MODIFICATIONS AND UPPER EXTREMITY ORTHOTICS FOR THE LOFSTRAND CRUTCH

BY

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THESIS

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ABSTRACT

According to the latest National Institute on Disability and Rehabilitation Research (NIDRR) Mobility Device Report, there are an estimated 566,000 persons who use crutches in the US. During Lofstrand crutch mobility, irregular loads are placed upon the upper extremities during quadrupedal gait. During self-load bearing while walking with Lofstrand crutches, the arms can experience periodic loads every 1 s of up to 50% of body weight for durations as short as 340 ms. Excessive loads and motion of the wrist increase chances for carpal tunnel syndrome. Lofstrand crutch users would clearly benefit by having these upper extremity demands reallocated or supported by an orthosis to allow for longer-term ambulation, reduction in pain, and injury avoidance.

This thesis contains two studies: the design of a passive orthosis for Lofstrand crutch gait, and the use of a pneumatic pump as an energy harvesting device. An orthotic attachment for the Lofstrand crutch was developed, in order to reduce wrist extension and redirect loads from the carpal tunnel region on the palm and toward the adductor pollicis. Pressure sensors were used at the handle of the crutch to locate and measure loads, while motion capture was used to calculate joint angles. Results show a decrease in the average force and mean pressure across a Lofstrand crutch handle when using the orthosis, although peak palmar pressures may be greater with the orthosis. Palmar load displacement toward the adductor pollicis was achieved.

There is motivation to extend this work by using soft robotic technology, which will be powered pneumatically. Therefore, a preliminary design for a pneumatic harvesting device that can accumulate pressure throughout the gait cycle was created and assessed. The harvesting device was a piston attached to the tip of the crutch. When compressed, the device stores the pneumatic energy into an accumulator. Data were collected using a pressure transducer. A mathematical model of pressure in the accumulator was developed in order to predict accumulated pressure as a function of effective volume of the piston and dead volume in the system. The model simulation results were compared to experimental values and ranged from 3.79-15.53 percent error. A second custom piston design to achieve higher stroke volume is also presented.

Bismillah; in the name of God.

ACKNOWLEDGEMENTS

I would like to thank my advisor, Elizabeth Hsiao-Wecksler, not only for her guidance and ideas throughout the stages of these projects, but for her eagerness and willingness to pursue this project from its onset. It has been an honor to be a part of the Human Dynamics and Controls Laboratory at the University of Illinois and I look forward to the amazing things to be accomplished there. I would also like to thank Brooke Slavens, Omid Jahanian, Alyssa Schnorenberg and all those who assisted at the University of Wisconsin Milwaukee for their expertise and technical assistance throughout this project. Mazharul Islam, Chenzhang Xiao, and Ye Lwin Oo among other plentiful lab-mates have been of great assistance in a variety of ways throughout this work and I thank them for their help and ideas. I would also like to thank my friends, family, and loved ones; especially my wife Shazia Siddiqi, and my parents, Julia and Mohammad Farooq. Without them, their love and support, none of this would be possible. Finally and most importantly, I thank, praise and glorify God for the blessings, honor, guidance, and knowledge that only He can bestow. Nothing can be accomplished without Him allowing it to be. God willing, this work will be of use, aid those in need, and allow peace, tranquility, and guidance to be spread throughout the world.

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1: INTRODUCTION & LITERATURE REVIEW

ABSTRACT

There were over 6 million crutch, cane, and walker users in the United States as of the year 2000 and over 500,000 of them use crutches (Table 1)[1]. Due to the nature of the crutch, there is repeated loading on the wrist, shoulder, and elbows. This thesis focuses on the use of the Lofstrand crutch, commonly known as the forearm crutch (Figure 1). Lofstrand crutch users are at a greater risk for developing injuries in their upper extremities. Repeated loading of the wrist and palmar region and high angles of deviation from the neutral wrist position can cause wrist strain, injury, and may lead to carpal tunnel syndrome [2]–[5].Populations who use Lofstrand crutches have repeated loading of the wrist and may have large deviations from the neutral wrist position and may experience loads at the wrist up to 50% of body weight [3]. Surgical treatment for carpal tunnel syndrome may become necessary if other intervention is not used. If a crutch user sustains an injury to the wrist, the quality of life experienced can be significantly decreased as they will no longer be able to use their crutches as a primary means of locomotion.

There are various commercial products available to assist with injury and crutch-gait. In order to determine effectiveness of any assistive support or crutch, one must understand the demands of different gait modes and their implications on gait kinematics. Therefore, a summary of crutch gait is presented further on. Hand braces and splints which keep ergonomic posture are found to be beneficial in treating carpal tunnel syndrome [5]. In order to improve crutch gait, a number of crutch designs have been developed. The benefits and drawbacks of these will be discussed. This thesis draws focus on the wrist and carpal tunnel syndrome (CTS) as prime examples of crutch mechanics and injury; however, it should be noted that there are other significant injuries that may occur at the shoulder, elbow, and upper back as well [1].

1.1 PATIENT POPULATIONS

This section will focus on the pathologies of patient populations of interest: crutch users. A focus on Carpal Tunnel Syndrome is used here as a primary example of common pain experienced by crutch users. The effects of this pain and factors contributing to pain will also be discussed.

1.1.1 Medical Conditions Experienced By Patient Populations

According to the latest National Institute on Disability and Rehabilitation Research (NIDRR) Mobility Device Report, there are an estimated 6.1 million users of walkers, crutches, canes, or other devices in the US as of 2000 [1]. Crutch use has been reported by 566,000 persons [1]. Cerebral palsy (CP), spinal cord injury (SCI), myelomeningocele (MM), and osteogenesis imperfecta (OI) are associated with these leading causes of assistive mobility device usage in children and adolescents [6]. Over all ages, the

leading conditions associated with the use of crutches are osteoarthritis, impairment, or loss of the lower extremities. Other conditions associated with crutches are included in the appendix (Table 2) [1]. There are many different medical conditions experienced by patient populations who use assistive devices; however, there is a lack of literature showing potential solutions to crutch-gait issues faced by these populations.

1.1.2 Carpal Tunnel Syndrome (CTS)

According to mayoclinic.org, there are more than 3 million cases of CTS in the United States every year and most cases occur after the age of 19 [7]. CTS occurs when the median nerve within the carpal tunnel becomes pressurized or squeezed (Figure 2). Although CTS does not have an agreed upon clinical definition, symptoms indicating CTS include tingling, itching numbness, frequent burning, and pain in the palm and/or fingers of the hand [8], [9]. Conservative measures such as splinting and cortical steroid injections may suffice in the treatment of CTS; however, strenuous use of the hand tends to aggravate symptoms. Symptoms may not become apparent until several hours after the activity [8]. This means that CTS may go unnoticed during an activity, thus not giving time to an individual to alter their behavior and avoid CTS.

If CTS is not prevented in its onset, according to Waring and Werner, post CTS treatments have no significant long term effect [10]. Therefore, preventative measures are the best course of action in the case of CTS. It has also been noted that there was a direct correlation between the preferred hand that typically holds the cane or crutch and the hand which carpal tunnel was developed [9], [10]. The chronic use of cane and crutch predisposes patients to the development of CTS. This further promotes preventative measures to take place in order to reduce the likelihood of CTS. With preventative measures, it may be possible to avoid the onset of carpal tunnel syndrome [7].

Factors that affect CTS and other forms of wrist injury include actions that place high loads on the wrist and palm region, as well as actions that involve repetitive movement of the wrist [2], [4], [8]. Chronic repetitive stresses have been linked to carpal instability [11]. Higher weight can lead to higher wrist pain frequency and severity [12]. It has been hypothesized that handedness may influence wrist pain rates; however, there were no findings which show differences in frequency of pain between dominant and non-dominant hands [12]. Factors other than wrist position and loading including metacarpophalangeal flexion and extension, forearm pronation and supination, certain finger postures and fingertip loading may also increase carpal tunnel pressure [13]. There are many complex anatomical factors that may affect CTS, a common and gross movement that seems to have a large effect on the onset of CTS is repetitive movement of the wrist. This means that persons who use crutches and place large repetitive loads on their wrists are constantly performing actions that may lead to CTS.

1.1.3 The Effects of Upper Extremity Pain on Quality of Life

Lofstrand crutches are generally prescribed for long term use. Persons who use crutches generally use them as their main mode of locomotion. Shoulder and/or wrist pain may develop from using these assistive devices. Persons with shoulder and wrist pain may face limitations in using crutches. While most individuals experiencing upper extremity (UE) pain are able to work around limitations that they may face, it was found that 28% of subjects tested with upper extremity pain reported limitations of independence and that the severity of pain experienced was significantly related to asking for help with daily activities [14]. Functional limitations of crutch users that may impact their quality of life include upright tasks and bending. Over all ages, 71.5 % of crutch users reported being unable or having difficulty walking ¼ mile, 61.7% reported similarly for standing 20 minutes, and 56.1% reported being unable or having difficulty in bending down (Table 3) [1]. Continuing to use crutches with upper extremity pain can become a hindrance and may increase the amount of pain experienced. Natural progression finds pediatric assistive device users often transitioning to using wheelchairs as they mature because their arms can no longer support their increasing body weight. This transition reduces mobility, fitness, independence, and quality of life. There is a reduction in quality of life for crutch users with UE pain. The reduction of the quality of life for crutch users motivates further studies on assistive devices to be utilized during gait.

1.2 EXISTING CRUTCH DESIGNS

Despite injury and other issues that may be faced with crutches, crutches provide greater independence and locomotion to the user, compared to not using crutches at all. To further increase the benefits of crutches, and minimize the risks and disadvantages, there have been several design variations to the crutch. There are advantages and disadvantages to each design, which is presented in this section. This section will stay contained to commercially available products within the broad scope of crutches; it is not comprehensive of all types of assistive crutches and is meant to give an overview of available options.

1.2.1 The Traditional Lofstrand Crutch

The Lofstrand crutch, commonly known as the forearm crutch, is often prescribed for long term use. There can be many pathological issues that lead to the recommendation of a Lofstrand crutch (Table 2). Although the traditional Lofstrand crutch allows patient populations to be upright and mobile, it may have adverse contributions such as, carpal tunnel syndrome, pain, and other injuries [1].

Traditional Lofstrand crutches are typically composed of a hollow aluminum shaft, rubber tip, a handle, and a plastic pliable cuff (Figure 2). In general the height of the crutch is adjustable as well as the distance from the handle to the cuff. The entire structure is rigid, with the exception of the cuff, which is often pliable around the forearm. There are different options of handle, tip, and cuff available for traditional Lofstrand crutches, which may have a range of benefits such as: padding, grip, and adjustability. Traditional aluminum adult Lofstrand crutches can support a weight limit of about 250 to

300 lbs. and are generally lighter weight than traditional auxiliary (under-arm) crutches. Modifications to the neck may exist for crutches that can carry larger loads. There are no classifications or categories of types of Lofstrand crutches; they are distinguished from auxiliary crutches by their cuff, which stabilizes the forearm during gait. They are also distinguishable by the angle at which the neck is tilted, which is approximately 15 degrees (Figure 1). In general, the crutch can be easily taken apart into three main components: cuff, mid shaft, and lower shaft. The cuff and lower shaft can generally be unattached via pin connectors and slid out of the main shaft, which allows for easy storage. The traditional Lofstrand crutch is generally more affordable than alternative Lofstrand crutches, which may have carbon fiber shafts and extra accessories. These alternative designs are presented in the next section.

1.2.2 Alternatives and Accessories to the Lofstrand Crutch

There are several variations of the traditional Lofstrand crutch. Most of the variations attempt to offer ergonomic supports, greater efficiency during gait, and/or greater adjustability.

Walkers and underarm crutches (also known as auxiliary crutches) are of the most commonly available alternatives to the Lofstrand crutch. Walkers are generally slower than other forms of upright assistive devices, and are also three times more commonly used than crutches. 78.1% of walker users are aged 65 and older, while 72.6% of crutch users are non-elderly [1]. The greater usage of walkers may be attributed to the greater usage rates of the elderly who use walkers for stability. Persons who use walkers may incur wrist and shoulder pain; however, the advantage of a walker is that the design allows the user greater stability during gait [15].

Auxiliary crutches are generally prescribed for shorter term injuries. However, the use of the auxiliary crutch is generally misused by applying too much pressure to the auxiliary support underneath the shoulder. This behavior may be due to fatigue of use with the auxiliary crutch or due to ease of use in this manner [16]. Extra padding is generally recommended under the arm for the auxiliary crutch.

Platform forearm crutches are used in a similar way to walkers; however, a platform crutch does not have four legs and has a resting area for the forearm (Figure 3 A) [17]. These are recommended for individuals who are unable to rest their body weight on their hands.

There are also crutches available that have extra adjustable parts, such as the SmartCRUTCH (SmartCRUTCH, USA),which can adjust the angle of the handle relative to the shaft (Figure 3 C) [17]. This design is meant to allow for comfort and ergonomic positioning of the upper extremities during gait. It should be noted that extra adjustability mechanisms may be difficult for certain pathological injuries and the angle should be determined by a physical therapist. The SmartCRUTCH design is intended to reduce palmar loads by redistributing loads to the forearm and allow the user to customize the forearm platform angle.

Accessories and spare parts can often be purchased separately. These include specialized tips, handles, cuffs, padding, as well as carrying cases [18]. Specialized tips are generally used for slippery surfaces

such as ice. Specialized handles and grips tend to have more padding and a curvature meant to align the wrist to an ergonomic position. Crutches may also be made of different materials such as carbon fiber to increase strength and reduce weight; however, these materials generally increase the price of the crutch. The author has not found any previous scientific data showing the effects of customized ergonomic grips or crutches with extra adjustable features, on UE loads during crutch gait. The author has not found any previous scientific data showing conclusive evidence on the effects of customized ergonomic grips on wrist posture. The lack of alternatives and accessories available for ergonomic wrist posture and palmar load distribution during crutch gait motivates a new orthotic design to support the wrist and redistribute palmar loads during crutch gait.

1.2.3 Spring-Loaded Crutches

Spring-loaded auxiliary crutches have been shown to exhibit small effects on gait velocity when compared to traditional auxiliary crutches [19]. When a subject uses spring-loaded crutches, there can be small changes in forward velocity during gait. Elastic potential energy is stored during the contact of the crutch tip and the ground. Some of the potential energy is converted to kinetic energy and transmitted to the user in a manner facilitating forward motion. This forward kinetic energy was found to be small and not affect preferred ambulation speed. The forward velocity gained as a result of the decompressing spring may be lost when the subject compresses the spring at the next gait cycle [19]. With spring-loaded crutches, stride time was increased and there was no significant differences found in stride length [20]. There may be slight increases in velocity during the gait cycle due a compression spring in a crutch; this increased velocity does not appear to affect preferred ambulation speeds.

When using spring loaded auxiliary crutches, a shock absorption effect takes place. It was found that people with disabilities may choose spring loaded crutches over traditional crutches due to the shock absorption [21]. Impulse of ground reaction forces of spring-loaded crutches were found to be reduced when comparing to traditional auxiliary crutches, although the peak ground reaction forces were larger [20]. Metabolic energy was not saved due to the use of spring-loaded crutches [22]. Due to the biomechanical advantage of spring-loaded crutches, it is likely that spring-loaded crutches may reduce overuse injury in crutch users [20]. This means that spring-loaded crutches are beneficial to crutch gait and spring-loaded mechanisms within crutch tips (such as energy harvesting devices) can provide extra benefit to crutch users.

Shock absorption and power assist crutches are also available options. These use spring actuation systems near the tip of the crutch to allow for a dampened impact and an assisted push during swing-through gait (Figure 3 B) [17]. Prior studies found no statistically significant differences in metabolic energy expenditure between spring-loaded crutches and standard ones; however, there were increased mechanical efficiencies and slightly higher peak velocities found when using spring-loaded crutches [19], [22]. The third chapter of this thesis will cover these types of crutches in more detail.

1.2.4 Pediatric Crutches

There are three main reasons that pediatric crutches are inherently different than adult crutches: their size, adjustability, and the lack of available alternatives. The size of pediatric crutches is the most defining feature as opposed to other crutches. Pediatric crutches tend to be smaller and hence, lighter in weight. The crutch shaft diameter tends to be smaller, which can lead to the maximum load of pediatric crutches to be smaller. Pediatric Lofstrand crutches are often less adjustable than adult crutches; they may have only two height settings and rigid cuffs, while adult crutches may have 10 height settings and pliable cuffs to adjust to the curvature of the forearm. There are relatively few available crutch models for the pediatric population when compared to adults; nearly all of the 'specialty' crutches are only available in adult sizes. This means that persons who use long-term pediatric crutches may be more susceptible to injury without the availability of specialized crutches to fit their needs.

1.3 UNDERSTANDING LOFSTRAND CRUTCH GAIT

In order to effectively design an orthotic attachment or any device to assist gait during crutch use, gait patterns must be understood and quantified. The process for doing so can be split into three main areas: understanding why different gait modes are used during crutch use, understanding how these modes can be improved and/or idealized in an ergonomic fashion, and understanding the forces and moments acting during gait. These three can, to some extent, be found in the literature and are presented below.

1.3.1 Different Gait Modes and Their Use

As opposed to underarm crutches, forearm crutches allow a user to adapt gait depending on the physical challenge presented. Personal preference and level of fatigue may also become factors when choosing a gait mode.

There are five types of Lofstrand crutch gaits: 4-point alternate gait, 2-point alternate gait (also known as reciprocal gait), 3-point alternate gait, swing-to gait, and swing through gait (Figure 4) [18]. 4-point alternate gait is a slow yet stable gait pattern which allows an individual to move each leg separately while alternating the crutch movement. Considerable weight may be placed on each foot. The sequence of 4-point alternate gait is: right crutch, left foot, left crutch, right foot. 2-point alternate gait requires more balance than 4-point, but is also slightly faster than 4-point gait. 2-point gait most closely resembles normal walking and its sequence is as follows: right crutch and left foot, left crutch and right foot. 3-point alternate gait requires arm strength to support significant amounts of body weight and maintain balance, but is more rapid than the above gait patterns. 3-point gait sequence is as follows: both crutches and the weaker leg move forward simultaneously; then the stronger extremity is moved forward while placing most of the body weight on the arms. Swing-to gait is more rapid than any of the

above gait patterns. For swing-to gait the movement of the legs is parallel and requires considerable upper body strength to support the body. Its sequence is as follows: bear weight on good leg (or legs); advance both crutches forward simultaneously, lean forward while swinging the body to a position even with the crutches. Swing-to gait is the natural progression to swing-through gait. Swing-through gait is the fastest gait and only differs from swing-to gait in that the feet land past the crutches with every stride. Its sequence is as follows: advance both crutches forward; lift both legs off the ground and swing forward landing in advance of the crutches; bring crutches forward rapidly [18].

During Lofstrand crutch-assisted gait, two types of gait patterns are often observed: reciprocal gait and swing-through gait. Swing-through gait places extremely large loads (>100% body weight) on the upper body between both arms of the user [6], [23]–[26]. There is an advantage to learning and utilizing several types of gait patterns. Different gait modes may bring about different levels of fatigue and combinations of muscles to perform. Once an individual becomes tired, they may change the gait type to still be mobile and expend less energy [18]. Users who prefer swing through gait place large loads on the upper extremities; this may lead to CTS and other wrist injury. The use of an orthotic device during swing through gait may assist in the prevention of CTS and other wrist injury.

1.3.2 Benefits of Having the Wrist in the Neutral Position During Crutch Gait

During crutch gait the wrist undergoes repetitive motions, relatively high loads, and does not remain near the neutral position [6], [27], [28]. The neutral position for the wrist is one of zero degrees flexion, extension, pronation, and supination. In one study, crutch handles with larger surface area were tested against normal sized handles [29]. The handles with larger surface area were found preferable by individuals; however, there was no statistical significance to show load differences in this study. Palmar loads were found to be highest in the adductor pollicis and carpal tunnel regions of the palm during this study [29]. Keeping the wrist near a neutral position while refraining from high loads or repetitive motions is the recommended dynamic situation to avoid wrist injury [2]–[4], [8], [30], [31].

1.3.3 Kinetics and Kinematics

Typical peak loads during crutch-assisted gait observed in the wrist, elbow, and shoulder can range from 5% of body weight for reciprocal gait to 50% for swing-through gait in one arm [23]–[26], [32]. These repetitive, high loads have been shown to lead to joint pain and injury, such as carpal tunnel syndrome, arthritis, or joint deformity [32]–[35]. Swing-through gait generally has higher peak loads in the shoulder, elbow, and wrist than reciprocal crutch gait. The anterior/posterior forces for the wrist were shown to be higher than the shoulder and elbow in any measured direction (Figure 4-Figure 6) [27], [28].

The shoulder was found to have a range of motion of 30 degrees, from about -5 to 25 degrees, between flexion and extension throughout a gait cycle. The elbow was shown to have a range of motion of 15-25 degrees between flexion and extension, while the wrist at most has a range of motion of about 17

degrees (Figure 7). However, it is notable that the wrist is in a consistent state of relatively high extension (-30 to -40 degrees) throughout the gait cycle. It can also be seen that the wrist has many minor deviations in flexion and extension throughout the gait cycle (Figure 7). Different pathologies may cause different kinematic behaviors during the gait cycle (Figure 8 and Figure 9) [27], [28]. The repetitive loading and high angles of extension for the wrist may lead to injury. If the repetitive loading and high angles of extension are prevented by an orthotic device, then injury may be prevented.

1.3.4 A Closer Look at the Wrist

The wrist was selected as the initial joint of study since joint ranges of motion observed during crutch use vary by only 5 to 17 degrees over a gait cycle. Furthermore, the forces experienced at the wrist may be higher than those experienced at other joints, while the wrists are always in a state of extension averaging between -30 to -40 degrees (Figure 7) [23], [27], [28], [32]. This consistent state of extension was previously shown to increase risk for CTS and other wrist injuries. It has also been found that increasing wrist extension increases the vertical loads experienced on the hand [30]. Further, repetitive motions involving wrist extension from assistive device use can result in injury [10]. Maintaining the wrist in neutral position, i.e. minimizing extension and flexion, has been recommended to prevent damage. Therefore, creating an orthosis that can change wrist configuration to reduce wrist extension while providing joint stability support during load bearing would be beneficial to any assistive mobility device user, who is susceptible to overuse injury and pain (those with acute limb injury or surgery, elderly, pediatric, or adult pathology populations).

1.4 POWER HARVESTING DEVICES

Energy harvesting devices use existing mechanical processes in order to store energy. Energy can be stored in various forms and can be used for a number of applications. This section will focus on devices that harvest energy from gait cycles and also give an overview of existing pneumatic pump and accumulator technology.

1.4.1 Existing Devices Harvesting Energy from Gait

A promising way to extract energy from persons is by using changes in kinetic and potential energy of body segments caused by their movement while walking; this is typically achieved by energy harvesting from the gait cycle [36], [37]. There are a variety of devices which harvest power from gait; these devices may use several types of mechanisms. Fluid power harvesting under foot bellows found significant pressure increases when placing bellows in the sole of a shoe; however, issues of storing the pressure required further investigation [38]. Mechanical gearing systems around the knee have also been shown to generate electricity [39]. Pneumatic cylinders and gas springs were found to be an effective form of transferring and storing energy in an orthotic device [40]. Several energy harvesting

devices are based on magnetic generators [41]. Magnetic generators for power harvesting during gait were proposed for low-power applications [41]. It has also been observed that energy harvesting through backpacks is a viable source of low power, especially in field work, with the use of piezoelectric actuators [42], [43]. These power harvesting devices focus on gait induced power harvesting. It has been shown that power harvesting through gait is feasible and effective; harvesting pneumatic energy from crutch gait should also be feasible and effective.

1.4.2 Pneumatic Pump and Accumulator Technology Pertaining to Crutch Gait

This sub-section reviews and analyzes the components and intent of existing pneumatic pump and accumulator technology pertaining to crutch gait. There are few literature items on pneumatic pump and accumulator technology pertaining to crutch gait, compared to walking gait. A 1980 patent describes a crutch which used a piston assembly that pressurizes a liquid-gas mixture in order to lengthen and shorten a crutch [44]. Another patent similarly describes a crutch that used compressed fluid to extend the length of a crutch to assist in sitting or standing positions [45].These patents lack an internal spring mechanism to return piston strokes. Furthermore, the intent of these designs was to adjust the height of a crutch using fluid power. Furthermore, a pneumatic assist device for gait restoration was incorporated into a customized crutch, which required strenuous pre-pumping before gait began [46]. Pneumatic foot pumps were used to inflate air bladders for a temporary prosthesis; however, these were not actuated during gait [47]. A hydraulic crutch was developed to assist in paraplegic crutch gait; however, the fluid was directly used for linear actuation and not stored [48]. The author has found no current ‘off the shelf’ piston assemblies that have the capability to spring return upon being loaded. Pneumatic cylinders for the purpose of harvesting pneumatic energy during Lofstrand crutch gait is an untapped source of energy that could allow for powered orthotic devices to be feasible and effective options during crutch gait.

A novel pneumatic energy harvesting device, actuated during crutch gait, via the use of an internal piston and elastomeric accumulator within the Lofstrand crutch is a proposed solution to storing pneumatic energy and powering pneumatic applications during crutch gait.

1.5 THESIS OVERVIEW

From the aforementioned information and motivation, two studies were conducted which are presented over the following three sections.

Chapter 2 focuses on a passive wrist orthosis for Lofstrand crutch users developed through the Human Dynamics and Controls Laboratory (HDCL) at the University of Illinois at Urbana-Champaign. The goal of this device was to allow crutch users to keep their wrist in an improved ergonomic position throughout the gait cycle, while distributing palmar loads away from the carpal tunnel and toward the adductor pollicis region, thus reducing the risk of CTS. The device design, validation, and data, as well as limitations of the study, are discussed in this chapter.

Chapter 3 analyzes the concept, design, and validation of a pneumatic energy harvesting device within the Lofstrand crutch. The system description and integration are discussed, as well as limitations of the

study and future work for the application and implementation of this system with a pneumatically actuated orthotic device.

Chapter 4 summarizes the findings of Chapters 2 and 3, while explaining possible future work for the over-arching project of using upper extremity orthotics to assist crutch gait.

Figures:

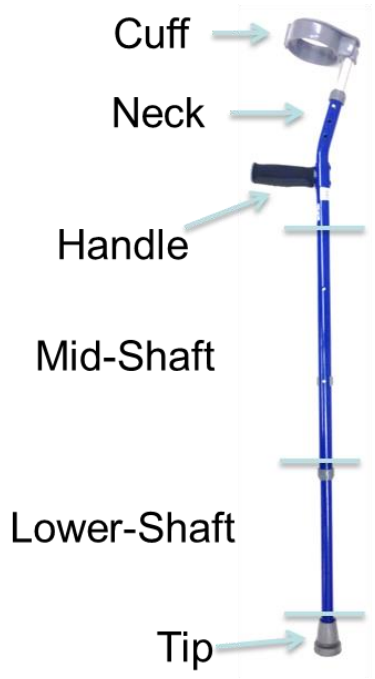

Figure 1: The traditional Lofstrand crutch

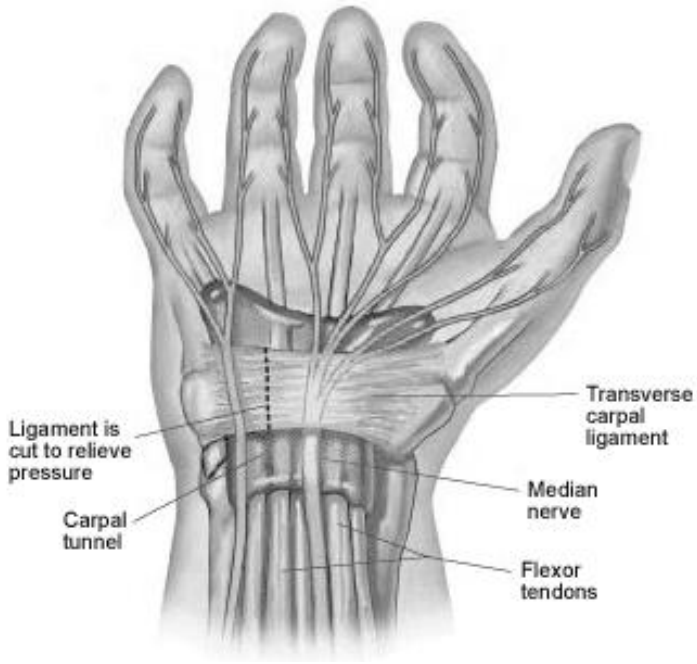

Figure 2: Carpal tunnel, median nerve and surrounding tissues. [49]

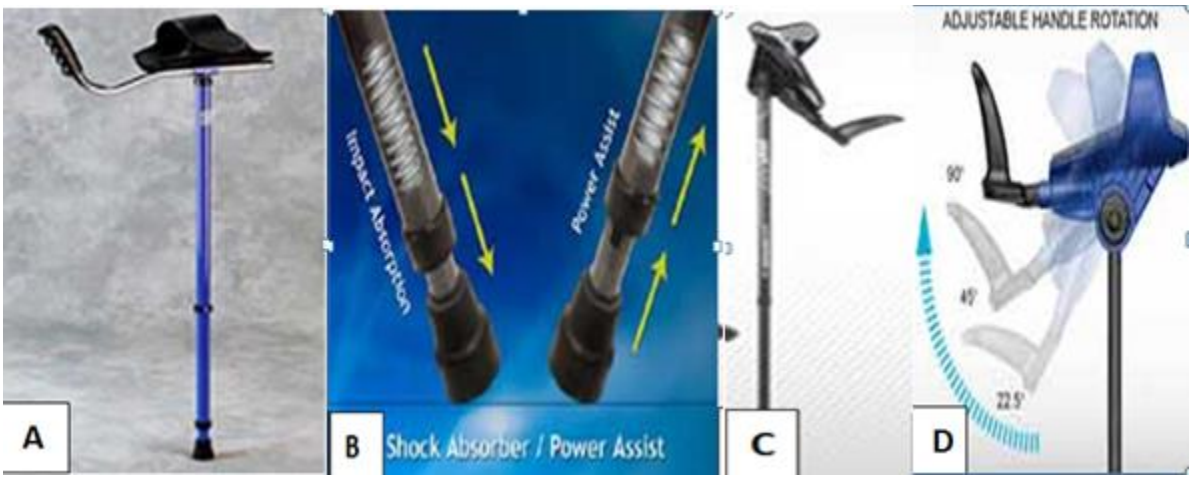

Figure 3: Alternative crutch designs. (A) Platform Crutch. (B) Power Assist Crutch. (C/D) SmartCRUTCH [50] **.**

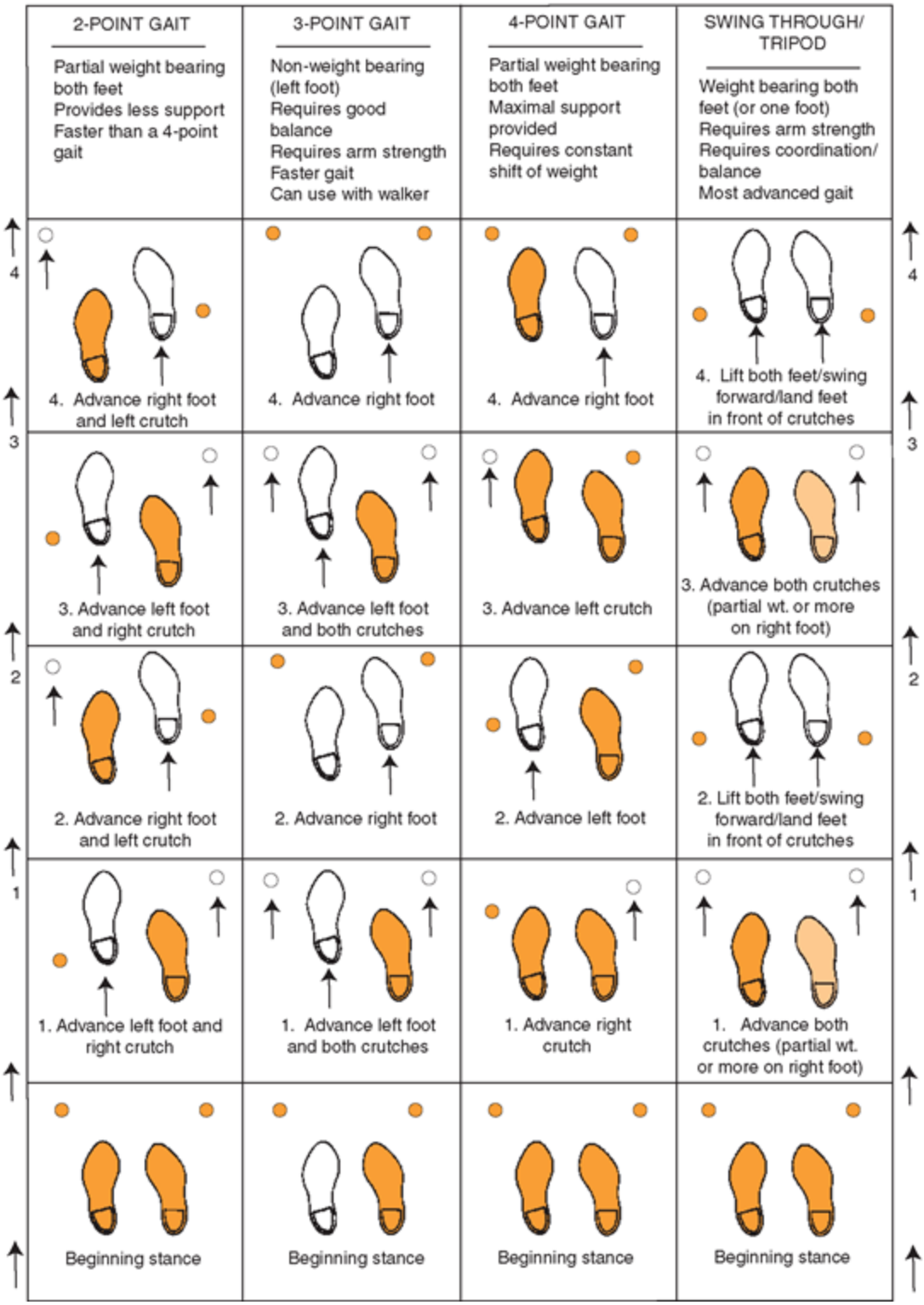

Figure 4: Types of crutch gait [51]

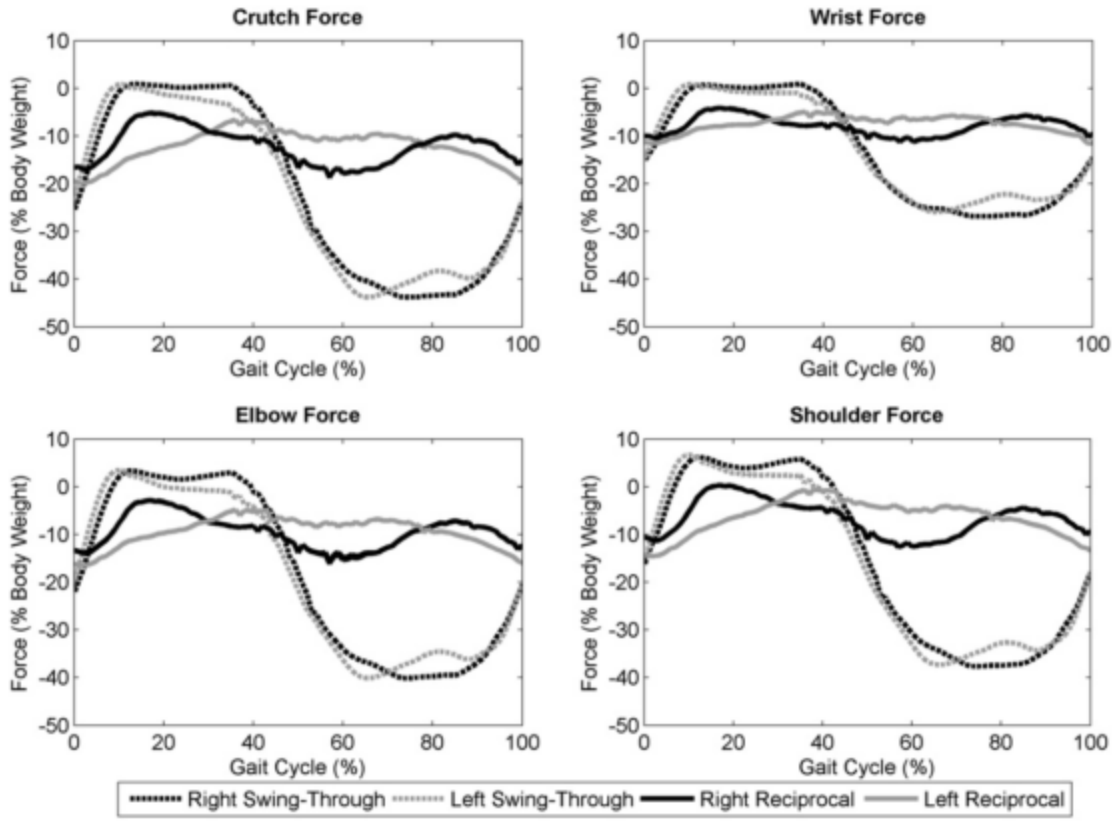

Figure 5: Force as %BW during Lofstrand crutch gait, showing swing through and reciprocal gait [27]**.**

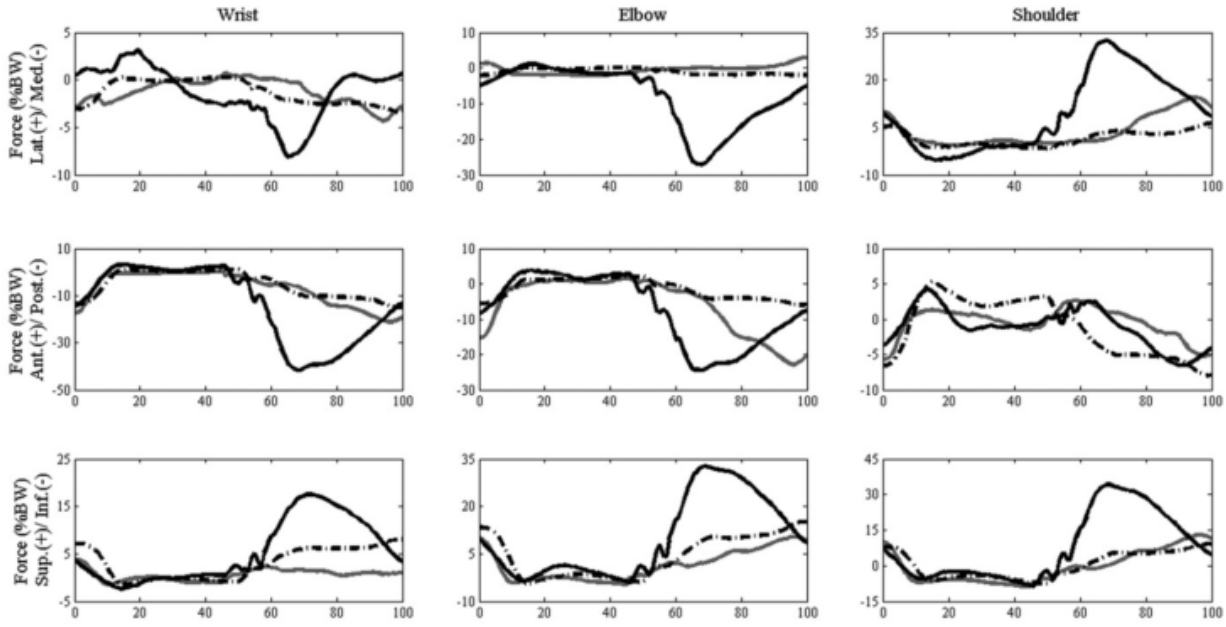

Figure 6: Forces along different planes as %BW for the wrist, elbow and shoulder during swing through gait [27]**.**

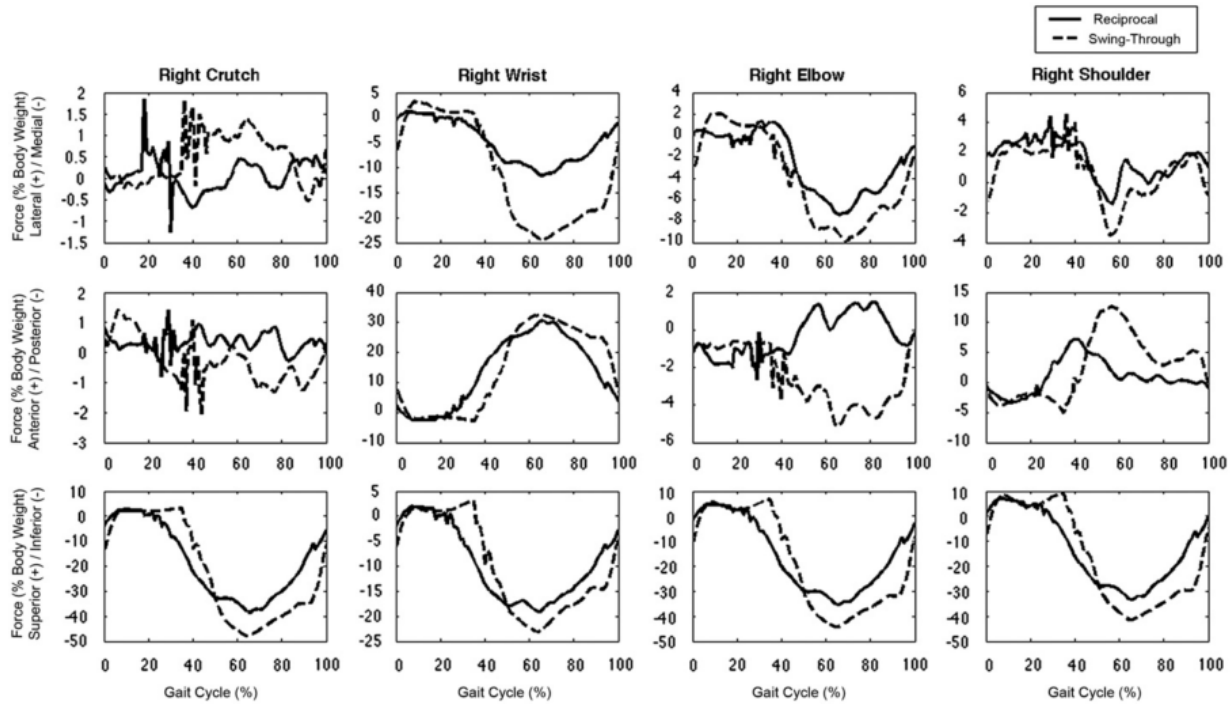

Figure 7: Forces along different planes and joints as %BW during Lofstrand crutch gait, showing swing through and reciprocal gait [6]**.**

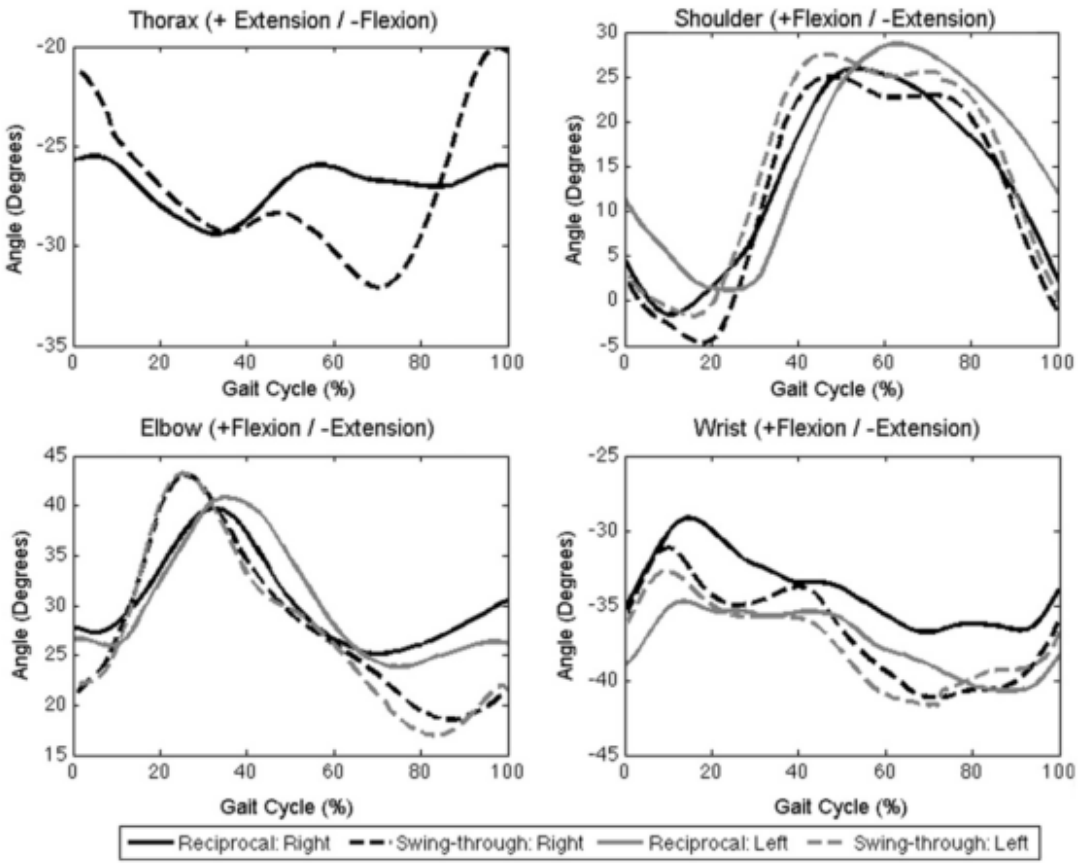

Figure 8: Flexion/Extension angles during swing through and reciprocal gait [27]

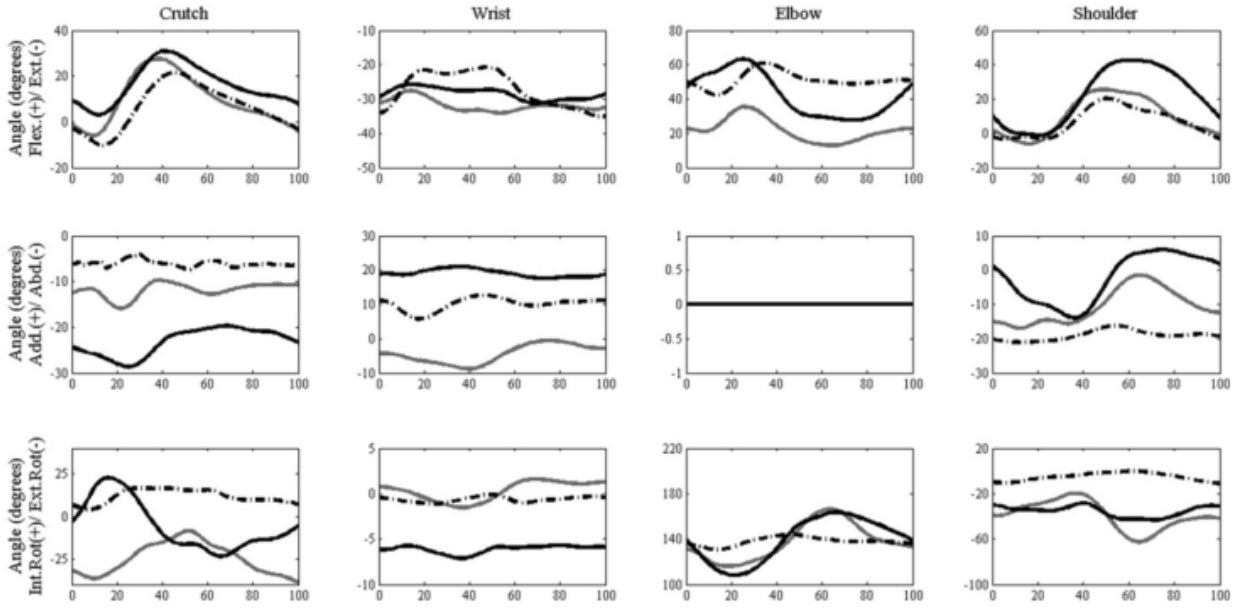

Figure 9: Flexion/Extension angles for different pathologies during swing through gait [27]**.**

Tables:

Table 1: Number of persons and proportion of population using mobility devices by age and device used [1].

	All Persons		Under 18		18–64		65 and over	
	Number (1000s)	Proportion (percent)	Number (1000s)	Proportion (percent)	Number (1000s)	Proportion (percent)	Number (1000s)	Proportion (percent)
Any mobility device	6,821	2.62	145	0.21	2,310	1.45	4,366	13.97
Wheelchair or scooter	1,679	0.64	88	0.12	658	0.41	933	2.99
Wheelchair	1,599	0.61	88	0.12	614	0.39	897	2.87
Manual wheelchair	1,503	0.58	79	0.11	560	0.35	864	2.76
Electric wheelchair	155	0.06	18	0.02	90	0.06	47	0.15
Scooter	142	0.05	0	0.00	78	0.05	64	0.21
Other mobility device	6,126	2.35	73	0.10	1,987	1.25	4,065	13.01
Cane	4,755	1.82	19 *	0.03	1,535	0.96	3,200	10.24
Crutches	566	0.22	36	0.05	375	0.24	155	0.50
Walker	1,820	0.70	27	0.04	373	0.23	1,421	4.55

*Standard error exceeds 30 percent of the estimate.

Table 2: Leading conditions associated with the use of crutches, all ages [1].

Condition	Persons (1000s)	Proportion of device users (%)
All conditions	492	100.00
1 Osteoarthrosis and allied disorders	59	11.92
2 Orthopedic impairment of lower extremity	55	11.09
3 Absence or loss of lower extremity	47	9.45
4 Chronic injuries or late effects of injuries	40	8.04
5 Orthopedic impairment of back or neck	25	4.98
6 Rheumatoid arthritis and other inflammatory polyarthropathies	21	4.33
7 Cerebral palsy	20	4.02
8 Orthopedic impairment of hip and/or pelvis	19	3.92
9 Intervertebral disc disorders	17	3.43
10 Other paralysis	11 *	2.32 *

†Conditions reported as the main cause of functional or activity limitation (see text).
*Standard error exceeds 30 percent of the estimate.

Table 3: Health and disability related characteristics of mobility device users and non-users by type of device: United States, all ages [1].

	Total		No mobility device		Any mobility device		Wheelchair		Scooter		Cane		Crutches		Walker	
	Number (1000s)	%	Number (1000s)	%	Number (1000s)	%	Number (1000s)	%	Number (1000s)	%	Number (1000s)	%	Number (1000s)	%	Number (1000s)	%
All persons	260,763	100.0	253,942	100.0	6,821	100.0	1,599	100.0	142	100.0	4,755	100.0	566	100.0	1,820	100.0
Health status																
Excellent	97,569	37.4	97,190	38.3	379	5.6	76	4.7	8 *	5.4 *	236	5.0	68	12.1	82	4.5
Very good	75,130	28.8	74,433	29.3	697	10.2	116	7.2	8 *	5.8 *	485	10.2	80	14.2	151	8.3
Good	60,045	23.0	58,323	23.0	1,722	25.2	346	21.6	32	22.5	1,200	25.2	161	28.4	374	20.6
Fair	18,603	7.1	16,654	6.6	1,949	28.6	407	25.5	38	26.7	1,438	30.3	119	21.0	520	28.5
Poor	7,214	2.8	5,199	2.0	2,015	29.5	638	39.9	52	36.4	1,355	28.5	136	24.1	667	36.6
Unknown	2,202	0.8	2,144	0.8	58	0.9	16	1.0	5 *	3.3 *	41	0.9	2 *	0.3 *	27	1.5
Hospitalization history																
Discharged in prior 6 months	11,252	4.3	9,776	3.8	1,476	21.6	478	29.9	35	24.7	898	18.9	154	27.2	556	30.6
Hospitalization in prior year	19,439	7.5	17,210	6.8	2,229	32.7	687	43.0	48	33.8	1,420	29.9	199	35.1	810	44.5
Perceived disability																
Self-perceived disability	17,557	6.7	13,312	5.2	4,246	62.2	1,285	80.4	122	85.7	2,796	58.8	357	63.1	1,239	68.1
Other-perceived disability	14,455	5.5	10,530	4.1	3,925	57.5	1,255	78.5	120	84.3	2,510	52.8	342	60.4	1,197	65.8
No perceived disability	241,444	92.6	239,171	94.2	2,273	33.3	240	15.0	14	10.2	1,754	36.9	183	32.4	492	27.0
Activity limitation																
Unable to do major activity	11,904	4.6	8,958	3.5	2,946	43.2	1,057	66.1	85	59.5	1,808	38.0	251	44.4	957	52.6
Only limited in major activity	14,725	5.6	12,970	5.1	1,755	25.7	349	21.8	39	27.1	1,248	26.2	146	25.7	471	25.9
Limited only in other activity	12,320	4.7	11,280	4.4	1,039	15.2	81	5.1	14	9.7	883	18.6	44	7.8	191	10.5
Not limited in activity	221,814	85.1	220,733	86.9	1,081	15.8	112	7.0	5 *	3.6 *	817	17.2	125	22.1	202	11.1
All persons aged 18 & over	190,414	100.0	183,738	100.0	6,676	100.0	1,511	100.0	142	100.0	4,736	100.0	530	100.0	1,794	100.0
Functional limitation																
Limited in 1 or more	25,103	13.2	19,183	10.4	5,920	88.7	1,453	96.2	140	98.4	4,136	87.3	433	81.8	1,711	95.4
Unable to perform 1 or more	7,595	4.0	3,970	2.2	3,625	54.3	1,294	85.7	109	76.8	2,182	46.1	240	45.4	1,336	74.5
No functional limitation	165,311	86.8	164,555	89.6	756	11.3	58	3.8	2 *	1.6 *	599	12.7	96	18.2	83	4.6
Specific functional limitations																
Lifting 10 lbs.	10,046	5.3	6,610	3.6	3,435	51.5	1,077	71.3	78	55.2	2,226	47.0	217	40.9	1,195	66.6
Unable	3,314	1.7	1,600	0.9	1,714	25.7	730	48.3	42	29.6	933	19.7	78	14.8	725	40.4
Difficulty only	6,732	3.5	5,010	2.7	1,721	25.8	347	23.0	36	25.6	1,293	27.3	139	26.2	470	26.2
Climbing stairs	11,020	5.8	6,654	3.6	4,365	65.4	1,336	88.4	116	81.8	2,826	59.7	315	59.5	1,464	81.6
Unable	2,519	1.3	738	0.4	1,781	26.7	962	63.7	63	44.3	760	16.1	107	20.2	794	44.3
Difficulty only	8,501	4.5	5,916	3.2	2,585	38.7	374	24.7	53	37.5	2,066	43.6	208	39.3	670	37.3
Walking 1/4 mile	14,558	7.6	9,412	5.1	5,147	77.1	1,424	94.2	138	96.9	3,469	73.3	379	71.5	1,603	89.4
Unable	4,936	2.6	2,061	1.1	2,875	43.1	1,186	78.5	102	72.0	1,596	33.7	183	34.5	1,132	63.1
Difficulty only	9,622	5.1	7,351	4.0	2,271	34.0	238	15.7	35	24.9	1,873	39.6	196	36.9	471	26.3
Standing 20 mins.	11,261	5.9	6,936	3.8	4,325	64.8	1,311	86.8	125	87.6	2,821	59.6	327	61.7	1,409	78.6
Unable	2,738	1.4	902	0.5	1,836	27.5	922	61.0	67	47.1	896	18.9	130	24.6	708	39.5
Difficulty only	8,524	4.5	6,035	3.3	2,489	37.3	389	25.8	58	40.5	1,924	40.6	197	37.2	701	39.1
Bending down	12,074	6.3	8,018	4.4	4,057	60.8	1,226	81.2	110	77.4	2,662	56.2	297	56.1	1,327	74.0
Unable	2,359	1.2	842	0.5	1,517	22.7	837	55.4	55	38.4	679	14.3	103	19.4	587	32.7
Difficulty only	9,715	5.1	7,176	3.9	2,539	38.0	389	25.8	55	39.0	1,983	41.9	195	36.8	740	41.3
Reaching up or out	5,272	2.8	3,520	1.9	1,752	26.2	563	37.3	59	41.6	1,158	24.5	112	21.2	572	31.9
Unable	916	0.5	455	0.2	460	6.9	227	15.0	11 *	8.0 *	242	5.1	15	2.9	168	9.3
Difficulty only	4,357	2.3	3,065	1.7	1,292	19.3	336	22.2	48	33.6	916	19.3	97	18.3	405	22.6
Grasping	4,695	2.5	3,255	1.8	1,440	21.6	510	33.7	47	33.4	941	19.9	79	14.9	452	25.2
Unable	309	0.2	129	0.1	180	2.7	137	9.1	6 *	4.6 *	65	1.4	4 *	0.8 *	40	2.2
Difficulty only	4,387	2.3	3,127	1.7	1,260	18.9	373	24.7	41	28.8	875	18.5	74	14.0	413	23.0
Holding pen	3,141	1.6	1,975	1.1	1,166	17.5	478	31.6	37	25.8	720	15.2	43	8.2	434	24.2
Unable	330	0.2	138	0.1	192	2.9	137	9.0	4 *	2.7 *	66	1.4	0	0.0	61	3.4
Difficulty only	2,811	1.5	1,838	1.0	974	14.6	341	22.6	33	23.2	654	13.8	43	8.2	374	20.8

2: DESIGN AND VALIDATION OF A PASSIVE WRIST ORTHOSIS

ABSTRACT

Lofstrand, or forearm, crutches are typically used when long-term gait assistance is needed. Repeated loading of the wrist and palmar region and continual hyperextension of the wrist during crutch use can cause wrist strain, injury, and possibly carpal tunnel syndrome. A novel wrist orthosis, which can be easily attached to any Lofstrand crutch, was developed with the intent of improving wrist posture and reducing/redirecting palmar loads from the carpal tunnel region to the adductor pollicis area.

Dominant-hand palmar loads and wrist extension angles of ten healthy able-bodied subjects were measured during swing-through Lofstrand crutch gait. Each subject performed 10 trials each without and with the orthosis. Peak pressure, mean pressure, maximum force, total force, contact area, and wrist extension from two healthy adult subjects are presented in this thesis.

Results suggested a decrease in total force, maximum force, mean pressure, peak pressure, contact area, and wrist extension when using the orthosis. Palmar loads were redirected toward the adductor pollicis when using the orthosis. With the use of the orthosis, total force decreased by 31%, maximum force decreased by 19%, mean pressure decreased by 31%, peak pressure decreased by 3.5%, contact area decreased from 58 cm^3 to 46 cm^3, and wrist extension decreased from 37.5 deg. to 26.5 deg.

The use of a wrist orthosis on a Lofstrand crutch can reduce wrist extension and redirect palmar loads, thus reducing potential sources of injury.

2.1 INTRODUCTION

Common injuries found among crutch users include wrist, elbow, and shoulder pain. Wrist pain may occur in the form of carpal tunnel syndrome (CTS). CTS and other forms of wrist pain are preventable by reducing loads experienced by the wrist, reducing the degree of flexion and extension experienced during loading, and reducing the number of times flexion and extension occur during loading [2], [4], [7], [8]. Since pressures at the palmar surface of the carpal tunnel increase carpal tunnel pressure more than any other areas of the palm, it may be beneficial to redirect loads from the carpal tunnel region to other areas on the palm [2], [4]. Traditional hand and arm orthoses may restrain movement in ways that interfere with crutch gait [52]. Alternative orthosis designs to assist in the prevention of wrist pain during crutch gait should be explored.

A novel wrist orthosis was proposed as a solution by improving wrist posture and reducing/redirecting palmar loads from the carpal tunnel to the adductor pollicis region. Goals for this design included: the ability to limit wrist hyperextension and reposition the wrist into neutral position during gait, the ability to reduce and redirect loads experienced at the carpal tunnel region of the palm, and to seamlessly integrate the device with the Lofstrand crutch. By contouring and supporting the forearm during crutch gait, the wrist orthosis allows for the wrist to be supported closer to the neutral position. The orthosis is expected to allow for loads to be displaced from the carpal tunnel region on the palm to the adductor

pollicis. The remainder of this chapter presents the design of the orthosis and the experimental protocol used to evaluate the device.

2.2 METHODS

2.2.1 Orthosis Design

The orthosis is lightweight, made from high impact polystyrene (HIPS), and is an easily attachable accessory to any Lofstrand crutch (Figure 10 and Figure 11). The device weighs 3.9 oz, which included screws, strap, coupling, padding, and worm gear clamp. The enclosed dimensions of the device once attached are approximately 6 in x 5.8 in x 2.5 in. The orthosis has two connection points to the Lofstrand crutch: at the handle and at the neck. Once fastened at both the handle and neck, the orthosis remains stable with little deformation. To add comfortable, a strap was added to provide compliant support at the forearm-orthotic interface (Figure 10). The strap allows for a small amount of extension, which may be changed by adjusting the tension of the strap.

The curvature of the wrist orthosis was an important feature. The curvature allowed for comfort of the forearm and wrist during gait, while acting as a barrier to high angles of extension. When an individual places the arm into the Lofstrand crutch cuff and holds the handle, there is a natural angle of radial deviation of the wrist. The curvature of the orthotic allowed the wrist to have natural radial deviation. Furthermore, the curvature and attachment sites were designed to allow for small to large forearms and hands. Larger hands were found to rub against previous design iterations of the orthosis; rounded and raised edges for areas around the thumb and palm were made to allow for comfortable use. The use of only one side of the handle for clamping, and the large width of the device allow for hands of up to 4.0 inches wide and forearms up to 3.6 inches wide.

2.2.2 Experimental assessment

An experimental study was conducted to assess the effectiveness of the wrist orthosis with regard to improved wrist posture and reducing and redirecting palmar loads.

2.2.3 Test subject demographics

Ten healthy able-bodied subjects were evaluated in this study. Five male and five female subjects (average height: 1.61 m, weight: 72 kg) participated in this study, two subjects were left handed. One left handed female and one left handed male subject are analyzed for this thesis, and they are referred to as Subject 10 and Subject 11, respectively, throughout this chapter. The institutional review boards at the University of Illinois and University of Wisconsin - Milwaukee approved all procedures, and all participants gave informed consent.

2.2.4 Testing Protocol

Subjects used a swing-through gait while walking with Lofstrand crutches at a self-selected gait speed. Subjects performed ten trials under one of two conditions: without and with the orthosis attached to

each crutch. One set of Lofstrand crutches was used; between conditions the orthosis was either attached or unattached. Attaching and removing the orthosis can be performed in less than five minutes. Conditions were randomized as to which condition would be performed first. Subjects performed swing-through gait down a 20 foot long walkway for a total of 20 trials. The subject was offered rest after each trial. The subject was given instruction on how to hold the crutches and perform swing-through gait. The subject was allowed up to 10 minutes of practice prior to the first test condition.

The Lofstrand crutch height was adjusted such that the handle of the Lofstrand crutch was level with the subject's standing wrist height. If the subject's wrist was between two heights, the subject was allowed to choose the height that felt most comfortable. The cuff of the Lofstrand crutch was placed 1-1.5 in below the elbow [53].

Thirty-nine motion capture markers were attached to the subject to record joint kinematic data. Kinematic data were captured at 100 Hz and processed using a 15 camera motion analysis system (T-Series, Vicon Motion Systems Ltd., Oxford, UK) and a Plug-in-Gait (PIG) model using Nexus 2.1.1 software. The motion capture marker placements used for the PIG model are labeled in Figure 12-Figure 14. A video recorder was also used during testing.

A static trial was recorded once per subject for calibration purposes of the motion capture system. The static trial was taken by having the subject stand in the anatomical neutral position while in view of the camera system for three seconds. Data were recorded for the static trial and were used for the PIG model to recognize and label each individual marker. After this trial, the subject's anthropometric measurements were put into the PIG software (Figure 15).

A 16 x 16 cm^2 Novel Pliance matrix array (S2129_010, Novel Inc., Munich, Germany) was used to record loads on the crutch handle under the palm of the dominant hand (Figure 16). The sensor data were captured at 120 Hz. The array was folded in half to fit on the crutch handle, rolled around the handle and adhered with double sided tape to capture pressure data around the surface area of the crutch handle (Figure 17). The sensors were masked such that only the cells touching the handle (i.e., the closest layer to the crutch handle) were collecting data. The array was plugged into the Novel Pedar transmitter, which was attached on the back of a belt worn by the subject (Figure 18). The belt also housed the Pedar battery pack. The Novel Pliance software was used to interpret and analyze data from these sensors.

Prior to the gait trials, pressure data were collected to locate the adductor pollicis relative to the crutch handle. The subject was instructed to hold the crutch handle as it would normally be held during gait and apply pressure to the adductor pollicis by leaning into the hand on that region. This process was repeated without and with the orthosis in use. Additionally, hand width and the perpendicular distance from a subject's adductor pollicis to the mid-palm were measured to identify the carpal tunnel area (Figure 19).

Subject commentary was recorded throughout the study. After every five trials, the subject was asked to describe the perceived exertion of the task on a Borg RPE scale (Figure 20) [54]. After the completion of all trials for a given test condition, the subject was asked to describe the feeling of the wrists, elbows,

palms, and shoulders. At the end of the testing session, participants were asked to describe differences between test conditions, and whether the device helped with the crutch-walking experience or not [Table 4].

Data were post-processed using MATLAB, Vicon Nexus, and Novel Pliance software. The kinematic marker trajectories were filtered directly in the Nexus software using a Woltring filter with a mean squared error setting of 20 [55]. Due to poor marker visibility, five trials were used for Subject 010, and four trials were used for Subject 011 to calculate wrist extension [Table 5]. Wrist angles were calculated using the PIG and Nexus software and averaged over all relevant trials without and with the orthosis for each subject. The cell arrays analyzed for the Novel sensor were only the cells physically touching the crutch handle when wrapped around the handle. Dominant hand maximum force, peak pressure, and contact area over the crutch handle were calculated using the Pliance software [Table 6 and 7]. MATLAB was used to process and plot all data including, total force, mean pressure, peak pressure locations, wrist extension, and respective averaged values over each trial (Figure 21-Figure 31). Total force and mean pressure were normalized to subject weight when computing percentage differences [Table 8].

2.3 RESULTS

The two subjects presented in this thesis displayed an overall decrease in total force, maximum force, mean pressure, peak pressure, contact area, and wrist extension while using the orthosis [Table 6-Table 8]. Palmar loads were observed to be redirected toward the adductor pollicis when using the orthosis (Figure 21-Figure 23). Results also show a decrease in the total force, mean pressure, and wrist extension when using the Lofstrand crutch handle with the orthosis (Figure 24-Figure 31).

On average between the two subjects with the use of the orthosis, total force decreased by 31%, maximum force decreased by 19%, mean pressure decreased by 31%, peak pressure decreased by 3.5%, contact area decreased from 58 cm^3 to 46 cm^3, and wrist extension decreased from 37.5 deg to 26.5 deg On average, total force and mean pressure over the handle decreased when using the orthosis.

Arrays of cells representing the matrix array of the Novel Pliance sensor are shown in Figure 21 and Figure 22. The cell array numbers correspond to the 'Cell Location' columns in Tables 5 and 6. The location of peak pressure can be seen to shift toward the adductor pollicis while using the orthosis (Figure 21 and Figure 22). An example of mapping particular cells to the adductor pollicis and mid-palm region for Subject 011 is shown in (Figure 23).

The averaged wrist extension angle decreased from 37.5 deg to 26.5 deg when using the orthosis (Figure 28-Figure 31).

2.4 DISCUSSION

The overall decrease in contact area and maximum force for the subjects had a direct effect on the maximum pressure readings while using the orthosis. Due to the decrease in area, there was an increase in maximum pressure. Although there was a decrease in maximum force experienced by the subjects,

this force may have been offloaded to the orthosis or otherwise affected by the subject's wrist and forearm posture. Differences in contact area without and with the orthosis may be due to gripping patterns on the crutch handle. Different gripping patterns resulting in less contact area and force may indicate improved ergonomics while using the orthosis (Figure 22).

The location of the peak pressure shifted toward the adductor pollicis for both subjects (Figure 21 and Figure 22). For carpal tunnel syndrome prevention, it is beneficial to load away from the mid-palm area. The orthosis caused a shift in maximum pressure to be further from the mid-palm area. It is notable that, for Subject 011, using the orthosis also shifted the location of the subject's adductor pollicis with respect to the crutch handle (Figure 22). The shift in the location of palmar regions by Subject 011 may indicate a new ergonomic posture of the palm due to the orthosis.

On average, the total force, maximum force, mean pressure, and peak pressure decreased when using the orthosis. The location of the peak pressure also shifted for both subjects. These data further indicate an offloading effect due to the orthosis.

2.4.1 Limitations

One limitation of the current wrist orthosis prototype was the size and fit of the orthosis. Although the orthosis was designed to allow for a wide range of hand sizes (up to 4 inches in width), the effectiveness and comfort of the device may have changed based on a subject's hand size, forearm size, or gait pattern. Some subjects felt as though the device was too large and others felt that they could not feel the device at all [Table 4]. Other subjects commented that they felt that the orthosis directly aided their wrist posture. None commented that the device was too small.

Another limitation was that some markers on the wrist, particularly RWRA and RWRB, were frequently blocked due to the orthosis and other hardware (Figure 12). With the markers blocked for some frames, interpolations were made throughout subject trials. For select trials, missing markers lead to the motion capture trials being omitted from the study.

A further limitation of the study was that only healthy adult subjects were tested. Additional studies within specific patient populations should be done to fully assess the benefit of the orthosis. Additional models of the device using the same design principles could be made for other crutch types, such as auxiliary crutches. Creating these additional models of the orthosis would allow for a full spectrum of subject populations and crutch types to be tested. A further limitation of the study was the lack of a sensor between the orthosis and the subject at the strap. This would have allowed for data on the offloading capabilities of the orthosis.

2.4.2 Future Work

Additional considerations should be taken into account for sizing of the orthosis. Whether this is in the form of large, medium, and small versions of the device or an adjustable device, the issue should be addressed to suit a particular user of the device. Despite this shortcoming, evidence that a wrist orthosis attached to a Lofstrand crutch handle can alter the location of palmar pressure and wrist angle has been

demonstrated. By addressing the above limitations surrounding the orthotic design, sets of orthoses can be implemented to improve the quality of life for crutch users throughout the world.

2.5 CONCLUSION

Two subjects were analyzed after performing 10 swing-through gait cycle trials without and with a novel wrist orthosis attached to a Lofstrand crutch. Wrist kinematics were captured with a 15 camera Vicon system. Palmar pressure data were captured using a Novel Pliance system. It was found that for these subjects, on average, the orthosis reduced palmar total force, maximum force, mean pressure, peak pressure, contact area, and wrist extension during swing-through gait. Palmar loads were observed to be redirected from the carpal tunnel region toward the adductor pollicis when using the orthosis. The intended goals of the wrist orthosis were achieved for the two subjects analyzed. The additional eight subjects will still be analyzed for a full reported study. Through the use of a wrist orthosis, it is possible to redirect loads away from the carpal tunnel area, reduce palmar load magnitudes, and improve wrist posture to mitigate hyperextension.

Figures:

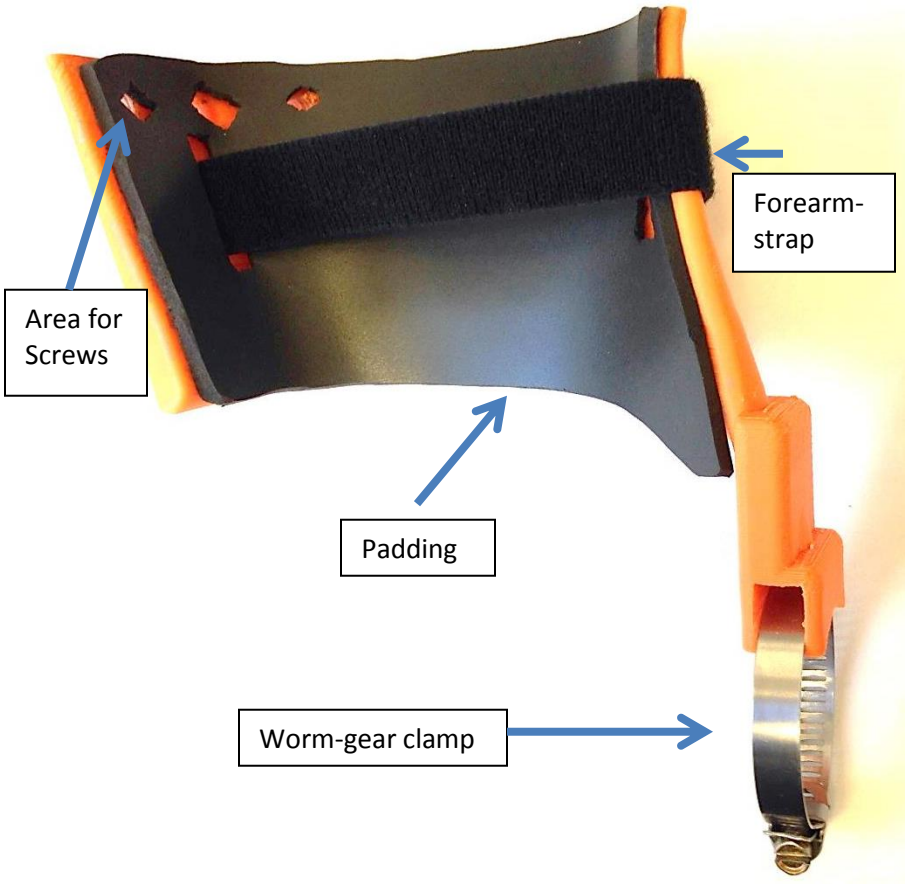

Figure 10: Single orthosis

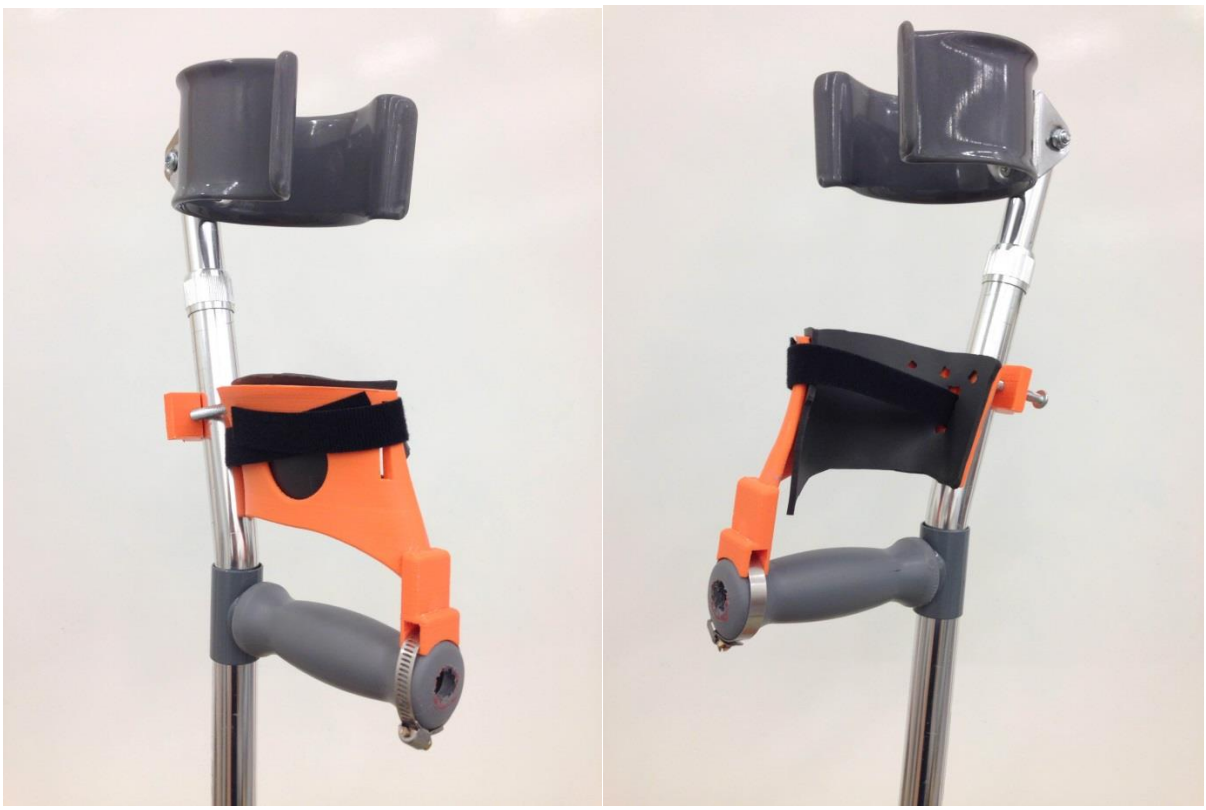

Figure 11: Upper extremity orthosis (orange) attached to a Lofstrand crutch

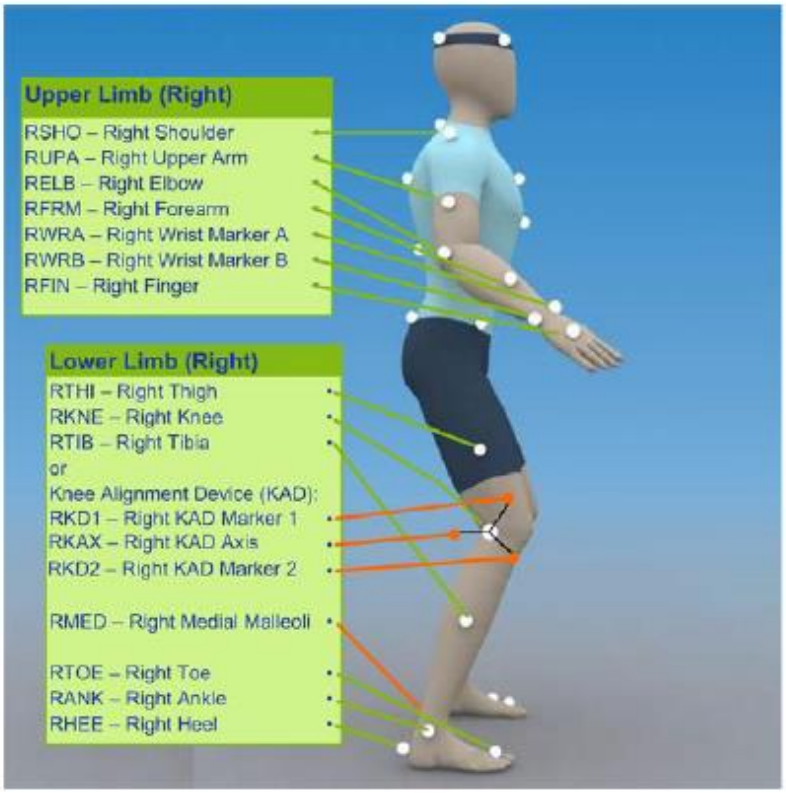

Figure 12: PIG marker placement (side)

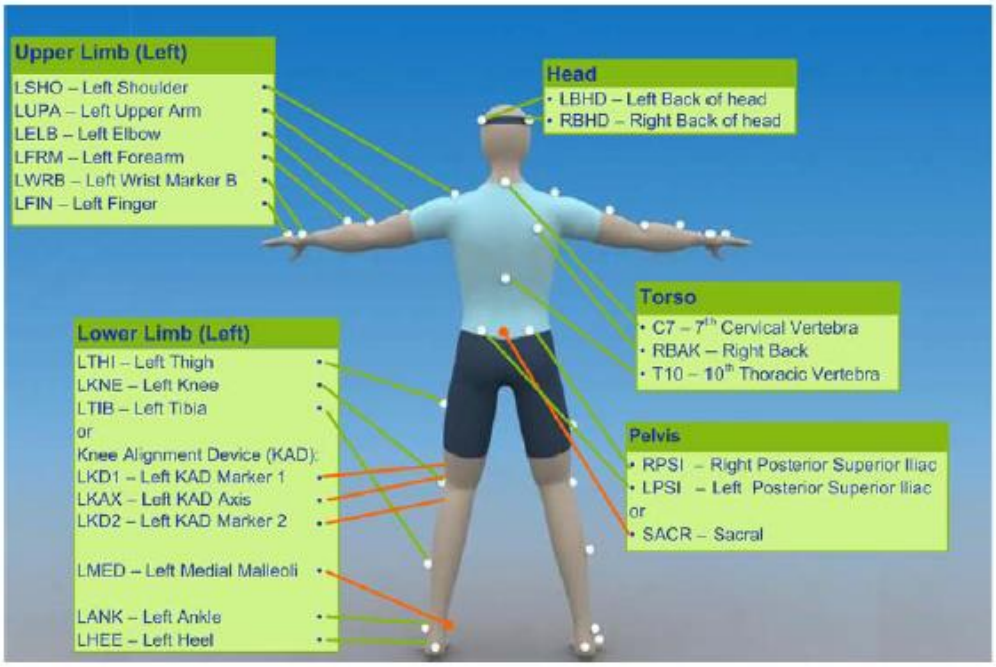

Figure 13: PIG marker placement (back)

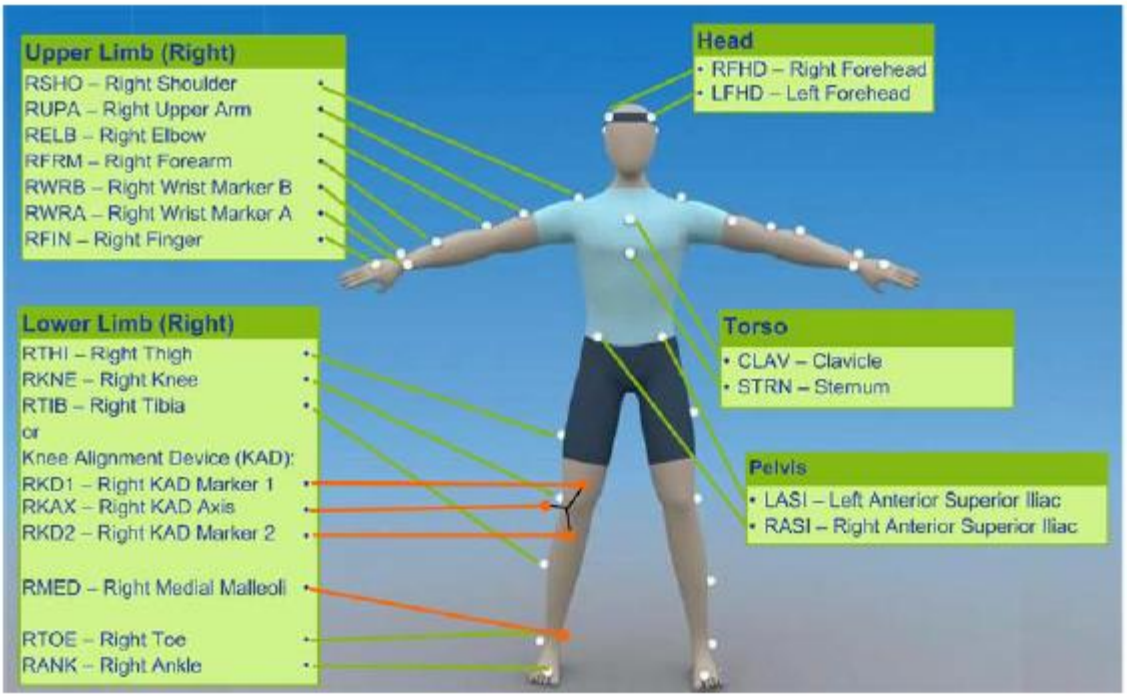

Figure 14: PIG marker placement (front)

Name	Description	Measurement	
		Left	Right
All			
Body Mass	Patient mass.	_____ kg	
Height	Patient height	_____ mm	
Lower Body			
Inter-ASIS distance*	ASIS-ASIS distance is the distance between the left ASIS and right ASIS. This measurement is only needed when markers cannot be placed directly on the ASIS, for example, in obese patients.	_____ mm	
Leg Length	Full leg length, measured between the ASIS marker and the medial malleolus, via the knee joint. Measure with patient standing, if possible. If the patient is standing in the crouch position, this measurement is NOT the shortest distance between the ASIS and medial malleoli, but rather the measure of the skeletal leg length.	_____ mm	_____ mm
Knee Width	The medio-lateral width of the knee across the line of the knee axis. Measure with patient standing, if possible.	_____ mm	_____ mm
Ankle Width	The medio-lateral distance across the malleoli. Measure with patient standing, if possible.	_____ mm	_____ mm
Sole Thickness Delta*	The difference in the thickness of the sole at the toe and the heel. A positive sole delta indicates that the patient's heel is raised compared with the toe.	_____ mm	_____ mm
ASIS-Trochanter Distance*	ASIS-greater trochanter distance is the vertical distance, in the sagittal plane, between the ASIS and greater trochanter when the patient is lying supine. Measure this distance with the femur rotated such that the greater trochanter is positioned as lateral as possible.	_____ mm	_____ mm
Upper Body			
Shoulder Offset	Vertical offset from the base of the acromion marker to shoulder joint center.	_____ mm	_____ mm
Elbow Width	Width of elbow along flexion axis (roughly between the medial and lateral epicondyles of the humerus).	_____ mm	_____ mm
Wrist Width	Anterior/Posterior thickness of wrist at position where wrist marker bar is attached.	_____ mm	_____ mm
Hand Thickness	Anterior/Posterior thickness between the dorsum and palmar surfaces of the hand.	_____ mm	_____ mm

*: optional measurements which Plug-in Gait can calculate

Figure 15: Subject measurements for Plug in Gait (PIG)

Figure 16: Novel Pliance sensor folded in half and taped.

Figure 17: Novel Pliance sensor wrapped around handle with orthotic.

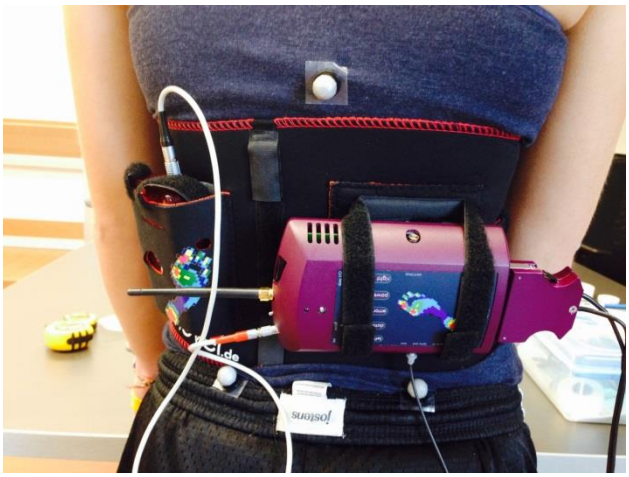

Figure 18: Pedar Pliance system (right) and battery pack (left) attached to belt.

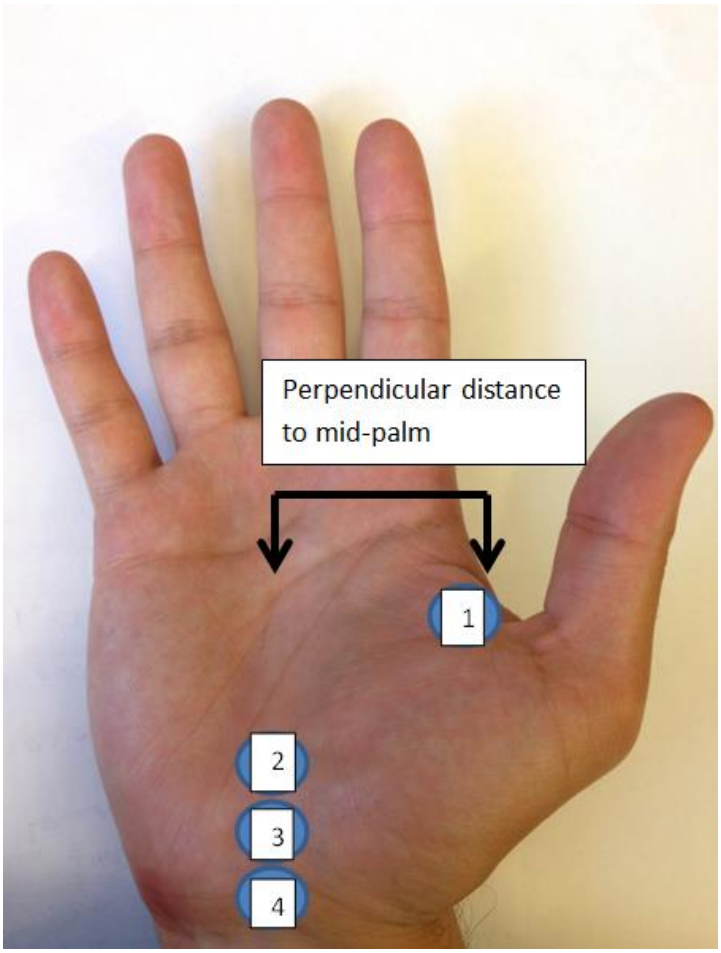

Figure 19: Areas of interest on hand, referred to as follows: 1) Adductor pollicis, 2) mid-palm, 3) lower palmar aponeurosis, and 4) carpal tunnel

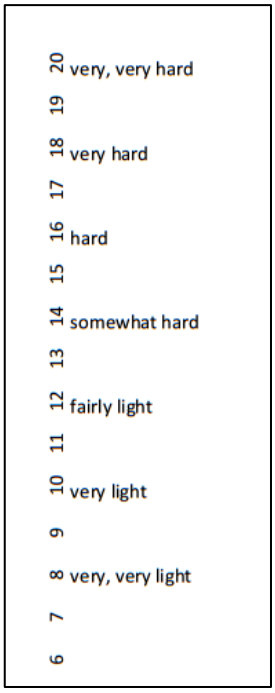

Figure 20: Borg RPE Scale

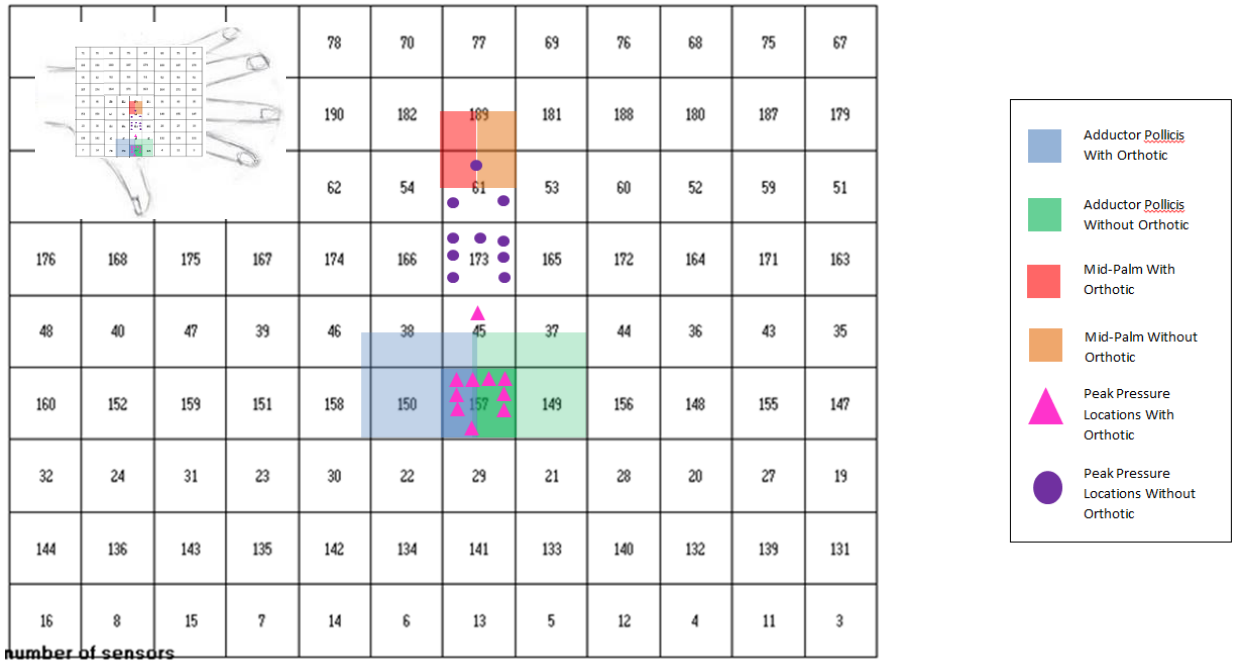

Figure 21: Subject 010 adductor pollicis and mid-palm cell locations with and without orthotic (each cell is 1 cm^2)

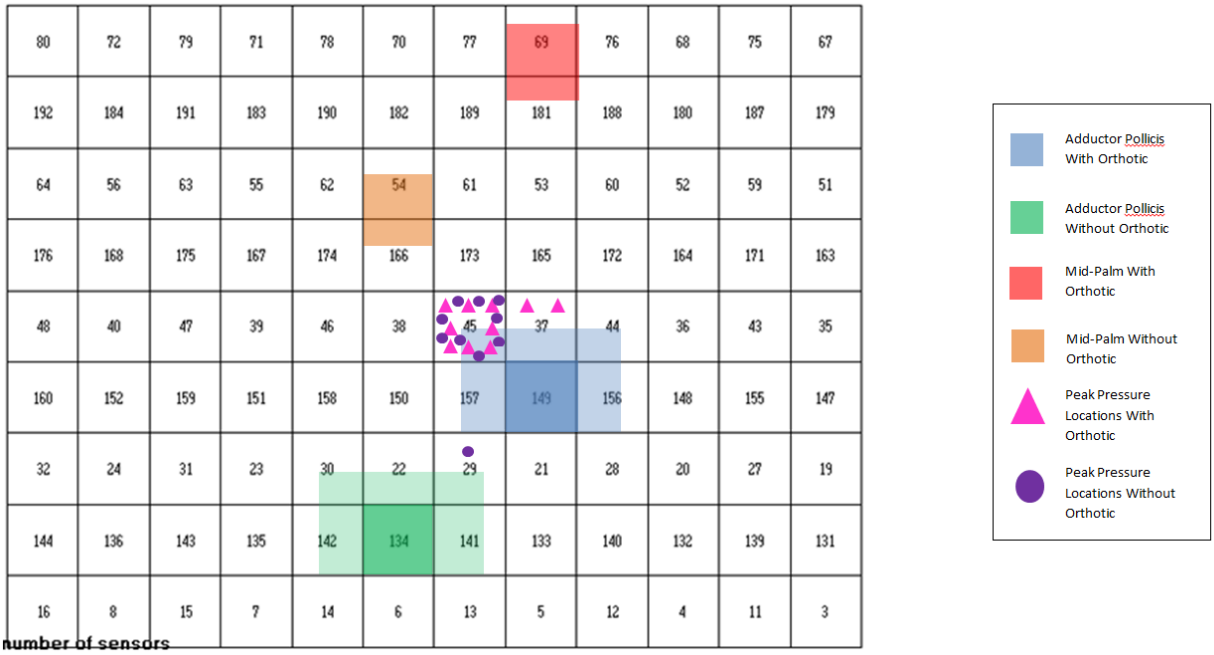

Figure 22: Subject 011 adductor pollicis and mid-palm cell locations without and with orthosis (each cell is 1 cm^2)

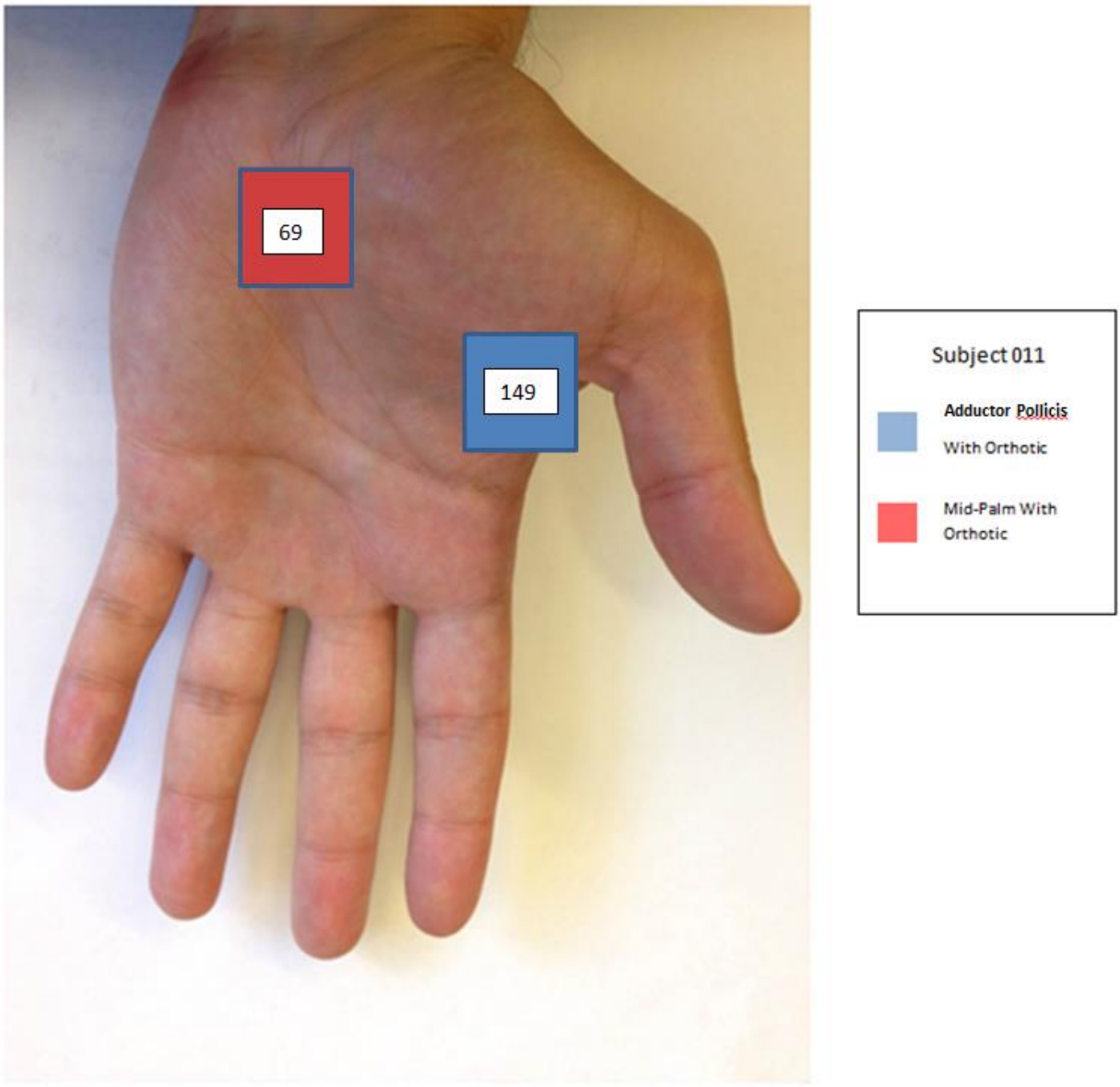

Figure 23: Example of cell mapping over to the adductor pollicis and mid-palm regions of Subject 011

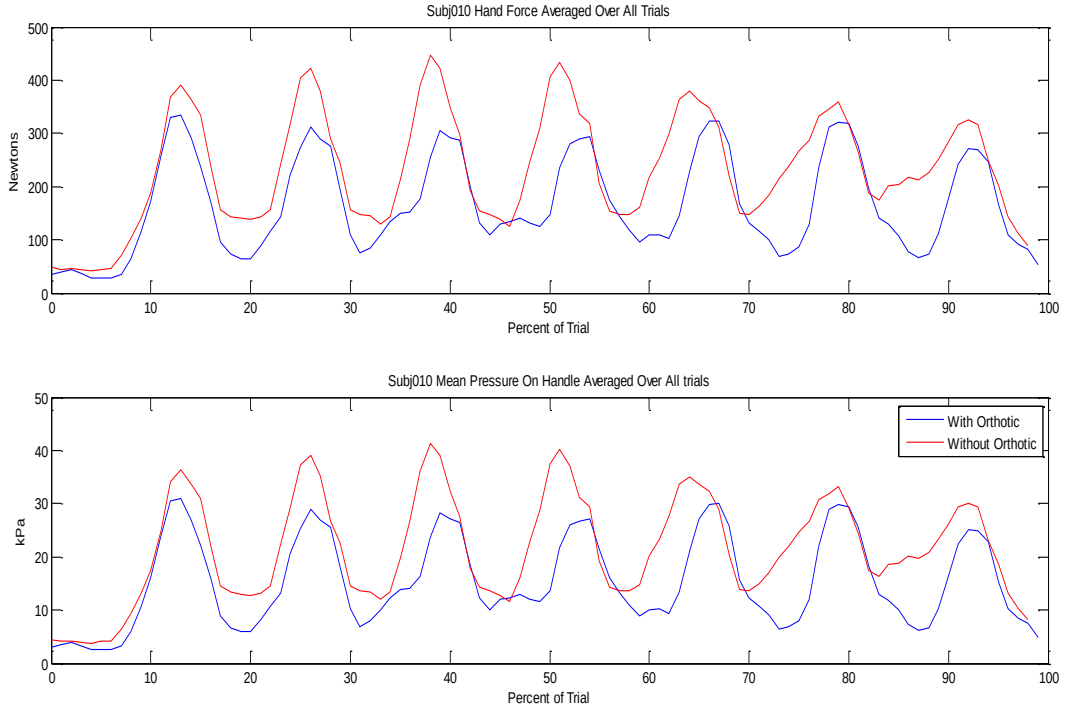

Figure 24: Subject 010 averaged force and mean pressure over all trials without and with orthosis

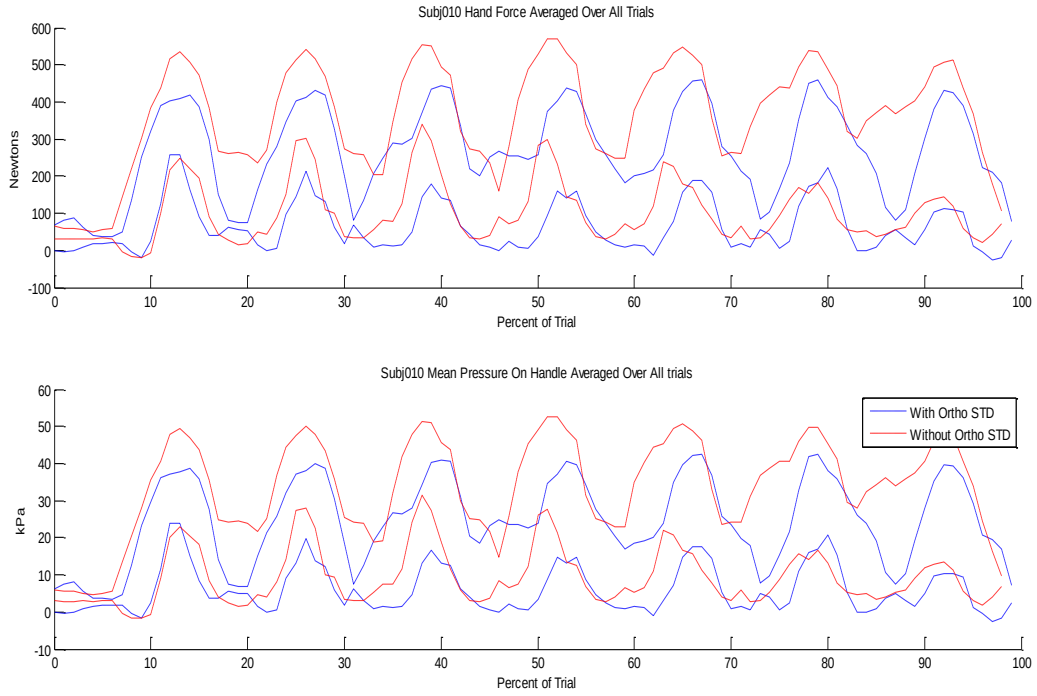

Figure 25: Subject 010 standard deviations of force and mean pressure over all trials without and with orthosis

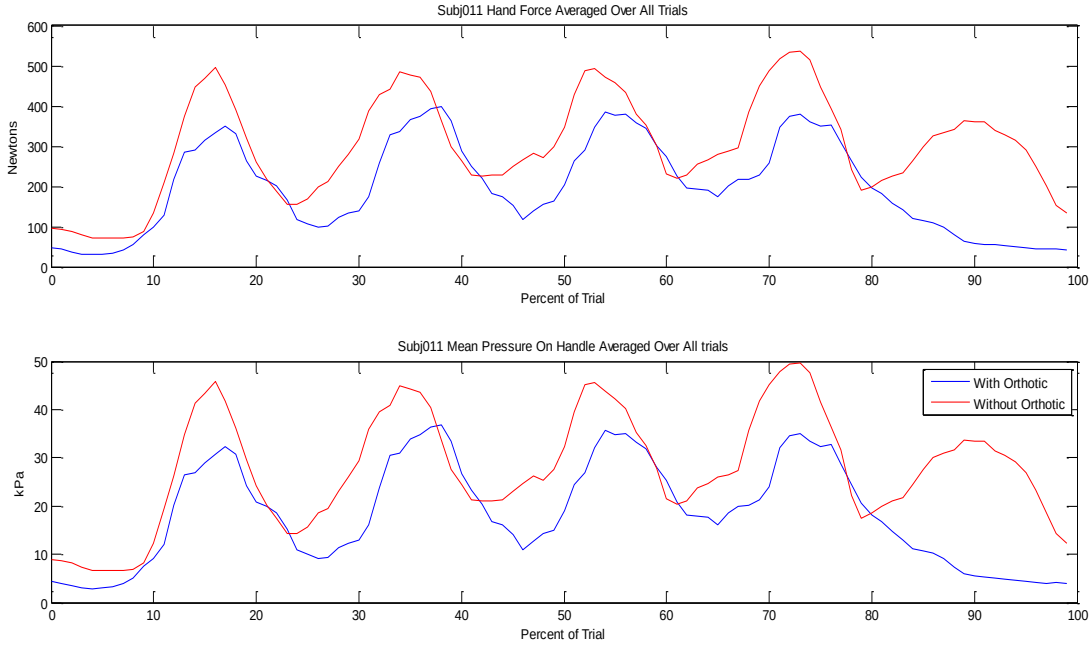

Figure 26: Subject 011 averaged force and mean pressure over all trials without and with orthosis

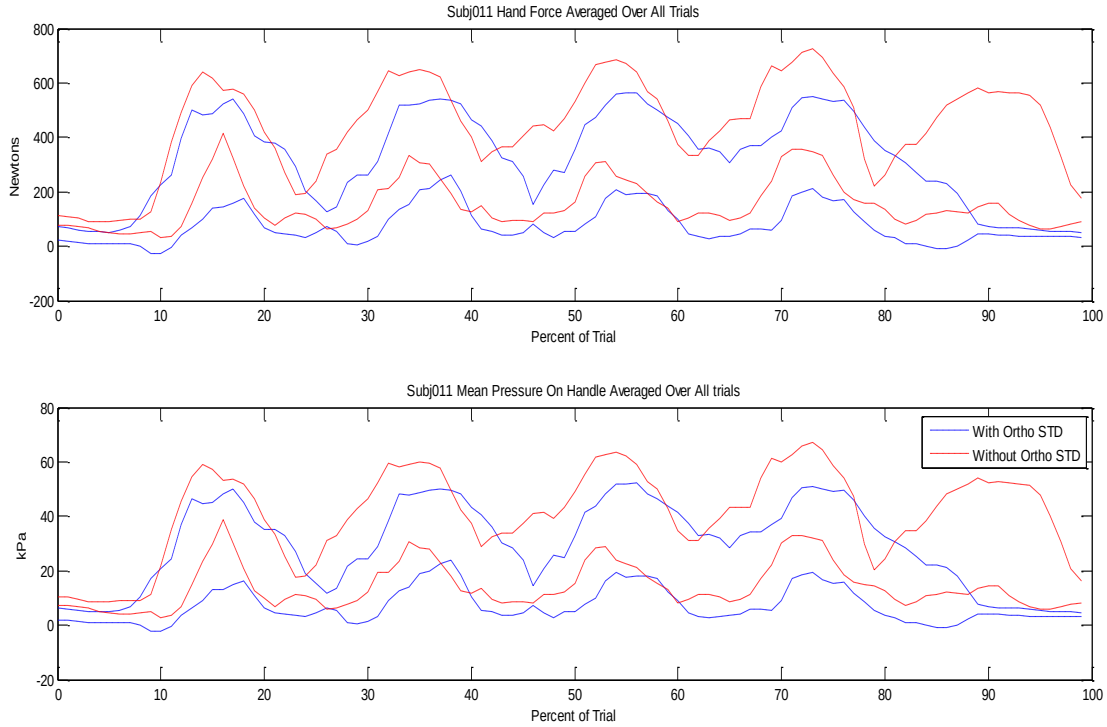

Figure 27: Subject 011 standard deviations of force and mean pressure over all trials without and with orthosis

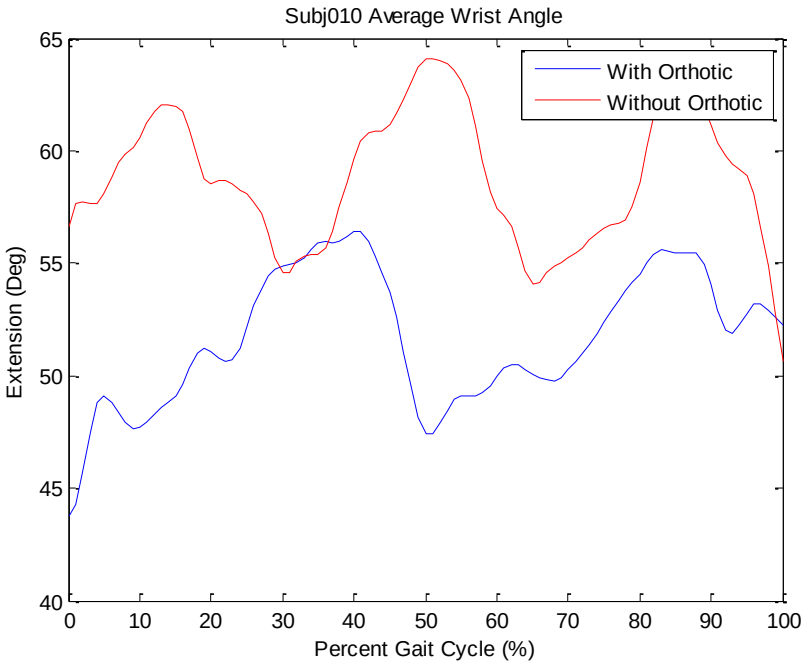

Figure 28: Subject 010 wrist extension averaged over all trials without and with orthosis

Figure 29: Subject 010 wrist extension standard deviations over all trials without and with orthosis

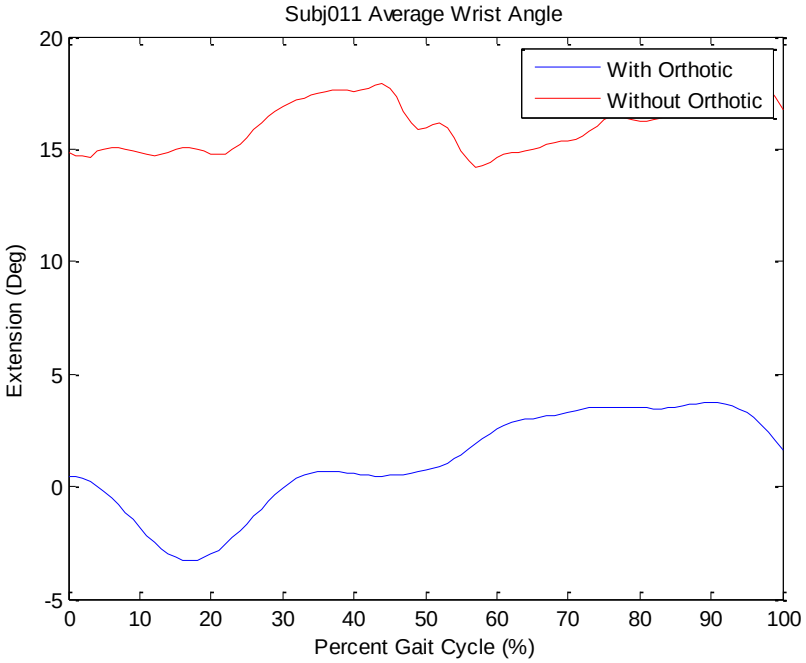

Figure 30: Subject 011 wrist extension averaged over all trials without and with orthosis

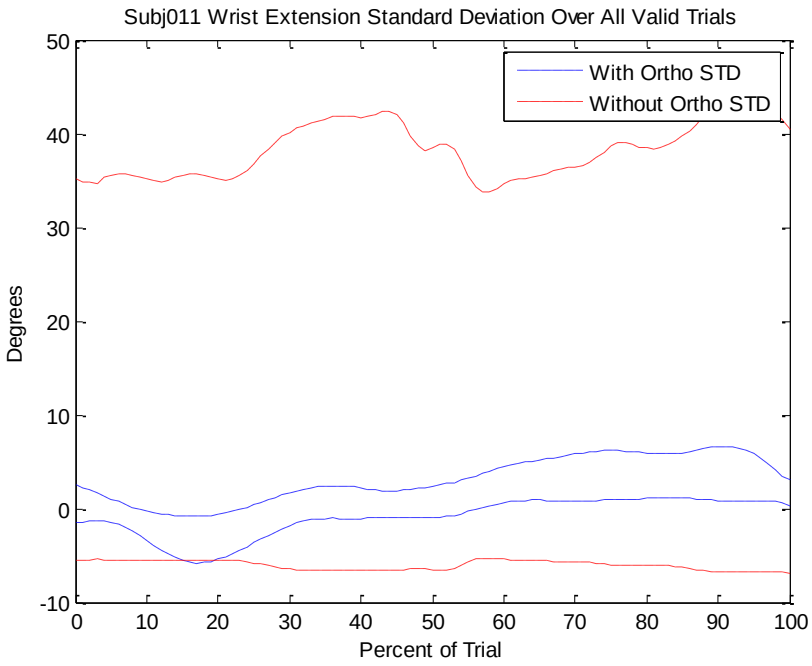

Figure 31: Subject 011 wrist extension standard deviations over all trials without and with orthosis

TABLES

Table 4: Subject commentary and Borg RPE ratings.

Subject	10	11
Order of test condition	Without orthosis first then with orthosis	With orthosis first then without orthosis
Comments About Lofstrand Crutches	Started feeling discomfort in hand on 5th trial; Triceps started to feel it	A lot of pressure on thenar region; Arms slip out of cuffs; Unstable compared to underarm crutches
Comments Using Crutches & Orthotic	More unsteady at elbows; Elbow stability helped with tightening forearm cuff	
Feeling Without Orthotic	Hurts (pain in adductor pollicis area); Elbows/shoulders feel shaky; Wrists felt loaded	Palms felt a lot of pressure in adductor pollicis arch; Elbows/shoulders/wrist good
Feeling With Orthotic	Left hand loading more than right; Felt pressure moving further back into thenar fatty area; Left wrist hurting; Shoulders tired	Elbow/shoulders/wrist fine; adductor pollicis area of palms feels lots of pressure
Which Condition Was Easier/Less Painful?	No notice to significant difference; Felt more tired going into trials w/ orthotic	Orthosis got in way; Easier without it; Don't know if orthosis was engaged at all
Overall Impressions of Orthosis	Seemed too big for wrist	Didn't think of it; Don't know if it helped or hurt
Average Perceived Exertion With Orthosis	13	12
Average Perceived Exertion Without Orthosis	14.5	13

Table 5: Subject 010 and 011 VICON trial quality. Only Good trials were used for wrist extension data.

Subject 010		
Trial	Quality	Comments
1 With	Good	
2 With	Poor	Missing wrist marker
3 With	Good	
4 With	Poor	Missing wrist marker
5 With	Poor	Missing wrist marker
6 With	Good	
7 With	Good	
8 With	Good	
9 With	Poor	Missing wrist marker
10 With	Poor	Missing wrist marker
1 Without	Poor	Missing wrist marker
2 Without	Poor	Missing wrist marker
3 Without	Poor	Missing wrist marker
4 Without	Good	
5 Without	Good	
6 Without	Good	
7 Without	Poor	Missing wrist marker
8 Without	Poor	Missing wrist marker
9 Without	Good	
10 Without	Good	

Subject 011		
Trial	Quality	Comments
1 With	Good	
2 With	Poor	Missing wrist marker
3 With	Good	
4 With	Poor	Missing wrist marker
5 With	Poor	Missing wrist marker
6 With	Poor	Missing wrist marker
7 With	Good	
8 With	Poor	Missing wrist marker
9 With	Poor	Missing wrist marker
10 With	Good	
1 Without	Good	
2 Without	Poor	Missing wrist marker
3 Without	Good	
4 Without	Poor	Missing wrist marker
5 Without	Poor	Missing wrist marker
6 Without	Good	
7 Without	Poor	Missing wrist marker
8 Without	Poor	Missing wrist marker
9 Without	Poor	Missing wrist marker
10 Without	Good	

Table 6: Subject 10 data without and with orthosis.

Subject 010				
Trials With Orthotic				
	Peak Pressure (kPa)	Cell Location	Contact area [cm²]	Max force [N]
Palpation Trial	160	157		
Trial 1	492.5	157	33	375.5
Trial 2	515	157	36	451.5
Trial 3	492.5	157	36	395
Trial 4	500	157	35	416.3
Trial 5	470	157	40	411
Trial 6	502.5	45	34	427.5
Trial 7	575	157	38	441.8
Trial 8	580	157	39	468
Trial 9	540	157	40	440.5
Trial 10	605	157	37	456.8
Trial Average (STD)	527.3 (43.0)		36.8 (2.3)	428.4 (27.6)
Trials Without Orthotic				
Palpation Trial	165	157		
Trial 1	400	173	53	523.5
Trial 2	502.5	61	50	512.5
Trial 3	447.5	173	49	536.3
Trial 4	477.5	61	48	540
Trial 5	475	173	49	556.3
Trial 6	500	61	53	545.8
Trial 7	517	173	53	556.3
Trial 8	520	173	48	558
Trial 9	557.5	173	48	541.3
Trial 10	542.5	173	50	524.3
Trial Average (STD)	494.0 (44.0)		50.1 (2.0)	539.4 (14.8)

Table 7: Subject 11 data without and with orthosis.

Subject 011				
Trials With Orthotic				
	Peak Pressure (kPa)	Cell Location	Contact area [cm²]	Max force [N]
Palpation Trial	225	149		
Trial 1	522.5	45	57	594.3
Trial 2	587.5	45	54	571
Trial 3	510	45	56	574
Trial 4	477.5	45	50	507
Trial 5	510	45	56	553.3
Trial 6	467.5	45	52	517.5
Trial 7	637.5	45	57	566.3
Trial 8	472.5	37	50	515.8
Trial 9	522.5	37	55	540.8
Trial 10	540	45	56	582.3
Trial Average (STD)	524.8 (50.6)		54.3 (2.6)	552.2 (29)
Trials Without Orthotic				
Palpation Trial	287.5	134		
Trial 1	580	45	64	624.3
Trial 2	637.5	29	63	674.3
Trial 3	600	45	67	640
Trial 4	637.5	45	66	688.8
Trial 5	617.5	45	68	661.5
Trial 6	607.5	45	68	645.5
Trial 7	532.5	45	65	663.5
Trial 8	600	45	67	673
Trial 9	592.5	45	71	754
Trial 10	587.5	45	66	685.5
Trial Average (STD)	599.3 (28.9)		66.5 (2.2)	671 (33.7)

Table 8: Subject 010 and 011 normalized total force and mean pressure data.

Subject & Measure	With Orthosis	Without Orthosis
S-10 Avg Total Force	166 N	233 N
S-10Avg Mean Pressure	15 kPa	22 kPa
S-11 Avg Total Force	199 N	299 N
S-11 Avg Mean Pressure	18 kPa	28 kPa
S-10 Normalized Total Force	2.38	3.35
S-10 Normalized Mean Pressure	0.22	0.31
S-11 Normalized Total Force	2.68	4.03
S-11 Normalized Mean Pressure	0.25	0.37

3: DESIGN AND VALIDATION OF A PNEUMATIC ENERGY HARVESTING DEVICE WITHIN THE TIP OF THE LOFSTRAND CRUTCH

ABSTRACT

Pneumatically powered orthoses need power sources; however, pressurized canisters can be bulky and replacing pneumatic supplies can be inconvenient. Power harvesting devices have attempted to solve the issue by tapping into the potential power source of the human gait cycle. During crutch gait, there is a potential source of energy to be harvested at the crutch tip. The purpose of this study was to evaluate the effectiveness of a piston pump within a Lofstrand crutch tip used as a pneumatic energy harvesting device.

Two piston pump designs with varying stroke volumes (2.4 ml and 12.5 ml) were tested and stored pneumatic pressure was collected in an elastomeric accumulator. The smaller piston pump was inserted into the bottom of a Lofstrand crutch and actuated by pressing down on the crutch handle. The larger piston pump was not installed in the crutch and actuated by hand. These designs were tested by loading the piston pump for 20-35 cycles and measuring the pressure accumulated.

It was found that the accumulated pressure depended on the maximum fluid volume pumped per stroke, as well as the dead volume in the cylinder and line. The amount of force applied was found to be negligible with regard to accumulated pressure. A 2.4 ml stroke volume with 2 ml of dead volume was found to accumulate a maximum of 15 psig after 20 strokes. A 12.5 ml stroke volume with 3.5 ml of dead volume were found to accumulate up to 40 psig within 20 strokes and could go to higher pressures with more strokes. Pneumatic modeling was performed to understand the relationship between accumulated pressure, stroke volume, dead volume, and different displaced volumes per stroke. Design constraints were determined for the construction of pneumatic energy harvesting devices at a crutch tip. Through use of the models, it is possible to design a pneumatic cylinder to fit the pressure and volume specifications of a variety of applications.

3.1 INTRODUCTION

One motivation to develop a pneumatic energy harvesting device is to power a pneumatic device used by the crutch user. The pneumatic harvesting device is proposed for use within a Lofstrand crutch for powering pneumatically powered active orthotic devices. The goal of this project was to design and validate an energy harvesting device that could accumulate pneumatic energy during ambulation. This device may allow for pneumatically actuated orthotics to be powered by energy harvesting during gait. The pneumatic harvesting device may also be used in other forms of ambulation, such as auxiliary crutches and non-rolling walkers. External power sources may be bulky, heavy, or inconvenient and the proposed design may allow the user to refrain from carrying an external power source as they walk. The addition of a pneumatic energy harvesting device to a Lofstrand crutch would allow for pneumatically powered assistive devices to be less bulky by utilizing a constantly replenishing energy source.

There are several possible benefits to having an internal piston pump assembly that utilizes a spring in a Lofstrand crutch. It has been shown that shock absorption, propulsion assistance, reduced ground reaction forces, and increased velocity are results of utilizing an internal spring during crutch gait. No metabolic efficiency of spring-crutches were found versus unmodified Lofstrand crutches [19]–[21]. This means that the use of an internal spring within a pneumatic energy harvesting device may provide extra benefit to a subject's gait.

The purpose of the device was to allow for pneumatic energy to be accumulated and stored into an accumulator with each stroke throughout crutch gait. The spring-loaded piston pump at the bottom of a Lofstrand crutch could improve the Lofstrand crutch by allowing for pneumatic energy to be harvested during each gait cycle.

The next sections describe the design of the device used during this study and its integration within a Lofstrand crutch. It continues to explain the analytical methods used to evaluate the piston, followed by the results collected and a discussion of their interpretation. Limitations of this study and future design work are also discussed.

3.2 METHODS

3.2.1 Device Design

The pneumatic piston pump design concept proposed that a single linear piston cylinder would be inserted into the end of the crutch shaft to act like a pneumatic pump. The first prototype design had two main attachment points. At the top of the piston, there were screws that were used as pegs to restrict the piston cylinder from moving further into the crutch (Figure 32). At the bottom, the crutch tip bumper held the shaft of the piston (Figure 32 and Figure 33).The shaft of the piston was secured into the crutch tip bumper via a PVC plug. A wye connector was situated at the top of the pneumatic piston cylinder (Figure 32). The wye valve was connected to two one-way valves which allowed for inlet and outlet pressures, respectively. The inlet pressure is atmospheric; the outlet goes to the accumulator. An elastomeric accumulator consisting of an inflatable balloon encased by a rigid plastic housing was utilized in this design (Figure 34). Pressure in the accumulator is measured by a pressure transducer (AST4000A00100P3B0000, 100 psig, American Sensor Technologies, Inc., Mount Olive, NJ).

When the crutch is loaded during swing phase, the piston compresses (forward stroke) and an internal shaft compresses an internal spring while compressing fluid into the accumulator (Figure 35). At the end of a swing phase, the piston is fully compressed and pressurized air has entered the accumulator. When the crutch is unloaded, during stance phase, the potential energy in the spring is released and pushes the piston to its initial position (back stroke). During a back stroke, atmospheric pressure is drawn into the pneumatic cylinder due to a negative pressure gradient (Figure 35). With successive gait cycles, pneumatic energy is stored in the elastomeric accumulator.

For this preliminary work, two pneumatic cylinder sizes were tested with effective volumes of 12.5 ml and 2.4 ml. Effective volume was defined to be the volume from the cylinder that is compressed and enters the accumulator upon a forward stroke (Figure 35). The pneumatic cylinders used were plastic store bought medical syringes. The two sizes of pneumatic cylinders used were of 10 ml and 60 ml total volume. The same compression spring and valves were used for both of the pneumatic cylinders. Pneumatic tubing (OD: 0.5 in., ID: 0.33 in.) was attached at the outlet of the cylinder to a wye connector (Figure 36). One path of the wye connector attached to the accumulator and pressure transducer, while the other was for inlet atmospheric pressure. Due to the different sizes of the cylinders, the larger cylinder could not fit inside the crutch shaft and was actuated by hand while holding the cylinder. The larger cylinder also had more dead volume than the smaller cylinder due to a larger surface area of the piston plunger. Dead volume was defined to be the volume in the cylinder and/or line that does not enter the accumulator upon a forward stroke (Figure 35).

3.2.2 Theoretical System Description

Pneumatic models predicting energy storage were derived using the ideal gas law and assuming isothermal behaviors. These models were created in order to provide design specifications for energy harvesting assemblies (piston pump and accumulator) to power pneumatically powered orthoses. For example, a pneumatic sleeve orthosis could be used to assist with wrist posture and reduction of palmar loading during crutch-gait. The concept, evaluation, and detail of the pneumatic sleeve orthosis are outside the scope of this thesis; however, the power specifications of the pneumatic sleeve orthosis are relevant. The accumulated pneumatic energy would need to be able to fill a 35 ml container at 40 psig to allow for full actuation. The theoretical pneumatic models presented in this sub section allow a designer to specify constraints on a pneumatic energy harvesting device.

The governing equations for the simulation involve several variables.

$$V_1 = Effective\ Volume$$

$$V_2 = Dead\ Volume$$

$$V_3 = Accumulator\ Volume$$

Accumulator volume was defined to be the current volume in the accumulator.

$$P_{init} = Initial\ Pressure\ of\ V_1 + V_2\ at\ the\ Beginning\ of\ a\ Stroke$$

$$P_{x(i)} = Current\ Accumulator\ Pressure\ After\ Forward\ Stroke\ i$$

$$P_f = Maximum\ Accumulator\ Pressure$$

For this model, we were interested in finding the final maximum accumulator pressure as our governing equation. We did this by taking the case where a virtual plug was inserted on the one-way valve to the accumulator (Figure 37.1). When the fluid was pressurized within the dead volume (while the virtual plug was in place), we assume that this pressure must eventually reach an equilibrium state with the accumulator pressure (after the virtual plug was removed, Figure 37.2). At the time that the equilibrium is reached, there can be no net flow through the one-way valve leading to the accumulator, therefore this pressure would be the final pressure achieved within the accumulator. The final accumulator pressure can be derived in a stroke-by-stroke analysis, as described below.

For the case of the piston pump (energy harvester assembly), we assumed isothermal conditions and the ideal gas law which give:

$$P_{init} = 1atm$$

$$P_{init}(V_1 + V_2) = P_f V_2 \quad (1)$$

Solving for P_f:

$$P_f = \left(1 + \frac{V_1}{V_2}\right) atm \quad (2)$$

We start the derivation of Equation (2) by assuming that the system starts at atmospheric pressure. We then continue by calculating the pressure at the end of each forward stroke, but before pressure enters the accumulator. We do this by taking the case where we put a virtual plug on the one-way valve to the accumulator, until the fluid is pressurized in the cylinder, and then we calculate the state after the fluid has flown between the cylinder and accumulator to equilibrium (Figure 37). We do this for every stroke.

For the first piston stroke, the left side of the equation represents the state after the removal of the virtual plug and the right side of the equation represents the state instantaneously before the virtual plug is removed. This process is repeated for successive strokes with the pressure increasing inside the accumulator per every stroke.

First piston forward stroke:

$$P_{x1}(V_3 + V_2) = P_{init} V_3 + P_f V_2$$

The initial pressure here is 1 atm. Furthermore, we plug in Equation (2) and find

$$P_{x1}(V_3 + V_2) = V_3 + \left(1 + \frac{V_1}{V_2}\right) V_2$$

$$P_{x1}(V_3 + V_2) = V_3 + V_2 + V_1$$

$$P_{x1} = 1 + \frac{V_1}{V_2 + V_3}$$

Second piston forward stroke:

$$P_{x2}(V_3 + V_2) = P_{x1}V_3 + P_f V_2$$

Plugging in the above value for P_{x1} and the final accumulator pressure, we find:

$$= V_3 + V_2 + V_1 + \frac{V_1 V_3}{V_3 + V_2}$$

$$P_{x2} = 1 + \frac{V_1}{V_3 + V_2} + \frac{V_1 V_3}{(V_3 + V_2)^2}$$

Third piston forward stroke:

$$P_{x3}(V_3 + V_2) = P_{x2}V_3 + P_f V_2$$

Plugging in the above value for P_{x2} and the final accumulator pressure, we find:

$$= V_3[1 + \frac{V_1}{V_3 + V_2} + \frac{V_1 V_3}{(V_3 + V_2)^2}] + V_2 + V_1$$

$$= V_3 + V_2 + V_1 + \frac{V_1 V_3}{V_3 + V_2} + \frac{V_1 V_3^2}{(V_3 + V_2)^2}$$

$$P_{x3} = 1 + \frac{V_1}{V_3+V_2} + \frac{V_1 V_3}{(V_3+V_2)^2} + \frac{V_1 V_3^2}{(V_3+V_2)^3} \quad (3)$$

Continuing the same process with the updated pressure in the accumulator, we find a pattern forming and that Equation (3) can be represented as a summation from zero to infinity:

$$P_{xn} = 1 + V_1 \sum_0^n [\frac{V_3^n}{(V_3+V_2)^{n+1}}] \quad (4)$$

When $n = \infty$ the series in Equation (4) converges to our governing Equation (2):

$$P_{xn} = \left(1 + \frac{V_1}{V_2}\right) atm = P_f$$

Simulated values were calculated from the above to predict the final accumulator pressure based on varying the effective stroke volume of a pneumatic cylinder (V_1) and the dead volume (V_2) (Figure 38 and Figure 39). The simulated values were then compared to experimental values.

In order to predict the number of strokes necessary to reach an accumulator pressure at a certain volume, it was necessary to use a P-V curve for the elastomeric accumulator (Figure 40). A linear model was based off of the fill phase of the P-V curve and can be seen in Equations (5) and (6).

From the P-V curve, for accumulator pressures below 32.6 psia (225 kPa on Figure 40) the accumulator volume can be modeled as Equation (5) depicts, while Equation (6) models accumulator volume above 32.6 psia.

$$m_1 = Slope\ of\ linear\ model\ below\ 32.6\ psia$$

$$m_2 = Slope\ of\ linear\ model\ above\ 32.6\ psia$$

Finding y-intercept values of 14.7 psia and 21.7 psia for two linear equations from the P-V curve based on end points at 1x10^{-5} m^3 and 2x10^{-5} m^3 leads to the following equations:

$$V_3 = \frac{(P_{x1} - 14.7)}{m_1} \quad (5)$$

$$V_3 = \frac{(P_{x1} - 21.7)}{m_1} \quad (6)$$

From Equations (4), (5), and (6) we can now iterate the accumulator pressure and have each iteration change as the accumulator volume changes. Each calculation of Equation (4), with updated accumulator volume, represents the number of cylinder strokes (n) required to reach the current elastomeric accumulator pressure.

3.2.3 Experimental Testbed and Procedure

Data were collected using a pressure transducer (AST4000A00100P3B0000, 100 psig range, American Sensor Technologies, Inc, Mount Olive, NJ) and processed with MATLAB. The loading was done manually by loading the piston of the cylinder by hand for 12.5 ml cylinder or with body weight through the crutch for 2.4 ml cylinder. Raw pressure data from the accumulator were recorded at 125 Hz every cycle for 20-35 cycles of loading.

Dead volume was measured by filling pneumatic tubing downstream of the pneumatic cylinder with water and pouring it into a volumetric marked cylinder. The pneumatic cylinder was also compressed while the internal spring was inserted and filled with water. This volume was also measured. The volume of the downstream tubing and compressed pneumatic cylinder were combined to calculate dead volume.

The first experimental setup consisted of the pneumatic cylinder being connected directly into a pressure transducer. The 12.5 ml cylinder with approximately 3.5 ml of dead volume was pumped by hand (Figure 41). Pumping by hand was performed by the same person for all trials. During hand pumping, the end of a stroke was determined to be when the internal spring was fully compressed. When the internal spring was fully compressed the plunging shaft was released. When the spring extension was complete, the next pumping cycle could begin.

Another test was done with a 2.4 ml cylinder with approximately 2 ml of dead volume and by pressing down on the crutch with approximately 80lbs of body weight (Figure 42). For these trials, the pneumatic cylinder was internal to the crutch and affixed inside the crutch tip. Upon loading the crutch, the internal spring of the pneumatic cylinder would compress. When unloaded, the spring would decompress and allow for the next loading cycle to begin. Trials with the pneumatic cylinder internal to the crutch shaft were not tested during gait. Trials with the pneumatic cylinder internal to the crutch were tested by loading the crutch vertically on the handle with body weight.

Additionally, tests were done while using the accumulator with the same procedures as outlined above for the 12.5 ml and 2.4 ml cylinders (Figure 43 and Figure 44, respectively). The accumulator was connected by adding an additional wye valve. The additional wye valve connected the pressure outlet of the pneumatic cylinder to the accumulator and pressure transducer.

3.3 RESULTS

Comparing experimental values to simulated ones, we found similar correspondence (Figure 38, Figure 44). The maximum pressure values for a 2.4 ml cylinder with 2 ml of dead volume were found to be approximately 14.9 psig. The simulation suggested a 15.53% error with a value of 17.64 psig for the maximum accumulator pressure for a 2.4 ml cylinder with 2 ml of dead volume. Also, for the 12.5 ml cylinder with 3.5 ml of dead volume the simulation shows a maximum accumulator pressure of 52.49 psig. Experimentally, the maximum accumulator pressure for a 12.5 ml cylinder with 3.5 ml of dead volume was found to be approximately 50.5 psig, which results in a 3.79% error.

The errors found between the experimental and simulated values may be due to assumption of the ideal gas law and isothermal conditions, as well as human error in experimental volume measurements (such as measuring dead volume).

The ballooning of the elastomeric accumulator only occurred at higher pressures (around 35 psig). At lower pressures, the elastic accumulator does not balloon. At higher pressures, the ballooning causes a pressure drop in our measurements; this pressure drop was hypothesized to be due to a step volume increase in the accumulator. The decrease in pressure was regained after more strokes of the piston.

The volume of the pneumatic elastomeric accumulator during fill and exhaust cycles are shown in Figure 40. The elastomeric fill curve allows us to predict the volume in the elastomeric accumulator at a given pressure when filling or exhausting the accumulator. For example during the fill phase, if pressures reach 42 psia (290kPa), the accumulator volume is approximately 20 ml. These data allow for calculations of number of strokes needed in order to reach desired accumulator pressures.

Comparing experimental stroke number values to the simulated model for the number of strokes required to reach a desired accumulator pressure, we found similar correspondence to the P-V curve presented in Figure 40. The simulation suggested 4 strokes to reach an accumulator volume of 28.9 ml at 53.2 psia for a 12.5 ml cylinder with 3.5 ml dead volume. The P-V curve shows 45.7 psia for 28.9 ml during a fill phase of the elastomeric accumulator, which resulted in a 14.7% error.

It should be noted that the amount of force applied to the piston had little to no effect on the accumulated pressure. This was an unintentional artifact found throughout the testing process. For example, 5 strokes of a cylinder with 25 lbs of force would show a similar pressure accumulated as 5 strokes with 80 lbs of force. This was tested by varying the forces between trials. The trials presented do not measure the amount of force that the piston was loaded with.

3.4 DISCUSSION

In order to increase the maximum pressure accumulated, it was found necessary to increase the effective stroke volume or decrease the dead volume. The application specification required 40 psig to use the FREE actuator. This can be achieved through many combinations of effective and dead volumes (Figure 39). This design specification could not be achieved with the current piston pump assembly within the crutch volume. The Future Work section describes a possible design solution to increasing the effective volume of the pneumatic cylinder.

In tests using the accumulator, there was a visible pressure drop due to the inflation of the accumulator (Figure 43). It is expected that the accumulator pressure would eventually reach the pressure measured when testing without the accumulator, although it may take more strokes of the pneumatic cylinder to reach this pressure. The extra strokes needed are likely due to conversion of pneumatic energy into the elastic potential energy of expanding the elastomeric accumulator.

The simulation for calculating the number of strokes to reach a desired accumulator pressure was found to be adequate, although it should be noted that the accumulator volume presented by the simulation changes. This means that the model should only be used for accumulators with changing volume as pressure increases.

It is notable that it is highly desirable to reduce the dead volume of the system when attempting to increase the efficiency of the piston pump and reach higher accumulator pressures. Furthermore, the use of the elastomeric accumulator may be undesirable due to the pressure drop created by an expanding accumulator and the extra piston strokes needed to accommodate the added elastic potential energy of the elastomeric accumulator.

There are a variety of limitations of the mathematical models presented in this chapter. It should be noted that the accuracy of the models appear to be lower with smaller effective volumes. Caution is advised when using the models with small pneumatic cylinders. Also, the simulation does not allow the designer to change the accumulator volume, but rather calculates the accumulator volume based on the P-V curve in Figure 40. The inability to choose the current accumulator volume as a parameter should cause caution when using the model for determining design specifications. The designer should be wary of further calculations required in order to determine the pressure exhausted from the accumulator.

3.4.1 Future Work

The most feasible option for increasing the efficiency of the pneumatic cylinder may be to re-design a crutch tip that incorporates a piston pump of greater effective volume to achieve a desired final accumulator pressure needed for the current generation of pneumatic FREE actuator (Figure 45 and Figure 46). There are many design solutions for this task; however, by redesigning the crutch tip and working within the crutch shaft, the user's own downward force can be harvested into pneumatic energy via the new crutch tip. The new crutch tip would act as a piston pump, allowing the pneumatic energy to be stored in an accumulator within the Lofstrand crutch shaft (Figure 45-Figure 47).

Furthermore, there is a smooth visual and geometric transition from the traditional Lofstrand crutch to an augmented pneumatic energy harvesting crutch. It is important to have similar visual and geometric characteristics to the traditional Lofstrand crutch so users do not feel uncomfortable using the augmented crutches.

The smaller design of the piston cylinder (2.4 ml effective volume) was internal to the crutch, but it did not generate high pneumatic pressures into the accumulator. This limitation was due to a design constraint of a 0.5 in stroke length, as well as the shaft diameter being 0.75 in; thus only allowing for 2.4 ml of volume in the piston cylinder. This design achieved around 15 psig in the accumulator; however 40 psig is the desired target. In order to achieve 40 psig in the accumulator, one option could be to use a larger piston cylinder with approximately 12 ml effective volume and 4.4 ml of dead volume (Figure 39). These specs were suggested by the aforementioned simulation and estimates of how much dead volume is expected for the new crutch tip design (Figure 39).

The proposed future tip design incorporates a linear bearing inside a custom crutch tip. The linear bearing slides on the outside of the crutch shaft (Figure 45-Figure 47). There is a piston cylinder inside the customized crutch tip which is held inside the shaft by a pin. The pin holds back the piston cylinder from going inside crutch. Upon a forward stroke, the crutch tip moves while the piston cylinder stays static. Conical internal springs are utilized inside the crutch tip to return the crutch tip to its original position. The conical springs decrease dead volume (as opposed to compression springs). There are inlet and outlet ports at the bottom of the crutch tip with one way directional valves to control flow from the atmosphere to the accumulator. The accumulator is internal to the crutch and is used to store the pneumatic energy (Figure 34). It is highly recommended that a 3D printed prototype is made before a machined or molded final version of the crutch tip.

Hardware required for the new tip design include: drill press, linear bearing, conical springs, torsional spring/clamp, two one way valves rated to at least 50 psig, metal pegs, piston cylinder, o rings, customized tip, wye connection, accumulator, and pneumatic tubing rated to at least 50 psig. This list is not extensive; however, these items will make it possible to construct a working prototype of the customized crutch tip.

Extra hardware which might be necessary for the new tip design to be integrated with the FREE actuator include: proportional valves to control flow to sleeve and an accumulator pressure release valve. This list is not extensive.

There are several potential issues of the new crutch tip design, which include: leaking at the piston O-ring interface, linear bearing sliding efficiently, clamping to the linear bearing without slipping, weak conical springs, inlet and outlet nipples, and changes in the accumulator volume.

For leaking at the O-ring interface, it should be noted that, with 3D printed material, there should be a metallic interface with the O-ring. This metallic interface helps create a better seal. For clamping to the linear bearing, there are many possible options such as torsional springs, clamps, and elastic bands. The author suggests using clamps to secure the customized crutch tip to the linear bearing as it slides along the crutch shaft. Conical springs with enough potential energy to move the crutch tip when compressed may be difficult to find; however, utilizing the uncompressed space in dead volume calculations may allow the design to accommodate springs that do not completely compress. The inlet and outlet nipples at the bottom of the customized crutch tip may need to be modified to allow for off-the-shelf fittings to be attached, these fittings will allow for greater usability with the accumulator and FREE actuators. The accumulator volume may need to change and be larger or smaller than for the current design. The changes in accumulator volume can be addressed in several ways: by allowing a larger accumulator within the shaft of the crutch, by switching to a non-elastomeric accumulator, or by increasing the thickness of the elastomeric tubing and upgrading the adhesives to allow for higher pressures to be stored within the current elastomeric accumulator design.

3.5 CONCLUSION

Pneumatically powered orthoses can greatly benefit from energy harvesting devices, which can replace bulky pneumatic canisters. The development of a pneumatic energy harvesting device was proposed for use within a Lofstrand crutch for powering pneumatically powered active orthotic devices. The specifications and advantages of a pneumatic cylinder within a Lofstrand crutch were discussed in this chapter. A pneumatic cylinder within a Lofstrand crutch was determined to be adequate for powering FREE actuator specifications of 35 ml and 40 psig. It was also shown that pressures and volumes achieved by a pneumatic cylinder and elastomeric accumulator can be much higher than those specified by the proposed FREE actuator. Mathematical models were derived to determine design specifications for a pneumatic cylinder. The models were shown to predict elastomeric accumulator pressure with accuracy up to 3.8% error. The accumulator pressure model created allows for design specifications to be identified for pneumatically powered orthoses. Another model to predict the number of strokes needed to reach a particular accumulator volume was created. This model also predicts the current accumulator volume per stoke of the pneumatic cylinder.

Figures:

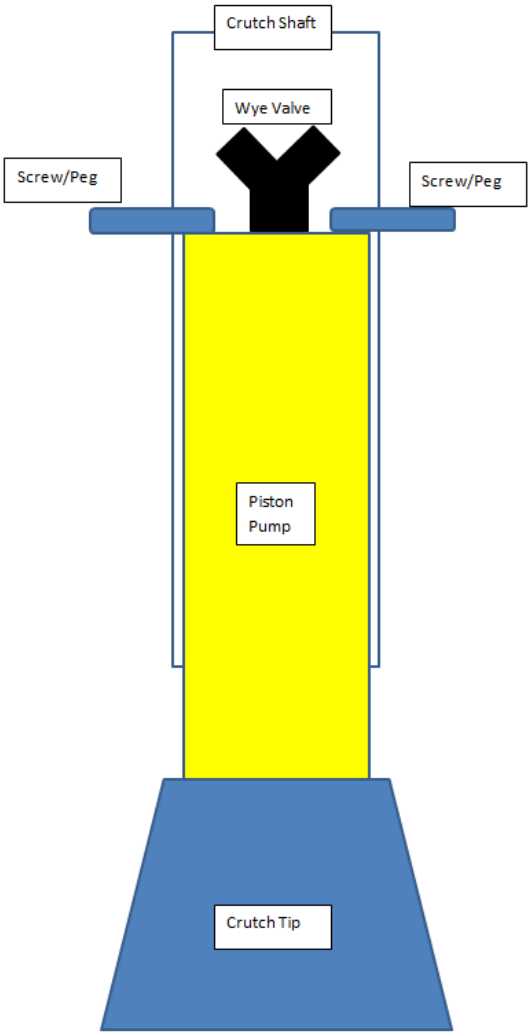

Figure 32: Piston pump configuration inside crutch shaft

Figure 33: Pneumatic cylinder (10 ml total, 2.4 ml effective volume) inserted into a crutch tip bumper

Figure 34: Elastomeric accumulator

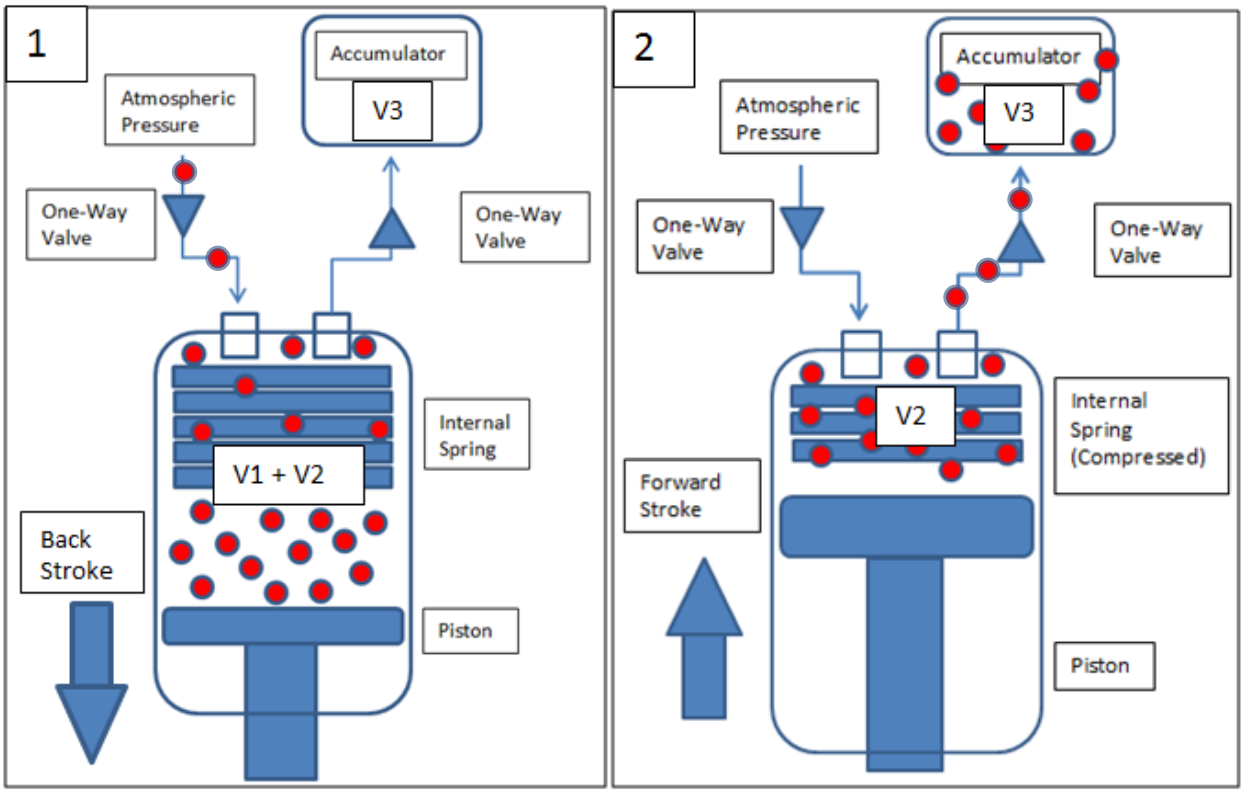

Figure 35: Back stroke (1) and forward stroke (2) of piston cylinder. Red circles represent air flow.

Figure 36: 60ml and 10 ml syringes with wye valve and pneumatic tubing

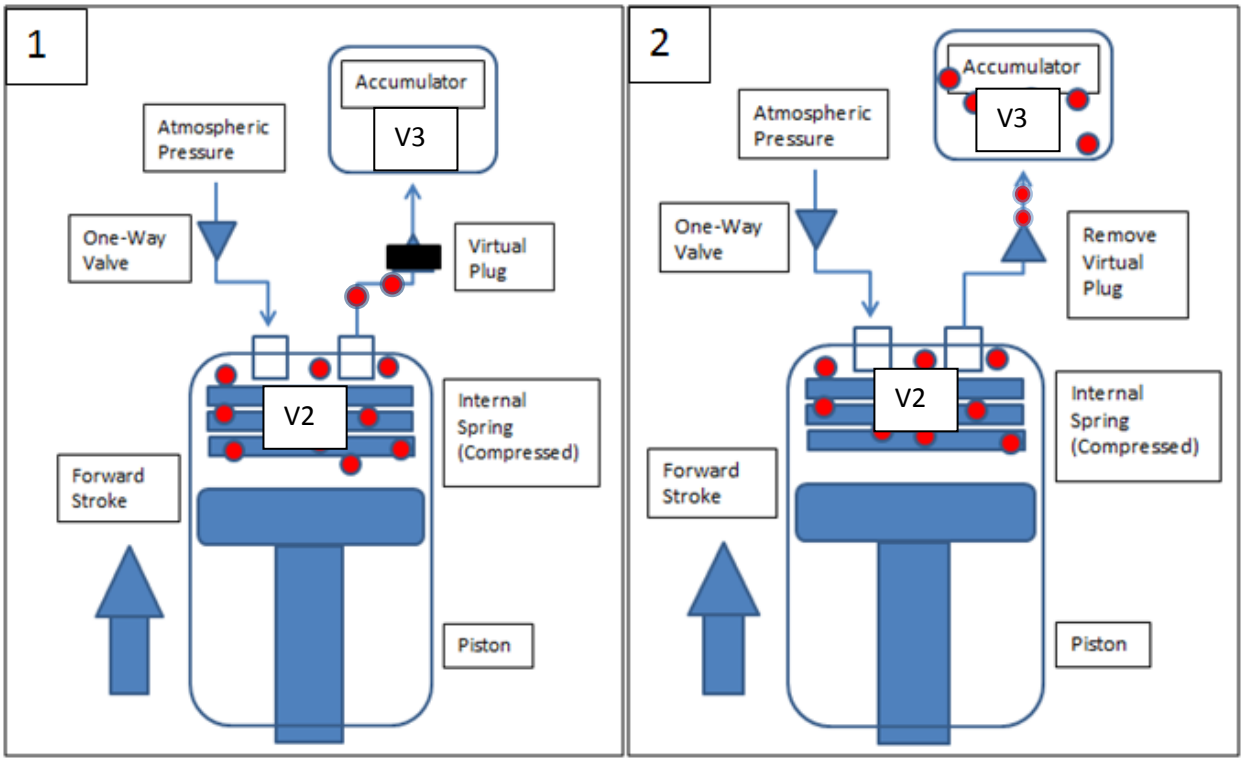

Figure 37: Forward stroke of piston showing the effect of a virtual plug between calculation steps.

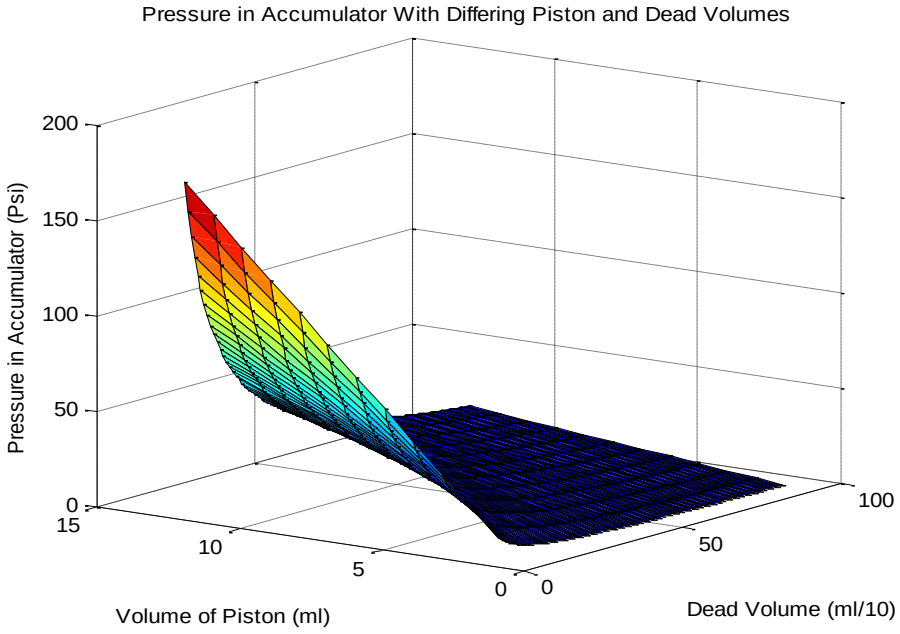

Figure 38: Simulated 3 dimensional view of rising accumulator pressure based on dead volume and effective stroke volume of the piston cylinder.

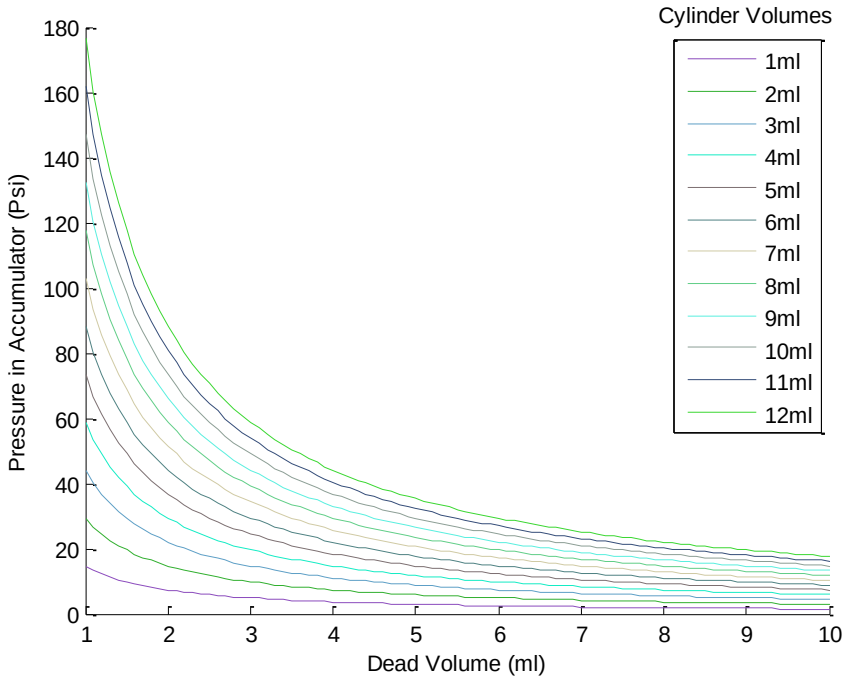

Figure 39: Simulated pressure in the accumulator as a relation of effective cylinder volume and dead volume.

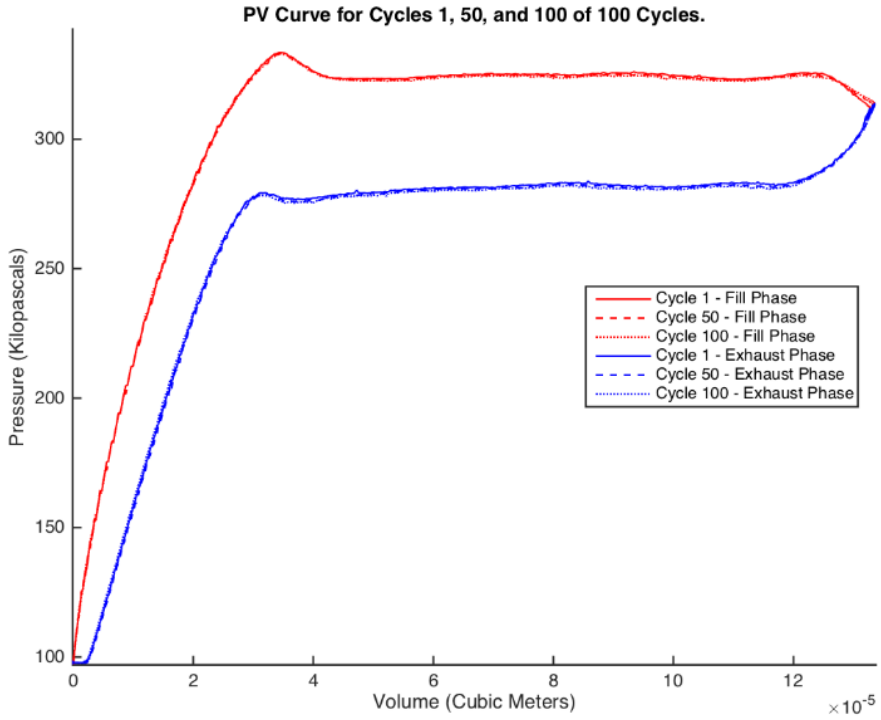

Figure 40: P-V curve for elastomeric accumulator (courtesy of Josh Cummings from Vanderbilt).

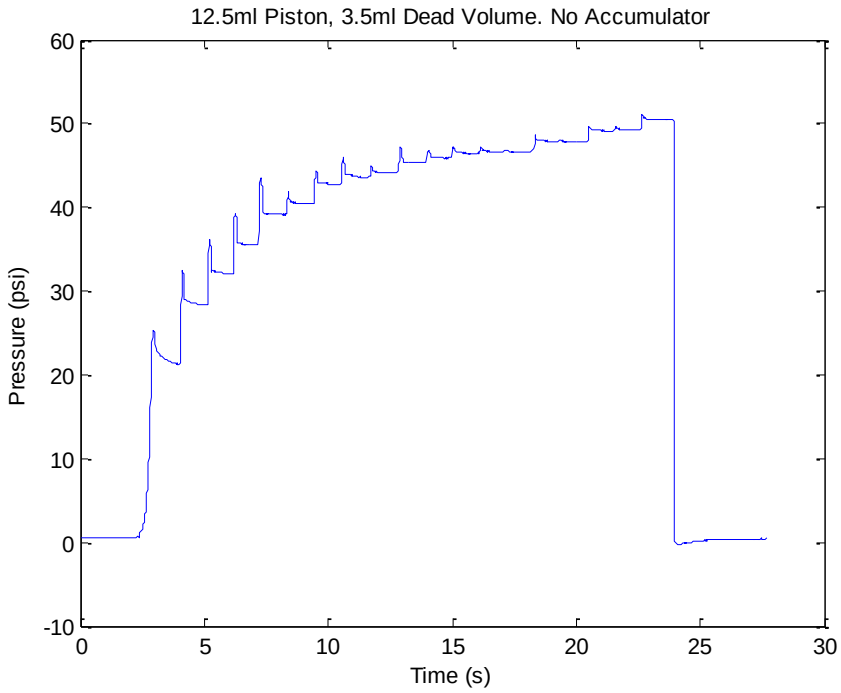

Figure 41: Pressure vs. time of a 12.5ml piston cylinder with 3.5ml dead volume pumped into a pressure transducer without an accumulator

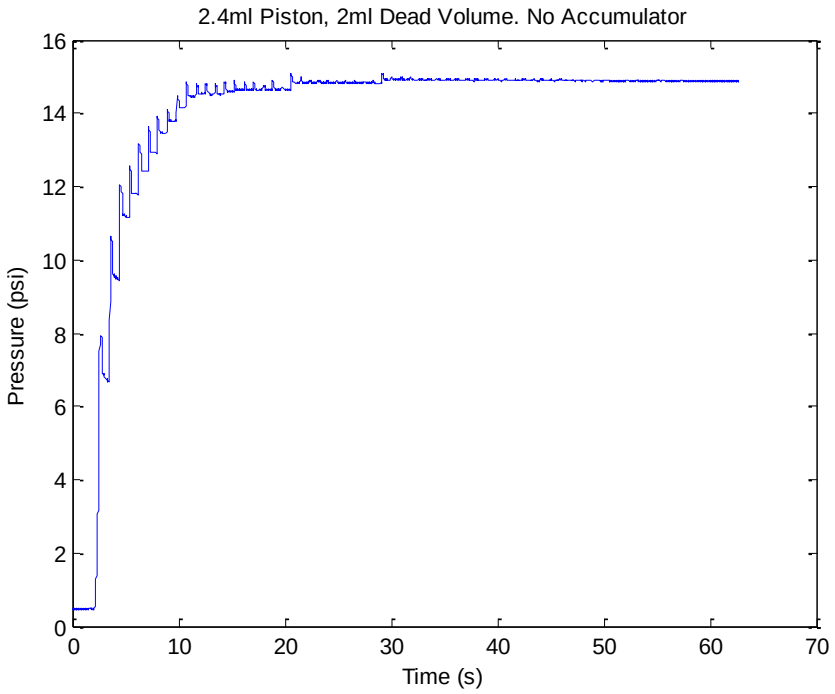

Figure 42: Pressure vs. time of a 2.4ml piston cylinder with 2ml dead volume pumped into a pressure transducer without an accumulator

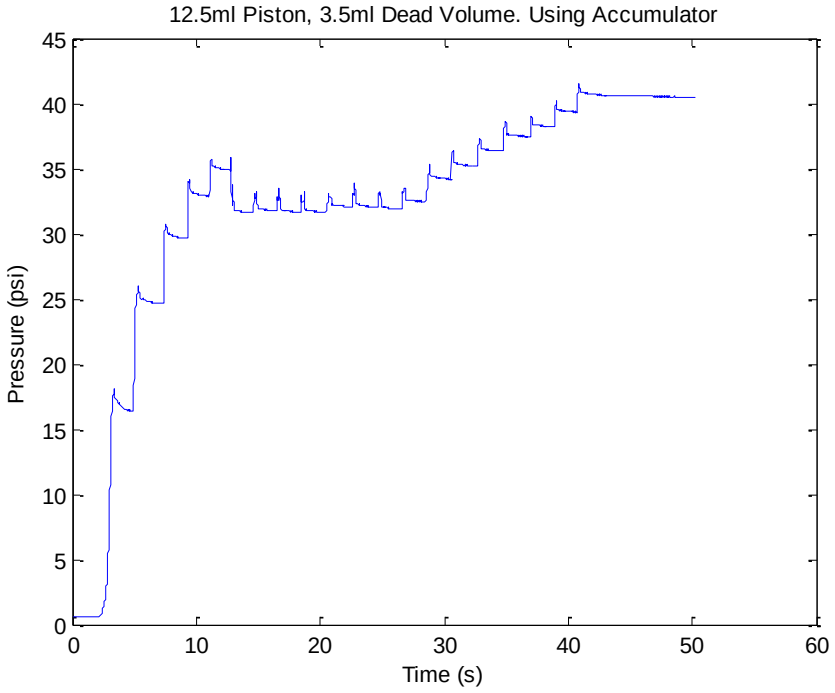

Figure 43: Pressure vs. time of a 2.4ml piston cylinder with 2ml dead volume pumped into an accumulator. Visible pressure drop from expanding accumulator.

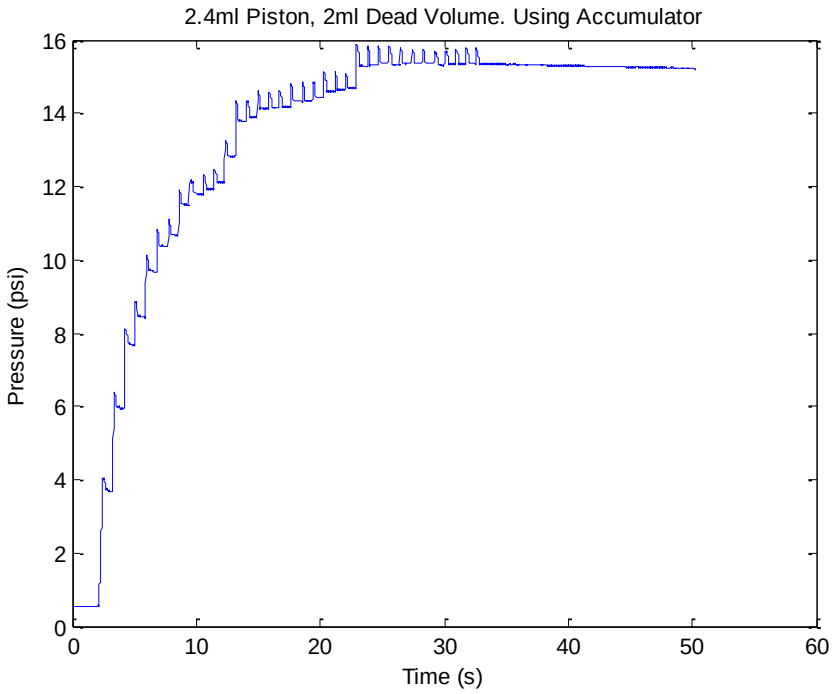

Figure 44: Pressure vs. time of a 2.4ml piston cylinder with 2ml dead volume pumped into an accumulator. Visible pressure drop from expanding accumulator.

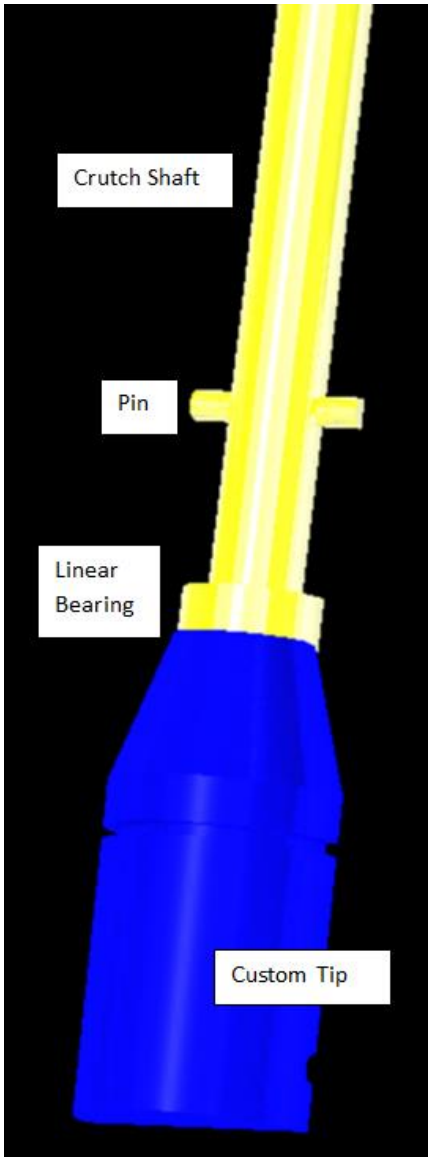

Figure 45: CAD drawing of customized crutch tip assembly.

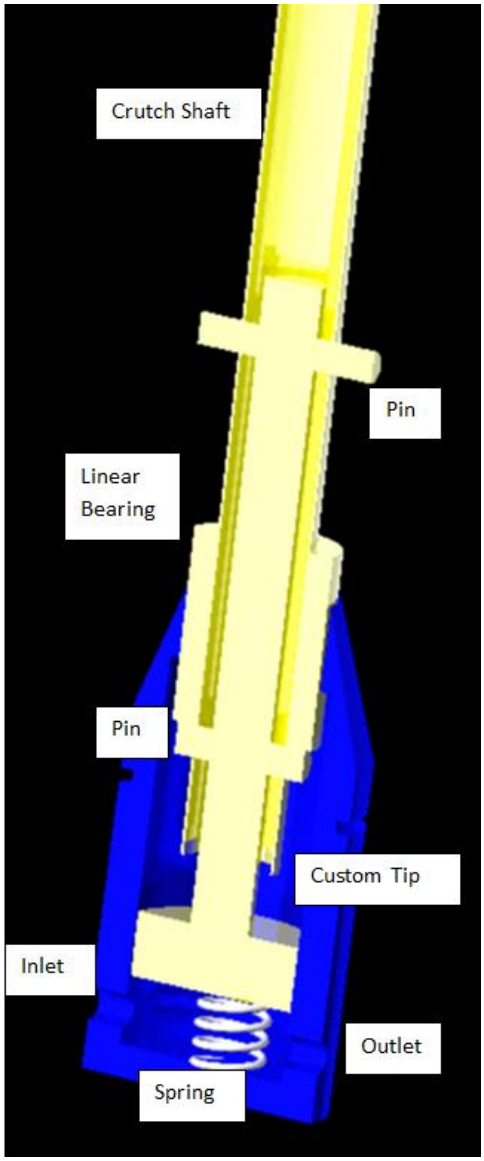

Figure 46: Section view of customized crutch tip assembly.

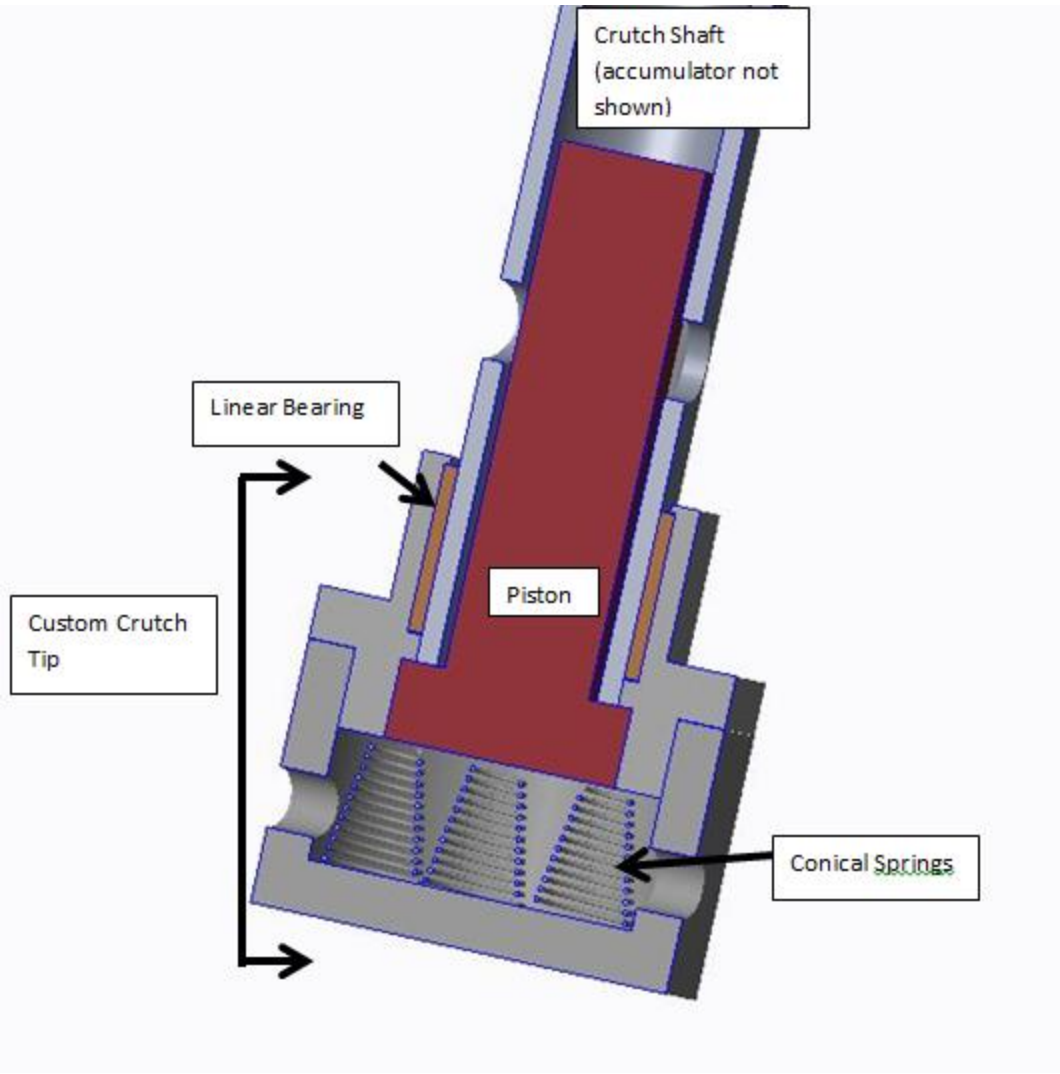

Figure 47: Section view of customized crutch tip assembly

4: CONCLUSIONS

4.1 REVIEW OF FINDINGS

4.1.1 Wrist Orthosis

Chapter 2 described the design of a wrist orthosis to be used during Lofstrand crutch gait. The design was contoured to the natural curvature of the forearm and had a strap to allow for gentle compliance to accommodate some extension of the wrist during crutch gait. Weight and speed of release design specifications were met for the orthotic device. The intent of the device was to decrease wrist hyperextension and redirect loads away from the carpal tunnel region of the palm during the crutch gait cycle.

Two out of ten tested subjects were analyzed for this thesis. Each subject performed 10 swing-through gait cycle trials without and with the orthosis attached to the crutch handle. Joint kinematics were captured with a 15 camera VICON system. Palmar pressure data on the handle of the crutch for the dominant hand were captured using Novel Pliance software. It was found for these subjects, on average, the orthosis reduced total force, maximum force, contact area, mean pressure of the hand, and wrist extension during swing-through gait. Peak pressure was observed to increase with the use of the orthosis for these subjects. Peak pressure was also observed to move toward the adductor pollicis area of the palm and away from the carpal tunnel region.

The intended goals of the upper extremity orthotic device were achieved for the two subjects analyzed in Chapter 2. Future work will include analysis of the remaining eight test subjects to confirm that the design will redirect carpal tunnel loads, reduce palmar loads, and improve wrist posture.

4.1.2 Pneumatic Energy Harvesting Device in the Lofstrand Crutch Tip

Chapter 3 described the design of a pneumatic energy harvesting device to be used in tandem with the tip of a Lofstrand crutch. A mathematical model was created to determine the amount of pressure stored in an elastomeric accumulator based on various design constraints of the pneumatic cylinder (effective volume and dead volume). Another model was created to determine the amount of cylinder strokes necessary to reach a desired volume within the pneumatic cylinder.

There are three design constraints for future construction of pneumatic energy harvesting devices: the effective volume of the pneumatic cylinder, the dead volume of the system, and the accumulator volume. With these constraints known, it is possible to design a pneumatic piston pump to fit the pressure and volume specifications of a variety of applications. The plausibility of a new pneumatic piston pump integrated into the tip of a Lofstrand crutch was discussed.

4.2 EXPANSION ON LIMITATIONS AND FUTURE WORK

Limitations and future work for the wrist orthosis were discussed in section 2.4; however, long term work on the project was not discussed. In addition to what was discussed in section 2.4, it is important to discuss the marketability of the orthotic device. For the device to be marketable, there are several

design changes that should take place. The first being an upgrade to the hardware used to attach the orthosis to the Lofstrand crutch. The current screws and worm-gear clamp are not customized to fit the orthosis, but rather the orthotic design was compromised in order to be attached securely to the Lofstrand crutch. Furthermore, the metallic edges of the worm gear clamp and screws are exposed. A new attachment system to the Lofstrand crutch is necessary in order to allow for a marketable product. The forearm strap should also be riveted to the orthosis to securely hold it in place. Injection molding is recommended as a manufacturing process to allow the orthosis to be mass produced. Finally, aesthetics should especially be considered for a finalized marketable product.

In addition to marketability of a passive orthotic design, further design steps may be considered to create a more advanced pneumatic system FREE actuators. This novel device would allow for a soft robot to not only move in one direction, but also twist, bend, and rotate multiple segments, using unique pneumatic pathways. These pathways would allow for corrective forces and torques to be applied to the upper extremities. This design could be applicable to other upper extremity motions such as reaching, feeding, or even manual wheelchair propulsion.

Future work for the pneumatic energy harvesting cylinder was discussed in section 3.4. A new crutch tip design was discussed and design specifications were recommended. One limitation of the study was the lack of experimental data points to verify the model output. Three dimensions were modeled (effective, dead, and accumulator volumes), while only two experimental data points were used in determining the effectiveness of the model. A stronger study would include more experimental data points to verify the model created. Furthermore, the model itself approximates the accumulator volume with two linear fits. Experimental data shows that the volume in an elastomeric accumulator expands nonlinearly. For a more accurate model of accumulator volume, a nonlinear representation of experimental data should be created.

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APPENDIX

Due to availability of sensors early on in the project, several pilot tests were performed. The first pilot test used two Flexiforce FSRs (A201, Tekscan, South Boston, USA) adhered to a subject's adductor pollicis and carpal tunnel of their dominant hand (Figure 19). A custom electrogoniometer was used to measure wrist flexion and extension. A subject would practice for two minutes with a traditional Lofstrand crutch with unrecorded data. Data were recorded for five trials of walking with the orthosis down a 40 ft tiled indoor pathway with optional rest between trials. After the first five trials, a five minute rest was given. Following the rest, data were recorded for five more trials of walking without the orthosis down a 40 ft tiled indoor pathway with optional rest between trials was administered. Data were recorded in ADC values; therefore only relative magnitudes were compared.

Preliminary FSR data showed an increase in loads over the carpal tunnel when using the orthosis (Figure 48). However, it was believed that there was misplacement of the FSRs during the trials, which may have not fully been in contact with the carpal tunnel region for trials without the orthotic device. This assumption was made due to periodic 'jumps' in ADC value read by the FSR (Figure 48). These 'jumps' were sudden increases in ADC value. These 'jumps' were due to contact being made by the sensor touching the handle, which would mean that there was little contact of the sensor before the 'jump.' This would mean that FSR sensor placement around the carpal tunnel region is extremely important to be accurate; otherwise the sensor would not make contact. In order to test this assumption, single swing-through gait cycles were tested on two subjects by placing FSR's on different areas of interest on the palm without and with the orthosis. Areas of interest are labeled in Figure 19 and are referred to as the carpal tunnel, palmar aponeurosis, mid palm, and adductor pollicis. These areas were systematically tested and key results are shown in (Figure 49). The results indicate up to three times reduction in loads along the mid-palm area when using the orthosis for a single subject. Data trials that tested the carpal tunnel area without the orthosis showed little to no loading at all (Figure 48). This was observed to be due to a lack of contact between the sensor and crutch, confirming the initial assumption.

The electrogoniometer used was developed in the Human Dynamics and Controls Laboratory (HDCL) at the University of Illinois and has not been validated for dynamic testing. However, the results may still be of interest (Figure 48). Over several gait cycles, it appeared that the angle of extension increased (decreasing ADC values) without the orthosis; while with the orthosis, there was a gradual change in angle toward the normal position, which is about 30-40 on the ADC scale (Figure 48). These results may indicate that wrist extension can change over a period of several gait cycles. This effect may be due to fatigue.

A second set of preliminary data was collected at the University of Wisconsin Milwaukee using two 4x4 cm. Novell Pliance matrix sensors, model number S2052. The procedure was similar to the aforementioned procedure with Flexiforce FSRs except no angle data were recorded during these sessions. The purpose of these preliminary tests was two-fold: to test the effectiveness of different sensor locations relative to the palm and crutch handle, and to assess preliminary results of the wrist orthosis.

It was decided that attaching the two Novell sensors directly to the dominant palm was the most effective placement in terms of determining the location of different loads (Figure 50). These trials held the clearest results with respect to the questions of interest for this study. There was also the added benefit of not needing to mask the location of the lower aponeurosis and adductor pollicis, since the sensors were directly placed on a subject's hand. Throughout this process, data were collected both unilaterally and bilaterally without and with the orthosis. The only trials of interest included here are the 'unilateral dual' sensor trials (Figure 51-Figure 54).

These data indicated a reduction in pressure along the carpal tunnel region of the palm, as well as the mid-palm and palmar aponeurosis regions when the orthosis is used. Maximum pressure decreased from 560 kPa to 383 kPa with and without the orthosis respectively (Figure 51-Figure 54). There was also an increase in load in the area of the adductor pollicis when the orthosis was used (Figure 52 and Figure 54). These results indicate a transfer of load from the carpal tunnel region of the palm to the region of the adductor pollicis during swing-through gait when using the orthosis. Also notable was the pattern of center of pressure (COP) of the palmar regions without and with the device (Figure 52 and Figure 54). With the orthosis, the swept area was contained to a smaller region, which may indicate an increase in stability or a decrease in wrist extension during swing-through gait. It should be noted that peak pressures may be higher for certain trials or subjects when using the orthosis; however, due to the change in location of the load, it may cause a decrease in the occurrence of carpal tunnel pressure [4]. These data suggest promising results for the use and further testing of the orthosis.

A third set of preliminary data was collected at the University of Wisconsin Milwaukee using two Novell Pliance matrix sensors. Testing was done in a similar way to the previous testing sessions; however, the walkway was shortened to about 20 ft. During testing one of the sensors was found to have been damaged and began to have 'time out' errors, which eliminated a row of data. However, data recorded were still consistent with previous data collected with Novell Pliance sensors without and with the orthosis (Figure 55 and Figure 56). The author still sees these data as valid, it can be seen that the 'time out' error occurs in the dark blue row with no pressure values when the device is in use (Figure 55 and Figure 56). This error simply meant that no data were collected from that row; however, with the surrounding rows showing little pressure over the carpal tunnel region, it can be assumed that the missing row would have a similar pattern. The peak pressure in the area of the adductor pollicis increased while using the orthosis from 272.5 kPa to 322.5 kPa, while the peak pressure in the carpal tunnel area decreased from 272.5 kPa to 190 kPa. Peak loads show a similar trend (Figure 55 and Figure 56). These results follow the previous data collected in that there was an increase in pressure in the area of the adductor pollicis, while there was a decrease in pressure along the carpal tunnel area of the palm when using the orthosis. These results illustrate a displacement of loads experienced at the carpal tunnel region of the palm toward the adductor pollicis, which was a design goal of the device.

Figures:

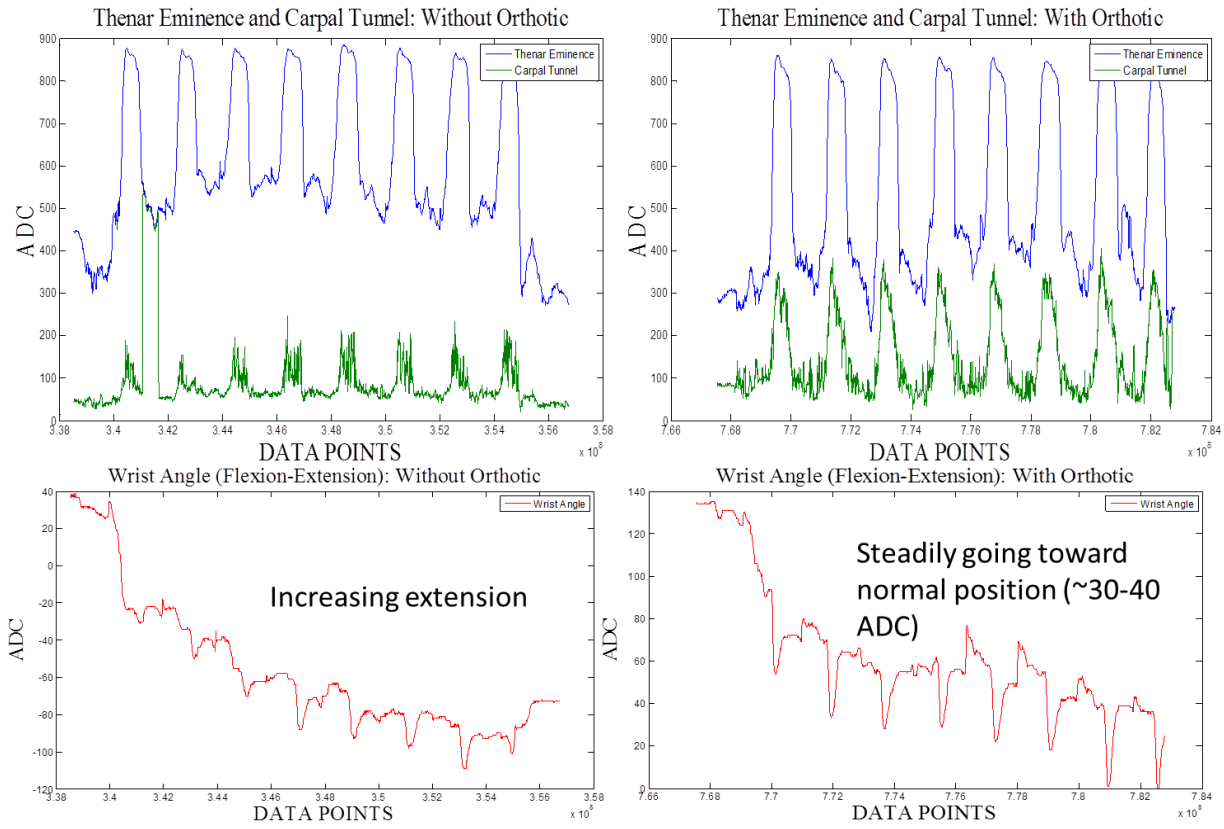

Figure 48: Preliminary data using an electrogoniometer to measure wrist flexion-extension, as well as FSRs at the adductor pollicis and carpal tunnel.

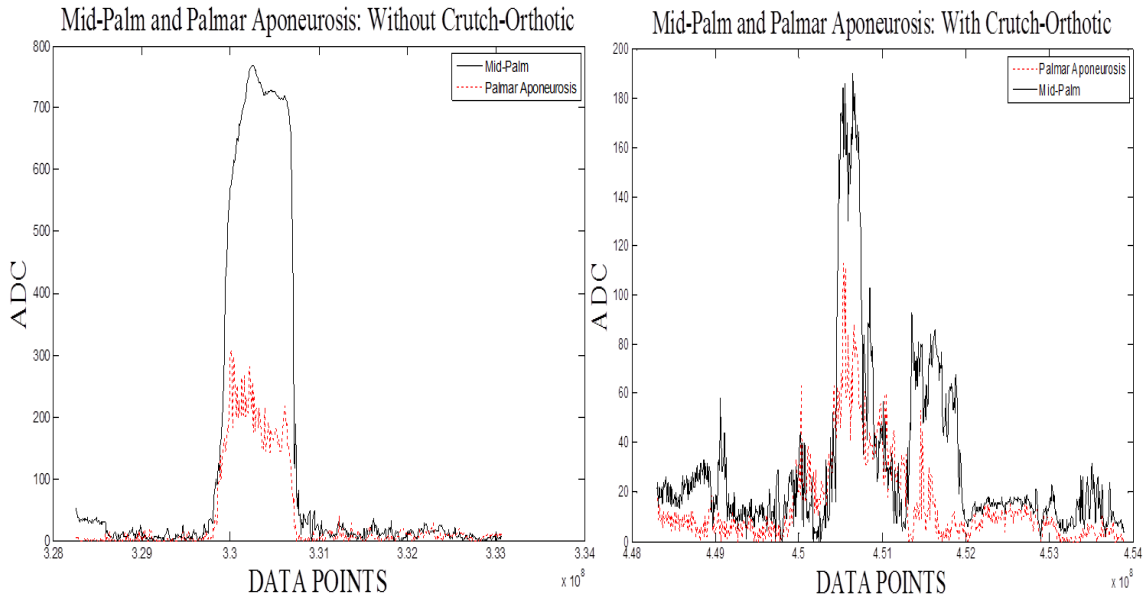

Figure 49: Data from one subject using FSRs at the mid-palm and lower palmar aponeurosis, with and without the orthosis

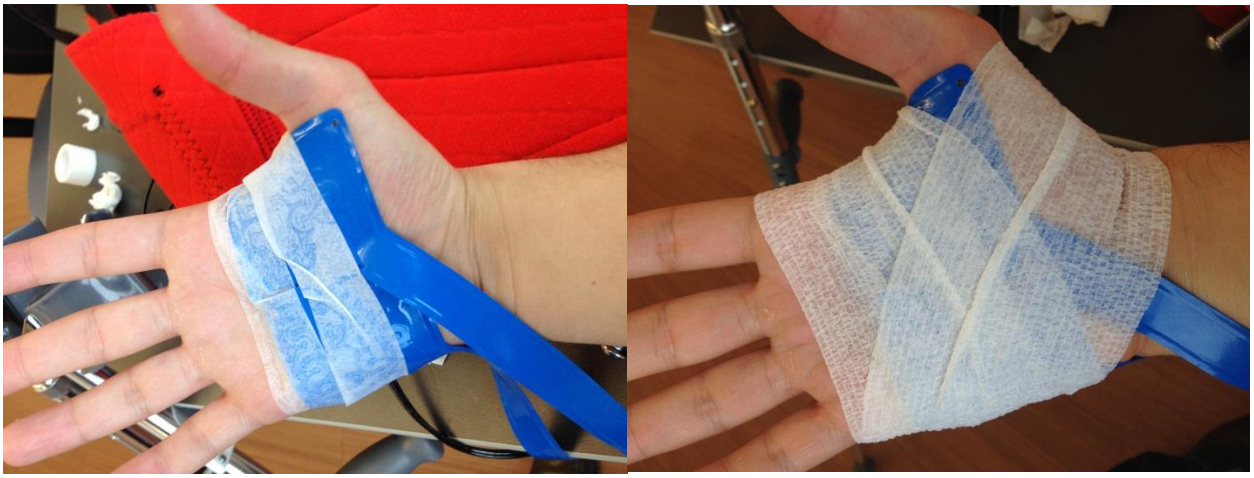

Figure 50: Location of Novell Pliance sensors on subject's palm. Placed on radial and ulnar sides of palm.

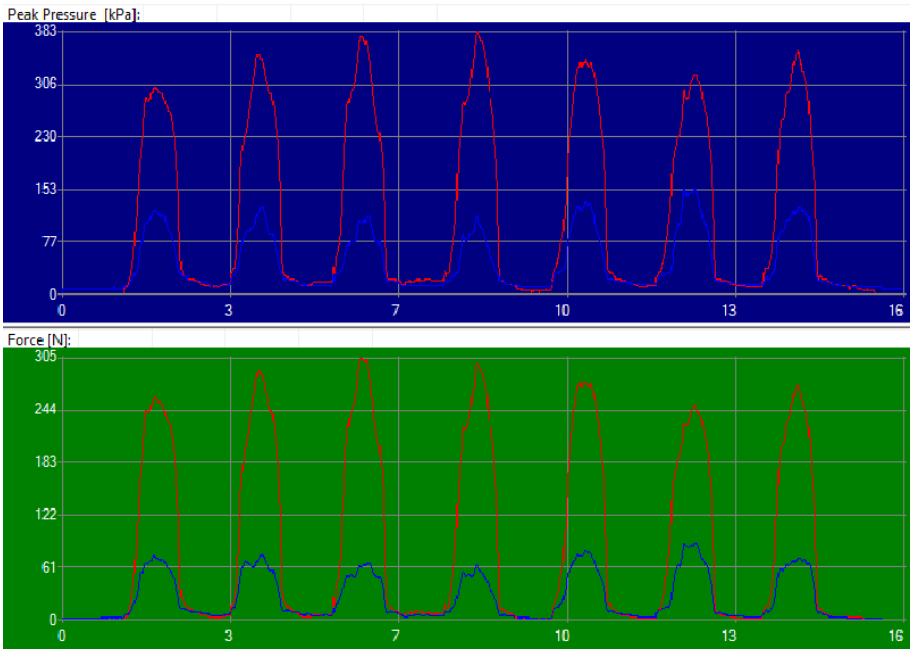

Figure 51: Dual sensor unilateral pressure and force over time: With orthosis. Red: Radial sensor. Blue: Ulnar sensor.

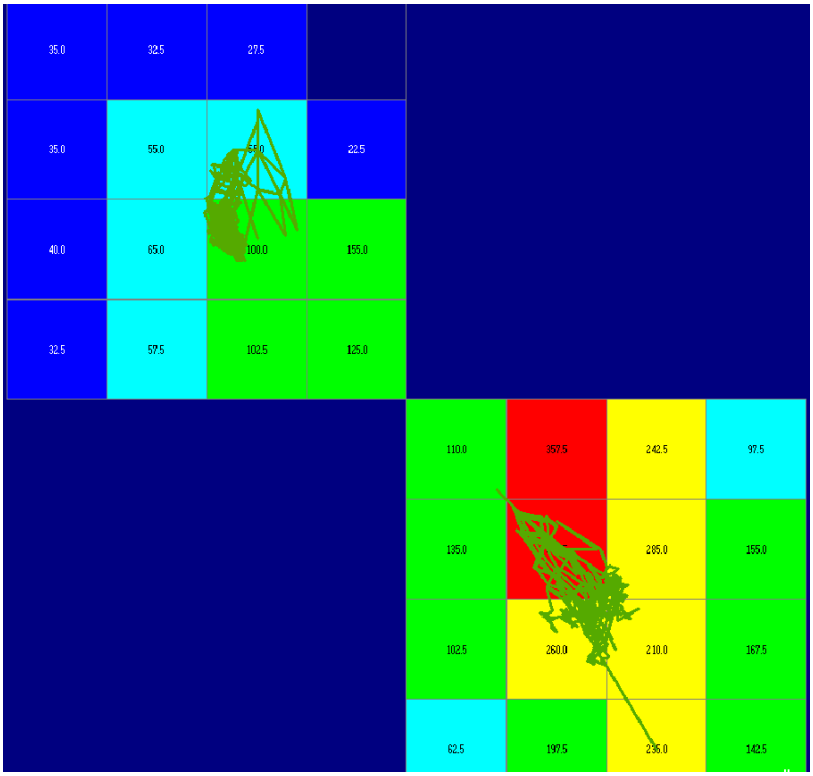

Figure 52: Dual sensor unilateral maximum pressure grid and COP swept path: With orthosis. Corresponding to Figure 51. Right: Radial sensor. Left: Ulnar sensor.

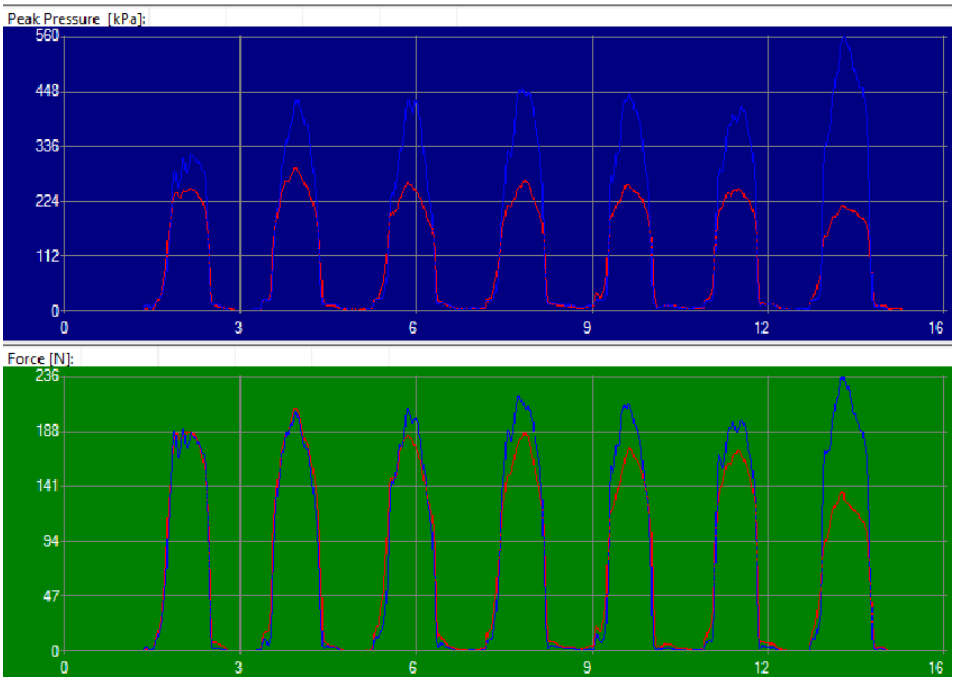

Figure 53: Dual sensor unilateral pressure and force over time: Without orthosis. Red: Radial sensor. Blue: Ulnar sensor.

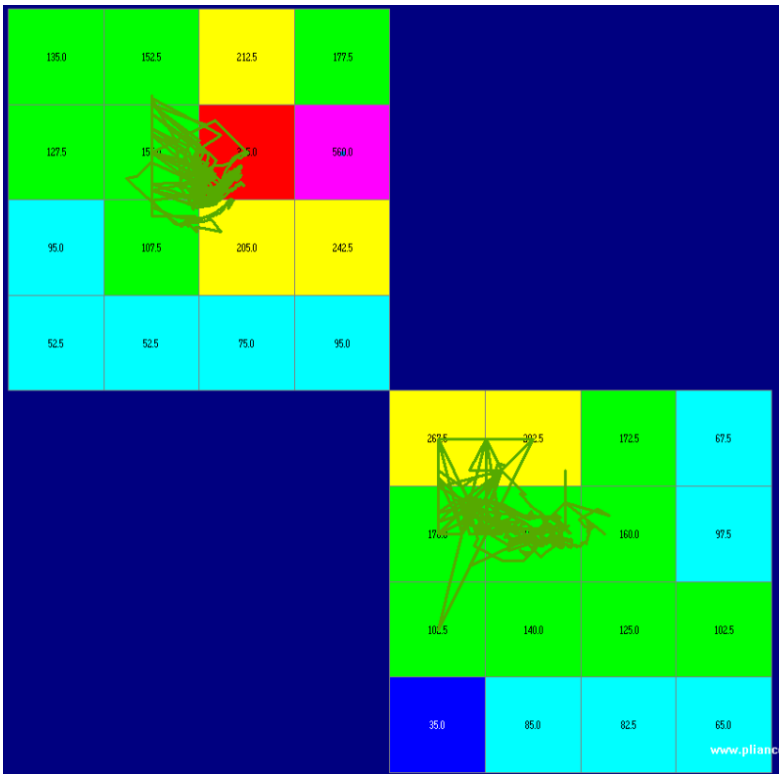

Figure 54: Dual sensor unilateral maximum pressure grid and COP swept path: Without orthosis. Corresponding to Figure 53. Right: Radial sensor. Left: Ulnar sensor.

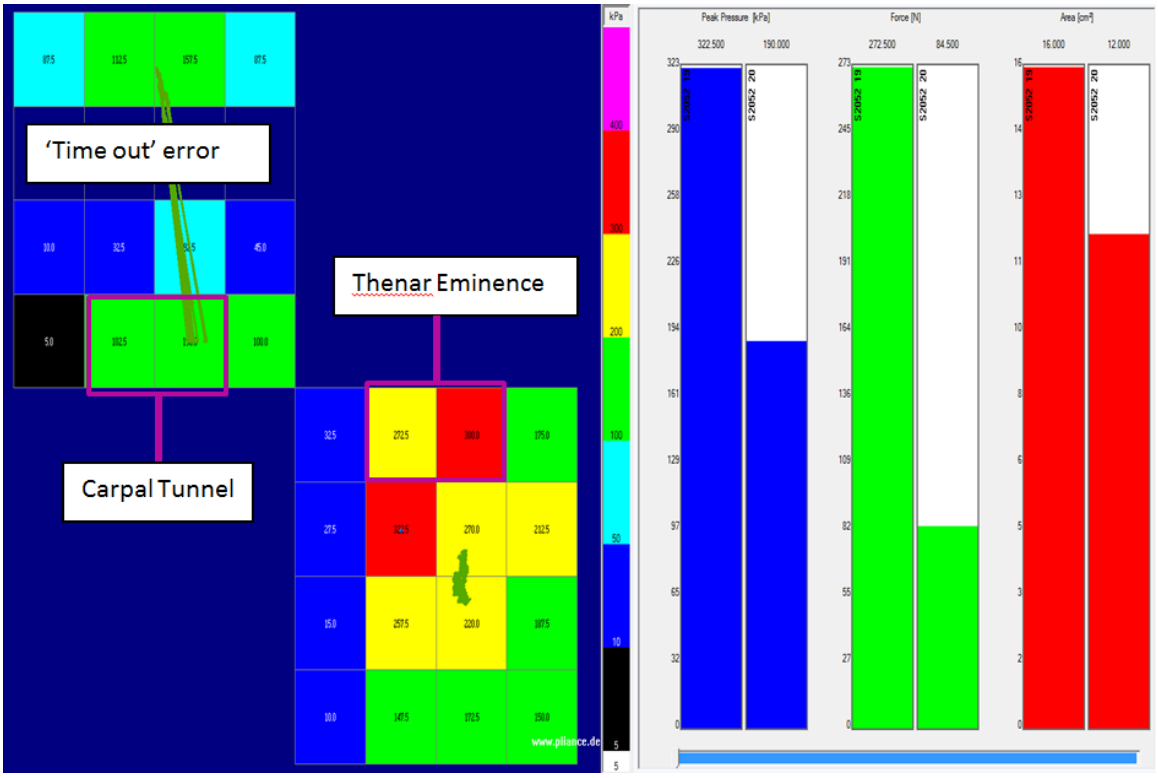

Figure 55: Peak pressure and force data using Novell Pliance sensors with orthotic device. Row without values is 'time out' error. Right: Radial sensor. Left: Ulnar sensor.

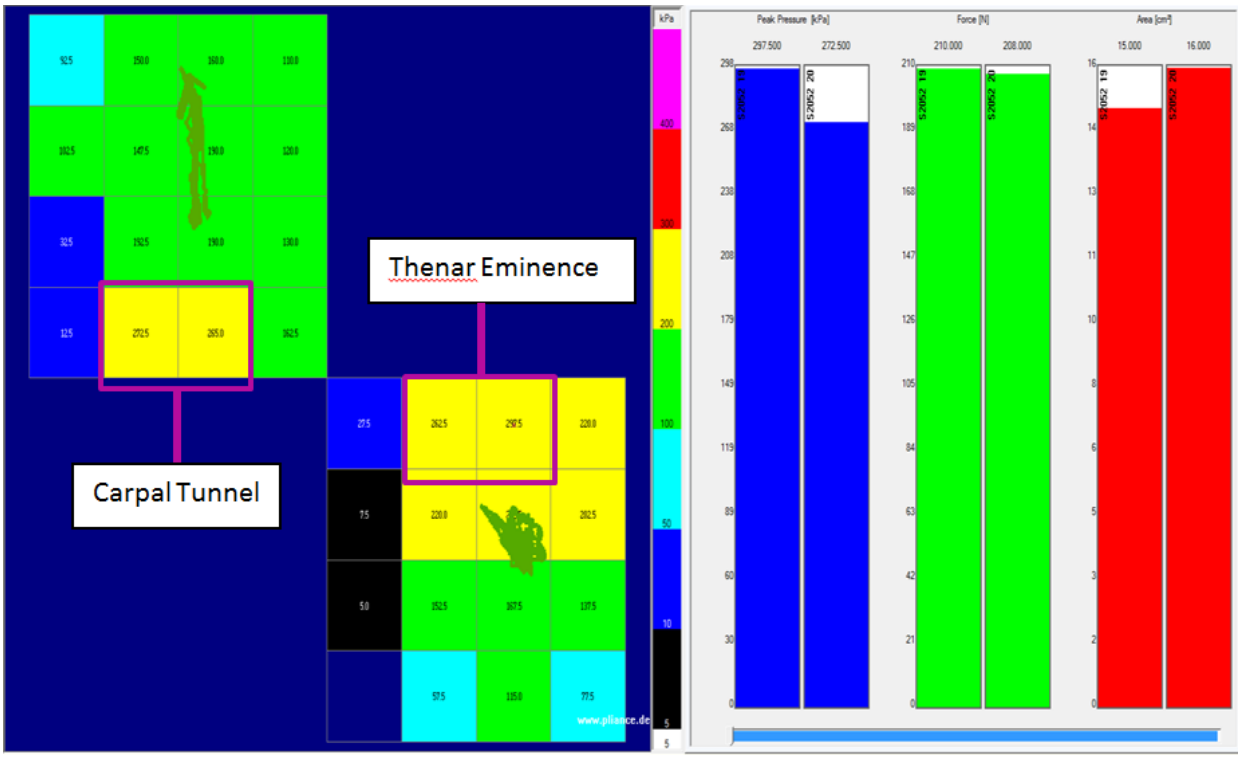

Figure 56: Peak pressure and force data using Novell Pliance sensors without orthotic device. Right: Radial sensor. Left: Ulnar sensor.