Trunk Rehabilitation Using Cable-Driven Robotic Systems

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Abstract

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Upper body control is required to complete many daily tasks. One needs to stabilize the head and trunk over the pelvis, as one shifts the center of mass to interact with the world. While healthy individuals can perform activities that require leaning, reaching, and grasping readily, those with neurological and musculoskeletal disorders present with control deficits. These deficits can lead to difficulty in shifting the body center of mass away from the stable midline, leading to functional limitations and a decline in the quality of activity. Often these patient groups use canes, walkers, and wheelchairs for support, leading to occasional strapping or joint locking of the body for trunk stabilization.

Current rehabilitation strategies focus on isolated components of stability. This includes strengthening, isometric exercises, hand-eye coordination tasks, isolated movement, and proprioceptive training. Although all these components are evidence based and directly correlate to better stability, motor learning theories such as those by Nikolai Bernstein, suggest that task and context specific training can lead to better outcomes. In specific, based on our experimentation, we believe functional postural exploration, while encompassing aspects of strengthening, hand-eye coordination, and proprioceptive feedback can provide better results.

In this work, we present two novel cable robotic platforms for seated and standing posture training. The Trunk Support Trainer (TruST) is a platform for seated posture rehabilitation that provides controlled external wrench on the human trunk in any direction in real-time. The Stand Trainer is a platform for standing posture rehabilitation that can control the trunk, pelvis, and knees, simultaneously. The system works through the use of novel force-field algorithms that are

modular and user-specific. The control uses an assist-as-needed strategy to apply forces on the user during regions of postural instability. The device also allows perturbations for postural reactive training.

We have conducted several studies using healthy adult populations and pilot studies on patient groups including cerebral palsy, cerebellar ataxia, and spinal cord injury. We propose new training methods that incorporate motor learning theory and objective interventions for improving posture control. We identify novel methods to characterize posture in form of the “8-point star test”. This is to assess the postural workspace. We also demonstrate novel methods for functional training of posture and balance.

Our results show that training with our robotic platforms can change the trunk kinematics. Specifically, healthy adults are able to translate the trunk further and rotate the trunk more anteriorly in the seated position. In the standing position, they can alter their reach strategy to maintain the upper trunk more vertically while reaching. Similarly, Cerebral Palsy patients improve their trunk translations, reaching workspace, and maintain a more vertical posture after training, in the seated position. Our results also showed that an Ataxia patient was able to improve their reaching workspace and trunk translations in the standing position. Finally, our results show that the robotic platforms can successfully reduce trunk and pelvis sway in spinal cord injury patients. The results of the pilot studies suggest that training with our robotic platforms and methods is beneficial in improving trunk control.

Table of Contents

List of Figures

List of Tables

Acknowledgments

"The more I learn, the more I realize how much I don't know" – Albert Einstein

Research has been a means to explore and unravel the secrets of the world. As I embarked on this long educational journey, I found myself intrigued by the intricate and hidden nature of the physics of the world and mind. Often as we progress, we are engulfed by new questions and problems and leave behind areas that first sparked our curiosity. It is important to remember the root of our curiosity and motivation. A speck of a contribution to our universe and society can render the challenges worthwhile.

I would like to express heartfelt appreciation to all those who have supported me in the long and arduous process of my educational pursuit. First off, I would like to thank my advisor and mentor, Dr. Sunil Agrawal, who believed in me and my abilities and gave me an opportunity to attempt at my childhood dream. He has been an inspiration through his passion for research and scientific curiosity. I highly value his patience in letting me explore and figure out solutions to problems and his indirect nature of challenging me to push harder and develop a sense of innate confidence.

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A series of milestones is what we call life. Each milestone, big or small, is the culmination of one’s dedication and perseverance, yet never possible without the contributions of those who invest in us. I thank each and every one of you who has inspired, motivated, and helped me explore my own capabilities.

Chapter 1

1 Introduction

1.1 Posture Control

Trunk stability is a prerequisite for balance, gait, and daily activities [1]. Trunk control is the ability of the trunk to maintain upright posture, respond to weight shifts and disturbances, and maintain the body center of mass (COM) within the base of support (BOS). Postural control requires the stabilization and control of the head, trunk, pelvis, and knees to complete many daily tasks, e.g., reaching for an object, walking, sports. Although sitting or standing statically upright can be less demanding with the COM of the upper body centered over the BOS, large and quick displacements of the upper body outside the pelvic or foot boundaries can cause a sudden shift of the COM away from the center of the BOS, creating a lack of a stable neutral upper body and pelvic position. In this situation, high volitional control of the upper body is required to recover verticality and maintain stability. Maintaining balance and posture requires muscle synchrony and kinematic coordination.

While healthy individuals can readily perform everyday activities, such as leaning, reaching, and grasping, patients with neurological and musculoskeletal disorders (MSDs) may have deficits in shifting their weight and moving their upper body within and beyond the BOS. These tasks are neurologically demanding requiring the primary motor cortex, somatosensory systems, and frontal and parietal areas to play an essential role to maintain balance [2]. The musculoskeletal structure within the human body responds to the gravitational forces and resists internal and external disturbances during motor tasks while providing mechanical support and balance [3]–[5]. Postural control can be divided into two tasks: (i) generation of direction specific kinematic movements and (ii) adaptation of these specific movements based on multi-sensorial afferent inputs [2], [6], [7]. Studies have shown that direction specificity emerges as a result of self-organization of intersegmental components during exposure to new positions [6], [8], [9]. Therefore, performing a new or more difficult task can show task-specific kinematic adaptations of posture. Posture control requires stabilization of all linked segments of the body through complex coordination of biomechanical, sensory, motor, and central nervous systems [10], [11].

1.2 Motor Control

In the study of motor control, researchers are interested in identifying how our neuromuscular system functions to activate and coordinate the muscles and limbs when performing a motor skill. A motor skill is a task or activity that has a specific purpose or goal to achieve. In the textbook "Motor Learning and Control: Concepts and Applications", authors Richard Magill and David Anderson identify and discuss various motor control theories and their applications [12].

Movement studies look at how specific patterns of motion among joint and body segments are achieved. These movements make up a motor skill. Therefore, movement components are what allow an action goal to be achieved. There are three major reasons as to why we need to distinguish between motor movements and skills. First is that people learn movements as they try to acquire a certain skill. Though the movement is the same for most individual, the skill or technique used will vary. If we consider a shooters technique in professional basketball, all shooters need to move their hand in the shooting motion to move the ball to the basket. However, each shooter has a different and unique form which makes their technique work best for them. The second is that people adapt movement characteristics to achieve a common action goal. This means that although people may move differently, the outcome of their movement produces the same action goal. Due to the different physical features and abilities of a person, the movement pattern may be different while still achieving the same action goal, such as putting the ball in the basket. This can be especially important for a patient population that may use different techniques to compensate for walking deficits and yet produce the same end goal of walking. The third and final reason is that people evaluate motor skill performance, movements, and neuromotor processes with different types of measures. While a motor skill is measured by its outcome such as distance walked, a movement is evaluated by displacements, rotations, velocities, accelerations, and forces that allow these movements.

As described by Antoinette Gentile, one way to classify a skill is by the stability of the physical environment or “environment context” in which a skill is performed [13]. This takes into account the physical location where a skill is performed. The areas encompassing the environmental context are the supporting surface, objects involved, and people involved. In understanding the movement a person carries out for completing a given action or task, it is important to consider the surface the task was performed on, the objects in interaction such as a ball or box, and people in the environment that serve as obstacles or cues. In certain situations, some environmental contexts may be stationary or moving. If objects and environment are stationary, this is considered as a closed motor skill. If objects and environment are in motion, this is an open motor skill.

Researchers believe that motor skill performance is influenced by three major areas: motor skills, performance environment, and physical and psychological factors of the person. In understanding these areas, the goal is to identify how one becomes skillful, going from infancy to old age. Motor control theories help us understand how the nervous system produces coordinated movement such that humans are able to perform a variety of motor skills in a variety of environmental context. These theories describe a large class of observations and make definitive predictions about the results of future observations. In other words, these theories describe how humans produce coordinated movements. Coordination is how humans pattern the head, body, and limbs relative to the environment. Since coordination involved the whole body, it becomes necessary to understand how the nervous system controls the muscles and joints involved in producing complex movement patterns. Many existing motor control theories aim to answer this problem using the bases of “degrees of freedom” problem first proposed by Nikolai Bernstein in 1930s. He proposed that to perform a coordinated movement, the nervous system had to solve the degrees of freedom. Degrees of freedom is the number of independent elements or components of a system and the number of ways each component can vary. To learn a new skill, it is necessary to solve a complex degrees of freedom problem.

In the degrees of freedom problem, a complex coordination between the body and joints needs to be solved. The nervous system must first determine the actual number of degrees of freedom that must be controlled for coordinated movement. Bernstein argued that the nervous system must break the problem of learning a new skill into stages. First it must determine the level of actions such as which motor segments need to be controlled first. Then the movements have to be coordinated to the external environment, followed by organizing muscle synergies and finally regulating muscle tone. The second stage of the problem solving requires developing a plan or strategy. This is how the performance would look from the outside or the environment. The third stage is where the most appropriate sensory corrections are made from the body's internal point of view. These stages incorporate standardization and stabilization. Bernstein stated that such a complex process requires repetition and follows the "law of practice". This law states that practice will lead to large initial changes which will taper off with practice.

Following Bernstein's theory of degrees of freedom and Gentile's two-stage model of motor learning, movement coordination can be studied further. Gentile's two stage model states that motor learning happens in two stages. In the first, a beginner must acquire a movement pattern that will allow some degree of success at achieving an action goal. The second stage is where the learner discriminates between regulatory and nonregulatory conditions in the environmental context. The regulatory context is one which will define the movement needs, such as the shape of the cup. The nonregulatory is one that is nonimportant, such as the color of the cup. In performing the first stage of Gentile's model, where some degree of success of a task is needed, a beginner may alter their movement strategy. Bernstein explains this as the locking or freezing of degrees of freedom. To gain initial control of the many degrees of freedom, a beginner may hold some degrees of movement rigid by coupling joint motion in tight synchrony. In most cases, distant joints are frozen together and the most proximal joint is moved to complete the task. An example may be writing with the nondominant hand. Usually the writing appears sloppy, yet readable due to the wrist and elbow being frozen, while the shoulder, most proximal joint is moved to achieve some level of action success. As the task is practiced, the joints are released or unfrozen, operating as a multisegmental unit. This creates what is known as "functional synergy" where the arm and hand work together in a cooperative manner to allow for optimal performance of a skill.

Many new theories have derived from these initial motor control theories and understanding. Theories can be divided into two major classifications based on how the nervous system controls coordinated movement in terms of relative importance given to movement instructions specified by central components of control and the environment. Theories that give primary importance to movement instructions by the central nervous system (CNS) have some form of memory representation or motor program for the basis of organizing and carrying out an action. The others give more importance to movement instructions specified by the environment and the dynamic interaction of the environment with the body, limbs, and nervous system.

The motor program theories are led by the work of Richard Schmidt, who proposed the generalized motor program as a mechanism that could account for the adaptive and flexible qualities of human coordinated movement behavior. This uses the fundamental pattern of the class of actions, which remains consistent from one performance of an action to another. To achieve success in a movement, one must retrieve appropriate coordination program from memory and then add movement specific parameters. This uses rhythmic time sequencing to achieve coordination. While a task duration may change, the proportions of required activations are invariant and can be used to complete a skilled task. In contract to this is the dynamical systems theory. This theory uses aspects of physics, biology, chemistry, and mathematics to explain human movement control. Here the human movement control is thought of as a complex system that behaves in similar ways to complex biological or physical system. This is similar to nonlinear dynamics. In this the system behavior changes over time and does not follow a continuous, linear progression. At different time instances, the behavior and its characteristics abruptly change due to varying factors such as the environment and internal body behavior. Experiments in literature often stem from these theories and work to identify which theory is most prevalent to human coordination and skill performance.

1.3 Traditional Posture Rehabilitation

Posture rehabilitation uses the physical and physiological understanding of the human body to help patients gain or regain optimal human performance. These training programs use a combination of scientific and practical application needs to help patient progress through recovery. In rehabilitation of patients with severe upper body motor dysfunction, postural training is usually conducted by fixing the trunk or pelvis using a rigid frame, strap, or cord [14], [15]. This allows practical interaction with the patient in a safe and secure environment. Though this can provide added stability while practicing a task, this method is passive and restrictive. This limits the patient's postural adjustments as the trunk is held fixed and also reduces postural sensorimotor experience due to constrained degrees-of-freedom (DOF), as described in the motor control section in Chapter 1.2. Locking the joints will allow an individual to achieve task or action success as necessary under Gentile's first stage of motor learning. Yet, the environmental context is lost as body coordination is not required.

As outlined in Fig. 1.1, strategies for postural rehabilitation are very limited, and often several components of balance and posture are isolated and trained. The components include muscle activation and synergies, kinematic coordination, and proprioception. This is done to retain or build muscle activity and composition, improve kinematic coordination, and train proprioception awareness. Balance and posture are a combination of proprioception, kinematic coordination, and muscle activation. Yet, traditional training paradigms isolate these components and fail to address functional rehabilitation, which is performing a task in an appropriate environmental context.

In patients with severe postural instabilities, joint locking and harnesses are used. While this may reduce sway in the patient, it does not actively allow the patient to control their trunk and posture. The motor control and planning are altered in this situation. This mean though a person may move a hand to complete a task, the locking of joints makes the task different from that in a real environment. A real environment is one where obstacles such as objects and humans are observed, and coordination changes are made to adhere to the task requirements. Though locking joints is beneficial if the task being trained will be performed in a constrained environment going forward in the patient's progress, it does not allow for recovery of the fixed body segment.

In rehab settings, muscle strengthening is strongly promoted. Isometric, concentric, and eccentric exercises build muscle mass by recruiting muscle neurons. Muscle mass is necessary for maintaining bone integrity and for generating motion. However, muscle mass alone does not result in achieving task or action success. Similarly, balance training on unstable surfaces may help develop coordination and balance, but the transfer to functional tasks may not occur. As we mentioned in the earlier section on motor control, mass practice of a task will allow an individual to learn the sequence of coordination requirements from an internal and external self-representation. This would allow improvements in the task that is being trained, whether it is an athlete swinging a golf club, or a patient relearning trunk control during a reach and grasp task.

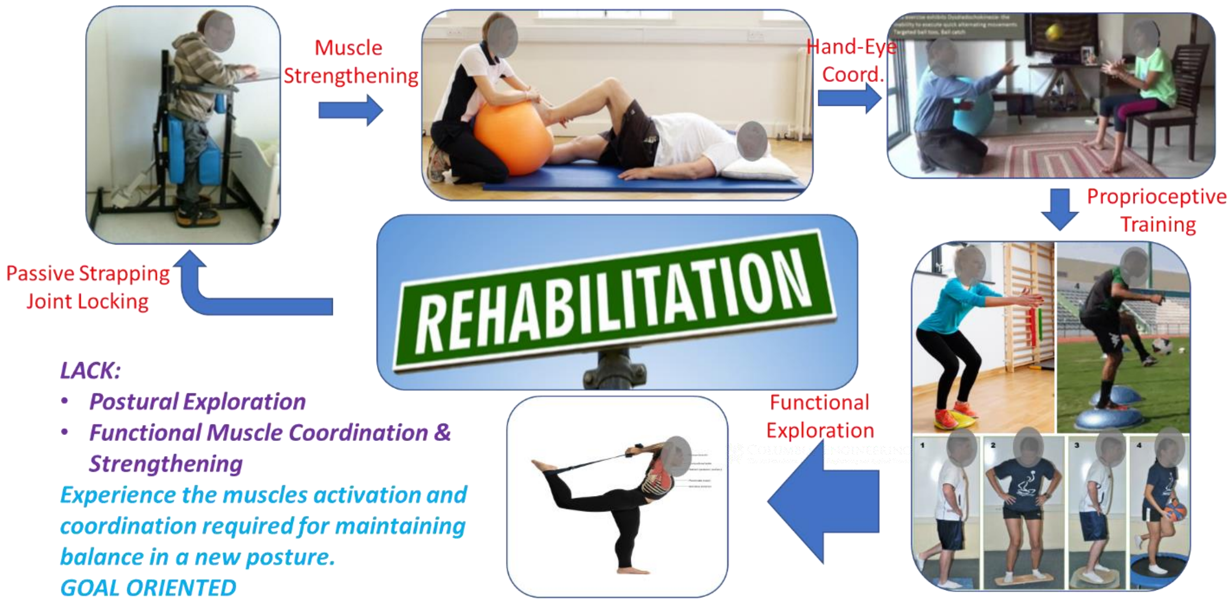


Fig. 1.1 Current rehab methods isolate various components of functional activities. This includes muscle strengthening, joint locking, hand eye coordination tasks, and proprioceptive training. Functional and goal oriented postural exploration is often lost.

1.4 Use of Technology for Posture Rehabilitation

In previous work, researchers have aimed to develop technology and rehab methods for balance training. Although each has its own benefits, most isolate aspects of balance training such as muscle strengthening, coordination, and proprioception are passively administered onto the patient. Specifically, in [18] researchers used virtual reality and a bicycle to stimulate visual and vestibular senses. Yet, this method was not task specific as is recommend by motor learning paradigms. The research geared towards training reaction from visual stimulus and proprioceptive awareness from the vestibular system, which provides spatial orientation. In [19], the “Spider-bot” was proposed, a cable driven system which provides a supporting structure during training (Fig. 1.2). This can be through of as a wheel-less and baseless walker. This type of system will allow support for those with adequate hand and feet functionality, during balance training programs. However, this method would not be suitable for those with muscular weakness, since it requires the patient to support their weight. It also does not target a specific body segment directly, as it is an external apparatus. No additional real-time sensing or training strategy was discussed. No human studies were conducted either. It was unclear as to why a supportive robotic structure would be more beneficial than a traditional cane or walker. In addition to assistive systems, many researchers have proposed balance platforms such as the Hunova by Movendo Technology, the AMBA, and bio-feedback systems, as shown in Fig. 1.2 [5]. These systems utilize visual and vibratory feedback for balance correction in form of virtual games. Feedback is provided at the feet, shank, thighs, and/or trunk using vibratory feedback. The user is guided to control their posture following a gaming strategy. To complete the game successfully, the user is required to maintain instructed postures. A visual and vibratory feedback is provided to report their performance and to improve proprioception. In the Hunova, a unstable surface can also be provided to challenge the user at the ankle. Though these biofeedback devices and methods provide a training environment, they do not directly provide assistance or resistance to the body and are not suitable for severely affected patient groups.

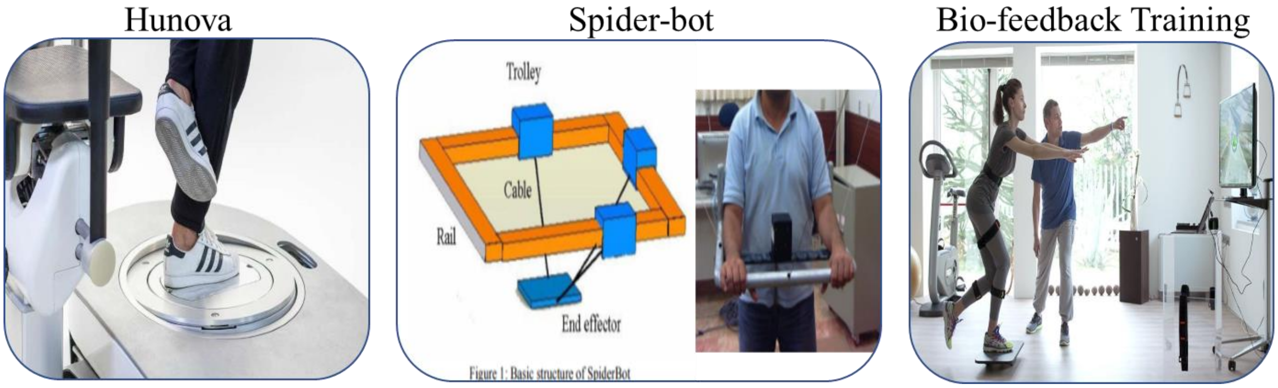


Fig. 1.2 Several technology platforms have been proposed for balance and gait training. The devices shown provide haptic and visual feedback, but do not provide active assistance.

1.5 Conclusion: Need for Assist-As-Needed Training Methods

Postural control requires a complex interaction between the sensory and motor systems. Among sensory inputs, proprioception plays a fundamental role in the fine control of postural movements. In some neuromotor disorders, defects in the proprioceptive system may be associated with unsteady, uncoordinated, and exacerbated upper body movements during activities of daily living that may secondarily result in imbalance, falls, and severe injuries. Passive or active proprioceptive interventions are implemented routinely as a critical part of rehabilitation programs that target postural balance disorders.

Considering patients with cerebellar ataxia, spinal cord injury, cerebral palsy, or even the elderly, devising rehab strategies that target a multimodal strategy can be most beneficial. These patient groups present with trunk and posture instability. Due to neurological and musculoskeletal deficits or paralysis, these patient show trunk and pelvis sway and difficulty maintaining balance during sitting, standing, and walking. Difficulty with muscle and body coordination forces the patient groups to rely on cane, walkers, and wheelchair, while often being strapped for safety. This leads to a diminished workspace, or movement area for the patient to perform daily tasks. Training that includes muscle coordination, muscle strengthening, and proprioceptive enhancement, may allow patients to improve their tasks performance [16]. While current rehab strategies rely on manual techniques to provide these modalities where the training is limited to a single component at a given time, robotic platforms permit the integration of multiple modes of therapeutic training at once.

When experiencing new tasks or postures, humans have to consciously synchronize muscle activations and body movements. If a new task is not practiced, it cannot be experienced, adapted to, or learned. In this dissertation, we propose a new robotic intervention for posture rehabilitation and training. The goal of this system and its associated algorithms is to allow postural exploration during functional tasks. The device provides user specific parameters, real-time assist-as-needed forces when a person is detected to be unstable, and haptic feedback. Through its assistive nature, the device allows for increased time when completing a challenging new task. We believe increased time for completing a new task allows for more time for motor planning and for self-promoted movement, muscle activations, and kinematic coordination. In the following sections, we discuss the device, healthy adults, and patient experiments to validate our hypothesis.

Chapter 2

2 Trunk Support Trainer (TruST) for Seated Posture Training

In this chapter we discuss the development of a novel cable-driven robotic platform, The Trunk Support Trainer (TruST), for posture training. Here we modify a similar system, the Active Tethered Pelvic Assist Device (A-TPAD) developed in the Robotics and Rehabilitation Laboratory (RoAR Labs) at Columbia University [17]. We discuss new force-field algorithms which enable training trunk posture using an assist-as-needed strategy. The algorithms allow for real-time detection of human motion while providing assistance unique to the need of each person in the device.

TruST is a cable driven robotic system for posture training. The TruST can apply forces and moments on its end-effector, which in this case is a soft belt on the trunk. Benefits of this system include low inertia on the human body and modular control over the desired forces and moments. This is particularly beneficial for low force applications where the force applied on the body is a small body weight percentage, such as on children populations. This is primarily due to the motor torque, low strength and light weight cables, and low system inertia. The low inertia allows free movement, without altering human performance, while allowing haptic feedback for assistance and perturbations. Another benefit of this system is that it does not require rigid links or fitting on the human body. This allows for quick fitting on a wide range of patients of different sizes. Since cables only require routing to an end-effector, the only adjustments required is the fitting of the belt and pulley positions. Since the belt is size adjustable and deformable, it can be fitted onto a wide range of human trunk sizes. It can also easily be modified for seated or standing applications by changing the pulley heights, at the cable exit points off the pulley. This allows control of cable vectors required for achieving a desired force. More on the details will be explained in this chapter. In addition, the software control allows quick modification of force applications. The force magnitude can be adjusted by entering the weight and body weight percent assistance requires for testing.

The specific setup of the TruST incorporates the use of four cables, each attached to a corner of the end effector or belt on the human trunk. This allows for direct control of planar forces and moment in the axial direction. In cable robotics, controlling certain number of degrees (n) of the end effector requires n+1 number of cables [17]. With four cables, three degrees at the end-effector can be controlled. The primary purpose of the TruST is to detect the position of the trunk during various reach and training activities and to provide assistance when a person is outside of their stability region. This is because outside of a person's stability, they experience postural failure where the trunk collapses over the pelvis. At this point, stability is lost and a task is not adequately experienced or completed. Therefore, a planar assistance towards the boundary center is provided. This assistance is a planar force and requires two degrees of control, in the antero-posterior and lateral directions. The TruST allows a minimal form of intervention, while providing real-time forces and objective outputs. A minimal form of intervention is one that does not hinder the movement of the individual and minimally assists a user's desired motion when needed.

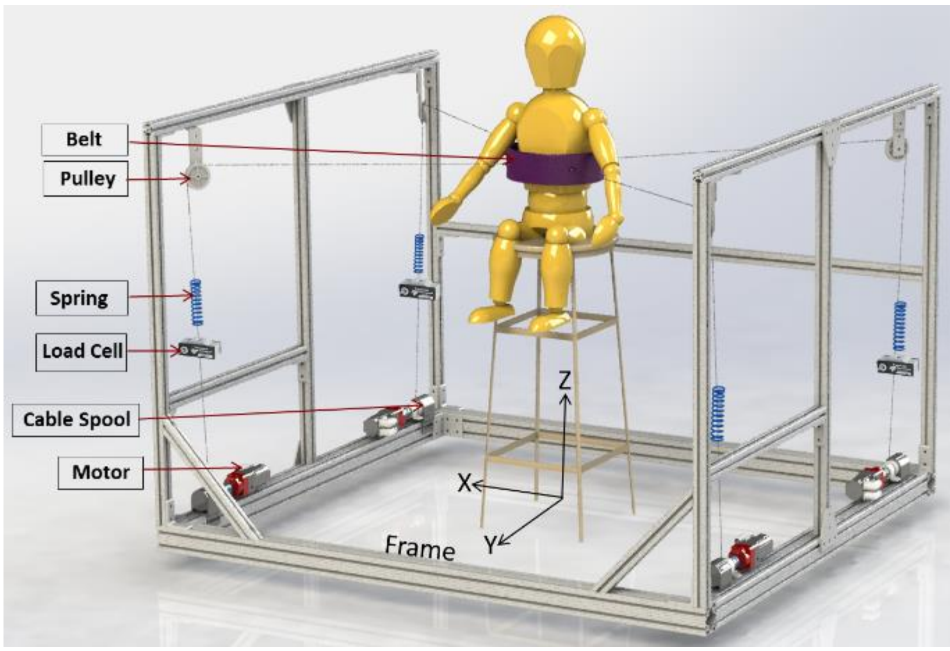


Fig. 2.1 Schematic of the TruST. Four cables are attached to a torso belt, along the transverse plane. Four motors are mounted on a stationary frame, while a spring and load cell is attached in series with the cable. A global reference frame is set at the middle of the sitting platform.

2.1 Mechanical Design

TruST utilizes four cables attached to each corner of an adaptable but rigid torso belt, while the other ends of the cables are connected to AC servo motors attached to a fixed frame. The cable attachment points on the belt are reinforced with thermoplastic to eliminate belt deformation during force application. The motors have encoders (AKM series motors and AKD drivers from Kollmorgen, Pennsylvania). A tension sensor (MLP-200 Transducer Techniques, California) and a spring (Stiffness 2.5N/mm) are attached in series to each cable to measure the tension. The spring acts as a filter in two ways. It reduces motor reaction due to unwanted end-effector vibration and reduces the noise from any motor vibration onto the human body. These tension sensors record force up to 890N and are powered by a 12V DC amplifier (TMO-1 Transducer Technique, California). Pulleys are used to route cables from the motors to the torso belt along the transverse plane of the trunk. A cable spool of 5 cm diameter is attached to the end of each motor shaft to prevent the cables from wrapping over themselves. The smaller the cable spool, the large the tensions that can be achieved by the motor output. A 5 cm cable spool allows for fitting over our 2.5cm motor shaft, while allowing a desired range of cable tensions. A motion capture system (Bonita-10 series from Vicon, Denver) is used to record the cable attachment points on the belt and pulley to calculate the force directions. A two-stage control is implemented using Labview, PXI real time controller and data acquisition cards (National Instrument, Austin). An illustration of the system can be seen in Fig. 2.1.

2.2 System Model & Tension Planner

The TruST, like the TPAD, is a cable driven parallel robotic system with each end of an actuated cable attached to an end effector. Four cables are connected to each corner of the end effector or torso belt to apply the desired force/moment in the transverse plane. If tensions in the cables are represented by $\boldsymbol{T} \in \boldsymbol{R}^{\boldsymbol{4\times 1}}$, then the force-moment vector $\boldsymbol{F} \in \boldsymbol{R}^{\boldsymbol{6\times 1}}$ applied at the trunk segment can be obtained by:

$$F = AT \tag{2.1}$$

$$F = [F_x \; F_y \; F_z \; M_x \; M_y \; M_Z]^T \tag{2.2}$$

and A is a 6×4 structure matrix based on the system geometry. The matrix A can be expressed as:

$$A = \begin{bmatrix} \cdots & \vec{\boldsymbol{I}}_i & \cdots \\ \cdots & \boldsymbol{r}_i \times \vec{\boldsymbol{I}}_i & \cdots \end{bmatrix}_{6 \times m} \tag{2.3}$$

where $\vec{I}_i$ is the i^th^ unit cable length vector away from the rigid body and r_i is the vector from the center of the rigid body to the cable attachment point i on the rigid body (Fig. 2.2).

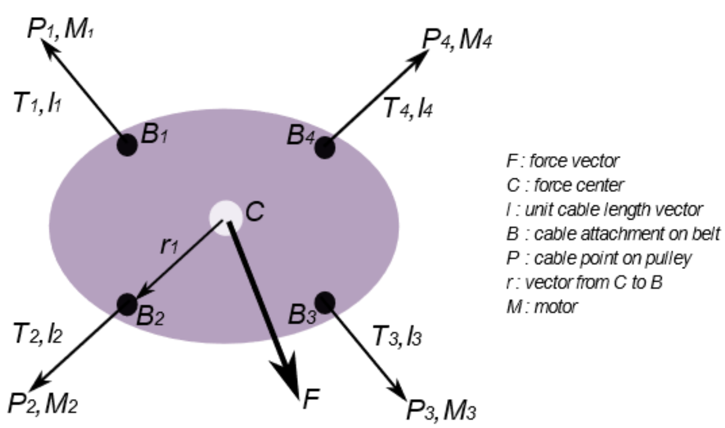


Fig. 2.2 Matrix A is determined using unit cable length I and direction vector r from the center of the rigid body inside the torso brace, to the cable attachment point on the brace. F is the desired force vector and T is the tension required to achieve F.

A quadratic programming-based optimization scheme was implemented to solve Eq. (2.1). This is to minimize discontinuities for all positive tensions T, often seen with linear programming, which finds the optimal solution at the corner of a convex hull of the feasible set [18]. The objective function minimizes deviations between T and T_p as follows.

$$\min f \tag{2.4}$$

$$f = \frac{1}{2}\left(T - T_p\right)^T\left(T - T_p\right)$$

such that,

$$F_x = F_{dx} \; ; F_y = F_{dy} \tag{2.5}$$

$$-25 < F_z < 25 \; ; -15 < M_{x,y,z} < 15$$

$$T_{min} < T < T_{max}$$

where, T_p is a positive tension value at the previous time instance, added to the objective function to ensure non-zero cable tension values and T_{min} and T_{max} are lower and upper bounds for the cable tension values. This applies to the tension in each cable. Inequalities were set to create upper and lower F_z, M_x, M_y, and M_z boundaries which allow the solver to solve for F_x and F_y forces within the inequality constraint. Since the movement of the end effector is not purely planar, the solver solves for the specified planar forces F_x and F_y, while the other DOF are constrained within the specified F_z, M_x, M_y, and M_z boundaries. These were determined through testing prior to the study.

The normal transverse force, $\boldsymbol{F_d} = [F_{\boldsymbol{dx}} \quad F_{\boldsymbol{dy}}]$, is computed by the high-level controller discussed in the next section.

2.3 Controller

The controller consists of two modules, the high-level and the low-level controller to implement a force control scheme. The high-level controller determines the desired force vector based on real time position information, then computes the desired cable tensions using the tension planner. The low-level controller achieves the desired tension using the feedforward and feedback terms. The feedforward term is determined through multiplication of a desired force in Newtons and an empirical motor constant relating the tension and voltage. The feedback terms are calculated from a proportional-integral-derivative (PID) controller. The PID is tuned using the Ziegler-Nichols method. In this, the proportional gain is increased until the motor becomes unstable and oscillates with a constant amplitude and period. Using the gain and period, a Ziegler-Nichols table is used to determine the PID gains. During testing, a command is given and the error is measured. If a steady-state error still exists, the integral gain is increased to decrease the stead-state error. If the response speed is slow or too fast, the derivative gain is adjusted. Using trial and error, the gains are found that minimize the error between the desired and actual tension values. The low-level controller runs at a 1000Hz, while the high-level operates at 200Hz.

As a general outline of the control diagram (Fig. 2.3), the high-level controller reads real-time information from the motion capture system. Based on the position of the end effector, a desired force is determined. The Cartesian force is used to plan for the appropriate cable tensions. The low-level controller sends a feed forward term to the motor, which is used to determine the motor output. Based on the errors, the feedback terms works to achieve the desired tension value precisely.

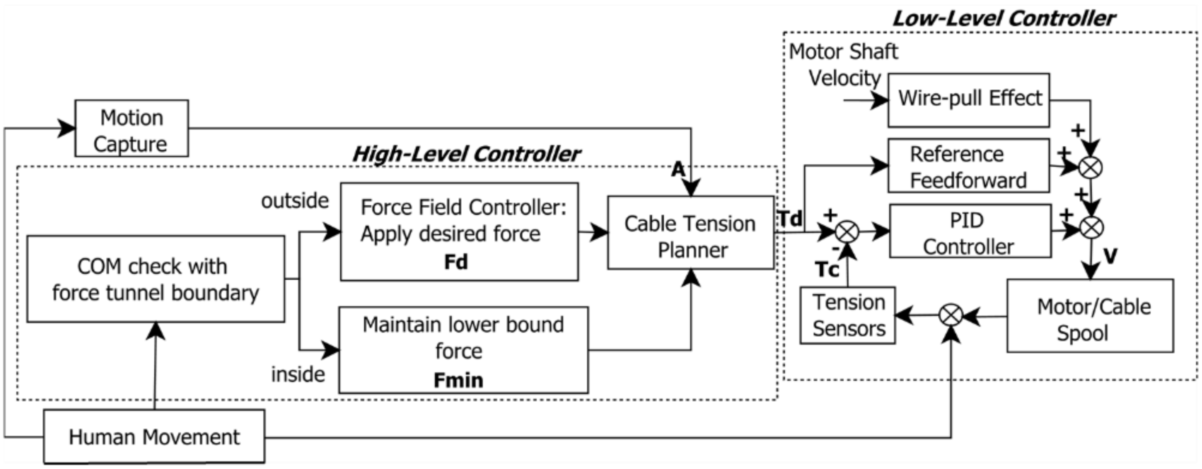


Fig. 2.3 Control system block diagram of the high and low level controllers. The high-level controller plans the desired cable tension vector T_d to apply desired force/moment F_d, once the COM is outside the boundary. The low- level controller implements these tensions using a feedback PID loop.

2.3.1 High Level Controller

As discussed previously, the high-level controller is used for tension planning. It is also used to determine the appropriate desired Cartesian force, based on real-time position data. The end effector is tracked in real-time and is used to determine the force vector. Specifically, the TruST uses force-field algorithms which are used to apply a desired force onto the human trunk.

2.3.1.1 Force-Field Algorithm

The high level controller consists of a force tunnel controller [19] which creates a virtual force field at a specified radius around the subject's trunk in the transverse plane. The center of the lower trunk is obtained by using a motion capture system and computing the estimated centroid of the left and right lateral points on the belt. The controller is designed such that when the trunk center is outside the specified circle of radius r, normal forces are applied to provide guidance to the center point back inside the circle, into a region of trunk stability (Fig. 2.4).

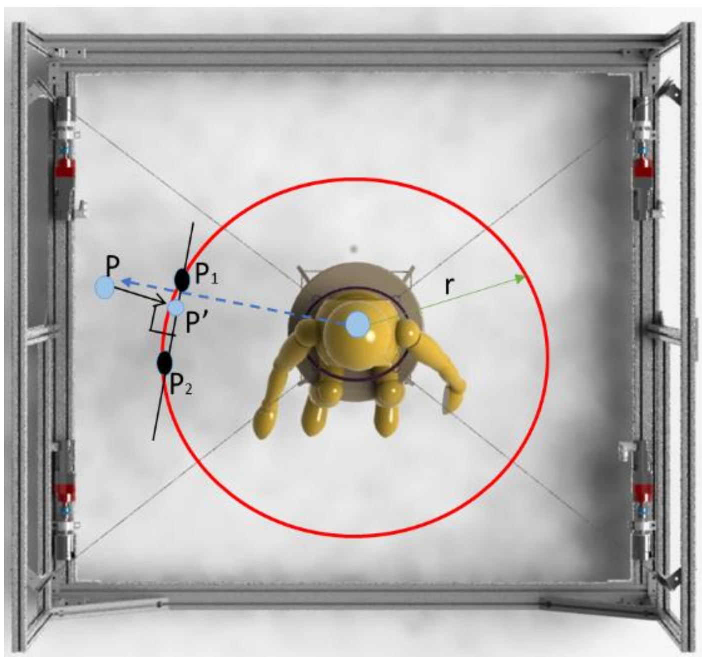


Fig 2.4 The circular force-field is applied around the lower trunk. When the center of the trunk P is outside the boundary, a normal force is calculated and applied based on neighboring points P_1 and P_2.

Let $\boldsymbol{P}$ be the current position of the trunk centroid, $\boldsymbol{P_1}$ and $\boldsymbol{P_2}$ be the two closest predefined points to $\boldsymbol{P}$ along the circular force tunnel with radius r. $\boldsymbol{P'}$ be the point perpendicular to $\boldsymbol{P}$ on the line between $\boldsymbol{P_1}$ and $\boldsymbol{P_2}$, with the force magnitude k.

$$\boldsymbol{F} = \boldsymbol{F_n} = k(\vec{\boldsymbol{n}}) \tag{2.6}$$

where the direction of the force, $\vec{\boldsymbol{n}}$ is found by:

$$\vec{\boldsymbol{n}} = \frac{(\boldsymbol{P}' - \boldsymbol{P})}{||\boldsymbol{P}' - \boldsymbol{P}||} \tag{2.7}$$

and the perpendicular point ***P'*** as,

$$(\boldsymbol{P_1} - \boldsymbol{P_2}) \cdot (\boldsymbol{P} - \boldsymbol{P}') = 0 \tag{2.8}$$

$$\boldsymbol{P}' = (1 - m)\boldsymbol{P_1} + m\boldsymbol{P_2} \tag{2.9}$$

$$m = \frac{(\boldsymbol{P_1} - \boldsymbol{P_2}) \cdot (\boldsymbol{P_1} - \boldsymbol{P})}{\|(\boldsymbol{P_1} - \boldsymbol{P_2})\|^2} \tag{2.10}$$

The force constant *k* was found by trial and error prior to the experiment. During the study, it varied between training blocks, decreasing from 60N (block 1) to 52N (block 5). This was chosen through trial and error prior to testing to ensure adequate assistive force, with a noticeable decrease through training blocks. In subsequent studies, we determined k as a percentage of body weight.

2.3.2 Low Level Controller

The low-level controller runs at 1k Hz using Labview PXI real time controller. An open loop feed forward term T_{FF} and a closed loop PID based feedback term T_{FB} are implemented to achieve the desired tension. The voltage provided to the motors yields

$$V = K_M(T_{FF} + T_{FB}) \tag{2.11}$$

where K_M is a pre-measured motor constant which is obtained from a linear relation between voltage and desired tension for each motor.

For our cable-driven systems, to keep tension in the cable $(T_{FF} + T_{FB}) > 0$, the lower bound of the feedback term is set as $T_{FB,low} = -T_{FF}$. However, if the subject pulls the cable away from the motor, extra negative input is required to compensate motor friction and unspool the cable reel. Only for this pulling case, the controller decreases the lower bound of the feedback term respect to the speed of the cable reel.

2.4 System Validation

A representation of the desired and actual Cartesian force can be seen in Fig. 2.5. The PID controller was able to follow the desired force trajectory with a root mean square error of 6.89N (3.6%) with standard deviation of 6.38N. The min error was $6.61\ X\ 10^{-5} N$ and max error 40.64N while any error above 10N lasted for a maximum of 0.135s, due to a small delay in real time response.

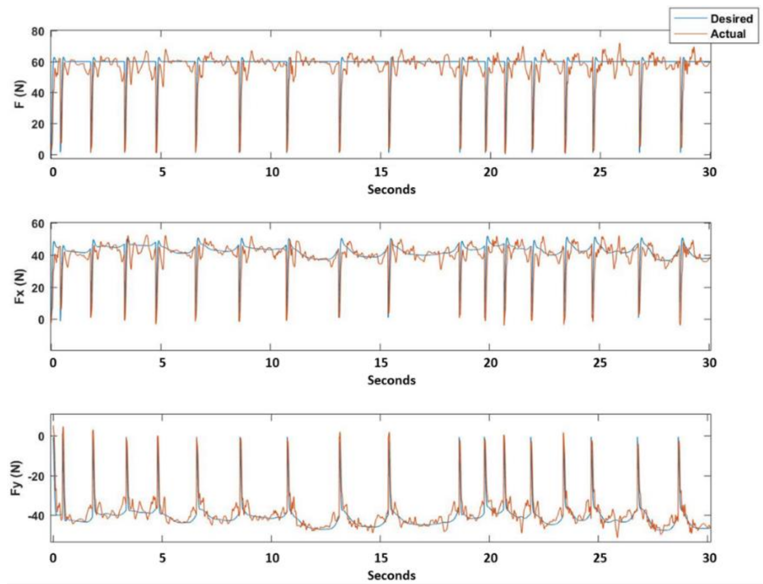


Fig. 2.5 The desired and actual Cartesian force vectors Fx, Fy, and resultant F demonstrate the capability of the low level controller to achieve the desired trajectory during dynamic movement.

2.5 Conclusion

The design of the TruST allows for control of planar forces at the end-effector. It utilizes four cables attached from an end-effector belt onto a pulley at each corner of a rectangular frame. The device utilizes a force-field algorithm, which applies real-time assistance to the human trunk, based on its position. The system is cable of generative over 800N of force in each cable. The control strategy applies a body-weight percentage assistance on the human trunk, once the trunk is detected to be outside of a specified boundary.

Of the unique capabilities of the system is its modular and user-specific force field control algorithm. This force-field shape and size is determined prior to the use of the device. The shape encompasses the unique trunk workspace of each individual. Thus, it allows full independent posture control by the user at the end-effector, while inside this force-field boundary. Once outside, a body-weight percentage assistance is provided to help the user maintain stability for a longer period of time and to experience new postures. This ability allows us to test new training modalities and answer scientific questions on the ability of patient groups to learn new tasks and trunk postures. The following chapters show case the studies and the new findings on trunk control.

Chapter 3

3 Seated Human Experiments with TruST

In this chapter, we present the protocol and results from training experiments with the TruST. These experiments were conducted on healthy and cerebral palsy patients in the seated position. The TruST was used to provided trunk assistance at each individual's stability boundary. The studies were first conducted on healthy individuals to document the changes in human performance, followed by a similar test in a virtual environment. Finally, we conduct a longitudinal study on children with cerebral palsy.

3.1 Kinematic Changes During Seated Training

Twenty healthy subjects (20-30yrs, 12 males, 8 females, 19 right-handed, and 1 left-handed) were recruited and randomly assigned to either the experimental or control group. The training protocol was approved by Columbia University's institutional review board and consisted of three stages: baseline (BL/pre-training), training (T), and post-training (PT). Before training, subjects were instructed about the training protocol, but were not told whether they were in the control or experimental groups. Retro-reflective markers were placed on the subject to record kinematics using a Vicon motion capture system. The subjects were seated on a stool at the center of the TruST (Fig. 3.1). An adaptable but rigid three-inch wide belt was attached to the lower trunk (lumbar) region. During the BL, the pre-training tasks consisted of the functional reach test to determine the maximum lower trunk displacement and define the point of stability failure, and a pre-training nine-hole peg task without assistive forces. During the T stage, five blocks of two consecutive trials of the nine-hole peg task was conducted (massed practice). This was followed by PT stage in which both, functional reach and nine-hole peg tasks, were re-assessed after removing the external assistance from TruST.

Fig. 3.1 Nine-hole peg test and the modified functional reach test, respectively, performed during the human experiment.

During the experiment, all subjects were asked to sit on a flat, wooden stool. The torso belt was firmly placed at the lower trunk. The subjects were asked not to use any foot or hand support while performing the functional reach task and the nine-hole peg task. However, they were allowed to move their body to complete all tasks to the best of their ability. All tasks started from a stable neutral position with the head and trunk centered over the pelvis, with elbows in external rotation and bent 90-degrees in the air. Subjects were instructed to perform each task as fast and accurately as possible, while maintaining postural control. During training, subjects were allowed to use a finger or the volar area of the wrist for support on the table, only if posture stability was lost during the placement of the pegs.

For the functional reach test, the subjects were asked to displace a wooden block anteriorly as far as they could in a controlled and self-paced manner. If the subject used any support or lost balance, the task was stopped and the point of stability failure was kinematically recorded. If the subject lost balance prematurely, they were allowed to repeat the task. We defined premature loss of balance if 1) the subject touched the table surface for support, or 2) there was premature foot-ground contact for displacing the piece of wood at further distance.

The reach test was performed at BL and PT, with the shoulders flexed at 90-degrees and arm parallel to the floor. The failure point was used to identify the boundary between postural stability and instability. This boundary specified the maximum anterior translation of the lower trunk before postural collapse.

The nine-hole peg board (3 x 3 holes of 4 mm diameter) was then placed in front of the subject, with the furthest row being in line with the position of the wooden block at the time of stability failure. The subjects were instructed to grab a peg from their dominant side and to place it onto the board from right to left, working from the closest row to the furthest. After inserting the nine pegs, subjects removed these in the same order. A complete cycle of inserting and removing pegs was identified as a single trial. Two consecutive trials conducted at a time were defined as a block.

Five blocks were conducted, with the assist-as-needed, error-based force for the experimental group at or beyond the predefined maximum lower trunk displacement (e.g. point of stability failure). Accordingly, the subjects moved independent of any assistance as long as they were inside the force tunnel but received assistance at and beyond their failure point. The assistive force was decreased by 2N (3.33%) after each block of training. The subjects were allowed as much rest time as they needed between sessions to a maximum of five minutes. The same protocol was followed for the control group but no assistive forces were provided.

3.1.1 Data Analysis

The data were analyzed to assess the spatiotemporal changes in head, upper trunk, lower trunk, and pelvis translation and rotation between pre and post functional reach test. As exploratory performance measures, we analyzed the percentage of time the experimental subjects required assistive forces provided by TruST and postural behavior of subjects during the T stage. This last variable was assessed as the number of times the subjects would require 1) foot/reaching support due to postural collapse or 2) hand-table support for more than 1000ms due to lack of upper body stability and/or inability to recover postural verticality.

The data were analyzed using Matlab (MathWorks, Natick). The COM of the lower trunk was estimated using right and left belt markers. Translation of this trunk segment was measured in the anterior-posterior direction, from start of the trial (neutral position) to the point of stability failure. All the subjects were video-recorded from a 45-degree angle. Datavyu software (http://datavyu.org/) was used to characterize postural behavior. The statistical analysis was conducted using SPSS 23 (SPSS, Chicago, Illinois). We conducted a two-factor mixed-design ANOVA with one within factor (Test Session: Pre-Training and Post-training) and one between factor (Group: Experimental and Control). Pairwise comparisons are reported if the interaction was significant and p values were adjusted using Bonferroni's procedure.

Postural behavior variables were not normally distributed, as it was indicated by the non-significant Shapiro-Wilk normality test ($p < 0.05$) and visual inspection of Q-Q plots. Therefore, a non-parametric Mann-Whitney U-test was applied to test the potential significant difference between control and experimental subjects in postural behavior during the T stage. The alpha value was set at 0.05 for both statistical procedures.

3.1.2 Results

The anterior COM lower trunk displacement (e.g. position of belt) during the pre and post-training stages is depicted in Fig. 3.2. The two groups started from a similar baseline in COM displacement (6.2 cm and 7.0 cm, p=0.368). There was a significant Test Session X Group interaction (F=11.33(1,18), p=0.003, η^2=0.89), with the pairwise comparison showing a significant increase of 61.4% in experimental subjects (p<0.001) compared to an increase of 14% in the case of controls (p>0.05) during the post-training.

The changes in head, upper trunk, lower trunk, and pelvis rotations are shown in Fig. 3.3 – 3.7. The pelvis rotation was measured relative to the global frame, while each segment was measured in relation to its caudal segment (e.g. the upper trunk relative to the lower trunk). There was a significant Test Session X Group interaction for lower trunk (F=5.32(1,18), p=0.033, η^2=0.59) and pelvic rotation (F=8.01(1,18), p=0.011, η^2=0.76) in the flexion/extension axes. These results indicate a significant increase in the rotatory component of both lower thorax and pelvis only for the experimental subjects that received the training with the TruST. Specifically, after training, lower trunk rotation increased towards flexion by 41.3% (p=0.028) and pelvic rotation increased towards extension by 81.0% (p<0.001) for the experimental group, while there was no significant change in the control group (p>0.05).

The number of reaching or foot placing reactions for postural support was not statistically different between the experimental and control groups (U = 26.5, $p = 0.06$). Nonetheless, controls (*Mdn*: 4, *Min*: 1, *Max:* 22) showed significantly greater number of hand contacts with the table for either postural support or recovering verticality (U = 23.5, $p = 0.04$) compared to experimental subjects (*Mdn*: 2, *Min*: 0, *Max: 8*). We should note that TruST was active and providing force assistance, on average, 97% of the total time of the seated reaching training across the 5 blocks of trials in the experimental group.

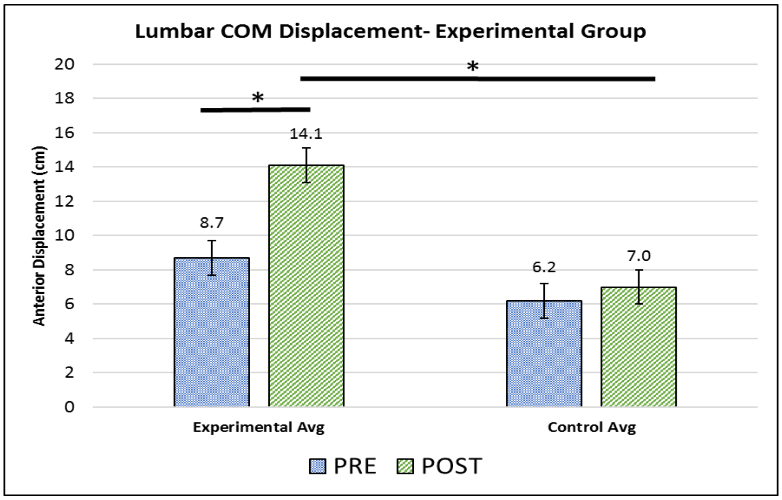


Fig. 3.2 Lower trunk COM displacement, measured from the stable/neutral position to the failure point during the functional reaching task, pre and post training for the experimental and control group (* = p < 0.001).

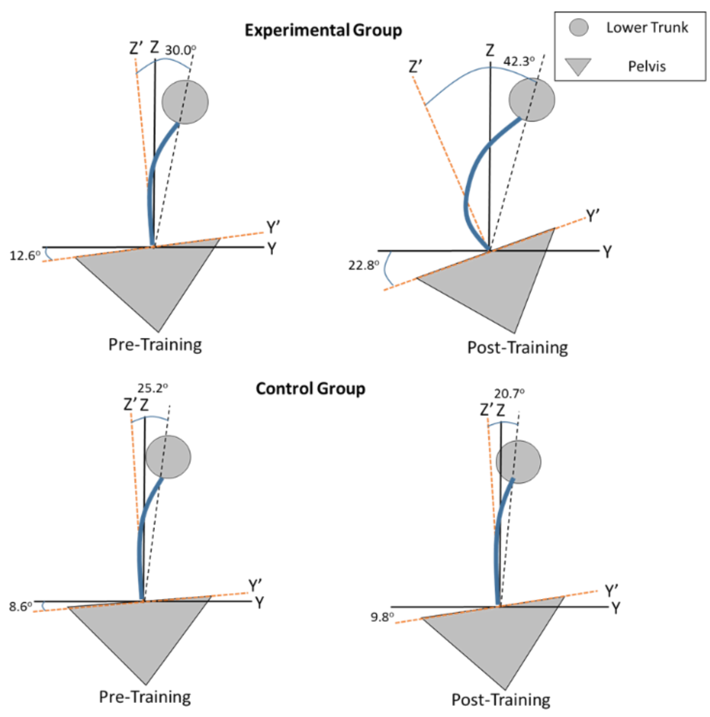


Fig. 3.3 Lower trunk and pelvis schematic, showing the average rotation in the anterior/posterior direction, pre and post training for both the experimental and control group.

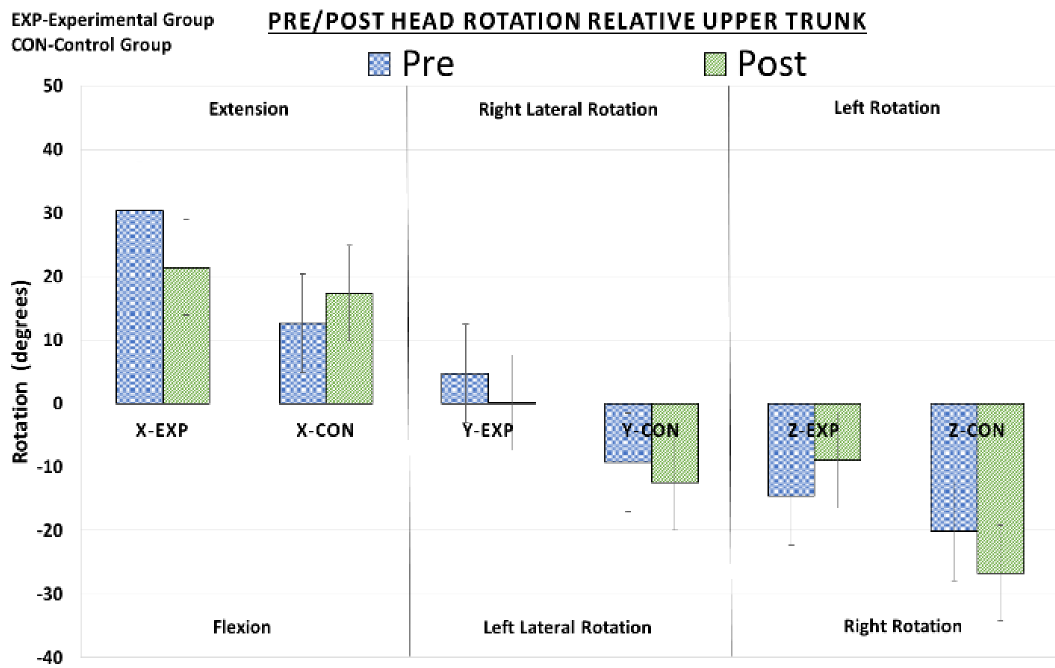


Fig. 3.4 Pre and post head rotation relative to the upper trunk, for the experimental and control group (in degrees).

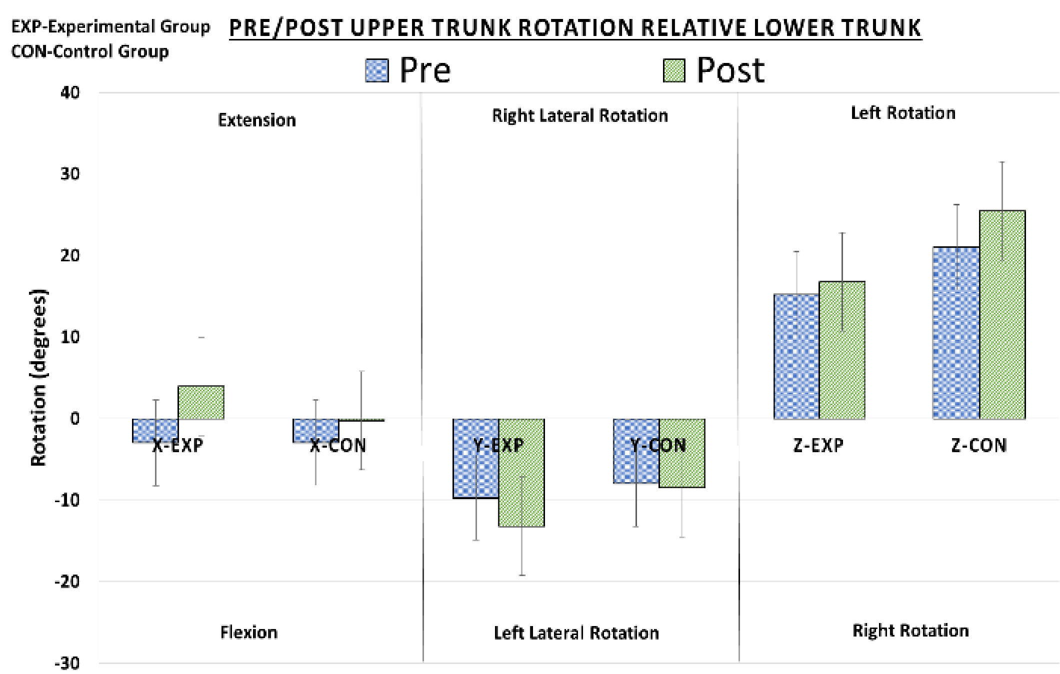


Fig. 3.5 Pre and post upper trunk rotation relative to the lower trunk, for the experimental and control group (in degrees).

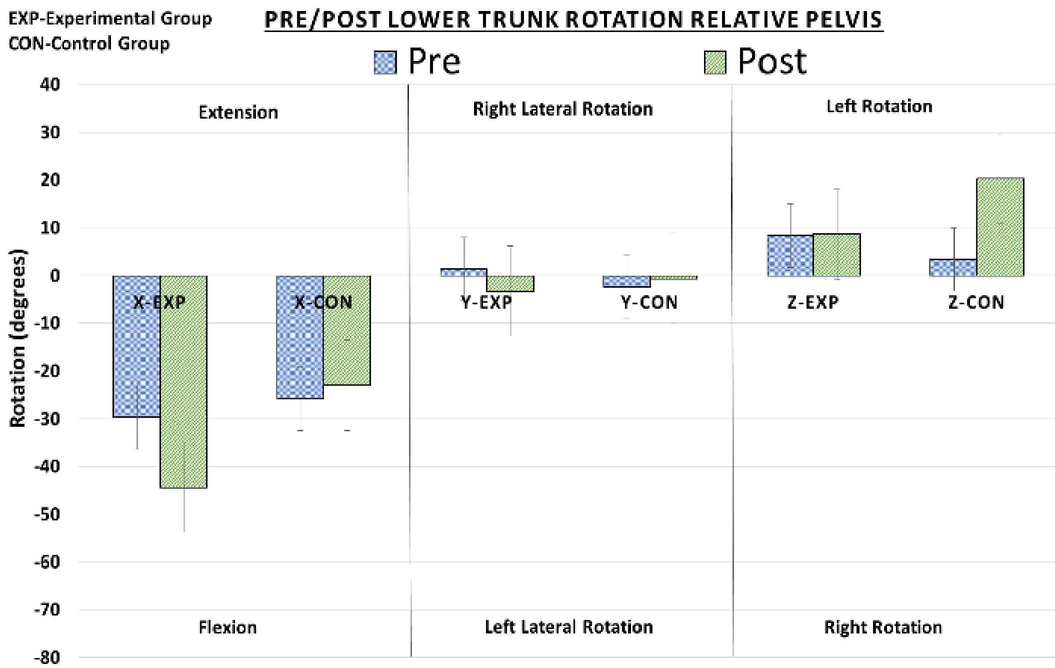


Fig. 3.6 Pre and post lower trunk rotation relative to the pelvis, for the experimental and control group (in degrees).

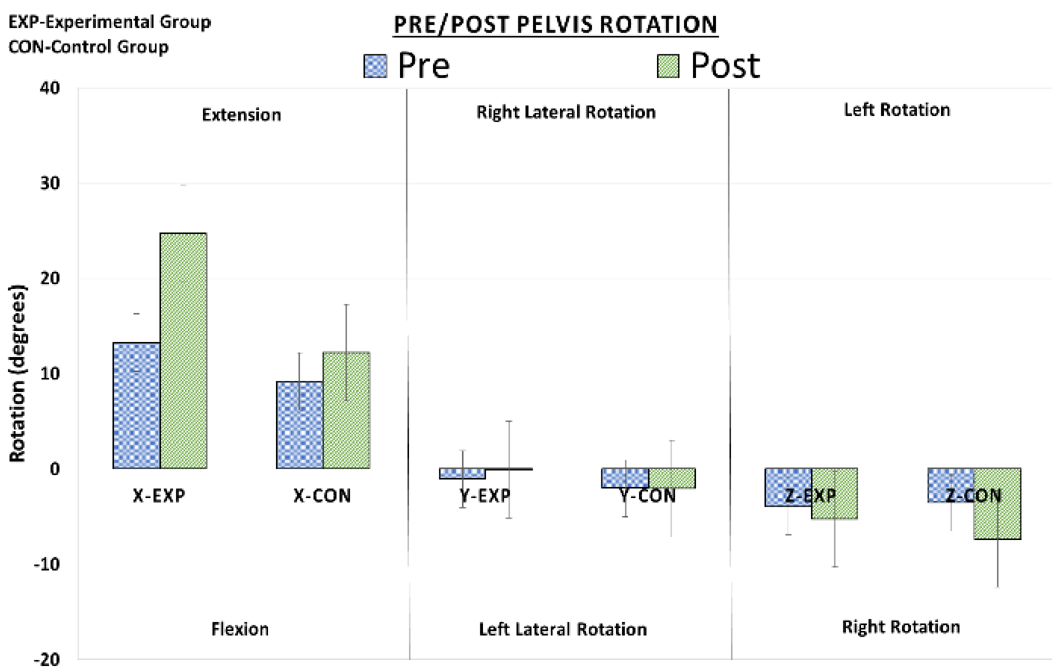


Fig. 3.7 Pre and post pelvis rotation relative to the global frame, for the experimental and control group (in degrees).

3.1.3 Discussion

In this proof-of-concept study, we tested if specific practice of a seated reaching task at and beyond the individual's point of postural stability failure, without foot support, could enhance volitional control of upper body and further the stability limits established by the configuration of the pelvis. Experimental results demonstrated that over a single training session, on average, the subjects were able to significantly translate their body further anteriorly from their neutral postural configuration and significantly increase the rotational profiles of the lower trunk and pelvic segments in the flexion-extension plane of motion following the assist-as-needed force training with the TruST. Additionally, the subjects of the experimental group did not require frequent and long-lasting hand contact with the table during the training phase.

As was seen with the control group, it was difficult for subjects to train at the border of stability failure without any external assistive forces. As subjects completed each trial of the nine-hole peg test, it was difficult for them to maintain dynamic postural stability with the motion of the arm during the reaching task at or beyond their stability boundaries without using their hand or wrist for support on the peg board. It is possible that without assistive forces, individuals often fail at their stability limits and are not able to explore a larger range of motion and a stable postural configuration required to maintain or recover seated balance while actively reaching to insert and remove the pegs. Therefore, the sensorimotor postural experience required for adjusting their postural kinematics to complete the task successfully may be diminished compared to the experimental group who were provided with the assist-as-needed force from TruST. It was seen that the experimental group spent on average 97% of their training task performance utilizing the assistive forces. This exploratory outcome suggests that reaching further past the maximum arm's extension and at and beyond the individual's point of postural failure requires continuous volitional shifting of the lower trunk to avoid falling or using hands for body support.

Nikolai Bernstein's theory of motor learning states that the central nervous system's (CNS) hierarchy of control mechanisms for posture and movement are organized with distributed and parallel processing [20]–[23]. Neural mechanisms that integrate posture with dexterity movements (e.g. reaching control) are distributed in the CNS and are recruited in patterns that are task and context dependent [20], [23]. The manner in which the CNS recruits the DOF depends on the variables it uses to plan, time, and control the movement [23]. By providing an assistive force at an individual's point of postural stability failure, the subjects experienced larger upper body displacements with the subsequent selective training in the control of key segments for maintaining postural stability during the reaching task practice. These segments were lower trunk and pelvis. Our results demonstrate that the control group was not able to enhance the degree of selective rotations of the lower segments (lumbar and pelvis) and these remain similar before and after the training. The movement of these lower segments followed an *'in-block'* mode of control, in which the DOF of these key segments are constrained during the linear translation of the upper body during the functional reach task at and beyond the postural limits. However, in the case of the experimental group, the subjects were able to release the lower thorax and pelvic DOF and exploit their rotational amplitude in the flexion-extension plane of motion after providing the assist-as-needed force control with the TruST. Thus, subjects in the experimental group were able to experience and control a set of postural kinematics that would have been impossible to undergo without the selective force field provided by the TruST.

The overall goal of this assist-as-needed control strategy via the robotic device TruST is to train subjects in an accurate reaching task beyond maximum arm extensions and at/beyond their point of postural stability failure. With TruST, the volitional neuromuscular control of lower thorax and pelvis is continuous, allowing them to actively participate in the postural learning process required to attain this specific reaching task context. The effect of training and lack of subject-TruST inter-dependence is observed because the assistive force was progressively reduced across the five blocks of practice. In addition to this, the TruST was inactive during the post-assessment and yet significant improvements were observed for the translational and rotational components in the voluntary control of seated posture beyond the point of stability failure.

The application of this device in combination with motor learning and control principles can be seen as potentially beneficial for use in patients with neurological or musculoskeletal disorders, such as in Cerebral Palsy or Spinal Cord Injury. In these pathological conditions, patients have little or no voluntary seated postural stability in both static and dynamic dimensions of control.

We cannot disregard other possible factors such as the level of fatigability of the paravertebral muscles, which could be reduced in experimental subjects compared to controls. Some of the technical limitations of this study may include the ability to distinguish between the changes in muscle activation, joint torques, and kinematics that lead to increased postural stability beyond the point of failure. Although posture is a combination of several intricate biomechanical processes, it would be beneficial to include these parameters in future experiments in order to identify which parameters are likely to play a larger role for adapting a more stable posture.

3.1.4 Conclusion

In this novel proof-of-concept study, we presented the first state-of-the-art, active posture training device, TruST. We also demonstrated benefits of training subjects at and beyond their point of stability failure for enhancing their maximum volitional control in seated trunk displacement with and without assist-as-needed forces. We defined postural stability as the ability to displace the estimated COM of the lower trunk segment further from the neutral postural configuration of reference without postural collapse of the upper body. The results demonstrate that the force field concept used in TruST has potential benefits in the rehabilitation of posture. Training with assistive forces at a person's point of postural stability failure can increase the rotational amplitude of pelvis and lower thorax in order to displace the upper body under volitional control further away from the center of the pelvic configuration. These outcomes are in accordance with: (a) the principle of practice specificity in motor learning when providing appropriate assistive forces with the TruST and (b) Bernstein's theory of motor learning, where the CNS can recruit and release the DOF to organize postural kinematics to attain the goal of reaching within and beyond the individual's point of stability failure. In future studies, we plan to test our hypothesis applying the TruST in patient populations characterized by trunk instability.

3.2 Translation of Training to a Virtual Environment

Virtual Reality (VR) offers a three dimensional user interface along with real-time computer simulation of an interactive environment [24], [25]. It can present complex multimodal sensory information to the user and can elicit the feeling of realness and engagement [25]. The capability of VR can potentially offer many advantages in rehabilitation. Although physical reality (PR) training is the traditional and proven method towards rehabilitation, the successful integration of VR can allow access to unexplored rehab paradigms, and provide a quantitative and qualitative comparison of VR and PR based methods.

An advantage of VR over PR training activities of daily living (ADL) during rehabilitation is the added layer of cognitive engagement due to gaming aspect [26], [27]. With specific gaming scenarios that require a specific movement, VR based systems can allow exploration of how the brain controls movement, learns new movements, and relearns movement skills after an injury [28]. VR also provides the ability to vary a training task in a small space, without having to physically alter the environment. Being able to train on a variety of different tasks can provide a better overall improvement in function than repetition of the same task [28]. Specifically, VR has potential in rehabilitation of patients with neurological deficits, where the cost of treatment is high, therapist time is limited, and repetitive training is shown to produce positive outcomes [27], [28]. The use of VR systems, integrated with novel robotics, paves the way for testing a larger range of patient training paradigms and enhancing scientific exploration. In return, use of such systems can reduce the cost of rehabilitation, allow therapists to be more productive in their training, and allow extended duration of rehabilitation [28]. VR can serve as a tool for sensory stimulation for neural and functional recovery by providing movement observation, imagery, repetitive massed practice, and imitation therapy [25].

Research also supports “task oriented training” for rehabilitation, where the motions relevant to a certain activity of daily living are part of the rehab training [28]. Studies have shown that quantity, duration, and intensity are important variables in learning and relearning motor skills and in changing neural architectures [25]. Evidence demonstrates that plasticity is use dependent and intensive massed and repeated practice may be necessary to modify neural organization [25]. Using VR and robotics, various areas of motor rehabilitation can be studied. One such area is seated postural control. Seated postural control requires the stabilization and control of the head and trunk to complete many daily tasks, e.g., reaching for an object beyond maximum arm's length [29]. Although sitting statically upright can be less demanding with the center of mass (COM) of the upper body centered over the base of support (BOS), large and quick displacements of the upper body outside the pelvic boundaries can cause a sudden shift of the COM away from the center of the BOS, creating a lack of a stable neutral upper body and pelvic position [29]. In this situation, a precise control of the upper body is required to recover verticality and maintain stability. In this work, we created a VR environment in correlation to a PR environment, and synchronized it with the TruST to compare the benefits of postural training using VR with TruST to PR with TruST and PR without TruST assistance (control group). In addition, assistive forces provide error-based haptic feedback for proprioceptive training. The TruST allows for self-control of postural adjustments during training.

In this work, we propose the use of VR based training tasks with TruST for improving volitional trunk displacement for direction specific kinematic adaptation. Our training activities consist of novel challenging postural tasks conducted without foot support, to challenge postural balance and coordination. We demonstrate our methods by training ten healthy adult subjects at their maximum trunk displacement, or failure point, in a VR environment and then compare with results in a real environment with physical objects [29], with and without the assist-as-needed forces of the TruST. A motion capture system was used to determine the subject's maximum seated lower trunk COM displacement and a force tunnel was created at that distance. During the experiment, subjects performed a total of five blocks of two trials each. For PR, this consisted of a nine-hole peg test placed at the subject's point of stability failure, measured during the baseline stage, while for VR, this consisted of a virtual nine-coin collection with coins placed at the same measured point of stability failure. Our study tests and supports the hypothesis that a single training session in a VR environment, at a maximum stability region, increases lower trunk COM displacement and shows no significant difference to training in a real environment with physical object manipulation.

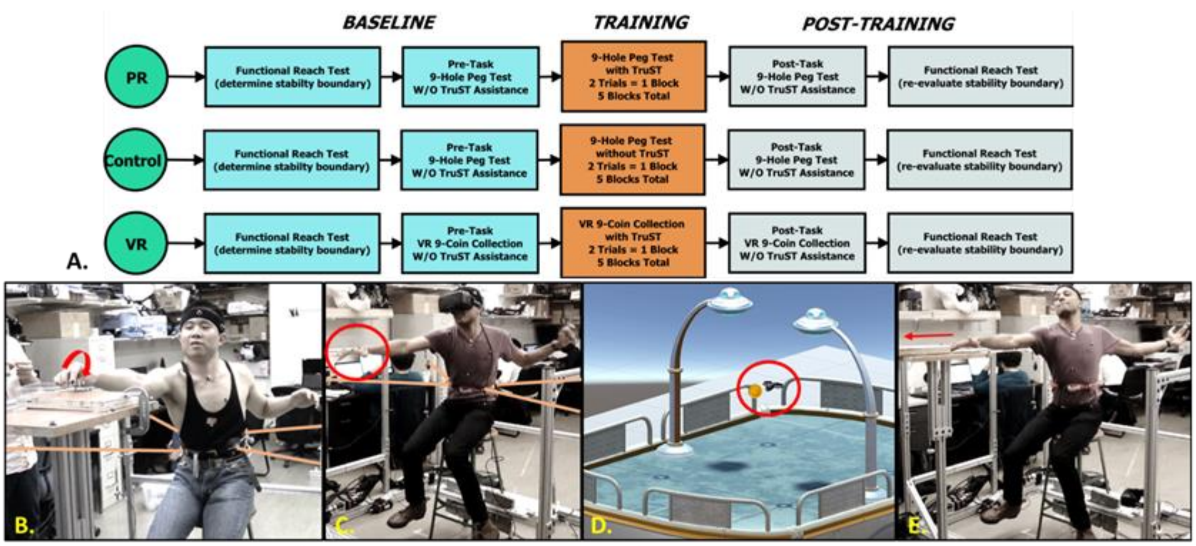


Fig. 3.8 A) The study protocol for the different study groups. B) PR task consisting of nine-hole peg test. C)VR task consisting of nine-coin collection task. D) VR environment experienced during VR testing. E) Modified functional reach test to determine maximum region of stability and force field boundary.

3.2.1 Protocol

Ten healthy adult subjects were recruited for the VR training group. They were compared to 20 healthy adult subjects, who were randomly assigned to either the physical environment/reality training (PR) or control group (PR with no TruST support). The VR group consisted of seven males and three females (Avg: 25.7yrs/155lbs/174cm, 9 right-handed and 1 left-handed). The PR and control group consisted of 12 males and eight females (20-30yrs, 19 right-handed and 1 left-handed). The training protocol was approved by Columbia University's institutional review board and consisted of three stages: baseline (BL/pre-training), training (T), and post-training (PT), as outlined in Fig. 3.8. Before training, subjects were instructed about the training protocol. The PR and control groups were randomly assigned and were not told which group they were in. The VR group was aware of their group as they had to wear a VR headset. Retro-reflective markers were placed on the subjects to record kinematics using a Vicon motion capture system. The subjects were seated on a stool at the center of the TruST. An adaptable but rigid three-inch wide belt was attached to the lower trunk (lumbar) region.

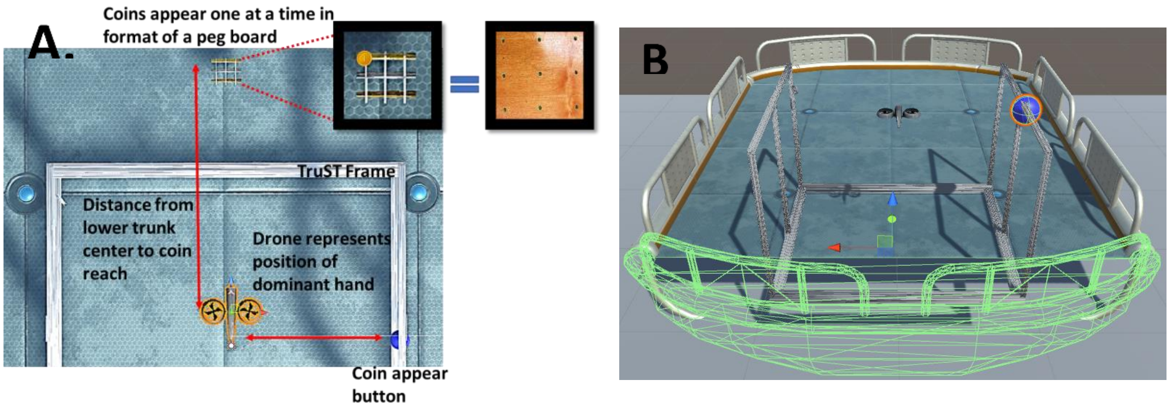


Fig. 3.9 A) Coins appear at the distance of the maximum BL reach. The coins appear in the formation as the 3x3 peg board used in PR. Coins appear one at a time, after a button placed at the dominant side, is touched. B) Overlay of the TruST frame and VR environment demonstrates the synchronization between real spatial frame parameters and correlation within VR. The origin of the global coordinate system is linked between the VR and PR using a motion capture system. The appearance of the blue ball is equivalent to the position of peg retrieval prior to initiating the reach towards the peg board hole.

The training activities were separated as PR training using a nine-hole peg task (Fig. 3.8B) or VR training using a virtual coin collecting game task (Fig. 3.8C & 3.8D) structured in the same format (collecting nine coins spaced out at the same distance as the peg board). The peg board had 3 x 3 holes of 4mm diameter spaced at 5 cm apart, and the subjects were instructed to pick a peg from their dominant side and place it into the peg board. For the VR group, to simulate the same training in a virtual environment, the subjects were instructed to touch a button on their dominant side before collecting the coins spaced in a 3 x 3 configuration at 5 cm apart (Fig. 3.9).

During the BL, the pre-training tasks consisted of the functional reach test to determine the maximum lower trunk displacement and define the point of stability failure for all three groups (Fig. 3.8A). The PR and control groups conducted a pre-training nine-hole peg task without assistive forces, while the VR group conducted the virtual coin collecting game task with no assistive force. During the training stage, five blocks of two consecutive trials of the nine-hole peg task was conducted for the PR and control groups, while the VR group performed the virtual coin collecting game task. This was followed by PT stage in which the PR and control groups performed the functional reach and nine-hole peg tasks, and the VR group performed the functional reach and coin collecting task. These PT tasks were for re-assessment of postural kinematics without external assistance from TruST.

The purpose of an assist-as-needed force strategy is to minimize subject dependency on assistive forces and to motivate self-initiation and self-correction of postural kinematics. With the force field control, the subject is not administered any supportive forces within their measured stability region, as this is their pre-intervention workspace. As the subject is challenged to tasks outside their stability boundary, the assistive forces are triggered to support the posture and allow adequate time to explore further areas outside of their stability boundary. The haptic feedback provided through the assistive forces notifies the subject of their proximity to their stability boundary. As the subjects experience this new workspace and realize their ability to maintain stability, they are naturally encouraged to explore new postural configurations to successfully occupy the new workspace. In contrast to physical therapy based interventions for postural training, our protocol was designed to provide minimal required assistance, determined through system testing, to assist subjects during training tasks described in the following sections.

During the experiment, all subjects were asked to sit on a flat, wooden stool. The torso belt was firmly placed at the lower trunk. The subjects were asked *not* to use any foot or hand support while performing the functional reach task and the nine-hole peg task. However, they were allowed to move their body freely as desired, to complete all tasks to the best of their ability. All tasks started from a stable neutral position with the head and trunk centered over the pelvis, with elbows in external rotation and bent 90-degrees in the air. Subjects were instructed to perform each task as fast and accurately as possible, while maintaining postural control. During training, subjects were allowed to use a finger or the volar area of the wrist for support on the table, only if posture stability was lost during the placement of the pegs.

For the functional reach test, the subjects were asked to displace a wooden block anteriorly as far as they could in a controlled and self-paced manner. If the subject used any support or lost balance, the task was stopped and the point of stability failure was kinematically recorded. If the subject lost balance prematurely, they were allowed to repeat the task. We defined premature loss of balance if 1) the subject touched the table surface for support, or 2) there was premature foot-ground contact for displacing the piece of wood at further distance.

The reach test was performed at BL and PT, with the shoulders flexed at 90-degrees and arm parallel to the floor. The failure point was used to identify the boundary of postural stability. This boundary specified the maximum anterior translation of the lower trunk before postural collapse. This maximum anterior translation was set as the radius of the force field circle, allowing for the TruST device to provide assistive forces when outside this radius.

For the PR and control groups, the nine-hole peg board was then placed in front of the subject. The furthest row of the board was positioned in line with the position of the wooden block where the stability failure happened during BL. For the VR group, the coin position was set to the wooden block position, or the stability failure point. The subjects were instructed to grab a peg from their dominant arm and place it onto the board from the dominant side to opposite side, working from the closest row to the furthest. For VR group, the subjects were instructed to touch a button at their dominant side and then reach to grab the coins, as these appeared one at a time in a specific order, from the dominant to the opposite side and from the closest to furthest away.

The coins also ranged in color from bronze, silver, and gold, to identify from closest to furthest in distance. After inserting the nine pegs, subjects removed these in the same order, while the VR group recollected the coins in the same order. A complete cycle of inserting and removing pegs or collecting coins twice was identified as a single trial. Two consecutive trials conducted at a time were defined as a block. Five blocks were conducted, with the assist-as-needed, error based force for the VR and PR groups at or beyond the predefined maximum lower trunk displacements (e.g. point of stability failure). Accordingly, the subjects moved independent of any assistance as long as they were inside the force tunnel but received assistance at and beyond their failure point. The assistive force was decreased by 2N (3.33%) linearly after each block of training, starting from 60N, determined through system testing. The subjects were allowed as much rest time as they needed between sessions to a maximum of five minutes. The same protocol with the peg board was followed for the control group but no assistive forces were provided.

3.2.2 Data Analysis

The data were analyzed to assess the spatiotemporal changes in head, upper trunk, lower trunk, and pelvis translation and rotation between pre and post functional reach test between the VR, experimental, and control group. The data were analyzed using MATLAB (MathWorks, Natick). The COM of the lower trunk was estimated using right and left belt markers. Translation of this trunk segment was measured in the anterior-posterior direction, from start of the trial (neutral position) to the point of stability failure. The statistical analysis was conducted using SPSS 23

(SPSS, Chicago, Illinois). We conducted a two-factor mixed-design ANOVA with one within factor (Two Groups: Pre-Training and Post-Training) and one between factor (Three Groups: VR, PR, Control). Pairwise comparisons are reported if the interaction was significant and p values were adjusted using Bonferroni's procedure. The alpha value was set at 0.05 for both statistical procedures.

3.2.3 Results

The anterior COM lower trunk displacement (e.g. position of belt) during the pre and post-training stages is depicted in Fig. 3.10. There was a significant Test Session X Group interaction (F=4.47(2,27), p=0.021, η^2=0.72), with the pairwise comparison showing a significant increase of 61.4% in the PR group (p<0.001) and 34.4% in VR group (p=0.004) compared to an increase of 14% in the case of controls (p>0.05) during the post-training. Both PR and VR showed significantly more range of motion in translation compared to the control, 515% (p=0.001) and 278% (p=0.006) respectively, while there was no significant difference between the PR and VR group (p>0.05).

The changes in head, upper trunk, lower trunk, and pelvis rotations are shown in Figs. 3.11-3.12. The pelvis rotation was measured relative to the global frame, while each segment was measured in relation to its caudal segment (e.g. the upper trunk relative to the lower trunk). There was a significant Test Session X Group interaction for lower trunk (F=3.41(2,27), p=0.048, η^2=0.59) and pelvic rotation (F=5.37(2,27), p=0.011, η^2=0.80) in the flexion/extension axes. These results indicate a significant increase in the rotatory component of both lower thorax and pelvic rotation for the PR group compared to the control, with a tendency of interaction effect for pelvic rotation (*p*=0.056) between the VR and control group.

Specifically, after training, lower trunk rotation increased towards flexion by 41.3% (*p*<0.001), while pelvic rotation increased towards extension by 81.0% (*p*<0.001) for the PR group. However, the VR group showed an increase toward flexion by 10.5% (*p*=0.516) and a pelvic rotation increase towards extension by 23.5% (*p*=0.123). There was no significant change in the control group (*p*>0.05) and no interaction effect between the PR and VR group.

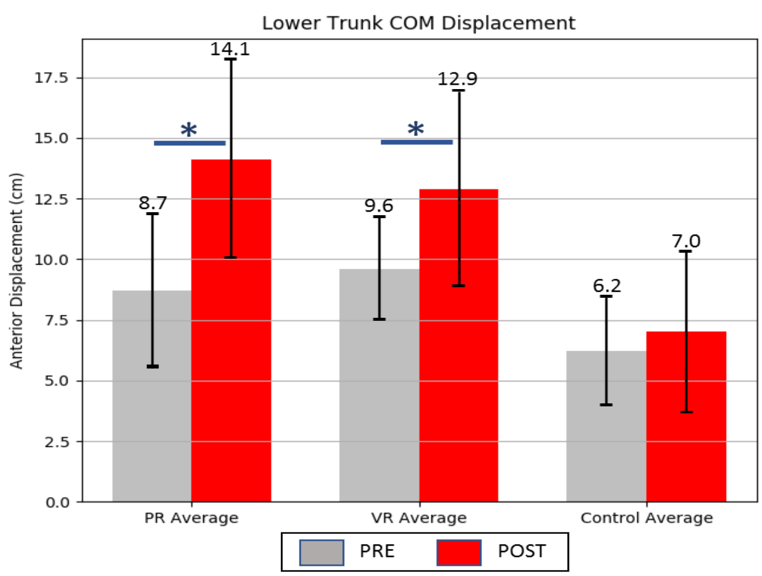


Fig. 3.10 Lower trunk COM displacement, measured from the stable/neutral position to the failure point during the functional reaching task, pre and post training for the experimental, VR, and control group (* = p < 0.001).

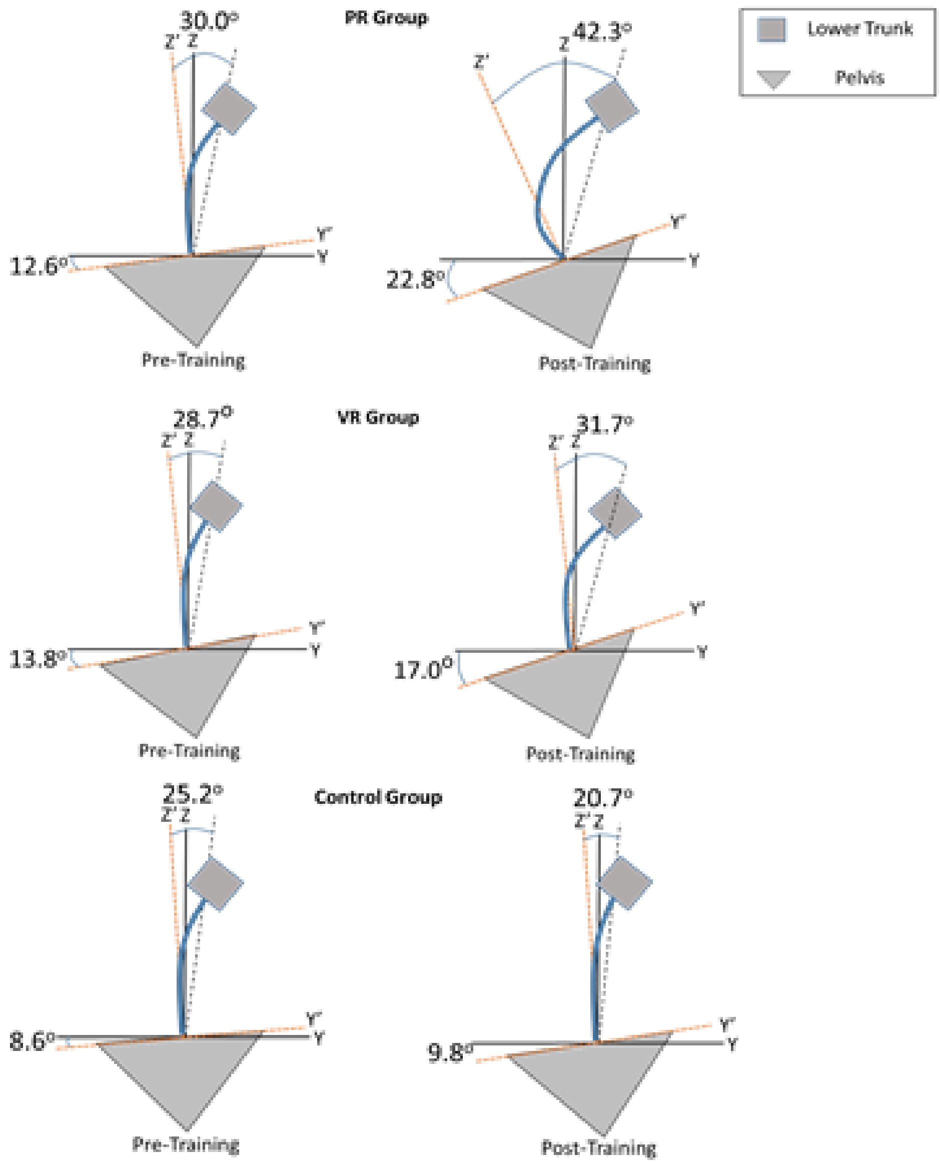


Fig. 3.11 Lower trunk and pelvis schematic, showing the average rotation in the anterior/posterior direction, pre and post training for the PR, VR, and control group.

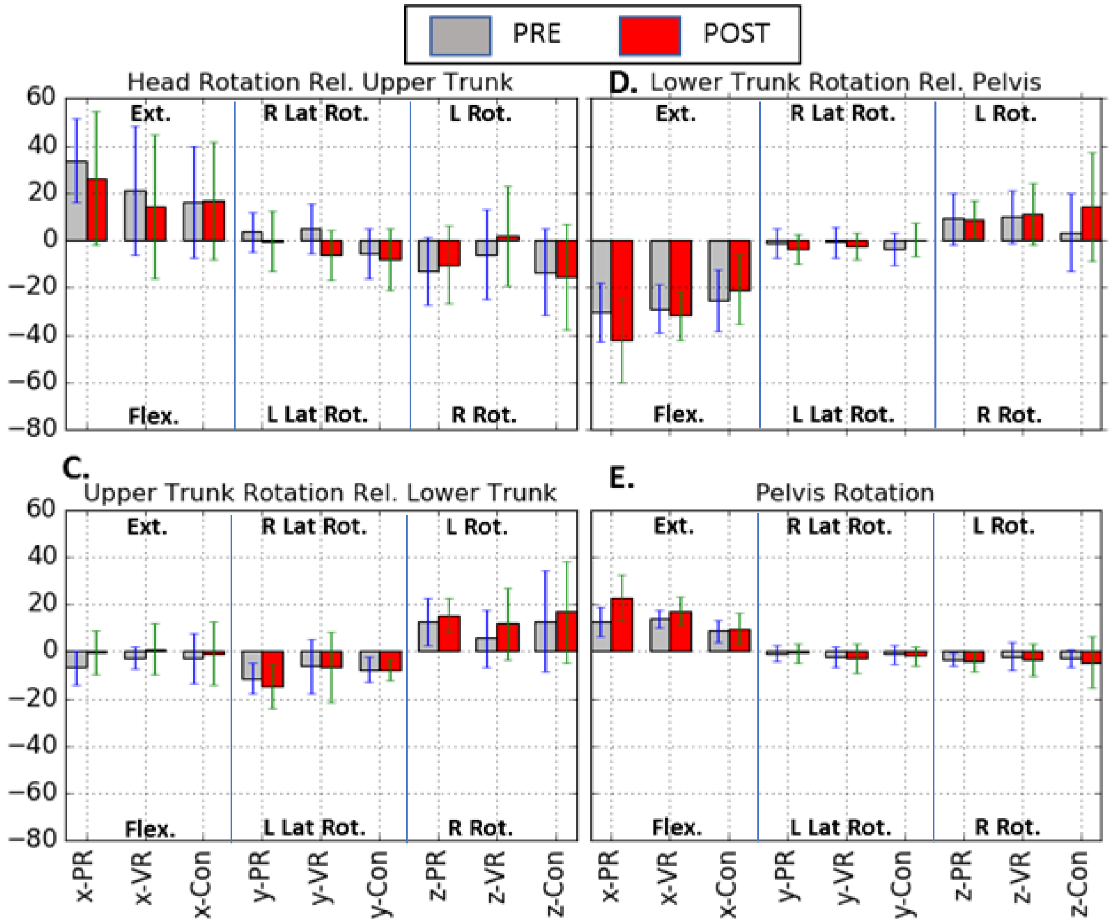


Fig. 3.12 Results on pre/post segmental rotations during the modified functional reach test across each group.

3.2.4 Discussion

In this proof-of-concept study, we tested if specific practice of a seated reaching task (without foot support) in a virtual environment, at and beyond the individual's point of postural stability failure, could enhance volitional control of upper body and extend the stability limits established by the configuration of the pelvis. We compared the results with experiments conducted in a real environment with physical object manipulation (PR) and with a control group to extract the advantages of a VR based training. Experimental results demonstrated that on average, over a single training session, the subjects were able to significantly translate their body further anteriorly from their neutral postural configuration with both physical reality and virtual reality training with TruST. Physical reality training also significantly increased the rotational profiles of the lower trunk and pelvic segments in the flexion-extension plane of motion. On the other hand, the control group (physical reality training without assist-as-needed forces) did not show any significant change.

The significant increase in anterior lower trunk translation seen with the PR and VR groups, but not with the control group demonstrates that the TruST robotic system has a significant beneficial effect in seated posture training. The assist-as-needed force strategy allows the subjects to train at their maximum stability limits without failing. This allows the subjects to explore a larger range of motion and a stable postural configuration required to maintain or recover seated balance. On the other hand, the control group does not get enough time to spend at the stability boundary without failing and thus has to use their hand for support. As a result, the sensorimotor postural experience required for adjusting their postural kinematics to complete the task successfully may be decreased compared to the PR and VR group who were provided with the assist-as-needed force from TruST. Furthermore, it was visually apparent that the control group had difficulty maintaining dynamic postural stability during the peg board test at positions beyond their stability limits. In this case, the subjects repetitively had to place their fingers for support while the PR group showed more consistency in their reaching, without often using the hand for support. Accordingly, the VR group did not have a table for support, as the experiments were virtual, and rarely placed their hands on the stool to regain stability during training.

In comparing the PR and VR groups, to observe the difference between training requiring physical object manipulation or through a gaming experience, it was seen that the PR based training produced slightly better outcomes. Although both PR and VR groups showed significant increases in lower trunk translation, only the PR group showed significant increases in lower trunk and pelvic rotational profiles, while the VR group showed non-significant increases. The training effect with the TruST showed that the pelvis increased towards extension while the lower trunk increased towards flexion, but only the changes in the PR group were significant. This suggests that although both PR and VR training showed significant increases in postural range of motion, PR training can potentially show better outcomes. Yet, there were no significant differences in results between PR and VR. It was also noted that in VR, there was a tendency for an interaction effect for pelvic rotation ($p=0.056$). This can mean that with a larger sample size, VR could also show significant improvements in rotational profiles. Since there was no difference in statistical significance between PR and VR, we cannot specifically conclude that one is better than the other, yet both individually showed improvements in postural kinematics. As discussed in [29], Nikolai Bernstein's theory of motor learning states that the central nervous system's hierarchy of control mechanisms for posture and movement are organized with distributed and parallel processing [20]–[23], where neural mechanisms that integrate posture with dexterity movements (e.g. reaching control) are recruited in patterns that are task oriented [20], [23]. By providing an assistive force at an individual's point of postural stability failure, the subjects experienced larger upper body displacements, and were better able to integrate the postural requirements within the context of the specified task [29]. As the task was practiced at the point of stability failure or maximum level of stability, the subjects were better able to control each segment separately, thus increased their range of motion.

The purpose of this study was to assess the improvements in seated postural range of motion with the use of an assist-as-needed force strategy from TruST, in a virtual and real training environment. For this reason, the virtual gaming experience was made to be less variable, and more repetitive in nature to mimic the training provided by a nine-hole peg test. Although controlling a drone and collecting coins add a gaming experience, the task can be seen as repetitive where a user can master the motion and timing required to successfully complete the training. For this reason, the level of cognitive engagement desired for a game based training may be diminished, leading to fewer improvements in outcomes as seen with the rotational profiles for the VR group. On the other hand, physical training such as the peg board, require adequately completing and witnessing an outcome (e.g. peg going into a hole) that feeds on visual and tactile stimulus, which might have led to an increased range of translational and rotational motion.

The main advantage that a VR can provide is variability of training. With variation of training, the cognitive demands and engagement are increased, as the user is required to make decisions under time constraints and possibly rewarded for improving through a scoring system. It maybe that due to a lack of variability in the training task in the VR group, the outcomes were not as large as those seen with the PR group. Yet, there was no significant interaction or difference between the PR and VR groups.

Even though the improvements are significant only in translation and not in rotation, VR training can be appreciated for its trade-offs. Although physical training would suggest a better outcome, VR training could be performed more readily without the need for added personnel, be more cost effective, and be extended to higher variability in training and in a low-cost environment, while physical (PR) training requires specific hardware and personnel for training. VR training can also be conducted at home and therefore can be done more often than PR training. It may be that more sessions of training can show far better improvements than a more challenging training conducted with fewer sessions. Another advantage of VR is that we can create direction specific, task oriented trajectories for training and modify the training task according to the subject's spatial position, recorded by a motion capture system. By using a virtual environment, we can adapt the difficulty of the training sessions progressively, challenging the subject and increasing the cognitive demand to complete the task successfully. This can be tested in future studies.

Our work shows, by direct comparison of similar tasks, that VR training can improve postural range of motion when used with TruST. With the adherent similarity between VR and PR towards postural improvements, we are inclined to believe that the ease of conducting training and the low cost make VR an excellent source of providing rehab training. In areas such as rehab of children with musculoskeletal disorders, this may be an optimal way to incorporate rehab in their daily lives, along with school and homework, where the remaining time between PT visits and leisure time is limited, by combining the two together. It may be that with increased normalcy in daily schedules and decreased mental fatigue towards PT visits, children patients may show improved outcomes. We are raising questions for future investigation through this work, where positive rehab outcomes may not only be a factor of training but also the normalcy in lifestyle and the feeling of compromise between rehab and leisure time. Seeing preliminary benefits from VR, we believe that VR based rehab might show improvements in rehab outcomes, due to not only to the primary intervention, but the secondary factors that also improve the quality of life.

3.3 Seated Posture Training with Cerebral Palsy Patients

Cerebral palsy (CP) is a permanent disorder caused by a non-progressive brain damage that impairs development of movement and posture, and disrupt reaching, grasping, and walking [30] [31]. CP is the most common childhood physical disability with 2.0-3.5 per 1000 births with a lifetime cost per person of $921,000 in the US [32], [33]. According to the Gross Motor Function Classification System-Expanded and Revised (GMFCS-E&R), children with GMFCS-E&R III-IV have limited ambulation and impaired sitting control that require the use of wheelchairs with pelvic-trunk seating adaptations to carry activities of daily living [34], [35]. They constitute about 25% of the CP population and their limited sitting abilities are associated with reduced life expectancy[36], [37].

Sitting is critical to independence in ADLs for children with GMFCS-ER III-IV [36]. These children do not acquired independent sitting and their postural control systems in this position are highly impaired—orientation to maintain the spatial relation among anatomical segments and between the body and the environment and stability to control the body's center of mass (COM) within the base of support (BOS)[38]. During development, trunk control in sitting is acquired progressively, from head to hips, throughout the first 8mos of life. Its acquisition promotes coordinated reaching, grasping, visual-manual-oral object exploration, and head orientation to direct gaze [39]–[41]. In this CP subpopulation, the neural insult disrupts this segmental progression of trunk control and sitting-related functions [42], [43]. Research has shown that children with CP GMFCS-E&R III-IV have trunk control deficits at low-thoracic and lumbar vertebral regions that hamper posture and reaching during sitting. However, when their torso is supported at mid-thoracic level, slightly above the more-affected trunk region, postural and arm performance significantly improve during reaching [43]. This finding suggests that an external support at the impaired trunk region can be applied to train seated postural and reaching abilities in children with GMFCS III-IV.

To date, conventional rehabilitation has failed to address this critical problem. In response to this long-standing need, we have developed the robotic Trunk-Support-Trainer (TruST). The TruST is a motorized-cable driven belt placed on the trunk. It creates a force field around the user that exerts active-assistive forces on the trunk when the body moves beyond the user's postural stability limits. The TruST-intervention is based on motor control and learning principles and consists of two blocks, each lasting six sessions. In the 1stblock, in addition to the TruST belt, pelvic strapping was required during sitting. This belt was no longer needed during the 2ndblock. Tolerance and feasibility of the TruST and postural training were tested with a single-subject study of a 13-year-old child with cerebral palsy. Then, we longitudinally investigated short-term and long-term postural improvements in three children (6-14y).

3.3.1 Protocol

Participants were children and adolescents (M = 11y, SE: 1.5y) diagnosed with bilateral CP GMFCS III-IV and trunk control deficits at the low-thoracic region (T9-T12), according to the Segmental Assessment of Trunk Control (SATCo = 4) [44]. Three of the four children had difficulty sitting independently due to the trunk impairment.

The GMFCS classified functionally all participants as wheelchair users [34] and The Manual Ability Classification System (MACS) and ABILHAND-kid indicated that participants had limited manual and bimanual abilities [45], [46]. The Gross Motor Function Measure Item-Set

(GMFM-IS) revealed motor limitations during sitting [47]. Manual dexterity and upper limb function was evaluated with the Box & Block test (B&B) in combination with video-coding to determine the rates of successful and unsuccessful responses, and to compute performance time (ms) of grasping and arm transport (upper extremity displacement) phases [48]. The GMFM-IS and B&B were used to report changes in gross motor and upper limb functionality after intervention.

We examined seated postural control with the TruST belt placed on the low-thoracic region. To test postural across training sessions, participants performed the functional reach test (forward direction) and the postural star-sitting test, a customized postural test that measures the area of stable sitting control [49]. This test is based on the Star Excursion Balance Test to measure standing balance, in which a person displaces the foot along 8 star-shaped lines during one-leg stance [50]. Similarly, in the star test, participants move the trunk as far as possible while displacing a large ball with the head and without hand assistance.

We collected upper body kinematics to explore postural improvements in both orientation (static control) and stability (active and proactive control). In static control, the participants had to orient upper body in the vertical plane and maintain sitting during 10s. Active control is the ability to move the upper body's COM over the BOS without losing balance. The participants had to maintain upright sitting while visually following a toy to rotate the head 90deg to both sides. Proactive control is the ability to anticipate and counteract external and self-triggered postural perturbations, such as those encountered during reaching [38]. TruST was used to statically support the low-thoracic region during baseline and post-training assessments. In the single-subject study, we tested assistive-forces equivalent to 10% body weight and the functional reach test to compute the force field boundaries (Fig. 3.13).

In the single-subject study, the participant underwent five baselines. Two post-training evaluations were performed after each treatment block, 1day and 1week after the last training session, to measure immediate and short-term improvements. We computed means (M), ± standard errors (SE), the Minimal Detectable Change (MDC) constructed on 95% confidence intervals (95%CI), and the net percentage change with respect to the mean baseline measurement (Δ%) to interpret substantial motor changes. In the longitudinal-group study (n = 3), two baseline assessments were averaged. Two post-training sessions were performed 1week and 3mos after the TruST intervention to address short-term and long-term postural improvements. We used Generalized Estimating Equations (GEE) for statistical analysis.

The postural program comprised two training blocks that were subdivided into 6 sessions each. In the 1sttraining block (1st - 6th), the child underwent the motor-learning-and-control postural program with TruST and pelvic strapping. Data from our single-subject study suggested that pelvic strapping could be removed in the 5th or 6th session; and thus in the 2ndtraining block (7th - 12th), the participant followed the postural training with TruST in independent sitting without pelvic assistance.

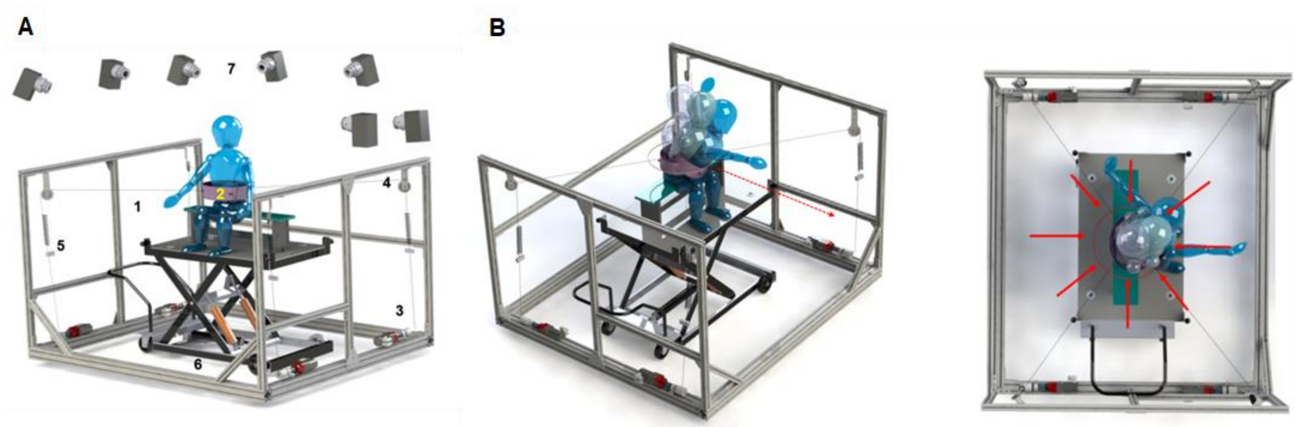


Fig. 3.13 Trunk Support Trainer (TruST) design and force field configuration with the functional reach test. (A) The main TruST components are four steel cables (1) connecting a pliable belt (2) with the motor/spools (3) through pulleys (4). The cable tensions were measured with springs and load cells (5). A lift table (6) was used to regulate the height of the sitting child and keep the belt and cables in the horizontal plane. Infrared cameras (7) were used to track the position of the belt and collect kinematics. (B) Participants were instructed to perform the functional reach test in forward direction (red dotted arrow, left panel) to define the circular force field (red circle, left and right panels). The shadows represent the body transition from sitting to maximum body displacement to define the force field boundaries (red small arrows, right panel).

The same pediatric physical therapist delivered the training in all sessions. Regardless of the task practiced, the training followed an organized set of learning and control parameters that could vary across participants and from one session to another. These training sessions were 90-120min long and consisted of massed practice with a 15min break. The training intensity was moderate-high, but never beyond maximum fatigue. When it was possible, problem solving and decision-making were part of the training to allow the child to experience trial-and-error and postural and reaching control strategies in complex actions. The task-oriented postural training was practiced sequentially, first within postural boundaries, then close to these boundaries, and finally beyond stability limits with active-assistive forces via TruST. Once the task was learned (≥ 50% success), movement variability was added and control parameters were modified.

3.3.2 Data Analysis

MATLAB (R2017b, Mathworks, 2017) was used for data processing. To examine seated static, active and proactive postural control, we applied kinematics (200Hz) and video-coding (60Hz) analyses. A LED light was used to synchronize video and kinematic data. We first trimmed through video-analysis the reaching task and then selected onsets/offsets with a kinematic-based analysis computing 5% of the reach peak velocity. Then, arm path profiles (i.e. resultant vector) were depicted to re-define reaching onsets/offsets [39]. The number of successful and unsuccessful reaches during the B&B was used to define the more-affected and less-affected hands of the participants. Additionally, Datavyu software (http://www.datavyu.org/) was used to video-code the behavioral frequency and the temporal windows of the grasping and arm transport components of the reach during the B&B. Grasping was defined as the temporal window from the moment the hand contacted the block to the time this was raised from the box surface. Arm transport accounted for the arm movement from one side of the box to the other over the partition; and it was computed from the end of the grasping to the release of the block (1st frame at which the block was not in contact with the hand palm).

Three-dimensional motion capture (VICON, Oxford Metrics) was used to collect upper body kinematics, rotations and translations. Data was smoothed with a zero time-lag 4thorder Butterworth filter with a 4Hz-cutoff [51]. Angular rotations were computed as inter-segmental angles, following the right-hand convention with an Euler sequence X-Y'-Z'': flexion(-) / extension(+) around x-axis (rightwards: +), right-lateroflexion(+) / left-lateroflexion(-) around y-axis (forward: +), and left-rotation(+) / right-rotation(-) around z-axis (upwards: +). We modeled the upper body as a 6 linked-segment system, including: head, upper-thorax, lower-thorax, pelvis and upper limbs. COM approximation of segments and upper body was based on anatomical landmarks and anthropometrics [52].

To examine postural control, we computed the absolute summation of upper body COM displacement (cm) across x-y-z axes. Postural orientation (static) was estimated as the averaged angular motion of each segment across planes of motion. Angular movements in each plane were computed by subtracting absolute maximum and minimum angles. Total angular motion was calculated as the absolute summation of angles. The SD of angles was used as a measurement of dispersion to address postural variability.

In the single-subject-design study, we applied a distribution-based approach to explore and interpret the data. We first calculated the standard error of measurement (SEM) to define the spread in computation error and distinguish reliable substantial changes after the postural program with TruST [53].

In the single-subject-design study, only postural/reaching control changes that met the MDC or were higher or close to 50Δ% were interpreted as substantial changes. We hypothesized that the presence of clinical, functional and biomechanical improvements would demonstrate tolerance and feasibility of the TruST intervention; and thus, the longitudinal-group study would be justified.

In the longitudinal-group study, the alpha rate was set at 0.05. Kinematic angles identified as extreme outliers (values three times greater than the interquartile data range) were removed. Data normality was examined with *Shapiro-Wilk* test and visual *Q-Q* plots. We originally planned to carry a within-subject ANOVA, but normality and sphericity were violated in most instances. Also, data were highly variable and required a trial-by-trial analysis. Thus, we applied GEEs to analyze events-in-trials following a repeated-measures procedure with subjects as clusters and training/evaluation sessions as the within-subject variable. A linear model was selected. An independent covariance structure was specified as correlation matrix based on the quasi-likelihood under independence criterion (QIC) goodness of fit coefficient, and because it is recommended in studies with small sample size [54]. Sequential *Holm-Bonferroni* method was used to increase the power of the statistical test while controlling the familywise error [55]. *Post-Hoc* tests were carried if the statistical model was significant.

3.3.3 Single Subject Results

In the single-subject study, the participant acquired greater motor capability to displace the trunk and reach further along the sagittal plane. During training, the child progressively enhanced his trunk control in the functional reach test. When the TruST intervention was completed, the child increased by Δ233% his ability to lean and reach without hand support or sitting failure (baseline: 6.3cm, 1week post-training: 21cm).

At the beginning of each training block (first 2-3 days), we observed a detriment in trunk control during the functional reach test immediately after the training. We believe that this within-session difference might be the minimum time that the child requires to adapt to the training features (pelvic support versus no pelvic assistance, and progressive increase in the force field diameter during active-assistive feedback) and to cognitive-muscle fatigue effects (low attention-to-task and poor tone, strength and endurance of torso muscles). However, the child was able to adjust to the training parameters and the discrepancy in trunk displacement faded across training sessions. Fig. 3.14 shows the ability to reach anteriorly during the baseline assessment, pre and post training each day, and during the follow-up session. This shows that the first patients increased their ability to reach from roughly 6cm up to above 20cm and without any pelvic strapping.

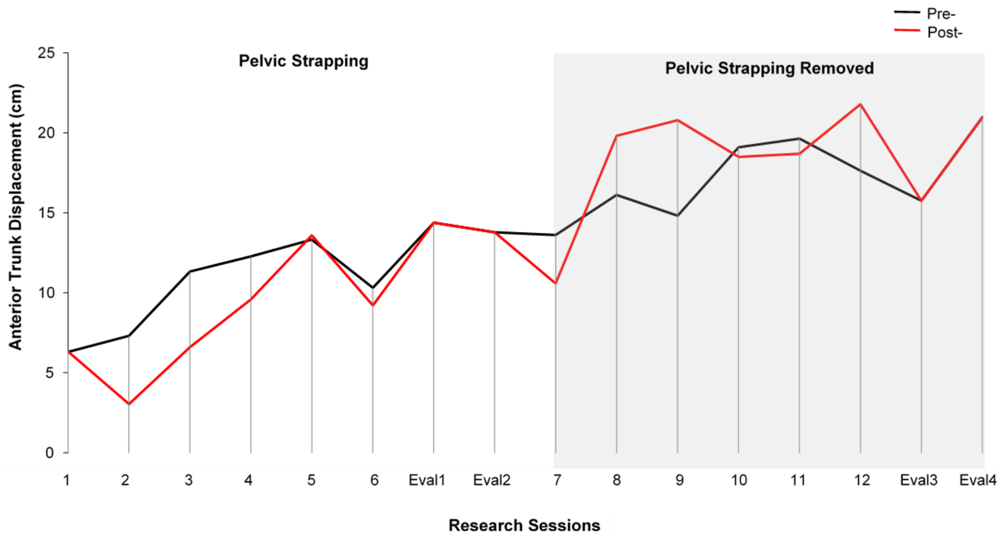


Fig. 3.14 Trunk control improvement with TruST. The functional reach test was evaluated before (black lines) and after (red lines) the postural intervention with TruST across training and evaluation sessions. Trunk control improved progressively across the 12days. With respect to BL (1stPre-training session = 6.31cm), the increase in trunk displacement was Eval1 = Δ128% (1day after 1sttraining block), Eval2 = Δ118% (1week after 1sttraining block), Eval3 = Δ150% (1day after 2ndtraining block) and Eval4 = Δ233% (1week after 2ndtraining block). Note the substantial trunk control improvement after the 7thtraining session during independent sitting (without pelvic strapping). Pre-, Pre-training. Post, Post-training. Eval1, Evaluation 1day after the 1sttraining block. Eval2, Evaluation 1week after the 1sttraining block. Eval3, Evaluation 1day after the 2ndtraining block. Eval4, Evaluation 1week after the 2ndtraining block.

After the TruST intervention, the child acquired more proficient upper body control in static upright sitting during 10s. The child obtained a stable posture with a Δ58% reduction in upper body COM translation across the anteroposterior, lateral and vertical axes (baseline: M = 26.7 ± 7.6cm, MDC = 7.4cm; PT: M = 11.3cm). More specifically, there was a Δ41% reduction in head displacement (baseline: M = 40.4 ± 7.6cm, MDC = 10.1cm; 1week post-training = 23.93cm) and Δ40% decrease in upper-thorax displacement (Baseline: M = 25.3 ± 3.9cm, MDC = 9.1cm; 1week post-training: M = 15.3cm). Moreover, the upper body segments above the TruST belt were more vertically aligned (Fig. 3.15).

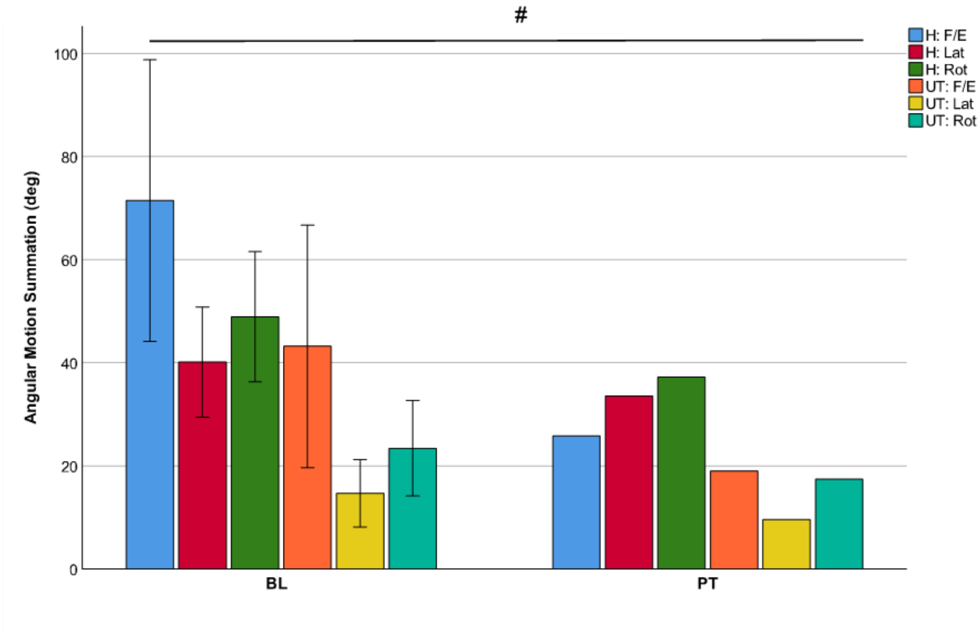


Fig 3.15 Seated postural control kinematics in our single-subject-design study. (A) A more vertical position of H was observed in F/E (BL: M = 71.47 ± 13.66º, MDC = 6.76º; PT: M = 25.84º, Δ64%), M/L (BL: M = 46.97 ± 7.99º, MDC = 6.27º; PT = 33.56º, Δ29%) and ROT (BL: M = 54.25º ± 7.21º, MDC = 7.43º; PT = 37.22º, Δ31%). This was also true for UT in the F/E (BL: M = 43.21 ± 8.32º, MDC = 11.78º; PT: M = 18.98º, Δ56%), M/L (BL: M = 20.82 ± 6.64º, MDC = 6.40º; PT = 9.57º, Δ54%) and ROT (BL: M = 27.40º ± 5.37º, MDC = 6.88º; PT = 17.47º, Δ36%). # = MDC was achieved.

3.3.4 Longitudinal Study Results

We trained three additional subjects in a longitudinal study. Two out of three children experienced functional improvements in the GMFM-IS after the postural TruST intervention. One participant (Subject 3) improved sitting control 1week post-training that returned to baseline values and another (Subject 4) demonstrated short-term and long-term gait improvements.

We used TruST as a measurement tool to examine trunk control during the functional reach test. All participants significantly and progressively improved trunk control across the two training blocks (Wald $\chi 2 = 41.63$, $P < 0.001$) and research stages (Wald $\chi 2 = 36.86$, $P < 0.001$) compared to baseline (M = 5.5 ± 1.4cm, 95%CI: 2.7-8.3). They enhanced trunk control during the functional reach test after the 1st training block (M = 12.5 ± 0.5cm, 95%CI: 11.5-13.5, $P < 0.001$) and 2nd training block (M = 16.4 ± 1.7cm, 95%CI: 13.1-19.6, $P < 0.001$). Similarly, they showed significant improvements 1week post-training (M = 15.9 ± 0.6cm, 95%CI: 14.7-17.2, $P < 0.001$) that were retained after 3mos (M = 14.6 ± 1.3cm, 95%CI: 12.0-17.3, $P < 0.001$).

Fig 3.16. shows how each subject in the longitudinal study improved their anterior reach. In addition, subjects were assessed for workspace and coordination, which showed improvements as well. Subjects were able to reach further in each direction around them, therefore able to increase their postural workspace. These findings were retained during the three month follow up.

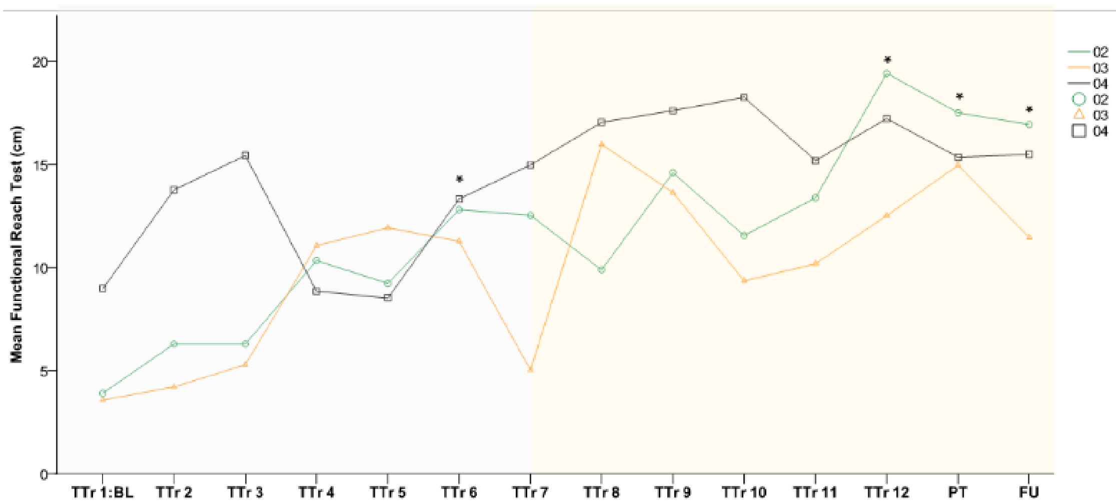


Fig. 3.16 Trunk control during the functional reach test in our longitudinal-group study. All participants improved gradually their trunk control. Note the reduction in trunk displacement after the 1st training block (TTr 6) when the pelvic straps were removed. This is particularly accentuated for participant 03, who had severe sitting control deficits. The three participants show a steady increase in trunk control and short-term and long-term improvements in independent sitting 1week (PT) and 3mos (FU) after the TruST intervention. BL, Baseline. TTr, Training. PT, 1week post-training. FU, 3mos follow-up. Numbers indicate the participant code. $^*P < 0.05$.

Similar to the results obtained in the functional reach test, all the participants improved their 360° of volitional trunk mobility and significantly expanded their area of stable sitting control during the postural training with TruST (Wald $\chi 2 = 41.63$, $P < 0.001$) and across research stages (Wald $\chi 2 = 49.35$, $P < 0.001$). A low increase rate and more variable trunk control responses were found during the 1sttraining block (Fig. 3.17 – 3.18). However, the participants significantly improved their trunk control after the 2ndtraining block (M = 409.9 ± 130.1cm^2, 95%CI: 154.8 - 665.0, $P < 0.001$) with respect to baseline (M = 127.6 ± 61.1cm^2, 95%CI: 7.9-247.2). Furthermore, the postural gains were observed 1week post-training (M = 395.3 ± 48.3cm^2, 95%CI: 300.7-490.0, $P < 0.001$) and at the 3mos follow-up session (M = 270.0 ± 30.4cm^2, 95%CI: 210.49-329.56, $P < 0.001$).

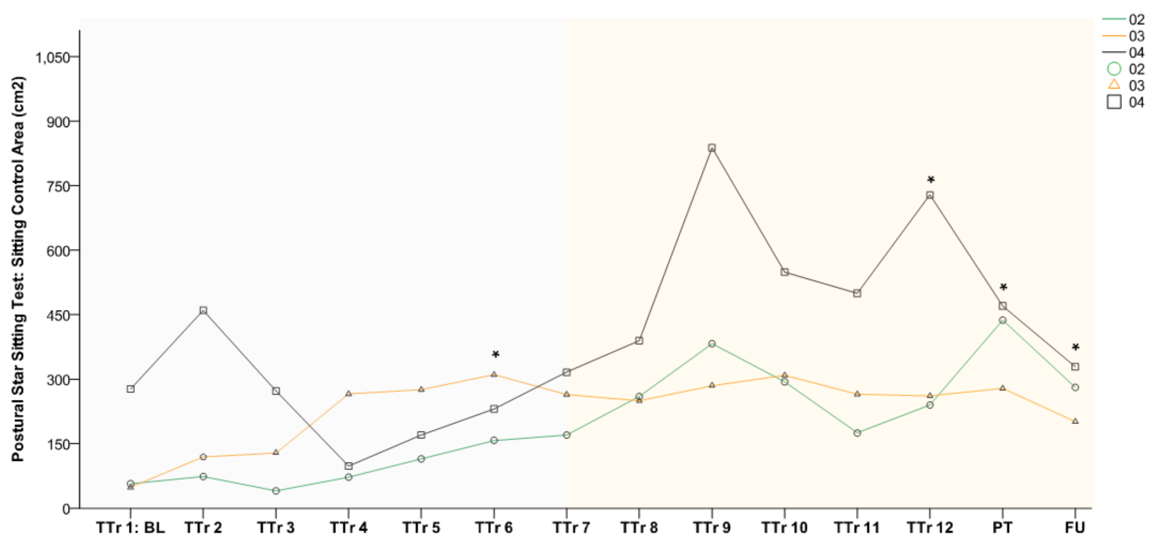


Fig. 3.17 Trunk control area in the postural star-sitting test in our longitudinal-group study. (A) A gradual increase in the area of stable sitting control is observed for the participants across training sessions with greater trunk control during the 2ndtraining block (without pelvic strapping). The participants 02 and 03 achieved their highest level of sitting control 1week post-training (PT). The participant 04 obtained his maximum level of sitting control during the training (9th-12th). They maintained their sitting control improvements 3mos after the TruST program; although, a slight decrease toward baseline, mainly for participant 04, was found ($P < 0.001$). BL, Baseline. TTr, Training. PT, 1week post-training. FU, 3mos follow-up. Numbers indicate the participant code. *$P < 0.001$.

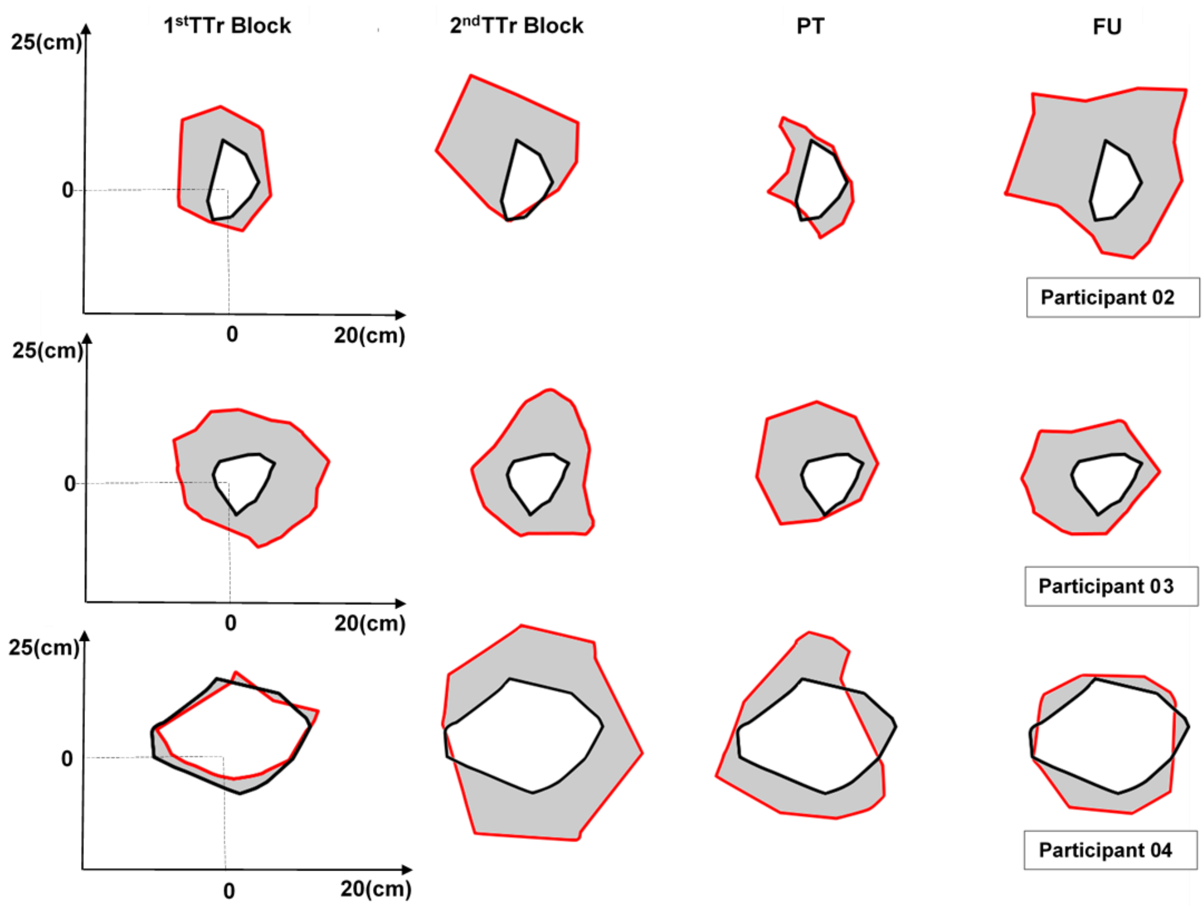


Fig. 3.18 The BL is represented by black lines and unfilled area. The 1stTTr, 2ndTTr, PT and FU are represented by red lines and grey area. The plots show that all participants improve symmetrically their trunk control area immediately after completing the TruST intervention (2ndTTr block). In general, this increased area of trunk control was still present 1week and 3mos post-training. Note that for participant 04 the area of trunk control overlapped between BL and 1stTTr block. BL, baseline. 1sttraining block (1st TTr), 2ndtraining block (2nd TTr), 1week post-training (PT), 3mos follow-up (FU). A, anterior. R, right.

In the seated static task during baseline, the participants displayed the typical hyperextended head-trunk posture found in children with CP and severe balance deficits during sitting. They enhanced their postural configuration with an upright alignment of upper-thorax and lower-thorax after the TruST intervention (Fig. 3.19). The group significantly improved postural balance (Wald χ^2 = 1567.78, P < 0.001), as demonstrated by a reduction in upper body COM displacement 1week post-training (P < 0.001). However, the 3mos follow-up assessment showed that upper body COM displacement returned to baseline values. Interestingly, there was a specific-training effect in lower-thorax orientation (flexion-extension plane) (Wald χ^2 = 16938.65, $P < 0.001$), which was the most impaired trunk region across all participants, according to the SATCo. The lower-thorax segment changed from an abnormal flexion in baseline to a vertically aligned position 1week post-training ($P = 0.003$). Notably, this vertebral alignment was retained after 3mos ($P < 0.01$). Moreover, the upper-thorax configuration was significantly different across research stages (Wald $\chi 2$ = 117.68, $P < 0.001$). Participants demonstrated a flexed upper-thorax orientation 1week post-training ($P < 0.001$) that was vertically realigned 3mos post-training ($P < 0.001$).

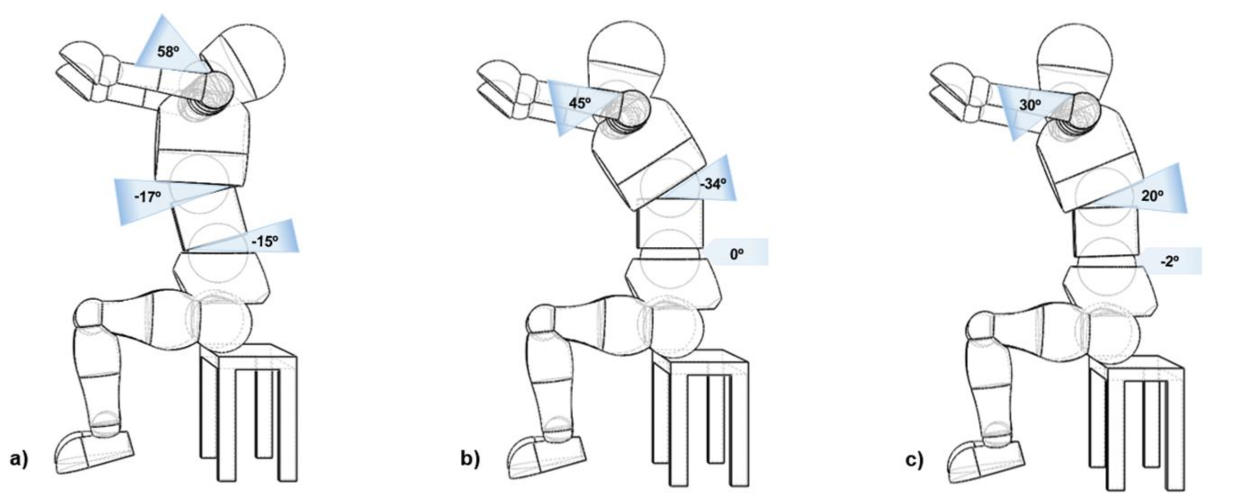


Fig 3.19 Postural and reaching kinematics in our longitudinal-group study. (A) The models represent averaged postural configuration of head, upper-thorax and lower-thorax. Before training in BL (a), participants presented a stereotypical CP sitting configuration with hyperextended head and upper-thorax to prevent sitting failure. After the training in PT (b), participants showed neutral alignment of lower-thorax with respect to the pelvis, and head and upper-thorax hyperextension disappeared.

3.3.5 Discussion

In historical CP reports, posture has been a central aspect in the motor disability that accompanies this condition [56]. However, most studies on postural interventions in CP involve high-functional ambulatory participants and neglect those who are wheelchair users and in critical need to improve their basic sitting abilities to fully participate in society. In this study, we show that robotics may be a solution to address this problem. We investigated the feasibility and potential of a robotic-mediated postural intervention with TruST. The overall outcomes of the current study indicate that the progressive application of haptic-force feedback on the torso, tailored to the user's trunk stability limits, in combination with a motor-learning-and-control training approach, promotes independent sitting and maximizes functional postural and reaching abilities in children with CP GMFCS-E&R III-IV.

To date, there are no successful conventional rehab programs (i.e. exercises or *"hands-on"* approaches) in children with CP GMFCS-E&R IV to promote independent sitting or with CP GMFCS-E&R III to obtain long-term improvements in seated postural and reaching abilities *(16, 17)*. Play-oriented therapy ensures motivation, high level of compliance, continuous attention-to-task and engagement [57]. Learning and control principles have been satisfactorily implemented in other therapeutic approaches to train paretic upper extremities [58], bimanual control of hands and arms [59] and combined upper-lower extremity movements [60]. However, rehab programs in sitting control have not accounted for motor learning and control parameters with a robotic platform that has been designed to maximize functional outcomes. There is evidence showing that robotic platforms, such as TruST, address engagement, repetition and intensity during practice of task-oriented movements that enhance impaired upper limb motion in children with hemiplegic

CP [61]. TruST promotes training, based on trial-error and haptic-feedback depending on the user's body position within or beyond stability limits, respectively.

Unlike other robotic systems, TruST has not been designed to substitute the role of the rehabilitation professional. In contrast, the goal of this robotic-rehab paradigm with TruST is to create a dynamic robot-child-clinician interface during the implementation of postural training, based on motor learning and control principles. The TruST system displays online visual feedback to the clinician about the location of the trunk's COM, where the belt is placed, while the participant practices goal-oriented postural tasks. The intensity of the haptic force-feedback on the torso corresponded to 10% of the child's body weight and was programmed to be an "*assist-as-needed force"* during training. Thus, when the participant was within the area of stable sitting control, the force field was inactive and the subject only experienced task-intrinsic feedback from visual, vestibular and somatosensory inputs during voluntary postural and reaching movements [12]. Only when the upper body was outside the predefined force field limits, the haptic-force feedback was applied to the trunk to actively train postural sitting recovery. A fundamental aspect of TruST is that the force field boundaries were progressively increased and systematically adjusted to the participant's trunk control capabilities across training sessions. This configuration was crucial to avoid child-TruST interdependence that would inhibit learning of the postural improvements.

In our single-subject and longitudinal-group studies, participants without sitting control acquired independent sitting after 12 training sessions. Three participants demonstrated gross motor improvements. Considering the GMFM minimum clinically important difference, two participants (03 and 04) showed a medium effect size change (1 point) and one participant (01) demonstrated a large effect size (2 points) between baseline and 1week post-training evaluation

[62], [63]. In the participant with GMFCS-III and sitting difficulties (04), the trunk control gains propagated to walking improvements. This is in line with previous work showing that trunk control is correlated with gross motor functions, and sitting is the strongest predictor in walking abilities compared to other factors such as spasticity, muscle strength and selectivity, and age [64], [65].

The progressive increase in sitting control measured by the functional reach and postural star-sitting tests shed light on relevant clinical improvements at the functional and neuromuscular levels. The functional reach test measures proactive postural control, because it takes into account motor planning and unidirectional control of the trunk in the sagittal plane during reaching [66]. When participants started the experiment, they could reach no further than 6cm. However, 1week and 3mos after the robotic-mediated postural training with TruST, they could obtain a maximum trunk distance of 16cm during the reach. Although there is no known cut-off value in the functional reach test in CP, the reaching improvements were higher than the MDC (6cm) defined for people with stroke [67]. The postural star-sitting test is a customized play-oriented postural assessment with TruST to measure the 360^{o} area of stable sitting control. This area was severely limited in our participants before the robotic-aided postural training with TruST, except for one participant (04) classified as GMFCS-E&R III. Children with CP and higher functional independence, but sitting control problems, could opt for a postural training with TruST in which pelvic strapping is removed early in the training. Our data show that there was not much difference in the area of stable sitting control for participant 04 between the baseline and end of the 1sttraining block. Accordingly, potential improvements in trunk control could have been expected earlier in the treatment if the pelvic straps had been removed from the beginning of the TruST intervention. Overall, participants tripled their area of stable sitting control after completion of the training and doubled it 3mos post-training due to potential muscle strengthening, enhanced proprioception, improved spatial-temporal postural strategies, and fine control of the torso muscles. Specifically, in dynamic sitting tasks, the stability limits are larger in front of the body (forward direction) compared to behind. Children with severe postural dysfunction cannot sit upright and they lean forward to lock their trunk in a non-functional position to maintain stability (Movie S3). Despite this stereotypical sitting posture, the use of haptic-feedback with the TruST force field in the posterior peripersonal space resulted in substantial improvements in posterior sitting balance [68]. The self-postural exploration and the expanded area of stable sitting control after the postural intervention with TruST could be explained by enhanced intrinsic sensorimotor feedback mechanisms [12].

Postural kinematics showed significant improvements in static, active and proactive dimensions, which are mechanistically characterized by different modes of neural and musculoskeletal control [69]. Static control requires muscle tone regulation and long-lasting isometric co-activation of neck-torso muscles to maintain upright sitting [70]. Participants obtained a more aligned lower-thorax with respect to the position of the pelvis, which was the trunk region with the greatest level of instability and where the TruST belt was placed during training. This improvement could be due to more efficient regulation of paravertebral muscle stiffness [71]. Moreover, participants obtained a more vertically aligned upper-thorax, which could be critical to prevent vertebral pain and respiratory problems due to sustained spinopelvic misalignments during passive sitting [72], [73]. Active control is a dimension assessed by numerous clinical tests and it requires online sensorimotor control to maintain the body stable during ongoing movements [44], [74]–[76]. Participants acquired the ability to perform head rotations with reduced compensatory trunk displacements and variability during independent sitting. In regards to proactive control, participants acquired shorter, straighter and less variable reaching paths with proficient task-dependent postural strategies while sitting independently [38],

[77]. In our single-subject study, the child showed improved reaching control with increased trunk displacements, which most likely were used as a compensatory postural strategy to overcome the lack of elbow extension, paresis, and spasticity of arm muscles [78], [79].

In summary, the single-subject-design study with a child with CP GMFCS-E&R IV proved tolerance, feasibility and short-term postural and reaching benefits after 12 postural training sessions with TruST. The longitudinal-group study replicated these results and added that these functional improvements were retained after a 3mos-washout period. A limitation was our sample size. Thus, cohort studies with large sample size and including different clinicians to drive the TruST intervention would serve to generalize our outcomes. Future research should address the efficacy of our postural intervention with TruST in randomized clinical trials compared to conventional therapy.

Chapter 4

4 Standing Human Experiments with TruST

In this chapter, we translate the experiments in Chapter 3 from the seated to the standing position. The goal of this chapter is to assess if humans can change their trunk coordination and movement after training in the standing position with TruST. When in the standing position, the pelvis. A person in the standing position needs to maintain balance and coordination at the trunk, pelvis, and knees. For this reason, we have applied a passive belt to the pelvis. We also incorporate an instable surface for challenging healthy patient during our initial pilot study. In the following sections, we will discuss the modifications to the TruST for stand training, a study with healthy adults, and then a three session training with a single patient with cerebellar ataxia.

4.1 Stand Training in Healthy Adults

Elderly and people with neurological and musculoskeletal pathologies present with lack of postural control. Postural control profiles are characterized by increased postural sway, diminished or excessive responses to perturbations, abnormal trunk oscillations, and poor control of equilibrium postures [16]. These balance deficits may be accompanied by exacerbated postural sway without a preferred direction and with increased intersegmental movement at the head, trunk, pelvis, and legs [16], [80]. In certain cases, movement is decomposed from a multi-joint strategy of control into a series of single joint motion at the trunk, hip, knee, and ankle [16]. Rehabilitation strategies have focused on muscle strengthening or coordination training alone to reduce intersegmental sway. However, some studies suggest that postural training that directly targets active balance control could be most beneficial for motor learning [16].

Balance maintenance requires a complex correlation between different sensory modalities and optimal multisensory processing [81]. This includes a sensorimotor interaction from the vestibular, proprioceptive, and visual systems for motor coordination [82], [83]. Considering patients with cerebellar ataxia, spinal cord injury, cerebral palsy, or even the elderly, devising rehab strategies that target a multimodal strategy can be most beneficial. This is to provide training that consists of muscle coordination, muscle strengthening, and proprioceptive enhancement [16]. While current rehab strategies rely on manual techniques to provide these modalities where the training is limited to a single component at a given time, robotic platforms permit the integration of multiple modes of therapeutic training at once.

Poor sensorimotor control of postural stability and orientation increases the load of the cognitive system, which has been associated with higher prevalence of potential falls due to the presence of a dual cognitive-motor processing for controlling the postural task [83]. Furthermore, balance platforms have shown to improve body position awareness to serve as a means to improved balance [83], [84]. Active training that directly controls a person’s center of pressure (COP) in combination with real-time feedback are thought to have beneficial impact on postural orientation and stability [83]. In addition, balance training that includes static and dynamic modes of control has proven to show beneficial effects [80].

4.1.1 Previous Design Concepts

In past work, researchers have aimed to identify novel methods to improve postural balance. In [85], researchers utilized virtual reality training with a bicycle and correlated riding velocity with path trajectories to determine postural balance control. Yet, the proposal was for using such parameters for indirect balance rehabilitation. A device called the "Spider-bot" was proposed in [86], which allowed a subject to hold an end-effector using their hands and to practice activities of daily living. This device does not directly assist or train any segment of the body, but provides a single barrier for holding in cases of falls. The work does not show human studies and states a possible use in balance and strengthening exercises.

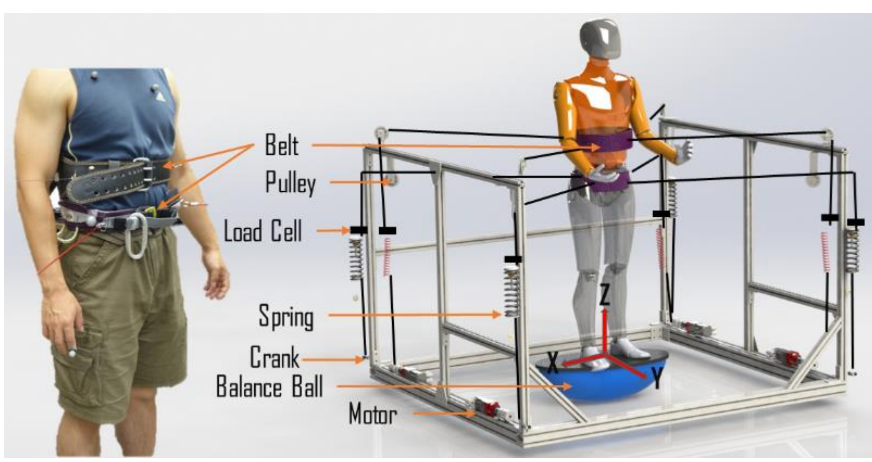


Fig. 4.1 The novel robotic platform with a dual-belt system for multimodal and intersegmental training. Device consists of an active actuated assistive segment and a passive spring-based segment for stability training.

We have modified our previous robotic platform for seated balance training, the Trunk Support Trainer (TruST) [29], to a stand balance training device with dual segmental control (Fig. 4.1). In addition to the active control, we have added resistive forces applied through a spring and pulley system. Four springs are mounted, one at each corner of the frame, and attached to a pelvic belt. A mechanical ratcheting crank is installed at each end to tighten and preload the spring forces as desired.

The passive pelvic belt is composed of four cables at each corner of the belt, attached in series with a load cell and an extension spring (Stiffness: k = 0.3N/mm), to a ratcheting crank. The crank is used to preload the springs to 5% body weight, with the subject in a neutral configuration. The force follows Hooke's law, $F = -k\Delta x$, at each spring or attachment point creating a resultant force in the opposing direction of human movement. $\Delta x = x - x_0$, where x_0 is the original length of the spring at 40cm and x is the current length of the spring.

4.1.2 Protocol

The research protocol was approved by Columbia University's institutional review board. Ten healthy adults (7 males, 3 females) with average age: 27.6y, weight: 164.9lbs and height: 68.1in. were recruited. Nine of ten subjects were right-handed. They were randomly assigned to either the experimental (EXP) or control (CON) group across three stages: baseline (BL), training (T), and post-training (PT).

The lower trunk was provided with actively controlled AAN forces using motors and load cells and a force-field control strategy, as done in our previous work on seated posture control [29], [87]. This was provided through a belt, placed firmly at the lower trunk. The subjects had no force administered to the lower trunk, as long they stayed within their predetermined workspace. As the lower trunk positioned out of the determined stability boundary, assistive forces provided haptic (proprioceptive) feedback and 10% body weight force to assist back towards the stability boundary.

The pelvis was controlled passively through resistive training with preloaded springs, load cells, and a ratcheting crank. A belt was placed at the pelvis and connected to springs using cables, one at each corner of the belt. The crank was used to pre-load the springs to 5% body weight, while the subject stood in their neutral configuration. The subjects performed all training standing on a balance ball, to simulate postural instabilities. This unstable surface required the subjects to provide additional balance at the knee, pelvis, and trunk level.

The EXP group wore both belts, for resistive forces at the pelvis and assistive forces at the lower trunk. The CON group wore both belts, but were not provided any assistance at the lower trunk or pelvis. The cables were removed to disengage the springs.

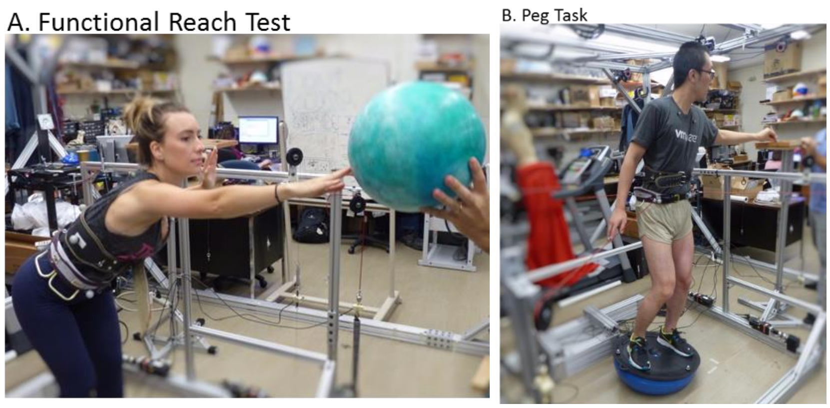


Fig. 4.2 The functional reach task to determine the lower trunk stability boundary and the peg task used for training and assessment, respectively.

During the experiment, subjects stood on a balance ball with a semi-sphere base while inserting and retrieving one peg (23mm x 6.4mm) in a small-diameter whole (7.8mm) that was placed at subject’s maximum reaching distance. This distance was obtained by having the subject displace a ball as far as possible in forward direction, while standing on a balance ball (Fig. 4.2A Forward Reaching Test).

In this unstable standing condition, all subjects were instructed to maintain standing balance and learn a new postural pattern during the peg task. It consisted of displacing first the pelvis with respect to the BOS (feet positioning) and then upper body in the direction of the reaching task. This was to consciously increase muscle activation at the pelvis for increased proprioception due to the resistive forces. In addition, it promotes experience of different postural configurations. Without such motion, the subject may focus solely on using the upper trunk, without possibly challenging themselves to experience more optimal pelvis-trunk configurations, under unstable conditions. This task was performed at the dominant side during BL stage. This task and the forward reach task were repeated again during the PT stage, after the training session.

During training, subjects performed two blocks of 10 trials in five different directions: right, 45deg right, forward, 45deg left and left. In each trial, the subjects had to insert and retrieve the peg from the peg board hole, while maintaining stability on the balance ball (Fig. 4.2B). The peg was placed at the maximum reach distance, or boundary of max stability, measured during the initial forward reach test.

Those subjects in the EXP group had to displace/rotate pelvis against an initial preloaded spring, equivalent to 5% of body weight resistance in each spring (passive component). This resistance however increased linearly as the subject pushed past their upright equilibrium point, due to the nature of a passive spring system. On the other hand, CON subjects had no resistance or assistance from the device. This allowed us to compare the effects of the training paradigm alone with that with the use of the robotic device.

4.1.3 Results

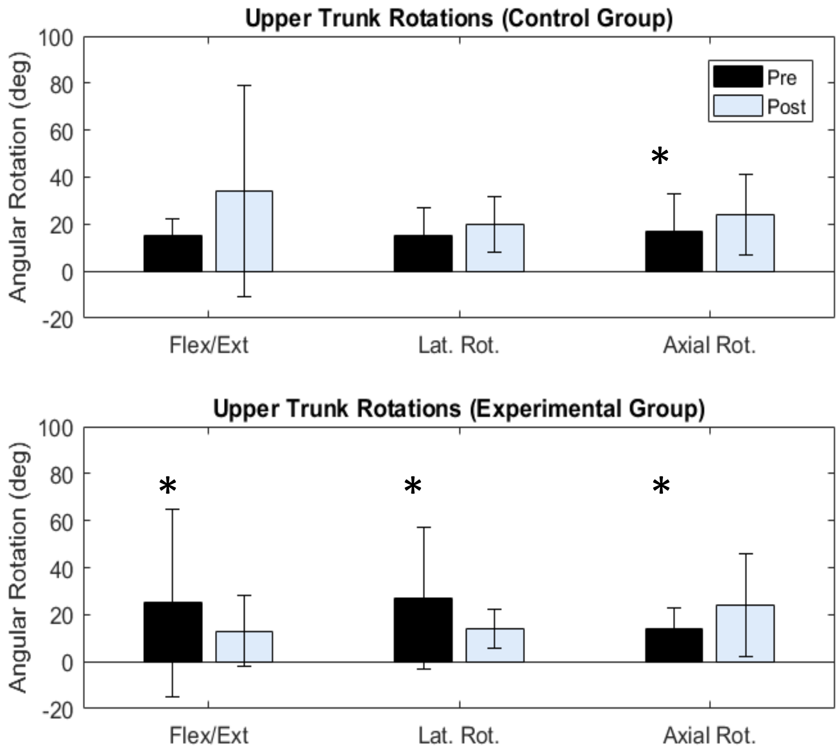


Fig. 4.3 Upper trunk rotation profiles. These results show that using the TruST, the experimental group decreased their rotations in flex/extension, lateral rotation, and increased movement in axial rotation. The control group increased their rotations.

The results in Fig. 4.3 show that the experimental group significantly decreased their upper trunk flexion and extension and lateral rotation to the dominant side. However, they were able to complete the same task by utilizing axial rotation. This was while standing on an unstable surface. The control group increased their rotations with the unstable surface. This shows that training with the system can significantly alter the trunk kinematics, making an individual more vertical during a task.

4.2 Stand Training in Cerebellar Ataxia Patient

Ataxia is a neurological disorder associated with uncoordinated movements, which results from damage to the cerebellum. This damage can be congenital, result from a stroke or tumor, or be caused by an injury. This damage leads to abnormal proteins that attack the brain and the spinal cord, affecting motion planning and balance. This can be considered a feed-forward issue, meaning it can be challenging for an individual with Cerebellar Ataxia to plan an accurate movement. Other symptoms include lack of coordination, abnormal eye movements, slurred speech, tremors, difficulty with walking and poor balance, irregular gait, and loss of fine motor skills.

The cerebellum contributes to proprioception during motion [88]. Proprioception, by definition, is the spatial and temporal awareness of body position during motion, and is inherently essential for generating accurate movements. Research suggests that motions which rely more on spatial information provide a better proprioceptive acuity than motions which rely heavily on temporal parameters [88]. Cerebellar integrity is also essential for motor adaptation when presented with new tasks [16]. This may mean that active motion which requires a predictive motor output, such as a moving target, may be more challenging for individuals when the cerebellum is affected.

In terms of clinical features, balance abnormalities are characterized by increased postural sway, abnormal response to perturbations, variable control of equilibrium, and abnormal oscillations of the trunk [16], [89]. The region of cerebellar injury has a direct effect on the individual’s sway. Anterior lobe atrophy causes an increase in center of pressure sway in the anterior posterior direction, while the vestibulocerebellar lesion shows an increase in omnidirectional movement. In particular, anterior lobe damage results in high velocity, low amplitude sway, while a vestibulocerebellar lesion results in low frequency, high amplitude sway [16].

Researchers have questioned whether patients can improve balance and coordination even though the damaged portions of their brains directly affect motor planning and motor learning [90]–[92]. Because of this, it has been traditionally considered that physiotherapy treatments that focus on remediation of impairments caused by cerebellar dysfunction would be ineffective [90]. Therefore, treatment strategies in the literature suggest compensatory approaches which use limb weight, gait aids, and coordination exercises [90], [93], [94]. However, recent studies suggest that motor learning is possible in patients with cerebellar damage, suggesting that physiotherapy interventions and motor retraining aimed at promoting neural plasticity may be beneficial [95]–[97].

A systematic review conducted by Martin et al. [90] documented the effectiveness of physiotherapy interventions on improving cerebellar based impairments. Majority of the studies were case studies. The interventions included proprioceptive neurofacilitation (PNF), balance exercises, vestibular habituation exercises, and Frenkel exercises. PNF is an intervention where muscles are stretched maximally to induce muscle activation and response. A prolonged and maximal stretch will loosen the muscle, preparing them for activation. In vestibular habituation exercises, a patient is asked to perform a series of bending, reaching, and alternating posture tasks. Among the many tasks are sitting, bending to the side, standing, and bending forward at the hip. The Frenkel exercises are coordination tasks which require reaching and aiming. This includes following a finger, touching the nose, touching a target, or other similar activities. These are progressed based on the needs of an individual. The basic concept includes constant repetition, sensory cues, increased task speed, assistance as needed, rest periods, new tasks with performance cues such as instructions or sensory stimulation. Majority of the studies documented were case studies. The results of studies using these metrics showed modest improvements in gait, subjective trunk control during silent standing, and self-reporting of balance improvements. Often, the documentation of the training methodology was unclear and unreproducible.

Although it is known that the trunk plays an integral role in daily tasks, it is often overlooked or not addressed in rehabilitation [89]. The trunk is essential for maintaining balance and limb control, and relies on muscle strength, endurance, and sensory-motor control for stability. The trunk muscles, such as the rectus abdominis, external oblique, internal oblique, transverse abdominis, and erector spinae, provide stability through kinetic chain activities [98]. Since persons with cerebellar ataxia have difficulty regulating the force and speed of trunk muscle contractions, rehabilitation aimed at trunk stabilization may improve limb coordination and balance [89].

Based on the literary findings described above, we have developed a novel approach for trunk posture rehabilitation. We have combined several aspects of physiotherapy that have shown benefits in posture and coordination for patients with cerebellar damage. Using the principles in Frenkel exercises, our training promotes several reaching and coordination tasks. As seen with vestibular habituation, our task also incorporates alternating the trunk and pelvis position, requiring one to maintain stability to perform a task. Finally, using a robotic intervention, we provide sensory feedback and assistance at the trunk and pelvis. This is to promote sensory awareness in terms of spatial positioning, and assistance at extreme postures. We believe that training modalities which provide haptic feedback on spatial position, challenge muscle activation, and provide postural exploration can be beneficial for rehabilitation. Specifically, the training tasks should include functional and goal-oriented performance. The nature of this training is described in detail in the next section.

4.2.1 Protocol

In this study, we were interested in assessing whether a patient with ataxia can improve their reach and trunk workspace, when training with our robotic intervention. To quantify the difference, we recruited a single 18 year-old individual with cerebellar ataxia, who performed the training with assistance from the robotic device and then again six months later without the robotic intervention. The six-month wait period was to washout any training effect. This allowed us to document the changes in performance with our training methodology, with (experimental) and without (control) the robotic intervention. The training allows the subject to train at their maximum trunk displacement, perform a reaching task, and receive assistance at the trunk and resistance at the pelvis.

This individual underwent three non-consecutive days of training in a single week, followed by a post training assessment after the week (Fig. 4.4). Roughly six months later, the patient served as his own control, following the training protocol without robotic intervention. The training protocol was approved by Columbia University's institutional review board.

Fig. 4.4 Study timeline for training and post-training sessions for ataxia patient.

In this study, we utilized the modified TruST platform in the standing position. This was based off our findings with a healthy adult population showing changes in reaching strategies [99], described in chapter 3.4. We replicated the healthy adult study with a few changes to the protocol. Since patients with ataxia have internal balance disturbances, the balancing platform was removed. The reach task was replaced from a peg board to a button. The button was a small 3 inch light up toy that was placed at 120% of maximum reach. This button light up if it was located and pressed properly. The individual was provided with actively controlled assist-as-needed forces through a belt placed firmly at the lower trunk, which was connected by cables to motors and load cells. A force-field control strategy was used, as done in our previous work on seated posture control [29], [87]. The patient had no force administered to the lower trunk within a predetermined workspace. As the lower trunk extended past the determined stability boundary, assistive forces provided haptic (proprioceptive) feedback through a 10% body weight force to assist the individual back towards the stability boundary. The force magnitude was chosen through trial and error and kept the same as previous healthy and patient studies. All tasks were performed in the standing position.

The pelvis was passively controlled through resistive training with preloaded springs, load cells, and ratcheting cranks, as shown in Fig. 4.1. A belt was worn at the pelvis and connected to springs using cables, one at each corner of the belt. The cranks were used to pre-load the springs to 5% body weight while the subject stood in their neutral configuration. The initial removed spring and cable slack. As the person moved away from their midline, the force increased in the opposite direction of movement. The patient stood on top of a force plate as they performed several reaching tasks. In the control experimentation, the subject underwent the same training protocol, but no cables or forces were applied to the trunk or pelvis.

The force field applied on the trunk was determined through the *8-point Star Test*. This is an assessment we devised to determine the force-field shape and boundary, as well as measure the changes in postural workspace. In this test, the patient is asked to displace a ball as far as they can in eight directions: anterior, posterior, right and left lateral, and four diagonals. An example of the force field shape is shown in Fig. 4.5, and shows the asymmetry in the patient's reaching ability in different directions. At the boundary, the patient received haptic feedback, or low amplitude vibrations, to signal they were at their boundary. The plot on the right in Fig. 4.5 shows that the force did not activate until the trunk center moved outside the boundary of measured stability.

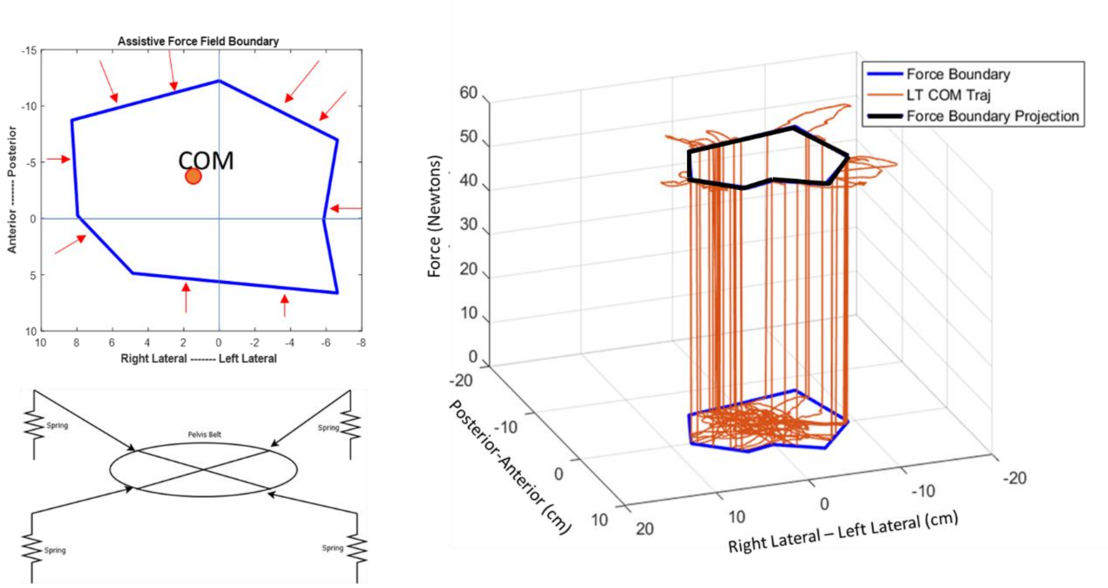


Fig. 4.5 Active force-field profile at the trunk and passive spring based belt at the pelvis. The right graph shows the force activation occurs when the center of the trunk (orange) is outside the asymmetric stability boundary (blue and black).

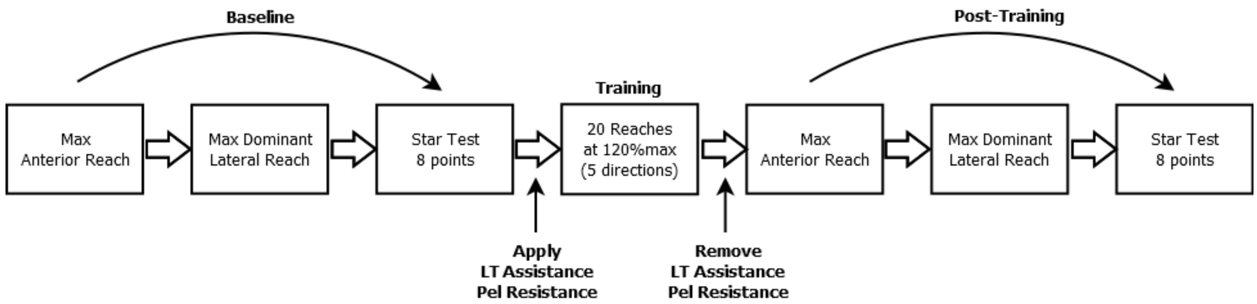


Fig. 4.6 Protocol schematic showing the task progression for each training day.

The study protocol can be seen in Fig. 4.6. During the experiment, the patient stood on a force plate and was asked to press a button (3-inch diameter) placed at 120% of his maximum reaching distance. This distance was obtained by having the subject displace a ball as far as possible in the anterior direction (anterior reach test). The 20% distance outside the maximum reaching distance was added to further challenge the individual.

In this unstable standing condition, the individual was instructed to maintain standing balance while learning a new upper body postural pattern during the reach task. The postural pattern consisted of first displacing the pelvis with respect to the BOS (feet positioning), then displacing the upper body in the direction of the reaching task. This reaching strategy afforded a conscious effort to displace past the resistive pelvis forces. Without such motion, the subject may focus solely on using the upper trunk, without challenging themselves to experience more optimal pelvis-trunk configurations under unstable conditions. This maximum reach test was performed at the anterior and dominant lateral side during both the baseline and post-training stage.

During training, the patient performed 20 reaches in five different directions: right, 45deg right, forward, 45deg left, and left. In each trial, the subject had to locate and press a button while maintaining stability (Fig. 4.7). During the experimental session, he had to displace/rotate his pelvis against an initial preloaded spring, equivalent to 5% body weight resistance in each spring (passive component). This resistance increased linearly as he pushed past his upright equilibrium point, due to the nature of a passive spring system. On the other hand, during the control sessions, the patient had no resistance or assistance from the device. This allowed us to compare the effects of the training paradigm alone against the addition of the use of the robotic device.

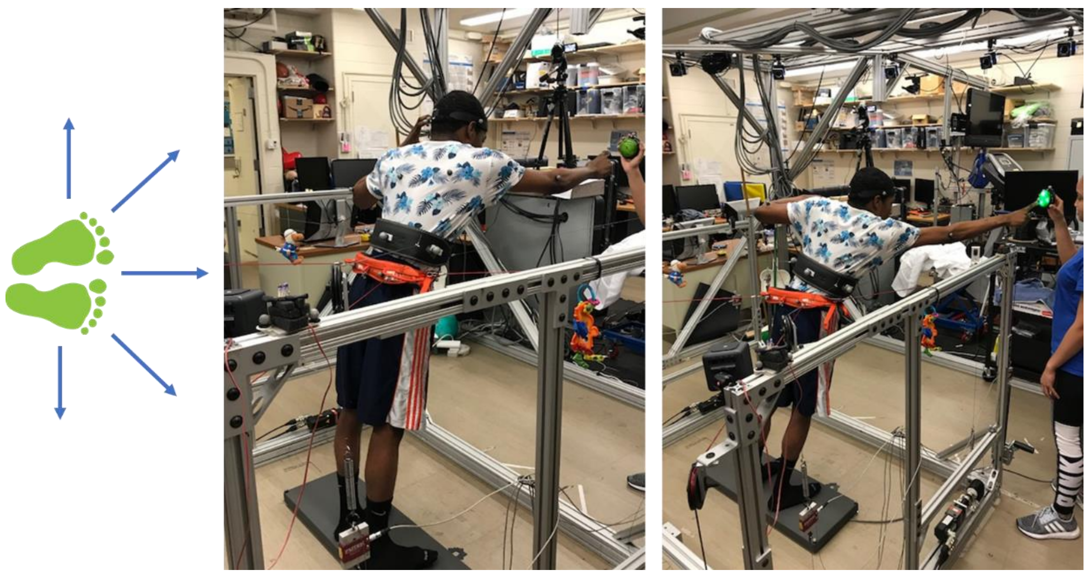


Fig. 4.7 Five reach directions during training and the task initiation and completion using the button.

4.2.2 Results

To assess the workspace changes throughout the training protocol, we performed the 8-point star test at the beginning of each session. This was done for both the experimental and control subject. During the experimental training with the robotic device, the workspace area increased each day, totaling 199% growth over three training sessions. Conversely, the control subject's workspace decreased after each session, totaling a 66% reduction over the three non-robotic training sessions. The results can be seen in the graph below (Fig. 4.8-4.9).

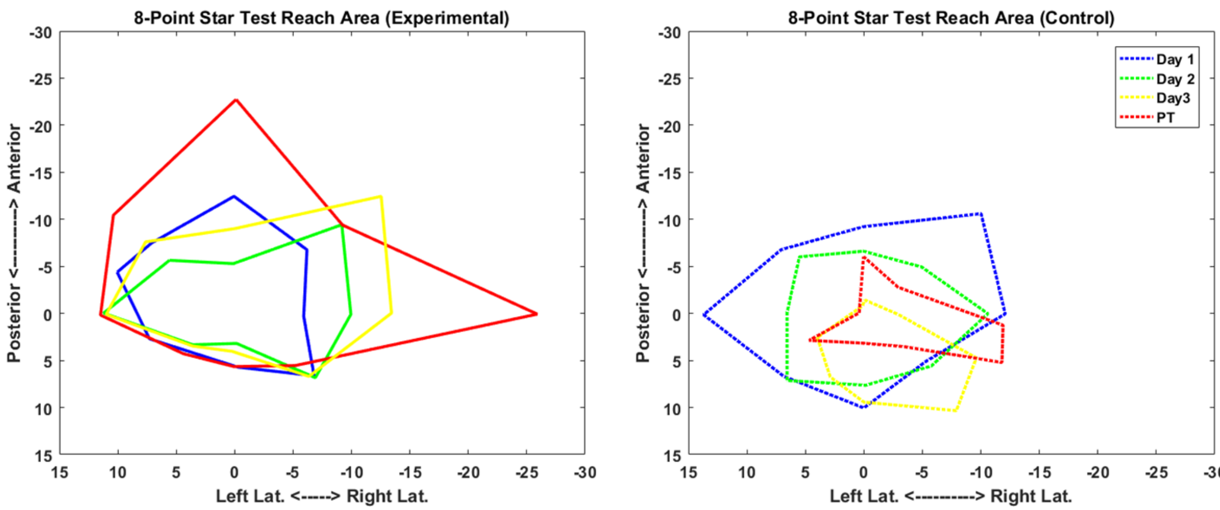


Fig. 4.8 The star test shows that the experimental subject improved the postural workspace measured at the beginning of each day, while the control decreased each day.

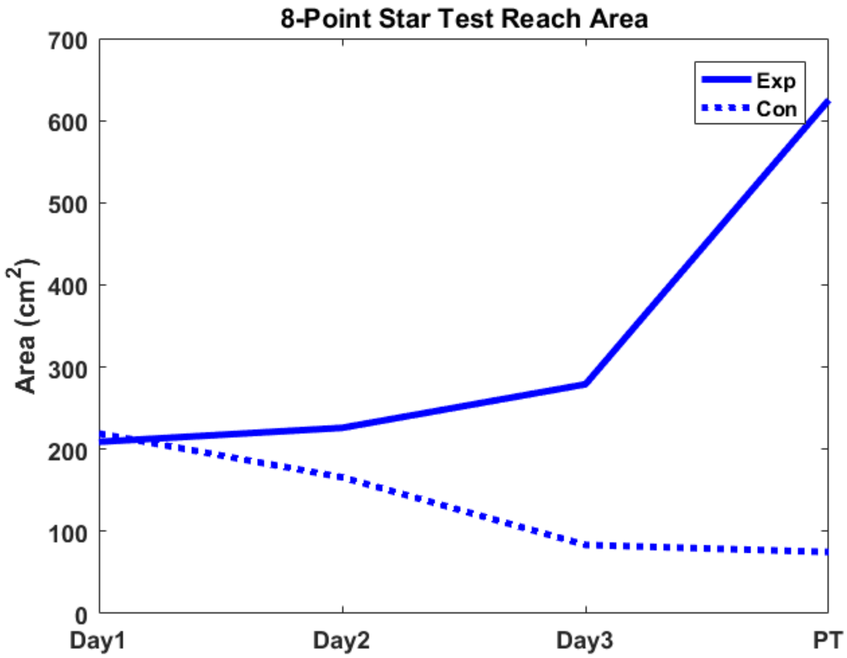


Fig. 4.9 The total area of the workspace was measured each day. The line graph shows the experimental subject improved the postural workspace area measured at the beginning of each day, while the control subject decreased each day.

The upper trunk (UT) and lower trunk (LT) translations were measured at the beginning of each training session as the patient performed the maximum reach test in the anterior and dominant lateral directions (Fig. 4.10). In the experimental training, the anterior LT and UT translations increased each session, elongating 161.8% and 75.7%, respectively. The control training showed a 17.1% decrease for the LT but a 2.7% increase for the UT.

For the lateral reach in the experimental training, the LT reach increased 54.5% and the UT reach increased 13.9%. On the other hand, the lateral reach in the control training decreased 50.5% for the LT and 18.4% for the UT.

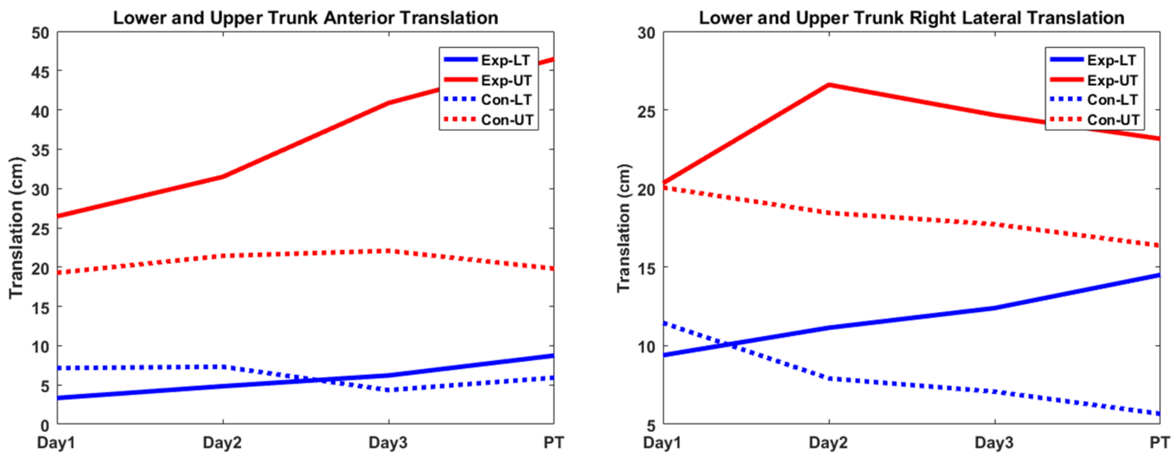


Fig. 4.10 The total anterior and posterior translations increased each day of training with the experimental sessions and decreased with the control sessions.

The differences in the trunk's center of mass (COM) and center of pressure (COP) were measured to account for the changes in reaching strategy (Fig. 4.11). In an ideal and stable configuration, the COM sits over the COP. In the experimental subject, as the patient reached further each day, he also decreased the deviation between the COM and COP. At the same time, the medial-lateral (ML), anterior-posterior (AP), and overall COP displacements also increased. This means that as the trunk moved farther, the COP moved farther, but stayed closer under the COM.

For the control subject, displacements were more variable. As the overall translation of the trunk decreased, the COP moved farther overall and in the ML direction. This may signify that the COP required a larger deviation to account for smaller trunk changes.

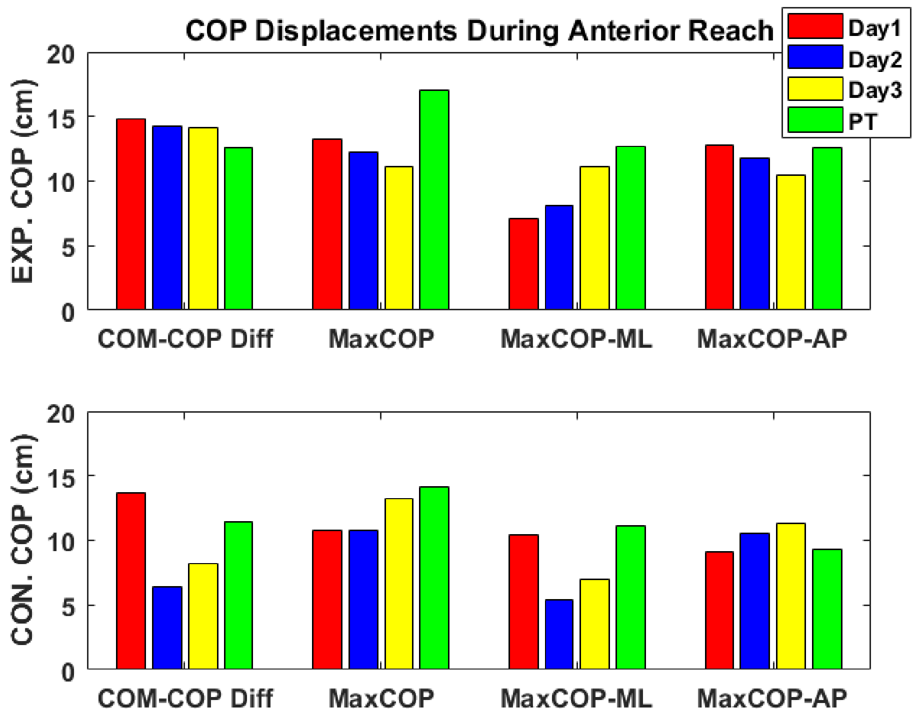


Fig. 4.11 The center of pressure (COP) translations during the anterior reach test.

Each body segment rotation was measured relative to its caudal segment. The rotational profiles associated with the anterior reach show that after the experimental training, the patient was able to increase the rotation of the upper and lower trunk, while rotating the pelvis back posteriorly (Fig. 4.12). After the control training, the upper trunk and pelvis rotated posteriorly, while the lower trunk rotated anteriorly, resulting in an overall decrease in effective reach. This presented as a “C-shaped posture” reach. The reaching posture for the experimental training can be seen in Fig. 4.13.

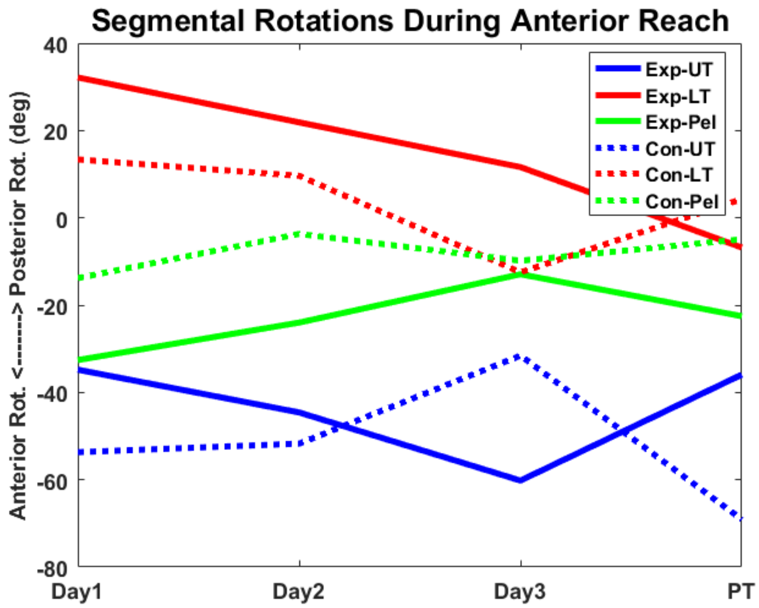


Fig. 4.12 Segmental rotations measured during the anterior reach test, at the beginning of each training session.

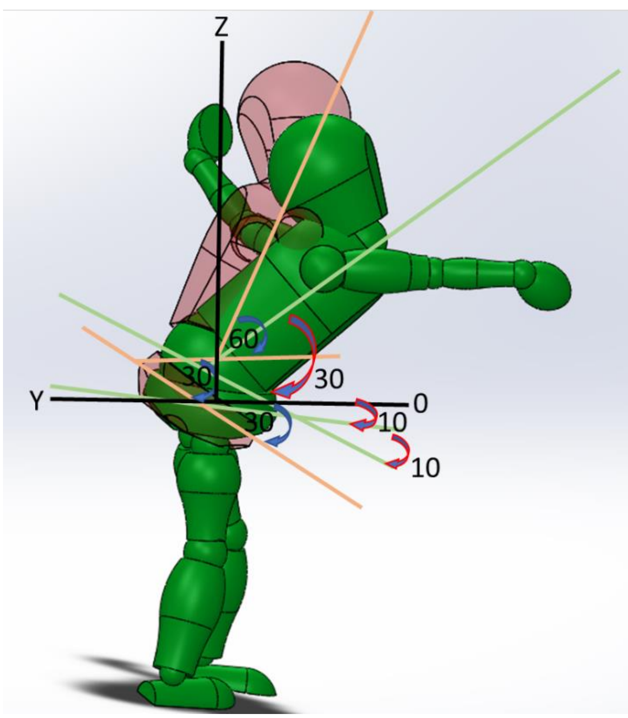


Fig. 4.13 The illustration shows the rotation change during the anterior reach for the experimental training sessions. The pink was the initial reaching posture prior to training and the green was during the last day.

4.2.3 Discussion

Several studies have attempted to determine the most effective medical interventions for patients with cerebellar damage [16], [88], [90], [93], [95]. While the treatments vary between medical, surgical, stimulation, and physical therapy, no definitive training has been identified. However, outcomes from non-invasive interventions have implied that task-specific training may produce the most beneficial results. A majority of the focus on task-specific therapy has been given to gait rehabilitation, body weight support training, and compensatory weight training, which all focus on lower limbs. Although it is accepted that the trunk plays a large role in postural balance, research on trunk rehabilitation is limited and often neglected.

In our training paradigm, we utilized our robotic platform, which showed promise in seated posture training. We modified the system to accommodate multimodal control of the trunk and pelvis in the standing position. A strong measure of trunk stability is the size of the workspace in which the trunk can maintain control. By control, we mean the ability to actively coordinate the trunk to perform a specified task, then successfully navigate back to a neutral configuration without external assistance. By including the trunk and pelvis in our training paradigm, we challenge muscle and kinematic coordination through both assistance and resistance. While the pelvis received passive resistance due to the spring-crank system, the trunk was actively assisted with haptic feedback and assist-as-needed forces. These coupled training components afforded postural exploration. Our implemented goal-oriented tasks challenge an individual to reach further, maintain stability, then successfully return to their neutral postural configuration. To assess the difference between our training with and without our robotic platform, we conducted the training using the same patient with a washout period of roughly six months.

The results from our experimental training show substantial improvements. The overall functional workspace improved 199% in just three training sessions. The translations in the anterior directions for the lower and upper trunk improved 161.8% and 75.7%, respectively. The translations in the lateral directions for the lower and upper trunk improved 54.5% and 13.9%, respectively. The range in the COP increased largely, while the difference between the COM and COP decreased. Finally, in terms of rotations, the trunk and pelvis dissociated from an *in-block* motion to opposing posterior pelvic rotation and anterior upper and lower trunk rotations. While the trunk and pelvis moved further away from the midline, the rotation between the segments provided a new reach strategy for the patient, leading to changes in the COP to account for the COM deviations. These findings clearly express that a person with cerebellar ataxia may possess the ability to learn or adapt to improve postural balance with the appropriate training.

On the contrary, the control training showed the opposite result. Over the training, the patient performed worse in their workspace assessment and reaches. They altered their reaching strategy by reducing pelvic and upper trunk rotation in the anterior direction. Yet the COP displacement from the center position increased. This may mean that the patient is using larger muscle activations at the ankle level to counter the perception of the COM shift. In general, COP changes are

Though the protocol was the same for both trainings, the patient performed all successful reaches in the experimental training sessions. This means that they were able to reach and press the button with each attempt. During the control sessions, the patient failed to touch the button in all of their attempted reaches, even though they attempted rigorously. This is likely due to the lack of assistance and resistance provided by our device during the reach. Without the device, the patient may have not experienced adequate time to adjust their posture and coordination, and therefore may not have explored new postural configurations. In addition, the failure to hit the button may have had a psychologically demotivating component, which could result in a loss of confidence. With the experimental sessions, each successful reach might have provided more confidence to the patient. This may not have been the case with the control sessions, although the patient worked and challenged themselves with each reach.

Although this is only a case report, this study provided encouraging findings to support that multimodal training paradigms may provide a beneficial environment for rehabilitation. One of the essential components to our training is our robotic platform, which provides customizable and modular capabilities to fit the specific needs of each patient. We believe that continued investigation is necessary to apply our findings to a larger patient population.

Chapter 5

5 Stand Trainer - Cable Robot Development

In this work, we have created a novel full-body cable robot for quantitatively engaging the trunk, pelvis, and the knees for training of balance and coordination (Fig. 5.1-5.2). Our system is called the “Stand Trainer” and it can apply assistive, resistive, or perturbation forces on the trunk and pelvis. This system builds upon our previous research on cable systems designed for training of sitting, standing, and walking [17], [29], [87], [99]–[101]. The system utilizes patient specific assist-as-needed force-fields while facilitating active postural recovery. In addition, the system can apply postural disturbances and train balance/coordination. The system details are explained in the next sections. Two separate systems of the same type were built for studies to be performed in our laboratory at the Columbia Medical Center and at the University of Louisville.

The primary novelty of this system is the ability to have full body control. The device allows real-time control of the trunk, pelvis, and knees. This is essential for someone in the standing position. By applying forces and moments at each segment, the body can be manipulated in the standing position. Furthermore, the device utilizes novel force-field algorithms for symmetric, asymmetric, and three-dimensional trunk and pelvis stability workspace, which are discussed later in this chapter.

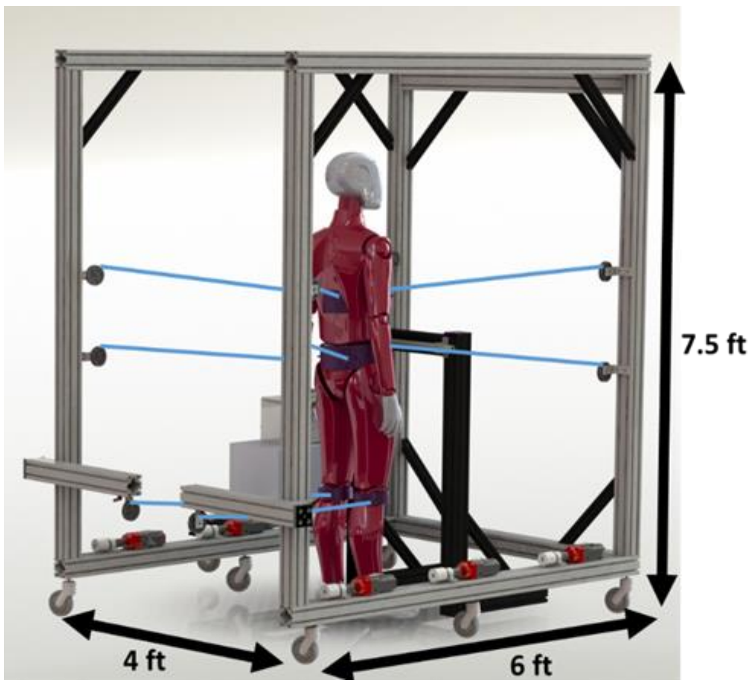


Fig. 5.1 Stand Trainer Robot CAD design

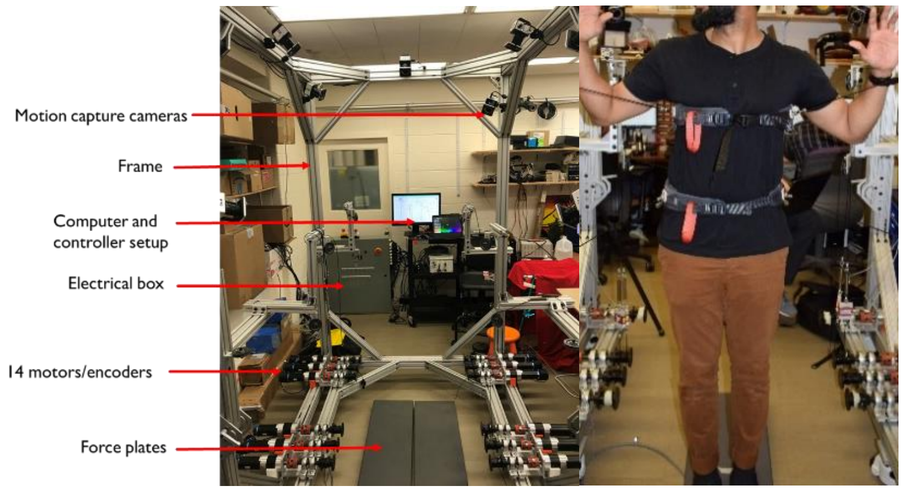


Fig. 5.2 A novel robotic platform for Stand Training. The device consists of cables actuated belts that allow control of the trunk, pelvis, and knees. This system allows training of balance through sophisticated algorithms. The device consists of 14 motors/encoders, 14 load cells, a real time motion capture system and controller, an electrical box housing the drivers and amplifiers, and two force plates.

5.1 Mechanical Design

The modular design allows control of six degrees-of-freedom at the pelvis, three degrees-of-freedom at the trunk, and extension of the knee joint. The system can apply assistive, resistive, and perturbation forces which can be adjusted in real-time using the software interface. The end effectors are belts that can be strapped on the pelvis or a region of the trunk as desired. Cables are attached to four points on the belt which is reinforced with thermoplastic for rigidity. The cables are routed through pulleys to DC motors mounted on an aluminum frame (80/20 Inc, Indiana).

The device consists of 14 DC motors with a 14:1 gearhead reduction and a three-channel encoder (Maxon Motor, Switzerland). Each motor is capable of generating up to 143.85Nm of output torque (8.81Nm nominal continuous torque). The motor shafts are fixed with a 4cm diameter winch spooled with a 1/16in PVC coated flexible steel cable allowing continuous tension of 440N (Fig. 5.3). Each cable has a load cell/tension sensor (LSB302 Futek, California) in series which can measure up to 300lb load (1334N). The load cells are connected to a signal conditioning amplifier (IAA100 Futek, California). The motors are controlled through 14 Epos2 digital positioning controllers (Maxon Motor, Switzerland) that are daisy chained and communicate via a controller area network (CAN) serial bus (Fig. 5.4). A motion capture system with nine infrared cameras (Vicon Vero 2.2 from Vicon, Denver) is used to record the cable attachment points on the belt and pulleys to calculate the force directions in the cables. A two-stage control is implemented using LabView, PXI real time controller and data acquisition cards (National Instrument, Austin). Each of the three body segments (trunk, pelvis, knees) has its own two stage control, which operate individually but simultaneously. Two individual six-axis force plates (Bertec, Columbus, Ohio) are instrumented for measuring under-foot force, moments, and center of pressure changes.

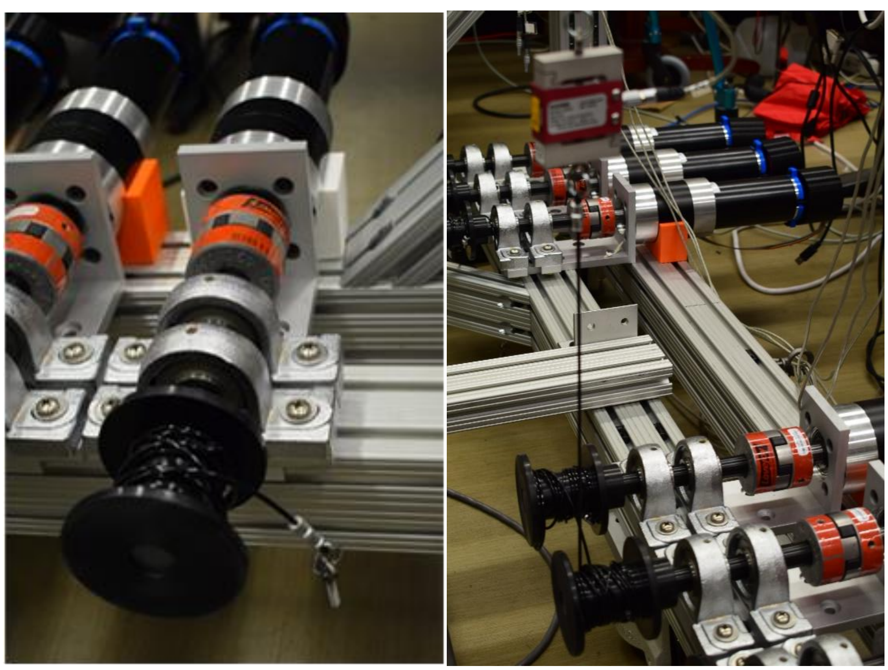


Fig. 5.3 Motor, encoder, gearhead, couple, bearings, motor shaft, and spool assembly can be seen here.

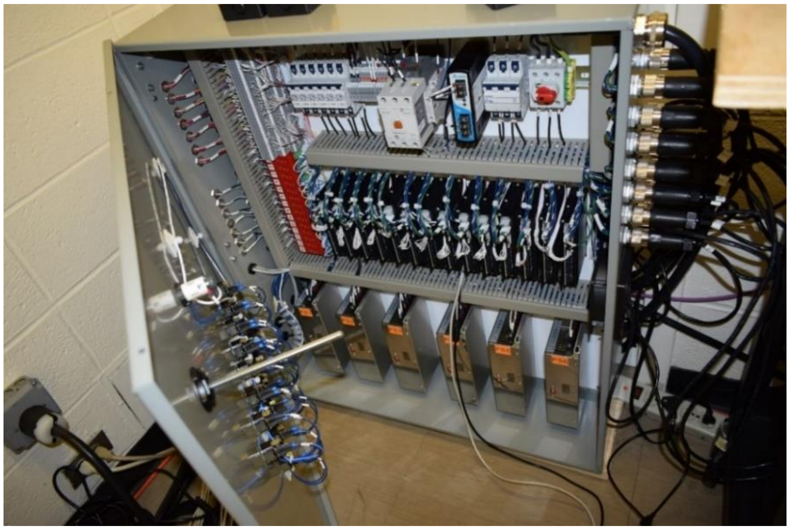
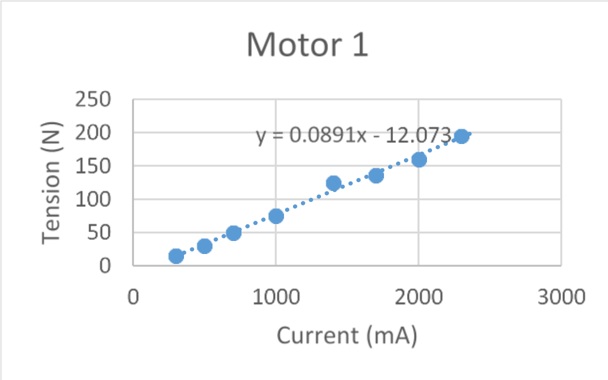


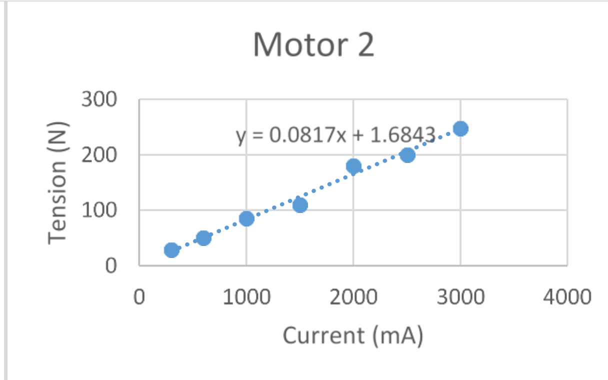
Fig 5.4 The motor driver, amplifier, and power supply enclosure box for powering and controlling the motors and tension sensors.

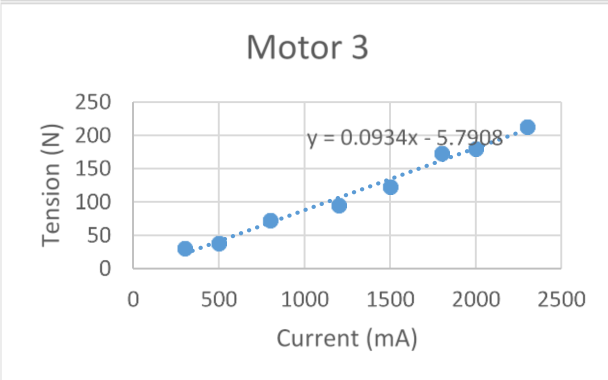
5.2 System Model

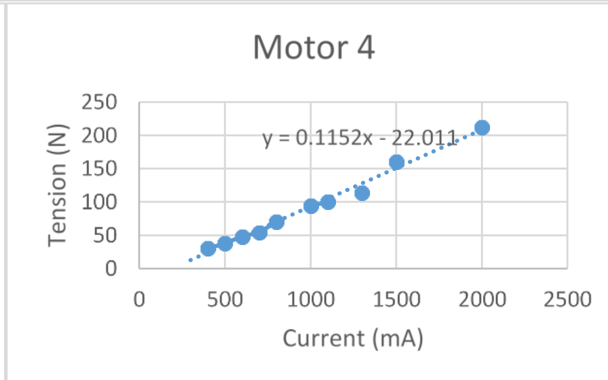
The Stand Trainer operates similar to that explained for the TruST in chapter 2. The system operates using a two stage controller and a tension planner. The two stages, high and low level controllers, are implemented separately for controlling the trunk, pelvis, and knees. The difference lies in the CAN communication between motor drivers and the controller speeds. To account for the large quantity of motors, and the real-time calculation speeds, the low-level controller operates at 500Hz and the high-level at 200Hz.

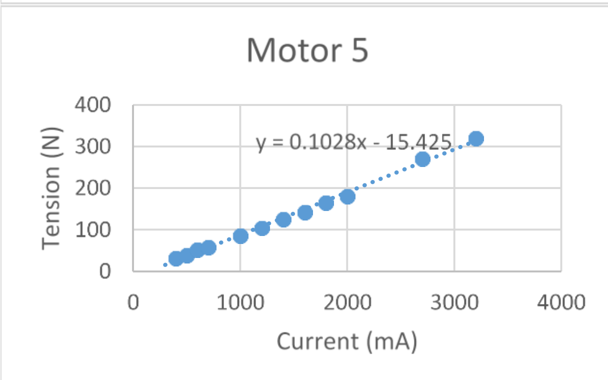
Prior to use, the tension sensors were calibrated and characterized to ensure proper performance. For the motors, a current (mA) and force (N) relationship was created using empirical data. A current was applied and the tension force was measured for each motor and sensor combination. A linear relationship was obtained and the slope was used as the feed forward term, multiplied by the desired force (N). Fig. 5.5 shows the motor constants for 14 motors.

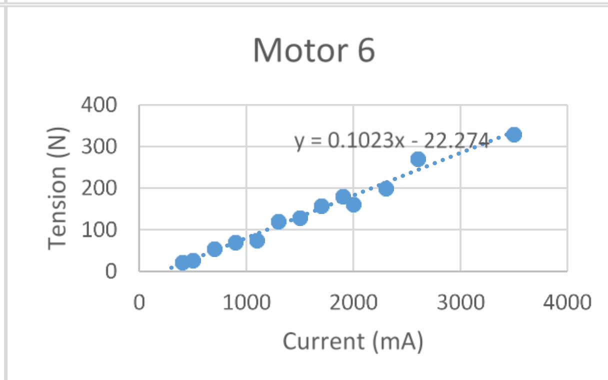

Motor 1
Tension (N)
250
200
150
100
50
0
0
1000
2000
3000
Current (mA)
y = 0.0891x - 12.073


Motor 2
Tension (N)
300
200
100
0
0
1000
2000
3000
4000
Current (mA)
y = 0.0817x + 1.6843


Motor 3
Tension (N)
250
200
150
100
50
0
0
500
1000
1500
2000
2500
Current (mA)
y = 0.0934x - 5.7908


Motor 4
Tension (N)
250
200
150
100
50
0
0
500
1000
1500
2000
2500
Current (mA)
y = 0.1152x - 22.011


Motor 5
Tension (N)
400
300
200
100
0
0
1000
2000
3000
4000
Current (mA)
y = 0.1028x - 15.425


Motor 6
Tension (N)
400
300
200
100
0
0
1000
2000
3000
4000
Current (mA)
y = 0.1023x - 22.274

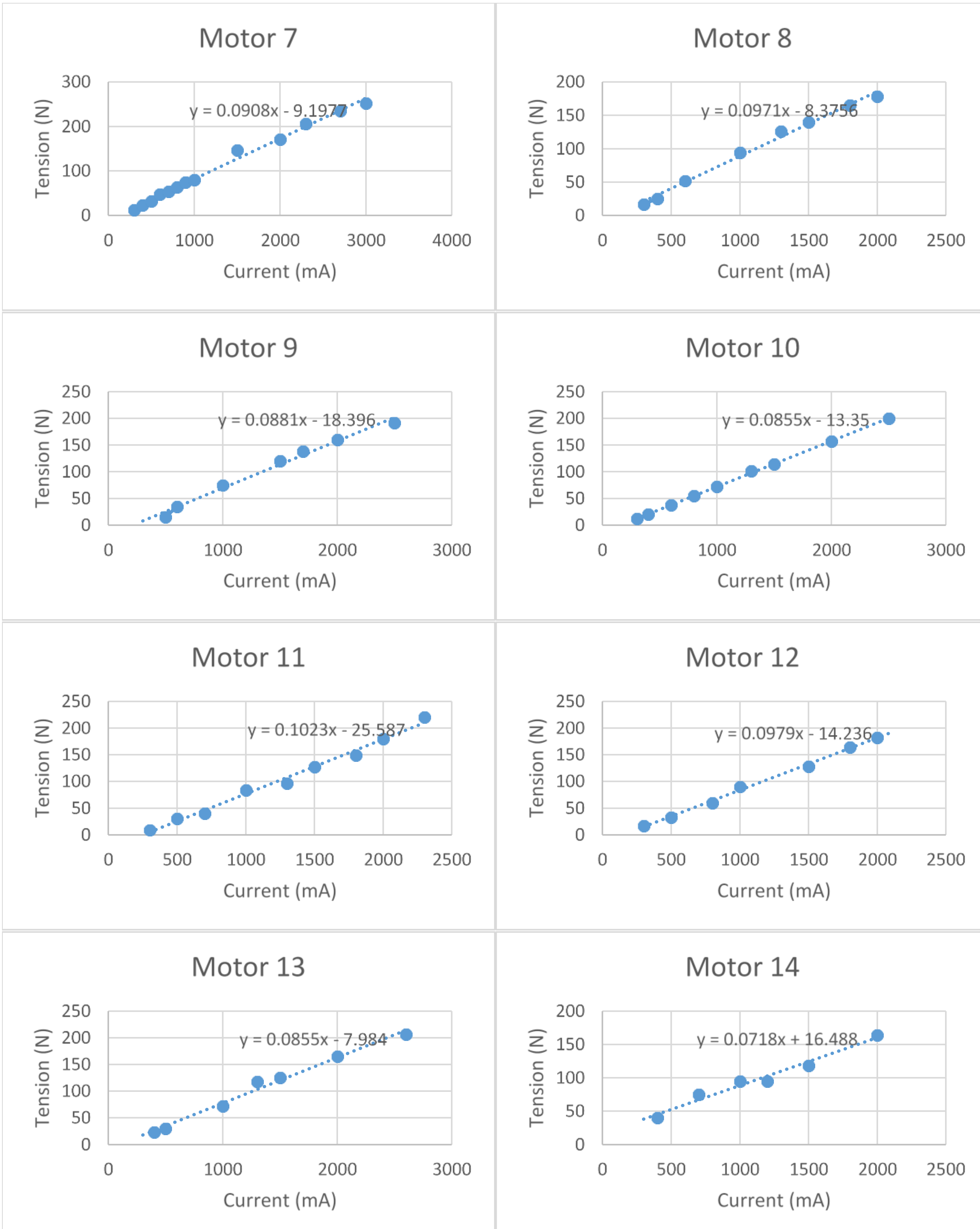


Fig. 5.5 Tension and current relationship modeled empirically for each motor.

5.3 Driver Communication

The device consists of 14 motors/encoders/drivers. We utilized a CAN serial bus to reduce the required number of cables between the drivers and PXI controller. In addition, the CAN protocol allows fault-tolerant real-time communication between the components. We divided the system into two CAN buses for communication, with seven drivers per bus. Using the CAN bus, we can read and write to the drivers using a unique node ID. Each high-speed CAN is set to transmit up to 1Mbits/sec (baud rate). The CAN devices send data across the network in packets called frames. We programmed the CAN bus to operate in stages to confirm communication prior to controlling the robot platform. During initialization, each driver confirms its node ID and enters into pre-operational state with the instructed baud rate and service data objects (SDO). These allow a device to be configured with the communication object dictionary. Each driver enters a disabled operational mode where it becomes capable of reading process data objects (PDO). These carry the information to be exchanged, such as motor current and encoder position. Finally, the drivers enter enabled operational mode where they begin sending PDOs. The bus remains active until it enters the stop state at the end of system use.

The controller can be broken into the same two stages as described for the TruST above. The high-level solver solves for the force vector and plans the cable tensions. The low-level uses a PID to achieve the desired tension by minimizing the errors. There is a separate high and low level for each of the body segments being controlled. This includes the trunk, pelvis, and knees.

5.4 System Validation

We tested the system to validate the end-effector forces. To do this, we fixed a six-axis force torque sensor (FT) at the center (Fig. 5.6). The sensor was mounted with an acrylic plate and cable attachment points (Fig. 5.7). Cable attachments and pulley positions were calculated using Vicon cameras. A desired force was generated with the robot and the accuracy of the tension values and end effector forces were analyzed.

Fig. 5.6 Device setup for system validation. Acrylic plates are mounted to a force-torque sensor for measuring the external forces at the end effector.

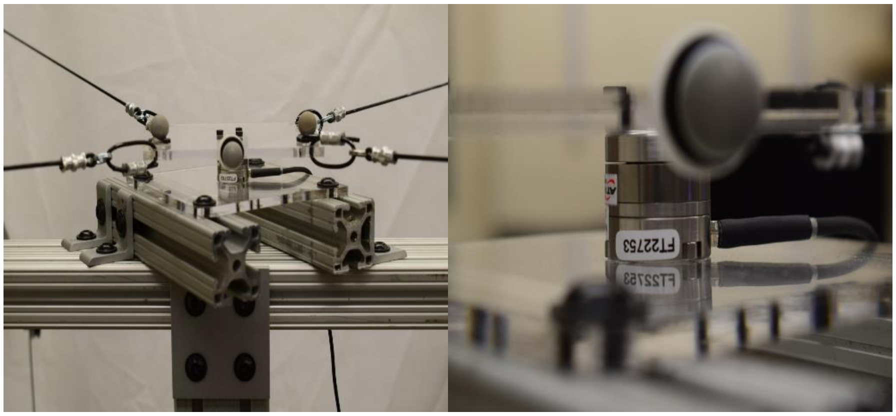


Fig. 5.7 The setup shows the use of a force-torque sensor mounted on a rigid platform for device validation.

For the cable tensions, the average error between the actual and desired tensions was 0.12N. In terms of the Cartesian forces, the average absolute error overall was 1.22% between the desired/actual and 1.73% between the desired and FT. All auxiliary forces were maintained within the desired constraints. This means that we were able to generate a desired force along a specific direction, while keeping forces in all other directions within a specified boundary. The individual breakdown can be seen in Table 5.1. Fig. 5.8 shows that the controller was able to follow the desired force.

ABSOLUTE ERRORS IN CABLE TENSION AND FORCE SENSORS		
Direction	**Current/Desired**	**FT/Desired**
Anterior	0.88%	0.68%
Posterior	1.37%	2.27%
Right	0.79%	1.25%
Left	1.85%	2.72%

Table 5.1 The table shows absolute errors with respect to the desired force at the end effector. These were calculated with comparison between the actual and desired force measured at the cable and the end effector level.

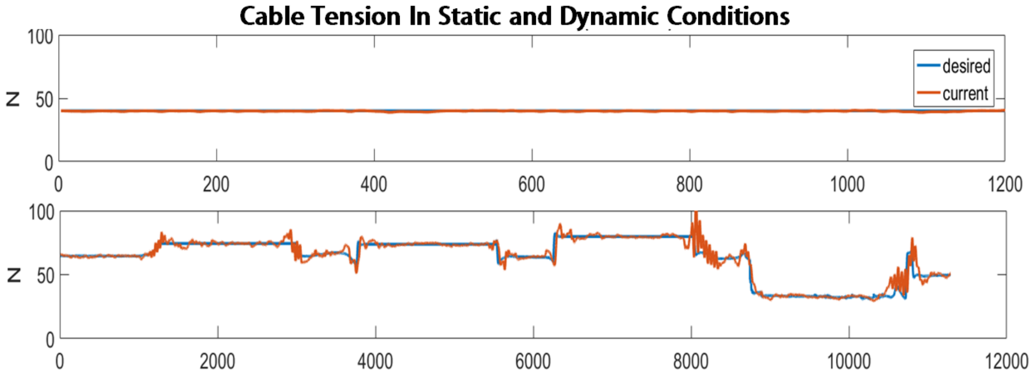


Fig. 5.8 Results show that the cable tensions were tracked accurately by the controller during both static and dynamic movement.

5.5 Force-Field Algorithms

The Stand Trainer is capable of applying various planar and three-dimensional forces. Specifically, the device is programmed to apply various force-field algorithms. It can generate the planar circular force field, two ring planar force field, irregular force field, and the three-dimensional ellipsoidal force field.

The planar force field creates a circle of a given radius around a body segment. If the body is within the circle, no force is applied at the end effector. If the person moves outside the circle, a force perpendicular to the circle is applied to assist the belt back inside the circle. This is done at a fixed body weight percentage magnitude.

The two ring force field uses two rings around the belt. Inside the first circle, no force is provided. Outside each of the rings, a different body weight percentage force is applied to assist the belt to the boundary of the ring. This may be effective for individuals with poor postural control the further they move.

The irregular force field works in a similar way but provides an irregular shape around the person. This shape is defined by the persons asymmetry, giving earlier assistance in certain, smaller workspace quadrants. This is particularly useful for individuals with postural asymmetry.

Finally, the three-dimensional ellipsoidal force field provides assistance in the anterior posterior, lateral, and vertical direction. This can be specified to allow more unassisted motion in a certain direction, while providing earlier assistance in others. Specifically, this also assists the user in the vertical direction, in situation where they are collapsing or falling down. This maybe useful in patient groups with lower limb atrophy or as a fall prevention.

Chapter 6

6 Human Experiments with Stand Trainer

Functional rehabilitation of patients with spinal cord injury remains a current challenge. Training these patients to successfully stand is the first step towards restoring advanced skills such as walking. To address this need, we developed a novel robotic stand trainer described in Chapter 4. The Stand Trainer can apply controlled forces on the trunk and the pelvis of a user, while controlling the knee angle. The Stand Trainer utilizes cables to apply assistive, resistive, or perturbation forces at the trunk, pelvis, and the knees, simultaneously. We have conducted a human study to validate the system. In this study, we applied multi-direction perturbation forces either at the pelvis or the trunk while assist-as-needed forces were applied to the other segment to keep balance. This study characterizes the human kinematics and measures of balance under the perturbations and assistive forces on the human body. Results shows that the level of force-field assistance (trunk or pelvis) directly affects the motion of the trunk, pelvis, and center of pressure. This provides a quantitative framework to restore balance in patients while providing assistance only when needed. This stand trainer can potentially free up therapists to attend to higher level rehabilitation goals and objectively assist patients to engage in interventions that challenge both their musculoskeletal and sensorimotor impairments.

6.1 Assistance and Perturbation Characterization in Healthy Adults

Balance control is essential for performing many every day functional tasks. It allows individuals to maintain their gross posture, while precisely tune the body to the needs of a task. Often posture control is simplified to a single degree inverted pendulum approximation, but recent studies have clarified that balance involves the entire kinematic chain [102]. Balance requires control of postural joints while coordinating the center of pressure (COP) and center of mass (COM) [102].

Impaired balance control increases the risk of falls, which can have serious health consequences (fractures and related complications), and impact health care costs significantly [84], [103], [104]. Poor balance is often accompanied by postural sway with increased intersegmental movement at the head, trunk, pelvis, and legs [16], [80]. Impaired balance control can be due to aging and disuse, as well as neurological and musculoskeletal deficits such as that seen in cerebral palsy, stroke, Parkinson’s, cerebellar ataxia, and spinal cord injury patients.

Balance control can be improved by activity-based training interventions aimed at strengthening lower limbs, balance exercises, and perturbation based training [105]–[107]. General rehabilitation strategies have focused on muscle strengthening of lower extremities or balance training to reduce intersegmental sway and improve functionality. However, studies suggest that postural training that directly targets active balance control could be quite beneficial for motor learning [16].

Many patient populations with balance and coordination deficits are trained using manual manipulation. Often the focus of these training strategies is to strengthen muscles and engage the ankle and knee joints. The rationale is that these are important for maintaining standing equilibrium. Specifically, researchers of Parkinson's and stroke have shown that balance training improves stability and gait reducing the risk of falling[108], [109]. In research with the elderly, computerized tests with foam or moving platforms have been used for balance assessment [110].

More severe neurological deficits such as motor complete SCI lead to paralysis, and compromise motor responses for postural control. Recent studies on complete animal models and humans suggest that activity-based training coupled with spinal cord epidural stimulation can promote partial recovery of postural control [111]–[113] . This can be of particular interest because recent findings suggest that standing and walking with assistance for balance control can be recovered after severe SCI by the combination of spinal epidural stimulation and activity-based training [112]. Therefore, a facility for multi-modal training that constrains, assists, perturbs or modulates stability could be important for task-oriented therapy in SCI.

Achieving postural equilibrium requires coordination of multiple body segments and postural strategies to stabilize the body COM during self-initiated reaching or external disturbances [114]. The strategies depend on the postural displacements, individual's expectations and goals [114]. Studies suggest that programs that involve motor learning principles in active balance control are more effective [16]. Furthermore, active training that directly controls a person's center of pressure (COP) have beneficial impact on postural orientation and stability [81].

The goal of rehabilitation is to advance patients along their course to regain balance, strength, and coordination, while moving from sitting, standing, to walking. The therapies may have aspects of strengthening through resistance, haptic feedback, proprioceptive and coordination training, and cognitive enrichment. While manual training has shown benefits in patients with SCI, it can require multiple therapists, induce training variability, and reduce motor learning.

In this work, we have created a novel full-body cable robot for quantitatively engaging the trunk, pelvis, and the knees for training of balance and coordination. Our system is called the Stand Trainer and it can apply assistive, resistive, or perturbation forces on the trunk and pelvis. This system builds upon our previous research on cable systems designed for training of sitting, standing, and walking [17], [29], [87], [99]–[101]. The system utilizes patient specific assist-as-needed force-fields while facilitating active postural recovery. In addition, the system can apply postural disturbances and train balance/coordination. In this work, we recruited 10 healthy adults to characterize the differences in COP and trunk/pelvis coordination, with and without an assistive force field at the trunk or the pelvis. We hypothesize that our system can reduce postural sway by use of a force-field. We further plan to explore the effects of perturbations with assistance on different body segments in terms of kinematics. Our results show that an assistive force field during postural perturbations decreases trunk and pelvis movement and COP trajectory excursions. The results of this study are beneficial in understanding the effects of assistance at different body segments and can provide insights into designing new training paradigms for different patient population groups.

6.1.1 Protocol

The research protocol was approved by Columbia University's institutional review board. Ten healthy adults (7 males, 3 females) with average age: 27y, weight: 161lbs and height: 68in. were recruited. All subjects were right dominant. Each subject underwent four conditions with a combination of perturbations and assistive force field. As shown in Fig. 6.1, the conditions included trunk perturbation (Tpert), trunk perturbation with an assistive pelvis force field (Tpert-Pff), pelvis perturbation (Ppert), and pelvis perturbation with assistive trunk force field (Ppert-Tff). For simplicity of force application on the body and to minimize the number of cables along each body segment, we applied the perturbations and assistance separately to either the trunk or pelvis. The perturbations represented both internal disturbances, as with a patient group, and added disturbance. Our choice of conditions allowed us to test if: 1) force field assistance is sufficient to minimize sway and 2) determine which combination of assistance/disturbance would provide the largest kinematic constrain or the most challenging condition.

Vicon motion capture system was used to record kinematics sampled at 100Hz. This was used to detect the real-time center position of the trunk and pelvis during training. Retro-reflective markers were placed on the trunk and pelvis belt to determine the estimated centers.

Two separate belts were placed on each subject, at the trunk and pelvis. Four planar cables were connected to each belt. The subjects stood on a force plate with feet apart. The subjects were told to raise their hands up and to the side for consistency in data collection. The subjects were instructed to move their hands as needed, without touching or holding on to surrounding structures.

In each of the four conditions, the subjects received a total of eight perturbations, two in each direction: anterior (N), posterior (S), and right lateral (E) and left lateral (W). The perturbations were randomized and occurred with a random time spacing. The subjects responded to the perturbations and returned back to their starting position after each perturbation. The perturbation was characterized as a 20% body weight trapezoidal force that had a 0.5s rise time, 0.5s constant value, and 0.5s fall time. The total force lasted 1.5s from start

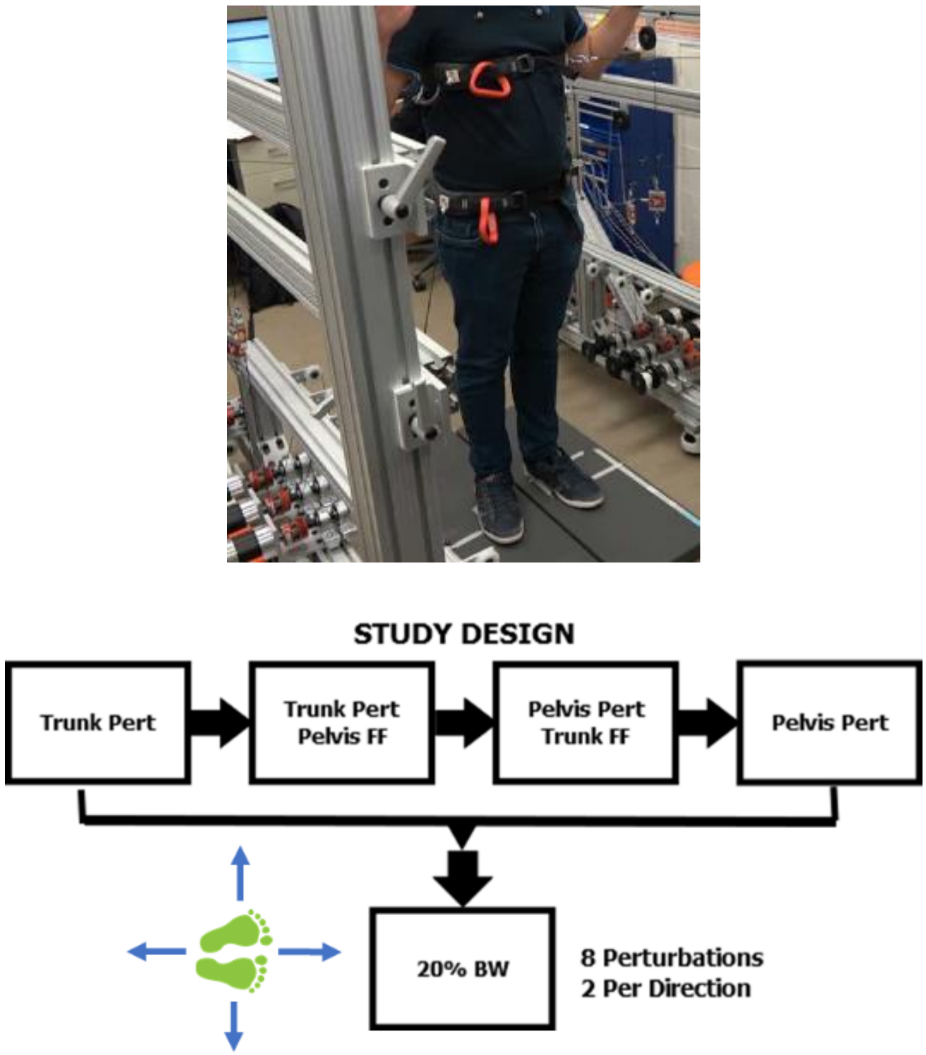


Fig. 6.1 The experimental design and the setup with two belts - one at the trunk and the other at the pelvis.

to finish. Our goal was to provide a force that was gradual and allowed time to initiate postural adjustments necessary to maintain a stable configuration. We also wanted the force to produce postural sway as one begins to lose balance, as opposed to be an unexpected “push”. We chose a percentage of body weight that was sufficient to challenge the user to adjust their kinematics to maintain stability. The choice of the perturbation force was determined after multiple trials.

The body level assist-as-needed force field was provided using motors and load cells [29], [87]. No force was applied if the subject stayed within a predetermined workspace. As the lower trunk moved out of the stability boundary, assistive forces were applied equivalent to 10% body weight to assist in movement towards the stability boundary. The force field provided a virtual circle of radius 10cm. This was determined based on anthropometric data reported in [115]. The 50^{th} percentile foot size is roughly 20cm. With a virtual circular force field of radius 10cm, we can assist the COM to remain within the BOS (boundary of the feet) in the anterior/posterior direction.

6.1.2 Data Analysis

Performance variables were analyzed via kinematics. Data was collected and low-pass filtered with a 2nd-order Butterworth filter. Angular motion of the trunk and pelvis was computed following an Euler sequential rotation x-y’-z’’ and right-handed convention (flexion (+)/extension(-); right latero-flexion (+)/left latero-flexion (-); left rotation (+) and right rotation (-). MATLAB (MathWorks, Natick) was used for data processing. For the analysis, we divided the perturbations into cycles from the onset of the force to the offset (1.5s) and an additional (0.5s) response time. These 2s trials were then grouped separately based on the direction of the perturbation. All perturbations in the same direction were grouped and averaged. The statistical analysis was conducted using SPSS 24 (IBM, Chicago, Illinois). Alpha rate was set at 0.05 to test significant differences with 95%CI. A one-way repeated measures (within subjects) Analysis of Variance (ANOVA) across 4 experimental conditions (A: Tpert, B: Tpert=Pff, C: Ppert & D: Ppert-Tff) was applied to test variables across directions (posterior & non-dominant lateral) that were dependent on on the maximum position of the body segment (max trunk and max pelvis). Bonferroni's inequality procedure was used to correct the P-values based on the number of comparisons. Post-hoc analyses were carried only if the ANOVA model was significant ("Omnibus Test").

6.1.3 Results

We analyzed the data comparing the various conditions of perturbations. We wanted to identify the changes in human performance when the perturbation was applied to the trunk versus the pelvis, and with and without a force-field constraint at the opposite level.

Fig. 6.2 shows the area encompassing the overall area of movement for the trunk and pelvis *geometrical* center in each test condition. There was a significant effect for trunk area [$F(3,27) = 24.3$, $P < 0.001$] and pelvis area [$F(3,27) = 17.1$, $P < 0.001$]. This takes into account eight consecutive perturbations for each, two in each of the N, S, E, and W, directions. For trunk, the results show that Tpert produced the highest area of movement compared to other conditions (M=273.8cm^2, S=81.6, P<0.05). The pelvis area was similar with Tpert (M=213.5 cm^2, S=56.7) and Ppert (M=199.9 cm^2, S=75.8). Yet, both trunk and pelvis area decreased with the use of a force field (P<0.05). For the trunk, using a force field assistance during trunk perturbations decreased the area by 43.6% (P<0.05) and during pelvis perturbations by 60% (P<0.05). For the pelvis, using a force field assistance during trunk perturbations decreased the area by 61.3% (P<0.05) and during pelvis perturbations by 54.4% (P<0.05).

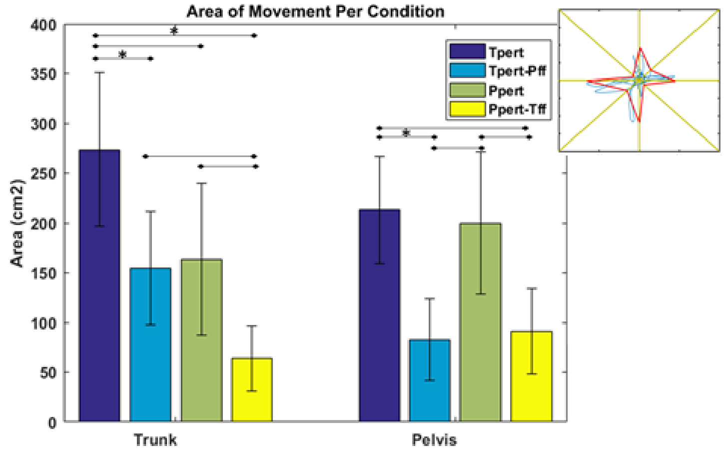


Fig. 6.2 Area of movement after eight perturbations under each condition. A sample data is shown below, identifying the boundary of movement. Lines indicate significant difference (P<0.05), where *<0.001).

We also analyzed the overall distance (path or trajectory) travelled by the trunk center, pelvis center, and the COP (RCOP-right force plate COP, LCOP-left force plate COP, and ACOP-average COP) during posterior and non-dominant side perturbations. There was a significant effect within subjects for trunk trajectory [$F(3,27) = 4.16$, $P < 0.001$] and pelvis trajectory [$F(3,27) = 24.7$, $P < 0.001$]. The results show that Tpert produced the longest path for the trunk and pelvis (M=35.5cm, S=5.0, P<0.001) and COP (M=7.4cm, S=3.1, P<0.05) compared to all other conditions. Applying a forcefield on the pelvis decreased trunk path (P<0.05), while there was no difference between Ppert with and without assistance. In terms of the pelvis, force field on the trunk or pelvis decreased the pelvis path (P<0.05). The COP path was longest with Tpert compared to all other conditions (P<0.05). In terms of non-dominant side perturbations, Tpert caused the trunk to travel the longest path compared to all other conditions (P<0.05). Pelvis assistance decreased trunk and pelvis trajectory during Tpert (P<0.05).

Fig. 6.3 shows the trunk and pelvis trajectory during the posterior and non-dominant lateral perturbations. The dark middle of each plot shows the average trajectory while the shaded shows the variance. It is seen that the variance in the perturbation response increases over time. The larger the displacement in the posterior or lateral direction, the more variance in compensation strategy. The trunk perturbation shows the most variance and largest displacement as was seen with Fig. 6.4. The time to maximum translation remains similar for all conditions (P>0.05).

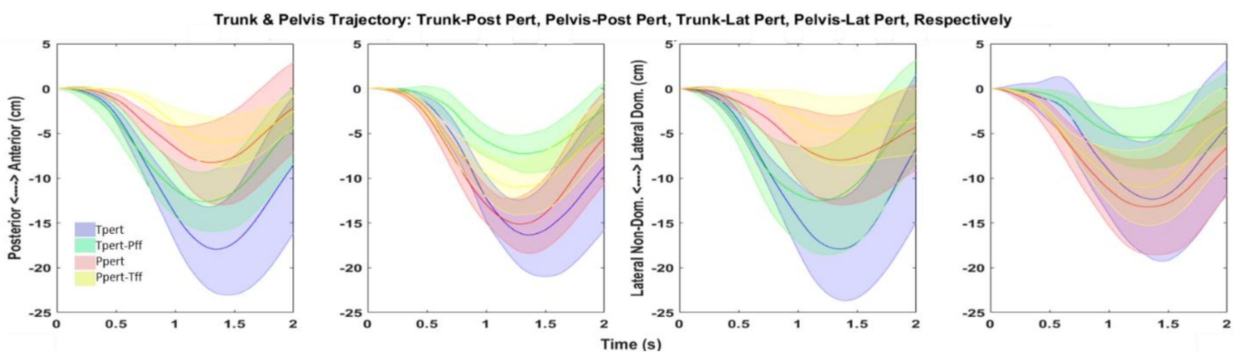


Fig. 6.3 The trunk and pelvis trajectories are shown during posterior and lateral perturbation in each of the experimental conditions. Trunk pert creates the largest postural deviations while the assistive conditions constrain the motion.

Fig. 6.4 shows the trunk and pelvis displacement in the direction of the posterior and non-dominant side perturbation at maximum segment translation. There was a significant effect within subjects for trunk [$F(3,27) = 34.4$, $P < 0.001$] and pelvis [$F(3,27) = 19.5$, $P < 0.001$] posterior perturbations, and also trunk [$F(3,27) = 29.8$, $P < 0.001$] and pelvis [$F(3,27) = 8.0$, $P = 0.001$] non-dominant perturbations. During posterior perturbation, the trunk translates most with Tpert compared to any other condition (M=20.1cm, S=5.1, P<0.05). Tpert-Pff decreases trunk translation (M=13.1cm, S=3.7, P<0.05), while there is no difference in the trunk translation when perturbing at the pelvis with (Ppert-Tff, M=6.1cm, S=2.7) and without trunk assistance (Ppert, M=9.2cm, S=5.0). Non-dominant side perturbations follow the same performance. Pelvis shows no difference in translations when the posterior perturbation is at the trunk (Tpert, M=17.2cm, S=4.2) or pelvis (Ppert, M=15.2cm, S=3.3). Pelvis translates less with trunk perturbations with pelvis assistance (Tpert-Pff, M=7.4cm, S=2.4, P<0.05) compared to no assistance (Tpert). Non-dominant side perturbations follow the same performance.

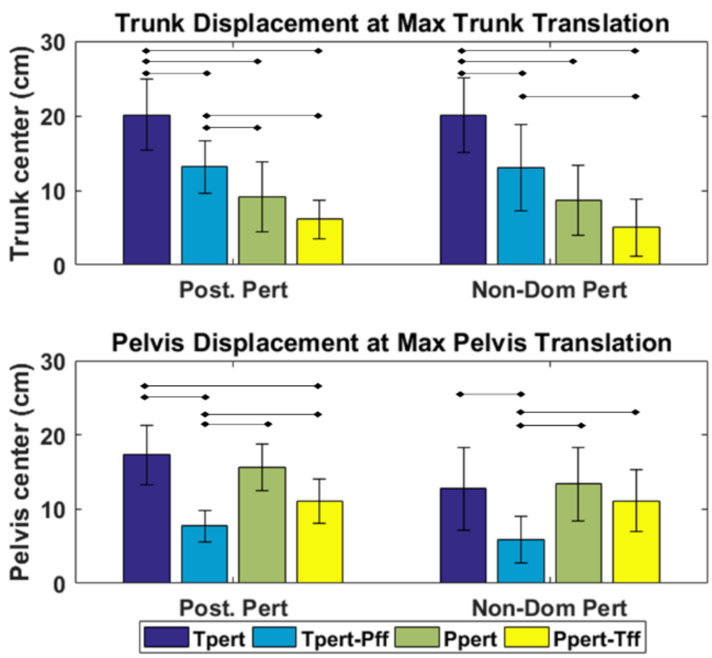


Fig. 6.4 Trunk and pelvis displacement at max translation in the direction of the perturbation. Lines indicate significant difference (P<0.05).

Fig. 6.5 shows the trunk and pelvis rotations in the direction of the posterior and non-dominant side perturbation at maximum segment translation. There was a significant effect within subjects for trunk [$F(3,27) = 9.9$, $P < 0.001$] and pelvis [$F(3,27) = 8.1$, $P = 0.001$] posterior perturbations, and also trunk [$F(3,27) = 7.7$, $P = 0.001$] and pelvis [$F(3,27) = 30.7$, $P < 0.001$] non-dominant perturbations. Pelvis perturbation (Ppert, M= 21.6^0, S=14.2) provided the largest change in trunk angle relative to the pelvis during posterior perturbation. This was significantly decreased with trunk assistance (Ppert-Tff, M=5.2^0, S=7.0, P<0.05). During posterior perturbation, the pelvis flexed more with trunk pert (Tpert, M=10.2^0, S=7.9, P<0.05) than pelvis pert (Ppert, M= 0.6^0, S=3.7, P<0.05). During non-dominant lateral perturbation, the trunk rotated right (dominant side) the least with trunk perturbation and pelvis assistance (Tpert, M=1.6^0, S=3.4). The pelvis rotated in the direction of the perturbation, and the most with just trunk perturbations (Tpert, M=-14.1^0, S=5.8, P<0.05) compared to all other conditions. The pelvis was most constrained with pelvis perturbations with trunk assistance (Ppert-Tff, M=-0.7^0, S=1.7).

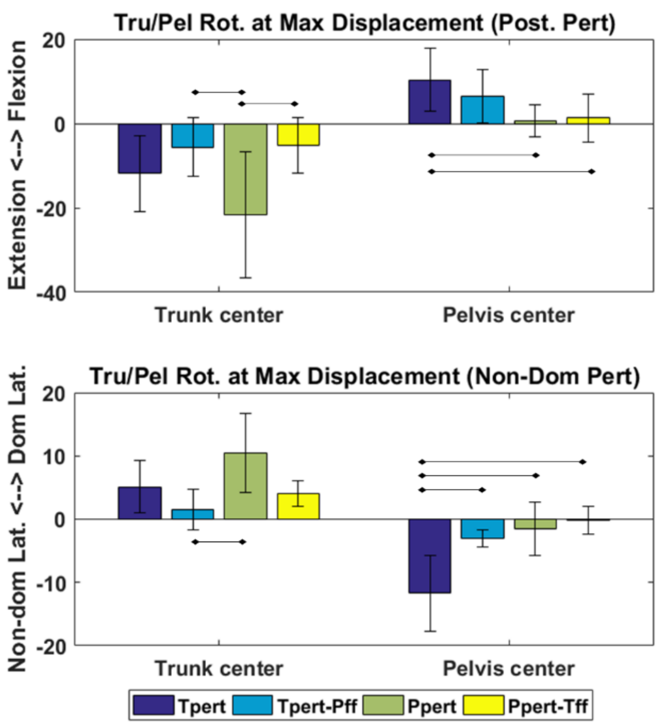


Fig. 6.5 The trunk and pelvis rotations are shown during posterior and lateral perturbation in each of the experimental conditions

In Table 6.1, we outline the number of steps a subject took with each condition. There was a significant effect within subjects for posterior perturbations [$F(3,27) = 7.2$, $P = 0.001$] and non-dominant lateral perturbations [$F(3,27) = 10.0$, $P < 0.001$]. During posterior perturbations, the largest number of steps were taken during trunk perturbation (M=1.7, S=0.7, P<0.05). This was decreased when a pelvis force field was applied (M=0.6, S=0.8, P<0.05) and furthermore, during pelvis perturbations with trunk assistance (M=0.4, S=0.8, P<0.05). In the non-dominant direction, the trunk perturbations also produced the largest number of steps (M=1.8, S=0.6, P<0.05). This was significantly reduced with pelvis perturbations (M=0.7, P=0.9, P<0.05) and furthermore with pelvis perturbations and trunk assistance (M=0.1, S=0.3, P<0.05).

	Average No. of Steps with Pert. (max 2)			
Pert Dir.	**Tpert**	**Tpert-Pff**	**Ppert**	**Ppert-Tff**
Post.	1.7**	0.6*	0.9	0.4*
Non-Dom Lat.	1.8**	0.9	0.7*	0.1*

Table 6.1 Average number of steps taken during perturbations in each of the conditions. The maximum number possible is two. Both pelvis and trunk assistance decrease the number of steps required to maintain stability. * indicate significant difference (P<0.05).

		Maximally Constrain				Maximally Challenge			
Goal	**Variable Assesed**	**Tpert**	**Tpert-Pff**	**Ppert**	**Ppert-Tff**	**Tpert**	**Tpert-Pff**	**Ppert**	**Ppert-Tff**
Overall trunk sway	Area				X	X			
Overall pelvis sway	Area		X		X	X		X	
Posterior trunk sway	Total distance			X	X	X			
Posterior pelvis sway	Total distance		X		X	X		X	
Non-dom lateral trunk sway	Total distance			X	X	X			
Non-dom lateral pelvis sway	Total distance		X			X		X	
Maximally decrease trunk or pelvis COP path	Total distance		X	X	X	X	X		
Trunk translation during posterior perturbation	Translations			X	X	X			
Pelvis translation during posterior perturbation	Translations		X			X		X	
Trunk translation during non-dom lateral perturbation	Translations			X	X	X			
Pelvis translation during non-dom lateral perturbation	Translations		X			X		X	X
Trunk rotation in flex/ext	Rotations	X	X		X	X		X	
Pelvis rotation in flex/ext	Rotations		X	X	X	X	X		
Trunk rotation in lateral	Rotations	X	X		X	X		X	X
Pelvis rotation in lateral	Rotations		X	X	X	X			
Total Count	Overall	2	10	7	12	15	2	7	2

Table 6.2 Summary of the findings showing the best condition to challenge or maximally constrain a specific movement. The overall analysis shows that Ppert-Tff is best to constrain movement, while Tpert produces the most challenging conditions.

6.1.4 Discussion

In this study, we characterized trunk and pelvis center and COP movement patterns during perturbations. We compared the changes in the movement when subjects were provided force field assistance. Our results show that the force field adequately decreases trunk and pelvis displacement during perturbations. Specifically, a force field assistance on the trunk will decrease the effect of a pelvis perturbation and a force field assistance on the pelvis will decrease the effects of a trunk perturbation. The overall area of movement (trunk and pelvis) is the largest with trunk disturbances compared to pelvis. This fits our hypothesis since disturbances at a higher body segment creates a larger moment arm at the feet. Ankle torques are the initial contributors for standing balance [116], and once the disturbance is too large, the ankles cannot sufficiently counter the postural imbalance. The force field limits the range of movement of the pelvis, which decreases the overall trunk movement. Since the force field is on the pelvis, we are directly constraining the pelvis. Therefore, largest decreases are seen at the pelvis. Applying a force field at the trunk further reduces the overall movement of the trunk compared to that of the pelvis.

The summary of the findings can be found in Table 6.2. This table shows several conditions and outlines the best training conditions to either maximally constrain or maximally challenge certain measurements. The X identifies the condition that either provided the maximum constrain to the variable (e.g. area of movement, etc.) or maximally challenged the individual for that variable. If the conditions had no significant difference, then multiple X are shown to signify that multiple conditions are appropriate. The summation of the results shows that best overall condition to constrain kinematic and biomechanical requirements is to use condition Ppert-Tff. This is where the assistance is provided at the trunk and perturbations are applied to the pelvis. In contrast, the best overall condition to maximally challenge a person is Tpert, where perturbations are applied to the trunk with no assistance.

The trunk and pelvis motion are most affected when a perturbation or disturbance is applied to the trunk. When the trunk is perturbed, and the pelvis is stabilized using an assistive force field, the trunk and pelvis motion both decrease significantly. It can be assumed that physically locking the trunk while applying a trunk perturbation will stop the disturbance. Though apparent, this does not allow the subject to actively control the segment of interest. On the other hand, the trunk moves significantly less with perturbation at the pelvis compared to those directly at the trunk. Yet, there is no difference to the pelvis movement if the perturbation is applied at the trunk or pelvis.

We also see that a pelvis constraint can significantly decrease the number of steps a person takes to account for a posterior trunk perturbation. A pelvis posterior perturbation with trunk constraint will also significantly reduce the number of steps one takes, compared to trunk perturbation. In terms of non-dominant lateral side perturbation, pelvis perturbation causes significantly less stepping than trunk perturbation. Furthermore, pelvis perturbation with trunk assistance produces significantly less stepping that pelvis pert alone. Recent studies have reported strategies that humans use to maintain standing equilibrium [114], [117]–[119]. In situations where there is a COM shift due to a sudden disturbance or a required functional task, the COP shifts in the opposite direction to maintain equilibrium. These postural adaptations get optimized to changes in environment, tasks, and subject's intention [116]. Specifically, to maintain postural equilibrium during sudden COM shifts, humans may exert a higher torque at the ankle or the hip, or use a combined strategy [116]. The ankle strategy uses torques about the ankle to control the anterior-posterior sway during quiet stance or slow body translations. This strategy moves the COP further beyond the COM. In cases of sudden or large amplitude perturbations, flexion of the trunk at the hip joint allows for early activation of the abdomen and quadriceps muscles. If a disturbance is too large in amplitude and fast, a stepping reaction may be used to move the base of support (BOS) under the falling COM. While ankle strategy is efficient to maintain an erect posture during quiet standing, the hip strategy is often used during large amplitude perturbations. If a step is not taken, then hip flexion is coupled with ankle and neck counter rotation. In addition, in situations where the floor surface has low friction and does not allow for large torques, early activation is produced at the rectus abdominis and rectus quadratus (neck flexors), with little ankle coactivation [116]. This specific outcome addresses the safety of our stand trainer system to prevent falls while it is applied to train trunk and pelvis stability in standing.

Our results show that assistive force fields, applied by our robotic platform, can significantly constrain pelvis and trunk motion. This is different than applying a rigid constraint as it provides a resistance to motion rather than rigidly locking a segment from moving. An individual is required to adjust various body segments such as the trunk, pelvis, and knees, to return to their stable configuration. Our results help us understand the kinematics and physical mechanisms by which we can alter the movement of an individual with internal trunk or pelvis disturbances due to balance deficits. The applied trunk and pelvis perturbations can represent the inherent disturbances one may have at the trunk or pelvis level. To minimize movement of a segment (trunk or pelvis), it may be best to apply a constraint directly to that region.

One of the primary benefits of our system and its use for training is its modular design and patient-specific assistance. The boundary of assistance around a subject (force field) can be easily adjusted to provide a larger or smaller range of independent movement. As the patient is able to increase the range of trunk motion, the force field radius can be increased. This will allow the subject to travel to a further distance before any force assistance is provided. In cases like SCI, in which people have very limited trunk control, we can also make the radius very small so the patient can successfully sway and configure their posture without being rigidly constrained. In addition, we can adjust the magnitude of assistance that is provided once the patient becomes unstable or goes outside the stability boundary. With the force field, we do not actively bring a subject upright, but provide a body weight percentage of assistance to prevent the subject from failing. This can give the patient increased time in adjusting their postural strategy for maintaining balance. The system also provides haptic feedback at the boundary of stability, signaling to the patient that they are at their limits. At this point, the subject can physically push past the boundary or configure themselves back to the neutral, upright position. As we mentioned in the introduction, studies show that the magnitude and speed of disturbance change the balance strategy from the ankle/feet, to the hip, or stepping strategy. With our system, we can provide assistance objectively, based on the patient's needs.

Our design largely stems from the need in rehab settings to implement balance training programs in patients with severely limited or inexistent volitional postural control such as SCI. As medical advancements are made to enable patients to regain control from paralysis or hemiplegia, training strategies are required to safely but independently explore postural configurations for maintaining upright stability. Our previous work on sitting balance training has shown that subjects can change their kinematics and movement strategies with assist as needed training [29], [99]. We have incorporated these capabilities into a full-body system that can provide an objective assessment of various training strategies. Our pilot study with healthy subjects show that the assistive nature of the device can significantly decrease the movement of the trunk and pelvis. In our future work, we plan to test the capabilities of the system on spinal cord injury patient populations and then initiate a longitudinal study for training patients with postural instability. We also plan to assess the effects of assistance and disturbance at the same level in both healthy and patient groups. We believe this system will be instrumental in conducting balance therapy, freeing therapists from hard manual labor, and providing a combination of neuromuscular and coordination improvements for patients with postural instabilities.

The goal of rehabilitation is to propel patients along their course to regain balance, strength, and coordination, moving from sitting, standing, to walking. Therapeutic modalities include strengthening through resistance, haptic feedback, proprioceptive and coordination training, and cognitive postural aspects. While manual patient manipulation has shown benefits, it can require multiple therapists, induce training variability, and reduce motor learning.

In this work, we have created a novel robotic system, the Stand Trainer, for rehabilitation of postural standing. In this study, we show the potential for objective assessment and selective training of stability of key body segments (trunk and pelvis) to maintain standing. Our system consists of trunk, pelvis, and knee belts for providing assistive, resistive, or perturbation forces. In this study, we recruited 10 subjects to assess the changes in trunk and pelvis movement and COP, when subjected to perturbations. This was conducted by providing assistance at two different levels, trunk and pelvis, and with and without assistance from a force field. Our results support our hypothesis that an assist-as-needed algorithm can constrain kinematics and COP movements. This is essential in understanding how to assist patients with various levels of weakness, without passively constraining their motion and coordination. In addition, our study identifies the mechanics used by healthy individuals to maintain stability. This gives us a comparison for characterizing the difference in patient strategies and will be instrumental in designing and training patient populations.

6.2 Characterization of Incomplete and Complete Spinal Cord Injury Patients

Spinal Cord Injury, or SCI, is a multi-systemic condition characterized by muscle paralysis and deficits in the cardiopulmonary, integumentary, gastrointestinal, genitourinary, and sensory systems. Changes in these functions can reduce mobility and quality of life [120]. There are roughly 270,000 individuals in the United States with a SCI, and 12,000 new cases yearly [121],

[122]. Injuries can be classified as complete (cSCI) or incomplete (iSCI). This is in reference to the spinal cord, which carries information to and from the brain and the extremities. In incomplete cases, part of the spinal cord is severed, creating various changes to motor and sensory function. In cases of complete injury, all motor and sensory information below the lesion is lost. The level of injury determines which sensory and motor functions remain intact. The higher the injury on the spinal cord, the more detrimental are its effects. Spinal cord injuries are classified using the ASIA Impairment Scale to grade the severity of neurological loss. This ranges from A, complete SCI where there is no motor or sensory function, to B-D, incomplete SCI where some motor and sensory function is impaired, to finally E, meaning motor and sensory function are normal.

Many researchers in the area believe that the dysfunction stems from impaired neural control of the involved musculature coupled with decreased sensory information being transmitted to the brain [121]. A majority of individuals with SCI do not recover functional walking, and there has been very little progress in rehabilitative strategies [121]. Current rehab strategies focus on providing compensatory strategies for improving mobility and strength above the level of the lesion or injury. More recent studies suggest that training which utilizes sensory information associated with motion may provide some benefits [121].

Recent studies on complete SCI animal models and humans suggest that activity-based training coupled with spinal cord epidural stimulation can promote partial recovery of postural control [111]–[113]. In particular, Angeli et. al's findings suggest that assistive training to improve balance control may recover standing and walking capabilities after severe SCI when combined with spinal epidural stimulation and activity-based training [112]. Therefore, a facility for multi-modal training that constrains, assists, perturbs, or modulates stability could be important for task-oriented therapy for individuals with a SCI.

Achieving postural equilibrium requires coordination of multiple body segments as well as postural strategies which stabilize the body's COM during self-initiated reaching or external disturbances [114]. The strategies depend on the postural displacements, individual's expectations, and the goals of the task [114]. Studies suggest that programs which involve motor learning principles in active balance control are more effective [16]. Furthermore, active training that directly influences a person's COP has a beneficial impact on postural orientation and stability [81].

In our work, we are interested in characterizing the motion of individuals with a SCI under several assistive and perturbation-based conditions. The purpose is to understand how assistance at the trunk or pelvis can affect the motion of the trunk, pelvis, and COP. In addition, we measure the trunk workspace changes to document the movement area under each of our conditions. This provides a baseline comparison between healthy adults, incomplete SCI, and complete SCI individuals. To perform this training, we utilize our Stand Trainer device.

6.2.1 Incomplete SCI Protocol

After obtaining protocol approval from Columbia University's Institutional Review Board, we recruited one patient with an incomplete spinal cord injury at the L4-5 level, AIS C. This was a 64-year-old male, 6'2" in height, and weighed 190 lbs. The patient participated in the same protocol defined previously in the healthy adult stand trainer study.

The patient underwent four conditions with a combination of perturbations and assistive force fields. The conditions included trunk perturbation (Tpert), trunk perturbation with an assistive pelvis force field (Tpert-Pff), pelvis perturbation (Ppert), and pelvis perturbation with assistive trunk force field (Ppert-Tff). For simplicity of force application on the body and to minimize the number of cables along each body segment, we applied the perturbations and assistance separately to either the trunk or pelvis (Fig. 6.6). The perturbations represented both internal disturbances, as experienced in a patient group, and added external disturbances. Our choice of conditions allowed us to: 1) test if force field assistance is sufficient to minimize sway and 2) determine which combination of assistance/disturbance would provide the largest kinematic constraint, or the most challenging condition.

A Vicon motion-capture system was used to record body kinematics sampled at 100 Hz. This was used to detect the real-time center positions of the trunk and pelvis during training. Retro-reflective markers were placed on belts firmly attached at the trunk and pelvis to estimate the center of each body region. Four planar cables were connected to each belt. The patient was instructed to move his hands as needed, without touching or holding on to surrounding structures.

In each of the four conditions, the subject received a total of eight perturbations, two in each direction: anterior (N), posterior (S), and right lateral (E) and left lateral (W). The perturbation directions were randomized and occurred with a random time spacing to ensure the patient did not have predictive motion. The subject responded to the perturbations and returned to the starting position after each perturbation ended. The perturbation profile was characterized as a 20% body weight trapezoidal force that had a 0.5 sec rise time, 0.5 sec constant value, and 0.5 sec fall time. The total force lasted 1.5 sec from the initial onset.

Fig. 6.6 Incomplete SCI patient setup with the neutral and perturbed posture shown, respectively.

6.2.2 Results

The iSCI patient underwent a series of eight randomized perturbations, two in each of the anterior, posterior, and both lateral directions. The total movements of the trunk and pelvis centers were measured to determine the overall workspace, or area of movement (Fig. 6.7). It was evident that both the trunk and pelvis showed the largest movement under the Tpert condition. The trunk was the most constrained during the Ppert-Tff condition. Similarly, the pelvis was most constrained during the Tpert-Pff condition.

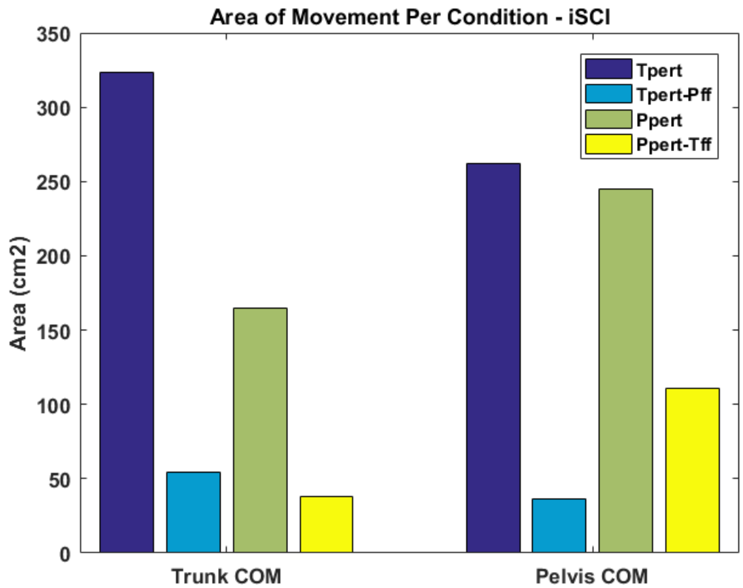


Fig. 6.7 Trunk and pelvis area of movement during each of the four conditions for the iSCI patient.

The effects of the posterior and non-dominant lateral perturbations were analyzed individually for the trunk and pelvis in each experimental condition (Fig. 6.8). The pelvis perturbation conditions yielded the largest path travelled by the trunk in the posterior and lateral directions and by the pelvis in the lateral direction. Conversely, when the pelvis was applied with a force field assistance (Tpert-Pff), the trunk and pelvis were more constrained than all other conditions.

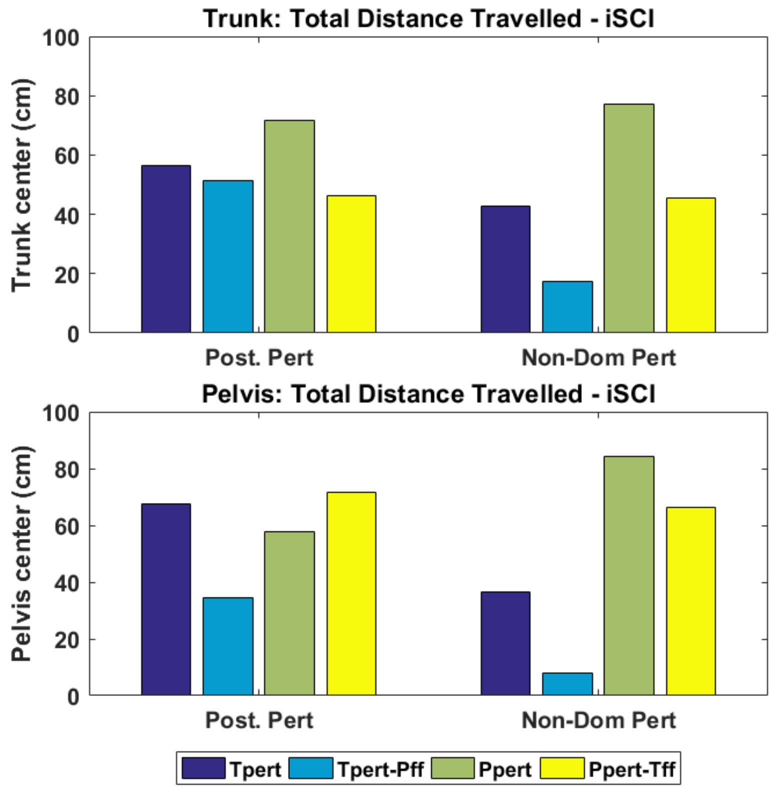


Fig. 6.8 Total distance travelled by the trunk and pelvis center during each of the four conditions, during posterior and lateral perturbations.

When looking at the maximum distance travelled by the trunk and pelvis in the posterior direction, Ppert condition produced the largest deviation (Fig. 6.9). On the other hand, there was no noticeable movement during Ppert-Tff condition with posterior perturbations. This may be due to an anticipatory reaction.

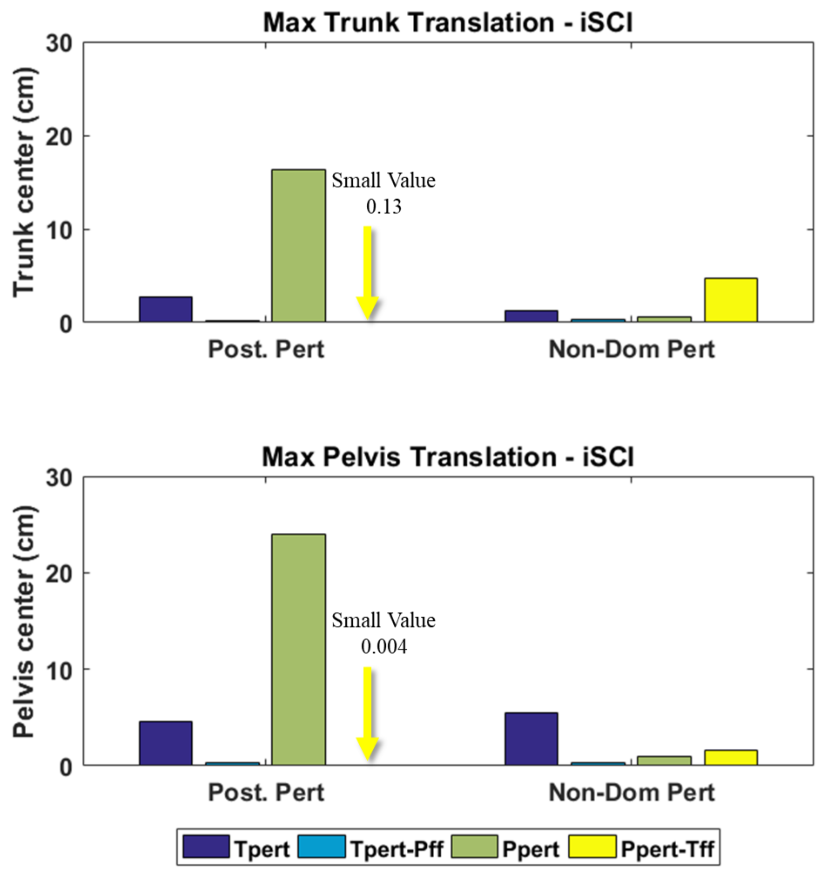


Fig. 6.9 Max translation by the trunk and pelvis center during each of the four conditions, during posterior and lateral perturbations.

The purpose of this study was to assess the kinematics of an iSCI patient under various assistance and perturbation conditions. The aim is to assess the changes in kinematics from the force field assistance and to compare the performance to healthy subjects. In the iSCI patient, the area of movement is similar to that of the healthy adult group previously studied. The trunk perturbation without any assistance provided the largest overall area of movement. This is quite fitting as trunk perturbations produce a larger moment arm about the ankle. The force field adequately decreases both trunk and pelvis motion. Enabling a force field around the trunk while perturbing the pelvis produces the most constrained environment in the healthy adults and the single iSCI patient tested. Further detailed analysis can be found in the following discussion section.

6.2.3 Complete SCI Protocol

After obtaining protocol approval from Columbia University's and University of Louisville's Institutional Review Boards, we recruited two patients with complete spinal cord injuries. This study was conducted at the University of Louisville's Spinal Cord Injury Research Center in Kentucky. The first patient was a 23-year-old male, AIS B, with a T8 level injury. The second patient was a 20-year-old male, AIS B, with a C4 level injury. Each patient presented with different motor and sensory deficits. Both patients were wheel chair bound and required assistance from a therapist to enter and exit the robotic stand trainer platform (Fig. 6.10).

In this protocol, the patient participated in four conditions. In the first, the pelvis was constrained using the motorized cables, the trunk received force field assistance, and the hands were supported on an instrumented handle bar directly in front of the patient. In the second condition, the pelvis was constrained using the motorized cables, the trunk received force field assistance, and the hands were not supported. In the third condition, both the trunk and pelvis were constrained by the motorized cables and the hands were not supported. The final condition was the traditional manual therapy setting, where the trunk and pelvis were supported by therapists and the hands were supported. In all cases, the knees were manually assisted by the therapist.

The force field algorithm consisted of two circular perimeters around the trunk (Fig. 6.11). The first ring had a radius of 2 cm and the second was 6 cm. Inside the first ring, no assistance was provided by the Stand Trainer. Between the first and second ring, the patient received 5% body weight assistance back towards the boundary of the first ring. Outside the second ring, the patient received 20% body weight assistance back towards the boundary of the second ring. These parameters were determined after several test sessions with research personnel. Due to the intrinsic instabilities within the patient, no external perturbations were applied.

Fig. 6.10 Complete SCI patient preparation and experimentation, respectively.

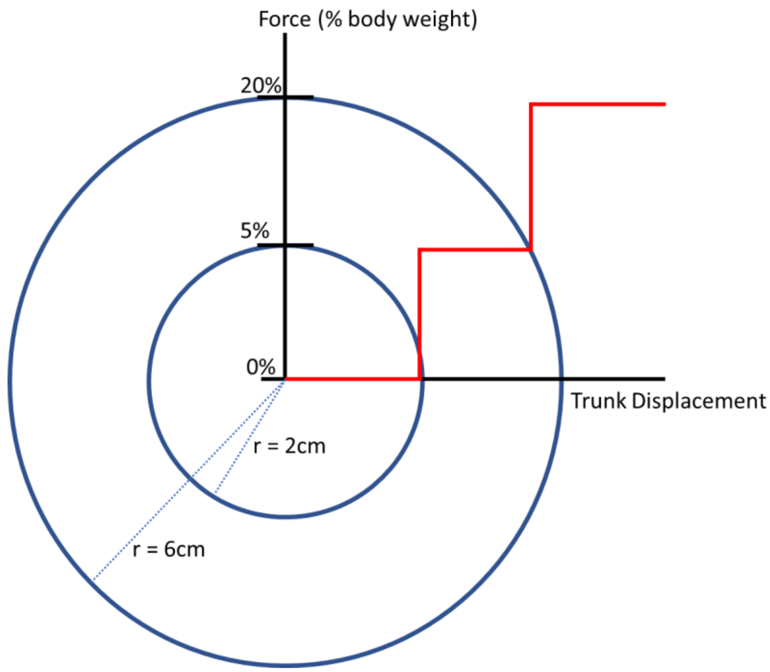


Fig. 6.11 Schematic of the force-field assistance algorithm based on trunk position

6.2.4 Results

Analysis of the workspace area (Fig. 6.12) shows that during manual therapy, the cervical injury patient (C4) exhibited the largest trunk and pelvis sway. This is while the therapists were manually working to prevent the segments from moving. Patient used the hand during this training to maintain stability, which alters the base of support. This is due to the increase in contact points. With four contact points, the base of support is increased, providing a larger stable workspace. This is important because such a condition may make it easier to balance, while would reduce the intrinsic need of the patient to maintain balance and stability. In addition, use of hand forces for support may create static tension at the arms and shoulder, possibly minimizing training time.

The patient with the thoracic injury (T8) showed the largest trunk and pelvis sway when the Stand Trainer provided the force field at the trunk and constant resistance at the pelvis with no hand support (Tff-Pcon-Nohands). This patient was also unable to perform the manual condition, where traditional PT assistance is provided, due to fatigue.

For both patients, using a large, constant force at the trunk and pelvis locked the segments in place. This condition (Tcon-Pcon-Nohands) produced the least movement. Also, both patients showed improved trunk and pelvis movement once the force field was applied to the trunk with constant pelvic assistance and hand support (Tff-Pcon-Hands). In both subjects, the device assistance was adequate and effective in reducing trunk and pelvis motion. While constant forces locked the segment's motion, the force field assistance allowed active motion while limiting the overall movement workspace.

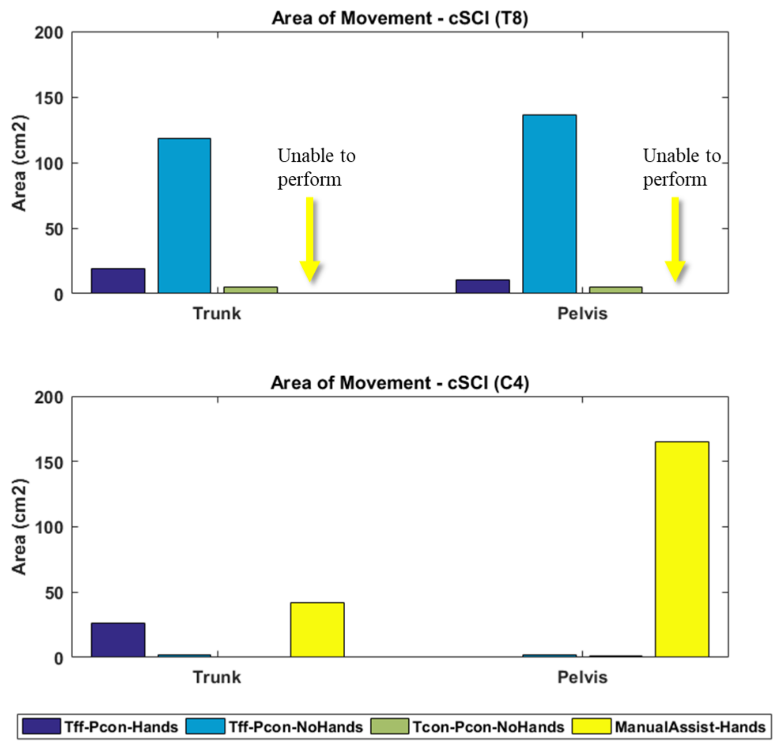


Fig. 6.12 Area or workspace of the trunk and pelvis center during each of the four conditions.

In analyzing the total distance travelled by the trunk, pelvis, and COP during 15 seconds of each condition (Fig. 6.13), it was noted that manual assistance had the most movement. This means that although the therapists' hands were used to maintain stability, the unstable nature of the trunk and pelvis coupled with any external motion from the therapists caused a large sway in segment and COP activity to maintain balance. The Stand Trainer adequately decreased trunk and pelvis motion for both patients. However, for the cervical injury patient, the trunk and pelvis had larger displacements when force field assistance and hands were used compared to without hand support. Perhaps this individual used the support from their hands to attempt to regulate trunk center, causing a larger area of displacement.

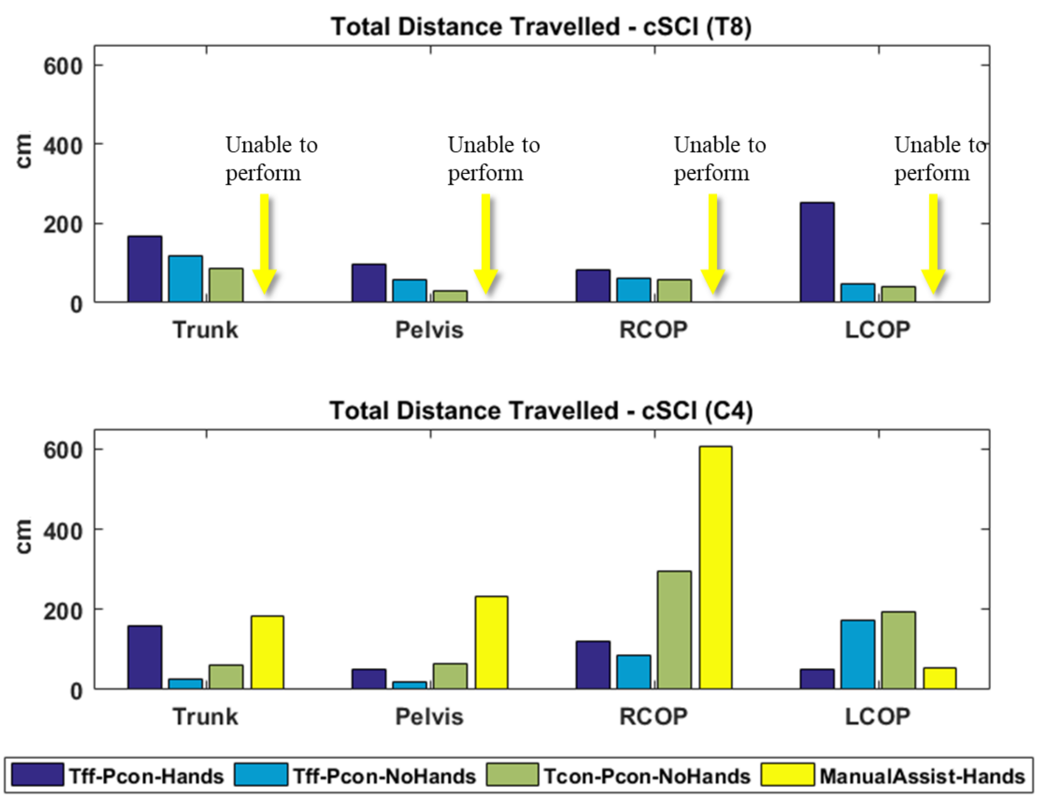


Fig. 6.13 Total distance travelled by the trunk, pelvis, and COP during each of the four conditions.

6.2.5 Discussion

Our analysis of both the iSCI and cSCI patients shows that the Stand Trainer conditions adequately and appropriately decreased trunk, pelvis, and COP motion. The purpose of the Stand Trainer system under the identified conditions is to constrain trunk and pelvis motion while providing the patient a safe environment in which to explore postural stability.

In the iSCI patient, the area of movement is similar to that of the healthy adult group previously studied. The trunk perturbation without any assistance provided the largest overall area of movement. This is quite fitting as trunk perturbations produce a larger moment arm about the ankle. The pelvis perturbation without assistance produced the second largest motion, which also manifested in the healthy adult population. The smallest areas of movement are conditions with the force field applied. The force field adequately decreases both trunk and pelvis motion. The therapist or engineer has full control in modifying the force field shape and boundary. Enabling a force field around the trunk while perturbing the pelvis produces the most constrained environment in the healthy adults and the single iSCI patient tested. This was expected, as providing support on both sides of the perturbation, with the force field at the trunk and ground contact at the feet, creates the most stable condition for the individuals being perturbed.

In analyzing the total distance travelled by the trunk and pelvis centers, it was noted that in the iSCI patient, the largest distance travelled is during the pelvic perturbation. Although only a single iSCI patient was tested, it is interesting to note this finding. The center of mass is estimated to be near the pelvis. Even though the largest trunk and pelvis path is expected during the trunk perturbation due to the larger moment arm about the ankle, it was seen during the pelvis perturbation. It may be that altering the COM directly has a greater affect in SCI patients. This can be evaluated further with a larger sample size. Similarly, the maximum distance travelled by the trunk and pelvis during posterior perturbation was seen with the pelvis perturbation condition. This corroborates the findings seen with the total distance travelled. Again, the Stand Trainer force fields adequately decrease the overall movement. This illustrates that a force field at the trunk level may be most appropriate for a severely affected patient.

From analysis of data from collected from healthy individuals and iSCI patients with the Stand Trainer, it is increasingly evident that a force field applied at the trunk can produce the most constrained condition for a patient with postural instability. Accordingly, when working with the cSCI patients we started with this condition. We provided force field assistance to the trunk and held the pelvis using constant forces of 80 newtons in each cable. In comparison to traditional manual therapy, where the trunk and pelvis are supported by physical therapists and the patient uses their hands on a walker for support, the force field conditions decreased trunk, pelvis, and COP motion. The COP activity is directly related to muscle activations from the ankle to maintain stability and counters the effects of the COM shift. In a neutral and stable configuration, the COM would sit directly over the COP. The decrease in COP motion indicates increased stability of the patient during the condition.

In another condition, the trunk and pelvis were constrained using constant forces from the Stand Trainer. This is the most constrained configuration, and the trunk and pelvis accordingly showed the least motion, illustrating again the Stand Trainer's capability of constraining motion.

This study provides both a characterization of the trunk and pelvis motions and the device feasibility in a patient population. The force field algorithms give the ability to modify the level of assistance through the size of the force field and the percentage of body weight assistance. Traditional manual therapy can be challenging both on the patient and therapists. It requires several therapists to hold the patient and stimulate the trunk, pelvis, and knees. In addition, the patient is being manipulated subjectively, with inherent variations in these non-exact forces. The Stand Trainer provides full control over the training, objective intervention, and might provide ease to the patient as well. With complete control of the various conditions, we can provide an easier training starting point and challenge the user as much as desired. These findings can also be recorded scientifically to better understand and characterize SCI rehabilitation from an objective standpoint.

Chapter 7

7 Conclusion

7.1 Contributions of the Current Work

In this dissertation, we have identified novel methods to characterize and train posture in the seated and standing position. These findings are documented through healthy and patient population studies, after the development of a seated and standing posture training robotic device. The following is a list of novel contribution of this dissertation.

1. Developed novel force-field algorithms for trunk posture training, which identify and account for the needs of patients with asymmetric trunk and pelvis workspace profiles, postural collapse, varying assistance over trunk and pelvis displacement.
2. Developed a novel "Stand Trainer" robot which can control the trunk, pelvis, and knees of individuals to train standing posture.

3. Developed a novel method called the “8-point star test” to measure trunk and pelvis workspace and area of controlled movement.
4. Tested and validated new posture training methods that train a person at and beyond their postural stability. Testing show kinematic changes in workspace and trunk displacement after training in the seated and standing position.
5. Show preliminary results that document changes in coordination patterns of patients with cerebellar deficits.
6. Show results that document how wheelchair bound cerebral palsy patients can improve postural stability by increasing their workspace boundary, translations, and vertical sitting.
7. In summary, patients trained at their stability boundary using assistance modify their kinematic coordination which are retained after the assistance is removed.
8. Test and document several assistance and perturbation based environments on spinal cord injury patients to show that a robotic force-field intervention adequately alters trunk and pelvis sway. Such a findings can be beneficial for future SCI training programs.

7.1.1 Development of TruST & Stand Trainer

Using current cable driven robotic technologies used for gait rehabilitation [17], we developed new methods to utilize the same system for posture training. In the seated position, the device allowed us to apply planar forces at the trunk. A new modified version was created for standing posture training. This is called the Stand Trainer. The device uses the same cable driven principles but provided control of the trunk, pelvis, and knees, simultaneously. Design of the system uses real-time human in the loop control and CAN communication to interact with 14 high torque motors.

In cable robotics, load cell sensors are often placed in series with cables to determine the tension in the cable. Although this is an effect method, the movement of the sensor along with the cable can create unwanted noise in the signal processing. It also requires consideration of the workspace, to prevent the sensor from crashing into the robotic environment. We mounted sensors onto the frames and used a pulley to connect the wires to the sensors. We characterized the sensors value to that of the tension in the cable. This reduces moving parts during robot use.

We designed new “force-field” algorithms for posture training. These are simple shapes, but unique to the user at the end effector. This consisted of planar circles, donut rings with different force characteristics outside each ring, and three-dimensional ellipsoids with different force assistance in each cartesian direction.

7.1.2 Novel Training Methods and Characterizations

The primary contribution of this dissertation is to identify and validate new posture characterization and training methods. Current posture rehabilitation methods are limited and not well documented. Often research primarily focuses on gait training. Yet, prior to train gait, it is necessary to train a patient to acquire independent sitting and independent standing. From this point forward, the patient should develop functional abilities such as leaning, reaching, and grasping which modulate the center of mass and the center of pressure. The oscillation of these components is what can lead to challenges in posture recovery, if the proper kinematic coordination and muscle activation is not achieved. Research suggests that task-oriented training and assist as needed interventions are best for motor adaptation and motor learning. Both are essential for the CNS to make new pathways and decisions regarding motor control. Although these are not analyzed in this work, changes in human performance shed light to the ability of the human to relearn after prominent injury or disease condition.

Prior to training and studying posture, it is important to first characterize posture. This is important to both assess the changes from a rehab intervention, as well as make the robot and system fit the specific needs of the patient. We developed what we call the “reach test” and “8-point star test”. These allow us to characterize the trunk and pelvis function in various direction and define the individual’s workspace, respectively. The important component of this test is to perform the test in full control and independently. This means being able to move to a maximal distance and then returning successfully and independently back to the neutral position. The force-field assistance then takes the shape of this workspace to provide “assist-as-needed” forces only when outside the boundary and haptic feedback at the boundary to signal postural limits. The assistance is modular and allows for postural exploration along with various components of the task-specific strengthening and coordination.

7.1.3 Postural Rehabilitation

Throughout the work in this dissertation, we explored the nature of individualized force field training on various patient populations. We also performed healthy adult validation studies prior to patient testing. Specifically, we tested on healthy adults and cerebral palsy, cerebellar ataxia, and spinal cord injury patients. Each required their own unique robotic assistance module.

In our initial studies, we validated the effects of the robotic intervention on seated training in healthy adults. The goal was to identify any motor adaptations which lead to a larger reach ability. Current literature does not show much evidence on postural rehab. Our results showed that healthy adults, when challenged and allowed adequate assistance for postural exploration, change their reaching strategies to allow a further reaching ability. The kinematics show increased trunk and pelvis rotation and increased trunk translation in the desired direction.

After obtaining confidence from reaching studies, we performed a longitudinal study with cerebral palsy patients. This study provided 12 training sessions across 4-6 weeks, with baseline and post-training assessments. Although the sample size was small (n=4), results proved that wheelchair bound patients can learn to improve their postural control. The changes were evident in all patients, leaving the patients with the ability to sit independently without any pelvis support or straps. We identified changes in kinematics such as increased rotations, translations, and workspace boundaries. We attribute this to postural exploration and motor learning of new postural configurations, not previously experienced.

We also outlined a similar training in the standing position. In this group, the subjects received active trunk assistance using the force field and passive resistance at the pelvis. We promoted instability in healthy adults and had them perform a series of reach tasks. The experimental group modified their reaching strategy while the control group without the robotic intervention developed more unstable configurations. The provided assistance was adequate to promote sensory cueing through haptic feedback and postural changes required for a more stable performance.

For our final set of studies, we developed a full body posture training system. In this, we were interested in characterizing the behavior of healthy adults and spinal cord injury patients. We wanted to see how various levels of assistance and perturbations can change the trunk, pelvis, and center of pressure movement. In our study, we provided assistance at the trunk, then pelvis, and provided perturbations at the opposite level. We then provided perturbations without assistance. It being evident that the force field assistance provided by the robotic platform was adequate in decreasing postural sway during external disturbances. We also noted that support at the trunk was more effective in reducing the sway, although the center of mass is approximated to be at the pelvis. We followed this with a single incomplete spinal cord injury patient and performed the same study. The patient performed similarly to the healthy adult group.

Finally, we characterized the effect of the device on two complete spinal cord injury patients. The patients, due to their degree of paralysis, were not provided external disturbances through perturbations. Instead, the study relied on internal disturbances due to balance deficits. The characterization compared traditional manual therapy with that given by the robot assistance, with and without hand support. Our results showed that the patients reduced their trunk and pelvis motion using the force field assistance at the trunk. Additionally, the patient was able to withstand self-standing without hand support. This is promising in terms of future rehab sessions.

Our studies give promise to the force field-based training paradigms in both the seated and standing positions. The device is modular and patient specific. All parameters are configured based on the specific needs of the patient. This can provide objective data in terms of rehab paradigms and also free up therapist so they can focus on patient interaction instead of physical support. Our results are promising, and we hope this dissertation provides a segue into new force-field based training paradigms and the understanding that patients have innate ways of adapting to their environment.

7.2 Suggestions for Future Work

Our studies provide validation of new posture rehabilitation methods. In this dissertation, we outlined a new robotic platform for safe and effective rehabilitation. Yet, our sample sizes and study durations were typically small. Although our work provides a strong affirmation to the benefits of postural exploration through robotic intervention, it is important to assess larger population sizes and analyze more information on kinematic coordination and muscle activation.

In our studies, sitting or standing, all subject groups showed motor adaptation and some degree of motor learning. Our only longitudinal study was one with cerebral palsy. With the strong evidence provided by the studies, it would be beneficial to invest 50-80 training sessions to study the changes in functional sitting or standing. In addition, a six month follow up may give better insight into whether the changes are retained or temporary. It is also recommended to obtain a control group. Through this requires large resources, it may serve as a strong method to tease out the effects of the training with those from the robotic intervention. Although many of our single session studies utilized a control for this very reason, a further investigation would give more strength to the findings.

In addition to the current methods, it would be beneficial to study changes at the muscle level. This includes looking at changes in muscle synergies during and after these robotic trainings. If possible, looking for methods to analyze changes in brain activity associated with the training can shed light to changes in brain stimulation.

It is also important to develop an outline and guidelines for post-training activities. Often after training, patients return to their sedentary lifestyles, where they become strapped to their wheelchairs or repeat use of walking aids. It would be beneficial to develop new physician or therapist monitored guidelines where the patient attempts to incorporate new reaching and workspace changes their have achieved. This may help retain the gains and develop them further. This includes gaining the appropriate confidence along with it.

We hope this dissertation serves as a strong motivator for researchers to investigate new methods for postural rehabilitation. Our results proved to be very promising and we hope future research helps define the scientific change more in depth.

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