Upper Extremity Electromyography during Activities of Daily Living in Control's and Children with Unilateral Limb Loss

by

Carly Genn

BSc Engineering, University of Guelph, 2007

A Thesis Submitted in Partial Fulfillment

of the Requirements for the Degree of

Master of Science in Engineering

in the Graduate Academic Unit of Mechanical Engineering

Supervisors: E. Biden, PhD, Faculty of Mechanical Engineering

V. Chester, PhD, Faculty of Kinesiology

M. Tingley, PhD, Faculty of Mathematics and Statistics

Examining Board: E. Hussein, PhD, Faculty of Mechanical Engineering, Chair

P. Parker, PhD, Faculty of Electrical and Computer Engineering

P. Kyberd, PhD, Faculty of Electrical Engineering

This thesis is accepted by the Dean of Graduate Studies

THE UNIVERSITY OF NEW BRUNSWICK

May, 2011

Your file Votre référence
ISBN: 978-0-494-91862-3

Our file Notre référence
ISBN: 978-0-494-91862-3

ABSTRACT

This study focused on the question of how a child with unilateral limb loss performed activities of daily living while wearing a prosthesis and how this compared to activities conducted while not wearing one.

The purpose of this study was to determine if patterns of muscle activity in a child with unilateral limb loss were within normal limits during three activities of daily living (swinging, biking, and walking) when compared to normally limbed children doing the same tasks. This was done by looking at bilateral muscle symmetry, muscle co-contraction, activation timing and length of activation for each muscle.

The methods used included comparing a sample of normally limbed children with a sample of children with unilateral limb loss. The data collected included motion data captured using a motion analysis system and a wireless EMG system. Each subject had reflective markers placed on anatomical landmarks in order to observe their motion as well as 10 electrodes placed on 5 muscles on each side of the body.

The importance of this work is identified by the need for further research examining overuse injuries in individuals with upper limb loss (particularly pertaining to their able limb). Also, the recent debate within literature on whether children with unilateral limb loss should be fitted with prostheses at all based on their functionality is a serious topic needing more information before making any radical decisions. Lastly, there is currently very little research available on children with upper limb loss performing activities of

daily living. These represent the need for this study and how it could be an important addition to current research.

This study showed that consistent muscle patterns were apparent during the swing task for the control group. Also, the swing task exhibited the most average muscle co-contraction (for all muscles), and muscle activation (for all muscles except the dominant erector spinae), out of the three tasks for the control group. The highest average muscle symmetry was during walking, and the lowest was during biking. The case studies revealed each prosthesis user was different from the control group in various ways, however all four preferred not wearing their prosthesis during swinging compared with wearing it. The group statistics revealed that the prosthesis users, under both conditions, had significantly lower muscle symmetry compared with the control group.

ACKNOWLEDGEMENTS

This thesis would not have been possible, or coherent without the help of family and friends at home (in particular mom, dad, Andrea, and Mike), but most notably the tireless efforts of my two supervisor's Dr. Ed Biden and Dr. Vicky Chester. A few unsung heroes who helped me a great deal even though there was nothing in it for them were Dr. Usha Kuruganti, Dr. Maureen Tingley, my lab/office mate Katelynn Craig, and the clinic OT, Wendy Hill. Lastly, there were many people within the Institute of Biomedical Engineering who helped out every day, and I appreciate everything that was done and truly believe the IBME is a great place for students, professionals, and academics to share, learn, and excel together.

Table of Contents

List of Tables

List of Figures

1.0 Introduction

Currently there are two very opposite schools of thought in terms of fitting children with unilateral, below elbow congenital deficiencies with electric prostheses. The first is to fit the child with a passive prosthesis when they can sit up and subsequently fit them with an electric device between 12 and 15 months (Gaebler-Spira & Uellendahl, 1999; Shaperman et al., 2003). The second is that since this group of children are able to perform most activities of daily living (ADL's) independently, the prosthesis provides little to no increase in functionality (Davids et al., 2006; James, 2006). An interesting factor to consider in this argument could be the risk level of these children for acquiring repetitive strain injuries. There have been some studies that have already indicated that upper limb amputees have had overuse problems (Jones and Davidson, 1999; Datta, 2004).

Upper extremity limb loss is uncommon compared to lower extremity limb loss, but it is more prevalent in children because of the incidence of congenital cases. In general, the rate of congenital upper-limb deficiency has been shown to be approximately two to three fold that of congenital lower-limb deficiency (Ephraim, et al., 2003). Upper extremity limb loss may occur traumatically, surgically, or congenitally. Between 1988 and 1996, Dillingham et al. (2003) studied the incidence of upper limb loss and found that the deficiency was approximately 24 out of every 100 000 people in 1988 and approximately 21 out of every 100 000 people in 1996. Of these, in 1988 60.7% were congenital, 33.5%

were traumatic, and 5.8% were acquired. In 1996, 75.7% were congenital, 18.1% were traumatic, and 6.5% were acquired.

The main purpose of this study was to describe the muscle activity in children with unilateral limb loss and to decipher whether this activity lies within normal limits during activities of daily living. We hypothesized that throughout the activities chosen for study, children with unilateral limb loss would have muscle symmetry and activation patterns that lie outside the normal range.

The tools that were used to identify these potential compensations are encompassed in the advanced motion laboratory located in the Institute of Biomedical Engineering at the University of New Brunswick. Included in this laboratory are an infrared eight-camera Vicon motion capture system as well as the Zero-Wire electromyography (EMG) system and four Kistler force plates that are embedded in the laboratory floor.

The Vicon motion capture system was utilized to observe how the patient moved while performing these gross motor tasks in terms of body segment angles and rotations. The surface electromyography system monitored the muscle activity of ten major upper body muscle groups while performing three activities. The force plates were used during the walking segment of the laboratory session.

The specific group of individuals included in the study were children between the ages of 5 and 14 years old. The control group was required to have two asymptomatic arms. The

prosthesis users were individuals in the same age group that had unilateral, below-elbow limb loss. The prosthesis users were tested once while wearing their myoelectric prosthesis on and once while not wearing it. Therefore, intra-individual comparisons were made when they were wearing their prosthesis and when they were not, and inter-individual comparisons were made between the prosthesis wearers and the control subjects under both wearing and not wearing conditions.

Electromyography during activities of daily living has not been studied extensively with the exception of gait analysis. In the case of electromyography in gait analysis, it is almost exclusively lower body muscles that have been studied. This study aims to yield a better understanding of the muscle activity for normally limbed children and children with limb loss while performing ADL's.

These experiments attempt to examine muscle activity in the upper limb child amputee and determine whether this activity lies outside normal limits. Currently, there is little to no research previously conducted that has reported the EMG activity of children with unilateral upper limb loss. There have been a few studies that have reported that these individuals are at a greater risk for repetitive strain injuries (Jones and Davidson, 1999; Datta, 2004), but there is not enough evidence currently to confirm this. This research attempts to initiate more interest in these problems in order to eventually provide a clearer prosthesis prescription plan for individuals who have unilateral limb loss particularly those for whom the loss is congenital.

2.0 Literature Review

Over the years, the development of myoelectric prostheses helped move the field of prosthetics forward through advances in technology and increased user control. However, myoelectric prostheses may be better suited for some groups of prosthesis wearers than others. For example, there has been recent scrutiny in fitting children with below-elbow congenital deficiencies with myoelectric prostheses. The main contributor for this argument is that children with unilateral congenital below-elbow deficiency are usually able to perform most activities of daily living (ADL's) independently and therefore the use of a prosthesis provides little to no increase in functionality (Davids et al., 2006). However, this is a topic of recent debate and some important factors that are for and against this argument are rejection rates, reported prosthesis wear, and risks of overuse injuries.

2.1 Upper Extremity Limb Loss

Upper extremity limb loss occurs either traumatically, acquired, or congenitally. Traumatic loss occurs when the individual is not expecting to lose the limb, acquired occurs when the individual has chosen to get the limb removed surgically, and congenital loss is when the individual is born with an upper limb deficiency. Limb loss can be a traumatizing event to go through and can result in psychological, vocational and physical problems (Dillingham et al., 2003). Between 1988 and 1996 Dillingham et al. (2003) studied the incidence of upper limb loss and deficiency was approximately 24 out of every 100 000 people in 1988 and approximately 21 out of every 100 000 people in 1996.

Of these, in 1988 60.7% were congenital, 33.5% were traumatic, and 5.8% were acquired. In 1996 75.7% were congenital, 18.1% were traumatic, and 6.5% were acquired. There seems to be an increase in congenital cases, however the causes of congenital limb deficiency are often unknown.

2.1.1 Congenital Limb Deficiency

Congenital limb deficiency is a condition where a child is born with a missing limb. The International Society for Prosthetics and Orthotics (ISPO) has a classification system to describe varying levels of congenital limb deficiencies. The two main classifications of limb deficiencies are transverse and longitudinal. For this project, the transverse ISPO classifications are of most use, because these are mainly used to classify congenital deficiencies. Figure 1 below describes the various levels of transverse limb deficiencies.

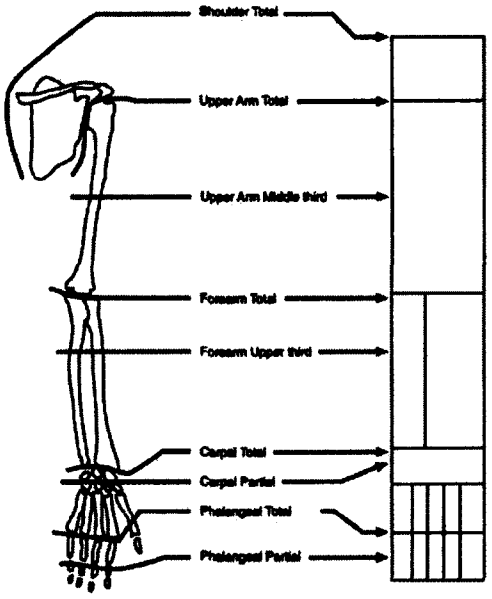


Figure 1: Describes the various levels of transverse upper limb loss possible (Day H., 2004).

Below-elbow limb loss will be the focus, which means the individual has an intact elbow joint. Therefore, based on Figure 1, this would be from the forearm upper third level and down. These are the most common upper arm injuries, and are also the group of individuals that have been deemed "too functional" to require myoelectric prosthesis assistance in recent literature (Davids et al., 2006).

It has been documented that individuals who have sustained limb loss in childhood or young adulthood experience aging effects superimposed on their existing impairments and disabilities (Flood et al., 2006). Therefore, treating young individuals with upper extremity limb loss must have emphasis on prevention and reduction of the incidence of secondary injuries that will leave them further disabled.

2.1.2 Prosthetic Options

There are many options available to an individual with upper extremity limb loss, most of which are classified as passive, body powered, externally powered, and hybrid prostheses.

Ideally, since every individual is unique physically, mentally, and socially, everyone would receive a different prosthesis catering to their desires and needs. It is difficult to satisfy all of these because of the current technology, time, labour, and money necessary. Each prosthesis has its advantages and disadvantages, and depending on the individual,

some are more important than others Table 1 summarizes these advantages and disadvantages.

Table 1: Summary of various prosthesis options, and their advantages and disadvantages.

	Advantages	Disadvantages
Passive	aesthetically pleasing lightweight low maintenance little harnessing required	No grip ability No pinch ability
Body-Powered	low cost lightweight high reliability	harness is uncomfortable and restricts movement a relatively large amount of energy is necessary to drive the system components not aesthetically pleasing
Externally Powered (ie. Myoelectric Prosthesis)	best available pinch and grip forces physical exertion to drive components is minimal compared with body-powered more natural to operate	heavy components high maintenance costly
Hybrid	same grip and pinch forces are attainable if an electrically controlled terminal device is used ability to simultaneously control movements of the elbow and terminal device/wrist lighter than a fully electrically controlled system	somewhat difficult to operate because they use two different control systems these two control systems could have interference

2.1.3 Myoelectrically Controlled Prostheses

Myoelectric control is the most common form of control for externally powered devices when a useable electromyography signal can be produced (Alley & Sears, 2004). The controller works by using an electrical signal generated by a muscle (usually found in the

residual limb) to operate the terminal device; pinching, grasping etc. Figure 2 below gives a good representation of how a myoelectric prosthesis functions.

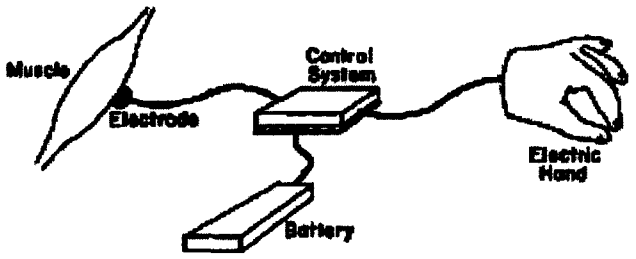


Figure 2: The main components that drive a myoelectric prosthesis (Scott, 1984).

Some problems with myoelectric control occur if the individual does not have the ability to produce a suitable myoelectric signal. This could happen because of a variety of reasons, including:

- Poor muscle strength,
- Too much tissue between muscle and electrode (ie. subcutaneous tissue),
- Poor electrode to skin interface (non-conductive),
- No muscles available (higher level limb loss).

The advantages of using a myoelectric prosthesis over the other options are that it eliminates the reliance on gross body movements, it gives proportional control (the harder they contract the muscle, the faster the hand moves/positioning of elbow), energy expenditure is minimal to operate, and it promotes muscle tone therefore preventing muscle atrophy (Alley & Sears, 2004). Overall, myoelectric prostheses yield a system of control that most closely mimics the natural human control in comparison to other types

of prostheses. One method of determining the quality of assistance a prosthesis provides to its user is by measuring the rejection rates of various prosthesis types.

2.2 Prosthesis Rejection

Prosthesis rejection has been recorded over the past 30-40 years, but has yet to be done in a consistent manner. Inconsistencies in sample, methodological approach, and characteristics of the clinical programs or prostheses involved (Biddiss & Chau, 2007) could be contributing to the vast differences recorded for rejection rates. Another large difference between various studies is the definition of prosthesis rejection. Some define prosthesis rejection when there has been no contact with the family/patient for a specified amount of time or the client has stated they do not use the prosthesis (Davids et al., 2006). A more complex way of defining rejection takes into account the time the prosthesis was used prior to rejection (Postema et al., 1999). The latter takes into account the value the prosthesis had when it was being used prior to rejection.

Rejection rates help quantify the success of prostheses for various groups of people using different prosthesis designs. To address the recent speculation on myoelectric prosthesis use for children with unilateral, below-elbow congenital deficiencies, rejection rates and prosthesis usage for this these individuals were examined. It has been widely reported that a myoelectric prosthesis for a child with below-elbow congenital limb loss does not provide additional functionality. Most of the literature expresses the fact that children with unilateral, below-elbow limb loss are almost fully functional (Glynn et al. 1986; James et al., 2006; Davids et al., 2006; Crandall & Tomhave, 2002; Kruger & Fishman,

1993; Postema et al., 1999; Pruitt et al., 1998). However, Nelson et al. (2006) notes that this group may need help with problem solving on how to perform activities such as riding a bike or playing on playground equipment. Another potential functional benefit for these individuals is when doing bimanual activities requiring body symmetry (Meurs et al., 2006).

As a child ages, they will go through lifestyle changes that can make their myoelectric prosthesis more or less useful to them. It has been shown that a high social adjustment in children with upper limb deficiencies occurs in contrast with the adult population (Hermansson et al., 2005; Tyc, 1992). Also, during puberty, patients can deal with increased psychological problems regarding their deficiency (Meurs et al., 2006). This emphasizes the fact that children go through drastic lifestyle changes and even if a prosthesis may be adding little to no functional benefit, it may help them psychologically.

The order of prostheses prescribed to an individual depends on their age when they first receive treatment, and each prosthetic clinic will have their own prosthesis prescription plan that they follow. One example is to provide a passive prosthesis once the child has achieved the ability to walk independently. If the child wears the prosthesis on a daily basis, and the parents express an interest in using a more complex design, a body-powered or a myoelectric prosthesis is offered when the child is between 2 and 4 years of age (Davids et al., 2006). In a study done by Biddiss & Chau (2007), it was observed that electric devices may be more widely accepted than body-powered devices by children. Therefore, if an institute tends to initially fit more children with body-powered devices

before switching to an electric device, they may be more susceptible to rejecting their prosthesis compared with a child who is fit with a myoelectric one first.

The average rejection rates over the past 25 years for adults using a passive device are 39% +/-35%; adults using a body powered are 27% +/- 13%; adults using an electric device are 22% +/- 12% (Biddiss & Chau, 2007b). The average rejection rates over the past 25 years for children using a body powered device are 45% +/- 17%; children using an electric device 32% +/-19% (Biddiss & Chau, 2007b). Therefore, adults are rejecting electric prostheses less often than children, but children seem to be rejecting the electric prosthesis less than body-powered devices. An important factor in reducing prosthesis rejection and increasing prosthesis usage is choosing a prosthesis that is right for the individual and their needs.

Another way to determine whether the prosthesis has made a successful integration into a person's life is to determine the amount of time and type of usage the individual has with the prosthesis. This is another quantity that is difficult to accurately obtain and thus has large variance in literature.

2.3 Prosthesis Usage and Wear Patterns

An individual's lifestyle, age, level of limb loss and desired functions will dictate the type of prosthesis used, what it is used for and how often it is used (Biddiss & Chau (b), 2007). A study done by Biddiss & Chau (2007b) determined the predisposing

characteristics linked with prosthesis use; level of limb loss emerged as the largest predisposing factor. Multiple limb amputees and unilateral shoulder injured patients saw themselves being more affected in leisure tasks than work tasks, and in contrast, unilateral upper limb and partial hand amputees saw themselves as having a greater disability affecting work tasks (Davidson, 2004). These feelings of disability can affect the way a person will accept their prosthesis. It is generally known in literature that most adults wear their prosthesis at work or at school for at least 8 hours daily (Pylatiuk et al., 2007), and wear on weekends and at home is reduced (Biddiss & Chau (b), 2007). Children, however, may only use their myoelectric hand for less than 4 hours a day (Pylatiuk et al., 2007). Passive and body-powered hands are the least consistently used and when used are primarily for social activities (Biddiss & Chau, 2007).

A client and clinic review was completed for the upper-limb prosthetics clinic based out of the Institute of Biomedical Engineering in Fredericton, New Brunswick. Figure 3 shows the trends for clients to decrease their number of visits per year as they age. The Y-axis is the average number of visits per client, per year. This yielded a yearly visit average for each type of prosthesis user at a particular age.

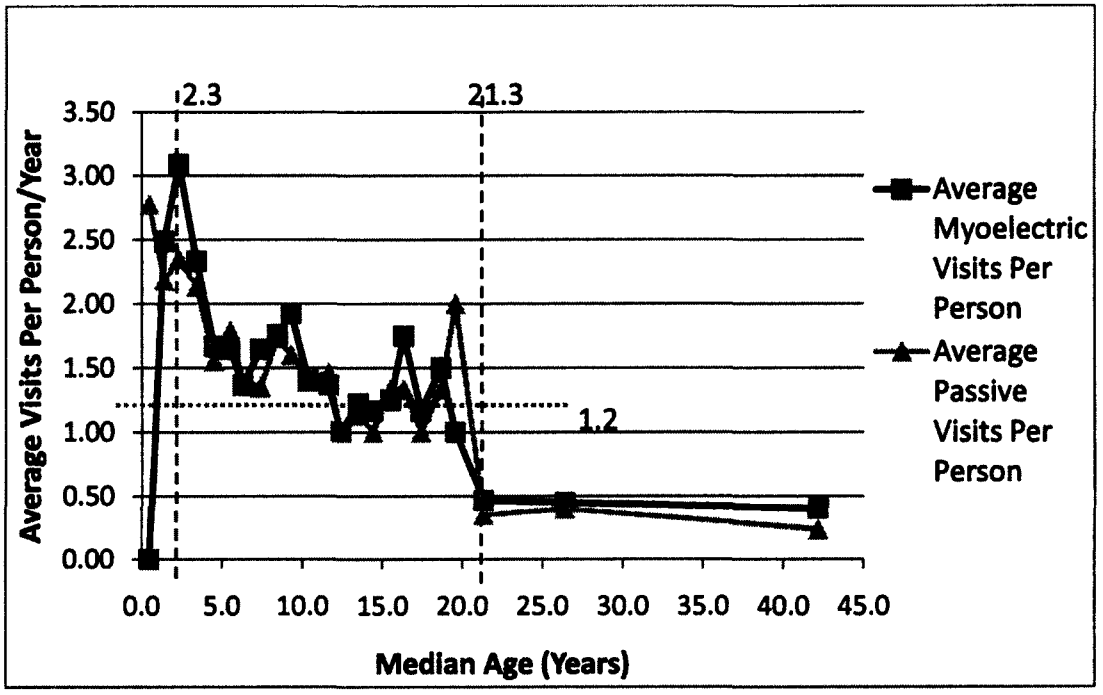


Figure 3: Average Visits per client/year for individuals attending the upper limb clinic at the Institute of Biomedical Engineering (Genn, et al. 2009).

Based on Figure 3, it is evident that as a client ages, their average visits per year also decreases and seems to settle around one visit every 2 years for adults. This may indicate less maintenance is necessary for the clients as they age as their growth slows and they get accustomed to prosthesis use. However, this could also point towards prosthesis use decreasing as the client ages. The notion that prosthesis wear decreases with age has been found by several paediatric studies (Herberts et al., 1980; Millstein et al., 1986; Ballance et al., 1986; Postema et al., 1999; Davids, et al., 2006).

This could be due to the fact that very young children perform mostly motor skills, in which case a prosthesis may be useful (Postema et al., 1999) such as climbing, bringing objects to the mouth, playing with blocks etc. Postema et al. (1999) notes that as a child grows older, their activities shift toward increasingly intellectual tasks where the prosthesis is not as useful. Therefore, even if an electric prosthesis use may eventually decrease, the prior use of the prosthesis could have been a critical part of their development. Additionally, one prosthesis may not provide all of the functional benefits desired by an individual when performing various activities. A study by Davids et al. (2006) noted that long term prosthesis use increased for individuals who had multiple prostheses. Overall paediatric use was recorded by Biddiss et al. (2007) in which 97 children were interviewed, 39% were full time users, 18% were part-time users, 12% were occasional users, 12 % were sporadic users, and 19% were non-users (past wearers).

One method of attempting to reduce rejection rates and increase prosthetic usage is to determine any pain associated with prosthesis use and take preventative measures against this pain development.

2.4 Chronic Pain Associated with the Upper Limb Amputee

Chronic pain affects a broad range of individuals and is generally defined as pain that persists beyond the expected period of healing (Turk, 2001; Merskey, 1986).
The main types of chronic pain an individual with upper limb loss experiences are phantom limb pain, residual limb pain, back and neck pain, and intact limb pain.

Chronic, persistent pain can lead to limitations in function in a physical and psychosocial manner (Ephraim et al., 2005). This pain can also alter the individual with limb loss's lifestyle, such as decreasing their chances of employment (Whyte & Carroll, 2002 ; Millstein et al., 1985) as well as reduce the likelihood of participation in social activities (Parks, 1973). Considering these individuals are already performing many tasks one-handed, chronic pain can greatly interfere with their everyday lives.

In various studies, phantom pain was often the most prevalent source of pain and ranged from 50% to 79% of upper limb loss respondents (Kooijman et al., 2000; Datta, 2004; Fraser et al.,2001; Hanley et al., 2009). Residual limb pain was reported between 48% and 71% in upper limb deficient respondents (Datta, 2004; Kooijman et al., 2000; Ephraim et al, 2005; Hanley et al., 2009). Neck and back pain were reported between 43-45% (Datta, 2004; Hanley et al., 2009) and 40-62% (Datta, 2004; Hanley et al., 2009; Ephraim et al., 2005) respectively. Pain associated with the intact limb (the limb contralateral to the side with the upper limb deficiency) was less widely reported. However, the studies that have researched this type of pain reported 33% (Hanley et al., 2009), 45% (Datta, 2004), and 50% (Ephraim et al., 2005) of respondents.

In a study performed by Hanley et. al. (2009), it was noted that pain in the non-amputated limb had the highest average pain interference and highest number of disability days compared to all other reported pain types (phantom limb pain, residual limb pain, back pain, neck pain). This was a striking finding that demonstrated how disabling pain in the

able limb can be. The same authors also questioned whether wearing a prosthesis contributed to or relieved phantom limb pain. In this study there seemed to be no other correlation between residual limb pain, back and neck pain, and non-amputated limb pain with whether the person wore a prosthesis or not. However, in another study the use of a myoelectric prosthesis was found to reduce phantom pain in upper limb amputees (Lotze, et al., 1999). Pain reported in the contralateral limb was the incentive to perform this study and to discover whether prosthesis users electromyography patterns lie within normal limits while wearing their prosthesis versus not wearing their prosthesis compared to normally limbed children. Even though there is not extensive literature that studies chronic pain in the sound limb, the current information available shows that this area of research deserves more attention.

Research highly relatable to an individual with upper limb loss is post stroke patients with a hemiparetic limb. Both of these groups have one able limb and are therefore prone to overusing it. In fact, evidence exists that individuals with a hemiparetic hand have longstanding decrements in motor performance (Smutok et al., 1989), as well as lower performance in manual dexterity, global performance, motor coordination, and thumb kinesthesia (Desrosiers et al., 1996) in the "uninvolved" limb. It is difficult to determine a common underlying cause for the decreased functionality in the able limb in these stroke patients experiencing hemiparesis. However, a study done by Yoshiro et al. (1999) determined that after stroke occurs, subclinical Carpal Tunnel Syndrome occurs in the "unaffected" side relative to the hemiparetic side or to control subjects. Specifically, "Tinsel's sign (lightly tapping the nerve to elicit a "pins and needles" feeling in the

distribution of the nerve) was significantly more prevalent in the disused hand group (57.7%) than the functioning hand group (31.1%). Due to the similarities between post-stroke patients with a hemiparetic limb and individuals with upper limb loss, it is possible the latter are also at risk of developing overuse injuries in their non-affected side.

2.5 Overuse Injuries

The definition of overuse injuries does not apply to just one injury, but is more of a general term describing many similar injuries. These injuries develop as a result of repetitive movements, awkward postures, and sustained force (Tulder, 2007). The term "repetitive strain injuries" is somewhat controversial is also referred to as "overuse syndromes", or "cumulative trauma disorders". The most common injuries of the upper extremity which were most relatable to upper limb amputees were identified by Gambrell (2008). These included; injuries of the shoulder: tendonitis, shoulder impingement, and bursitities; the elbow: medial and lateral epicondylitis; the wrist/forearm: carpal tunnel syndrome, and tendonitis of forearm flexors and extensors; the hand/finger: De Quervain's Syndrome and Trigger Finger.

Repetitive strain injuries are most often diagnosed based on history, and clinical examination (Tulder, 2007). A good way to manage these kinds of injuries is prevention and education on the subject for the individual at risk (Fry, 1986; Krivickas, 1997; McCarroll, 2001; Shafer-Crane, 2006).

The pathophysiology of overuse injuries is diverse, yet relatively unproven with scientific evidence. One of specific interest is the "Cinderella Hypothesis" (Hagg, 1991). This hypothesis basically suggests that if long periods of muscle activity occur without sufficient rest, muscle fibre damage and pain will be the result. Some studies that use electromyography to verify this hypothesis include the detection of gaps in muscle activity, or periods of muscle rest; where fewer incidence of muscle activity gaps have been found in individuals with trapezius muscle pain (Veiersted et al., 1993; Hagg & Astrom, 1997) and shoulder myalgia (Sandsj, et al., 2000). The "Cinderella hypothesis" is beginning to be used to prevent and treat low intensity static cases such as computer related strain injuries (Hermens & Hutten, 2002).

This helped shape the hypothesis that abnormal muscle activity and postures during various ADL's could indicate the potential risk individuals with upper extremity limb loss have for repetitive strain injuries. The analysis of the current study focused mainly on the muscle symmetry between the same muscles on each side, as well as the percentage of time each muscle is on per cycle. The antagonist versus agonist muscle pairings were also analysed based on evidence suggesting that prolonged antagonist and agonist muscle contraction increases the risk of obtaining focal hand dystonia (an overuse disorder of the hand) (Chen, 1998).

To date, few studies have examined repetitive strain injuries in individuals with upper limb loss. Therefore more evidence is needed before concrete conclusions can be drawn. However, despite the current lack of empirical evidence, there have been

recommendations to include patient awareness and prevention of repetitive strain disorders into prosthetic management plans (Flood, et al., 2006; Smurr, et al., 2008; Gambrell, 2008; Lake & Dodson, 2006; Datta, et al., 2004).

One study that has looked directly at the incidence of repetitive strain disorders in individuals with upper limb loss was by Jones and Davidson (1999). Their study revealed that 50% of upper limb amputee respondents had overuse problems of varying severity and type. Some of the repetitive strain injuries seen were tenosynovitis, carpal tunnel syndrome, shoulder impingment, and epicondylitis.

Whether individuals with upper limb loss are experiencing chronic pain or repetitive strain injuries in their intact limb, both of these scenarios will lead to increases in disability for the individual. Therefore, already there are striking trends in the literature on the risk of overuse injuries in the sound limb of individuals with upper limb loss. However, further knowledge on the subject is important. This potential increased risk of overuse injury, as well as the questions pertaining to fitting children with unilateral, below elbow limb loss with prostheses gives this project the opportunity to be valuable to the prosthetic community.

2.6 Purpose and Hypothesis

The purpose of this study was to determine the muscle patterns of children with unilateral, below-elbow limb loss, and compare these with normative muscle patterns. These normative muscle patterns were also collected and examined. The hypothesis was that children with limb loss would have muscle activity that lies outside normal limits. Additionally, children with limb loss would have activity closer to normal while wearing their prosthesis compared to the same individual while not wearing their prosthesis.

3.0 Experimental Methods

The main focus of this study was to describe the muscle activity in children with unilateral limb loss and to decipher whether this activity lies within normal limits during activities of daily living. To achieve this, the use of motion analysis, electromyography, and force plate data were required.

3.1 Sample

The subjects who participated in this study were divided into two groups between the ages of 4 and 14 years; a histogram of their ages are described below in Figure 4.

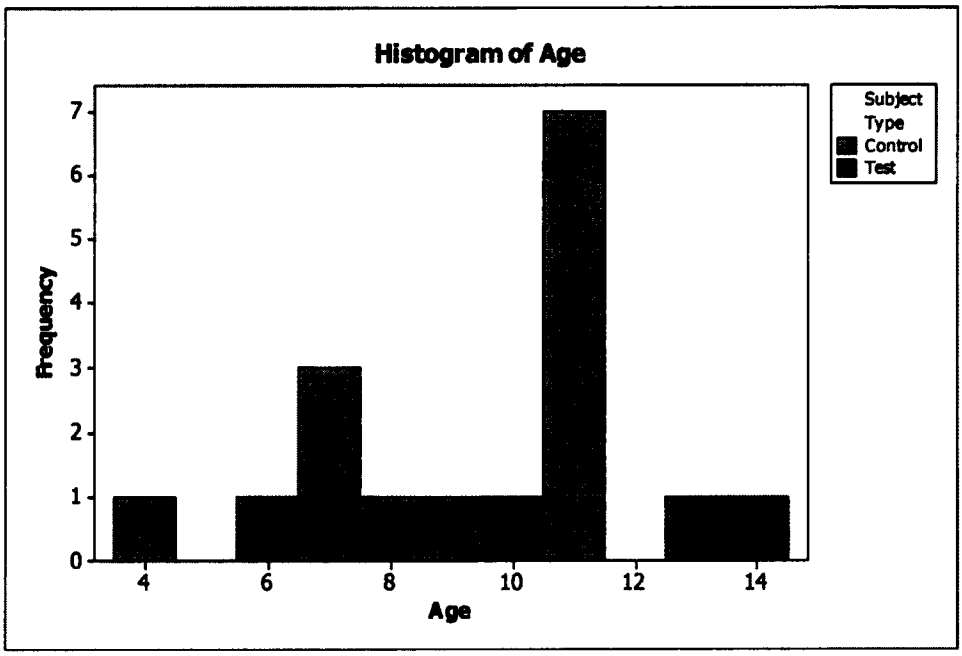


Figure 4: Histogram of age for subjects included in the test and control groups.

The control group consisted of normally limbed children and were recruited by email throughout the faculty, staff, and students at the University of New Brunswick as well as through the Scout Association in Fredericton. The group of individuals in the test group were clients of the upper-extremity fitting centre located in the Institute of Biomedical

Engineering. The occupational therapist at the clinic asked children and their parents in the appropriate age bracket if they would consider being a part of the study. This led to four participants. Therefore, there were 16 children in the control group and four in the test group. Some additional demographic information about subject gender and dominant hand side are displayed in Figure 5, and information regarding weight, height, age, and residual limb length for the control group and test group are displayed in Tables 2 and 3.

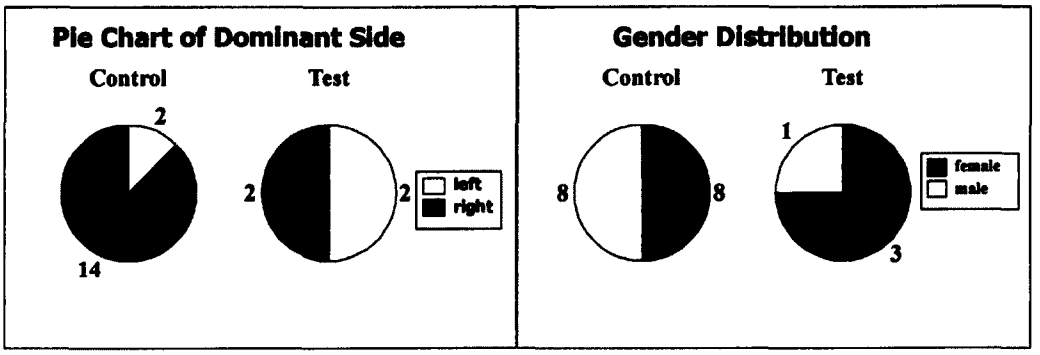


Figure 5: Subject gender and dominant hand side information for the control and test groups.

The subject and parent/guardian were given a detailed explanation of what the lab session would entail and had the option of withdrawing if they were not comfortable for any reason. The consent form used for this study is located in Appendix A. Once the consent form was filled out, they were escorted into lab and the parent/guardian had the option of sitting in on the lab session.

Table 2: Demographic information on control group subjects included in study.

Subject	Subject Group	Age (yr)	Weight (kg)	Height (cm)	Gender	Dominant Side	Non-Dominant Side
AMNFR11a	Control	11	40.1	138	female	right	left
BGNMR10	Control	10	29.9	133	male	right	left
CCNFR11	Control	11	36.1	146	female	right	left
DKNFR07	Control	7	22.9	124	female	right	left
EJNMR11	Control	11	37.9	154.5	male	right	left
FPNMR14	Control	14	51.3	177	male	right	left
GZNML11	Control	11	56.2	155	male	left	right
HJNMR08	Control	8	36.7	143	male	right	left
LSNMR09	Control	9	27.2	128	female	right	left
MWNMR06	Control	6	25.9	122	male	right	left
NVNFR11	Control	11	55.6	162	female	right	left
OSNFR11	Control	11	44.2	145	female	right	left
SMNFR07	Control	7	29.9	133.5	female	right	left
TMNMR04	Control	4	20.2	113	male	right	left
UKNFR07	Control	7	20.6	121.5	female	right	left
VTNML11	Control	11	31.5	143	male	left	right

Table 3: Demographic information on test group subjects included in study.

Subject	Subject Group	Age	Weight (kg)	Height (cm)	Gender	Dominant Side	Non-Dominant Side	Level of Loss
ICPML 14	Test	14	44.2	162	male	left	right	Transradial
PAPFL 10	Test	10	46.3	147	female	left	right	Wrist Disartic.
RCPFR 06	Test	6	19.7	102	female	right	left	Elbow Disartic.
WRPFR 13	Test	13	42.6	165.5	female	right	left	Transradial

3.2 Data Instrumentation

The two main components that were used throughout this study were the Vicon motion analysis system as well as the zero-wire wireless EMG system. Both of these systems, along with the 32 channels of analog information provided by the force plates were fed into the Analog-to-Digital box provided by the Vicon station, which allowed for 64 channels in total. The Vicon station was then connected to the computer in the lab. Figure 6 below shows a schematic of the data collection system in the lab.

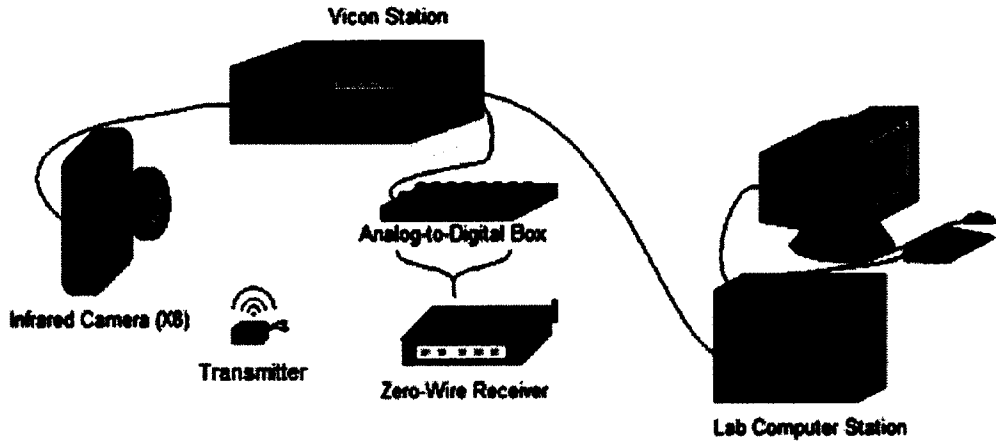


Figure 6: Various data collection systems coordinated together

3.2.1 Motion Capture System

The Vicon motion capture system (from Oxford Metrics Group, UK) uses eight infrared cameras to locate reflective marker balls in three dimensional space. Each camera emits infra-red rays that reflect off the markers, and it is these reflections that are captured by the cameras. An example of one of the eight cameras used for this system is shown below in Figure 7.

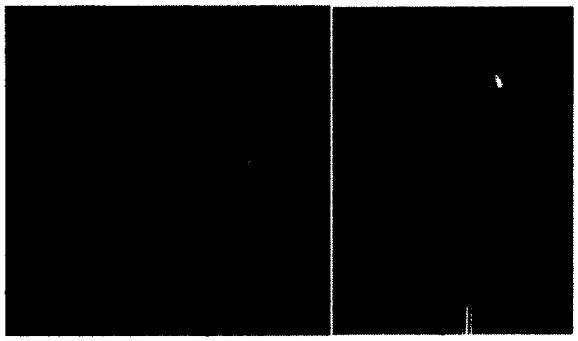


Figure 7: Two of 8 infrared cameras belonging to the Vicon motion capture system.

The cameras were calibrated to determine their location away from one another. The first step taken was to perform an initial static and dynamic calibration of the system. The static calibration was done by placing an "L" shaped rod at the desired origin of the

data collection site. For this study, the origin was determined to be the corner of the force plates in the middle of the lab floor. Figure 8 below shows the L rod at the origin of the system.

Figure 8: Static calibration rod at origin of data capture space and global coordinate system axis designation.

The reflective markers on this L-rod define the x,y, and z global coordinates for the Vicon system. The dynamic calibration was done by waving a rod with two large markers placed 500 mm apart around the desired capture space. These data are combined to allow the system to calculate the locations and orientation of each camera.

Once calibration was completed, the eight cameras could not be shifted or altered in any way. After this point, for any subsequent capture, two cameras had to be able to "see" each marker at all times throughout data capture to ensure it's visibility to the system.

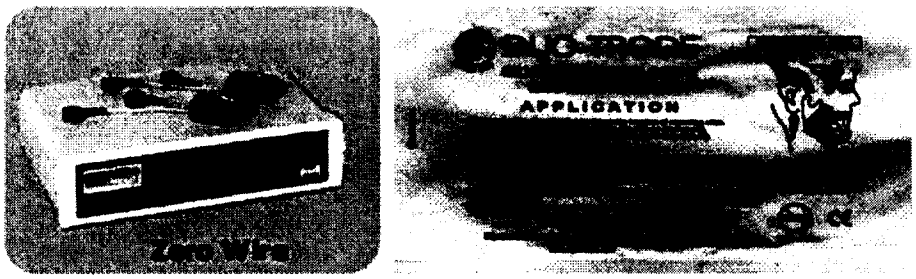
For this study, the video system was calibrated at a sample rate of 60 Hz, which was more than double the frequency that the subject's movements were occurring at. This satisfies the Nyquist Theorem requirements that any frequencies being captured must be half the sample rate of the system or lower.

3.2.2 Electromyography

Surface electromyography was used exclusively to determine the timing of the muscle activation during gross motor activities. This allowed for determination of muscle activation symmetry as well as timing and length of muscle activation.

The analog system was calibrated separately, as it had a different sample rate of 1080 Hz. All analog trials connected to the 64-channel Analog-to-Digital (A/D) board were calibrated simultaneously. The analog calibration was completed when the subject was lying face down on a massage table to ensure minimum muscle activity.

The Zero-Wire EMG system was the second main system used in this study, with three main components, the duotrodes, transmitter, and the receiver, which were integrated through Vicon's 64-channel A/D board. Figure 9 below shows these components.



A B

Figure 9: A shows the transmitting electrodes and the receiver (Noraxon, 2010), B shows the duotrodes used in this study (Myotronics, 2010).

Duotrodes were used and consisted of silver-silver chloride, bi-polar, disk-shaped electrodes. Pre-gelled contact surfaces were 12.5 mm diameter and were mounted in pairs 19 mm from center to center. The Zero-Wire system had an EMG bandwidth of 10-500 Hz, probe gain of 1000, and operative range within 20 meters of the electrodes. Four factors to consider with EMG related instrumentation (Soderberg and Cook, 1984):

- signal source,
- transducer used in detection,
- amplifier and,
- signal processing

The signal (EMG signal) is based upon the addition of motor units (MUs) that are activated during a muscle contraction. The two most important mechanisms that influence the magnitude and density of the EMG signal are the recruitment of the MUs and their firing frequency (Konrad, 2005). These mechanisms are altered by the physiology of the individual being measured, as well as the orientation of the electrodes on the skin. The large differences between physiological properties of individuals

studied coupled with the endless possibilities for variations in methodology by the lab practitioner is ultimately what determines the validity of the EMG signal. Some of the main factors that will affect the EMG signal are as follows: anatomical and biochemical characteristics of the muscle, amount of subcutaneous fatty tissue, muscle cross talk, external noise, and electrode placement/configuration (Konrad, 2005; De Luca, 1997). The maximum amplitude of such a complex EMG wave form is approximately 3 mV (Soderberg & Cook, 1984). When capturing EMG, there are generally two types of muscle contractions available. The first is a maximum muscle contraction, which is the maximum at which a subject can contract their muscles. This usually involves a proven system which isolates the muscle contraction such that the maximum muscle contraction is not contributing to any movement of a limb segment. This type of contraction is usually used when the amplitude is being monitored and thus needs to be normalized. The second type of contraction is a nominal muscle contraction, which is a normal contraction from any subject that is not regulated by the researcher (not regulated between subjects either). Nominal muscle contractions, without any measurement of a maximum muscle contraction, do not allow for the amplitude to be normalized. However, they are adequate for research focusing on the timing of the muscle contractions. Based on these stipulations, nominal muscle contractions were measured in this study, and no maximum muscle contraction was measured.

The transducer used was provided by the wireless EMG system from Aurion. This system allows for the conversion of the digital telemetry data coming from the wireless electrodes, to analog output signals. The amplifier in this system is the gain of 1000

within the wireless electrodes. This gain came into consideration when calculating the EMG amplitude. A detailed description of the signal processing is included in the Data Analysis section. An example of raw (unfiltered) EMG signal of three nominal trapezius muscle contractions is shown below in Figure 10.

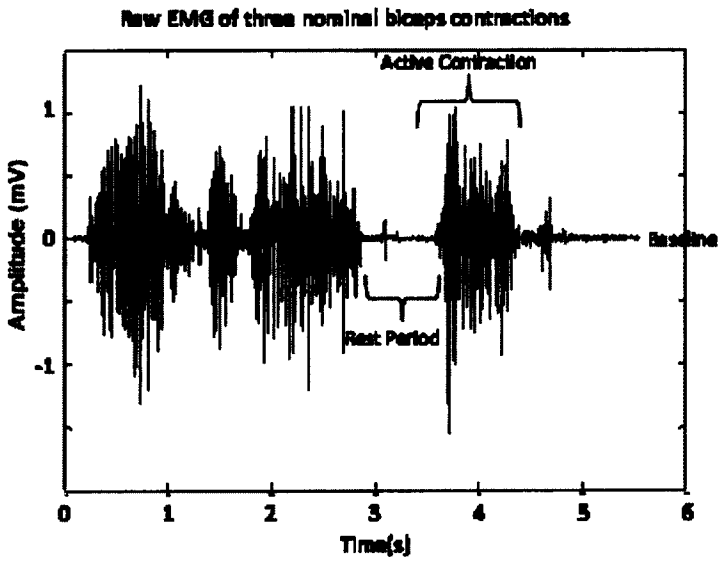


Figure 10: Raw EMG of three nominal trapezius muscle contractions

The protocol into the determination of muscles examined was as follows:

- Muscle had to be on the upper limbs,
- Muscle had to be above the elbow or on the trunk (due to the individuals with limb loss would not have certain muscles below the elbow and this would defeat the purpose),
- Muscle had to be easy to locate, and therefore were usually large muscles close to the surface of the skin,
- Muscle was thought to be active in at least two out of the three activities chosen.

According to these factors, ten muscles were selected for observation; both sides for each of the following muscles: the biceps brachii, triceps brachii, trapezius transversalis, deltoideus medius,and erector spinae. It was possible to collect data on 16 muscles, however it was decided that more than ten muscles being monitored at once may have caused interference on the children's performance during the three activities. In addition, since the age group of the children was relatively young, they had smaller muscles that were closer together making electrode placement more difficult. Out of the ten muscles observed, the erector spinae was the most difficult muscle to accurately capture due to its development stage in children.

Since there is no documented observation of upper body muscles in children during gross motor activities, there was no literature to guide these muscle choices other than one article stating that the erector spinae muscles are activated primarily just before foot-strike of each limb during gait (Steven et al., 2002). The muscle activity patterns of these ten selected muscles were described based on the cycle for each activity.

A European group called SENIAM has in place guidelines for electrode placement that is meant to increase the consistency and validity of EMG studies (Seniam, 2010). Figure 11 shows the SENIAM guidelines for electrode placement on the biceps, triceps, trapezius, deltoidus, and erector spinae muscles.

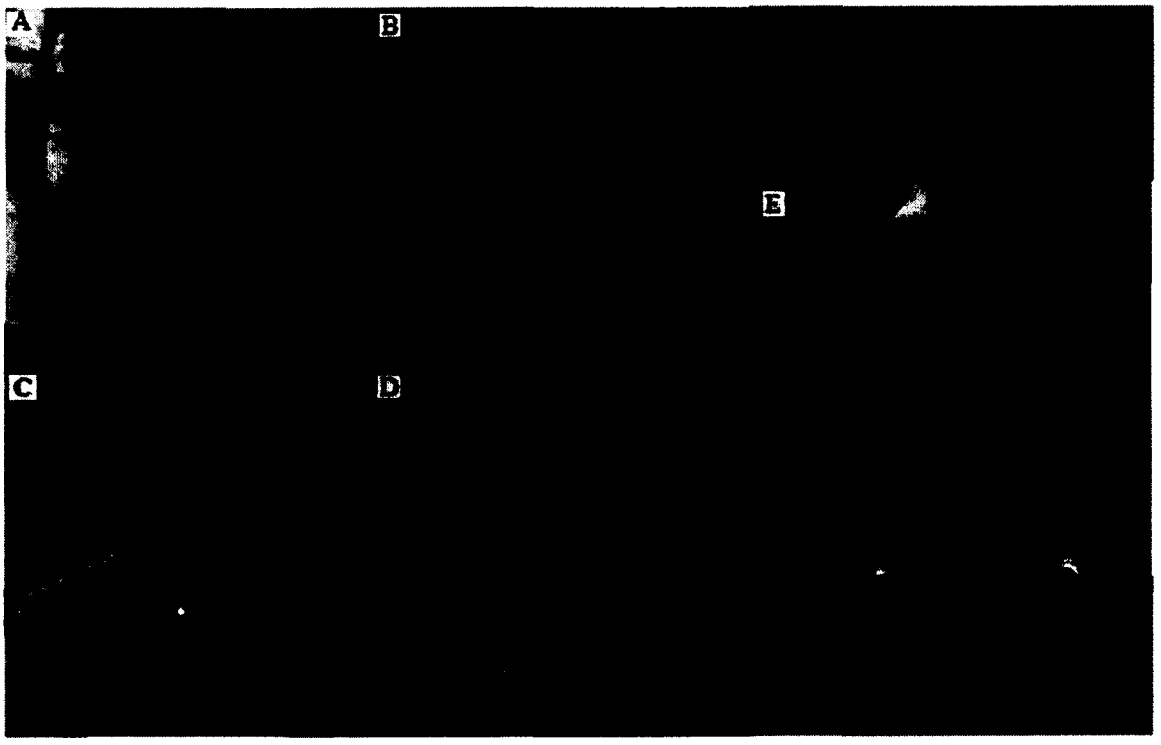


Figure 11: SENIAM guidelines for electrode placement on the A) biceps, B) triceps, C) trapezius medius muscles, D) erector spinae, and E) deltoideus medius (Seniam, 2010).

To simplify data analysis, each muscle had an electrode transmitted to the same analog channel for all subjects. Therefore, Table 4 below describes the setup that was used for all subjects for the electrode number and receiving analog channels for each muscle. The only variable that may have changed for each subject was the electrode number; since there were 16 electrodes to choose from, and ten needed, the electrodes that were fully charged were the ones used each day data were collected. However, this had no effect on the analog channel that was used to output the EMG signal into the VICON workstation on the computer.

Table 4: Corresponding analog channels for each muscle and transmitting electrode used

Muscles on Right Side	Transmitting Electrode	Analog Channel Acquired On	Muscles on Left Side	Transmitting Electrode	Analog Channel Acquired On
Biceps	1	33	Biceps	2	34
Triceps	3	35	Triceps	4	36
Trapezius	5	37	Trapezius	6	38
Deltoids	7	39	Deltoids	8	40
Erector Spinae	9	41	Erector Spinae	10	42

3.2.3 Force Plates

Force plates were used to determine the key incidents during the gait cycle; foot strike, opposite toe off, opposite foot strike, toe off, foot strike. There are four piezoelectric force plates embedded in the center of the laboratory floor. These force plates were supplied by Kistler Group, and are used in dynamic biomechanics applications. Each plate is type 9281C, the technical data is shown below in Table 5; taken from the user manual.

Table 5: Kistler force plate technical data obtained from user manual

Technical Data

Dimensions		mm	600x400x100
Measuring range	F_x, F_y	kN	–10 ... 10
	F_z	kN	–10 ... 20
Overload	F_x, F_y	kN	–15/15
	F_z	kN	–15/25
Linearity	%FSO		<±0,5
Hysteresis	%FSO		<0,5
Crosstalk	F_x <-> F_y	%	<±1,5
	F_x, F_y -> F_z	%	<±1,5
	F_z -> F_x, F_y	%	<±1,0[1]
Rigidity	x-axle (a_y = 0)	N/µm	≈250
	y-axle (a_x = 0)	N/µm	≈400
	z-axle		
	(a_x = a_y = 0)	N/µm	≈30
Natural frequency	f_n (x, y)	Hz	≈1 000
	f_n (z)	Hz	≈1 000
Operating temperature range		°C	0 ... 60
Weight		kg	16
Degree of protection	EN 60529:1992		IP65

[1] inside sensor rectangle

3.3 Test Procedure

Upon arrival the subject was asked to change into a specific top that had most of the back missing so that there was minimal to no rubbing of clothing against the electrodes. After this was done, some preliminary data was collected by taking some anthropometric measurements of particular limb segment widths and lengths, as well as the subject's height and weight. Additionally, there were a few questions asked pertaining to the subject's dominant limb side and how they usually performed biking, and swinging (for subjects in the test group). A copy of the data sheet used to gather this information is included in Appendix B . An important step in the process for subject confidentiality was to produce a unique code for each subject, in which all consequent information and data

was filed under. This unique code contained important information about the subject including:

1. Data collection order with respect to other subjects (i.e. the first subject tested was "a", second was "b", etc)
2. First letter of the subject's first name
3. Whether or not the subject was in the control group (N), or test group (P)
4. Gender (M or F)
5. The subject's dominant side (R or L)
6. Subject age
7. Extra letter index to track multiple tests for the same subject (not applicable for this study, so all were "a")

Sample subject codes are shown in Figure 12.

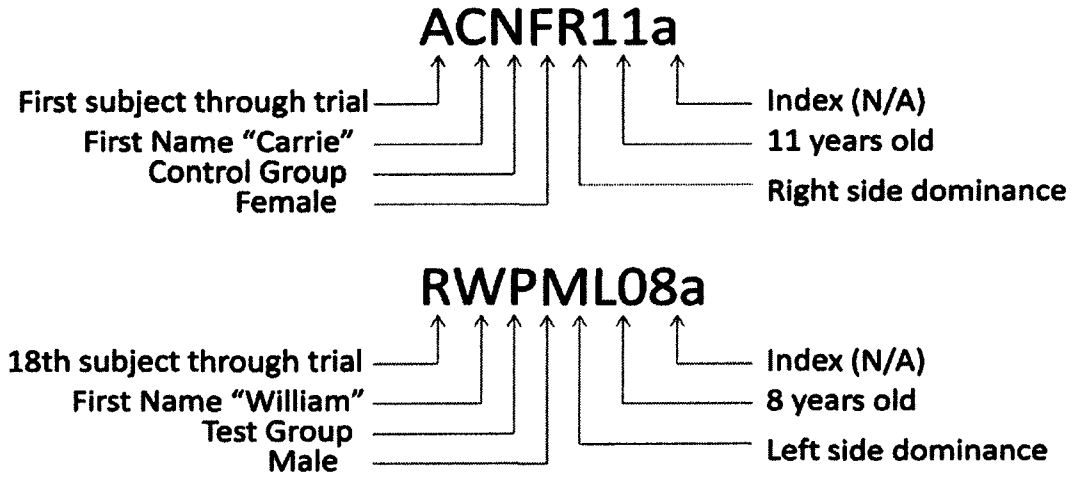


Figure 12: Unique subject code used for subject confidentiality

3.4 Marker set

The marker set used is shown in Figure 13 and explained further in Table 6. Reflective markers were placed in pre-determined anatomical locations based on the marker set used for this study. Choosing a marker set was an important step in the experimental methods based on the following factors:

- if the marker set is too complex, the activity can obstruct the reflective markers from view of the camera's,
- if the markers are too close together ghost markers or marker switching can occur,
- if the markers are placed on landmarks that are prone to high skin movement, the markers could move not relative to the task the subject is performing.

The first two scenarios outline possibilities for gaps in the data referring to the reflective markers position within the global coordinate system. If these gaps are too large, interpolation cannot be used because too much critical information was lost or the data are not cyclic. The third scenario can reduce the accuracy of marker positions relative to the subject's tasks. This is the nature of motion capture systems, and all of these factors came into consideration when choosing the marker set.

3.5 Surface Electrodes

In addition to the reflective markers, the ten electrodes that monitored EMG activity were placed based on the SENIAM guidelines for surface electrodes (refer to Figure 11). Figure 13 shows where the electrodes were placed, and Table 7 describes them in greater detail. The subject's skin was cleansed with alcohol prior to placing the electrodes, and since these were children there was minimal hair growth, so they were not shaved. This

increases the EMG signal strength because there is less resistance for the EMG signal to go through before reaching the electrode.

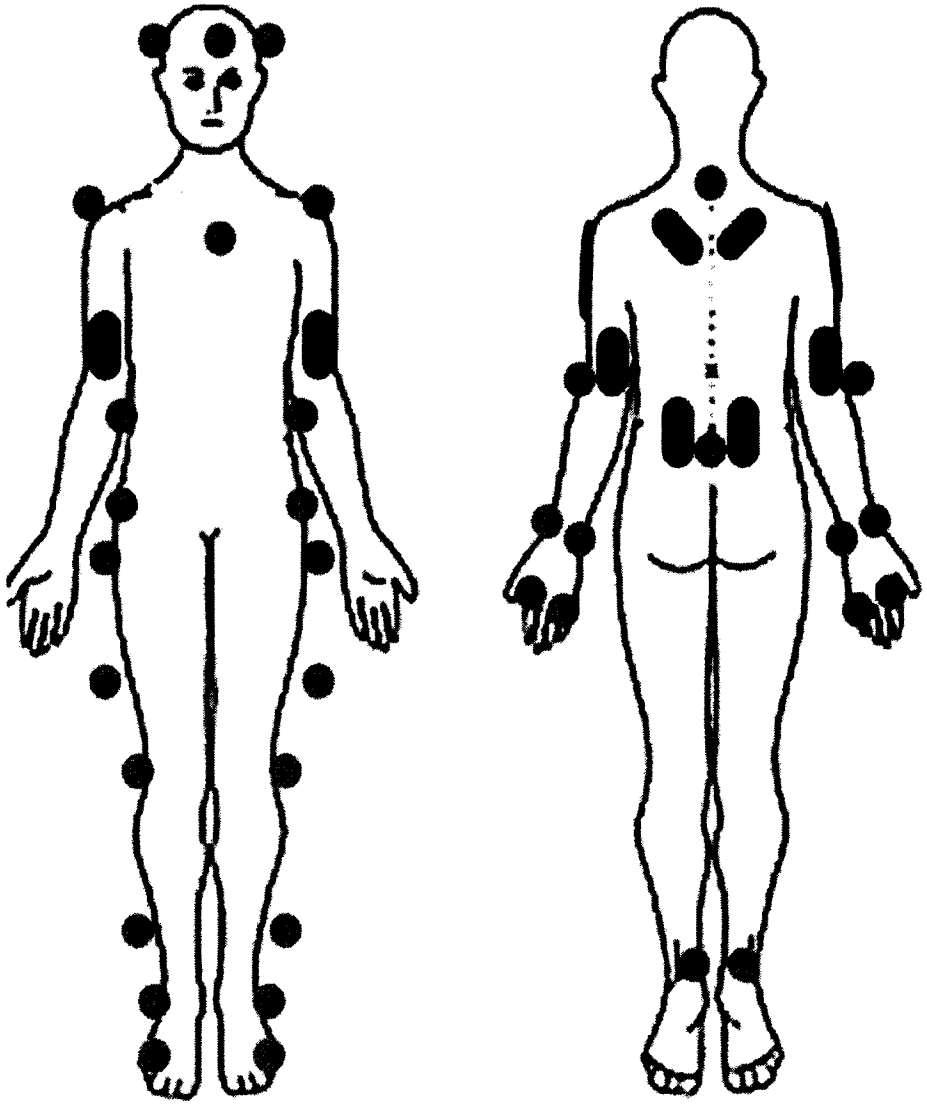


Figure 13: Reflective marker and electrode placement

Table 6: Description of anatomical marker positions and their respective sizes

Anatomical Landmark	Marker Size (mm)	Notes
Front of head	25	On band
Left side of head	25	
Right side of head	25	
Disc C7 in spine	25	
Sternoclavicular	25	Midway between left and right sternoclavicular joints
Left shoulder	25	
Left medial epicondyle of humerus	16	
Left lateral epicondyle of humerus	16	
Near left styloid process of radial	25	Placed directly on skin,
Near left styloid process of ulna	25	
Left 2nd metacarpal head	16	
Left 5th metacarpal head	16	
Front of left ASIS	25	
Left greater trochanter	25	
Left thigh	25	On band with wand
Left lateral epicondyle of femur (knee)	25	
Left leg (shin)	25	On band with wand
Left lateral malleolus of fibula (ankle)	25	
Left posterior calcanous (heel)	25	
Left 5th metatarsal (pinky toe)	25	
Right shoulder	25	
Right medial epicondyle of humerus	16	
Right lateral epicondyle of humerus	16	
Near right styloid process of radial	25	Placed directly on skin
Near right styloid process of ulna	25	
Right 2nd metacarpal head	16	
Right 5th metacarpal head	16	
Front of right ASIS	25	
Flat part of sacrum	25	On band with wand
Right greater trochanter	25	
Right thigh	25	On band with wand
Right lateral epicondyle of femur (knee)	25	
Right leg (shin)	25	On band with wand
Right lateral malleolus of fibula (ankle)	25	
Right posterior calcanous (heel)	25	
Right 5th metatarsal (pinky toe)	25	

Once the electrodes and reflective markers were applied, the subject was taken to the "Slicer Platform" where 2 simultaneous pictures were taken equidistant from the platform; one of the sagittal plane and one of the frontal plane. Figure 14 below shows the platform. This data was later used in data analysis for another study to estimate each limb segment's weight and inertial properties.

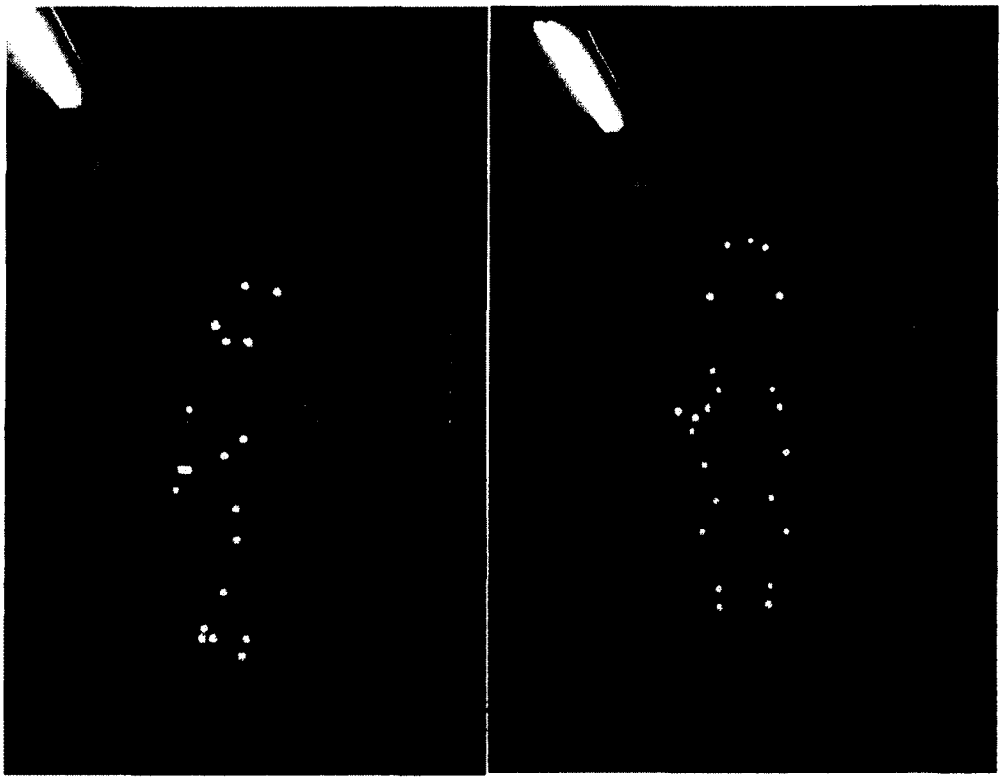
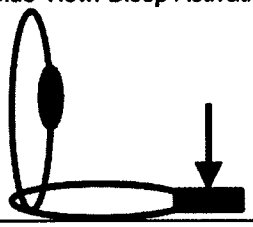
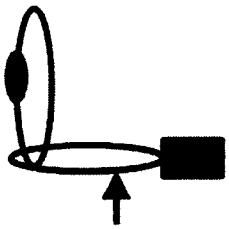
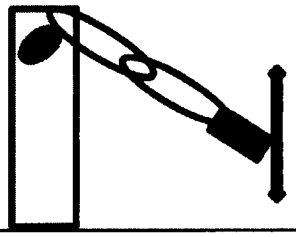
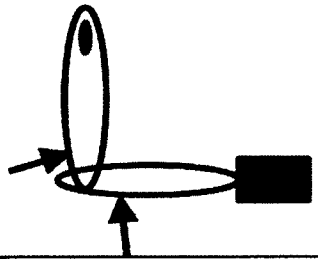
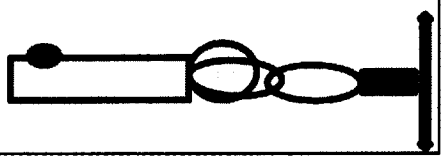


Figure 14: Picture taken on the left is the sagittal plane of a subject within the normal group, and the picture on the right is the frontal plane of a subject from the test group

Each muscle was tested using a recognized clinical test (Seniam, 2010) listed in Table 7 below, this was done to ensure appropriate electrode placement.

Table 7: Electrode placement and clinical test protocol

Muscle	Electrode Placement	Test Electrode Placement	Illustration
Biceps	2/3 the way from the acromion to the fossa cubit	Elbow bent at 90^0 with palm facing up. Resist downward force on palm.	Side View: Bicep Activation 
Triceps	1/2 way between the dorsa acromion and the end of the elbow; 2 finger widths towards backside of arm	Elbow bent at 90^0 with palm facing down. Resist force applied across body on forearm.	Top View: Tricep Activation 
Trapesuis Medialis	1/2 way between the imaginary line made by the scapula and T3	Sitting, have arms at sides, raise each arm keeping it straight out to the side.	Back View: Trapezuis Medialis Activation 
Deltoidus Medialis	On line between lateral acromion and lateral epicondyle on largest muscle bulge	Elbow bent at 90^0 with palm facing down. Resist downward diagonal force on either side of elbow.	Top View: Deltoidus Medialis Activation 
Erector Spinae	2 finger widths on either side of L1	Lying down on stomach, lift arms and stomach off platform and repeat.	Side View: Erector Spinae Activation 

3.6 Activities of Daily Living

The purpose of this experiment was to monitor the subjects while performing two to three activities of daily living. These activities had to conform to the following criteria:

- Tasks that children normally perform
- Tasks that are age appropriate (4 years to 14 years of age)
- Tasks which require the use of the larger muscles in the upper body to encourage gross motor movement
- Tasks that require minimal props which could block the view of cameras
- Tasks which can be performed in a short amount of time
- Tasks that could be as symmetric as possible regarding movement of both sides of the body
- Tasks as cyclic as possible

These criterion were constructed based on previous work in the clinic (Ross, 2005; Zinck, 2008).

Based on these criterion, the three activities of daily living chosen were:

- swinging on a swing,
- biking on a stationary bike and,
- walking along a level walkway.

For the control group, each subject was only required to perform one round of each activity. One round consisted of the subject performing each activity between four and eight times for the swing and bike tasks, depending on their attention span and whether or

not they were performing the tasks as they usually would. The subject's performed between 10 and 20 walking trials because clean foot strikes on two out of the four force plates were necessary and difficult to obtain at times. The subject's parents were able to give a good indication of whether or not they were performing the task as they usually would or not. These activities were performed in the same order for each subject so that the walking task was first, followed by the swing, and ending with the bike task. This was done for two reasons, one so that the most interesting and fun tasks were performed last to attempt to keep the subject attentive when performing all tasks. The second reason was that if any fatigue factored into each subject it would be relatively consistent for each task for different individuals because they were performed in the same order.

The test group had to perform two rounds of each activity, one while wearing their prosthesis and another while not wearing it. For this group, the task order and whether or not the prosthesis was used was randomized. This was done to decrease the effect that fatigue may have when comparing the two conditions (wearing versus not wearing) within the same subject. For example, if the tasks were not randomized, the prosthesis was worn for all tasks and then taken off and the process repeated, there could be a fatigue factor that could make the subject alter what they would normally do when not wearing their prosthesis.

3.6.1 Swing Task

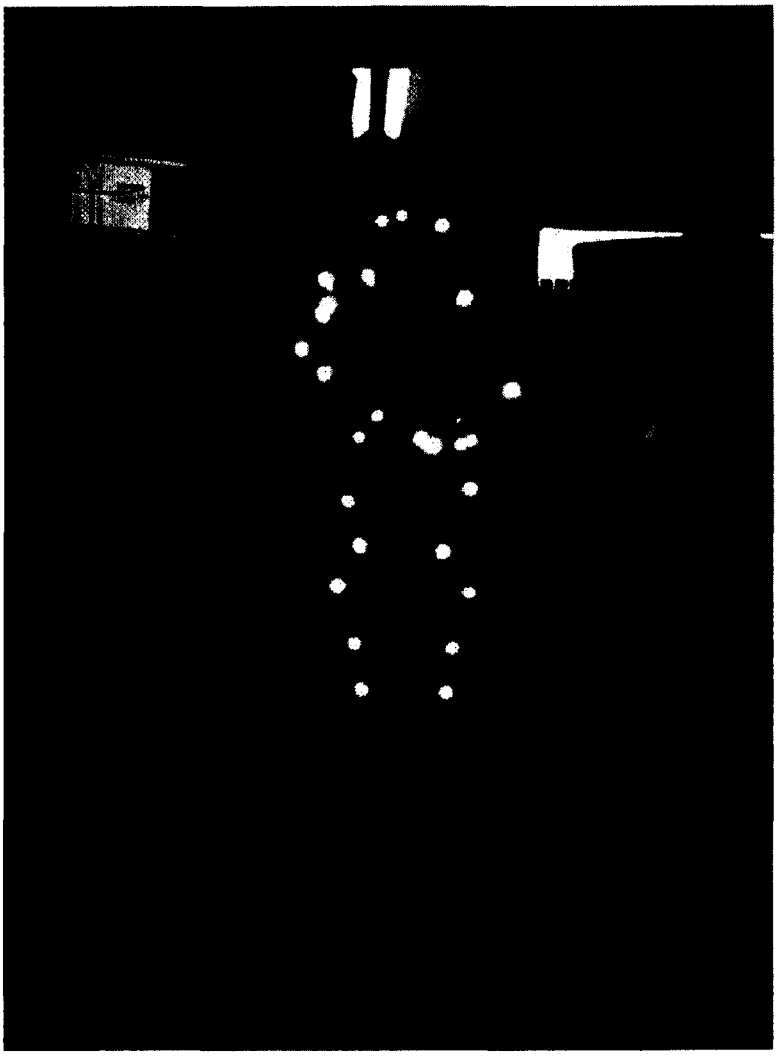


Figure 15: Test subject, wearing a prosthesis while performing the swing task.

The swing task is something that children may do on a regular basis and provides an activity that is highly cyclic, making it relatively simple to observe. In addition, swinging requires upper body muscle activity, making it an ideal task for this experiment.

3.6.2 Bike Task

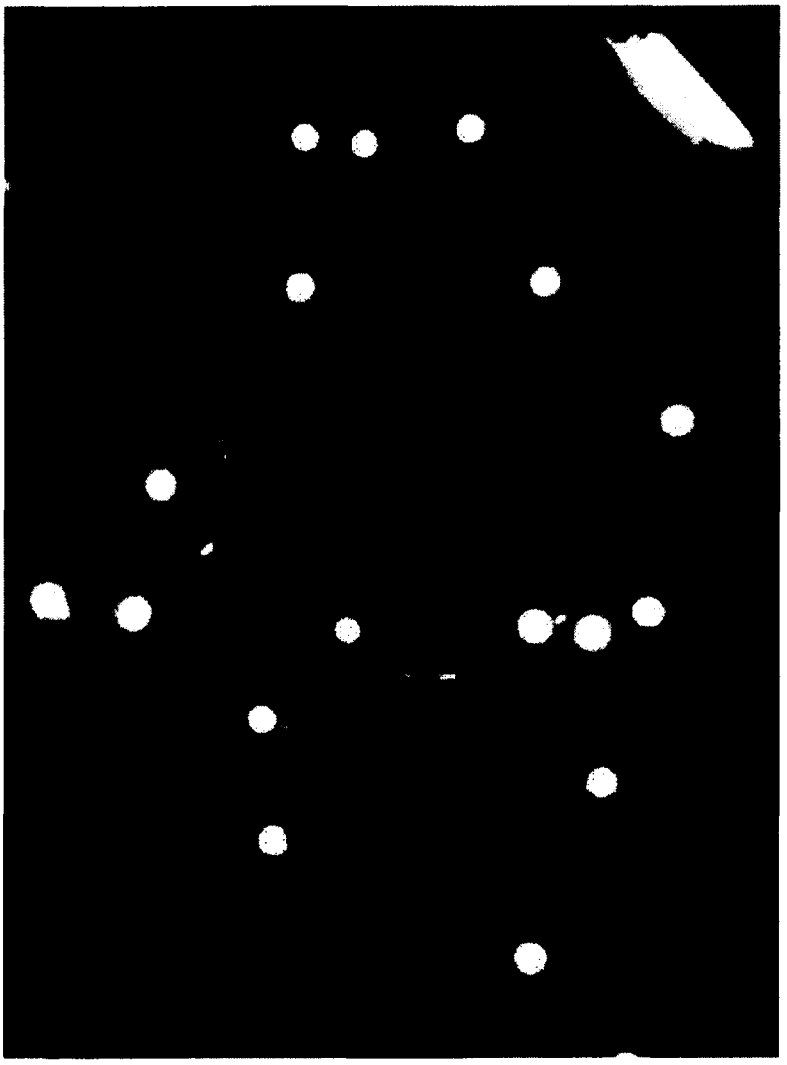


Figure 16: Test subject, wearing a prosthesis while performing the bike task.

Biking is another task that may be performed on a daily basis by children, especially if they ride their bike to school. During biking, it was fairly consistent that children did not use their upper body muscles as much as they did during the swinging task. The bike was made stationary by a custom stand that was fabricated specifically for this task. Figure 17 below shows the stationary bike used for these experiments.

Figure 17: Stationary bicycle that was used in lab.

The stand was attached to the back tire and had two stiff springs that allowed for a more realistic biking experience, allowing the subject's to lean a little during turning. The stationary bike was positioned in the center of the lab to ensure maximum visibility of the reflective markers for the cameras. The lab technician gave the subject an overview of what they were supposed to do. For example, first bike straight, then turn left, then bike straight, then turn right, then bike straight, then stand up and pedal; each for five seconds. Once the subject knew what to expect, the subject was asked to start and given cues every five seconds on what to do. Therefore, each bike trial lasted approximately 30 seconds (6 segments times 5 seconds each).

3.6.3 Walking Task

This was an excellent task for this study, as it is familiar, is cyclic and offers symmetry. Additionally, there is little literature currently available on upper body muscle activity during gait (especially in the prosthesis user group).

The gait cycle is defined as the sequence of events that occur between two successive initial contacts of the same lower extremity (Seymour, 2002), or from foot-strike to foot-strike of the same limb. The various phases of the gait cycle are described below in Figure 18.

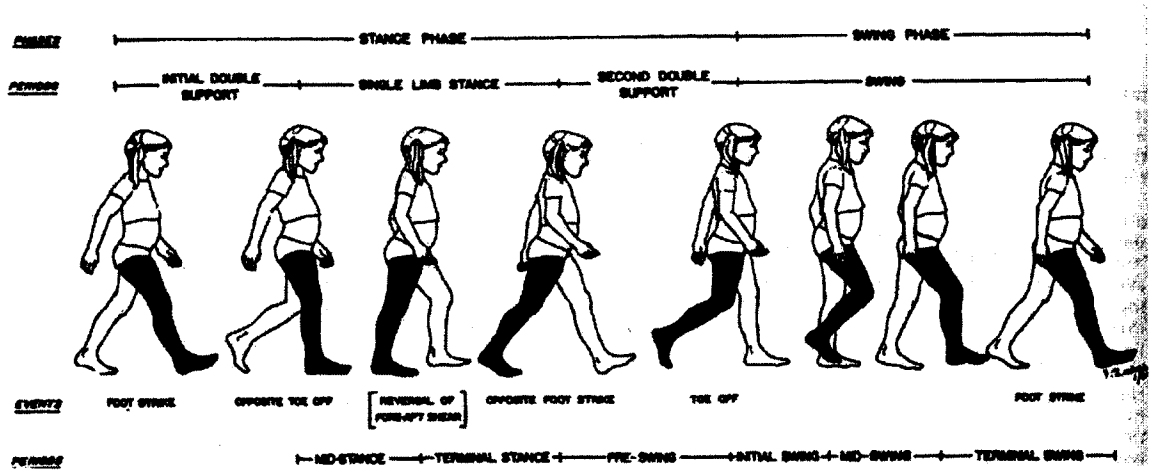


Figure 18: describes the various phases of the gait cycle (Sutherland, 1981).

Stance phase refers to the amount of time the limb is touching the ground (60% of gait cycle), and the swing phase refers to the amount of time the same limb is not touching the ground. Clinical gait analysis aims to quantify and assess the mechanics of walking, and facilitates the identification of deviations from normal movement patterns (Chester et al., 2005). This was done as a small segment of this project, being one of the three gross

motor tasks analyzed for each subject. The lower extremity movements during the gait cycle are highly cyclic and repeatable, therefore knee flexion was captured to assist in identifying the timing of the gait cycle. Foot strike and toe off gait events will be determined by markers on the foot coupled with the force plate data. Unlike the lower extremities, the variety, complexity, and range of upper-extremity movements is a big challenge to assessment and interpretation of data, and even more complicated in clinical application (Rau et al., 2000). Therefore, the EMG data collection in conjunction with the motion data is critical, because it allows for comparisons of upper extremity muscle activation patterns between individuals. Also, the addition of EMG to gait analysis provides information about muscular coordination and muscle recruitment by the central nervous system (Rau et al., 2000; Schmidt-Rohlfing et al., 2006). Force plate data allows for the determination of foot-strike as well as providing information on the ground reaction forces and center of pressures acting during gait

3.7 Post Collection

Once all of the tasks had been completed, the removal of the electrodes and reflective marker balls took place. The children were allowed to take the reflective markers off themselves, however, due to the cost of the EMG sensors, they were taken off by the lab technician. The parents and child were then able to leave and asked if they would like to be informed of the results of the project.

4.0 Data Analysis

Electromyography data were used to determine what muscles were active during an entire cycle of each task. This analysis was simplified for the gait and swing tasks because they were both highly cyclic in comparison with the bike task. The bike procedure was intended to be as predictable as possible, by having five various segments (biking straight, turning left, biking straight, turning right, biking straight, standing up) go on for approximately five seconds each. The person conducting the test verbally instructed the subject to change at the beginning of each five second interval.

The muscle activity was determined to be "on" or "off" based on a calculated threshold value (this is described in more detail in the Muscle Onset Timing section). The MATLAB programming code for reading the EMG signal and performing these calculations was completed. Once the activation timing for each muscle was determined it was described as a percentage of each activity's overall time. For example, for the walking task, the muscle activity of the 10 muscles was described in reference to the gait cycle. The force plates were on during the walking task and were used in conjunction with the EMG and motion data when analyzing gait.

Software used for data analysis include, Vicon workstation software, MATLAB, Minitab, and Excel. The analog data were sampled at a rate of 1080 Hz, which was at least two times the bandpass output of the Zero-wire electrodes (between 10 and 500 Hz). The video data were captured at a rate of 60 Hz, which was at least double the highest

frequency occurring from the gross motor activities. To determine the highest frequency for the motion data during the three tasks, the point with the highest and quickest movement during each task was chosen and a Fast Fourier Transform (FFT) was performed. For the swing task, the motion of the right toe was chosen and seems to occur at a frequency between 0 and 2 Hz. The motion of the right knee was chosen for both the biking and walking tasks, and occurs at approximately 1and 2 Hz respectively.

4.1 Vicon Processing

The first step, post data capture, was to use the Vicon software provided, called the Vicon Workstation. Using this software, the marker-set (Figure 13 & Table 6) was added to each trial; for every subject, and the corresponding trajectories were labelled using this marker-set. Any additional trajectories not corresponding to the marker-set (reflections off other camera's or on objects within the lab workspace) were deleted. Figure 19 shows a snapshot of the Vicon workstation display before and after processing.

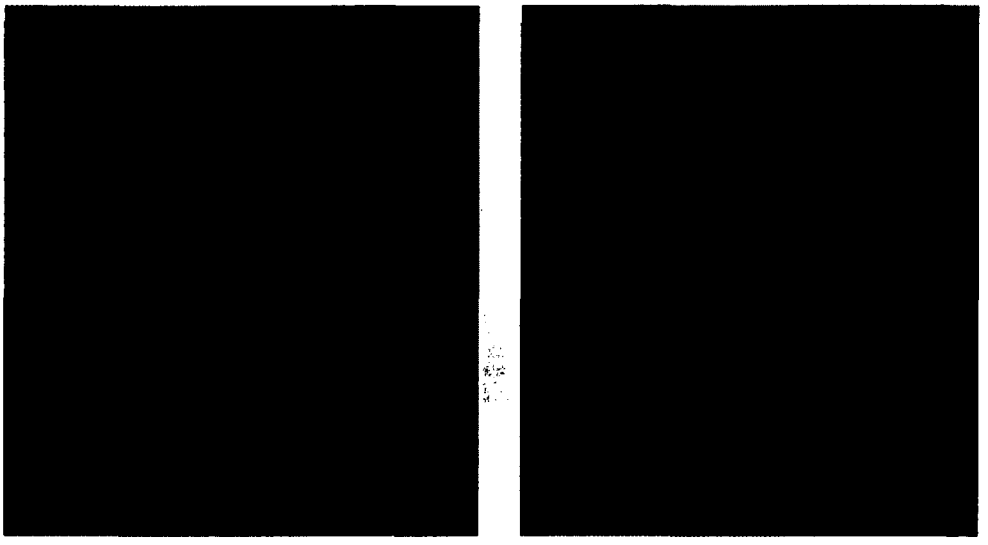


Figure 19: The left side of this figure is a raw VICON file prior to processing. The right side of this figure shows the same trial fully labelled.

4.2 Trial Selection

As mentioned previously, each task was completed approximately five times for the bike and swing activities and approximately 10 times for the walking task. The walking task was done more times in order to capture a good trial. Certain factors that were considered when choosing the trial were:

- maximum visibility of all markers
- amount of background noise (this was not very likely to change between trials within the same person as the system settings were never altered)
- muscle signal strength (sometimes during a trial an electrode may have slipped and had to be reattached)
- electrodes did not come off or slip during trial
- subject was performing the task the way they would "normally" do so. For the control group this was monitored by their parents and for the test group it was monitored by the occupational therapist at the clinic
- subject was attentive throughout the entire trial

Each of these factors was noted throughout each trial, if they were not met, then the trial was not considered for selection. For each task, part of the data analysis was segmenting the trials into cycles and calculating an average of these cycles for the task. Table 8 below shows each trial chosen for all three tasks.

Table 8: Describes the trials analyzed for the Swing, Bike and Walking tasks

Subject Code	Walking	Swing	Bike
amNFR11a	*Walk12*	*Swing07*	*Bike02*
bgNFR10a	*Walk05*	*Swing02*	-
ccNFR11a	*Walk09*	*Swing01*	*Bike02*
dkNFR07a	*Walk11*	*Swing01*	*Bike02*
ejNMR11a	*Walk09*	*Swing03*	*Bike01*
fpNMR14a	*Walk06*	*Swing02*	*Bike07*
gzNML11a	*Walk09*	*Swing04*	*Bike04*
hjNMR08a	*Walk13*	*Swing05*	*Bike04*
icPML14p	*Walk10*	*Swing04p*	*Bike02p*
icPML14n	*Walk09*	*Swing04n*	*Bike02n*
lsNMR09a	*Walk11*	*Swing02*	*Bike03*
mwNMR06a	*Walk10*	*Swing04*	*Bike05*
nvNFR11a	*Walk10*	*Swing02*	*Bike03*
osNFR11a	*Walk04*	*Swing05*	*Bike04*
paPFL10p	*Walk03*	*Swing04p*	*Bike04p*
paPFL10n	*Walk12*	*Swing01n*	*Bike02n*
rcPFR06p	*Walk09*	*Swing02p*	*Bike01p*
rcPFR06n	*Walk10*	*Swing03n*	*Bike03n*
smNFR07a	*Walk11*	*Swing03*	*Bike03*
tmNMR04a	*Walk07*	*Swing04*	*Bike04*
ukNFR07a	*Walk07*	*Swing04*	*Bike02*
vtNML11a	*Walk08*	*Swing02*	*Bike01*
wrPFR13p	*Walk06*	*Swing03p*	*Bike01p*
wrPFR13n	*Walk05*	*Swing08n*	*Bike04n*

4.3 MATLAB Programming

MATLAB (MathWorks Laboratory) was used to write code that was specifically tailored for the data collected. Previously written code (Chester, 2005) was used to determine the gait cycle and segment the analog data to match. New code was written in order to analyse the muscle activity in all three tasks as well as the determination of the swing and bike cycles.

A detailed description of the necessary steps taken during data analysis is as follows:

4.3.1 Converting VICON trial files to C3D files

The next step was using a previously written MATLAB program called "getc3d" (See Appendix CI) which opened the Vicon trial to be analysed and then called the function "labelc3d4p" (See Appendix CII) which required the marker-set used. The marker-set varied depending on the task and the condition of the subject. For example, the bike task required the use of shoes, so the ankle and toe markers were no longer included; only the heel marker. Also, if a prosthesis user was performing the experiment while not wearing his/her prosthesis, there were no markers in the absence of a limb. Figure 20 below shows three examples of variations of marker-sets used throughout the experiment.

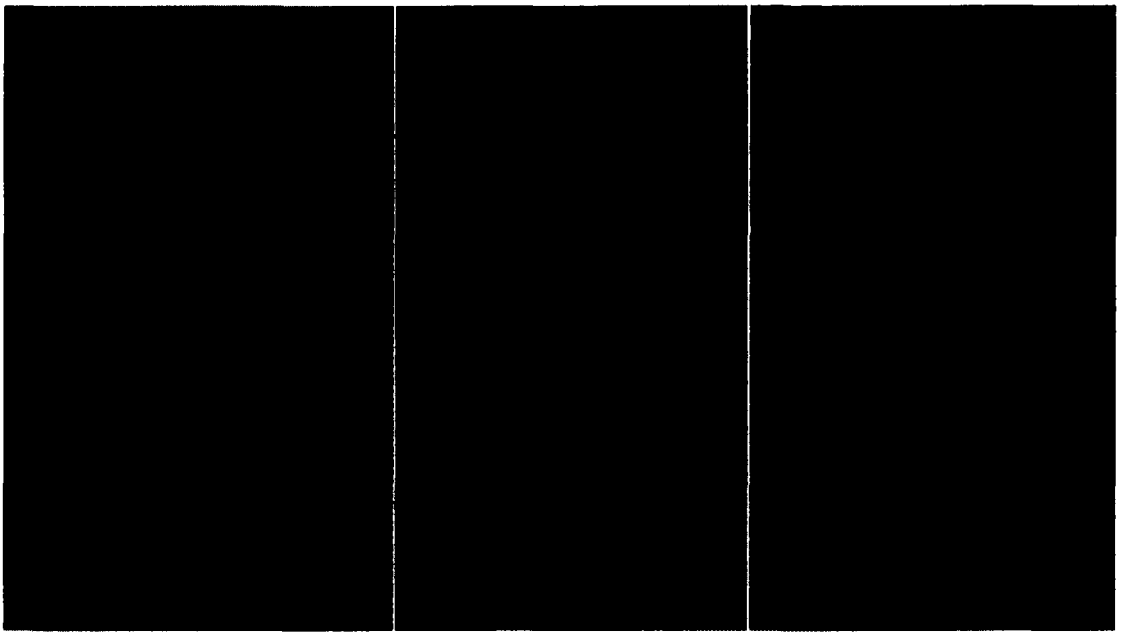


Figure 20: Full marker-set for either a normally limbed child or a child with limb loss while wearing his/her prosthesis (Left), Full marker-set on a child with limb loss while he/she was not wearing prosthesis (Middle), Full marker-set of normally limbed child during biking (Right).

After the correct marker-set was input into the "labelc3d4p" program, "getc3d" was able to fully convert the VICON files into .c3d files. All variables calculated during the conversion were saved in the format of a .mat file, and used later on in analysis. From this stage on, the process was task dependent.

4.3.2 Converting Bits to Volts

Before any data was analysed, the muscle output had to be converted from bits to volts. This was done based on the following:

0 Volts =32768 bits

10 Volts = 62536 bits

Figure 21 represents the conversion from bits to volts, which is an illustration of Equation 1 below.

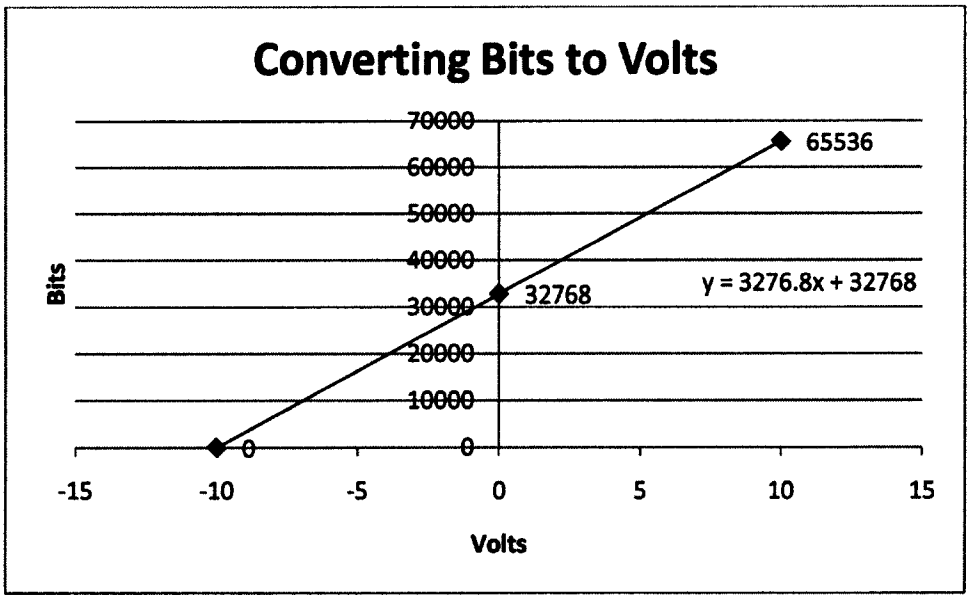


Figure 21: Graph showing the conversion of bits to volts

Equation 1: Conversion from bits to volts

$$x = (Y - 32768)/3276.8$$

where, x=volts; Y=bits

Each analog output (EMG), therefore, was first subtracted by 32768 and then divided by 3276.8 to convert from bits to volts.

4.3.3 Resolution

The resolution for detecting muscle onset was approximately 3 mV, which was directly related to the analog-to-digital conversion. The voltage resolution of an analog to digital converter (ADC) is equal to the voltage range divided by the number of discrete voltage intervals. Equation 2 below describes this, which is how the resolution of the system was found.

The voltage resolution of an ADC is equal to its overall voltage measurement range divided by the number of discrete voltage intervals:

Equation 2: Voltage resolution of an analog to digital converter

$$Q = \frac{V_{hi} - V_{low}}{2^m}$$

where, Q= Voltage Resolution; m=number of bits,

Therefore,

$Q=(10-(-10))/2^{16}$

Q=20/65536 = 0.00305 Volts, or **Q=3.05 mV**

4.3.4 Filtering

The bandpass frequency the zero-wire EMG system operated at is 10-500 Hz. In order to satisfy the Nyquist theorem (capture rate must be at least twice the highest frequency), the capture rate had to be at least 1000 Hz. Therefore, analog data was collected at a frequency of 1080 Hz. This capture rate was chosen as opposed to 1000 Hz so that matching to the video sample rate (60 Hz) was an integer; for example, for every 1 video frames, there was 18 (1080/60) analog frames. If 1000 Hz was used for the analog data, this would not have resulted in an integer.

The raw EMG signal was too noisy, and could pick up individual motor action potentials. For determining whether a muscle was "on" or "off", this was inappropriate based on the criteria that muscle activity must stay above the threshold level for a specified amount of time. For this experiment 50 ms was used (Hodges & Bui, 1996; Soderberg & Cook, 1984; DiFabio, 1987). If the noise level of a signal is high (i.e. raw EMG signal), individual motor units firing on and off would skew the outcome of the muscle activity. An illustration of this is displayed in Figure 22 below; the left side showing the fully rectified, raw EMG signal (noisy dark line), as well as the same signal filtered (lighter line within raw signal), and the muscle threshold level (horizontal line along bottom). The right side is the same graph, zoomed into the point just after the muscle is activated. The right side of this figure shows that the filtered signal is consistently above the muscle threshold level that determines whether the muscle is in fact on or off. Based on Equation 3, for a muscle to be considered on, it must be above the threshold level for at least 25 milliseconds. The raw signal dips below the threshold numerous times within

this 25 ms period. Therefore, if the raw signal was used to determine muscle activity, this portion of the muscle signal would be determined as off, when in fact it is on.

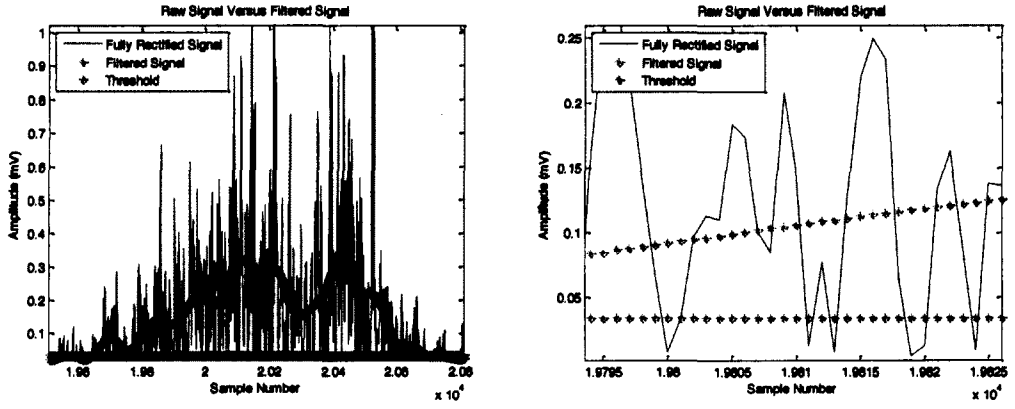


Figure 22: Raw EMG signal versus Filtered EMG signal

The raw EMG signals were full wave rectified in order to perform EMG onset calculations. Due to the fact that Zero-Wire operated within the bandpass frequencies of 10-500 Hz, the data were filtered with a bandpass filter of 10-500 Hz as the lower and upper limits (using a 4^{th} order Butterworth filter with zero phase-shift). Additionally, each signal was low-pass filtered (6^{th} order elliptical filter with 0 phase-shift) with a cut-off frequency of 50 Hz. This process was done based on previous works (Hodges & Bui, 1996; Soderberg & Cook, 1984; DiFabio, 1987).

When determining EMG onset timing, the amount of filtering is critical. Too much filtering will result in a loss of pertinent information including potentially informative EMG peaks and valleys. Filtering can also affect the apparent timing of muscle activity. The usage of two differing Matlab functions for filtering is shown in Figure 23. The top portion of the graph was filtered using the MATLAB function of FiltFilt, which results in zero phase shift. The bottom portion of the graph was filtered using the MATLAB

function Filter, which had forward phase shift. This phase shift is seen as the difference between the two lines close together, which can be calculated for either of the two examples (two sets of two lines close together) shown on the graph. This figure describes the subtle differences in muscle onset timing when using the function FiltFilt as opposed to Filter. When using Filter, there was a slight phase shift in the data on the x-axis. This occurred because Filter filters the data only forward, which means the data is filtered and then displayed; so there is a small shift in the real time data and the filtered data displayed. After filtering data in the forward direction, FiltFilt then reverses the filtered sequence and runs it back through the filter, which results in zero-phase distortion of the original signal. For consistency and accuracy, the FiltFilt function was used to ensure no phase-shift in the time series occurred.

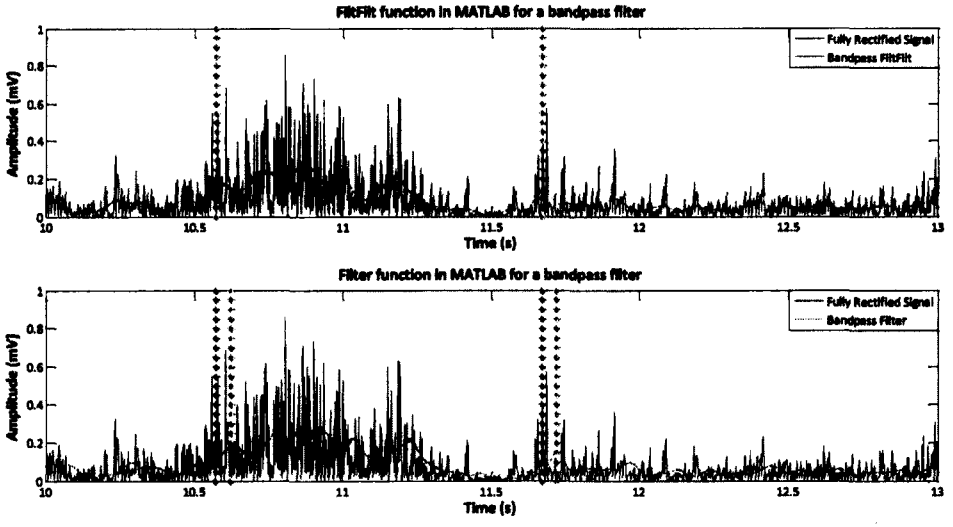


Figure 23: This figure describes varying filtering options for processing the EMG signal

4.3.5 Trial Segmenting

The purpose of this study was to determine muscle activity relative to each task's cycle. Since the swing and walking tasks were highly cyclic, it was not necessary to segment the data prior to cycle determination. The bike task, however, was not a cyclic task, and therefore, needed to be segmented first.

For segment determination, there were four markers that were simultaneously graphed: the right or left hand marker, the sacral marker or the front head marker, and the right knee marker. By observing the sacral, hand marker, and front head marker, it was simple to determine when the subject turned, and when the subject was standing up. The beginning and end of the turning and standing up events were the points at which the data was segmented. The right knee marker was tracked to ensure there were no gaps in the data, as this is the marker that would later be used for determination of the bike cycle.

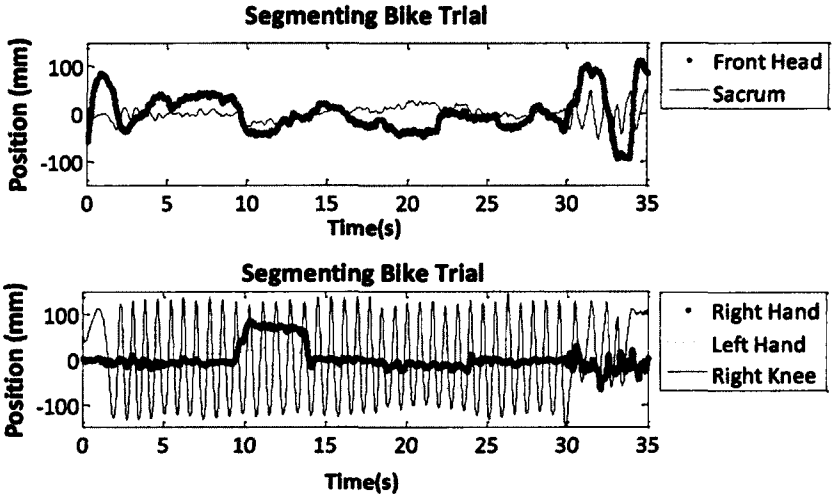


Figure 24: Top shows the right knee marker throughout the entire bike trial; Bottom shows the right hand marker as well and the segmentation points on the bike trial.

The top portion on Figure 24 tracks the right knee marker during an entire bike trial; the bottom shows the right hand marker and the subsequent segmenting of the bike trial. Figure 25 describes the same information with the determined points to segment the data.

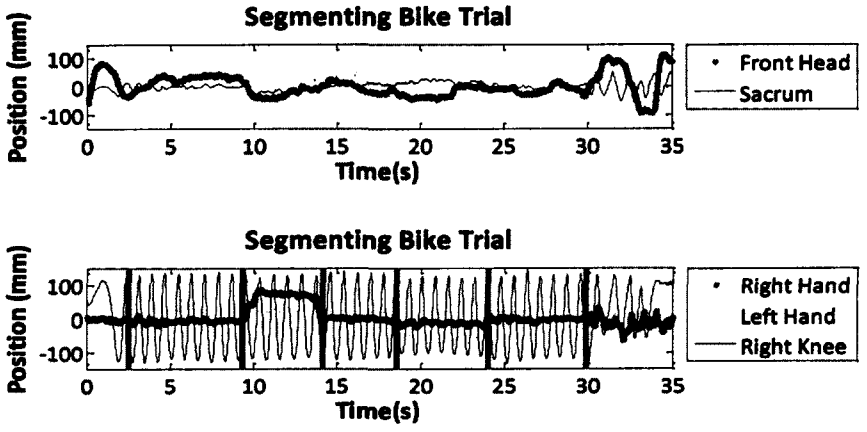


Figure 25: Shows the determination of each segment (Red line) based on the right and left hand, front head, sacrum and right knee markers.

Once the points of segmentation were chosen, each segment was saved as a separate variable; Figure 26 illustrates the six data sequences based on a full bike trial.

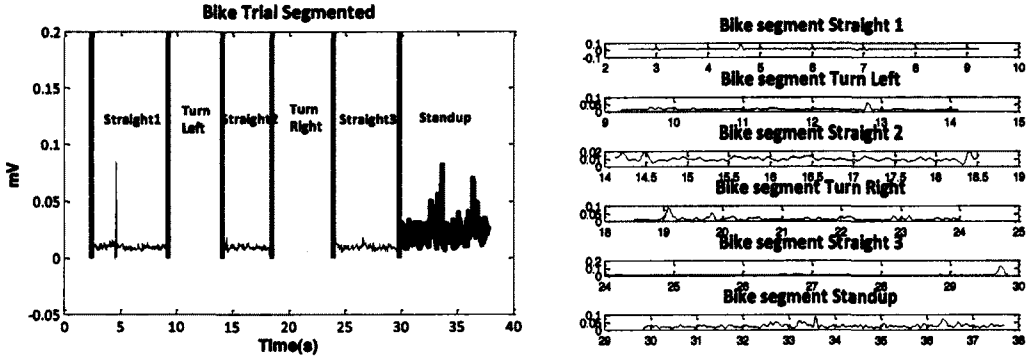


Figure 26: The left shows an entire bike task, each colour representing a new segment; the right shows each of these segments graphed separately.

4.3.6 Cycle Selection

After the data were filtered and segmented, a cycle was defined for each task.

Swing

The swing cycle was determined to start at the furthest point the toe (right or left) travelled in the backswing; the halfway point in the cycle was determined to be the furthest point the toe travelled in the forward swing; the end of the cycle was the same as the next cycle's starting point, the furthest point the toe travelled in the backswing. Figure 27 below describes the swing cycle.

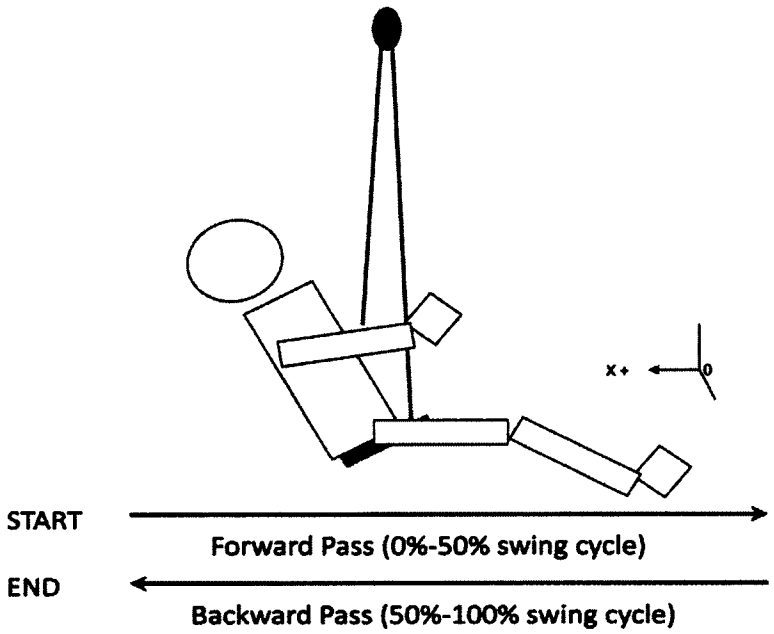


Figure 27: Description of swing cycle and coordinate system relative to swinging

Figure 27 describes the coordinate system relative to the swing task. Based on this, the maximum points the right toe travelled were actually the points where the toe is furthest back, which corresponds to the starting and ending of each swing cycle.

An example of the oscillations made by the right toe during swinging is shown in Figure 28. This particular subject slowly increased his/her swinging amplitude with each swing. This was generally the case, however some subjects got up to their highest amplitude (or steady state amplitude) faster.

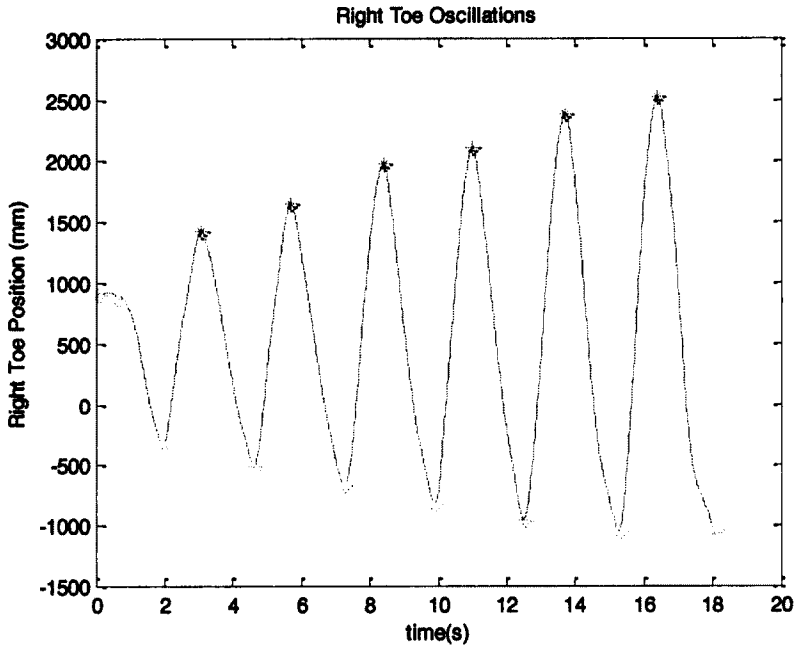


Figure 28: Right toe oscillations during swinging

For each trial, the most typical swing cycle was chosen to be analyzed. This was done by comparing all cycles occurring within each trial; taking the average of these cycles, and then choosing the cycle that most closely followed the same pattern as the average cycle. To determine the average cycle, each had to have the same number of data points. After each cycle was determined, the shortest one was measured and the rest of the cycles were re-sampled in order to have the same number of data points. The resample function in MATLAB applies a lowpass filter on the data by using "n" terms on either side of the

current sample. The the higher the value of "n", leads to greater accuracy in the resampled signal. This created longer computation times. In this case, "n" was equal to the shortest average cycle length, therefore, "n" was very large, and represented an accurate estimate.

In Figure 29, all of the cycles for Subject E were arranged on the same plot as well as the calculated average out of these cycles. The overall average was subtracted from each individual cycle to determine which cycle was closest to the average .

The closest cycle to 0 (one that most closely represented the average) was chosen; this process is illustrated in Figure 30. The selection was done by taking the difference between each cycle and subtracting the average from it. The cycle yielding the value closest to zero was the one chosen. Therefore, for subject E, trial 2, it was cycle 4 that was chosen to be analysed.

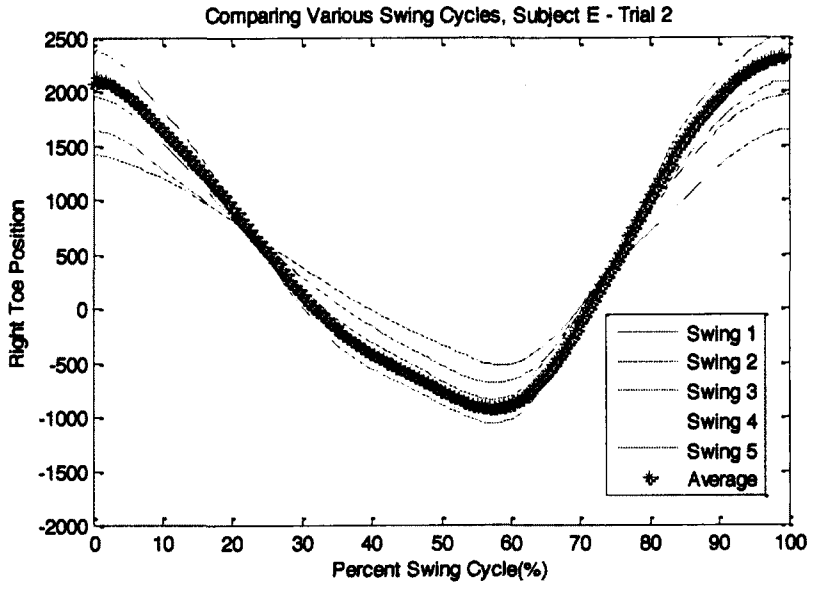


Figure 29: All cycles occurring for Subject E, trial 2 during swinging

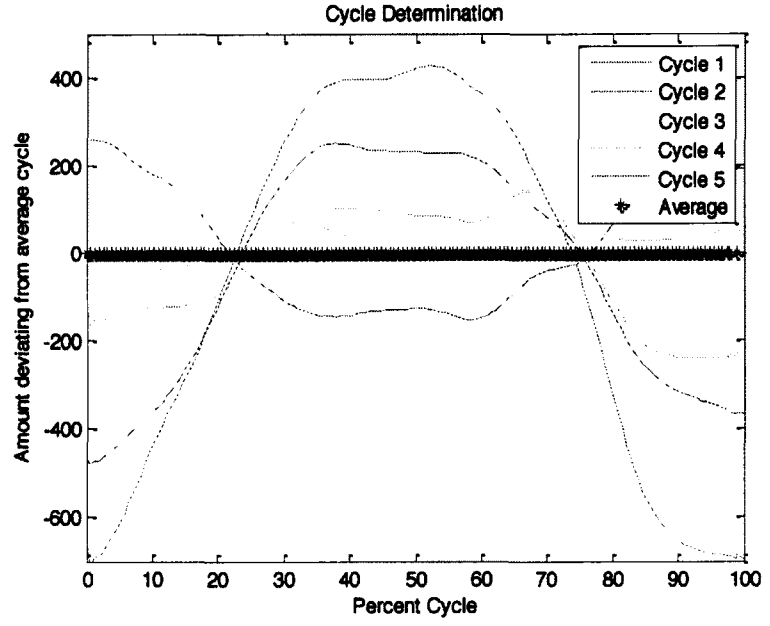


Figure 30: Determining the cycle that most closely follows the average cycle.

Once cycle selection took place, this cycle was the only one analysed for each child from this point on.

Bike

After the bike trial was segmented, the cycle for each segment was determined. This was defined as a full revolution by the pedal of the right side. The beginning of the cycle was when the right knee was lowest to the ground; 50% of the cycle occurred when the knee was at its highest point during pedaling, and the end of the cycle was the same as the next cycle's starting point which was the lowest point the knee was to the ground. Figure 31 below describes the bike cycle.

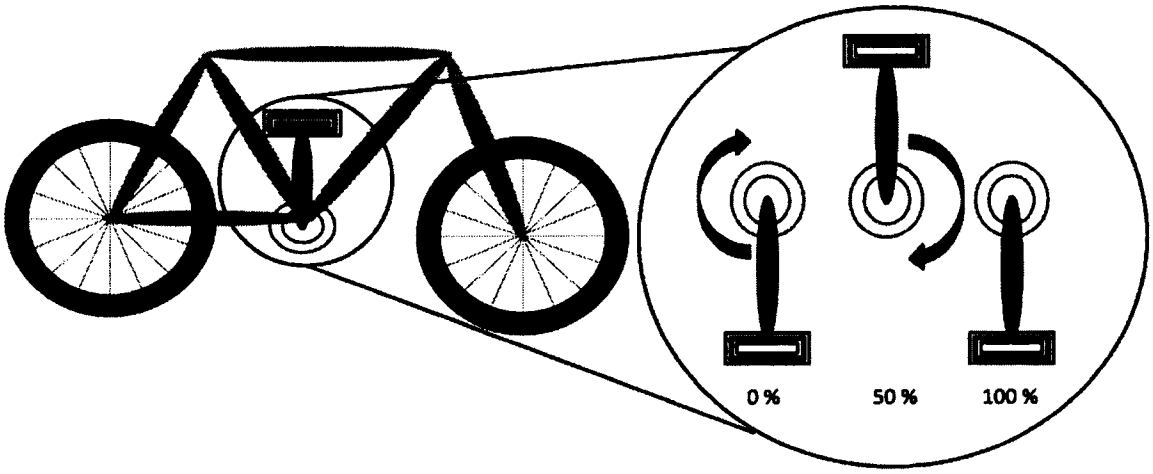


Figure 31: Illustration of the bike cycle

For the second segment of biking straight, Figure 32 shows the right knee while pedalling. The minimum points determined the start and end of the cycle, whereas the maximum points were the halfway point in each cycle.

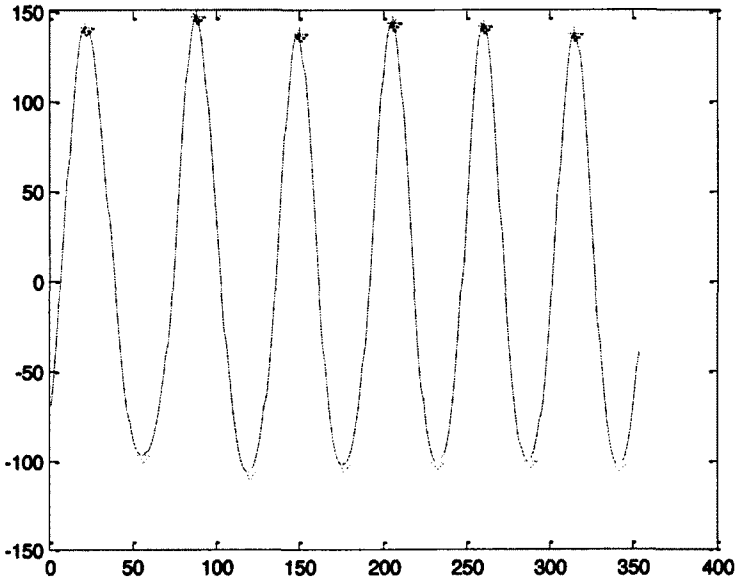
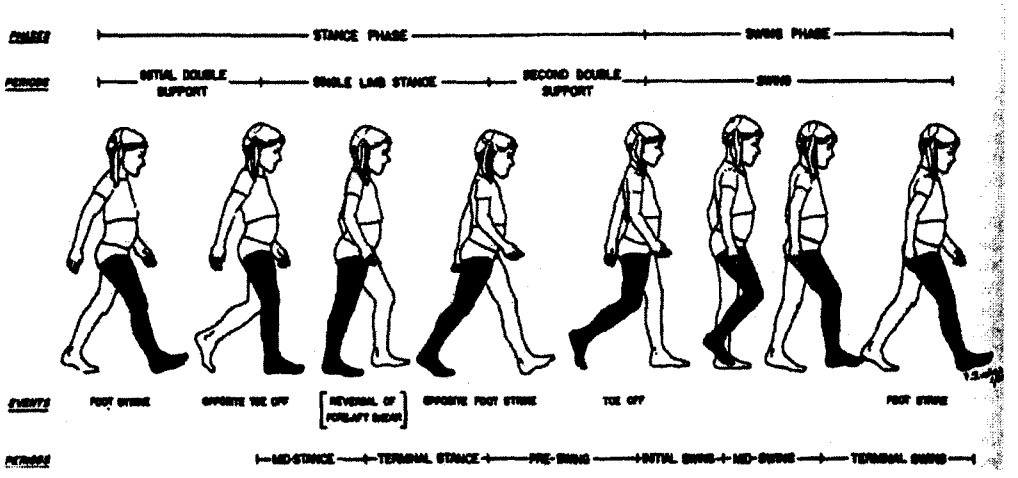


Figure 32: Right knee oscillations during bike segment 2

Similar to the swing task, the most typical bike cycle was chosen for each bike segment and analysed. The bike task was slightly more consistent because the pedal length used for the bike was the same for all subjects. The only difference between subjects was the length of their legs which would determine how high the knee started. The swing task segmenting, however, was based on the position of the right toe, whose height depended on how high the subject was able to get going during swinging.

Walking

The gait cycle is defined as the sequence of events that occur between two successive initial contacts of the same lower extremity (Seymour, 2002), or from foot-strike to foot-strike of the same limb. Figure 33 below shows the major events occurring during a typical gait cycle.



EVENT	AGE 1	3	5	7	ADULT <10
Opp. Toe-Off (%)	17	16	13	12	13
Opp. Foot Strike (%)	50	51	50	50	50
Duration of Single Stance (%)	33	35	37	38	37
Toe-Off (%)	67	66	63	62	63

Figure 33: describes the major events that occur during gait as well as time spent in single and double support (Sutherland, 1981).

The force plates on the floor of the lab were used to determine these major gait events. The muscle activity levels were observed alongside the gait cycle. The reason more trials had to be taken for gait (between 10 and 20) was that most children took some time to relax and walk in a normal fashion. Their parents were able to determine whether each trial was normal or not.

For each walking trial, the five gait events for the left and right legs were recorded (Table 9) and each cycle was determined using these points. This process is described for the left leg on the left side of Figure 34, and for the right leg on the right side of Figure 34.

Table 9: Frame numbers of gait events for the left and right legs for Subject F, Trial 06.

	Foot Strike	Opposite Toe Off	Opposite Foot Strike	Toe Off	Foot Strike
Left Leg(Trial06)	269	275	303	310	339
Right Leg(Trial06)	233	239	269	275	303

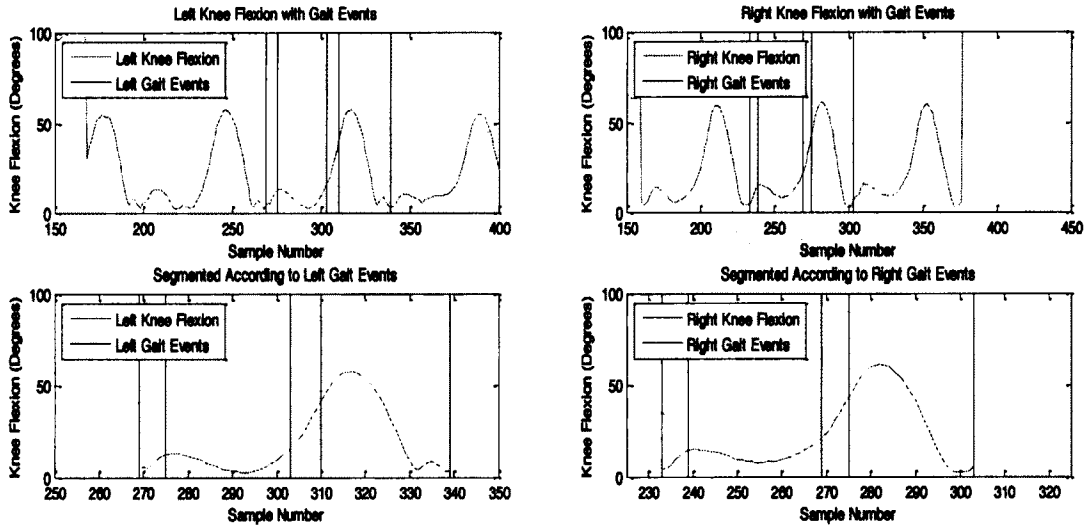


Figure 34: Gait cycle determination based on the five gait events; left leg on the left, right leg on the right.

Since there was only one cycle with definitive gait events, as determined by the force plates for each leg, this was the cycle that was analysed. Once each task cycle was determined, the muscle activity analysis was completed.

4.3.7 MATLAB Program Flow Chart

GASP PROGRAM

This program was developed by Dr. Vicky Chester in the Faculty of Kinesiology.

Steps of this program are as follows:

1. Subject folder must be created

2. Gait events are inputted

3. Subject Trial is entered

4. Subject Parameters are inputted (i.e. Knee & ankle width, height, weight etc.)

5. File conversion (from VICON to .c3d)

6. Marker re-location (move markers to center of joints)

7. Calculation of relative angles

8. Print out graph of gait

9. Additional EMG section added to calculate onset times

10. Save data

4.3.8 EMG PROGRAM

Steps of this program are as follows:

1. C3D Convert Program

   Converting raw data in preparation

   for analysis using Matlab

2. Filter

   Using Matlab FiltFilt function

   to filter Muscle activity

3. Cycle Determination

   Determine cycles for each of the

   three activities

4. EMG Baseline activity was determined

   Necessary for the calculation of

   muscle "on"/ "off" threshold value

5. EMG onset/offset times were calculated

   Descriptive values were

   extracted from onset information

6. Statistical Analysis

4.3.9 Muscle Onset Timing

Muscle activity was quantified using bilateral muscle symmetry, agonist/antagonist muscle pairs, muscle activation patterns, and percent muscle activation for each muscle. Muscle onset timing has been determined in many ways, including, visually, as a percent of the peak magnitude, and higher than a threshold value (Bullock & Saxton, 1993 &1994; Badke & Duncan, 1983; Carey & Allison, 1983; Chanaud & Macpherson, 1991; De Luca, 1993; Difabio,1987; Greenisen et al,1979; Happee, 1992; Karst & Willet, 1995; Leader, et al., 1998; Konrad, 2005; Nashner et al, 2978; Neafsey,1978; Steele,1994; Studenski et al.,1991; Thompson & McKinley, 1995; Woollacott et al, 1988). The magnitude in this study was not normalised, and so no calculation of muscle onset could be relative to this value. A study done by Hodges & Bui (1996) evaluated the accuracy of these various methods to determine muscle onset timing and additionally identified the most consistently accurate combination of these parameters for muscle detection. Using signals with both high and low noise levels, the best combination of parameters to detect accurate muscle onset is shown in Equation 3. Since this equation was used in this analysis, the most important calculation necessary for the determination of muscle activation patterns was the threshold level that determined whether the muscle is on or off, depending on whether the muscle activity was above or below the threshold value respectively.

Equation 3: Muscle "on/off" threshold calculation (Hodges & Bui, 1996)

$$\tau = 3 * \sigma x$$

σ: Standard Deviation (of the rectified baseline noise)

τ: Threshold value

x: is the baseline noise (of the rectified signal) for a segment of at least 50ms (DiFabio, 1987; Hodges and Bui, 1996) prior to muscle activation when the muscle is off.

Therefore, for a muscle to be considered "on", the muscle signal had to exceed the standard deviation of the baseline noise by a factor of 3 for at least 25 ms.

The baseline noise could have been determined in two different ways. The first was to measure the activity of each muscle during nominal contractions when it was not active. The problem with this method was that it was a recording that was separate from each trial, and therefore, new system noise could be introduced while performing the gross motor activities. This meant that the baseline noise chosen for this experiment was a segment of muscle activity during the task trial where the muscle was relatively inactive. The baseline noise had to be at least 50 ms long (Hodges and Bui, Bui, 1996; Di Fabio, 1987). Figure 35 below shows the baseline noise level (magnified in light grey) for the right trapezius muscle during swinging. This signal has been full wave rectified, but no filtering has been applied at this point.

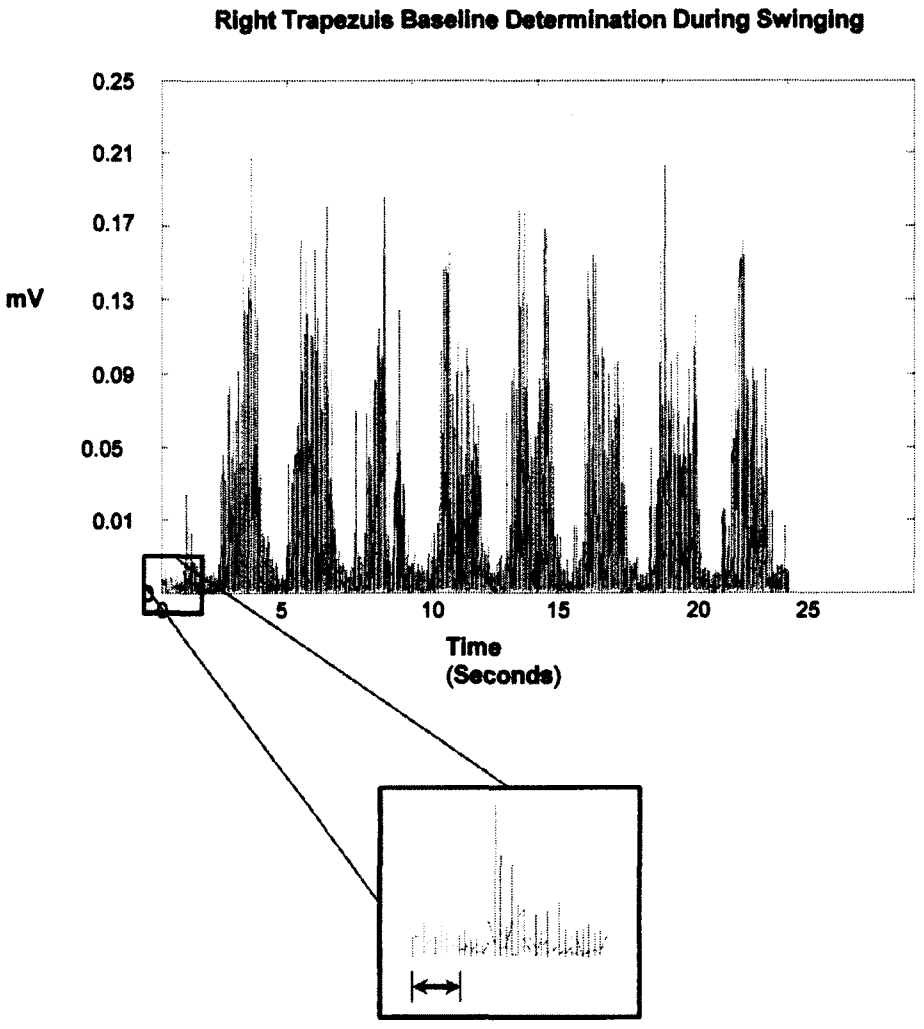


Figure 35: Definition of baseline noise for the right Trapezius muscle for a subject during a swing trial.

For consistency, all threshold calculations were determined using Equation 3. This was modelled on the results of the study done by Hodges and Bui (1996) and also to keep repeatability as high as possible.

4.4 Key Variables

For each trial, five values that described muscle activity were calculated and are described in Table 10. Each of these variables assists in describing muscle symmetry and activation timing.

Table 10: Explains each of the six values describing muscle activity, calculated for each subject

Variable Name	Definition	Matrix Size
EMG Onset	Stores the onset and offset information for all 10 muscles. This is a matrix with only two values, either 1.0 for muscle "on", or 0.0 for muscle "off".	[10xL] L=Signal Length)
EMG Basic	Stores the filtered EMG signal for all 10 muscles during the cycle that was chosen for analysis.	[10xL] L=Signal Length
EMG Percentage	Stores the percentage of the task cycle that each muscle is "on".	[1x10]
EMG Symmetry	Stores the percentage of time the same muscles on the left and right side are both "on", or both "off".	[1x5]
EMG Agonist	Stores the percentage of time the agonist and antagonist muscles are co-contracting.	[1x4]

The top two variables in Table 10 were used to describe muscle patterns in all three tasks. To describe muscle activity in patterns, similar principals to a confidence interval were used. For the control group, there were 16 control group subjects for walking and swinging, and 15 for biking. In order to describe muscle activation trends, a certain number of individuals out of the control group had to have their muscle activation "on" at

the same point in the task cycle to be counted as "on" for the control group. For example, for the swing task, 14 out of the 16 (88%) individuals in the control group had to have muscle activation "on" at exactly the same point in the swing cycle for it to be counted as "on" for the group. In order to decrease the limitations of only using the one value of 14/16, a range of plus or minus two subjects were added onto this number. This was done by creating a program in MATLAB that counted each date point on the x-axis (percentage of swing cycle) that was recorded as "on" for all 16 subjects. If 14 out of the 16 subjects had the same data point "on", then the muscle was considered "on" for the group. This was done for 12/16 and 16/16, to add the upper and lower limits of the muscle activity. Figure 43 illustrates the muscle activation patterns of the control group during swinging. For example, Figure 43 shows that the dominant deltoid was "on" for at least 14 out of the 16 control subjects between 39 and 41%, and again between 81 and 86 % of the swing cycle.

Total muscle activity was described using the result from EMG percentage. This value was calculated by measuring the total time a muscle was active and dividing it by the length of the task cycle. Therefore, EMG percentage represents the percentage of a task cycle that the muscle was "on".

Muscle symmetry, between dominant and non-dominant sides was described using the variable EMG symmetry. This value was calculated by comparing the dominant and non-dominant sides of an individual during each task cycle. For example, to calculate the biceps symmetry during the swing cycle, each data point (at the same percent of the

swing cycle) for the dominant side was simultaneously compared to the corresponding data point for the non-dominant side. If the two sides showed the same activity (ie. both were either "on", or both were "off"), then this data point was counted as symmetrical. EMG symmetry was then calculated as a scalar, recording the percentage of the cycle for which the muscles were symmetrical.

The last variable calculated in Table 10 was used to describe muscle co-contraction, and was stored in the variable EMG agonist. EMG agonist was calculated similarly to EMG symmetry. However, muscle co-contraction was recorded when the agonist and antagonist muscles (from the same agonist/antagonist muscle pair) were simultaneously "on". For example, if the dominant biceps was "on" and the dominant triceps was "on" during the same percent of the swing cycle, it was counted as co-contraction. EMG agonist was recorded as a scalar; the time spent in co-contraction as a percentage of time either the agonist or antagonist muscle was on. This is different than the EMG symmetry calculation because the co-contraction is not simply divided by the length of the entire cycle to get a percentage of the task in co-contraction. Instead, co-contraction was divided by the length of cycle minus the times when both agonist/antagonist muscles were "off". This ensured that co-contraction was not over/under represented.

There were 4 agonist/antagonist muscle pairs, because the erector spinae muscle is the antagonist muscle for the rectus abdominus muscle, and abdominal muscles were not captured in this study due to difficulties locating this muscle in children, and the fact that

it would be much more personally invasive to capture the abdominal muscles (the children would have to wear less clothing).

4.5 Z-Scores

Each prosthesis user was treated as a case study and compared against the control group because there were four individuals in the test group, which was not enough to yield strong statistical group results. The control group consisted of 16 normally limbed children between the ages of 4 and 14. Data captured during the bike task did not record for 1 control subject. The differences between the control sample and individuals in the test group under both conditions (wearing a prosthesis versus not wearing), were described using Z-scores. The Z-scores indicate the number of standard deviations the test subject is from the mean of the control group.

For example, for subject ICPML14p, muscle symmetry of the biceps compared to the control group during biking was described as:

(ICPML14p biceps muscle symmetry - mean control biceps muscle symmetry)/(control biceps symmetry standard deviation)

⟶ (-86.8-81.1)/27.2 =+0.2

ICPML14 displayed 0.2 standard deviations more biceps symmetry than the control group while wearing a prosthesis during biking. This means that there was normal biceps

activity during biking for subject ICPML14 while wearing a prosthesis during biking. Positive values indicate a greater value compared with the control group, and negative values indicate a lesser value.

Only the Z-score equal or greater than 2.4 was recognized as being different than the control group, because for any normal sample, 98-99% of observations should lie within 2.4 standard deviations of the mean. For example, for each prosthesis user, there are 10 Z-scores calculated for each variable (symmetry, co-contraction, percent activation), and each task (biking, walking, swinging). The probability that all Z-scores for each variable lie within +/- 2.4 standard deviations of the control is 0.985^10=0.86. This means there is an 86% chance that all 10 Z-scores lay within the "normal" limits of +/- 2.4 standard deviations. If one Z-score falls outside normal limits for each variable during a specific task, this is interesting, and significant. All individual entries for the control and test groups can be found in Appendix D. Each individual in the test group is described under two differing conditions. One condition being when the individual was wearing their prosthesis; which was designated by the word "prosthesis" underneath the task title, and the second condition was when the individual was not wearing their prosthesis; which was designated by the word "no prosthesis" underneath the task title.

5.0 Results and Discussion

5.1 Control Group

5.1.1 Muscle Activation Patterns

The muscle activation for the control group during biking is described in Figure 38, and it is obvious that muscle activation among the control group was not consistent (this is also demonstrated by the relatively high standard deviations for the biceps, triceps, trapezius and deltoids found in Table 13). The only muscles that demonstrated consistent activity during the bike cycle were the dominant and non-dominant trapezius muscles. In order to describe muscle activation trends, a certain number of individuals out of the control group had to have their muscle activation "on" at the same point in the task cycle to be counted as "on" for the control group. For example, for the swing task, 14 out of the 16 (88%) individuals in the control group had to have muscle activation "on" at exactly the same point in the swing cycle for it to be counted as "on" for the group. This was done by creating a program in MATLAB that counted each data point on the x-axis (percentage of swing cycle) that was recorded as "on" for all 16 subjects. If 14 out of the 16 subjects had the same data point "on", then the muscle was considered "on" for the group. Even so, the selection criteria for the bike muscle activation was only 11 out of the 15 control subjects (73%), as opposed to the 88% (14/16) used in the swing task. Therefore, these trends were not as strong as ones described during swinging and walking. However, it appears that there is consistent dominant trapezius activation between 15% and 21% of the bike cycle, which is when the dominant heel is lifting back and up towards the pedal being vertical (25% of the bike cycle); this is described in

Figure 36. The non-dominant trapezius was activated shortly between 34 and 35% and again between 42 and 52% of the bike cycle. Figure 37 shows, that this is when the non-dominant heel is heading down and back towards until the pedal reaches the lowest point in the cycle, and continues to be activated just after that point on the way back and up.

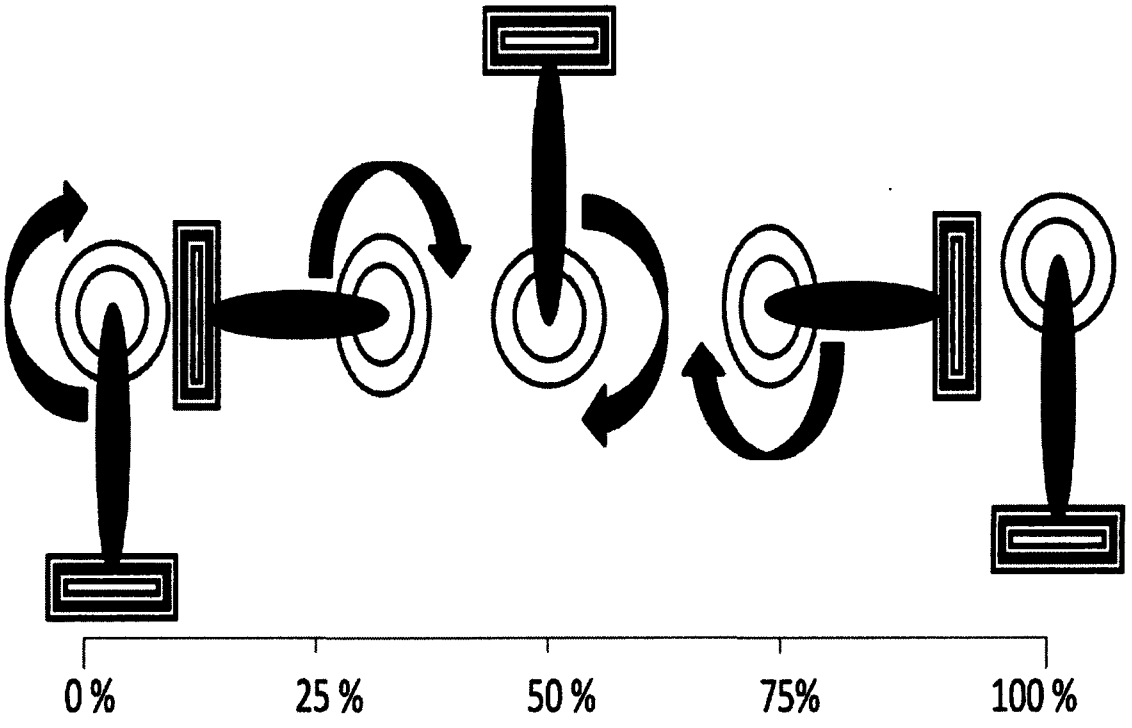


Figure 36: Illustration of the bike cycle; the dominant side pedal progression throughout the cycle (starting and ending with the pedal in the lowest position).

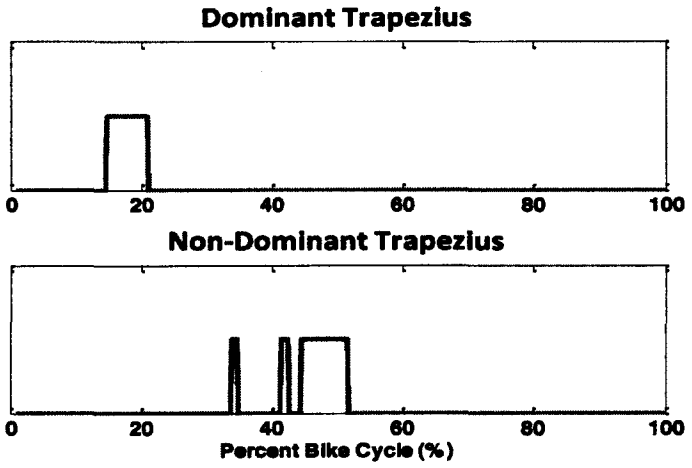


Figure 38: Consistent muscle activation patterns for the control group during biking ((13 +/-2)/15)).

To date, there have been no reported upper body muscle activation patterns during cycling. This could be due to the fact that upper body muscle activations may not exhibit any patterns relative to a lower body driven activity like cycling. This seems to be the case for the control group, as there was no consistent muscle activation found within the middle and upper limits of the selection criteria (when 13 and 15 out of 15 subjects, respectively, had to have muscles "on" for the patterns to be counted as "on"). There were dominant and non-dominant trapezius muscle activation when the lower selection criterion was used (only 11/15 subjects). However, the dominant trapezius muscles were activated when the dominant pedalling foot was lifting backwards and up towards the pedal's highest point in the cycle. In contrast, the non-dominant trapezius activation occurred when the non-dominant heel was heading down and backwards towards the pedal's lowest point in the cycle.

The middle of the selection criteria for the walking task was 14 out of 16 subjects (88%), the upper and lower limits were 16/16 and 12/16 respectively. The consistent muscle

patterns for the dominant gait cycle are shown in Figure 40 which demonstrates that all muscles, with the exception of the non-dominant and dominant erector spinae muscles, were not consistently active during the gait cycle. There was slightly more activity shown for the dominant erector spinae when the lower selection criterion was used, however this muscle timing was the same as was when 14/16 subjects were counted (middle of the selection criterion range). The non-dominant erector spinae also had more consistent muscle activation when the lower limits were used, and even had an additional burst of muscle activity at the end of the gait cycle that was not present when 14/16 subjects were counted. The middle of the selection criterion showed the non-dominant erector spinae was consistently active between 2% and 9% of the dominant gait cycle. This corresponds to the point just after dominant foot-strike event up until just prior to non-dominant toe-off, and additionally for a brief time between 49% and 52% which is illustrated in Figure 39. The dominant erector spinae was consistently active between 46% and 55% which corresponds to the period just before and after non-dominant foot-strike.

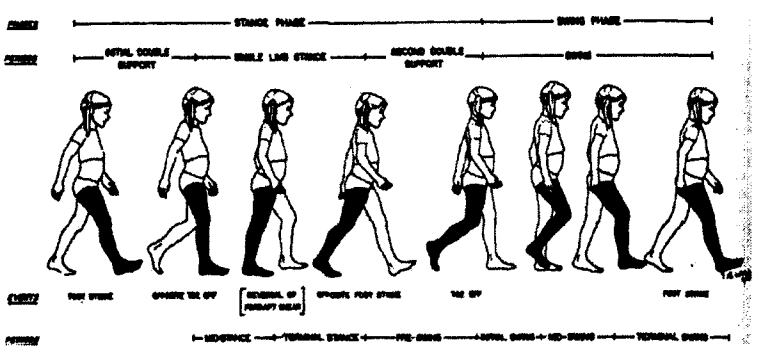


Figure 39: Illustration of the major gait events during the gait cycle (Sutherland, 1981).

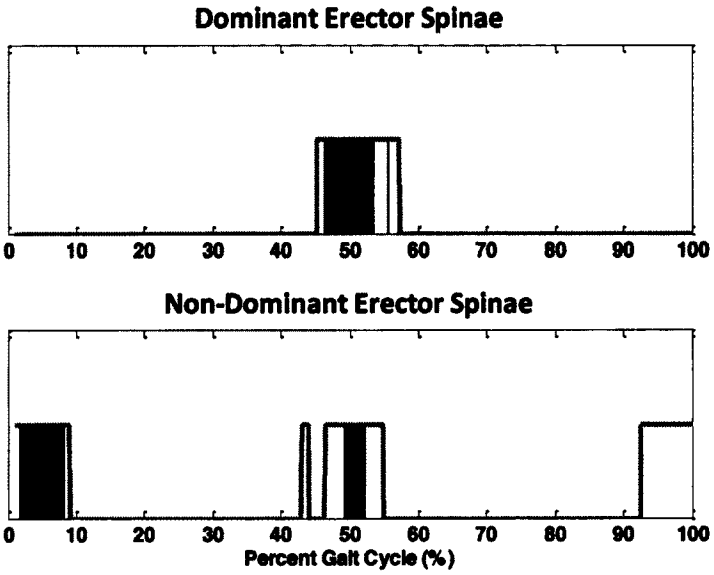


Figure 40: Consistent muscle activation patterns for the control group during the walking task for the dominant gait cycle ((14 +/-2)/16) subjects).

This is consistent with Steven et. al. (2002), which provided Figure 41 below, and shows erector spinae activity between approximately 48 % and 62 %, which starts just before opposite foot strike and ended just after toe off.

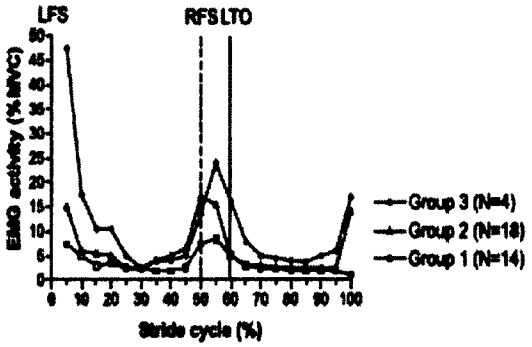


Figure 41: Figure from (Steven et. al., 2002) describing erector spinae muscle activity during gait

The presence of a peak in muscle activity at foot-strike was consistent with other research as well (Callaghan, et al., 1999; Murray, et. al., 1984; Murray, et. al., 1985; Waters, et.

al., 1972; Winter & Yack, 1987). There were some varying opinions on why this was occurring; Winter & Yack (1987) believed the erector spinae was activated at this point to control forward rotation of the trunk while it decelerated to during weight acceptance at footstrike. However, Shiavi (1990) suggested that the activity was related to weight transference between limbs and to the reversal in direction of pelvic and thoracic rotation. During swinging, the same selection criteria was used as the walking task, where 14 out of 16 subjects (88%) had to have the muscle "on" for it to be counted "on" for the group. Additionally, the range of plus or minus 2 individuals (out of 16) were added in order to describe what changes in this criterion occur within the group muscle patterns of the control group. Overall, there was more consistent muscle activity for the swing task compared with the other two tasks, which is probably due to the cyclic nature of the task and the fact that it requires more upper body muscle activation.

The following section will report on the muscle patterns of the control group in the middle of the range given by the number of individuals included for muscle to be considered "on" (14/16). The dominant and non-dominant biceps were consistently activated between approximately 15% and 55% of the swing cycle. Based on Figure 42, this means that the biceps were activated midway through the forward swing until just after the beginning of the downswing (refer to Figure 42 for swing segment terminology). There was slightly more activity shown for the non-dominant biceps. The dominant and non-dominant triceps were similar except for a portion between approximately 28% and 36% of the cycle, where only the non-dominant side was activated. The non-dominant triceps were consistently activated during the upswing, and they were both consistently

active between approximately 73% and 88% of the swing cycle. This corresponded to activity during the downswing of the cycle (when their arms are pulling backwards on the swing chain). The dominant and non-dominant trapezius muscles were also similarly activated, however the non-dominant trapezius started and ended approximately 4% before the dominant one. They were consistently activated between approximately 26% and 57%, which corresponded to being activated at the beginning of the upswing, until approximately a third of the way through the downswing. Similarly to the trapezius muscles, the non-dominant deltoid seemed to have started and ended prior to the dominant deltoid by approximately 6%. Figure 43 shows that the non-dominant deltoid muscle was consistently active for a larger percent of the swing cycle than the dominant deltoid. The deltoid muscles were active between approximately 35% and 42% and again between 75% and 85%. This corresponded to activation that started in the middle of the upswing until ¾ of the way through the upswing, as well as activation that started at the beginning of the backswing and ended halfway through the backswing.

The lower limits (12/16), which is the open bar extruding out from the solid bar on figure 44, of the muscle activation patterns within the control group show there are much more consistent muscle activity for the biceps, triceps, deltoids, and trapezius muscles. Similarly for the upper limits (16/16), which is the light coloured open bar inside of the solid bar on figure 44, of the muscle activation patterns within the control group show there are much less consistent muscle activity when the selection criterion is increased to encompass all 16 out of 16 individuals.

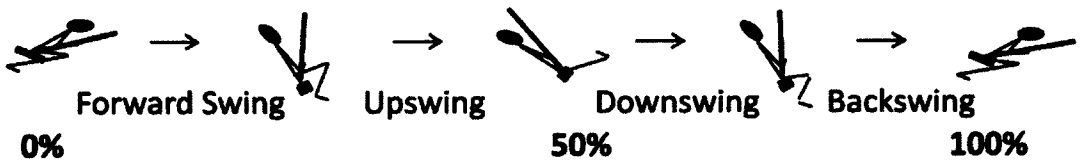


Figure 42: Illustration of swing cycle.

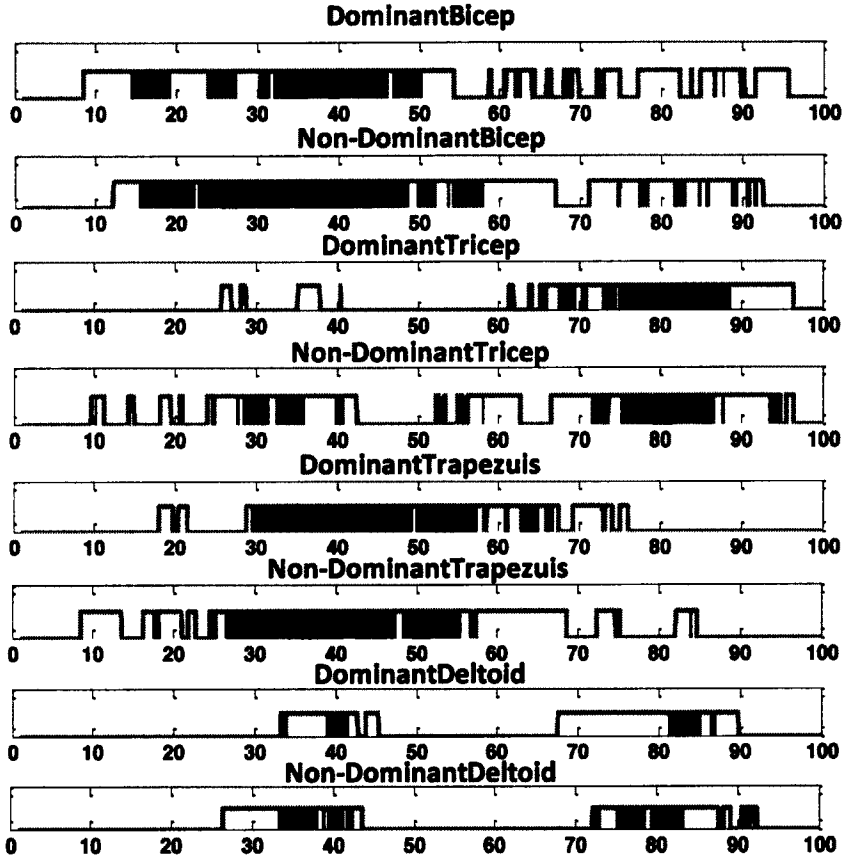


Figure 43: Muscle activation pattern for the control group during swinging.

To date, there has been no reported upper body muscle activation patterns during swinging, so no comparisons can be made. However, this study shows that there appears to be consistent, symmetrical muscle activation (for the biceps, triceps, trapezius, and deltoids) during swinging in normally limbed children.

5.1.2 Percent Symmetry

Percent symmetry was calculated by finding the time at which both of the same muscles (ie. dominant biceps and non-dominant biceps) were simultaneously active or not active, and dividing this number by the total length of the task cycle. This resulted in the percentage of time the muscles were acting symmetrically. Values for percent symmetry throughout all three activities can be found in Table 11.

For the bike task, the muscle symmetry had to be calculated slightly different than the swing task because of the asymmetrical nature of the bike cycle definition. That being that throughout the bike task, the definition of the bike cycle was only related to the dominant leg. Therefore, the non-dominant bike cycle was found and the non dominant muscles were related to this. Muscle symmetry was then calculated by comparing the non-dominant cycle with the dominant one. Thus, all non-dominant muscles were calculated (for muscle symmetry) relative to Figure 37, and all dominant muscles were calculated relative to Figure 36.

During biking, the control group exhibited muscle symmetry for more than 50% of the cycle for all muscles. The lowest average muscle symmetry was by the trapezius (53.7%), and the highest average muscle symmetry occurred for the biceps (muscle active or not active together 81.1% during the bike cycle). This may have been due to the fact that the trapezius muscles were the most active during biking (Table 13), and less active muscles appear to be more symmetrical because they are both "off" together for longer periods of time.

Muscle symmetry for walking was calculated the same as it was for the bike task muscle symmetry. During walking, the control group exhibited the highest average muscle symmetry for the biceps (91.7%), as seen in Table 11; the triceps, trapezius and deltoids all had symmetrical activity for at least 70% of the gait cycle, whereas the erector spinae only showed muscle symmetry throughout 59% of the cycle. The high symmetry during walking is probably attributed to the fact that there was very little overall muscle activity (see Figure 43 & Table 13), and therefore, if both muscles were "off" at the same time, they were acting symmetrically. It should be noted that only the dominant gait cycle was chosen for analysis (on all other key variables) for each person, as this was consistently the best side in terms of force plate contact for gait events.

The control group had consistent muscle symmetry for all 5 muscles during swinging; ranging from 72 and 78% of the swing cycle. It is evident that the swing task was highly symmetrical based on Figure 43, where the muscle pairs that exhibited consistent activation patterns (biceps, triceps, trapezius, and deltoids) were "on" and "off" simultaneously for large portions of the swing cycle.

Table 11: Control group percent muscle symmetry for three tasks: mean (standard deviation)

Percent Symmetry	Bicep	Tricep	Trapezius	Deltoids	Eresp
Bike	81.1 (27.2)	63.2 (25.5)	53.7 (20.3)	60.4 (21.9)	80.3 (13.4)
Walk	91.7 (13.0)	80.6 (23.1)	72.2 (20.5)	80.7 (20.7)	59.9 (19.8)
Swing	77.3 (7.8)	77.5 (15.3)	75.4 (10.7)	73.7 (11.6)	72.2 (21.9)

5.1.3 Percent Co-Contraction

The average percent co-contraction was calculated for the control group and these values are found in Table 12. For all three activities, the dominant and non-dominant agonist/antagonist pair of the trapezius and deltoid had more co-contraction than the biceps and triceps agonist/antagonist pair for the control group. The most co-contraction occurred during the swing activity, and the least co-contraction occurred during walking.

The control group had an average co-contraction under 35% during biking. The lowest average co-contraction was seen by the non-dominant trapezius/deltoid pair at 8%, and the highest average co-contraction by the dominant trapezius/deltoid pair at 33%.

During walking, there was an average of approximately 0% co-contraction for the dominant and non-dominant biceps/triceps pairs, but the dominant and non-dominant trapezius/deltoid pairs had a higher average percent co-contraction with 20% and 14% respectively.

Swinging had the highest average percent co-contraction out of all three tasks, probably due to the increased muscle activity for all muscles (except the erector spinae muscles) during this activity; see Table 13. These average percent co-contractions ranged between 35% (non-dominant bicep/tricep) to 44% (dominant trapezius/deltoid) agonist/antagonist muscle pairs.

It is important to note that Table 11 has high standard deviations relative to the mean muscle activity for all tasks. Therefore, when comparing the prosthesis user case studies to the control group, no significant deviations could be reported on, as these data show that during all tasks, muscle co-contraction is sporadic for the control's.

Table 12: Control group percent co-contraction for three tasks: mean (standard deviation)

Percent Co-Contraction	Dom Bicep-Tricep	Non-Dom Bicep-Tricep	Dom Trap-Delt	Non-Dom Trap-Delt
Bike	11.1 (16.0)	7.7 (16.7)	33.0 (29.4)	22.4 (25.6)
Walk	0.1 (0.3)	0.0 (0.0)	20.0 (20.9)	14.1 (20.9)
Swing	38.5 (17.5)	35.3 (18.6)	43.8 (18.3)	43.3 (16.3)

5.1.4 Percent Activation

The average percent activation for the control group was calculated by averaging each subject in the control group's percent activation for each muscle. This was done by measuring the amount of time a particular muscle was "on" and dividing this number by the length of the task cycle. The average percent activation of all 10 muscles for individuals in the control group are displayed in Table 13.

The highest average muscle activity during biking for the control group was for the non-dominant trapezius, which was typically active for 51% of the bike cycle (Table 13). The lowest average percent activity for the control group during biking was the non-dominant biceps at 17%.

The non-dominant trapezius muscle also had the highest average percent activity during walking at 50% of the gait cycle. All average percent activity during walking, was lower than the other two tasks except for the dominant and non-dominant erector spinae, which were both more active during walking compared with biking and swinging.

The swing task had the highest average muscle activity for all muscles except the dominant and non-dominant erector spinae muscles compared with the bike and walking tasks. The most active muscle on average during swinging was the non-dominant biceps (77%), and the muscle with the lowest average activity was the dominant erector spinae (12%).

It is important to note that Table 13 has high standard deviations relative to the mean muscle activity for the bike and walking tasks (except for the erector spinae activity during walking). Therefore, when comparing the prosthesis user case studies to the control group, no significant deviations could be reported on (in these instances), as these data show that during biking and walking, muscle activity is sporadic for the control's.

Table 13: Control group percent activation for three tasks: mean (standard deviation)

Percent Activation	Dom Bicep	Non Dom Bicep	Dom Tricep	Non Dom Tricep	Dom Trap	Non Dom Trap	Dom Delt	Non Dom Delt	Dom Eresp	Non Dom Eresp
Bike	20.8 (25.6)	16.8 (28.8)	38.0 (28.3)	37.8 (30.1)	39.8 (26.9)	50.8 (31.9)	45.7 (34.5)	27.7 (26.3)	22.7 (21.4)	23.0 (22.4)
Walk	2.9 (6.7)	2.2 (5.2)	15.6 (23.2)	5.0 (7.4)	33.6 (31.1)	49.9 (28.4)	19.9 (20.4)	16.8 (24.6)	34.7 (18.6)	42.4 (28.7)
Swing	73.7 (18.1)	77.3 (17.0)	63.5 (27.3)	70.8 (21.4)	63.0 (13.1)	68.8 (16.0)	58.8 (27.9)	58.0 (23.7)	12.2 (14.1)	28.6 (28.7)

5.2 Prosthesis Users

5.2.1 Subject ICPML14

This test subject was 14 years old, and male, with left hand dominance and limb loss on the right side (transradial).

5.2.1.1 Muscle Activation Patterns

Amongst the control group during biking (Figure 38), the only consistent muscle activation was dominant trapezius activation between 15% and 21%, and non-dominant trapezius activation between 43% and 52%. Muscle activation for all 10 muscles are described in Figure 44 for subject ICPML14 while wearing a prosthesis, and Figure 45 while not wearing a prosthesis.

This subject showed different muscle activity compared to the control group. Since there was only consistent activation in the control group for the trapezius muscles, any absence of trapezius activation at the same point in the bike cycle found in subject ICPML14 would be considered outside the normal levels of activity. The non-dominant biceps was activated between 78% and 91%, the dominant triceps was "on" almost entirely between 19% and 100%, the dominant erector spinae was "on" between 2% and 10%, again between 31% and 40%, and lastly between 72% and 100%, and the non-dominant erector spinae was "on" from 1% to 55%, again from 62% to 70% and lastly from 82% to 100%.

Without a prosthesis during biking, muscle activation patterns were not similar to the control group for the trapezius muscles. Also, compared to this subject wearing a

prosthesis, each muscle pair seemed more symmetrical. The biceps, triceps and erector spinae muscles were more active on the non-dominant (prosthesis) side, compared with the dominant side when wearing a prosthesis. This also appeared to be true when not wearing a prosthesis, although there seemed to be much more similar activation patterns between non-dominant and dominant sides compared with the subject wearing their prosthesis.

This can be explained by the manner in which the subject typically bikes. Subject ICPML14 stated that he preferred to bike one-handed, instead of wearing a prosthesis during biking. Substantial forward trunk lean was required to perform the bike task with the prosthesis on, and this could have caused the increase in muscle activity.

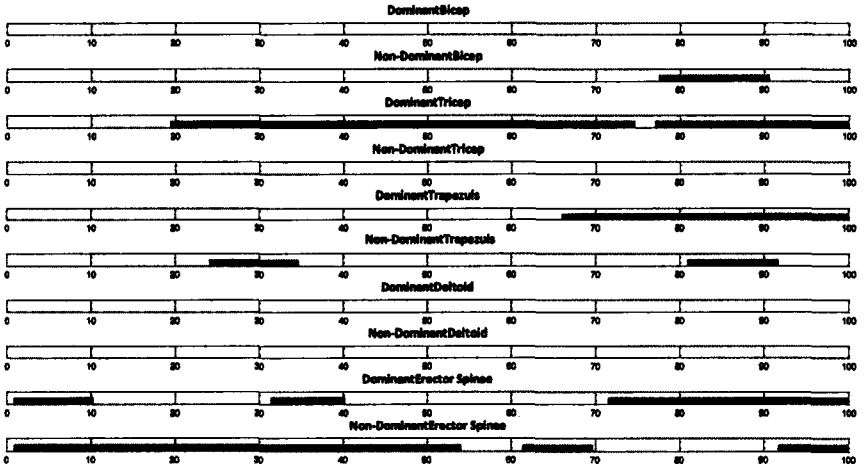


Figure 44: Muscle activation for subject ICPML14 while wearing a prosthesis during the bike task.

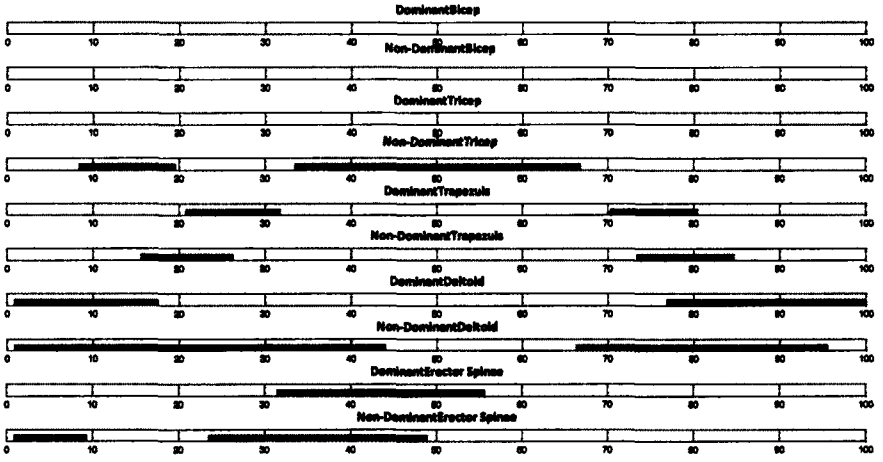


Figure 45: Muscle activation for subject ICPML14 while not wearing a prosthesis during the bike task.

During the walking task, there were not many differences between subject ICPML14 and activation patterns measured for the control group. This was illustrated when comparing the control group (Figure 40) with the test condition wearing a prosthesis (Figure 46), and the test condition not wearing a prosthesis (Figure 47). The dominant and non-dominant erector spinae muscles had activation at the same points of the control group during the gait cycle. Both conditions (while wearing a prosthesis and not), appear to have much higher activity compared with the controls, however, there is no statistical evidence that supports this from Table 16 (no significant Z-scores).

The two prosthesis conditions were similar to one another in terms of which muscles were active and when, however, while wearing a prosthesis subject ICPML14 had slightly more activation compared to when he was not wearing one during walking.

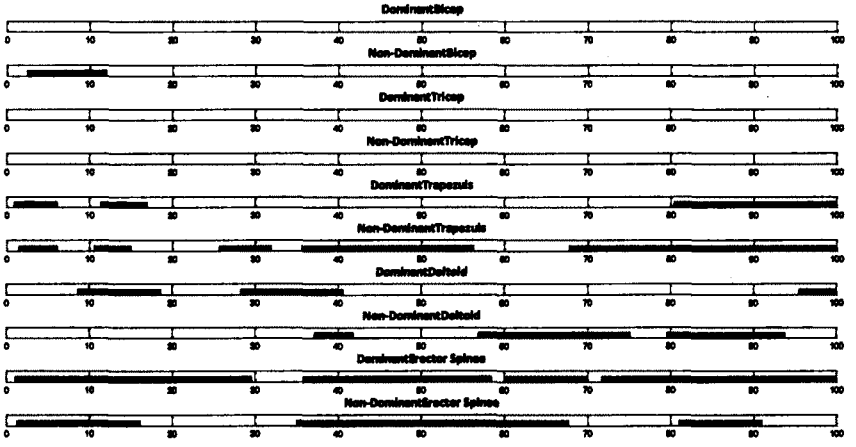


Figure 46: Muscle activation for subject ICPML14 while wearing a prosthesis during walking.

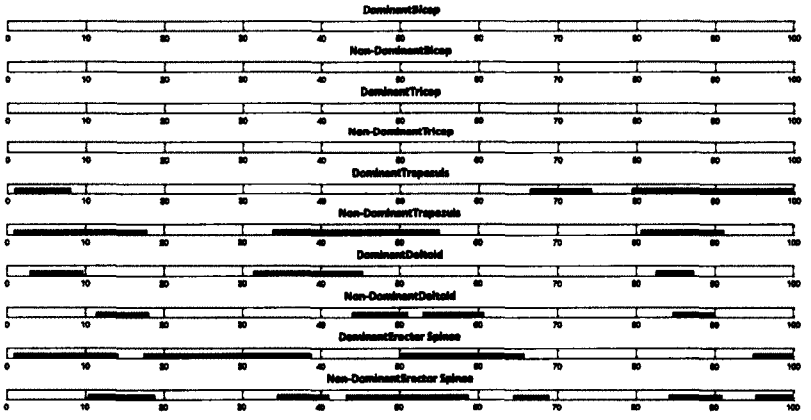


Figure 47: Muscle activation for subject ICPML14 while not wearing a prosthesis during walking.

During swinging, there was little difference between subject ICPML14 wearing a prosthesis (Figure 48) versus not wearing one (Figure 49), except that the non-dominant and dominant trapezius muscles were noticeably more active when wearing a prosthesis compared to not wearing it. In addition, the non-dominant and dominant deltoid muscles seemed to be more active when not wearing a prosthesis compared with wearing one.

During both scenarios (wearing a prosthesis versus not wearing one), since there was so much muscle activation, it is difficult to say if these patterns were not similar to the control group (Figure 43).

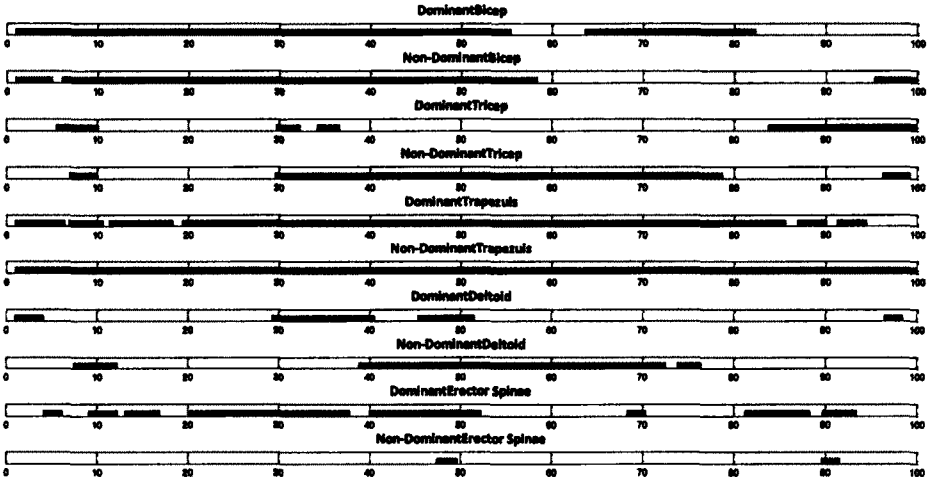


Figure 48: Muscle activation patterns for subject ICPML14 while wearing a prosthesis during the swing cycle.

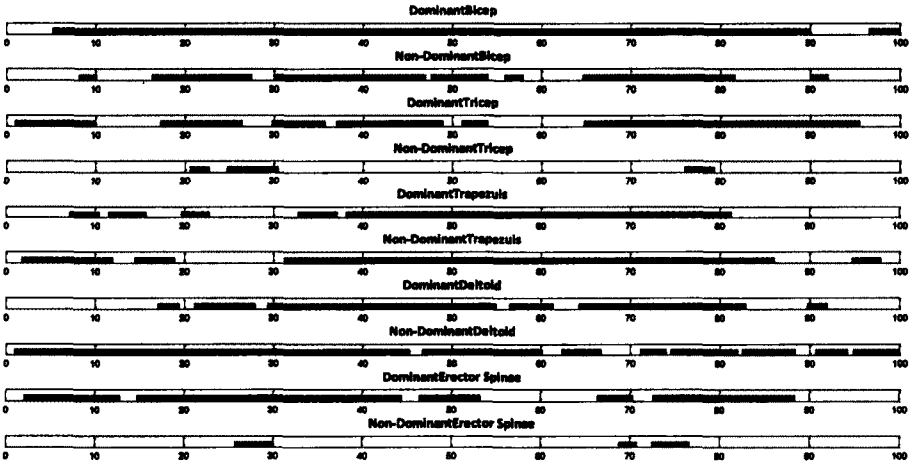


Figure 49: Muscle activation patterns for subject ICPML14 while not wearing a prosthesis during the swing cycle.

5.2.1.2 Percent Symmetry

Refer to Table 14 for all muscle symmetry information on subject ICPML14.

Even with the high variability amongst the control group for the triceps muscle (standard deviation of 25.5), there was still a Z-score of -2.4 when comparing the triceps of the test condition wearing their prosthesis with the control group. Therefore, this subject's triceps muscles were activated symmetrically for a smaller percentage of time than did children in the control group during biking.

The test group under both conditions (not wearing a prosthesis and wearing a prosthesis) had the highest muscle symmetry from the biceps and triceps during walking. Although, this may only be the case because both muscles were both "off" at the same time for a large part of the gait cycle (this is evident by the low levels of percent activation for the biceps and triceps during walking in Table 16). There were no large differences observed in muscle symmetry between the control group and this subject while wearing a prosthesis and not wearing one during walking.

During the swing task, the condition while wearing the prosthesis had lower triceps symmetry than the control group (Z= -2.4). The strongest trend towards asymmetry was found to be the triceps while swinging without a prosthesis (Z = -2.8).

Table 14: Subject ICPML14 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Symmetry (Z-Score)

ICPML14 Percent Symmetry	Bicep	Tricep	Trapezius	Deltoids	Eresp
Bike					
Prosthesis	86.8 (0.2)	2.6 **(-2.4)**	56.6 (0.1)	100.0 (1.8)	60.0 (-1.5)
no prosthesis	100.0 (0.7)	3.5 (-2.3)	100.0 (2.3)	100.0 (1.8)	100.0 (1.5)
Walking					
Prosthesis	87.2 (-0.4)	95.4 (0.6)	65.3 (-0.3)	66.4 (-0.7)	64.1 (0.2)
no prosthesis	100.0 (0.6)	100.0 (0.8)	61.5 (-0.5)	62.4 (-0.9)	56.1 (-0.2)
Swing					
Prosthesis	72.4 (-0.6)	40.4 **(-2.4)**	89.8 (1.4)	51.7 (-1.9)	51.5 (-0.9)
no prosthesis	65.0 (-1.6)	34.1 **(-2.8)**	73.9 (-0.1)	55.5 (-1.6)	41.9 (-1.4)

5.2.1.3 Percent Co-Contraction

Refer to Table 15 for all muscle co-contraction information on subject ICPML14.

As early stated, any large Z-scores in this section cannot be credited based on the very high standard deviations seen by the control group (Table 12) in comparison to the mean

values. Therefore, even though there were 2.4 standard deviations more co-contraction within the agonist/antagonist muscle pair of the dominant biceps and triceps while wearing a prosthesis during swinging, this should be discounted.

Overall, ICPML14 showed normal levels of co-contraction over all activities, both with and without the prosthesis. The bike task had the lowest percent muscle co-contraction for all agonist/antagonist muscle pairs, and the swing task had the most muscle co-contraction for all agonist/antagonist muscle pairs.

During swinging, while wearing a prosthesis subject ICPML14 had 2.4 standard deviations more percent muscle co-contraction for the dominant biceps/triceps pair when compared to the control group of the same agonist/antagonist pair. Otherwise, all other agonist/antagonist muscle co-contractions were at similar levels displayed by the control group during all tasks.

Table 15: Subject ICPML14 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Co-Contraction (Z-Score)

IcPML14 Percent Muscle Co-Contraction	Dominant Bicep-Tricep	Non-Dominant Bicep-Tricep	Dominant Trapezius-Deltoid	Non-DominantTrapezius-Deltoid
Bike				
prosthesis	0.0 (-0.7)	0.0 (-0.5)	6.4 (-0.9)	30.7 (0.3)
no prosthesis	0.0 (-0.7)	0.0 (-0.5)	0.0 (-1.1)	0.0 (-0.9)
Walking				
prosthesis	0.0 (-0.3)	0.0 (-0.3)	38.1 (0.9)	27.3 (0.6)
no prosthesis	0.0 (-0.3)	0.0 (0.2)	31.2 (0.5)	22.6 (0.4)
Swing				
prosthesis	80.3 **(2.4)**	47.2 (0.6)	71.0 (1.5)	58.4 (0.9)
no prosthesis	38.0 (-0.0)	52.1 (0.9)	29.2 (-0.8)	32.7 (-0.7)

5.2.1.4 Percent Activation

Refer to Table 16 for all muscle activity information on subject ICPML14.

During swinging, the dominant erector spinae for both test conditions were more active than the control group (Z=2.9 and Z=3.9 respectively). This means the subject was activating his erector spinae muscle on the non-affected side during swinging under both test conditions. This may be due to the twisting of the torso under both conditions while swinging. Figure 50 shows ICPML14 from a bird's eye view (X-Y Plane), and shoulders are significantly twisting under both conditions, but very much so while wearing a prosthesis. This figure is a snapshot of ICPML14 during the backswing, while pulling back on the swing chain. Just as ICPML14 gets to the furthest point in the backswing, this twisting would have to be stopped (by the non-dominant erector spinae muscles).

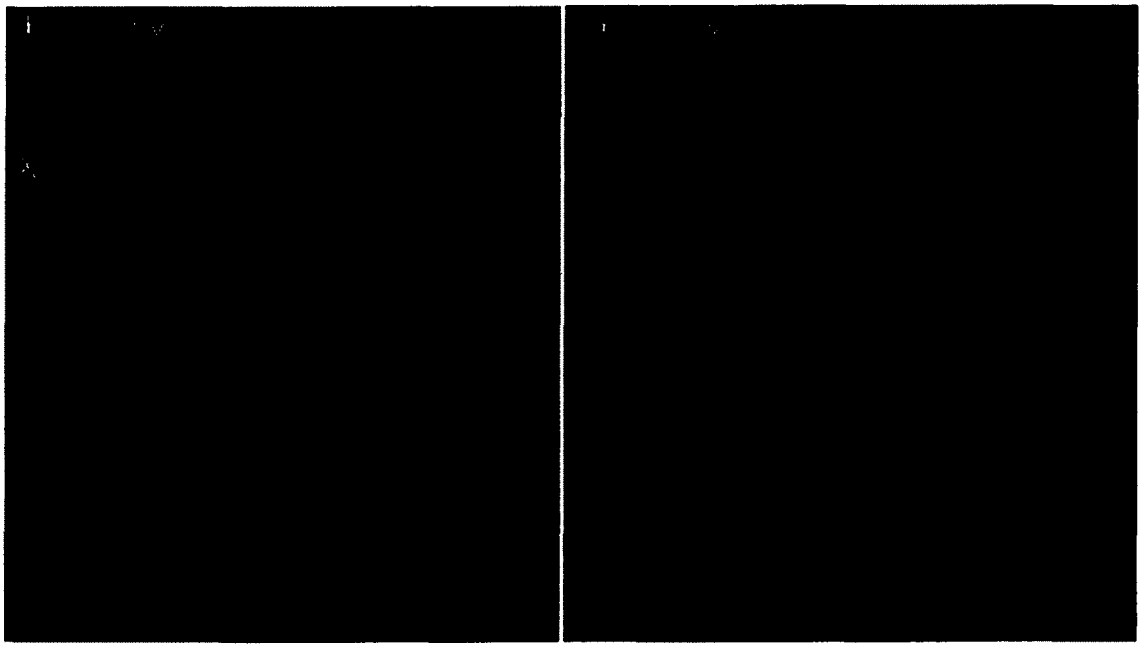


Figure 50: Snapshot of ICPML14 swinging without a prosthesis (left), and with a prosthesis (right).

The non-dominant triceps muscle was less active during the swing cycle under the test condition of wearing their prosthesis compared with the control group (Z=-2.8); so in contrast with the dominant erector spinae, the non-dominant triceps (triceps on the side

with limb loss) had a much lower percentage of activation, while wearing a prosthesis during swinging than was illustrated by the control group.

Table 16: Subject ICPML14 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Activation (Z-Score)

IcPML14 Percent Activation	Dom Bicep	Non-Dom Bicep	Dom Tricep	Non-Dom Tricep	Dom Trap	Non-Dom Trap	Dom Delt	Non-Dom Delt	Dom Eresp	Non-Dom Eresp
Bike										
prosthesis	0.0 (-0.8)	13.2 (-0.1)	78.9 (1.4)	0.0 (-1.3)	34.3 (-0.2)	21.8 (-0.9)	0.0 (-1.3)	0.0 (-1.1)	47.2 (1.1)	70.6 (2.1)
no prosthesis	0.0 (-0.8)	0.0 (-0.6)	96.5 (2.1)	0.0 (-1.3)	0.0 (-1.5)	0.0 (-1.6)	0.0 (1.3)	0.0 (-1.1)	0.0 (-1.1)	0.0 (-1.0)
Walking										
prosthesis	0.0 (-0.4)	12.8 (2.0)	0.0 (-0.7)	4.6 (-0.1)	27.7 (-0.2)	58.6 (0.3)	32.7 (0.6)	33.1 (0.7)	74.2 (2.1)	64.5 (0.8)
no prosthesis	0.0 (-0.4)	0.0 (-0.4)	0.0 (-0.7)	0.0 (-0.7)	27.8 (-0.2)	51.0 (0.0)	28.8 (0.4)	12.0 (-0.2)	65.9 (1.7)	49.0 (0.2)
Swing										
prosthesis	73.9 (0.0)	61.7 (-0.9)	26.4 (-1.4)	55.7 (-0.7)	89.8 (2.1)	100.0 (1.9)	23.0 (-1.3)	41.6 (-0.7)	52.7 **(2.9)**	4.5 (-0.8)
no *prosthesis*	89.3 (0.9)	58.1 (-1.1)	70.8 (0.3)	11.4 **(-2.8)**	58.8 (-0.3)	73.6 (0.3)	61.2 (0.1)	88.2 (1.3)	67.7 **(3.9)**	10.7 (0.6)

In general for subject ICPML14, the triceps muscle seemed to deviate from normal muscle patterns with regards to muscle symmetry for two different tasks (biking and swinging). In addition, the swing task had the most abnormal muscle symmetry, co-contraction, and total muscle activation out of the three tasks. Although there were large differences between the two conditions of the test group compared to the control group, ICPML14 tended to have more differences from normal while wearing his prosthesis versus not wearing it.

5.2.2 Subject PAPFL10

This test subject was 10 years old, female, with left hand dominance and limb loss on the right side (carpal partial).

5.2.2.1 Muscle Activation Patterns

Amongst the control group during biking (Figure 38), the only consistent muscle activation was dominant trapezius activation between 15% and 21%, and non-dominant trapezius activation between 43% and 52%. Muscle activation for all 10 muscles are described in Figure 51 for subject PAPFL10 while wearing a prosthesis, and Figure 52 while not wearing a prosthesis.

During the bike task, the dominant and non-dominant trapezius muscles were very different for the test condition while wearing a prosthesis compared with the control group. Subject PAPFL10 had no non-dominant trapezius activity throughout the entire bike cycle. Subject PAPFL10 not wearing a prosthesis (Figure 52) had both dominant

and non-dominant trapezius muscle activity much closer to the typical activation patterns of the control group. The muscle was active between 33% and 40%, and again between 47% and 64%. Therefore, there were two bursts of muscle activity similar to the non-dominant trapezius activity of the control group (Figure 38), however, the first burst happens much later for the test subject compared with the first burst of the control group (15% to 21%). The second muscle burst was similar.

There was more triceps activity when wearing a prosthesis compared with not wearing one, however the deltoids seemed to be more active when not wearing a prosthesis, and the erector spinae muscles were similar between the two test conditions.

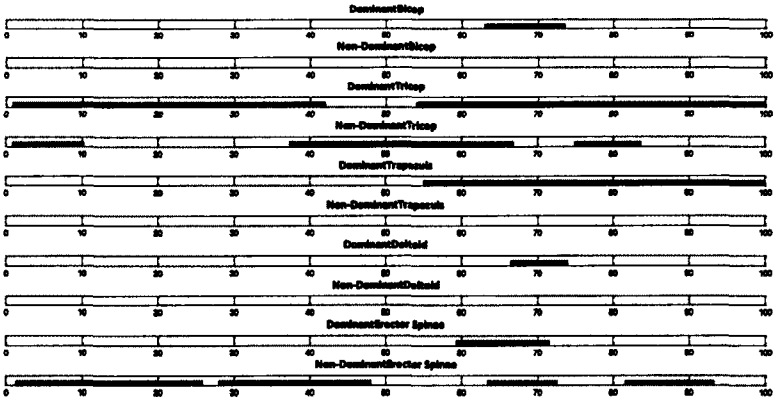


Figure 51: Muscle activation for Subject PAPFL10 while wearing a prosthesis during biking.

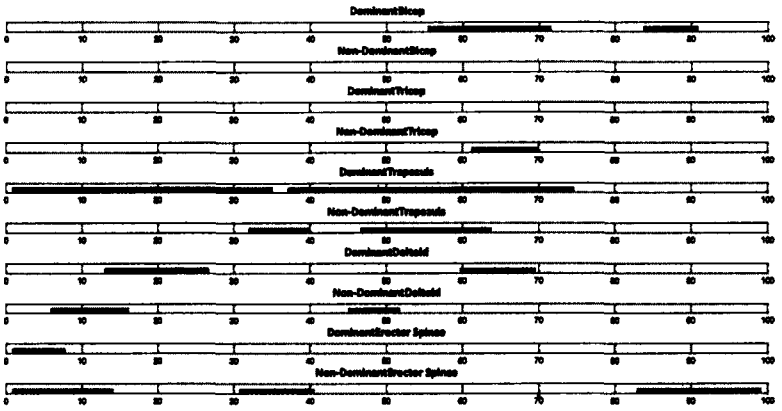


Figure 52: Muscle activation for Subject PAPFL10 while not wearing a prosthesis during biking.

During walking, the dominant erector spinae muscle in subject PAPFL10, while wearing her prosthesis (Figure 53), had similar muscle activation as the patterns displayed by the control group. The muscle activity patterns of the dominant erector spinae for the control group (Figure 40) had one burst of activity between 46% and 55% of the gait cycle. Similarly, the dominant erector spinae for subject PAPFL10 had muscle activation between 43% and 52% of the gait cycle. However, there was additional activity displayed by the dominant erector spinae between 2% and 8% of the gait cycle. The non-dominant erector spinae for subject PAPFL10 while wearing her prosthesis (Figure 53), was active between 1% and 23%, again between 34% and 78%, and finally between 85% and 100%. This appears to be much more active than the non-dominant erector spinae activation patterns of the control group. This is confirmed by comparing the average percent activation levels of the non-dominant erector spinae muscles in the control group in Table 13 (42.4%), with the percent activation level of the non-dominant erector spinae for subject PAPFL10 during walking, while wearing a prosthesis in Table 19 (78.2%).

The test condition while not wearing a prosthesis showed less similarities between the erector spinae muscles compared with the control group. All muscles except the non-dominant deltoids were more active during the test condition while not wearing a prosthesis in comparison to the test condition while wearing a prosthesis. This is re-enforced by Table 19.

.

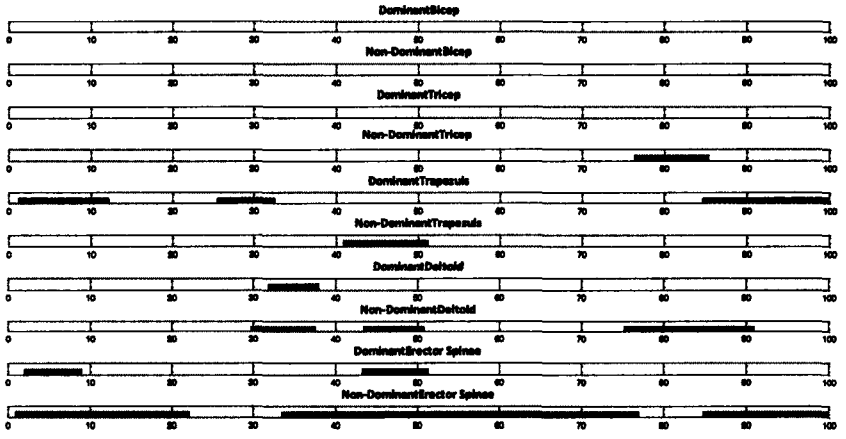


Figure 53: Muscle activation for Subject PAPFL10 while wearing a prosthesis during walking.

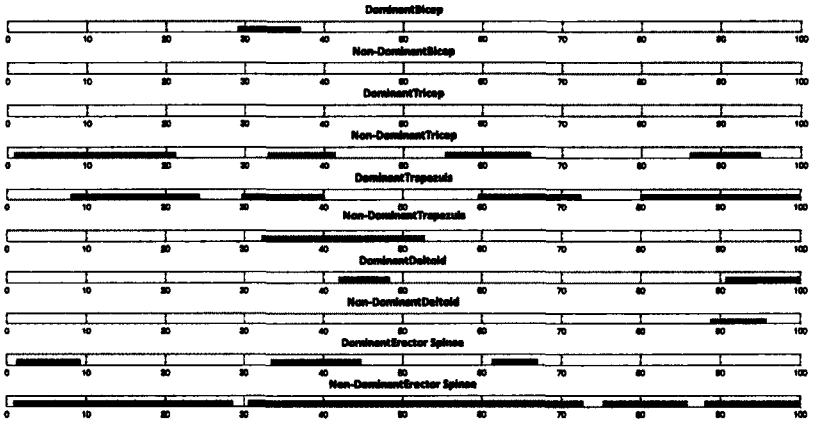


Figure 54: Muscle activation for Subject PAPFL10 while not wearing a prosthesis during walking.

The dominant biceps and trapezius muscles while wearing a prosthesis, were "on" during the entire swing cycle. This is almost double the activity shown by the control group muscle patterns. However, Table 19 describes that there is in fact no difference between the percent activation between the dominant biceps, and trapezius muscles compared with the control group. Figure 55 suggests this is because only one individual's activation timing is recorded here. However, in order to be recorded as "on" for the control group, 14 (+/-2) out of the 16 individuals had to have their muscle "on" at the same time. Therefore, it is important to realize that when looking at the muscle activation patterns, it is necessary to note the Z-scores for the prosthesis user to get a full picture of what is occurring. The deltoid muscle appears to be the most asymmetric while wearing a prosthesis. In general, wearing a prosthesis appears to reduce the muscle symmetry compared with not wearing one during swinging. This is confirmed in Table 17.

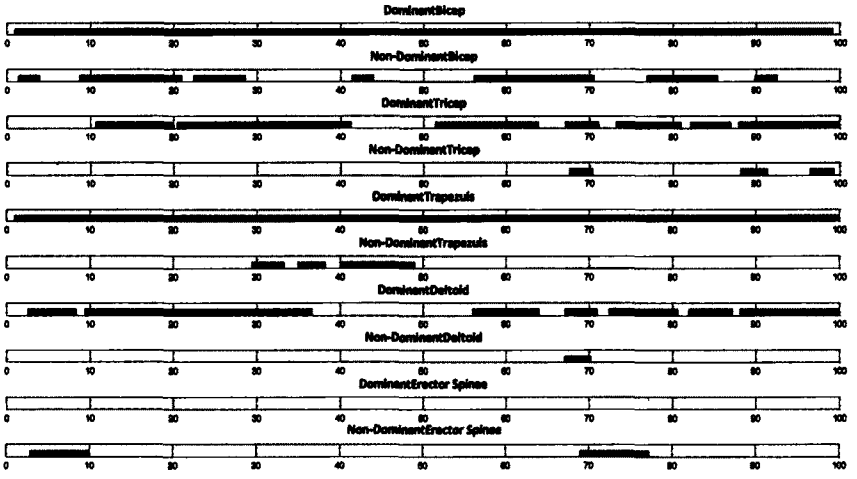


Figure 55: Muscle activation for Subject PAPFL10 while wearing a prosthesis during swinging.

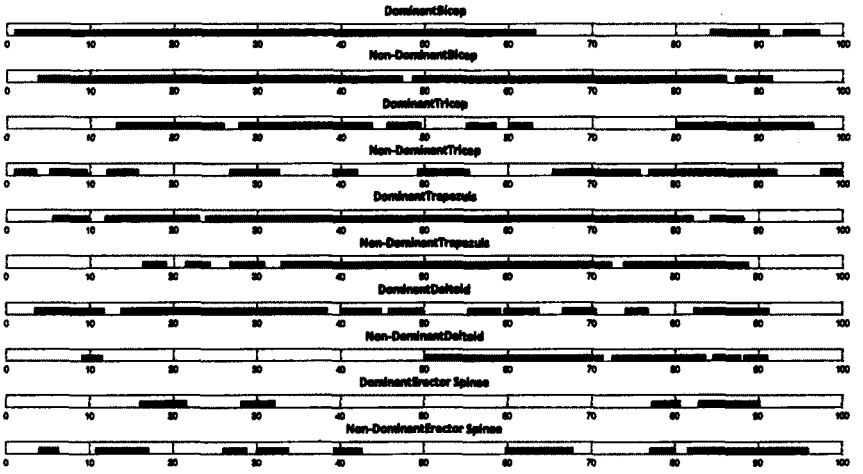


Figure 56: Muscle activation for Subject PAPFL10 while not wearing a prosthesis during swinging.

5.2.2.2 Percent Symmetry

Refer to Table 17 for all muscle symmetry information on subject PAPFL10.

For subject PAPFL10, while wearing a prosthesis, there was high muscle symmetry shown by the biceps, trapezius, and deltoid muscles (all at 100% symmetric during the entire bike cycle). This could be due to some of these muscles being "off" for most of the cycle at the same time. The triceps muscle was symmetric during approximately half of the bike cycle (51.7%), and the erector spinae muscles had the lowest symmetry during this task at 33%. In fact, the erector spinae muscle was 3.5 standard deviations below the erector spinae symmetry of the control group during biking. With no prosthesis, there was lower symmetry demonstrated by the biceps, trapezius, and deltoids compared with the test condition wearing a prosthesis, however the triceps and erector spinae were much

more symmetrical. Overall, without the prosthesis, there were normal levels of percent muscle symmetry for all five muscles during biking.

Throughout both test conditions, subject PAPFL10 had similar muscle symmetry as the control group during walking. The biceps and erector spinae had the highest and lowest muscle symmetry respectively under both test conditions.

The swing task appeared to create the most abnormal muscle symmetry patterns under both test conditions, but particularly while wearing the prosthesis. In this case, the biceps, triceps, trapezius, and deltoid muscles were all much less symmetrical than the control group (Z=-3.4,-2.6,-5.5,-3.6 respectively). Therefore, while wearing a prosthesis, four out of the five muscle pairs examined were much more asymmetric during swinging compared with the same muscles of the control group.
Additionally, while subject PAPFL10 was not wearing a prosthesis during swinging, the triceps and deltoids had lower muscle symmetry than the control group; both having a Z score of -2.7.

Therefore, subject PAPFL10 performed the walking task within normal limits for muscle symmetry, only deviated from the normal with the erector spinae muscle symmetry during biking, but had a large amount of abnormal muscle symmetry for the swing task while wearing a prosthesis and not wearing one.

Table 17: Subject PAPFL10 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Symmetry (Z-Score)

PaPFL10	Bicep	Tricep	Trapezius	Deltoids	Eresp
Bike					
prosthesis	100.0 (0.7)	51.7 (-0.5)	100.0 (2.3)	100.0 (1.8)	33.4 **(-3.5)**
no prosthesis	76.4 (-0.2)	88.0 (1.0)	46.2 (-0.4)	68.2 (0.4)	57.0 (-1.7)
Walking					
prosthesis	100.0 (0.6)	75.9 (-0.2)	72.9 (0.0)	63.5 (-0.8)	36.8 (-1.2)
no prosthesis	91.9 (0.0)	47.0 (-1.5)	49.7 (-1.1)	84.3 (0.2)	31.3 (-1.4)
Swing					
prosthesis	50.7 **(-3.4)**	37.0 **(-2.6)**	16.2 **(-5.5)**	31.9 **(-3.6)**	84.4 (0.6)
no prosthesis	69.2 (-1.0)	36.2 **(-2.7)**	79.7 (0.4)	42.9 **(-2.7)**	62.8 (-0.4)

Refer to Table 18 for all muscle co-contraction information on subject PAPFL10.

Subject PAPFL10 displayed no large differences in muscle co-contraction for all three activities comparing the test conditions with the control group. In general, the swing task had much higher levels of co-contraction compared with the bike and walking task.

Table 18: Subject PAPFL10 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Co-Contraction (Z-Score)

PaPFL10	Dominant Bicep-Tricep	Non-Dominant Bicep-Tricep	Dominant Trapezius-Deltoid	Non-DominantTrapezius-Deltoid
Bike				
prosthesis	12.0 (0.1)	0.0 (-0.5)	17.0 (-0.5)	0.0 (-0.9)
no prosthesis	0.0 (-0.7)	0.0 (-0.5)	32.6 (-1.1)	13.7 (-0.9)
Walking				
prosthesis	0.0 (-0.3)	0.0 (-0.3)	0.0 (-1.0)	41.5 (1.3)
no prosthesis	0.0 (-0.3)	0.0 (-0.3)	11.2 (-0.4)	0.0 (-0.7)
Swing				
prosthesis	28.7 (-0.6)	50.1 (0.8)	28.8 (-0.8)	19.5 (-1.5)
no prosthesis	29.9 (-0.5)	45.1 (0.5)	31.7 (-0.7)	36.1 (-0.4)

5.2.1.4 Percent Activation

Refer to Table 19 for all percent muscle activity information on subject PAPFL10.

During biking and swinging, there were no differences in muscle activity between both test conditions and the control group. During walking, the non-dominant triceps was more active while both wearing and not wearing a prosthesis compared with the control group (Z=2.6,6.5 respectively). However, this should be disregarded based on Table 13, which shows the control group activation of the triceps with higher standard deviations than the mean value for these activations.

Table 19: Subject PAPFL10 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Activation(Z-score).

PaPFL10	Dom Bicep	Non-Dom Bicep	Dom Tricep	Non-Dom Tricep	Dom Trap	Non-Dom Trap	Dom Delt	Non-Dom Delt	Dom Eresp	Non-Dom Eresp
Bike										
prosthesis	10.6 (-0.4)	0.0 (-0.6)	88.0 (1.8)	48.2 (0.3)	45.3 (0.2)	0.0 (-1.6)	7.7 (-1.1)	0.0 (-1.1)	12.3 (-0.5)	66.6 (1.9)
no prosthesis	23.4 (0.1)	0.0 (-0.6)	0.0 (-1.3)	8.9 (-1.0)	72.4 (1.2)	25.1 (-0.8)	23.6 (-0.6)	17.0 (-0.4)	7.0 (-0.7)	39.7 (0.7)
Walking										
prosthesis	0.0 (-0.4)	0.0 (-0.4)	0.0 (-0.7)	24.1 **(2.6)**	16.4 (-0.6)	24.6 (-0.9)	0.0 (-1.0)	42.7 (1.1)	10.6 (-1.3)	78.4 (1.3)

PaPFL10	Dom Bicep	Non-Dom Bicep	Dom Tricep	Non-Dom Tricep	Dom Trap	Non-Dom Trap	Dom Delt	Non-Dom Delt	Dom Eresp	Non-Dom Eresp
no prosthesis	18.2 (2.3)	0.0 (-0.4)	6.3 (-0.4)	53.0 **(6.5)**	77.6 (1.4)	24.4 (-0.9)	10.0 (-0.5)	0.0 (-0.7)	37.6 (0.2)	91.9 (1.7)
Swing										
prosthesis	99.2 (1.4)	49.9 (-1.6)	72.1 (0.3)	9.1 **(-2.9)**	100 **(2.8)**	16.2 **(-3.3)**	71.2 (0.4)	3.3 (-2.3)	0.0 (-0.9)	15.6 (-0.5)
no *prosthesis*	74.6 (0.0)	86.5 (0.5)	56.4 (-0.3)	55.2 (-0.7)	78.8 (1.2)	65.2 (-0.2)	66.5 (0.3)	41.6 (-0.7)	20.8 (0.6)	45.3 (0.6)

Overall, subject PAPFL10 had no large differences from the control group for co-contraction throughout all tasks, as well as no differences in overall muscle activity for the bike tasks. During swinging, while wearing a prosthesis, the non-dominant tricep (-2.9) and trapezius (-3.3) muscles were both less active compared to the control group. Additionally, the dominant trapezius muscle was more active than the control group (2.8).

The most prominent deviations from the control group were apparent within percent muscle symmetry while wearing a prosthesis during swinging. The biceps, triceps, trapezius and deltoid muscles were all much less symmetrical than the control group, and none of them were symmetrical for greater than 50% of the swing cycle. Therefore, half of the time during the swing cycle, the right and left muscles were acting opposite to one another. While not wearing a prosthesis, the triceps and deltoids were also much less symmetrical than the control group. This subject had relatively normal muscle activity

except for muscle symmetry during swinging, where both wearing a prosthesis and not wearing one had large deviations from the control group. This is intuitive, especially for the condition while wearing a prosthesis, because she had difficulty getting her prosthetic wrist to rotate enough to have a normal grip on the swing chains. This caused her to have an unbalanced posture while swinging with her prosthesis on. This is evident in Figure 57; which was the point just before she reached her furthest point forward in the forward swing. Under both conditions (wearing a prosthesis versus not wearing it), she was twisting her right shoulder towards the back left corner of the room. It was more pronounced when she was wearing her prosthesis.

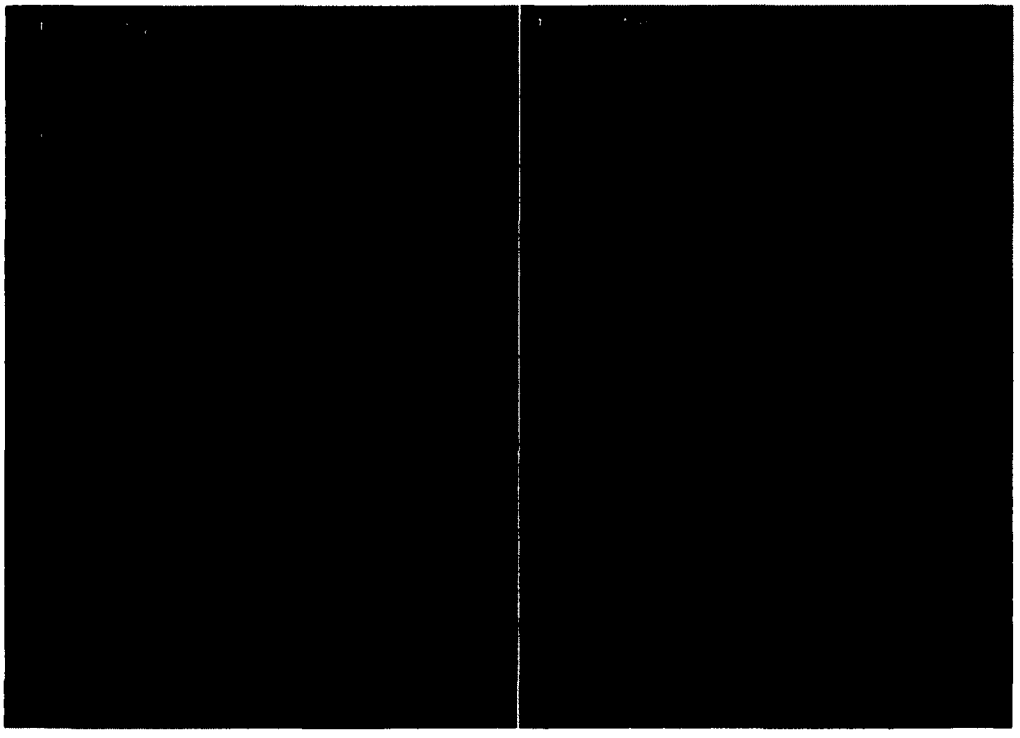


Figure 57: Snapshot of PAPFL10 performing the swing task without prosthesis (left) and with prosthesis (right)

5.2.3 Subject RCPFR06

This test subject was 6 years old, and female, with right hand dominance and limb loss on the left side (elbow disarticulation).

5.2.3.1 Muscle Activation Patterns

During biking, the control group (Figure 38) only exhibited consistent muscle patterns for the dominant and non-dominant trapezius muscles. Subject RCPFR06, while wearing a prosthesis (Figure 58), had no activity from the dominant and non-dominant trapezius muscles, and therefore displayed no similarities with the control group muscle patterns. However, subject RCPFR06 not wearing a prosthesis (Figure 59) had closer to normal activation levels for the dominant trapezius and non-dominant trapezius muscles. Subject RCPFR06 had dominant trapezius activation between 45% and 50% of the bike cycle, compared with 16% to 21% in the control group. Subject RCPFR06 had non-dominant trapezius activation between 58% and 64%, and again between 84% and 88%, compared with 34% to 35%, and again between 41% and 52%. Therefore, subject RCPFR06 was having activation levels that were similar, but the timing of them was occurring much later in the bike cycle compared with the control group. This might be due to the fact that (unlike the control group) this subject did not lean forward while biking. The trunk sway was probably similar to the control group (yielding similar activation), but since they were upright, the timing may have been altered.

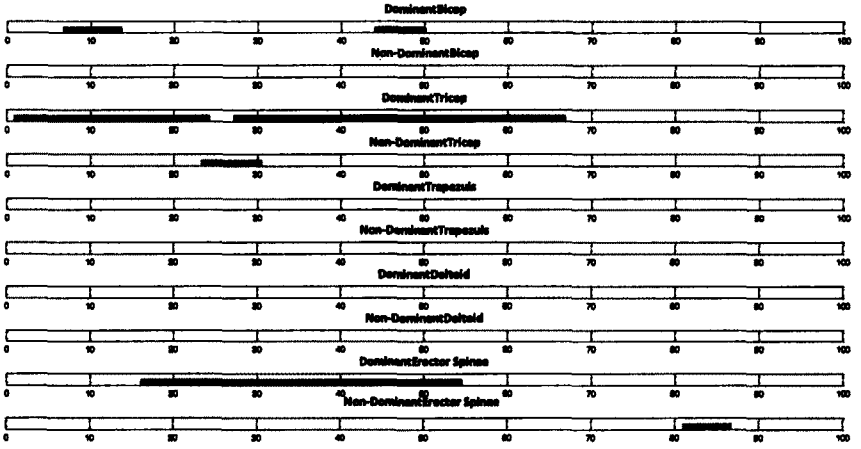


Figure 58: Muscle activation for Subject RCPFR06 while wearing a prosthesis during biking.

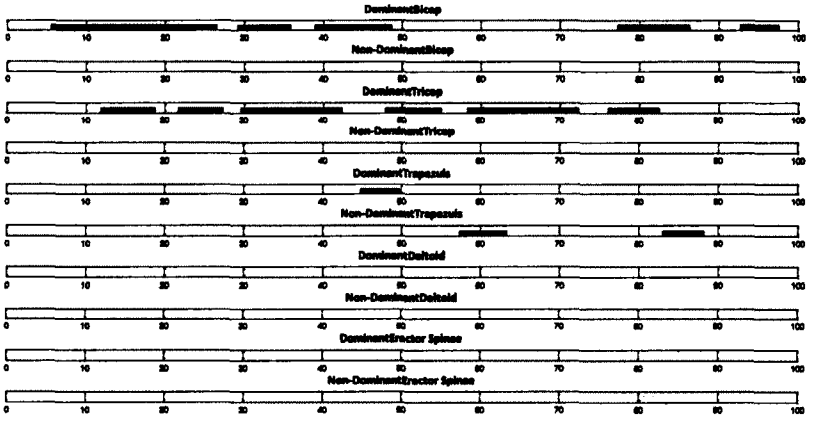


Figure 59: Muscle activation for Subject RCPFR06 while not wearing a prosthesis during biking.

The walking task had activation in six muscles while wearing a prosthesis (Figure 60), which included the dominant triceps and dominant deltoid, dominant and non-dominant trapezius, and erector spinae. While not wearing a prosthesis (Figure 61), subject RCPFR06 had activation in four muscles, including the dominant and non-dominant trapezius and erector spinae muscles. There was similar activation levels in the dominant erector spinae, and less than normal activation by the non-dominant erector

spinae compared with the control group average. The activation patterns of the dominant and non-dominant erector spinae were similar to those shown by the control group.

While not wearing a prosthesis, there was lower activation in all muscles compared to the control group except for the non-dominant trapezius muscle and the dominant erector spinae. The non-dominant erector spinae showed similar activation timing as that of the control group, with a longer period of activation around the 50% point of the gait cycle.

When comparing the two test conditions (Table 21 indicates that) while wearing a prosthesis, there was a higher percent activation in all muscles with greater than 0% activation except for the two erector spinae muscles (which were both more active while not wearing a prosthesis). Additionally, based on Figure 62 and Figure 63, there was more activation in the dominant and non-dominant trapezius muscles while wearing a prosthesis compared with not wearing one.

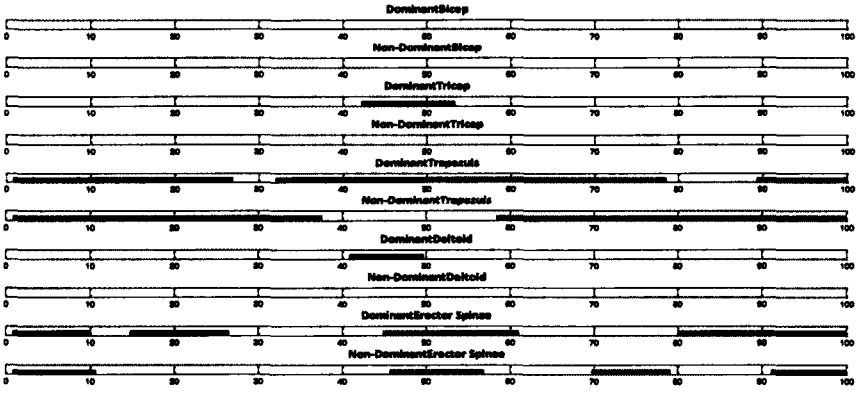


Figure 60: Muscle activation patterns for Subject RCPFR06 while wearing a prosthesis during walking.

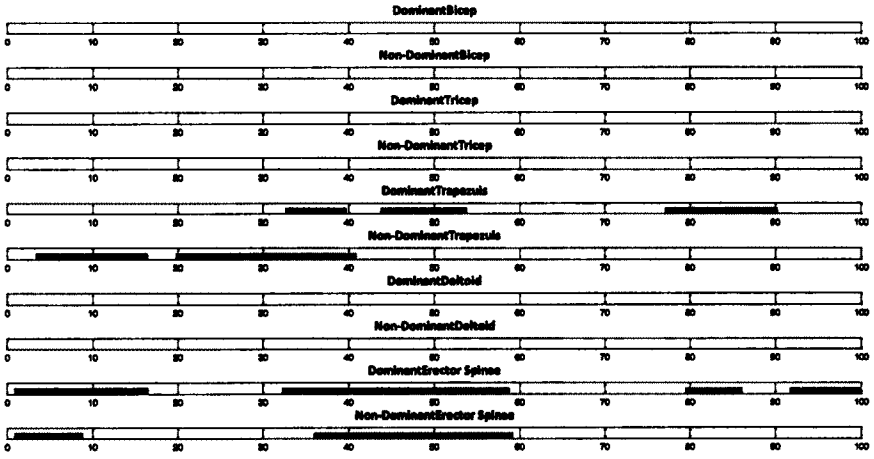


Figure 61: Muscle activation for Subject RCPFR06 while not wearing a prosthesis during walking.

During swinging, subject RCPFR06 seems less symmetrical than the control group when comparing Figure 62 with Figure 43. This is re-enforced by Table 20 as well, showing some large Z-scores under both test conditions. The only muscle pair that was similar to the control group was the dominant and non-dominant biceps while not wearing a prosthesis during swinging. Even so, there is much more activation after the 50% point in the cycle compared with the control group muscle patterns. However, the test condition wearing a prosthesis, shows little similarities to the control group for all muscles.

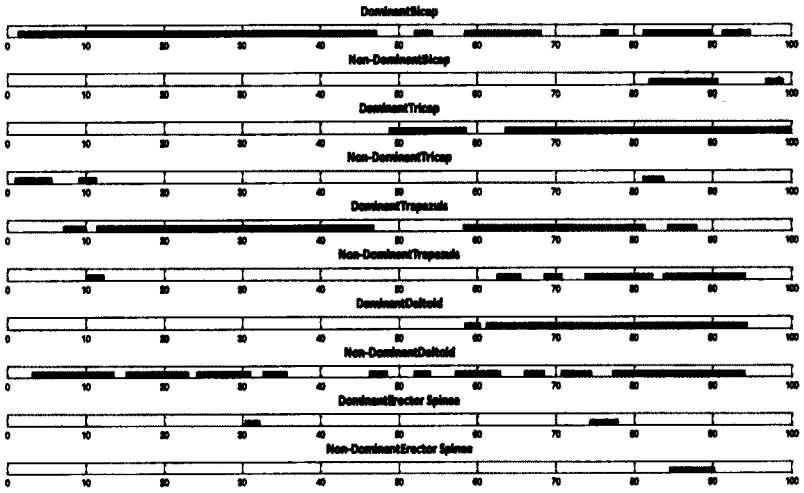


Figure 62: Muscle activation for Subject RCPFR06 while wearing a prosthesis during swinging.

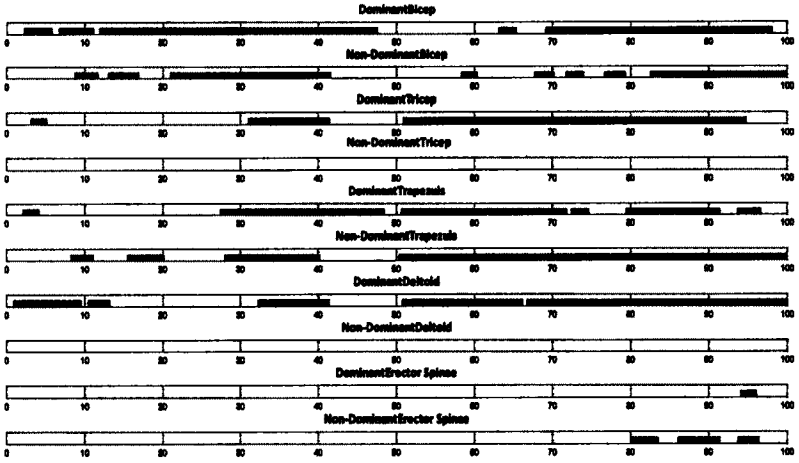


Figure 63: Muscle activation for Subject RCPFR06 while not wearing a prosthesis during swinging.

5.2.3.2 Percent Symmetry

Refer to Table 20 for all muscle symmetry information on subject RCPFR06.

Subject RCPFR06 had more muscle symmetry for all muscles not wearing a prosthesis compared with wearing one except for the trapezius and deltoid muscles (which were equal). However, both test conditions displayed normal levels of muscle symmetry throughout the bike task.

Comparing the two test conditions for subject RCPFR06 illustrates that the biceps muscles had equal percent muscle symmetry (100%), but when wearing a prosthesis there was more symmetry for the trapezius and erector spinae muscles and less symmetry for the triceps and deltoids. Similar to the bike task, both test conditions showed normal levels of muscle symmetry for all five muscles.

During swinging, this subject had less biceps and trapezius symmetry compared with the control group while wearing a prosthesis (Z=-5.9,-3.0 respectively). While not wearing a prosthesis, this subject had less deltoid muscle symmetry compared with the control group during swinging (Z-score=-3.8). Overall, the bike and walking tasks, had higher muscle symmetry than the swing task.

Table 20: Subject RCPFR06 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Symmetry (Z-Score)

RcPFR06	**Bicep**	**Tricep**	**Trapezius**	**Deltoids**	**Eresp**
Bike					
prosthesis	86.8 (0.2)	36.2 (-1.1)	94.4 (2.0)	100.0 (1.8)	67.3 (-1.0)
no prosthesis	91.5 (0.4)	61.6 (-0.1)	78.9 (1.2)	100.0 (1.8)	100.0 (1.5)
Walking					
prosthesis	100.0 (0.6)	52.8 (-1.2)	73.4 (0.1)	91.6 (0.5)	69.0 (0.5)
no prosthesis	100.0 (0.6)	100.0 (0.8)	65.1 (-0.3)	100.0 (0.9)	64.0 (0.2)
Swing					
prosthesis	31.7 **(-5.9)**	49.0 (-1.9)	43.3 **(-3.0)**	56.3 (-1.5)	88.4 (0.7)
no prosthesis	67.8 (-1.2)	43.1 (-2.2)	70.0 (-0.5)	30.2 **(-3.8)**	90.5 (0.8)

5.2.3.3 Percent Co-Contraction

Refer to Table 21 for all muscle co-contraction information on subject RCPFR06. Although there were normal co-contraction levels by all three tasks, the swing task had the highest levels of muscle co-contraction compared with the walking and bike tasks.

Table 21: Subject RCPFR06 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Co-Contraction (Z-Score)

RcPFR06	Dominant Bicep-Tricep	Non-Dominant Bicep-Tricep	Dominant Trapezius-Deltoid	Non-DominantTrapezius-Deltoid
Bike				
Prosthesis	20 (0.6)	0.0 (-0.5)	0.0 (-1.1)	0.0 (-0.9)
no prosthesis	36.1 (1.6)	0.0 (-0.5)	0.0 (-1.1)	0.0 (-0.9)
Walking				
Prosthesis	0.0 (-0.2)	0.0 (-0.3)	9.0 (-0.5)	9.8 (-0.2)
no prosthesis	0.0 (-0.3)	0.0 (-0.3)	0.0 (-1.0)	0.0 (-0.7)
Swing				
Prosthesis	76.2 (2.2)	17.0 (-1.0)	48.6 (0.3)	51.2 (0.5)
no prosthesis	50.9 (0.7)	55.2 (1.1)	32.9 (-0.6)	70.2 (1.6)

5.2.3.4 Percent Activation

Refer to Table 22 for all muscle activity information on subject RCPFR06.

Subject RCPFR06 showed normal levels of percent muscle activation during biking and swinging, and therefore had no differences between test conditions and control group for these two tasks.

During swinging, the non-dominant bicep, non-dominant tricep, and non-dominant trapezius had less activity while wearing their prosthesis compared with the control group. Therefore, the side with limb loss had less muscle activation while wearing a prosthesis than the control group did on their non-dominant side. Similarly, while not wearing a prosthesis the non-dominant triceps and non-dominant deltoid had less activity than the control group.

For subject RCPFR06, the swing task had the most abnormal muscle activity, including muscle symmetry and overall activity. While wearing a prosthesis the deltoids and erector spinae muscles were the only ones who had muscle symmetry for more than 50% of the swing cycle. The most abnormal muscle symmetry occurred when this subject was wearing a prosthesis during the swing task, where the biceps and trapezius muscles were much less symmetrical than the control group. While not wearing a prosthesis, there seemed to be slightly higher muscle symmetry overall compared with her wearing her prosthesis, however there was still much less deltoid muscle symmetry compared with the control group.

Overall muscle activity was less than normal during swinging, both while wearing a prosthesis and not wearing one. This probably was the case because this subject stated that at home she does not normally swing with her prosthesis on, and during this task, her prosthesis was in fact not grabbing the grip on the swing chain, but instead it was at her side, wrapped around the bottom of the chain. This subject also normally bikes with her

prosthesis on, and when she was asked to perform the bike task without her prosthesis, she chose to bike sitting up straight because she did not feel comfortable leaning over and placing her residual limb on the handle bar. This subject seemed to have similar deviations from normal comparing wearing a prosthesis versus not, with slightly closer to normal muscle activity while not wearing a prosthesis.

Table 22: Subject RCPFR06 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Activation (Z-Score)

RcPFR06	**Dom Bicep**	**Non-Dom Bicep**	**Dom Tricep**	**Non-Dom Tricep**	**Dom Trap**	**Non-Dom Trap**	**Dom Delt**	**Non-Dom Delt**	**Dom Eresp**	**Non-Dom Eresp**
Bike										
prosthesis	13.3 (-0.3)	0.0 (-0.6)	63.6 (0.9)	7.2 (-1.0)	0.0 (-1.5)	0.0 (-1.6)	0.0 (-1.3)	0.0 (-1.1)	38.9 (0.8)	5.9 (-0.8)
no prosthesis	51.9 (0.1)	0.0 (-0.6)	53.4 (-1.3)	0.0 (-1.0)	4.9 (1.2)	11.6 (-0.8)	0.0 (-0.6)	0.0 (-0.4)	0.0 (-0.7)	0.0 (0.7)
Walking										
prosthesis	0.0 (-0.4)	0.0 (-0.4)	47.2 (1.4)	0.0 (-0.7)	94.3 (2.0)	77.0 (1.0)	8.4 (-0.6)	7.6 (-0.4)	24.5 (-0.6)	8.6 (-1.2)
no prosthesis	0.0 (-0.4)	0.0 (-0.4)	0.0 (-0.7)	0.0 (-0.7)	20.5 (-0.4)	58.8 (0.3)	0.0 (-1.0)	0.0 (-0.7)	67.3 (1.7)	39.0 (-0.1)

RcPFR06	Dom Bicep	Non-Dom Bicep	Dom Tricep	Non-Dom Tricep	Dom Trap	Non-Dom Trap	Dom Delt	Non-Dom Delt	Dom Eresp	Non-Dom Eresp
Swing										
prosthesis	73.1	11.0	46.7	9.7	65.6	27.0	35.7	63.0	5.7	5.9
	(0.0)	**(-3.9)**	(-0.6)	**(-2.9)**	(0.2)	**(-2.6)**	(-0.8)	(0.2)	(-0.5)	(-0.8)
no prosthesis	75.5	55.2	56.9	0.0	62.3	70.2	69.8	0.0	2.1	11.6
	(0.1)	(-1.3)	(-0.2)	**(-3.3)**	(-0.1)	(0.1)	(0.4)	**(-2.4)**	(-0.7)	(-0.6)

5.2.4 Subject WRPFR13

This test subject was 13 years old, and female, with right hand dominance and limb loss on the left side (transradial).

5.2.4.1 Muscle Activation Patterns

During biking, subject WRPFR13 had a lot of triceps activity while both wearing a prosthesis (Figure 64) and not wearing one (Figure 65), however these high activation levels were not much different than those of the control group (see Table 25). While not wearing a prosthesis, there was slightly more activation in all muscles except the non-dominant triceps and both erector spinae muscles. Figure 65 also suggests that there was more asymmetry while not wearing a prosthesis compared with wearing one. This is confirmed by Table 23, which shows that all muscles except for the erector spinae's are more symmetrical while wearing a prosthesis compared with not wearing one.

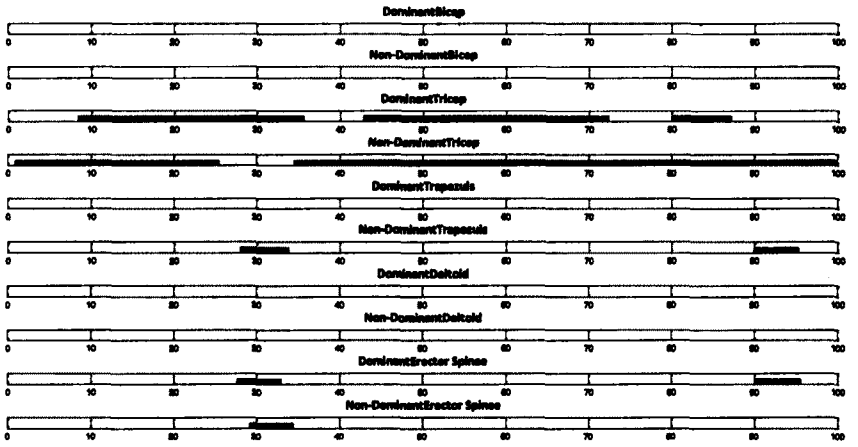


Figure 64: Muscle activation for Subject WRPFR13 while wearing a prosthesis during biking.

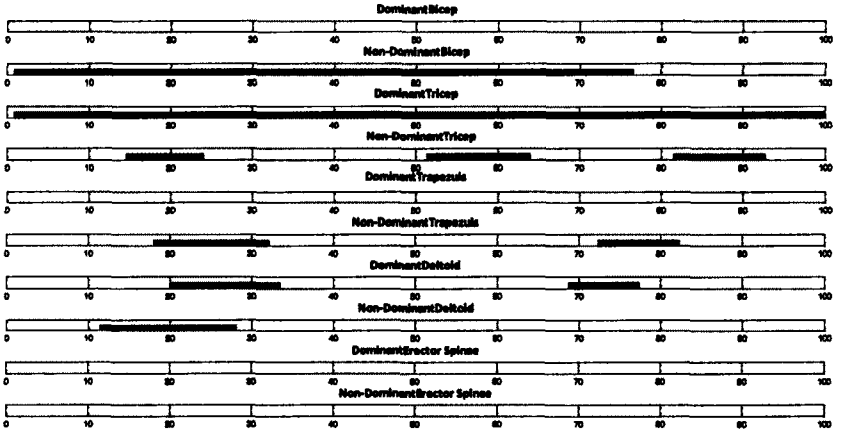


Figure 65: Muscle activation for Subject WRPFR13 while not wearing a prosthesis during biking.

During walking, while wearing a prosthesis (Figure 66), the erector spinae activation timing was closer to normal than the test condition not wearing a prosthesis (Figure 67); the dominant erector spinae being "on" between 45 and 59% and again between 85 and 100% (this portion was "off" for the control group), and the non-dominant erector spinae being "on" between 1 and 9% as well as between 39 and 51% of the gait cycle.

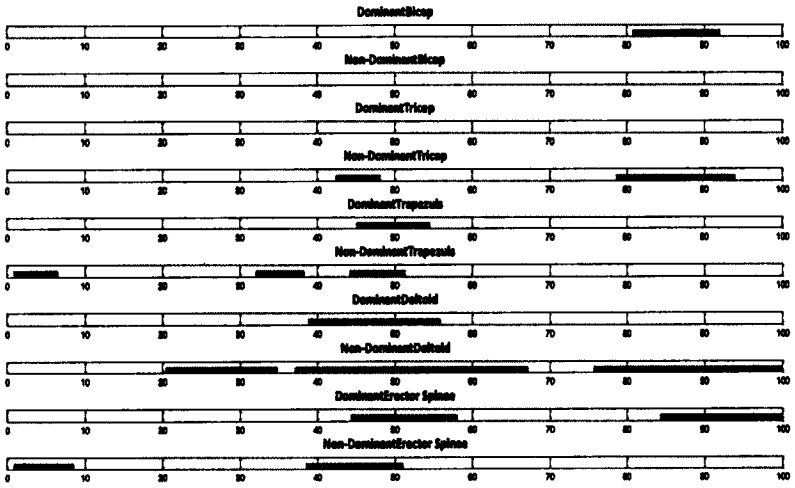


Figure 66: Muscle activation for Subject WRPFR13 while wearing a prosthesis during walking.

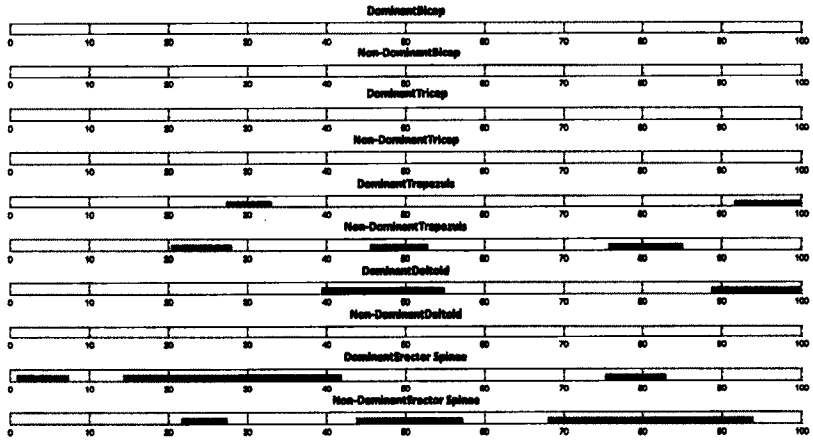


Figure 67: Muscle activation for Subject WRPFR13 while not wearing a prosthesis during walking.

During swinging, Figure 68 and Figure 69 illustrate the different timing the muscles are activated compared with the control group (Figure 43). Under both test conditions, there is a lot of muscle activation after the 60% point in the cycle for the dominant and non-dominant biceps and trapesuis muscles, which is not present for the control group.

Similarly, there is also a lot of muscle activation prior to the 50% point in the cycle in the dominant and non-dominant triceps, which does not occur to the same extent amongst the controls. Lastly, there seems to be more erector spinae activity under both test conditions compared with the control group. This is true for the dominant erector spinae as seen by two large and positive Z-scores in Table 25. There seems to be little difference between the two test conditions for the swing task for subject WRPFR13.

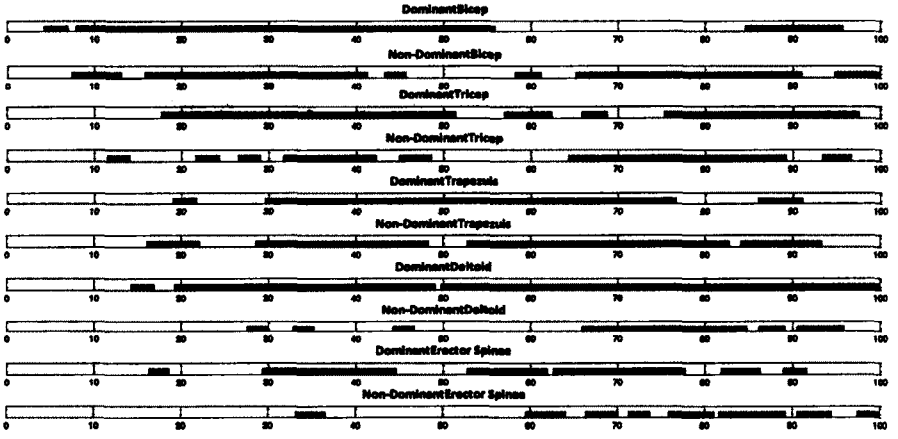


Figure 68: Muscle activation for Subject WRPFR13 while wearing a prosthesis during swinging.

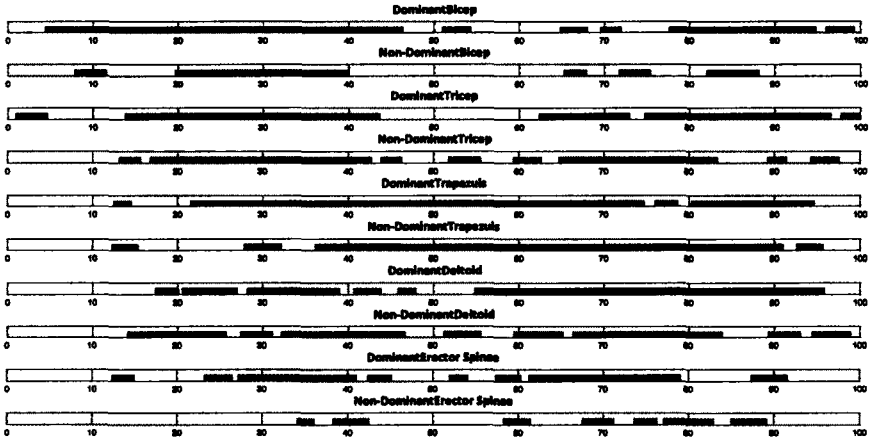


Figure 69: Muscle activation for Subject WRPFR13 while not wearing a prosthesis during swinging.

5.2.4.2 Percent Symmetry

Refer to Table 23 for all muscle symmetry information on subject WRPFR13.

Subject WRPFR13 had higher muscle symmetry while wearing a prosthesis for the biceps, triceps, trapezius and deltoids compared with her performing the task without the prosthesis. Normal muscle symmetry was displayed by subject WRPFR13 while wearing a prosthesis for all muscles, and for four out of the five muscles while not wearing the prosthesis. However, while not wearing the prosthesis, this subject had very low biceps symmetry (only symmetrical for 8% of the swing cycle), and was less symmetrical compared to the control group (Z=-2.7).

There were high levels of muscle symmetry during walking, which were within normal limits under both test conditions.

The highest deviation from normal muscle symmetry occurred during biking while wearing a prosthesis for the biceps muscle (Z-score=-3.4). However, while not wearing a prosthesis, the triceps had much lower levels of muscle symmetry when compared to the control group as well (Z-score=-2.8). In general, the swing task had the lowest percent muscle symmetry out of all three tasks.

Table 23: Subject WRPFR13 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Symmetry (Z-Score)

WRPFR13	Bicep	Tricep	Trapezius	Deltoids	Eresp
Bike					
Prosthesis	100.0 (0.7)	69.8 (0.3)	94.4 (2.0)	100.0 (1.8)	89.0 (0.6)
no prosthesis	7.8 **(-2.7)**	25.6 (-1.5)	85.3 (1.6)	78.0 (0.8)	100.0 (1.5)
Walking					
prosthesis	100.0 (0.6)	78.6 (-0.1)	69.7 (-0.1)	43.8 (-1.8)	78.3 (0.9)
no prosthesis	100.0 (0.6)	100.0 (0.8)	84.5 (0.6)	73.0 (-0.4)	73.1 (0.7)
Swing					
prosthesis	50.8 **(-3.4)**	64.6 (-0.8)	80.7 (0.5)	51.4 (-1.9)	59.8 (-0.6)
no prosthesis	57.9 (-1.6)	68.6 **(-2.8)**	83.1 (-0.1)	69.7 (-1.6)	58.6 (-1.4)

5.2.4.3 Percent Co-Contraction

Refer to Table 24 for all muscle co-contraction information on subject WRPFR13. Throughout all three tasks, subject WRPFR13 displayed normal levels of co-contraction between agonist/antagonist muscle pairs. In general, the swing task had higher levels of muscle co-contraction compared with the other two tasks.

Table 24: Subject WRPFR13 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Co-Contraction (Z-Score)

WrPFR13	Dominant Bicep-Tricep	Non-Dominant Bicep-Tricep	Dominant Trapezius-Deltoid	Non-DominantTrapezius-Deltoid
Bike				
prosthesis	0.0 (-0.7)	0.0 (-0.5)	0.0 (-1.1)	0.0 (-0.9)
no prosthesis	0.0 (-0.7)	25.5 (1.1)	0.0 (-1.1)	32.9 (0.4)
Walking				
prosthesis	0.0 (-0.3)	0.0 (-0.3)	38.9 (0.9)	43.7 (1.4)
no prosthesis	0.0 (-0.2)	0.0 (-0.3)	25.8 (0.3)	0.0 (-0.7)
Swing				
prosthesis	37.0 (-0.1)	32.3 (-0.2)	29.8 (-0.8)	40.4 (-0.2)
no prosthesis	29.8 (-0.5)	43.2 (0.4)	21.4 (-1.2)	38.7 (-0.3)

5.2.4.4 Percent Activation

Muscle activation levels were normal throughout the bike and walking tasks. The walking task had higher activation levels for the dominant and non-dominant trapezius, deltoids, and erector spinae muscles compared with the bike task.

During swinging, while wearing a prosthesis, the dominant erector spinae was much more active than the control group (Z-score=2.6). While not wearing a prosthesis, the non-dominant biceps was much less active (Z-score=-2.4), and the dominant erector spinae was much more active (Z-score=2.7) compared with the control group.

Refer to Table 25 for all muscle activity information on subject WRPFR13.

For subject WRPFR13, the swing task had the greatest deviation from the control group for muscle symmetry and total muscle activity. However, during biking, there was one large difference in regards to muscle symmetry for subject WRPFR13. While not wearing a prosthesis, this subject had much less biceps muscle symmetry compared with the control group. Therefore, this subject seemed to demonstrate more abnormal muscle activity while not wearing a prosthesis compared wearing one. This makes sense, as compared with the other test subjects this prosthesis user normally wore her prosthesis for both swinging and biking when she was at home.

Table 25: Subject WRPFR13 under both test conditions (prosthesis/no prosthesis) during all three tasks: Percent Muscle Activation (Z-Score)

WrPFR13	Dom Bicep	Non-Dom Bicep	Dom Tricep	Non-Dom Tricep	Dom Trap	Non-Dom Trap	Dom Delt	Non-Dom Delt	Dom Eresp	Non-Dom Eresp
Bike										
Prosthesis	0.0 (-0.8)	0.0 (-0.6)	64.7 (0.9)	91.0 (1.8)	0.0 (-1.5)	11.6 (-1.2)	0.0 (-1.3)	0.0 (-1.1)	11.0 (-0.5)	5.5 (-0.8)
no prosthesis	0.0 (-0.8)	76.3 (2.1)	100.0 (2.2)	33.6 (-0.1)	0.0 (-1.5)	24.4 (-0.8)	22.5 (-0.7)	16.8 (-0.4)	0.0 (-1.1)	0.0 (-1.0)
Walking										
Prosthesis	0.0 (-0.4)	0.0 (-0.4)	0.0 (-0.7)	9.0 (0.5)	16.8 (-0.5)	27.2 (-0.8)	17.9 (-0.1)	52.3 (1.4)	34.2 (-0.0)	27.6 (-0.5)
no prosthesis	0.0 (-0.4)	0.0 (-0.4)	0.0 (-0.7)	0.0 (-0.7)	14.2 (-0.6)	24.6 (-0.9)	27.0 (0.3)	0.0 (-0.7)	42.3 (0.4)	45.5 (0.1)

WrPFR13	Dom Bicep	Non-Dom Bicep	Dom Tricep	Non-Dom Tricep	Dom Trap	Non-Dom Trap	Dom Delt	Non-Dom Delt	Dom Eresp	Non-Dom Eresp
Swing										
prosthesis	62.5 (-0.6)	68.7 (-0.5)	65.1 (0.1)	51.2 (-0.9)	55.6 (-0.6)	65.8 (-0.2)	83.6 (0.9)	35.0 (-1.0)	49.5 **(2.6)**	34.2 (0.2)
no prosthesis	72.2 (-0.1)	37.2 **(-2.4)**	69.0 (0.2)	63.1 (-0.4)	73.2 (0.8)	66.2 (-0.2)	67.2 (0.3)	66.8 (0.4)	50.6 **(2.7)**	26.2 (-0.1)

5.3 Test Subjects: Group Comparisons with the Controls

After calculating the Z-scores for each person in the test group, there were enough emerging trends to follow-up with some group statistics. The case studies show some consistent trends. In this section, the non-parametric Mann-Whitney test (also called the Wilcoxon Rank-Sum test) is used to investigate statistical significance of such trends. It is important to use a test that does not assume input data are Normal (Gaussian) because Normality cannot be checked with small samples. The test compares median response for control and treatment groups, assuming that each group is a random selection from the population it represents.

5.3.1 Bike

The only segment analyzed during biking was the portion where each subject was biking straight.

For muscle symmetry during biking, the group findings suggest that there is more muscle symmetry for the deltoids while both wearing a prosthesis (P=0.007) and not wearing one (P=0.05) compared with the control group. This suggests that the test group under both conditions (wearing a prosthesis/not wearing one) had more deltoid percent muscle symmetry compared with the control group. This can be explained by the low activity of the deltoids muscles by the deltoids during biking for most of the prosthesis users. Since there was low deltoid activity, these muscles would seem highly symmetrical because they would be "off" at the same time.

In general, these findings do not agree with all other deviations from muscle symmetry compared to the control group for the swing task and most of the case study deviations from muscle symmetry. These being, that in all cases (except this one), muscle symmetry in the prosthesis user's were less than the control group.

These data are presented in Figure 70. The boxplot's represent the control group values for each muscle and the red crosses represent the outliers of the control group. The individual test subject values are also displayed under both conditions (prosthesis "on" are the solid circles, prosthesis "off" are the open circles).

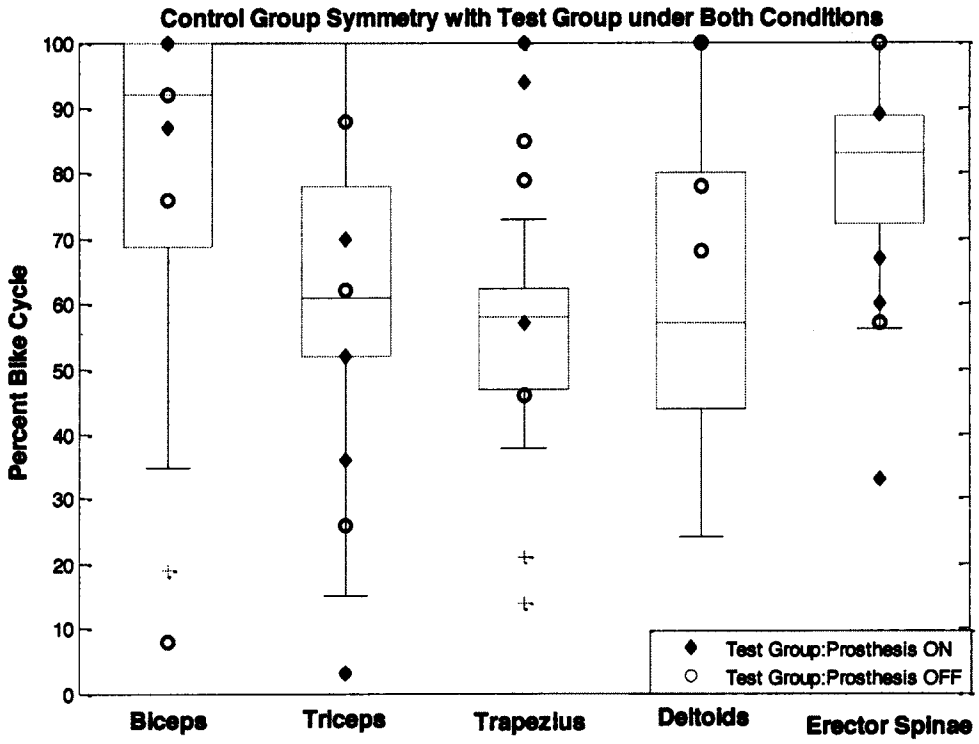


Figure 70: Boxplot of muscle symmetry for all 5 muscles during biking straight

There were no significant differences in muscle co-contraction between the two conditions of the test group and the control group. These data are presented in Figure 71. The boxplot's represent the control group values for each agonist/antagonist muscle pair and the red crosses represent the outliers of the control group. The individual test subject values are also displayed under both conditions (prosthesis "on" are the solid circles, prosthesis "off" are the open circles).

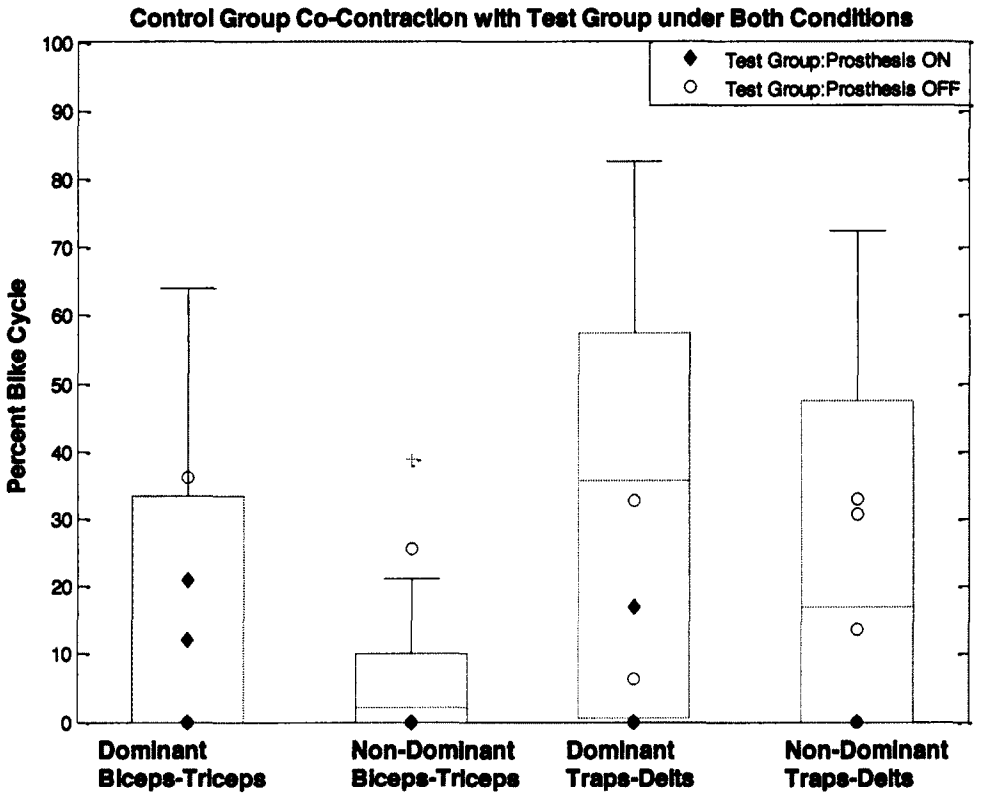


Figure 71: Boxplot of co-contraction between agonist/antagonist muscle pairs during biking straight

The total percent each muscle was activated had much more statistically significant findings than the other two variables. Figure 72 describes the percent activation by the control group and the test group. The median of muscle activity for the control group in descending order is: dominant deltoids (49%) , non-dominant trapezius (48%) , non-dominant triceps (42%), dominant triceps (36%), dominant trapezius (31%), non-dominant deltoids (23%), dominant erector spinae (20%), non-dominant erector spinae (18%), dominant biceps (10%), non-dominant biceps (0%).

During biking, there were some significant differences for total muscle activity; including:

- The dominant triceps of the control group compared with the dominant triceps of the test group wearing a prosthesis; P=0.04
    - Referring to Figure 72, this P value suggests that the dominant triceps of the control group is less active than the dominant triceps of the test group wearing a prosthesis.
- The non-dominant trapezius of the control group compared with the non-dominant trapezius of the test group wearing a prosthesis; P=0.024
    - Referring to Figure 72, this P value suggests that the non-dominant trapezius of the control group is more active than the non-dominant trapezius of the test group wearing a prosthesis.
- The non-dominant trapezius of the control group compared with the non-dominant trapezius of the test group not wearing a prosthesis; P= 0.04
    - Referring to Figure 72, this P value suggests that the non-dominant trapezius of the control group is more active than the non-dominant trapezius of the test group not wearing a prosthesis.
- The dominant deltoid of the control group compared with the dominant deltoid of the test group wearing a prosthesis; P=0.028
    - Referring to Figure 72, this P value suggests that the dominant deltoid of the control group is more active than the dominant deltoid of the test group wearing a prosthesis.
- The non-dominant deltoid of the control group compared with the non-dominant deltoid of the test group wearing a prosthesis; P=0.032

- Referring to Figure 72, this P value suggests that the non-dominant deltoid of the control group is more active than the non-dominant deltoid of the test group wearing a prosthesis.

- The dominant erector spinae of the control group compared with the dominant erector spinae of the test group not wearing their prosthesis; P=0.05
  - Referring to Figure 72, this P value suggests that the non-dominant erector spinae of the control group is more active than the non-dominant erector spinae of the test group not wearing a prosthesis.

- The dominant erector spinae of the test group wearing a prosthesis compared with dominant erector spinae of the test group not wearing their prosthesis; P=0.03
  - Referring to Figure 72, this P value suggests that the dominant erector spinae of the test group wearing a prosthesis is more active than the dominant erector spinae of the test group not wearing a prosthesis.

These data are presented in Figure 72. The boxplot's represent the control group values for each muscle and the red crosses represent the outliers of the control group. The individual test subject values are also displayed under both conditions (prosthesis "on" are the solid circles, prosthesis "off" are the open circles).

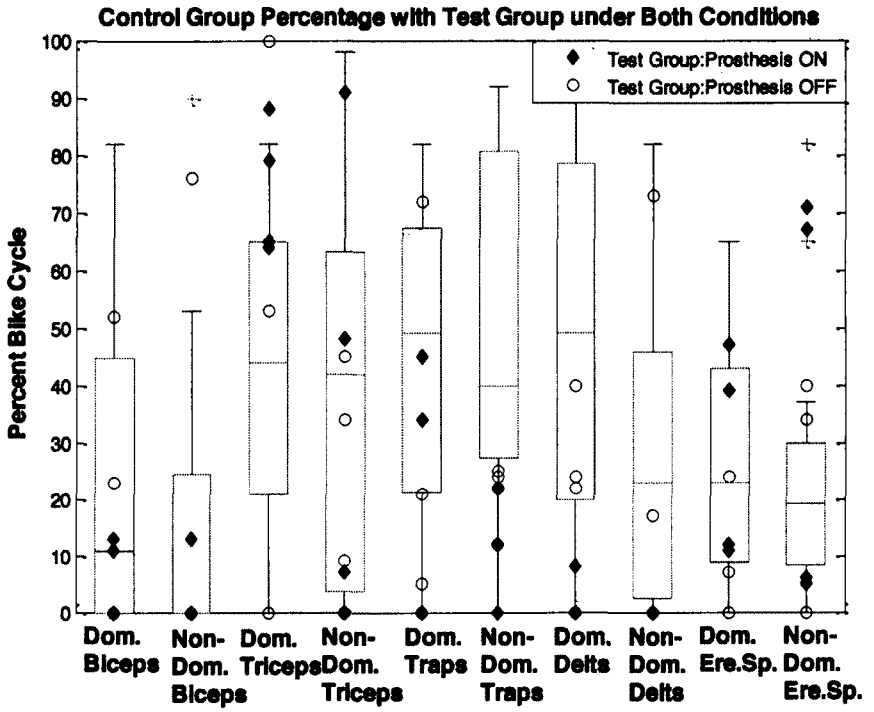


Figure 72: Boxplot of total muscle activity for each muscle during the bike cycle

Since the bike task did not have consistent muscle activity within the control group as shown by the high standard deviations in Tables 11, 12 and 13, the significant findings here are not as strong as differences found within the swing and walking tasks. However, the Mann-Whitney statistical tests does not assume that the control group population was normally distributed. Therefore, these findings cannot be discounted. Both conditions of wearing versus not wearing a prosthesis showed significant differences from the control group. This included the dominant erector spinae, and the non-dominant trapezius. It seems as though wearing a prosthesis contributed to lower muscle activity in the able limb deltoid and triceps and not wearing one contributed to lower muscle activity in the affected limb deltoid.

5.3.2 Walking

There were no significant differences for the walking cycle between the two conditions of the test group and the control group for all three variables (percent muscle symmetry, percent co-contraction, and total muscle percent activation). The data are displayed below in Figures 73 to 75.

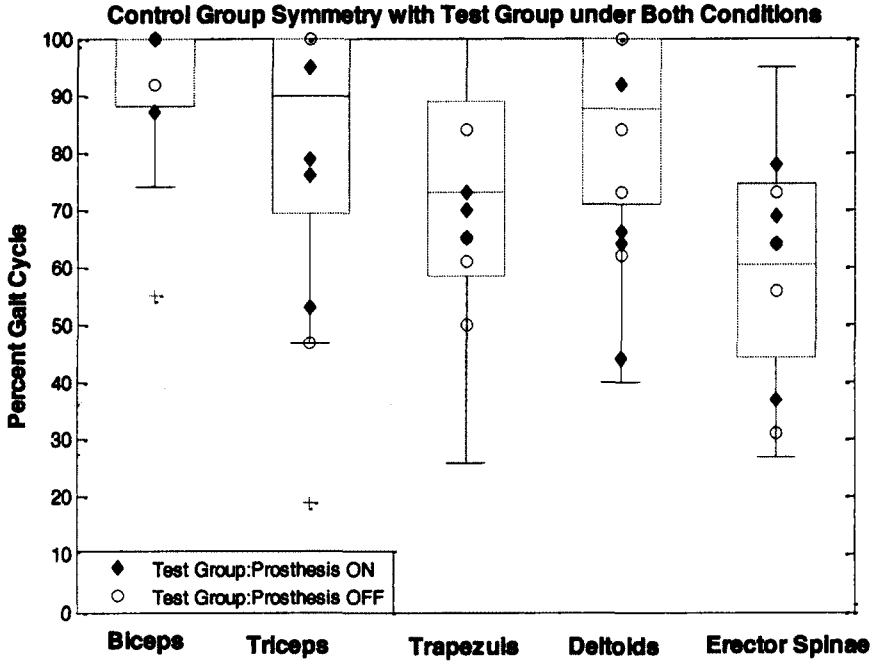


Figure 73: Boxplot of percent muscle symmetry for each muscle pair during walking for the control group and the two conditions of the test group.

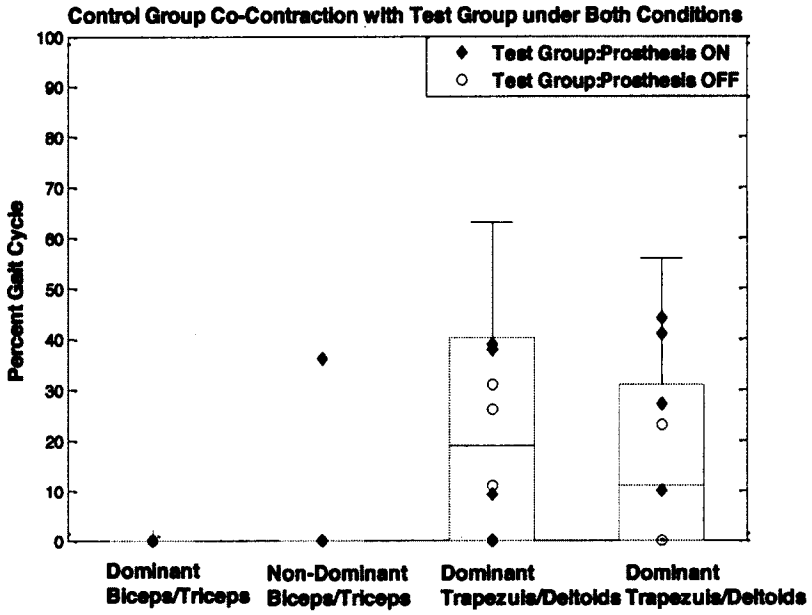


Figure 74: Boxplot of percent muscle co-contraction for each agonist/antagonist muscle pair during walking for the control group and the two conditions of the test group.

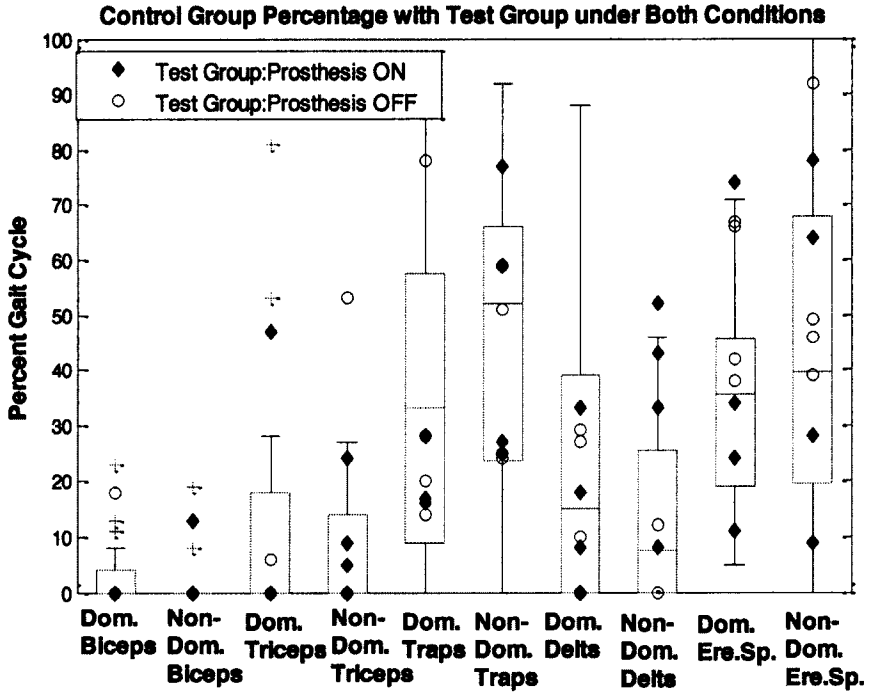


Figure 75: Boxplot of total percent muscle activation for each muscle during walking for the control group and the two conditions of the test group.

5.3.3 Swing

A Mann-Whitney statistical test was performed for the percentage of time each muscle pair (dominant and non-dominant sides) were simultaneously "on" or "off" throughout the swing cycle. This corresponds to the symmetry between right and left sides for all 5 muscles. This may be the best indicator of compensations made by the test groups when performing the swing activity. After the Mann-Whitney statistical test was performed, it was found that there were some statistically significant differences between the test and control groups;

- The biceps of control group compared with test group wearing their prosthesis; P=0.009
    - Referring to Figure 76, this P value suggests that the biceps of the control group have a higher percent symmetry than the biceps of the test group wearing a prosthesis.
- The biceps of control group compared with test group not wearing their prosthesis; P=0.012
    - Referring to Figure 76, this P value suggests that the biceps of the control group have a higher percent symmetry than the biceps of the test group not wearing a prosthesis.
- The triceps of control group compared with test group wearing their prosthesis; P=0.016
    - Referring to Figure 76, this P value suggests that the triceps of the control group have a higher percent symmetry than the triceps of the test group wearing a prosthesis.

- The triceps of control group compared with test group not wearing their prosthesis; P=0.0094
    - Referring to Figure 76, this P value suggests that the triceps of the control group have a higher percent symmetry than the triceps of the test group not wearing a prosthesis.
- The deltoids of control group compared with test group wearing their prosthesis; P=0.004
    - Referring to Figure 76, this P value suggests that the deltoids of the control group have a higher percent symmetry than the deltoids of the test group wearing a prosthesis.
- The deltoids of control group compared with test group not wearing their prosthesis; P=0.021
    - Referring to Figure 76, this P value suggests that the deltoids of the control group have a higher percent symmetry than the deltoids of the test group not wearing a prosthesis.

These data are presented in Figure 76. The boxplot's represent the control group values for each muscle and the red crosses represent the outliers of the control group. The individual test subject values are also displayed under both conditions (prosthesis "on" are the solid circles, prosthesis "off" are the open circles).

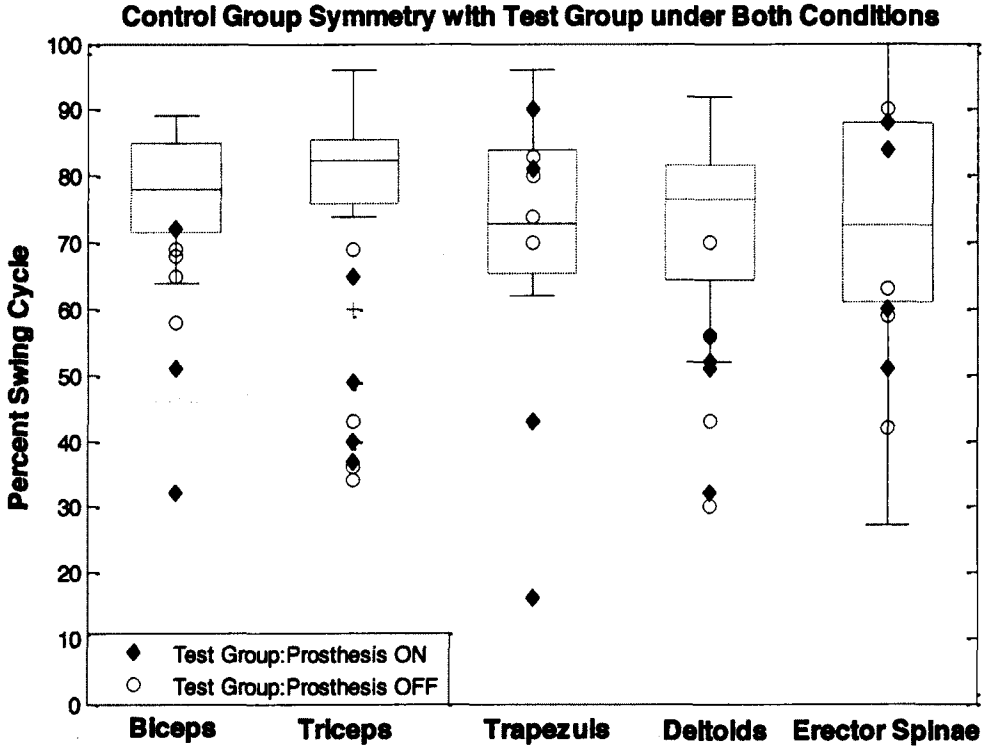


Figure 76: Boxplot of the total percentage of symmetry for each during swinging

Therefore, for muscle symmetry, 3 out of the 5 muscle groups examined showed significant differences between the control group and both the prosthesis wearers and non-wearers. The muscle symmetry for all muscles in the control group was at approximately between 70-80% of the entire swing cycle. The muscles in the test group that displayed significant differences from the control group were always lower for the test groups than the control group, demonstrating less symmetry among children with limb loss (which was expected).

This supports the hypothesis that prosthesis users would have different muscle activity than normal, however this also showed that one test condition (wearing a prosthesis versus not wearing one) did not have muscle symmetry closer to normal than the other.

The second variable that was statistically analysed was the occurrence of co-contraction between agonist/antagonist muscles on the same side of the body. Figure 77 describes the boxplot for the occurrence of co-contraction throughout the swing cycle. Most of the muscle pairs had co-contraction throughout approximately 40% of the swing cycle. Therefore, the normal activity levels showed almost half the swing cycle where agonist and antagonist muscle pairs were active at the same point during the cycle. There were no significant differences between the control group, and either of the test groups, for all agonist/antagonist muscle pairs. These data are presented in Figure 77. The boxplot's represent the control group values for each agonist/antagonist muscle pair and the red crosses represent the outliers of the control group. The individual test subject values are also displayed under both conditions (prosthesis "on" are the solid circles, prosthesis "off" are the open circles).

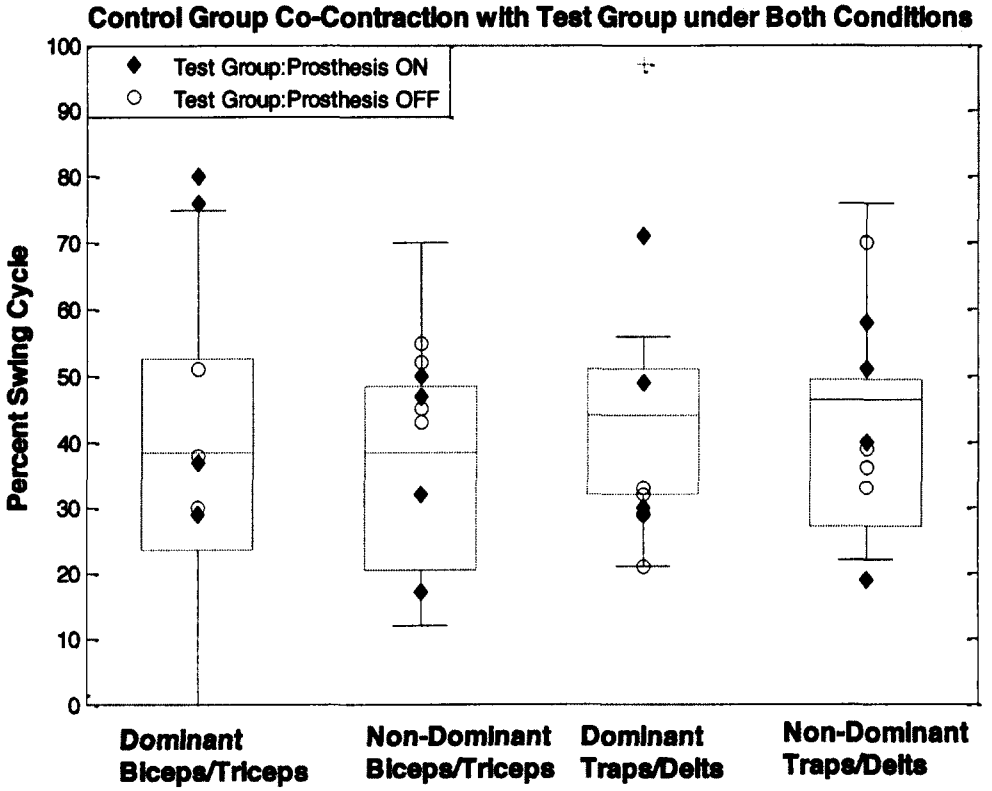


Figure 77: Boxplot of the total percentage of co-contraction per swing cycle for each Agonist/Antagonist Muscle pair

A Mann-Whitney statistical test was performed for the percentage of time each muscle was active throughout the swing cycle, the results for all muscles are shown in the boxplot in Figure 78. There were a few significant differences with P-Value's less than 0.05, corresponding to a 95% confidence interval, including:

- The non-dominant biceps of the control group compared with the non-dominant biceps of the test group wearing a prosthesis; P=0.021

- Referring to Figure 78, this P value suggests that the non-dominant biceps of the control group is more active than the non-dominant biceps of the test group wearing a prosthesis.

- The non-dominant triceps of the control group compared with the non-dominant triceps of the test group wearing a prosthesis; P=0.016

    - Referring to Figure 78, this P value suggests that the non-dominant triceps of the control group is more active than the non-dominant triceps of the test group wearing a prosthesis.

- The non-dominant triceps of the control group compared with the non-dominant triceps of the test group not wearing a prosthesis; P=0.016

    - Referring to Figure 78, this P value suggests that the non-dominant triceps of the control group is more active than the non-dominant triceps of the test group not wearing a prosthesis.

Therefore, the non-dominant side (side with limb loss) for the biceps and triceps had less activity than the control group during swinging. While wearing a prosthesis, there were two muscles that exhibited less than normal activity, and not wearing one only had one muscle less active than normal.

These data are presented in Figure 78. The boxplot's represent the control group values for each muscle and the red crosses represent the outliers of the control group. The individual test subject values are also displayed under both conditions (prosthesis "on" are the solid circles, prosthesis "off" are the open circles).

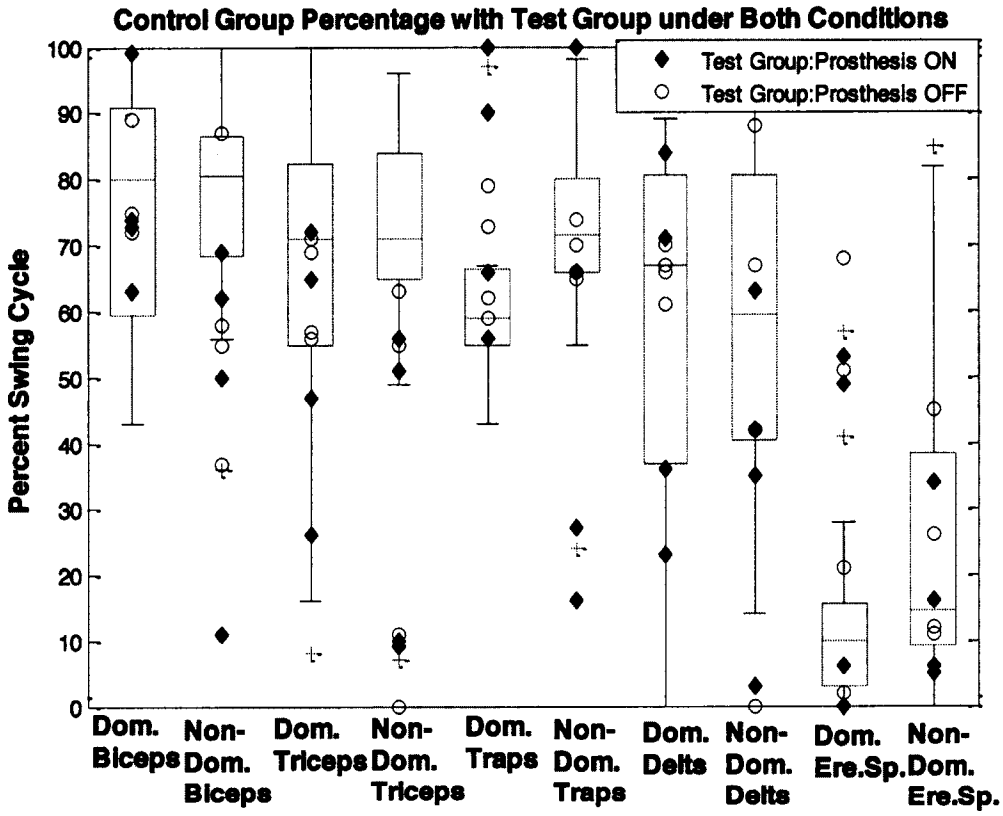


Figure 78: Boxplot of each muscle's total activity per percentage swing cycle in the control group, test group wearing their prosthesis, and test group not wearing their prosthesis

6.0 Conclusions and Recommendations

6.1 Conclusions

The data presented here show strong evidence that during activities of daily living, individuals with upper limb loss have muscle activity that lie outside the normal limits both while wearing a prosthesis and not wearing one. This data, however, does not clarify the role of the prosthesis (whether it was on or off) on the muscle activity of the prosthesis user.

The control group exhibited consistent muscle activity during biking (dominant and non-dominant trapezius), walking (dominant and non-dominant erector spinae), and swinging (dominant and non-dominant biceps, triceps, trapezius, and deltoids). The consistent muscle activity for the control group was presented as a range in selection criterion. The swing and walking tasks had excellent muscle consistency throughout those tasks when the middle selection criterion was used (14/16 subjects). The bike task, however, only had consistent muscle activity under the lowest limit of the selection criteria (11/15 for the bike task). The swing task exhibited the highest average muscle co-contraction (for all muscles), and muscle activation (for all muscles except the dominant erector spinae), out of the three tasks for the control group. The highest average muscle symmetry was during walking, and the lowest was during biking.

Refer to Tables 26 to 29 for a summary of significant deviations from the control group for each case study. The case studies revealed each prosthesis user was different from the

control group in various ways. This supports the hypothesis that prosthesis users would have muscle activity outside normal limits, but does not support the hypothesis that while wearing a prosthesis, muscle activity is closer to normal.

Table 26: Summary Subject ICPML14 significant deviations from control

Task	Prosthesis	Z score	Muscle	Variable
Swing	ON	-2.4	Triceps	% Symmetry
Swing	OFF	-2.8	Triceps	% Symmetry
Swing	ON	2.9	D. Eresp	% Activation
Swing	OFF	3.9	D. Eresp	% Activation
Swing	OFF	-2.8	N.D Triceps	% Activation

Table 27: Summary Subject PAPFL10 significant deviations from control

Task	Prosthesis	Z Score	Muscle	Variable
Bike	ON	-3.5	Eresp	% Symmetry
Swing	ON	-3.4	Biceps	% Symmetry
Swing	ON	-2.6	Triceps	% Symmetry
Swing	ON	-5.5	Trapezius	% Symmetry
Swing	ON	-3.6	Deltoids	% Symmetry
Swing	OFF	-2.7	Triceps	% Symmetry
Swing	OFF	-2.7	Deltoids	% Symmetry
Swing	ON	-2.9	N.D Triceps	% Activation
Swing	ON	-3.3	N.D. Traps	% Activation
Swing	ON	2.8	D. Traps	% Activation

Table 28: Summary Subject RCPFR06 significant deviations from control

Task	Prosthesis	Z Score	Muscle	Variable
Swing	ON	-5.9	Biceps	% Symmetry
Swing	ON	-3.0	Trapezius	% Symmetry
Swing	OFF	-3.8	Deltoids	% Symmetry
Swing	ON	-3.9	N.D Biceps	% Activation
Swing	ON	-2.9	N.D Triceps	% Activation
Swing	ON	-2.6	N.D Traps	% Activation
Swing	OFF	-3.2	N.D Triceps	% Activation
Swing	OFF	-2.4	N.D Delts	% Activation

Table 29: Summary Subject WRPFR13 significant deviations from control

Task	Prosthesis	Z score	Muscle	Variable
Bike	OFF	-2.7	Biceps	% Symmetry
Swing	ON	-3.4	Biceps	% Symmetry
Swing	OFF	-2.8	D. Triceps	% Symmetry
Swing	ON	2.6	D. Eresp	% Activation
Swing	OFF	-2.4	N.D Biceps	% Activation
Swing	OFF	2.7	D. Eresp	% Activation

Refer to Table 30 for a summary of significant deviations from the control group that each prosthesis user group had. The group statistics revealed that the prosthesis users, under both conditions, had significantly lower muscle symmetry compared with the control group. This also supported the hypothesis that prosthesis users had muscle activity outside normal limits. These findings are a great indication that further work should be completed in order to find out whether or not these children are at risk for any overuse injuries because of this difference in muscle symmetry compared with normally limbed children.

Table 30: Summary of significant deviations from the control group by the prosthesis user group

Task	Prosthesis	P Value	Muscle	Variable
Bike	ON	0.007	Deltoids	↑ % Symmetry
Bike	OFF	0.05	Deltoids	↑ % Symmetry
Bike	ON	0.04	D. Triceps	↑ % Activation
Bike	ON	0.02	N.D. Trapezius	↓ % Activation
Bike	OFF	0.04	N.D. Trapezius	↓ % Activation
Bike	ON	0.03	D. Deltoids	↓ % Activation
Bike	ON	0.03	N.D. Deltoids	↓ % Activation
Bike	OFF	0.05	D. Erector Spinae	↓ % Activation
Swing	ON	0.009	Biceps	↓ % Symmetry
Swing	ON	0.02	Triceps	↓ % Symmetry
Swing	ON	0.0004	Deltoids	↓ % Symmetry
Swing	OFF	0.01	Biceps	↓ % Symmetry
Swing	OFF	0.009	Triceps	↓ % Symmetry
Swing	OFF	0.02	Deltoids	↓ % Symmetry
Swing	ON	0.02	N.D Biceps	↓ % Activation
Swing	ON	0.02	N.D Triceps	↓ % Activation
Swing	OFF	0.02	N.D Triceps	↓ % Activation

6.2 Recommendations

Review methods that were used to construct muscle activation patterns in the control group as the ones used may have been too restrictive and left out important information. Another recommendation for the data analysis would be to perform the same research while adding in a maximum voluntary muscle contraction (MVC) in order to analyze the muscle signal amplitude. This may slightly change some of the activation, symmetry and co-contraction results, as a constraint could be made to only include muscle activity above a certain MVC %.

A continuation of this study could be to begin performing research to determine whether the individuals attending the upper limb clinic at the institute of biomedical engineering

are at a greater risk of overuse injuries compared with normal. A clinical test for RSI's/overuse injuries or increased risks of them could be integrated into yearly visits with patients. In that way, a longitudinal study could be done at a level that is not time or cost intensive.

6.3 Limitations of Study

The first substantial limitation of this study is that there were only four prosthesis users amongst the test group. This is a product of the small global population of young individuals with upper limb loss. However, a more vigorous recruitment strategy from the very beginning of this study may have slightly improved these numbers. The second limitation of the study is the lack of current literature available to make comparisons/validate any results.

Another limitation present was the method in which the data was analysed for the muscle patterns. It would have been less limiting if muscle patterns (within the control group) were not only presented as a range of the number of individuals necessary for a "pattern" to be counted (the selection criterion range), but to also present a small range in the actual timing of the muscle onset's themselves. For example, if a subject had activation +/- 1% (of the task cycle) from another subject, they could be considered occurring at the same time. This would be difficult, but if a good method were used it would be helpful because EMG signals (especially off of children), are noisy and a little unpredictable. Therefore, a range in this activation timing may have revealed some more meaningful results.

7.0 References

Anders, C., et al. Trunk muscle activation patterns during walking at different speeds. Journal of Electromyography and Kinesiology. 2007; 17: 245-252.

Alley, R., and Sears, H. 2004. Powered Upper Limb Prosthetics in Adults. In A. Muzumdar, Powered Upper Limb Prostheses. 2004 ed., 117-145. Berlin Heidelberg, Germany: Springer-Verlag.

Ballance,R. et al.. Factors affecting myoelectric prosthetic use and wearing patterns in the juvenile unilateral below-elbow amputee. Occupational Therapy. 1989;56:132 – 137.

Biddiss, E., et al. Consumer design priorities for upper limb prosthetics. Disability and rehabilitation: Assistive Technology. 2007; 2:346-357.

Biddiss, E., and Chau, T. Upper limb prosthesis use and abandonment: A survey of the last 25 years. Prosthetics and orthotics international. 2007; 31:236-257.

Biddiss & Chau (b). The roles of predisposing characteristic, established need, and enabling resources on upper extremity prosthesis use and abandonment. Disability and Rehabilitation: Assistive Technology. 2007; 2:71-84.

Black, Nancy et al. "Using potential energy to measure work related activities for persons wearing upper limb prostheses." *Robotica*, 2005: 319-327.

C.J. De Luca. The use of surface electromyography in biomechanics, Journal of Applied Biomechanics. 1997; 13: 135-16.

Callaghan J, Patla A, McGill S. Low back three dimensional joint forces, kinematics, and kinetics during walking. Clin Biomech 1999;14:203–16.

Chen, R. Focal Dystonia and Repetitive Motion Disorders. Clinical Orthopaedics and related research. 1998; 35: 102-106.

Chester, V., Biden, E., and Tingley, M. Gait Analysis. Biomedical Instrumentation and Technology. 2005;64 - 74.

Crandall, R., and Tomhave, W. Pediatric unilateral below-elbow amputees: retrospective analysis of 34 patients given multiple prosthetic options. Journal of Pediatric Orthotics. 2002;22:380-3.

Datta, D. Functional outcome of patient with proximal upper limb deficiency – acquired and congenital. Clinical Rehabilitation. 2004;18:172-177.

Davids, J., et al. Prosthetic management of children with unilateral congenital below-elbow deficiency. Journal of Bone and Joint Surgery. 2006; 88:1284-1300.

Davidson, J. A comparison of upper limb amputees and patients with upper limb injuries using hte Disability of the Arm, Shoulder and Hand (DASH). Disability and rehabilitation. 2004; 26:917-923.

Day, H. The ISO/ISPO Classification of Congenital Limb Deficiency. Atlas of Amputations and Limb Deficiencies. 2004; Chapter 33.

Deci, E. L., & Ryan, R. M. A motivational approach to self: Integration in personality: Perspectives on motivation. Nebraska Symposium on Motivation, University of Nebraska Press. 1991; 38; 237-288

Desrosiers, J. Et al. Performance of the "unaffected" upper extremity of elderly stroke patients. Stroke. 1996; 27:1564-1570

Dillingham, T., et al. Limb amputation and limb deficiency: epidemiology and recent trends in the united states. Southern medical journal. 2003; 95:875-883.

Elphraim, P., et al. Epidemiology of limb loss and congenital limb deficiency: A review of literature. Archives of physical medicine and rehabilitation. 2003; 84:747-761.

Ephraim, P., Wegener S., MacKenziem E., et al: Phantom pain, residual limb pain, and back pain in amputees: Results of a national survey. Archives of physical medical rehabilitation 2005;86:1910-9

Flood, K., et al. Limb deficiency and prosthetic management. 2. Aging with limb loss. Archives of Physical Medicine and Rehabilitation. 2006; 87: S10-S14.

Fry,H. Overuse syndrome of the upper limb in musicians. Med J Aust 1986;144:182–185.

Freivalds, A. Biomechanics of the upper limbs, Mechanics, Modeling, and Musculoskeletal Injuries. 2004; USA(Florida): CRC Press.

Genn, C., et al. Congenital upper limb patient history at the institute of biomedical engineering. Proceedings of ACPOC Conference. 2009; 17-18.

Glynn MK, Galway HR, Hunter G, Sauter WF. Management of the upper-limb-deficient child with a powered prosthetic device. Clin Orthop. 1986; 209:202 – 205

Hagg, G.M. Static workload and occupational myalgia— a newexplanation model. In: Anderson, P., Hobart, D., Danoff, J. (Eds.), Electromyographical Kinesiology. Elsevier Science Publishers, Amsterdam. 1991; 141–144.

Hagg, G.M., Astrom, A. Load pattern and pressure pain threshold in the upper trapezius muscle and psychosocial factors in medical secretaries with and without shoulder/neck

disorders. International Archives of Occupational and Environmental Health. 1997; 69: 423–432.

Hanley, M., et al. Chronic Pain Associated with Upper-Limb Loss. American Journal of Physical Medical Rehabilitation. 2009; 88-9; 742:751.

Heintz, S., and Gutierrez-Farewik, E. Static optimization of muscle forces during gait in comparison to EMG-to-force processing approach. Gait and Posture. 2007; 26: 279-288.

Herberts P, et al. Rehabilitation of unilateral below-elbow amputees with myoelectric prostheses. Scandanavian Journal of Rehabilitative Medicine. 1980;12:123 – 128.

Hermens, H. and Hutten, M. International Journal of Industrial Ergonomics. 2002; 30: 325-336.

Hodges, P. and Bui, B. A comparison of computer-based methods for the determination of onset of muscle contraction using electromyography. Electroencephalography and Clinical Neurophysiology. 1996; 101: 511-519.

Jones, L. and Davidson, H. Save that arm: a study of problems in the remaining arm of unilateral upper limb amputees. Prosthetics and Orthotics International. 1999; 23:55–58.

Kooijman, C., Djikstra, P., Geertzen, J., et al:Phantom pain and phantom sensations in upper limb amputees:An epidemiological study. Pain 2000;87:33-41

Konrad, P. The ABC of EMG, A Practical Introduction to Kinesiological Electromyography. 2005.

Krivickas L. Anatomical factors associated with overuse sports injuries. Sports Medicine. 1997; 24:132–146.

Kruger, L. and Fishman, S. Myoelectric and body-powered prostheses. J Journal of Pediatric Orthotics. 1993;13:68-75.

Lento, P. Congenital Limb Deficiency. Located: http://www.disaboom.com/Health/amputationspediatric/Printable/pediatric-amputee-congenital-limb-deficiency.aspx#. Date accessed: June 2008.

McCarroll J. Overuse injuries of the upper extremity in golf. Clinical Sports Medicine. 2001; 20:469–479.

MacPhee, B. Examination of prosthetic function and body behaviour of prosthetic users performing activities of daily living with two different wrist systems. Master's Thesis., University of New Brunswick. 2007.

Merskey, H. Classification of chronic pain: Descriptions of chronic pain syndromes and definitions of pain terms. Pain. 1986; 3: 226.

Meurs, M., et al. Prescription of the first prosthesis and later use in children with congenital unilateral upper limb deficiency: A systematic review. Prosthetics and orthotics international. 2006; 30:165-173.

Millstein, et al. A review of employment patterns of industrial amputees: Factors influencing rehabilitation. Prosthetics and Orthotics International 1985;9:69-78

Millstein et al. Prosthetic use in adult upper limb amputees: a comparison of the body powered and electrically powered prostheses. Prosthetics and Orthotics International.1986;10:27 – 34.

Murray M, Mollinger L, Gardner G, Sepic S. Kinematic and EMG patterns during slow, free, and fast walking. J Orthop Res 1984;2:272–80.

Murray M, Spurr G, Sepic S, Gardner G, Mollinger L. Treadmill vs floor walking: kinematics, electromyogram, and heart rate. J Appl Physiol 1985;59:87–91.

Myotronics, 2010. Myotronics Products, Duotrodes. http://www.myotronics.com/
Nelson, V., et al. Limb deficiency and prosthetic management. 1. Decision making in prosthetic prescription and management. Archives of physical medicine and rehabilitation. 2006; 87:S3-S9.

Noraxon, 2010. ZeroWire, True wireless EMG.
http://www.noraxon.com/aurion/zerowire.php. Date accessed: June 2009.

Parks determining the persistence of phantom limb pain in the amputee. J Psychosom Res 1973;17:97-108.

Postema, K., et al. Prosthesis rejection in children with a unilateral congenital arm defect. Clinical rehabilitation. 1999; 13:243-249.

Pruitt, S. Et al. Functional status in limb deficiency: development of an outcome measure for preschool children. Archives of Physical Medical Rehabilitation. 1998;79:405-11.

Pylatiuk, C., et al. Results on an internet survey of myoelectric prosthetic hand users. Prosthetics and orthotics international. 2007; 31:362-370.

Rau, G., et al. From cell to movement: to what answers does EMG really contribute?, Journal of Electromyography and Kinesiology. 2004; 4: 611-617.

Rau, G., Disselhorst-Klug, C., and Schmidt, R. Movement biomechanics goes upwards: from the leg to the arm. Journal of Biomechanics.2000; 33: 1207-1216.

Ross, M. Development of a Quantitative Test for Prosthetic Function Using Activities of Daily Living and Motion Analysis. Master's Thesis.University of New Brunswick. 2005.

Sandsj, L., Melin, B., Rissen, D., Dohns, I., Lundberg, U. Trapezius muscle activity, neck and shoulder pain, and subjective experiences during monotonous work in women. European Journal of Applied Physiology.2000; 83, 235–238.

Schulte, E, et al. Comparison of the electromyographic activity in the upper trapezius and biceps brachii muscle in subjects with muscular disorders: a pilot study. European Journal of Applied Physiology. 2006; 96: 185-193.

Schmidt-Rohlfing, B. et al. Interpretation of surface EMGs in children with cerebral palsy: an initial study using a fuzzy expert system. Journal of orthopaedic research. 2006; 24: 438-447.

Scott, R. An Introduction to Myoelectric Prostheses. In A. Muzumdar, U.N.B. Monographs on Myoelectric Prostheses. University of New Brunswick, Canada: Bio-engineering Institute. 1984; 1-19.

Seniam. Recommendations on electrode placement. http://www.seniam.org/. Date accessed: June 2010.

Seymour, R. Prosthetics and Orthotics, Lower limb and spinal. USA: Lippincott Williams & Wilkins. 2002.

Shafer-Crane G. Repetitive stress and strain injuries: preventative exercises for the musician. Phys Med Rehabil Clin N Am 2006;17:827– 842.

Shiavi R. Electromyographic patterns in normal adult locomotion. In: Smidt GL, editor. Gait in rehabilitation. New York: Churchill Livingstone. 1990; 97–119.

Smutok M., et al, Effects of unilateral brain damage on contralateral side and ipsilateral upper extremity function in hemiplegia. Physical Therapy. 1989;69:195-203

Soderberg, G., and Cook, T. Electromyography in Biomechanics. Journal of Physical therapy. 1984; 64: 1813-1820.

Steven, G. et al. Abdominal and erector spinae muscle activity during gait: the use of cluster analysis to identify patterns of activity. Clinical Biomechanics. 2002; 17: 177-184.

Sutherland D. The events of Gait" Bulletin of prosthetics research. 1981; 10-35: 281-82

Tulder, M., et al. Repetitive Strain Injury. The Lancet. 2007;369: 1815-1822.

Turk, D.C.; Okifuji, A. Pain terms and taxonomies. in Loeser, D.; Butler, S. H.; Chapman, J.J. et al.. Bonica's management of pain (3 ed.). Lippincott Williams & Wilkins.2001; 18–25

Tyc VL. Psychosocial adaptation of children and adolescents with limb deficiencies: A review. Clin Psychol Rev 1992;12:275 – 291.

Veiersted, K.B., Westgaard, R.H., Andersen, P. Electromyographic evaluation of muscular work pattern as a predictor of trapezius myalgia. Scandinavian Journal of Work Environment and Health 19.1993;284–290.

Waters RL, Morris JM. Electrical activity of muscles of the trunk during walking. J Anat 1972;111:191–9.

White, B. Development of a Quantitative Test for Prosthetic Function Using Activities of Daily Living and Motion Analysis. Master's Thesis. University of New Brunswick.2006.

Whyte, A. & Carroll, L. A preliminary examination of the relationship between employment, pain and disability in an amputee population. Disability and Rehabilitation. 2002; 24:462-470.

Winter DA, Yack HJ. EMG profiles during normal human walking: stride-to-stride and inter-subject variability. Electroenceph Clin Neurophysiol. 1987;67:402–11.

Williams, S., et al. An upper body model for the kinematical analysis of the joint chain of the human arm. Journal of Biomechanics.2006;39: 2419-2429.

Wright, F. Et al. The Prosthetic Upper Extremity Functional Index: development and reliability testing of a new functional status questionnaire for children who use upper extremity prostheses. Journal of Hand Therapy. 2001;14:91-104.

Yoshihiro, S., Masahide, K., Toshiomi T., Kotaro O. Carpal Tunnel Syndrome Involving Unaffected Limbs of Stroke Patients. 1999.

Zinck, A. Investigation of compensatory movements in prosthesis users and the design of a novel wrist. Master's Thesis. University of New Brunswick. 2008.

Appendix A

Upper Extremity Movement in Children with Unilateral Below Elbow Prostheses during Gross Motor Tasks

Location: Institute of Biomedical Engineering
University of New Brunswick
Fredericton, NB

Principal Investigators: Katelynn Craig and Carly Genn, UNB Graduate Students
Institute of Biomedical Engineering, UNB
Tel: (506) 453-4966

Supervisor: Dr. Edmund Biden
School of Graduate Studies / Department of Mechanical Engineering
University of New Brunswick
(506) 458-7762

Invitation: You are invited to participate in a research project entitled "Upper Extremity Movement in Children with Unilateral Below Elbow Prostheses during Gross Motor Tasks". The objectives, purposes, background, experimental methods, as well as the risks will be described below.

Objective: To compare how a prosthesis-using child performs gross motor tasks with and without his or her prosthesis, and subsequently compare these findings with the normative data collected on able-limbed children performing the same tasks.

Purpose and Potential Benefits: This project will incorporate position, force and inertial data to develop joint load models, which will lead to a better understanding and assessment of prosthetic arms in children, as well as evaluate the long term risk of repetitive strain injury due to prosthesis use. Compensatory motions will also be examined using a gait task to study the possibility of postural problems occurring from not wearing a prosthesis.

Experimental Method: M-CAM, a motion analysis system at the Institute of Biomedical Engineering, will be the method of data capturing. Before the data capturing, you will be asked to fill in a questionnaire including your age and your dominant hand. You may decline to answer any of the questions if you choose. Measurements, including diameter and length, of your upper limbs will be taken. Approximately twenty small marker balls, coated with reflective tape, will be attached to bony landmarks on your upper torso and arms using two-sided tape. Eight EMG surface electrodes will also be positioned over specific muscles. The markers are tracked by the cameras as you move and the EMG electrodes measure muscle activity. You will be asked to perform various

gross motor tasks, including swinging on a swing, bicycling and walking. Before you perform the activities, you will have the opportunity to practice. You will be required to wear a tank shirt, to ensure maximum visibility of markers placed on the shoulders. Tank shirts will be provided; however, you are welcome to bring your own as long as the front and back of your neck and shoulders are visible. In addition to capturing the movements of the markers with the motion analysis cameras, you will be video taped for reference during the capture of the motion data. You may decline to be video taped for the experiment. The entire test will take about 90 minutes.

Data Security: All data, including video tapes, are stored securely and accessible only by researchers. Filenames are coded so people's identities cannot be determined. The data will be stored in a data bank accessible only by researchers working on this study and will be destroyed when it no longer has clinical or scientific value. When the results of this research are reported, participating individuals will not be identifiable.

Risks: The risks of associated with the experiment of this kind are considered to be minimal. Occasionally, a skin irritation will develop because of the two-sided tape. Contact the researcher if the problem persists.

Questions or Concerns: When the results of this experiment are reported, the results will be available to you. Please indicate below if you would like to see the results. If you have any questions about the project you are welcome to contact any of the indicated researchers. Bernie Hudgins, Director of The Institute of Biomedical Engineering, is also available for questions or concerns about the research. He may be contacted by phone at (506) 458-7094.

Consent: I have read and understood the procedure and risks associated in volunteering for this experiment. I have had any questions or concerns addressed or explained by the researcher. I understand that my involvement in this experiment is completely voluntary and that I can withdraw myself or my data from this experiment at any time. I also understand that my identity will be protected in the reporting of this research.

_____	_____	_____
Subject Name (print)	Signature	Date
_____	_____	_____
Witness Name (print)	Signature	Date

I would like a summary of the results after they have been reported: Yes:____ No:____
If yes, please indicate how you may be contacted:_____

Appendix B

Questionnaire (Control Group)

Code: _____

Today's Date: _____

Date of Birth: _____

Gender (M/F): _____

Dominant side (R/L):_____

Standing Height:_____ cm

Body Weight: _____ lbs

Distance from acromium along humerus axis to CoR: _____(mm)

Distance from lateral to medial epicondyle: _____(mm)

Hand Thickness (R)_____(mm)

Hand Thickness (L)_____(mm)

Wrist Width (R)_____(mm)

Wrist Width (L)_____(mm)

Knee Width (R)_____(mm)

Knee Width (L)_____(mm)

Ankle Width (R)_____(mm)

Ankle Width (L)_____(mm)

Left Hand Grip Force _____ (kg)

Right Hand Grip Force _____ (kg)

Questionnaire (Test Group)

Code: _____

Today's Date: _____

Date of Birth: _____

Gender (M/F): _____

Prostheses side (R/L):_____

Date of First Fitting: _____

Standing Height: _____ inches
Body Weight (wearing prosthesis): _____ lbs
Body Weight (not wearing prosthesis): _____lbs

Distance from acromium along humerus axis to CoR: _____(mm)

Distance from lateral to medial epicondyle: _____(mm)

Dominant Hand Grip Force _____ (kg)

Prostheses Hand Grip Force _____ (kg)

Distance from wrist crease to CofR: _____(mm)

Hand Thickness (R)_____(mm)

Hand Thickness (L)_____(mm)

Wrist Width (R)_____(mm)

Wrist Width (L)_____(mm)

Knee Width (R)_____(mm)

Knee Width (L)_____(mm)

Ankle Width (R)_____(mm)

Ankle Width (L)_____(mm)

Appendix CI

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%
%                                      getc3d.m
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%
%
% This file converts c3d files to matlab arrays (coord, analog).
% The forefront of the program belongs to Alan Morris from the Hugh
MacMillan Centre in Toronto
% The force plate code added on was written by V. Chester
% This code was validated against the csv file which automatically
produces the force plate variables which must
% be manually calculated when using the c3d file.
%
% Created March 2001
% Last Modification:   February 06th, 2002, V. Chester
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%
%
function[coord,analog,nframes,sframe,eframe,frame_rate,nmarkers,adcperf
rame,nchannel]=getc3d(fname)
% GETC3D:         Getting 3D coordinate/analog data from .C3D binary
file (V370 version)
%
% Input:         fname - file name to be read (or path)
%
% Output:         coord - coordinate data [m] (nframes x (nmarkers*4)
matrix)
%                 analog - analog data 16 bit [volts] (nfames x
(nchannel*ADCframes/VICON) matrix)
%                 nframes - total number of VICON frames
%                 sframe - starting frame of data
%                 eframe - ending frame of data
%                 frame_rate - frame rate of VICON camera data
%                 nmarkers - number of coordinate markers
%                 adcperframe - number of analog frames per VICON frame
%                 nchannel - number of analog channels sampled
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%

clear all
close all

%function[coord,analogdata,EMG_data]=getc3d4pnew(fname,addrs,trial)
[fn,pn]= uigetfile('*.c3d','Specify VICON data file');  %%added this
for easier processing
fname= [pn fn];
%fname = ['C:\Users\biden\Desktop\Working Data\Subject C\bike4.c3d'];
```

```
% Last line of code saves analog and coord data based on the log name
% Log command is based on a filename of 10 letters (eg. jenpilot05.c3d)
excluding the 3 letter extension
log=fname(length(fname)-9:length(fname)-4);

% Create list of markers and analog channels
[markerlist,analoglist]=c3dlabel4pworking;

% Open the file and skip the first 2 bytes
%  Reading file in VAX D format
fid=fopen(fname,'r+','d');

%  Reading record number of parameter section
key1=fread(fid,1,'int8');
key2=fread(fid,1,'int8');

fseek(fid,2,'bof');
scale=0;
%  Getting all the necessary parameters from the header record
nmarkers=fread(fid,1,'int16');
%number of markers
nanalog=fread(fid,1,'int16');
%number of analog channels x #analog frames per video frame
sframe=fread(fid,1,'int16');
%# of first video frame
eframe=fread(fid,1,'int16');
%# of last video frame
nframes=eframe - sframe + 1;
%number of frames
intgap=fread(fid,1,'int16');
%maximum interpolation gap allowed (in frame)
scale=fread(fid,1,'float32');
%floating-point scale factor to convert 3D-integers to ref system units
nstart=fread(fid,1,'int16');
%starting record number for 3D point and analog data
adcperframe=fread(fid,1,'int16');
%number of analog channels per video frame
fseek(fid,0,'cof');
frame_rate=fread(fid,1,'float32');
%frequency of video data
nchannel=nanalog/adcperframe;
%number of analog channels

% Group Data/Parameter Records
fseek(fid,512,'bof');

% 1st group data only
dat1=fread(fid,1,'int8');
dat2=fread(fid,1,'int8');
records=fread(fid,1,'int8');
proctype=fread(fid,1,'int8');
```

```
% The end of the parameter record is indicated by 0 characters for
group/parameter name
characters=fread(fid,1,'int8');
%characters in group/parameter name
idnumber=fread(fid,1,'int8');
%id number -ve=group / +ve=parameter

while characters~=0

   if idnumber<0 %Group data
      name=fread(fid,[1,characters],'char');
      group(abs(idnumber)).name=cellstr(char(name));              %group
name
      offset=fread(fid,1,'int16');
%offset in bytes
      deschars=fread(fid,1,'int8');
%description characters
      description=fread(fid,[1,deschars],'char');
      group(abs(idnumber)).description=cellstr(char(description));
%group description

      index=0;
      fseek(fid,offset-3-deschars,'cof');


   else %parameter data
      clear dimension;
      index=index+1;
%index all parameters within a group
      name=fread(fid,[1,characters],'char');
%name of parameter

      % read parameter name
      if size(name)>0
         parameter(idnumber,index,1).name=cellstr(char(name));
%save parameter name
      end

      % read offset and type
      offset=fread(fid,1,'int16');
%offset of parameters in bytes
      filepos=ftell(fid);
%present file position
      nextrec=filepos+offset(1)-2;
%position of beginning of next record
      type=fread(fid,1,'int8');

      if type==-1
         datatype='char';
%characters
      elseif type==1
         datatype='int8';
%1-byte for boolean
```

```
      elseif type==2
         datatype='int16';
%integer
      else %type==4
         datatype='float';
%floating-point number
      end

      parameter(idnumber,index,1).datatype=type;        %type of data:
-1=char/1=byte/2=integer*2/4=real*4

      % read number of dimensions
      dimnum=fread(fid,1,'int8');
      parameter(idnumber,index,1).numdim=dimnum;                %number
of dimensions in parameter

      for j=1:dimnum
         dimension(j)=fread(fid,1,'int8');
         parameter(idnumber,index,j).dim=dimension(j);  %save parameter
dimension data
      end

      if dimnum==0 %number of dimensions of the parameter
         datalength=abs(type);
%length of data record
      else
         mult=1;
         for j=1:dimnum
            mult=mult*parameter(idnumber,index,j).dim;
         end

         datalength=abs(type)*mult;
%length of data record for multi-dimensional array
      end


      if type==-1 %datatype=='char'  if data is a character string

         wordlength=parameter(idnumber,index,1).dim;        %length of
character word

         if dimnum==2 & datalength>0 %&
parameter(idnumber,index,2).dim>0
            for j=1:parameter(idnumber,index,2).dim
               data=fread(fid,[1,wordlength],datatype);
%character word data record for 2-D array
               parameter(idnumber,index,j).data=cellstr(char(data));
            end

         elseif dimnum==1 & datalength>0

            data=fread(fid,[1,wordlength],datatype);
%numerical data record of 1-D array
```

```
            parameter(idnumber,index,1).data=cellstr(char(data));

         end

      elseif type==2 & datalength>0
%integer

         datum=datalength/abs(type);
%number of data of numerical type

         for j=1:datum
            parameter(idnumber,index,j).data=fread(fid,1,datatype);
%parameter data
         end

      elseif type==4 & datalength>0

         datum=datalength/abs(type);

         for j=1:datum
            parameter(idnumber,index,j).data=fread(fid,1,datatype);
%parameter data
         end

      else
         datum=0;
      end

      deschars=fread(fid,1,'int8');
%description characters

      if deschars>0
            description=fread(fid,[1,deschars],'char');
            parameter(idnumber,index,1).description=description;
      end

      %moving ahead to next record
      fseek(fid,nextrec,'bof');

   end

   % check group/parameter characters and idnumber to see if more
records present
   characters=fread(fid,1,'int8');
        idnumber=fread(fid,1,'int8');

end

%-----------------------------------------------------------------------

%  Get the scaled coordinate data [mm] and analog data
%  using 3-dimensional coordinate arrays
```

```
fseek(fid,(nstart-1)*512,'bof');
hwaitbar = waitbar(0,'Getting scaled coordinate data');
for i=1:nframes
        waitbar(i/nframes);
   for j=1:nmarkers
      for k=1:4
         if scale < 0
            tempcoord(i,j,k)=fread(fid,1,'float32');
         else
            tempcoord(i,j,k)=fread(fid,1,'int16')*scale;
         end
      end;
   end;

   for l=1:adcperframe
      for m=1:nchannel
         tempanalog(i,m,l)=fread(fid,1,'int16')-32768;
      end;
   end;

end;
close(hwaitbar)

fclose(fid);



[groups,parameters,data]=size(parameter);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%
% Find the labels for the markers
for i=1:groups
   name=char(group(i).name);

   if strcmp(name,'POINT')
      labelgrp=i;
      break
   end
end

for j=1:parameters
   name=char(parameter(labelgrp,j).name);

   if strcmp(name,'LABELS')

      labelpar=j;
      break
   end
end
```

```
% Finding order of markers to arrange in sequence
clear landmark k l
listlen(1) = size(markerlist,2);

for k=1:parameter(labelgrp,labelpar,2).dim
   for l=1:listlen(1)
      if strcmp(parameter(labelgrp,labelpar,k).data,markerlist(l))
         landmark(l)=k;
      end
   end
end
%
coord(1:nframes,1,1:3)=0;         %Zero first marker data

% Re-order marker data into sequence
hwaitbar = waitbar(0,'Re-ordering marker data');
for i=1:nframes
   waitbar(i/nframes);
        for j=2:listlen(1)
           for k=1:3
                coord(i,j,k)=tempcoord(i,landmark(j),k);
      end
   end
end
close(hwaitbar);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%
%Analog data 12-bit scaled to +/-10 Volts, not scaled to amplifier
ranges
% Kistler scaling is done in forceplate calculation program

% Find the labels for the markers
for i=1:groups
   name=char(group(i).name);

   if strcmp(name,'ANALOG')
      labelgrp=i;
      break
   end
end

for j=1:parameters
   name=char(parameter(labelgrp,j).name);

   if strcmp(name,'LABELS')

      labelpar=j;
      break
   end
end

% Finding order of markers to arrange in sequence
```

```
listlen(2) = size(analoglist,2);

for k=1:parameter(labelgrp,labelpar,2).dim
   for l=1:listlen(2)
      if strcmp(parameter(labelgrp,labelpar,k).data,analoglist(l))
         channel(l)=k;
      end
   end
end

if (nchannel~=0)  % When there is analog data...

        % Re-order analog data into sequence
        %hwaitbar = waitbar(0,'Re-ordering analog data');
        for i=1:nframes,
           waitbar(i/nframes);
           for j=1:nchannel,
              for k=1:adcperframe,
                 analog(i,j,k)=tempanalog(i,channel(j),k);
              end
           end
        end
        %close(hwaitbar);


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
%                                                Vicky's Addition to
Code
%                                Last Modified February 6th, 2002 VC
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

% This section of the code performs A/D conversion, scaling and offset
operations (eg. volts to Newtons)
% This section of the code performs the appropriate calculations for
Kistler plate which does not output force and moments directly
% This section of the code performs moving average smoothing of analog
data
% This section of the code removes the DC component from the analog
data
% This section of the code saves all analog data to a matrix called
"analogdata"

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
%                                    Force Plate Section
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
% sums the channels as needed for Kistler
% Calculates rGRF, COP etc
% Outputs a file of parameters called 'analogdata'
%
% channels 1-8 = new Kistler 9281CA
```

```
% channels 9-14 = AMTI
% channels 15-22 = old Kistler 9281 B11
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

%%% APPLY OFFSETS TO ANALOG DATA BEFORE SCALING %%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%Plate 1 is the old plate 9281 B11

analog(:,1,:) = analog(:,1,:) - (parameter(4,7,1).data - 32768);
%previously was parameter(4,6&8,1).data
analog(:,2,:) = analog(:,2,:) - (parameter(4,7,2).data - 32768);
analog(:,3,:) = analog(:,3,:) - (parameter(4,7,3).data - 32768);
analog(:,4,:) = analog(:,4,:) - (parameter(4,7,4).data - 32768);
analog(:,5,:) = analog(:,5,:) - (parameter(4,7,5).data - 32768);
analog(:,6,:) = analog(:,6,:) - (parameter(4,7,6).data - 32768);
analog(:,7,:) = analog(:,7,:) - (parameter(4,7,7).data - 32768);
analog(:,8,:) = analog(:,8,:) - (parameter(4,7,8).data - 32768);

analog(:,9,:)  = analog(:,9,:)  - (parameter(4,7,9).data - 32768);
analog(:,10,:) = analog(:,10,:) - (parameter(4,7,10).data - 32768);
analog(:,11,:) = analog(:,11,:) - (parameter(4,7,11).data - 32768);
analog(:,12,:) = analog(:,12,:) - (parameter(4,7,12).data - 32768);
analog(:,13,:) = analog(:,13,:) - (parameter(4,7,13).data - 32768);
analog(:,14,:) = analog(:,14,:) - (parameter(4,7,14).data - 32768);
analog(:,15,:) = analog(:,15,:) - (parameter(4,7,15).data - 32768);
analog(:,16,:) = analog(:,16,:) - (parameter(4,7,16).data - 32768);

analog(:,17,:) = analog(:,17,:) - (parameter(4,7,17).data - 32768);
analog(:,18,:) = analog(:,18,:) - (parameter(4,7,18).data - 32768);
analog(:,19,:) = analog(:,19,:) - (parameter(4,7,19).data - 32768);
analog(:,20,:) = analog(:,20,:) - (parameter(4,7,20).data - 32768);
analog(:,21,:) = analog(:,21,:) - (parameter(4,7,21).data - 32768);
analog(:,22,:) = analog(:,22,:) - (parameter(4,7,22).data - 32768);
analog(:,23,:) = analog(:,23,:) - (parameter(4,7,23).data - 32768);
analog(:,24,:) = analog(:,24,:) - (parameter(4,7,24).data - 32768);
analog(:,25,:) = analog(:,25,:) - (parameter(4,7,25).data - 32768);
analog(:,26,:) = analog(:,26,:) - (parameter(4,7,26).data - 32768);
analog(:,27,:) = analog(:,27,:) - (parameter(4,7,27).data - 32768);
analog(:,28,:) = analog(:,28,:) - (parameter(4,7,28).data - 32768);
analog(:,29,:) = analog(:,29,:) - (parameter(4,7,29).data - 32768);
analog(:,30,:) = analog(:,30,:) - (parameter(4,7,30).data - 32768);
analog(:,31,:) = analog(:,31,:) - (parameter(4,7,31).data - 32768);
analog(:,32,:) = analog(:,32,:) - (parameter(4,7,32).data - 32768);

analog(:,33,:) = analog(:,33,:) - (parameter(4,7,33).data - 32768);
analog(:,34,:) = analog(:,34,:) - (parameter(4,7,34).data - 32768);
analog(:,35,:) = analog(:,35,:) - (parameter(4,7,35).data - 32768);
analog(:,36,:) = analog(:,36,:) - (parameter(4,7,36).data - 32768);
```

```
analog(:,37,:) = analog(:,37,:) - (parameter(4,7,37).data - 32768);
analog(:,38,:) = analog(:,38,:) - (parameter(4,7,38).data - 32768);
analog(:,39,:) = analog(:,39,:) - (parameter(4,7,39).data - 32768);
analog(:,40,:) = analog(:,40,:) - (parameter(4,7,40).data - 32768);
analog(:,41,:) = analog(:,41,:) - (parameter(4,7,41).data - 32768);
analog(:,42,:) = analog(:,42,:) - (parameter(4,7,42).data - 32768);
analog(:,43,:) = analog(:,43,:) - (parameter(4,7,43).data - 32768);
analog(:,44,:) = analog(:,44,:) - (parameter(4,7,44).data - 32768);
analog(:,45,:) = analog(:,45,:) - (parameter(4,7,45).data - 32768);
analog(:,46,:) = analog(:,46,:) - (parameter(4,7,46).data - 32768);
analog(:,47,:) = analog(:,47,:) - (parameter(4,7,47).data - 32768);
analog(:,48,:) = analog(:,48,:) - (parameter(4,7,48).data - 32768);
analog(:,49,:) = analog(:,49,:) - (parameter(4,7,49).data - 32768);
analog(:,50,:) = analog(:,50,:) - (parameter(4,7,50).data - 32768);
analog(:,51,:) = analog(:,51,:) - (parameter(4,7,51).data - 32768);
analog(:,52,:) = analog(:,52,:) - (parameter(4,7,52).data - 32768);
analog(:,53,:) = analog(:,53,:) - (parameter(4,7,53).data - 32768);
analog(:,54,:) = analog(:,54,:) - (parameter(4,7,54).data - 32768);
analog(:,55,:) = analog(:,55,:) - (parameter(4,7,55).data - 32768);
analog(:,56,:) = analog(:,56,:) - (parameter(4,7,56).data - 32768);
analog(:,57,:) = analog(:,57,:) - (parameter(4,7,57).data - 32768);
analog(:,58,:) = analog(:,58,:) - (parameter(4,7,58).data - 32768);
analog(:,59,:) = analog(:,59,:) - (parameter(4,7,59).data - 32768);
analog(:,60,:) = analog(:,60,:) - (parameter(4,7,60).data - 32768);
analog(:,61,:) = analog(:,61,:) - (parameter(4,7,61).data - 32768);
analog(:,62,:) = analog(:,62,:) - (parameter(4,7,62).data - 32768);
analog(:,63,:) = analog(:,63,:) - (parameter(4,7,63).data - 32768);
analog(:,64,:) = analog(:,64,:) - (parameter(4,7,64).data - 32768);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

%%% APPLY SCALE FACTORS TO ANALOG DATA %%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

analog(:,1,:) = analog(:,1,:) * (parameter(4,6,1).data); %previously
parameter(4,5&7,1).data
analog(:,2,:) = analog(:,2,:) * (parameter(4,6,2).data);
analog(:,3,:) = analog(:,3,:) * (parameter(4,6,3).data);
analog(:,4,:) = analog(:,4,:) * (parameter(4,6,4).data);
analog(:,5,:) = analog(:,5,:) * (parameter(4,6,5).data);
analog(:,6,:) = analog(:,6,:) * (parameter(4,6,6).data);
analog(:,7,:) = analog(:,7,:) * (parameter(4,6,7).data);
analog(:,8,:) = analog(:,8,:) * (parameter(4,6,8).data);

analog(:,9,:)  = analog(:,9,:)  * (parameter(4,6,9).data);
analog(:,10,:) = analog(:,10,:) * (parameter(4,6,10).data);
analog(:,11,:) = analog(:,11,:) * (parameter(4,6,11).data);
analog(:,12,:) = analog(:,12,:) * (parameter(4,6,12).data);
analog(:,13,:) = analog(:,13,:) * (parameter(4,6,13).data);
analog(:,14,:) = analog(:,14,:) * (parameter(4,6,14).data);
analog(:,15,:) = analog(:,15,:) * (parameter(4,6,15).data);
analog(:,16,:) = analog(:,16,:) * (parameter(4,6,16).data);
```

```
analog(:,17,:) = analog(:,17,:) * (parameter(4,6,17).data);
analog(:,18,:) = analog(:,18,:) * (parameter(4,6,18).data);
analog(:,19,:) = analog(:,19,:) * (parameter(4,6,19).data);
analog(:,20,:) = analog(:,20,:) * (parameter(4,6,20).data);
analog(:,21,:) = analog(:,21,:) * (parameter(4,6,21).data);
analog(:,22,:) = analog(:,22,:) * (parameter(4,6,22).data);
analog(:,23,:) = analog(:,23,:) * (parameter(4,6,23).data);
analog(:,24,:) = analog(:,24,:) * (parameter(4,6,24).data);

analog(:,25,:) = analog(:,25,:) * (parameter(4,6,25).data);
analog(:,26,:) = analog(:,26,:) * (parameter(4,6,26).data);
analog(:,27,:) = analog(:,27,:) * (parameter(4,6,27).data);
analog(:,28,:) = analog(:,28,:) * (parameter(4,6,28).data);
analog(:,29,:) = analog(:,29,:) * (parameter(4,6,29).data);
analog(:,30,:) = analog(:,30,:) * (parameter(4,6,30).data);
analog(:,31,:) = analog(:,31,:) * (parameter(4,6,31).data);
analog(:,32,:) = analog(:,32,:) * (parameter(4,6,32).data);

analog(:,33,:) = analog(:,33,:) * (parameter(4,6,33).data);
analog(:,34,:) = analog(:,34,:) * (parameter(4,6,34).data);
analog(:,35,:) = analog(:,35,:) * (parameter(4,6,35).data);
analog(:,36,:) = analog(:,36,:) * (parameter(4,6,36).data);
analog(:,37,:) = analog(:,37,:) * (parameter(4,6,37).data);
analog(:,38,:) = analog(:,38,:) * (parameter(4,6,38).data);
analog(:,39,:) = analog(:,39,:) * (parameter(4,6,39).data);
analog(:,40,:) = analog(:,40,:) * (parameter(4,6,40).data);
analog(:,41,:) = analog(:,41,:) * (parameter(4,6,41).data);
analog(:,42,:) = analog(:,42,:) * (parameter(4,6,42).data);
analog(:,43,:) = analog(:,43,:) * (parameter(4,6,43).data);
analog(:,44,:) = analog(:,44,:) * (parameter(4,6,44).data);
analog(:,45,:) = analog(:,45,:) * (parameter(4,6,45).data);
analog(:,46,:) = analog(:,46,:) * (parameter(4,6,46).data);
analog(:,47,:) = analog(:,47,:) * (parameter(4,6,47).data);
analog(:,48,:) = analog(:,48,:) * (parameter(4,6,48).data);

analog(:,49,:) = analog(:,49,:) * (parameter(4,6,49).data);
analog(:,50,:) = analog(:,50,:) * (parameter(4,6,50).data);
analog(:,51,:) = analog(:,51,:) * (parameter(4,6,51).data);
analog(:,52,:) = analog(:,52,:) * (parameter(4,6,52).data);
analog(:,53,:) = analog(:,53,:) * (parameter(4,6,53).data);
analog(:,54,:) = analog(:,54,:) * (parameter(4,6,54).data);
analog(:,55,:) = analog(:,55,:) * (parameter(4,6,55).data);
analog(:,56,:) = analog(:,56,:) * (parameter(4,6,56).data);
analog(:,57,:) = analog(:,57,:) * (parameter(4,6,57).data);
analog(:,58,:) = analog(:,58,:) * (parameter(4,6,58).data);
analog(:,59,:) = analog(:,59,:) * (parameter(4,6,59).data);
analog(:,60,:) = analog(:,60,:) * (parameter(4,6,60).data);
analog(:,61,:) = analog(:,61,:) * (parameter(4,6,61).data);
analog(:,62,:) = analog(:,62,:) * (parameter(4,6,62).data);
analog(:,63,:) = analog(:,63,:) * (parameter(4,6,63).data);
analog(:,64,:) = analog(:,64,:) * (parameter(4,6,64).data);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
```

```
%%% TRANSLATE ANALOG DATA TO ZERO %%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

% Take the first 100 samples and zero the data (here adcperframe is 10
- this changes depending on the sampling frequency)
% see adcperframe definition above - original analog matrix is 10 deep
(eg. 300 X 14 X 10) - 14 is # of channels, 10 is the adcperframe, and
302 is the number of rows.
% So 302 * 10 is 3020 which is the number of analog samples, while the
number of video samples is 302. Analog data points 1:10 are along the
depth dimension (eg. 1 X N X10)
% the second set of 10 analog data points would be on row 2, along the
depth dimension (eg, 2 X N X 10), where N is the channel number.

% for i=1:length(analog)
%       meanch1 = mean(analog(1:10,1,1:adcperframe));
%           meanch1 = mean(meanch1);
%           analog(i,1,:) = analog(i,1,:) - meanch1;
%
%       meanch2 = mean(analog(1:10,2,1:adcperframe));
%           meanch2 = mean(meanch2);
%           analog(i,2,:) = analog(i,2,:) - meanch2;
%
%           meanch3 = mean(analog(1:10,3,1:adcperframe));
%           meanch3 = mean(meanch3);
%           analog(i,3,:) = analog(i,3,:) - meanch3;
%
%           meanch4 = mean(analog(1:10,4,1:adcperframe));
%           meanch4 = mean(meanch4);
%           analog(i,4,:) = analog(i,4,:) - meanch4;
%
%           meanch5 = mean(analog(1:10,5,1:adcperframe));
%           meanch5 = mean(meanch5);
%           analog(i,5,:) = analog(i,5,:) - meanch5;
%
%           meanch6 = mean(analog(1:10,6,1:adcperframe));
%           meanch6 = mean(meanch6);
%           analog(i,6,:) = analog(i,6,:) - meanch6;
%
%           meanch7 = mean(analog(1:10,7,1:adcperframe));
%           meanch7 = mean(meanch7);
%           analog(i,7,:) = analog(i,7,:) - meanch7;
%
%           meanch8 = mean(analog(1:10,8,1:adcperframe));
%           meanch8 = mean(meanch8);
%           analog(i,8,:) = analog(i,8,:) - meanch8;
%
%           meanch9 = mean(analog(1:10,9,1:adcperframe));
%           meanch9 = mean(meanch9);
%           analog(i,9,:) = analog(i,9,:) - meanch9;
%
%           meanch10 = mean(analog(1:10,10,1:adcperframe));
%           meanch10 = mean(meanch10);
```

```
%           analog(i,10,:) = analog(i,10,:) - meanch10;
%
%           meanch11 = mean(analog(1:10,11,1:adcperframe));
%           meanch11 = mean(meanch11);
%           analog(i,11,:) = analog(i,11,:) - meanch11;
%
%           meanch12 = mean(analog(1:10,12,1:adcperframe));
%           meanch12 = mean(meanch12);
%           analog(i,12,:) = analog(i,12,:) - meanch12;
%
%           meanch13 = mean(analog(1:10,13,1:adcperframe));
%           meanch13 = mean(meanch13);
%           analog(i,13,:) = analog(i,13,:) - meanch13;
%
%           meanch14 = mean(analog(1:10,14,1:adcperframe));
%           meanch14 = mean(meanch14);
%           analog(i,14,:) = analog(i,14,:) - meanch14;
%
%              meanch15 = mean(analog(1:10,15,1:adcperframe));
%           meanch15 = mean(meanch15);
%           analog(i,15,:) = analog(i,15,:) - meanch15;
%
%              meanch16 = mean(analog(1:10,16,1:adcperframe));
%           meanch16 = mean(meanch16);
%           analog(i,16,:) = analog(i,16,:) - meanch16;
%
%       meanch17 = mean(analog(1:10,17,1:adcperframe));
%           meanch17 = mean(meanch17);
%           analog(i,17,:) = analog(i,17,:) - meanch17;
%
%       meanch18 = mean(analog(1:10,18,1:adcperframe));
%           meanch18 = mean(meanch18);
%           analog(i,18,:) = analog(i,18,:) - meanch18;
%
%       meanch19 = mean(analog(1:10,19,1:adcperframe));
%           meanch19 = mean(meanch19);
%           analog(i,19,:) = analog(i,19,:) - meanch19;
%
%       meanch20 = mean(analog(1:10,20,1:adcperframe));
%           meanch20 = mean(meanch20);
%           analog(i,20,:) = analog(i,20,:) - meanch20;
%
%       meanch21 = mean(analog(1:10,21,1:adcperframe));
%           meanch21 = mean(meanch21);
%           analog(i,21,:) = analog(i,21,:) - meanch21;
%
%       meanch22 = mean(analog(1:10,22,1:adcperframe));
%           meanch22 = mean(meanch22);
%           analog(i,22,:) = analog(i,22,:) - meanch22;
%
%       meanch23 = mean(analog(1:10,23,1:adcperframe));
%           meanch23 = mean(meanch23);
%           analog(i,23,:) = analog(i,23,:) - meanch23;
%
%       meanch24 = mean(analog(1:10,24,1:adcperframe));
```

```
%           meanch24 = mean(meanch24);
%           analog(i,24,:) = analog(i,24,:) - meanch24;
%
%       meanch25 = mean(analog(1:10,25,1:adcperframe));
%           meanch25 = mean(meanch25);
%           analog(i,25,:) = analog(i,25,:) - meanch25;
%
%       meanch26 = mean(analog(1:10,26,1:adcperframe));
%           meanch26 = mean(meanch26);
%           analog(i,26,:) = analog(i,26,:) - meanch26;
%
%       meanch27 = mean(analog(1:10,27,1:adcperframe));
%           meanch27 = mean(meanch27);
%           analog(i,27,:) = analog(i,27,:) - meanch27;
%
%       meanch28 = mean(analog(1:10,28,1:adcperframe));
%           meanch28 = mean(meanch28);
%           analog(i,28,:) = analog(i,28,:) - meanch28;
%
%       meanch29 = mean(analog(1:10,29,1:adcperframe));
%           meanch29 = mean(meanch29);
%           analog(i,29,:) = analog(i,29,:) - meanch29;
%
%       meanch30 = mean(analog(1:10,30,1:adcperframe));
%           meanch30 = mean(meanch30);
%           analog(i,30,:) = analog(i,30,:) - meanch30;
%
%       meanch31 = mean(analog(1:10,31,1:adcperframe));
%           meanch31 = mean(meanch31);
%           analog(i,31,:) = analog(i,31,:) - meanch31;
%
%       meanch32 = mean(analog(1:10,32,1:adcperframe));
%           meanch32 = mean(meanch32);
%           analog(i,32,:) = analog(i,32,:) - meanch32;
%
%   end


% Changing from a 3D matrix to a 2D matrix
B = analog;
ndim1 = size(B,1);
ndim2 = size(B,2);
ndim3 = size(B,3);

nrow = ndim1*ndim3;
ncol = ndim2;

newanalog = zeros(nrow,ncol);
for i = 1:ndim3
    newanalog(i:ndim3:nrow,:) = B(:,:,i);
end
```

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
                              %%% OLD KISTLER ANALOG DATA - PLATE 1 %%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

% Sum Kistler data
Kofx1 = newanalog(:,1) +  newanalog(:,2); % ie. channel 1 + channel 2
gives X-dir data
Kofy1 = newanalog(:,3) +  newanalog(:,4); % ie. channel 3 + channel 4
gives Y-dir data
Kofz1 = newanalog(:,5) +  newanalog(:,6) + newanalog(:,7) +
newanalog(:,8); % ie. channel 5+6+7+8 gives Z-dir data

%for i=1:length(Kgrf)
%   if (Kgrf(i)==0)
%           Kalpha(i) = -200;
%           Kbeta(i)  = -200;
%      Kgamma(i) = -200;
%   end
%end

% define distances to origin of plate from transducers in mm
ao = 120;
bo = 200;

% Components of the resulting moment vector
Komx1 = bo*[newanalog(:,5) +  newanalog(:,6) - (newanalog(:,7)) -
(newanalog(:,8))];
Komy1 = ao*[-(newanalog(:,5)) +  newanalog(:,6) + newanalog(:,7) -
(newanalog(:,8))];
Komz1 = bo*[-(newanalog(:,1)) +  newanalog(:,2)] + ao*[newanalog(:,3) -
newanalog(:,4)];

% Depth of transducer from surface of plate (az) in mm
azo = -52;

Komx1 = Komx1 + Kofy1*azo;
Komy1 = Komy1 - Kofx1*azo;

% Center of Pressure Data
Koax1 = (-Komy1)./Kofz1;
Koay1 = (Komx1)./Kofz1;

%%CHECK HERE
% Call the corner parameters for the Kistler plate
yBo = parameter(5,4,1).data + 300; % 300 is half the length of the
Kistler in mm
xBo = parameter(5,4,2).data + 200; % 200 is half the width of the
Kistler in mm

Koax1 = xBo - Koax1;
Koay1 = yBo - Koay1;
```

```
KoTz1 = Komz1 - (Kofy1.*Koax1) + (Kofx1.*Koay1);

% In Vicon Coordinates
Koaxp1 = Koay1;
Koayp1 = Koax1;

Komxp1 = -Komy1;
Komyp1 = -Komx1;
Komzp1 = -Komz1;

Kofxp1 = -Kofy1;
Kofyp1 = -Kofx1;
Kofzp1 = -Kofz1;

KoTzp1 = -KoTz1;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
% Smooth Analog Data
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

% Force and Moment Data
% Hamming 20pt Window (really 10 pts on either side of centre point =
21 points)
% clear a b
% a = 1;
% b = hamming(21);
% FD1 = filter(b,a,Kofx)/sum(b);
% FD2 = filter(b,a,Kofy)/sum(b);
% FD3 = filter(b,a,Kofz)/sum(b);
% FD4 = filter(b,a,Komx)/sum(b);
% FD5 = filter(b,a,Komy)/sum(b);
% FD6 = filter(b,a,Komz)/sum(b);
% FD8 = filter(b,a,Koax)/sum(b);
% FD9 = filter(b,a,Koay)/sum(b);
%
% % Remove Phase shifts
% l_FD1 = length(FD1); % length of smoothed data
% Kofx = FD1(11:l_FD1); % remove first ten points (half the window)
% Kofy = FD2(11:l_FD1);
% Kofz = FD3(11:l_FD1);
% Komx = FD4(11:l_FD1);
% Komy = FD5(11:l_FD1);
% Komz = FD6(11:l_FD1);
% Koax = FD8(11:l_FD1);
% Koay = FD9(11:l_FD1);
%
% %figure
% %plot(1:N,Kfz,'r', (1:N)-10,FD3,'g')   % -10 here to remove phase
shift
%
```

```
% % KTz data - using smaller window for higher frequency data if
necessary
% % Will leave COP data as is
% clear a b
% a = 1;
% b = hamming(21);
% l_FD1 = length(FD1); % length of smoothed data
% FD7 = filter(b,a,KoTz)/sum(b);
% KoTz = FD7(11:l_FD1);

% Calculate resultant ground reaction force
Kogrfp1 = sqrt(Kofxp1.^2 + Kofyp1.^2 + Kofzp1.^2);

% Calculate direction cosines of resultant grf
Koalphap1 = acos(Kofxp1./Kogrfp1);
Kobetap1 = acos(Kofyp1./Kogrfp1);
Kogammap1 = acos(Kofzp1./Kogrfp1);

% Take one tenth of the data
Kofxp1=Kofxp1(1:adcperframe:length(Kofxp1));
Kofyp1=Kofyp1(1:adcperframe:length(Kofyp1));
Kofzp1=Kofzp1(1:adcperframe:length(Kofzp1));

Komxp1=Komxp1(1:adcperframe:length(Komxp1));
Komyp1=Komyp1(1:adcperframe:length(Komyp1));
Komzp1=Komzp1(1:adcperframe:length(Komzp1));

Koaxp1=Koaxp1(1:adcperframe:length(Koaxp1));
Koayp1=Koayp1(1:adcperframe:length(Koayp1));
KoTzp1=KoTzp1(1:adcperframe:length(KoTzp1));

Koalphap1=Koalphap1(1:adcperframe:length(Koalphap1));
Kobetap1=Kobetap1(1:adcperframe:length(Kobetap1));
Kogammap1=Kogammap1(1:adcperframe:length(Kogammap1));
Kogrfp1 = Kogrfp1(1:adcperframe:length(Kogrfp1));

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
                                             %%% NEW KISTLER ANALOG DATA - PLATE 2 %%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
%PLATE 2
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

% Sum Kistler data
Kfx2 = newanalog(:,9) +  newanalog(:,10); % ie. channel 1 + channel 2
gives X-dir data
Kfy2 = newanalog(:,11) +  newanalog(:,12); % ie. channel 3 + channel 4
gives Y-dir data
```

```
Kfz2 = newanalog(:,13) +  newanalog(:,14) + newanalog(:,15) +
newanalog(:,16); % ie. channel 5+6+7+8 gives Z-dir data

%for i=1:length(Kgrf)
%   if (Kgrf(i)==0)
%           Kalpha(i) = -200;
%           Kbeta(i)  = -200;
%      Kgamma(i) = -200;
%   end
%end

% define distances to origin of plate from transducers in mm
clear a2 b2
a2 = 120;
b2 = 200;

% Components of the resulting moment vector
Kmx2 = b2*[newanalog(:,13) +  newanalog(:,14) - (newanalog(:,15)) -
(newanalog(:,16))];
Kmy2 = a2*[-(newanalog(:,13)) +  newanalog(:,14) + newanalog(:,15) -
(newanalog(:,16))];
Kmz2 = b2*[-(newanalog(:,9)) +  newanalog(:,10)] + a2*[newanalog(:,11)
- newanalog(:,12)];

% Depth of transducer from surface of plate (az) in mm
az2 = -52;

Kmx2 = Kmx2 + Kfy2*az2;
Kmy2 = Kmy2 - Kfx2*az2;

% Center of Pressure Data
Kax2 = (-Kmy2)./Kfz2;
Kay2 = (Kmx2)./Kfz2;

% Call the corner parameters for the Kistler plate
yB2 = parameter(5,4,13).data + 300; % 300 is half the length of the
Kistler in mm
xB2 = parameter(5,4,14).data + 200; % 200 is half the width of the
Kistler in mm

Kax2 = xB2 - Kax2;
Kay2 = yB2 - Kay2;

KTz2 = Kmz2 - (Kfy2.*Kax2) + (Kfx2.*Kay2);

% In Vicon Coordinates
Kaxp2 = Kay2;
Kayp2 = Kax2;

Kmxp2 = -Kmy2;
Kmyp2 = -Kmx2;
Kmzp2 = -Kmz2;
```

```
Kfxp2 = -Kfy2;
Kfyp2 = -Kfx2;
Kfzp2 = -Kfz2;

KTzp2 = -KTz2;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
% Smooth Analog Data
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

% Force and Moment Data
% Hamming 20pt Window (really 10 pts on either side of centre point =
21 points)
% clear a b
% a = 1;
% b = hamming(21);
% FD1 = filter(b,a,Kfx)/sum(b);
% FD2 = filter(b,a,Kfy)/sum(b);
% FD3 = filter(b,a,Kfz)/sum(b);
% FD4 = filter(b,a,Kmx)/sum(b);
% FD5 = filter(b,a,Kmy)/sum(b);
% FD6 = filter(b,a,Kmz)/sum(b);
% FD8 = filter(b,a,Kax)/sum(b);
% FD9 = filter(b,a,Kay)/sum(b);
%
% % Remove Phase shifts
% l_FD1 = length(FD1); % length of smoothed data
% Kfx = FD1(11:l_FD1); % remove first ten points (half the window)
% Kfy = FD2(11:l_FD1);
% Kfz = FD3(11:l_FD1);
% Kmx = FD4(11:l_FD1);
% Kmy = FD5(11:l_FD1);
% Kmz = FD6(11:l_FD1);
% Kax = FD8(11:l_FD1);
% Kay = FD9(11:l_FD1);

%figure
%plot(1:N,Kfz,'r', (1:N)-10,FD3,'g')   % -10 here to remove phase shift

% KTz data - using smaller window for higher frequency data if
necessary
% Will leave COP data as is
% clear a b
% a = 1;
% b = hamming(21);
% l_FD1 = length(FD1); % length of smoothed data
% FD7 = filter(b,a,KTz)/sum(b);
% KTz = FD7(11:l_FD1);

% Calculate resultant ground reaction force
Kgrfp2 = sqrt(Kfxp2.^2 + Kfyp2.^2 + Kfzp2.^2);
```

```
% Calculate direction cosines of resultant grf
Kalphap2 = acos(Kfxp2./Kgrfp2);
Kbetap2 = acos(Kfyp2./Kgrfp2);
Kgammap2 = acos(Kfzp2./Kgrfp2);

% Take one tenth of the data
Kfxp2=Kfxp2(1:adcperframe:length(Kfxp2));
Kfyp2=Kfyp2(1:adcperframe:length(Kfyp2));
Kfzp2=Kfzp2(1:adcperframe:length(Kfzp2));

Kmxp2=Kmxp2(1:adcperframe:length(Kmxp2));
Kmyp2=Kmyp2(1:adcperframe:length(Kmyp2));
Kmzp2=Kmzp2(1:adcperframe:length(Kmzp2));

Kaxp2=Kaxp2(1:adcperframe:length(Kaxp2));
Kayp2=Kayp2(1:adcperframe:length(Kayp2));
KTzp2=KTzp2(1:adcperframe:length(KTzp2));

Kalphap2=Kalphap2(1:adcperframe:length(Kalphap2));
Kbetap2=Kbetap2(1:adcperframe:length(Kbetap2));
Kgammap2=Kgammap2(1:adcperframe:length(Kgammap2));
Kgrfp2 = Kgrfp2(1:adcperframe:length(Kgrfp2));

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
%PLATE 3
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

% Sum Kistler data
Kfx3 = newanalog(:,17) +  newanalog(:,18); % ie. channel 1 + channel 2
gives X-dir data
Kfy3 = newanalog(:,19) +  newanalog(:,20); % ie. channel 3 + channel 4
gives Y-dir data
Kfz3 = newanalog(:,21) +  newanalog(:,22) + newanalog(:,23) +
newanalog(:,24); % ie. channel 5+6+7+8 gives Z-dir data

%for i=1:length(Kgrf)
%   if (Kgrf(i)==0)
%           Kalpha(i) = -200;
%           Kbeta(i)  = -200;
%       Kgamma(i) = -200;
%   end
%end

% define distances to origin of plate from transducers in mm
clear a3 b3
a3 = 120;
b3 = 200;

% Components of the resulting moment vector
Kmx3 = b3*[newanalog(:,21) +  newanalog(:,22) - (newanalog(:,23)) -
(newanalog(:,24))];
```

```
Kmy3 = a3*[-(newanalog(:,21)) +  newanalog(:,22) + newanalog(:,23) -
(newanalog(:,24))];
Kmz3 = b3*[-(newanalog(:,17)) +  newanalog(:,18)] + a3*[newanalog(:,19)
- newanalog(:,20)];

% Depth of transducer from surface of plate (az) in mm
az3 = -52;

Kmx3 = Kmx3 + Kfy3*az3;
Kmy3 = Kmy3 - Kfx3*az3;

% Center of Pressure Data
Kax3 = (-Kmy3)./Kfz3;
Kay3 = (Kmx3)./Kfz3;

% Call the corner parameters for the Kistler plate
yB3 = parameter(5,4,25).data + 300; % 300 is half the length of the
Kistler in mm
xB3 = parameter(5,4,26).data + 200; % 200 is half the width of the
Kistler in mm

Kax3 = xB3 - Kax3;
Kay3 = yB3 - Kay3;

KTz3 = Kmz3 - (Kfy3.*Kax3) + (Kfx3.*Kay3);

% In Vicon Coordinates
Kaxp3 = Kay3;
Kayp3 = Kax3;

Kmxp3 = -Kmy3;
Kmyp3 = -Kmx3;
Kmzp3 = -Kmz3;

Kfxp3 = -Kfy3;
Kfyp3 = -Kfx3;
Kfzp3 = -Kfz3;

KTzp3 = -KTz3;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%
% Smooth Analog Data
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%

% Force and Moment Data
% Hamming 20pt Window (really 10 pts on either side of centre point =
21 points)
% clear a b
% a = 1;
% b = hamming(21);
% FD1 = filter(b,a,Kfxp3)/sum(b);
```

```
% FD2 = filter(b,a,Kfyp3)/sum(b);
% FD3 = filter(b,a,Kfzp3)/sum(b);
% FD4 = filter(b,a,Kmxp3)/sum(b);
% FD5 = filter(b,a,Kmyp3)/sum(b);
% FD6 = filter(b,a,Kmzp3)/sum(b);
% FD8 = filter(b,a,Kaxp3)/sum(b);
% FD9 = filter(b,a,Kayp3)/sum(b);
%
% % Remove Phase shifts
% l_FD1 = length(FD1); % length of smoothed data
% Kfxp3 = FD1(11:l_FD1); % remove first ten points (half the window)
% Kfyp3 = FD2(11:l_FD1);
% Kfzp3 = FD3(11:l_FD1);
% Kmxp3 = FD4(11:l_FD1);
% Kmyp3 = FD5(11:l_FD1);
% Kmzp3 = FD6(11:l_FD1);
% Kaxp3 = FD8(11:l_FD1);
% Kayp3 = FD9(11:l_FD1);

%figure
%plot(1:N,Kfz,'r', (1:N)-10,FD3,'g')   % -10 here to remove phase shift

% KTz data - using smaller window for higher frequency data if
necessary
% Will leave COP data as is
% clear a b
% a = 1;
% b = hamming(21);
% l_FD1 = length(FD1); % length of smoothed data
% FD7 = filter(b,a,KTz)/sum(b);
% KTz = FD7(11:l_FD1);

% Calculate resultant ground reaction force
Kgrfp3 = sqrt(Kfxp3.^2 + Kfyp3.^2 + Kfzp3.^2);

% Calculate direction cosines of resultant grf
Kalphap3 = acos(Kfxp3./Kgrfp3);
Kbetap3 = acos(Kfyp3./Kgrfp3);
Kgammap3 = acos(Kfzp3./Kgrfp3);

% Take one tenth of the data
Kfxp3=Kfxp3(1:adcperframe:length(Kfxp3));
Kfyp3=Kfyp3(1:adcperframe:length(Kfyp3));
Kfzp3=Kfzp3(1:adcperframe:length(Kfzp3));

Kmxp3=Kmxp3(1:adcperframe:length(Kmxp3));
Kmyp3=Kmyp3(1:adcperframe:length(Kmyp3));
Kmzp3=Kmzp3(1:adcperframe:length(Kmzp3));

Kaxp3=Kaxp3(1:adcperframe:length(Kaxp3));
Kayp3=Kayp3(1:adcperframe:length(Kayp3));
KTzp3=KTzp3(1:adcperframe:length(KTzp3));
```

```
Kalphap3=Kalphap3(1:adcperframe:length(Kalphap3));
Kbetap3=Kbetap3(1:adcperframe:length(Kbetap3));
Kgammap3=Kgammap3(1:adcperframe:length(Kgammap3));
Kgrfp3 = Kgrfp3(1:adcperframe:length(Kgrfp3));


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%
%PLATE 4
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%

% Sum Kistler data
Kfx4 = newanalog(:,25) +  newanalog(:,26); % ie. channel 1 + channel 2
gives X-dir data
Kfy4 = newanalog(:,27) +  newanalog(:,28); % ie. channel 3 + channel 4
gives Y-dir data
Kfz4 = newanalog(:,29) +  newanalog(:,30) + newanalog(:,31) +
newanalog(:,32); % ie. channel 5+6+7+8 gives Z-dir data

%for i=1:length(Kgrf)
%   if (Kgrf(i)==0)
%           Kalpha(i) = -200;
%           Kbeta(i)  = -200;
%      Kgamma(i) = -200;
%   end
%end

% define distances to origin of plate from transducers in mm
clear a4 b4
a4 = 120;
b4 = 200;

% Components of the resulting moment vector
Kmx4 = b4*[newanalog(:,29) +  newanalog(:,30) - (newanalog(:,31)) -
(newanalog(:,32))];
Kmy4 = a4*[-(newanalog(:,29)) +  newanalog(:,30) + newanalog(:,31) -
(newanalog(:,32))];
Kmz4 = b4*[-(newanalog(:,25)) +  newanalog(:,26)] + a4*[newanalog(:,27)
- newanalog(:,28)];

% Depth of transducer from surface of plate (az) in mm
az4 = -52;

Kmx4 = Kmx4 + Kfy4*az4;
Kmy4 = Kmy4 - Kfx4*az4;

% Center of Pressure Data
Kax4 = (-Kmy4)./Kfz4;
Kay4 = (Kmx4)./Kfz4;

% Call the corner parameters for the Kistler plate
```

```
yB4 = parameter(5,4,37).data + 300; % 300 is half the length of the
Kistler in mm
xB4 = parameter(5,4,38).data + 200; % 200 is half the width of the
Kistler in mm

Kax4 = xB4 - Kax4;
Kay4 = yB4 - Kay4;

KTz4 = Kmz4 - (Kfy4.*Kax4) + (Kfx4.*Kay4);

% In Vicon Coordinates
Kaxp4 = Kay4;
Kayp4 = Kax4;

Kmxp4 = -Kmy4;
Kmyp4 = -Kmx4;
Kmzp4 = -Kmz4;

Kfxp4 = -Kfy4;
Kfyp4 = -Kfx4;
Kfzp4 = -Kfz4;

KTzp4 = -KTz4;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%
% Smooth Analog Data
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

% Force and Moment Data
% Hamming 20pt Window (really 10 pts on either side of centre point =
21 points)
% clear a b
% a = 1;
% b = hamming(21);
% FD1 = filter(b,a,Kfx)/sum(b);
% FD2 = filter(b,a,Kfy)/sum(b);
% FD3 = filter(b,a,Kfz)/sum(b);
% FD4 = filter(b,a,Kmx)/sum(b);
% FD5 = filter(b,a,Kmy)/sum(b);
% FD6 = filter(b,a,Kmz)/sum(b);
% FD8 = filter(b,a,Kax)/sum(b);
% FD9 = filter(b,a,Kay)/sum(b);
%
% % Remove Phase shifts
% l_FD1 = length(FD1); % length of smoothed data
% Kfx = FD1(11:l_FD1); % remove first ten points (half the window)
% Kfy = FD2(11:l_FD1);
% Kfz = FD3(11:l_FD1);
% Kmx = FD4(11:l_FD1);
% Kmy = FD5(11:l_FD1);
% Kmz = FD6(11:l_FD1);
```

```
% Kax = FD8(11:l_FD1);
% Kay = FD9(11:l_FD1);

%figure
%plot(1:N,Kfz,'r', (1:N)-10,FD3,'g')    % -10 here to remove phase shift

% KTz data - using smaller window for higher frequency data if
necessary
% Will leave COP data as is
% clear a b
% a = 1;
% b = hamming(21);
% l_FD1 = length(FD1); % length of smoothed data
% FD7 = filter(b,a,KTz)/sum(b);
% KTz = FD7(11:l_FD1);

% Calculate resultant ground reaction force
Kgrfp4 = sqrt(Kfxp4.^2 + Kfyp4.^2 + Kfzp4.^2);

% Calculate direction cosines of resultant grf
Kalphap4 = acos(Kfxp4./Kgrfp4);
Kbetap4 = acos(Kfyp4./Kgrfp4);
Kgammap4 = acos(Kfzp4./Kgrfp4);

% Take one tenth of the data
Kfxp4=Kfxp4(1:adcperframe:length(Kfxp4));
Kfyp4=Kfyp4(1:adcperframe:length(Kfyp4));
Kfzp4=Kfzp4(1:adcperframe:length(Kfzp4));

Kmxp4=Kmxp4(1:adcperframe:length(Kmxp4));
Kmyp4=Kmyp4(1:adcperframe:length(Kmyp4));
Kmzp4=Kmzp4(1:adcperframe:length(Kmzp4));

Kaxp4=Kaxp4(1:adcperframe:length(Kaxp4));
Kayp4=Kayp4(1:adcperframe:length(Kayp4));
KTzp4=KTzp4(1:adcperframe:length(KTzp4));

Kalphap4=Kalphap4(1:adcperframe:length(Kalphap4));
Kbetap4=Kbetap4(1:adcperframe:length(Kbetap4));
Kgammap4=Kgammap4(1:adcperframe:length(Kgammap4));
Kgrfp4 = Kgrfp4(1:adcperframe:length(Kgrfp4));

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

%%% EMG DATA %%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

% Name EMG channels
Ch1 = newanalog(:,33);
%Ch1=Ch1(1:adcperframe:length(Ch1));
Ch2 = newanalog(:,34);
```

```
%Ch2=Ch2(1:25:length(Ch2));
Ch3 = newanalog(:,35);
%Ch3=Ch3(1:25:length(Ch3));
Ch4 = newanalog(:,36);
%Ch4=Ch4(1:25:length(Ch4));
Ch5 = newanalog(:,37);
%Ch5=Ch5(1:25:length(Ch5));
Ch6 = newanalog(:,38);
%Ch6=Ch6(1:25:length(Ch6));
Ch7 = newanalog(:,39);
%Ch7=Ch7(1:25:length(Ch7));
Ch8 = newanalog(:,40);
Ch9 = newanalog(:,41);
Ch10 = newanalog(:,42);
%Ch8=Ch8(1:25:length(Ch8));
% Ch9 = newanalog(:,41);
% Ch10 = newanalog(:,42);

% EMG_data = [Ch1 Ch2 Ch3 Ch4 Ch5 Ch6 Ch7 Ch8];% Ch9 Ch10];
% emg_loc = strcat(addrs,'\',trial(1:5),'\',trial,'.emg');
% %emg = fopen(emg_loc,'w');
% csvwrite(emg_loc,EMG_data);


% if(emg == -1)
%    clc;
%    fprintf('\nThe file %s was not created!\n.',emg_loc);
%    fprintf('\nPress any key');
%    pause;
%    return;
% end



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% *** VERY IMPORTANT *** - there are negatives placed on some of the
variables below before saving to file.  This is because of the lab
setup -
% patients walked towards origin (towards negative) - we rotated 180
deg to place the subject in the diagonal quadrant which would now have
the person walking
% towards origin - but towards + direction.  All force plate data was
changed accordingly to match this new quadrant.
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% Changing from a 3D matrix to a 2D matrix in order to write to *.tda
[x,y,z] = size(coord);
for (i=1:x)
   for (j=1:(y-1))
      for (k=1:z)
         new_coord(i,(3*(j-1)+k)) = coord(i,(j+1),k);
```

```
      end
   end
end

%temp_array = zeros(1,((y-1)*z));
%temp_array(1) = ISIDE;
%temp_array(2) = x;
%temporary_matrix = [temp_array;new_coord];

[x,y] = size(new_coord);

% This is where the x and y coordinates are negated because of the lab
setup (eg. in hypotonic study 2002 - we were walking towards origin)
***** VERY IMPORTANT ****
% Last Modified by V. Chester, February 28th, 2002

% for i=1:(y/3)
%    new_coord((1:x),3*(i-1)+1) = -new_coord((1:x),3*(i-1)+1);
%    new_coord((1:x),3*(i-1)+2) = -new_coord((1:x),3*(i-1)+2);
% end

%csvwrite('c:\swin2.csv',new_coord);

analogdata = [-Kofxp1 -Kofyp1 Kofzp1 Komxp1 Komyp1 Komzp1 Kogrfp1
Koalphap1 Kobetap1 Kogammap1 -Koaxp1 -Koayp1 KoTzp1...
              -Kfxp2 -Kfyp2 Kfzp2 Kmxp2 Kmyp2 Kmzp2 Kgrfp2 Kalphap2
Kbetap2 Kgammap2 -Kaxp2 -Kayp2 KTzp2...
              -Kfxp3 -Kfyp3 Kfzp3 Kmxp3 Kmyp3 Kmzp3 Kgrfp3 Kalphap3
Kbetap3 Kgammap3 -Kaxp3 -Kayp3 KTzp3...
              -Kfxp4 -Kfyp4 Kfzp4 Kmxp4 Kmyp4 Kmzp4 Kgrfp4 Kalphap4
Kbetap4 Kgammap4 -Kaxp4 -Kayp4 KTzp4];
else

analogdata = '';
EMG_data = '';
end
save walk13.mat
return;
```

Appendix CII

```
function[markerlist,analoglist]=c3dlabel4pworking()
% READMKR: Output desired marker labels and analog channel labels
%
% Input:
%
% Output:        markerlist - list of markers in a character array
%                analoglist - list of analog channels in a character
array

% Author:          Alan Morris
% Date:                September 1999
% Institution:        Bloorview MacMillan Centre (Gait Laboratory)
%                 350 Rumsey Road
%                 Toronto, Ontario
%                  Canada M4G-1R8
%                 Tel (416) 425-6220 x3508 / Fax (416) 425-1634


%Marker list
% markerlist = {'BLNK' 'frhd' 'lfth' 'rth' 'c7' 'Lcla' 'Rcla' 'Lsh'
'Llel' 'Lmel' 'Lrad' 'Luln' 'L2mc' 'L5mc' 'Lasi' 'Lgtr' 'Lthi' 'Ltib'
'Lhee' 'Ltoe' 'Lank' 'Lkne' 'Rsh' 'Rlel' 'Rmel' 'Rrad' 'Ruln' 'R2mc'
'R5mc' 'Rasi' 'Rgtr' 'Rthi' 'Rtib' 'Rhee' 'Rtoe' 'Rank' 'Rkne' 'sacr'};
markerlist = {'BLNK' 'NECK' 'LSHO' 'LASI' 'LGRT' 'LTHI' 'LTIB' 'LHEE'
'LTOE' 'LANK' 'LKNE' 'RSHO' 'RASI' 'RGRT' 'RTHI' 'RTIB' 'RHEE' 'RTOE'
'RANK' 'RKNE' 'SACR'};

%markerlist = {'BLNK' 'ILAC' 'ASIS'        'GTRO'        'THIW'
'KNEE' 'TIBW'        'ANK'        'HEEL' 'TOE'};
%markerlist = {'BCP' 'ULN' 'SHO'};
%markerlist = {'BLNK' 'RGRT' 'RKNE' 'RANK'};
%markerlist = {'BLNK' 'SACR' 'LASI' 'RASI' 'LGRT' 'LTHI' 'LKNE' 'LANK'
'LHEE' 'LTOE' 'RGRT' 'RTHI' 'RKNE' 'RANK' 'RHEE' 'RTOE'};
%markerlist = {'BLNK' 'SACR' 'LASI' 'RASI' 'LTHI' 'RTHI' 'LKNE' 'RKNE'
'LTIB' 'RTIB' 'LANK' 'RANK' 'LTOE' 'RTOE'};

%Analog channel list
%Old one - when we had two plates
%analoglist = {'X12' 'X34' 'Y14' 'Y23' 'Z1' 'Z2' 'Z3' 'Z4' 'Fx' 'Fy'
'Fz' 'Mx' 'My' 'Mz'};
%with EMG analoglist = {'Fx1+2' 'Fx3+4' 'Fy1+4' 'Fy2+3' 'Z1' 'Z2' 'Z3'
'Z4' 'Fx' 'Fy' 'Fz' 'Mx' 'My' 'Mz' 'Fx12' 'Fx34' 'Fy14' 'Fy23' 'kZ1'
'kZ2' 'kZ3' 'kZ4' 'Ext1' 'Ext2' 'Ext3' 'Ext4' 'Ext5' 'Ext6' 'Ext7'
'Ext8'};
%analoglist= {'Fx1+2(P1)' 'Fx3+4(P1)' 'Fy1+4(P1)' 'Fy2+3(P1)' 'Z1(P1)'
'Z2(P1)' 'Z3(P1)' 'Z4(P1)' 'Fx1+2(P2)' 'Fx3+4(P2)' 'Fy1+4(P2)'
'Fy2+3(P2)' 'Z1(P2)' 'Z2(P2)' 'Z3(P2)' 'Z4(P2)' 'Fx1+2(P3)' 'Fx3+4(P3)'
'Fy1+4(P3)' 'Fy2+3(P3)' 'Z1(P3)' 'Z2(P3)' 'Z3(P3)' 'Z4(P3)' 'Fx1+2(P4)'
'Fx3+4(P4)' 'Fy1+4(P4)' 'Fy2+3(P4)' 'Z1(P4)' 'Z2(P4)' 'Z3(P4)' 'Z4(P4)'
'Ext01' 'Ext02' 'Ext03' 'Ext04' 'Ext05' 'Ext06' 'Ext07' 'Ext08'};
```

```
analoglist= {'Fx1+2(P1)' 'Fx3+4(P1)' 'Fy1+4(P1)' 'Fy2+3(P1)' 'Z1(P1)'
'Z2(P1)' 'Z3(P1)' 'Z4(P1)' 'Fx1+2(P2)' 'Fx3+4(P2)' 'Fy1+4(P2)'
'Fy2+3(P2)' 'Z1(P2)' 'Z2(P2)' 'Z3(P2)' 'Z4(P2)' 'Fx1+2(P3)' 'Fx3+4(P3)'
'Fy1+4(P3)' 'Fy2+3(P3)' 'Z1(P3)' 'Z2(P3)' 'Z3(P3)' 'Z4(P3)' 'Fx1+2(P4)'
'Fx3+4(P4)' 'Fy1+4(P4)' 'Fy2+3(P4)' 'Z1(P4)' 'Z2(P4)' 'Z3(P4)' 'Z4(P4)'
'Ext01' 'Ext02' 'Ext03' 'Ext04' 'Ext05' 'Ext06' 'Ext07' 'Ext08' 'Ext09'
'Ext10' 'Ext11' 'Ext12' 'Ext13' 'Ext14' 'Ext15' 'Ext16' 'CH49' 'CH50'
'CH51' 'CH52' 'CH53' 'CH54' 'CH55' 'CH56' 'CH57' 'CH58' 'CH59' 'CH60'
'CH61' 'CH62' 'CH63' 'CH64'};
%also tried (to account for space 7 and 8 on AMTI which are blank)
%ie difference between type 2 and type 3 in Force Platform: Group
'CHANNEL'
%analoglist = {'Fx12' 'Fx34' 'Fy14' 'Fy23' 'Fz1' 'Fz2' 'Fz3' 'Fz4'
'amtiFx' 'amtiFy' 'amtiFz' 'amtiMx' 'amtiMy' 'amtiMz' 'na' 'na'};
```

Appendix D

Bike Task

Symmetry

Subject Name	Bicep	Tricep	Trapezius	Deltoids	Eresp
AmNFR11a	88.8	95.5	51.9	47.1	73.2
CcNFR11a	92.1	51.8	57.9	42.7	68.1
DkNFR07a	100.0	75.2	50.2	56.8	72.1
EjNMR11a	100.0	72.0	38.0	32.8	79.9
FpNMR14a	100.0	15.3	45.6	54.0	85.3
GzNML11a	35.3	47.4	58.4	67.9	55.6
HjNMR08a	100.0	100.0	73.3	100.0	100.0
LsNMR09a	100.0	79.1	62.8	24.4	85.4
MwNMR06a	45.2	52.4	62.6	36.1	59.5
NvNFR11a	64.4	57.4	60.2	82.2	95.6
OsNFR11a	82.6	17.7	14.3	60.8	72.8
SmNFR07a	18.6	62.7	21.0	74.2	100.0
TmNMR04a	100.0	60.7	100.0	82.3	82.8
UkNFR07a	100.0	61.2	57.6	56.6	89.7
VtNML11a	89.3	100.0	50.9	87.7	83.9
Mean	81.1	63.2	53.7	60.4	80.3
Standard Deviation	27.2	25.5	20.3	21.9	13.4

Subject Name	Bicep	Tricep	Trapezius	Deltoids	Eresp
IcPML14p	86.8	2.6	56.6	100.0	60.0
# stdev's from mean	-0.2	2.4	-0.1	-1.8	1.5
PaPFL10p	100.0	51.7	100.0	100.0	33.4
# stdev's from mean	-0.7	0.5	-2.3	-1.8	3.5
RcPFR06p	86.8	36.2	94.4	100.0	67.3
# stdev's from mean	-0.2	1.1	-2.0	-1.8	1.0
WrPFR13p	100.0	69.8	94.4	100.0	89.0
# stdev's from mean	-0.7	-0.3	-2.0	-1.8	-0.6
IcPML14n	100.0	3.5	100.0	100.0	100.0
# stdev's from mean	-0.7	2.3	-2.3	-1.8	-1.5
PaPFL10n	76.4	88.0	46.2	68.2	57.0
# stdev's from	0.2	-1.0	0.4	-0.4	1.7

mean					
RcPFR06n	91.5	61.6	78.9	100.0	100.0
# stdev's from mean	-0.4	0.1	-1.2	-1.8	-1.5
WrPFR13n	7.8	25.6	85.3	78.0	100.0
# stdev's from mean	2.7	1.5	-1.6	-0.8	-1.5

Co-Contraction

Subject Name	RBicep-RTricep	LBicep-LTricep	RTrap-RDelt	LTrap-LDelt
AmNFR11a	32.4	6.1	70.5	72.5
CcNFR11a	12.7	0.0	38.1	0.0
DkNFR07a	0.0	0.0	2.2	16.9
EjNMR11a	0.0	0.0	12.9	15.6
FpNMR14a	0.0	0.0	0.0	0.0
GzNML11a	38.8	64.0	47.1	42.5
HjNMR08a	0.0	0.0	0.0	0.0
LsNMR09a	0.0	9.1	58.8	24.7
MwNMR06a	36.3	0.0	82.7	48.9
NvNFR11a	11.2	10.9	67.4	71.4
OsNFR11a	0.0	21.1	26.7	16.9
SmNFR07a	35.6	0.0	35.7	0.0
TmNMR04a	0.0	4.0	0.0	0.0
UkNFR07a	0.0	0.0	53.1	25.8
VtNML11a	0.0	0.0	0.0	0.0
Mean	11.1	7.7	33.0	22.4
Standard Deviation	16.0	16.7	29.4	25.6

Subject Name	RBicep-RTricep	LBicep-LTricep	RTrap-RDelt	LTrap-LDelt
IcPML14p	0.0	0.0	0.0	0.0
# stdev's from mean	0.7	0.5	1.1	0.9
PaPFL10p	12.0	0.0	17.0	0.0
# stdev's from mean	-0.1	0.5	0.5	0.9
RcPFR06p	20.9	0.0	0.0	0.0
# stdev's from mean	-0.6	0.5	1.1	0.9
WrPFR13p	0.0	0.0	0.0	0.0
# stdev's from mean	0.7	0.5	1.1	0.9
IcPML14n	0.0	0.0	0.0	0.0
# stdev's from mean	0.7	0.5	1.1	0.9
PaPFL10n	0.0	0.0	32.6	13.7
# stdev's from mean	0.7	0.5	0.0	0.3
RcPFR06n	36.1	0.0	0.0	0.0

# stdev's from mean	-1.6	0.5	1.1	0.9
WrPFR13n	0.0	25.5	0.0	32.9
# stdev's from mean	0.7	-1.1	1.1	-0.4

Percentage

Subject Name	Dominant Bicep	Non-Dominant Bicep	Dominant Tricep	Non-Dominant Tricep	Dominant Trapezius	Non-Dominant Trapezius	Dominant Deltoid	Non-Dominant Deltoid	Dominant Eresp	Non-Dominant Eresp
AmNFR11a	36.9	90.0	82.5	15.1	61.9	88.6	83.1	81.9	59.8	82.0
CcNFR11a	7.9	0.0	61.9	98.3	27.6	31.1	70.7	10.3	48.3	12.4
DkNFR07a	0.0	0.0	11.7	33.7	11.7	57.8	8.8	22.9	28.2	20.9
EjNMR11a	0.0	0.0	29.8	0.0	19.3	82.0	85.5	12.8	17.2	16.5
FpNMR14a	10.9	0.0	20.2	0.0	30.5	17.5	0.0	23.7	0.0	18.6
GzNML11a	40.9	63.4	67.2	61.6	61.0	68.3	42.0	48.6	65.0	58.8
HjNMR08a	0.0	0.0	0.0	0.0	12.2	39.7	0.0	0.0	0.0	17.5
LsNMR09a	0.0	27.2	59.3	64.0	64.7	76.7	94.1	47.3	20.3	0.0
MwNMR06a	46.4	0.0	65.5	45.4	81.7	33.3	67.6	51.7	28.3	37.0
NvNFR11a	55.5	53.2	24.0	42.7	71.8	92.5	80.9	66.6	23.2	0.0
OsNFR11a	14.7	12.9	0.0	61.0	14.5	91.9	49.2	15.6	0.0	27.2
SmNFR07a	81.9	0.0	44.3	42.2	73.7	0.0	26.3	0.0	0.0	0.0
TmNMR04a	0.0	5.6	68.4	67.6	0.0	0.0	17.7	0.0	7.8	7.3
UkNFR07a	0.0	0.0	35.8	35.8	40.0	33.8	58.9	33.8	11.0	20.1
VtNML11a	16.8	0.0	0.0	0.0	25.8	48.5	0.0	0.0	31.1	26.9
Mean	20.8	16.8	38.0	37.8	39.8	50.8	45.7	27.7	22.7	23.0
Stdev	25.6	28.8	28.3	30.1	26.9	31.9	34.5	26.3	21.4	22.4

Subject Name	Dominant Bicep	Non-Dominant Bicep	Dominant Tricep	Non-Dominant Tricep	Dominant Trapezius	Non-Dominant Trapezius	Dominant Deltoid	Non-Dominant Deltoid	Dominant Eresp	Non-Dominant Eresp
lcPML 14p	0.0	13.2	78.9	0.0	34.3	21.8	0.0	0.0	47.2	70.6
# stdvs from mean	0.8	0.1	-1.4	1.3	0.2	0.9	1.3	1.1	-1.1	-2.1
PaPFL 10p	10.6	0.0	88.0	48.2	45.3	0.0	7.7	0.0	12.3	66.6
# stdvs from mean	0.4	0.6	-1.8	-0.3	-0.2	1.6	1.1	1.1	0.5	-1.9
RcPFR 06p	13.3	0.0	63.6	7.2	0.0	0.0	0.0	0.0	38.9	5.9
# stdvs from mean	0.3	0.6	-0.9	1.0	1.5	1.6	1.3	1.1	-0.8	0.8
WrPFR 13p	0.0	0.0	64.7	91.0	0.0	11.6	0.0	0.0	11.0	5.5
# stdvs from mean	0.8	0.6	-0.9	-1.8	1.5	1.2	1.3	1.1	0.5	0.8
lcPML 14n	0.0	0.0	96.5	0.0	0.0	0.0	0.0	0.0	0.0	0.0
# stdvs from mean	0.8	0.6	-2.1	1.3	1.5	1.6	1.3	1.1	1.1	1.0
PaPFL 10n	23.4	0.0	0.0	8.9	72.4	25.1	23.6	17.0	7.0	39.7
# stdvs from mean	-0.1	0.6	1.3	1.0	-1.2	0.8	0.6	0.4	0.7	-0.7
RCPFR 06n	51.9	0.0	53.4	0.0	4.9	11.6	0.0	0.0	0.0	0.0
# stdvs from	-1.2	0.6	-0.5	1.3	1.3	1.2	1.3	1.1	1.1	1.0

mean										
WrPFR 13n	0.0	76.3	100.0	33.6	0.0	24.4	22.5	16.8	0.0	0.0
# stdvs from mean	0.8	-2.1	-2.2	0.1	1.5	0.8	0.7	0.4	1.1	1.0

Walking Task

Symmetry

Subject Name	Bicep	Tricep	Trapezius	Deltoids	Eresp
AmNFR11a	100.0	68.2	26.3	40.4	95.2
BgNMR10a	100.0	100.0	59.3	66.2	34.8
CcNFR11a	100.0	100.0	54.8	100.0	85.9
DkNFR07a	100.0	89.7	57.5	75.8	76.8
EjNMR11a	100.0	100.0	100.0	100.0	32.8
FpNMR14a	100.0	100.0	100.0	100.0	73.2
GzNML11a	91.6	71.1	47.9	100.0	63.7
HjNMR08a	100.0	47.5	62.5	87.8	39.9
LsNMR09a	88.9	77.8	79.7	93.4	60.0
MwNMR06a	54.9	90.7	87.4	78.2	49.0
NvNFR11a	73.5	100.0	67.4	51.2	50.7
OsNFR11a	94.1	89.7	93.5	81.8	60.8
SmNFR07a	100.0	19.4	91.3	87.4	72.6
TmNMR04a	86.5	62.8	79.2	100.0	59.9
UkNFR07a	78.2	100.0	81.3	87.6	76.2
VtNML11a	100.0	72.9	67.3	41.5	26.7
Mean	91.7	80.6	72.2	80.7	59.9
Standard Deviation	13.0	23.1	20.5	20.7	19.8

Subject Name P	Bicep P	Tricep P	Trapezius P	Deltoids P	Eresp P
IcPML14p	87.2	95.4	65.3	66.4	64.1
# stdev's from mean	0.4	-0.6	0.3	0.7	-0.2
PaPFL10p	100.0	75.9	72.9	63.5	36.8
# stdev's from mean	-0.6	0.2	0.0	0.8	1.2
RcPFR06p	100.0	52.8	73.4	91.6	69.0
# stdev's from mean	-0.6	1.2	-0.1	-0.5	-0.5
WrPFR13p	100.0	78.6	69.7	43.8	78.3

# stdev's from mean	-0.6	0.1	0.1	**1.8**	-0.9
IcPML14n	100.0	100.0	61.5	62.4	56.1
# stdev's from mean	-0.6	-0.8	0.5	0.9	0.2
PaPFL10n	91.9	47.0	49.7	84.3	31.3
# stdev's from mean	0.0	1.5	1.1	-0.2	1.4
RCPFR06n	100.0	100.0	65.1	100.0	64.0
# stdev's from mean	-0.6	-0.8	0.3	-0.9	-0.2
WrPFR13n	100.0	100.0	84.5	73.0	73.1
# stdev's from mean	-0.6	-0.8	-0.6	0.4	-0.7

Co-Contraction

Subject Name	Dominant Bicep-Tricep	Non-Dominant Bicep-Tricep	Dominant Trapezius-Deltoid	Non-Dominant Trapezius-Deltoid
AmNFR11a	0.0	0.0	19.1	5.6
BgNMR10a	0.0	0.0	38.6	55.7
CcNFR11a	0.0	0.0	0.0	0.0
DkNFR07a	0.0	0.0	44.1	29.3
EjNMR11a	0.0	0.0	0.0	0.0
FpNMR14a	0.0	0.0	0.0	0.0
GzNML11a	0.0	0.0	49.3	0.0
HjNMR08a	0.0	0.0	15.1	0.0
LsNMR09a	0.0	0.0	38.1	0.0
MwNMR06a	0.0	0.0	21.7	30.6
NvNFR11a	0.0	0.0	56.7	24.9
OsNFR11a	0.0	0.0	0.0	0.0
SmNFR07a	0.0	0.0	0.0	16.5
TmNMR04a	1.4	0.0	0.0	0.0
UkNFR07a	0.0	0.0	0.0	0.0
VtNML11a	0.0	0.0	37.2	62.6
Mean	0.1	0.0	20.0	14.1
Standard Deviation	0.3	0.0	20.9	20.9

Subject Name P	Dominant Bicep-Tricep P	Non-Dominant Bicep-Tricep P	Dominant Trapezius-Deltoid P	Non-DominantTrapezius-Deltoid P

IcPML14p	0.0	0.0	38.1	27.3
# stdev's from mean	0.3	0.3	-0.9	-0.6
PaPFL10p	0.0	0.0	0.0	41.5
# stdev's from mean	0.3	0.3	1.0	-1.3
RcPFR06p	0.0	0.0	9.0	9.8
# stdev's from mean	0.2	0.3	0.5	0.2
WrPFR13p	*0.0*	*0.0*	*38.9*	*43.7*
# stdev's from mean	*0.3*	*0.3*	*-0.9*	*-1.4*
IcPML14n	0.0	0.0	31.2	22.6
# stdev's from mean	0.3	-0.2	-0.5	-0.4
PaPFL10n	0.0	0.0	11.2	0.0
# stdev's from mean	0.3	0.3	0.4	0.7
RCPFR06n	0.0	0.0	0.0	0.0
# stdev's from mean	0.3	0.3	1.0	0.7
WrPFR13n	0.0	0.0	25.8	0.0
# stdev's from mean	0.2	0.3	-0.3	0.7

Percentage

Subject Name	Dominant Bicep	Non-Dominant Bicep	Dominant Tricep	Non-Dominant Tricep	Dominant Trapezius	Non-Dominant Trapezius	Dominant Deltoid	Non-Dominant Deltoid	Dominant Eresp	Non-Dominant Eresp
AmNFR11a	0.0	0.0	25.8	0.0	41.3	89.4	59.7	5.0	4.8	0.0
BgNMR10a	0.0	0.0	0.0	0.0	46.2	69.3	52.7	39.3	19.1	58.7
CcNFR11a	0.0	0.0	0.0	0.0	0.0	41.4	0.0	0.0	18.6	11.1
DkNFR07a	0.0	0.0	0.0	16.2	18.6	54.4	24.3	45.6	39.2	74.6
EjNMR11a	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	11.5	72.4
FpNMR14a	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	30.4	12.6

GzNML 11a	0.0	8.4	14.9	6.4	18.4	41.4	9.8	0.0	33.2	31.8
HjNMR 08a	0.0	0.0	53.4	0.0	37.4	36.4	6.9	0.0	40.0	100.0
LsNMR 09a	11.1	0.0	0.0	22.8	10.6	18.2	27.9	23.1	47.8	14.9
MwNM R06a	0.0	18.9	9.3	0.0	93.1	81.6	20.2	26.4	39.6	73.1
NvNFR1 1a	22.6	0.0	0.0	12.7	28.8	59.8	49.8	25.2	53.3	25.5
OsNFR1 1a	0.0	0.0	10.3	8.7	6.5	29.5	18.2	0.0	15.3	23.9
SmNFR 07a	0.0	0.0	80.6	0.0	94.2	91.9	0.0	15.1	22.8	26.2
TmNM R04a	13.5	0.0	27.6	12.6	70.9	63.4	0.0	0.0	71.3	64.1
UkNFR0 7a	0.0	7.5	0.0	0.0	19.7	53.1	12.4	0.0	56.5	45.8
VtNML 11a	0.0	0.0	27.4	0.0	51.3	68.5	37.1	88.5	52.4	42.8
Mean	**2.9**	**2.2**	**15.6**	**5.0**	**33.6**	**49.9**	**19.9**	**16.8**	**34.7**	**42.4**
Standard Deviation	**6.7**	**5.2**	**23.2**	**7.4**	**31.1**	**28.4**	**20.4**	**24.6**	**18.6**	**28.7**

Subject Name P	**Dominant Bicep P**	**Non-Dominant Bicep P**	**Dominant Tricep P**	**Non-Dominant Tricep P**	**Dominant Trapezius P**	**Non-Dominant Trapezius P**	**Dominant Deltoid P**	**Non-Dominant Deltoid P**	**Dominant Eresp P**	**Non-Dominant Eresp P**
IcPML 14p	0.0	12.8	0.0	4.6	27.7	58.6	32.7	33.1	74.2	64.5
# stdev's from mean	0.4	**-2.0**	0.7	0.1	0.2	-0.3	-0.6	-0.7	**-2.1**	-0.8
PaPFL 10p	0.0	0.0	0.0	24.1	16.4	24.6	0.0	42.7	10.6	78.4
# stdev's from mean	0.4	0.4	0.7	**-2.6**	0.6	0.9	1.0	-1.1	1.3	-1.3
RCPFR 06p	0.0	0.0	47.2	0.0	94.3	77.0	8.4	7.6	24.5	8.6

# stdev's from mean	0.4	0.4	-1.4	0.7	**-2.0**	-1.0	0.6	0.4	0.6	1.2
WrPFR 13p	0.0	0.0	0.0	9.0	16.8	27.2	17.9	52.3	34.2	27.6
# stdev's from mean	0.4	0.4	0.7	-0.5	0.5	0.8	0.1	-1.4	0.0	0.5
IcPML 14n	0.0	0.0	0.0	0.0	27.8	51.0	28.8	12.0	65.9	49.0
# stdev's from mean	0.4	0.4	0.7	0.7	0.2	0.0	-0.4	0.2	-1.7	-0.2
PaPFL 10n	18.2	0.0	6.3	53.0	77.6	24.4	10.0	0.0	37.6	91.9
# stdev's from mean	**-2.3**	0.4	0.4	**-6.5**	-1.4	0.9	0.5	0.7	-0.2	-1.7
RCPFR 06n	0.0	0.0	0.0	0.0	20.5	58.8	0.0	0.0	67.3	39.0
# stdev's from mean	0.4	0.4	0.7	0.7	0.4	-0.3	1.0	0.7	-1.7	0.1
WrPFR 13n	0.0	0.0	0.0	0.0	14.2	24.6	27.0	0.0	42.3	45.5
# stdev's from mean	0.4	0.4	0.7	0.7	0.6	0.9	-0.3	0.7	-0.4	-0.1

Swing Task

Symmetry

Subject Name	Bicep	Tricep	Trapezius	Deltoids	Eresp
AmNFR11a	73.6	73.9	73.7	59.0	100.0
BgNMR10a	65.7	89.9	82.7	81.4	46.8

CcNFR11a	85.8	88.8	88.0	91.5	82.9
DkNFR07a	81.4	59.5	63.0	62.5	100.0
EjNMR11a	88.6	85.1	81.2	79.6	36.5
FpNMR14a	79.2	86.3	70.5	82.2	69.8
GzNML11a	84.8	96.0	87.2	91.6	62.7
HjNMR08a	63.6	84.2	68.8	78.8	73.3
LsNMR09a	69.5	84.1	85.5	66.6	89.9
MwNMR06a	72.6	81.3	81.4	51.8	68.2
NvNFR11a	73.5	48.7	63.3	63.6	71.6
OsNFR11a	86.4	77.6	68.1	70.0	27.5
SmNFR07a	70.2	81.1	72.0	74.6	81.7
TmNMR04a	77.1	39.6	95.9	77.9	59.4
UkNFR07a	84.9	85.3	61.8	83.2	85.6
VtNML11a	80.3	79.3	63.2	64.5	100.0
Mean	77.3	77.5	75.4	73.7	72.2
Standard Deviation	7.8	15.3	10.7	11.6	21.9

Subject Name	**Bicep**	**Tricep**	**Trapezius**	**Deltoids**	**Eresp**
IcPML14p	72.4	40.4	89.8	51.7	51.5
# stdev's from mean	0.6	2.4	-1.4	1.9	0.9
PaPFL10p	50.7	37.0	16.2	31.9	84.4
# stdev's from mean	3.4	2.6	5.5	3.6	-0.6
RcPFR06p	31.7	49.0	43.3	56.3	88.4
# stdev's from mean	5.9	1.9	3.0	1.5	-0.7
WrPFR13p	50.8	64.6	80.7	51.4	59.8
# stdev's from mean	3.4	0.8	-0.5	1.9	0.6
IcPML14n	65.0	34.1	73.9	55.5	41.9
# stdev's from mean	1.6	2.8	0.1	1.6	1.4
PaPFL10n	69.2	36.2	79.7	42.9	62.8
# stdev's from mean	1.0	2.7	-0.4	2.7	0.4
RCPFR06n	67.8	43.1	70.0	30.2	90.5
# stdev's from mean	1.2	2.2	0.5	3.8	-0.8
WrPFR13n	57.9	68.6	83.1	69.7	58.6
# stdev's from mean	1.6	2.8	0.1	1.6	1.4

Co-Contraction

Subject Name	RBicep-RTricep	LBicep-LTricep	RTrapezius-RDeltoid	LTrapezius-LDeltoid
AmNFR11a	29.4	22.2	41.8	48.3
BgNMR10a	28.1	40.7	50.5	45.3
CcNFR11a	11.3	18.2	21.2	24.0
DkNFR07a	74.6	55.9	48.2	50.5
EjNMR11a	18.0	12.0	27.8	28.4
FpNMR14a	35.1	15.6	45.2	49.3
GzNML11a	18.9	0.3	25.9	21.7
HjNMR08a	36.2	70.1	42.7	68.8
LsNMR09a	51.9	48.7	50.5	33.1
MwNMR06a	41.6	37.9	26.7	21.8
NvNFR11a	48.6	46.4	49.1	60.3
OsNFR11a	19.1	24.8	41.5	32.5
SmNFR07a	52.5	47.9	56.3	49.4
TmNMR04a	52.9	30.5	97.4	76.3
UkNFR07a	59.5	52.4	52.1	47.7
VtNML11a	38.6	40.9	24.0	35.7
Mean	38.5	35.3	43.8	43.3
Standard Deviation	17.5	18.6	18.3	16.3

Subject Name	RBicep-RTricep	LBicep-LTricep	RTrapezius-RDeltoid	LTrapezius-LDeltoid
IcPML14p	80.3	47.2	71.0	58.4
# stdev's from mean	-2.4	-0.6	-1.5	-0.9
PaPFL10p	28.7	50.1	28.8	19.5
# stdev's from mean	0.6	-0.8	0.8	1.5
RcPFR06p	76.2	17.0	48.6	51.2
# stdev's from mean	-2.2	1.0	-0.3	-0.5
WrPFR13p	37.0	32.3	29.8	40.4
# stdev's from mean	0.1	0.2	0.8	0.2
IcPML14n	38.0	52.1	29.2	32.7
# stdev's from mean	0.0	-0.9	0.8	0.7
PaPFL10n	29.9	45.1	31.7	36.1
# stdev's from mean	0.5	-0.5	0.7	0.4
RCPFR06n	50.9	55.2	32.9	70.2

# stdev's from mean	-0.7	-1.1	0.6	-1.6
WrPFR13n	29.8	43.2	21.4	38.7
# stdev's from mean	0.5	-0.4	1.2	0.3

Percentage

Subject Name	Right Bicep	Left Bicep	Right Tricep	Left Tricep	Right Trapezius	Left Trapezius	Right Deltoid	Left Deltoid	Right Eresp	Left Eresp
AmNFR11a	70.0	79.5	55.4	75.8	52.3	64.1	53.2	39.9	0.0	0.0
BgNMR10a	95.4	70.4	76.5	76.0	63.1	74.1	69.4	80.5	16.2	52.9
CcNFR11a	96.0	83.5	89.4	94.4	67.5	69.3	88.5	80.0	10.2	10.1
DkNFR07a	80.8	80.6	15.8	49.4	48.8	68.9	29.2	54.5	0.0	0.0
EjNMR11a	86.8	96.9	83.4	89.3	66.0	71.0	84.5	77.3	5.8	69.3
FpNMR14a	79.2	100.0	81.8	84.4	66.0	78.9	70.2	58.8	7.2	23.0
GzNML11a	84.8	100.0	96.3	99.7	67.6	66.8	91.2	87.7	9.1	41.4
HjNMR08a	52.6	69.8	17.8	7.0	55.3	82.5	20.9	14.3	27.7	12.8
LsNMR09a	42.8	68.2	70.7	69.3	43.1	54.7	64.7	41.4	14.7	9.5
MwNMR06a	70.1	87.9	70.4	66.8	67.1	72.3	46.0	94.2	12.6	24.0
NvNFR11a	85.8	68.8	59.4	65.8	58.5	23.5	77.4	62.4	14.2	18.9
OsNFR11a	96.0	87.2	78.6	84.0	56.4	81.3	83.7	60.1	10.3	82.1
SmNFR07a	46.4	56.4	88.8	73.4	60.5	75.1	39.7	41.4	10.0	15.0
TmNMR04a	52.4	35.6	8.4	64.2	97.4	98.5	0.0	22.1	57.2	84.8
UkNFR07a	81.3	86.2	55.0	61.4	55.1	61.7	34.2	37.4	0.0	14.4
VtNML11a	58.2	66.4	67.5	71.3	83.1	58.3	88.4	76.1	0.0	0.0
Mean	**73.7**	**77.3**	**63.5**	**70.8**	**63.0**	**68.8**	**58.8**	**58.0**	**12.2**	**28.6**
Standard Deviation	**18.1**	**17.0**	**27.3**	**21.4**	**13.1**	**16.0**	**27.9**	**23.7**	**14.1**	**28.7**

Subject Name	Right Bicep	Left Bicep	Right Tricep	Left Tricep	Right Trapezius	Left Trapezius	Right Deltoid	Left Deltoid	Right Eresp	Left Eresp
IcPML14	73.9	61.7	26.4	55.7	89.8	100.0	23.0	41.6	52.7	4.5

p										
# stdev's from mean	0.0	0.9	1.4	0.7	**-2.1**	**-1.9**	1.3	0.7	**-2.9**	0.8
PaPFL10 p	99.2	49.9	72.1	9.1	100.0	16.2	71.2	3.3	0.0	15.6
# stdev's from mean	-1.4	1.6	-0.3	**2.9**	**-2.8**	**3.3**	-0.4	**2.3**	0.9	0.5
RcPFR06 p	73.1	11.0	46.7	9.7	65.6	27.0	35.7	63.0	5.7	5.9
# stdev's from mean	0.0	**3.9**	0.6	**2.9**	-0.2	**2.6**	0.8	-0.2	0.5	0.8
WrPFR13 p	62.5	68.7	65.1	51.2	55.6	65.8	83.6	35.0	49.5	34.2
# stdev's from mean	0.6	0.5	-0.1	0.9	0.6	0.2	-0.9	1.0	**-2.6**	-0.2
IcPML14 n	89.3	58.1	70.8	11.4	58.8	73.6	61.2	88.2	67.7	10.7
# stdev's from mean	-0.9	1.1	-0.3	**2.8**	0.3	-0.3	-0.1	-1.3	**-3.9**	0.6
PaPFL10 n	74.6	86.5	56.4	55.2	78.8	65.2	66.5	41.6	20.8	45.3
# stdev's from mean	0.0	-0.5	0.3	0.7	-1.2	0.2	-0.3	0.7	-0.6	-0.6
RCPFR06 n	75.5	55.2	56.9	0.0	62.3	70.2	69.8	0.0	2.1	11.6
# stdev's from mean	-0.1	1.3	0.2	**3.3**	0.1	-0.1	-0.4	**2.4**	0.7	0.6
WrPFR13 n	72.2	37.2	69.0	63.1	73.2	66.2	67.2	66.8	50.6	26.2
# stdev's from mean	0.1	**2.4**	-0.2	0.4	-0.8	0.2	-0.3	-0.4	**-2.7**	0.1

Curriculum Vitae

Candidate's full name: Carly Genn

Universities attended (with dates and degrees obtained):

2008-2011 University of New Brunswick: MSc (Mechanical Engineering)

2003 – 2007 University of Guelph: BSc (Biological Engineering)

Conference Presentations:

Genn, C., et al. EMG during swinging in children with unilateral below elbow limb loss; Wearing a prosthesis versus not. Proceedings of ACPOC Conference, 2011.

Genn, C., et al. Congenital upper limb patient history at the institute of biomedical engineering. Proceedings of ACPOC Conference, 2009; 17-18.