DESIGN AND VALIDATION OF WEARABLE TECHNOLOGIES FOR CHILDREN WITH PHYSICAL DISABILITIES

by

Ben Greenspan

A dissertation submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Doctor of Philosophy in Biomechanics and Movement Science

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ABSTRACT

This dissertation aims to design and validate novel wearable technologies for children with physical disabilities. The first aim is to develop and test the validity and reliability of technology that could allow for better measurement of movement across prolonged periods of time and a variety of natural settings. The second aim is to develop and test the feasibility, reliability, and validity of a smart wearable to monitor infant positioning in early development when it is determined by parents and impacts future developmental outcomes. The third aim is to develop and test the effectiveness of an assistive device that can allow for easier human computer interaction for video game play. Therefore, this dissertation develops and tests a variety of technologies with the potential to impact rehabilitation by improving measurement of movement across time in natural settings, providing feedback for intervention to change end users' behaviors in ways that could positively impact developmental outcomes, and altering daily tasks to increase independence and participation for people with physical disabilities.

Chapter 1

INTRODUCTION

1.1 Introduction

The purpose of this dissertation was to develop and test novel wearable technologies with the potential to improve monitoring and performance of physical activity for children with physical disabilities. These technologies were designed to monitor movement or to allow for control of external stimuli via movement and have potential for use outside of laboratory settings within natural settings. According to the 2010 Census, in the US there are 5.2 million children under 15 years old with some kind of disability. Of those 5.2 million, 980,000 specifically have some form of physical disability or movement impairment [1]. There are a variety of intervention techniques and rehabilitative technologies designed to improve function and participation for these children. However, it is difficult to quantify how effective these techniques and technologies are, because most testing and data collection methods can only be performed in a laboratory setting or in the presence of a therapist or experienced researcher. This limits our knowledge to only a snapshot of the child's performance, leaving out important influences of the child's everyday environment. Thus, there was a need for technology that could monitor and possibly alter physical activity in natural settings for children with physical disabilities. This technology could provide a broader view of the child's activity in the natural environment. It could be used:

1) to track behavior for the purpose of early identification of delays or to measure the effectiveness of intervention, 2) to provide feedback to users to encourage or alter the performance of specific activities, and/or 3) to provide a mechanism through which users can control their external environment for increased participation and independence. All of the devices in this project were made using open sourced electronic components and very heavily incorporated user feedback during all prototyping iterations to make more accessible and better rehabilitation tools. The projects associated with the three aims highlight ways that open source wearable technology can be used in rehabilitation for individuals with physical disabilities to monitor activity, change activity by providing feedback, or to control external devices.

1.2 Specific Aims

The aims for this dissertation were:

Aim 1: To design and validate a smart compression garment for measurement of upper extremity physical activity. Currently, visual motion capture is the gold standard for recording kinematic or physical activity data. This type of data collection can only be performed in a laboratory environment or using expensive equipment. To address this problem, motion capture through inertial measurement units (IMUs) attached to clothing is becoming more common, but still requires rigid sensors to be fixed to each body segment [2,3]. Building off of Gioberto and Dunne's work [4], we will determine the optimal integration pattern for a conductive thread stitched stretch sensor. This sensor will then be strategically stitched across the elbow and shoulder joints in a

compressive garment allowing for a comfortable form of motion capture. A custom algorithm will be written in LabVIEW to translate stretch levels into orthopedic angles. Data will be validated against the TrakSTAR motion capture system. This type of stretch sensor may provide a comfortable, more affordable method to monitor physical activity in the natural environment, and could serve as an input to control external devices.

Aim 2: To design and validate a device to track the amount of time infants are in distinct positions. From Lobo and Galloway's work [3], it was found that infants exposed to advanced handling techniques (more time spent upright and in prone, or "tummy time") reached developmental milestones sooner [6,7,8]. Despite this knowledge, there have never been quantitative data recorded on the amount of time infants are placed in distinct positions throughout the day. To collect these data, we will develop a comfortable garment that houses an accelerometer and microcontroller, continuously recording infants' body orientation (prone, inclined, upright, reclined, supine). We will assess 10-15 infants in their natural environments to determine the validity and reliability of data from the garment. This device will allow for the collection of normative data on infants with or without risk for future developmental delays to determine differences in positioning experiences. This device could also be used to provide feedback to parents to encourage them to place infants more often in upright or prone rather than in the supine position.

Aim 3: To design and test a user driven video game controller adapter that may facilitate inclusive play. Video games have become extremely popular in

recent history. They allow for social interaction and participation with peers and can impact quality of life for children with physical disabilities [9]. The traditional video game controller has not drastically changed throughout the last 20 years and can be very challenging to use for many children with physical disabilities [10,11,12,13]. We will interview children with physical disabilities to identify if and how they currently interface with video game controllers and to determine their key barriers to playing video games with their peers. We will then create an adapted video game controller based on the abilities and needs of two children with physical disabilities. This will allow for fun, inclusive play in the same games their peers are playing.

1.3 Overview of this Dissertation Document

The components of this thesis document are as follows:

Chapter 2: Wearables for Pediatric Rehabilitation: How to Optimally Design and Utilize Products to Meet the Needs of Users. This chapter has been accepted for publication in a special issue of the Physical Therapy journal [Lobo, M.A., Hall, M.L., Greenspan, B., Rohloff, P., Prosser, L.A., & Smith, B.A. (in press). Wearables for pediatric rehabilitation: How to optimally design and utilize products to meet the needs of users. Physical Therapy, Special Issue “Infant and Child Development: Innovations and Foundations for Rehabilitation}. Additional contributors to this chapter were Beth A. Smith, Martha L. Hall, Peter Rohloff, Laura A. Prosser, and Michele Lobo. This chapter presents a novel approach for designing wearables and discusses ways in which existing and emerging wearables may be used in rehabilitation and habilitation for the purposes of

assessment and intervention. This chapter provides a broad overview of wearables that frames the following chapters and work performed in this thesis. Ben Greenspan took part in the decision making for the structure of the paper, and supportive wearables and smart wearables sections.

Chapter 3: Development and Testing of a Stitched Stretch Sensor with the Potential to Measure Movement. This chapter was published in The Journal of The Textile Institute [Greenspan, B., Hall, M.L., Cao, H., and Lobo, M.A. (2018). Development and testing of a stitched stretch sensor with the potential to measure human movement. The Journal of the Textile Institute, 109 (11), 1493-1500]. Additional contributors to this chapter were Martha Hall, Huantian Cao, and Michele A. Lobo. This chapter describes the development and bench testing of a textile-based sensor using conductive thread (Aim 1).

Chapter 4: Design and Validation of a Garment with Conductive Paint Stretch Sensors to Measure Physical Activity [Greenspan, B., & Lobo, M.A. (in review). Journal for the Measurement of Physical Behaviour]. An additional contributor to this chapter was Michele A. Lobo. This chapter describes the development and testing on human subjects of a textile-based sensor using conductive paint (Aim 1). Conductive paint sensors were selected for this chapter rather than the conductive thread sensors from Chapter 3 because it was discovered thread sensors had poor long term reliability due to a rapid increase in resistance after multiple uses.

Chapter 5: Design and Validation of a Smart Garment to Record Infant Positioning at Home. This chapter has been prepared for submission to the

Journal of Early Human Development. Additional contributors to this chapter were Andrea Baraldi Cunha and Michele A. Lobo. This chapter describes the design and validation of a smart garment to measure the time infants spend in different positions (e.g. supine, prone, upright, inclined, reclined) during the first year of life (Aim 2).

Chapter 6: Video Game Play in Children with Movement Challenges: Identifying & Addressing Barriers Through the Development and Testing of Adaptive Video Game Controllers. An additional contributor to this chapter was Michele A. Lobo. This chapter describes the development and testing of adaptive video game controllers for children with physical disabilities (Aim 3).

Chapter 7: Conclusion

Chapter 2

WEARABLES FOR PEDIATRIC REHABILITATION: HOW TO OPTIMALLY DESIGN AND UTILIZE PRODUCTS TO MEET THE NEEDS OF USERS

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2.2 Abstract

This paper will define "wearables" as objects that interface and move with users, spanning clothing through smart devices. A novel design approach merging information from across disciplines and considering users' broad needs will be presented as the optimal approach for designing wearables that maximize usage.

Then, three categories of wearables applicable to rehabilitation and habilitation will be explored: 1) inclusive clothing (e.g. altered fit, fasteners), 2) supportive wearables (e.g. orthotics, exoskeletons), and 3) smart wearables (e.g. with sensors for tracking activity or controlling external devices). For each category, we will provide examples of existing and emerging wearables and potential applications for assessment and intervention, with a focus on pediatric populations. We will discuss how these wearables may change task requirements and assist users for immediate effects and how they may be used with intervention activities to change users' abilities across time. It is important for rehabilitation clinicians and researchers to be engaged with the design and use of wearables so they can advocate and create better wearables for their clients and so they can determine how to most effectively utilize wearables to enhance their assessment, intervention, and research practices.

2.3 Introduction

A variety of wearables are available for rehabilitation and habilitation (hereto referred to collectively using the term rehabilitation). Some of these, such as clothing and orthotics, have existed in a variety of forms for centuries, while others, such as exoskeletons and smart wearables with sensors, have evolved more recently. We define “wearables” as objects that interface and move with users, spanning clothing through smart devices.

This paper proposes that wearables, just like treadmills and other equipment, can be effective rehabilitation tools if they are used in ways to impact users’ abilities, task demands, or the environment to promote function and

participation. With a recent increase in variety of wearables available, it is critical that rehabilitation professionals: 1) are aware of the existing wearables; 2) understand how they can positively impact the design process for wearables; and 3) critically evaluate how wearables can be used to improve the lives of users. This perspective paper aims to provide readers with the information and framework to meet these three goals.

Traditionally, rehabilitation tools have been designed with a primary focus on function. Because wearables are donned and transported with users, they must meet a broad range of user needs in addition to function. A novel design approach merging interdisciplinary perspectives with users' needs is presented here as the optimal approach for designing wearables for rehabilitation.[1] Then, three categories of rehabilitation wearables are explored: 1) inclusive clothing (e.g. altered fit, fasteners), 2) supportive wearables (e.g. orthotics, exoskeletons), and 3) "smart" wearables (e.g. with sensors). For each category, we provide examples of existing and emerging wearables and describe potential applications for assessment, intervention, and research. We discuss how these wearables may change function and participation at the level of the individual, task, or environment.

2.4 FEA2 Model

The advantages of wearables for rehabilitation are that they can assist users in a variety of environments during daily activities, and they can offer the high doses of practice often required for meaningful rehabilitation outcomes.[2,3] The traditional approach to the design of products for rehabilitation has focused

on addressing impairments or limitations in users' functional abilities.[4] However, other key needs must be addressed for a wearable device to be accepted by a user.[2,3] Devices created using the traditional, function-focused model can be uncomfortable to wear physically, psychologically, and socially, as they often focus attention on users' impairments.[5] In addition, devices may provide the desired assistance for function but may be cost prohibitive.[6] It is critical that wearable rehabilitation designs be overtly guided by a model emphasizing the broad needs of users. In the case of pediatric and other patient populations, the role and reliance on a caregiver should be considered when assessing user needs. The caregiver may both play the primary role as informant for the assessment of user needs as well as ultimately mediate device usage.[7-9] Therefore, it is important to discern whose needs are being addressed during the design process. As advocates for our clients, and as liaisons between our clients and device manufacturers, rehabilitation professionals have significant potential to transform the way in which rehabilitation wearables are designed and distributed.

We propose the FEA2 (Function, Expressiveness, Aesthetics, Accessibility) Model to guide rehabilitation wearable design. We developed this model in our Supersuits Program at the Move To Learn Innovation Lab, University of Delaware, to fill the need for a model to guide the design of rehabilitation products. The model is an extension of the existing FEA apparel design user needs model integrated with the design approach of engineering and the philosophy of patient-centered health care.[10,11] The FEA2 model incorporates

user needs related to function, expressiveness, aesthetics, and accessibility (Figure 2.1).[1,10]

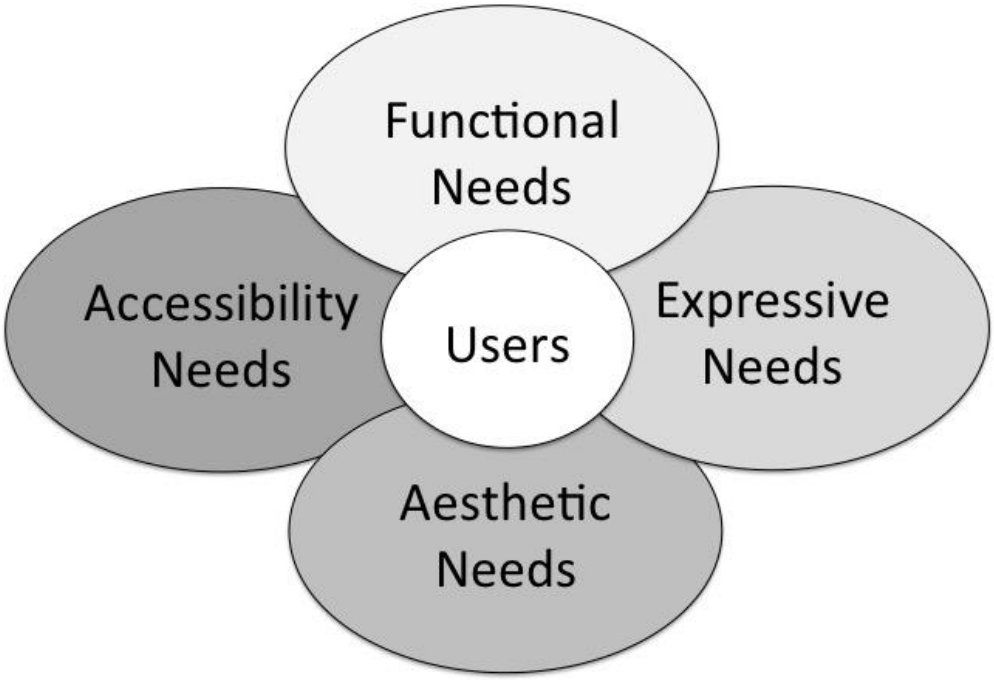

Figure 2.1: The FEA2 model was created to guide the design of rehabilitation products so they address the broad needs of users in relation to function, expressiveness, aesthetics, and accessibility.

Functional needs for a product pertain to its fundamental use. Does the device facilitate and/or improve movement, function, or participation? Does it fit the user appropriately? Is it safe to wear or use? When designing a device, one must consider how the device will be used, how that use determines device demands and constraints, and ultimately the metrics the device must meet in order to be successful.[11-13]

Expressive needs relate to the socially communicative or symbolic aspects of wearables.[10] Wearables communicate varied social messages about the wearer, such as social group membership, personal identity, and sense of self.[14] The expressive quality of a device may vary across users yet is important for conveying personal expression and self-concept.[15]

Aesthetic needs for products relate to the users' interest in the product as an object of beauty.[10] For example, does the object have a pleasing shape, color, or pattern? Different users will have different aesthetic criteria and this may change over time or among cultures.[14] Aesthetics are a vehicle by which a wearer can mediate the social environment.[14,16] For example, the appearance of a wearable can be aesthetically engaging to others, creating opportunities for social interaction and acceptance.

Accessibility needs for products deal with availability, affordability, and inclusivity.[1] It is common for relatively limited choices of rehabilitation devices to be available in the traditional retail environment[17] or through medical manufacturing companies.[1,18] Moreover, these products are often cost prohibitive in a manner that limits users to those with sufficient financial means.[19,20] A recent shift towards more inclusive design practices that focus on serving the maximum number of users, regardless of age or ability level,[21] has the potential to allow users with disabilities to access wearables via the general retail market rather than through niche medical venues. Retail marketing of inclusive designs has the potential to significantly impact availability and affordability of wearables for people with disabilities. The do-it-yourself (DIY) and maker-movements are also

impacting the way devices are designed and obtained as well as empowering end users and members of the broader community.[22]

The FEA2 multidisciplinary framework presents a model for developing innovative and impactful rehabilitation devices for use in the real world of clients. For example, the FEA2 model was recently used to design the Playskin Lift™, the first exoskeletal garment for pediatric users with upper extremity impairment.[23] Our team involved children who had discontinued use of a hard exoskeleton throughout the design process.[23] The final solution improved functional ability for children while also being aesthetically pleasing, personally expressive, easy to use, and accessible via a DIY manual.[1,23]

Rehabilitation professionals should be aware of the FEA2 Model for use in their own designs and so they can guide manufacturers to create products that address clients' broad needs. Improved design may increase adherence with prescribed device use[24] leading to potentially better client outcomes and overall satisfaction.[25,26]

2.5 Inclusive Clothing

Clothing itself is a form of a wearable rehabilitation tool. In the early 20th century, the U.S. medical community began to identify clothing as a rehabilitation tool, using tests of donning and doffing of clothing to assess physical and cognitive improvement of clients.[27] It can be challenging to find clothing that meets desired criteria for style, fit, price point, and self-expression, especially for users with disabilities.[27] The mainstream apparel industry historically neglects this market, due to misconceptions that the market is small (and therefore not

profitable), is not interested in style or aesthetics,[21,28] and requires a drastically different production process.[27] Clothing designed for people with physical disabilities has traditionally fallen into the apparel industry category of "functional clothing."[10] User satisfaction with function-focused designs have been mixed.[29,30]

The newly emerging inclusive clothing design perspective may better serve the needs of people with disabilities than the traditional functional or adaptive clothing perspectives. Inclusive clothing is designed with a broad range of users in mind, with no population excluded, making it more universally appealing and marketable.[21] Inclusively designed clothing has the potential to not only allow for assessment of clients' changing abilities, but also to impact rehabilitation outcomes at the level of the individual, task, or environment (Figure 2.2).

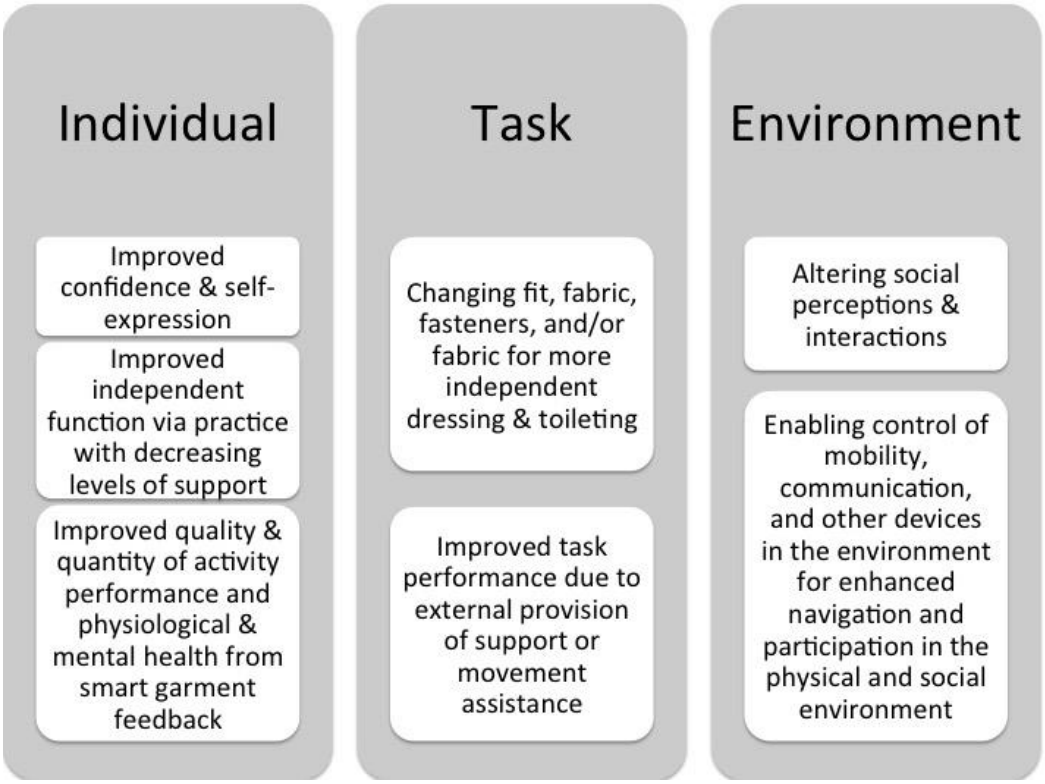

Figure 2.2: Examples of ways wearables can be used to improve the lives of users by impacting the individual, the task, or the environment.

Inclusive clothing can be designed such that the garment facilitates donning and doffing. Individuals with disabilities, medical conditions, or age-related impairments comprise a heterogeneous group, yet have common dressing challenges that can be addressed through inclusive design.[17,31,32] Most research in this area is either dated[33-35] or strictly theoretical,[17,36] but proposed solutions include altering the fit to accommodate comfortable positioning in wheelchairs,[17] altering the location and type of fasteners,[13,14] and constructing clothing using comfortable, non-restrictive material.[17]

Inclusive clothing is becoming more readily available in the retail marketplace. Two examples are the Adaptive Clothing Collection by Tommy Hilfiger in collaboration with the Runway of Dreams organization[37] and the Cat & Jack line available at Target.[38] The Tommy Hilfiger collection features adjustable sleeves and pant lengths for improved fit and magnets that replace challenging buttons and zippers.[37] The Target collection is specifically designed for children with sensory sensitivity.[38] The line features tagless garments, soft fabric, and side openings for donning and doffing.[38] These collections represent two industry leaders bringing positive change to the historically overlooked special needs market, by making available inclusively designed clothing that is functionally appropriate for the user, while also being aesthetic, expressive, and accessible.

Finally, inclusive clothing can be designed to facilitate social participation, thereby changing the social environment of the user. Like any wearable, clothing has a psychosocial component.[14,39] Choice of clothing presents the wearer's sense of self in the social environment, thus impacting interaction with others. Clothing can communicate one's culture, socioeconomic status, group identification, gender identity, and other key perceiver variables.[14] Individuals with disabilities may feel socially stigmatized due to distorted body image and/or negative self-image.[14] For this reason, clothing can be used to mediate potential social stigmatization by concealing physical differences, reducing perceived impairments, or enhancing self-expression[15] thereby increasing self-confidence and security.[14] Therefore, rehabilitation professionals can consider clothing as a tool to assess change across time, to increase

independence with ADLs, or to positively impact user confidence, self-expression, and social interaction.

2.6 Supportive Wearables

Rehabilitation professionals often use supportive wearables that aim to provide movement assistance to users. Examples include passive and active orthoses or exoskeletons. Passive orthoses or exoskeletons provide a fixed amount of movement support at any given moment. Active orthoses or exoskeletons can be controlled to provide variable levels of movement support.[40] While the distinction between orthoses and exoskeletons is blurred, we use the term "orthosis" here to refer to wearable devices with support as their main function and the term "exoskeleton" for devices that augment movement.

Passive orthoses generally provide support for a specific region of the body. One common orthosis for the lower extremities is the ankle foot orthotic (AFO). The goal of an AFO is to stabilize the ankle in dorsiflexion. This may allow users to achieve a normal gait velocity or prevent falls, facilitating greater participation and safety in daily activity. While AFOs usually improve function in clients with poor volitional dorsiflexion,[41] they are made of rigid thermoplastic and may not address user needs related to comfort, expressiveness, and aesthetics. A recent systematic review showed that AFO adherence is often quite low, in some cases as low as 20%, due to complaints related to pain, discomfort, and appearance of the wearable.[42,43] To improve patient adherence, common orthoses should be re-designed using the FEA2 model that accounts for users' needs. Interestingly, some companies have already begun to alter their designs to

incorporate softer, comfortable, more aesthetically pleasing materials. For example, knee braces have traditionally utilized metal components on the medial and lateral aspects for stability. In recent years, companies like DonJoy, known for high stability knee braces, have been offering softer fabric- and silicone-based brace options (https://www.donjoyperformance.com/knee).

In contrast to traditional plastic orthoses, dynamic elastomeric fabric orthoses, such as Theratogs and Dynamic Movement Orthoses,[44] create compression forces on a specific part of the body for added stability. These orthoses can be fit to most body segments, but are primarily used for postural support and stability around the torso and hip. Because these garments are fabric based, they can be more comfortable than traditional rigid orthotics, and may result in higher doses of use. Preliminary evidence demonstrates some potential to improve comfort, posture, or gait.[44-47]

Passive exoskeletons augment how a user is able to move. The Wilmington Robotic Exoskeleton (WREX) is an example that uses 3D-printed components, metal components, and elastic bands to assist with elbow and shoulder flexion (Figure 2.3A).[48] The device improves arm function for children with muscle weakness due to muscular dystrophy, arthrogryposis multiplex congenita, or spinal muscular atrophy.[48] However, it has limitations associated with bulkiness, hardness, aesthetics, and affordability that can constrain families' access to the device and limit use once obtained. The Playskin Lift™ is a passive upper extremity exoskeletal garment that incorporates supportive bundles of

flexible metal wires running through vinyl tunnels underneath each arm (Figure 2.3B).[23]

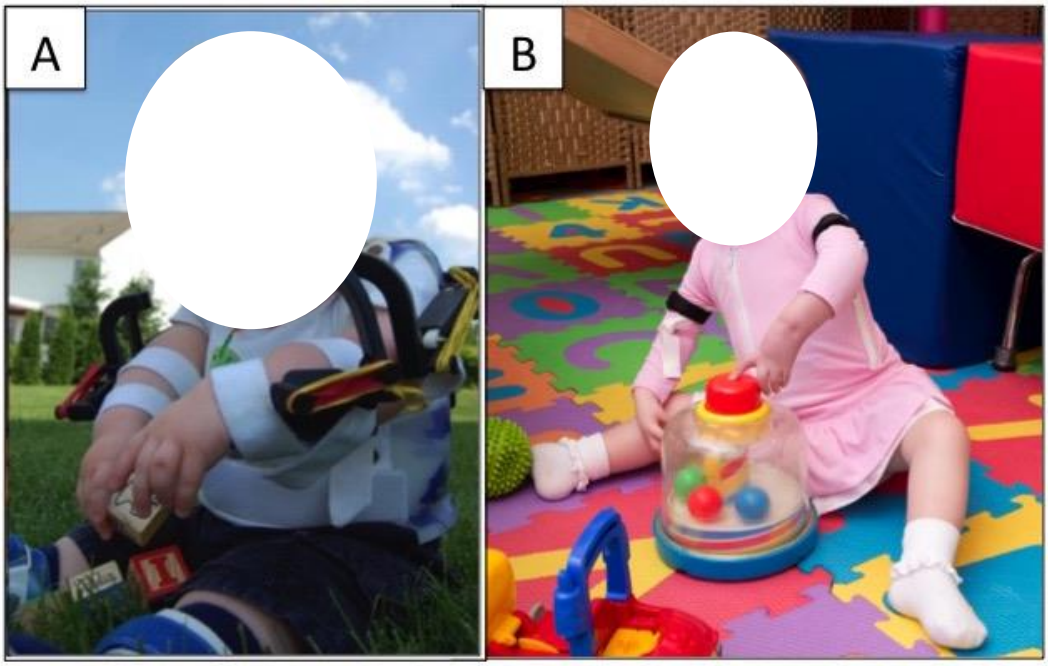

Figure 2.3A & 2.3B: Examples of two passive exoskeletons currently available to assist upper extremity function for children with arm movement impairments. The Wilmington Robotic Exoskeleton (WREX; A) uses plastic or metal components for support and elastic bands to lift the arms. It is available through Nemours A.I. DuPont Hospital for Children and JAECO Orthopedic. The Playskin Lift™ (B) is a garment with vinyl tunnels under each arm to house bundles of elastic wires to lift the arms. It is available via an open-access DIY manual (http://sites.udel.edu/move2learn/how-todiy/).

Pilot data suggest it can improve arm function in children with arm weakness within a session.[23] Because the garment looks and feels like a typical shirt and is easy to don, doff, and clean, users report high rates of adherence. Low

profile soft exoskeletons like this have a variety of benefits including improved comfort, aesthetics, and affordability. However, current designs have key limitations. For instance, the Playskin Lift™ cannot support the arms of children over 3 years of age. Passive exoskeletons are also limited in that they do not incorporate mechanisms allowing users to regulate the amount of support they receive.

Active/powered orthoses and exoskeletons incorporate mechanisms to allow users to customize and regulate the level of support. Robotic exoskeletons have been created to enhance human performance, as assistive devices for individuals with disabilities, and as therapeutic devices for rehabilitation.[49] The Defense Advanced Research Projects Agency (DARPA) has funded a variety of exoskeleton projects since 2000[50] to enhance human performance. Many of the exoskeletons produced aim to allow users to carry a heavier load with greater ease. Unfortunately, problems such as bulkiness, power demands, and human energy exertion from altered kinematics have prevented these exoskeletons from becoming mainstream.[49] The military is now shifting from hard to soft exoskeletons like the exosuit from Harvard University, which consists of two motors that pull cables to assist in hip extension and ankle plantar flexion.[51] Using this device, participants' net metabolic use decreased by 7.3%. Reductions in metabolic cost of this magnitude could have a significant impact on endurance, fatigue, and agility when walking over large distances.

Powered exoskeletons, like the Ekso and ReWalk, are also currently being tested for individuals with disabilities. ReWalk was the first exoskeleton to

receive U.S Food and Drug Administration approval for people with spinal cord injury in 2014 (http://rewalk.com/rewalk-robotics-announces-expansion-to-india-with-saimed-innovations-2-2-2-2-3-2-3/). This exoskeleton allows people with paraplegia to walk while standing upright because it supports their body weight and augments their leg motion through motors at the hip and knee. While they are becoming increasingly lower in profile, powered exoskeletons remain expensive and as a result are typically clinic-based. The LiteRun Company has begun to shift the design of rehabilitation exoskeletons from hard to soft, mirroring the shift observed in military exoskeleton design, creating a lower extremity pneumatic exoskeleton using soft air bladders (http://www.literun.com/). Soft exoskeletons like this have the potential to more broadly address needs related to aesthetics, comfort, and ease of use. They also can be more affordable, increasing the likelihood for use in natural environments.

Supportive wearables have the potential to serve as assistive devices, improving function by providing support and/or movement assistance to perform various tasks. If used in conjunction with intervention activities, supportive wearables can also be used as rehabilitative devices. By grading the level of assistance provided during intervention across time, exoskeletons have the potential to facilitate permanent improvements in independent performance and function beyond the wear time.[52]

2.7 Smart Wearables

Smart wearables, or wearables that incorporate sensors, afford the opportunity to collect continuous data across time, tasks, and environments.[53]

This contrasts with standard clinical tools that provide a snapshot of behavior at a specific time in a particular environment, often failing to capture true performance or variability in performance. Smart wearables are becoming more common and a variety are available both for the research market and general population. These include wearables that track body temperature, heart rate, step number, foot pressure, position, and acceleration.[54] The data measured depend on the sensor type(s) integrated in the wearable. Accelerometers have been commonly used in smart wearables in recent years. They provide information about the acceleration of the person or body part to help us understand physical activity and can be especially useful for rehabilitation professionals.

Both commercial and research wearables incorporating accelerometers exist. Commercial devices include the Fitbit and other fitness trackers sold for personal use. Research devices include accelerometer-only devices, such as Actical and Actigraph, designed to measure the intensity of physical activity.[55] Other research devices, for example, the Stepwatch, have embedded software to calculate the number of steps taken.[56] Finally, some research wearables measure acceleration as well as other signals, such as angular velocity, electromyography, cardiac activity, respiration, and/or galvanic skin response (for example, MC10 and Shimmer).

A key consideration for rehabilitation professionals using smart garments/sensors is the validity of those wearables for measuring the variable of interest. One challenge relates to how the products process the sensor data. Activity monitors like Actigraph and Actical use proprietary software to

determine activity counts and classify physical activity intensity. In effect, the type of movement required to produce an activity count is unknown to users. Different types of data output may have varying levels of validity. For instance, Actigraph data in 31 ambulatory toddlers 12-30 months old showed a positive correlation between number of Actigraph activity counts and observed activity level measured by the Observational System for Recording Physical Activity in Children - Preschool, but poor to fair relations between classification of activity levels based on activity counts.[57] It is important to understand the process and accuracy of software classification systems associated with wearables, especially if they are going to be used to prescribe or evaluate effectiveness of interventions.

Another challenge to the validity of wearables for measuring the variable of interest is that a device designed and calibrated to measure physical activity in a healthy adult may not be accurate for an adult with a disability or for pediatric populations. Accelerometer thresholds (cutpoints) to classify intensity of physical activity validated by visual observation have been established for 5-6-year-olds but do not yet exist for other ages to the best of our knowledge.[58] Actigraph activity counts have been measured for non-ambulatory infants, but no validation of activity counts to a gold standard measure has been reported.[59,60, 61] Step-counting devices such as FitBit and Stepwatch have been designed and calibrated to measure steps in ambulatory persons. Whether they are able to capture kicking or stepping movements in non-ambulatory infants is unknown. Stepwatch has been validated for use in ambulatory youth with cerebral palsy, Gross Motor Function Classification System Levels I to III between 10 and 13 years of age.[62]

In regard to non-ambulatory populations, a recent study showed three commercially available wrist-worn activity trackers generally performed poorly for accuracy/precision in counting strokes per minute when users were propelling a manual wheelchair.[63] The Apple Watch is one of the only step-counting devices that has a separate wheelchair mode to measure activity while rolling.[64] Other studies have also shown that participants didn't find current activity tracker data relevant to their specific activities.[65,66]

While using data from accelerometers to classify stationary versus movement periods or to measure number of steps can be fairly straightforward, identifying or classifying other types of movements is not as clear-cut.[67] For many sensor-derived metrics, variability is high, both within and across participants (e.g., arm movement characteristics in infants[68] and adults[69]). Alternatively, sensor data may look similar across a variety of activities, making it difficult to classify activity performance. Therefore, interpretation and classification of data can be quite complicated (Figure 2.4).

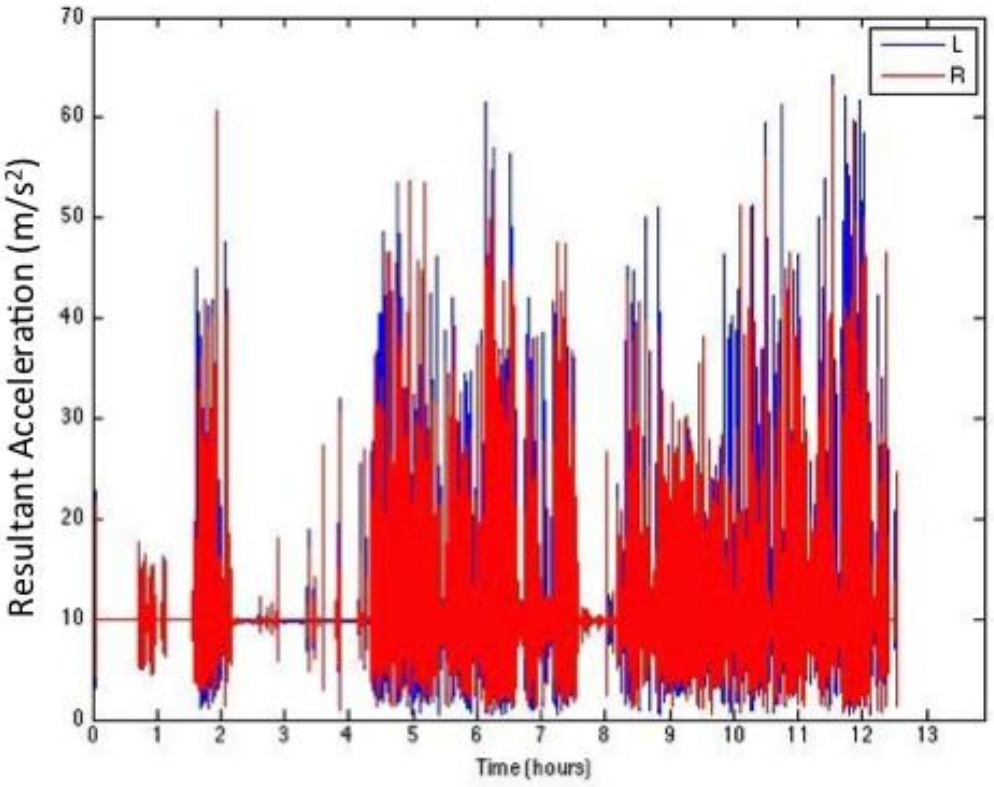

Figure 2.4: One full day (~11 hours) of acceleration data from a 7-month-old with wearable sensors on both ankles. Each sensor contains tri-axial accelerometers collecting actively synchronized data at 20 samples per second. Left and right leg data are overlaid. The sensors were put on around the 1.5-hour mark and stayed on until around the 12.5-hour mark. The infant made a variety of movements across different activities during the data capture window, but this is not readily apparent by viewing the acceleration time series alone; advanced analyses are needed.

As these examples demonstrate, research to determine the validity of the different types of data output by wearables is critical. While algorithms for processing data from wearables need to be created and validated for specific

disabilities, we will not be able to validate all types of movements produced by each unique type or category of disability. As smart wearables become more integrated within practice, rehabilitation professionals should drive the design and testing not only of the wearables but also of the associated data processing approach and software-user interface to ensure that products meet the key needs of clients and rehabilitation professionals as end users.

Once the validity of smart wearables has been established for the intended use, these wearables can provide useful information about activity in the natural environment. For instance, in the Infant Neuromotor Control Laboratory at the University of Southern California, we use wearable sensors to capture the variability and repertoire of limb movements that infants produce across full days. We used wearable sensors containing tri-axial accelerometers and gyroscopes (Opals, APDM, Inc.) to create and validate algorithms to count the thousands of leg movements infants make in a day,[70] as well as the duration, peak acceleration, average acceleration, and type (unilateral or bilateral) of each movement produced.[71] We have taken a similar approach to create and validate an algorithm to identify bouts of arm movements.[68] We have measured infants with typical development (1-12 months of age) and infants at risk for developmental delay (1-15 months post-term age). These are modest samples at this point (12-20 per group) but we are demonstrating the ability to use data from wearables to discriminate between typical and at-risk groups as well as to predict neurodevelopmental outcomes. Preliminary outcomes demonstrate that leg movements in infants at risk show smaller accelerations,[72] more repeatability (less

variability), and computationally less information.[73,74] Further, infants at risk for delays and poor neurodevelopmental outcomes at 24 months of age demonstrated smaller acceleration values in their leg movements than did infants at risk for delays and good neurodevelopmental outcomes at 24 months of age.[72] These types of data from smart wearables may facilitate early identification of atypical neuromotor development and assessment of change across time in response to intervention.

Collecting repeated measures of activity from large, culturally diverse samples of infants and young children over extended periods in daily life will improve our understanding of typical and atypical health and development and how to optimize the delivery of early intervention services. Collecting full day and week datasets from wearable sensors will also increase our understanding of the influence of environmental and cultural contexts on infant development in support of global health and development efforts. In low- and middle-income countries, 43% of all young children are at risk of not achieving their developmental potential due to experiencing malnutrition, poverty, and lack of early stimulation.[75] Improving the ability of health workers in low-resources settings to efficiently identify which children are most at risk and could benefit from intervention is an important global health priority. However, existing psychometric tools are time-consuming, require significant training and support for testers, and may not always measure culturally appropriate developmental endpoints.[76] Smart wearables may help to address these practical needs.

To this end, the Infant Neuromotor Control Laboratory and Maya Health Alliance collaborated to collect wearable sensor data from infants in rural Guatemala at risk for stunting of growth and developmental delay. For example, pilot data on a full day of leg movement from a one-month-old indigenous infant in a remote part of Guatemala were compared to a one-month-old infant in the U.S. with typical development. The infant in Guatemala had a movement rate of 400 movements per hour awake, compared to 800 movements per hour awake for the U.S. infant. This preliminary observation suggests that infants experiencing stunting of growth due to undernutrition may exhibit a greater proportion of sleep time and lower quantity of movement when awake. At the same time, however, infants in rural Guatemala are traditionally worn swaddled against the caregiver's body and this would be expected to impact physical activity of the limbs. Smart wearables can serve as useful tools to gather data in a variety of populations like the one in Guatemala to better understand how movement experience may impact future development and to identify movement characteristics that are early indicators of stunted development.

Novel wearable sensors have the potential to make smart wearables more comfortable and common in the commercial market. Stretch sensors made from conductive threads using a coverstitch,[77] or using a zig-zag stitch[78] are being developed and tested to measure movement as alternatives to hard, bulky sensors. These sensors can be stitched directly into clothing across joints to measure movement. Novel silicone capacitive sensors that can be integrated with fabric are also being developed and tested.[79] Soft sensors like these can address user needs

related to comfort, aesthetics, and expressiveness to facilitate motion data capture in natural settings over longer periods of time.

In addition to measuring activity and identifying differences or delays, smart garments have the potential to assist in changing an individual's function. For example, sensors can provide feedback to users, informing them how to change aspects related to the quality of their movement, such as pattern, speed, sway, or pressure distribution.[80-82] Smart garment feedback can potentially shape the quality of users' movement practice so that users learn to perform behaviors in more functional, energy efficient, and safe manners. Smart garments may also provide feedback about amount of activity performance that can be used to motivate users to improve their endurance, and physiological and mental health.[83,84] Finally, when utilized in conjunction with other devices, including active exoskeletons, mobility devices, and communication devices, sensors integrated in smart garments can empower users to more independently and fully navigate and participate in their physical and social environment (Figure 2.5).[85-87]

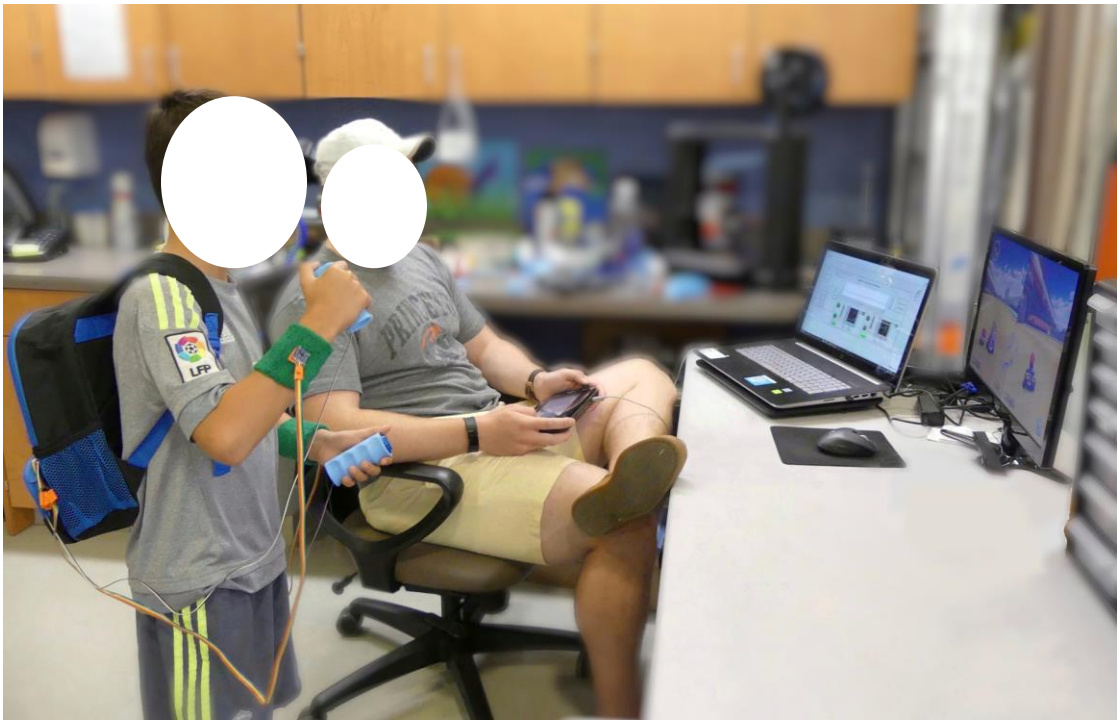

Figure 2.5: An example of how smart wearables can allow children to control their environment in meaningful ways. This prototype allows a child to use gross arm movement rather than more challenging fine manual movements to engage in video game play.

2.8 Conclusions

A variety of novel wearables have become available to rehabilitation professionals and clients in recent years. Wearables spanning clothing through smart garments can be used in a variety of ways for assessment and intervention to impact users' abilities, task demands, or the environment to promote functional performance and participation (Figure 2.6).

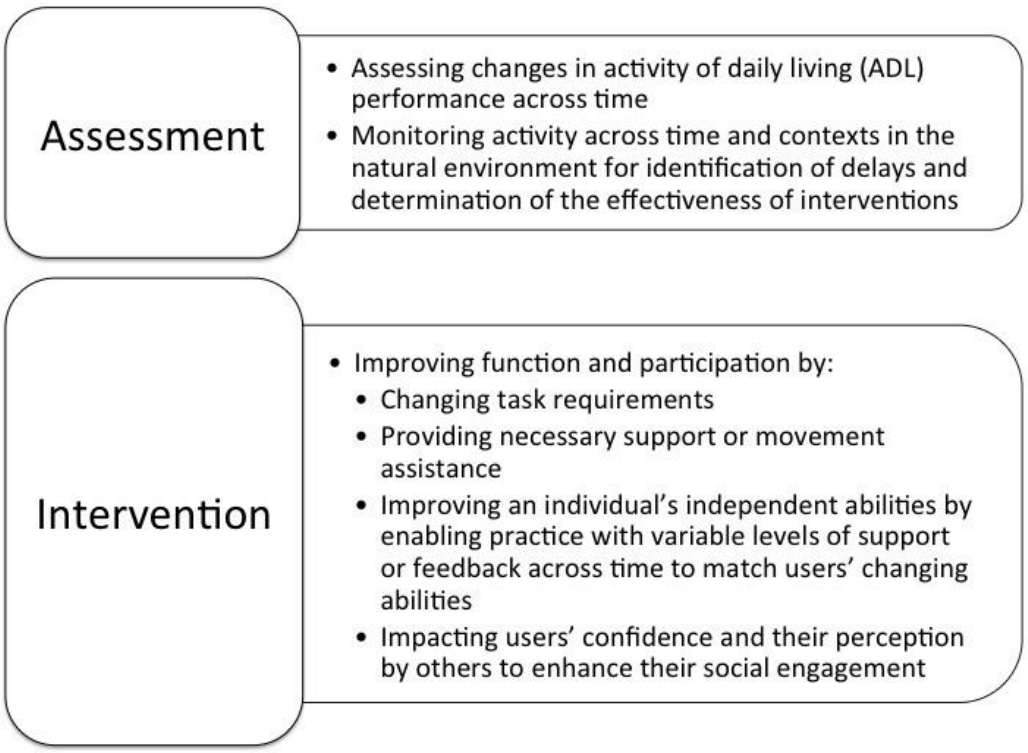

Figure 2.6: Potential uses of wearables for rehabilitation assessment and intervention.

They have the potential to be used as assistive devices to aid functional performance for users when they are worn and as rehabilitation devices incorporated in activity programs aimed at improving independent function across time when they are not donned.[52] There is a significant need for additional, high-quality research on all wearables to determine their effectiveness in relation to these potential uses.

Smart wearables provide the additional ability to continuously monitor a range of variables and therefore can be helpful in tracking performance across time in natural settings and identifying potential delays or impairments in performance and participation. Smart wearables will likely be most effective

when clinicians and clients use them to measure or provide feedback on targeted outcomes based on scientifically supported models for change. In other words, although smart garments may allow us to monitor and provide feedback on any number of variables, we should focus on those variables that the literature suggests are likely to allow us to impact our clients' goals.[88,89] More research is needed not only to determine the validity of the available smart garments but also to determine how to optimally set goals and provide feedback to users based on data from these devices.

To maximize usage, wearables and other rehabilitation tools should not be designed with function as the sole focus but should consider the broad needs of users related to function, expressiveness, aesthetics, and accessibility, noting the role caregivers play in mediating the assessment of user needs and device usage. Physical therapists are trained to focus on the needs of users for intervention design and they liaise between users and device manufacturers. Thus, they are uniquely positioned to advocate for and to create better wearables for those they serve.

2.9 Acknowledgments

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Chapter 3

DEVELOPMENT AND TESTING OF A STITCHED STRETCH SENSOR WITH THE POTENTIAL TO MEASURE MOVEMENT

3.1 Authors

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3.2 Abstract

Human movement in health sciences is typically measured using motion capture laboratory equipment, but this method has inherent limitations including cost, specialized knowledge requirements, and inability to measure activity in natural settings. This study presents a novel textile based stretch sensor with potential for use for measurement of human movement in smart clothing. The sensor is composed of silver plated thread stitched into a textile substrate. Multiple samples varying in thread type, length, stitch geometry, and textile substrate were systematically fabricated and tested. Elongation and recovery testing revealed the optimal sensor configuration for providing the largest and

most repeatable change in resistance. Proper placement of these sensors in clothing may allow for a variety of applications, including kinematic measurement, activity monitoring, movement impairment detection, activity feedback, or control of environmental and assistive technologies.

3.3 Introduction

Smart and e-textiles are an emergent area of apparel engineering in which electronic technology is integrated with textile material, creating a platform in which to sense and respond to stimuli (Zhou, Huosheng, & Tao, 2006). In the domains of physical therapy, sports medicine, biomechanics, and kinesiology, these stimuli are often in the form of specific changes in position. Kinematic, or movement, analysis is a common practice for clinicians seeking to assess, monitor, or alter the amount and patterns of movement for individuals. Traditionally, kinematic analysis has been performed using motion capture systems housed in research laboratories (Papi, Osei-Kuffour, Chen, & McGregor, 2015). These systems use high-speed cameras to record an individual's movement (Schmitz, Ye, Shapiro, Yang, & Noehren, 2014). These systems are expensive to use and maintain (Kim, Gravunder, & Park, 2015), are inaccessible for many researchers and clinicians, require specialized training, and do not allow for activity assessment in natural settings (Potter, 2015). To address these limitations, researchers and clinicians have begun utilizing wearable devices, primarily those incorporating inertial measurement units (IMUs), to measure human movement (Fong & Chan, 2010). In order to accurately measure movement, IMUs must be aligned with fixed limb segments and attached to the

body via straps or plastic housings secured to clothing. The sensors are rigid, can be difficult to integrate into clothing (Papi et al., 2015; Patel, Park, Bonato, Chan, & Rodgers, 2012; Zhou et al., 2006) and often result in uncomfortable user experience (Chan, Estève, Fourniols, Escriba, & Campo, 2012; Dunne, Ashdown, & Smyth, 2005). Further limitations of these types of sensors and electronic components are related to product care, in that hardware must be removed for laundering and require special skills for proper maintenance (McAdams et al., 2011). Flexible, smart textiles have the potential to address these key limitations of existing motion analysis systems, allowing for activity monitoring via comfortable, more affordable clothing used in natural settings (Yi, 2015).

Commercially available soft sensors currently exist, and are made from conductive polymers that flex and stretch to measure movement ("Flexible stretch sensors", 2016). As the sensor is stretched and the contacts between the conductive material change, the conductive properties of the sensor change (Atalay, Kennon, & Hussain, 2013). Measurement of those changes in conductive properties can provide information about physical activity. However, integrating soft sensors into clothing safely, comfortably, and aesthetically is challenging (Papi et al., 2015). Textile-based stretch sensors consist of two components: conductive material and a textile substrate (Gioberto & Dunne, 2012). The conductive material can be in the form of an electroactive polymer, piezoelectric ceramic, or conductive coating (Gioberto & Dunne, 2012). The textile substrate should be one that allows for comfortable clothing fit, while being elastic to allow the sensor to elongate and then relax (return to the original state). For this study,

the conductive material was a silver-coated thread that was stitched into a knit textile substrate. Conductivity of the thread (Moradi, Bjorninen, Ukkonen, & Rahmat-Samil, 2012), geometry of the stitch structure (Gioberto & Dunne, 2012; Moradi et al., 2012), sensor length, and properties of the textile substrate (Atalay et al. 2013; Gimpel, Mohring, Muller, Neudeck, & Scheibner, 2004) were all expected to contribute to the characterization and responsiveness of the soft sensor.

To date, there is only one known design study of a textile-based stitched stretch sensor (Gioberto & Dunne, 2012). In this study, a silver plated nylon thread was stitched using a coverstitch structure into a polyester based fabric substrate to test a looped conductor approach. In this approach, when the sensor is stretched, more loops come in contact with one another, causing the resistance to decrease. In contrast, when the sensor is relaxed, no loops are in contact and the resistance is very high. This change in resistance can then be mapped to stretch in inches. The aim of the present study was to expand on this previous research by Gioberto and Dunne (2012, 2013) via systematically testing combinations of conductive threads, stitch geometries, sensor lengths, and textile substrates to determine the optimal integration pattern. In this way, the current study offers novel insights into the integration of soft sensors within textiles. This study also explores the reliability of soft sensors, or consistency of performance over multiple elongation and retraction cycles. The results suggested that a predictable change in resistance could be detected from elongating the thread across 5, then 250 cycles.

3.4 Experimental

Materials

Four factors were taken into account during material selection process: conductive thread, stitch geometry, sensor length, and textile substrate.

Conductive Thread

Fourteen conductive threads were used for initial bench testing: twelve silver plated synthetic threads from Shieldex Trading, Inc. (Palmyra, NY) and two stainless steel threads from Adafruit Industries (New York, NY). To identify if the threads could be used for further sensor testing, they needed to be compatible for standard machine stitching, meaning their thread size needed to be between 140 and 700 decitex (dtex; grams per 10,000 meters). The thread size was provided by each company. To measure a recordable change in sensor resistance, the thread needed to have a linear resistance greater than 0.01 Ω/cm. Resistance was measured using a digital multimeter (Adafruit Industries, Model 9205B+), providing the accuracy within 1 ohm (Ω). Four threads met these criteria and were determined to be viable for sensor testing (Table 3.1).

Table 3.1. Conductive thread technical data

Thread Number	Fiber Content	Thread Size	Linear Resistance
1	Silver-plated Nylon 6 & PET	22/1 dtex & 113/32 dtex 2ply	<300 Ω/cm
2	Silver-plated Nylon 6 & PET	44/12 dtex & 113/32 dtex	<300 Ω/cm
3	Silver-plated Nylon 6	110/34 dtex	<30 Ω/cm
4	Silver-plated Nylon	280 dtex 2ply	<0.025 Ω/cm

Stitch Geometry

Four stitch geometries were selected for sensor testing based on the knitted structure of the textile substrate and the elongation characteristic of the geometry. Table 3.2 illustrates each of the four stitch geometries and provides its stitch class, number of threads used to create the stitch (thread count), and the name for each stitch configuration. Stretch stitches included 300, 400, and 500 American Society for Testing and Materials (ASTM) classifications ("Stitch and seam guide," 2004). These four geometries represent all ASTM geometries that will stretch. Previous studies used stitch 406 geometry (Gioberto & Dunne, 2012; 2013).

Table 3.2. Stitch geometry ASTM classifications

Stitch Geometry	Stitch Class	Thread Count	Name
	304	Two	Zig-zag Stitch
	402	Two	Chainstitch
	406	Three	Bottom Coverstitch
	514	Four	Overedge Stitch

Sensor Length

Two sensor lengths (2” and 6”) were selected for testing based on the length of a sensor with potential to cross joints and capture a variety of human movements.

Textile Substrate

Three textile substrates were selected for the testing. First was a polyamide(nylon)/elastane blend (90% nylon/10% spandex blend) which is commonly used in knitted sportswear fabrics (Umar, Hussain, & Maqsood, 2016). This type of textile has been extensively tested to understand its material properties related to factors such as compression, stretch, and recovery (Dunne et al., 2005). Second was a polyester/elastane (90% polyester/ 10% spandex blend), the other most common material used in compression sportswear (“Under

Armour," 2017). Third was a cotton-based substrate (95% cotton/5% spandex blend) used to compare synthetic vs. natural fibers.

Preliminary Testing: Identification of the optimal conductive thread

To determine which of the four conductive threads would show the largest change in resistance when stretched, each thread was stitched using each of the 4 stitch geometries. A sensor length of 6" and the nylon based fabric substrate were kept constant for this round of testing resulting in a total of 16 samples fabricated. Protocol for sensor fabrication included textile substrate orientation in the horizontal (course) direction, conductive thread stitched along the technical face of the course direction of the fabric, stitch tension consistent on both sides of the textile substrate, and consistent stitch length and width. Each of the sixteen samples was manually stretched to 30% elongation (1.8 inches) and its resistance at its relaxed and elongated states was recorded using a digital multimeter (Adafruit Industries, Model 9205B+). The 30% elongation value was selected based on the initial testing of textile displacement measured in a jersey knit compression garment during shoulder adduction/abduction. Thread #1 (silver-plated 22/1 dtex Nylon 6 and 113/32 dtex PET 2-ply) had a change in resistance two orders of magnitude greater than the other three threads, so all further testing reported utilizes the thread #1. One stitch geometry (ASTM classification 514) did not yield a change in resistance for any of the thread types in preliminary conductive thread testing, and thus was eliminated from further testing.

Sensor Integration Testing: Identification of the optimal stitch geometry and sensor length

Testing began with the aim of determining optimal stitch geometry and sensor length. This testing incorporated 1 thread type (silver-plated 22/1 dtex Nylon 6 twisted with 113/32 dtex PET 2-ply), 3 stitch geometries (ASTM classifications 304, 402, and 406), 2 sensor lengths (2" and 6"), and 3 textile substrates (90% nylon/10% spandex blend, 90% polyester/ 10% spandex blend, and 95% cotton/5% spandex blend). Each stitch geometry was incorporated into each fabric substrate for both sensor lengths, resulting in a total of 18 samples being fabricated for sensor integration testing (all combinations are described in Table 3.3).

In order to minimize the influence of environmental factors, all sensor integration testing samples were treated according to ASTM standard D1776 for conditioning and testing of textiles ("ASTM", 2016). Samples were conditioned at 21°C and 65% relative humidity for 24 hours prior to testing in an environmental chamber (Lunaire, Model CEO910-4).

Tensile testing machines are the standard apparatus for testing stretch properties of a textile ("ASTM," 2015). These machines are intended to measure textile recovery and distortion based on the effect from a specified mechanical load. The stitched stretch sensor necessitated a unique testing apparatus that would repeat a specified displacement and recovery over the course of multiple cycles, while also measuring the resistance of the sensor. A custom made testing apparatus (Figure 3.1A) was designed using Autodesk Inventor (Autodesk, San

Rafael, CA) and 3D printed on a Printerbot Plus 3D Printer (Printrbot, Lincoln, CA) to stretch the sensor textile samples by 30% (1.8” for the 6” sample, and 0.6” for the 2” sample) at a rate of roughly 1 inch per second across five cycles for the purpose of this testing. The first elongation was excluded from the analysis to allow the sensor to reach a thermal equilibrium (initial peak after having a voltage running through the thread), and the last retraction was excluded so there would be an equal number of elongations and retractions. This results in four cycles to be analyzed. Each sample was connected to the apparatus by two spring clamps, and readings were taken through two banana clips situated 5.3” apart for the 6” sample, and 1.3” apart for the 2” sample.

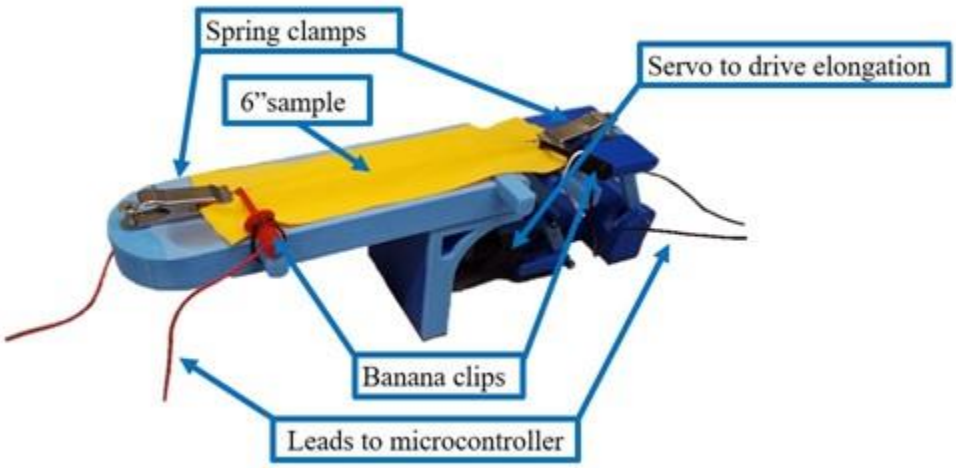

Figure 3.1A. Testing apparatus shown with example 6-inch sensor swatch

Figure 3.1B. Testing apparatus showing 1.8” elongation

While each sensor was elongated, the data was collected through an Arduino Uno microcontroller using a custom written LabVIEW program (LabVIEW 2015, National Instruments, Austin, TX). The LabVIEW program controlled the elongation of the sensor using a servo motor. When the servo turned, it moved a four-bar linkage, elongating the apparatus (Figure 3.1B). This elongation was measured using digital calipers (model Pittsburgh ITEM 47257) accurate to 0.001". The sample was elongated over the course of five trials at the speed of approximately 1 inch/sec. The microcontroller sent a constant 5 volt (V) voltage through the sensor, and when the resistance of the sensor changed, a voltage drop was detected and recorded by the microcontroller. The data from the sensor were sampled at 75 Hz in the form of a voltage between 0 and 5 V. Data analysis and signal processing for these data are described below.

Sensor Reliability Testing: Identification of the optimal fabric substrate

After the optimal stitch geometry and sensor length were identified, three samples (one of each substrate) were fabricated and tested with the same testing protocol as in the sensor integration testing. These samples were elongated and retracted for 250 cycles rather than the previously tested five cycles from sensor integration testing to determine the reliability of sensor results over time. An early, middle, and late, four cycle snapshot were compared for each of the three samples. Four cycles were chosen because it represents the same number of cycles that were analyzed during sensor integration testing. Data analysis and signal processing for these data are described below.

Data analysis and signal processing

For sensor integration and sensor reliability testing, data were initially recorded as voltage drop resulting from the resistance change when the sensor was elongated. To compare sensor output with results reported in the literature (Gioberto & Dunne, 2013), sensor data needed to be converted from voltage to the resistance measure. For this reason, the voltage drop from commercially available resistors were measured and recorded. Commercial resistors have a variability of ± 5% resistance, so resistance was measured using a digital multimeter (Model 9205B+). Sixteen commercial resistors with a range in resistance of 0.534 kilo ohms (KΩ) to 71.63 KΩ were tested, and their corresponding voltages recorded. This range represented the voltage range recorded from the stretch sensor. Once change in voltage was known for each resistor, a scatter plot correlating voltage to resistance was created. Non-linear regression analysis was performed and two third order polynomial fits yielded the best lines of fit, $y = -0.5798x3 + 7.4307x2 - 34.386x + 58.353$ for resistance values from 0.534 KΩ to 14.32 KΩ and $y = -26.709x3 + 132.36x2 - 232.9x + 164.71$ for resistance values from 14.32 KΩ to 71.63 KΩ (Figure 3.2).

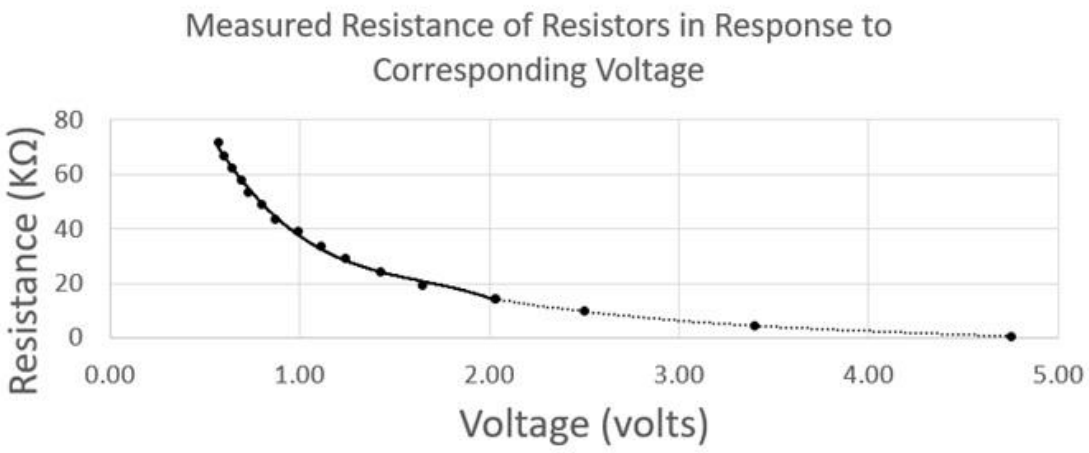

Figure 3.2. Voltage resistance scatter plot and corresponding best line of fit. One polynomial fit is shown as a solid line, and the other as a dotted line.

After the polynomial fit lines were calculated, they were then used to map the measured voltages to the corresponding resistances. A custom filter was used to replace data points when a slope greater than a specified KΩ/sample was detected. The waveform was then run through two, alternating direction, 2nd order low pass Butterworth filters. Both the cutoff slope and low pass frequency were set specifically for each sample to most accurately represent the data. The cutoff slope ranged from 0.133 to 13.673 KΩ/sample and the low pass frequency ranged from 2.35 to 10 Hz. Next, the first elongation and last retraction were cut from the analysis to allow the thread to reach thermal equilibrium as mentioned in the methods section.

After the signal had been filtered, the elongations and retractions were then linearly mapped to the sensor length. A third order polynomial regression was performed for the elongations and for the retractions. Next, the average absolute error from the polynomial fit was found for the elongations and for the

retractions. Because each sample had a different baseline and change in resistance, the average absolute error was normalized by dividing the average absolute error for each sample by its total change in resistance. The average R2 value was taken for the elongation cycles, then the retractions, and their averaged value was recorded for each sample. This signal processing sequence can be seen in Figure 3.3. For the sensor integration testing, this analysis was performed for each of the sensor samples tested across the 4 cycles. For sensor reliability testing (250 elongation-relaxation cycles), this analysis was performed for 4 early cycles (cycles 2-5), 4 middle cycles (cycles 123-126), and 4 late cycles (cycles 246-249) for each of the sensor samples to represent performance throughout different phases of the 250-cycle test. For the sensor reliability testing an intraclass correlation coefficient (ICC) was also found using SPSS (IBM, Armonk, NY). An individual ICC will be found comparing early to middle cycles, and early to late cycles for each fabric substrate.

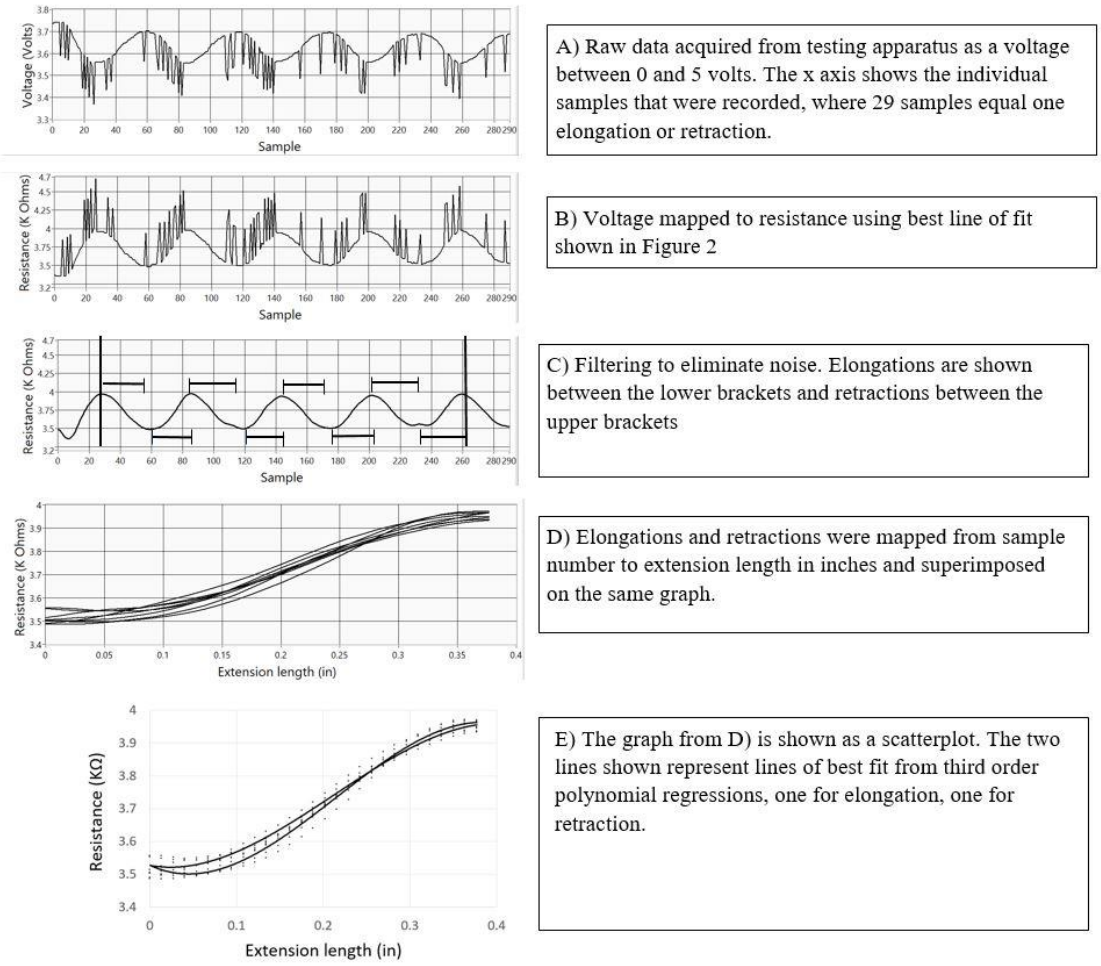

Figure 3.3. An example of signal processing sequence for a sample of the data.

3.5 Results and discussion

Sensor Integration Testing: Identification of the optimal stitch geometry and sensor length

The sensor integration testing results are shown in Table 3.3. Samples with the lowest average absolute percent errors (%) and the highest R^2 values represent samples with optimal performance, and are shaded in grey. The sample

with the overall best performance (6”, stitch class 304, cotton substrate) is marked in bold.

Table 3.3. Sensor Integration Testing Results

6” Samples		Nylon		Polyester		Cotton	
		%	R^2	%	R^2	%	R^2
Stitch Class	304 Zig-Zag	3.05	0.990*	5.30	0.958*	1.701	0.995*
	406 Cover	25.757	0.500*	331.931	0.629	29.337	0.586*
	402 Chain	5.079	0.952*	7.140	0.927*	7.158	0.912*

2” Samples		Nylon		Polyester		Cotton	
		%	R^2	%	R^2	%	R^2
Stitch Class	304 Zig-Zag	4.504	0.969*	3.336	0.988*	3.752	0.980*
	406 Cover	15.325	0.723*	21.850	0.857	29.025	0.603
	402 Chain	9.495	0.895*	24.609	0.653*	17.695	0.858*

* p value < 0.001

The p values for the polynomial regressions were below 0.001 for all samples using stitch geometries 304 and 402. Stitch geometry 304 had the lowest percent error and the highest R^2 value across both sample lengths and all fabric substrates. The mean average absolute percent error across all samples incorporating stitch geometry 304 was 3.61%, which was 8.26% less than the mean for all 402 samples, and 71.93% less than the mean for all 406 samples. The mean R^2 value across all samples incorporating the 304 Zig-Zag stitch was 0.98, which was 0.11 higher than the mean for all 402 samples, and 0.33 higher than the mean for all 406 samples. For these reasons, stitch geometry 304, Zig-Zag, was selected as the optimal stitch geometry.

While excluding the stitch geometry 406 due to its drastically inferior performance, the average percent error and R^2 values were calculated for all 6” and 2” length samples. The mean average absolute percent error for the 6” samples was 4.905%, which was 5.661% less than that for the 2” samples; the

mean R^2 value for the 6” samples was 0.96, which was 0.07 higher than the mean R^2 for the 2” samples. For this reason, the 6” samples were selected as the optimal sensor length.

Sensor Reliability Testing: Identification of the optimal fabric substrate

After testing across 250 cycles, it became clear that all three samples displayed an increased resistance over time. A custom correction filter was applied, so that the valleys of the waveform would remain level throughout the testing. Three sections of data, each four cycles long, were selected from the data for analysis: early cycles 2-5, middle cycles 123-126, and late cycles 246-249 to represent performance throughout the 250-cycle test. Data from each set of cycles were analyzed using the same protocol as described for the sensor integration testing, except the third order polynomial regressions calculated for the early cycles were also applied to the middle and late cycles. Keeping the regression equation the same allowed us to highlight if/how the sensor changed over time (Table 3.4) The R^2 could only be reported for the early cycles because it was a product of the best line the polynomial regression. Because this equation was inputted rather than derived for the middle and late stages, the R^2 was not reported.

Table 3.4. Sensor reliability (250 cycle) testing results

6” Samples 250 cycle Testing		Nylon		Polyester		Cotton	
		%	R^2	%	R^2	%	R^2
Cycle Stage	304 Zig-Zag early	2.760	0.990*	2.492	0.988*	4.088	0.970*
	304 Zig-Zag middle	27.520	na	4.172	na	13.317	na
	304 Zig-Zag late	51.682	na	5.707	na	16.206	na

* p value < 0.001

The p values for the polynomial regression were below 0.001 for all three samples. At the late phase of 250 elongations and retractions, there was only a 5.707% error for the polyester sample compared to 16.206% error for the cotton sample and 51.682% error for the nylon sample.

To further compare how these samples changed over time, a two-way mixed effect ICC with absolute comparison was found comparing the early to middle cycles, then early to late cycles for all three samples (Table 3.5).

Table 3.5. Sensor reliability ICC results

6” Samples 250 cycle Testing ICC		Nylon	Polyester	Cotton
Cycle Stage	Early to Middle	0.690*	0.986*	0.813*
	Early to Late	0.493*	0.977*	0.710*

* p value < 0.001

The reported ICC values are single measures, and all have a p value below .0001. As it can be seen in Table 3.5, the sensor with a polyester based fabric substrate significantly outperformed both the nylon and cotton based substrates. The polyester sample had an ICC value of 0.986 between early and middle stages, and 0.977 between early and late stages.

Because Polyester had resistance measures that were consistent across all phases of testing, and had the lowest % error and highest ICC across all phases, it was selected as the optimal textile substrate.

Altogether, the results suggested that the optimal integration configuration utilized conductive thread 1 (silver-plated 22/1 dtex Nylon 6 twisted with 113/32 dtex PET 2-ply), stitch geometry 304, 6" sensor length, and 90% polyester/ 10% spandex fabric substrate (Figure 3.4).

The total change in resistance for this configuration was just over 2.5 KΩ, which was three orders of magnitude greater than that reported in previous developmental studies (Gioberto & Dunne, 2012), with soft sensors that showed a total change in resistance of 9Ω. We believe that this drastic difference in the reported results is due to the elongation of the actual thread, rather than the loop conductor method, which changes resistance due to multiple "shorts" of the thread, making contact with itself when elongated (Gioberto & Dunne, 2012). Stitch geometry 304 may have performed better than stitch geometries 402 and 406 because when the fabric is stretched, the configuration requires the thread to stretch more than the other two geometries. Stitch geometry 304 also has less "slack" between stitches, which may lead to more repeatable results. We also believe that polyester performed better than nylon and cotton substrates because polyester has a higher recovery percentage (Umar, 2016). Therefore, a stitch geometry that stretches the thread more, and a fabric that has a higher recovery percentage leads to greater sensor reliability.

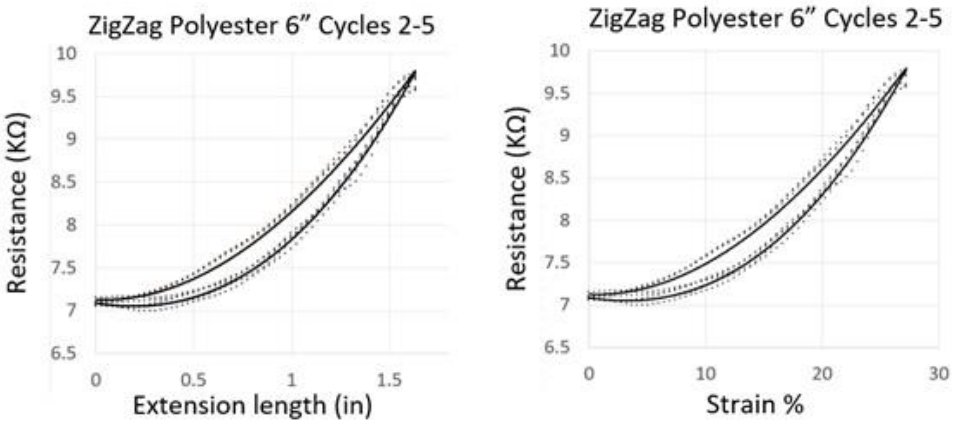

Figure 3.4: Scatterplots with polynomial regression lines for resistance & extension length (A) and resistance & strain (B) in the sample determined to represent the optimal integration configuration. The strain graph is displayed because it is commonly shown graph for stretch related sensors. Elongation is the bottom curve, and retraction is the top curve for both (A) and (B).

This paper was the first to test the reliability of soft sensor performance across 250 elongation-relaxation cycles, but further testing is needed to determine how the sensor will perform over a much larger cycle number, cycles of different velocities, cycles in different strain ranges, and after the exposure to normal wear, tear, and laundering. We hypothesize that an increase in resistance observed over time may have been due to the cross sectional area of the thread being continuously contracted and expanded, thus slowly having permanent deformation, causing the thread to settle in a semi-contracted position.

Textile-based stitched stretch sensors have a variety of potential smart clothing applications. Wearable motion capture can lead to a seamless form of biofeedback outside of a lab environment. This can have uses ranging from

helping individuals improve their exercise form, providing physical therapists quantitative data on the progress of their patients, or providing warning about joint movement into ranges associated with musculoskeletal injury after surgery (e.g., avoiding excessive hip flexion after hip arthroplasty) or for athletes (e.g., minimizing shoulder strain for baseball pitchers). Stretch sensors could also be used as an input mechanism for virtual reality experiences or for users to control aspects of their environment, such as lighting, mobility devices, communication devices, and other electronic systems.

3.6 Conclusion

Based on the current results, there is evidence that a soft stitched stretch sensor made from silver-plated conductive thread can be used as an accurate and reliable measure of change in resistance due to elongation. The optimal integration configuration utilized conductive thread 1 (silver-plated 22/1 dtex Nylon 6 twisted with 113/32 dtex PET 2-ply), stitch geometry 304, a 6" sensor length, and a 90% polyester/10% spandex fabric substrate. Textile-based stitched stretch sensors have a variety of potential smart clothing applications including biofeedback, injury assessment, input for virtual reality experiences and many more. Future studies should aim to test the effectiveness of these sensors across longer periods of use, in different environmental conditions, and after laundering. Another important future direction is to integrate these sensors into clothing, placing them strategically across joints and validating them with typical motion capture measures to determine whether they are valid for measuring human

movement. It will be critical during these tests to develop a user-friendly integration solution that is comfortable and appealing to users.

3.7 Declaration of Conflicting Interests

The authors have applied for a patent, Application No: 15/289,465, based on the technologies developed from this study.

3.8 Funding

The authors gratefully acknowledge the funding support from the Delaware Economic Development Office Matching Grant Program, Grant Number 109.

Chapter 4

DESIGN AND VALIDATION OF A GARMENT WITH CONDUCTIVE PAINT STRETCH SENSORS TO MEASURE PHYSICAL ACTIVITY

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4.2 Abstract

Motion capture and the measurement of physical activity are common practices in the fields of physical therapy, sports medicine, biomechanics, and kinesiology. Traditional motion capture systems are very expensive any only allow for data collections to be performed in a lab environment. In our previous research, we have tested the validity of a novel stitched stretch sensor using conductive threads. This paper furthers that research by validating a conductive paint stretch sensor that was integrated into a compression garment to measure movement. When these sensors are paired with an open sourced wireless microcontroller, it can lead to a much less restrictive form of motion capture. There is a need for affordable, accessible, comfortable, and wireless form of

motion capture like the one tested in this study to understand how meaningful, functional activities are performed in a natural setting.

4.3 Introduction

Motion capture and the measurement of physical activity are common practices in the fields of physical therapy, sports medicine, biomechanics, and kinesiology [1-2]. The majority of motion capture systems are expensive and use high-speed cameras positioned around the perimeter of a room that require data collections to be done inside of a laboratory [3-5]. Motion capture systems using inertial measurement units (IMUs) have become more common and allow for data collection outside of a lab setting, but these systems require bulky sensors to be attached to the body [3, 6-8]. These bulky sensors can make motion capture uncomfortable and lead to unnatural movement [3, 7-10]. A variety of textile-based sensors are currently being researched, but none have made their way into a common commercial product [11]. Textile sensors have been made from silver coated nylon thread [12-14], conductive knit fabrics [15], carbon nanotubes [16], dielectric elastomers [17], and P(VDF-TrFE) fibers [18]. All of these sensors track motion via measured changes in resistance or capacitance.

Most of the above sensors can be difficult to acquire or to fabricate. Because our lab focuses on affordable, open-sourced technology, we have been working within the domain of conductive threads, and more recently conductive paint, for our stretch sensors. Previous testing has shown that conductive thread could be stitched to produce a stretch sensor with a large change in resistance that was repeatable within a testing session [14]. Over time, the silver coating oxidized

quickly, leading to the formation of micro cracks in the silver coating, and a resistance too large to read. In order to investigate an alternative, even more affordable option, this study aimed to test the validity and reliability of carbon-based conductive paint sensors integrated within commercially available compression garments to measure flexion activity at the elbow and shoulder. We aimed to test the smart garment system on an adult and child participant to determine the accuracy of scaled versions of this affordable system relative to the established, more expensive trakSTAR motion analysis system [19].

There is a need for affordable, accessible, comfortable, and wireless forms of motion capture like the one tested in this study [20]. Data collections can track more meaningful, functional activities when they can be performed in natural, real-world environments where expensive and wired forms of motion capture aren't feasible. A low cost, wireless form of motion capture has the potential to allow for activity classification and continuous tracking of kinematics. This can support or replace existing data collection systems in the natural environment that typically rely on time-intensive coding of behavior from video. In order to document clinically relevant with differences patient populations, a wearable system should be accurate in tracking joint angles within roughly 15 degrees [21].

4.4 Materials and Methods

Fabrication and integration of the conductive paint sensors: The conductive paint (Electric Paint, Bare Conductive, United Kingdom) we purchased is originally intended to "draw" circuits, replacing where copper wire would be for short distances, or for cold soldering for those without soldering

experience. If the conductive paint was painted on to a sheet of paper using a ½" wide brush, the resistance would be roughly 52 ohms (Ω) / inch [22]. From our previous study, it was found that a 90% polyester/ 10% spandex jersey knit (Polyester Jersey Knit, Denver Fabrics, USA) was the best fabric substrate for a stretch sensor, due to its advantageous mechanical recovery, so we utilized this fabric substrate for the current project as well [23-26]. Different methods of applying the conductive paint to this fabric were tested, but the best results came from first cutting the fabric to a desired length (6 inches long by 2 inches wide, in the lengthwise or weft direction, then pre stretching the fabric to the maximum length it would be stretched when worn by the user, roughly 30% strain. The fabric was clamped at this distance using metal alligator clips and a custom 3D printed curved piece was placed underneath the fabric to further expose the individual threads. This part was printed with PLA plastic on a (Printrbot Plus, Printrbot, USA) 3D printer. Next, the conductive paint was applied as a thin bead down the center of the fabric and a plastic squeegee was used to make a roughly ½" wide line of paint down the center of the fabric sensor (Figure 4.1). The paint was left to dry while the fabric was still stretched for 30 minutes. Once the paint was dry in this configuration, the resistance was approximately 0.5 kΩ/inch. A 6" sensor where 5" was painted had an unstretched resistance of approximately 2.6 kΩ. Preliminary resistance data was measured on a multimeter (Model 9205B+, Adafruit Industries, USA) This sensor was cyclically elongated and retracted for 250 cycles on the same testing apparatus as in Greenspan et al's 2018 paper. When stretched to a 30% strain the sensor reached roughly 3.4 kΩ (Figure 4.2).

Figure 4.1. Conductive paint drying on stretched fabric

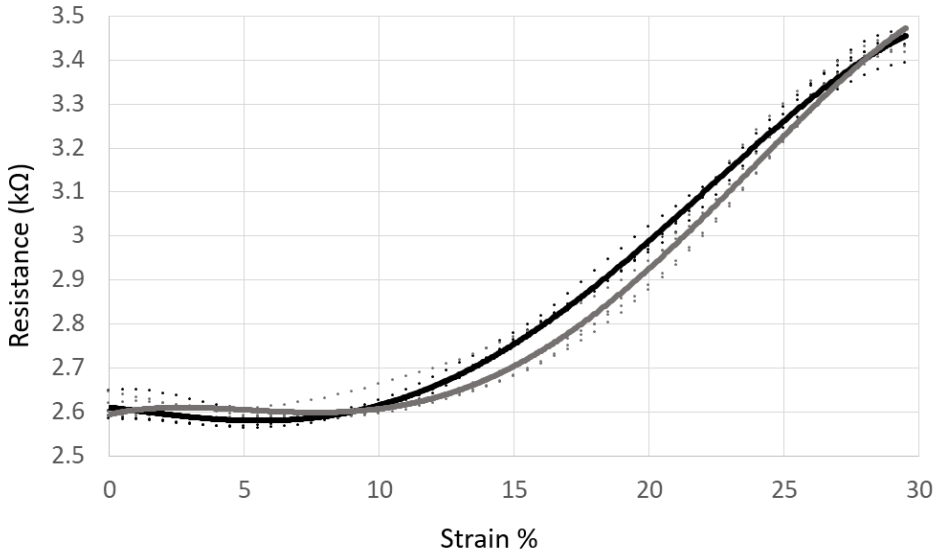

Figure 4.2. Conductive paint resistance v strain graph, cycles 123-126 of 250 cycles, dotted lines represent raw data with a 4th degree polynomial trendline for elongations (black line) and retractions (grey line)

For this study, we aimed to measure elbow and shoulder flexion. Therefore, one sensor was placed across each joint angle. For the elbow, the sensor was placed directly across the joint on the posterior side, and for the

shoulder, just under the axilla, running parallel to the transverse plane towards the scapula (Figure 4.3).

On both ends of each conductive paint sensor a 5mm tin-plated brass snap (Sewable Snaps – 5mm, Adafruit Industries, USA) was sewn into the sensor using conductive thread (Stainless Thin Conductive Thread – 2 ply, Adafruit Industries, USA). The other side of the button snap was soldered to 22 gauge stranded copper wire that went from the sensor to the microcontroller worn on the users back. Sensor data was read by a WiFi enabled microcontroller (WF32: WiFi Enabled PIC32 Microcontroller, Digilent, USA). For improved accuracy, sensor data was read through a 16 – Bit analog to digital converter (ADC) (ADS1115, Adafruit Industries, USA) rather than the 10 – Bit on board ADC. A constant 5 volt signal was sent out by the microcontroller and the voltage drop was recorded. Data was transmitted from the microcontroller through a router (Belkin N600 Router, Belkin, USA) over a Local Area Network (LAN) to a laptop with a custom written LabVIEW (LabVIEW, National Instruments, USA) program.

Garment Testing: For this study we tested two participants, one child (11 year old male) and one adult (24 year old male) to evaluate the scalability of the design. Both participants filled out an informed consent that was IRB approved. Each participant had a compression garment (Long-Sleeve Compression Tee, Badger, USA) purchased for his size. Then the placement for the sensors was marked on the garment while the user was wearing it. After the fitting, each stretch sensor was stitched to the compression garment at both ends of the sensor, perpendicular to the direction of stretch, in the locations described above.

Participants returned for testing in the laboratory when the garment was ready. Throughout all testing, participants wore the garment as well as three IMUs from the trakSTAR (Ascension, USA) motion analysis system to simultaneously track shoulder and elbow motion. A mid-range transmitter with Model 180, 6 degree of freedom sensors were used. One IMU was placed on the anterior side of middle of the forearm, the next on the anterior side of the middle of the upper arm, and the last on the T3 vertebra (Figure 4.3).

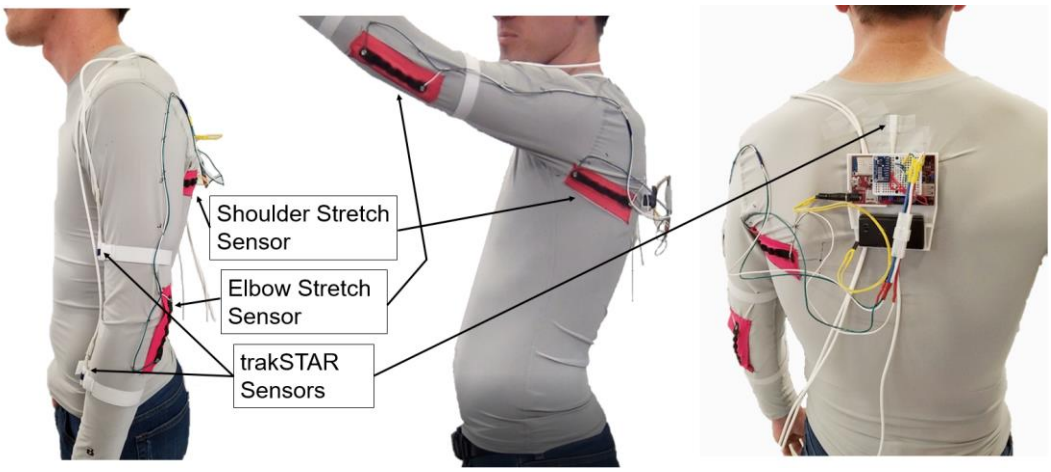

Figure 4.3. Sensor placements

Testing Protocol: Each participant went through three trials consisting of structured phases. Each trial consisted of a Calibration Phase and Static 1, Dynamic, Functional, and a second Static (Static 2) Phase. Data from Trial 1 of one of the participants for both the trakSTAR and conductive paint sensor smart garment from all phases can be seen in Figure 4.4.

Calibration: For all phases, the range of motion for the elbow was between 0 and 90 degrees, and for the shoulder was between 0 and 135 degrees. We

selected to use a dynamic calibration where the participant moved their arm as they would in the dynamic assessment over 10 cycles of 0-90 degrees of flexion for the elbow and 10 cycles of 0-135 degrees of flexion for the shoulder. Then the minimum and maximum values were recorded from both sensors. For the elbow, the minimum and maximum values were linearly mapped to 0 and 90 degrees respectively, and for the shoulder, they were mapped to 0 to 135 degrees respectively. These minimum and maximum values were kept constant for the remainder of the trial.

Static 1 Phase: The Static Phase consisted of a 10-second hold at distinct positions for both the elbow and the shoulder. One 10-second hold occurred at each of the following positions: 0°, 45° and 90° of elbow flexion, and 0°, 45°, 90° and 135° of shoulder flexion. Readings from the trakSTAR were used as the true reading.

Dynamic Phase: The Dynamic Phase consisted of moving the shoulder or elbow through a specific range of motion cyclically. The participant completed 10 cycles of elbow flexion between 0 and 90 degrees at 1 Hz then 10 cycles of shoulder flexion at 0.7 Hz. The different speeds were selected so that the angular velocity participants moved at was similar. Participants coordinated their movement speed to a metronome.

Functional Phase: The Functional Phase incorporated a functional task that involved both elbow and shoulder flexion. Each participant repeatedly lifted an empty box from waist level to a shelf that was eye level. This task was repeated for 10 cycles at 0.7 Hz. Participants coordinated their actions to a metronome.

Static 2 Phase: The protocol for the Static 1 Phase was repeated a second time for a Static 2 Phase so that reliability within a trial could be evaluated.

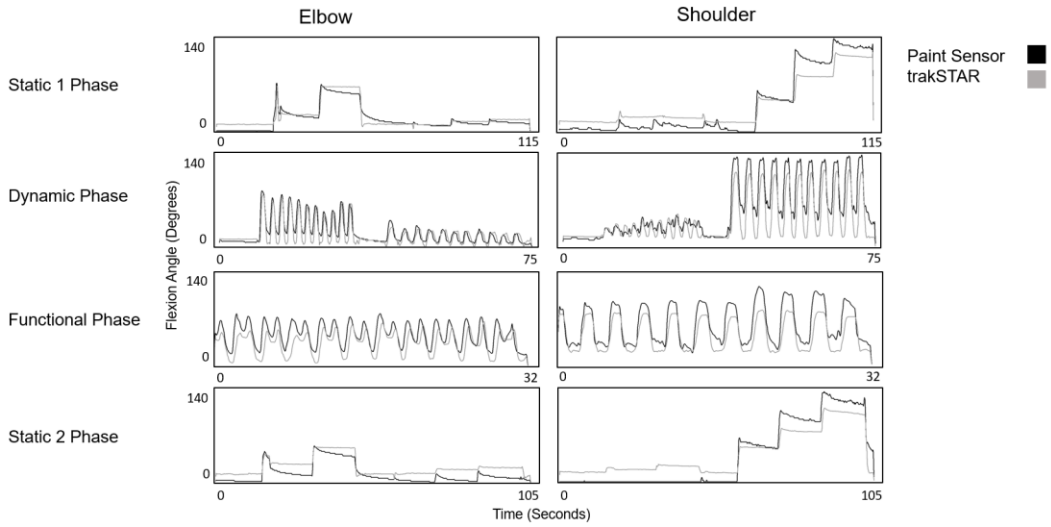

Figure 4.4. Graphs comparing data between the smart compression garment and trakSTAR from all phases for one trial

4.5 Results

Data was filtered through two, reversing direction, 2nd order, low pass Butterworth Filters with the cutoff frequency set at 3 Hz. For Static Phase trials, the middle 6 seconds were selected from the 10 second hold for analysis. For Dynamic and Functional Phase trials, the middle 6 cycles of the 10 cycles were analyzed.

A variety of statistical tests were used to analyze the data. Using the trakSTAR system as the true angle, first the average absolute error between the two systems was found to identify the accuracy during individual study phases in the data collection. These results can be seen in Table 4.1A and were Static 1

Phase 9.98, Dynamic Phase 22.21, Functional Phase 25.02, and Static 2 Phase 16.49 degrees of error. Both Static Phases produced better results than the Dynamic and Functional Phases because in the Dynamic and Functional Phases. During fast movements, the resistance value would increase quickly, and because of hysteresis from the sensor, this value wouldn't return to its original resistance before the next cycle began. This led to high resistance values throughout the trial, and furthermore inflated angle readings.

Next the average absolute error between the two systems was taken across all phases within each of the 3 trials to assess reliability over time. These results can be seen in Table 4.1B and were, Trial 1: 11.54, Trial 2: 13.63, and Trial 3: 21.49 degrees of error. These numbers show that the sensor decreased in accuracy over time, which is consistent with the Static 1 Phase producing better results than Static 2 Phase. A large portion of the increased error over the three trials came from the resistance exceeding the maximum resistance set during the calibration phase. With a more robust algorithm and calibration protocol, this error could be minimized. From averaging these results, it was also found that the elbow produced an average absolute error of 14.43 degrees and the shoulder 16.45 degrees. Averaging all of the data resulted in an average absolute error of 15.55 degrees.

Despite the Dynamic and Functional Phases producing less accurate results than the Static Phases, the two systems were still highly correlated in these phases of movement. To show this, an Intraclass Correlation Coefficient (ICC) for consistency was run for each trial of the Dynamic and Functional Phases. *P* value

was less than 0.001 for all tests. Averaging all the Dynamic Phase trials resulted in an ICC of 0.912 and averaging all of the Functional Phase trials resulted in an ICC of 0.846 (Table 4.1C). Despite the hysteresis of the system preventing a low mean error, there was an extremely strong correlation between the trakSTAR and the Smart Compression Garment.

Lastly, because the end goal was to see if the smart compression garment could record activity, not just kinematic data, the ability of the garment to record activity counts was also tested. We used the threshold method and defined an activity count as an angular velocity greater than 50 degrees per second. Angular velocity was filtered again, through two, reversing direction, 2^{nd} order, low pass Butterworth Filters with the cutoff frequency set at 2 Hz. Activity counts were only analyzed during the Dynamic and Functional Phase trials because Static Phase trials inherently lacked activity. The Dynamic Phase trials resulted in 79.39% accuracy and the Functional Phase trials resulted in 75.00% accuracy (Table 4.1D). While these accuracy values are already high, they would be expected to be significantly higher over a longer data collection that has more than 10 cycles for individual trials. There were roughly 10 activity counts registered over 10 cycles, thus not registering one activity count dropped the accuracy of the trial by 10%.

Table 4.1. Results across all trials

	Average Absolute Error	Static 1		Dynamic		Functional		Static 2	
		Elbow	Shoulder	Elbow	Shoulder	Elbow	Shoulder	Elbow	Shoulder
A	Participant 1	10.85	13.12	15.16	31.98	29.68	24.11	17.14	16.62
	Participant 2	6.69	9.26	18.75	22.94	9.73	36.54	17.83	14.36
	Participant Average	8.77	11.19	16.96	27.46	19.71	30.33	17.49	15.49
	Total Average	9.98		22.21		25.02		16.49	

	Average Absolute Error	Trial 1	Trial 2	Trial 3
B	Participant 1	12.63	14.80	23.21
	Participant 2	10.45	12.46	19.76
	Total Average	11.54	13.63	21.49

	ICC	Dynamic		Functional	
		Elbow	Shoulder	Elbow	Shoulder
C	Participant 1	0.949	0.960	0.800	0.933
	Participant 2	0.847	0.892	0.749	0.901
	Participant Average	0.898	0.926	0.775	0.917
	Total Average	0.912		0.846	

all data: p value < 0.001

	Activity Counts	Dynamic		Functional	
		Elbow	Shoulder	Elbow	Shoulder
D	Participant 1	96.97	81.08	87.10	75.00
	Participant 2	98.42	69.57	61.29	77.78
	Participant Average	97.70	75.33	74.20	76.39
	Total Average	79.39		75.00	

4.6 Discussion

In summary, there is a need for affordable, accessible, comfortable, and wireless forms of motion capture and activity classification that allows for data collections to be performed in natural settings during functional activities. Having a tool that meets these needs would allow researchers to have a more thorough understanding of movement and physical activity for a variety of patient populations outside of a lab setting. Building off of research that utilized conductive threads, we created a novel stretch sensor using conductive paint that changes in resistance when stretched. When this sensor was incorporated into a compression garment and placed across the shoulder and elbow joints, it accurately measured physical activity for both an adult and child, supporting the scalability of the solution. Activity counts were calculated throughout Dynamic Phase trials with 79.39% accuracy and throughout Functional Phase trials with 75.00% accuracy. With a combined average overall absolute error of 15.55 degrees compared to the trakSTAR data, this first prototype suggests conductive paint sensors hold high potential for motion and activity tracking. Further research should focus on a more robust calibration protocol and mapping algorithm to minimize error over time.

4.7 Acknowledgements

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Chapter 5

DESIGN AND VALIDATION OF A GARMENT WITH CONDUCTIVE PAINT STRETCH SENSORS TO MEASURE PHYSICAL ACTIVITY

5.1 Authors

Ben Greenspan, Andrea B. Cunha, Michele A. Lobo

5.2 Abstract

Before infants can transition between positions on their own, the positions they experience (e.g., supine, prone, upright) are reliant on parents/ caregivers. Parental education for advanced handling techniques (e.g., increased time in prone or upright positions) has been shown to facilitate future motor and cognitive development. The Get Around Garment (GG) a smart garment designed at the University of Delaware to track infants' body position (prone, supine, reclined/inclined, or upright) via a small sensor (accelerometer) incorporated into an adjustable belt, provides data regarding infants' position experiences. Tracking infants' positions throughout the day in the natural environment could allow for the collection of important normative developmental data. The aim of this study was to evaluate the feasibility, reliability, and construct validity of the GG for infants less than 6 months old. Moreover, to gather data showing the typical positioning activities of infants in the natural environment across a typical 1-2 day

period. **Methods:** Eighteen infants were enrolled in this study: seven for development of the garment (age: *Mean* = 3.6, *SD* = 1.2 months), and 11 to assess the psychometric properties of the final garment (age: *Mean* = 3.2, *SD* = 1.2 months). The visit consisted of: 1) *Structured Play Assessment*: the infant was placed by the researcher for 15-30 seconds into each of the following 5 positions: supine, prone, inclined about 45 degrees from upright, reclined about 45 degrees from upright, and seated upright (total of 2.5-minutes); 2) *Free Play Assessment*: parents were encouraged to go about their daily activities and interact with their infants in their typical manner for 40-60 minutes; and 3) a second 2.5-minute *Structured Play Assessment*. Infants wore the GG throughout all of the assessments. The entire visit was also video recorded. Positions (prone, supine, reclined/inclined, and upright) were coded from video Datavyu using software and were identified from sensor data using LabView programming. After the first visit, the GG was left with families. Parents/caregivers were asked to have their infants wear the garment during awake periods under adult supervision for 1-2 days. The validity of the smart garment was assessed by comparing data from the garment's movement sensor to data from behavioral coding of video. The reliability of the smart garment was assessed by comparing the garment's movement sensor data between the first and second Structured Position Assessments. Total percent time infants were placed in each position when the garment was left with parents was reported, and a Friedman's ANOVA was performed to check for differences. **Results:** The GG was considered comfortable, aesthetically pleasing, and easy to don and doff. Agreement between

the video and garment data (validity of the garment data) was greater than 89% and a very strong positive correlation was obtained (r (55) = .953; p < .0001). The reliability of the sensor garment was demonstrated to be excellent. The ICC was 0.997 (95% CI:.989-.996). Infants were placed in prone for 3.89%, inclined 5.20%, upright 37.77%, reclined 29.16% and supine 23.99% of the time. From Friedman's AVOVA, there were differences between these positions [X^2(4) = 25.309, p <.0001]. **Conclusions and Implications:** The GG might be a feasible, valid and reliable device to track the orientation of an infant in a natural setting. These data can better inform us about infants' typical positional experiences. Feedback from the device could potentially be used to alter caregiver-infant interactions, therefore optimizing infants' future motor and cognitive outcomes.
Key-words: infants, accelerometer, body position, wearable, connected-health

5.3 Introduction

At early ages when infants are not yet independently mobile, their experience of different body positions (e.g., supine, prone, reclined, inclined and upright) is reliant on parents/ caregivers. Being placed in a variety of positions and advanced handling techniques provided by parents, primarily having the infant spend more time in prone and upright positions, have been shown to improve future motor and cognitive outcomes (Fetters & Huang, 2007; Fogel et al., 1999; Lobo & Galloway, 2008, 2012). Thus, it is important to educate parents with infants at risk for developmental delays to engage in these practices to close the gap between these infants and those with typical development. Feedback from wearable sensors might have the potential to shape infant-caregiver interactions

and optimize the trajectory of infants' motor and cognitive development (Sgandurra et al., 2016, 2017).

Accelerometers are examples of wearable sensors that measure accelerations of a body segment typically in three dimensions (Lang et al., 2013, Migueles et al., 2017). They have been widely used and shown to be feasible, valid and reliable tools in research and clinical settings to measure physical activity, as well as sedentary time in healthy children and adults and in individuals with disabilities (Bailey et al., 2013; Bassett et al., 2012; Lang et al., 2013; Sarker et al., 2015; Sasaki et al., 2011; Waddell et al., 2016). In infants, wearable sensors have begun being used only in recent years (Hauck et al., 2013; Sgandurra et al., 2016).

Studies available using wearable sensors in infants with typical development or with risk for developmental delay have focused on upper-extremity and lower-extremity movement activity (Oghi et al., 2008; Smith et al., 2015; Shida-Tokeshi et al., 2018; Trujillo-Priego et al., 2017). Measurement of this activity is aimed at establishing normative values to assist with early identification of activity delays and to assist in the evaluation of the effectiveness of early intervention programs. However, none of these studies to our knowledge has aimed to assess the psychometric properties of accelerometers for measuring body positioning activity. One study aimed to determine the minimal number of days and hours of sensor wear necessary to reliably measure daily physical activity in infants (Pitchford et al., 2017). Accelerometers placed on the right wrist and ankle were compared against caregiver report, in order to remove

activity counts not produced by the infant (e.g., external sources of motion such as parents carrying infants or infants moved in a stroller or bouncer chair). The authors showed that at least three days of wear during the daytime hours (including nap times) can produce reliable physical activity measurement (Pitchford et al., 2017).

Wearable sensor technology can provide the opportunity to capture full-day movement data from infants across tasks and contexts in the natural environment (Shida-Tokeshi et al., 2018; Smith et al., 2015; Trujillo-Priego et al., 2017). However, it is unknown whether wearable sensors are accurate and consistent for measuring body position of infants. A few studies have assessed the psychometric properties of accelerometers for measuring children's body position. Accelerometers (e.g., ActiGraph and activPAL) have been compared between one another or with measures from direct observation, showing the sensors to be valid for assessing children's sitting and standing time in a school classroom setting (Ridgers et al., 2012; Ridley et al., 2016). These tools have been implemented to determine the amount of time that children spend sedentary (Hänggi et al., 2013; Pate, O'Neil, & Mitchell, 2010).

To our knowledge, only one study aimed to design and develop a wearable sensor for tracking infants' body position in order to measure adherence to Kangaroo Mother Care; infants' body position was measured for several days in the hospital and out of the hospital. The results showed that the sensor was feasible and acceptable to parents for wear, but validity and reliability of the sensor were not evaluated (Rao et al., 2018). Thus, there is a lack of evidence

regarding the validity and reliability of wearable accelerometers for measuring infants' body position. We also lack normative values related to time infants spend in different body positions and how this might relate to developmental outcomes. The purpose of this project was to design a smart garment to measure the amount of time infants are placed in distinct positions throughout the day via an integrated 3-axis analog accelerometer. Then, we aimed to evaluate the feasibility, reliability, and construct validity for use in natural settings for infants less than 6 months old. Finally, we aimed to gather data showing the typical positioning activities of infants in the natural environment across a typical 1-2 day period. We hypothesized that the smart garment, named the (GG) would be a feasible, valid, and reliable tool for assessing infants' body position in natural environments, and that it could allow for the collection of important normative developmental data with the potential to be of clinical utility for identification of developmental delays and evaluating the effectiveness of interventions.

5.4 Methods

Study Design and Setting

The data reported here represent a subset of data from a larger longitudinal intervention study. Recruitment of participants, informed consent, and data collection were completed in accordance with the regulations set by the University of Delaware's Institutional Review Board.

Participants

A convenience sample of 18 infants was enrolled in this study: seven for development of the garment (age: *Mean*= 3.6, *SD* = 1.2 months; gestational age: *Mean* = 38.8, *SD* = 1.1 weeks), and 11 to assess feasibility, validity, and reliability of the final garment (age: *Mean*= 3.2, *SD* = 1.2 months; gestational age: *Mean* = 38.5, *SD* = 2.1 weeks).

All participants were recruited via advertisement in the local community. Infants were eligible if they were (1) able to safely engage in the data collection activities, 2) between 1-5 months old and not consistently rolling; and 3) without gross motor delays [>10th percentile on the Alberta Infant Motor Scale (AIMS)], visual impairments, or medical equipment that would restrict positioning. Additional sample characteristics are presented in Table 5.1.

Table 5.1. Participants' demographic and health-related information (percentage, *Mean* ± *SD*).

Infants	Garment development (n=7)	Final garment testing (n=11)
African-American	0%	0%
Caucasian	71.4%	72.7%
Asian	0%	9.1%
Multiple races	28.6%	18.1%
Gestational age (weeks)	38.5±1.1	38.5±2.1
Age (months)	3.6±1.2	3.2±1.2

Motor development		
AIMS total score	14.0±5.5	12.6±3.8
AIMS percentile	48.6±20.4	52.3±23.2

The Smart Garment: The Get Around Garment

The original design of the garment included a 3-axis analog accelerometer (ADXL335, Adafruit Industries, USA), stored on a SD card in a data logging shield (Assembled Data Logging shield for Arduino, Adafruit Industries, USA), and processed through a microcontroller (Arduino Uno - R3 microcontroller, Adafruit Industries, USA). This device was powered by a 9-volt (V) battery contained in a belt the infant wore over a soft onesie of their choosing (Figure 5.1A). In terms of function, this device worked well, but was ultimately too large for wearing for an extended period of time. This large size also led to more sensor drift because it couldn't be as easily integrated into the onesie.

We aimed to create a garment that would allow for safe, comfortable integration of an accelerometer while minimizing sensor drift. For the final prototype, a 3-axis analog accelerometer was selected (ADXL335, Adafruit Industries, USA) with a small microcontroller (Feather 32u4 Adalogger, Adafruit Industries, USA). The dimensions of the final sensor unit were 2 inches long by 0.95 inches wide and 0.45 inches tall (Figure 5.1B).

By using the Feather microcontroller, the device could also be powered by a small lithium ion polymer battery rather than a traditional 9V battery. A low profile custom 3D printed case, PLA plastic printed on a 3D printer (Printrbot

Plus, Printrbot, USA) was then designed using SolidWorks (Dassault Systems, France) to house the electronics. The sensor unit (electronics and case) was inserted into a vinyl pocket stitched into the side of a soft onesie (0-6 Month, Short-Sleeve Unisex Adjustable Snap-N-Grow 100% Cotton Bodysuit, MiracleWear, USA). Because this study aimed to identify infant positioning before infants were rolling, the position of infants primarily changes within the sagittal plane. For this reason, the sensor was placed on the side of the onesie where it would be the most comfortable for the infant as infants least often place weight on their sides. The onesie had a snug fit and a crotch strap that together minimized sensor drift. Furthermore, the vinyl pocket was secured with an elastic belt to keep the sensor in place. The sensor could easily be removed from the vinyl pocket to allow for facility of cleaning the onesie. This final prototype can be seen in Figure 5.1C. One of two different sized batteries (Lithium Ion Polymer Battery, Adafruit Industries, USA) was used, 150 milliamp hours (mAh) or 350 mAh, to allow for different lengths of testing, just under 12 hours or 28 hours respectively. Raw accelerations from the accelerometer sampled at 1 Hz were recorded on all three axes through an integrated 10bit analog to digital converter (ADC) which was then converted to a pitch angle from the formula: $pitch\ \theta = \arctan(\frac{y-510}{\sqrt{(x-510)^2+(z-510)^2}})$.

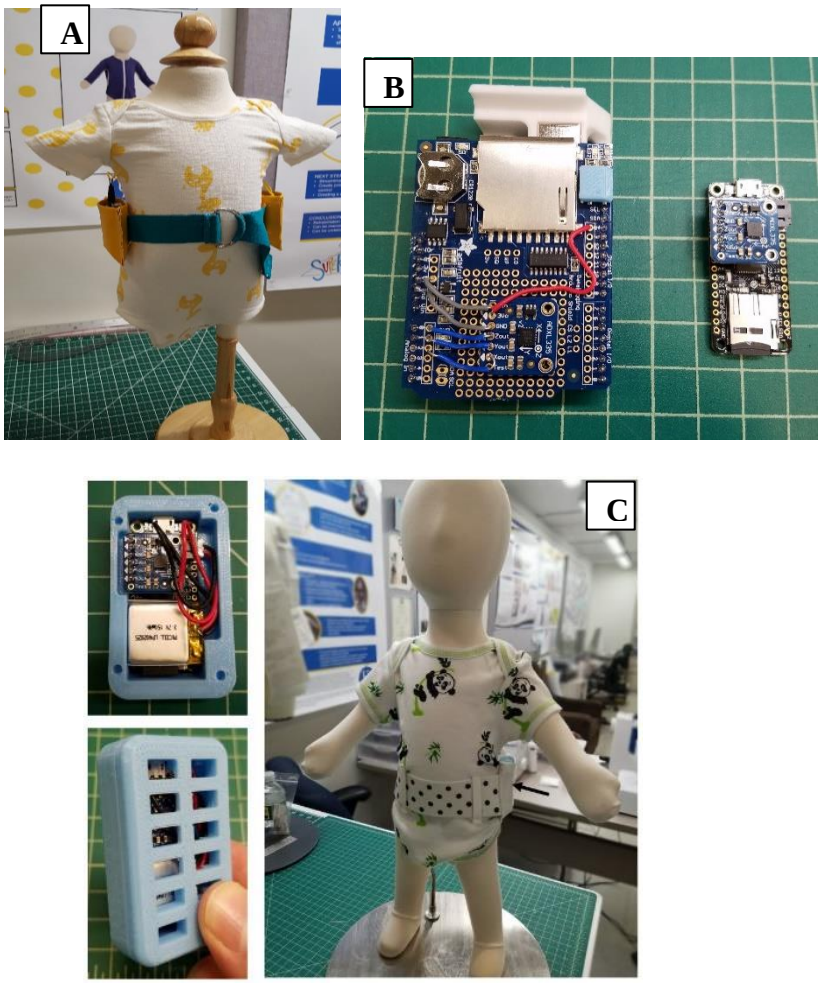

Figure 5.1. A) Initial prototype of Get Around Garment; B) Size reduction of microcontroller from the initial prototype (on the left) to the final prototype (on the right); C) Final prototype of Get Around Garment

Assessment Procedures

Infants were assessed across two visits. The first visit began with infants donning the GG. Then, infants engaged in: 1) one 2.5-minute *Structured Play Assessment* during which the infant was placed by the experimenter for 15-30 seconds in each of the following 5 positions: supine, reclined about 45 degrees from upright, seated or standing upright, inclined about 45 degrees from upright,

and prone; 2) one 40-60 minute *Free Play Assessment* during which parents were asked to go about their typical daily activities (*Mean*= 44.2, *SD* = 4.5 minutes); and 3) a second 2.5-minute *Structured Play Assessment* similar in protocol to the first (Figure 5.2 A-B*).* Data from the accelerometer in the GG was collected simultaneously with video recorded data throughout all assessments.

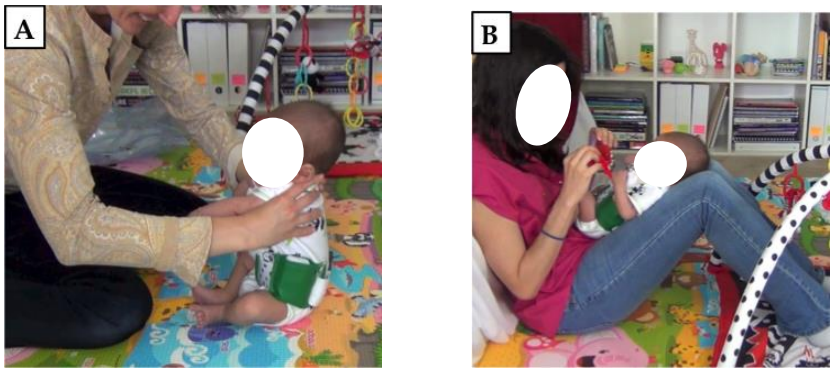

Figure 5.2. A) Examples of activities for the Structured Play Assessment and B) Free Play Assessment

After the first visit, the GG was left with families. Parents/caregivers were asked to have their infants wear the garment during awake periods under adult supervision for 1-2 days. To exercise an abundance of caution, parents/caregivers were instructed not to leave the garment on infants during periods of sleep or periods without adult supervision. Families were also asked to document the times the garment was worn and infants' general activity (periods the infant was awake, asleep, eating, having a diaper change) in a Daily Log (see Appendix 1). Experimenters returned after 2 days for a second visit, during which the GG was retrieved and parents were asked to fill out a Garment Perception Questionnaire related to the comfort, ease of use, and aesthetics of the garment.

Assessment of Feasibility, Validity, Reliability, and Activity

Feasibility of the Get Around Garment for Caregiver Use in the Home. Measures to assess feasibility included the times parents reported in the Daily Log that infants wore the garment and parents' ratings on the Garment Perception Questionnaire (Table 5.2). These data were analyzed descriptively.

Construct Validity of the Get Around Garment. Time spent in each of five positions (supine, reclined, upright, inclined and prone) during each of the assessments was coded from video by 2 independent reliable coders using Datavyu software. Prior to coding the data for analysis, the coders were trained until they reached >90% intra- and inter-rater reliability. These levels of reliability were assessed for maintenance across 20% of the data. Behavioral coders had intra-rater reliabilities of 95.9 ± 5.3% and 96.1 ± 5.1% and an inter-rater reliability of 95.1 ± 5.3%. The total time spent in each of the five positions was exported into an Excel file. Sensor data were saved on a memory card and transferred to a computer for categorization using a custom LabVIEW program (LabVIEW, National Instruments, USA). The LabVIEW program exported the total time spent in each of the five positions. Data from the garment's sensor was time synced to video data by matching the first frame where the infant transitioned from supine to reclined during the first structured play assessment. Cutoff angles were defined as 0°-30° for supine, 30°-55° for reclined, 55°-115° for upright, 115°-150° for inclined, and 150°-180° for prone. To evaluate the construct validity of the GG, we compared and examined correlation between the data categorized from the garment's sensor with data coded from video

throughout the entire first visit within each infant. The data was correlated using bivariate Pearson two-tailed correlation in SPSS. A correlation coefficient (r) describes the strength of the correlation, with values of r = .40-.59 considered as "moderate", r = .60-.79 as "strong", and r = .80-1.0 as "very strong" (Field, 2013).

Reliability of Data from the Get Around Garment. Sensor data between the first and second Structured Play Assessments were compared to assess the reliability of the garment for measuring body position within a session. First, we averaged the angular data for the middle 10 seconds for each trial (position) of both the first and second Structured Play Assessments. The difference between the values for each position was then calculated and averaged across participants to find the total average difference between the first and second Structured Play Assessments for each position. The intraclass correlation coefficient (ICC) between the angular data from the first and second structured assessments was calculated. Values greater than .80 were considered "good", and those greater than .90 "excellent" (Portney & Watkins, 2009).

Positioning Activity. Sensor data from the one to two days the GG was left in the home was analyzed to describe the positioning activity of the participants. These data were analyzed descriptively. In addition, the time spent in each position was compared using Friedman's ANOVA using Wilcoxon signed rank post-hoc tests with Bonferroni corrections (significance level was set at p <.010) to compare time in prone, the position we expected to be least experienced, with time in each of the other positions.

5.5 Results

Feasibility

The total time that each infant wore the garment across a typical 2-day awake period was 3.13 ± 0.5 hours. Parents' perception ratings of the garment are presented in the Table 5.2. Parents found the garment to be comfortable, easy to don and doff, and attractive. They reported they would be likely to use the garment again.

Table 5.2. Questions, rating Scale, and outcomes from the parent perception questionnaire

Perception Questions	Rating scale	Rating outcomes
How comfortable do you think you child found the garment?	1= very uncomfortable; 2= uncomfortable; 3= neutral; 4= comfortable; 5= very comfortable	4.5 ± 0.5
How easy was to put on the garment?	1= extremely difficult; 2= difficult; 3= neutral; 4= easy; 5= very easy	4.2 ± 0.9
How easy was to take off the garment?	1= extremely difficult; 2= difficult; 3=neutral; 4= easy; 5= very easy	4.4 ± 0.7
What is the likelihood you would choose to use this garment?	1= not wear; 2=probably not wear; 3= neutral; 4= wear; 5= wear often	4.2 ± 0.4
How attractive do you find the garment?	1= very unattractive; 2= unattractive; 3=neutral; 4= attractive; 5= very attractive	4.2 ± 0.7

Construct Validity

Comparing the total time infants spent in each position was evaluated by the sensor data from the GG with data from the video coding, there was an agreement of 89.36 ± 2.62% (6 hours and 6 minutes of data in total). There was a very strong positive correlation between the total time obtained from sensor data and from the video coding ($r(55) = .953$; $p < .0001$).

Reliability

Reliability between the First and Second Structured Assessments were assessed. A visual representation of what the structured assessments look like from the entire data collection can be seen in Figure 5.3.There was a 6.87 ± 0.74° average difference between measurements during the First and Second Structured Play Assessments. The ICC was 0.997 (95% CI:.989-.996) between the total angular data from the first and second structured assessments, indicating excellent reliability.

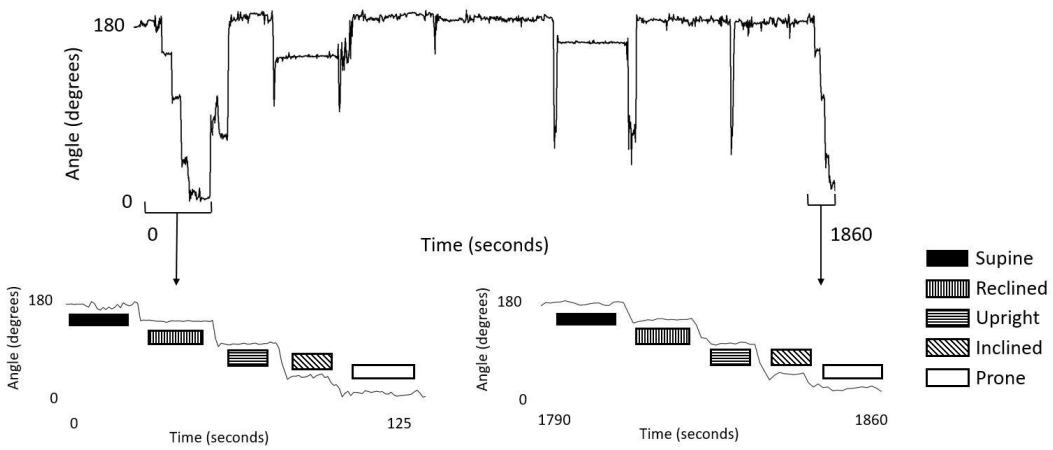

Figure 5.3. Example of data from the First and Second Structured Play Assessments. Pitch angle from the sensor throughout both assessments can be seen on the top; the First Structured Play Assessment can be seen on the bottom left, the Second on the bottom right.

Positioning Activity

We collected a total of 34 hours and 29 minutes of data from wear at home led by parents between the first and second visits. During this time, infants were placed in prone 3.89% (*SE* = 2.13%) of the time, in the inclined position 5.20% (*SE* = 1.43%) of the time, upright 37.77% (*SE* = 4.75%) of the time, reclined 29.16% (*SE* = 4.83%)of the time, and supine 23.99% (SE = 4.54%) of the time (Figure 5.4). After analyzing these data using Friedman's ANOVA it was found that there was a difference in the time spent among the positions [$X^2(4) = 25.309$, $p < .0001$]. Post-hoc testing comparing time in prone to time spent in each of the other positions showed significant differences between prone vs. upright position ($z = -2.934$; $p = .003$); prove vs. reclined position ($z = -2.578$; $p = .010$); prone vs. supine position ($z = -2.578$; $p = .010$).

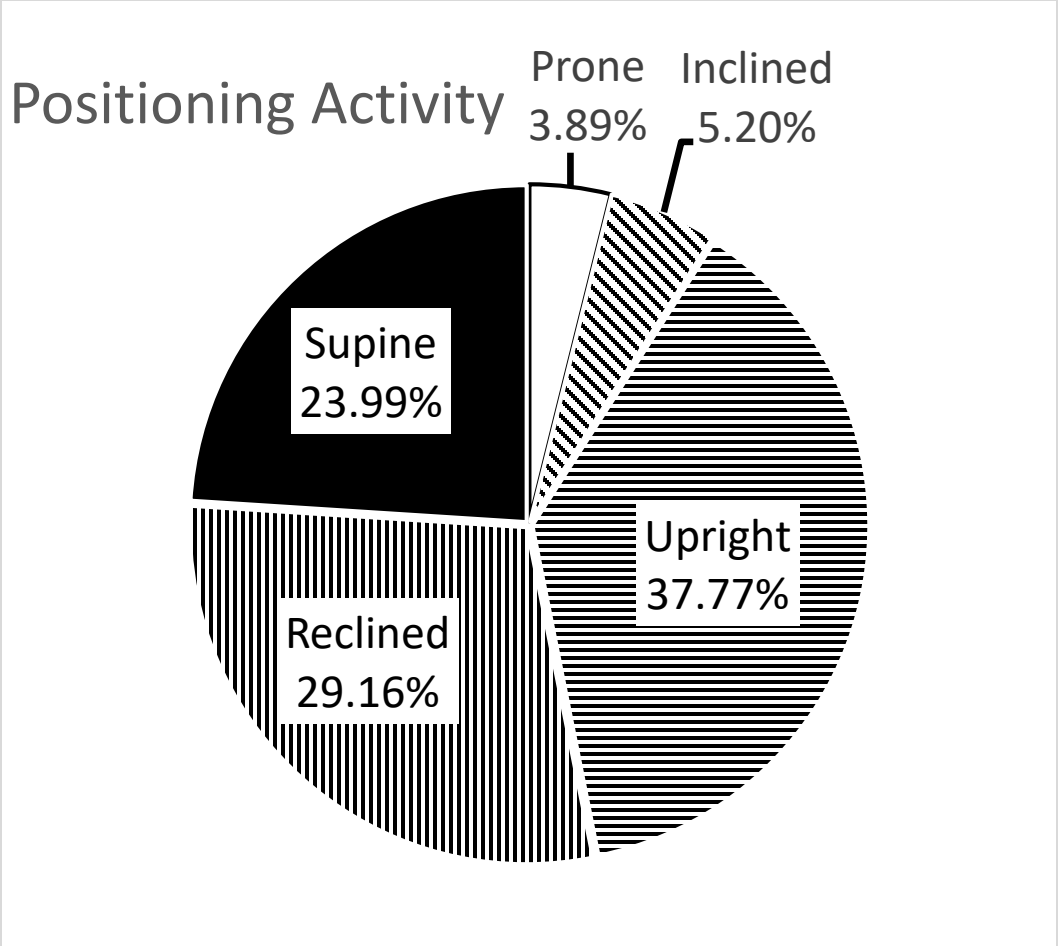

Figure 5.4. Percent time spent in each position during typical daily activity (based on data collection in the home between visits)

4. Discussion

The purpose of this study was to assess the feasibility, reliability, and construct validity of the GG as well as to gather data showing the typical positioning activities of infants in their natural environment. This study is the first to evaluate the use of a smart garment with an integrated accelerometer for the measurement of body position in infants less than 6 months of age.

Our first hypothesis that the garment would be feasible, valid, and reliable tool for assessing infants' body position in natural environments was confirmed.

The results suggest the GG was considered comfortable, aesthetically pleasing, and easy to don and doff. Parents were able to safely and appropriately use it without the supervision of a researcher across a 1-2-day period in their home setting. Data collected from the garment's sensor and processed with our custom program showed a high level of accuracy (<89 %) and very strong and positive correlation compared to the gold standard, behavioral video coding. Also, the reliability of the GG was demonstrated to be excellent. Thus, this smart garment might be an appealing tool because it allows to accurately and reliably track infant's body position in the natural environment.

The data collected and analyzed when the researchers were present suggest drift of the sensor in the GG was minimal (<7°) even after multiple changes in position and daily activities. This is encouraging as most commercial physical activity monitors have an error between 10 and 15% for activity measurement such as recording steps (Lee, Kim, & Welk, 2014). Moreover, the use of accelerometers has been shown to be a feasible, reliable and valid alternative for measuring sedentary time and physical activity instead of self-report questionnaires, which are unable to capture movements during the day especially in younger populations (Bassett et al., 2012; Sasaki et al., 2011; Sirard et al., 2001; Warren et al. 2010). Thus, monitoring and classification of body position in infants may not only be possible to perform accurately, but the level of accuracy may be greater than that observed for other activity classification systems currently available.

Our second hypothesis was that the GG would allow collecting important normative developmental data, showing the typical positioning activities of infants in the natural environment across a 1-2 day period. In this initial study, infants were only placed in prone ("tummy time") for 3.79% of the time while awake. These data suggest that, despite the efforts from the National Back to Sleep, Prone to Play Campaign, parents remain reluctant to place their infants in prone in the early months of life. Also, despite the American Academy of Pediatrics (American Academy of Pediatrics, 2006) advising that caregivers should dedicate time for active play every day and limit time spent in devices that restrict movement (e.g., car seat, stroller), the infants spend more than half of their time in supine or reclined (likely supported in an infant container, such as a bouncing seat or car seat).

A tool like the GG that can reliably and accurately measure body position has the potential to assist early intervention providers in tracking the performance of positioning behaviors that are associated with future developmental outcomes. Studies have been shown that infants with typical development at 1-6 months old who moved more their arms (measured via wearable sensors) had higher increases in language and cognitive performance and larger chances in movement's characteristics were related to higher motor scores across time (Shida-Tokeshi et al., 2018). Moreover, wearable sensors have been considered an important tool to characterize spontaneous movements and physical activity in infants with typical development and at risk for motor delays; and these data might be a reference standard for detecting delays (Gima et al., 2011; Oghi et al., 2008; Smith et al.,

2015; Trujillo-Priego et al., 2017). Despite wearable sensors for measuring position from infants in research and clinical settings are still unknown, three weeks of enhanced handling and positioning experiences provided to infants by caregivers showed short-term effects for improving behaviors in prone, head control, reaching, and sitting behaviors, as well as long-term effects for object exploration, crawling and walking behaviors (Lobo & Galloway, 2012). Therefore, data from the GG or similar smart garments may be helpful in allowing us to establish normative values describing parent-child positioning practices in low- and high-risk developmental groups.

Feedback from sensors have been shown to help modify behavioral performance in other populations. For example, it has been shown that simple feedback from a Fitbit increased wellness and physical activity within a workplace (Lee et al., 2019). It has also been shown that wearable sensors are accurate and useful for self-monitoring within health care settings (Appelboom et al., 2014). They might further be helpful in providing caregivers with feedback that can encourage them to change their daily positioning practices to provide infants greater opportunities to experience body positions, such as prone, that can positively impact developmental outcomes.

Future studies should focus on generating larger normative datasets to characterize positioning behavior in the home environment, as the sample size of the current study is modest. Future studies should also evaluate whether feedback provided to parents can change their infant positioning practices. Important aspects for evaluation include the manner in which parents prefer to receive this

feedback and levels of activity that should be encouraged for achievable, behavior-changing goal-setting. For example, it is not clear whether it is necessary to encourage parents to attain a specific threshold of positioning performance or it might be sufficient to encourage them to meet or exceed their performance from the days prior. These are important and interesting questions for future research.

5.6 Discussion

Accelerometers can be integrated within infant clothing to create comfortable garments that are feasible for home use and reliable in measuring infants' body position across time in natural environments as early as the first months of life. Algorithms can be created to correctly classify body position into five categories (supine, reclined, upright, inclined, and prone). The smart garment used in this study, the Get Around Garment (GG), and similar garments have the potential to serve as tools to support the evaluation of early parent-child positioning activities for the purpose of creating normative datasets and providing feedback to change parent-child positioning behaviors.

5.7 Acknowledgements

The authors would like to thank all the participants and the research assistants for coding the data. This study was funded by the University of Delaware Research Foundation, the State of Delaware Federal Research and

Development Grant Program (#109), and the National Institutes of Health: Delaware Clinical and Translational Research Program ACCEL Pilot Grant Award.

Chapter 6

VIDEO GAME PLAY IN CHILDREN WITH MOVEMENT CHALLENGES: IDENTIFYING & ADDRESSING BARRIERS THROUG THE DEVELOPMENT AND TESTING OF ADAPTIVE GAME CONTROLLERS

6.1 Authors

Ben Greenspan and Michele A. Lobo

6.2 Abstract

Video games have become extremely popular across the globe and are played by millions of children every day. Traditional video game controllers haven't changed in nearly two decades and can be very difficult for children with upper extremity movement challenges to use. Playing video games, especially with friends, can be an important social outlet for gamers with movement challenges, and outdated video game controllers should not be a barrier to play. Nine children were interviewed to identify common challenges children with arm movement impairments encounter when trying to play video games using standard systems. It was found that the triggers on the backside of the controller and small buttons are consistently challenging and that buttons on the far side of the controller are challenging for children that play with one hand. Based on these findings, video game controller adapters were designed and evaluated for two participants. The adaptive controllers allowed the participants to play games that

were previously too challenging and improved their ability to utilize challenging buttons within the games they were currently playing.

6.3 Introduction

Video games have become extremely popular across the globe and are played by millions of children every day. From a study by The National Purchase Diary (NPD) Group in 2018, it is estimated that over 50 million children between the ages of 2 and 17 play video games [1,2]. For most children, video game is a fun activity to allow for pleasure and relaxation with friends; for children with disabilities, socialization with friends is often limited and video games have the potential to serve as an important social outlet [2]. Among the major video game consoles (Xbox One, PlayStation 4 and Nintendo Switch) the button layout is very consistent. This is especially true for Xbox and PlayStation, whose basic controller designs haven't changed for nearly two decades (Figure 6.1) so that players can gain experience with the controller and transitions between versions of controllers are smooth [4,5]. For children with upper extremity mobility impairments or difficulty with fine motor skills, holding traditional controllers and activating their small buttons may be very challenging.

Figure 6.1. Left: PlayStation 2 controller released in 2000. Right: PlayStation 4 controller currently sold in 2019

A standard video game controller costs between $60 and $80. Adaptive video game controllers exist but can be very expensive, up to $1,945 [5-8]. Some charities and organizations such as The AbleGamers Charity, Warfighter Engaged, Special Effect, OneSwitch and The Controller Project have do-it-yourself options or grants to subsidize some of these extreme costs, but these adaptive controllers still remain difficult for many people to access [9-13]. Adaptive controllers have previously been designed and sold by third party companies until the Xbox Adaptive controller was released in September of 2018 [14]. This controller is priced at a more reasonable $100 and encompasses the same style guidelines as the rest of the Xbox products.

Existing adaptive video game controllers support individuals with a variety of functional abilities. Adaptations exist that employ larger inputs for more challenging buttons for individuals with limited hand control, movement of buttons to locations where they can be pressed using the head, mouth, or other body parts for individuals with limited arm movement, or placement of all of the buttons in locations where they can be accessed using one hand for individuals with limited use of one upper extremity [15]. While current adaptive controllers can help individuals with varying abilities, there is limited research published on how individuals with upper extremity impairments currently interact with standard or adaptive controllers, which buttons are more challenging than others, and how helpful adaptive controllers can be.

Video game play research has instead focused on evaluating ways that existing video game systems may be utilized to improve health and behavioral

outcomes. Current research has primarily focused on using existing video games in order to promote health outcomes for people with challenges such as obesity or aggression [16-28]. This can involve using outcomes in existing video games as goals for different types of interventions or instrumenting a therapy device with simple sensors that can be translated into an interactive game so the therapy seems more exciting. A number of clinical research studies have used commercial products, such as playing Wii Sports, to improve cognitive health, evaluate exercise intensity levels, or promote physical activity [21-23].

Research has also focused on developing and evaluating novel software for individuals with physical disabilities, designing "special" video games that aim to serve as rehabilitation tools to target and improve the movement abilities of individuals. One example of these "special" games included tilting a joystick left or right to play a pong game for in a social, competitive arm rehabilitation activity. Participants reported higher levels of enjoyment when playing the game with others [30]. Another example involved having a participant shift his/her weight on a pressure mat to control location selection while playing Tic-Tac-Toe for balance rehabilitation [31]. In this study participants were able to practice for longer durations leading to improved outcomes. Only one non-peer-reviewed product was identified that aimed to understand which games were being played by gamers with disabilities and which assistive technologies they used to play them [32]. Survey findings suggested that gamers with disabilities were playing primarily mainstream games; the primary reason they played games was for pleasure rather than for rehabilitation. Out of 154 gamers surveyed, 91 had upper

limb physical disabilities. It was found that a personal computer was the most common gaming platform. Only 18 participants used a customized controller, 3 used a one-handed controller, and 7 used an alternative controller. A customized controller referred to a standard controller with adaptations; an alternative controller referred to a controller designed for the general consumer without disabilities but that was not an official controller from the manufacturer. The most common physical accessibility adaptation for the gamers was key remapping on a keyboard (83 participants). Thus, there is a need to understand what the common challenges are for players with movement impairments during video game play.

The aims of the current study were: 1) to understand how children with upper extremity movement impairments interface with standard video game controllers; 2) to identify challenges standard controllers present for these children; and 3) to develop and evaluate the effectiveness of adaptive video game controllers aimed at addressing these challenges to facilitate video game play.

6.4 Methods

This study was conducted in two phases. During the first phase, participants were interviewed to understand how children with upper extremity movement impairments interface with standard video game controllers and to identify challenges standard controllers present for these children. During the second phase, custom video game controller adapters were developed and their effectiveness was evaluated relative to standard controllers for a subset of the participants. In this phase, we compared game play performance between children with movement impairments and age-matched peers.

Participants

Phase 1: Interview

A total of 9 participants, average age of 10.9 ± 3.1 years old, with self-reported upper extremity movement impairments that impact their video game play were interviewed (Table 6.1). Inclusion criteria for the study were that the children were between 7 and 17 years old, reported having trouble moving their arms and/or hands when using video game controllers, were able to communicate using spoken English, and enjoyed playing video game systems they owned. Participants were recruited through local community connections and social media.

Table 6.1. List of participants that participated in Phase 1

#	Age	Gender	Upper Extremity Impairment Reportedly Related to:
1	7	Male	Cerebral palsy, left side hemiplegia
2	7	Female	Spinal muscular atrophy & cystic fibrosis
3	9	Male	Limb difference, right arm below elbow
4	9	Male	Symbrachydactyly
5	10	Male	Right arm amputation
6	12	Male	Cerebral palsy *
7	13	Female	Cerebral palsy, right side hemiplegia *

8	15	Male	Cerebral palsy
9	16	Male	Cerebral palsy

*continued to Phase 2 of the study

Phase 2: Adaptive Controller Development and Evaluation

Two participants, ages 12 and 13 (identified in Table 6.1), transitioned into the second phase of the study to have a custom video game controller adapter designed for them based on their movement level and functional needs. These participants were selected based on their willingness to continue and their availability for future, in-person visits. The 12-year old child had cerebral palsy and was unable to hold a standard controller due to hand contractures. The 13-year old had hemiplegic cerebral palsy affecting the right side of her body as well has right hand contractures. As a result, she only utilized her left hand for video game play. Two age matched peers, within 1 year in age, without reported movement impairments were also recruited from the local community through word of mouth. These age-matched peers with typical movement ability were included so we could evaluate whether video game performance differed between children with and without movement impairments when children with movement impairments used the standard or the adaptive controllers.

Methods

Phase 1: Interview

Six interviews were conducted in person in participants' homes, and three interviews were conducted via video chat. Participants were asked a series of

questions that aimed to inform us how they currently play video games and use video game controllers. Questions related to which consoles, controllers, controller adaptations, and games participants use and play, as well as to what challenges present when they play video games. Specific questions aimed to inform us which buttons were easier versus harder to activate and about the facility of gaming with peers. A full list of the interview questions can be seen in Appendix 2. These data were analyzed descriptively through the reporting of frequencies and themes.

Phase 2: Adaptive Controller Development and Evaluation

The two participants with movement impairments were assessed over three data collections. During the first session, participants completed the Phase 1 interview. They then completed a Button Click Assessment using the standard video game controller they typically utilize for game play. This test was custom designed in our lab using LabVIEW (LabVIEW, National Instruments, USA). It displayed an image of the controller on the screen. Then, each button on the controller would light up for a time the researcher could define and participants were requested to click each button as soon as they saw it illuminate. Each button was highlighted one time during a trial and the order of illumination remained consistent across trials. In total, three trials were conducted with the button illumination time progressively decreasing across trials. Trial 1 illuminated buttons for three seconds each, Trial 2 for two seconds each, and Trial 3 for one second each. Next, a general assessment was performed by a trained physical therapist to classify the participant's level of upper extremity and gross motor

function. Hand function was classified using the Manual Ability Classification System (MACS), which describes hand function in daily activities of children aged 4-18 with cerebral palsy (CP). Children are classified into one of five levels, based on their self-initiated ability to handle objects and need for assistance or adaptation to perform manual activities in everyday life. Level I: handles objects easily and successfully; Level II: handles most objects but with somewhat reduced quality and/or speed of achievement; Level III: handles objects with difficulty; needs help to prepare and/or modify activities; Level IV: handles a limited selection of easily managed objects in adapted situations; Level V: does not handle objects and has severely limited ability to perform even simple actions [33]. Gross motor function was classified using the *Gross Motor Function Classification System* (GMFCS). It is a 5-level classification system that describes the gross motor function of children and youth with cerebral palsy based on their self-initiated movement with emphasis on sitting, walking, and wheeled mobility. Distinctions between levels are based on functional abilities, the need for assistive technology. Level I: walks without limitations; Level II: walks with limitations; Level III: walks using a mobility device; Level IV: self-mobility with limitations; may use powered mobility; Level V: transported in a manual wheelchair [34].

Following this, the participant played 30 minutes of his/her favorite video game. Three video cameras were set up to capture play activity, one directed at the television to assess game activity, one directed at the controller to observe hand/ controller interactions, and the last one directed at the participant's face to assess affect. Video cameras were time synced using a defined visual and auditory

action in the video game, and then overlaid using video editing software (Wondershare Filmora, Wondershare, China). We reviewed the recording of play and the interview results to identify key barriers to play with standard controllers for the two participants with movement impairments.

Based on the user-identified barriers, we designed and fabricated a custom video game controller adapter unique to each child between the first and second session. During the second session, the participants were first taught how to use the new controller adapter. Participants then completed the Button Click Assessment using both controllers. Following this, participants engaged in video game play through four, ten-minute bouts of the same game alternating every ten minutes between the standard and adaptive controllers. These ten-minute bouts were then split up into five, two-minute sections, allowing for ten, 2-minute sections of play with each controller. Video recording was performed as during the first session. Outcome measures for the Button Click and video game play assessments included: 1) total number of button activations, calculated per minute and then averaged across each 10-minute period; 2) number of activations for each individual button, calculated per minute and then averaged across each 10-minute period; 3) average time a button was pushed (only calculated for the left and right thumbsticks as these controlled longer lasting behaviors in the games selected for play by each participant), calculated as an average for each button across each 10-minute period; and 4) button click variability, calculated within each minute and then averaged across each 10-minute period, defined as the number of unique buttons activated by the user within that minute based on the

buttons involved in the game (5 buttons for the game for Participant 1 and his age-matched peer, 10 buttons for the game for Participant 2 and her age-matched peer).

These outcome measures could not directly be retrieved from the controller during participant gameplay because the controller could only communicate with one device, the XBOX One or LabVIEW program, at any given time. Instead, these data were gathered by having a researcher replaying video footage of all game play while replicating the game play in real time on an XBOX controller that was read into a separate LabVIEW program that recorded all button clicks. Participants kept their unique adaptive controllers for two weeks and then the protocol from the second session was repeated during a third session. Results from the second session aimed to inform us whether the adaptive controller had an immediate impact on performance. Results from the third session aimed to inform us whether video game performance with the adaptive controller improved after a 2-week practice period. Parents and participants were instructed to fill out an activity log stating how long they played video games and what controllers they used for play over the 2-week period. This log was collected at the third session. At the end of the third session, a device perception survey (see Appendix 3) was also provided to participants and their parents to determine what participants and their parents liked and disliked about the controller.

The two age-matched peers without movement impairments completed the assessments described above for the first session with the exception that only grip and pinch strength were assessed by the physical therapist for comparison, since

these children did not have sensorimotor impairments. For all children, data collection was performed when children were in a positive or neutral affective state. No child exhibited periods of negative affect during data collections.

Data from Phase 2 were analyzed using descriptive statistics as well as single subject data analysis using Simulation Modeling Analysis (SMA) [32] to report and compare results for all of the outcomes listed above. To evaluate for immediate effects, performance with the adaptive controller and with the standard controller were compared from the second session within children with movement impairments. To evaluate for practice effects, performance with the adaptive controller and with the standard controller were compared from the third session within children with movement impairments. In addition, performance with the adaptive controller at the second session was compared with the third session within children with movement impairments. To compare performance between children with and without movement impairments, performance with the standard controller for the age-matched child without movement impairments was compared with performance of the child with movement impairments using: 1) the standard controller at the second session; 2) the adaptive controller at the second session; and 3) the adaptive controller at the third session.

6.5 Results

Phase 1: Interview

The interview data showed that all the participants had a large motivation to play video games. The minimum number of hours of video game play per week was reported as 6 hours, with additional hours reported weekly in watching others

play video games; the maximum number of hours of weekly video game play reported was 28. All participants (100%) reported having a favorite console or device and had tried others as well. Five participants (55.6%) primarily played on consoles, and four (44.4%) participants played on mobile devices. Only three participants (33.3%) had tried an adaptive solution; two of those were commercial products rather than a home hack even though all nine participants reported trouble using their hands for play. Six participants (66.7%) reported that hitting the triggers was challenging and all nine participants (100%) played with the controller in their lap or on a flat surface instead of holding the controller. Participants were additionally asked to rate the ease of button use for the easiest button to click (scale from 1-5 with 5 being easiest) and the hardest button to click (scale from 1-5 with 5 being most difficult). The average response for the easiest button was a 4.83 and the hardest button was a 3.44. The buttons on the top of the controller or on one side of the controller for participants with hemiplegia or limb differences were overwhelmingly rated the easiest to access. The most difficult buttons to access included the triggers, buttons on the far side of the controller, and smaller buttons. Lastly, all of the participants (100%) reported playing with friends, but two participants (22.2%) reported they could only fully participate when they were playing at home and often opted to transition from playing to watching and coaching their friends play when game play became too fast-paced.

Phase 2: Design Outcome, Participant 1

Participant 1 had cerebral palsy with hand contractures that prevented him from holding a standard video game controller. His hand function level was a

MACS Level III, handles objects with difficulty and needs help to prepare and/or modify activities. His gross motor function level was GMFCS Level IV, self-mobility with limitations, use of powered mobility. Currently, the solution he had adopted was to tape the controller to a flat surface in front of him (Figure 6.2). This solution allowed him to hit all of the buttons on the top of the controller, but made it extremely challenging to hit the bumpers and triggers on the back of the controller.

Figure 6.2. Participant 1's gaming position with standard controller

The custom solution for this participant was a 3D printed box designed in CAD (Autodesk Inventor, Autodesk, USA) and printed on a 3D printer (Printrbot Plus V2, Printrbot, USA) that the controller was placed into. On the back of the box was a mechanism that translated the buttons on the back of the controller (left bumper (LB), right bumper (RB), left trigger (LT), and right trigger (RT)) onto buttons in the same plane as the rest of the buttons on the top of the controller (Figure 6.3).

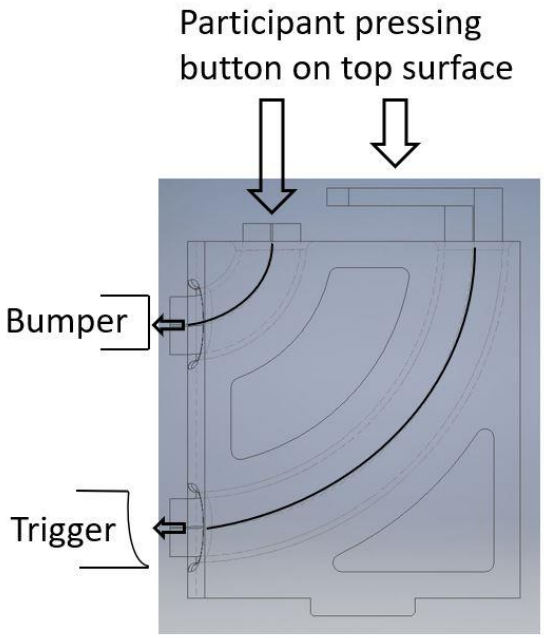

Figure 6.3. Cross section of mechanism that translates triggers and bumpers to buttons on the top of the controller

This adapter had suction cups integrated into the bottom of the box to easily and securely attach and detach the adapter from the flat surface. The adapter also had an interchangeable ergonomic cushion to support the participant's wrists. An additional early challenge reported by the participant was that the diameters of the thumbsticks were too small and made moving them difficult. Therefore, a flexible 3D-printed (Semiflex, NinjaTek, USA) cap was fabricated to increase the diameter of the thumbsticks as well. This adapter can be seen in Figures 6.3 and 6.4.

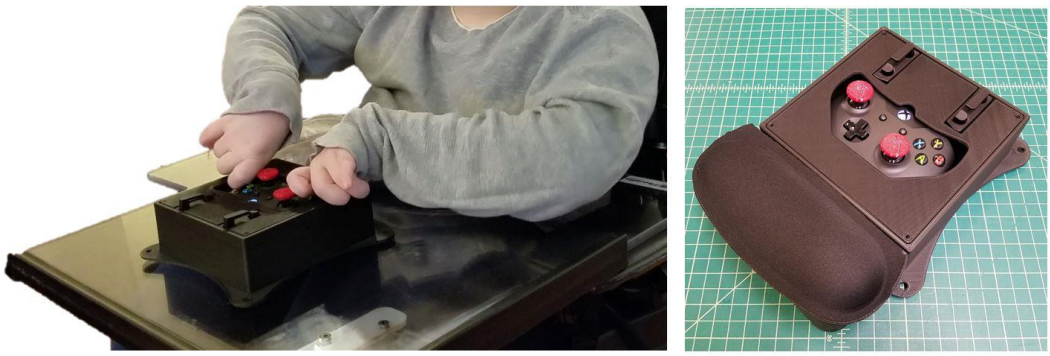

Figure 6.4. Left: Participant 1 gaming position with controller adapter. Right: Controller adapter. The controller sits in a 3D printed box with suction cups on the bottom to securely fix it to the clear surface. The buttons that translate inputs from the top plane to the bumpers and triggers can be seen on the top right. The two thumbstick caps can be seen in red. The interchangeable ergonomic cushion can be seen on the bottom left.

Phase 2: Design Outcome, Participant 2

Participant 2 also had cerebral palsy, as well as hemiplegia affecting her right side. Her hand function level was a MACS Level IV, handles a limited selection of easily managed objects in adapted situations. Her gross motor function level was GMFCS Level II, walks with limitations. She was unable to use her right hand at all for video game play. She generally played in a reclined position with her left hand on top of the controller (Figure 6.5). She pressed the controller into her stomach to anchor it, and needed to adjust her hand position frequently to hit all of the buttons.

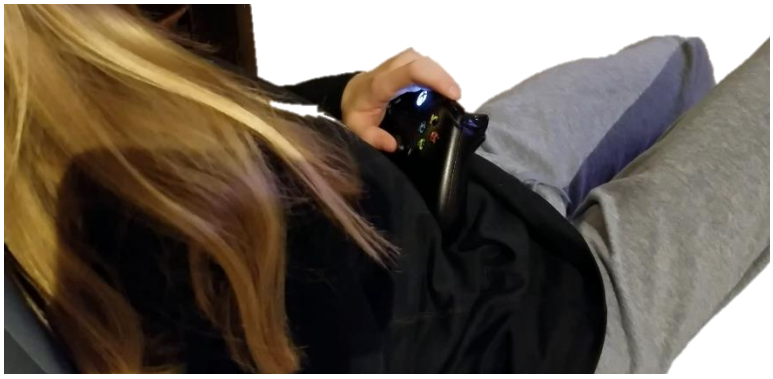

Figure 6.5. Participant 2 gaming position with standard controller

The custom solution for this participant was a 3D-printed adapter that interfaced with the two traditional handles where a player would typically hold the controller. This controller snugly fit into this rigid adapter which then had a flexible curved surface on the bottom to rest flush with her stomach for improved comfort and stability. This adapter provided a significantly larger surface area for the controller to sit on, as well as a flexible material, not rigid plastic resting against the participant's stomach. During the participant's preferred games, first person shooter games, with the standard controller, she used the palm of her hand to access the left thumbstick, her thumb to access the right thumbstick, and she shifted hand position frequently to access all of the buttons, specifically the triggers. She reported and we observed that she did not access the right trigger at all with the standard controller. To address her needs, the adaptive controller moved the left thumbstick to the back of the controller where her middle finger naturally rested for greater dexterity when moving. Trigger extenders were then 3D printed so there was a smaller distance between the buttons. This adapter can be seen in Figure 6.6.

Figure 6.6. Top left: Participant 2 thumbstick position selection. Top right: Participant 2 gaming position with adaptive controller. Bottom left: Adaptive controller showing the 3D printed adapter that goes from the controller handle to a wide smooth bottom to match the contour of stomach. Bottom right: Adaptive controller showing new thumbstick position on the back of the controller and trigger extenders in white.

Phase 2: Outcome Measures

Button Click Assessment

Both participants and their age matched peers completed the button click assessment during all sessions. Some limitations in the outcomes exist. After the first session, Participant 1 thought the fastest speed, each button illuminated for 1 second, was too fast and chose not to participate in this part of the assessment for

sessions two and three. The custom adaptive controller that was designed for Participant 2 had the left thumbstick moved to the back of the controller. The button click assessment is a wired assessment and the placement of the micro USB port on the controller directly interfered with the participant's ability to hit the new joystick. For this reason, we recorded data for this assessment for Participant 2 from video coding rather than via direct connection to the computer. This allowed us to analyze data for all trials except the 1-second trial. Results from this assessment can be seen in Table 6.2.

Relative to an age-matched peer, Participant 1 was able to activate fewer inputs across all trials of the Button Click Assessment and he required more time to activate each input. These results were observed with both the standard and adaptive controller; however, the participant was able to hit the left trigger (LT) input within the required time for both of the 3 second trials using the adaptive controller. This was the only time the participant was able to activate a trigger successfully within the time limit.

Participant 2 performed similarly in terms of the number of inputs activated with the standard and adaptive controller. Her performance with the standard controller was poorer during the 1-second trial. Across all trials, Participant 2 required more time to activate the inputs in this assessment relative to her age-matched peer. These higher click times were likely a result of the participant having to repeatedly adjust her hand position to hit the appropriate buttons while playing with only one hand.

Table 6.2. Button Click Assessment Results

			Illuminated 3 seconds		Illuminated 2 seconds		Illuminated 1 second	
Participant	Session	Controller Type	Total (and Percentage) of Inputs Clicked	Time to Click (sec ± *SE*)	Total (and Percentage) of Inputs Clicked	Time to Click (sec ± *SE*)	Total (and Percentage) of Inputs Clicked	Time to Click (sec ± *SE*)
1	Session 1	Standard	14 (63.6%)	2.24 ± 0.13	10 (45.5%)	1.49 ± 0.09	2 (9.1%)	0.40 ± 0.40
	Session 2	Standard	17 (77.3%)	1.81 ± 0.12	14 (63.6%)	1.35 ± 0.10	N/A	N/A
	Session 2	Adaptive	13 (59.1%)	2.05 ± 0.13	7 (31.8%)	1.41 ± 0.12	N/A	N/A
	Session 3	Adaptive	18 (81.8%)	1.75 ± 0.13	12 (54.5%)	1.42 ± 0.10	N/A	N/A
	Session 3	Standard	16 (72.7%)	1.58 ± 0.12	14 (63.6%)	1.50 ± 0.08	N/A	N/A
1 age matched peer		Standard	19 (86.4%)	1.40 ± 0.10	21 (95.5%)	1.07 ± 0.05	14 (63.6%)	0.76 ± 0.04
2	Session 1	Standard	12 (54.5%)	1.20 ± 0.09	20 (90.9%)	1.04 ± 0.06	13 (59.1%)	0.79 ± 0.02
	Session 2	Adaptive	19 (86.4%)	N/A	16 (72.7%)	N/A	N/A	N/A
	Session 2	Standard	21 (95.5%)	1.28 ± 0.12	22 (100%)	1.11 ± 0.08	13 (59.1%)	0.70 ± 0.03
	Session 3	Standard	22 (100%)	1.10 ± 0.06	19 (86.4%)	0.99 ± 0.09	14 (63.6%)	0.73 ± 0.04
	Session 3	Adaptive	22 (100%)	N/A	20 (90.9%)	N/A	N/A	N/A
2 age matched peer		Standard	20 (90.9%)	0.88 ± 0.05	21 (95.5%)	0.76 ± 0.04	19 (86.4%)	0.65 ± 0.03

Gameplay Results

The number of individual button clicks per minute, average time the left and right thumbsticks were depressed, total clicks per minute and button variability per minute were recorded and can be seen in Table 6.3. SMA was used to compare each of these four variables between the standard controller and adaptive controller within session 2 and within session 3, as well as between the adaptive controller from session 2 to session 3. SMA for individual button clicks was only used to compare buttons that had been modified.

Participant 1's favorite game was Lego The Incredibles. This game was in part his favorite because it didn't require him to use the triggers or bumpers that were modified in the adaptive controller. As a result, there was no significant difference in his performance with the two controllers for any of the variables. After reminding the participant that he should now be able to use the triggers as a control instead of the A and B buttons, he demonstrated the ability to use the triggers. Perhaps with repeated reminders and practice, improvements in use of these inputs might be noted. A noteworthy improvement not reflected in the data below was that the participant also played a racing game, Forza Horizon 4, which requires a steering wheel and pedal controller. He was able to control the steering wheel and his parents would need to hit the pedals for him. While using the adaptive controller, he was able to hit the triggers independently and furthermore was able to play the game without any parental assistance.

Participant 2's favorite game was Fortnite. The individual buttons LT, RT, and left thumbsticks were compared using SMA because modifications had

been made to those buttons. The left thumbstick was at first activated significantly less with the adaptive controller than the standard controller during session 2) (*R = -0.491, p = .034)* but was then activated more during session 3 *R = +0.686, p = .009).* In addition, there was a significant increase in number of activations of the left thumbstick on the adaptive controller from sessions 2 to 3 *R = +0.615, p = .028).* The left thumbstick at first was activated on average for significantly longer with the adaptive controller compared to the standard controller in session 2 (*R = +0.449, p = .017*) but then activated for less time in session 3 (*R = -0.592, p .010)*; its duration of activation with the adaptive controller also decreased from session 2 to session 3 (*R = -0.434, p = .041).* In this case, a shorter amount of time the left thumbstick was advantageous because it meant the participant gained control to enact precise movements of her character rather than long, slow movements. There were no significant differences in performance noted with the other inputs or variables but it should be noted that during session 3, the participant did use RT for the first time during any gameplay.

Table 6.3. Gameplay button click data. Cells in row 2 are individual buttons on the controller. Left bumper (LB), right bumper (RB), left trigger (LT), right trigger (RT), left thumbstick (Lts), and right thumbstick (Rts). Controller periods A, A1, and A2 were performed with the standard controller; periods B1 and B2 with the adaptive controller. Cells shaded in light grey relate to the inputs adapted for each participant.

			Number of Button Clicks Per Minute, Average										Duration Pressed (seconds), Average		Per Minute	
Participant	Session	Controller	A	B	X	Y	LB	RB	LT	RT	Lts	Rts	Lts Pressed	Rts Pressed	Total Clicks	Variability
1	2	A1	3.1	2.2	3.7	0.5	0.0	0.0	0.0	0.0	10.5	0.8	3.7	0.6	20.9	3.4
		B1	3.8	1.7	5.0	2.7	0.0	0.1	0.0	0.0	14.9	0.5	1.9	0.5	28.6	3.2
		A2	4.3	2.9	9.4	1.0	0.2	0.3	0.1	0.0	11.3	2.5	3.2	1.4	31.9	3.4
		B2	4.5	1.0	1.5	0.6	0.0	0.0	0.0	0.0	19.1	2.0	1.0	1.2	28.8	2.9
	3	B1	5.6	2.4	1.3	1.6	0.1	0.1	0.7	1.0	13.5	1.7	2.7	1.0	27.9	3.1
		A1	5.7	3.2	0.2	0.9	0.0	0.0	0.0	0.0	15.7	3.1	1.4	1.5	28.7	3.3
		B2	4.5	2.4	0.1	0.3	0.0	0.2	0.0	0.0	12.8	3.9	2.4	1.4	24.5	3.0
		A2	6.0	3.5	0.2	1.9	0.2	0.2	0.0	0.0	12.5	2.8	1.9	1.5	27.2	3.6
1 Age Match		A	5.7	4.8	8.1	1.6	0.0	0.0	0.4	0.8	11.6	0.1	3.1	0.8	33.0	4.0
2	2	B1	1.4	0.3	1.7	0.2	0.1	0.9	1.5	0.2	7.1	9.8	6.0	0.4	23.2	4.1
		A1	3.2	0.4	2.7	0	0	0.7	1.6	0.1	11.6	11.3	3.5	0.5	31.7	4.2
		B2	3	0.8	0.9	0.1	0.7	0.8	1.9	0.2	6.9	8.5	6.9	0.8	24.0	4.4
		A2	2.5	0.2	3.6	0.5	0.6	2.7	2.1	0.3	13.5	15.7	3.1	0.4	41.7	5.3
	3	A1	1.8	0.2	1.8	0	0	0.6	0.8	0	7.1	12.5	6.8	0.6	24.7	3.9
		B1	0.8	0.4	1.7	0	0.2	1.2	1.7	0.2	11.9	9.1	3.6	0.8	27.3	4.1
		A2	1.9	0.2	1.5	0.2	0	2.2	1.1	0	7	17.6	6.7	0.7	31.8	4.0
		B2	1.4	0.3	3	0.3	0	2.7	1.1	0.3	13.1	13.1	3.2	0.5	35.2	4.9
2 Age Match		A	6.3	0.6	4.9	0.8	1.6	3.7	2.1	2.1	10.5	27.6	5.1	0.8	60.3	7.0

Activity log

Between the first and second sessions, Participant 1 reported using the adaptive controller for play 6 hours and 35 minutes over 19 days. Participant 2 reported using it for 12 hours and 30 minutes over 14 days. Both participants reported only selecting the adaptive controllers for play between visits rather than the standard controller.

Device Perception Survey

At the end of the third session, participants were asked several questions to understand their impressions about the adaptive controller. Both participants reported the best outcome (5) out of a 1-5 scale for how much they liked the controller, how easy it was to press the previously most difficult button to press now, and the likelihood they would continue to use the controller after the study. Results from this survey can be seen in Table 6.4.

Table 6.4. Device Perception Survey Results

1. Question	Participant 1	Participant 2
How much do you like the video game controller adapter? Don't like (1) through Like it a lot (5)	5	5
How easy was it to learn to use the video game controller adapter? Very hard (1) through Very easy (5)	3	4
How hard was the hardest button to press before?	1	3
How hard is that same button to press now? Very hard (1) through Very easy (5)	5	5
How cool do you find the controller? Not cool (1) through Very cool (5)	4	5
Will you continue to use this controller outside of the study? Will not use (1) through Will use often (5)	5	5

6.6 Discussion

Based on previous research, it has been shown that people with movement impairments have a strong desire to play video games for pleasure, but there are barriers that limit their ability to play with ease. All of the participants with upper extremity impairments that were interviewed reported having trouble using their hands to play video games. On a scale from 1 to 5, it was found that the hardest button to hit on the controller was 1.39 points harder to hit than the easiest button, and common difficult buttons to hit included the triggers, buttons away from the primary hand playing for gamers with unilateral impairments, and smaller buttons. Adaptive video game controllers should aim to create ways users with a variety of abilities can more successfully access these buttons.

Starting with a user-centered approach, it was shown that two different video game controller adapters helped to improve ease of play for two participants. One controller aimed to make hitting the bumpers and triggers on the back of the controller easier, and the other moved the left thumbstick to position that would allow for ease of control with one hand. Both participants did not return to using the standard controller during the two weeks they kept the adaptive controllers at home for play. One video game controller adapter allowed a participant to play a game independently when he previously needed parental assistance to play. The other adapter allowed for more natural, controlled movement within a game, due to significantly increased frequency of button clicks with shorter durations of pressing, allowing for more precise control of the

game character, potentially leading to more competitive gameplay. These results suggest that Participant 1 may be expected to show increased independence in games that utilize triggers he was previously unable to access. Participant 2's self-selected game for play utilized the inputs that were challenging for her. We initially observed poorer performance during gameplay when Participant 2 first used the adaptive controller, but later observed significant improvements in her performance with the adaptive controller after a 2-week period of practice. She was likely learning how to use the left thumbstick in its altered location during session 2 and had mastered how to access it during gameplay by session 3.

This research highlights common challenges that exist using current video game controllers and shows that adaptive controllers can impact level of independence and success in game play while remaining affordable and accessible. There are do-it-yourself instructions for the adaptive controllers described in this article available online (http://sites.udel.edu/move2learn/how-todiy/). While these controller adapters were made to fit the specific needs of the two participants, the solutions presented in this paper can cater to a wider range of needs such as a common challenge of hitting the triggers or difficulty during one handed play. Further studies should aim to interview more participants, to test the effects of a variety of adaptive controllers more longitudinally for a larger number of participants, and to evaluate the ability of users within the broader community to fabricate and utilize these designs.

Chapter 7

CONCLUSION

7.1 Conclusion

The goal of this research was to develop and test novel wearable technologies with the potential to improve monitoring and performance of physical activity for children with physical disabilities.

Aim 1: To design and validate a smart compression garment for measurement of upper extremity physical activity.

The optimal conductive thread stitched stretch sensor to measure movement was first determined based on conductive thread type, stitch geometry, sensor length, and textile substrate. This sensor was accurate but became less reliable over time. To solve this, a novel conductive paint stretch sensor was then created that changed in resistance when stretched. This sensor was placed across the shoulder and elbow joints and tested on both a child and an adult through a series of static, dynamic, and functional assessments. When validated against a commercial motion capture system, this smart compression garment could record activity counts with 75% accuracy and had an average percent error of 15.55 degrees. This smart garment shows it has the potential to record physical activity within a natural setting.

Aim 2: To design and validate a device to track the amount of time infants are in distinct positions.

After multiple iterations, a small wearable device was designed to track infant positioning. The final prototype included infants wearing a soft onesie with a vinyl pocket on the side that housed a small microcontroller capable of measuring orientation. The final prototype was tested on 11 infants who were not yet rolling. It had an accuracy of 89.36% for categorizing body position compared with visual coding. When the device was left with families, it was found that infants were only placed in prone for 3.89% of the time providing the first quantitative baseline for this important early positioning activity in natural contexts. A tool like this that measures and could provide feedback on positioning activity with infants could be useful for early intervention assessment and intervention. Providing feedback to parents on this activity may encourage them to place their children more often in in positions, such as prone and upright, that have been shown to advance future motor and cognitive outcomes.

Aim 3: To design and test a user driven video game controller adapter that may facilitate inclusive play.

First, nine children with upper extremity physical disabilities were interviewed to find common challenges faced when trying to play video games. It was found that all of the children had trouble holding or using traditional video game controllers, and that the easiest button to press (scale from 1-5 with 5 being the easiest) was a 4.83 and the hardest button (scale from 1-5 with 5 being the most difficult) was a 3.44. For two of these children, a custom user driven video game controller adapter was designed to make it easier to press the hardest buttons. After designing an open sourced custom controller adapter, one participant was able to play a new game with complete independence where parental assistance was previously needed and the other participant was able to move the left thumbstick with more precision leading to more competitive

gameplay. Both participants never returned to playing with the standard controller once the controller adapter was given to them. The results from this study will hopefully influence the design of future video game controllers.

Each aim of this dissertation had the goal of designing novel, open source wearable technology that can be used in rehabilitation for individuals with physical disabilities. Each aim strived to do this in different methods which were to monitor activity, change activity by providing feedback, and to control external devices leading to an increased understanding of movement and improved quality of life.

REFERENCES

Chapter 1

1. Brault, M. (2012). Americans with disabilities: 2010. Retrieved from https://www.census.gov/content/dam/Census/library/publications/2012/demo/p70-131.pdf

2. Zhou, H., Huosheng, H., & Tao, Y. (2006). Inertial measurements of upper limb motion. *Medical & Biological Engineering & Computing, 44*, 479-487.

3. Papi, E., Osei-Kuffour, D., Chen, Y-M. A., & McGregor, A. H. (2015). Use of wearable technology for performance assessment: A validation study. *Medical Engineering and Physics, 37*(7), 698-704.

4. Gioberto, G., & Dunne, L. (2012). Theory and characterization of a top-thread coverstitched stretch sensor. *IEEE Conference on Systems, Man, and Cybernetics*, 3275-3280.

5. Lobo, M. A., & Galloway, J. C. (2008). Postural and object-oriented experiences advance early reaching, object exploration, and means-end behavior. *Child Development, 79*, 6.

6. Sgandurra, G., Bartalena, L., Cecchi, F., Cioni, G., Giampietri, M., Greisen, G., ... Dario, P. (2016). A pilot study on early home-based intervention through an intelligent baby gym (CareToy) in preterm infants. *Research in Developmental Disabilities*, 32-42.

7. Sgandurra, G., Lorentzen, J., Inguaggiato, E., Bartalena, L., Beani, E., Cecchi, F., ... CareToy Consortium. (2017). A randomized clinical trial in preterm infants on the effects of a home-based early intervention with the 'CareToy System'. *Plos One*, *12*, 3.

8. Ustad, T., Evensen, K. A., Campbell, S. K., Girolami, G. L., Helbostad, J., Jørgensen, L., ... Øberg, G. K. (2016). Early Parent-Administered Physical Therapy for Preterm Infants: A Randomized Controlled Trial. *Pediatrics, 138*, 2.

9. Rowland, J. L., Malone, L. A., Fidopiastis, C. M., Padalabalanarayanan, S., Thirumalai, M., & Rimmer, J. H. (2016). Perspectives on Active Video Gaming as a New Frontier in Accessible Physical Activity for Youth With Physical Disabilities. *Physical Therapy, 96*(4), 521-32.

10. Gamespot Staff (2000). Sony announces PS2 launch date and price. Retrieved from https://www.gamespot.com/articles/sony-announces-ps2-launch-date-and-price/1100-2568701/

11. LukieGames (2017). Original Sony Playstation 2 Dualshock Controller!. Retrieved from https://www.lukiegames.com/Used-Playstation-2-PS2-Dualshock-Controller.html

12. BestBuy (2017). Sony – DualShock 4 Wireless Controller for PlayStation 4 – Jet black. Retrieved from http://www.bestbuy.com/site/sony-dualshock-4-wireless-controller-for-playstation-4-jet-black/5580915.p?skuId=5580915

13. CATEA (2011). Instructables-Wiimote Wireless Modification for Persons with Disabilities. Retrieved from http://www.instructables.com/id/Wiimote-Wireless-Modification-for-Persons-with-Dis/

Chapter 2

1. Hall ML, Lobo M. Design and development of the first exoskeletal garment to enhance arm mobility for children with movement impairments. *Disabil Rehabil Assist Technol.* 2017.

2. de Ana FJ, Umstead KA, Phillips GJ, Conner CP. Value driven innovation in medical device design: A process for balancing stakeholder voices. *Ann Biomed Eng.* 2013;41(9):1811-1821.

3. DeMarco CT. Medical Device Design and Regulation. ASQ Quality Press; 2011.

4. Gonzalez-Villanueva L, Cagnoni S, Ascari L. Design of wearable sensing system for human motion monitoring in physical rehabilitation. *Sensors.* 2013;13(6):7735-7755.

5. Parette P, Scherer M. Assistive Technology Use and Stigma. *Educ Train Dev Disabil.* 2004;39(3):217.

6. Yan T, Cempini M, Oddo CM, Vitiello N. Review of assistive strategies in powered lower-limb orthoses and exoskeletons. *Rob Auton Syst.* 2015;64:120-136.

7. Kling A, Campbell PH, Wilcox J. Young children with physical disabilities: Caregiver perspectives about assistive technology. (Clinical report). *Infants & Young Children*. 2010;23(3).

8. Cardon TA, Wilcox MJ, Campbell PH. Caregiver perspectives about assistive technology use with their young children with autism spectrum disorders. (Report). *Infants & Young Children*. 2011;24(2).

9. Arntzen C, Holthe T, Jentoft R. Tracing the successful incorporation of assistive technology into everyday life for younger people with dementia and family carers. (Report). *Dementia*. 2016;15(4):646-662.

10. Lamb JM, Kallal MJ. A Conceptual Framework for Apparel Design. *Clothing and Textiles Research Journal*. 1992;10(2):42-47.

11. Cynthia LR, Doris HK, Gwen S. Applicability of the Engineering Design Process Theory in the Apparel Design Process. *Clothing and Textiles Research Journal.* 1998;16(1):36-46.

12. Rosenblad-Wallin E. User-oriented product development applied to functional clothing design. *Appl Ergon.* 1985;16(4):279-287.

13. Watkins SM, Dunne LE. Functional clothing design: From sportswear to spacesuits. 2015.

14. Kaiser SB. The social psychology of clothing: Symbolic appearances in context. New York: Fairchild Publications; 1997.

15. Hall ML, Orzada BT. Expressive Prostheses: Meaning and Significance. *Fashion Practice*. 2013;5(1):9-32.

16. King IW. The aesthetics of dress. 2017: https://link.springer.com/book/10.1007%2F978-3-319-54322-2.

17. Carroll KE, Kincade DH. Inclusive Design in Apparel Product Development for Working Women With Physical Disabilities. *Family and Consumer Sciences Research Journal*. 2007;35(4):289-315.

18. Mincer AB. Assistive devices for the adult patient with orthopaedic dysfunction. Why physical therapists choose what they do. *Orthop Nurs*. 2007;26(4).

19. Scherer M, Jutai J, Fuhrer M, Demers L, Deruyter F. A framework for modelling the selection of assistive technology devices (ATDs). *Disabil Rehabil Assist Technol.* 2007;2(1):1-8.

20. Schraner I, de Jonge D, Layton N, Bringolf J, Molenda A. Using the ICF in economic analyses of Assistive Technology systems: Methodological implications of a user standpoint. *Disabil Rehabil.* 2008;30(12-13):916-926.

21. Keates S, Clarkson J. Countering design exclusion: An introduction to inclusive design. London; New York: Springer; 2003.

22. Awori J, Lee JM. A maker movement for health: A new paradigm for health innovation. *JAMA Pediatr*. 2017;171(2).

23. Lobo MA, Koshy J, Hall ML, et al. Playskin Lift: Development and Initial Testing of an Exoskeletal Garment to Assist Upper Extremity Mobility and Function. *Phys Ther.* 2016;96(3):390-399.

24. Trish W, Jenny S. Compliance with Prescribed Adaptive Equipment: A Literature Review. *Br J Occup Ther*. 2000;63(2):65-75.

25. Tsai K-H, Yeh C-Y, Lo H-C. A novel design and clinical evaluation of a wheelchair for stroke patients. *Int J Ind Ergon*. 2008;38(3):264-271.

26. Huang IC, Sugden D, Beveridge S. Assistive devices and cerebral palsy: Factors influencing the use of assistive devices at home by children with cerebral palsy. *Child Care Health Dev.* 2009;35(1):130-139.

27. Gwilt A. Fashion design for living. 2015.

28. Lamb JM. Disability and the Social Importance of Appearance. *Clothing and Textiles Research Journal*. 2001;19(3):134-143.

29. Freeman CM, Kaiser SB, Wingate SB. Perceptions of Functional Clothing By Persons with Physical Disabilities: A Social-Cognitive Framework. Clothing and Textiles *Research Journal.* 1985;4(1):46-52.

30. Wingate SB, Kaiser SB, Freeman CM. Salience of Disability Cues in Functional Clothing: A Multidimensional Approach. *Clothing and Textiles Research Journal* 1986;4(2):37-47.

31. Carroll K, Gross K. An Examination of Clothing Issues and Physical Limitations in the Product Development Process. *Family and Consumer Sciences Research Journal*. 2010;39(1):2-17.

32. Stokes B, Black C. Application of the Functional, Expressive and Aesthetic Consumer Needs Model: Assessing the clothing needs of adolescent girls with disabilities. *International Journal of Fashion Design, Technology and Education*. 2012;5(3):179-186.

33. Rusk HA, Taylor EJ. Functional fashions for the physically handicapped. *JAMA* 1959;169(14):1598-1600.

34. Shannon E, Reich N. Clothing and related needs of physically handicapped persons. *Rehabil Lit.* 1979;40(1):2-6.

35. Lamb JM. Family use of functional clothing for children with physical disabilities. *Rehabil Lit.* 1984;45(5-6):146-150.

36. Chang W-M, Zhao Y-X, Gup R-P, Wang Q, Gu X-D. Design and study of clothing structure for people with limb disabilities. *J Eng Fiber Fabr*. 2009;2(2):61-66.

37. Kratofil C. Tommy Hilfiger Launches First Collection of Adaptive Clothing for the Differently-Abled Community. 2016. Accessed April 19, 2018.

38. Pittman T. Target is Releasing Adaptive Apparel for Kids with Disabilities 2017. Accessed April 19, 2018.

39. Kaiser SB, Freeman CM, Wingate SB. Stigmata and negotiated outcomes: Management of appearance by persons with physical disabilities. *Deviant Behav*. 1985;6(2):205-224.

40. Blaya JA, Herr H. Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait. *IEEE Trans Neural Syst Rehabil Eng*. 2004;12(1):24-31.

41. Ferreira LA, Neto HP, Grecco LA, et al. Effect of Ankle-foot Orthosis on Gait Velocity and Cadence of Stroke Patients: A Systematic Review. *J Phys Ther Sci.* 2013;25(11):1503-1508.

42. Swinnen E, Kerckhofs E. Compliance of patients wearing an orthotic device or orthopedic shoes: A systematic review. *J Bodyw Mov Ther.* 2015;19(4):759-770.

43. Vinci P, Gargiulo P. Poor compliance with ankle-foot-orthoses in Charcot-Marie-Tooth disease. *Eur J Phys Rehabil Med.* 2008;44(1):27-31.

44. Matthews MJ, Watson M, Richardson B. Effects of dynamic elastomeric fabric orthoses on children with cerebral palsy. *Prosthet Orthot Int.* 2009;33(4):339-347.

45. Betts L. Dynamic movement lycra orthosis in multiple sclerosis. *Br J Neurosci Nurs.* 2015;11(2):60-64.

46. Maguire C, Sieben JM, Frank M, Romkes J. Hip abductor control in walking following stroke - the immediate effect of canes, taping and TheraTogs on gait. *Clin Rehabil*. 2010;24(1):37-45.

47. Serrao M, Casali C, Ranavolo A, et al. Use of dynamic movement orthoses to improve gait stability and trunk control in ataxic patients. *Eur J Phys Rehabil Med*. 2017;53(5):735-743.

48. Rahman T, Sample W, Seliktar R, et al. Design and Testing of a Functional Arm Orthosis in Patients With Neuromuscular Diseases. *IEEE Trans Neural Syst Rehabil Eng IEEE Transactions on Neural Systems and Rehabilitation Engineering*. 2007;15(2):244-251.

49. Young AJ, Ferris DP. State of the Art and Future Directions for Lower Limb Robotic Exoskeletons. *IEEE Trans Neural Syst Rehabil Eng*. 2017;25(2):171-182.

50. Garcia E, Sater JM, Main J. Exoskeletons for Human Performance Augmentation (EHPA): A Program Summary. *Adv Robot*. 2002;20(8):822-826.

51. Asbeck AT, Schmidt K, Galiana I, et al. Multi-joint soft exosuit for gait assistance. *Proc IEEE Int Conf Rob Autom Proceedings - IEEE International Conference on Robotics and Automation.* 2015;2015-June:6197-6204.

52. Babik I, Kokkoni E, Cunha AB, Galloway JC, Rahman T, Lobo MA. Feasibility and Effectiveness of a Novel Exoskeleton for an Infant With Arm Movement Impairments. *Pediatr Phys Ther*. 2016;28(3):338-346.

53. Cho G, Lee S, Cho J. Review and Reappraisal of Smart Clothing. *Int J Hum Comput Interact*. 2009;25(6):582-617.

54. Patel S, Park H, Bonato P, Chan L, Rodgers M. A review of wearable sensors and systems with application in rehabilitation. *J Neuroeng Rehabil*. 2012;9.

55. John D, Freedson P. ActiGraph and Actical Physical Activity Monitors: A Peek under the Hood. *Med Sci Sports Exerc.* 2012;44(1S Suppl 1):S86-S89.

56. Bassett DR, Jr., Toth LP, LaMunion SR, Crouter SE. Step Counting: A Review of Measurement Considerations and Health-Related Applications. *Sports Med.* 2017;47(7):1303-1315.

57. Van Cauwenberghe E, Gubbels J, De Bourdeaudhuij I, Cardon G. Feasibility and validity of accelerometer measurements to assess physical activity in toddlers. *Int J Behav Nutr Phys Act*. 2011;8(1):67.

58. Van Cauwenberghe E, Gubbels J, De Bourdeaudhuij I, Cardon G. Calibration and comparison of accelerometer cut points in preschool children. *Pediatr Obes.* 2011;6(2):e582-e589.

59. Angulo-Barroso R, Burghardt AR, Lloyd M, Ulrich DA. Physical activity in infants with Down syndrome receiving a treadmill intervention. *Infant Behav Dev.* 2008;31(2):255-269.

60. Ketcheson L, Pitchford EA, Kwon H-J, Ulrich DA. Physical Activity Patterns in Infants With and Without Down Syndrome. *Pediatr Phys Ther*. 2017;29(3):200-206.

61. Pitchford EA, Ketcheson LR, Kwon HJ, Ulrich DA. Minimum Accelerometer Wear Time in Infants: A Generalizability Study. *J Phys Act Health*. 2017;14(6):421-428.

62. Bjornson KF, Belza B, Kartin D, Logsdon R, McLaughlin JF. Ambulatory physical activity performance in youth with cerebral palsy and youth who are developing typically. *Phys Ther.* 2007;87(3):248-257.

63. Kressler J, Koeplin-Day J, Muendle B, Rosby B, Santo E, Domingo A. Accuracy and precision of consumer-level activity monitors for stroke detection during wheelchair propulsion and arm ergometry. *PLoS One*. 2018;13(2):e0191556-e0191515.

64. Accessibility - Apple Watch. 2018; http://www.apple.com/accessibility/watch/. Accessed August 23, 2018.

65. Malu M, Findlater L, EAI International Conference on Pervasive Computing Technologies for Healthcare P. Toward accessible health and fitness tracking for people with mobility impairments. Pervasive Health: Pervasive Computing Technologyl Healthcare Pervasive Health: Pervasive Computing Technologies for Healthcare. 2016;2016-May:2016.

66. Carrington P, Chang K, Mentis H, Hurst A. "But, I don't take steps:" Examining the inaccessibility of fitness trackers for wheelchair athletes. Paper presented at: International ACM SIGACESS Conference on Computers & Accessibility. 2015.

67. Preece SJ, Goulermas JY, Kenney LP, Howard D, Meijer K, Crompton R. Activity identification using body-mounted sensors--a review of classification techniques. *Physiol Meas.* 2009;30(4):1-33.

68. Trujillo-Priego I, Lane C, Vanderbilt D, et al. Development of a Wearable Sensor Algorithm to Detect the Quantity and Kinematic Characteristics of Infant Arm Movement Bouts Produced across a Full Day in the Natural Environment. *Technologies*. 2017;5(3):39.

69. Bailey RR, Klaesner JW, Lang CE. An accelerometry-based methodology for assessment of real-world bilateral upper extremity activity. *PloS one*. 2014;9(7).

70. Smith B, Trujillo-Priego I, Lane C, Finley J, Horak F. Daily Quantity of Infant Leg Movement: Wearable Sensor Algorithm and Relationship to Walking Onset. *Sensors.* 2015;15(8):19006-19020.

71. Trujillo-Priego I, Smith B. Kinematic characteristics of infant leg movements produced across a full day. *J Rehabil Assist Technol Eng.* 2017;4.

72. Abrishami MS, Nocera L, Trujillo-Priego IA, Purushotham S, Shahabi C, Smith BA. Identification of developmental delay in infants using wearable sensors: Whole-day leg movement feature analysis. *Technologies in prep.* ;5(3):1-16.

73. Smith B, Vanderbilt D, Applequist B, Kyvelidou A. Sample Entropy Identifies Differences in Spontaneous Leg Movement Behavior between Infants with Typical Development and Infants at Risk of Developmental Delay. *Technologies*. 2017;5(4):55.

74. Torres EB, Smith B, Mistry S, Brincker M, Whyatt C. Neonatal Diagnostics: Toward Dynamic Growth Charts of Neuromotor Control. *Front Pediatr*. 2016;4.

75. Lu C, Black MM, Richter LM. Risk of poor development in young children in low-income and middle-income countries: an estimation and analysis at the global, regional, and country level. *Lancet Glob Health.* 2016;4(12):e916-e922.

76. Henrich J, Heine SJ, Norenzayan A. The weirdest people in the world? *J Behav Brain Sci*. 2010;33(2-3):2-3.

77. Gioberto G, Dunne L. IEEE International Conference on Systems M, Cybernetics SMC. Theory and characterization of a top-thread coverstitched stretch sensor. *Conf Proc IEEE Int Conf Syst Man Cybern*. 2012:3275-3280.

78. Greenspan B, Hall ML, Lobo MA, Cao H. Development and testing of a stitched stretch sensor with the potential to measure human movement. *J Text Inst*. 2018:1-8.

79. Atalay A, Sanchez V, Atalay O, et al. Batch Fabrication of Customizable Silicone-Textile Composite Capacitive Strain Sensors for Human Motion Tracking. *Adv Mater Technol.* 2017:1700136.

80. Sensoria Fitness. http://www.sensoriafitness.com/technology. Accessed April 30, 2018.

81. Lumo. 2016; https://www.lumobodytech.com/. Accessed April 30, 2018.

82. Moticon Sensor Insole. https://www.moticon.de/. Accessed April 30, 2018.

83. Cadmus-Bertram LA, Marcus BH, Patterson RE, Parker BA, Morey BL. Randomized Trial of a Fitbit-Based Physical Activity Intervention for Women. *Am J Prev Med*. 2015;49(3):414-418.

84. Suorsa K, Tackett A, Mullins A, et al. The Use of a Fitbit Intervention to Increase Physical Activity among Children and Adolescent Paediatric Cancer Survivors: An N-Of-1 Randomized Controlled Trial. *Pediatr Blood Cancer*. 2016;63:S58-S59.

85. ReWalk™ Personal Exoskeleton System Cleared by FDA for Home Use. 2014; http://rewalk.com/rewalk-robotics-announces-expansion-to-india-with-saimed-innovations-2-2-2-2-3-2-3/. Accessed April 19, 2018.

86. Ekso Bionics. 2018; https://eksobionics.com/. Accessed April 30, 2018.

87. Wu J, Sun L, Jafari R. A Wearable System for Recognizing American Sign Language in Real-Time Using IMU and Surface EMG Sensors. *IEEE J Biomed Health Inform*. 2016;20(5):1281-1290.

88. Thelen E. Motor Development: A New Synthesis. *Pediatr Ann*. 1996:32.

89. Thelen E. Dynamic Systems Theory and the Complexity of Change. *Psychoanal Dialogues*. 2005;15(2):255-283.

Chapter 3

1. Atalay, O., Kennon, W.R., & Husain, D.M. (2013). Textile-based weft knitted sensors: Effect of fabric parameters on sensor properties. *Sensors, 13*(8), 11114-11127.

2. ASTM D6614-07 (2015). Standard Test Method for Stretch Properties of Textile Fabrics – CRE Method, ASTM International, West Conshohocken, PA, 2015, http://dx.doi.org/10.1520/D6614-07R15

3. ASTM D1776/D1776M-16 (2016). Standard Practice for Conditioning and Testing Textiles, ASTM International, West Conshohocken, PA, 2016, http://dx.doi.org/10.1520/D1776_D1776M-16

4. Chan, M., Estève, D., Fourniols, J-Y., Escriba, C., & Campo, E. (2012). Smart wearable systems: Current status and future challenges. *Artificial Intelligence in Medicine, 56*, 137-156.

5. Dunne, L. E., Ashdown, S. P., & Smyth, B. (2005). Expanding garment functionality through embedded electronic technology. *Journal of Textile and Apparel, Technology and Management, 4*(3).

6. Flexible stretch sensors. (2016). Retrieved from http://www.imagesco.com/sensors/stretch-sensor.html

7. Fong, D.T.P., & Chan, Y.Y. (2010). The use of wearable inertial motion sensors in human lower limb biomechanics studies: A systematic review. *Sensors, 10*(12), 11556-11565.

8. Gimpel, S., Mohring, U., Muller, H., Neudeck, A., & Scheibner, W. (2004). Textile-based electronic substrate technology. *Journal of Industrial Textiles, 33*(3), 179-189.

9. Gioberto, G., & Dunne, L. (2013). Overlock-stitched stretch sensors: Characterization and effect of fabric property. *Journal of Textile and Apparel, Technology and Management, 8*(3), 1-14.

10. Gioberto, G., & Dunne, L. (2012). Theory and characterization of a top-thread coverstitched stretch sensor. *IEEE Conference on Systems, Man, and Cybernetics*, 3275-3280.

11. Kim, J., Gravunder, A., & Park, H.S. (2015). Commercial motion sensor based low-cost and convenient interactive treadmill. *Sensors*, *15*(9), 23667-23683

12. McAdams, E., Krupaviciute, A., Gehin, C., Dittmar, A., Delhomme, G., Rubel, P., Fayn, J., & McLaughlin, J. (2011). Wearable electronic systems: Applications to medical diagnostics/monitoring. In A. Bonfiglio & D. De Rossi (Eds.), *Wearable monitoring systems* (179-203). New York: Springer.

13. Moradi, E., Bjorninen, T., Ukkonen, L., & Rahmat-Samil, Y. (2012). Effects of sewing pattern on the performance of embroidered dipole-type RFID tag antennas. *IEEE Antennas and Wireless Propagation Letters, 11*, 1482-1485.

14. Papi, E., Osei-Kuffour, D., Chen, Y-M. A., & McGregor, A.H. (2015). Use of wearable technology for performance assessment: A validation study. *Medical Engineering and Physics, 37*(7), 698-704

15. Patel, S., Park, H., Bonato, P. Chan, L., & Rodgers, M. (2012). A review of wearable sensors and systems with application in rehabilitation. *Journal of Neuroengineering and Rehabilitation, 9*(21), 1-17.

16. Potter, D. Accessible motion capture set to revolutionize medical device development. Retrieved from http://www.mdtmag.com/blog/2015/11/accessible-motion-capture-set-revolutionize-medical-device-development (2015, accessed 2 February 2017).

17. Schmitz, A., Ye, M., Shapiro, R., Yang, R., & Noehren, B. (2014). Accuracy and repeatability of joint angles measured using a single camera markerless motion capture system. *Journal of Biomechanics, 47*, 587-591.

18. Stitch and seam guide to the ASTM Standard D-6193. Retrieved from http://www.garmento.org/751Astitchesandseams/apparelsearchs&s.html (2004, accessed 7 January 2017)

19. Umar, J., Hussain, T., & Maqsood, M. (2016). Modeling the mechanical and compression properties of polyamide/elastane knitted fabrics used in compression sportswear. *Journal of the Textile Institute, 107*(10), 1240-1252.

20. Umar, J. (2016, October). Comparison of Stretch and Recovery Properties of Polyester and Polyamide Knitted Fabrics for Compression Sportswear. Presented at The Fiber Society 2016 Fall Meeting and Technical Conference, Ithaca, New York.

21. Under Armour - Product DNA. Retrieved from https://www.underarmour.com/en-us/mens-ua-football-armour-short-sleeve-compression-shirt/pcid1276826-600 (2017, accessed 12 January 2017).

22. Yi, W. (2015). Flexible fabric strain sensors. In X. Tao (Ed.), *Handbook of smart textiles* (293-312). Singapore: Springer.

23. Zhou, H., Huosheng, H., & Tao, Y. (2006). Inertial measurements of upper limb motion. *Medical & Biological Engineering & Computing, 44*, 479-487.

Chapter 4

1. Valevicius, A. M., Jun, P. Y., Hebert, J. S., & Vette, A. H. (2018). Use of optical motion capture for the analysis of normative upper body kinematics during functional upper limb tasks: A systematic review. *Journal of Electromyography and Kinesiology, 40*, 1-15.

2. van, . K. E., & Reijne, M. M. (2018). Accuracy of human motion capture systems for sport applications; state-of-the-art review. *European Journal of Sport Science, 18*(6), 806-819

3. Papi, E., Osei-Kuffour, D., Chen, Y-M. A., & McGregor, A. H. (2015). Use of wearable technology for performance assessment: A validation study. *Medical Engineering and Physics, 37*(7), 698-704

4. Schmitz, A., Ye, M., Shapiro, R., Yang, R., & Noehren, B. (2014). Accuracy and repeatability of joint angles measured using a single camera markerless motion capture system. *Journal of Biomechanics, 47*, 587-591.

5. Potter, D. Accessible motion capture set to revolutionize medical device development. Retrieved from http://www.mdtmag.com/blog/2015/11/accessible-motion-capture-set-revolutionize-medical-device-development (2015, accessed 2 February 2017).

6. Fong, D. T. P., & Chan, Y. Y. (2010). The use of wearable inertial motion sensors in human lower limb biomechanics studies: A systematic review. *Sensors, 10*(12), 11556-11565.

7. Zhou, H., Huosheng, H., & Tao, Y. (2006). Inertial measurements of upper limb motion. *Medical & Biological Engineering & Computing, 44*, 479-487.

8. Patel, S., Park, H., Bonato, P. Chan, L., & Rodgers, M. (2012). A review of wearable sensors and systems with application in rehabilitation, *Journal of Neuroengineering and Rehabilitation, 9*(21), 1-17.

9. Chan, M., Estève, D., Fourniols, J-Y., Escriba, C., & Campo, E. (2012). Smart wearable systems: Current status and future challenges. *Artificial Intelligence in Medicine, 56*, 137-156.

10. Dunne, L. E., Ashdown, S. P., & Smyth, B. (2005). Expanding garment functionality through embedded electronic technology. *Journal of Textile and Apparel, Technology and Management, 4*(3).

11. Yi, W. (2015). Flexible fabric strain sensors. In X. Tao (Ed.), *Handbook of smart textiles* (293-312). Singapore: Springer.

12. Gioberto, G., & Dunne, L. (2012). *Theory and characterization of a top-thread coverstitched stretch sensor.* IEEE Conference on Systems, Man, and Cybernetics, 3275-3280.

13. Gioberto, G., & Dunne, L. (2013). Overlock-stitched stretch sensors: Characterization and effect of fabric property. *Journal of Textile and Apparel, Technology and Management, 8*(3), 1-14.
14. Greenspan, B., Hall, M. L., Cao, H., & Lobo, M. A. (2018). Development and testing of a stitched stretch sensor with the potential to measure human movement. *Journal of the Textile Institute, 109*(11), 1493-1500.

15. Atalay, A., Sanchez, V., Atalay, O., Vogt, D. M., Haufe, F., Wood, R. J., & Walsh, C. J. (2017). Batch Fabrication of Customizable Silicone-Textile Composite Capacitive Strain Sensors for Human Motion Tracking. *Advanced Materials Technologies, 2,* 9, 1700136.
16. Doshi, S. M., & Thostenson, E. T. (2018). Thin and Flexible Carbon Nanotube-Based Pressure Sensors with Ultrawide Sensing Range. *ACS Sensors, 3*(7), 1276-1282

17. O'Brien, B., Gisby, T., Anderson, I., O'Brien, B., Gisby, T., Anderson, I., Anderson, I., ... Electroactive Polymer Actuators and Devices, EAPAD 2014. (2014). Stretch sensors for human body motion. *Proceedings of Spie - the International Society for Optical Engineering, 9056.*

18. S. Ma, T. Ye, T. Zhang, Z. Wang, K. Li, M. Chen, J. Zhang, Z. Wang, S. Ramakrishna, L. Wei. (2018). Highly Oriented Electrospun P(VDF-TrFE) Fibers via Mechanical Stretching for Wearable Motion Sensing. *Adv. Mater. Technol., 3*, 1800033.

19. Korshøj, M., Skotte, J. H., Christiansen, C. S., Mortensen, P., Kristiansen, J., Hanisch, C., Ingebrigtsen, J., ... Holtermann, A. (2014). Validity of the Acti4 software using ActiGraph GT3X+accelerometer for recording of arm and upper body inclination in simulated work tasks. *Ergonomics, 57*(2), 247-253.

20. Kim, J., Gravunder, A., & Park, H. S. (2015). Commercial motion sensor based low-cost and convenient interactive treadmill. *Sensors*, *15*(9), 23667-23683.

21. Curtis, C., Stephens, D., Clarke, H. M., & Andrews, D. (2002). The active movement scale: An evaluative tool for infants with obstetrical brachial plexus palsy. *Journal of Hand Surgery-American Volume, 27A*(3):470-478.

22. Bare Conductive - Electric Paint Application Notes. (2016) Retrieved September 4, 2018, from https://www.bareconductive.com/wp-content/uploads/2016/05/ElectricPaint_ApplicationNotes.pdf

23. Atalay, O., Kennon, W. R., & Husain, D. M. (2013). Textile-based weft knitted sensors: Effect of fabric parameters on sensor properties. *Sensors*, *13*(8), 11114-11127.

24. Gimpel, S., Mohring, U., Muller, H., Neudeck, A., & Scheibner, W. (2004). Textile-based electronic substrate technology. *Journal of Industrial Textiles*, *33*(3), 179-189.

25. Umar, J., Hussain, T., & Maqsood, M. (2016). Modeling the mechanical and compression properties of polyamide/elastane knitted fabrics used in compression sportswear. *Journal of the Textile Institute, 107*(10), 1240-1252.

26. Umar, J (2016, October). *Comparison of Stretch and Recovery Properties of Polyester and Polyamide Knitted Fabrics for Compression Sportswear*. Presented at The Fiber Society 2016 Fall Meeting and Technical Conference, Ithaca, New York.

Chapter 5

1. Fetters, L., & Huang, H. H. (2007). Motor development and sleep, play, and feeding positions in very-low-birthweight infants with and without white matter disease. *Developmental Medicine and Child Neurology, 49*(11), 807-13.

2. Fogel, A., Messinger, D. S., Dickson, K. L., & Hsu, H. (1999). Posture and gaze in early mother-infant communication: Synchronization of developmental trajectories. *Developmental Science, 2*(3), 325-332.

3. Lobo, M. A., & Galloway, J. C. (2008). Postural and Object-Oriented Experiences Advance Early Reaching, Object Exploration, and Means-End Behavior. *Child Development, 79*(6), 1869-1890.

4. Lobo, M. A., & Galloway, J. C. (2012). Enhanced Handling and Positioning in Early Infancy Advances Development Throughout the First Year. *Child Development, 83*(4), 1290-1302.

5. Sgandurra, G., Bartalena, L., Cecchi, F., Cioni, G., Giampietri, M., Greisen, G.,... Dario, P. (2016). A pilot study on early home-based intervention through an intelligent baby gym (CareToy) in preterm infants. *Research in Developmental Disabilities*, 32-42.

6. Sgandurra, G., Lorentzen, J., Inguaggiato, E., Bartalena, L., Beani, E., Cecchi, F., ... CareToy Consortium. (2017). A randomized clinical trial in preterm infants on the effects of a home-based early intervention with the 'CareToy System'. Plos One, 12, 3.

7. Lang, C. E., Bland, M. D., Bailey, R. R., Schaefer, S. Y., & Birkenmeier, R. L. (2013). Assessment of upper extremity impairment, function, and activity after stroke: Foundations for clinical decision making. *J Hand Ther, 26*(2), 104-115.

8. Migueles, J. H., Cadenas-Sanchez, C., Ekelund, U., Delisle Nyström, C., Mora-Gonzalez, J., Löf M, et al.(2017). Accelerometer Data Collection and Processing Criteria to Assess Physical Activity and OtherOutcomes: A Systematic Review and Practical Considerations. *Sports Med, 47*(9):1821-1845. doi: 10.1007/s40279-017-0716-0.

9. Bailey, R. R., & Lang, C. E. (2013). Upper-limb activity in adults: referent values using accelerometry. *J Rehabil Res Dev, 50*(9), 1213-1222.

10. Bassett, D. R., Rowlands, A., & Trost, S. G. (2012). Calibration and validation of wearable monitors. *Med Sci Sports Exerc, 44*(1), S32-38. doi: 10.1249/MSS.0b013e3182399cf7.

11. Sarker, H., Anderson, L. N., Borkhoff, C. M., Abreo, K., Tremblay, M. S., Lebovic, G., et al.(2015). Validation of parent-reported physical activity and sedentary time by accelerometry in young children. *BMC Res Notes. 30*(8), 735. doi: 10.1186/s13104-015-1648-0.

12. Sasaki, J. E., John, D., & Freedson, P. S. (2011). Validation and comparison of ActiGraph activity monitors. *J Sci Med Sport, 14*(5), 411-416. doi: 10.1016/j.jsams.2011.04.003.

13. Waddell, K. J., et al. (2016). Does task-specific training improve upper limb performance in daily life post-stroke? *Neurorehabil Neural Repair.*

14. Hauck, J. L.,& Ulrich, D. A. (2013). Developmental trajectory of physical activity for infants ages 0-6 months. *Res Q Exerc Sport,84,* A32.

15. Ohgi, S., Morita, S., Loo, K. K., & Mizuike, C. (2008). Time series analysis of spontaneous upper-extremity movements of premature infants with brain injuries. *Phys Ther, 88*(9), 1022-1033. doi: 10.2522/ptj.20070171.

16. Smith, B. A., Trujillo-Priego, I. A., Lane, C. J., Finley, J. M., & Horak, F. B. (2015). Daily Quantity of Infant Leg Movement: Wearable Sensor Algorithm and Relationship to Walking Onset. *Sensors, 15*(8), 19006-19020.

17. Shida-Tokeshi, J., Lane, C. J., Trujillo-Priego, I. A., Deng, W., Vanderbilt, D. L., Loeb, G. E., & Smith, B. A. (2018). Relationships between full-day arm movement characteristics and developmental status in infants with typical development as they learn to reach: An observational study. *Gates Open Research, 2*.

18. Trujillo-Priego, I. A., Lane, C. J., Vanderbilt, D. L., et al. (2017). Development of a Wearable Sensor Algorithm to Detect the Quantity and Kinematic Characteristics of Infant Arm Movement Bouts Produced across a Full Day in the Natural Environment. *Technologies (Basel), 5*(3), 39.

19. Pitchford, E. A., Ketcheson, L. R., Kwon, H. J., & Ulrich, D. A. (2017).

Minimum Accelerometer Wear Time in Infants: A Generalizability Study. *J Phys Act Health, 14*(6), 421-428. doi: 10.1123/jpah.2016-0395.

20. Ridgers, N. D., Salmon, J., Ridley, K., O'Connell, E., Arundell, L., & Timperio, A. (2012). Agreement between activPAL and ActiGraph for assessing children's sedentary time. *The International Journal of Behavioral Nutrition and Physical Activity, 9*.

21. Ridley, K., Ridgers, N. D., & Salmon, J. (2016). Criterion validity of the activPALTM and ActiGraph for assessing children's sitting and standing time in a school classroom setting. *The International Journal of Behavioral Nutrition and Physical Activity, 13*, 1.

22. Pate, R. R., O'Neill, J. R., & Mitchell, J. (2010). Measurement of physical activity in preschool children. *Medicine and Science in Sports and Exercise, 42*(3), 508-512.

23. Rao, S., Thankachan, P., Amrutur, B., Washington, M., & Mony, P. K. (2018). Continuous, real-time monitoring of neonatal position and temperature during Kangaroo Mother Care using a wearable sensor: A techno-feasibility pilot study. *Pilot Feasibility Stud, 21*(4), 99. doi: 10.1186/s40814-018-0293-5.

24. Lee, J. M., Kim, Y., & Welk, G. J. (2014). Validity of consumer-based physical activity monitors. *Medicine and Science in Sports and Exercise, 46*(9), 1840-1848.

25. Warren, J. M., Ekelund, U., Besson, H., Mezzani, A., Geladas, N., Vanhees, L., et al. (2010). Assessment of physical activity—a review of methodologies with reference to epidemiological research: A report of the exercise physiology section of the European Association of Cardiovascular Prevention and Rehabilitation. *Eur J Cardiovasc Prev Rehabil, 17*(2), 127-139. doi: 10.1097/HJR.0b013e32832ed875.

26. Sirard, J. R., Trost, S. G., Pfeiffer, K. A., & Dowda, M. (2005). Calibration and evaluation of an objective measure of physical activity in preschool children. *J Phys Act Health, 2*, 345-357. https://doi.org/10.1123/jpah.2.3.345

27. American Academy of Pediatrics. (2006). Active healthy living: Prevention of childhood obesity through increased physical activity. *Pediatrics, 117*(5),1834-1842. doi:10.1542/peds.2006-0472 PubMed

28. Gima, H., Ohgi, S., Morita, S., Karasuno, H., Fujiwara, T., & Abe, K. (2011). A dynamical system analysis of the development of spontaneous lower extremity movements in newborn and young infants. *Journal of Physiological Anthropology, 30*(5), 179-86.

29. Lee, S. H., Ha, Y., Jung, M., Yang, S., & Kang, W. S. (2019). The Effects of a Mobile Wellness Intervention with Fitbit Use and Goal Setting for Workers. *Telemedicine Journal and E-Health : the Official Journal of the American Telemedicine Association.*

30. Appelboom, G., Camacho, E., Abraham, M. E., Bruce, S. S., Dumont, E. L., Zacharia, B. E., ... Connolly, E. S. J. (2014). Smart wearable body sensors for patient self-assessment and monitoring. *Archives of Public Health = Archives Belges De Sante Publique, 72,* 1.

Chapter 6

1. (2018, September 11). Over Half of The 211.2 Million Video Gamers in The U.S. Play Games Across Multiple Platforms, According to NPD. Retrieved from https://www.npd.com/wps/portal/npd/us/news/press-releases/2018/over-half-of-the-211-2-million-video-gamers-in-the-u-s--play-games-across-multiple-platforms--according-to-npd/

2. (2019). Age breakdown of video game players in the United States in 2018. Retrieved from https://www.statista.com/statistics/189582/age-of-us-video-game-players-since-2010/

3. Rowland, J. L., Malone, L. A., Fidopiastis, C. M., Padalabalanarayanan, S., Thirumalai, M., & Rimmer, J. H. (2016). Perspectives on Active Video Gaming as a New Frontier in Accessible Physical Activity for Youth With Physical Disabilities. *Physical Therapy, 96(*4), 521-532.

4. Gamespot Staff (May 13, 2000). Sony announces PS2 launch date and price. Retrieved from https://www.gamespot.com/articles/sony-announces-ps2-launch-date-and-price/1100-2568701/

5. (2019). PS4 DualShock 4 Wireless controller. Retrieved from https://www.playstation.com/en-us/explore/accessories/gaming-controllers/dualshock-4/

6. (2019). Xbox Wireless Controller – Black. Retrieved from https://www.xbox.com/en-US/xbox-one/accessories/controllers/xbox-black-wireless-controller

7. (2019). Create your favorite color combo. Retrieved from https://www.nintendo.com/switch/choose-your-joy-con-color/

8. (2019). Head Gamer Sip-n-Puff Mouth Joystick Game Controller on Robo Arm Mount. Retrieved from https://www.broadenedhorizons.com/head-gamer-roboarm

9. (2019). The AbleGamers Charity. Retrieved from https://ablegamers.org/

10. (2019). Warfighter Engaged. Retrieved from https://warfighterengaged.org/

11. (2019). Special Effect The gamers' charity. Retrieved from https://www.specialeffect.org.uk/

12. (2019). OneSwitch.org.uk. Retrieved from http://www.oneswitch.org.uk/

13. (2019). The Controller Project. Retrieved from http://thecontrollerproject.com/

14. (2019). Xbox Adaptive Controller. Retrieved from https://www.xbox.com/en-US/xbox-one/accessories/controllers/xbox-adaptive-controller

15. (2019) Accessible Game Controls. Retrieved from https://www.broadenedhorizons.com/products/accessible-game-controls

16. Primack, B. A., et al. (June 2012) Role of Video Games in Improving Health-Related Outcomes. *American Journal of Preventive Medicine , Volume 42*(6) , 630-638.

17. Klasen, M. A., Lasky, J., & Atkins, W. A. (2018). Video games. In D. S. Blanchfield (Ed.), *The Gale Encyclopedia of Children's Health: Infancy through Adolescence* (3rd ed.). Farmington Hills, MI: Gale.

18. Brown, D. (2006). Playing to win: Video games and the fight against obesity. *Journal of the American Dietetic Association, 106*, 2.

19. Ivory, J. D., & Kalyanaraman, S. (2007). The Effects of Technological Advancement and Violent Content in Video Games on Players' Feelings of Presence, Involvement, Physiological Arousal, and Aggression. *Journal of Communication, 57*(3), 532-555.

20. Sparrowhawk, K., Kumar, R., & Harrison, J. (2014). Adaptive Video Games can Assess and Enhance Cognitive Health. *Value in Health, 17*, 7.

21. O'Donovan, C., Greally, P., Canny, G., McNally, P., & Hussey, J. (2014). Active video games as an exercise tool for children with cystic fibrosis. *Journal of Cystic Fibrosis, 13*(3), 341-346.

22. Robert, M., Ballaz, L., Hart, R., & Lemay, M. (2013). Exercise intensity levels in children with cerebral palsy while playing with an active video game console. *Physical Therapy, 93*(8), 1084-1091.

23. Howcroft, J., Klejman, S., Fehlings, D., Wright, V., Zabjek, K., Andrysek, J., & Biddiss, E. (2012). Active Video Game Play in Children With Cerebral Palsy: Potential for Physical Activity Promotion and Rehabilitation Therapies. *Archives of Physical Medicine and Rehabilitation, 93*(8), 1448-1456.

24. Arbour-Nicitopoulos, K. P., Grassmann, V., Orr, K., McPherson, A. C., Faulkner, G. E., & Wright, F. V. (2018). A Scoping Review of Inclusive Out-of-School Time Physical Activity Programs for Children and Youth With Physical Disabilities. *Adapted Physical Activity Quarterly, 1*, 1-28.

25. Widman, L. M., McDonald, C. M., & Abresch, R. T. (2006). Effectiveness of an upper extremity exercise device integrated with computer gaming for aerobic training in adolescents with spinal cord dysfunction. *The Journal of Spinal Cord Medicine, 29*(4), 363-370.

26. Li, W., Lam-Damji, S., Chau, T., & Fehlings, D. (2009). The development of a home-based virtual reality therapy system to promote upper extremity movement for children with hemiplegic cerebral palsy. *Technology and Disability, 21*(3), 107-113.

27. Wiemeyer, J., Deutsch, J., Malone, L. A., Rowland, J. L., Swartz, M. C., Xiong, J., & Zhang, F. F. (2015). Recommendations for the Optimal Design of Exergame Interventions for Persons with Disabilities: Challenges, Best Practices, and Future Research. *Games for Health Journal, 4*(1), 58-62.

28. Jannink, M. J., Van Der Wilden, . W. G. J., Navis, D. W., Visser, G., Gussinklo, J., & Ijzerman, M. (2008). A low-cost video game applied for training of upper extremity function in children with cerebral palsy: A pilot study. *Cyberpsychology & Behavior: the Impact of the Internet, Multimedia and Virtual Reality on Behavior and Society, 11*(1), 27-32.

29. Ronald, R. M., Maura, D. I., Yingzi, L., & Jeffrey, B. (2010). Biosensor-Based Video Game Control for Physically Disabled Gamers. *Proceedings of the Human Factors and Ergonomics Society Annual Meeting, 54*(28), 2383-2387.

30. Goršič, M., Cikajlo, I., Goljar, N., & Novak, D. (2017). A multisession evaluation of an adaptive competitive arm rehabilitation game. *Journal of Neuroengineering and Rehabilitation, 14*(1), 1-15.

31. Betker, A. L., Szturm, T., Moussavi, Z. K., & Nett, C. (2006). Video Game-Based Exercises for Balance Rehabilitation: A Single-Subject Design. *Archives of Physical Medicine and Rehabilitation, 87*(8), 1141-1149. Borckardt, J. J., & Nash, M. R. (2014). Simulation modelling

analysis for small sets of single-subject data collected over time. *Neuropsychological Rehabilitation, 24*, 492-506.

32. Beeston, J., Power, C., Cairns, P., Power, C., Cairns, P., Barlet, M., & 16th International Conference on Computers Helping People with Special Needs, ICCHP 2018. (2018). Accessible player experiences (APX): The players. *Lecture Notes in Computer Science (including Subseries Lecture Notes in Artificial Intelligence and Lecture Notes in Bioinformatics)*, 245-253.

33. Eliasson, A. C., Krumlinde-Sundholm, L., Rösblad, B., Beckung, E., Arner, M., Öhrvall, A. M., & Rosenbaum, P. (2006). The Manual Ability Classification System (MACS) for children with cerebral palsy: Scale development and evidence of validity and reliability. *Developmental Medicine & Child Neurology, 48*(7), 549-554.

34. Palisano, R. J., Rosenbaum, P., Bartlett, D., & Livingston, M. H. (2008). Content validity of the expanded and revised Gross Motor Function Classification System. *Developmental Medicine and Child Neurology, 50*(10), 744-750.

APPENDIX

APPENDIX A

SENSOR WEARING LOG

Please write down the start and stop times when your child is wearing the garment while awake. The garment can be taken off at any time, and should be taken off before naps, but please record when the garment is not being worn.
Time and date we left the garment: ___ : ___ , ___ / ___ / _____

	Start	Stop
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___

Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___
Awake and playing	___ : ___	___ : ___
Changing, feeding, sleeping	___ : ___	___ : ___

Time and date we picked up the garment: ___ : ___ , ___ / ___ / _____

APPENDIX B

VIDEO GAME PERCEPTION INTERVIEW QUESTIONS

For child:

- How old are you in years?
- What video game console/system do you currently own and play on the most?
- What video game controller do you use to play?
- Do you have any trouble using your hands to play with video game controllers?
- What video game consoles/systems have you owned in the past, such as Xbox, Play Station, etc.?
- Have you tried other controllers?
    - If yes, what kinds have you tried?
    - What did you like about them?
    - What didn't you like about them?
- Have you used an adaptive game controller – one that has changes that are supposed to make it easier for you to play?
- Do your hands get tired while playing video games with controllers?
- Do you get frustrated when playing video games with controllers? If so, why?
- What are the 5 most common video games you currently play?
- What is your favorite video game?
    - What buttons are easy to press/ reach in that game?
        - How easy is it to press that button?
            - Very difficult 1 2 3 4 5 Very easy
    - What button is the most difficult to press/ reach in that game?
        - How difficult is it to press that button?
            - Very easy 1 2 3 4 5 Very difficult
- Do you play video games with your friends?
- Would you like to play more video games with your friends? If so, what is keeping you from doing this?
- Would you like to be better at playing video games? If so, what is keeping you from being better?
- Is there anything else you would like to let me know about how you use video game controllers to play or about your ability to use your hands for other things?

For parents:

- Does your child have any diagnoses or identified disabilities? If so, could you please share them with us?

- Do you allow your child to play video games on a regular basis alone? With friends?
- How many minutes of video games does your child usually play each day? Each week?
- Is your child able to participate fully when playing video games alone or with friends? If not, can you please elaborate on what you see as the obstacles to full play?
- Are there any video games your child excels in?
- Are there any video games your child struggles with but would like to better play?
- Are there any video game controllers that work well for your child?
- Are there any video game controllers your child struggles with but would like to better use?
- Are there any functional fine motor tasks you notice your child struggling with, both video game and not video game related?
- Is there anything else you would like to let me know about how you use video game controllers to play or about your child's hand use in general?

APPENDIX C

ACCESSIBLE VIDEO GAME CONTROLLER EXIT SURVEY

- How much do you like the video game controller adapter?
  - Don't like 1 2 3 4 5 Like it a lot
- How easy was it to learn to use the video game controller adapter?
  - Very hard 1 2 3 4 5 Very easy
- How hard was the hardest button to press before? What buttons was that: _____
  - Very hard 1 2 3 4 5 Very easy
- How hard is that same button to press now?
  - Very hard 1 2 3 4 5 Very easy
- How cool do you find the controller?
  - Not cool 1 2 3 4 5 Very cool
- Will you continue to use this controller outside of this study?
  - Will not use 1 2 3 4 5 Will use often
- What do you think having a controller like this will let you do? _____

_____

_____

_____

- What recommendations would you make to improve the controller adapter, for instance making it more comfortable, activating different buttons, or easier to use? _____

_____

_____

_____

APPENDIX D

IRB APPROVAL LETTERS

Chapter 4 IRB Approval Letter

UNIVERSITY OF DELAWARE
RESEARCH OFFICE
210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE:	July 26, 2018
TO:	Michele Lobo, PT, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1110233-2] Smart Wearable Systems to Support and Measure Movement in Individuals With and Without Mobility Impairments
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
APPROVAL DATE:	July 26, 2018
EXPIRATION DATE:	August 15, 2019
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category 9

Chapter 5 IRB Approval Letter

RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE:	October 17, 2018
TO:	Michele Lobo, PT, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1152018-3] A Novel Parent Education Program for Early Intervention
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
APPROVAL DATE:	October 17, 2018
EXPIRATION DATE:	November 14, 2019
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category 9

Chapter 6 IRB Approval Letter 1

RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE:	April 27, 2018
TO:	Mark Greenspan
FROM:	University of Delaware IRB
STUDY TITLE:	[1234312-1] Identifying challenges impacting video game play for individuals with physical disabilities affecting arm function
SUBMISSION TYPE:	New Project
ACTION:	APPROVED
APPROVAL DATE:	April 27, 2018
EXPIRATION DATE:	April 26, 2019
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(6,7)*
	Research with Children- Subpart D Determination 45 CFR 46.404

Chapter 6 IRB Approval Letter 2

RESEARCH OFFICE

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DATE:	January 31, 2019
TO:	Mark Greenspan
FROM:	University of Delaware IRB
STUDY TITLE:	[1368449-1] Design and testing of an accessible video game controller adapter for children with challenges moving their arms
SUBMISSION TYPE:	New Project
ACTION:	APPROVED
APPROVAL DATE:	January 31, 2019
EXPIRATION DATE:	January 15, 2020
REVIEW TYPE:	Full Committee Review

Sub Part D Determination 45 CFR 46.404

APPENDIX E

RELATED PATENT

Chapter 3:

Conductive Thread Stitched Stretch Sensor, 32/343,899 & 15/289,465

Mark Greenspan, Martha Hall, Michele Lobo, & Huantian Cao