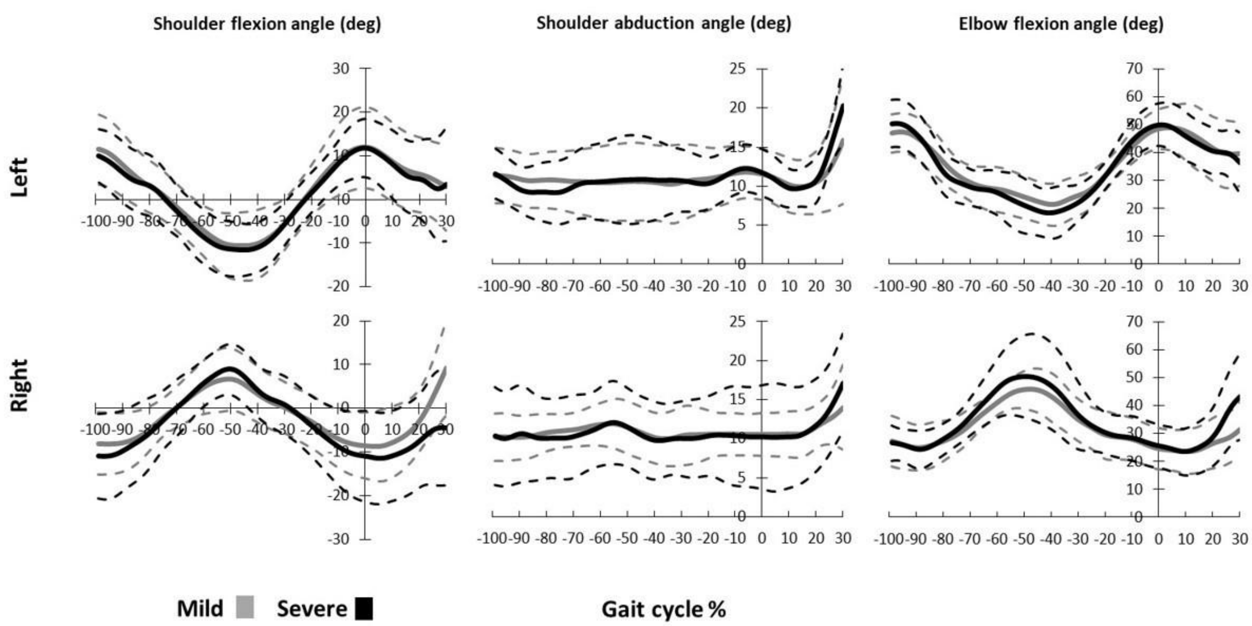

Figure 3 Upper body kinematics for mild and severe slippers for a full gait cycle prior to slip (-100% to 0%) and 30% of the gait cycle time length during slipping. The bold lines represent the average values, while the dashed lines indicate the standard deviation. (reprint from Nazifi et al., 2020)

The SPM analysis indicated that mild and severe slippers differ in their COM height and dimensionless sagittal angular momentum after slip initiation. The independent *t*-tests showed that the duration of single/double support differs in different severity groups following slip initiation. Preceding the heel contact on slippery contaminant (i.e., walking), the mild and severe slippers did not differ in COM height; however, from 24%-30% of the gait cycle into slipping, COM height became significantly higher in mild slippers (p-value < 0.05) (Figure 4).

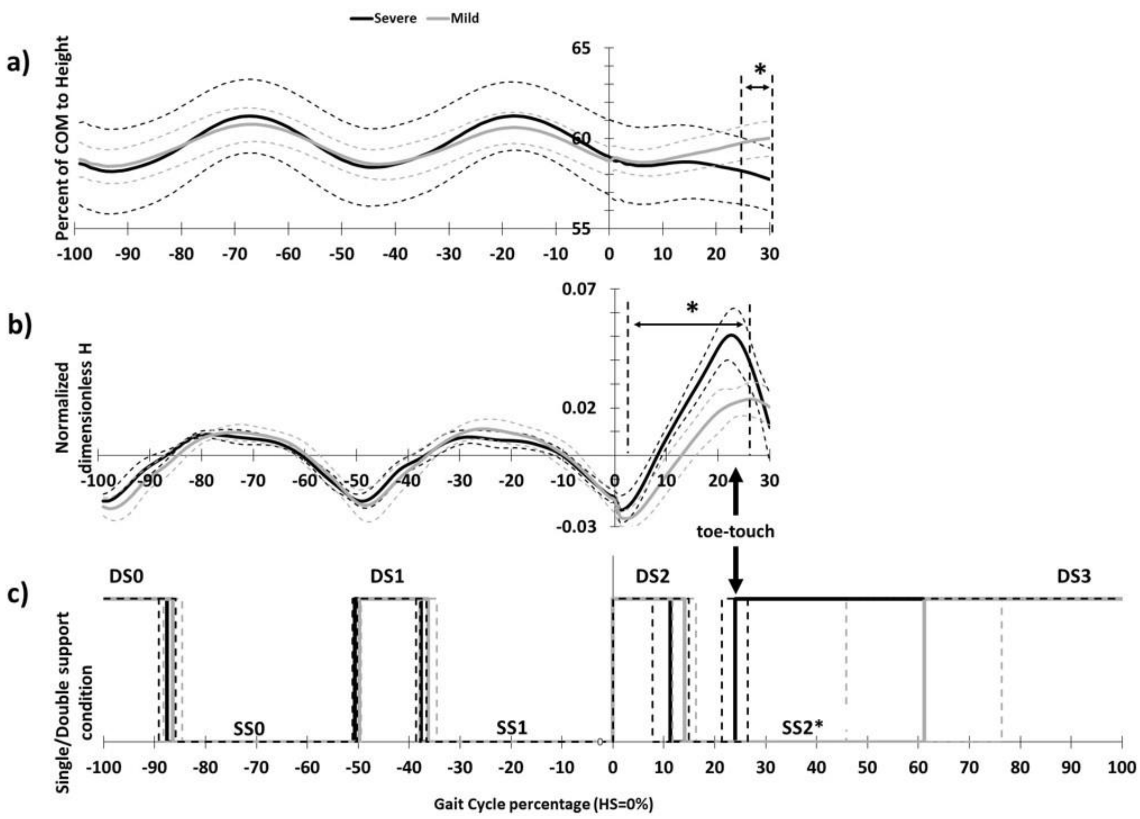

Figure 4 COM height, sagittal H, and single/double support phase duration for mild and severe slippers. The bold lines represent the average values, while the dashed lines indicate the standard deviation. Asterisks indicate significant difference. (reprint from Nazifi et al., 2020)

Moreover, for the dimensionless sagittal angular momentum, mild and severe slippers showed a significant difference from 4%-26% into slipping (p-value<0.001) (Figure 4). Lastly, statistical analysis indicated that severe slippers have a shortened single stance phase compared to their mild slipper counterparts after slip initiation (p-value<0.001) (Figure 4, SS2).

2.4. Discussion

The significant discrepancies in COM height post-slipping could be interpreted as a strong correlation between slip severity and deviation of COM height. In severe slippers, COM height was significantly lower following a slip compared to normal gait, while mild slippers maintained their post-slip COM height somewhat similar to COM height during normal walking (Figure 4). A sudden decrease in the COM height was associated with severe slipping and hence, falling. Consequently, controlling COM could be a useful yardstick in the identification of people with a high risk of falling and may result in the development of rehabilitative/preventative anti-fall devices. This finding is consistent with previous articles that claimed the height drop could be used as an indicator of falls in the presence of harness (Yang and Pai, 2011). However, another possible interpretation for the observed deviation between pre-slip and post-slip COM height (in severe slippers) can be a potential safety strategy. In other words, it is possible that due to the severe slip, the CNS changes its strategy from “maintaining the COM height” to deliberately “lowering the COM” in order to take a safer fall. This interpretation, however, requires further investigation.

Furthermore, the severe slippers experienced a shortened single stance phase following a slip. “Toe-touch” response is a known method to increase the base of support during slipping (Marigold et al., 2003; Nazifi et al., 2017a). Toe-touch is responsible for disrupting the gait while slipping to place the swing limb on the ground and is beneficial in reestablishing a wider base of support, providing weight support, and regaining balance. However, it seems that this strategy is only used in more severe slips since all mild slippers avoided using this strategy while slipping and continued countering slip on one limb without a toe-touch. Considering this strong association, it is likely that only severe slips required this response to maintain their balance. A more focused study is required to examine this hypothesis and to see if a toe-touch response has a higher trigger for its activation, using an accelerating treadmill that could induce slips with desired intensities.

Analysis of the sagittal angular momentum showed that mild and severe slippers differ in their H early after the onset of the slip at 4% until 26% of slipping (*p*-value<0.001, Figure 4). Human gait exhibits a periodic angular momentum pattern (Figure 4), and the gait pattern has evolved in a way to match the dynamics of the body while walking, restrain the H by countering the upper body movements (i.e., moving limbs in opposite directions), and using the impact of heel strikes to continue the gait cycle (Herr and Popovic, 2008; Vistamehr, 2014). Modulating the H values throughout walking is of crucial importance (Herr and Popovic, 2008; Vistamehr, 2014). According to our findings, it seems that severe slippers could not modulate H or counter their excessive body rotation caused by slipping from 4%-26% into slipping. On the other hand, mild slippers have been able to maintain their angular momentum significantly lower (and more similar to normal walking), which made them more successful in maintaining their balance following a slip.

Association of an excessive H with severe slipping and falling suggests that falling does not only happen as a vertical COM drop but also as a backward rotational. More importantly, the deviation observed in H values (onset at 4% into slipping, Figure 4) had a significant time-lead over the significant drop observed in COM height (at 24%, Figure 4). As mentioned before, the COM height drop has been introduced as one of the main indicators of falls (Yang and Pai, 2014, 2011, 2010). Since the deviations in H happen before the main indicator of falling (i.e., COM height), we suspect the angular momentum of the body to be an earlier indicator of falls and one of the critical variables in controlling slips. This finding matches the existing literature that showed a higher hip flexion angle and knee extension angle to be associated with more severe slips (Moyer, 2006) as both contribute to a higher backward angular momentum and hence, a potential backward falling.

In postural balance studies, it has been shown that the CNS has the potential to choose different control strategies and employ them for situations with different intensities (i.e., ankle strategy, hip strategy, stepping strategy (Hur et al., 2010)). Hence, one may speculate that the CNS would react differently to slips with different severities as well (Moyer, 2006). We argue that angular momentum can potentially be a deciding variable in post-slip control, meaning that the CNS may choose different control methods based on H value. This hypothesis is substantiated by the pattern observed in the single/double support phase duration. As mentioned, only the severe slippers utilized a 'toe-touch' response to their slips. This 'toe-touch' response (completed at 23%, Figure 4) could not have been triggered by COM height drop due to its time-lead (onset at 24%, Figure 4). Hence, we suggest that this toe-touch response may be enforced by the CNS to constrain and regulate the excessive H because angular momentum can only be changed by the exertion of an external moment around the body's COM (which is done by the toe-touch). This is observable in Figure 4,b-c, where the excessive positive H values in severe slippers (i.e., backward falling) dropped significantly following their toe-touch response that widens the base of support to provide a moment to prevent backward falling. Further validation of our theory about H and slip control will be an open question for examination for our future studies. Also, we are interested in investigating the angular momentum in other planes in our future studies to substantiate the current findings further.

The upper extremity kinematics stayed consistent with the previous kinematic studies. An arm elevation strategy, as described by (Marigold and Patla, 2002), was deployed by all participants (i.e., Figure 3, shoulder abduction happening from 0% to 30%) in response to a slip. This strategy helps to move the COM forward to prevent backward falls. Hence participants tend to move their arms to a more anterior and superior position (i.e., shoulder abduction and flexion, Figure 3, from 0% to 30%) to avoid falls (Tang et al., 1998; You et al., 2001). However, there were no discrepancies detected between the upper body kinematics for different severities. This indicates that the upper extremity kinematics and control during normal walking and early slipping (up to 30% of the cycle) has little to no significant effect on the slip severity outcome, although aging has shown to be an essential factor in the arm reaction and slip outcome (Merrill et al., 2017). Nonetheless, considering our theory of the importance of H, we suspect the rapid, countermovement of the hands to be a measure to lower whole-body angular momentum. This fact and the timing of this drop in H stays consistent with existing literature that suggests upper extremity movements as a strategy to prevent falling (Marigold et al., 2003; Nevisipour et al., 2019).

There were a few limitations associated with this chapter. First, despite the efforts to hide the audible and visual clues of the contaminated surface, the possibility of anticipation of the slip still exists. Moreover, the scope of this chapter is limited to only 30% of the gait cycle following the slip initiation. Also, this chapter is limited to the uncertainty and the accuracy provided by the motion capture system rather than the force plate system. Lastly, this chapter did not consider the timing of angular momentum deviations relative to other biomechanical variables (i.e., foot force) (Beschorner et al., 2013) that also deviate early after slipping onset. Thus, the importance of H relative to the other parameters is currently unknown.

2.5. Conclusion

This chapter examined several kinematic and dynamic measures in mild and severe slippers to identify the inter-group differences. We found that mild and severe slippers differ in their control of COM height, sagittal angular momentum, and duration of the single/double support phase, mainly after slip initiation. Also, the time sequence of the deviations substantiated angular momentum to be a relevant variable in controlling slips. These findings can substantiate that healthy young mild and severe slippers have no difference in their pre-slip control, and the higher severity is potentially caused by their post-slip response and probably their angular momentum regulation. Such studies are useful in the identification of the underlying causes of severe slipping, which is a primary step in fall prevention. Further studies are required to examine these variables in older adults to generalize the findings of this chapter possibly.

3. SHARED AND TASK-SPECIFIC MUSCLE SYNERGIES DURING NORMAL WALKING AND SLIPPING[2]

3.1. Introduction

Muscle synergies could potentially be shared across activities. Studies done on animals (e.g. frogs) (d'Avella et al., 2003; d'Avella and Bizzi, 2005) suggested that a few synergies were being shared between walking, jumping, and swimming. As muscle synergies of a motor-task correspond to its physical sub-tasks, having the same mechanical goals and sub-functions in different motor-tasks may result in having the same structure of muscle combinations and ratios, or shared muscle synergies. In other words, if two different motor-task include a common mechanical sub-task, it is likely for a common muscle synergy to appear in both of the synergy sets. On the contrary, a task-specific mechanical goal is more likely to be reflected in a task-specific muscle synergy, which will not appear in the synergies of any other motor-task. For example, Chvatal et al. (2013) found that there exist shared synergies between unperturbed and perturbed standing and walking as well as the other non-shared, task-specific synergies. Also, Martino et al. (2015) investigated the synergies of normal walking as well as unstable gait conditions such as walking on a slippery surface and studied similarities of those motor behaviors. However, to our knowledge, no studies have investigated synergies

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during slipping (as an exclusive motor-task) and their existence, nor have any comparisons been made between synergies of normal walking and slipping to date. Thus, a muscle-synergy perspective may provide insights into whether a modular control strategy (synergies) is being used in response to slips or not, and if so, how similar or dissimilar those synergies are compared to normal walking synergies. Interlimb coordination could also be studied by extracting muscle synergies of both legs during these motor-tasks. Muscle synergies could be useful in diagnosis and rehabilitation process (Roh et al., 2013). As synergies could present a sub-task of the main motor-task, extraction of slipping muscle synergies helps determining the sub-tasks of slipping. Also, activation coefficients of muscle synergies are informative as they represent the timing of activation for muscle synergies, and the corresponding sub-tasks. For the case in which a shared synergy exists, similarities in activation coefficients of the shared synergy would indicate an identical mechanical goal and timing performed in both motor-tasks. Thus, a muscle-synergic approach could provide a foundation for a comparison between healthy participants' synergies and those of patients who are unable to recover from slips, hastening the diagnosis of impaired synergy and subsequently, identifying the impaired sub-task in them. It also may result in design of more efficient therapeutic interventions and targeted motor rehabilitation specifically intended to recover the malfunctioning sub-task (Dipietro et al., 2007; Roh et al., 2013).

The objective of this chapter is to examine and compare muscle synergies and time-series activation coefficients for two conditions: normal walking and slipping. We hypothesize that there exist both shared and task-specific muscle synergies between the two conditions. The rationale underneath this hypothesis is that a person may use a modularized lower limbs' control strategy while recovering from a slip, and some of these modules might be similar to those of normal walking. Accordingly, we hypothesize that the activation coefficients of the corresponding shared synergies that represent the timing of activation for these synergies would be similar for both normal walking and slipping.

3.2. Methods

3.2.1. Participants

Eleven healthy young adults (6 males and 5 females, and age range: 22-33 years) free of balance disorders participated in this study. Everyone whose age was outside of the range of 18-35 and who might have issues with normal walking, e.g., pregnant women, were excluded from the study. Participants provided informed consent prior to participation in the study and the study was approved by the University of Wisconsin-Milwaukee Institutional Review Board.

3.2.2. Measurements, Experimental protocol, and Data Processing

Participants were fitted with a safety harness and surface electromyography (EMG) electrodes (Trigno, Delsys, Natick, MA) for four bilateral muscles and were asked to walk on a floor with four force plates (BP400600, AMTI, Watertown, MA) embedded. Using force plates, the kinetic data was collected at 1000 Hz. Collected ground reaction forces were used later to detect heel contacts using visual techniques. EMG data was sampled at 1000 Hz from four muscles for each side: medial hamstring (MH), tibialis anterior (TA), rectus femoris (RF), and medial gastrocnemius (MG).

High-pass filtering and wrapping electrodes are commonly practiced to remove the possible movement artifacts. In this chapter, the electrodes were secured and stabilized using extra bandages around the electrodes to avoid artifacts due to the movement of the electrodes. Since the EMG sensors were wireless, cables might not contribute to movement artifact.

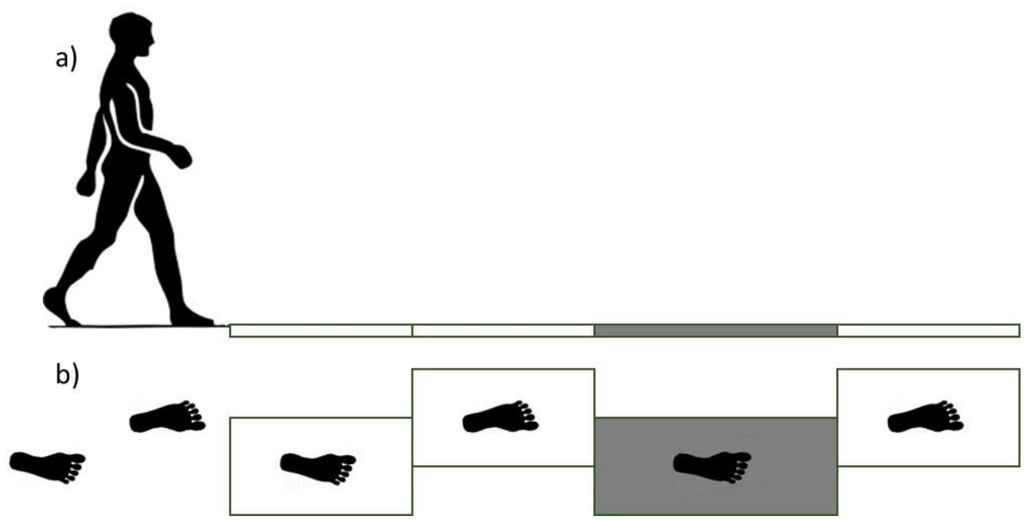


Figure 5 Side view of force plates (a) and top view (b). For the slipping trials, the third force plate was contaminated by a diluted glycerol. Note right and left foot strikes. (reprint from Nazifi et al., 2017b)

The force plates were place in the middle of a 12 m long pathway in order to ensure that at least 5 steps were taken before stepping on the force plates. Participants were asked to walk at their comfortable pace, step length, and cadence. Prior to the start of the gait trials, the participants' starting position was adjusted to ensure that the participants hit their right foot (leading foot, referred to as slipping/leading foot) on the third force plate (Figure 5). Participants completed five unperturbed walking trials on the dry floor followed by one unexpected slip trial on the contaminant. During the slip trial, the third force plate (Figure 5) was contaminated to be slippery via applying a diluted glycerol (90% glycerol and 10% water) solution on it without informing the participants. The third force plate was longer than the other force plates (0.4 m × 0.8 m compared to 0.4 m × 0.6 m) to minimize the risk that the participant would slip completely off of it during the slip (Figure 5).

Only the first interval of 300ms starting from heel strike on the third force plate was used for data analysis in both normal walking and slipping trials, since activation onset time of the EMG data for aforementioned muscles typically occur within the first 300ms or 50% of stance after heel contact (Cham and Redfern, 2002; Hur and Beschorner, 2012; Marigold et al., 2003; Marigold and Patla, 2002; Moyer et al., 2009). The EMG activities for the eight muscles were recorded and processed via a full wave rectification and low-pass filtering (fourth-order Butterworth, cut-off frequency at 30Hz) using MATLAB (v2014a, Mathworks, Natick, MA). Data were normalized to the maximum activation level among all trials within the same participant for each muscle. Finally, the data were integrated over every 10ms interval, resulting in 30 data points for the whole 300ms (Figure 8 and Figure 9). This interval was determined by d'Avella et al. (2003). As there existed several normal walking trials, the average of the all trials were used. Although averaging trials may affect the variance-covariance structure of the data, in order to avoid having different time step size and timing at each data point, the trials were averaged. For each participant, the resulting processed EMG data were then assembled into a matrix, M, that had 30 rows corresponding to each time interval and eight columns corresponding to each muscle (i.e., $M \in R^{30\times 8}$).

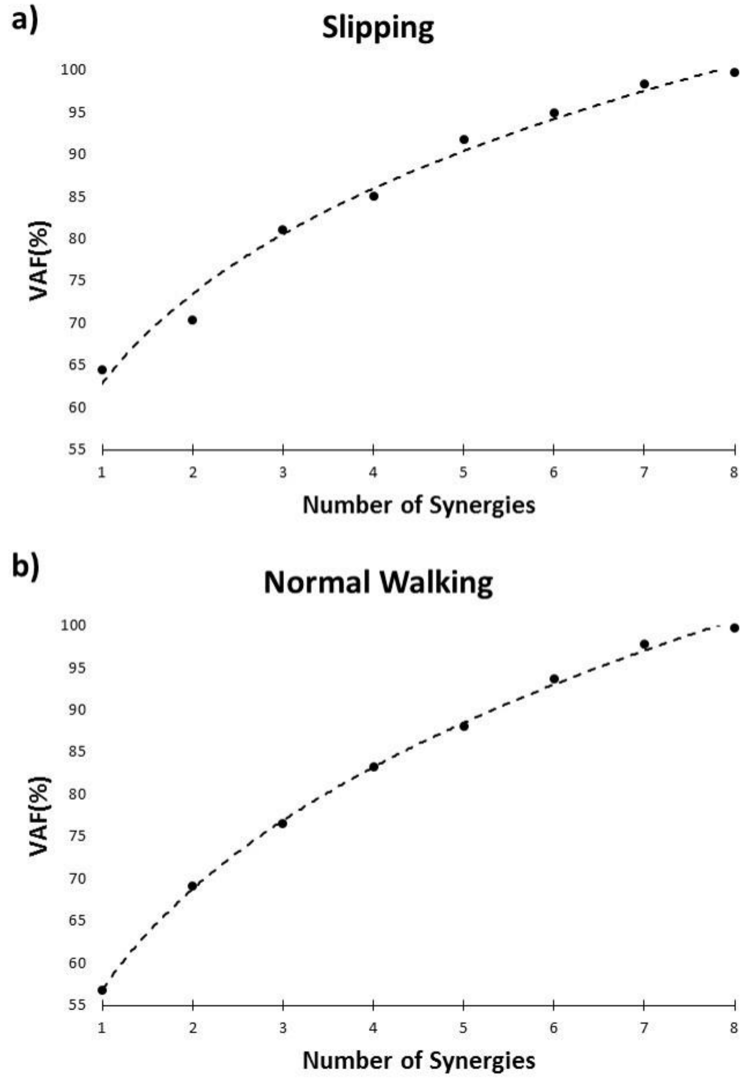


Figure 6 VAF versus number of synergies curve for slipping (a), and walking condition (b), based on the pooled data set. (reprint from Nazifi et al., 2017b)

3.2.2.1. Synergy extraction

Muscle synergy was considered to be a row vector, $w_i \in R^{1\times 8}$, where each of the elements corresponded to each muscle's contribution to build that specific synergy. Also, time-series activation coefficient of the corresponding synergy was noted with a column vector, $c_i \in R^{30\times 1}$ with each element corresponding to a time step (=30 in this chapter). Using the same iterative nonnegative matrix decomposition algorithm introduced by Ting and Macpherson (2005) (via MATLAB functions *fmincon* and *isqnonneg*), *n* muscle synergies ($W_{n\times 8} = \begin{bmatrix} w_1 \\ \vdots \\ w_n \end{bmatrix}$) and the corresponding *n* activation coefficients ($C_{30\times n} = \begin{bmatrix} c_1 & \cdots & c_n \end{bmatrix}$) were extracted for each participant's data during unperturbed walking condition and slipping condition, respectively. This algorithm identified the muscle synergies and time-series activation coefficients that best fit the resulting processed EMG data ($M_{30\times 8}$). Note that n denotes the number of extracted synergies and can vary from 1 to 8 (= total number of muscles).

$$M_{30\times 8_{rebuilt}} = \sum_{i=1}^{n} c_i w_i = C_{30\times n} \times W_{n\times 8}$$

Equation 2

The number of synergies was chosen in a way to maximize the efficacy of the reproduced data using the lowest number of synergies possible. *Variability Accounted For* (VAF) was utilized as the metric (Clark et al., 2010; Neptune et al., 2009; Ting and Macpherson, 2005) to do so. VAF was defined (Eq. 2) according to previous research (Clark et al., 2010; Neptune et al., 2009). The number of synergies was chosen using two criteria. The number of muscle synergies was the minimum of 1) The minimum number that could account for at least 75% of the variability of the data (Torres-Oviedo and Ting, 2010), and 2) at the minimum number at which adding an extra synergy did not contribute more than 5% in rebuilding the processed EMG data (Figure 6 and Figure 7) (Clark et al., 2010).

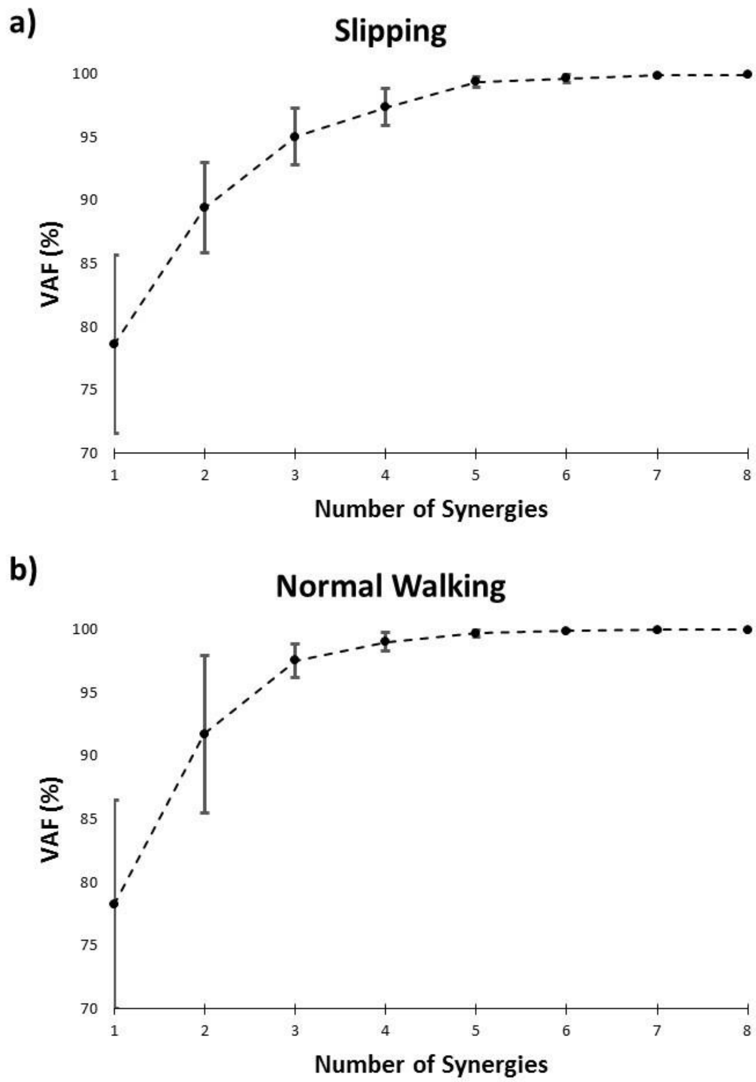


Figure 7 VAF (averaged) versus number of synergies curve for slipping (a), and normal walking condition (b), based on each individual's data. Error bars indicate one standard deviation. (reprint from Nazifi et al., 2017b)

$$VAF = 1 - \frac{\left\| M_{30\times 8,processed} - M_{30\times 8,rebuilt} \right\|_F^{\,2}}{\left\| M_{30\times 8,processed} \right\|_F^{\,2}}$$

Equation 3

where the subscript F indicates the Frobenious norm.

3.2.2.2. Reference participant selection and synergy ordering

The synergies extracted using the abovementioned method would not have any pre-specified sequential order; meaning that a sorting is crucial to have all the similar synergies (i.e., with highest correlation (dot product) to each other, explained in details in the next paragraph) in the same order among participants (Figure 10). Hence, once synergies were calculated for each individual, synergy referencing, and ordering were performed to group similar synergies across participants. Referencing and ordering were performed as follows: First, a reference participant was chosen for each condition (normal walking and slipping) whose synergies best described the synergies of all participants (i.e., the participant whose synergies showed the maximum value of similarity to all other participants' synergies). Similarity was quantified by uncentered correlation coefficients (d'Avella et al., 2003; Torres-Oviedo and Ting, 2010), ($r_{ij} = \cos\theta = \frac{w_i . w_j}{|w_i| . |w_j|}$ *for two given synergies)*, for every possible pairs of synergies of a reference participant (i.e., participant *i*) and all the others participants (i.e. participant *j*, *j*≠*i*) (Roh et al., 2013; Torres-Oviedo and Ting, 2010). A pair of synergies were considered significantly similar if *r*>0.7, and marginally similar if *r*>0.45 (Torres-Oviedo and Ting, 2010). The reference participant was selected to be the one with the greatest number of significantly similar synergies (pairs with r>0.7) with all the other participants.

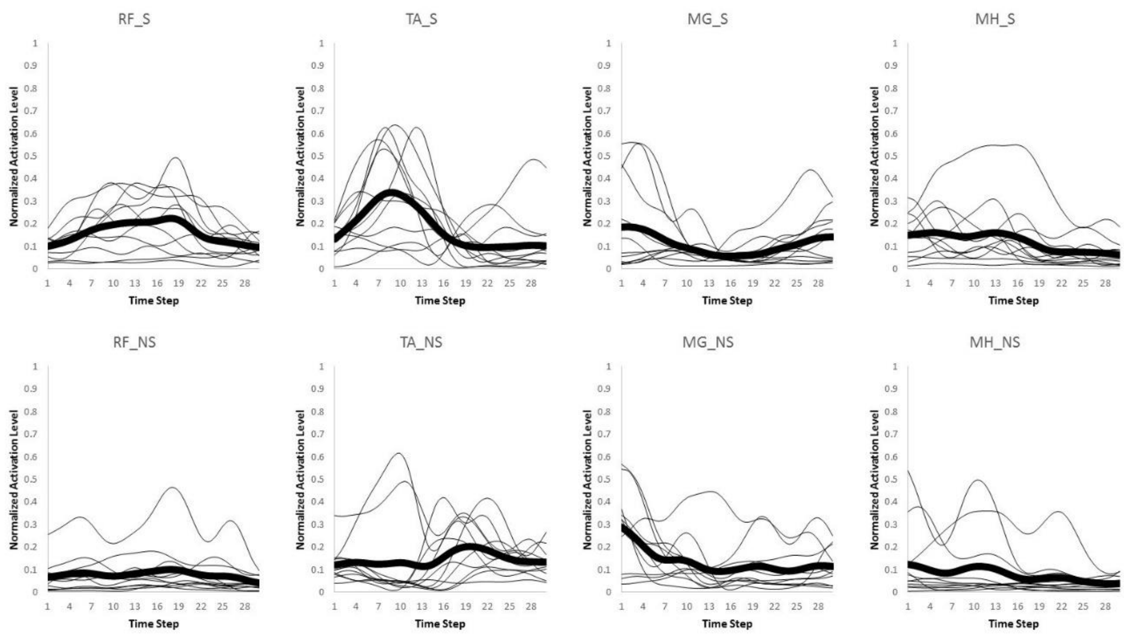


Figure 8 Original muscle activation patterns during normal walking trials for the first 300 ms (integrated every 10 ms). The thick line represents the average value for every individual (thin lines). (reprint from Nazifi et al., 2017b)

Once the reference participant was determined, the ordered list of synergies, (w_1, w_2, …, w_n), in the reference participant were changed such that the most common synergy comes the first. We chose the most common synergy as the one that significantly correlated with the maximum number of participants (e.g., maximum number of r>0.7). Once synergies were reordered in the reference participant, the orders of synergies of the other participants were systematically modified as follows in order to match with the most similar synergies of the reference participant (d'Avella et al., 2003) (Figure 11). The correlation r was computed for every possible pair of synergies between the reference participant and any other participant. Then, we picked the pair with highest similarity value, and synergies involved in that pair were removed from the set. Then, the highest similarity value among the remaining set was selected and again the pair was removed. This procedure was repeated until all the synergies were matched with their best matching pair. This step was performed so that similar synergies were in the same order across all participants. In this way, the ordered list of synergies, (w_1, w_2, …, w_n), would always present a unique set of synergies irrespective of the participant. Finally, the ordered lists of synergies were averaged across the participants for presentation purpose (Figure 12, Figure 14, Figure 13, and Figure 15).

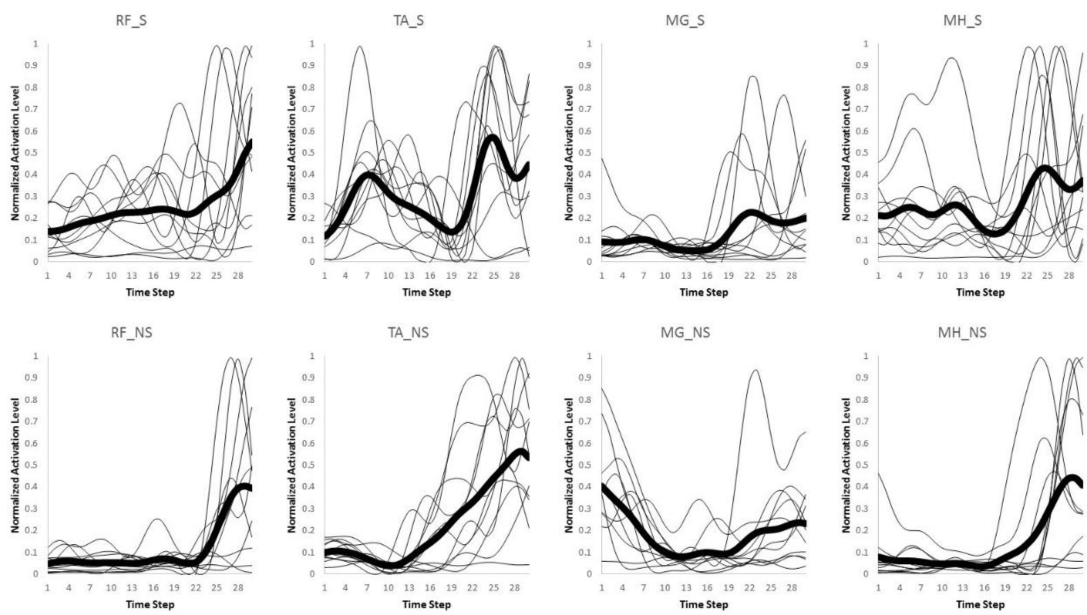


Figure 9 Original muscle activation patterns during slipping trials for the first 300 ms (integrated every 10 ms). The thick line represents the average value for every individual (thin lines). (reprint from Nazifi et al., 2017b)

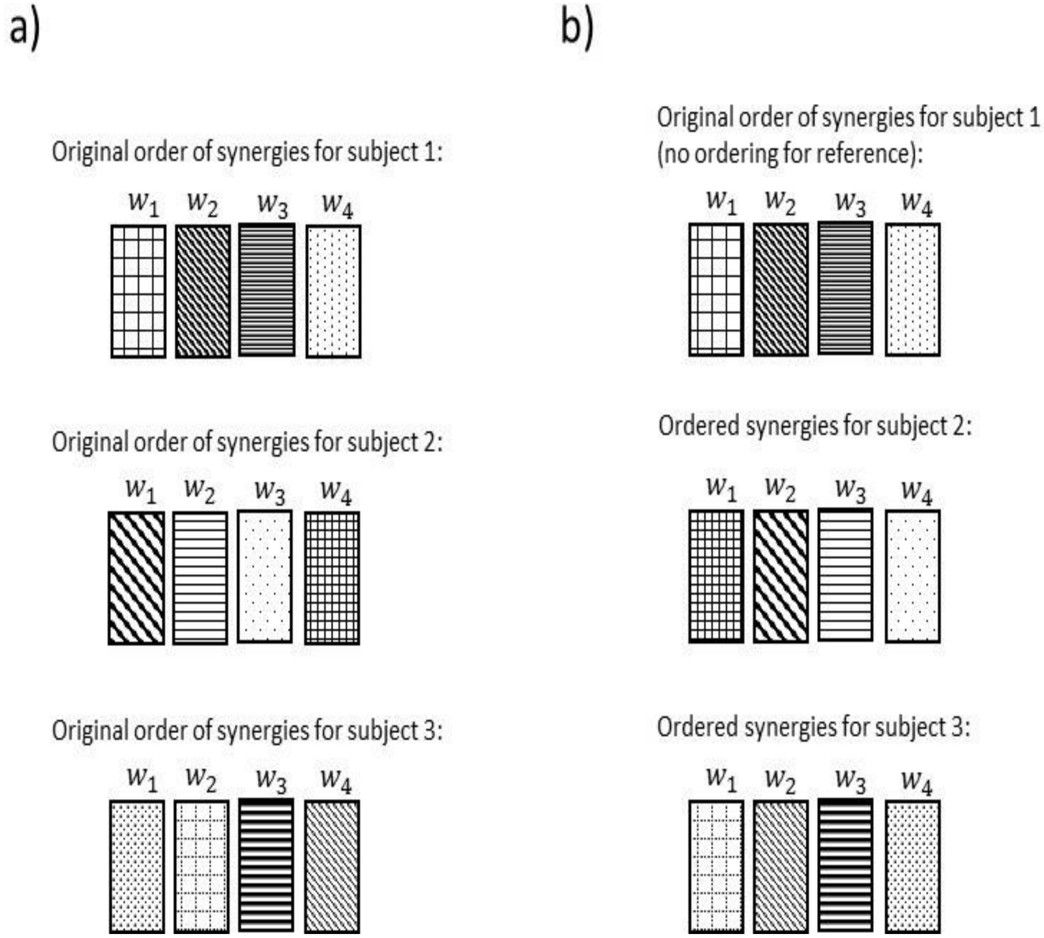
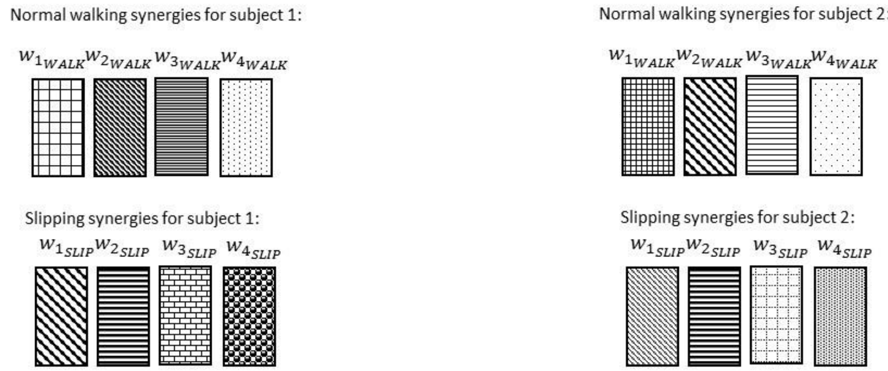


Figure 10 Order of the normal walking synergies in different participants before ordering (a) and after choosing participant 1 as the reference and ordering the synergies accordingly (b). Discrepancies of synergies are symbolized via hatch patterns. Note that after ordering, w_1 for each participant would always refer to a synergy with the same hatch pattern (hatch pattern symbolizes characteristics). (reprint from Nazifi et al., 2017b)



$$r_{sub1} = \begin{array}{c|cccc} & w1_{WALK}\downarrow & w2_{WALK}\downarrow & w3_{WALK}\downarrow & w4_{WALK}\downarrow \\ \hline w1_{SLIP}\rightarrow & r_{11} & r_{12} & r_{13} & r_{14} \\ w2_{SLIP}\rightarrow & r_{21} & r_{22} & r_{23} & r_{24} \\ w3_{SLIP}\rightarrow & r_{31} & r_{32} & r_{33} & r_{34} \\ w4_{SLIP}\rightarrow & r_{41} & r_{42} & r_{43} & r_{44} \end{array} \qquad r_{sub2} = \begin{array}{c|cccc} & w1_{WALK}\downarrow & w2_{WALK}\downarrow & w3_{WALK}\downarrow & w4_{WALK}\downarrow \\ \hline w1_{SLIP}\rightarrow & r_{11} & r_{12} & r_{13} & r_{14} \\ w2_{SLIP}\rightarrow & r_{21} & r_{22} & r_{23} & r_{24} \\ w3_{SLIP}\rightarrow & r_{31} & r_{32} & r_{33} & r_{34} \\ w4_{SLIP}\rightarrow & r_{41} & r_{42} & r_{43} & r_{44} \end{array}$$

Figure 11 Correlation coefficients are calculated after ordering the synergies according to a reference participant. Note the same pattern and order in normal walking synergies (and slipping synergies) in different participants. The intra-participant correlation of normal walk and slip synergies was determined via correlation coefficient matrix, *r*. Same elements of *r* matrix in different participants always show the correlation of a specific pair of synergies. (reprint from Nazifi et al., 2017b)

3.2.2.3. Investigation for shared and task-specific synergies between different gait conditions and their roles

To investigate if some synergies are shared between normal walking and slipping, normal walking synergies and slipping synergies were compared for every individual. As before, uncentered correlation coefficients (*r*) were used to determine the similarity between the synergies of two tasks. For each participant, there were n^2 possible pairs of synergies between normal waking and slipping conditions (Figure 11). For example, r_{32} represents the correlation between the third slip synergy and second normal walking synergy for all participants. Once the correlation coefficient was calculated for each individual, one sample *t*-test (SPSS v21, IBM, Chicago, IL) was performed on the same r value of all participants to investigate if any of these pairs were significantly larger than the critical value across all participants $(r_{ij} > r_{critical}, p_{value} < \alpha)$. The significance level was fixed to be α=0.05. The critical *r* values were set to be 0.7 and 0.45, respectively (Torres-Oviedo and Ting, 2010). The pairs of synergies that were correlated (either significantly or marginally) across all the participants were considered shared synergies between two tasks, while the pairs that were not correlated were considered *task-specific* synergies (d'Avella et al., 2003). Gender effects were not included as a variable in the statistical analysis, since insufficient number of members in each group (6 members in male group versus 5 in female group) would discredit the analysis.

This comparison method was repeated for time-series activation coefficients. However, activation coefficients were compared only for the shared synergies. The reason for this constraint was that for muscle synergies, if considered as building blocks of the nervous system, one can expect independent activation for each block (synergy) in general case. Thus, comparison of the similarity of activation patterns between different blocks (synergies) would not be meaningful, unless for the same blocks (the shared muscle synergies). We suspected that shared synergies have the same activation patterns since shared synergies are technically the same building blocks and might be activated with a similar pattern. However, we could not expect the similar activation patterns for task-specific synergies as they represent the activation of two totally different blocks.

Additionally, since studies suggest a 200ms latency for postural response to a slip (Cham and Redfern, 2001; Chambers and Cham, 2007), we also compared time-series of activation coefficients between two tasks for the first 200ms.

At last, a simulation was run on each of the extracted synergies using OpenSim (SimTK, Stanford, CA) in order to observe their mechanical effect. The resulting muscle activations of every individual synergy was fed to a generic musculoskeletal system in OpenSim and the resulting movements were observed to conclude the role of each muscle synergy. Subsequently, based on the contribution weights of each muscle in the synergies, one could postulate the sub-task each synergy performs.

3.3. Results

The setup effectively induced slip incidents on all participants. The mean and the standard deviation for PHS during slipping and Slipping Distance were measured to be (0.90±0.50 m/s) and (163.62±101.89 mm), respectively. The EMG data was processed and prepared for synergy extraction. The original muscle activation patterns for walking and slipping are presented in Figure 8 and Figure 9, respectively. Using the aforementioned iterative nonnegative matrix decomposition technique and varying the number of extracted synergies, corresponding VAFs were calculated for the pooled data from all participants (Figure 6). Different research groups have used different values and techniques to find the thresholds for VAF (Chvatal et al., 2011; Roh et al., 2013; Torres-Oviedo and Ting, 2010). In this chapter, four synergies considered to be enough to account for variability of the normal walking and slipping data (Figure 6) as they successfully reconstructed more than 75% of the original pooled data (VAF≥0.75) and also addition of an extra synergy did not contribute in reconstruction of more than 5% of the original data (Figure 6) (Chvatal and Ting, 2013; Eskandari et al., 2016). The local VAF curves (for each individual's data) also substantiated our choice of four muscle synergies , accounting for more than 95%, for both walking and slipping condition (Figure 7) (Ting and Macpherson, 2005). The number of synergies used in this chapter

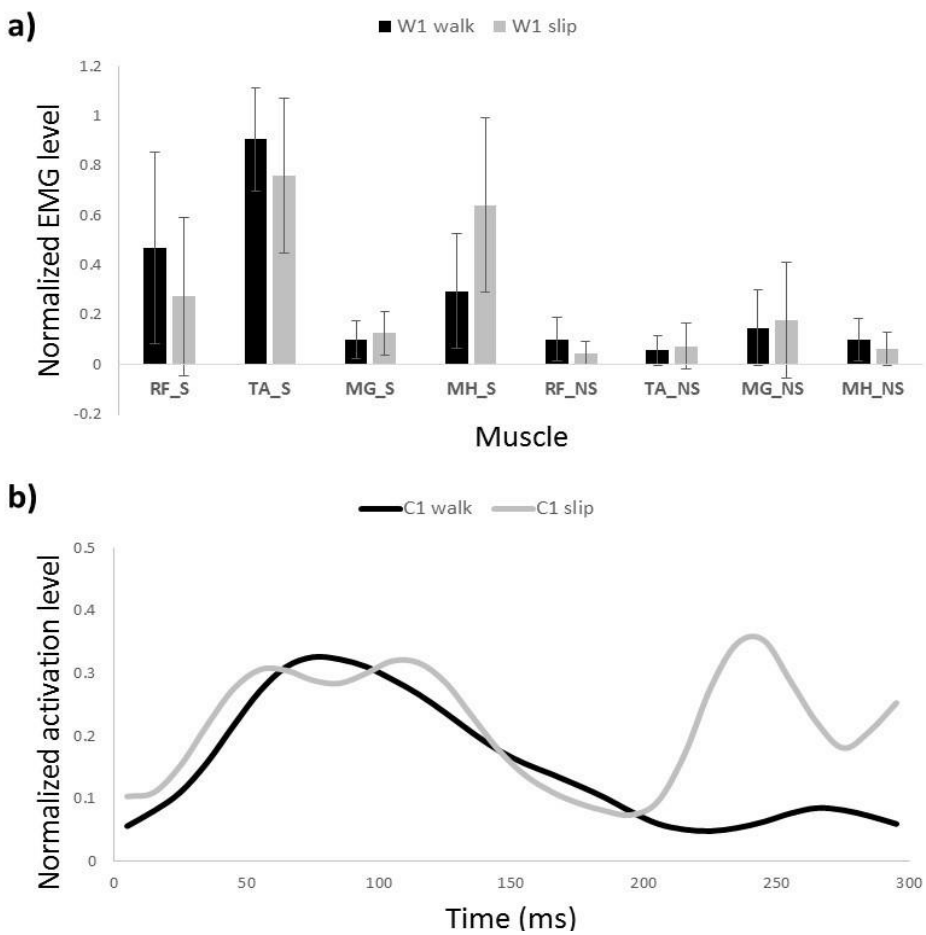


Figure 12 Muscles synergies (a) and their corresponding time-series activation coefficients (b) for the first shared muscle synergy between normal walking and slipping. Error bars indicate one standard deviation. Note that muscles belonging to slipping foot are shown by S while muscles of non-slipping foot are shown by NS. (reprint from Nazifi et al., 2017b)

(four) also matches with the number of synergies used in similar studies with the same dimensionality (number of involved muscles) and motor-task (walking and its sub-functions) (Clark et al., 2010; Neptune et al., 2009; Roh et al., 2013). Subsequently, four normal walking synergies and four slip synergies were extracted (Figure 12, Figure 14,

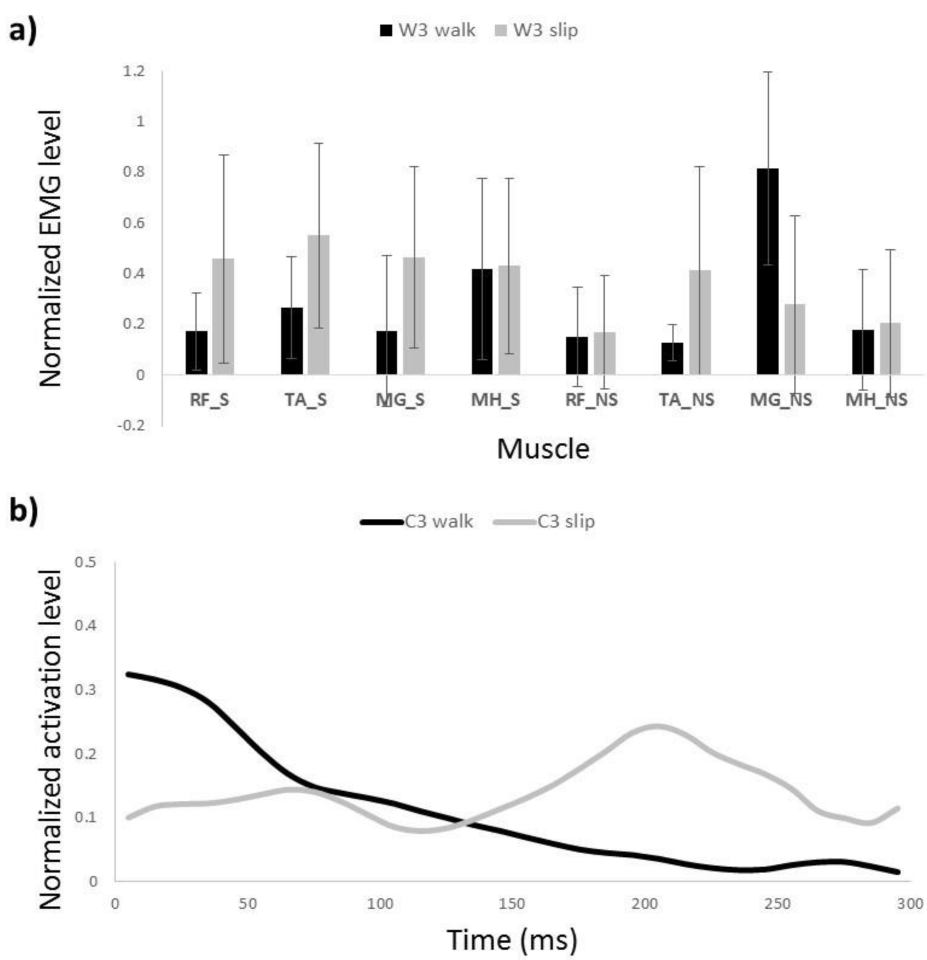


Figure 13 Muscles synergies (a) and their corresponding time-series activation coefficients (b) for the third muscle synergy of normal walking and slipping. Error bars indicate one standard deviation. Note that muscles belonging to slipping foot are shown by S while muscles of non-slipping foot are shown by NS. (reprint from Nazifi et al., 2017b)

Figure 13, and Figure 15). The corresponding time-series activation coefficients of those synergies were extracted as well.

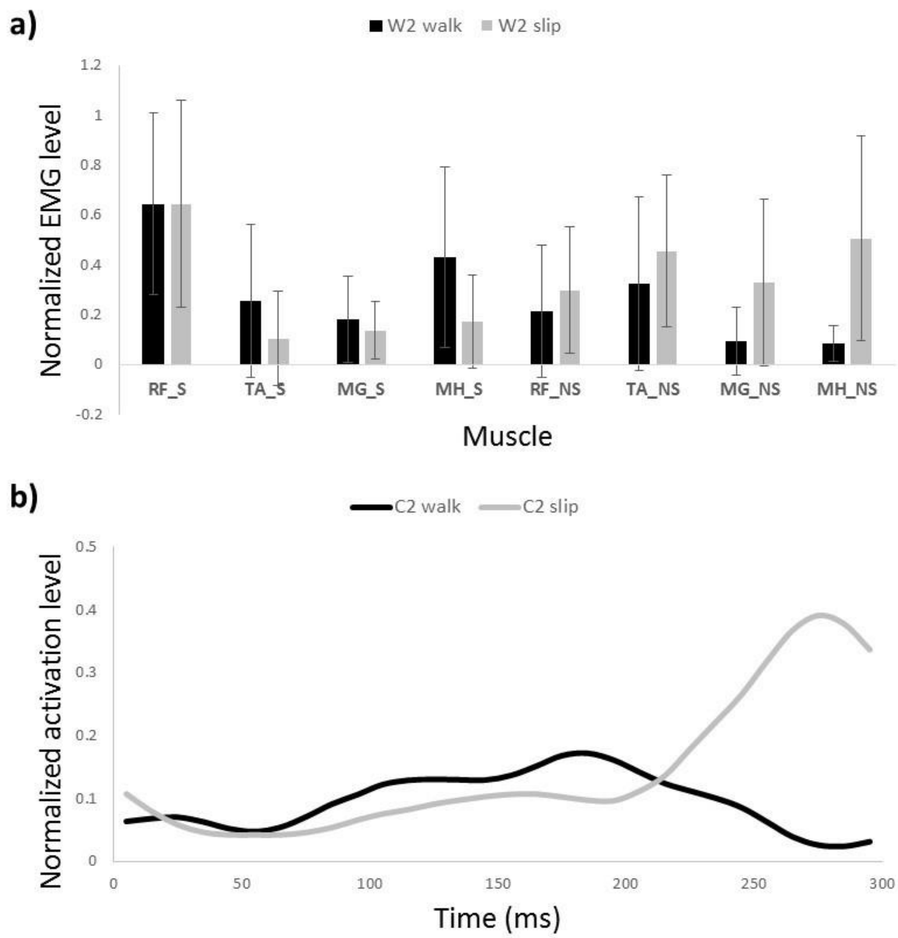


Figure 14 Muscles synergies (a) and their corresponding time-series activation coefficients (b) for the second shared muscle synergy between normal walking and slipping. Error bars indicate one standard deviation. Note that muscles belonging to slipping foot are shown by S while muscles of non-slipping foot are shown by NS. (reprint from Nazifi et al., 2017b)

One sample *t*-test results revealed that there are two pairs of synergies shared between normal walking and slipping. The first walking synergy was found strongly correlated with the first slipping synergy among participants (r_{11}=0.82±0.13> $r_{\text{strong correlation}}$ =0.7, $t(10)$=3.10 , p-value<0.01,Figure 12), while the second walking synergy was marginally correlated to the second slipping synergy (r_{22}=0.62±0.23>$r_{\text{marginal correlation}}$=0.45, $t(10)$=2.37, p-value =0.02, Figure 14). Hence, there are two pairs of

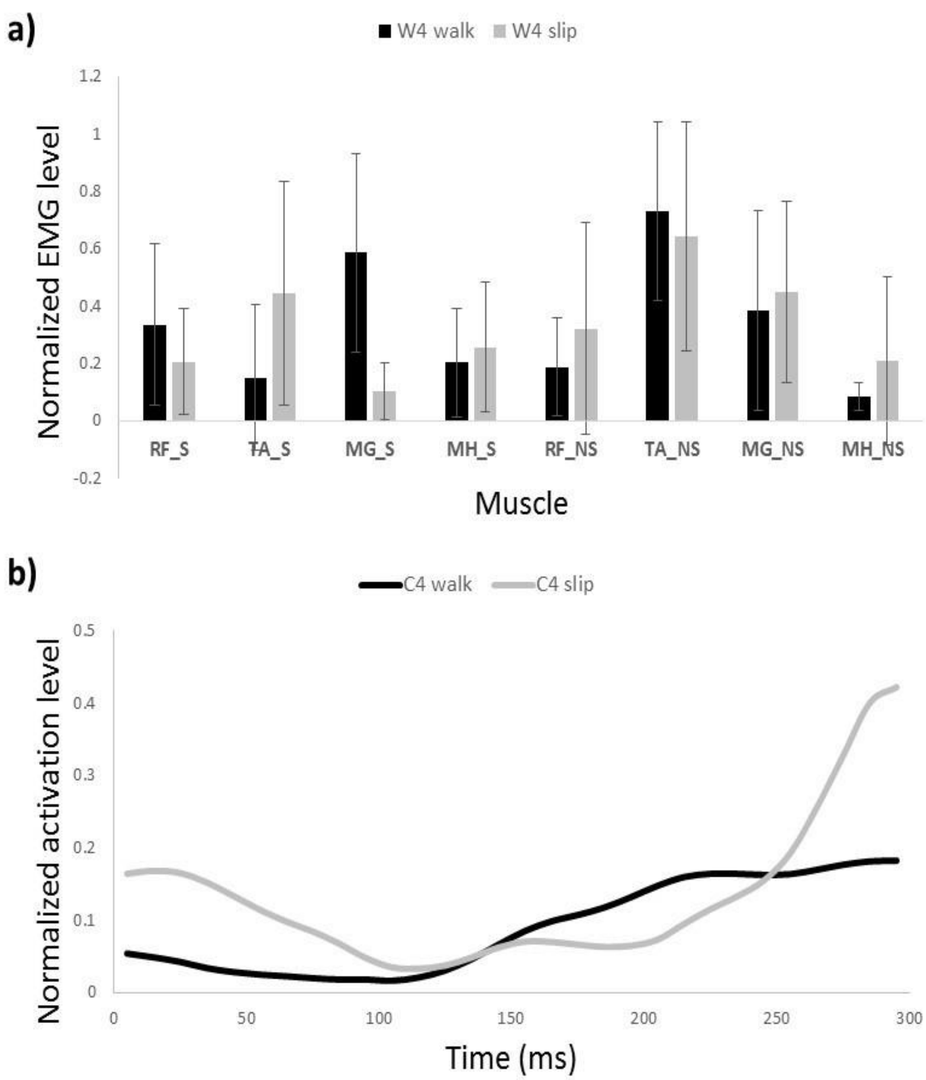


Figure 15 Muscles synergies (a) and their corresponding time-series activation coefficients (b) for the fourth muscle synergy of normal walking and slipping. Error bars indicate one standard deviation. Note that muscles belonging to slipping foot are shown by S while muscles of non-slipping foot are shown by NS. (reprint from Nazifi et al., 2017b)

shared synergies between normal walking and slipping. The other synergies were not correlated to each other, and these synergies were considered task-specific.

The complete 300ms of activation coefficients of the shared synergies were tested to identify potential correlations. However, there was no strong correlation observed between the activation coefficients of the shared synergies. Only the activation coefficients of the first shared synergy showed a marginal correlation (r=0.71±0.18, $t(10)=4.62$, p-value<0.001).

Interestingly, comparing the time-series of activation coefficients of the shared synergies for the first 200ms revealed two significant similarities. The activation pattern of the first shared synergies were significantly correlated between normal walking and slipping conditions for the first 200ms after heel contact (r=0.84±0.17, $t(10)=2.72$, , p-value=0.01, Figure 12, first 200ms). The time courses of activation coefficients for the second shared synergy were also significantly correlated between normal walking and slipping conditions for the first 200ms after heel contact (r=0.59±0.21, $t(10)=2.25$, , p-value= 0.02, Figure 14, first 200ms).

3.4. Discussion

Muscle activities for both lower limbs during gait and slipping were successfully presented. This study found four muscle synergies for each condition, of which two were shared between normal walking and slipping tasks, suggesting similarities between the required sub-tasks during normal walking and slipping tasks. As stated before, different research groups have used a wide range of VAF values and standards in their muscle synergy studies to decide number of synergies. This fact shows that there is no commonly accepted VAF threshold and one might simply find other criterions conservative or flexible. In this study, we tried to accommodate local, global, and "less than 5% growth" VAF conditions which are the most prevalent criteria introduced by different groups. Yet, other researchers may still prefer other values due to the existing uncertainties about this issue. Furthermore, limited number of muscle synergies (i.e. four synergies in comparison to eight muscles) used by the CNS during slipping shows the

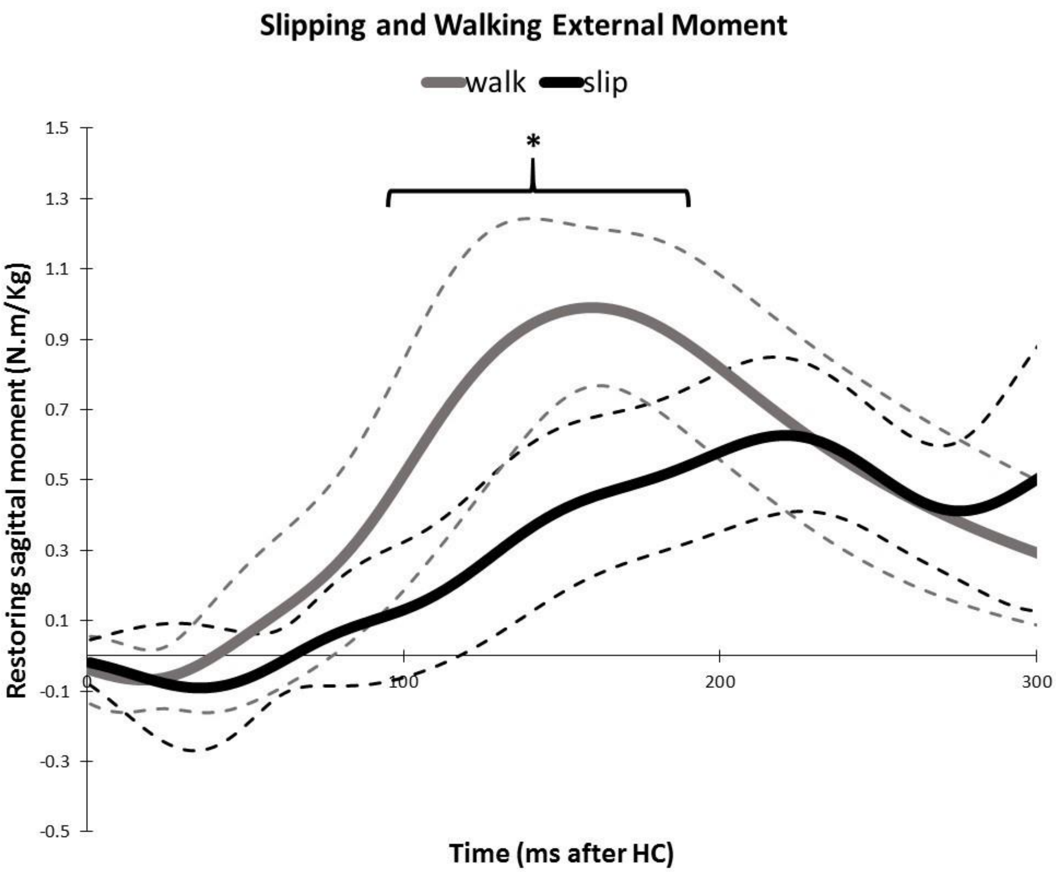


Figure 16 Average amount of the external mechanical effect (restoring moment) induced on the body after the heel strike for slipping and walking. Dashed lines indicate one standard deviation. The astriks indicate statistically significant differences (p<0.05). (reprint from Nazifi et al., 2017b)

efficacy of muscle synergies in accounting for variation of a motor-task using a low dimensional modular organization, since the degrees of freedom are reduced to as low as four out of eight potentially available muscles in response to a slip. Finding of 6-7 synergies to control both walking and slipping stay consistent with the concept of synergies as a declarative and descriptive mean. Finally, only one of the time-series activation coefficients for the two shared synergies were correlated for the first 300ms after heel contact. Interestingly however, both of the time-series activation coefficients for the two shared synergies were correlated during the first 200ms after heel contact and deviated afterward according to the latencies and sub-functions reported for postural response to a slip (Cham and Redfern, 2001).

The synergies could have a specific functionality (Ting and Macpherson, 2005) and possibly could be interpreted as physical sub-tasks of the original motor behavior (Clark et al., 2010; d'Avella et al., 2003; Ting and Macpherson, 2005). Considering this fact along with the extracted muscle synergies of slipping and walking, one could postulate the sub-task each synergy performs based on the contribution weights of each muscle in the synergies.

The possible role of the first shared synergy was to decelerate the leading limb. This mechanical goal stays consistent with the known sub-tasks of the gait cycle at terminal swing phase. Pretibial and hamstring muscles group are known to be activated at the end of swing phase and in the early stance phase (Basmajian and De Luca, 1985; Medved, 2000; Rose and Gamble, 2005), in order to decelerate the leading limb and position the foot and arrange the contact. These sub-tasks are needed in both terminal swing phase and also in response to slips. The primary response to slip is to bring the slipping leg back near the body and shifting the COM forward (Cham and Redfern, 2001) which is possibly achieved by activation of this synergy. This common mechanical goal explains this synergy being shared between slipping and walking. The sub-tasks are produced by superpositioning the role of the main activated muscles in this synergy, namely TA_S and MH_S (Figure 12). Also, a relatively high activation of RF_S was observed in this synergy. The co-activation of MH_S and RF_S would result in a stiffer knee joint on the slipping limb in order to avoid knee buckling while enduring the body weight. The simulation also verified the aforementioned role for this synergy in generating a hip extension as well as dorsiflexion on the leading leg. Considering the role of the hamstring in decelerating the lower limb (Lockhart and Kim, 2006; Rose and Gamble, 2005), activation of hamstring in early-stance phase results in deceleration of the slipping leg (Qu et al., 2012; Yang and Pai, 2010). Moreover, activation of TA_S causes a dorsiflexion to elevate anterior part of the leading foot and arrange the heel to strike. It also helps to store energy and prevents from foot drop or foot slapping phenomenon (Rose and Gamble, 2005).

The second shared synergy seemed to prepare the weight transfer to the leading limb. This sub-function happens at early stance phase of the gait and are mainly achieved by activation of the quadriceps muscles. As stated by Medved (2000), shortly after the heel strike, the quadriceps muscles group (RF in this dissertation) contract in order to absorb the shock and provide more support to stabilize the knee and pelvis joint on the leading leg. This stabilization prepares the leading leg for weight transfer.

Abovementioned sub-functions are also required in response to a slip. The secondary response to slip is to extend the knee and flex the hip of the slipping leg to avoid knee buckling and continue gait (Figure 14) (Cham and Redfern, 2001). Once again, the common mechanical goal substantiates this synergy being shared between the two conditions. Expectedly, the main activated muscle in this synergy was RF_S that contributes in knee extension, hip flexion, and weight acceptance. However, there was a slight difference between slipping and walking synergies (Figure 14). A larger knee flexion angle was observed in the slipping synergy (most probably due to activation of MH_NS and MG_NS on the trailing foot) matching with previous studies. As the swing phase of the trailing limb is disturbed by the slip, these activations prevent the fall as the leading limb is not ready to accept body weight (Moyer et al., 2009). The simulation also resulted in knee extension and hip flexion on the leading foot, verifying the abovementioned arguments. Appearance of these muscle activations from both legs in a single synergy substantiates that the interlimb coordination in slip recovery might be rooted in synergies.

Other two synergies and their functionalities are considered task-specific. Although having similar muscle synergies and activation between normal walking and slipping may seem to substantiate synergies as a neural control mechanism, having dissimilar synergies is more likely to support muscle synergies as a descriptive tool. That is because if synergies were a control mechanism, we would see identical synergies and activations during walking and the first 200 ms of slipping due to the reaction time. However, as W3 and W4 of walking are not used during early slipping (which is the same as walking), we claim that synergies are rather a descriptive tool instead of a neural control mechanism. Our findings show that the third normal walking synergy can be considered as the propulsion provider on the non-slipping leg according to the activation of MG_NS (Figure 13). The simulation result also showed a high plantarflexion and knee flexion substantiating this expectation. This phasic sub-function happens at late stance of the trailing limb to provide the propulsive force and accelerate the body.

However, as the terminal stance phase is interrupted during a slip, it seems cogent to lack this synergy during slipping. while the fourth synergy is responsible for dorsiflexion of the non-slipping foot to clear foot and avoid the toe from hitting the ground during swing and accomplishing foot flat when swing is terminated (Moyer et al., 2009)(Figure 15).

On the other hand, the third slipping synergy seems to stabilize and stiffen joints on both legs via activating almost all of antagonist muscles equally (Figure 13). Moreover, the fourth slipping synergy contributes to dorsiflexion of the non-perturbed limb and might show the measure to avoid tripping during the slip (Marigold et al., 2003)(Figure 15).

Similarity of activation coefficients of the shared synergies for both conditions (Figure 12 and Figure 14, before 200ms) agrees with the hypothesis of having the same activation level for the shared synergies before the corrective motor response to slip. This result seems cogent since before the reaction of the body to slip, normal walking and slipping should be dealt with identically and should have the same muscle activation patterns. As a result, before the reaction to a slip, same control blocks (shared muscle synergies) of these “temporarily same” tasks should be activated with the same activation pattern (same activation coefficients). A question may also arise here: The external mechanical effects of the slip (moments imposed on the body) might not deviate from normal walking moments instantly. Hence, can these observed similarities be interpreted as result of the similar moments during “early slipping” and “walking”? To address this question, we calculated the mechanical effect of each condition (i.e., restoring torque in sagittal plane after the heel contact) on the body right using the shear forces generated by the leading limb. We found that the deviations between walking and slipping moments start well before 300 ms post-heel-strike. Using an independent *t*-test, we also found that the restoring moments are significantly different between slipping and walking conditions from 70 to 202 ms (p-value<0.05). Thus, the studied interval (300 ms) encompasses different external mechanical effects of slipping and walking on the body. Consequently, we claim that the observed similarities show that the control blocks used for these motor-tasks (synergies) are common, rather than that the motor-tasks are similar. Characteristics of the extracted activations for slip synergies match with previous studies. The body started to react to the slip after 200ms (Cham and Redfern, 2001) via activating the appropriate control blocks, indicated by peaks in activation levels of slip synergies (Figure 12 and Figure 14). Timing of the peaks is in accordance with the known primary and secondary motor response to slip (Cham and Redfern, 2001). The first peaks seen in the slip synergies belong to the third and the first synergies (leg decelerator synergy) (Figure 12 and Figure 138),dorsiflexing ankle, flexing knee and extending hip of the slipping foot, bringing the slipping leg back near the body, matching with the introduced primary response to slip by Cham and Redfern (2001). The next peak belonged to the second shared synergy, that extends knee and flexes hip of slipping leg (Figure 14), or the secondary response to a slip according to Cham and Redfern (2001).

These findings stay consistent with the existing literature. For example, in a study by Chvatal and Ting (2013) it was found that a common set of muscle synergies is utilized to achieve task-level goals during perturbed and unperturbed walking and standing. Although most of the studies that examined eight muscles for walking reported four muscle synergies (Clark et al., 2010; Neptune et al., 2009), Chvatal and Ting found five to eight synergies for unperturbed walking (average six), three of which were shared with perturbed walking synergies. Their results, however, do not dispute our findings since that study used 16 lower extremity muscles, all from one leg (unilateral). Since muscle synergy analysis is sensitive to the original dimension of the data set (i.e. number of the studied muscles), a direct comparison of the number of synergies would not be feasible among these studies. Yet, their results substantiate the notion that similar biomechanical demands between perturbed walking and normal walking is likely to result in the CNS recruiting similar muscle synergies for both tasks. Furthermore, this article only examines the first 300 ms after the slipping, which captures only the reactive response of the CNS to slipping. However, Chvatal and Ting (2013) studied a larger time period, enabling them to investigate both reactive and voluntary response to the perturbations. Finally, Chvatal and Ting (2013) perturbed participants in different directions while walking. However, slipping typically happens in the forward direction, which complicates the comparison of these studies. Needless to say, both studies reported that perturbed walking would evoke similar motor patterns to those of unperturbed walking.

A muscle synergy approach is significant as it potentially could establish a basis for a more direct motor rehabilitation (Roh et al., 2013). Having the slipping muscle synergies of healthy individuals as a reference, identification of the impaired synergies in patients would be facilitated. Consequently, one could design appropriate therapies and trainings, conducted toward the damaged synergy (sub-task) to reestablish it in order to perform the required phasic mechanical goals and sub-functions.

Limitation of this chapter was that there was no classification performed on participants based on their slipping severity. It is probable for participants to choose different strategies while countering slips with different severity. Thus, classifying the participants based on their slip severity would be legitimate. However, the number of individuals in each group could prove insufficient for a cogent statistical analysis, preventing further groupings. In the future studies, we would study participants' slipping synergies for larger number of classified (based on the severity of slips) groups. By doing so, a conclusion could be made whether the discrepancies in the muscle synergies are or are not significant among groups with different slipping severity and how slipping synergies would help diagnosing the possible cause of severe slips. Another interesting aspect would be studying the modifications of the synergies with repeated perturbed trials to look for possible evolutions in synergies and slipping strategies (Ison and Artemiadis, 2015).

3.5. Conclusion

This chapter extracted muscle synergies for normal walking and slipping among young healthy participants. We found two shared and two task-specific muscle synergies among eight lower limb muscles for these two tasks. The activation levels for the shared synergies were identical before the onset of the motor response to slip. Also, the sub-tasks executed by the synergies matched with the known sub-tasks of the gait and slip. The significance of our approach in studying slip, is the identification of the synergies used during this motor-task. This identification would form a foundation for a novel diagnosis and rehabilitation method, based on the impaired synergies of motor-tasks. Future works will include investigation of the inter-participant deviations and discrepancies of slip synergies and its correlation with the severity of slips, which lead to understanding of the factors causing sever slips to happen.

4. ASSOCIATION BETWEEN SLIP SEVERITY AND MUSCLE SYNERGIES OF SLIPPING[3]

4.1. Introduction

Several studies suggested that each muscle synergy may represent a sub-task of the original motor-task (d'Avella et al., 2003; Neptune et al., 2009). Previous chapter have discussed the beneficial aspects of a muscle synergy approach in studying motor-tasks, like walking and slipping. A muscle synergy approach highly facilitates analysis of the coordination of the interlimb muscles since the muscle synergy hypothesis claims that all muscles with the same neurological origin that are activated together appear in the same synergy. However, traditional EMG analysis fails to decompose co-activated muscles into the same control block (synergy) (Chambers and Cham, 2007; Qu et al., 2012). Moreover, another main advantage of the muscle synergy approach is that it would help identify the sub-tasks of the original motor-task. Not only would these sub-tasks facilitate diagnosis of the severe slippers, but also, they might result in designing of a targeted motor rehabilitation based on the impaired sub-tasks (Allen et al., 2013; Roh et al., 2013).

Although previous chapters have extracted and studied slipping response muscle synergies in young adults , no study tried to relate slipping muscle synergies to slip severity. In this sense, this chapter proposes the first step to investigate the cause of

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severe slips or discrepancies between the interlimb coordination of the mild slippers compared to severe slippers while experiencing a slip. The objective of this chapter is to compare the slipping muscle synergies and activation coefficients of “severe slippers” and “mild slippers” to quantify differences in coordination between the two groups. Such differences in muscle synergies, if found, can potentially be related to severity index of an individual. Also, the function of each synergy would be investigated to reveal the sub-function of each synergy during slipping. We hypothesize that the slipping muscle synergies would differ between mild slippers and severe slippers, indicating the malfunctioning synergies responsible for the adverse slip response of the severe slippers. Also, as the previous chapters have revealed similarities between the control of the gait and slipping we hypothesize that the physical sub-functions of some synergies (after being revealed via forward simulation) would be common with known sub-functions of the gait.

4.2. Methods

4.2.1. Participants

A total number of twenty young adults (nine females and eleven males, age (mean ± SD): 23.6 ± 2.52) were recruited for this experiment. Participants were excluded in case of a history of neurological, orthopedic, cardiovascular, pulmonary, and gait abnormalities. The experiment took place upon approval of Institutional Review Board at the University of Pittsburgh. The deidentified dataset was then transferred to Texas A&M University for further analysis with approval from IRB of both Universities. All participants gave written consent before their participation.

4.2.2. Measurements, Experimental protocol, and Data Processing

Participants were asked to walk in a pathway at their self-selected speed. There were two force plates embedded in the pathway. To have each of the force plates receive exactly one foot-strike (the right foot first, and then the left foot second), the starting location of each participant was adjusted (Figure 17). To induce an unexpected slip, participants were assured that the surface would be dry during trials. However, after two to three normal walking trials, the surface of the second force plate was contaminated by applying a solution (75% glycerol, 25% water). To minimize the inter-participant variation of friction force, all participants wore the same brand/model of polyvinyl chloride hard-soled shoes that matched their sizes. The lights were dimmed throughout the experiment to minimize the visual clues about the slippery surface. Also, to catch the participants in case of a total loss of balance after experiencing a slip, a safety harness was provided.

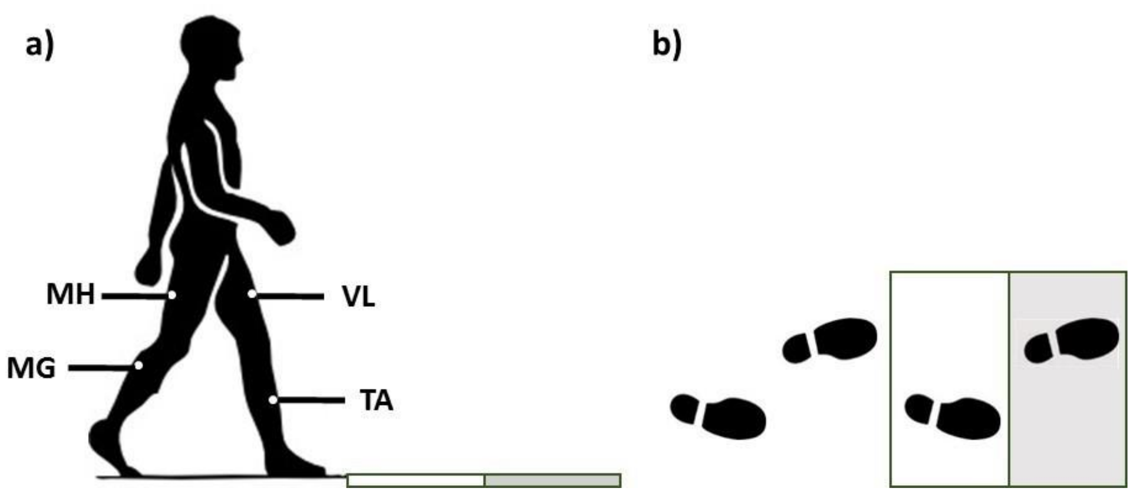

Figure 17 The side (a) and top view (b) of the experimental setup for the walkway and force plates. Gray surface indicates the slippery (contaminated) force plate in slip trials. (reprint from Nazifi et al., 2017a)

EMG data were recorded at 1080 Hz to extract the muscle synergies. Surface EMG electrodes were used to record the activation of four major leg muscles according to (Chambers and Cham, 2007): medial hamstring (MH) (i.e. the primary knee flexor/hip extensor), tibialis anterior (TA) (the main ankle dorsiflexor), vastus lateralis (VL) (hip flexor/knee extensor), and medial gastrocnemius (MG) (knee flexor/ankle plantarflexor). The data were recorded from both right/trailing/non-slipping leg (NS) and left/leading/slipping leg (S). Joint kinematics and PHS was captured using a motion capture system (Vicon 612, Oxford, UK) at 120 Hz. Also, kinetic data and ground reaction forces were collected at 1080 Hz using the force plates.

The EMG data were demeaned, rectified, filtered (4th order low-pass Butterworth filter, cut-off: 15 Hz), normalized (to the maximum activation recorded for each muscle of every individual), and integrated for every 10 ms of the activity (d'Avella et al., 2003). Previous studies have suggested that the aforementioned four muscles have an activation onset time of less than 175 ms in response to an unexpected slip (Cham and Redfern, 2002; Hur and Beschorner, 2012; Marigold et al., 2003; Nazifi et al., 2017b). Hence, the first 300 ms after the slip initiation (i.e., the heel strike moment) was used in slipping muscle (i.e., the synergies observed while the participants were experiencing a slip or 'post-slip-initiation' muscle synergies) synergy extraction. Using an iterative non-negative matrix factorization (MATLAB 2014a, Mathworks, Natick, MA) consistent with previous research and chapters (Clark et al., 2010; Nazifi et al., 2017b; Roh et al., 2013; Ting and Macpherson, 2005). Since previous chapter has shown that four synergies are enough to reconstruct slipping data with a VAF>%95, in this chapter four slipping synergy were extracted and sorted using a reference participant that had the most similar behavior to all other individuals (Nazifi et al., 2017b). Then, using the markers' data (that includes the 3-D position of the heel), the instantaneous heel velocity was calculated. Then, using the PHS criterion, the participants were classified into the mild and the severe slippers. Slips with a PHS of 1.44 m/s or greater were considered "severe," and the rest were counted as "mild" (T. Lockhart et al., 2003). Once the participants were separated into severity sub-groups, the synergies of each group were reordered and sorted according to their similarity to each other (similarity was assessed via correlation coefficient, *r*). To detect significant inter-group differences, an independent *t*-test (α=0.05) was used for each of the muscle synergies and every time point of the activation coefficients using SPSS (v21, IBM, Chicago, IL).

Lastly, to reveal the role of each synergy, OpenSim (SimTK, Stanford, CA) was used to perform a forward simulation of each synergy. The activations resulting from each muscle synergy was separately fed to a generic model to observe the resulting joint torques. Using the provided generic ten degree-of-freedom gait model in OpenSim, the model was first scaled to match to the anthropometric parameters of the reference participant (weight: 52.5 kg, height:1.64 m). Then, the 300 ms time course data of muscle activities resulting from each individual synergy were fed to the corresponding muscles in OpenSim while holding the lower limb joints in a static position (i.e., the same posture at the heel contact of the slipping limb). Finally, the resulting joint moments were studied.

4.3. Results

Four muscle synergies and their corresponding activation coefficients were extracted from the processed data according to our previous chapter. Based on PHS, twelve participants were classified as mild slippers (PHS<1.44 m/s) while the other eight were severe slippers (PHS≥1.44 m/s). There was no difference observed in age, height,

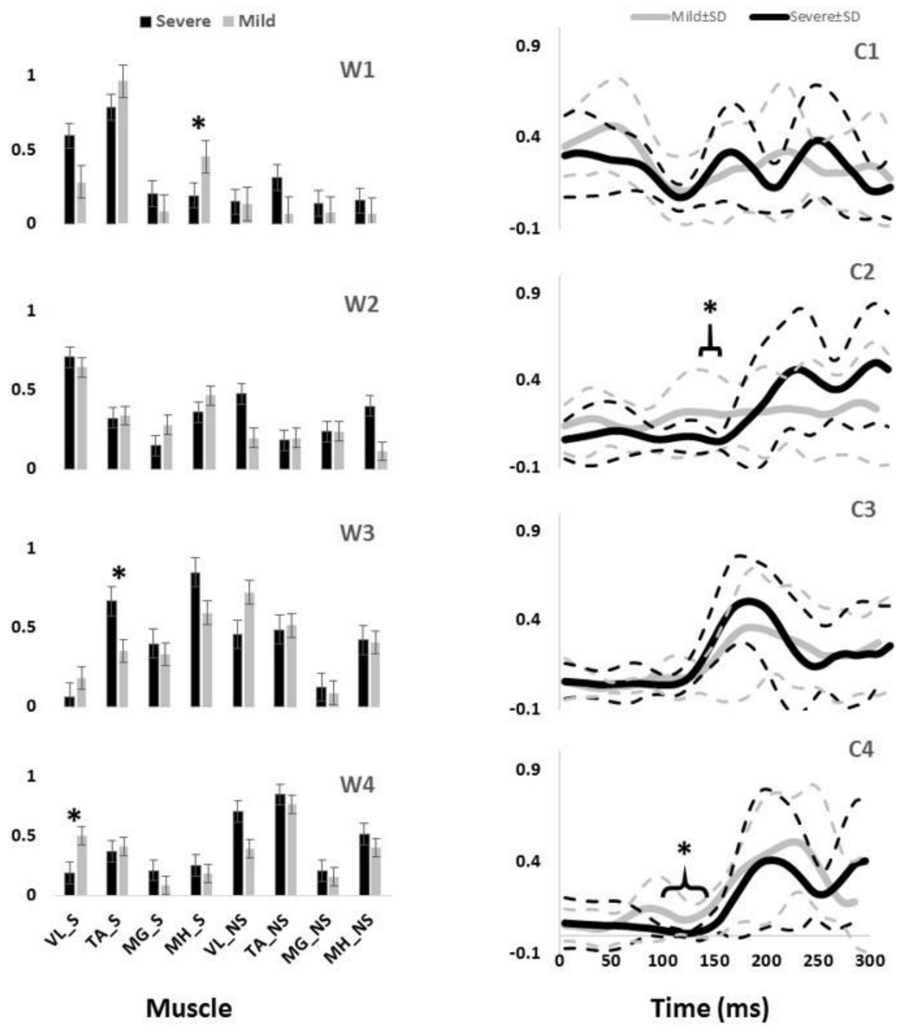

Figure 18 The slipping muscle synergies (W's) and the corresponding activation coefficients (C's) for both severity groups. Solid lines show the average value while dashed line shows one standard deviation. Asterisks indicate significant inter-group differences. (reprint from Nazifi et al., 2017a)

and mass of the severe slippers versus mild slippers (Information on each group is provided in Table 1). The averaged synergies and activation coefficients for each group are provided in Figure 18.

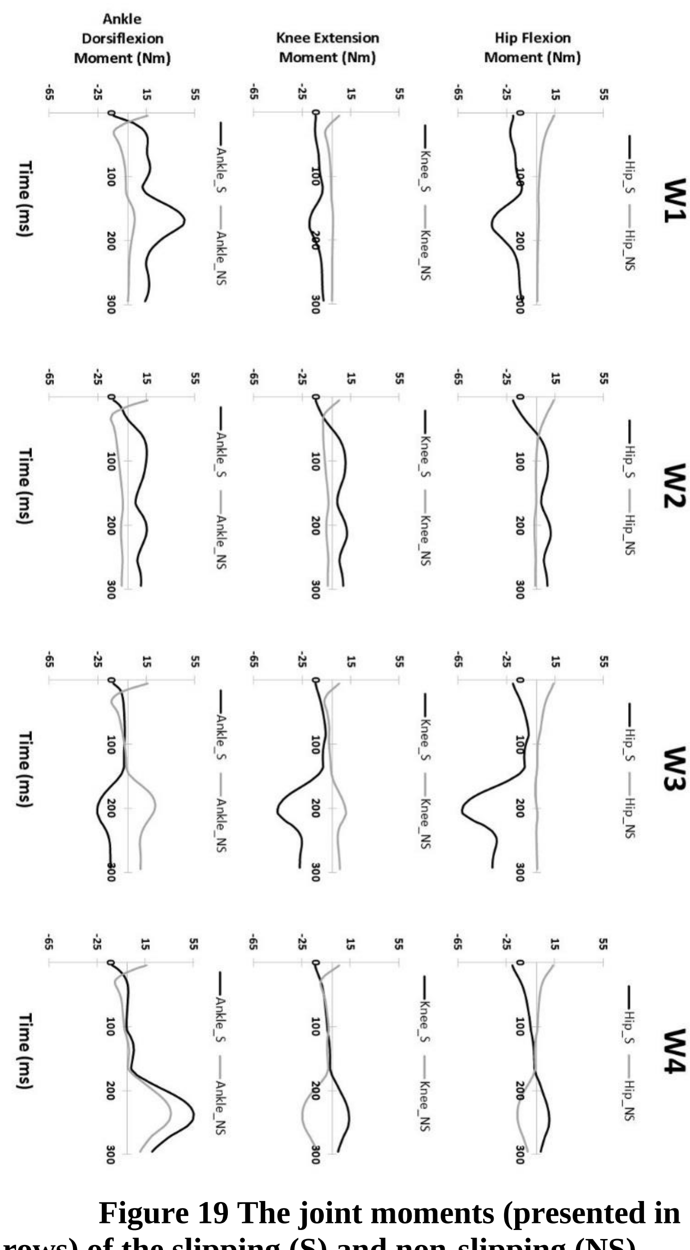

Figure 19 The joint moments (presented in rows) of the slipping (S) and non-slipping (NS) limb calculate via simulation for all slipping synergies (presented in columns) (reprint from Nazifi et al., 2017a)

Independent *t*-test detected several inter-group differences. Higher activation of MH_S in the first synergy (W1) was significantly different between mild and severe slippers (Table 2). Higher MH_S activation was associated with mild slips (Figure 18). Also, activation of the VL_S in the fourth slipping synergy (W4) was found to be different between mild and severe slippers (Table 2). Mild slippers showed a higher contribution of VL_S during their slips (Figure 18). Lastly, activation of the TA_S was different in the third muscle synergy (W3) (Table 2). Higher activation of TA_S was associated with severe slips (Figure 18).

Significant differences were also observed in the activation coefficient of two synergies. Mild slippers had significantly higher activations for the second synergy (C2) (Table 2) from 130 ms to 150 ms after the slip initiation (Figure 18). Additionally, mild slippers had higher activations for the fourth synergy (C4) from 100 ms until 140 ms after the heel strike on the slippery surface (Figure 18). These differences may indicate that the mild slippers activated their corresponding muscle synergies faster (earlier by 30-50 ms) than their severe slipper counter parts in response to a slip. The simulation results revealed the role of each muscle synergy during slipping (Figure 19). The first slipping synergy caused a significant hip extension, knee flexion, and dorsiflexion moment on the slipping limb (Figure 19). The second synergy mainly prompted hip flexion and knee extension moment on the slipping limb (Figure 19). The third muscle synergy resulted in a considerable hip extension, knee flexion, and ankle plantar flexion moment in the slipping limb as well as knee extension moment on the non-slipping limb. However, the fourth muscle synergy induced a substantial ankle dorsiflexion moment on the unperturbed limb. It also caused a distinct hip extension and knee flexion on the unperturbed limb.

Variable		Mild	Severe	p-value
W1	MH_S	0.45 (0.29)	0.19 (0.19)	0.040
W3	TA_S	0.35 (0.31)	0.67 (0.28)	0.032
W4	VL_S	0.50 (0.29)	0.19 (0.20)	0.017
C2	130^{ms}-140^{ms}	0.22 (0.24)	0.06 (0.07)	0.045
	140^{ms}-150^{ms}	0.12 (0.12)	0.01 (0.02)	0.010
C4	100^{ms}-110^{ms}	0.14 (0.16)	0.02 (0.03)	0.026
	110^{ms}-120^{ms}	0.12 (0.12)	0.01 (0.02)	0.010
	120^{ms}-130^{ms}	0.09 (0.09)	0.01 (0.02)	0.012
	130^{ms}-140^{ms}	0.09 (0.09)	0.02 (0.04)	0.043

Table 2 Variables that showed statistically significant differences between groups (reprint from Nazifi et al., 2017a)

4.4. Discussion

The significant inter-group differences for muscle contributions and their activation coefficients show important aspects of post-slip-initiation responses. The similarities between the general trend of muscle contribution ratios in the synergies of different severity groups suggest that mild slippers and severe slipper use the same strategies in response to a slip. However, the differences observed in the activation coefficients may indicate that mild slippers can respond in a faster (Figure 18) or stronger (Figure 18) way, as opposed to having an overall stronger response throughout the slip. The first muscle synergy may be responsible for hip extension, knee flexion, and dorsiflexion of the perturbed limb (Figure 19). This synergy is likely to be responsible for the terminal swing phase of the gait. Moreover, a higher activation of the MH_S was observed in mild slippers in this synergy (p value = 0.04) (Figure 18). Considering the role of the MH muscle in deceleration of the limb, it is suggested that the mild slippers can generate a greater deceleration at the terminal swing phase. The role of the second synergy in generating hip flexion and knee extension (Figure 19) matches with the secondary response to slips (Cham and Redfern, 2001) in which the mild slippers had a higher activation (Figure 18). This indicates that there is an association between severity mitigation and stronger activation of the secondary response to slips. The function of the third muscle synergy is likely to be hip extension, knee flexion, and ankle plantarflexion (Figure 19). This is a known sub-task during slipping, namely, the primary response to a slip (Cham and Redfern, 2001). However, excessive activation of the TA_S muscle was observed among severe slippers (p value = 0.03, Figure 18), resulting in an excessive Foot Floor Angle (FFA) and thus severe slips (Moyer et al., 2006). Lastly, the fourth muscle synergy caused significant effects on the unperturbed limb. It induced a distinct hip extension and dorsiflexion (Figure 19). This sub-function may be interpreted as another strategy to counter the slips called the “toe-touch”. Toe-touch is commonly practiced as an effective way to increase the base of support while slipping (Marigold et al., 2003). Moreover, the mild slippers showed faster activation for their toe-touch synergy (Figure 18). This may suggest that faster recruitment of toe-touch strategy is associated with less severity.

The results found in this research stay consistent with the previous chapter studies. We claimed that a muscle synergy is strongly shared between walking and slipping. The aforementioned muscle synergy has very similar muscle contribution patterns to the first slipping synergy found in this chapter substantiating our claim about this synergy to be the shared muscle synergy between walking and slipping (so called, "deceleration synergy"). Additionally, Cham and Redfern (Cham and Redfern, 2001) claimed that the response to an unexpected slip can be decomposed into two fundamental components, namely, the primary and the secondary response. The primary

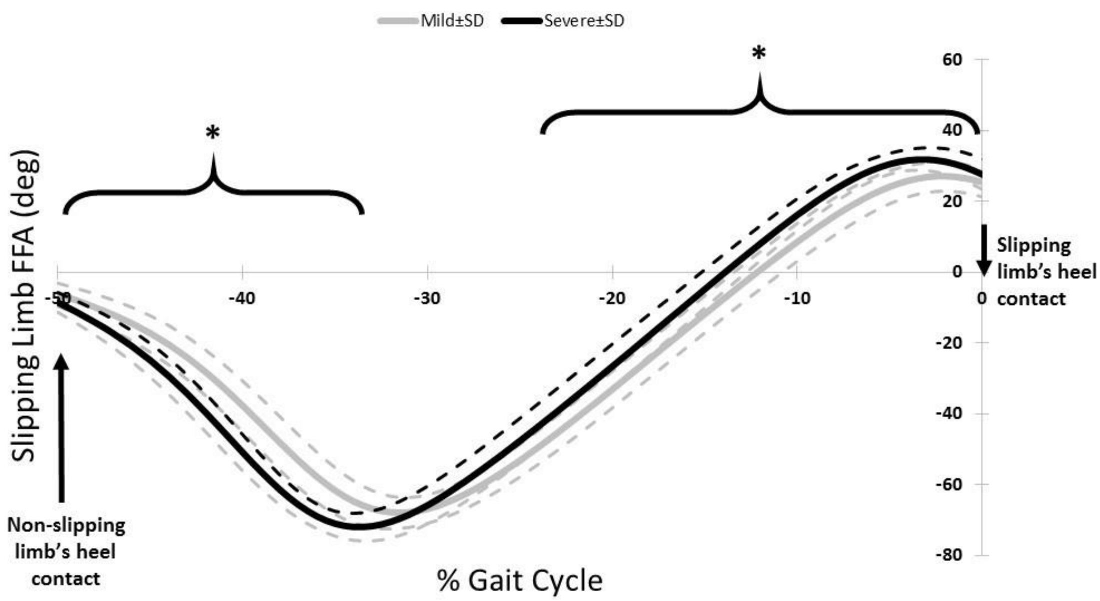

Figure 20 The FFA for the slipping limb for right before the heel strike of the slipping limb. Note that the slipping heel strike happens at 0%. The asterisks indicate significant differences while dashed lines represent one standard deviation (reprint from Nazifi et al., 2017a)

response is responsible for bringing the slipping limb back near the center of mass while the secondary response tends to extend the slipping limb to maintain forward weight progression by shifting the center of mass over the base of support. The observed muscle activations and simulation results verify the concluded functionality of these muscle synergies in response to a slip. Finally, Marigold (Marigold et al., 2003; Marigold and Patla, 2002) found the toe-touch response as a principal strategy required to maintain balance after a slip. Furthermore, he claimed that higher fall incidences in the elderly may be due to their inability in generating a fast toe-touch response, which further substantiates our finding about the faster activation pattern associated with mild slippers in the fourth muscle synergy.

Our conclusion about the first slipping synergy belonging to the terminal swing phase of the gait cycle and being the "deceleration synergy" comes from several observations: First, this muscle synergy (W1, Figure 18) has a dominant activation of TA, VL, and MH of the slipping limb. According to Rose and Gamble (Rose and Gamble, 2005), these muscles are activated during the final stage of the swing phase of the gait cycle. An eccentric (while lengthening) contraction of the MH (due to the activation of its antagonist, VL) should result in a smooth and effective deceleration of the swing limb. Also, the tibialis group will undergo an eccentric contraction to coordinate landing of the foot on the floor that verifies this interpretation. Hence, the observation of these muscles contributions in the first slipping synergy would result in the same physical sub-function as the deceleration of the limb in the terminal swing phase. Second, the activation patterns of the first slipping synergy also stay consistent with our suggested conclusion that the first slipping synergy is the "limb decelerator synergy," because there is a considerable activation, compared to other muscle synergies in Figure 18, immediately after the heel contact (0-100 ms). This immediate activation after the heel contact proves that this muscle synergy is decelerating the limb in the terminal swing phase of the gait. That is because the muscular corrective responses to a slip happen 120-170 ms after the heel strike rather than immediately after heel strike (Cham and Redfern, 2001; Chambers and Cham, 2007; Hur and Beschorner, 2012). Consequently, observation of this significant activation between 0-100 ms post heel strike indicates that this synergy is active even before the corrective response to a slip begin, and hence, belongs to the teminal phase of the gait cycle. The kinetics induced by this synergy (hip extension, knee flexion, ankle dorsiflexion) form the simulation study verifies the suggested functionality of this synergy in terminal swing phase and throughout slipping (Figure 19). Importantly, mild slippers showed a significantly higher activation of the MH muscle, which play the key role in decelerating the limbs in their terminal swing phase (Lockhart and Kim, 2006; Medved, 2000; Rose and Gamble, 2005). Hence, the association found between activation of hamstring muscle group and the mitigation of the slip severity suggests that the mild slippers possess a higher contribution of the "limb decelerator muscle" (hamstring) in their "limb decelerator synergy" (first slipping muscle synergy) prior to slip initiation. Also, another study suggested that higher knee flexion moment leads to more deceleration of the heel and reduced the risk of severe slips (Beschorner and Cham, 2008). This higher contribution indicates the higher capacity of mild slippers in slowing down their base of support and their slipping limbs right before the heel strike.

In the second synergy, the dominant activation of the VL_S resulted in significant hip flexion, knee extension, and plantarflexion (Figure 19, compare with other synergies). As VL plays an important role in supporting body weight, having a high activation level for VL_S suggests that the expected role of this synergy is the weight support on the slipping limb. This subfunction, known as the secondary response to a slip (Cham and Redfern, 2001), is crucial in slip responses since it can be considered as an attempt to continue gait and the forward weight progression on the slipping limb. This weight transfer to the slipping limb helps prevent knee buckling on the unperturbed limb. While there was no significant difference observed in the second muscle synergy (W2), the mild group showed a significantly higher level of activation for this muscle synergy (C2, Figure 18) between 130 ms to 150 ms after the heel strike. Considering the role of this synergy in the forward weight transfer, having a higher activation level offers a stronger weight support provided by the mild slippers compared to severe slippers. This results in a more effective weight transfer of the center of mass over base of support (Chambers and Cham, 2007). Failing to provide enough activation on VL of the slipping limb has been reported to be an involving factor in slip severity in other studies as well (Cham and Redfern, 2001; Chambers and Cham, 2007). The timing of the peak of activation in this synergy (about 200ms post heel strike, Figure 18) stays consistent with our speculation about its sub-task (i.e., the secondary slip response). The simulation results further verified the proposed sub-task for this synergy.

The third muscle synergy was assumed to generate the known “primary slip response” (Cham and Redfern, 2001). This assumption was made based on the following reasons: First and foremost, the primary response tries to retrieve the slipping limb under the body which is achieved by the exertion of a knee flexion and hip extension moment (Cham and Redfern, 2001). The simulation indicates the same moments on the slipping limb (Figure 19), supporting the suggested function. Secondly, the timing of the peak activation of this synergy can provide further evidence about the proposed function. The activation becomes distinct around 160 ms after the heel strike on the slippery surface (C3, Figure 18). According to Cham and Redfern (Cham and Redfern, 2001), the active corrective responses becomes distinct about 150-200 ms after the perturbation; hence substantiating the proposed mechanical goal for the third muscle synergy. In other words, the peak of activation for primary response happens after the “terminal swing synergy” (W1) and before the “secondary response” (W2) (refer to Figure 18). On the other hand, higher activation of TA_S was observed in third slipping muscle synergy for severe slippers (W3, Figure 18) (p value = 0.03). Pretibial muscles are highly activated during the early stance and terminal swing phase (Medved, 2000; Rose and Gamble, 2005). However, an excessive activation of TA muscle on the slipping limb is associated with severe slipping due to an excessive dorsiflexion of the foot. This finding can be also approached by point of view of the FFA. Moyer et al. (Moyer et al., 2006) claimed that severe slippers had a significantly higher FFA compared to their mild slipper counterparts at the heel strike moment. To quantify the FFA in our experiment, the markers data were used to study the angle of the slipping limb right before the heel strike

(Figure 20). The calculated angles for both mild and severe slippers were examined for inter-group differences using an independent *t*-test (SPSS v21, IBM, Chicago, IL). Interestingly, the results verified that the severe slippers had a higher FFA prior to their heel contact (p value < 0.05) (Figure 20). Although unlike Moyer's study, our experiment has mainly focused on investigating the post-slip-initiation incidents rather than pre-slip parameter, there is a high possibility that the association of the high TA_S activation with severe slips stays in the same line with Moyer's claim about the higher FFA in severe slippers.

In the fourth muscle synergy, the most activated muscles belong to the non-slipping limb. Considering the dominant activations (TA, MH, VL), the function of the fourth synergy is to prevent trips and generate a toe-touch response for the non-slipping limb. The trip avoidance happens due to the high activation of TA_NS (Nazifi et al., 2017b) (also supported with the simulation results, compare ankle moments in Figure 19), while the toe-touch is achieved by flexing the hip (Figure 19). Toe-touch is commonly practiced as an effective way to increase the base of support while slipping via extension of the unperturbed limb to touch the ground. On the other hand, a significant higher activation of VL_S (p value $= 0.017$) in the "toe-touch" synergy for mild slippers (W4, Figure 18) suggests that the slipping limb supports the body weight when the trailing limb has not yet touched the ground to provide any weight support. One interpretation could be that the severe slippers were unable to maintain their weight support on the slipping limb to secure enough time for the toe-touch to happen and increase their stability. Moreover, the activation pattern for the fourth synergy (C4,

Figure 18) was also significantly different between different severity groups. Mild slippers were able to recruit their “toe-touch” synergy faster than severe slippers (Figure 18). This result suggests that not only were mild slippers able to provide a better weight support (VL_S activation), but also they could execute the toe-touch strategy faster. This interpretation stays consistent with currently existing literature suggesting that a slow toe-touch in elderly is responsible for more frequent fall incidents (Marigold et al., 2003).

The findings of this paper may facilitate development of a synergy-based targeted motor rehabilitation, which may be a highly convenient and effective rehabilitation method. Targeted motor rehabilitation tries to design interventions that only stimulates and rehabilitates the impaired sub-function of a given motor-task to re-establish the sub-tasks and improve the overall motor-skill. This technique has already been proven to be beneficial in improving motor skills in patients (Dipietro et al., 2007). Subsequently, our findings about the sub-optimally performing sub-functions in severe slippers could be used in developing novel interventions that only stimulates the lost or impaired sub-tasks of slipping in order to transform severe slippers to mild slippers. Future studies will assess the extent of improvements in severity index of severe slippers after exposure to the aforementioned training method.

There were also a few limitations associated with this chapter. First, although this study revealed the association between severe slipping and adverse post-slip-initiation response, it is still unclear if this relation is causal or not. More investigations are required to clarify if there is a causal relation between slip severity and adverse post-slip-initiation response. To resolve this limitation, in future studies we will use interventions to improve the slip-response in participants to see if it results in mitigation of severity. We believe that ‘severe slipping synergies’ will evolve to ‘mild slipping synergies’ as the participants undergo slip trainings (Alnajjar et al., 2013). Also, a correlation analysis would further clarify the relation between slipping muscle synergies and different slip severities and result in a relation the severity index to the level of deviations observed from the synergies of reference mild slippers (Cheung et al., 2012)l., 2012). Lastly, model-based experiments can be performed to easily modulate experimental conditions and examine the causal relationship. In future works a wider range of age would be considered to recruit older adults. Also, future studies can perform kinematic analysis (only kinetic analysis was used in this chapter) in order to further investigate the functionality and importance of each muscle synergy of slipping.

4.5. Conclusion

This chapter has investigated the inter-group differences in slipping muscle synergies of the mild and severe slippers and identified several significant differences. This chapter also utilized a forward dynamic simulation in order to study the sub-task that each synergy is responsible for. Finally, using the physical interpretation of each synergy, along with the discrepancies observed between group, this study determined the possible malfunctioning sub-tasks in severe slippers which cause persons to experience more severe slips rather than mild slips. Also, while there were no differences in age, height, and mass observed between the two severity groups, there were several significant differences in the slip responses (reflected as differences in muscle synergies) and motor control of mild and severe slippers. Consequently, these points together suggest that the slip severity outcome may be associated with the slip response of the individual rather than other physical differences.

The results of this chapter could potentially result in development of a targeted motor-rehabilitation based on the deficient muscle synergies. Such trainings will aim at re-establishing the lost or impaired muscle synergies (and the corresponding sub-tasks). The efficacy of such a training will be tested in future studies. Synergy-based targeted motor-rehabilitation, if found effective, would be more convenient and practical, as it addresses only the lost sub-task (less complex to practice), instead of the original motor-task (more complex to practice).

5. DO WALKING MUSCLE SYNERGIES INFLUENCE PROPENSITY OF SEVERE SLIPPING?[4]

5.1. Introduction

Prior research has indicated the potential influence of motor control during gait on an individual's risk of falling. Moyer et al. (2006) used kinematic metrics of human gait (e.g., cadence) to evaluate an individual's risk of experiencing a severe slip, indicating the link between gait parameters and slip severity. Given that muscle activations can imperatively affect the resulting kinematics, one may suspect that a similar link might relate slip severity to the lower extremity muscle activation patterns during walking. This speculation is also substantiated by our previous chapter claiming that the CNS uses the same patterns/modules to control both human walking and slipping based on a muscle synergy approach (Nazifi et al., 2017b). Although our previous studies extracted walking muscle synergies, it is still unknown if the walking muscle synergies differ for individuals with different slip severity. Such knowledge is valuable as it may potentially result in a novel diagnosis method that only relies on walking behavior of participants to eventually predict their slip severity.

In sum, this chapter intends to understand how muscle synergies observed during walking differ for the individuals who were classified as severe slippers compared to

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those who were classified as mild slippers. We hypothesize that muscle synergies of walking will differ between mild and severe slippers. Since the walking muscle synergies represent the neural control of the gait, the observed differences in different severities may potentially show the effect of the neural control of the gait on slip severity. Such knowledge is valuable since each muscle synergy is shown to be associated with a physical sub-task of a gross motor-task (Nazifi et al., 2017a). Hence, comparing walking muscle synergies is equivalent to identifying the walking sub-tasks that differ between mild and sever slippers. These differences in walking muscle synergies may help pinpoint the underlying limb coordination and walking habits that may contribute to a higher risk of fall on the slippery surface and can potentially be used in the development of programs for slip/fall prevention, diagnosis, and rehabilitation.

5.2. Methods

5.2.1. Participants

A total number of twenty healthy young adults (9 females, 11 males) with an average age of 23.6 years old (SD = 2.52 years) participated in this study. Participants had no history of illnesses affecting gait (e.g., musculoskeletal, neurological, cardiovascular). All participants signed the written consent forms prior to participation in this IRB-approved experiment at the University of Pittsburgh. Upon a secondary approval from IRBs of University of Pittsburgh and Texas A&M University, the anonymized data were analyzed in Texas A&M University for the current study.

5.2.2. Measurements, Experimental protocol, and Data Processing

Participants were asked to walk at their comfortable speed along a ten-meter pathway with an embedded force plate at the middle. There were two or three practice trials before the main walking trial (i.e. data recording trial). The starting point was adjusted in each trial to make participants step on the force plate with their right limb. All participants were provided with Polyvinyl chloride (PVC) soled shoes in their size to control the coefficient of friction across all the participants. Participants donned a harness to protect them from any potential injuries due to slipping during testing.

After the practice trial, participants performed a normal walking trial during which the EMG signals and markers data were recorded for synergy extraction. After the normal walking trial, without informing the participants of a change in walkway condition, a slippery solution was applied to the force plate. The slippery solution was a diluted glycerol solution, with 75% glycerol and 25% water, that has shown promise in providing a slippery surface by other researchers (Beschorner and Cham, 2008; Chambers and Cham, 2007; Moyer et al., 2006). The coefficient of friction was 0.53 and 0.03 for the dry and slippery conditions, respectively. Then, participants performed an unexpected "slip trial" to classify the participants into different severity groups (Figure 21). To minimize the audible and visible cues and ensure an unexpected slip, we administered the following: we dimmed the lights during the whole experiment. Also, between the trials, participants were asked to look away from the walkway, while listening to loud music for one minute. Next, participants were asked to turn around to face the walkway again, place their feet on the instructed location, and start walking on the signal. Participants were also asked to keep their eyes at an eye-level target on the wall (i.e., horizon). The slipping trial was only recorded to classify participants into either mild or severe slippers while the main data analyzed in this chapter was the walking behavior of the participants.

Throughout the walking trials, bilateral EMG signals were recorded at 1080 Hz from medial hamstring (MH), tibialis anterior (TA), vastus lateralis (VL), and medial gastrocnemius (MG) (right/leading/slipping leg (L) and left/trailing/non-slipping leg (T)) (Figure 21). A motion capture system (Vicon 612, Oxford, UK) was utilized to capture heel kinematics at 120 Hz.

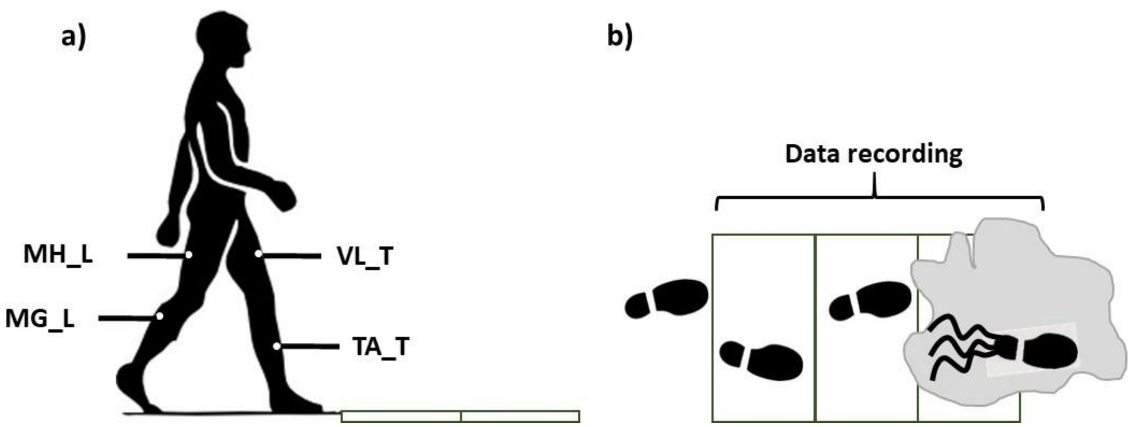

Figure 21 The side view (a) and the top view (b) of the walkway in the final trails (slip). The gray zone indicates the contaminant (reprint from Nazifi et al., 2019)

The PHS of each participant was used as representative of slip severity using the markers data. Upon recording the walking data, the slip data was used to classify participants into severe and mild slippers. Persons with a PHS higher than 1.44 m/s were considered severe slippers (T. E. Lockhart et al., 2003) while others were labeled as mild slippers. *t*-test was used to identify potential inter-group differences in weight, height, and age between mild and severe slippers. Also, a Pearson's Chi-squared test was performed to examine if there was a significant difference between genders of mild and severe slippers (Table 1).

EMG signals were processed (demeaned, rectified, filtered) for the walking trial according to previously-described procedures. EMG was then normalized to the maximum activity of each muscle for each participant throughout all of his/her walking and slipping trials. The force plate data was only used to detect the heel strike moment and the gait duration was normalized to 100 points (0 being the first right heel strike, 50 being left heel strike, 100 being the second right heel strike) and an iterative non-negative matrix factorization technique was used to extract walking muscle synergies and their coefficients from the normalized gait cycle for each participant (Nazifi et al., 2017b, 2017a). Prior chapters indicated that for walking, four muscle synergies are enough to reconstruct the EMG signals of walking and reach a VAF larger than 95%, hence the number of synergies were fixed to four in this chapter. Muscle synergies of each severity group were then sorted and re-ordered according to their similarity using correlation coefficients (r) (d'Avella et al., 2003; Nazifi et al., 2017b; Torres-Oviedo and Ting, 2010) to have the same synergies (i.e., ones with the highest correlation) in different participants in the same order (Figure 22). This step was necessary as our method extracted synergies in a random order for each participant. An independent *t*-test (α=0.05) was used (SPSS v21, IBM, Chicago, IL) to detect the significant differences in muscle synergies between mild and severe slippers. We then used Bonferroni's 95%

confidence interval to examine if the time courses of activation coefficients diverge between the mild and severe slippers.

5.3. Results

From the unexpected slip trial, PHS measurement classified twelve participants as mild slippers and eight participants as severe slippers. No significant differences were found in sex, height, mass, and age across severity levels (Table 1). According to our previous chapter, four muscle synergies were enough to account for more than 95% of the EMG variability during walking. Hence, four walking muscle synergies were extracted from each participant (Figure 22). We wish to emphasize that the slipping trial was only performed to classify participants into potential mild and severe slippers, and the synergy analysis was performed only on walking trials.

Statistical analysis revealed differences in both walking muscle synergies and their activation coefficients (Figure 22). Significant differences contributions of three muscles were found. The different contributions belonged to MH_T, TA_T, and VL_L muscles. MH_T and VL_L muscle had a significantly higher activation in mild slippers, while a higher activation of TA_T was associated with severe slipping (Figure 22, walking muscle synergy 1 and 3, i.e. W1 and W3). The inter-group comparison also found differences in activation of the first and the third walking muscle synergies (Figure 22, C1 and C3). Bonferroni's 95% confidence interval showed a divergence in the first synergy's coefficient 'C1' between mild and severe slippers from 11th until 15th percent (Figure 22) of the gait cycle. A higher activation of C1 in the aforementioned period was associated with mild slips. Also, the activation coefficient of the third muscle synergy, ‘C3’was higher in mild slippers from 37th percent to 45th percent (Figure 22) of the gait cycle according to the same analysis.

5.4. Discussion

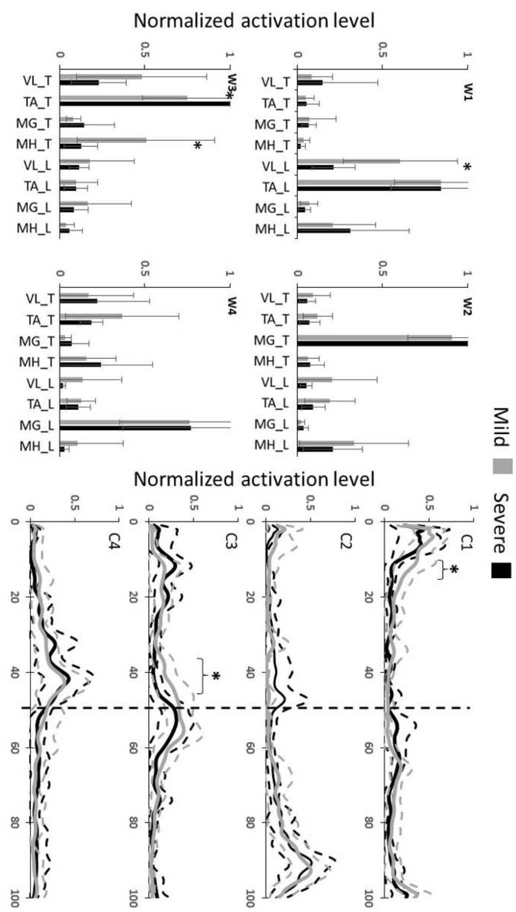

Figure 22 The walking muscle synergies and their activation coefficients for both severity groups. Dashed lines and error bars indicate SD. Asterisks represent significant differences. (0% coincides with leading limb’s heel strike and the trailing limb’s heel strike happens at 50%, i.e. the vertical dashed line). (reprint from Nazifi et al., 2019)

The results have indicated that a higher activation of the MH muscle right before the heel strike is associated with less severe slips (Figure 22, MH_T right before the trailing limb's heel strike at 50%). This point can be seen in both higher contribution of MH_T in the third synergy (W3), and in its higher activation coefficient (C3), right before the trailing limb's heel strike (which happens around 50%). Hamstring muscle is known to be involved in deceleration of the swing (same as trailing here) leg in right before the heel strike (Basmajian and De Luca, 1985; Lockhart and Kim, 2006; Medved, 2000; Rose and Gamble, 2005), hence, contributing to less slip severity by reducing the heel velocity at the moment of heel strike. According to a prior chapter (Nazifi et al., 2017a), a relatively similar sub-task was found to be associated with one of the slipping muscle synergies, where participants with a higher contribution of Hamstring group during slipping experienced less severe slips (Nazifi et al., 2017b). This fact further clarifies the key role that Hamstring group play in fall prevention and slip recovery. Interestingly, as the MH contribution is higher both after a novel slip initiation according to (Yang and Pai, 2010) and before slip initiation (i.e., in this chapter, walking trials), it is probable that not only do the hamstring group has a reactive role in fall prevention, but also it may have a proactive role as well.

Moreover, prior studies on walking muscle synergies found one of the synergies be responsible for the 'load acceptance' synergy(Nazifi et al., 2017b). This synergy prepared the weight to be shifted from the trailing limb to the leading limb. Despite the different scope of the studies, a comparable pattern was found in W1 in this chapter and is considered to be associated with the load acceptance. Statistical analysis of this synergy shows that a higher activation of the VL muscle right after the heel strike is associated with less severity in slips. This conclusion was made upon the observation of a higher contribution of VL_L in W1 along with a higher activation in C1, right after the leading limb's heel strike (which happens around 0%, Figure 22). Considering the role of VL in load acceptance, this conclusion stays consistent with existing studies claiming that a late activation of the VL may reduce the forward velocity of the center of mass relative to the base of support, resulting in less stability (Cham and Redfern, 2001; Chambers and Cham, 2007). In other words, mild slippers had a higher activation on their load acceptor muscle (VL_L) shortly after leading limb's heel contact enabling them to transfer their weight with more support.

The third muscle synergy suggests a toe lift behavior. Based on the contribution of each muscle it is likely that this synergy contributes to elevation of the toes right before the heel strike, probably to avoid tripping or foot drop. However, there was an association between higher activation of the TA muscle before the heel strike and high slip severity. TA_T muscle had a higher contribution in the third muscle synergy (W3). It was previously shown that severe slippers' high activation of TA right before their heel strike increases their foot-floor-angle significantly (FFA) and was found to be associated with their severe slips (Nazifi et al., 2017a). An excessive dorsiflexion and FFA right before the heel strike also challenge achievement of flat-foot and recovery (Chambers and Cham, 2007). It is also known that a reduced FFA (i.e., flat-foot walking) is a strategy used by individuals to increase dynamic stability of the gait (Bhatt et al., 2006; Gao and Abeysekera, 2004; Marigold and Patla, 2002; Strandberg and

Lanshammar, 1981). Finding of an excessive contribution from TA muscle in walking muscle synergies of severe slippers verifies our findings about a correlation between FFA and propensity to falls while normal walking.

5.5. Conclusion

This chapter examined the walking muscle synergies and their differences for different slip severities. We found significant differences in the walking muscle synergies of mild and severe slippers. This study provides a basis for a potential diagnosis method to identify the vulnerable population and people with high risk of fall based on solely their walking pattern and improves their safety and consequently, quality of life. Such a diagnosis method will be valuable as it does not require an actual slip trial once a predictive model is developed. There were a few limitations to our study. Our study was only performed on the young adults and can be significantly improved by including older populations. Also, as falls impose more detrimental consequences on the older adults, our future studies would test the differences between mild and severe slippers in older populations. Moreover, this study only targeted unexpected slips. A potential different can be present between the response to unexpected and expected slips that can be addressed in future. Also, only eight major muscles (i.e. four muscles per limb) were studied. Future studies can resolve this limitation by studying more muscles that may contribute to human gait. Future studies also will develop and study the effectiveness of a predictive model in identifying severe slippers. Another limitation of this chapter is the limited number of participants. A future study is needed to confirm the findings in a larger group. Also, despite that our statistical analysis did not find the slip severity to be gender-related, since prior studies have shown gender-related discrepancies in walking patterns (Cho et al., 2004), gender's association with slip severity will be studied in a larger data set. Lastly, this study has not controlled for footedness of the participants and it can be improved by controlling the footedness of each participant upon heel strike on the slippery surface.

6. A MUSCLE SYNERGY APPROACH IN EVALUATION OF THE GAIT COMPLEXITY FOLLOWING SURGICAL ALIGNMENT

6.1. Introduction

Adult degenerative scoliosis (ADS) is generally defined as an abnormal 3D (mainly lateral) curvature of the spine caused by asymmetric degeneration of discs (Birknes et al., 2008; Cho et al., 2014). ADS is a common musculoskeletal problem in the elderly, affecting up to 68% of their population with an average age of 70 years old (Carter and Haynes, 1987; Cho et al., 2014; Schwab et al., 2005; Silva and Lenke, 2010). ADS patients often suffer from back pain, shooting leg pain, and unnatural spine curvature that affects their gait; hence, surgical intervention is one of the potential treatment strategies to reestablish the disk spacing that causes the deformity (Cho et al., 2014; Dakwar et al., 2010). Although surgical treatment has resulted in a significant improvement in measures such as back curvature, low back pain, and quality of life, given the invasive nature of the operation and exposure of the spine, secondary complications such as infection, neurological deficit, and risk of death are possible (Birknes et al., 2008; Dakwar et al., 2010). Most importantly, due to the significant dissection and trauma, patients have to be hospitalized and immobilized for an extensive amount of time until their recovery (Benglis et al., 2008; Dakwar et al., 2010; Wang et al., 2008). Not only would the slow recoveries along with the lengthy lack of mobility put patients at higher risk of secondary conditions, but also it would make it challenging to evaluate their improvements following surgery.

Improvement in ADS patients could be measured with numerous variables such as pain levels, the severity of the deformity (commonly quantified via measuring the angle of curvature using Cobb's method (Cobb, 1948)), and gait symmetry index. These variables try to provide measures that indicate an enhancement in the and quality of daily life, and how advanced and complex the patients could perform specific motor tasks, such as gait. Nonetheless, due to the high prevalence and subject-specific nature of ADS, patients can vary in numerous aspects such as the affected side (i.e., left or right), Cobb angle, number and location of the affected disks (e.g., T7-L3), and unique 3D curvature of the spine. This high variation can further hinder the tracking of the gait complexity (i.e., a measure of gait quality) following surgery, since subjects may show higher/lower activations in different muscle groups on different sides to different degrees. Hence, a novel measure is required that considers all muscles and provides a generalized complexity index robust to the aforementioned subject-specific variations.

The concept of entropy might serve this purpose. Entropy has existed for a considerable amount of time in fields like thermodynamics, statistics, and information theory. Entropy is a quantity presenting randomness, disorder, and lack of information. Although entropy has shown promise in explaining the behavior of systems in different areas, it has been seldom used in the field of biomechanics (Friston, 2010; Hur et al., 2019). According to (Hur et al., 2019), a decrease in entropy is associated with an enhancement in postural balance, since less entropy expresses less chaotic and more deterministic control to move the center of mass only in directions ensuring a more secure balance. Similarly, one may expect to observe a decrease in gait's entropy values upon surgical interventions indicating potential improvements. As a result, the entropy of the muscles would be robust to the variations observed in ADS patients as it considers all muscles at the same time (Figure 23).

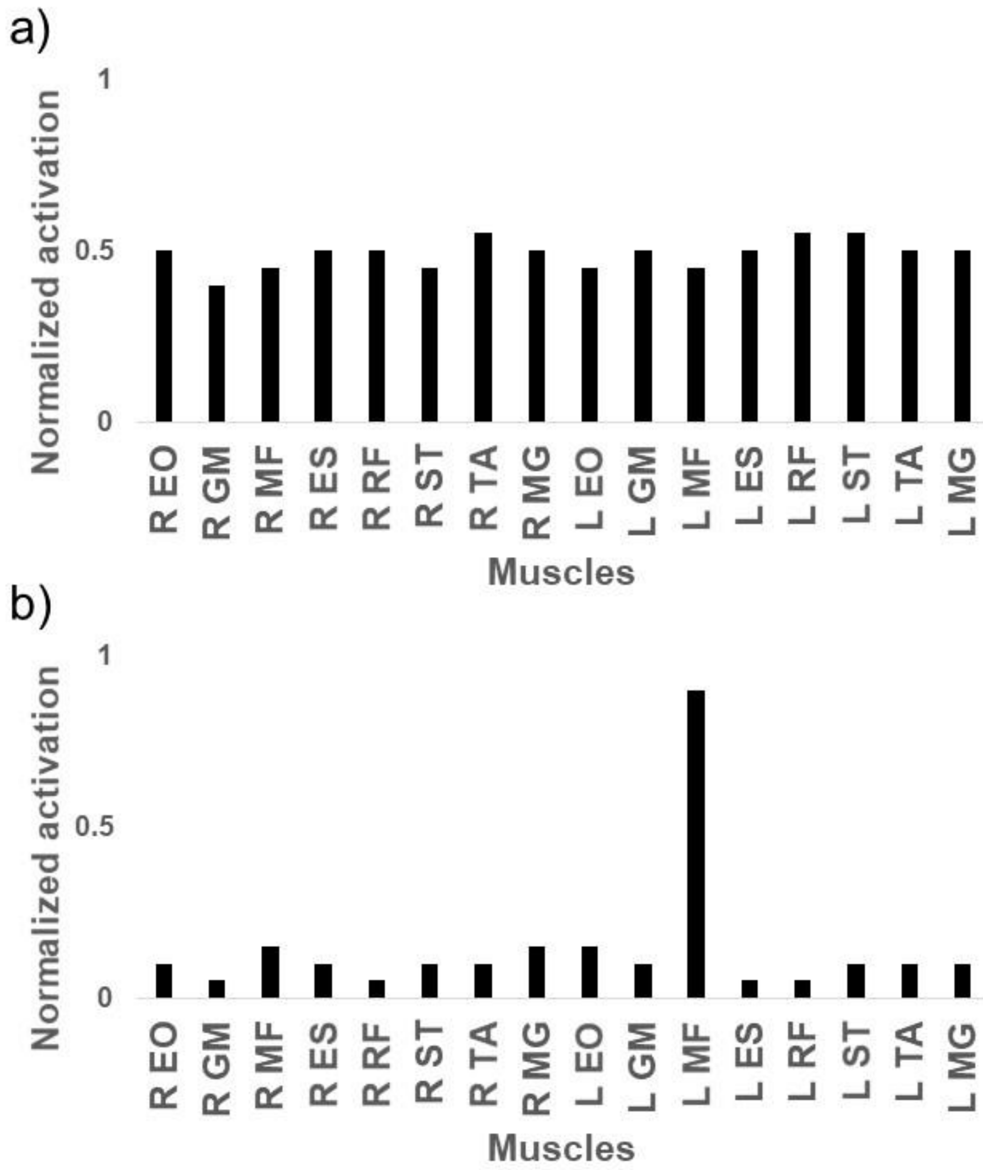

Figure 23 Two generic muscle synergies with high entropy (a) and low entropy (b) values. Note the randomness and disorder in (a).

As importantly, the concept of muscle synergies could be another candidate for indicating complexity. Researchers often fix the number of used muscle synergies to account for 95% of the variation observed in the collected EMG signals from the target muscles. The number of muscle synergies is of importance as studies have indicated an association between a greater number of required muscle synergies and higher quality and complexity of the performed motor task (Clark et al., 2010; Shuman et al., 2016). In other words, participants who require a higher number of synergies for the same task often present a higher quality of the motor task. For example, studies have shown that post-stroke patients who had a superior gait quality and residual function utilized a greater number of walking muscle synergies (Clark et al., 2010), indicating their CNS ability in generating independent control signals. Also, previous studies have interpreted an increase in the number of muscle synergies as a measure to indicate post-surgery improvements (Shuman et al., 2016). Since muscle synergies have shown more promise and robustness in explanation of the motor tasks (Chvatal et al., 2011; Nazifi et al., 2019, 2017b, 2017a; Ting and Macpherson, 2005), we suspect that using the number of muscle synergies can facilitate the tracking of gait improvements following surgical alignments despite the subject-specific conditions discussed above.

The objective of this study is to examine the number of muscle synergies and their associated entropy as a measure of gait improvement/complexity following surgical interventions in ADS patients. Both the number and the entropy of synergies would be compared both before and three months after surgery to detect significant improvements. We hypothesize that the number of utilized walking muscle synergies would show a significant increase indicating a more complex and richer gait control, while their associated entropy would decrease, showing a more deterministic control towards the gait.

6.2. Methods

6.2.1. Participants

Clinical gait analysis was performed on thirteen ADS patients (Average age: 61.77 (SD=11.19), Sex: 8 female/5 male), one week prior, and three months post-surgery. Subjects gave written consent before participation in this IRB-approved study. To maintain a moderate severity of ADS, participants were excluded in the case of a Cobb angle larger than 50 degrees (as a controlling factor for their ADS severity). Subjects had at least four fused levels in thoracic, lumbar, and thoracolumbar parts.

6.2.2. Measurements, Experimental protocol, and Data Processing

Five walking trials were performed at the self-selected speed of the patients for data collection. Patients were asked to walk normally in a walkway with three force plates embedded. The starting position was adjusted to secure one heel-strike per force plate. Surface EMG data were collected from sixteen trunk and lower extremity muscles: External Oblique, Gluteus Maximus, Multifidus (at the level of L5), Erector Spinae (at the level of L1), Rectus Femoris, Semitendinosus, Tibialis Anterior, Medial Gastrocnemius, bilaterally. The EMG and force data were collected at 2000 Hz.

The EMG data were high-pass filtered at 20 Hz, low-passed at 450 Hz, demeaned, rectified, and low-passed at 35 Hz with zero-lag Butterworth filter via

MATLAB (v2017b, The MathWorks, Natick, MA). The data was then normalized to the maximum activity observed among all trials of each participant. Then the processed EMG was used in a non-negative matrix factorizer to extract walking muscle synergies and their activation coefficients using the methods described in (Nazifi et al., 2019, 2017a). The number of synergies were varied, starting from one to sixteen (maximum possible due to the number of recoded muscles). Then, “the required number of synergies” was found as the minimum number of synergies that could reconstruct EMG signals with a VAF higher than 95%, defined as Equation 3.

This procedure was done for every participant, both pre- and post-surgery. A paired *t*-test was used to test the hypothesis (α=0.05) using SPSS (v21, IBM, Chicago, IL).

In order to perform a one-to-one comparison of the extracted muscle synergies, the same number of muscle synergies had to be extracted for each subject. This number was set to ensure the encompassment of the most synergies of the majority of the participants. To do so, using the mean and standard deviation value for the number of synergies in different individuals, a confidence interval was built, and the number was selected to be the upper bound of a 95% confidence interval, using the following equation:

$$\mu \leq \bar{x} + \sigma \times z$$

Equation 4

where μ is the upper bound for the true mean, x is the sample mean value, σ is the standard deviation, and z is the critical value for the normal distribution to ensure a 95%

encompassment with a significance of 0.05. Then, the bound for the true mean was rounded to the closest whole number.

Upon fixing the number of muscles, synergies entropy was calculated. The concept of entropy is often used on probability functions in which the summation of all possible events is equal to one. To follow the same procedure, first, the summation of all activations in a synergy was normalized to one, to enforce each synergy to resemble a probability density function (Figure 24).

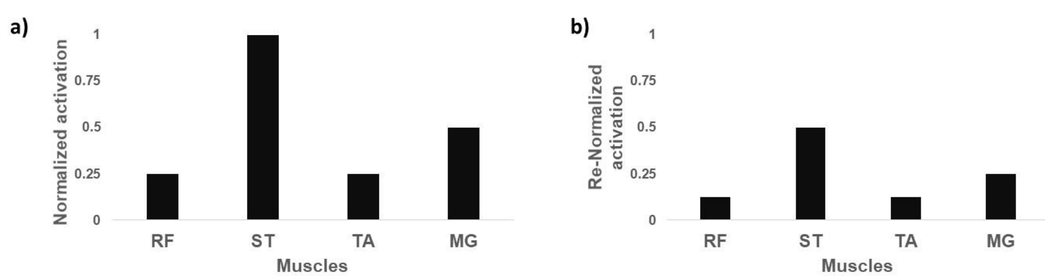

Figure 24 Ordinary (a) and re-normalized (b) muscle synergy. Note that sum of all activations is one in (b).

Then, the entropy of each muscle synergy of a participant was defined as:

$$H(W_i) = -\sum_{j=1}^{n} P(j) \log_2 P(j)$$

Equation 5

where H is the entropy value, W_i is the ith muscle synergy, n is the total number of muscles, and P is a probability which was replaced by re-normalized muscle synergy value (Figure 24). Then, the pre- and post-surgery entropy was compared using a paired *t*-test via SPSS (v21, IBM, Chicago, IL) to identify potential differences at a significance of 0.05.

6.3. Results

First, the number of the required synergies (i.e., Equation 3) for each participant was compared before and after surgery. The *t*-test revealed a significant increase in the number of required muscle synergies for walking in ADS patients following a surgical procedure (Table 3).

	Pre-surgery	**Post-Surgery**	***p*-value**
Number of synergies (Mean ± SD)	4.46 ± 1.33	5.07 ± 1.44	0.04

Table 3 Required number of walking synergies pre- and post-surgery, and *t*-test results.

Second, to run the one-to-one comparison of the synergies, an equal number of synergies needed to be extracted for all subjects. The z-value was extracted to be 1.64 from statistics references. Then, according to the values reported in Table 3 and using Equation 4, it was found that <u>seven</u> muscle synergies would serve as an upper bound for 95% of subjects both before and after surgery. Consequently, in our secondary analysis, irrespective of Equation 1 and the number of the <u>required</u> synergies, seven muscle synergies were extracted for every individual (Fig. 2).

The paired *t*-test also showed a highly significant reduction in the measured entropy following surgical interventions presented in Table 4.

Muscle synergy	**W1**	**W2**	**W3**	**W4**	**W5**	**W6**	**W7**
Pre-surgery entropy	3.26 ± 0.24	3.41 ± 0.17	3.22 ± 0.25	3.31 ± 0.19	3.30 ± 0.18	3.36 ± 0.21	3.30 ± 0.22
Post-surgery entropy	2.99 ± 0.30	3.22 ± 0.17	2.87 ± 0.28	3.09 ± 0.23	3.14 ± 0.22	3.17 ± 0.22	3.02 ± 0.25
***p*-value**	<0.0001	<0.0001	<0.0001	<0.0001	0.0013	0.0003	<0.0001

Table 4 Significant differences in entropy values of muscle synergies before and after surgery

Also, a clinical comparison following surgery showed that all participants had improved significantly in different variables such as Cobb angle, cadence, step length, single support time, and gait deviation index, all with p-values<0.05 (Haddas; et al., 2020).

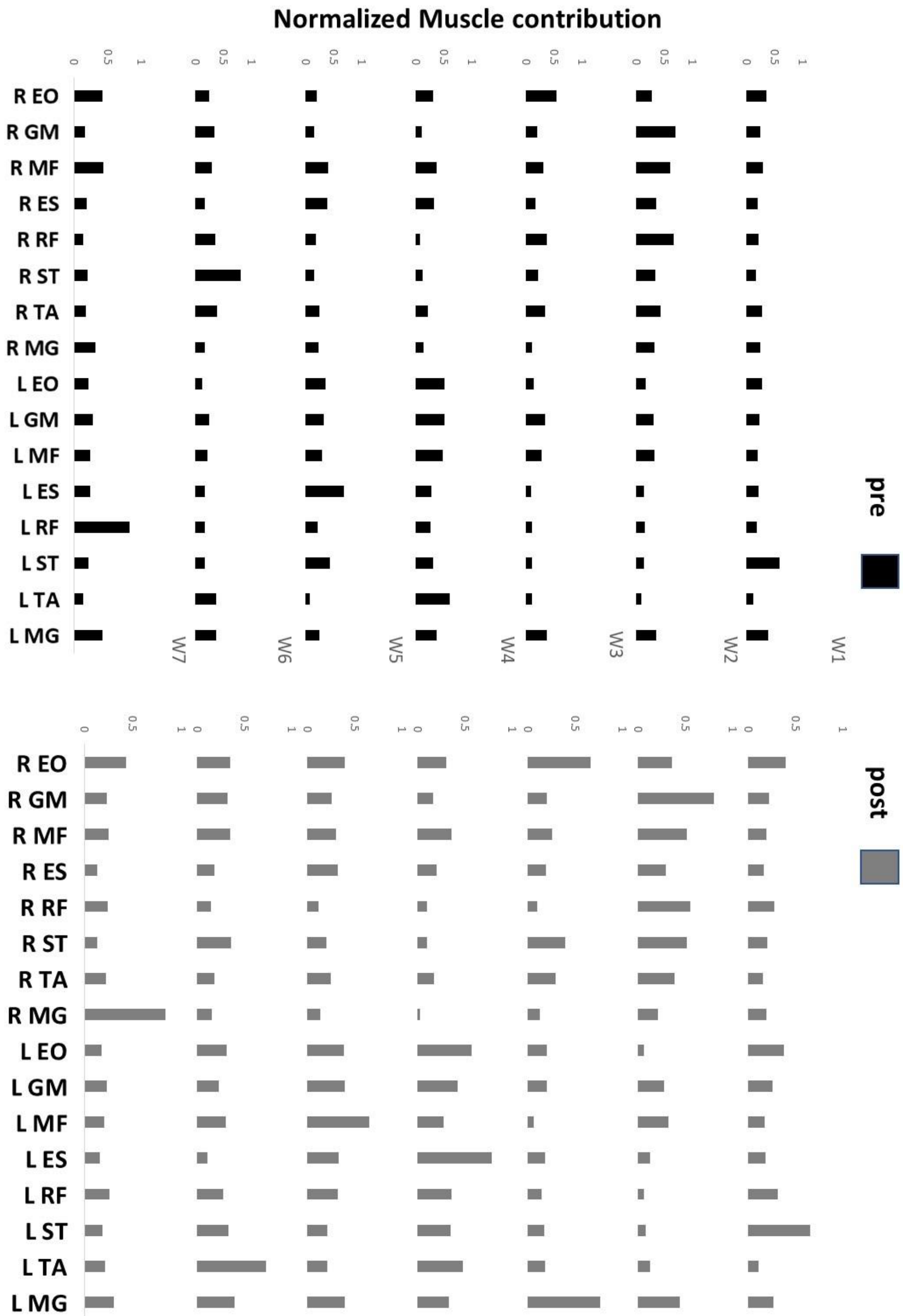

Figure 25 Pre- and Post-surgery walking muscle synergies.

6.4. Discussion

The results indicated that ADS patients had been clinically improved upon surgery. The prolonged single stance phase of the gait along with longer step lengths indicate the ability of the subjects to maintain their balance on one limb for longer periods. These results confirm our hypothesis about the number of required muscle synergies. ADS patients required a higher number of synergies to present their walking following a surgical alignment. As discussed before, a higher number of synergies is associated with a richer control signal that requires more vectors (i.e., synergies) to rebuild (Clark et al., 2010; Shuman et al., 2016). Hence, we suggest that the surgery has improved the gait quality for ADS patients at their three-months follow up. We also suspect that the observed changes are probably due to a reduction in pain levels of participants. Although the relation between motor control and pain is an ongoing research topic, pain adaption theories claim that pain affects motor control (Farina et al., 2003). Researchers believe that low back pain (which is highly prevalent in ADS patients) has the potential to change the co-activation patterns in a way to restrict spine movements as a measure to minimize the pain (Hodges and Jull, 2007; Hodges, 2011; Hodges and Moseley, 2003; Lund et al., 1991).

Furthermore, the same researchers claimed that pain does not necessarily cause inhabitation or excitation of the muscles; however, it may bring a redistribution of the activity within or between synergist muscles (Hodges, 2011). Hence, observation of changes in muscle synergies following surgery should be inevitable. Our results support this hypothesis by presenting a redistribution in muscle activities throughout muscle synergies following surgical interventions in ADS patients (Figure 25). Moreover, studies claim that in order to protect the body from pain, the central nervous system may decrease the activation levels in agonist muscle groups and increase the activation in the antagonist group as a measure to increase the joint stiffness in order to restrict the range of motion in joints (Hodges, 2011; Hodges and Moseley, 2003; Rudolph et al., 1998). Excessive activation of the antagonist muscles in synergies would increase the entropy by definition, as it requires activation of a greater number of muscles that is equivalent to less deterministic control. This theory stays highly consistent with our results that show a significant decrease in the entropy of walking synergies of ADS patients following surgery, which can be due to a reduction in the unnecessary co-activations enforced to reduce the pain. Our future studies would try to correlate a pain level questionnaire with the observed entropies to further verify our hypothesis.

The decrease in entropy also substantiates our initial hypothesis. We speculated that a lower entropy would be associated with a less random and more deterministic control of the muscles by the CNS that ultimately results in an improved gait. In other words, the CNS may more deliberately choose muscle activations in a way to secure a more stable gait. A similar concept had been presented before on balance studies (Friston, 2010; Hur et al., 2019). However, our study is the first to examine the concept of entropy in human gait. Due to the association between the clinical gait parameters and the decrease in entropy, we claim that studying the entropy can potentially be used as a yardstick to track the improvements in gait, specifically in the presence of high subjective variations.

6.5. Conclusion

This study examined the potential of the number of muscle synergies and their associated entropy in tracking the gait improvements of ADS patients following surgery. The patients who had already shown clinical gait improvements showed an increase in the number of required walking muscle synergies. Requiring a higher number of synergies could be an indicator of requiring more control signals to control the gait, which is associated with more advanced control. Also, the entropy associated with those synergies dropped significantly. A lower entropy may represent a more deterministic control of the limbs and maybe a beneficial tool in tracking the rehabilitation of patients. Future studies would try to examine the association between the observed clinical improvements and the evolution of muscle synergies. Also, we are interested in performing the same analysis on the subject at their one-year follow-up to observe the course of changes in the entropy and number of muscle synergies.

7. CONCLUSION

This dissertation aimed to study walking and slipping, and more specifically, severe slipping. Several kinematic variables were compared between the mild and the severe slippers. We found that angular momentum has the earliest deviations between the mild and the severe group. All other kinematic variables differed between the two severity groups during slipping; however, none were different during walking. The time lead on deviations in angular momentum showed the importance of this variable in maintaining balance. Next, a muscle synergy approach was chosen to study walking and slipping. We found that walking and slipping each required four muscle synergies to describe, from which two were shared between the two tasks.

Moreover, the observed differences between the muscle synergies of slipping and walking between the two severity groups suggested that both gait control and slip control are different between mild and severe slippers. The adverse gait control and slip control may be the underlying issue that puts severe slippers at a higher risk of falls. Lastly, muscle synergies and their associated entropy were used to track the gait improvements in Adult degenerative scoliosis patients. Both measures successfully predicted the observed clinical improvements. The number of the required muscle synergies and their associated entropy proved to be a promising tool in tracking enhancements in the motor task with potential benefits in the rehabilitation field.

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