THE PERCEPTION AND CONTROL OF WEIGHT DISTRIBUTION DURING SIT-TO-STAND IN HEMIPARETIC INDIVIDUALS: CAN ASYMMETRY BE ATTRIBUTED TO A SENSE OF EFFORT?

by

Jennifer Lindsay Sullivan

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Abstract

Hemiparetic stroke survivors often produce asymmetric forces when performing bilateral tasks, despite their perception that the forces are equal. It has been hypothesized that this asymmetry is due to the use of effort – as opposed to force magnitude – as the controlled parameter in bilateral force-matching. That is, human perception of force and weight seems to be based more on the intensity of the outgoing motor command than on afferent feedback. This thesis is focused around an experiment that investigated whether this sense of effort (SOE) plays a dominant role in the control and perception of weight distribution during a functional task, sit-to-stand (STS). Eight chronic stroke survivors and eight healthy controls performed a series of STS trials using a robotic assist device, which employed a rate-controlled, 1-degree-of-freedom rotating seat to allow users to perform the STS movement without having to support their entire body weight. The amount of assistance provided by the device was varied across trials in order to measure STS weight distribution in the context of large, medium, and small load magnitudes. The influence of SOE on the control strategy was assessed by evaluating whether or not the proportion in which the load was distributed between limbs was constant across all load magnitudes. Two types of linear models were fit to each group's data to quantify the relationship between weight distribution and load: one treating the slope as a fixed parameter, and one incorporating an interaction term. Results suggest that while SOE does influence the employed sensory-motor strategy, afferent feedback is a factor as well. Furthermore, the relative contributions of centrally-generated versus peripherally-generated signals varies among individuals: specifically, SOE has a larger influence on the control strategy of individuals who are more symmetric than those who are more asymmetric. Based on these results, we recommend that improving stroke survivors' awareness of their movement asymmetries and targeting their perceptual inaccuracies in therapy may serve to facilitate and expedite the rehabilitation process.

Preface

The author was primarily responsible for the direction and work involved in this thesis, including the background literature review, identification and formulation of the research question, experimental design, subject recruitment & screening, data collection, and data analysis. Usage of the first person in this document refers to the author and her supervising professors, Dr. Elizabeth Croft (UBC Mechanical Engineering), Dr. Antony Hodgson (UBC Mechanical & Biomedical Engineering), and Dr. Machiel Van der Loos (UBC Mechanical Engineering).

The experiment described in this thesis was conducted with the approval of the UBC Clinical Research Ethics Board under application number H13-01271, "Evaluation of Weight Distribution Asymmetry during Assisted Sit-to-Stand." The sit-to-stand assistive device used in the experiment was originally designed, constructed, and programmed as part of a prior master's project by Jeswin Jeyasurya (Jeyasurya, 2011). Modifications to the hardware and the code were made by the author to accommodate this experimental design.

The statistical analysis method, particularly related to equivalence testing and model fitting, was chosen based on the counsel of a statistics consulting service offered by the UBC Department of Statistics.

A paper reporting preliminary results and analysis of this work was accepted and presented at the Canadian Medical and Biological Engineering Society Conference 2014 in Vancouver, BC. The manuscript was written by the author and presented by Dr. Machiel Van der Loos.

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Abbreviations & Glossary

Afferent	Signal direction, ascending from periphery towards central nervous system
Asymm	Asymmetry *(computed metric)*
BWD	Body weight distribution *(computed metric)*
CNS	Central nervous system
D, ND	Dominant, non-dominant (leg/side)
Dev	Deviation *(computed metric)*
DF	(Ankle) dorsiflexors
DLR	Distribution-load relationship *(computed metric)*
DOF	Degrees of freedom
EB (+/-)	Equivalence Bounds (upper/lower)
Efferent	Signal direction, descending from central nervous system to lower motor system
KE	Knee extensors
MMT	Manual Muscle Testing
MVC	Maximum voluntary contraction
Paretic	Affected by brain injury, characterized by weakness (paresis)
PF	(Ankle) plantarflexors
SO	Seat-off
SOE	Sense of effort
STS	Sit-to-stand
VR	Variability *(computed metric)*

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I extend a most sincere thank you to my lab mates and friends, within and outside of the department, for your tips and tricks, suggestions, generosity, and unwavering encouragement; to my roommates, in particular, for putting up with the sprawl of papers occupying our dining room table; and of course, to Ron and Bailey, for your consistently uplifting spirits.

And finally, a very special thank you to my parents, for a whole lot more than could ever be articulated here.

For N.W.H. –

Here's your robot.

Chapter 1:

Introduction

Stroke has become one of the leading causes of disability in Canada, draining an estimated $3.6 billion in health care costs and lost economic output (Public Health Agency of Canada, 2011). Although fatality rates have declined over the last 40 years, there are still more than 300,000 Canadians living with the debilitating and life-changing side effects of stroke (Public Health Agency of Canada, 2011). Many stroke survivors need help with even simple activities of daily living, and more than half require some sort of formal rehabilitation (Public Health Agency of Canada, 2011). This residual impairment puts an enormous burden on friends and family members and jeopardizes stroke survivors' ability to return to an independent lifestyle.

One of the most common physical side effects of stroke is hemiparesis, which refers to a unilateral weakness and motor impairment. As a result, hemiparetic stroke survivors often demonstrate weight distribution asymmetry during bilateral movements, relying on the unaffected (non-paretic) side more than on the impaired (paretic) side. However, studies involving bilateral force-matching and weight-discrimination tasks have revealed that hemiparetic individuals are unaware of the extent to which they are asymmetric; that is, they seem to have an inaccurate perception of force production, as exemplified by their tendency to overestimate weights and forces on the paretic side (Bertrand, Mercier, Shun, Bourbonnais, &

Desrosiers, 2004; Brière, Lauzière, Gravel, & Nadeau, 2010; Gandevia & McCloskey, 1977c; D.I. McCloskey, Gandevia, Potter, & Colebatch, 1983; Rode, Rossetti, & Boisson, 1996; Simon, Kelly, & Ferris, 2009). In the absence of other sensory signals, it appears that bilateral forces are perceived to be equal when *effort*, quantified as force relative to maximum strength, is equal in both sides. In other words, when each side is exerting a force that is (for example) 50% of its maximum strength, the individual would perceive them to be equal, regardless of the actual force magnitudes produced. In individuals with significant bilateral strength differences, this reliance on a "sense of effort" (SOE) (Gandevia & McCloskey, 1977c) would explain the discrepancy between actual and perceived force production. Indeed, researchers have used SOE to explain force-matching inaccuracies in hemiparetic individuals in both upper (Gandevia & McCloskey, 1977c; Mercier, Bertrand, & Bourbonnais, 2004) and lower limbs (Brière, Nadeau, Lauzière, Gravel, & Dehail, 2013; Simon et al., 2009), as well as in healthy controls when fatigued (Carson, Riek, & Shahbazpour, 2002; McCloskey, Ebeling, & Goodwin, 1974) or partially curarized (Gandevia & McCloskey, 1977c).

While there are a few studies that have examined muscular effort in dynamic (Simon et al., 2009) or functional tasks (Brière et al., 2013; Milot, Nadeau, Gravel, & Requião, 2006), most investigate SOE in the context of isometric tasks in order to isolate the motor strategy employed. In practice, however, it is rare that tasks are isometric or devoid of other sensory inputs such as visual or vestibular information, so it is unclear to what extent the results of these studies can be extrapolated to everyday functional tasks. Therefore, it is relevant to investigate how hemiparetic individuals perceive and control force asymmetry in the context of activities of daily living: whether SOE is still heavily relied upon, or if other sensory signals are invoked as well. For the work described in this thesis, weight distribution in the sit-to-stand (STS) movement was chosen

as the focus because STS is an important functional task for mobility and independence among older populations. Furthermore, it has been shown that hemiparetic individuals exhibit asymmetric weight distribution during STS, even when they perceive their weight to be distributed evenly, and that the degree of asymmetry is related to strength (Brière et al., 2010; Roy et al., 2007). These observations allude to an influence of SOE on weight distribution perception in STS.

The research question guiding this work considers whether or not SOE has the most dominant influence on the control and perception of weight distribution in STS in both stroke subjects and healthy controls. Investigating this question provides the opportunity to understand more specifically what obstacles stroke survivors face on their path to recovery, knowledge of which could ideally be used to improve the efficiency and effectiveness of rehabilitation programs. As inpatient therapy lasts, on average, only 38 days (Public Health Agency of Canada, 2011), and intense doses of task-specific therapy are required for full recovery of pre-morbid abilities (Van Peppen et al., 2004), efficient and effective therapy regimens are crucial for rehabilitating stroke survivors.

1.1 Thesis Outline

This thesis is outlined as follows:

Chapter 2 provides a more thorough motivation for the research question by reviewing literature on the influence of SOE in force perception and how this mechanism affects hemiparetic individuals. Following the review, a case is made for choosing STS as the functional task of interest for this study.

Chapter 3 describes the experimental approach taken to answer the research question, the details of the study design, and the analysis process. Older healthy individuals as well as stroke survivors were recruited to participate in the experiment. Subjects were asked to perform the STS movement while being assisted by a robotic device. The purpose of the device was to vary the self-generated load supported by the user during the STS motion and measure how users distributed this load between their legs. If people relied primarily on SOE, the proportion of the load taken through each side was expected to remain constant, regardless of how much assistance was provided by the device – that is, weight distribution was expected to be independent of supported load.

Chapter 4 presents the results of the experiment described in Chapter 3. Linear models were fit to the data for each subject group, and the relationships between supported load and weight distribution (distribution-load relationship, DLR) were evaluated as the slopes of the models. These slope values were initially compared to Equivalence Bounds to determine whether the magnitudes could be attributed to movement variability or if they represented a true covariation between load and weight distribution. However, correlation analysis indicated that an interaction model accounting for a significant relationship between slope and asymmetry was a more appropriate formulation. The results of the data analysis overall indicated that the distribution-load relationship could not be summarized as a single value, but varied among individuals.

Chapter 5 discusses the implications of the results presented in Chapter 4. That fact that some individuals erroneously perceived their weight distribution to be symmetric implied that SOE does have a significant influence on weight distribution perception and control in STS. However, the tendency for weight distribution symmetry to improve at higher load magnitudes, among all subjects, implied that afferent feedback has a noteworthy contribution as well. Moreover,

afferent information may be more valuable to people who are more asymmetric overall than to the more symmetric individuals. We suggest that this difference between symmetric and asymmetric individuals is perhaps due to the CNS retuning the relative priorities of afferent and efferent signals.

Chapter 6 concludes this thesis by summarizing the work as a whole, and reflecting on the research question in light of the experimental results detailed in the previous chapters.

Chapter 2:

Background and Literature Review

In this chapter, a review of the literature on SOE and force perception is presented. Although the idea that our perception of weight does not stem primarily from afferent feedback is more than a century old, the details of exactly how the mechanism works are still unknown. Many studies have shown that the intensity of the outgoing motor command plays an important role (possibly the most important role), but the extent to and the manner in which sensory feedback contributes to our perception is unclear. Studying the effect of this mechanism in the context of hemiparetic stroke survivors, particularly those without sensory impairment, reveals aspects of the underlying functionality that are difficult to reproduce in healthy subjects. Moreover, it provides a clearer understanding of some of the obstacles stroke survivors face during rehabilitation.

This chapter is laid out as follows: Section 2.1 provides background information on stroke and hemiparesis and their effect on bilateral asymmetry. Sections 2.2 and 2.3 focus on the mechanism behind force perception and the influences of afferent and efferent signals. Motivation for studying perception and control of weight distribution in STS specifically is

discussed in 2.4, which then leads into the formation of the guiding research question presented in Section 2.5.

2.1 Stroke and Hemiparetic Asymmetry

A stroke occurs when the blood flow to a part of the brain is interrupted, either by a clot (ischemic stroke) or a rupture (hemorrhagic stroke) in the blood vessels. As a result, that part of the brain becomes deprived of oxygen and cells start to die. As these brain cells die, the body function they control becomes impaired. The severity and type of impairment depends on the extent and location of damage in the brain.

One of the most common side effects of stroke is a motor impairment called *hemiparesis*, affecting approximately 80% of stroke survivors (National Stroke Association, 2006, 2012). Hemiparesis refers to a unilateral (*hemi-*) weakness (*-paresis*) that affects the side of the body contralateral to the brain lesion and occurs as a result of damage to the neurons that form the motor pathways between the central nervous system (CNS) and the muscles. Damage to these pathways impedes the transmission of the motor signal responsible for muscle actuation, which limits (or altogether prevents) the individual's ability to move and control the paretic side of the body. Consequently, the paretic side muscles fall into disuse and atrophy over time, leading to a strength asymmetry between the paretic and the non-paretic side. Between the impaired muscle control and strength asymmetry, individuals with hemiparesis can have difficulty balancing, walking, grasping objects, making precise movements, and coordinating movements (National Stroke Association, 2012). Regaining pre-morbid movement patterns can be achievable, but generally requires extensive physical therapy and thousands of repetitions to reestablish neural traffic along the motor pathways (Van Peppen et al., 2004).

Given this unilateral impairment in strength and control, hemiparetic individuals often demonstrate asymmetric movement dynamics when performing bilateral tasks. This includes how they distribute their weight: instead of distributing it evenly between both legs, they often load a larger portion of their body weight on the non-paretic side. This tendency is perhaps not surprising in itself – it is reasonable to think that individuals would limit the load on the paretic leg due to insufficient strength or simply to a lack of confidence in their motor skills on the affected side. While these factors may be contributing, studies have revealed that this asymmetry is actually not an entirely conscious choice, as hemiparetic individuals are typically unaware of the fact that their weight distribution is uneven. Moreover, when they are asked to distribute their weight evenly or match forces between sides, they are often unable to do so accurately. Experimenters have found this to be the case even in the absence of sensory impairment and when controlling for weakness; that is, even in conditions in which hemiparetic individuals are physically capable of matching bilateral forces, they not only fail to do so but also are unaware of their inaccuracy.

These findings lay the foundations for the SOE hypothesis, as there is no a priori reason to think that damage to the motor pathways would affect sensory feedback. The next section describes basic muscle physiology responsible for sensing movement and force, followed by a review of the literature supporting the influence of SOE on force perception.

2.2 Force Perception: Afference or Efference?

2.2.1 Muscle Sensors

The two main sensory receptors in muscles are muscle spindles and Golgi tendon organs. The muscle spindles are capsules of fibers ("intrafusal fibers") that lie within and parallel to the

muscle belly fibers ("extrafusal fibers"). The spindles function as stretch receptors: due to the parallel configuration, when the muscle is stretched, the spindles get stretched as well and transmit a signal back to the CNS through afferent pathways. Certain fibers are sensitive to muscle length and others to lengthening velocity. Type Ia afferents relay both static (length) and dynamic (lengthening velocity) information, while Type II afferents relay only static information. Since the spindles only activate when stretched, they need to be under constant tension in order to transmit a signal. Thus, in addition to afferent nerve endings, muscle spindles also have efferent nerve endings (γ motor neurons) that allow the CNS to control the tension of the spindles. This tension maintenance is particularly important when the muscle is contracting – otherwise, the shortening of the muscle would release the tension in the spindles. This system of afferent and efferent connections between the CNS and the muscle spindles is known as the fusimotor system.

The other main sensory receptors in the muscle are the Golgi tendon organs (GTOs), which are responsible for measuring muscle tension (i.e. force). The GTOs are connected in series between the muscle tendon and a small group of muscle fibers. Given this series configuration, the muscle fibers pull on the GTO when they contract. This tension causes the GTO to activate and transmit a signal to the CNS via a Type Ib afferent nerve. Larger muscle contractions involve more muscle fibers, resulting in the activation of more GTOs. When the muscle is passively stretched, most of the elongation is taken by the muscle fibers themselves, so the GTOs remain quiet.

The flow of efferent (descending) and afferent (ascending) signals involved in producing and sensing muscle contraction is summarized in Figure 2-1.

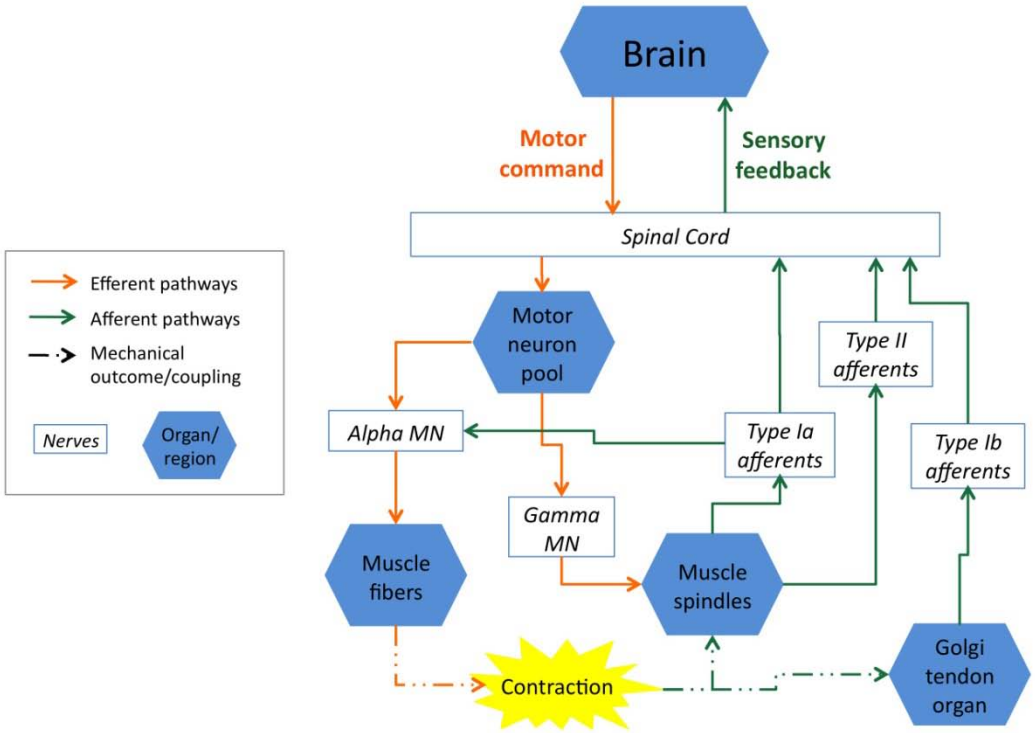

Figure 2-1: Summary of signal flow producing muscle contraction and corresponding sensory feedback. MN = motor neuron.

2.2.2 Dependence on Sense of Effort

Given the existence of these sensory receptors measuring muscle tension and elongation, one would expect that our perception of force and movement would stem primarily – if not entirely – from these afferent signals. However, we know that this is not actually the case. A simple example from everyday life is fatigue: when a person is asked to hold an object for an extended period of time, the object feels increasingly heavier as the muscle weakens. Although the actual weight of the object has not changed, the person's *perception* of the load has. Thus, it is evident that our perception of heaviness is not based solely on the afferent signals reflecting the magnitude of the muscle contraction – otherwise, the object would feel just as heavy after two

hours as it would after two minutes. Instead, the perceived heaviness of a load seems to be based at least partly on the magnitude of the outgoing (efferent) motor command. So as a muscle becomes increasingly fatigued, a larger motor signal is required to maintain the same level of contraction; consequently, the load it is supporting seems increasingly heavier, despite the fact that the load itself remains unchanged. In other words, the sensation is actually more of a reflection of *effort,* or exertion, than of load magnitude. This feeling of relative exertion has come to be known as a sense of effort (SOE) (Gandevia & McCloskey, 1977c).

For the average healthy person, this is not a poorly-designed mechanism – it is more important for our bodies to know when it is approaching the boundaries of its physical strength than to merely function as a scale and provide a consistent measurement of an external load. However, for stroke survivors with bilateral strength asymmetries, this mechanism can be problematic because some tasks, particularly those related to balance, require equal forces, which may or may not correspond to equal effort. Therefore, if hemiparetic individuals are unaware of this discrepancy and are relying on their natural sense of effort to coordinate complex functional movements, their risk of falls or injury could increase. In order to help stroke survivors avoid these risks and improve their bodily awareness, however, a better understanding of this perception mechanism is required.

2.3 Evidence for Efferent Influence: Literature Review

Evidently, our force perception is not based solely on afferent signals, and the intensity of the motor command plays a rather dominant role; this idea is at least 150 years old, dating back to the mid-19th century (see review by McCloskey, 1981). The theory behind the mechanism that is most supported by experimental results is that the efferent motor command is "read off"

somewhere along the descending pathway and is then "fed back" into higher sensory processing levels, as depicted in Figure 2-2. The terms "corollary discharge" (Sperry, 1950) and "efference copy" (von Holst, 1954) are often used to refer to this ascending branch of the motor command. This theory is supported by a number of studies in healthy individuals who have been experimentally fatigued (Carson et al., 2002; McCloskey et al., 1974), partially curarized (Gandevia & McCloskey, 1977c), or administered vibration in the antagonist muscles[1] (McCloskey et al., 1974), as all of these conditions limit the muscle's ability to produce the typical level of tension given a motor command of a certain intensity. In other words, a larger motor command is required to produce the same muscular tension as when the muscle is in its normal state. The increase in motor command is accompanied by an increased feeling of heaviness or muscular effort.

This theory is also supported by observations of individuals with lesions affecting the motor pathways, for example, due to a stroke. A lesion affects the neural signal in the same way as a curare, blocking part (or all) of it from reaching the muscle and thus limiting (or altogether prohibiting) the muscle contraction. Consequently, a larger motor command is required on the paretic side to produce the same magnitude of contraction on the non-paretic side; this is accompanied by an increased feeling of heaviness on the paretic side (Gandevia & McCloskey, 1977c; Gandevia, 1982; Mach, 1914/1906; Rode et al., 1996). Studying individuals with lesions also provides an indication of where the motor command is "read-off" along the descending pathway, as neural traffic needs to pass that point in order to produce any sensation of heaviness:

[1] When vibration is applied to a muscle, a sub-conscious tonic contraction is gradually produced (De Gail, Lance, & Neilson, 1966). If vibration is applied to the antagonist muscle while a subject is attempting to maintain a target force magnitude, a larger motor command is required in order to counteract the force of the co-contraction.

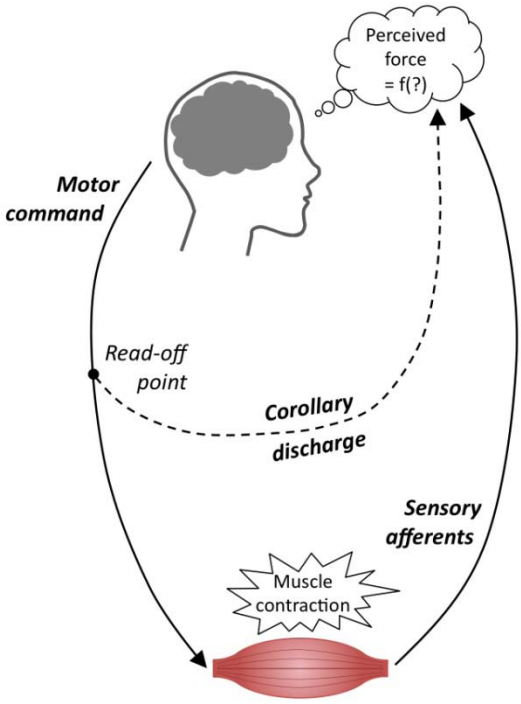

Figure 2-2: Perceived force seems to be based primarily on a corollary discharge of the motor command, which is read off at some point along the descending pathway and fed back into higher sensory areas. The exact location of the read-off point, as well as how the signal interacts with afferent information, is not known.

in individuals with upper motor neuron lesions, when their paretic side is fully paralyzed and they are entirely unable to move their limbs, they do not experience any feelings of heaviness. Once recovery progresses and they regain some ability to move the paretic limbs, the movements are accompanied by intense feelings of heaviness (Gandevia & McCloskey, 1977c; Gandevia, 1982; Mach, 1914/1906). Mach notes this specifically in his account of his own recovery from a stroke:

> "I can only describe my condition during the period of complete paralysis by saying that when I formed the intention of moving my limbs I felt no effort, but that it was absolutely impossible for me to bring my will to the point of executing the movement. On the other hand, during the phases of imperfect paralysis, and during the period of convalescence, my arm and leg seemed to me enormous burdens which I could only lift with the greatest effort." (Mach, 1914/1906, p. 175)

Presumably, in these situations, the lesions are above (rostral to) the motor command read-off point. Consequently, during the period of full paralysis, the lesion entirely blocks the neural traffic from reaching both the muscle and the read-off point, thus resulting in no contraction and no sense of muscular effort. Once the lesion begins to heal and some neural traffic is able to continue along the descending pathway, it reaches both the muscle and the read-off point, producing both a contraction as well as a feeling of muscular effort (see Figure 2-3). On the other hand, in individuals with more distal lesions, even full paralysis is accompanied by intense feelings of heaviness; this is presumably because neural traffic is able to pass the read-off point, producing a sense of muscular effort, but is blocked somewhere between that point and the muscle, thus prohibiting contraction (Gandevia, 1982; Rode et al., 1996). It has been suggested that this read-off point is perhaps in (Gandevia, 1982) or upstream of (Carson et al., 2002) the motor cortical areas.

It is apparent that the intensity of the efferent motor command plays a dominant role in sensations of muscular effort, more so than afferent sensory information. However, there are indications that afferent signals do still play a role in perception: for example, anesthetizing or stimulating various fingers has direct effects on the perceived heaviness of a load assessed through finger flexion (Gandevia & McCloskey, 1977a, 1977b). It seems that the afferent feedback serves as a checkpoint: muscular effort is mostly based on the efferent command, provided that the corresponding sensory information matches what is "expected." In the case of a mismatch, perceived heaviness is altered based on the afferent signals. Von Holst and Mittelstaedt (1950) proposed a more generalized form of this mechanism with regard to movement, positing that afference can be broken into two parts: the anticipated afferent feedback due to execution of the motor command ("reafference"), and any residual afferent feedback due

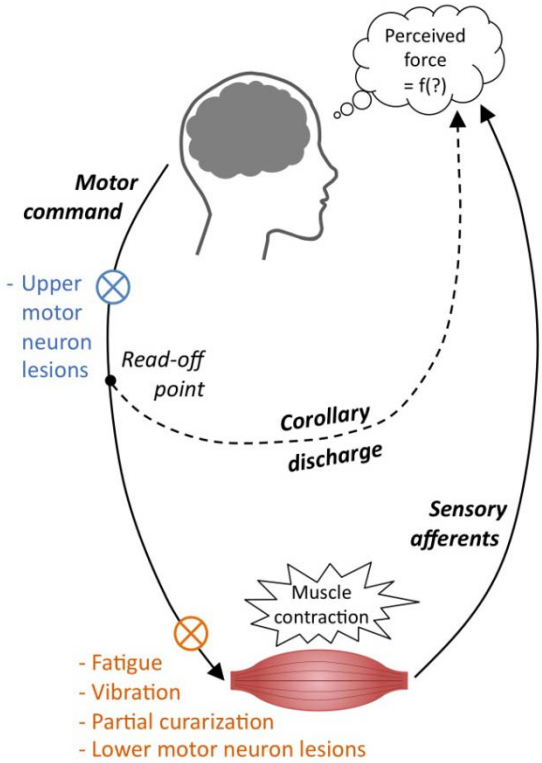

Figure 2-3: The effect of various lesion locations and stimuli on sense of effort. Upper motor neuron lesions (blue) resulting in paralysis produce no feeling of effort or heaviness, presumably because they block neural traffic above the read-off point. Lower motor neuron lesions and stimuli affecting more distal portions of the motor pathway (orange) produce altered feelings of effort relative to the actual muscle tension achieved.

to external disturbances ("exafference"). The purpose of the efference copy was to generate the reafference, which was then compared to the ascending afferent information to determine the exafference. In this way, exafference served as an error signal that was fed back into higher processing centers so that the CNS could determine appropriate compensation. Furthermore, the presence of sensory feedback may be necessary to beget a sensation of heaviness at all: deafferented individuals (e.g. due to sensory neuropathy) tend not to have any feeling of muscular effort, or if so, very little (Fleury et al., 1995; Lafargue, Paillard, Lamarre, & Sirigu, 2003; Miall, Ingram, Cole, & Gauthier, 2000). However, these observations are based on very few individuals, as conditions resulting in deafferentation without motor impairments are rare. That said, McCloskey et al. (1974) found that people *can* distinguish between a sense of

muscular effort and muscular tension: when people were asked to compare weights or, in a bilateral force-matching task, to "make both arms equal," they naturally based their judgments on muscular effort; yet, when specifically asked to keep tension constant – versus keep effort constant – they were able to do so with an accuracy of 80% or more. However, subjects did not seem to be acutely aware of *how* they adjusted their perceptual focus to distinguish tension from effort, as some subjects shifted back to sensing effort during the trials and, in those instances, were more confident in their performance. Thus, it appears that afferent feedback can and does contribute to force perception on some level, but the magnitude of that contribution and the way in which it interacts with the corollary discharge is unclear. A point of distinction is that while our perception of weight and heaviness seems to depend primarily on the efferent motor command, afferent feedback does seem to be the main contributor to our sense of movement and joint position (McCloskey, 1981).

More recent studies have focused on modeling perceived effort quantitatively: that is, they attempt to determine what parameter *is* equal between sides, if not force magnitude. If force perception were based entirely on the intensity of the motor command, we would expect that the ability to match bilateral forces accurately would be strongly related (if not directly proportional) to the degree of strength asymmetry between sides. The reasoning is as follows: when the CNS commands a muscle to contract, the information encoded in the signal is not a specific force magnitude. Instead, the signal is more of an indicator of activation level, or how much to "turn on" – for example, we would not intrinsically know how to produce 50 N of force, but we do have a sense of how to exert 50% of our maximum strength. Thus (continuing with the hypothetical situation that force perception is based entirely on the motor command intensity), in the context of a force-matching task, we would need to rely on effort, or activation level, as a

proxy for force. Consequently, a significant bilateral strength discrepancy would produce asymmetric force magnitudes that were proportional to the degree of strength asymmetry. For example, say an individual has a strong side (S) and a weak side (W) such that the maximum strength on the strong side, $F_{S,max}$, is much larger than the maximum strength on the weak side, $F_{W,max}$:

$$F_{S,\max} >> F_{W,\max} \qquad (2\text{-}1)$$

The individual is then asked to exert a force on the strong side equal to some reference value, F_s. If this reference value is 40% of the strong side's maximum strength, the required effort level, ε_s, would be of 40%:

$$\frac{F_S}{F_{S,\max}} = \varepsilon_S = 40\% \qquad (2\text{-}2)$$

The individual is then asked to produce the same force on the weak side, with no external feedback provided. If force perception were based entirely on the intensity of the motor command, in this case 40%, we would expect the individual to exert a force on the weak side, F_w, equal to 40% of its maximum strength, $F_{W,max}$:

$$\frac{F_S}{F_{S,\max}} = \varepsilon_S = 40\% = \varepsilon_W = \frac{F_W}{F_{W,\max}} \qquad (2\text{-}3)$$

However, these conditions would result in unequal force magnitudes since one side is much stronger than the other:

$$F_{S,\max} >> F_{W,\max} \quad (2\text{-}4a)$$

$$\rightarrow \quad 0.4 * F_{S,\max} >> 0.4 * F_{W,\max} \quad (2\text{-}4b)$$

$$\rightarrow \quad F_S >> F_W \quad (2\text{-}4c)$$

Of course, this expression is a simplified and idealized model of force-matching based purely on effort, which we know is not the case. Interestingly, however, a number of studies have found evidence of this relationship through experiments involving bilateral force-matching tasks (Bertrand et al., 2004; Carson et al., 2002; Gandevia & McCloskey, 1977c; Mercier et al., 2004; Simon & Ferris, 2008; Simon et al., 2009). Bertrand et al. (2004) were one of the first groups to suggest and test this model of expressing effort as force relative to maximum strength among hemiparetic individuals. In their study, stroke subjects matched bilateral grip forces at three sub-maximal levels. Linear regressions were done to evaluate whether the force ratio ([paretic side]/[non-paretic side]) at each sub-maximal level matched the ratio of maximum grip forces (note that in equation 2-3 above, a motor strategy consistent with matching effort would produce equal force ratios at the sub-maximal level and the maximum level: $F_w/F_s = F_{max,w}/F_{max,s}$). They found significant relationships between force ratios at all force levels, supporting the use of a motor strategy based on scaling the intensity of the motor command. The same group also investigated isometric force-matching in the context of a multi-joint contraction (elbow flexion/extension combined with shoulder flexion/extension) to evaluate a more complex model, which involved predicting the force magnitude on the paretic side based on scaling individual joint torques on the non-paretic side by the respective joint's bilateral maximum force ratio (Mercier et al., 2004). However, they found that the force magnitude discrepancy between sides was better predicted by the maximum force ratio measured in a multi-joint condition, indicating

that the motor strategy of force-matching across multiple joints is perhaps not a simple linear combination of the individual joints. Regardless, their results still support the involvement of a centrally-generated perception of force that is proportional to maximum strength.

Similar results have also been found for lower limbs in isometric and isotonic leg extension (Simon & Ferris, 2008; Simon et al., 2009). Among healthy individuals with leg strength asymmetry, significant differences in force magnitude were demonstrated between sides, despite subjects' perception that the forces were equal (Simon & Ferris, 2008). When forces were normalized to each side's *unilaterally*-measured maximum strength, the significant differences persisted; when forces were normalized to each side's *bilaterally*-measured maximum strength, however, there were no significant differences between sides. Similarly, among hemiparetic individuals, force magnitudes between sides were significantly different in both the isometric and isotonic conditions, but the significance disappeared when forces were normalized to maximum bilateral strength values (measured isometrically and isotonically, respectively) (Simon et al., 2009). Simon et al. (2009) also found that maximum paretic leg strength was significantly greater in the unilateral condition than in the bilateral condition, and that force distribution symmetry improved when stroke subjects were specifically instructed to exert equal forces versus when they were not given any instructions.

There are a few important takeaways from these results overall: first, that there is evidence supporting the hypothesis that force perception is influenced by SOE, which is centrally-generated and proportional to maximum strength. Secondly, that motor strategies are different among unilateral, bilateral, single-joint, and multi-joint force production – that is, while SOE does seem to have a substantial influence on the motor strategy, the maximum strength values to which the forces are relatively scaled are task-specific. Finally, stroke subjects are capable of

better bilateral force symmetry than they habitually produce, as evidenced by larger paretic side forces in unilateral conditions as well as improved symmetry when specifically instructed to match forces. These considerations will be revisited in the context of the experimental design used in this work, discussed in Chapter 3.

Although there is substantial evidence in favor of force perception being strongly – if not predominantly – influenced by SOE, in most of these studies, this influence is evaluated in the absence of many other sensory signals: the task is often isometric or otherwise constrained so that movement is limited and vestibular information is irrelevant, and no visual feedback is provided. This is a logical experimental design for attempting to isolate how the intensity of the motor command affects perception. In practice, however, it is rare that everyday movements and tasks have the same constraints and limitations, and it is reasonable to expect that all of the sensory-motor information we have available is factored into our perception. Thus, from a practical standpoint, it is relevant to investigate the influence of SOE in the context of a functional task to better understand how perceptual inaccuracy manifests itself in everyday life for hemiparetic individuals. Have they perhaps learned to prioritize other sensory signals more than SOE to compensate for its inaccuracy? Although afference is thought to have a dominant role in sensing movement (McCloskey, 1981), in a study by Milot et al. on gait (2006), hemiparetic individuals demonstrated similar levels of effort (quantified using Muscular Utilization Ratio of EMG signals) between sides in the hip flexors, extensors, and plantarflexors, implying the use of a motor strategy at least partly based on matching the intensity of the efferent signal. In this work, the investigation of SOE in bilateral force matching is extended to another functional task, sit-to-stand (STS).

2.4 The Case for Sit-to-Stand

The sit-to-stand (STS) movement was chosen as the focus of this research because it is one of the most important activities of daily living. As a crucial element of independent living and mobility, it is a particularly valuable skill among the aging population, which is the relevant demographic for stroke survivors. It is also deceptively demanding from a biomechanical standpoint: STS entails higher joint torques than walking or stair climbing (Berger, Riley, Mann, & Hodge, 1988), which can be as much as 87% of available knee torque for older adults (Alexander, Gu, & Branch, 1995). It also involves significant coordination and balance, as it requires that the horizontal momentum generated by initial trunk flexion subsequently be redirected vertically, and finally that the center of mass be stabilized over the feet (Riley, Schenkman, Mann, & Hodge, 1991). In addition, stroke survivors have been shown to distribute their weight asymmetrically during STS, loading more weight on their non-paretic side than their paretic side (Brière et al., 2010; Cheng et al., 1998; Cheng, Wu, Liaw, Wong, & Tang, 2001; Engardt & Olsson, 1992; Roy et al., 2006). They do this despite being physically capable of distributing their weight more evenly, for example by receiving feedback (Cheng et al., 2001; Engardt, Ribbe, & Olssen, 1993) or by altering foot position in the anterior-posterior direction (Roy et al., 2006). Furthermore, it seems that people's perception of their weight distribution during STS is influenced by SOE, as hemiparetic individuals are not only unaware of their asymmetric weight distribution (Brière et al., 2010), but there is also a relationship between their knee extensor strength and their weight-bearing distribution (Brière et al., 2010) as well as their knee moment asymmetry (Roy et al., 2007). As postural stability may be the most significant limiting factor in STS ability among older adults (Alexander, Schultz, & Warwick, 1991), it is important to address this weight distribution imbalance. Furthermore, improvement in weight

distribution has been linked to a decrease in falls (Cheng et al., 2001), which, in 2011, was reported to be the leading cause of injury-related deaths among individuals over 65 in the US (Centers for Disease Control and Prevention, 2013).

2.5 Research Question: Is Sense of Effort Influencing Sit-to-Stand?

In summary, stroke survivors demonstrate bilateral weight distribution asymmetry during STS, thereby decreasing their postural stability during a biomechanically complex, crucial activity of daily living. As they tend to do this unknowingly and often have the physical capability to improve their weight distribution, it is possible that this asymmetry is due not to weakness or the motor impairment itself, but to inaccurate perception. Thus, the guiding research question for this work is as follows:

Is SOE the dominant factor in the control and perception of weight distribution during STS among hemiparetic individuals?

Alternatively, is there perhaps a different reason altogether for why their weight distribution is asymmetric and their perception inaccurate?

The influence of SOE on STS weight distribution among hemiparetic individuals has been recently investigated (Brière et al., 2013). In this study, however, effort was quantified as the average Electromyographic Muscular Utilization Ratio (EMUR), which was computed as a ratio of EMG during the STS task to maximal EMG measured statically. EMUR values for the rectus femoris, vastus lateralis, and vastus medialis were then averaged as a "global" measure of effort for each side. Although similar levels of effort between sides were found for controls and for some of the stroke subjects, the more moderately-impaired hemiparetic individuals demonstrated

asymmetry in both effort level and weight distribution. Although this study certainly provides a basis for comparison, there are aspects of the study design that neglect key findings from past experiments. First, the EMUR, used as the effort metric, compares EMG during the STS movement to maximal EMG measured *statically*. This inconsistency is not ideal, since force magnitudes can be perceived differently depending on whether the force is produced isometrically or anisometrically (Mai, Schreiber, & Hermsdörfer, 1991). Furthermore, given the force-length and force-velocity relationships of muscles, the achievable tension differs between static and dynamic contexts. When using relative torque as an indication of effort in STS, Bieryla et al. (2009) showed that basing calculations on maximal isometric torque measurements can be less accurate than taking into account joint angle and angular velocity. Secondly, it seems that EMUR values were calculated based on *unilateral* isometric strength. As seen in past studies, unilateral and bilateral motor strategies differ, since normalizing a bilateral task to a bilateral maximum provided a better agreement than to a unilateral maximum (Simon & Ferris, 2008; Simon et al., 2009). It has also been shown that force distribution symmetry changes depending on whether subjects were asked to exert equal forces or whether they were given no instructions (Simon & Ferris, 2008; Simon et al., 2009). In Brière et al.'s experiment, subjects were not asked to distribute their weight evenly, nor were they asked afterwards about their perceived weight distribution. Lastly, results from Mercier et al. (2004) imply that a model based on a simple scaling of a few muscles does not provide enough of a "global" measure of effort, which presumably stems from many muscles.

Thus, despite looking to investigate the same question, we took a different experimental approach than Brière et al. did. Instead, we used a study design similar to that of Bertrand et al. (2004) and compared weight distribution across a range of sub-maximal force levels during STS.

A constant proportion of weight distribution between the paretic and non-paretic leg indicated a scaling of each side's maximum strength and a matching of the intensity of the motor command. A more detailed discussion of the experimental approach is presented in Chapter 3.

2.6 Chapter 2 Summary

Studies have shown that people seem to base their perception of force and weight primarily on the intensity of the efferent motor command as opposed to afferent sensory feedback. As hemiparetic stroke survivors have strength asymmetries and damage to their motor pathways, one would expect the functionality of this perception mechanism to cause a discrepancy between perceived force production and actual force production between the paretic and non-paretic sides. Indeed, studies have shown that hemiparetic individuals distribute bilateral loads unevenly, even when they are physically capable of matching forces between limbs, and that they are generally unaware of this asymmetry. Studies involving simple isometric and weight-matching tasks have demonstrated that hemiparetic individuals tend to distribute bilateral loads proportionately relative to the maximum strength of each side, indicating that muscular effort seems to be the dominant factor in weight perception. Less is known, however, about whether SOE is still the dominant factor in the context of functional tasks, when other sensory information such as visual and vestibular signals is available. Of particular interest is the STS movement: not only is it an important activity of daily living, but study results indicate that hemiparetic individuals distribute their weight unevenly during the movement and are unaware of doing so. Thus, the research question driving this work is whether SOE is the dominant factor in the control and perception of weight distribution during STS among hemiparetic individuals. Understanding how hemiparesis affects weight distribution and balance during STS and other

activities of daily living would help to guide rehabilitation programs towards a more efficient and effective impact on stroke survivors' recovery.

Chapter 3:

Methods and Implementation

Chapter 2 provided background and motivation for investigating the influence of SOE in STS among hemiparetic individuals. To answer this research question, stroke survivors as well as older healthy individuals participated in an experiment using a robotic STS assist device. The device supported a portion of the user's body weight, allowing them to perform the movement while bearing a smaller load than normally required for the task. The amount of assistance provided was varied across trials to see if load magnitude had an effect on weight distribution. If SOE had the most dominant influence on subjects' perceived weight distribution, the percentage of the load taken through the ND side should have remained constant across trials. If afference had a significant influence on weight distribution perception, subjects should have been able to distribute their weight evenly between sides, or if not, they should have been aware that they had failed to do so.

This chapter is laid out as follows: Section 3.1 describes participant recruitment and subject pool characteristics. Section 3.2 explains the rationale behind the experimental approach. Sections 3.3 and 3.4 detail the study design, including the control algorithm used to vary the amount of assistance provided as well as the experimental protocol. Metrics used to quantify

weight distribution characteristics and the statistical analysis used to evaluate the relationship between weight distribution and load magnitude are presented in Sections 3.5. and 3.6.

3.1 Participants

Participants were recruited through advertisements in community centers in the greater Vancouver area, as well as through local stroke recovery groups. Before participating in the experiment, volunteers were given the option to answer screening questions (Appendix A) over the phone or through an online survey in order to assess inclusion/exclusion criteria. All subjects were able to rise from a standard chair without using their hands; walk without a cane, walker, or other aid; and both communicate in and follow verbal instructions in English. Exclusion criteria included unilateral pain or conditions (aside from stroke) that might cause weight distribution asymmetry; balance disorders or other neurological conditions; inability to provide informed consent; most recent stroke within 6 months (stroke group) or history of stroke (control group); or any other conditions that might cause pain or discomfort while performing STS trials.

Data collection took place in the Collaborative Advanced Robotics and Intelligent Systems Lab on the UBC Vancouver campus. Study design and procedures were approved by the University of British Columbia Clinical Research Ethics Board under application number H13-01271. All subjects provided written consent to participate.

Data were collected on a total of 18 participants (10 stroke, 8 control). Two participants' datasets were excluded from the analysis due to their inability to control their rise motion while being assisted by the machine. Thus, the analysis comprised 16 participants in total with eight in each group. Mean demographic data are presented in Table 3-1.

SUBJECT GROUP DEMOGRAPHICS

	Age, *Years*	**Gender**, # (Male / Female)	**ND side**, # (Left / Right)	**Time post-stroke**, *Years*
Control	61.9 (16.4)	3 / 5	8 / 0	N/A
Stroke	66.5 (11.9)	6 / 2	3 / 5	5.8 (4.8)

Table 3-1: Mean demographics for each subject group. Standard deviations indicated in parentheses where applicable. ND = non-dominant side, taken as the paretic side for stroke subjects and self-reported for controls.

3.2 Experimental Approach

The most common experiment design for evaluating effort is a bilateral force-matching task, in which the subject exerts a force in one limb equal to some prescribed reference and then is asked to match the force using the other limb. This task is performed at a sub-maximal level – that is, the reference force is intentionally chosen to be much less than the individual's maximum strength to ensure that the subject is physically capable of matching forces. While some sort of feedback is often provided to ensure the *reference* limb achieves the target reference force, no external feedback is provided on how accurately the *matching* limb achieves the target force. Figure 3-1 depicts this experimental design. Effort is then quantified as force produced during the sub-maximal task relative to the individual's maximum strength.

However, as described in Chapter 2, quantifying effort is task-specific, meaning maximum strength should be measured in the same way that the task is performed. When the goal is to measure effort in an isometric context, one can easily measure the maximum strength in the appropriate configuration and quantify effort as a simple ratio. For a functional task, however, it becomes less clear how to measure an analogous maximum – for example, how "strongly" can

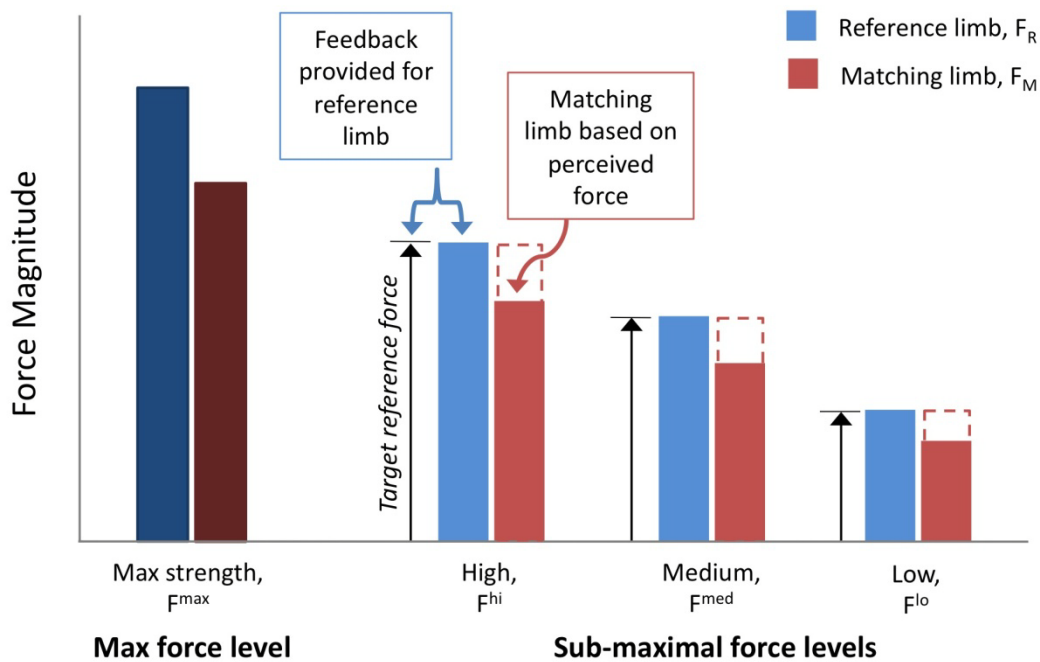

Figure 3-1: Typical bilateral force-matching experimental design. Max strength is measured for each limb (dark colors). Target sub-maximal force levels are chosen such that they are less than the weaker side's max strength. The subject is given feedback in order to produce the target reference force (black arrow) with the reference limb (blue); the subject then attempts to produce the target force in the matching limb (red), without receiving feedback. As the force levels are sub-maximal by design, the subject is physically capable of producing the reference force with the matching limb (indicated by dashed outline). However, any bilateral asymmetry displayed at the max force level is often reflected at the sub-maximal levels as well, indicating the influence of SOE.

a person sit-to-stand? Consequently, a different approach is needed to evaluate effort in STS.

Simon et al. (2008; 2009) used the typical force-matching experiment design with a leg press/extension task, and indeed found that when subjects perceived their bilateral forces to be correctly matched, it was actually force *normalized to maximum strength* that was (approximately) equal between sides:

$$\frac{F_R}{F_{R.\max}} \approx \frac{F_M}{F_{M.\max}} \qquad (3\text{-}1)$$

where R/M indicate reference and matching legs, F is the force produced during the sub-maximal task, and F_{max} is the force corresponding to each side's maximum strength. They found this relationship at multiple sub-maximal force levels among asymmetric healthy subjects (Simon & Ferris, 2008) and at a single sub-maximal force level for hemiparetic subjects (Simon et al., 2009). By cross-multiplying, this equation can be rearranged to produce the following expression:

$$\frac{F_M}{F_R} \approx \frac{F_{M,\max}}{F_{R,\max}} \qquad (3\text{-}2)$$

indicating that the ratio of forces *between* limbs during the sub-maximal task is approximately equal to the ratio of maximum strength values. It is noteworthy that, in these experiments, the target force levels for the sub-maximal task were arbitrarily chosen (although intentionally less than the paretic side's maximum strength), implying that the relationship should hold across *all* sub-maximal force levels, whether the reference force is set at 5% or 95% of the paretic side maximum. In other words, the load distribution strategy is a function of relative bilateral strength and is independent of force level. This relationship is what Bertrand et al. (2004) found for grip force matching, and serves as the key theory behind the experimental design used in this work.

However, using the bilateral force ratio shown in (3-2) as the weight distribution metric is not ideal because its value is undefined if the force is taken entirely through the matching side (i.e., $F_R = 0$). Instead, the weight distribution metric used in our STS study was defined as the percentage of the total load taken through the non-dominant side:

$$F_{ND}^{\%} \equiv \frac{F_{ND}}{\left(F_{ND} + F_D\right)} = \frac{F_{ND}}{F_{tot}} \qquad (3\text{-}3)$$

where F_{ND} is the force through the non-dominant leg, F_D the force through the dominant leg, and $F_{ND}^{\%}$ is the variable used to symbolize this ratio. If the bilateral force-matching strategy is based on effort, this metric should stay constant with respect to force level in the same way as the bilateral force ratio. Given the equal-effort relationship from Simon et al. (2008; 2009):

$$\frac{F_{ND}}{F_{ND,\max}} \approx \varepsilon \approx \frac{F_D}{F_{D,\max}} \qquad (3\text{-}4)$$

where ε represents the effort level, two expressions can be written:

$$F_{ND} \approx \varepsilon * F_{ND,\max} \qquad (3\text{-}5a)$$
$$F_D \approx \varepsilon * F_{D,\max} \qquad (3\text{-}5b)$$

Substituting into the expression for the weight distribution metric, $F_{ND}^{\%}$, yields:

$$F_{ND}^{\%} \approx \frac{F_{ND}}{F_{ND} + F_D} \approx \frac{\varepsilon * F_{ND,\max}}{\varepsilon * F_{ND,\max} + \varepsilon * F_{D,\max}} \approx \frac{F_{ND,\max}}{F_{ND,\max} + F_{D,\max}} \qquad (3\text{-}6)$$

Since the effort term, ε, gets cancelled out, the value of the weight distribution metric, $F_{ND}^{\%}$, should be independent of force level. Moreover, as the final term on the far right is a function of maximum strength values only, the fraction as a whole is constant for any individual; thus, $F_{ND}^{\%}$ should be a constant value related to the asymmetry of the individual's maximum strength.

What these mathematical manipulations imply is that if the weight distribution strategy is determined by matching effort, $F_{ND}^{\%}$ can be expected to remain constant regardless of the level of effort exerted during the task. That is, when a given individual perceives their weight distribution to be symmetric, their actual weight distribution should be such that $F_{ND}^{\%}$ (for example, 40%) of the load is taken through the non-dominant leg (and $F_D^{\%}$, or 60%, is taken

through the dominant leg), and that the value of that percentage should not be affected by the magnitude of the total load ($F_{ND} + F_D$).

Given this relationship, an experimental approach was chosen that involved varying the total load supported by the subject to assess whether the non-dominant leg maintained the same relative proportion of that load. This design is similar to the typical experimental design, except that instead of stipulating a reference force and asking subjects to match it, the total force is specified and subjects are asked to distribute it evenly between sides. Furthermore, performing the STS task at multiple sub-maximal levels and comparing among them provides an alternative to calculating effort directly (that is, with respect to a maximum strength value). Although there are forces acting in the transverse plane (both medial-lateral and anterior-posterior directions), only vertical forces were used in the analysis since they are much larger and primarily responsible for allowing the body to rise. This experimental design is compared with those of Simon et al. (2008) and Bertrand et al. (2004) in Figure 3-2.

3.3 Experimental Setup

Since STS is a functional task, performing the movement successfully requires subjects to support the load of their full body weight. In order to vary the total load supported by the subject, a STS assist device, slightly modified from that described in Jeyasurya's work (Jeyasurya, Van der Loos, Hodgson, & Croft, 2013; Jeyasurya, 2011), was used. The device allows users to perform the STS movement without having to support their full body weight. The following two sections describe the physical setup and equipment used for the experiment as well as the control algorithm and feedback loop.

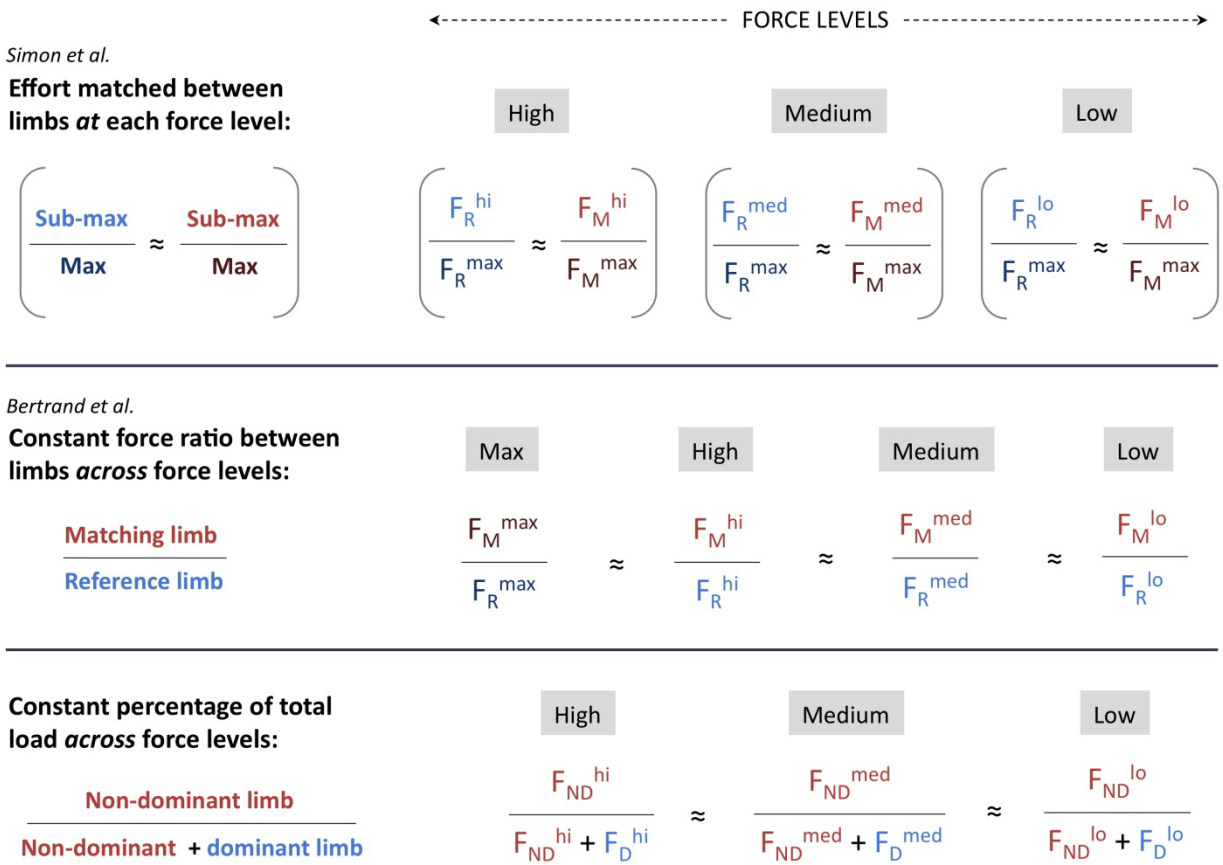

Figure 3-2: Comparison of experimental designs for evaluating role of SOE in bilateral force-matching. *Top*: Simon et al. (2008) showed forces on each side were the same percentage of max strength; *Middle*: Bertrand et al. (2004) showed bilateral force ratio is not significantly different across force levels; *Bottom*: STS study design, evaluating whether the proportion of the total load on the non-dominant side stays constant across force levels. Superscripts = force level; subscripts = limb; colors correspond to Figure 3-1.

3.3.1 STS Assist Device

The robotic STS assist used in the experimental setup is shown in Figure 3-3. The device comprised a 1-DOF seat, which was actuated by a single motor through a pulley system and rotated forward to help push the user up to standing. Physical limit switches were triggered when the seat reached the end of its movement and when it returned to the starting position. Two force plates (AMTI, Watertown, MA), one under each foot, measured the forces transmitted through

the legs. Instead of moving the whole seat mechanism to account for each user's height, adjustable foot platforms were installed on top of the force plates.

Two orientation sensors (Xsens Technology Inc.) were used to measure the thigh angle and the inclination angle of the chest, both in the sagittal plane. They were attached to the user with Velcro straps, one worn around the upper thigh of the dominant leg and the other around the upper chest, and were positioned so that the roll angle captured rotation in the sagittal plane.

For safety considerations, handrails were installed on either side of the seat. A safety harness was worn around the waist but did not provide any rise assistance. During all assisted STS trials, the user was required to hold a deadman switch, which stopped the movement of the assist if released.

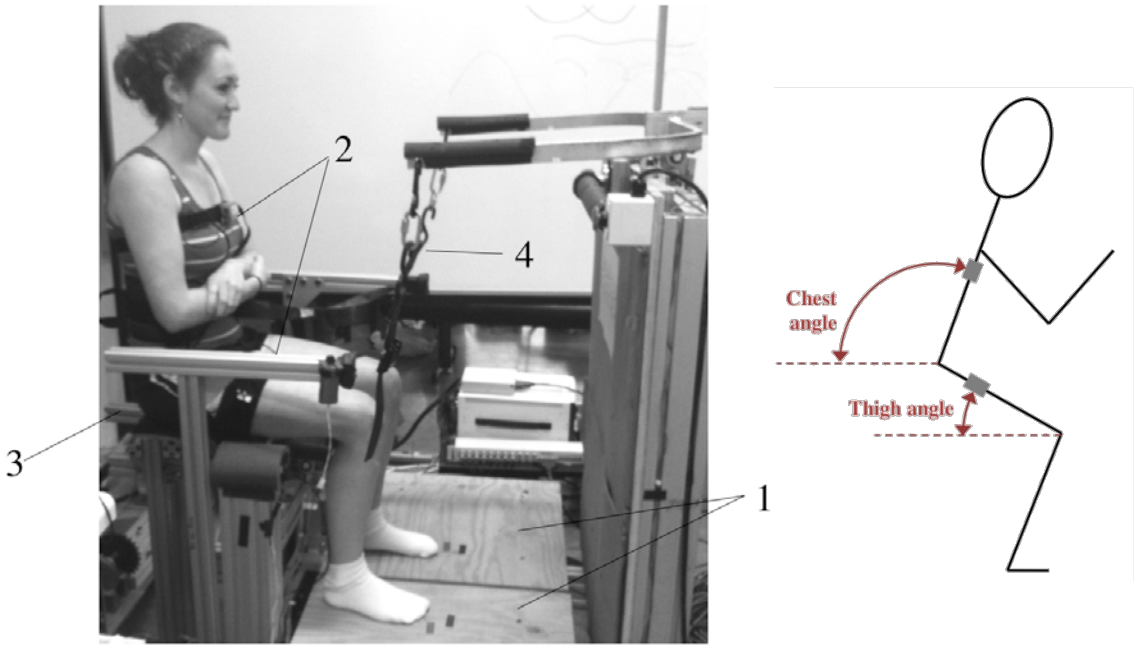

Figure 3-3: Sit-to-stand assist device. 1: Force plates, one for each foot 2: Orientation sensors, worn on thigh and upper chest, to measure inclination in the sagittal plane 3: Rotating, load-sharing seat. 4: Safety harness.

3.3.2 STS Assist Control System

A custom LabVIEW program collected and synchronized force and orientation data (50 Hz) and controlled the velocity and position of the seat. The control algorithm was designed to measure the magnitude of the load supported by the user and to manipulate it to match a target value using a human-in-the-loop feedback system: during the rise motion, the thigh angle, θ_i, was continually read from the sensor, and the total force ($F_{ND} + F_D$) was calculated from the force plates. The force value was compared to the force from a reference (unassisted) trial at the corresponding thigh angle by computing the ratio shown in (3-7). Since the reference function was made up of discrete thigh angle values, the reference thigh angle closest to θ_i was used to extract the reference force. This ratio was defined as the instantaneous load ratio, $r_{load,i}$, because it quantified the magnitude of the load supported by the user during the assisted trial relative to the total force required to rise without assistance:

$$r_{load,i} = \frac{F_{tot}(\theta_i)}{F_{ref}(\theta_i)} \qquad (3\text{-}7)$$

where θ_i is the instantaneous thigh angle. For example, a load ratio of 0.6 indicates that the user is only supporting 60% of the load normally required to rise unassisted at that thigh angle, and the device is supporting the remaining 40%. In order to drive the instantaneous load ratio towards a target value, r_{load}^*, cues were provided to the user by changing the rotation speed of the seat in real time. The speed of the motor was adjusted according to the following control law:

$$\text{Motor speed} = \begin{cases} 100\% & \text{if } r_{load}^{*} \le r_{load,i} \\ \dfrac{\left(r_{load,i} - LB\right)}{\left(r_{load}^{*} - LB\right)} & \text{if } LB \le r_{load,i} \le r_{load}^{*} \\ 0\% & \text{if } r_{load,i} \le LB \end{cases} \qquad (3\text{-}8)$$

where LB is the pre-specified lower bound value, and motor speed is expressed as a percentage of the its maximum speed. This relationship is depicted graphically in Figure 3-4. If the value of the load ratio was too low with respect to the target, the velocity of the seat declined linearly – and stopped altogether if the load ratio dropped below the lower bound – until subjects increased the force through their legs. Although the user's dynamics could not be directly controlled, the load ratio could be indirectly manipulated by cuing the subjects to adjust their interaction with the assist. The feedback loop is summarized as a block diagram in Figure 3-5.

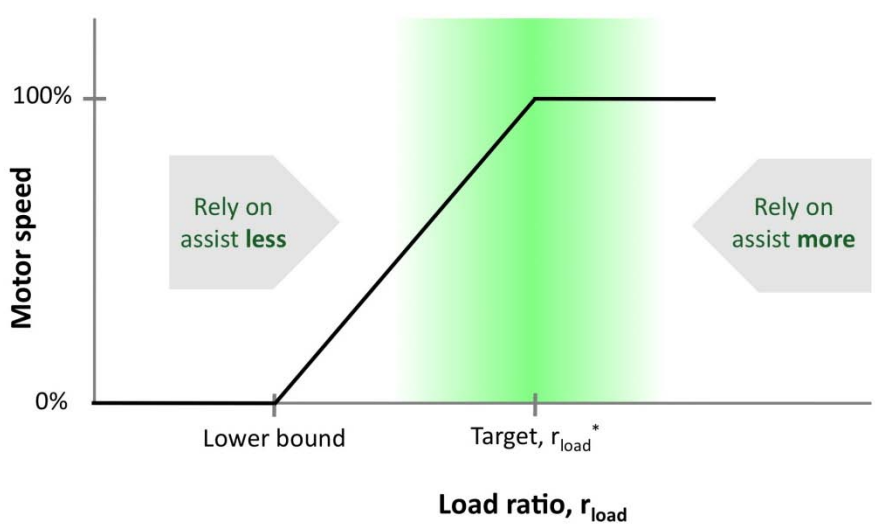

Figure 3-4: Velocity limiter. For each trial, subjects aimed to achieve a specific target load ratio, r_{load}^{*}. If their load ratio was at or above the target value, the assist operated at maximum speed. If their load ratio was below the target value, the motor speed was scaled down linearly until a specified lower bound (LB) value. Below the lower bound, the assist stopped altogether until the user increased the force through their legs. The green region indicates the ideal load ratio for the trial.

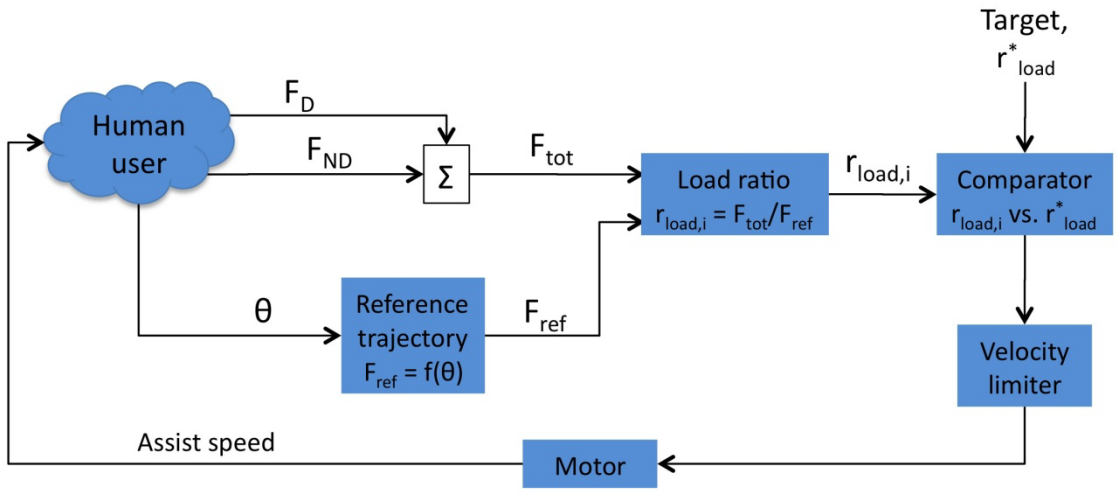

Figure 3-5: Human-in-the-loop feedback system. The goal was to drive the value of $r_{load,i}$ to r^*_{load}. Although the user could not be directly controlled, the speed of the assist provided cues for adjusting the load ratio. F_D, F_{ND} refer to the forces produced by the dominant and non-dominant legs. θ is the user's thigh angle. $F_{ref}(\theta)$ is the reference function produced from the unassisted calibration trials relating total force (dominant + non-dominant) to thigh angle.

3.4 Experiment Protocol and Design

The experiment consisted of two parts: clinical assessment and STS trials. The purpose of the clinical assessment was to evaluate subjects' overall functional ability as well as the degree of strength asymmetry between the dominant and non-dominant sides. The majority of the experiment was focused on the actual STS trials.

3.4.1 Background Information and Clinical Assessments

Basic demographic information was collected from each subject (see Appendix B). For stroke subjects, the dominant leg was taken to be the non-paretic leg. For controls, the dominant leg was taken to be the leg they would use to kick a ball (Bohannon, Waters, & Cooper, 1989).

The first clinical assessment performed was the Berg Balance Scale (K. O. Berg, Wood-Dauphinee, Williams, & Maki, 1992) (see Appendix B), which is one of the most widely-used evaluation tools for stroke survivors (Blum & Korner-Bitensky, 2008). The evaluation consists of 14 tasks intended to test balance and coordination in the context of functional movements, such as STS, transfers, turning around, balancing on one foot, etc. Each task is scored from 0-4 based on specific criteria, resulting in a maximum score of 56. A score of 20 or below indicates balance impairment or high fall risk; 21-40, acceptable balance or moderate fall risk; 41-56, good balance or low fall risk (Blum & Korner-Bitensky, 2008). Although all participants were ambulatory and community-dwelling, the Berg Balance Scale was used to provide a more specific score of general functional ability.

The second clinical assessment performed was Manual Muscle Testing (MMT), which was used as a measure of strength asymmetry between sides. Testing was done on the knee extensors (KE), ankle dorsiflexors (DF), and ankle plantarflexors (PF), since they are the most relevant muscle groups for STS. The evaluation and scoring was based on the counsel of a physiotherapist (C. Hilderman, personal contact, June 2013) as well as Kendall et al. (Kendall, McCreary, Provance, Rodgers, & Romani, 2005) and Daniels & Worthingham (1986). Each leg was tested individually: for the knee extensors, subjects were asked to extend their leg out in front of them while seated in a chair. They were instructed to keep their leg in that position while resisting the experimenter's attempt to bend their knee. Similarly, for the ankle dorsiflexors, subjects were instructed to keep their foot flexed off of the ground while resisting the experimenter's attempt to push their toes back down. Ankle plantarflexors were tested using single-leg heel raises. Subjects were given a score between 0 and 5 for each leg and for each movement, depending on how well they could resist the applied force or how may repetitions

they could perform. Specific scoring details are provided in Appendix B. Both clinical assessments were performed on all subjects.

3.4.2 Sit-to-Stand Trials

The second portion of the experiment comprised the STS trials. Subjects were fitted with the orientation sensors and positioned on the seat of the assist so that their legs were approximately parallel to the floor, and their knees were as close as possible to the rotation point of the seat, while their ankles were still on the force plates and at a comfortable level of flexion. Foot and hip positions were marked for consistency across trials. An additional safety harness was worn around the waist but did not provide any rise assistance. Three types of STS trials were then performed in four blocks:

<u>Blocks 1 & 4: Unassisted trials</u>

In Block 1, subjects first performed three unassisted trials for calibration purposes. These trials were averaged to create a reference trajectory of total vertical force as a function of thigh angle ($F_{ref} = f(\theta_i)$), which was used in subsequent assisted trials to compute load ratio values. The thigh angle corresponding to seat-off, θ_{SO}, was also defined from this trajectory as the thigh angle at peak vertical force.

In Block 4, after all other data were collected, subjects performed two additional unassisted trials. These trials were done at the end in a post-test fashion in case the weight distribution dynamics were significantly altered by the assisted trials (for example, due to fatigue). All five unassisted trials were pooled together for the analysis.

Block 2: Maximum voluntary contraction (MVC) trials

The purpose of the MVC trials was to acquire an *isometric* measurement of maximum strength in the seat-off position to serve as a comparison point. The seat of the assist was positioned on an incline such that the subject's thigh angle was approximately equal to θ_{SO}. The seat was left in this semi-elevated position for the duration of the MVC trials. The subject was then tightly strapped to the seat using a lap belt. Once signaled by the experimenter, subjects pushed their feet into the force plates as hard as they could using their knee extensors and ankle plantarflexors, as if they were trying to stand up (despite being restrained by the belt). This contraction was held for approximately three seconds and repeated two times.

Block 3: Assisted trials

At the beginning of the third block, the experimenter explained how to use the assist device and reviewed the safety features. Subjects then performed practice trials at a target load ratio value of 0.50 to familiarize themselves with the rise movement and the timing of the assist. In order to obtain data points across a wide range of load ratio values, three target load ratio values were chosen: 0.4, 0.55, and 0.7. These values were chosen based on pilot testing: load ratios below 0.4 were difficult to achieve due to the static weight of the subject's legs on the force plates, as were values above 0.7 due to the lack of momentum generated when using the assist. The target load ratios were done in blocks, the order of which was randomized across subjects. The goal was to conduct 18 trials for each subject across the aforementioned range of values. Subjects were aware of the target load ratio and were given feedback on the actual load ratio value they achieved after each trial (in addition to the real-time feedback provided via the motor speed). As weight distribution

calculations were computed in post-processing, those data were not available to the participants or the experimenter during testing. Some participants had difficulty achieving the target load ratio values in the specified order; in those cases, to avoid excessively long experiment times and fatigue, the experimenter focused on collecting data across a wide range of load ratio values regardless of the order of the target values. As some participants had an easier time achieving low load ratios while others naturally tended towards the higher values, this still produced a fairly arbitrary order of trials. While performing the assisted trials, subjects were instructed to "make it feel like their legs were working equally hard." After each trial, they were then asked how they felt their weight distribution was: left leg working a lot harder, left leg working a little harder, both legs doing equal work, right leg working a little harder, or right leg working a lot harder. These options were presented visually, as shown in Figure 3-6. It was emphasized that their *perception* was the parameter of interest; whether or not they were accurate was not important.

Subjects performed all trials without using their hands for assistance, e.g. by pushing on the handrails or swinging arms to generate momentum. Subjects were permitted to rest at any time during the data collection.

Figure 3-6: Visual representation of options to describe perceived weight distribution.

3.5 Data Analysis

As STS is a functional, dynamic movement, it is reasonable to expect that weight distribution may not be constant over the course of the rise motion. For example, a hemiparetic individual may be able to distribute their weight evenly when standing, but may not be able to do so in a nearly-seated position due to insufficient actuation of the paretic-side knee extensors. Thus, the analysis was focused on the portion of the movement around seat-off, when force peaks, joint torques are high (Jeyasurya, 2011; Mak, 2003), and assistance is likely most needed. Furthermore, since seat-off generally occurs within the first 10°-15° of thigh rotation, it also corresponds with the portion of the STS movement during which the device can provide the most assistance. For the assisted trials, the θ_{SO} used was the value calculated from the reference trajectory produced from the calibration trials. For the unassisted trials, the θ_{SO} value was the thigh angle corresponding to peak force for each individual trial.

To obtain a single value for load ratio and weight distribution for each trial, an average load ratio value, r_{load}, and an average weight distribution value, $F_{ND}^{\%}$, corresponding to seat-off were computed. These metrics were calculated as averages with respect to thigh angle since the timing of the assisted motion varied from trial to trial. Specifically, a 15° interval around θ_{SO} was used:

$$r_{load} = \frac{1}{15} \int_{\theta_{SO}-5}^{\theta_{SO}+10} r_{load,i}(\theta) d\theta \qquad (3\text{-}9)$$

$$F_{ND}^{\%} = \frac{1}{15} \int_{\theta_{SO}-5}^{\theta_{SO}+10} F_{ND,i}^{\%}(\theta) d\theta \qquad (3\text{-}10)$$

θ_{SO} - 5° was set as the lower bound because thigh angles less than that were often too close to the beginning of the motion to provide reliable data. Beyond θ_{SO} + 10°, the upper bound, it became

difficult to keep the load ratio low due to the inclination of the seat. For unassisted trials, this seat-off interval tended to be approximately 0.5 – 0.75 seconds of the rise motion; for assisted trials, this interval was closer to 1 second, although there was significant variability depending on how well the subject matched the target load ratio. These movement periods were similar to those used by Brière et al. (2010, 2013). Example plots with traces of F_{ref}, F_{tot}, r_{load}, and $F_{ND}^{\%}$ from a single trial, along with θ_{SO} and the seat-off interval, are shown in Figure 3-7.

The average weight distribution for MVC trials was calculated with respect to time (one-second intervals) since there was no change in thigh angle. Intervals were chosen by visual inspection as the period when total force was the most constant.

As the topic of interest is the internal mechanism when subjects *perceive* their weight distribution to be symmetric, assisted trials in which this was not the case (approximately 10% of trials across all subjects) were excluded from the analysis.

3.5.1 Descriptive Metrics

In order to quantify the relationship between weight distribution and load ratio, two regression metrics were calculated for each subject based on their individual STS trials:

***Distribution-load relationship (DLR Slope)*:**

The primary measure of interest, how absolute weight distribution is affected by load ratio. The DLR Slope is the slope of the regression line fit to the subject's data points ($F_{ND}^{\%}$ vs. r_{load}). A positive slope value corresponds to an increase in load percentage taken through the non-dominant side as load ratio increases.

***Deviation-load relationship (Dev Slope)*:**

How weight distribution asymmetry (that is, deviation from 50%) in general is affected by

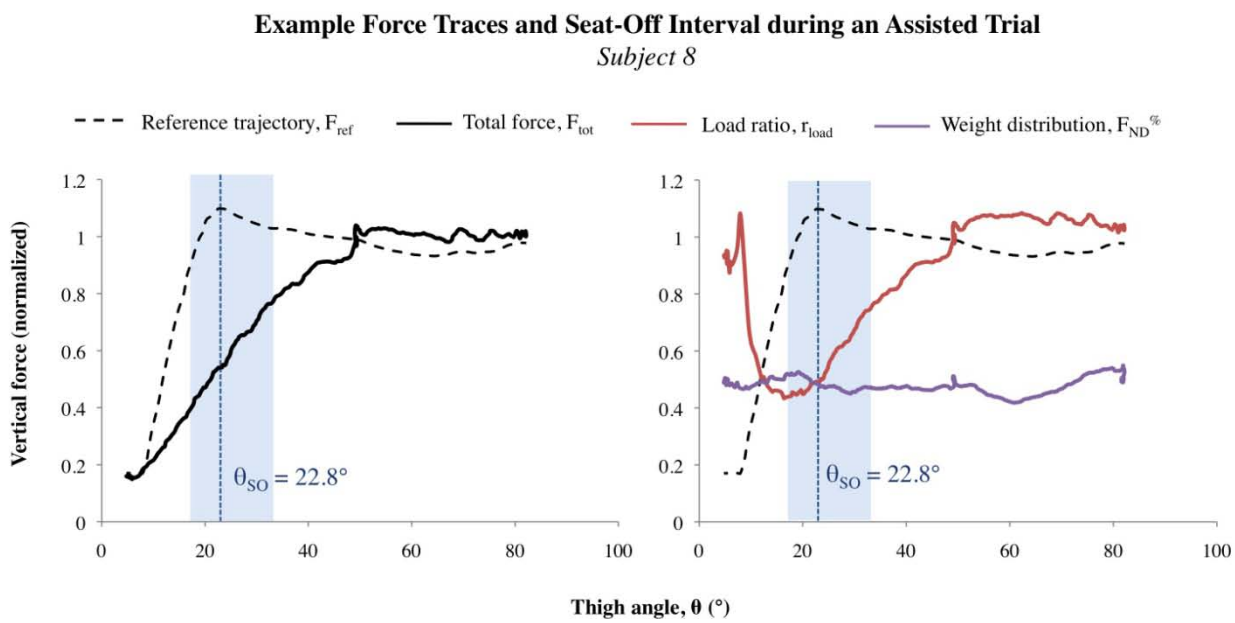

Figure 3-7: Example of force traces for single assisted trial with seat-off interval and thigh angle (θ_{SO}). The reference trajectory indicates the force-angle relationship for unassisted STS. In this example, θ_{SO} (blue line) is 22.8° and the seat-off interval (shaded region) is 17.8° - 32.8°. *Left*: Black line shows total force produced by subject, F_{tot}, measured during the trial. *Right*: Red line shows load ratio, r_{load}, calculated as the ratio of F_{tot}/F_{ref}. Purple line shows weight distribution, $F_{ND}^{\%}$, calculated as the ratio of F_{ND}/F_{tot}. Load ratio and weight distribution were averaged across the SO interval to produce single r_{load} and $F_{ND}^{\%}$ values for the trial.

load ratio. Deviation values are calculated for each trial as Dev = $|F_{ND}^{\%} - 50\%|$ and then plotted with respect to r_{load}. Dev Slope is the slope of the regression line fit to the data (Dev vs. r_{load}). A positive slope corresponds to an increase in deviation – that is, weight distribution becomes more asymmetric – as load ratio increases.

In addition to the two regression metrics, nine summary metrics were calculated for each subject:

***Mean body weight distribution (BWD_U, BWD_A, BWD_{MVC})*:**

Subject's mean weight distribution ($F_{ND}^{\%}$) averaged across all unassisted (U), assisted (A), or MVC trials. A BWD value of 0.5 or 50% corresponds to a perfectly symmetric weight

distribution; greater than 50% implies more weight is taken through the non-dominant side than the dominant side.

***Variability (VR_U, VR_A)*:**

The dispersion of the subject's data points, calculated as two times the standard deviation of $F_{ND}^{\%}$ for unassisted (U) or assisted (A) trials. Thus, VR provides a measure of inter-trial variability: assuming a normal distribution of $F_{ND}^{\%}$ values, approximately ⅔ of the data points are expected to fall within a range of VR (that is, ±VR/2).

***Overall asymmetry ($Asymm_U$, $Asymm_A$)*:**

The magnitude of the difference between the subject's average weight distribution (BWD) and 50%. Asymm thus provides a measure of how asymmetric a subject is overall, regardless of which side is favored. An Asymm value of 0 indicates perfectly symmetric (average) weight distribution.

***Average deviation (Dev_U, Dev_A)*:**

Each trial's deviation from 50% ($|F_{ND}^{\%} - 50\%|$), averaged across all unassisted (U) or assisted (A) trials. Thus, Dev provides a measure of how far from 50% $F_{ND}^{\%}$ is expected to be for a subject's average trial. The distinction between Dev and Asymm is subtle: Dev is the average deviation (from 50%), whereas Asymm is the deviation of the average. For example, if a subject's BWD is 50%, Asymm is 0%; however, Dev would only be 0% if FND% = 50% for every trial. Dev and Asymm are equal only when a subject's weight distribution is greater than 50% on all trials or less than 50% on all trials.

A summary of these metrics is presented in Table 3-2 and visually depicted with hypothetical data in Figure 3-8.

SUMMARY OF STS METRICS

Metrics	Description	Calculation
STS Regression		
DLR Slope	Regression line slope: $F_{ND}^{\%}$ vs. r_{load}	$\Delta F_{ND}^{\%} / \Delta r_{load}$
Dev Slope	Regression line slope: Dev vs. r_{load}	$\Delta Dev / \Delta r_{load}$
STS Summary		
BWD_U, BWD_{MVC}, BWD_A	Average body weight distribution	$Mean(F_{ND}^{\%}{}_i)$
VR_U, VR_A	Inter-trial variability	$2*SD(F_{ND}^{\%})$
$Asymm_U$, $Asymm_A$	Overall asymmetry	$\lvert BWD - 50\% \rvert$
Dev_U, Dev_A	Average deviation	$Mean(Dev_i)$

Table 3-2: Summary of STS regression and summary metrics with descriptions and mathematical expressions. Dev_i calculated as $|F_{ND}^{\%}{}_i - 50\%|$, where the index *i* corresponds to the trial number.

3.6 Statistical Analysis

3.6.1 Simple Linear Model

If SOE has the most dominant influence on the perception and control of weight distribution, weight distribution should be independent of load ratio – in other words, $F_{ND}^{\%}$ should remain constant across all levels of r_{load}. If this is the case, the DLR regression lines should be approximately horizontal, as shown in Figure 3-9 (top), with slope values near 0. Alternatively, if the weight distribution strategy is determined by a parameter other than effort, there should be a relationship between weight distribution and the magnitude of the total load. Figure 3-9 (bottom) shows an example of the expected relationship if the motor strategy were based on weakness: the load would be evenly distributed up until the weaker side's maximum strength, after which point the stronger side would taken on any additional load. $F_{ND}^{\%}$ would decline from 50%, resulting in

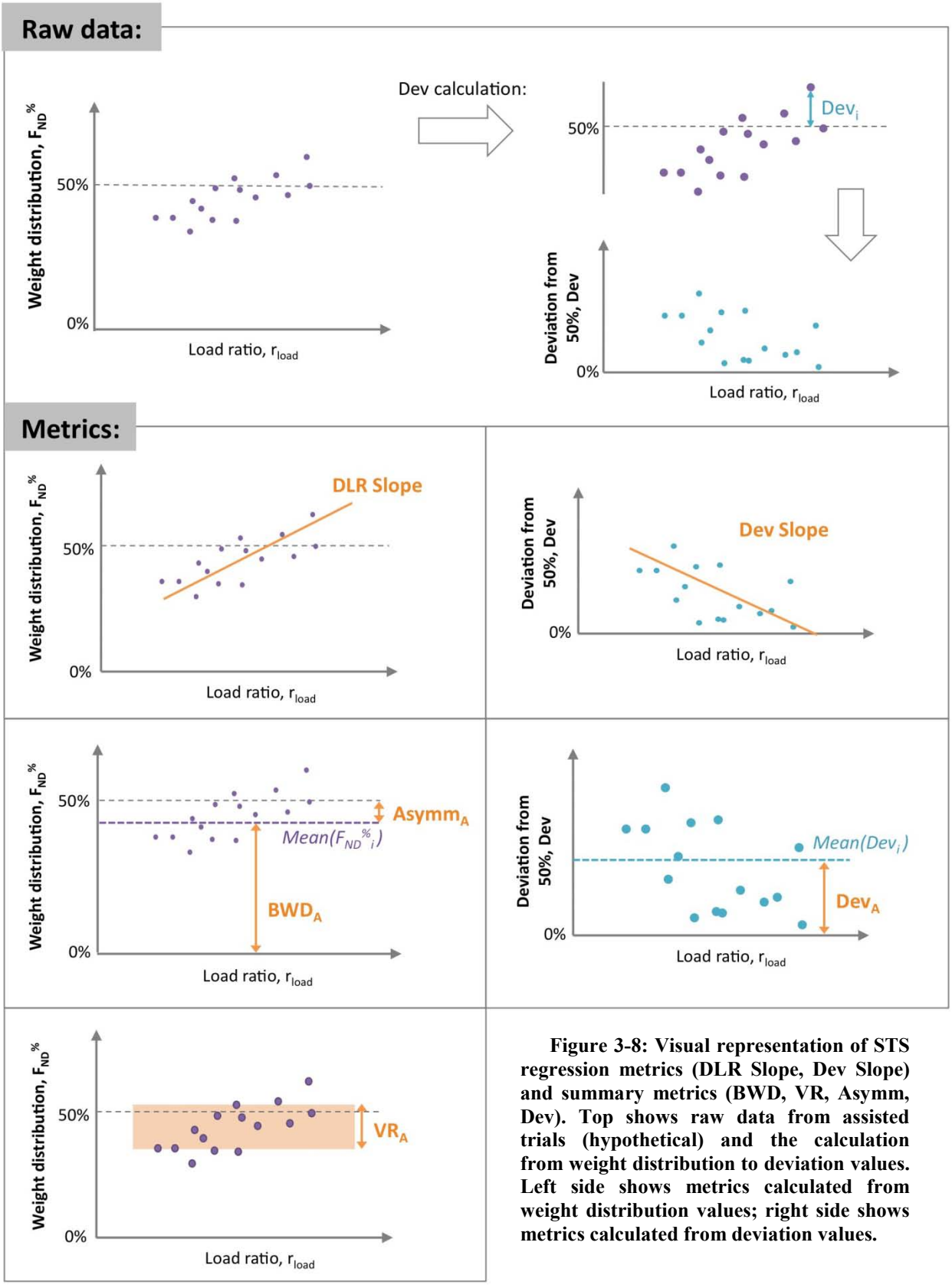

Figure 3-8: Visual representation of STS regression metrics (DLR Slope, Dev Slope) and summary metrics (BWD, VR, Asymm, Dev). Top shows raw data from assisted trials (hypothetical) and the calculation from weight distribution to deviation values. Left side shows metrics calculated from weight distribution values; right side shows metrics calculated from deviation values.

a negative DLR Slope. In this case, however, individuals should be aware of their weight distribution asymmetry, even if they are physically incapable of fixing it.

To assess the average slope for each group overall, the following linear mixed model was fit to each group's data using the lme4 function (Bates, Maechler, Bolker, & Walker, 2014) with lmerTest (Kuznetsova, Brockhoff, & Christensen, 2013) within R statistical analysis software:

$$F_{ND,ij}^{\%} = m * r_{load,ij} + b_i + \varepsilon_{ij} \qquad (3\text{-}11)$$

where *i* and *j* are the indices corresponding to subject number and trial number, respectively. In this model, the slope, *m*, is treated as a fixed effect, which means that the fundamental relationship between $F_{ND}^{\%}$ and r_{load} is assumed to be the same for all members within a group. That is, although the degree of asymmetry varies among individuals (captured in the random intercept term, b_i), the way in which weight distribution *changes* with respect to load ratio is assumed to be the same.

3.6.1.1 Slope Evaluation & Equivalence Bound Calculation

If the DLR regression lines are indeed horizontal, they should have a slope of 0; thus, after calculating the global slope estimate (*m*) for each group, two statistical tests were performed to determine how close to 0 each of those slope values was. The first was a standard t-test to evaluate whether or not the slope was statistically different from 0. However, since standard hypothesis testing can only show that a parameter is *different* from 0, an equivalence test was also performed to determine whether or not the parameters were bounded within specified limits. In this thesis, these limits are referred to as Equivalence Bounds (EB), and are defined as the magnitude of slope that could be attributed to natural movement variability. This definition was

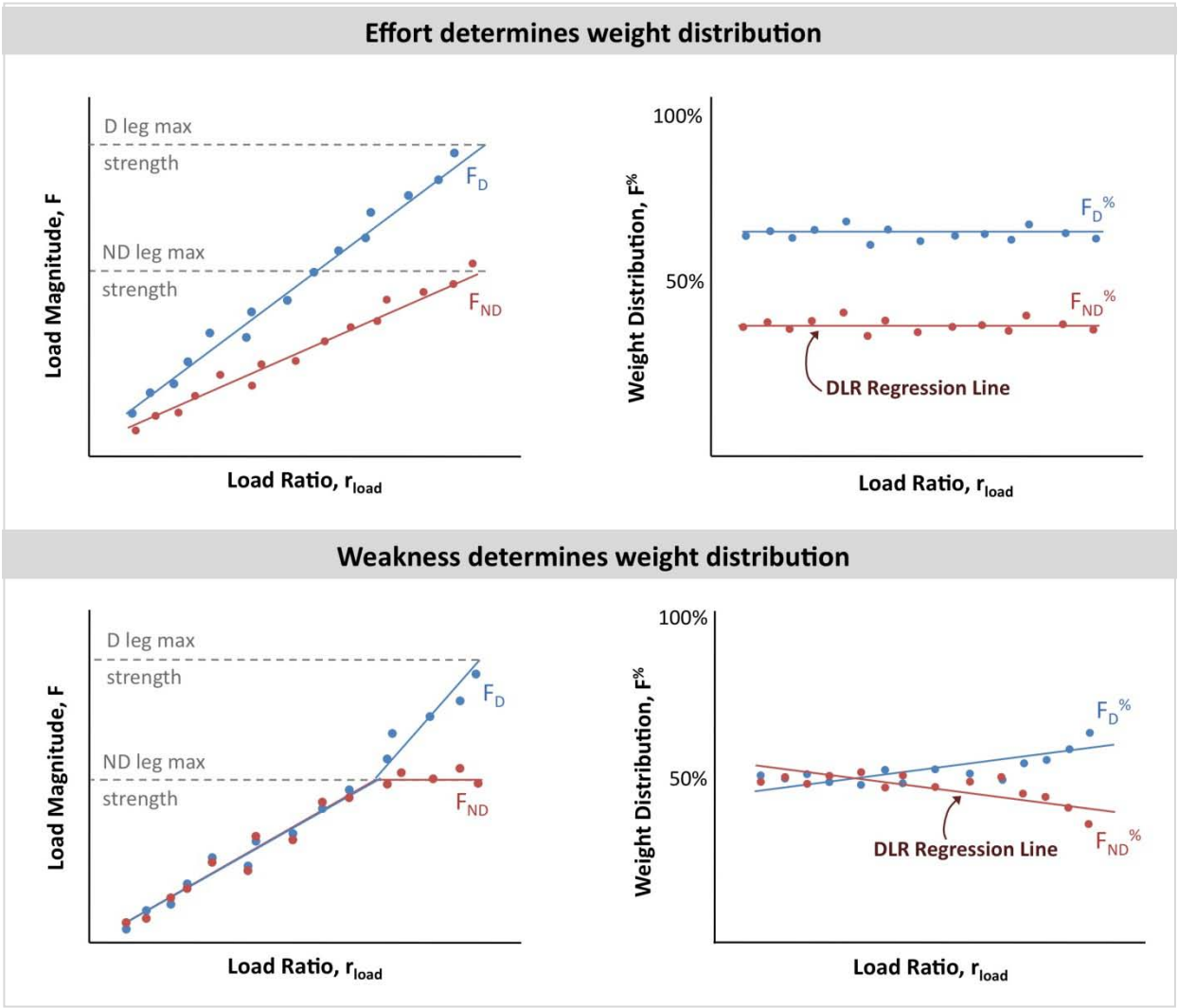

Figure 3-9: Hypothesized relationships between weight distribution and load magnitude depending on the factor that has the most dominant influence on force perception. *Top*: If perception is mostly based on a sense of effort, one would expect the load on each leg to be scaled proportionately from maximum strength values, resulting in each side supporting a constant percentage of the total load across force levels. *Bottom*: If the weight distribution strategy is limited only by weakness, one would expect weight distribution to be even up until a certain point, after which any additional increase in load is taken through the dominant side, and $F_{ND}^{\%}$ declines from 50%.

chosen because even if weight distribution *is* constant and independent of load ratio, subjects could not be expected to maintain precisely the same weight distribution throughout all of the trials. As such, this movement variability would tend to produce a small but non-0 DLR slope, which was used as the Equivalence Bounds. Two 1-sided t-tests were then performed for each

group to evaluate whether the slope was statistically bounded within the EB (H_{01}: $m < -EB$; H_{02}: $m > +EB$). Accepting the alternative hypothesis, that the slope falls within the Equivalence Bounds (H_A: $-EB \leq m \leq +EB$), requires the rejection of *both* null hypotheses. Rejecting both null hypotheses at a significance level of 0.05 corresponds to a significance level of 0.05 for H_A overall. The alternative hypothesis would also be accepted at the 0.05 significance level if the 90% confidence interval[2] were to fall entirely within the EB.

The value of the Equivalence Bounds were calculated for each group using a Monte Carlo simulation. To represent weight distribution values, a normal distribution of points with a mean of 0.5 and a standard deviation of $VR_U/2$ was created, where VR_U is the average *unassisted* variability of all the subjects in the group. *n* observations were then sampled from that distribution, where *n* was the average number of trials of the subjects in the group. To represent load ratio values, *n* points were generated, sampled from a uniform distribution with a respective maximum and minimum of $0.55 + r/2$ and $0.55 - r/2$, where *r* was the average range of r_{load} values for subjects in the group. A regression line was then fit to the simulated data points (weight distribution vs. load ratio) and the slope of the line was determined. This process was repeated 1000 times to generate a distribution of simulated DLR Slopes. The magnitude of these slopes, by design, were due strictly to random sampling and natural variability; they could not have been due to change in load ratio since weight distribution values were sampled independently of load ratio values. The ±EB values were then computed such that approximately 95% (950/1000) of the simulated slopes were contained in the interval [-EB, +EB], as depicted in Figure 3-10. Thus, if weight distribution is truly independent of load ratio, the actual regression

[2] For equivalence testing, rejecting the overall null hypothesis at a significance level of α requires rejecting both 1-sided t-tests at a significance level of α, which is the same as the 1-2*α confidence interval falling within the designated bounds.

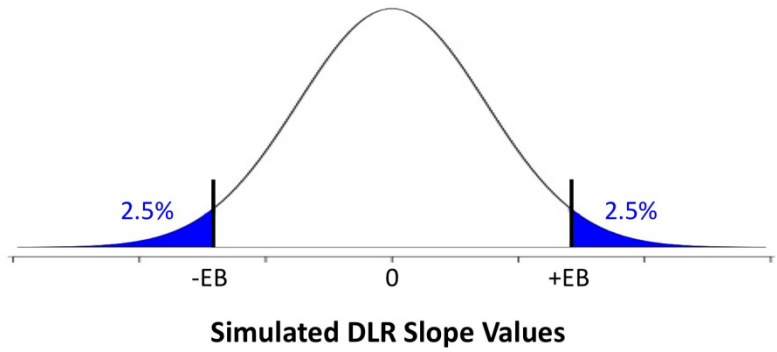

Figure 3-10: Equivalence Bound (EB) definition. Based on group averages for unassisted variability, $F_{ND}^{\%}$ and r_{load} values were simulated and sampled and a DLR Slope value was calculated. This process was repeated 1000 times, producing a distribution of simulated DLR Slope values corresponding to an independent relationship between weight distribution and load ratio. The +/- EB were defined such that 5% of the slope values fell outside of the Equivalence Bound values.

slope was expected to fall within the Equivalence Bounds (-EB ≤ *m* ≤ +EB) 95% of the time. If the regression slope fell outside of the Equivalence Bounds (*m* < -EB or *m* > +EB), it would be possible, but unlikely, that the true mean slope value was still 0. If the regression slope fell within the Equivalence Bounds – even if it was not 0 – the effect of the relationship between weight distribution and load ratio was reasoned to be small enough to be overshadowed by natural movement variability and was therefore insignificant. This latter case was considered to be evidence that weight distribution and load ratio are independent and that SOE is dominating the motor control strategy.

Thus, the following two questions were used to guide the statistical analysis for evaluating each group's slope value with respect to 0 and to the Equivalence Bounds:

Question 1: *Is the slope value statistically different from 0?*

Question 2: *Does the slope value fall within the Equivalence Bounds?*

Question 1 was addressed with a standard t-test and Question 2 with the equivalence testing. Accordingly, there were four possible outcomes based on the results of each test:

1. **Fail to reject H_0 for both Question 1 and Question 2**

   This combination would occur if the confidence interval for the group slope were very large and overlapped both 0 and one (or both) of the EB. It would be difficult to draw any strong conclusions since the true slope value may be 0, supporting the influence of SOE, or it could be larger than the EB, opposing the influence of SOE.

2. **Reject H_0 for Question 1, fail to reject for Question 2**

   This combination would occur if the confidence intervals did not contain 0 and also extended past one of the EB. This outcome would be the strongest evidence *against* the influence of SOE since it would imply a relationship between load ratio and weight distribution.

3. **Reject H_0 for both Question 1 and Question 2**

   This combination would occur if the confidence intervals were small enough to not contain 0, but also fall entirely within the EB. This outcome would be interpreted as evidence *for* the influence of SOE since the magnitude of the slope value, even though it was not 0, could still be attributed to movement variability and not the changing load ratio.

4. **Fail to reject H_0 for Question 1, reject for Question 2**

   This combination would occur if the confidence intervals contained 0 and also fell entirely within the EB. This outcome would be our strongest evidence *for* the influence of SOE because it would imply that the true slope value could in fact be 0, but even if it were not, the magnitude would be small enough to be attributable to movement variability.

Hypothetical examples of DLR Slope values and confidence intervals corresponding to each outcome are depicted in Figure 3-11.

	H_0	H_A
Question 1: Is slope value statistically different from 0?	→ Slope possibly = 0 *$p \geq 0.05$* *95% CI contains 0*	→ Slope is not 0 *$p < 0.05$* *95% CI does not contain 0*
Question 2: Does slope value fall statistically within EB?	→ Slope possibly > EB *p_{UB} or $p_{LB} \geq 0.05$* *90% CI extends past EB*	→ Slope within EB *p_{UB} and $p_{LB} < 0.05$* *90% CI within EB*

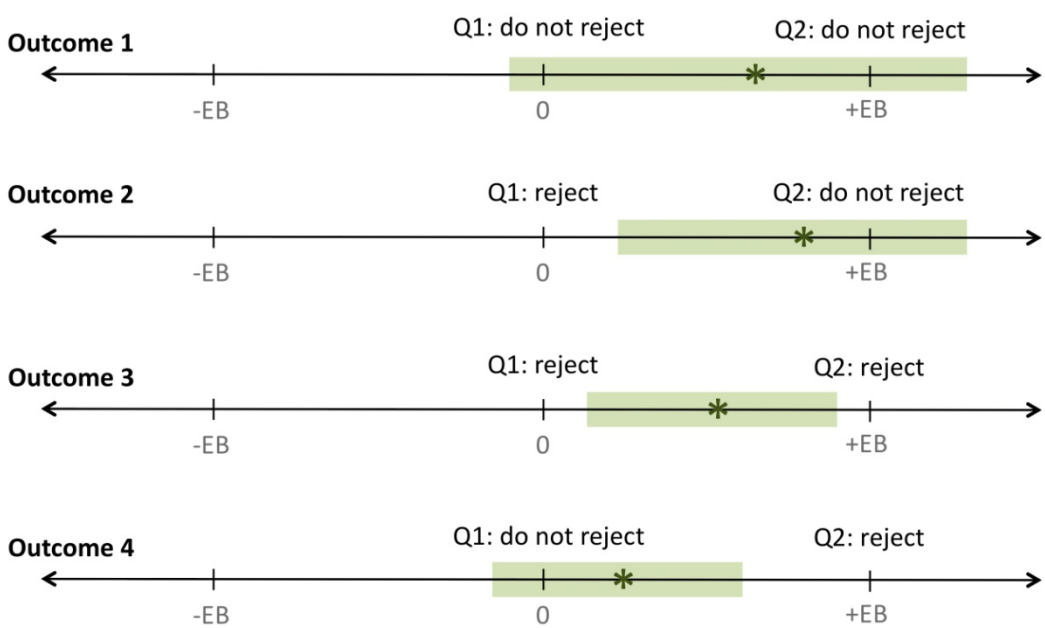

Figure 3-11: *Top*: Guiding questions for statistical analysis with possible results. p_{UB} and p_{LB} indicate *p*-values for each one-sided t-test of the equivalence test, where UB corresponds to the upper Equivalence Bound (+EB) and LB corresponds to the lower Equivalence Bound (-EB). Both questions are evaluated at a significance level of 0.05, so the 95% CI is used for Question 1 and the 90% CI is used for Question 2. *Bottom:* Hypothetical examples of DLR Slope values (stars) and confidence intervals (green) that would lead to each of the four outcomes based on statistical analysis results. Rejecting Question 1 (Q1) indicates statistical difference from 0. Rejecting Question 2 (Q2) indicates that the slope falls within the EB.

3.6.2 Correlations

In order to evaluate the relationships among the clinical and STS metrics, linear correlations were calculated between all combinations of the parameters in Table 3-3.

CORRELATION ANALYSIS METRICS

Clinical Metrics	Regression Metrics	STS Summary Metrics
Berg score	DLR Slope	BWD_U
MMT scores:	Dev Slope	BWD_{MVC}
D, ND KE	DLR Slope Magnitude	BWD_A
D, ND DF		VR_U
D, ND PF		VR_A
KE diff		$Asymm_U$
DF diff		$Asymm_A$
PF diff		Dev_u
		Dev_A

Table 3-3: Linear correlations performed between all clinical, regression and summary metrics. Diff = D – ND.

3.7 Chapter 3 Summary

This chapter detailed the experimental design, including setup, data metrics, and statistical analysis. The goal was to determine whether or not SOE had the most dominant influence on perception and control of weight distribution during STS. To answer that question, it was first shown that if people perceive even weight distribution by matching effort between sides, the percentage of the total force that is taken through the non-dominant side ($F_{ND}^{\%}$) should remain constant regardless of the magnitude of the total load. Based on that relationship, an experiment was designed using a robotic STS assist device that enabled the manipulation of the total load supported by the user by varying the amount of assistance provided (load ratio). Subjects

performed assisted STS trials such that they perceived their weight distribution to be symmetric. Figure 3-12 shows a visual representation of the dynamic between weight distribution and load ratio in this experimental approach.

If effort were the primary control parameter for weight distribution symmetry, weight distribution ($F_{ND}^{\%}$) would be expected to remain constant with respect to load ratio (r_{load}), and, consequently, the slope of the regression line fit to each subject's data would be expected to be approximately 0. A linear, fixed-slope model was used to evaluate the slopes with respect to each group's Equivalence Bounds, which were calculated based on normal movement variability. If the slope's 90% confidence interval were to fall entirely within the group's EB, it would be taken as evidence of the influence of SOE. A flowchart of the logic behind the statistical analysis is presented in Figure 3-13.

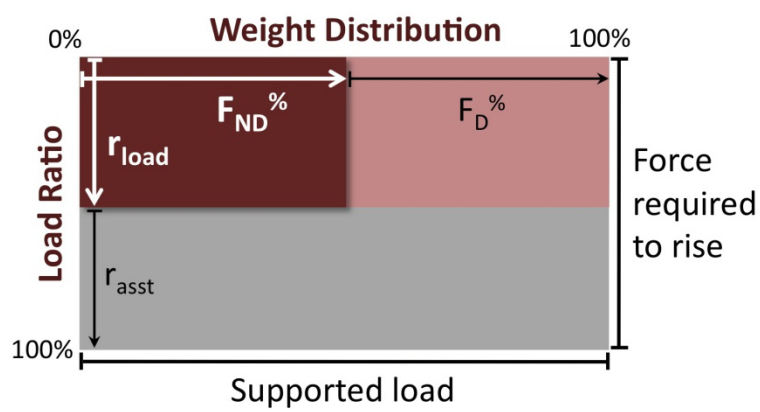

Figure 3-12: Experimental approach used for our study. The height of the block represents the total force required to rise; the percentage of that total force that is provided by the subject is the load ratio, r_{load}. The remaining portion of the required force is provided by the assist (r_{asst}). The load ratio is manipulated experimentally as the independent variable. Of the load supported by the subject, represented by the width of the block, the subject chooses how to distribute the forces between dominant (D) and non-dominant (ND) sides, attempting to do evenly. The percentage of this supported load that is taken through the ND side, $F_{ND}^{\%}$, is measured as the dependent variable.

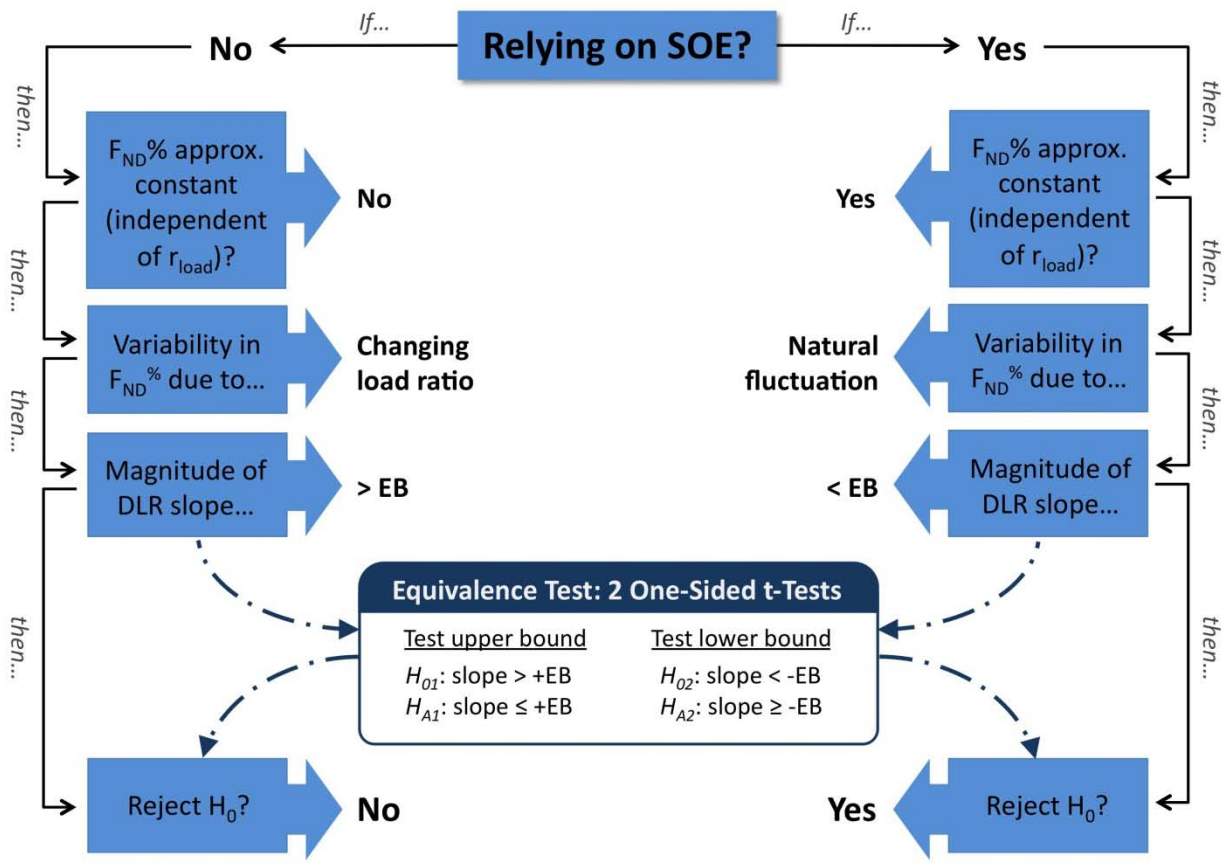

Figure 3-13: Logic flow of quantitative and statistical analysis used for determining the influence of SOE. If bilateral forces are matched primarily on effort (right side), the weight distribution metric, $F_{ND}^{\%}$, should remain approximately constant with respect to the load ratio, r_{load}, and any variability in this measure would be due to natural movement imperfections ands sampling error. Consequently, the magnitude of the DLR Slope should be close to 0 and within the Equivalence Bounds (EB), which could proven statistically by rejecting both H_{01} and H_{02}. If forces are matched based on a parameter other than effort (left side), weight distribution, $F_{ND}^{\%}$, should be affected by changing the load ratio, r_{load}. The magnitude of the DLR Slope would not fall within the Equivalence Bounds (EB), resulting in the failure to reject one or both of H_{01} and H_{02}.

Chapter 4:

Results

This chapter presents the results of the STS experiment described in Chapter 3. Data from 16 subjects, 8 stroke and 8 control, were included in the analysis. Given the assumption that the distribution-load relationship was the same for within-group individuals, a fixed-slope model was initially fit to the data, and the resulting slope estimate was compared to the group EB. However, after correlation analysis revealed a relationship between DLR Slope and BWD_A, an interaction model accounting for this correlation was fit to the data. This model was a statistically better fit than the fixed-slope model.

This chapter is laid out as follows: Section 4.1 presents descriptive statistics for the clinical assessments, STS summary metrics (BWD, VR, Dev, Asymm), and STS regression metrics (DLR Slope, Dev Slope), providing an overall characterization of STS behavior for groups and individuals. Section 4.2 describes the statistical analysis, including correlations and model fitting. The end of the chapter focuses on the trends and patterns revealed by the interaction model.

4.1 Descriptive Statistics

4.1.1 Clinical Measures

Group means of the clinical metrics are presented in Table 4-1. As a measure of individual asymmetry, MMT score differences between sides (diff = dominant – non-dominant) were also calculated for each subject and averaged within-group.

Due to the inclusion criteria, particularly the independent STS requirement, the subject pool was a very high-functioning group overall. All subjects in the control group scored 56/56 on the Berg Balance Scale except for one, who scored 54. Controls' MMT scores for all movements were no less than 4/5, and they demonstrated very little asymmetry between sides. For stroke subjects, Berg scores ranged from 46 to 56, so even the most impaired subjects fell into the good balance/low fall risk category (> 40). Knee extensor MMT scores were high, especially on the dominant side, although subjects demonstrated a greater degree of asymmetry between sides than the controls. Stroke subjects also had more trouble with the ankle movements, particularly the plantarflexion task. Strength asymmetries were most apparent in these movements. One subject (Subject 8) was barely able to dorsiflex or plantarflex the paretic ankle; another subject (Subject 7) was unable to move the paretic ankle altogether.

4.1.2 Sit-to-Stand Summary Metrics

Average summary metrics (BWD, VR, Asymm, and Dev) for each group are shown in Table 4-2. Perfectly even weight distribution for BWD was 50%; above 50% implies more weight was taken through the non-dominant (paretic, for stroke) side and below 50% implies more weight was taken through the dominant side. BWD averages for the stroke group were below 50%, although still relatively high. For the control group, BWD averages for the unassisted and MVC

CLINICAL ASSESSMENT SCORES

	Berg	KE			DF			PF		
		ND	D	Diff	ND	D	Diff	ND	D	Diff
Control	55.8 (0.7)	4.9 (0.4)	4.9 (0.4)	0.0 (0.0)	4.9 (0.4)	4.9 (0.4)	0.0 (0.0)	4.6 (0.5)	4.8 (0.5)	0.1 (0.4)
Stroke	50.9 (4.4)	4.4 (0.5)	5.0 (0.0)	0.6 (0.5)	3.4 (1.8)	4.6 (0.5)	1.3 (1.8)	2.8 (1.6)	4.0 (0.9)	1.3 (1.4)

Table 4-1: Average raw scores and standard deviations from Berg Balance Scale (max score = 56) and Manual Muscle Testing (max score = 5), as well as within-subject score differences between sides for Manual Muscle Testing. Diff calculated as D – ND. Both assessments were scored using whole numbers only. KE = knee extensors, DF = dorsiflexors, PF = plantarflexors; D = dominant side, ND = non-dominant side.

STS SUMMARY METRICS

	Weight Distribution (BWD), %			Variability (VR), %		Asymmetry (Asymm), %		Deviation (Dev), %	
	UA	A	MVC	UA	A	UA	A	UA	A
Control	50.4 (2.8)	53.2 (4.8)	50.3 (4.9)	3.1 (1.3)	5.4 (0.5)	2.3 (1.4)	4.9 (2.8)	2.6 (1.1)	5.3 (2.2)
Stroke	46.4 (5.8)	46.6 (5.7)	47.6 (4.3)	4.5 (2.3)	5.8 (1.9)	4.8 (4.7)	4.0 (5.3)	5.1 (4.5)	4.8 (4.8)

Table 4-2: Mean group STS metrics with standard deviations in parentheses. UA = unassisted, A = assisted.

conditions were very close to 50%. Unexpectedly, BWD values tended to be above 50% in the assisted condition, indicating that people loaded more weight on their reported non-dominant side. Within-subject, however, it was not necessarily the case that BWD was higher in the assisted condition than in the unassisted or MVC conditions (see Figure 4-1). As apparent from Figure 4-1, there are no obvious trends generalizing how body weight changes with condition for subjects overall.

As Asymm and Dev were calculated relative to 50%, perfect symmetry corresponds to a value of 0% for both metrics. Most subjects favored the same side across trials, so there was very little

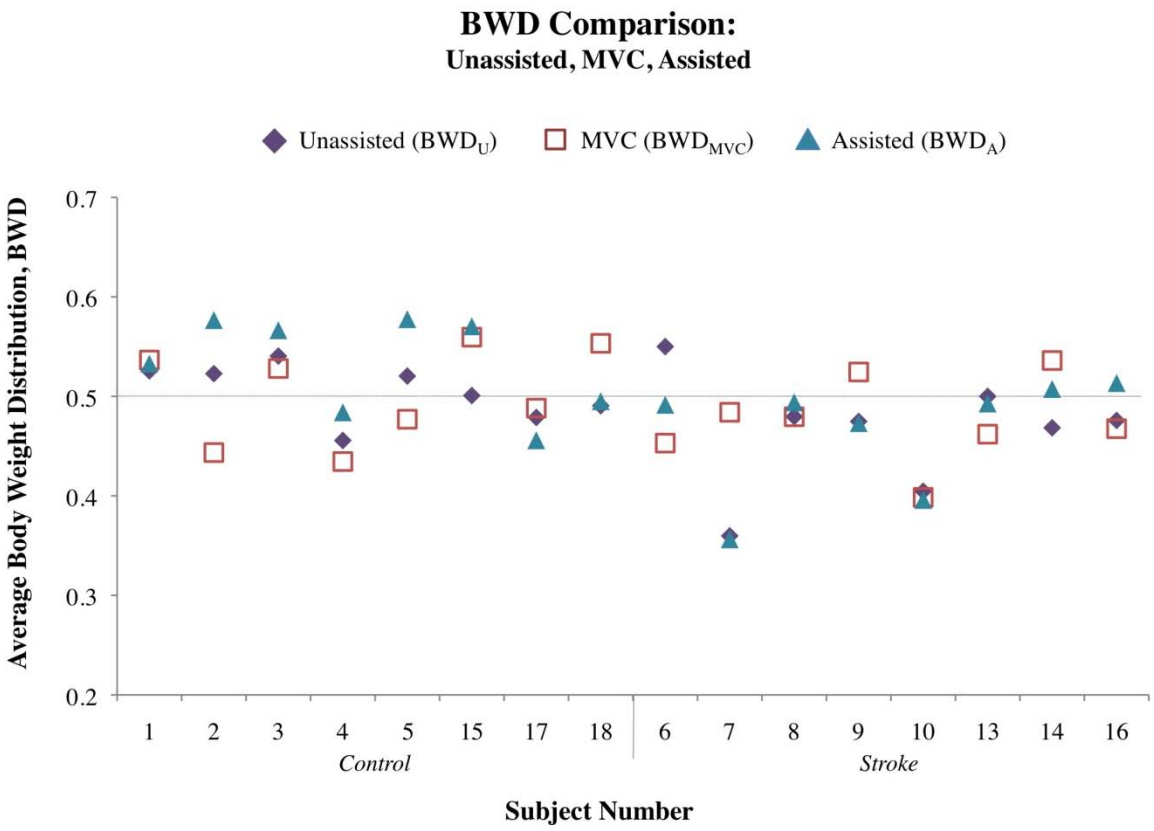

Figure 4-1: Comparison of BWD values for each subject for unassisted, MVC, and assisted trials.

difference within-subject between Asymm and Dev (the magnitude of difference was statistically less than 1%; see Appendix E).

4.1.3 Sit-to-Stand Regression Metrics

Examples of raw data from the assisted STS trials of individual subjects are presented in Figure 4-2 along with the associated summary metrics and DLR regression line. Regression lines were fit to each subject's data individually (separate from the model). Only assisted trials in which the subject reported even weight distribution were used.

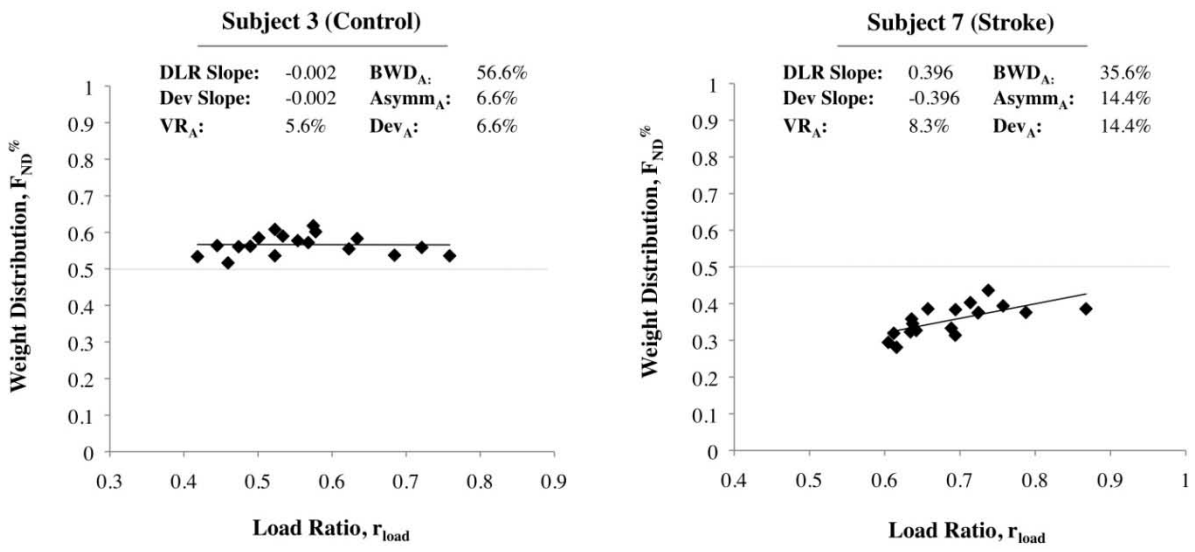

Figure 4-2: Example data from assisted trials for 1 control and 1 stroke subject. Fitted regression line indicates DLR Slope. Regression metrics and summary metrics for assisted trials listed.

DLR regression lines for all subjects are shown together in Figure 4-3, and slope values and regression fit characteristics are listed in Table 4-3. Not all subjects had load ratio values exactly between 0.4 and 0.7, but the range of values spanned at least 0.25 for all subjects (average range was 0.33). DLR Slopes for the controls ranged from -0.151 to 0.202, although all but two of the slope values fell between -0.084 and 0.061. The distribution of these slope values was relatively centered about 0. Stroke DLR Slopes were noticeably larger, ranging from -0.183 to 0.396. Unlike the control slopes, all but two were positive. In accordance with the BWD_A averages for each group, control regression lines tended to fall above 50%, and stroke regression lines tended to fall below. The two stroke regression lines at the bottom, farthest away from the rest of the group, were from Subjects 7 and 10. These were two of the most impaired subjects who had the least amount of feeling in their paretic limbs. The only control regression line with a large upward slope is from Subject 1, who reported mild scoliosis (not visibly detectable, and

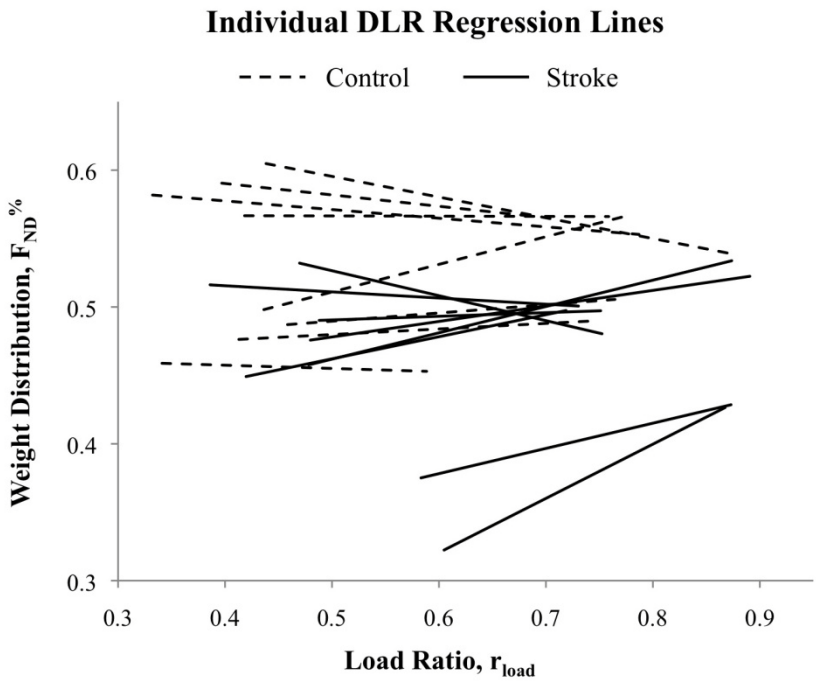

Figure 4-3: DLR regression lines fit to individual subjects' data points. Weight distribution quantified as percentage of load taken through non-dominant leg. If effort is the controlled parameter, lines are expected to be horizontal.

otherwise in good physical shape) and a natural tendency to position the left foot slightly behind the right when standing up. Since this is known to affect weight distribution (Roy et al., 2006), she was asked to maintain an even foot position during the trials.

Group averages of the regression metrics are presented in Table 4-4. DLR Slope Magnitude was also included to account for the fact that side dominance in controls may not have any bearing on direction of weight distribution. Overall, control subjects had much smaller slopes, with respect to both average value and average magnitude. Dev Slope values were negative (or very close to 0 – less than 0.005) for all subjects except Subject 1, the control group outlier, who had a Dev Slope of 0.174. Without Subject 1, the average Dev Slope for the controls would have been -0.051 (SD = 0.054). The average Dev Slope for both groups pooled together was -0.071,

INDIVIDUAL DLR SLOPE STATISTICS

Subject	DLR Slope	Slope SE	Residual SD
	Control		
1	0.202	0.031	0.014
2	-0.151	0.031	0.018
3	-0.002	0.074	0.028
4	0.041	0.068	0.029
5	-0.083	0.066	0.024
15	-0.063	0.040	0.020
17	-0.024	0.068	0.025
18	0.060	0.077	0.028
	Stroke		
6	0.193	0.035	0.019
7	0.396	0.111	0.031
8	0.027	0.067	0.024
9	0.161	0.038	0.016
10	0.184	0.121	0.038
13	0.114	0.026	0.016
14	-0.045	0.028	0.013
16	-0.183	0.071	0.024

Table 4-3: Individual subject DLR Slope values with standard errors (SE) and residual standard deviations (SD).

REGRESSION METRICS

	DLR Slope	Slope Magnitude	Dev Slope
Control	-0.003 (0.107)	0.078 (0.067)	-0.023 (0.094)
Stroke	0.106 (0.175)	0.163 (0.114)	-0.120 (0.127)

Table 4-4: Group means and standard deviations for DLR Slope, (DLR) Slope magnitude, and Dev Slope. Regression slopes were calculated from assisted STS trials only.

which was significantly less than 0 (1-tailed t-test, $t(15) = -2.39$, $p = 0.015$). The tendency for Dev Slopes to be negative implies that subjects generally got *more* symmetric as load ratio increased.

4.2 Statistical Analysis

4.2.1 Correlations

Table 4-5, Table 4-6, and Table 4-7 list the significant ($p < 0.05$) correlation values, categorized by parameter. Of particular interest is Table 4-5, which shows correlation relationships for the regression metrics. The fact that DLR Slope significantly interacted with a number of parameters suggests that the slope value was *not* constant among individuals within a group, which challenged the appropriateness of a fixed-slope model and will be discussed further in Section 4.2.2. The correlations indicate that as DLR Slope increased (became steeper and more positive), $Asymm_U$, Dev_U, DF diff, and VR_A also increased, while BWD_A, Berg score, and ND DF score decreased. This trend implies that people with more positive DLR Slopes:

- were generally more asymmetric in unassisted STS,
- had a larger strength difference in their dorsiflexors,
- had more inter-trial variability in assisted STS,
- loaded less weight on their non-dominant side,
- had lower Berg scores, and
- were weaker in their non-dominant dorsiflexors.

Taking into account only slope steepness and not direction, correlations indicate that $Asymm_U$, Dev_U, VR_A, $Asymm_A$, Dev_A, and DF diff all increased with Slope Magnitude, while Berg score and BWD_A decreased. This trend implies that people with larger Slope Magnitudes (i.e., those for whom changing load ratio had a larger effect on weight distribution):

- were generally more asymmetric in both unassisted and assisted STS,
- had more inter-trial variability in assisted STS,

REGRESSION METRIC CORRELATIONS

DLR Slope		DLR Slope Magnitude		Dev Slope	
***p* < 0.01**	***p* < 0.05**	***p* < 0.01**	***p* < 0.05**	***p* < 0.01**	***p* < 0.05**
+ $Asymm_U$	- Berg	+ $Asymm_U$	+ $Asymm_A$	- $Asymm_A$	+ Berg
- BWD_A	+ DF diff	+ Dev_U	- Berg	- $Asymm_U$	+ BWD_A
+ Dev_U	- ND DF	+ VR_A	- BWD_A	+ BWD_U	+ ND DF
	+ VR_A		+ Dev_A	- Dev_A	+ ND PF
			+ DF diff	- Dev_U	- PF diff
				- DF diff	

Table 4-5: Significant correlations with regression metrics. Groups pooled. +/- indicates positive/negative correlation, respectively.

STS SUMMARY METRIC CORRELATIONS

BWD_A		BWD_U		$Dev_U/Asymm_U$		$Dev_A/Asymm_A$
***p* < 0.01**	***p* < 0.05**	***p* < 0.01**	***p* < 0.05**	***p* < 0.05**		***p* < 0.05**
+ Berg	+ Dev Slope	+ Dev Slope	+ Berg	- Berg	+ DF diff	- Dev Slope
- DF diff	- |DLR|	- DF diff	+ ND DF	- Dev Slope	- ND DF	+ DF diff*
- DLR	+ ND DF	- PF diff	+ ND PF	+ |DLR|	- ND PF	+ |DLR|
	+ ND PF			+ DLR	+ PF diff	
	- PF diff					

Table 4-6: Significant correlations with noteworthy STS summary metrics. Groups pooled. Dev and Asymm grouped together due to insignificant within-subject difference and similar correlation patterns. +/- indicates positive/negative correlation, respectively. |DLR| = DLR Slope magnitude.
***$Asymm_A$ vs. DF diff: *p* = 0.057**

CLINICAL METRIC CORRELATIONS

Berg		DF diff	
***p* < 0.01**	***p* < 0.05**	***p* < 0.01**	***p* < 0.05**
- $Asymm_U$	+ BWD_U	+ $Asymm_U$	+ Dev_A
+ BWD_A	+ Dev Slope	- BWD_A	+ |DLR|
- Dev_U	- |DLR|	- BWD_U	+ DLR
- VR_A	- DLR	+ Dev_U	
	- VR_U	- Dev Slope	
		+ VR_U	

Table 4-7: Significant correlations with noteworthy clinical metrics. Groups pooled. +/- indicates positive/negative correlation, respectively. |DLR| = DLR Slope magnitude.

- had larger strength asymmetries in their dorsiflexors,
- had lower Berg scores, and
- loaded less weight on their non-dominant side in assisted STS.

These relationships are depicted in Figure 4-4.

Whereas DLR Slope represents the change in absolute weight distribution with respect to load ratio, Dev Slope represents the change in *symmetry* with respect to load ratio. Correlation analysis indicates that BWD_U, BWD_A, Berg score, ND DF, and ND PF all increased with Dev Slope, while $Asymm_A$, $Asymm_U$, Dev_A, Dev_U, DF diff, and PF diff all decreased. Since almost all of the Dev Slope values were negative, increasing Dev Slope refers to the slope becoming less negative, or closer to horizontal. In general, this trend indicates that the people with more negative Dev Slopes (i.e., those for whom increasing load ratio had a larger effect on improving weight distribution symmetry):

- had lower Berg scores,
- loaded less weight on their non-dominant side in both unassisted and assisted STS,
- were weaker in their non-dominant dorsiflexors and plantarflexors,
- were generally more asymmetric in both unassisted and assisted STS, and
- had larger strength asymmetries in their dorsiflexors and plantarflexors.

These relationships are depicted in Figure 4-5.

Some of these relationships, particularly with DLR Slope and DLR Slope Magnitude, seemingly conflict with each other – for example, BWD_A. BWD_A has an indirect relationship with both DLR Slope and DLR Slope Magnitude, which means that BWD_A values increase as DLR Slope values and magnitudes decrease. The discrepancy arises when slopes become increasingly negative, in which case the value decreases but magnitude increases. The reason the

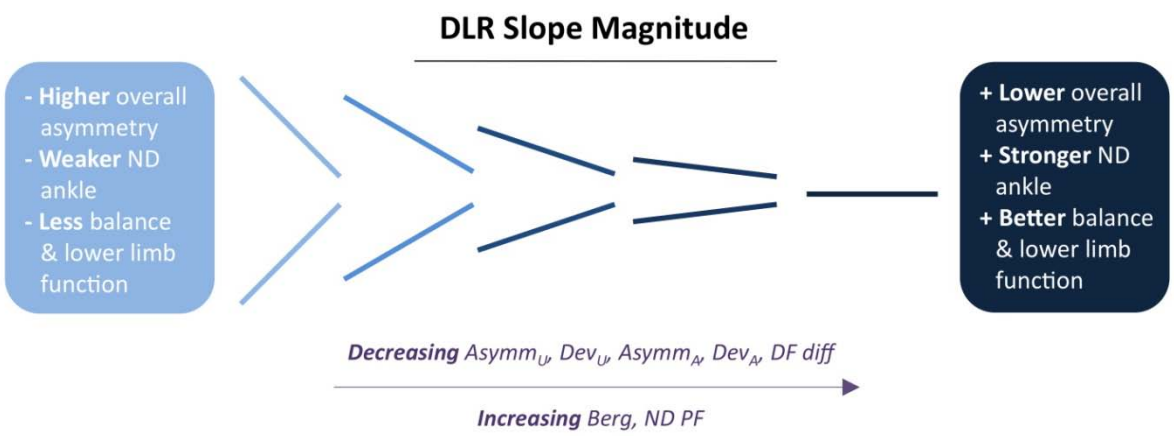

Figure 4-4: Correlations with DLR Slope magnitude. Lines represent DLR Slope magnitude, where light blue corresponds to larger (steeper) slope magnitudes and dark blue to smaller (flatter) magnitudes.

statistics display these inconsistencies is that the data itself are skewed: a few stroke subjects loaded significantly more weight on their dominant side (BWD_A values < 40%), but there were no subjects who displayed the same degree of asymmetry in the other direction (BWD_A values > 60%). This unevenness is not surprising, as we would not expect healthy subjects to be as asymmetric as stroke subjects, and we certainly would not expect stroke subjects to support the majority of the load on their paretic side. The other reason is that there are two types of metrics: bi-directional and uni-directional. Bi-directional metrics are those for which the range of values extends past the point of symmetry in both directions (e.g., BWD values can be above or below 50%, DLR Slopes can be positive or negative, etc.). Uni-directional metrics are those for which the range of values extends past the point of symmetry in only one direction (e.g., Asymm and Dev can only be above 0%, DLR Slope Magnitude can only be positive, etc.). For consistency, relationships between metrics of the same type will be the main focus of the analysis. Thus, in the case of BWD_A, more emphasis will be placed on its correlation with DLR Slope, since they are both bi-directional, than with DLR Slope Magnitude.

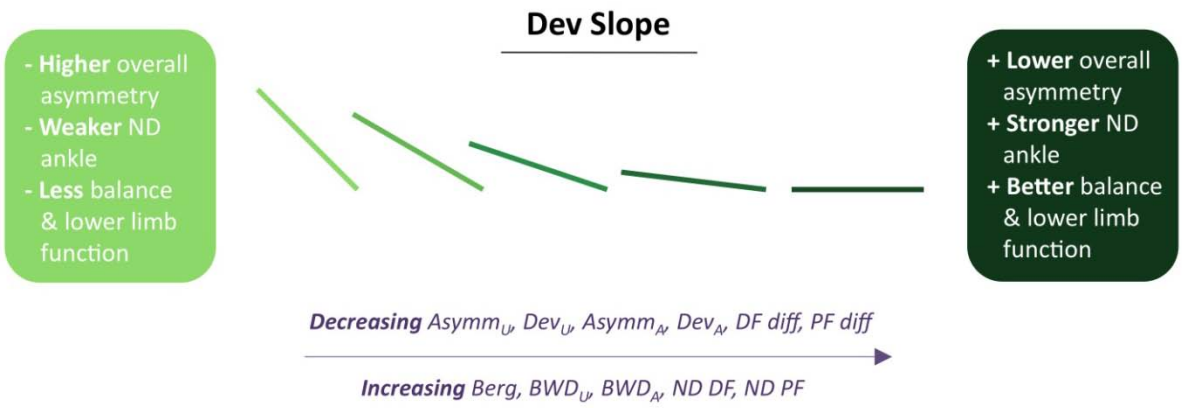

Figure 4-5: Correlations with Dev Slope. Lines represent Dev Slope values, where light green corresponds to more negative (steeper) slopes and dark green to more positive (flatter) slopes.

As expected, this choice is also reflected in the strength of the correlation (BWD_A vs. DLR Slope: $\rho = -0.72$; BWD_A vs. DLR Slope Magnitude: $\rho = -0.55$).

4.2.2 Linear Models

Parameters from fitting the fixed-slope model, $F_{ND}{}^{\%}{}_{ij} = m^* r_{load,ij} + b_i + \varepsilon_{ij}$, are presented in Table 4-8. This model was intended to quantify the distribution-load relationship for each group as a whole, assuming a fixed slope, while treating the intercept term as the random parameter to account for different degrees of overall asymmetry among individuals. The *p*-value associated with the slope corresponds to the basic t-test (H_0: slope = 0), which answers Question 1 of the analysis questions. If SOE were strongly influencing weight distribution perception and control, the fitted slope values were expected to be close to 0.

For the control group, since *p* is much larger than 0.05, the fitted slope value of -0.015 is not significantly different from 0, and H_0 cannot be rejected. In the case of the stroke group, the fitted slope value is 0.098 and *p* is much smaller than 0.05, so we do reject the null hypothesis

FITTED PARAMETERS: FIXED-SLOPE MODEL

	Slope, m		Intercepts, b_i	
	Avg. (SE)	p-value	Avg. (SE)	SD
Control	-0.015 (0.021)	0.464	0.54 (0.021)	0.048
Stroke	0.098 (0.022)	< 0.001	0.40 (0.026)	0.061

Table 4-8: Parameters resulting from fitting the fixed-slope model ($F_{ND}{}^{\%}{}_{ij} = m^* r_{load,ij} + b_i + \varepsilon_{ij}$) to each group's data. Mean slope and intercept values presented with standard errors in parentheses. The slope p-value corresponds to standard hypothesis testing where H_0: $m = 0$.

and accept that the slope is significantly different from 0. Figure 4-6 shows the model regression line for each group ($F_{ND}{}^{\%} = m^* r_{load} + b_0$) overlaid on the individual subject regression lines for comparison.

Simply because no significant difference was found between the model slope for the controls and 0 does not prove that it *is* 0, or even that it is close to 0. On the other hand, simply because there was a significant difference between the model slope for the stroke group and 0 does not necessarily mean that the magnitude of the relationship between weight distribution and load ratio is large enough to be meaningful. Hence, equivalence testing was used to compare the magnitude of the DLR Slopes to normal movement variability.

Table 4-9 shows the 90% confidence intervals for the fitted slope values and Equivalence Bound values calculated for each group. The p-values refer to the one-sided t-tests comparing the slope to each Equivalence Bound (H_0: m < lower bound or m > upper bound). To accept the alternative hypothesis and show that the slope value is bounded by the EB at a significance level of 0.05, H_0 must be rejected at both the upper and the lower bound; equivalently, the 90% confidence interval must fall entirely within the EB range. For the control group, both p-values

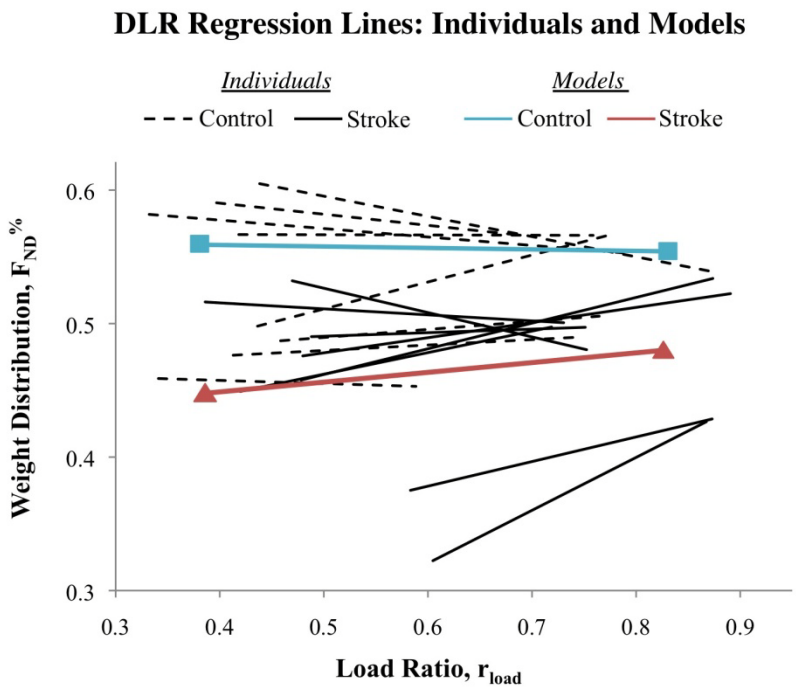

Figure 4-6: Group DLR regression lines from fitted model along with individual subject DLR regression lines. Control DLR = -0.015, stroke DLR = 0.098

were less than 0.05 and the confidence interval was well within the EB; thus, we accept the alternative hypothesis that the fitted slope value of -0.015 is within the Equivalence Bounds.

For the stroke group, the *p*-value on the lower bound is significant, but the *p*-value on the upper bound is slightly larger than 0.05. This deviation past 0.05 is reflected in the fact that the confidence interval falls well within the lower EB, but extends just past the upper EB (0.134 vs. 0.131).

Figure 4-7 graphically shows the model slopes for each group relative to their respective EB. While the slope line for the control group is fairly horizontal and in the middle of the EB, the slope line for the stroke group is larger in magnitude and very close to the upper EB.

EQUIVALENCE TEST RESULTS

	Slope	90% CI	Group EB	*p*-value, LB	*p*-value, UB
Control	-0.015	(-0.050, 0.019)	±0.075	0.0028	< 0.0001
Stroke	0.098	(0.062, 0.134)	±0.131	< 0.0001	0.068

Table 4-9: Equivalence test results of the slope parameter (*m*) from the simple linear model. EB = upper (+) and lower (-) group Equivalence Bounds, as calculated from the Monte Carlo simulation. *P*-values for the upper (UB) and lower bounds (LB) refer to 1-sided t-tests comparing the slope to +EB and –EB, respectively. If the p-values at both the upper and lower bound are significant, we accept the alternative hypothesis (H_A: -EB < *m* < +EB) for that group.

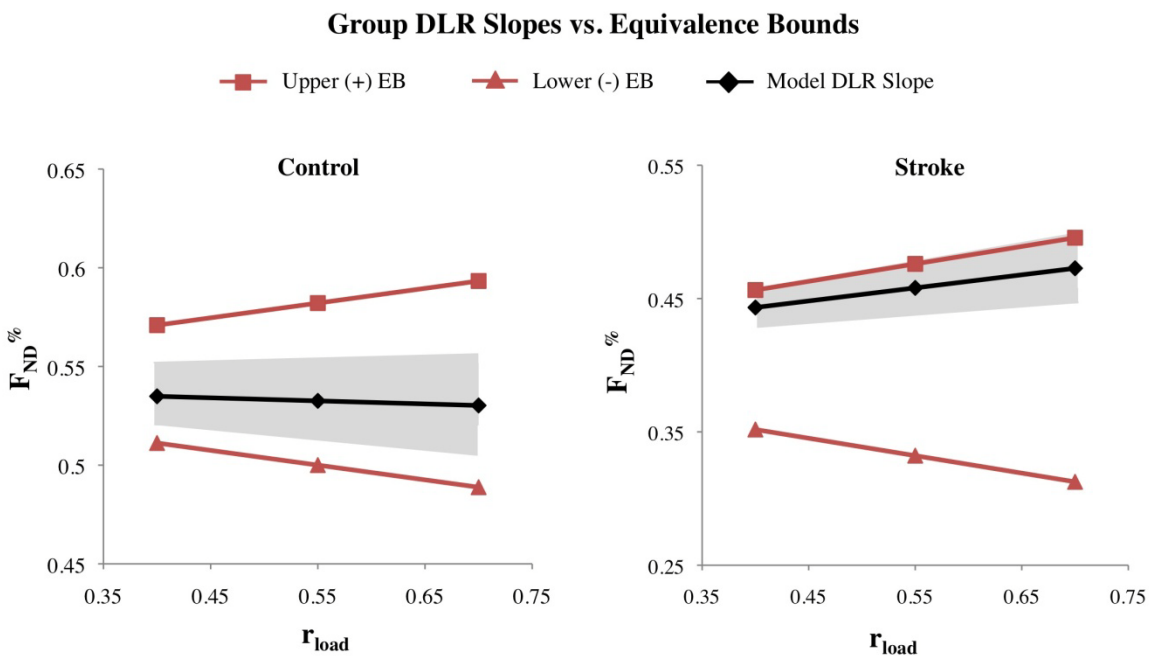

Figure 4-7: Group DLR Slopes from fitted models relative to upper (+) and lower (-) Equivalence Bounds (EB). The slope values are represented by the slopes of the black lines and the EB values by the slopes of the red lines. Shaded areas indicate 90% confidence intervals for DLR Slopes. For the controls, the CI falls entirely within the EB; for the stroke group, the CI extends slightly past the upper EB.

4.3 Data Summary and Reevaluation

The purpose of this slope analysis was to determine how close the model slope values were to 0. If effort were being used as the primary parameter for perceiving and controlling weight distribution, $F_{ND}^{\%}$ and r_{load} should be independent, and the true value of the regression line slope should be 0. Two statistical analysis questions were used to evaluate how close the slope values are to 0: 1) Are the slope values significantly different from 0? and 2) Are the slope magnitudes big enough that they could not be attributed to movement variability and/or sampling error?

Table 4-10 shows a summary the results of the statistical analysis for the control group. The model slope value was -0.015, which ostensibly was quite close to 0. For Question 1, we do *not* reject the null hypothesis, so the slope is not significantly different from 0. For Question 2, we *do* reject the null hypothesis, which indicates that the magnitude of the slope, even if it is not exactly 0, is still small enough to be attributed to factors other than an interdependence between weight distribution and load ratio. Thus, these results indicate a potential reliance on SOE.

Table 4-11 shows a summary the results of the statistical analysis for the stroke group. The value of the model slope was 0.098, which was not only considerably larger than the value for the control group, but also significantly different from 0. Thus, we reject the null hypothesis for Question 1. The interpretation of results for Question 2 is not as straightforward: strictly based on *p*-values, we *cannot* reject the null hypothesis because the upper bound *p*-value is not less than 0.05 ($p = 0.068$), and the 90% confidence interval does not fall within the EB. However, the *p*-value is only slightly larger than 0.05, which should be taken into account when interpreting these results: as the EB are calculated via simulation, re-running the simulation produces slightly different EB values and, consequently, slightly different *p*-values. Indeed, doing so produces *p*-

Statistical Results for Control Group *(m = -0.015)*

	H_0	H_A
Question 1: Is slope value statistically different from 0?	→ Slope possibly = 0 *$p \geq 0.05$* *95% CI contains 0*	→ Slope is not 0 *$p < 0.05$* *95% CI does not contain 0*
Question 2: Does slope value fall statistically within EB?	→ Slope possibly > EB *p_{UB} or $p_{LB} \geq 0.05$* *90% CI extends past EB*	→ Slope within EB *p_{UB} and $p_{LB} < 0.05$* *90% CI within EB*

Table 4-10: Results of statistical analysis questions for the control group.

Statistical Results for Stroke Group *(m = 0.098)*

	H_0	H_A
Question 1: Is slope value statistically different from 0?	→ Slope possibly = 0 *$p \geq 0.05$* *95% CI contains 0*	→ Slope is not 0 *$p < 0.05$* *95% CI does not contain 0*
Question 2: Does slope value fall statistically within EB?	→ Slope possibly > EB *p_{UB} or $p_{LB} \geq 0.05$* *90% CI extends past EB*	→ Slope within EB *p_{UB} and $p_{LB} < 0.05$* *90% CI within EB*

Table 4-11: Results of statistical analysis questions for the stroke group. For Question 1, $p < 0.05$, so we reject H_0. For Question 2, $p < 0.07$, so we cannot officially reject H_0; however, re-running the simulation did occasionally produce significant *p*-values, so we acknowledge the outcome associated with the alternative hypothesis as well.

values approximately in the range of 0.045 – 0.075, which means the results could have been statistically significant. Even given that they are not significant, the confidence interval extended past the upper bound by only 0.003, approximately 1% of the entire EB interval. So while we cannot reject the null hypothesis due to the stipulated significance criteria, we should acknowledge that there is evidence (albeit weak) supporting the alternative hypothesis. Thus, SOE seems to be influential, but there are perhaps other influences factoring in as well.

4.3.1 Interaction Model

Given the lack of conclusive findings from the group data model fitting, further analysis was done on individual data using the same two analysis questions as were used for the group analysis. Individual Equivalence Bounds were calculated (via simulation) using each subject's BWD_A and VR_U for the $F_{ND}^{\%}$ distribution, their range of r_{load} values for the r_{load} distribution, and their number of trials for the sample size. Thus, the Equivalence Bounds represented potential slope magnitudes based on the subject's own movement variability, given the range of load ratios they achieved and number of trials they performed. Standard difference testing (Question 1) was done in the same way as for the groups.

Figure 4-8 shows individual subject DLR slopes, with 90% confidence intervals, relative to their respective EB intervals. What becomes strikingly apparent from this plot is that while 9/16 subjects had slopes that were not significantly different from 0, only one person (Subject 8) had a slope that fell (significantly) within her EB. 10/16 confidence intervals overlapped some portion of their EB interval; of those, there were 7 in which the slope value itself fell within the EB. For 5 subjects, the DLR confidence interval fell entirely outside of the EB. Given that the group results suggested that the DLR Slopes were bounded within the EB, it is surprising that on an individual level, this was the case for only one subject.

Table 4-12 lists DLR Slope values and confidence intervals relative to the EB for each subject. There were no obvious patterns that could explain how or why the relationship between DLR Slope and EB intervals varied to such an extent among individuals.

As mentioned previously, an interaction was found between Average Assisted Weight Distribution and DLR Slope value (BWD_A vs. DLR Slope, $\rho = -0.72$, $p = 0.0016$). Those who loaded more weight on their non-dominant side tended to have steeper negative slopes, while

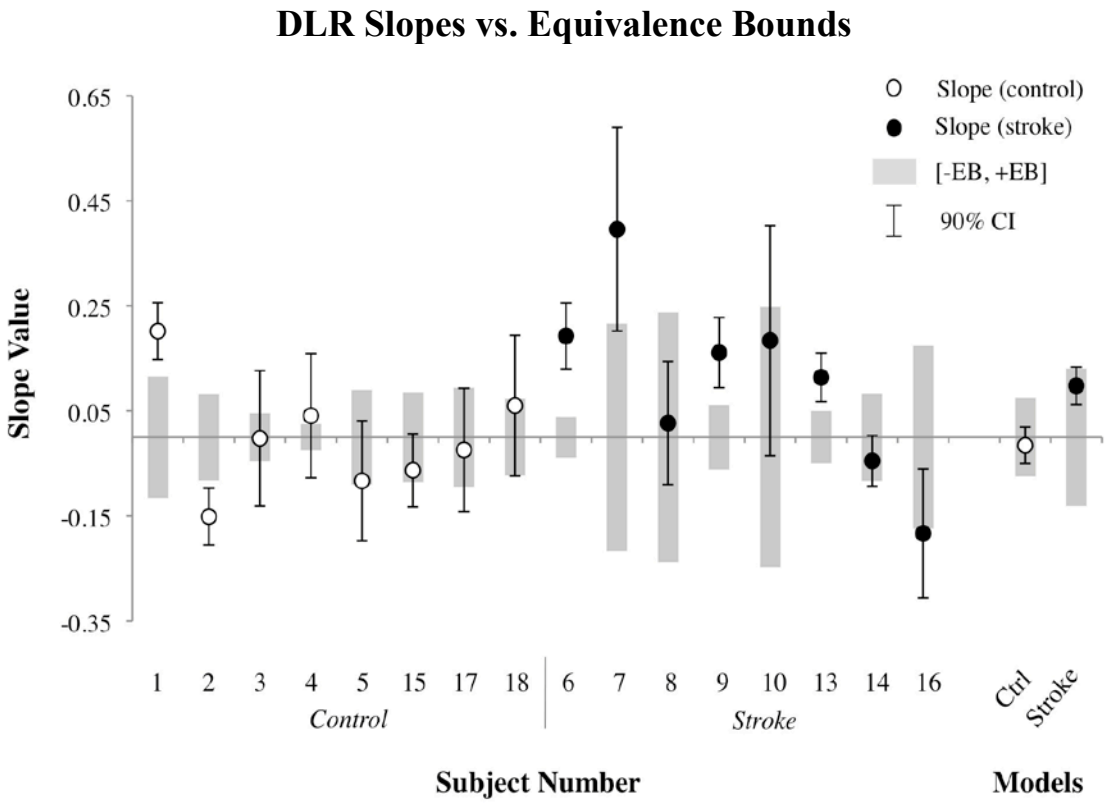

Figure 4-8: DLR Slope values compared to respective Equivalence Bounds (EB) for both individual subjects as well as from the simple linear model. Error bars indicate 90% confidence intervals (CI). The slope value is statistically bounded within its EB if the CI falls entirely within EB interval.

those who loaded more weight on their dominant side tended to have steeper positive slopes. The subjects with relatively even weight distribution tended to have flatter slopes. Figure 4-9 illustrates this relationship: individual subject DLR regression lines are plotted, but they have been shifted so that the height of the center point represents their average assisted weight distribution (BWD_A). In this diagram, it is more clearly apparent that the highest regression lines, which indicate the largest BWD_A values, tend to have steeper, more negative slopes, while the lowest regression lines, which correspond to the lowest BWD_A values, tend to have steeper, more positive slopes.

EQUIVALENCE TESTING RESULTS: INDIVIDUALS

Subject	DLR Slope value	CI within EB?	DLR value within EB?	CI overlaps EB?
		Control		
1	0.202	No	No	No
2	-0.151	No	No	No
3	-0.002	No	**Yes**	**Yes**
4	0.041	No	No	**Yes**
5	-0.083	No	**Yes**	**Yes**
15	-0.063	No	**Yes**	**Yes**
17	-0.024	No	**Yes**	**Yes**
18	0.060	No	**Yes**	**Yes**
		Stroke		
6	0.193	No	No	No
7	0.396	No	No	**Yes**
8	0.027	**Yes**	**Yes**	**Yes**
9	0.161	No	No	No
10	0.184	No	**Yes**	**Yes**
13	0.114	No	No	No
14	-0.045	No	**Yes**	**Yes**
16	-0.183	No	No	**Yes**

Table 4-12: Equivalence testing for individual subjects. The last three columns refer to the DLR Slope value and confidence interval (CI) relative to the individual's Equivalence Bounds: if the entire CI falls within the EB, we can be confident that the true value of the DLR Slope falls within the EB; if the entire CI does not fall within the EB but the DLR Slope value does, there is a chance that the true value of the DLR Slope is within the EB, but we cannot claim that with sufficient confidence; if the CI falls entirely outside of the EB, the statistics indicate that there is actually a significant difference between the DLR Slope value and the EB.

What this correlation indicates is that 1) people seem to have different DLR Slopes, and 2) that the value of the slope relates their weight distribution overall. Furthermore, it indicates that the assumption made in the model – that everyone in the same group has the same DLR Slope – is perhaps invalid. Accordingly, a new model that accounts for this interaction effect was fit to the data:

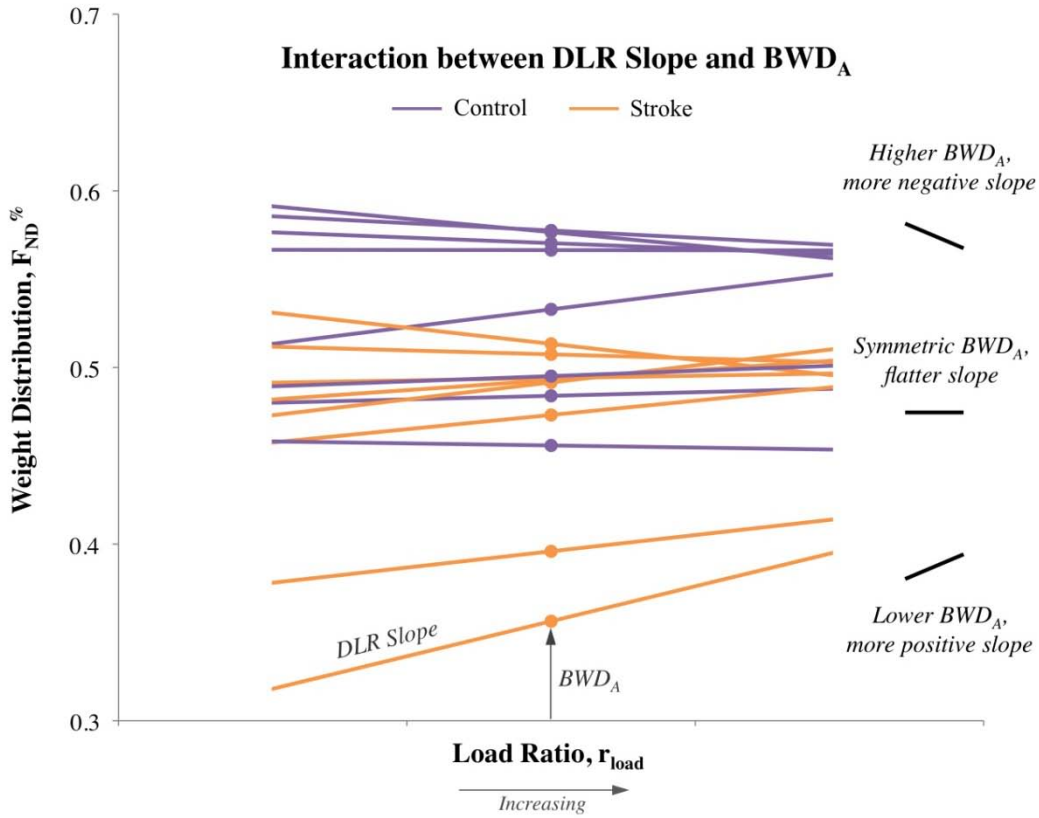

Figure 4-9: Relationship between DLR Slope value and Average Assisted Weight Distribution (BWD_A). The slope of the lines corresponds to each subject's DLR Slope value, and the height of the center point represents the subject's BWD_A. Subjects with larger BWD_A values tend have steeper, more negative slopes, while subjects with smaller BWD_A values tend to have steeper, more positive slopes. Subjects with the flattest slopes tend to have BWD_A values around 50%.

$$F_{ND,ij}^{\%} = \alpha * r_{load,ij} + \beta * BWD_{A,i} + \gamma * r_{load,ij} * BWD_{A,i} + b_i + \varepsilon_{ij} \qquad (4\text{-}1a)$$

$$= \left(\alpha + \gamma * BWD_{A,i}\right) * r_{load,ij} + \left(\beta * BWD_{A,i} + b_i\right) + \varepsilon_{ij} \qquad (4\text{-}1b)$$

in which *i* and *j* are indices referring to subject number and trial number, respectively. Where this model differs from the other model is in the incorporation of BWD_A and its interaction with r_{load}. Rearranging the equation into basic slope-intercept form, as in (4-1b), shows more clearly that the slope term is no longer a constant *m*, but is a function of BWD_A. Fitted parameters are listed in Table 4-13.

INTERACTION MODEL PARAMETERS

	α (SE)	β (SE)	γ (SE)	b_i	
				Avg. (SE)	SD
Control	0.51 (0.22)	1.53 (0.22)	-0.97 (0.40)	-0.27 (0.12)	0.00
Stroke	1.12 (0.23)	2.50 (0.33)	-2.13 (0.49)	-0.78 (0.16)	0.00

Table 4-13: Fitted parameters from the interaction model ($F_{ND}{}^{\%}{}_{ij} = \alpha * r_{load,ij} + \beta * BWD_{A,i} + \gamma * r_{load,ij} * BWD_{A,i} + b_i + \varepsilon_{ij}$). SE = standard error, SD = standard deviation.

Figure 4-10 shows a comparison of DLR Slope values with slopes calculated from the fixed-slope model and from the interaction model for individuals. Although the magnitude of the difference between models is small overall, the interaction model fits the data significantly better than the fixed-slope model for both groups (log-likelihood ratio test, control group: $\chi = 45.6$, $p < 1e\text{-}9$; stroke group: $\chi = 59.4$, $p < 1e\text{-}12$).

Figure 4-11 shows how the slope term from the interaction model varies with average weight distribution for each group. The BWD_A value corresponding to a slope of 0 (x-intercept of Figure 4-11) is 0.520 for the control group and 0.524 for the stroke group. Slope magnitude increases as BWD_A deviates from those values for both groups, although the rate of increase is higher for the stroke group than for the controls. Thus, given a perfect fit to the model, a control subject with a BWD_A of 52% and a stroke subject with a BWD_A of 52.4% would both have DLR Slopes of 0; a control subject with a BWD_A of 40% would have a DLR Slope of 0.12, while a stroke subject with the same BWD_A, 40%, would have a DLR Slope of 0.26.

This experiment and data analysis was designed to assess the influence of SOE on weight distribution by evaluating the relationship between $F_{ND}{}^{\%}$ and r_{load}. Specifically, an independence between the two parameters, or a DLR Slope value close to 0, was taken to be an indication of

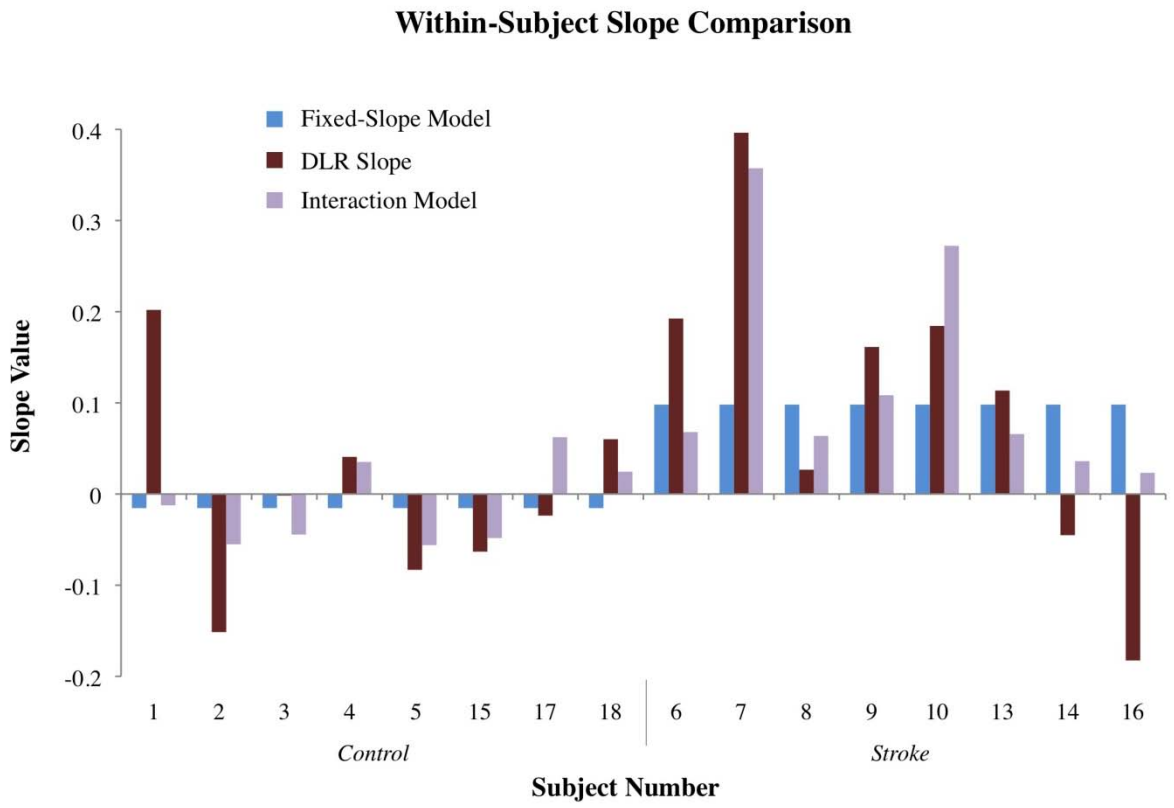

Figure 4-10: Comparison of DLR Slope values with those calculated from the fixed-slope and interaction models. The interaction model takes into account Average Assisted Weight Distribution (BWD_A) and is a statistically better fit to the data.

efferent involvement in the sensory-motor strategy. However, the correlation between DLR Slope and BWD_A, in tandem with the fact that the interaction model was a better fit to the data, implies that the relationship between weight distribution and load magnitude varies among individuals in a way that is related to overall asymmetry. That is, some individuals had slope values close to 0, suggesting a dominant influence of SOE, while others had slope values of much larger magnitudes, suggesting a substantial contribution of afferent feedback. This trend among DLR Slopes prevents us from drawing a simple conclusion about the influence of SOE in STS in general, instead introducing a layer of complexity to our interpretation of the results.

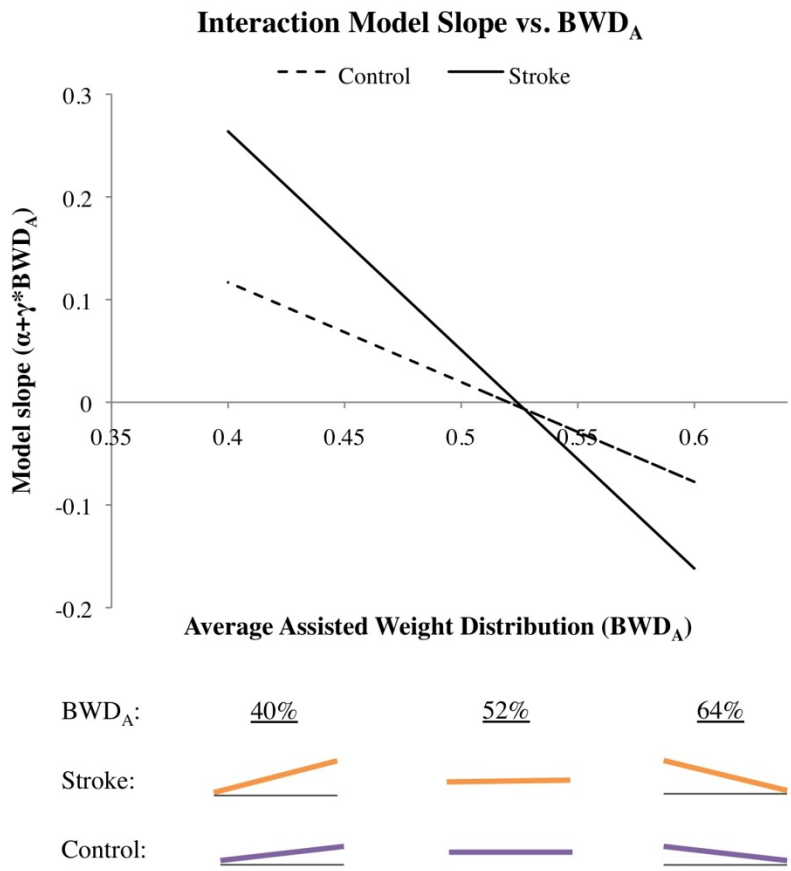

Figure 4-11: *Top*: Slope term from the interaction model plotted over a range of Average Assisted Weight Distribution (BWD$_A$) values for each group. BWD$_A$ has a larger effect on slope for the stroke group than for the control group. X-intercepts are 0.520 and 0.524 for the control and stroke groups, respectively. α and γ values are listed in Table 4-13. *Bottom*: Visual representations of DLR Slopes, according to interaction model, corresponding to specified BWD$_A$ values.

4.4 Chapter 4 Summary

This chapter detailed the data analysis results from the STS experiment. Based on clinical metrics, stroke subjects had more functional impairment and larger degrees of strength asymmetry than control subjects. The most significant strength asymmetries were evident in the ankle movements (plantar flexors and dorsiflexors). Stroke subjects also tended to load more

weight on their dominant (non-paretic) side during assisted STS, whereas the controls tended to load slightly more weight on their reported non-dominant side.

DLR Slope values were smaller among controls than stroke subjects, both in value and in magnitude. Control regression lines tended to be above 50% and downward-sloping, while stroke regression lines tended to be below 50% and upward-sloping. Dev slopes were negative (downward-sloping) or close to 0 for all subjects except one, implying symmetry improves as load ratio increases.

Correlation analysis showed a number of interaction effects among regression metrics, summary metrics, and clinical metrics. Of particular note was that DLR Slope Magnitude had significant correlations with many of the measures relating to asymmetry: specifically, those subjects who were more asymmetric overall tended to have larger slope magnitudes.

A linear mixed model was fit to each group's data to evaluate an average slope based on the assumption that the fundamental distribution-load relationship was the same for members of the same group. The model produced fitted slope values of -0.015 for the control group and 0.098 for the stroke group. Equivalence bounds were calculated for each group as an indication of how much movement variability could be contributing to the slope value. Equivalence test results showed that the control DLR Slope value was bounded within the control EB, which suggested that SOE had a strong influence. The DLR Slope value for the stroke group fell just short of being statistically bounded by the EB, but the majority of the confidence interval fell within the EB. However, when the same analysis procedure was carried out for individual subjects, only one subject had a DLR Slope value that was statistically bounded by her Equivalence Bounds. The confidence intervals for the DLR Slopes of the other 15 subjects fell either partially or

entirely outside of their respective EB. These individual results suggest the opposite of the group results: that SOE is not the single most dominant influence on force perception.

Given the relationship between DLR Slope and BWD_A, as well as the lack of agreement between group and individual results, a more complex interaction model was fit to the data. In this model, slope was a function of BWD_A, which resulted in a significantly better fit to the data than the fixed-slope model. Thus, it seems that for both stroke and control subjects, people with higher BWD_A values had steeper, more negative slopes, while those with lower BWD_A values had steeper, more positive slopes. Furthermore, extrapolation of the results of the interaction model indicates that as BWD_A values get more asymmetric, slope magnitudes for the stroke group increase at a faster rate than for the control group.

Chapter 5:

Discussion

Chapter 4 presented the results of the STS experiment: although a fixed-slope linear model was initially used to quantify the relationship between load ratio and weight distribution for each group, subsequent correlation analysis indicated that the fixed-slope assumption did not accurately describe the within-group variation of slope values. The data were better reflected by an interaction model, which incorporated a relationship between DLR Slope and Average Assisted Weight Distribution. This chapter discusses how these results suggest that while SOE does have a strong influence on weight distribution perception, afferent feedback seems to make a noteworthy contribution as well. Moreover, the patterns describing the variation of slope magnitude among subjects in each group imply that individuals may use different sensory-motor strategies depending on their overall level of asymmetry, and that the relative magnitudes of the efferent and afferent contributions appear to differ among those strategies.

This chapter is laid out as follows: Section 5.1 reviews the results of fitting the fixed-slope model and the data trends that led to the reformulation of the slope parameter. The tendency for Dev Slope values to be negative and the statistically better fit of the interaction model imply that the weight distribution strategy is influenced both by afferent feedback as well as the individual's overall level of asymmetry. Section 5.2 discusses the implication of these results with respect to

the research question, suggesting that while both efferent and afferent signals contribute to the perception of weight distribution, the relative magnitude of those influences is not the same for all individuals. Section 5.3 presents the limitations of this study, acknowledging that stronger conclusions could be made given a larger and more diverse subject pool.

5.1 The Relationship between Weight Distribution and Load Ratio

5.1.1 Preliminary Analysis: the Appropriateness of a Fixed-Slope Model

It was initially assumed that the fundamental Distribution-Load Relationship was the same for members of the same group; accordingly, a fixed-slope model of the following form was fit to the data: $F_{ND}{}^{\%}{}_{ij} = m^* r_{load,ij} + b_i + \varepsilon_{ij}$. This model produced fitted DLR slope values of $m = -0.015$ for the control group and $m = 0.098$ for the stroke group. The control group slope was not statistically different from 0 and was also bounded within the group equivalence bounds (EB = ±0.075). Thus, even if the true mean slope were not exactly 0, its value was small enough to be attributable to natural movement variability as opposed to changing load ratio. In the case of the stroke group, the fitted DLR Slope *was* significantly different from 0 and *not* statistically bounded by the group EB (EB = ±0.131). However, since the confidence interval was well within the lower EB and extended just slightly past the upper EB (CI upper bound = 0.134; +EB = 0.131), the DLR Slope likely falls within the Equivalence Bounds for the stroke group as well ($p < 0.10$). Although these results were not as definitive as those for the control group, they still suggest that weight distribution is largely independent of load ratio. Thus, it seems that people recovering from stroke do rely on SOE in the perception and control of weight distribution in STS.

Interestingly, however, the same patterns were not as evident among individuals as they were for the groups. That is, when individual DLR Slopes were compared to individual Equivalence Bounds calculated based on each subject's own Unassisted Variability (VR_U), only one subject had a slope statistically bounded within her EB. Given the group results, more (if not most) of the DLR Slopes were expected to fall within their respective EB. Moreover, there were no consistent trends as to how people's slopes related to their EB – that is, some DLR Slope values fell within the subject's EB, but the confidence interval extended past the upper or lower bound, while some confidence intervals fell entirely outside of the subject's EB. It seems, instead, that more subtle individual differences were present that were not captured in the fixed-slope model.

Two unexpected trends in the data suggested that a fixed-slope model was not, in fact, the most appropriate model: first, that symmetry improved with load ratio, and second, that DLR Slope value correlated with BWD_A. These trends also have meaningful implications with regard to the influence of SOE, as they suggest that while efference does seem to have a substantial influence, afference also has a noticeable contribution.

5.1.2 Symmetry Improves with Load Ratio

That weight distribution symmetry improved at higher load ratios was evident from the fact that Dev Slope values tended to be negative (mean = -0.071, $p = 0.015$; excluding one outlier, range = [-0.396, 0.005]). As Dev measured deviation from perfect symmetry (50%), a negative Dev Slope indicates that deviation decreases as load ratio increases – that is, subjects were better able to distribute their weight symmetrically when they were supporting larger portions of their body weight and getting less assistance from the STS device. This trend indicates that weight distribution symmetry was not limited by weakness; if it were, $F_{ND}^{\%}$ would have decreased away

from 50% as load ratio increased (see Figure 3-9), producing a positive Dev Slope. Furthermore, it suggests that weight distribution symmetry was limited by perceptual ability and influenced by afference: since larger muscle forces imply larger neural signals – which are easier to perceive than smaller ones – people should be able to match those signals between sides more accurately. Indeed, nearly all of the subjects commented that at the lowest load ratios, they felt like they were not doing anything and the assist was doing all the work for them. Similarly, Lafargue et al. (2003) found that both their control subjects and their deafferented subject were more accurate at a bilateral matching task at higher force levels than at lower force levels. That said, other studies that have focused on force and effort matching in bilateral tasks have found that subjects were equally or more accurate at lower force levels (Bertrand et al., 2004; Carson et al., 2002). This discrepancy in results may be related to a subtlety in the experimental design: whereas some studies required subjects to first produce a target force with the reference limb and then match with the other side, others required subjects to perform the task with both limbs simultaneously. The slight difference is that in the latter condition, subjects could attempt to send equal motor commands down to each limb at the same time. The former condition was more closed-loop in that the CNS had to determine what magnitude of motor command produced the target reference force, measure it, and while monitoring it, reproduce the same motor command with the matching side. Studies with deafferented individuals have found that while these subjects seem to be able to match the initial motor command, or "pulse," fairly well (Fleury et al., 1995; Lafargue et al., 2003), they have trouble maintaining a constant force output (Lafargue et al., 2003). This difficulty implies that sustaining a contraction involves a slightly different mechanism than initially producing it – presumably, a larger reliance on afferent feedback. Thus, slightly different results depending on how the bilateral matching is performed would not be

unreasonable. That said, this theory does not provide an explanation for the discrepancies in the aforementioned studies, as Lafargue et al. (2003) and Carson et al. (2002) used reference-matching paradigms, while Bertrand et al. (2004) used a simultaneous exertion model. Still, this methodological difference is worth considering when comparing studies that ask subjects to *produce* equal bilateral forces versus indicating *when* bilateral forces become equal (e.g. in a weight-matching task).

Regardless, the tendency for Dev Slopes to be negative implies involvement of afferent feedback, at least to some extent, as subjects were better able to perceive and distribute their weight evenly when the load magnitude was larger and stronger sensory feedback signals were available. This contribution of afferent information is not surprising given that our sense of movement and proprioception originates peripherally (McCloskey, 1981) and that STS is a dynamic, multi-joint movement. Even though people were focusing on weight distribution and muscular effort, proprioceptive cues are likely factored in at a subconscious level: Bertrand et al. (2004) found no influence of somatosensory impairment on isometric grip force ratios; however, when the same group looked at a multi-joint, directional isometric contraction, they found that they were better able to predict bilateral force asymmetries by accounting for both strength asymmetry and proprioceptive impairment (Mercier et al., 2004). McCloskey et al. (1974) also found that in a weight-matching task, subjects were able to judge relative heaviness more accurately when they were allowed to move their arms versus when they had to assess the weights isometrically. Thus, sensory information stemming from movement and proprioception may not be a fundamental factor in force perception, but it does seem to supplement the centrally-generated signals and improve the accuracy of an individual's perception, particularly in the context of more complex movements.

5.1.3 People Who Are Less Symmetrical Have Steeper Slopes

The second unexpected finding was the correlation between DLR Slope and BWD_A, which led us to reformulate the model and incorporate an interaction term: $F_{ND}{}^{\%}{}_{ij} = \alpha * r_{load,ij} + \beta * BWD_{A,i} + \gamma * r_{load,ij} * BWD_{A,i} + b_i + \varepsilon_{ij}$. The fact that data were better described by this interaction model, which modeled the slope as a function of BWD_A, contradicts the assumption made in the fixed-slope model that all members of the same group have the same distribution-load relationship (DLR). It seems, in fact, that slope values vary among individuals and that they are related to overall asymmetry. Based on the fitted parameters of the interaction model, a person with an average assisted weight distribution (BWD_A) of approximately 52% would have a DLR Slope of 0. Interestingly, this value was nearly the same for both controls and stroke subjects (52.0% and 52.4%, respectively), despite the fact that the model was fit to each group independently. As BWD_A increases from 52%, the model predicts that DLR Slopes get increasingly negative, and as BWD_A decreases, slopes get increasingly positive; more generally, as BWD_A values become increasingly *asymmetric*, DLR Slopes become steeper, oriented such that symmetry improves as load ratio increases. This pattern holds for both groups, except that the *rate* at which DLR Slopes get steeper is higher for the stroke group than the controls. That is, based on the model, a control subject with an Average Assisted Weight Distribution of 40% would theoretically have a DLR Slope of 0.122, while a stroke subject with the same BWD_A (40%) would have a DLR Slope of 0.268 (see Figure 4-11). The similarity of this trend between groups suggests that the fundamental internal mechanism responsible for slope variation based on asymmetry may be similar for all subjects, but that it is perhaps exaggerated in stroke survivors.

It is possible that large slope magnitudes could have arisen from large variability and perhaps still have fallen within individual equivalence bounds; the data, however, do not support that explanation. The DLR Slope magnitudes that were greater than their +EB (not considering confidence intervals) were significantly larger than the slope magnitudes that were within their EB (one-sided t-test, H_A: [|DLR| > +EB] > [|DLR| ≤ +EB]; $t(10.9) = 2.9$, $p < 0.01$), as shown in Figure 5-1. Furthermore, there was a significant correlation between slope magnitude and distance outside of EB: that is, larger slope magnitudes tended to fall farther outside of their EB than smaller slope magnitudes (|DLR| - EB vs. |DLR|; $\rho = 0.73$, $p < 0.0015$). These patterns rule out the possibility that larger DLR Slopes were due to higher natural (that is, unassisted) variability, because if that were the case, the corresponding Equivalence Bounds would be larger as well.

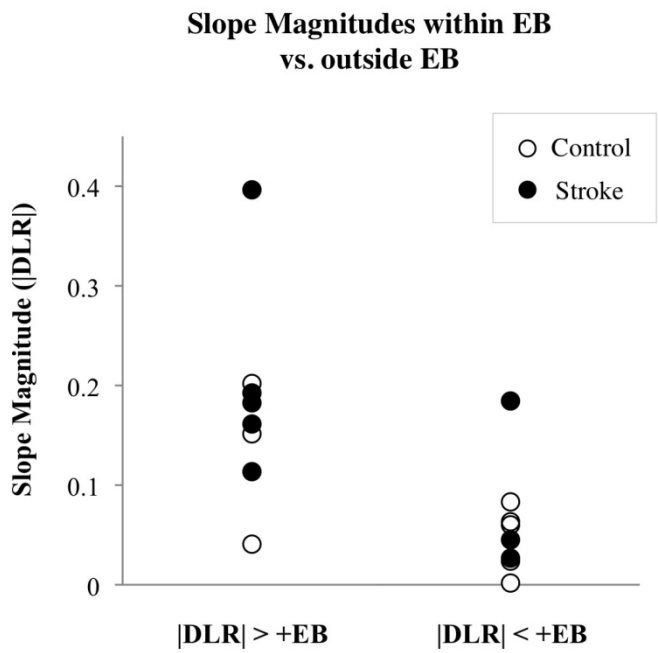

Figure 5-1: Comparison of DLR Slope magnitudes that are within the individual subject's Equivalence Bounds (|slope| < +EB) and slope magnitudes that fall outside of subject's Equivalence Bounds (|slope| > +EB). Slopes outside of EB are significantly larger than slopes within EB (one-sided t-test, $p < 0.01$).

The other aspect of the data that supports a connection between weight distribution and DLR Slope is that DLR Slope Magnitude significantly correlates with a number of the metrics relating to general asymmetry and functional ability: specifically, $Asymm_A$, $Asymm_U$, Dev_U, Dev_A, and DF diff (positive correlation); and Berg score (negative correlation). Nearly significant is also ND PF (negative correlation; $p = 0.0502$). These relationships suggest that the subjects with steeper DLR Slopes were also the ones who were more *asymmetric* in both assisted and unassisted STS, had *larger* strength asymmetries in their dorsiflexors, had *lower* Berg scores, and had *weaker* plantarflexors on the non-dominant side (see Figure 5-2). These relationships

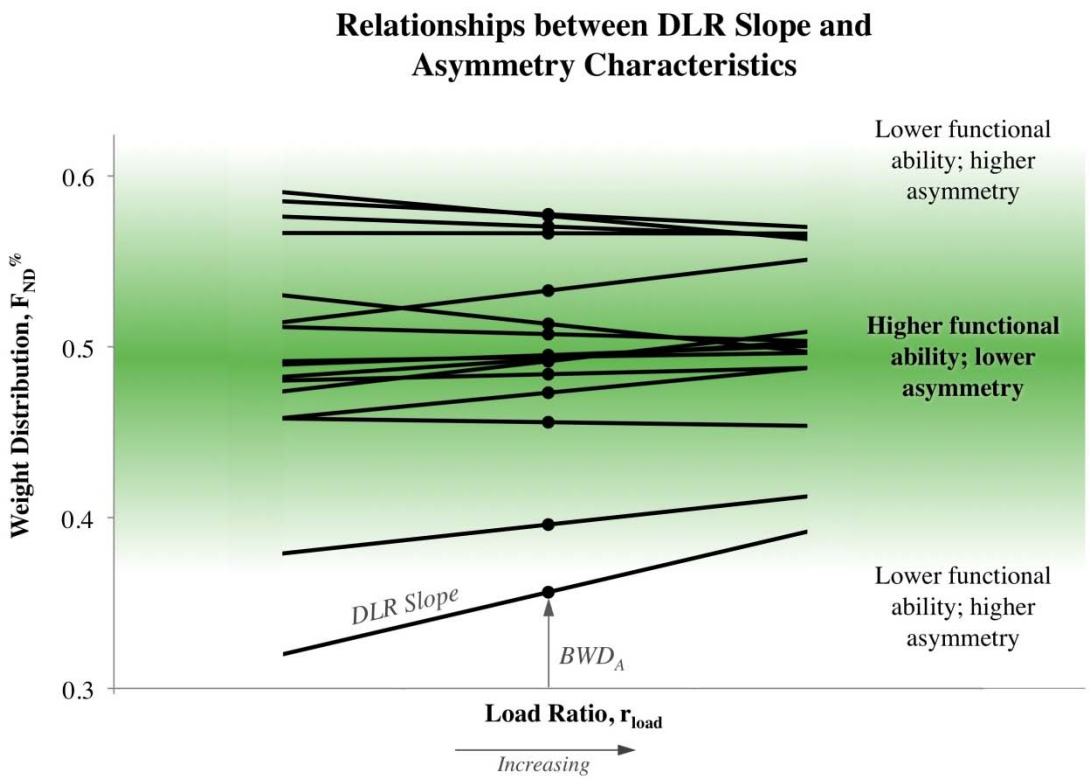

Figure 5-2: Correlation trends among DLR Slopes, BWD_A, and asymmetry metrics. Slopes of black lines show DLR Slope value of individual subjects, shifted so that the height of the middle point corresponds to the subject's BWD_A value. Subjects with flatter DLR Slopes (green zone) tended to have higher Berg scores and less asymmetry overall; those with larger DLR Slope magnitudes (white zone) tended to have higher values of asymmetry metrics and lower Berg scores.

applied to all subjects, regardless of group. Thus, it seems that the people with steeper DLR Slopes tended to be more asymmetric and impaired than those with flatter DLR Slopes.

5.2 Do People Rely Primarily on SOE?

The guiding research question for this work, as presented in Chapter 2, was whether SOE were the dominant factor in the control and perception of weight distribution during STS among hemiparetic individuals. That is, do people rely primarily on a corollary of the outgoing motor command as opposed to afferent information reflecting actual degrees of muscle contraction? Results do support that SOE is *a* factor in weight perception because, unlike any of the theories favoring the dominance of afferent feedback, it provides a logical explanation for why people perceived their weight distribution to be symmetric even when it was not. However, the results definitely do not support the idea that SOE is the *sole* factor in weight perception. The fact that people tended to become more symmetric at higher load ratios (Dev Slopes were negative) implies an afferent contribution as well, at least to some extent: as load ratio increases, muscle forces and corresponding afferent signals also increase, which likely makes them easier to perceive and thus easier to match between sides. Other sources of peripheral information, such as larger cutaneous signals from increased pressure between the subjects' feet and the force plates, likely contribute as well. Consequently, people's deviation from symmetry decreases. As this trend held for all subjects in both groups, it seems that people in general *do* make use of afferent signals, and weight distribution perception is not exclusively based on SOE. Results from a multitude of other studies support this dual influence as well (Fleury et al., 1995; Gandevia & McCloskey, 1977b; Lafargue et al., 2003; Mai et al., 1991; McCloskey et al., 1983; Miall et al., 2000).

5.2.1 The Influence of Afference: Symmetric vs. Asymmetric Individuals

Interestingly, it seems that perhaps some people make use of those afferent signals more so than others do. Specifically, the asymmetric subjects seemed to be more dependent on the afferent information than the symmetric subjects were. This hypothesis is based on the interaction effect found between DLR Slope and BWD_A, as well as on the correlations between asymmetry measures and DLR Slope Magnitude: what the DLR Slope fundamentally measures is how large a weight distribution change, $\Delta F_{ND}^{\%}$, is expected in response to a change in load ratio, Δr_{load}. A steeper DLR Slope – either positive or negative – means that for a given change in load ratio, a larger change in weight distribution would be expected than in the case of a flatter DLR Slope. Consequently, among the subjects in this experiment, changing the load ratio produced a *larger change* in weight distribution in the individuals who were generally more *asymmetric*. More specifically, given that Dev Slopes were mostly negative, *increasing* the load ratio produced a larger *improvement* in weight distribution symmetry for the more asymmetric people, as shown in Figure 5-3. This in turn implies that the most asymmetric people were the ones whose weight distributions, and perception of symmetry, were most affected by a change in load ratio. Apparently, the increase in load ratio is accompanied by an information source on which the asymmetric people are more dependent for accurately perceiving and distributing their weight evenly – presumably, stronger afferent signals corresponding to larger muscle forces.

Why would the more asymmetric people have a greater dependence on peripheral and afferent inputs to perceive their weight distribution than the more symmetric people? It is known that neuroplasticity allows the central nervous system (CNS) to tune the relative priority of various sensory signals in order to extract the most accurate information about how the body is interacting with an environment. For example, Fleury et al. (1995) conducted a study involving a

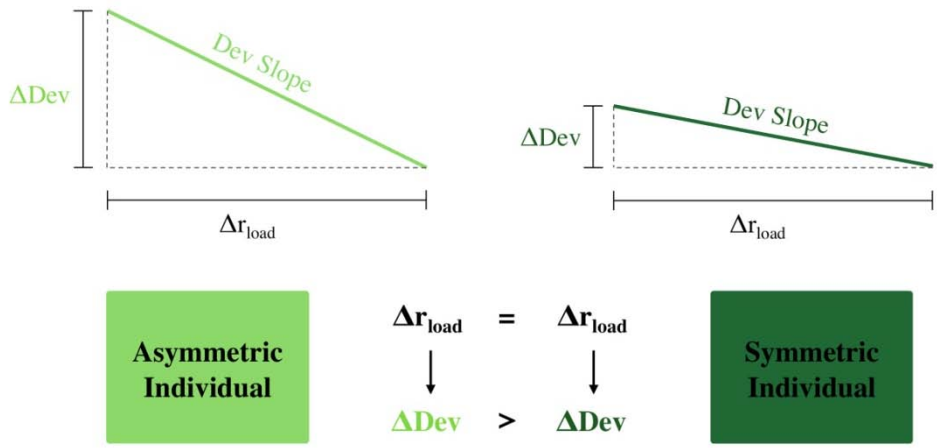

Figure 5-3: Comparison of relationship between weight distribution symmetry and load ratio for symmetric and asymmetric individuals. Steeper negative Dev Slopes indicate that given a change in load ratio, the improvement in weight distribution symmetry is larger than for a flatter Dev Slope. As steeper slopes corresponded to more asymmetric individuals, it seems that those individuals are more affected by a change in load magnitude than more symmetric individuals are.

weight discrimination task in which a deafferented subject performed nearly as well as controls when she was able to compensate for a lack of afference by relying mostly on visual and vestibular feedback. In another study, a deafferented subject seemed to be able to interpret a gradual weight change of a supported load by using the direction of hand drift (Miall et al., 2000). A more common example is the development of keener hearing, smell, and other senses in blind people to compensate for lack of sight. Perhaps, in asymmetric people, the CNS has de-prioritized SOE signals because they do not provide accurate information about bilateral weight distribution; instead, it relies more on afferent inputs because the information provided is more accurate, despite being slower and delayed (Wolpert & Ghahramani, 2000). If this were the case, it would make sense that these asymmetric people become more uneven at lower load ratios: smaller loads and muscle forces produce smaller afferent signals, on which they are more dependent. Thus, these people have more difficulty perceiving their weight distribution

accurately. Accordingly, when the load ratio is increased and larger afferent signals become available, a more dramatic improvement in symmetry is produced.

5.2.2 The Influence of Afference: Stroke vs. Control

The rate of change between DLR Slope and BWD_A from the interaction model was larger for the stroke group than the controls, which might be due to the fact that the most asymmetric stroke subjects were also the ones with the most sensory and proprioceptive deficits. Although they were not completely deafferented, they clearly were not getting as much sensory feedback at a given load ratio as less impaired and control subjects, making it even more difficult for them to correctly assess and equalize their weight distribution. Furthermore, it is possible that their perception was further skewed by a mismatch between "anticipated" sensory feedback resulting from the motor signal commanded and the actual sensory feedback transmitted. Gandevia & McCloskey (1977b) showed that limiting afferent feedback via finger anesthetization resulted in an increase in perceived heaviness at the same level of force production. If the same phenomenon were happening among the stroke subjects, it would make sense that they demonstrate larger errors in weight distribution symmetry as well as a larger change in weight distribution for a given change in load ratio than do healthy controls, who may have strength asymmetries but essentially no sensory deficits.

An important point of distinction is that the data suggest that asymmetric people are dependent on afferent information *more than symmetric people are,* which is *not* to say that asymmetric people are more dependent on afferent information than they are on SOE (see Figure 5-4). If that were the case, the afferent signals indicating asymmetry would overrule the SOE signal, which would result in the subject either adjusting their weight distribution to be

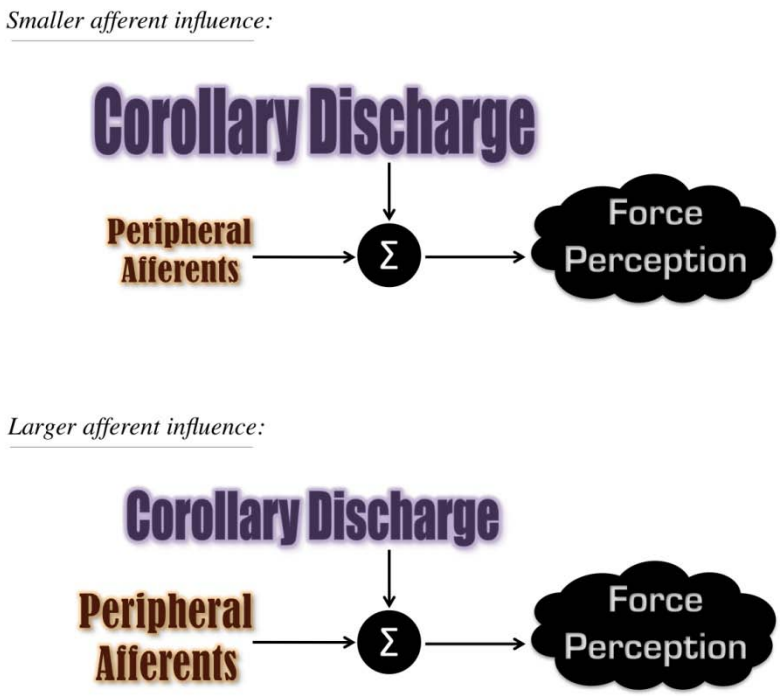

Figure 5-4: Qualitative depiction of the relative influences of efferent and afferent information on force perception. It seems that while a corollary discharge related to efferent signals plays a dominant role in general, some people rely on afferent feedback more than others do.

symmetric, or at the very least, perceiving that it was not symmetric. All trials included in this analysis were executed such that subjects *perceived* their weight distribution to be even. Not one person commented that they could tell their weight distribution was uneven but that they were unable to correct it.

5.2.3 Literature Supporting Individual Differences in Sensory-Motor Strategies

These theories are based mostly on the trends predicted by the models, as the inclusion criteria limited the subject pool to a fairly high-functioning group of stroke survivors, many of whom were not exceptionally asymmetric. That said, other studies have also found indications that different people use different sensory-motor strategies for bilateral force matching – that is,

while some people seem to rely predominantly on SOE, others appear to rely heavily on other signals as well. Bertrand et al. (2004), who looked at grip force matching across multiple force levels, found that when they divided their stroke group into two sub-groups based on the degree of strength asymmetry, the mildly weak group showed less consistency in bilateral force ratio than the controls or the severely weak sub-group. Specifically, the mildly weak group produced higher force ratios at the lowest force level than at the medium or high force level. Since a consistent force ratio across force levels indicates a matching strategy based on scaling the motor command, the implication is that the mildly weak group perhaps used a slightly different strategy.

Brière et al. (2013) also found evidence of differing motor strategies corresponding to the degree of strength asymmetry. Their study, as previously described in Chapter 2, was similar to our study in that they investigated the influence of SOE on weight distribution in hemiparetic individuals during STS, but they focused specifically on knee effort. What they found was that the motor strategies differed among the stroke subjects based on raw strength asymmetries: the mild subgroup (< 10% strength asymmetry) behaved similarly to controls, demonstrating equal weight distribution and effort levels between sides; the moderate subgroup (11%-24% strength asymmetry) showed significant differences between sides for both weight distribution and effort level; and the severe subgroup (> 25% strength asymmetry) showed significant differences between sides for weight distribution, but not for effort level. Thus, only the mild and severe subgroups showed indications of a strategy based on effort-matching, and only the severe subgroup data supported their hypothesis that stroke subjects would have *asymmetric* weight distribution but *equal* levels of effort between sides. Although the moderate subgroup also demonstrated asymmetric weight distribution, their effort levels on the non-paretic side were

significantly higher than on the paretic side. One simple and quite reasonable explanation that the authors propose is that effort *is* being matched, but across other or additional muscle groups such as the hip extensors or ankle plantar flexors, which were not accounted for in their effort metric.

Despite the fact that they took a very different approach to answer a similar research question, our results are consistent with Brière et al.'s. On a qualitative level, we both found that subjects' weight distribution strategies seem to vary among individuals, and that the variation correlated with the overall degree of asymmetry. Among our subjects, this correlation was reflected in the fact that the more symmetric people overall had flatter DLR Slopes, indicating a motor strategy based on effort-matching, while the more asymmetric people had steeper DLR Slopes, indicating the influence of variables other than effort. Among Brière et al.'s subjects, those with the most mild and most severe strength asymmetries seemed to employ a motor strategy based on matching knee extensor effort, while those with moderate strength asymmetries were either matching effort based on additional muscle groups, or perhaps matching a different parameter altogether. Although our measurement of knee extensor strength was not precise enough to determine which of their subgroups each of our subjects would fall into, it is very possible that all of our subjects would have been considered mild or moderate, in which case the trend we found would align with Brière et al.'s findings.

5.3 Limitations and Considerations

As mentioned, the interpretation of these results is based in large part on extrapolating the results of the fitted interaction model. Access to a larger subject pool, particularly stroke survivors with more moderate to severe disability and asymmetry, would have allowed us to see

if the trends displayed in the data would persist over a wider range of impairment levels. Although the difference between the fixed-slope model and the interaction model was statistically significant, the magnitude of the difference between fitted slopes for individual subjects was generally small, except for the subjects with the largest slopes. Of course, these subjects were also the most impaired and most asymmetric. Unfortunately, the requirement of independent STS and the ability to commute to the experiment location imposed a lower bound on the impairment level we could use in the subject pool. We could not require much stricter criteria, such as an absence of sensory impairment, as that would have limited the subject pool even further. At the same time, having a less-specific subject pool allows for generalization to a broader community.

Another consideration is that the assisted STS motion was not exactly the same as the unassisted motion. The unassisted rise necessitates momentum, which is generated by the initial trunk rotation and hip flexion, to move the center of mass from over the seat to over the base of support of the feet. The assisted rise motion is slower, so it does not precisely match the flow of the dynamics of the unassisted rise. The slower movement does allow the user to focus more on effort and weight distribution instead of stability and balance, which is beneficial from an experimental perspective, but somewhat hinders the ability to generalize to the full dynamic movement. That said, even though the assisted rise is not a perfect representation of the unassisted rise, it is still a dynamic movement and reflects the fundamental nature of a functional task.

5.4 Chapter 5 Summary

This chapter discussed the results of the STS experiment and their implications for the role of SOE in weight distribution perception and control. As the distribution-load relationship was initially assumed to be constant among subjects in each group, a fixed-slope model was fit to each group's data and the slope values were compared to their respective Equivalence Bounds. For the control group, equivalence test results were statistically significant, indicating that the fitted model slope was small enough in magnitude to be attributed to natural movement variability as opposed to a change in load ratio. For the stroke group, equivalence test results were borderline significant, indicating that the magnitude of the fitted model slope *could* have been due to natural movement variability; however, the significant non-zero-ness of the slope value alluded to the presence of some degree of underlying relationship.

The existence of this underlying relationship was supported by the negative correlation found between DLR Slope and BWD_A. Accordingly, the linear model was reformulated as an interaction model, expressing the slope parameter as a function of BWD_A. This model was a significantly better fit to the data for both groups. The fact that there was a relationship between subjects' slope values and average asymmetry not only fails to support the idea that weight distribution is independent of load ratio, but also suggests that the *degree* to which an individual's weight distribution is affected by load ratio is related to their overall asymmetry. Furthermore, the fact that Dev Slope values were significantly less than 0 indicates that subjects were better able to distribute their weight evenly at higher loads. While this trend does oppose the idea that weakness is the limiting factor in weight distribution symmetry, it also suggests that afference has some influence on perception.

Although the results did not show that SOE makes an overwhelmingly dominant contribution to weight distribution perception, they by no means suggest that it is absent altogether. Both afference and efference seem to provide meaningful contributions to the sensory-motor strategy invoked for perceiving and controlling weight distribution. This finding is not surprising, as it echoes a well-articulated perspective offered by McCloskey more than 30 years ago:

> "That one must attribute various aspects of kinesthetic or other sensibility only to central, command-related signals or only to sensory signals arising peripherally is an extreme view. Too frequently the claimed demonstration of a contribution of one type of signal to some aspect of sensation is taken as evidence against the participation of the other type. Such arguments are based on an assumed exclusiveness of contribution that does not exist" (McCloskey, 1981, p. 1415).

Moreover, the variation in the distribution-load relationships among individuals suggests that perhaps the relative importance of SOE versus afferent signals is tuned on an individual basis: while the CNS in more symmetric individuals can rely predominantly on SOE (because it does not provide misleading information on bilateral force symmetry), the CNS in more asymmetric individuals reduces its reliance on SOE and increases its dependence on afference to acquire a more accurate sense of weight distribution. These results reflect those of other studies that found, among subject groups divided based on strength asymmetry, slightly different sensory-motor strategies and dependence on SOE.

Chapter 6:

Conclusion

The question motivating this work was whether or not people's perception and control of their STS weight distribution is based predominantly on a sense of effort – which stems from a corollary discharge of the internal motor command – as opposed to on afferent signals produced from interacting with the environment. If effort were the primary controlled parameter, it could explain why individuals with hemiparesis distribute their weight unevenly during STS, despite their perception that their weight distribution is equal.

The role of SOE in STS weight distribution was evaluated by conducting an experiment using a robotic STS assist device. The device was used to share the load in order for the user to perform the STS movement at sub-maximal force levels – that is, perform a functional task without having to support their entire body weight. The control system was designed to provide feedback in a way that allowed for the manipulation of the load ratio (r_{load}) so that specific force levels could be targeted. Both stroke subjects and healthy controls participated in the study and performed assisted STS repetitions while supporting between 40% and 70% of the normal body force required for them to rise. Subjects were asked to distribute their weight such that they perceived it to be equal between sides. If SOE had the strongest influence on weight distribution, the percentage of the load taken through the non-dominant side ($F_{ND}^{\%}$) was expected to remain

constant within-subject across all load ratio values. If this were the case, the regression lines fit to $F_{ND}^{\%}$ vs. r_{load} (the distribution-load relationship) should have been horizontal, with slopes approximately equal to 0. If afference were playing the most dominant role in weight distribution perception, subjects should have been able to maintain an even weight distribution, at least up until a point of maximum strength. Accordingly, $F_{ND}^{\%}$ values should have been approximately equal to 50%, and Dev Slopes either greater than or equal to 0 (see Figure 3-9).

Data analysis indicated that a simple, fixed-slope linear model was insufficient for capturing key trends in subjects' slope values. Instead, the distribution-load relationship across individuals was better described by the interaction model that accounted for the correlation between DLR Slope and BWD_A. This correlation, along with other trends in the data, implied that the most asymmetric people overall were the ones whose weight distribution was the most strongly affected by load magnitude.

This thesis concludes by summarizing an overall response to the research question as put forward by the data. Finally, recommendations are provided for how these results can be utilized for improving stroke rehabilitation.

6.1 Is Weight Distribution Perception during Sit-to-Stand based on Sense of Effort?

Weight distribution perception during STS is definitely influenced – to an extent – by SOE. The data offered no reason to believe that SOE was completely absent; furthermore, theories citing afference as the major influence on perception do not provide an explanation as to why some people perceived their weight distribution to be symmetric when it actually was not.

Unsurprisingly, however, the results did not lay out a simple explanation as to exactly how dominant SOE is, as afferent information does seem to contribute as well. The correlations found between the asymmetry metrics and both the DLR Slope magnitude as well as Dev Slope values indicated that the subjects who were in general the most asymmetric were the ones with the steepest slopes: that is, a change in load ratio produced a larger change in weight distribution symmetry among asymmetric subjects than among symmetric subjects. This effect on asymmetric individuals could be due to a stronger dependence on afferent information, as sensory feedback is enhanced at higher load ratios given the larger load magnitude. Indeed, it would make sense that in asymmetric individuals, the CNS has adapted to rely more on peripheral afferent information since the corollary discharge of the motor command does not provide accurate information on relative bilateral force magnitude. In stroke subjects with sensory deficits, this increased dependence on afference could also be due to sensory "mismatch" between what the CNS "anticipates," given the motor command, and what actually gets transmitted back to the sensory processing areas. Although it is not explicitly clear how afferent information is used in conjunction with or relative to SOE, these results corroborate those of other studies that different sub-groups of stroke survivors, divided by strength asymmetry, seem to use slightly different sensory-motor strategies in bilateral matching tasks.

6.2 Recommendations & Future Work

This study joins a growing body of literature supporting the idea that the integration of sensory and motor signals into a single sense of perception is quite complex. Given the degree of individual differences, both among healthy and stroke subjects, as well as the apparent task-specific dependence of sensory-motor perception and control, it is reasonable to expect that

significantly more work needs to be done before all the pieces contributing to the internal control system can be accurately modeled. Further investigation of the influence of SOE relative to other afferent signals in functional tasks would provide helpful information towards reaching this milestone.

However, improvements in stroke rehabilitation require immediate attention and cannot be postponed until that milestone is reached. An alarming statistic revealed that many stroke survivors consider a severe disability to be worse than death; in fact, a health state preference assessment estimated that more than 80% of people overall would prefer to die than to have a major stroke (Public Health Agency of Canada, 2011). Clearly, merely surviving a stroke is not a high enough standard, and post-stroke quality of life needs to be dramatically improved.

Although the results of this work on the relative contribution of SOE in STS are not explicitly clear, one thing they have provided is further affirmation that many stroke survivors are unable to accurately perceive force exertion and weight distribution during a bilateral task. Moreover, many of them are unaware of the extent to which their perception is inaccurate, making it difficult for them to take ownership of their recovery. Indeed, many of the participants in this study wanted to know "how they did" with respect to their weight distribution, and were motivated by the task of working with the assist to achieve the target load ratio. Given that a full motor recovery often requires many more hours of therapy than are available (and affordable) to most people, it is worthwhile to give stroke survivors as much information about their progress and recovery status as possible so that they can focus on making small efforts towards improvement in their daily lives. Even in the specific case of weight distribution in STS, it has been shown that people are able to rise more symmetrically when they consciously focus on their weight distribution than when they are given no instruction (Engardt & Olsson, 1992). How can

people be expected to improve if they are not cognizant of what they are doing wrong? Making hemiparetic individuals aware that what they perceive their body to be doing is inaccurate can, at the very least, give them the opportunity to work towards improvement on their own. Even more helpful would be to incorporate perception and awareness into therapy itself: to encourage people to self-evaluate instead of merely executing repetitions and relying only on the therapist to give feedback. Lang et al. (2009) reported that on average only 11 transfers were performed in a therapy session; meanwhile, among community-dwelling older adults, the STS movement is performed 70 times over the course of the average day (Grant, Dall, & Kerr, 2013). How much more improvement would stroke survivors demonstrate if they focused on their weight distribution – or other movement parameters – during every rise motion, instead of just the 11 reps performed in the presence of a therapist? Not only might this empower people to take ownership of their recovery, but perhaps could also serve as a rehabilitation exercise in itself by working to reestablish neural traffic.

Another valuable research direction to pursue would be to investigate how best to *teach* people to improve their weight distribution. In addition to task-specific practice and strength training, the incorporation of biofeedback can be an effective tool in motor learning. Both Engardt (1994) and Cheng et al. (2001) have conducted studies in which auditory and/or visual feedback reflecting weight distribution asymmetry was provided to stroke subjects during the STS movement. While both groups found that feedback improved weight distribution symmetry over the course of training, their results differed with respect to the long-term retention of those improvements. The difficulty of evaluating the effectiveness of feedback in general lies in the fact that a variety of parameters are needed to define any single form of feedback, resulting in a vast number of permutations and combinations, all of which cannot be assumed to be equally

impactful. Some of these feedback parameters include type (e.g. visual, auditory, Haptic), schedule (e.g. 50% of trials with feedback, 50% without), timing (e.g. concurrent, bandwidth, terminal), and information type (e.g. knowledge of performance vs. knowledge of results). Even with these parameters specified, the feedback form is not uniquely defined: for example, in the case of STS weight distribution, visual, terminal feedback providing knowledge of results could be communicated as a numerical metric or be depicted as a bar chart. While the bar chart is perhaps a more intuitive representation of weight distribution, the numerical metric might allow the subject to compare between trials more easily. And unfortunately, more is not necessarily better: too much information can cause cognitive overload and impede motor learning; alternatively, too much information could lead to feedback dependence, impeding the transferability of the "learned" motor skills. Thus, research into the relative effectiveness of different feedback forms would not only be beneficial for improving stroke victims' movement dynamics and body awareness, but also for improving the efficiency of rehabilitation programs themselves.

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Appendix

CONTENTS

Appendix A

RECRUITMENT, SCREENING, & CONSENT FORMS

CONTENTS

A.1 Recruitment & Screening

To recruit participants for our study, advertisements were posted in community centers, included in newsletters, and distributed to senior centers, retirement communities, and private rehabilitation clinics. Two versions of the flier were used, one directed towards controls (Figure A-1) and one directed towards stroke survivors (Figure A-2). Those interested in participating were asked a series of screening question in order to assess inclusion and exclusion criteria (Figure A-3). These questions were asked over the phone or in the form of a private online survey.

November, 2013

Collaborative Advanced Robotics
and Intelligent Systems Lab

Department of Mechanical Engineering
University of British Columbia
2054-6250 Applied Science Ln
Vancouver, BC, V6T 1Z4

Interested in technology? Like helping people?

→ Seeking volunteers: Assisted Sit-to-Stand Research Study

We are conducting a study on weight distribution during sit-to-stand using our assistive device to improve stroke rehabilitation. We are seeking **healthy volunteers** as control subjects who meet the following criteria:

- Able to **rise from a chair** repeatedly and **without using hands**
- Have **never had a stroke** or other significant brain injury
- Able to **walk without a walker or cane**

Participants will be asked to come to the UBC Point Grey campus and perform a series of chair-rise motions while being assisted by our device. The study will take approximately 1.5-2 hours. Volunteers will be required to complete a consent form before participating. Participants will receive compensation for transportation expenses, as well as a small gift as a token of our appreciation.

For more information, or to volunteer for this study, please contact:
Jenny Sullivan
< omit >

Thank you for your interest in our research.

Jenny Sullivan | Graduate Student
Drs. Elizabeth Croft, Antony Hodgson, Mike Van der Loos | Professors
Department of Mechanical Engineering, UBC

Figure A-1: Recruitment flier for controls

Collaborative Advanced Robotics and Intelligent Systems Lab

Department of Mechanical Engineering
University of British Columbia
2054-6250 Applied Science Ln
Vancouver, BC, V6T 1Z4

Recovering from a stroke?

→ Seeking volunteers: Assisted Sit-to-Stand Research Study

We are conducting a study on weight distribution during sit-to-stand using our assistive device. We are seeking volunteers who are **recovering from a stroke** and who meet the following criteria:

- Able to **rise from a chair** repeatedly and **without using hands**
- Have **one leg that is noticeably stronger** than the other
- Most recent stroke more than **6 months ago**

Participants will be asked to come to the UBC Point Grey campus and perform a series of chair-rise motions while being assisted by our device. The study will take approximately 1.5-2 hours. Volunteers will be required to complete a consent form before participating. Participants will receive compensation for transportation expenses, as well as a small gift as a token of our appreciation.

For more information, or to volunteer for this study, please contact:

Jenny Sullivan
< omit >

Thank you for your interest in our research.

Jenny Sullivan | Graduate Student
Drs. Elizabeth Croft, Antony Hodgson, Mike Van der Loos | Professors
Department of Mechanical Engineering, UBC

Figure A-2: Recruitment flier for stroke

---- Initial screening questions ----

Inclusion criteria

1) Age: 2) How did you hear about the study?

2) What is your understanding of the study?

3) Have you had a stroke?

- If yes, in the last 6 months?

4) Do you normally use a cane, walker or other aid for walking?

- If yes, are you able to walk without it?

5) Can you get up out of a chair without using your hands (unassisted)?

- If yes, can you do so repeatedly?

6) Is one leg stronger than the other?

Exclusion criteria

8) Do you have any balance disorders (that started before your stroke)?

9) Do you have any neurological disorders (aside from stroke) e.g. Parkinson's disease?

10) Do you have any joint pain in your knees or hips?

11) Have you had any recent surgeries?

12) Have you had any recent leg, hip, or back injuries?

Other

13) What transportation method would you use to get to campus? Home address?

Height: Assess: Informed consent? [ ] English? [ ]

Scheduling: Send consent form to:

Screening questions
Version 1: June 3, 2013

Figure A-3: Screening questions used to assess inclusion/exclusion criteria

A.2 Consent Form

In order to participate in the experiment, subjects were required to sign the following consent form. The form was sent to them in advance of their scheduled experiment time and then reviewed with them in person.

THE UNIVERSITY OF BRITISH COLUMBIA

CARIS
Collaborative Advanced Robotics and Intelligent Systems Lab
< omit >

Department of Mechanical Engineering
University of British Columbia
2054-6250 Applied Science Ln
Vancouver, BC V6T 1Z4
< omit >

Participant Information and Consent Form
Evaluation of Weight Distribution Asymmetry during Assisted Sit-to-Stand

Principal Investigator:

Dr. Elizabeth A. Croft, Professor, Dept. of Mechanical Engineering, UBC
Phone: < omit >

Co-investigators:

Jenny Sullivan, Graduate Student, Dept. of Mechanical Engineering, UBC
Phone: **< omit >**
Email: **< omit >**

Dr. Mike Van der Loos, Professor, Dept. of Mechanical Engineering, UBC
Phone: < omit >

Dr. Antony Hodgson, Professor, Dept. of Mechanical Engineering, UBC
Phone: < omit >

Funding for this project is provided by the Natural Sciences and Engineering Council of Canada and the Department of Mechanical Engineering, University of British Columbia.

Figure A-4: Consent form (page 1 of 9)

Clinical participants:

You are being invited to participate in this study to help investigate weight distribution asymmetry (how people distribute their weight between one leg and the other) and perception during assisted sit-to-stand (the motion of rising from a chair). We are seeking adults who have suffered a stroke and meet the following criteria:

- Most recent stroke was more than 6 months ago
- One leg noticeably stronger than the other
- Able to rise repeatedly from a standard chair height without using hands
- Community-dwelling
- Able to walk without a walker, cane, or other aid
- Can understand verbal instructions and communicate in English
- Able to give informed consent

If you have any of the following, you may **not** participate in the study:

- Balance or neurological disorders not related to stroke (e.g. Parkinson's Disease)
- Significant cognitive impairment
- Inability to rise unassisted from a standard chair height repeatedly
- Any conditions, apart from stroke, that cause uneven weight distribution, uneven movement patterns, or discomfort while rising from a chair, such as:
  - Significant joint pain in legs
  - Recent significant leg, hip, or back injury or surgery
  - Long-term leg, hip, or back injury

All participants who choose to voluntarily participate in this study may withdraw from the experiment at any time by contacting Jenny Sullivan at < omit > or < omit >. You do not need to provide a reason for your withdrawal if you do not wish to do so.

Control participants:

You are being invited to participate in this study to help investigate weight distribution asymmetry (how people distribute their weight between one leg and the other) and perception during assisted sit-to-stand (the motion of rising from a chair). We are seeking healthy adults who meet the following criteria:

Figure A-5: Consent form (page 2 of 9)

- Never suffered a stroke or other significant brain injury
- Able to rise repeatedly from a standard chair height without using hands
- Community-dwelling
- Able to walk without a walker, cane, or other aid
- Can understand verbal instructions and communicate in English
- Able to give informed consent

If you have any of the following, you may **not** participate in the study:

- Balance or neurological disorders
- Significant cognitive impairment
- Inability to rise from a standard chair height unassisted
- Any conditions that cause uneven weight distribution, uneven movement patterns, or discomfort while rising from a chair, such as:
  - Significant joint pain in legs
  - Recent significant leg, hip, or back injury or surgery
  - Long-term leg, hip, or back injury

All participants who choose to voluntarily participate in this study may withdraw from the experiment at any time by contacting Jenny Sullivan at < omit > or < omit >. You do not need to provide a reason for your withdrawal if you do not wish to do so.

Purpose:

We are investigating how people distribute their weight between their legs after having suffered a stroke. We are particularly interested in the motion of standing up from a chair, known as "sit-to-stand," because it is important for mobility and independent living. The purpose of this study in particular is twofold: 1) to evaluate how an assistive device affects weight distribution asymmetry (how much weight is on one leg compared to the other leg) during the sit-to-stand (STS) movement, and 2) to see if that weight distribution asymmetry changes depending on how much assistance the device is providing to the user. The assistive device (or "test bed") consists of a wooden seat that rotates upward to help the user stand up by pushing from under the buttocks. Participants will be asked to perform a series of STS repetitions, some assisted and some unassisted, during which the forces under each foot will be measured, as well as movement of the trunk and thigh.

Figure A-6: Consent form (page 3 of 9)

Results from this study will be used to direct the design of a future study on how to improve weight distribution asymmetry during STS in stroke survivors. Results from this study may also provide insights into stroke rehabilitation in general.

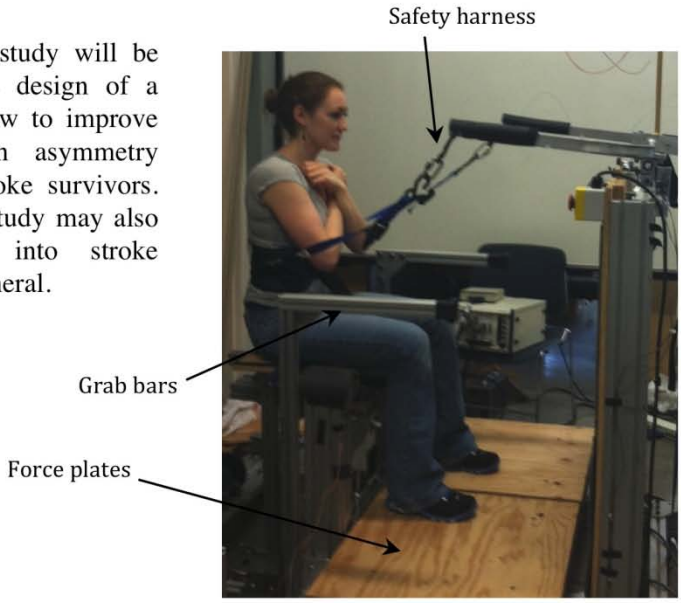

Figure 1: STS test bed

Study Procedures:

If you decide to participate, you will be requested to come to the Collaborative Advanced Robotics and Intelligent Systems (CARIS) laboratory, Room X015, ICICS building, 2366 Main Mall, at the University of British Columbia. **You will be asked to partake in a set of experiments that will take approximately 1.5-2 hours.** The study procedures are as follows:

Clinical Assessment

You will be asked a handful of basic questions about your health. You are not required to provide any information you don't feel comfortable providing. You will then be asked to do the Berg Balance Assessment, which consists of a number of functional tasks such as walking, turning, standing up from and sitting down on a chair. Afterwards, the investigator will perform manual muscle testing (straightening the leg, bending the leg, flexing the foot, and pointing the foot) on both legs to assess strength and range of motion.

Figure A-7: Consent form (page 4 of 9)

Equipment Setup

Your weight will be measured as well as the lengths of individual body segments. You will then be fitted with motion sensors that will be attached using velcro straps to your thigh and chest. You will be instructed on safety features of the assist device and protocol for device use.

Trial Blocks

You will then be asked to perform 6 blocks of STS trials. All trials will be performed on the test bed force plates, although some trials will not use the actual assist mechanism. During all trials, you will be asked to distribute your weight as evenly as possible between your right and left legs. After each trial, the experimenter will ask how evenly you felt your weight was distributed. **You may rest as often as you need to between trials.** Some trials may be video recorded or photographed.

Block 1 - Unassisted trials: you will be asked to perform 3 unassisted STS trials without using your hands
Block 2 – Isometric (stationary) maximum strength trials: your thighs will be strapped to the seat so that you remain stationary in the seated position. You will then be asked to push down onto the force plates with your feet as hard as you can for 5 seconds. You will be asked to perform 2 trials.
Blocks 3, 4, 5 - Assisted trials: 3 blocks of trials will use the assist mechanism of the test bed. In each block, you will be asked to support a certain percentage of your total body weight while rising, and to rely on the assist to do the rest of the work. The experimenter will guide you through these trials, and you will have the opportunity to do practice trials at the beginning of each block to familiarize yourself.
Block 6 - Unassisted trials: 2 final unassisted trials will be performed in the same way as in block 1.

Risks:

There are no major risks involved in participating in this study. The only minor risk is the possibility of losing balance while performing the assisted sit to stand motion, which we have mitigated through the following means: the range of motion of the seat assist will be constrained by limit switches so that you will not

Figure A-8: Consent form (page 5 of 9)

be placed into a position that extends beyond your natural range of motion. You will also wear a waist harness attached to supports, and support bars have been placed in the test bed to grab onto if necessary. The test bed can be stopped at any time by an experimenter-controlled emergency stop button. You will have control of the motion of the assist using a handheld joystick button, which only allows the assist to move when you are pressing it down. In all cases, if you let go of the joystick button before the STS is completed, you will be supported in the current position by the test bed. If the emergency stop button is pressed, you will be slowly lowered to the start position of the assist. With the exception of the maximum strength trials, during which the assist is not moving, you are never physically connected to the assist and therefore remain in control of your own movement.

Benefits and payment for participation in this study:

You will be compensated for your travel expenses based on transportation method and the distance between your home and the UBC campus. We will have this reimbursement ready for you when you come in for the study. However, if you prefer, you can bring your travel receipts when you come for the study, and we will send you a reimbursement for the exact amount in the mail afterwards. In addition to travel compensation, you will also receive a small thank you gift for participating.

Although your participation in this study may potentially lead to improved stroke rehabilitation methods in the future, there are no other direct benefits to you as the participant for taking part.

Confidentiality:

Your confidentiality will be respected. No information that discloses your identity will be released or published without your specific consent to the disclosure. However, research records and medical records identifying you may be inspected in the presence of the Investigator or her designate by representatives of the Natural Sciences and Engineering Research Council of Canada, Health Canada, and the UBC Research Ethics Board for the purpose of monitoring the research. No information or records that disclose your identity will be published without your consent, nor will any information or records that disclose your identity be removed or released without your consent unless required by law.

You will be assigned a unique study number as a subject in this study. Only this number will be used on any research-related information collected about you during the course of this study, so that your identity (i.e. your name or any other

Figure A-9: Consent form (page 6 of 9)

information that could identify you) as a subject in this study will be kept confidential. Information that contains your identity will remain only with the Principal Investigator and/or designate. The list that matches your name to the unique study number that is used on your research-related information will not be removed or released without your consent unless required by law.

Your rights to privacy are legally protected by federal and provincial laws that require safeguards to ensure that your privacy is respected and also give you the right of access to the information about you that has been provided to the sponsor and, if need be, an opportunity to correct any errors in this information. Further details about these laws are available on request to your study doctor.

After the study is finished:

Data and Results

Any personal information obtained from this research study will be kept confidential. All documents will be identified only by a code number and kept in a locked filing cabinet in the principal investigator's research office. Digital data including photographs and video records (if applicable) will be stored on password-protected computers that are accessible only by the investigators involved in this study. Individual data in reports or scientific publications will be referred to by participant number only.

Photographs and video may be presented at scientific conferences and or published in reports and journals, but identifying features in both video and photographs will be blurred using digital image and video blurring tools, and participants will be identified by participant number only. Access to the photographs and video will be restricted to the investigators involved in this study and used only in relation to the study the data was collected for. Data obtained from this study will be stored for at least 5 years after the dissemination of results, after which the data will be destroyed to assure anonymity.

When all data has been analyzed and results are available for dissemination, we would be happy to share them with you. If you would like us to contact you when these results are available, please check the appropriate box on page 9.

Future studies

We will eventually be conducting other studies for this project that require volunteers as well. If you would be interested in receiving more information about these future studies, please check the appropriate box on page 9.

Figure A-10: Consent form (page 7 of 9)

If something goes wrong:

Signing this consent form in no way limits your legal rights against the sponsor, investigators, or anyone else, and you do not release the study doctors or participating institutions from their legal and professional responsibilities.

The risk involved in participating in this study is very low, and measures have been taken to mitigate the effects of any potential harms. However, should something go wrong and you would like us to contact your doctor, please provide his/her information on page 9 (optional).

Contact:

If you have any questions or concerns, or if you would like further information about this study, you may contact Dr. Elizabeth Croft or **Jenny Sullivan < omit >**. We appreciate your time and willingness to participate, and we will do whatever we can to accommodate your needs.

If you have any concerns about your treatment or rights as a research participant you may contact the Research Subject Information Line at the University of British Columbia at (604) 822-8598.

Figure A-11: Consent form (page 8 of 9)

Participant's Consent to Participate

Evaluation of Weight Distribution Asymmetry during Assisted Sit-to-Stand

I understand that my participation in this study is entirely voluntary and that I may refuse to participate or I may withdraw from the study at any time without consequences.

Signing this consent form in no way limits my legal rights against the sponsor, investigators, or anyone else.

I have had sufficient time to consider the information provided and to ask for advice if necessary.

I have been told that I will receive a signed and dated copy of this consent form for my own records.

I consent to participate in this study by signing in the space provided below:

_____

Participant's Signature **Printed name of participant** Date

_____

Experimenter's Signature **Printed name of Experimenter** Date

OPTIONAL INFORMATION:

i) Yes, please contact me:

[ ] when study results are available

[ ] with information about participating in future studies

Phone/email: _____

ii) Doctor's name & phone number: _____

Figure A-12: Consent form (page 9 of 9)

Appendix B

DATA COLLECTION FORMS

CONTENTS

B.1 Basic Demographic and Background Information Questions

Before the clinical assessments or STS trials were performed, basic background information was collected on all subjects. For controls, the dominant leg was taken to be the leg they would use to kick a ball. For stroke subjects, their affected side was their paretic side. The majority of this information did not ultimately get factored into the analysis, as the level of detail (or lack thereof) was inconsistent among subjects. For example, some stroke subjects knew whether their stroke was due to a clot or a bleed, but rarely knew anything more. All subjects were community-dwelling, living either on their own or with a spouse. Some had a caretaker visit a few times a week. All subjects were also independently mobile – some could drive, but many relied on public transportation. Many of the stroke subjects used a cane or walker outside of their homes as a safety precaution. The "other medical conditions" of interest were those that might

cause asymmetry or discomfort during STS trial (e.g. arthritis, as opposed to diabetes or high cholesterol).

-------- BASIC INFO --------

Date & time: _____ Subject number: _____

Affected/dominant side: R L Time post-stroke: _____ Age: _____

Rehab program? _____

_____

Knowledge of stroke: _____

_____

Independence level (living situation, mobility): _____

_____

Other medical conditions: _____

Figure B-1: Basic background and demographic information questions

B.2 Berg Balance Scale Assessment

The first clinical assessment performed was the Berg Balance Scale (K. O. Berg et al., 1992; K. Berg, Wood-Dauphinee, Williams, & Gayton, 1989), described in Chapter 3. The 14 tasks involved in the assessment and the scoring criteria are presented below.

---

1) STS
"When I say go, stand up from the chair, and try not to use your hands for support"

4 – no hands
3 – hands
2 – hands, several tries
1 – minimal aid (stand/stabilize)
0 – moderate/maximal aid

2) Standing unsupported – 2 mins
"Stand still for 2 mins, stare straight ahead, keep feet on tape lines"

4* – 2 mins, safely → *skip #3
3 – 2 mins with supervision
2 – 30 secs, unsupported
1 – 30 secs, several tries
0 – < 30 secs

3) Sitting with back unsupported – 2 mins
"Sit with your arms folded or out in front of you. Try not to touch back to chair"

4 – 2 mins, safely & securely
3 – 2 mins, with supervision
2 – 30 secs
1 – 10 secs
0 – < 10 secs

4) StTS (standing to sitting)
"When I say go, sit down in chair as you normally do"

4 – controls descent, minimal hands, safely
3 – controls descent with hands
2 – controls descent with backs of legs
1 – uncontrolled descent
0 – needs assistance

5) Transfers
"Sit down in chair. When I say go, move over to other chair, and then back to this chair."

4 – independent transfer, minor hands, safely
3 – independent transfer, definite need of hands, safely
2 – independent transfer, needs supervision/verbal cueing
1 – needs 1 person to assist
0 – needs 2 people to assist

6) Standing, eyes closed – 10 secs
"When I say go, close eyes and stand still, keeping feet in same place"

4 – 10 secs, eyes closed, safely
3 – 10 secs, eyes closed, with supervision
2 – 3 secs, eyes closed
1 – eyes open after 3 secs, keeps balance
0 – needs help to avoid falling

7) Standing, feet together – 1 min
"When I say go, place feet together, and stand still for 1 min staring straight ahead"

4 – feet together independently, 1 min, safely
3 – feet together independently, 1 min, with supervision
2 – feet together independently, < 30 secs
1 – help putting feet together, > 15 secs
0 – help putting feet together, < 15 secs

8) Reaching forward
"Stand here, arms stretched out. When I say go, reach as far forward as you can and come back. Try to keep hands along line"

4 – 10" confidently
3 – 5"
2 – 2"
1 – reaches but needs supervision
0 – needs assistance/loses balance

9) Pick up object
"When I say go, pick up the foam from the floor"

4 – pick up safely, easily
3 – pick up with supervision
2 – unable to pick up, keeps balance within 1-2" of object
1 – unable to pick up, needs supervision
0 – won't try/needs assistance

10) Look over shoulders
"Stand here with feet on tape. When I say go, turn around to look at [ ] behind you, then turn in other direction and look at [ ]. Keep feet still whole time."

4 – all the way, both sides, shifts weight
3 – all the way, 1 side
2 – part way, both sides
1 – needs supervision
0 – needs assistance to not lose balance

11) Turn 360° - (4 secs)
"Turn around in a circle, pause, then turn back around in other direction"

4 – < 4 secs, both directions
3 – < 4 secs, one direction
2 – > 4 secs, both directions
1 – needs supervision/verbal cueing
0 – needs assistance

12) Step touch – (8x, 20 secs)
"When I say go, touch top of box with one foot, then other. Alternate until you've done 8 in total. Make sure foot touches red tape"

4 – 8x, < 20 secs
3 – 8x, > 20 secs
2 – 4x, no help, with supervision
1 – > 2x, minimal help
0 – needs help to avoid falling/won't try

13) Standing, one foot in front – 30 secs
"If you can, put one foot in front of the other on the line, and stand still for 30 secs"

4 – feet in line independently, hold 30 secs
3 – foot ahead independently, hold 30 secs
2 – small step independently, hold 30 secs
1 – help to step, hold 15 secs
0 – loses balance while stepping/standing

14) One leg stand – 10 secs
"When I say go, stand on one leg and balance for 10 seconds (or as long as you can)"

4 – lifts leg independently, holds >10 secs
3 – lifts leg independently, holds 5-10 secs
2 – lifts leg independently, holds > 3 secs
1 – can't hold leg independently 3 secs, keeps balance
0 – won't try/needs help to avoid falling

B.3 Manual Muscle Testing

As described in Chapter 3, the second clinical assessment performed was Manual Muscle Testing, which was done on the knee extensors, ankle dorsiflexors, and ankle plantarflexors. The scoring criteria for each movement are presented below.

Knee Extensors

5 – Subject able to maintain leg extension against external flexion moment
4 – Subject somewhat able to maintain leg extension against external flexion moment
3 – Subject able to extend leg through full range of motion against gravity only
2 – Subject able to extend leg though partial range of motion against gravity only
1 – Contraction visible when subject attempts to extend leg
0 – No movement visible when subject attempts to extend leg

Ankle Dorsiflexors

5 – Subject able to maintain foot flexion against external extension (plantarflexion) moment
4 – Subject partially able to maintain foot flexion against external extension (plantarflexion) moment
3 – Subject able to flex foot through full range of motion against gravity only
2 – Subject able to flex foot though full range of motion with leg elevated
1 – Subject able to flex foot though partial range of motion with leg elevated
0 – No movement visible when subject attempts to flex foot

Ankle Plantarflexors

5 – Subject able to perform ≥ 20 reps of single-leg heel raises
4 – Subject able to perform 10-19 reps of single-leg heel raises
3 – Subject able to perform 1-9 reps of single-leg heel raises, OR
Subject able to extend (plantarflex) foot against external resistance with leg elevated
2 – Subject able to extend (plantarflex) foot through full range of motion with leg elevated
1 – Subject able to extend (plantarflex) foot through partial range of motion with leg elevated
0 – No movement visible when subject attempts to extend (plantarflex) foot with leg elevated

---

Appendix C

INDIVIDUAL SUBJECT DATA

CONTENTS

C.1 Individual Metrics: Clinical, STS Summary, & STS Regression

Table C-1 shows demographic information and clinical scores for individual subjects, broken down by group, as well as group averages and standard deviations. The difference in MMT scores between sides was also calculated for each subject. The Berg Balance Scale and MMT were scored using whole numbers only. Time post-stroke was rounded to the nearest year. Group averages and standard deviations were rounded to two decimal places.

Table C-2 shows unassisted and assisted STS summary metrics (Average Body Weight Distribution, Variability, Average Asymmetry, and Average Deviation) for individual subjects, as well as group averages and standard deviations. In addition, the ranges (RNG) of weight distribution data points and load ratio values were included. Values were rounded to three decimal places.

Table C-3 shows STS regression metrics for individual subjects, as well as group averages and standard deviations. DLR Slope and Dev Slope values were calculated from assisted trials in which the subject perceived their weight distribution to be symmetric. These metrics are identical when weight distribution values were either greater than 50% for all trials or less than 50% for all trials. DLR Slope Magnitude is the absolute value of the DLR Slope, and is indicative of the magnitude of the relationship between weight distribution and load ratio.

SUBJECT DEMOGRAPHIC INFO & CLINICAL ASSESSMENT SCORES

Subject	ND side	Age	Gender	Yrs post-stroke	Berg	ND KE	D KE	ND DF	D DF	ND PF	D PF	KE diff	DF diff	PF diff
							Control							
1	L	49	F	-	56	5	5	5	5	5	5	0	0	0
2	L	33	M	-	56	5	5	5	5	4	5	0	0	1
3	L	77	F	-	56	5	5	5	5	4	4	0	0	0
4	L	50	F	-	56	5	5	5	5	5	5	0	0	0
5	L	63	M	-	56	5	5	5	5	4	4	0	0	0
15	L	73	F	-	54	4	4	4	4	5	5	0	0	0
17	L	80	M	-	56	5	5	5	5	5	5	0	0	0
18	L	70	F	-	56	5	5	5	5	5	5	0	0	0
Avg:	**-**	**61.9**	**-**	**-**	**55.75**	**4.88**	**4.88**	**4.88**	**4.88**	**4.63**	**4.75**	**0.00**	**0.00**	**0.13**
SD:	**-**	**16.4**	**-**	**-**	**0.71**	**0.35**	**0.35**	**0.35**	**0.35**	**0.52**	**0.46**	**0.00**	**0.00**	**0.35**
							Stroke							
6	L	84	M	5	46	4	5	4	4	3	3	1	0	0
7	R	61	M	13	46	4	5	0	5	0	3	1	5	3
8	R	56	F	12	51	4	5	1	4	1	4	1	3	3
9	R	75	M	1	56	4	5	4	5	5	5	1	1	0
10	R	81	F	4	46	4	5	4	5	3	5	1	1	2
13	R	54	M	8	56	5	5	5	5	4	4	0	0	0
14	L	56	M	1	53	5	5	4	4	3	5	0	0	2
16	L	65	M	2	53	5	5	5	5	3	3	0	0	0
Avg:	**-**	**66.5**	**-**	**5.75**	**50.88**	**4.38**	**5.00**	**3.38**	**4.63**	**2.75**	**4.00**	**0.63**	**1.25**	**1.25**
SD:	**-**	**11.9**	**-**	**4.77**	**4.36**	**0.52**	**0.00**	**1.85**	**0.52**	**1.58**	**0.93**	**0.52**	**1.83**	**1.39**

Table C-1: Background information and clinical assessment scores for individual subjects with group averages and standard deviations. Max Berg score = 56, max MMT score (each movement) = 5; Diff = D – ND.

SUBJECT STS SUMMARY METRICS

Subject	BWD_U	BWD_{MVC}	BWD_A	VR_U	VR_A	RNG_U	RNG_A	Dev_U	Dev_A	$Asymm_U$	$Asymm_A$	RNG_{r_load}
						Control						
1	0.526	0.537	0.533	0.046	0.055	0.053	0.094	0.027	0.036	0.026	0.033	0.342
2	0.523	0.444	0.577	0.045	0.056	0.060	0.105	0.024	0.077	0.023	0.077	0.441
3	0.541	0.528	0.566	0.018	0.056	0.028	0.102	0.041	0.066	0.041	0.066	0.341
4	0.456	0.435	0.484	0.010	0.058	0.010	0.099	0.044	0.028	0.044	0.016	0.327
5	0.520	0.477	0.578	0.032	0.051	0.039	0.094	0.020	0.078	0.020	0.078	0.291
15	0.501	0.560	0.570	0.045	0.044	0.045	0.072	0.020	0.070	0.001	0.070	0.455
17	0.479	0.488	0.456	0.028	0.051	0.034	0.103	0.021	0.046	0.021	0.044	0.247
18	0.491	0.553	0.495	0.027	0.058	0.033	0.092	0.012	0.024	0.009	0.005	0.307
Avg:	**0.504**	**0.503**	**0.532**	**0.031**	**0.054**	**0.038**	**0.095**	**0.026**	**0.053**	**0.023**	**0.049**	**0.344**
SD:	**0.028**	**0.049**	**0.048**	**0.013**	**0.005**	**0.016**	**0.011**	**0.011**	**0.022**	**0.014**	**0.028**	**0.071**
						Stroke						
6	0.550	0.453	0.492	0.016	0.068	0.016	0.113	0.050	0.029	0.050	0.008	0.396
7	0.360	0.484	0.356	0.069	0.083	0.084	0.156	0.140	0.144	0.140	0.144	0.262
8	0.480	0.480	0.494	0.075	0.047	0.097	0.093	0.026	0.020	0.020	0.006	0.263
9	0.475	0.525	0.473	0.023	0.048	0.030	0.075	0.025	0.027	0.025	0.027	0.299
10	0.405	0.399	0.396	0.064	0.084	0.087	0.118	0.095	0.104	0.095	0.104	0.290
13	0.500	0.462	0.493	0.025	0.046	0.035	0.072	0.008	0.020	0.000	0.007	0.411
14	0.468	0.536	0.507	0.035	0.028	0.041	0.047	0.032	0.013	0.032	0.007	0.344
16	0.476	0.468	0.513	0.055	0.057	0.062	0.087	0.028	0.027	0.024	0.013	0.282
Avg:	**0.464**	**0.476**	**0.466**	**0.045**	**0.058**	**0.057**	**0.095**	**0.051**	**0.048**	**0.048**	**0.040**	**0.318**
SD:	**0.058**	**0.043**	**0.058**	**0.023**	**0.020**	**0.030**	**0.034**	**0.045**	**0.048**	**0.047**	**0.054**	**0.059**

Table C-2: Summary metrics calculated from STS trials for individual subjects with group averages and standard deviations.

SUBJECT STS REGRESSION METRICS & AVERAGES

Subject	DLR Slope	Dev Slope	DLR Slope Magnitude
		Control	
1	0.202	0.174	0.202
2	-0.151	-0.151	0.151
3	-0.002	-0.002	0.002
4	0.041	-0.022	0.041
5	-0.083	-0.083	0.083
15	-0.063	-0.063	0.063
17	-0.024	0.004	0.024
18	0.060	-0.038	0.060
Avg:	**-0.003**	**-0.023**	**0.078**
SD:	**0.107**	**0.094**	**0.067**
		Stroke	
6	0.193	-0.011	0.193
7	0.396	-0.396	0.396
8	0.027	-0.049	0.027
9	0.161	-0.161	0.161
10	0.184	-0.184	0.184
13	0.114	-0.041	0.114
14	-0.045	-0.045	0.045
16	-0.183	-0.073	0.183
Avg:	**0.106**	**-0.120**	**0.163**
SD:	**0.175**	**0.127**	**0.114**

Table C-3: STS regression metrics calculated for individual subjects with group averages and standard deviations.

C.2 Individual Subject Data Plots

The following figures present plots of individual STS trials for each subject. Figure. C-1 & Figure C-2 show control subjects, and Figure C-3 & Figure C-4 show stroke subjects. All data points correspond to weight distribution values ($F_{ND}^{\%}$). Assisted trials are plotted with respect to corresponding r_{load} values, calculated as described in Chapter 3. Unassisted trials are plotted at a load ratio value of 1, as a load ratio of 100% indicates no help from the assist. Load ratio values for MVC trials were calculated according to the following formula:

$$r_{load,MVC} = \frac{\left(F_D + F_{ND}\right)_{MVC}}{F_{ref}\left(\theta_{SO}\right)} \qquad \text{(C-1)}$$

that is, the total force exerted during the MVC trial normalized to peak force exerted during the unassisted STS trial. As the MVC trials were performed isometrically at a thigh angle approximately equal to θ_{SO}, this formulation is analogous to the regular calculation of the load ratio.

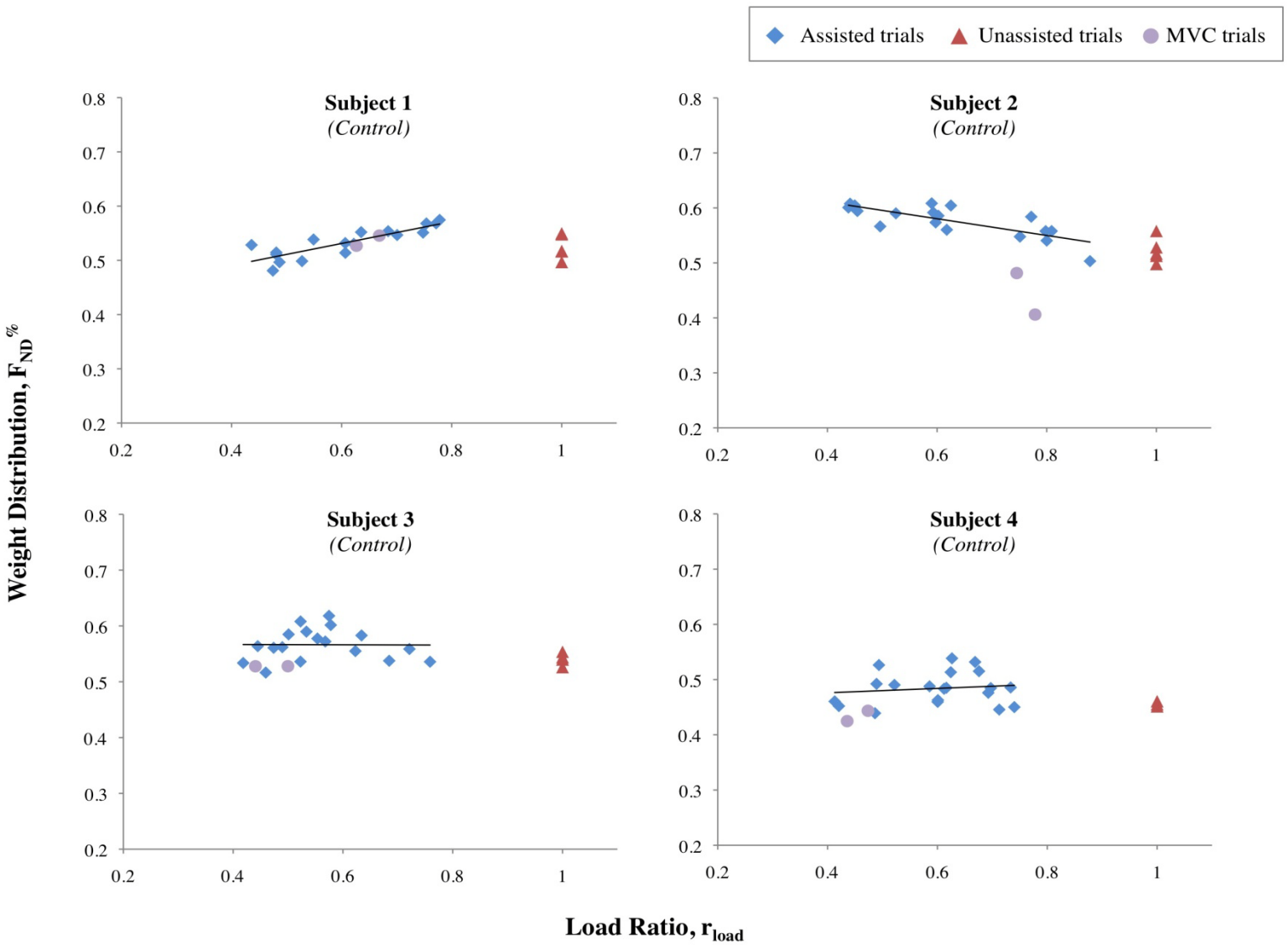

Figure. C-1: Individual trial data plots for Subjects 1-4 (control group).

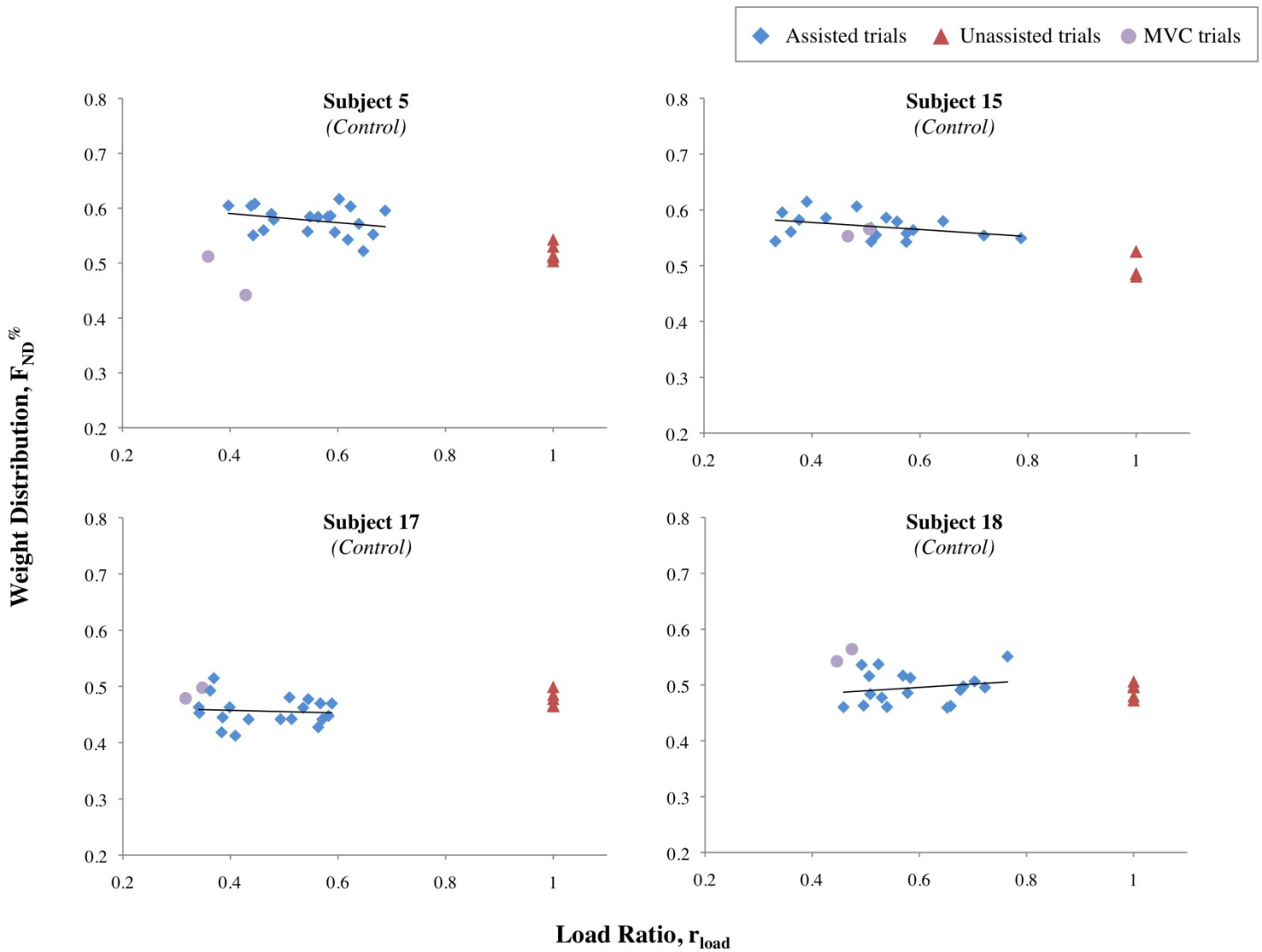

Figure C-2: Individual trial data plots for Subjects 5, 15, 17, & 18 (control group).

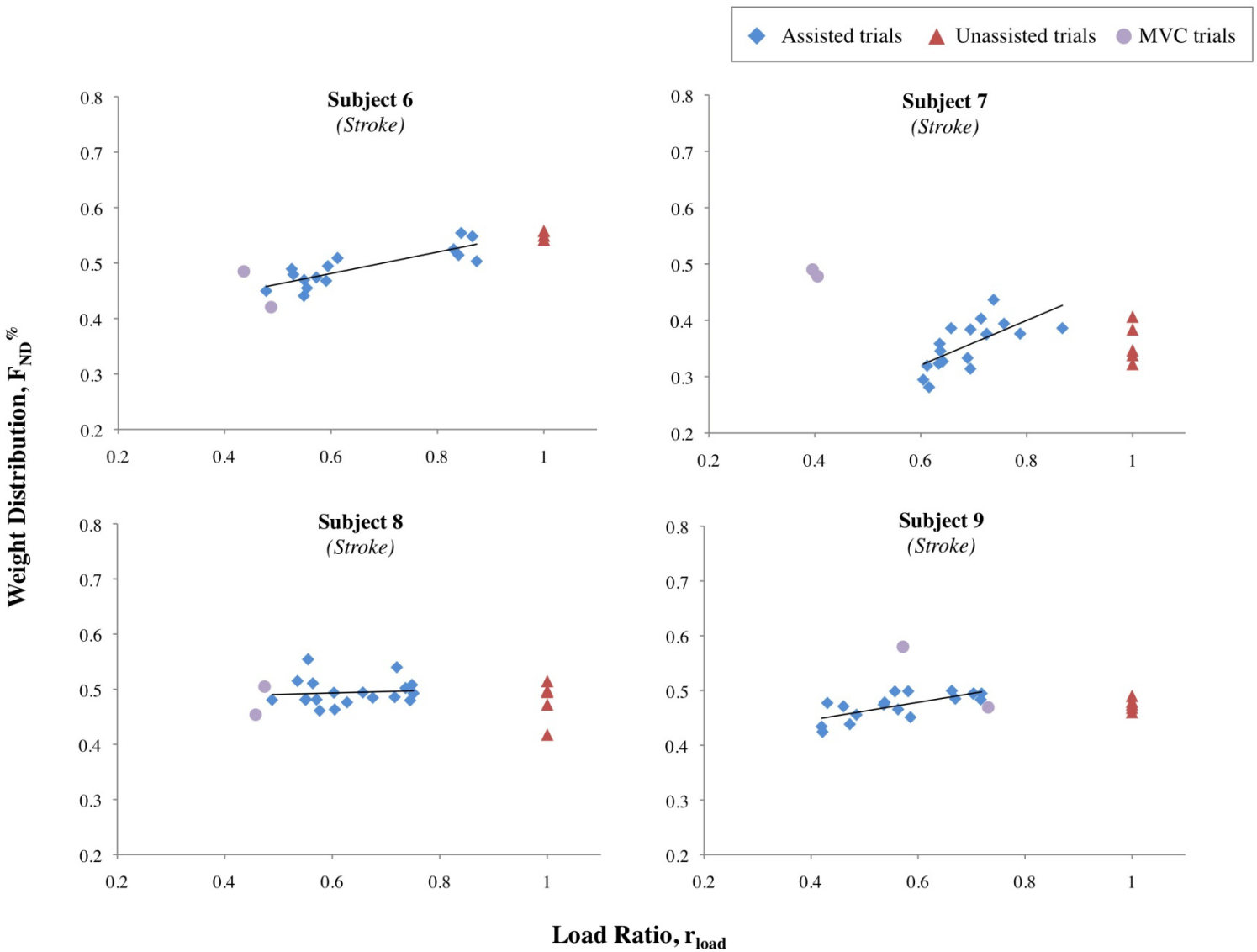

Figure C-3: Individual trial data plots for Subjects 6-9 (stroke group).

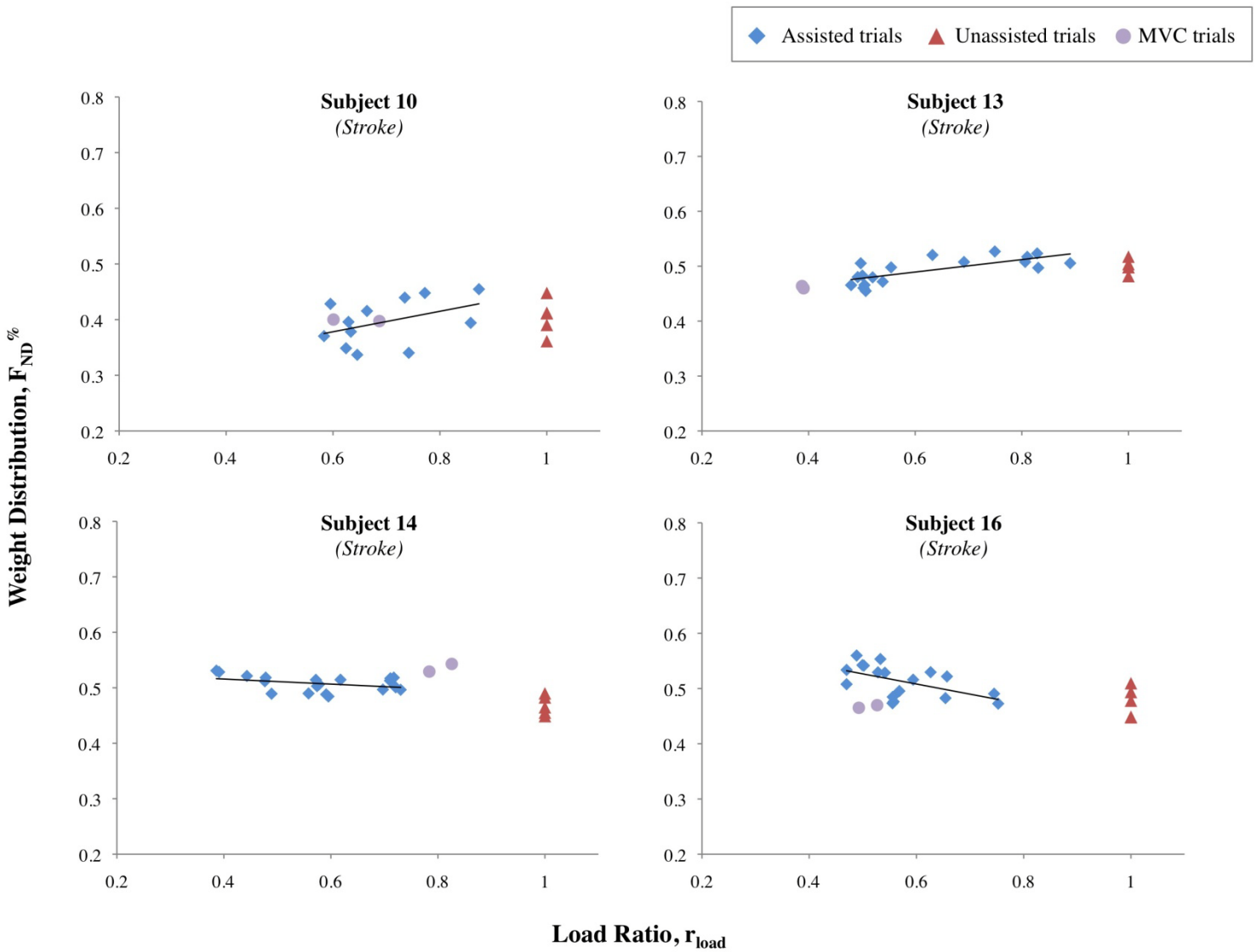

Figure C-4: Individual trial data plots for Subjects 10, 13, 14, & 16 (stroke group).

Appendix D

MODELS & REGRESSION STATISTICS

CONTENTS

D.1 DLR Regression Line Fits & Statistics

Initially, the relationship between weight distribution and load ratio was assessed overall by fitting the fixed-slope model to each group's data, and then performing statistical tests to determine whether each slope value was significantly different from 0 and/or bounded within the respective Equivalence Bounds. When these results suggested neither a clear acceptance nor clear rejection of the null hypothesis, the same analysis procedure was done for individual subjects, using their DLR Slopes and comparing those values to 0 and to upper and lower Equivalence Bounds (also calculated individually). The results of those statistical tests, as well as regression fit data for the DLR Slopes, are shown for each subject in Table D-1. For the DLR Slope to be significantly different from 0, the corresponding *p*-value had to be less than 0.05 (H_A: DLR Slope $\neq 0$); for the DLR Slope to be statistically bounded within the subject's

Equivalence Bounds, the *p*-values at *both* the upper and lower EB had to be less than 0.05 (H_A: -EB < DLR Slope < +EB).

SUBJECT DLR REGRESSION FITS & STATISTICS

	Linear Fit		*DLR Slope*			*Equivalence Bounds*		
Subject	**Intercept**	**(DLR) Slope**	**St. Error**	**SD Resid.**	***p*-val, 0**	**Equiv. Bound**	***p*-val, +EB**	***p*-val, -EB**
				Control				
1	0.410	0.202	0.031	0.014	0.000	±0.115	0.993	0.000
2	0.671	-0.151	0.031	0.018	0.000	±0.082	0.000	0.980
3	0.567	-0.002	0.074	0.028	0.982	±0.045	0.265	0.283
4	0.459	0.041	0.068	0.029	0.557	±0.025	0.591	0.173
5	0.623	-0.083	0.066	0.024	0.222	±0.090	0.009	0.459
15	0.603	-0.063	0.040	0.020	0.131	±0.085	0.001	0.290
17	0.467	-0.024	0.068	0.025	0.730	±0.094	0.049	0.157
18	0.460	0.060	0.077	0.028	0.445	±0.073	0.434	0.051
				Stroke				
6	0.366	0.193	0.035	0.019	0.000	±0.039	1.000	0.000
7	0.083	0.396	0.111	0.031	0.003	±0.216	0.937	0.000
8	0.477	0.027	0.067	0.024	0.696	±0.237	0.003	0.001
9	0.381	0.161	0.038	0.016	0.001	±0.061	0.990	0.000
10	0.268	0.184	0.121	0.038	0.159	±0.248	0.305	0.003
13	0.421	0.114	0.026	0.016	0.001	±0.050	0.987	0.000
14	0.534	-0.045	0.028	0.013	0.121	±0.083	0.000	0.093
16	0.618	-0.183	0.071	0.024	0.020	±0.174	0.000	0.550

Table D-1: Individual subject regression ($F_{ND}^{\%}$ vs. r_{load}) fit characteristics. Linear Fit (*left*): intercept and (DLR) slope values corresponding to linear regression fit. DLR Slope (*middle*): standard error of DLR Slope value and standard deviation of residuals. *P*-value < 0.05 indicates DLR Slope value is significantly different from 0. Equivalence Bounds (*right*): subject's Equivalence Bound (EB) values, with *p*-values corresponding to one-sided t-tests at each of the upper (+) and lower (-) bounds.

D.2 Comparison of Model DLR Slopes with Individual Regression Fits

Table D-2 shows a comparison of all slopes calculated for each subject: the DLR Slope, fit to each individual's data; the slope from the fixed-slope model, which was the same value within each subject group; and the slope from the interaction model, which was calculated from group data but dependent on the individual's Average Assisted Weight Distribution (BWD_A). Although the interaction model was a statistically better fit than the fixed-slope model, for the majority of subjects, the difference in slope values was small.

WITHIN-SUBJECT DLR SLOPE COMPARISON

Subject	DLR Slope	Fixed-slope Model	Interaction Model
		Control	
1	0.202	-0.015	-0.012
2	-0.151	-0.015	-0.055
3	-0.002	-0.015	-0.044
4	0.041	-0.015	0.035
5	-0.083	-0.015	-0.056
15	-0.063	-0.015	-0.048
17	-0.024	-0.015	0.062
18	0.060	-0.015	0.025
		Stroke	
6	0.193	0.098	0.068
7	0.396	0.098	0.357
8	0.027	0.098	0.064
9	0.161	0.098	0.108
10	0.184	0.098	0.272
13	0.114	0.098	0.066
14	-0.045	0.098	0.036
16	-0.183	0.098	0.023

Table D-2: Comparison of slope values for individual subjects.

Appendix E

RELATIONSHIPS AMONG METRICS

CONTENTS

E.1 Within-Subject Comparison of Dev vs. Asymm

As described in Chapter 3, a subject's average deviation (Dev) and average asymmetry (Asymm) would only be different values if their weight distribution were greater than 50% on some trials and less than 50% on other trials. As this turned out to be an infrequent occurrence, Dev and Asymm values were the same or very similar for nearly all subjects. Consequently, these metrics were lumped together ($Dev_U/Asymm_U$ and $Dev_A/Asymm_A$) when presenting and interpreting the correlation results. To support this decision, a brief statistical analysis showing the within-subject similarity of these metrics. First, a one-sided paired t-test was done to show that the within-subject difference between the metrics was significantly less than 1%. In addition,

a linear correlation was performed to show not only that a significant relationship exists between the metrics, but also that the slope of the regression line is close to unity. Subjects were pooled into a single group. The following presents the results of these statistical tests (coded in R). Meaningful values are bolded.

UNASSISTED

Paired t-test

Data: Dev_U and $Asymm_U$

H_0: true difference in means is greater than or equal to 0.01

H_A: true difference in means is less than 0.01

Mean of the differences: **0.002625**

t = -5.8785, df = 15, p-value = **1.516e-05**

95 percent confidence interval: (-Inf, 0.00482)

Linear Correlation

y = $Asymm_U$

x = Dev_U

```
Call: lm(formula = y ~ x)

Residuals:
     Min          1Q          Median       3Q          Max
-0.015508   -0.001212      0.002116     0.002604     0.003492

Coefficients:
                Estimate      Std. Error    t value      Pr(>|t|)
(Intercept)     -0.004438     0.001896      -2.341       0.0345 *
x               1.047322      0.037632      27.831       1.17e-13 ***
---
Signif. codes:  0 '***' 0.001 '**' 0.01 '*' 0.05 '.' 0.1 ' ' 1

Residual standard error: 0.004924 on 14 degrees of freedom
Multiple R-squared:  0.9822,      Adjusted R-squared:  0.981
F-statistic: 774.5 on 1 and 14 DF,  p-value: 1.174e-13
```

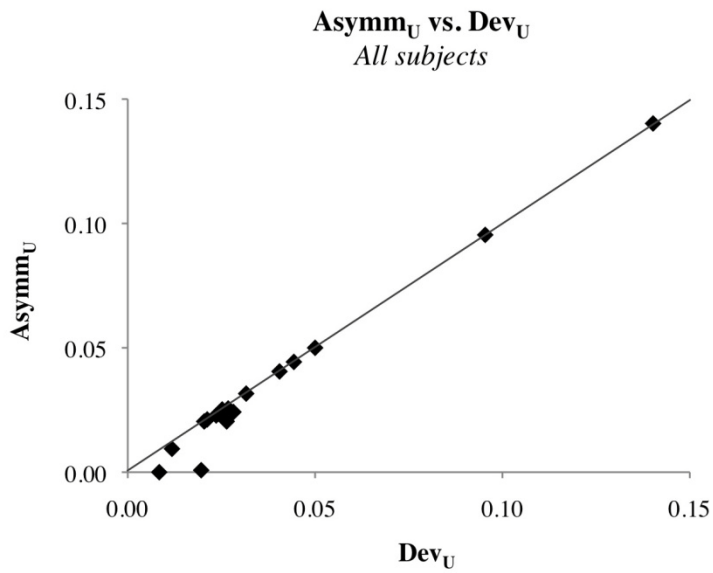

Figure. E-1: Within-subject comparison of Asymm$_U$ vs. Dev$_U$. Identity line (Asymm$_U$ = Dev$_U$) overlaid. Only a few subjects had Dev and Asymm values that were not exactly the same; of those subjects, the largest difference in values was 0.02.

ASSISTED

Paired t-test

Data: Dev$_A$ and Asymm$_A$

H_0: true difference in means is greater than or equal to 0.01

H_A: true difference in means is less than 0.01

Mean of the differences: **0.0065**

t = -1.8278, df = 15, p-value = **0.04377**

95 percent confidence interval: (-Inf, 0.00986)

Linear Correlation

y = Asymm$_A$

x = Dev$_A$

Call: lm(formula = y ~ x)

Residuals:

Min	1Q	Median	3Q	Max
-0.011624	-0.003657	0.001107	0.004608	0.009643

Coefficients:

	Estimate	Std. Error	t value	Pr(>|t|)
(Intercept)	-0.013243	0.002674	-4.953	0.000212 ***
x	**1.133369**	0.043351	26.144	**2.77e-13 *** **

Signif. codes: 0 '***' 0.001 '**' 0.01 '*' 0.05 '.' 0.1 ' ' 1

Residual standard error: 0.006124 on 14 degrees of freedom
Multiple R-squared: 0.9799, Adjusted R-squared: 0.9785
F-statistic: 683.5 on 1 and 14 DF, p-value: 2.773e-13

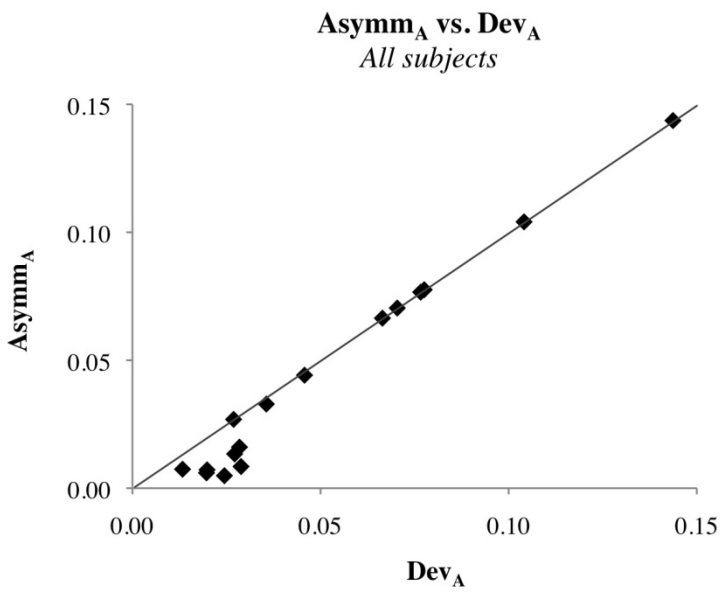

Figure. E-2: Within-subject comparison of Asymm$_A$ vs. Dev$_A$. Identity line (Asymm$_A$ = Dev$_A$) overlaid. Only a few subjects had Dev and Asymm values that were not exactly the same; of those subjects, the largest difference in values was 0.021.

E.2 Correlations

Linear correlations were calculated between all metrics (clinical metrics, STS summary metrics, and STS regression metrics; see Table 3-3). Correlation coefficients (Pearson's ρ) and corresponding *p*-values are listed below, where the significance of the *p*-value reflects the probability of a relationship between the two metrics (H_0: X and Y are independent, $\rho = 0$). The first sub-section lists the metrics with the most significant relationships, and the second sub-section includes correlations for all metrics.

E.2.1 Correlations of Highest Statistical Significance

The correlation values are separated into three tables according to *p*-value: $p < 0.01$, $p < 0.05$, and $p < 0.10$. Within each table, correlations are listed in order of decreasing significance.

SIGNIFICANT CORRELATIONS ($P < 0.01$)

Y	X	*p*-value	ρ
Dev_U	Berg	0.0004	-0.7800
$Asymm_U$	Berg	0.0007	-0.7540
VR_U	PF diff	0.0007	0.7560
BWD_U	Dev slope	0.0014	0.7270
Dev_U	DF diff	0.0016	0.7210
BWD_A	DLR slope	0.0016	-0.7210
Dev_U	DLR slope mag	0.0022	0.7080
BWD_U	DF diff	0.0024	-0.7040
Dev slope	DF diff	0.0026	-0.6990
Dev_U	Dev slope	0.0026	-0.6980
$Asymm_U$	DLR slope mag	0.0026	0.6980
$Asymm_U$	DF diff	0.0029	0.6940
Dev_A	Dev slope	0.0029	-0.6940
VR_U	ND PF	0.0035	-0.6830
$Asymm_U$	Dev slope	0.0043	-0.6730
Dev_U	DLR slope	0.0049	0.6650
Dev_U	ND PF	0.0050	-0.6640
$Asymm_U$	DLR slope	0.0050	0.6640
$Asymm_A$	Dev slope	0.0051	-0.6630
VR_A	DLR slope mag	0.0051	0.6630
VR_U	ND DF	0.0054	-0.6600
VR_U	DF diff	0.0054	0.6600
BWD_A	DF diff	0.0056	-0.6580
VR_A	Berg	0.0058	-0.6560
BWD_U	PF diff	0.0066	-0.6490
$Asymm_U$	ND PF	0.0069	-0.6460
BWD_A	Berg	0.0071	0.6440
Dev_U	ND DF	0.0071	-0.6440
DLR slope mag	DLR slope	0.0081	0.6360
DLR slope	DLR slope mag	0.0081	0.6360

Table. E-1: Linear correlations between metrics with corresponding *p*-values (H_0: ρ = 0) and correlation coefficients (Pearson's ρ).

SIGNIFICANT CORRELATIONS ($P < 0.05$)

Y	X	*p*-value	ρ
Dev_U	PF diff	0.0113	0.6150
Dev slope	ND DF	0.0117	0.6130
BWD_U	ND DF	0.0131	0.6050
$Asymm_U$	ND DF	0.0137	-0.6020
$Asymm_U$	PF diff	0.0144	0.5980
DLR slope	DF diff	0.0154	0.5930
Dev slope	ND PF	0.0156	0.5930
DLR slope mag	Berg	0.0160	-0.5910
DLR slope mag	Dev slope	0.0182	-0.5810
Dev slope	DLR slope mag	0.0182	-0.5810
VR_A	DLR slope	0.0187	0.5790
Dev slope	PF diff	0.0211	-0.5700
BWD_A	ND DF	0.0213	0.5690
DLR slope mag	DF diff	0.0221	0.5670
BWD_A	Dev slope	0.0231	0.5630
Dev_A	DLR slope mag	0.0252	0.5560
BWD_A	DLR slope mag	0.0277	-0.5490
DLR slope	Berg	0.0293	-0.5440
DLR slope	ND DF	0.0350	-0.5290
BWD_A	PF diff	0.0361	-0.5270
Dev_A	DF diff	0.0363	0.5260
BWD_U	ND PF	0.0375	0.5230
$Asymm_A$	DLR slope mag	0.0436	0.5100
Dev slope	Berg	0.0442	0.5090
BWD_U	Berg	0.0466	0.5040
BWD_A	ND PF	0.0485	0.5000
VR_U	Berg	0.0490	-0.4990

Table. E-2: Linear correlations between metrics with corresponding *p*-values (H_0: ρ = 0) and correlation coefficients (Pearson's ρ).

SIGNIFICANT CORRELATIONS ($P < 0.10$)

Y	X	*p*-value	ρ
DLR slope mag	ND PF	0.0502	-0.4970
VR_A	Dev slope	0.0522	-0.4930
BWD_U	DLR slope mag	0.0536	-0.4910
$Asymm_A$	DF diff	0.0565	0.4860
DLR slope mag	D PF	0.0693	-0.4650
DLR slope mag	ND DF	0.0705	-0.4640
BWD_U	DLR slope	0.0709	-0.4630
VR_A	DF diff	0.0807	0.4490
Dev_A	Berg	0.0843	-0.4450
VR_U	DLR slope mag	0.0931	0.4340
VR_U	Dev slope	0.0982	-0.4280

Table. E-3: Linear correlations between metrics with corresponding *p*-values (H_0: ρ = 0) and correlation coefficients (Pearson's ρ).

E.2.2 All Correlations

Correlations for all metrics are listed below, categorized by metric (X). The significance levels are coded as follows: ** $p < 0.01$; * $p < 0.05$; + $p < 0.10$

ALL CORRELATIONS

Y	X	*p*-value	ρ	Significance
BWD_A	Berg	0.0071	0.6440	**
BWD_U	Berg	0.0466	0.5040	*
BWD_{MVC}	Berg	0.1055	0.4200	
VR_A	Berg	0.0058	-0.6560	**
VR_U	Berg	0.0490	-0.4990	*
Dev_A	Berg	0.0843	-0.4450	+
Dev_U	Berg	0.0004	-0.7800	**
$Asymm_A$	Berg	0.1659	-0.3640	
$Asymm_U$	Berg	0.0007	-0.7540	**

DLR slope	Berg	0.0293	-0.5440	*
Dev slope	Berg	0.0442	0.5090	*
DLR slope mag	Berg	0.0160	-0.5910	*
BWD_A	D DF	0.5481	-0.1620	
BWD_U	D DF	0.4871	-0.1870	
BWD_{MVC}	D DF	0.3904	-0.2310	
VR_A	D DF	0.1449	0.3810	
VR_U	D DF	0.6188	-0.1350	
Dev_A	D DF	0.2809	0.2870	
Dev_U	D DF	0.6812	0.1110	
$Asymm_A$	D DF	0.2518	0.3040	
$Asymm_U$	D DF	0.5390	0.1660	
DLR slope	D DF	0.7299	0.0940	
Dev slope	D DF	0.5877	-0.1470	
DLR slope mag	D DF	0.3930	0.2290	
BWD_A	ND DF	0.0213	0.5690	*
BWD_U	ND DF	0.0131	0.6050	*
BWD_{MVC}	ND DF	0.8953	0.0360	
VR_A	ND DF	0.2453	-0.3080	
VR_U	ND DF	0.0054	-0.6600	**
Dev_A	ND DF	0.1164	-0.4080	
Dev_U	ND DF	0.0071	-0.6440	**
$Asymm_A$	ND DF	0.1643	-0.3650	
$Asymm_U$	ND DF	0.0137	-0.6020	*
DLR slope	ND DF	0.0350	-0.5290	*
Dev slope	ND DF	0.0117	0.6130	*
DLR slope mag	ND DF	0.0705	-0.4640	+
BWD_A	D PF	0.4985	0.1830	
BWD_U	D PF	0.8447	0.0530	
BWD_{MVC}	D PF	0.3689	0.2410	
VR_A	D PF	0.1758	-0.3560	
VR_U	D PF	0.4637	-0.1970	
Dev_A	D PF	0.5103	-0.1780	
Dev_U	D PF	0.1683	-0.3620	
$Asymm_A$	D PF	0.7352	-0.0920	
$Asymm_U$	D PF	0.1807	-0.3520	
DLR slope	D PF	0.4747	-0.1930	
Dev slope	D PF	0.2713	0.2930	
DLR slope mag	D PF	0.0693	-0.4650	+
BWD_A	ND PF	0.0485	0.5000	*
BWD_U	ND PF	0.0375	0.5230	*
BWD_{MVC}	ND PF	0.2588	0.3000	

VR_A	ND PF	0.1473	-0.3790	
VR_U	ND PF	0.0035	-0.6830	**
Dev_A	ND PF	0.1267	-0.3980	
Dev_U	ND PF	0.0050	-0.6640	**
$Asymm_A$	ND PF	0.2087	-0.3320	
$Asymm_U$	ND PF	0.0069	-0.6460	**
DLR slope	ND PF	0.1849	-0.3490	
Dev slope	ND PF	0.0156	0.5930	*
DLR slope mag	ND PF	0.0502	-0.4970	+
BWD_A	DF diff	0.0056	-0.6580	**
BWD_U	DF diff	0.0024	-0.7040	**
BWD_{MVC}	DF diff	0.6814	-0.1110	
VR_A	DF diff	0.0807	0.4490	+
VR_U	DF diff	0.0054	0.6600	**
Dev_A	DF diff	0.0363	0.5260	*
Dev_U	DF diff	0.0016	0.7210	**
$Asymm_A$	DF diff	0.0565	0.4860	+
$Asymm_U$	DF diff	0.0029	0.6940	**
DLR slope	DF diff	0.0154	0.5930	*
Dev slope	DF diff	0.0026	-0.6990	**
DLR slope mag	DF diff	0.0221	0.5670	*
BWD_A	PF diff	0.0361	-0.5270	*
BWD_U	PF diff	0.0066	-0.6490	**
BWD_{MVC}	PF diff	0.4064	-0.2230	
VR_A	PF diff	0.3597	0.2450	
VR_U	PF diff	0.0007	0.7560	**
Dev_A	PF diff	0.1286	0.3960	
Dev_U	PF diff	0.0113	0.6150	*
$Asymm_A$	PF diff	0.1576	0.3710	
$Asymm_U$	PF diff	0.0144	0.5980	*
DLR slope	PF diff	0.2246	0.3220	
Dev slope	PF diff	0.0211	-0.5700	*
DLR slope mag	PF diff	0.2235	0.3220	
BWD_A	DLR slope	0.0016	-0.7210	**
BWD_U	DLR slope	0.0709	-0.4630	+
BWD_{MVC}	DLR slope	0.8098	-0.0650	
VR_A	DLR slope	0.0187	0.5790	*
VR_U	DLR slope	0.6314	0.1300	
Dev_A	DLR slope	0.2034	0.3360	
Dev_U	DLR slope	0.0049	0.6650	**
$Asymm_A$	DLR slope	0.2682	0.2940	
$Asymm_U$	DLR slope	0.0050	0.6640	**

Dev slope	DLR slope	0.2348	-0.3150	
DLR slope mag	DLR slope	0.0081	0.6360	**
BWD_A	Dev slope	0.0231	0.5630	*
BWD_U	Dev slope	0.0014	0.7270	**
BWD_{MVC}	Dev slope	0.2598	0.2990	
VR_A	Dev slope	0.0522	-0.4930	+
VR_U	Dev slope	0.0982	-0.4280	+
Dev_A	Dev slope	0.0029	-0.6940	**
Dev_U	Dev slope	0.0026	-0.6980	**
$Asymm_A$	Dev slope	0.0051	-0.6630	**
$Asymm_U$	Dev slope	0.0043	-0.6730	**
DLR slope	Dev slope	0.2348	-0.3150	
DLR slope mag	Dev slope	0.0182	-0.5810	*
BWD_A	DLR slope mag	0.0277	-0.5490	*
BWD_U	DLR slope mag	0.0536	-0.4910	+
BWD_{MVC}	DLR slope mag	0.3543	-0.2480	
VR_A	DLR slope mag	0.0051	0.6630	**
VR_U	DLR slope mag	0.0931	0.4340	+
Dev_A	DLR slope mag	0.0252	0.5560	*
Dev_U	DLR slope mag	0.0022	0.7080	**
$Asymm_A$	DLR slope mag	0.0436	0.5100	*
$Asymm_U$	DLR slope mag	0.0026	0.6980	**
DLR slope	DLR slope mag	0.0081	0.6360	**
Dev slope	DLR slope mag	0.0182	-0.5810	*

Table. E-4: Linear correlations between metrics with corresponding *p*-values (H_0: $\rho = 0$) and correlation coefficients (Pearson's ρ).