SENSORY AUGMENTATION FOR BALANCE REHABILITATION USING SKIN STRETCH FEEDBACK

A Dissertation

by

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ABSTRACT

This dissertation focuses on the development and evaluation of portable sensory augmentation systems that render skin-stretch feedback of posture for standing balance training and for postural control improvement.

Falling is one of the main causes of fatal injuries among all members of the population. The high incidence of fall-related injuries also leads to high medical expenses, which cost approximately $34 billion annually in the United States. People with neurological diseases, e.g., stroke, multiple sclerosis, spinal cord injuries, and the elderly are more prone to falling when compared to healthy individuals. Falls among these populations can also lead to hip fracture, or even death. Thus, several balance and gait rehabilitation approaches have been developed to reduce the risk of falling. Traditionally, a balance-retraining program includes a series of exercises for trainees to strengthen their sensorimotor and musculoskeletal systems. Recent advances in technology have incorporated biofeedback such as visual, auditory, or haptic feedback to provide the users with extra cues about their postural sway. Studies have also demonstrated the positive effects of biofeedback on balance control.

However, current applications of biofeedback for interventions in people with impaired balance are still lacking some important characteristics such as portability (in-home care), small-size, and long-term viability. Inspired by the concept of light touch, a light, small, and wearable sensory augmentation system that detects body sway and supplements skin stretch on one's fingertip pad was first developed. The addition of a shear tactile display could significantly enhance the sensation to

body movement. Preliminary results have shown that the application of passive skin stretch feedback at the fingertip enhanced standing balance of healthy young adults. Based on these findings, two research directions were initiated to investigate i) which dynamical information of postural sway could be more effectively conveyed by skin stretch feedback, and ii) how can such feedback device be easily used in the clinical setting or on a daily basis.

The major sections of this research are focused on understanding how the skin stretch feedback affects the standing balance and on quantifying the ability of humans to interpret the cutaneous feedback as the cues of their physiological states. Experimental results from both static and dynamic balancing tasks revealed that healthy subjects were able to respond to the cues and subsequently correct their posture. However, it was observed that the postural sway did not generally improve in healthy subjects due to skin stretch feedback. A possible reason was that healthy subjects already had good enough quality sensory information such that the additional artificial biofeedback may have interfered with other sensory cues. Experiments incorporating simulated sensory deficits were further conducted and it was found that subjects with perturbed sensory systems (e.g., unstable surface) showed improved balance due to skin stretch feedback when compared to the neutral standing conditions. Positive impacts on balance performance have also been demonstrated among multiple sclerosis patients when they receive skin stretch feedback from a sensory augmentation walker. The findings in this research indicated that the skin stretch feedback rendered by the developed devices affected the human balance and can potentially compensates underlying neurological or musculoskeletal disorders, therefore enhancing quiet standing postural control.

DEDICATION

Dedicated to the power of kindness and the strength of gentleness.

"You cannot hope to build a better world without improving the individuals. To that end, each of us must work for our own improvement." - Maria Skłodowska-Curie

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NOMENCLATURE

COM	Center of Mass
COP	Center of Pressure
CNS	Central Nervous System
BOS	Base of Support
AP	Anterior-Posterior
ML	Mediolateral
IMU	Inertia Measurement Unit
PID	Proportional-Integral-Derivative
SAD	Sensory Augmentation Device
ICP	Instantaneous Capture Point
PWM	Pulse-width Modulation
TRAD	Traditional Measures
SDA	Stabilogram Diffusion Analysis
IDA	Invariant Density Analysis
MS	Multiple Sclerosis

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1. INTRODUCTION

1.1 Motivation

Imagine you are walking on a wood log. Instinctively, you would extend your arms on both sides to maintain your balance while walking on it. This can be explained by the dynamics of rotational motion; by extending both arms outwards your moment of inertia with respect to the rotational axis is increased, and hence your body is more resistant to rotation. Now, imagine as you raise your arms, you can touch an object at your fingertips. The object is fixed in position such as a wall or railings. Under these circumstances your body also becomes more stable. But why? The mystery behind this can be explained by the effect of "light touch" — the cutaneous feedback perceived through mechanoreceptors in our skin. It is found that by lightly touching an external, rigid surface with your fingertip, postural stability would be significantly improved. This fingertip cue is an additional source of sensory information, providing body sway and arm position information. Such information can be used to identify the direction of postural sway and to allow anticipatory muscle activation, and therefore improve postural control performance [4] [5] [6] [7] [8]. Not surprisingly, with more useful sensory information, one may perform better in motor tasks. Maintaining balance seems like a simple motor task in humans, and we do it almost every day. However, when one or more sensory systems are impeded, such as standing on a unstable surface or walking inside a dark room, the information received by us may be wrong or insufficient, with the consequence of increasing the risks of falling. For the elderly or people with neurological disorders (e.g., stroke, multiple sclerosis, Parkinson's disease, etc), falls are common due to degraded or impaired sensorimotor functions. Loss of sensation and muscle weakness could severely impact one's sense of balance, and eventually lead to serious falls.

Preventing falls is important, especially for people with degraded balance or impaired strength such as elderly or people suffering from neuromuscular diseases. Studies have shown that elderly people tend to sway more and postural control declines physiologically with age. Similarly, increased risk of falling observed from neuromuscular disease patient population is also associated with the balance and gait impairment, muscle weakness or other sensory disturbance. Balance impairment is a critical risk factor for falling which makes balance retraining necessary for preventing falls in these population. Over the past three decades, strength and balance retraining programmes/interventions have been developed. It is found that programs incorporating motor, sensory, and cognitive systems are more effective than strength training alone. How to effectively improve the sense of balance is a key question that researchers and therapists should think about when designing rehabilitation programs. Robotic assistive devices have been developed to improve postural stability and mobility. Such assistive devices augment muscle strength by coupling itself with the target body segment or augment the sensory information by providing additional biofeedback sources. The former can be illustrated by robotics exoskeletons for treadmill training (e.g., Lokomat) or overground training (e.g, ReWalk, Ekso Bionics, REX, etc). These devices can offer sufficient weight support and can increase the stability of patients' stance or gait. They also reduce labor effort for physical therapists. However, the "2010 Veterans Administration/Department of Defense (VA/DoD) guidelines" recommends against the use of robotics for the lower extremity [9]. There is evidence that such training methods are ineffective and do not have a better outcome when compared to conventional therapies. One reason could be that robotic training does not provide adequate challenge to the users to facilitate their motor learning. Therefore, instead of a device that only mitigates the undesired event, the event information such as users' training performance should also be provided in real time to adjust appropriate difficulty level for users. This concept echoes the biofeedback-based devices which can augment or substitute weak or missing sensory signals of the user with additional visual, auditory, haptic or multimodal feedback.

In this thesis, biofeedback-based devices using haptic feedback for balance retraining are presented. The reasons that haptic feedback is favored are as follows. First, this research is inspired by the concept of *light touch*. As mentioned in the first paragraph, light touch contact on an external stationary surface could stabilize body posture. The effects of light touch has been widely investigated in standing and walking but none of these previous studies have used an ungrounded, portable or wearable device as a source for light touch sensation. Specifically, the light touch sensation is recreated by inducing shear force at one's fingertips. It is referred to as *skin stretch feedback* in this thesis. Second, compared to vision and hearing, haptic feedback offers a unique feedback strategy with which one can identify, interpret and respond to cues received in the physical world. This could be useful in optimizing the subject performance. Haptic feedback is also a natural way for communicating information between humans or between humans and our surrounding environment, whereas audiovisual cues may be harder to identify under noisy environments or improper lighting conditions. More details about the different sensory feedback devices from previous studies are discussed in Chapter 2. The research questions to be answered by this work are i) Whether the sensory augmentation using skin stretch feedback can improve control of balance during quiet standing? ii) What kinds of information should be conveyed for improving postural balance? iii) How can such information be delivered in an intuitive way via skin stretch feedback? iv) Can such feedback device be easily used in the clinical setting or in home or on a daily basis? Research findings are summarized in Chapter 7.

1.2 Outline of Thesis

This thesis presents the motivation, methodologies, and experimental results for using augmented skin stretch feedback in real-time postural control. The current chapter introduces the motivation behind this research, the outline of thesis, and the main contribution from this research. Chapter 2 provides background information about human balance, the importance of balance retraining, and a summary of existing research in the fields of biofeedback-based rehabilitation and the studies that support the view of using skin stretch feedback in improving motor learning.

Chapter 3 presents the development of a portable sensory augmentation device that can induce skin stretch feedback on the fingertip. The motivation for reproducing the effect of *light touch* and the relevant studies are presented. Balance experiments were conducted to study the effectiveness of the skin stretch feedback device in improving balance control. In these experiments, different sensory deficit conditions were simulated in healthy young subjects to investigate how the additional sensory information could improve their balance.

Chapter 4 investigates whether velocity information of body sway is a more useful sensory feedback than position information in quiet standing balance. The findings from previous modeling studies have emphasized the importance of velocity information in the postural control system. The feedback control model that investigated the contribution of sensory information and the stochastic behavior of quiet standing along with the balance metrics derived based on these models are introduced. The position and velocity information of body sway was rendered by the device described in Chapter 3. Experiments were conducted on healthy young subjects to study the effects of augmented bodily information including position, velocity, and a combination of the two on quiet standing performance. Three different balance assessments (traditional measures, stabilogram diffusion analysis and invariant density analysis) were applied to better quantify the quiet standing behavior and to compare the results from these assessment methods.

Chapter 5 presents a wrist-worn device that induces lateral skin stretch on the wrist. In our everyday life, wearing a fingertip device might hinder the ability to grasp objects, touch a surface, or perform many routine activities. This motivates the design of a wrist-worn device. Ultimately, such a sensory augmentation device was developed not only for use in the rehabilitation setting, but also everyday use. In this chapter a pilot study was conducted on healthy young subjects to evaluate their abilities to dynamically shift their weights by following the skin stretch cues received through the wrist-worn device. Visual feedback of posture information was also incorporated to investigate the effectiveness of visual, skin stretch and multimodal feedback in the dynamic stance balance control.

Chapter 6 introduces the design and development of a sensory augmentation walker for balance rehabilitation. A novel skin stretch device was built into the handle of a conventional four-wheeled walker. Walking aids such as canes and walkers are essential for the elderly or people with neuromuscular disorders to increase their stability and mobility. While these devices provide partial weight support or feedback of surrounding environment, a potential fall may still occur due to the lack of attention or impaired sense of balance. To improve the sense of balance of users while using a walking aid, a sensory augmentation walker that renders postural cues was developed. Various state of the art on smart mobility aids incorporating sensors and actuators and hand-held haptic devices are presented. In the first half of this chapter, methods, experimental setup and results of a perceptual study for discerning the skin stretch cues at the fingertip and palm are presented. The latter half examines the use of this sensory augmentation walker in improving the sense of balance among people with multiple sclerosis. Overall, subjects could follow the skin stretch cues perceived at their fingertips while actively correcting their posture. The results highlight the potential benefit of incorporating such sensory feedback into mobility aids for enhancing user's control of balance.

This thesis concludes in Chapter 7. A summary presents the findings and highlights the important insights of this research. Recommended future directions and applications of this work are also presented.

1.3 Contributions

The primary contribution in this thesis is establishing the feasibility and value of portable sensory augmentation devices using skin stretch feedback for balance retraining and postural control. Specific contributions are as follows.

- A novel, wearable, lightweight fingertip-worn device that provides light skin stretch feedback of postural sway was developed. Unlike previous approaches that require a stable surface or grounded devices for light touch contact of the finger, this device is portable and compact while also applying light touch cues on the fingertip.
- This study evaluated fingertip skin stretch feedback in providing postural sway

information and comparison of position, velocity and combined information of postural sway. The results from these experiments can provide a better understanding of which bodily information is more effective for control of standing balance. The augmented velocity information of postural sway has been found to improve quiet standing balance more effectively compared to position information.

- A novel, wearable, light-weight wrist-worn device that provides lateral skin stretch on postural sway was developed. The device reproduces the physical contact of the environment that a swaying person would experience, which has established a new mapping strategy for balance control and waypoint navigation. The haptic actuation is relatively simple and offers an intuitive way for conveying posture information. This wrist-worn device is also a viable alternative when performing the activities of daily living that require use of the hands.

- A framework incorporating a sensory feedback device into a conventional walking aid was proposed. This device renders users' posture information as a means of improving their sense of balance while supported by a mechanical device. Unlike the existing walking aids that primarily offer partial weight-support and force feedback of surrounding obstacles, the sensory feedback rendered by this device can encourage walker users actively engage in postural control. The experimental results and feedback obtained from the multiple sclerosis subjects further demonstrated the device's potential for in-home self retraining and daily use.

2. BACKGROUND

2.1 Human Balance

An upright human body is inherently unstable, since two-thirds of the body mass is distributed over two-thirds of body height above the ground [1]. To maintain an upright posture, our central nervous system (CNS) needs to continuously fuse multisensory inputs and simultaneously control different motor outputs, which involves interplay among all levels of the CNS, from the spinal cord to cerebral cortex. Appropriate corrective torques must be generated to resist the torque due to gravity, this kind of corrective torque can be classified as "active" or "passive" torque. For "active torque", multiple channels of inputs from sensory systems are necessary, and it involves a time delay due to dynamic sensorimotor processes [10]. While the "passive torque", acting without time delay, is set by joint stiffness through the CNS at specific balance control sites [11]. Results from the study by Peterka [12] have supported the view that "active torque" generated from feedback control mechanism only is sufficient to account for postural control behavior.

Feedback control mechanism adapts the sensory information for us to correct movements. Three major sensory systems contribute to the postural balance including visual, vestibular and somatosensory systems [1] [10] [12]. Vision allows us to detect the surrounding environments and relative orientation to the physical world. In general, visual input dominates over other sensory inputs and is a primary source to account for postural adjustment at low frequency. It also plays an important role for learning new balancing skills under various tasks and conditions. The vestibular system consists of receptors in macular otoliths and semicircular canals which detect the linear and angular acceleration of the head, respectively. It is responsible for our sense of "verticality" and for triggering the response to unexpected falling. The somatosensory system includes the cutaneous mechanoreceptors that perceive the pressure of the object on skin and proprioceptors that detect the joint position and movement. Different from the visual and vestibular receptors which locate in the head and control human body independently, somatosensory receptors distribute all over the body, informing the qualities of the support surface and the forces exerting on the surface during standing. Combining with vestibular information, proprioceptors at the neck detect the position and velocity of trunk movement, providing another key information for the postural control system. How these sensory systems contribute individually to balance control has been studied over the past three decades using developed biomechanical models and experiments [13] [14] [10] [15] [12] [16]. Results from these studies have supported the view of "feedback control mechanism" in postural control behavior that the postural control system is able to re-weight sensory inputs depending on the context and task, and also have extended our knowledge on human stance control for further development of rehabilitation approaches.

2.1.1 Standing Balance

Maintaining stability in quiet standing is a fundamental motor skill that human has explored and acquired in early childhood. This seemingly simple task requires highly-coordinated CNS to transform mulitsensory inputs into appropriate bodily information. Using this information, the body orientation relative to the prior state can be estimated, and muscles of different segments are activated to control the body movement and prevent from falling. There are several sources leading to the postural imbalance including internal forces from the body's own movement and external forces from the surrounding environment or due to gravity. These forces

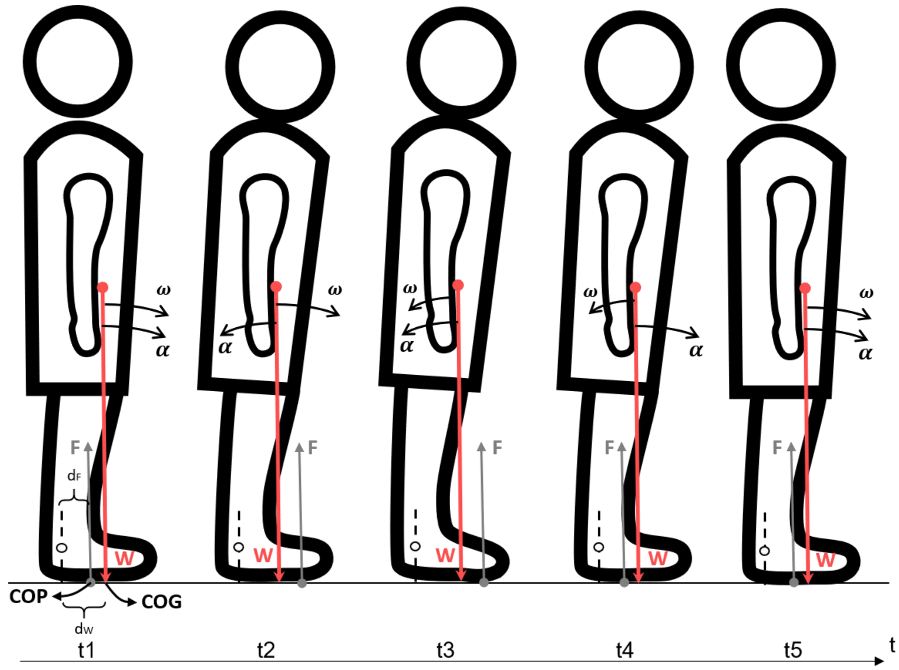

Figure 2.1: Postural sway in anterior-posterier direction during quiet standing, adapted from [1]

accelerate the center of mass (COM) of the body and cause the postural sway. Fig. 2.1 depicts the postural control behavior during quiet standing over time using an inverted pendulum model. At the initial time t1, subject stands quietly in a upright posture, with the body's center of gravity (COG) ahead of the center of pressure (COP). COG is the point where the COM projects onto and COP is the point where the ground reaction force exerting on. Body weight W is equal to the ground reaction force F and in an opposite direction. d_F and d_W represent the distances from the ankle joint to the COP and COG respectively. Assuming the human body anchored at the ankle joint as an inverted pendulum, a clockwise moment equals to Iα will be produced due to the offset distance between COP and COG:

$$W * d_W - F * d_F = I\alpha \tag{2.1}$$

where I represents the moment of inertia of the whole body about the ankle joint and α is the angular acceleration of the body. Forward sway occurs due to the clockwise moment as described in Eq. 2.1. In order to correct this moment, the subject performs plantarflexion at the ankle joint to move the COP ahead of the COG (see time point t2, Fig. 2.1). At this timing Fd_F is greater than Wd_W which reverses the direction of α and results in a reversal of the angular velocity ω at t3. Now both α and ω are counterclockwise thus a backward sway occurs. Such low frequencies and relatively small amplitudes dynamics will be continuously acting over time that characterizes the motion of postural sway.

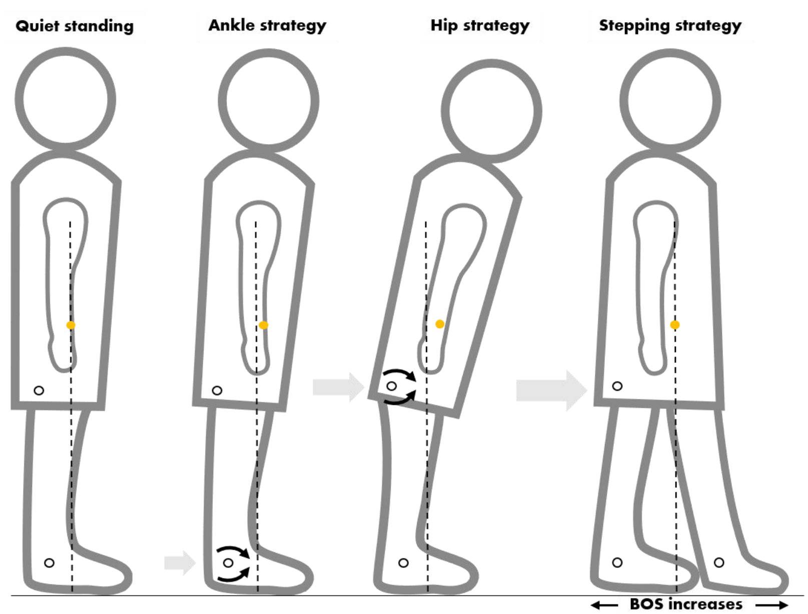

Figure 2.2: Movement strategies for recovering equilibrium in response to an external perturbation. Grey arrows represent the perturbation and the location it applies on, adapted from [1].

A fall may occur when subjects are not able to control their COM within the base of support (BOS). In stance, the BOS is the region between feet which is the quadrangle bounded by the toes and heels. The CNS has an internal representation of the BOS to allow subjects to move around their neutral position and maintain equilibrium [17]. The BOS of elderly or people with multisensory disorders are often small or their CNS have inaccurate internal representation of this region that affect their abilities to maintain equilibrium. When experiencing an external perturbation, persons move their COM to stay within the BOS. In the anterior-posterior (AP) direction, three movement strategies have been identified based on the biomechanical constraint of the BOS: the ankle, hip and stepping strategies. When standing persons sway naturally or are perturbed by a small amount of external forces, they exert an torque at the ankle as a inverted pendulum (see Fig. 2.2) to drive the COM back to the neutral position and maintain balance. This is called the ankle strategy, which is the most commonly used response during quiet standing. As the level of perturbation increases or the support surface becomes small or unstable, a hip strategy would be utilized (Fig. 2.2). This strategy is seen more often in elderly individuals when moving their bodies in the AP direction [17]. Persons may mix ankle and hip strategies in different perturbed situations. When both ankle and hip strategies can not return COM within the BOS, a stepping strategy would be initiated, to increase the BOS and maintain vertical alignment of the trunk and head (Fig. 2.2). The postural goals or the characteristics of task could also influence the selection among strategies for the maintenance of equilibrium and fall prevention.

In this thesis, the use of the ankle strategy is primarily studied for understanding the control of balance during quiet standing. In this way the body is modeled as an inverted pendulum with a single head-trunk-leg segment rotating about the ankle joint. This model captures the relationship between the COP-COM and the horizontal acceleration of the COM. Deriving from Eq. 2.1 and considering h as the

COM height, the following equation can be obtained [11]:

$$COP_x - COG_x = (-I/Wh)C\ddot{O}G_x \tag{2.2}$$

where COP_x and COG_x are the COP and the COG in the AP direction, respectively. $C\ddot{O}G_x$ is the COG acceleration in the AP direction. The (COP-COM) signal is considered as the error signal that the postural control system is sensing, since it is directly related the COM acceleration. The cross-correlation has also been validated in [11] and been used for modeling multisensory integration based on feedback control by researchers [12] [18]. The majority of research have emphasized the differences between COP and COM and these two measures should not be interpreted as the same thing or even interchanged, which has been seen in many of the previous studies.

2.1.2 Walking Balance

Although standing balance has been intensively researched in the last four decades, the control of balance during walking is still not clear. Maintaining balance while in motion is considered a way more challenging motor task and invloves many more relevant degrees of freedom compared to static balance. As described in the previous section, the stance balance control is to keep the COM within the BOS, whereas the study have shown that during walking, the COM moves forward along the medial border of each support foot and never moves within the base of the foot [1]. The COM also heads toward the direction opposite to the direction of the COM acceleration. Therefore the simple inverted pendulum model introduced in standing balance is challenged. Ankle muscles alone are not enough for maintaining balance, one should more safely place the swing foot on the desired location to prevent a fall. Several control strategies have been introduced such as foot placement shift, lateral ankle and hip strategy, and push-off modulation. Two important mechanisms among those are the stepping/foot placement strategy (as briefly described in Section 2.1.1) and lateral ankle strategy [19]. The control of the medial-lateral (ML) balance becomes essential during walking due to the mechanically less stability in this direction [20]. More details regarding the balance strategies during walking can be found in a recent comprehensive review [21]. In this thesis we will focus on both static and dynamic balance during standing while the dynamic balance during locomotion will be discussed in the future work.

2.2 Balance Retraining

Why is retraining balance so important?

Falls are common in elderly and are associated with morbidity, mortality and significant public health problems. About one third to one half of elderly fall each year [22], and study shows about two-thirds of chance to fall again in the subsequent year among those fallers [23]. Falls could lead to different levels of consequences, from pain, impaired fuction, fractures to loss of indpendence and even death. It is the commonest cause of injury-related death in elderly aged over 75 [24]. Many of these falls are due to postural imbalance. Studies have shown that elderly people tend to sway more and postural control declines physiologically with age [25] [26]. Costs of health care also increase as falls increase. Falls are common not only in older people, but also in people suffer from neuromuscular diseases such as stroke, hemiplegia, multiple sclerosis (MS), and spinal cord injury, etc. For example, in the literature, falls are the leading complication after acute stroke and even remain a major health concern for post stroke survivors [27]. More than half of the MS subjects fell at least once every six months [28] [29]. Increased risk of falling has been observed among these patient population and it can be associated with the balance and gait impairment, muscle weakness and spasticity, or other sensory disturbance.

From the vast majority of studies, balance impairment has been found to be an critical risk factor for falling which makes balance retraining necessary for preventing falls in elderly and people with neuromuscular disorders.

What are the existing balance retraining approaches?

Conventional balance retraining includes exercise programs designed in home and/or in clinics. It consists of a set of exercise to strengthen the leg muscle, retrain balance, increase flexibility or endurance, and reduce rate of falling. In addition, robot-aided therapy has been incorporated into clinical practices to deliver high-intensity, reproducible sensorimotor training for the past two decades. These robotic exoskeletons allow treadmill or over ground training and actuating the user's leg joints or segments through a set of robotic links, which can also support user's body weight. Biofeedback devices are also widely seen in such training programs providing real-time information about user's performance. More realistic training environment can also be realized by integrating Nintendo's Wii Balance Board or Virtual Reality (VR) into the conventional training approaches. Those biofeedback-based training approaches have been shown positive improvement in balance across elderly or people with neurological disorders. Examples and the effectiveness of these three methods in improving postural balance are described as follows.

2.2.1 Conventional Therapy

A conventional balance training programme includes static and dynamic balancing tasks. For static balance training, maintaining a quadruped position, kneeling, standing, and shifting weight to balance on one leg are some of the typical tasks. For dynamic balance training, perturbation is introduced such as unreliable support surface under the feet in order to examine trainee's ability to move within a given posture without loss of balance. Such training approach can be conducted in a clinical setting or in home. A home exercise programme that has been widely used is the Otago Exercise Program (OEP) [30]. It aims to prevent falls in elderly and community-dwelling people by performing a series of exercise that strengthens muscles and retraining balance. Home exercise programme targeting strength and balance retraining has been found to be effective in reducing falls and injuries in elderly people [31, 32].

However, a common problem of the conventional training is that they are not incorporating the human in a natural way. There is limited information about the user's performance. Therefore, the appropriate difficulty level for the training cannot be adjusted.

2.2.2 Robot-assisted Therapy

Robotic therapy devices were designed to provide mechanical assistance to help individuals complete various training tasks [33]. This strategy aims to reproduce the active assistance used by physical therapists (PT) during the rehabilitation training to help the patient to complete the movement. For example, PT provides balance and weight support during overground walking training; or helps moving the upper extremity for reaching tasks. Initial devices for the lower extremity, such as the Lokomat (Hocoma, Zurich, Switzerland) [34] and the Gait Trainer [35], assist subjects in maintaining postural stability and achieving gait-like motions. These devices can be physically attached to the subject and essentially tried to work in harmony to achieve desired movements.

However, over assisting a patient using such devices could decrese the amount of

Figure 2.3: Lokomat gait orthosis for lower-limb balance and gait training. Reprinted from [2].

learning that could occur during training. In two key robotic therapy studies [36] [37], the Lokomat was used for gait training by patients with stroke who were already ambulatory and compared two control groups that trained with conventional gait training techniques. While patients improved their gait speed through training with the Lokomat, they improved less than via conventional training. One interpretation is that the Lokomat created a training environment with too low of challenge by over assisting the trainee.

Robotic exoskeletons have been developed to augment motor function and provide physical support during training of the lower extremities for patients with paralysis caused by SCI or stroke [38]. These devices can actuate one or more of the user's lower extremity joints and support upright posture in standing and walking. While exoskeletons can provide some active balance control, it is important for users to know how to shift their weight to stay balanced. For some users, this may be intuitive, but for some, additional sensory information is needed depending on the type of injury, level of injury and/or sensory impairment.

2.2.3 Biofeedback-based Therapy

Additional sensory information can be provided by a physical therapist, or by incorporating additional biofeedback sources into the balance training. An existing technique using biofeedback approach is adopting the Nintendo Wii Fit^{TM} exergames for dynamic balance training, and several studies have shown its feasibility for improving balance among healthy [39], elderly [40] [41] and neurologically-impaired populations [42] [43]. This kind of game-based system incorporates both visual and auditory feedback to both the subject and therapist about the subject's performance in real-time. The additional biofeedback could increase the subject's engagement on repetitive movements and provide therapist immediate quantitative outcomes, therefore improving the training process. VR systems such as CAREN (Motek Medical BV, Amsterdam, Netherlands) have been developed [3] to simulate dynamic environments and challenge users to perform realistic motor tasks (Fig. 2.4). It is been shown that such VR-based system can be a good feature for the traditional balance intervention programs in improving postural control.

In addition to visual and auditory feedback, haptic feedback which can be applied to those parts of the body with complete sensation have shown effectiveness on improving balance performance among healthy individuals [44] [45], individuals with vestibular disorders [46] [47] or those with Parkinson's disease [48]. Haptic feedback, when placed in close proximity to the skin and in regions where the user experiences adequate sensation, offers a unique feedback strategy: the opportunity to identify, interpret and respond to cues that can be received in environments where audiovisual cues may be harder to identify (i.e. noisy environments and/or improper lighting

Figure 2.4: The virtual reality based system for balance rehabilitation. Reprinted from [3].

conditions). The simplicity and safety characteristics make it a preferable option in augmenting sensory information.

These rehabilitative interventions integrating additional biofeedback that could compensate for the missing or weak sensory signals due to the impaired sensorimotor system have demonstrated promising results in improving postural stability and further preventing fall-related injuries among the elderly population.

3. SKIN STRETCH FOR BALANCE CONTROL*

3.1 Introduction

Neurological disorders are the leading causes of poor balance. Previous studies have shown that biofeedback can compensate for weak or missing sensory information in people with sensory deficits. These biofeedback inputs can be easily recognized and converted into proper information by the central nervous system (CNS), which integrates the appropriate sensorimotor information and stabilizes the human posture. In this chapter, we introduce a form of cutaneous feedback which stretches fingertip pad with a rotational contactor, so called skin stretch. Skin stretch at a fingertip pad can be simply perceived and its small contact area makes it favored for small wearable devices. Taking advantage of skin stretch feedback, we developed a portable sensory augmentation device (SAD) for rehabilitation of balance. SAD was designed to provide postural sway information through additional skin stretch feedback.

In this study, our first objective is to develop a portable sensory augmentation device that can induce skin stretch feedback at the index fingertip pad in response to postural sway. Skin stretch feedback in this research aims to mimic the directional friction that swaying subjects may experience at their fingertip when they are lightly touching a stationary surface with their fingertip. Instead of actively touching a fixed surface, subjects are passively provided light touch information about their body sway by our developed wearable device. The second objective is to evaluate the feasibility of the developed device as a sensory augmentation device that can

*This chapter is based on the article ©2017 IEEE. Reprinted, with permission, from Yi-Tsen Pan, Han UI Yoon, and Pilwon Hur, "A Portable Sensory Augmentation Device for Balance Rehabilitation Using Fingertip Skin Stretch Feedback", IEEE Transactions on Neural Systems and Rehabilitation Engineering, vol. 25, no. 1, pp 31-39.

effectively reduce postural sway. As a feasibility study, postural sways of healthy young adults with simulated sensory deficit were investigated. It was hypothesized that augmented sensation via induced skin stretch feedback enhances quiet standing balance more effectively when more sensory modalities are removed or not reliable.

3.2 Prior Work

Biofeedback systems translate bodily function information to sensory inputs such as vision, hearing, or somatosensation so that individuals are provided extra cues of their physiological states [49]. This concept utilizes biofeedback as a substitute for, or as an augmentation to, the existing sensation so that the sensory signals transferred to the CNS can be processed and recognized in more efficient ways [50]. Biofeedback has been known as an essential technique in rehabilitation for stroke survivors [51] [52] and the elderly [53]. Therefore, how to enhance the impaired sensory systems or how to substitute the lost information with biofeedback is an important issue for both clinicians and researchers.

A number of rehabilitation techniques and devices for maintaining standing balance or performing a qualified mobility task using additional sensory information have been proposed and evaluated [50]. An audio-biofeedback system has been used to show the capability of correcting postural sway by providing trunk orientation information via auditory signal to subjects [54] [55]. There have been several studies that aimed at enhancing human postural control for individuals with disabilities, especially for people with visual or hearing impairments via vibrotactile feedback [56] [57] [44] [58]. Due to its simplicity and safety characteristics, many biofeedback applications for postural control using tactile vibration have been growing rapidly over the past decade.

Skin stretch feedback can also be used to convey biofeedback signals to the CNS [59]. The addition of this kind of simple shear tactile display would significantly enhance the friction sensation to a haptic device. Moreover, such light skin stretch could be easily perceived [60] especially at a fingertip pad, since a fingertip pad is more sensitive to skin stretch than vertical skin deformation [61]. Its easy perception, large contact surface, and the capability to provide both shear and normal forces may make the cutaneous skin stretch a more attractive alternative for sensory augmentation when compared to other types of biofeedback.

3.2.1 Light Touch

Another type of cutaneous cue, a light touch contact (contact force < 1N) of a fingertip on a fixed surface, has been shown, in several studies, to be capable of reducing body sway in standing [5] [62] [6] [8] and walking [63]. The light touch works as an additional tactile sensory input instead of a mechanical support [4]. Krishnamoorthy et al. [64] showed that light touch can be applied on different body parts other than fingertips to stabilize posture. Enders et al. [65] showed that subthreshold vibrotactile noises at various locations of the upper extremity improves light touch sensation in stroke survivors. Therefore, with the help of augmented sensation, individuals with sensory deficits may improve their balance in daily activities, which eventually could lead to enhanced quality of life.

While many studies have demonstrated the effectiveness of skin stretch feedback in improving task performance using haptic devices in a virtual environment [26] [27] or perceived friction magnitude [16], few studies have evaluated the efficacy of the skin stretch feedback at a fingertip pad in improving standing balance. Additionally, portability is a useful factor because wearable sensors attached to the human bodies can provide accurate and reliable information about humans' activities and behaviors in their daily lives [28]. Since portable and wearable sensors are not limited by operation place (e.g. laboratory) and cable length, they have great potential in home rehabilitation for patients such as elderly adults and stroke survivors.

3.3 System Overview

The system consists of a sensory augmentation device (SAD) that induces the skin stretch at an index fingertip pad. A control unit, motor driver, and IMU are enclosed in a waist belt. The schematic diagram and the fabricated device of our sensory augmentation system are illustrated in Fig. 3.1 and Fig. 3.2 respectively. The device's detailed design and related control strategy are described in the following subsections.

3.3.1 Skin Stretch Device Design

SAD was designed to induce skin stretch at an index fingertip pad (Fig. 3.2 a, b). The DC motor (1524T009SR, Faulhaber, Germany) was mounted inside the SAD's housing where the subject's index finger was inserted (Fig. 3.2 b). Skin stretch feedback was therefore provided by the shearing between the contactor, operated by the DC motor, and the fingertip pad (Fig. 3.2 b). Several contactors and housings of various sizes were fabricated to accommodate various subjects' finger sizes; we created these using a 3D printer (Replicator 2X, Makerbot, Brooklyn, NY). The weight of the entire device which subjects wore on their index fingers, including the contactor, housing, and DC motor was approximately 20 g. An IMU (MPU-9150, InvenSense Inc., San Jose, CA) was attached at the back of the waistline of each subject, which is the approximated location of the human body's center of mass (COM) (Fig. 3.1, 3.2 a). The data from the IMU were then used to monitor the postural

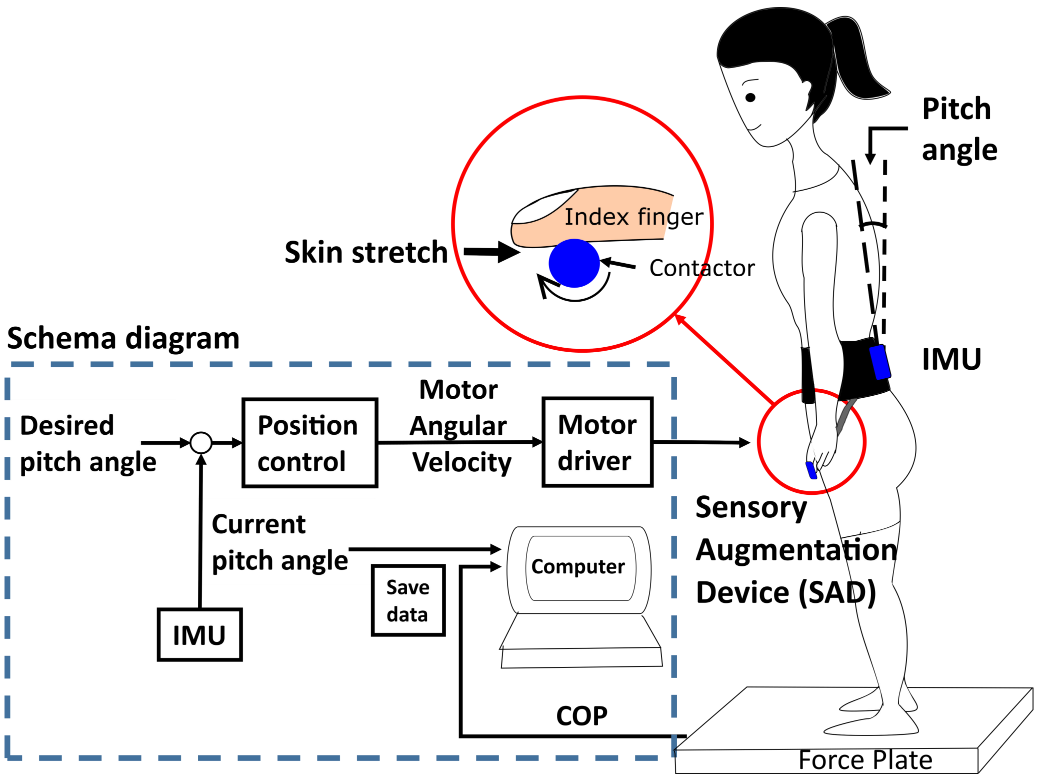

Figure 3.1: Schematic diagram of sensory augmentation system. Inertia measurement unit (IMU) measures the pitch angle of body sway while subject stands quietly on the force plate. Contactor's angular velocity is defined to be proportional to angular deviation of pitch angle from the desired pitch angle (reference angle). When subject tilts forward, the contactor rotates in clockwise direction, and vice versa. The skin stretch feedback is then provided at subject's index fingertip pad. Subjects' pitch angles and COP data are saved to evaluate the efficacy of SAD.

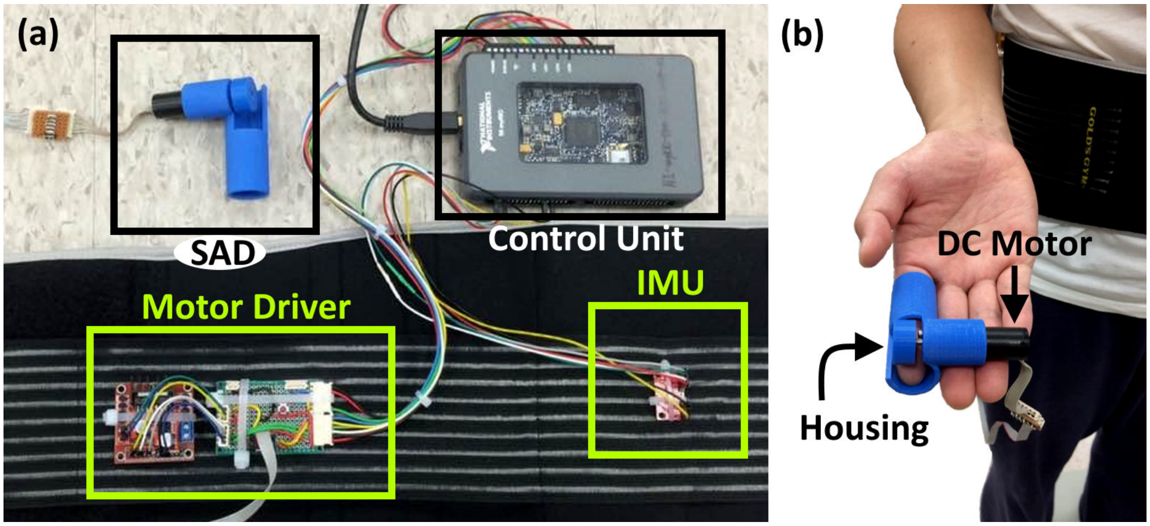

Figure 3.2: System overview. (a) The system consists of a sensory augmentation device (SAD) that induces the skin stretch at an index fingertip pad. A control unit, motor driver, and IMU are enclosed in a waist belt. (b) DC motor is mounted at the housing of SAD where subject's index finger is inserted. Cutaneous skin stretch feedback is therefore provided by the shearing between contactor operated by the DC motor and fingertip pad.

sway of the subject during quiet standing. An algorithm developed by Madgwick [66] was used to calculate pitch, roll, and yaw angles efficiently from the IMU data. In this study, only pitch angle was considered to measure the subject's postural sway in anterior-posterior (AP) direction. An embedded control unit (myRIO, National Instruments, Austin, TX) took the IMU data, computed pitch angle of a subject, calculated the desired contactor angular velocity, and controlled the DC motor so that the contactor maintained the desired angular velocity (Fig. 3.1, 3.2, 3.3). We used an h-bridge type motor driver (L298N, STMicroelectronics, Italy) to provide appropriate amount of power for the DC motor. (Fig. 3.2 a).

The IMU, embedded control unit, and motor driver were enclosed in a waist belt so that it could easily be worn by subjects. The overall weight to be worn on the waist is approximately 200 g. The IMU was fixed in the belt for acquiring stable

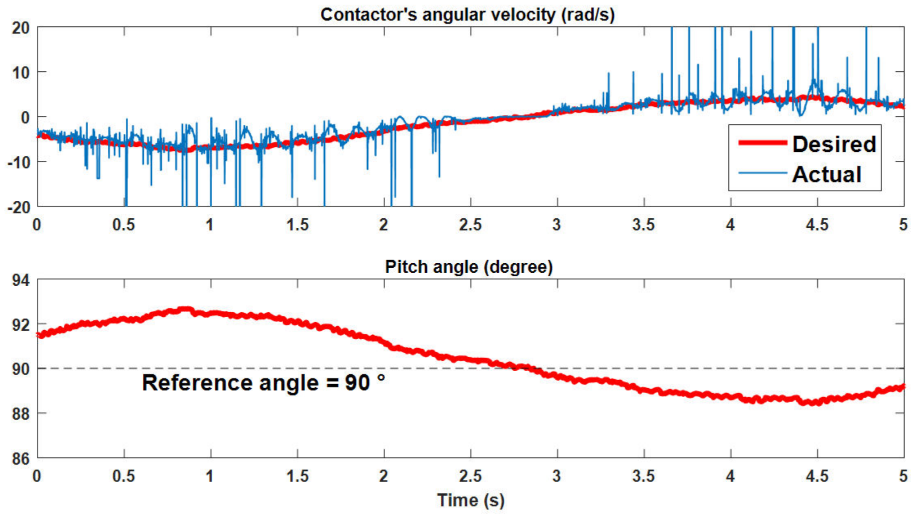

Figure 3.3: Relationship between contactor's angular velocity and pitch angle of subject. In this example plot, the reference angle was set to 90°.

estimate of COM displacement. Sampling rates of SAD, and IMU were 1 kHz, and 500 Hz, respectively (Fig. 3.1, 3.2 a).

3.3.2 Tactile Coding Scheme

To determine the desired angular velocity for the DC motor, a PID feedback controller was implemented. The desired contactor's angular velocity was defined to be proportional to angular deviation of pitch angle from a reference angle which is defined as the subject's averaged pitch angle during upright standing. For example, when a subject leaned forward, the contactor rotates clockwise so that the fingertip pad is stretched backward, and vice versa. In this way, subjects were provided with additional sensory cue (or augmented sensory feedback) of their postural sway. Fig. 3.1 and Fig. 3.3 shows the relationship between contactor's angular velocity and pitch angle. As expected, the contactor's angular velocity tracked the desired angular velocity determined by body postural sway (pitch angle). The reasons for noise presents in actual velocity (Fig. 3.3) are due to i) numerical differentiation and ii) encoder noise. However, implementing an online low-pass filter induced time delay in the system. Therefore, to avoid the detrimental effect of the delay on the stability of the velocity tracking, no filtering was applied to the output signals.

3.4 Quiet Standing Balance Experiment

3.4.1 Subjects

Fifteen healthy young adults (four females and eleven males; mean age ± s.d.: 26.4 ± 5.6 years) with neither neurological nor musculoskeletal impairments participated in this study. Prior to the experiment, subjects were given the instructions about the whole experimental procedure by the investigator and the written consent was obtained from each subject. Subjects were not informed of the function of device. This study was approved by the Texas A&M University Institutional Review Board.

3.4.2 Experimental Protocol

Subjects were asked to stand quietly on a force plate (OR6, AMTI, Watertown, MA) for 30 s with three sensory modalities and two sensory augmentation conditions. The three sensory modality conditions included: i) No Deficit (ND), ii) Visual Deficit (VD), and iii) Visual and Vestibular Deficit (VVD). Other than these, no other instructions, e.g., trying to reduce skin stretch while standing, were given to subjects. For VDD, subjects' vision was eliminated by closing their eyes, and the vestibular system was perturbed by tilting their head backwards for at least 45° in the sagittal plane, which made the tasks more challenging [67] [68] [69] [70]. Under such a head-extension condition, the plane of the vestibular organ is elevated relatively to its normal horizontal orientation, which puts the utricle otoliths into improper position. The vestibular sensory system is then perturbed and causes postural imbalance [67] [68]. Subjects were put on an overhead safety harness for the protection against unexpected falls. The two sensory augmentation conditions included: i) SAD is turned on (ON), and ii) SAD is turned off (OFF). Subjects wore the SAD on their right index fingers (Fig. 3.2 b) and their arms were hung naturally by their sides. When the SAD was turned on, the contactor rotated to induce light skin stretch on the fingertip pad. The skin stretch produced by the SAD was mild such that subjects felt neither pain nor discomfort at the fingertip pad. The belt enclosing an IMU and an embedded control unit was wrapped around waist of subjects (see Fig. 3.1 and Fig. 3.2 a).

The experiment consisted of two parts: i) practice session, and ii) main session. In the practice session, subjects were instructed to stand quietly barefoot on a force plate under three sensory modality conditions: i) ND-OFF, ii) VD-OFF, and iii) VVD-OFF. Each condition was repeated five times. The purpose of practice session was to measure the subject's averaged reference angle while standing quietly. In addition, subjects would familiarize themselves with the testing environment in this session. During the main session, subjects were asked to perform the same quiet standing tasks as in the practice session, with six sensory conditions: i) ND-OFF, ii) VD-OFF, iii) VVD-OFF, iv) ND-ON, v) VD-ON, and vi) VVD-ON. Each condition was repeated 10 times to remove random effects; there were a total of 60 trials in main session. The order of the trials was fully randomized. A two-minute rest was provided between every five trials to avoid muscle fatigue. Upon request, a five-minute break was provided. The whole experiment lasted about two hours. Note that in both practice session and main session, each subject wore the SAD at all times even if there was no cutaneous stimulus provided.

3.4.3 Assessment of Balance

A force plate (OR6, AMTI, Watertown, MA) and a data acquisition system (DAQ) (USB-6002, National Instruments, Austin, TX) with a computer were prepared to measure center of pressure (COP) and pitch angle data, sampled at 1 kHz and 500 Hz, respectively. The processed data was used to evaluate the efficacy of the SAD system.

To quantify the postural sway during quiet standing, we examined multiple traditional COP-based measures [71]. Many studies have evaluated the postural steadiness based on a single measurement [25] [4] [72]. However, it may not be sufficient since some postural sway measures are not sensitive enough to distinguish various aspects of postural impairment [73]. In this study, multiple traditional COP measures were investigated both in time domain and frequency domain [71]. For time domain measures, we calculated the range, mean velocity (MV) and mean frequency (MF) of COP in both AP and medio-lateral (ML) directions. MF is proportional to ratio of *Total Excursion* to *Mean Distance* or equivalently to ratio of MV to *Mean Distance*. *Mean Distance* represents the average distance from the centroid of COP [71]. In frequency domain, centroidal frequency (CF), referred to as the zero crossing frequency, was also computed to characterize the power spectral density of the COP time series in both AP and ML directions.

3.4.4 Statistical Analysis

A two-way repeated-measures analysis of variance (ANOVA) was performed to study the effect of availability of sensation and SAD on quiet standing balance. Significance level was set to α =0.05 (SPSS, v21, Chicago, IL). The cross-correlation (XCORR) function was used to identify the time delay between contactor's angular velocity and COP_{AP} time series. Correlation coefficient between two time series was also calculated using MATLAB (R2015a, MathWorks, Natick, MA).

3.5 Experimental Results

Fig. 3.4 shows the correlation between skin stretch (SAD) and COP in AP direction. Fig. 3.5 shows four postural sway measures of COP data in both AP and ML directions across fifteen healthy young subjects under three sensory modality conditions with SAD on and off. The mean values of these measures across three sensory modality conditions and across two sensory augmentation conditions are grouped and listed in Table 3.1, respectively. In the following, we will first present how skin stretch feedback could successfully control standing postural sway. We will then show how the postural sway measures among different sensory conditions. We will finally present how the SAD affected standing balance and how sensory deficits and sensory augmentation interacted.

3.5.1 Correlation between Skin Stretch and COP_{AP}

The time series of COP displacement in AP direction and angular velocity of a contactor, denoted by $\omega_{contactor}$, is depicted in Fig. 3.4. The example data was selected from one of the subjects (subject no. 6) in VVD condition. Skin stretch on the fingertip pad was generated by the contactor as it rotated at $\omega_{contactor}$. Hence the level of skin stretch can be represented by $\omega_{contactor}$. Since the skin stretch was applied only based on AP direction, we examined COP displacement in AP direction only. The result shows that COP_{AP} movement correlates $\omega_{contactor}$ with $r = 0.88$ and time lags of 172 milliseconds. The average correlation r and time lag of fifteen subjects are 0.82 (s.d. = 0.15) and 150 ms (s.d. = 22 ms) respectively. It indicates that skin stretch ($\omega_{contactor}$) is ahead of COP_{AP} movement suggesting that SAD led the postural sway of the subject in quiet standing.

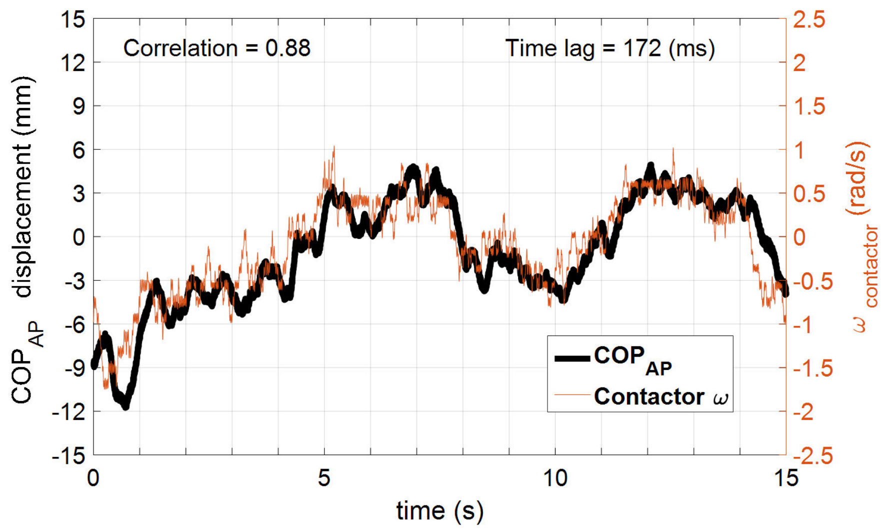

Figure 3.4: Time series of COP displacement in AP direction (black bold line) and contactor's angular velocity over 15-s period. The data was obtained from the same subject (subject no. 6) in VVD condition. Positive correlations (r = 0.88) and positive time lag (172 ms) are shown indicated that skin stretch ($\omega_{contactor}$) is ahead of COP_{AP} displacement. Mean correlation r and time lag are 0.82 (s.d. = 0.15, n = 15) and 150 ms (s.d. = 22, n = 15) respectively.

3.5.2 Effect of Sensory Deficits

All parameters except MF_{ML} ($p > 0.05$) indicated significant differences among three sensory modality conditions as shown Table 3.1. $Range_{AP}$ and MV_{AP} of postural sway were the smallest when all sensory information was available (ND), followed by when vision was removed (VD), and followed by when both vision and vestibular information was removed (VVD) ($Range_{AP}$: $p < 0.001$; MV_{AP}: $p < 0.001$). $Range_{ML}$ of postural sway was greater in VVD compared to ND ($Range_{ML}$: $p = 0.007$). MV_{ML}, MF_{AP} and CF_{AP} showed greater values in VD and VVD conditions than in ND (MV_{ML}: $p = 0.001$; MF_{AP}: $p = 0.014$; CF_{AP}: $p = 0.001$). However, reverse order was shown in CF_{ML}, as CF_{ML} was greater in ND than in VVD (CF_{ML}: $p = 0.009$).

Table 3.1: Measures of postural sway

Parameters	Sensory Modality Conditions			Sensory Augmentation		Interaction
	No Deficit	Visual Deficit	Visual & Vestibular Deficit	SAD ON	SAD OFF	p-value
	(A)	(B)	(C)	(D)	(E)	
Range_{AP} (mm)	21.40 $(1.31)^{BC}$	24.30 $(1.40)^{AC}$	29.00 $(1.90)^{AB}$	25.62 (1.70)	24.14 (1.35)	**0.019**
Range_{ML} (mm)	10.09 $(0.93)^{C}$	11.16 (1.00)	12.33 $(1.31)^{A}$	11.38 (1.19)	11.01 (0.97)	0.181
Mean Velocity_{AP} (MV_{AP}) (mm/s)	7.29 $(0.40)^{BC}$	9.06 $(0.62)^{AC}$	10.61 $(0.78)^{AB}$	9.10 (0.71)	8.88 (0.49)	**0.044**
Mean Velocity_{ML} (MV_{ML}) (mm/s)	3.86 $(0.33)^{BC}$	4.20 $(0.37)^{A}$	4.42 $(0.37)^{A}$	4.10 (0.37)	4.22 (0.34)	**0.027**
Mean Frequency_{AP} (MF_{AP}) (Hz)	0.360 $(0.026)^{BC}$	0.395 $(0.022)^{A}$	0.391 $(0.027)^{A}$	0.370 $(0.026)^{E}$	0.395 $(0.024)^{D}$	0.421
Mean Frequency_{ML} (MF_{ML}) (Hz)	0.461 (0.028)	0.441 (0.03)	0.430 (0.029)	0.421 (0.03)E	0.466 $(0.028)^{D}$	**0.029**
Centroid Frequency_{AP} (CF_{AP}) (Hz)	0.400 $(0.018)^{BC}$	0.441 $(0.018)^{A}$	0.436 $(0.020)^{A}$	0.414 $(0.018)^{E}$	0.436 $(0.019)^{D}$	0.195
Centroid Frequency_{ML} (CF_{ML}) (Hz)	0.209 $(0.023)^{C}$	0.196 (0.017)	0.177 $(0.016)^{A}$	0.181 $(0.014)^{E}$	0.207 $(0.023)^{D}$	0.84

Value represents mean (standard deviation) for three sensory modality conditions and two sensory augmentation conditions, and the interaction (sensory modality × sensory augmentation) p-values. Superscript denotes significant differences from indicated main effect condition ($p < .05$).

Removing sensory information (VD) or challenging balance condition (VVD) significantly increased postural sway, which agrees with the previous studies [74] [14] [12] [75] [76]. As expected, when all the sensory systems are functional, individuals' postural control was significantly better, compared to when there were any sensory deficits. However, only CF_{ML} showed the opposite result. CF_{ML} was greater when all sensory information was available, compared to when both visual and vestibular systems were deprived. CF_{ML} is proportional to the number of zero-crossing points of the detrended data in the ML direction [34]. Prieto et al. [71] reported that CF was positively correlated with the level of difficulties in standing balance. Also CF was reported to be higher with the elderly than young adults. These may suggest that when the quality of sensory information gets worse, more corrective movements of COP may happens in more inefficient ways. However, it is still not clear why CF_{ML} became smaller when all sensory information was removed. The only possible explanation may be that tilting one's head backward with eyes closed somehow helped CF_{ML} since it is not the same as completely removing vestibular information. Future studies are needed to investigate this phenomenon.

3.5.3 Effect of Sensory Augmentation

From Table 3.1, no significant differences between SAD ON and SAD OFF were found in the distance-based measures (Range, and MV) in either AP or ML directions. MF significantly decreased in both AP and ML directions when sensory augmentation was provided. (MF_{AP}: $p = 0.035$; MF_{ML}: $p = 0.005$). CF significantly decreased in both AP and ML directions when sensory augmentation was provided. (CF_{AP}: $p = 0.04$; CF_{ML}: $p = 0.002$).

The effect of induced skin stretch feedback at the fingertip may seem to be con-

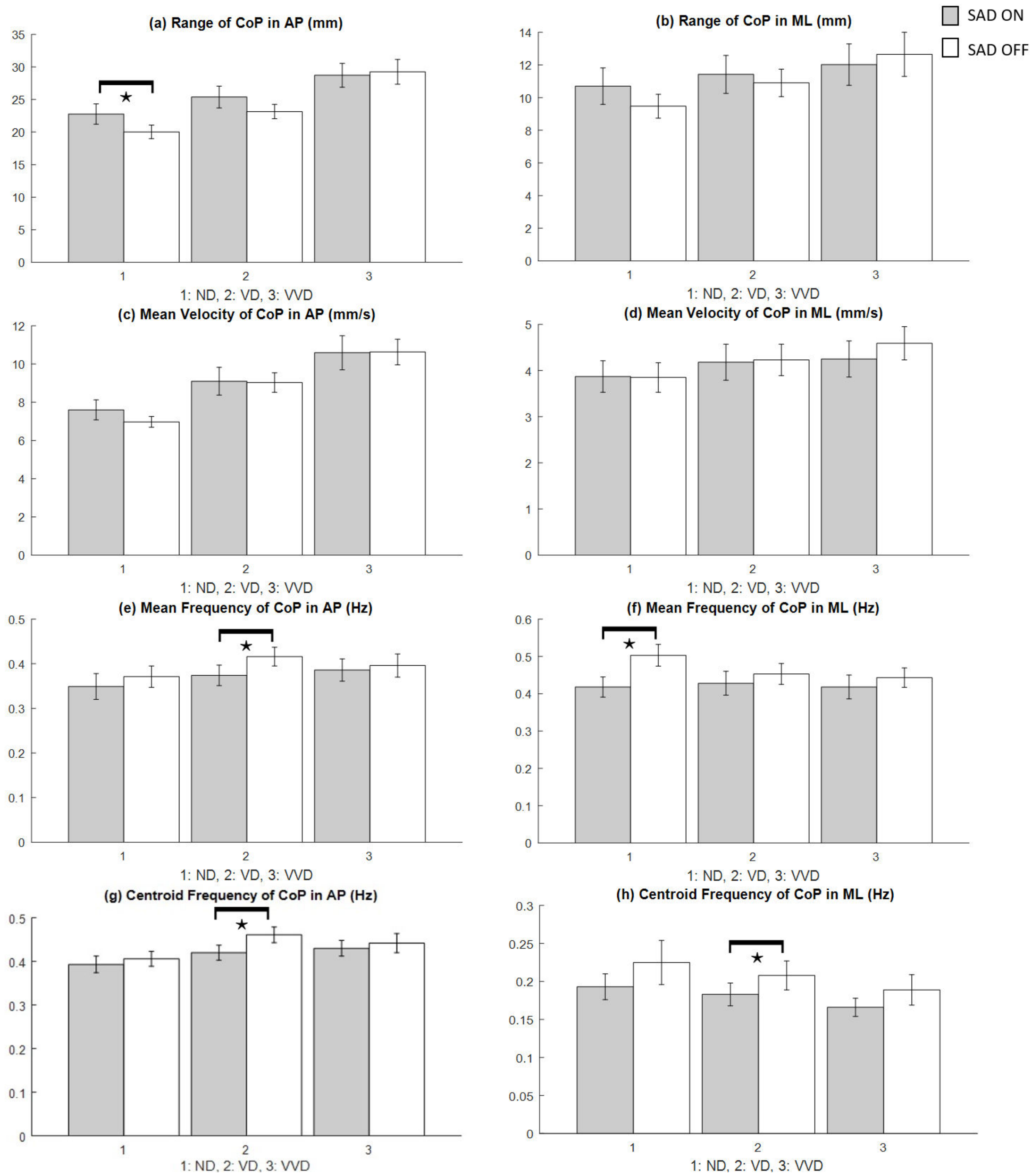

Figure 3.5: Mean values for range, mean velocity, mean frequency, and centroid frequency of COP in both AP and ML directions for each of ten trials under three sensory deficit conditions (1: ND, 2: VD, 3: VVD). Each condition shows when SAD is turned on (Grey) and SAD is turned off (White). Error bars indicate one standard deviation. Significant effects are indicated for $p < .05$ ($\star$) for comparison between two SAD conditions within each of the three levels of sensory deficit conditions.

tradictory. For example, Range_{AP} for ND significantly increased with skin stretch at the fingertip pad, whereas MF_{AP} for VD, MF_{ML} for ND, CF_{AP} for VD, and CF_{ML} for VD decreased significantly with skin stretch at the fingertip pad (Fig. 3.5). Since the objective of this study was to examine the feasibility of the developed sensory augmentation system for balance rehabilitation, we wanted to carefully investigate how the proposed sensory augmentation system can enhance the balance of the people with the simulated sensory deficits.

For ND condition, Range_{AP} increased, suggesting that skin stretch feedback may have worsened balance in the AP direction when no sensory deficit was present. Similar trends without statistical significance were found for Range_{ML} and MV_{AP} (Fig. 3.5). These results seemed to disprove the feasibility of the device for balance rehabilitation. However, it was worthwhile to investigate the trends of these variables when more sensory modalities were removed. For VVD condition, both directions in Range and MV were observed to become smaller when SAD was on compared to when SAD was off (Fig. 3.5). This was captured by the interaction effects. There were significant interaction effects between sensory modality and SAD for Range_{AP} ($p = 0.019$), MV_{AP} ($p = 0.044$) and MV_{ML} ($p = 0.027$) (See Table 3.1).

For frequency measures (i.e., MF and CF), it is interesting to note that mean values of MF and CF are always smaller for SAD ON condition compared to SAD OFF condition (Fig. 3.5). These results suggest that additional skin stretch feedback induced at the fingertip pad corrected postural sway. The decrease of MF_{AP} and MF_{ML} in all sensory conditions due to sensory augmentation may imply that the sensory augmentation due to SAD reduced the effective postural sway that may not be captured by the mean values of some postural sways (e.g., Range, MV and Mean

Distance). This may indicate that subjects reduced oscillatory movements of COM in the presence of skin stretch feedback while making more total COP movement that is proportional to the distance-based measures (e.g., Range, MV and Mean Distance). Increase in the total COP movement (proportional to MV) may indicate a higher regulatory balancing activity required during quiet standing [77] [78]. Thus we may speculate that subjects more actively controlled their posture while additional skin stretch feedback was provided. CF is associated with muscle and joint stiffness [11]. CF may give us an insight on how well the postural control could be achieved under different task constraints [72]. The significant decrease in CF_{AP} and CF_{ML} with sensory augmentation may imply that a sensory augmentation via skin stretch feedback compensates some underlying neurological or musculoskeletal disorders [78], therefore enhancing quiet standing postural control.

In the literature, MV was suggested as the most significant measure for separating different groups (e.g. age) [71] and the most reliable among traditional parameters [79]. In our study, no significance was found for the sensory augmentation effects in MV, suggesting that MV may not be sensitive to sensory augmentation. However, MF was found to be sensitive to sensory augmentation. Since the definition of MF is the ratio of MV to Mean Distance, MF was able to capture the effective postural sway that could not be interpreted by single variables such as MV and Range.

3.5.4 Interaction Effects of Sensory Deficits × Sensory Augmentation

The analysis revealed a significant interaction effect between sensory modality and sensory augmentation in Range_{AP} ($p = 0.019$) as presented in the rightmost column in Table 3.1. While applying skin stretch feedback, Range_{AP} tended to decrease in VVD whereas it tended to increase Range_{AP} in ND and VD when compared to when

SAD was turned off. Pairwise comparisons revealed that SAD significantly increased Range_{AP} ($p = 0.037$) in ND. Similarly, SAD provided positive effect on MV_{AP} for VVD as it was slightly lower for VVD, but slightly went up for ND and VD conditions with SAD on (MV_{AP}: $p = 0.044$). SAD also tended to enhance MV_{ML} for VVD condition ($p = 0.06$) whereas SAD did not seem to affect MV_{ML} for ND and VD conditions (MV_{ML}: $p = 0.027$). Significant interaction effects were also shown in MF_{ML} of postural sway (MF_{ML} $p = 0.029$). Pairwise comparisons of the interaction categories showed that MF_{ML} tended to decrease more in ND ($p < 0.001$) than in VD and VVD conditions when skin stretch feedback was applied. SAD significantly decreased MF_{AP} ($p < 0.001$), CF_{AP} ($p = 0.002$), CF_{ML} ($p = 0.023$) in VD condition. There were no significant interaction effects observed from Range_{ML}, MF_{AP}, CF_{AP} and CF_{ML}.

Range_{AP} for the ND condition worsened due to skin stretch feedback. A possible reason may be that during the ND condition, healthy young subjects already had good enough quality sensory information in maintaining balance such that the additional artificial biofeedback inputs may have interfered with the visual or other sensory cues. In other words, skin stretch feedback may have caused distractions to subjects during the ND condition. This is consistent with previous studies including attention and control studies of posture and gait [80] [81]. Therefore, we postulate that there could be a threshold of postural sway above which the additional artificial biofeedback may enhance the postural sway. On the contrary, when a person's postural sway is less than the threshold, the additional artificial biofeedback may worsen the postural sway. Since healthy young subjects are assumed to be optimal in postural control, their postural sway can be assumed to be less than the threshold. Therefore, the additional artificial biofeedback can be distracting. However, when more sensory information is removed, their postural sway may become greater than the threshold, and the additional artificial biofeedback may enhance the postural sway. The existence of a threshold needs to be examined in the future work.

3.6 Discussion

In this chapter we presented a sensory augmentation system that is able to detect the body sway angle by the integrated IMU and provides an additional cue of postural sway by SAD. Unlike the existing techniques that require a reachable fixed surface, our system offers a wearable device (SAD), which is lighter, smaller, less expensive, more flexible and has better wearability compared to current laboratory-based postural control systems. Skin stretch feedback levels were regulated by the amount of deviated pitch angle from a reference angle that could be detected by the IMU. This light somatosensory feedback seemed to correlate with COP positively and was in phase with body sway, which may demonstrate the feasibility of this sensory augmentation system.

The correction of postural control with sensory augmentation at the fingertip can be caused by sensorimotor integration at either spinal (i.e., spinal cord) or supraspinal (i.e., somatosensory cortex) level [65] [82] [83]. Manjarrez et al. [84] reported that random tactile feedback applied to the fingertip of a cat has increased spinal and cortical evoked field potentials, suggesting both spinal and supraspinal level sensorimotor integration. Similarly, vibrotactile stimulation at the human fingertip pad enhanced upper limb motor performances possibly due to the enhanced sensorimotor integration at the spinal or supraspinal level [65] [82] [83]. Jeka et al. found that COP displacement [5] [6] and left leg EMG activity [62] followed the lateral fingertip force with a time lag of approximately 300 ms and 150 ms respectively, suggesting that the response may be a supraspinal long-loop pathway [85] [86]. Nashner [87] found that a long-latency postural reflex (120 ms) helps to reduce postural sway, which is usually classified as a supraspinal pathway [88] [7]. In our study, the time lag was approximately 150 ± 22 ms (mean value ± s.d.) hence we consider that the enhancement of postural control via skin stretch feedback may be due to sensorimotor integration at the supraspinal level.

There may be several reasons why using a SAD for balance rehabilitation can be useful. First of all, small size and light weight make this design a favorable wearable application to neurologically impaired and physically weak patients. The weight to be put on finger is approximately 20 g; the overall weight to be worn on the waist is approximately 200 g. Therefore, the additional inertia added to the postural control system is so small that it does not affect natural conditions of a subject [89]. Moreover, body sway angle is measured by the IMU which is small, light, and highly accurate on measuring body orientation. Second, the whole system is portable so that patients are not limited to the working space. Previous studies [5] [6] [8] [64] [7] required reachable fixed surfaces or sizable laboratory equipment to obtain additional somatosensory cues from fingers. It is not practical in their home rehabilitation. The proposed SAD in our study allows patients to perform self-training in home or any other place they prefer, which can help patients increase the dose and convenience of the balance rehabilitation.

3.6.1 Limitation

Some limitations and potential future works of this study are illustrated as follows. As mentioned before, SAD may be a distraction to subjects with good quality sensory information while they are performing balancing control. This is partially due to the artificial nature of the augmented sensory signals. A different control strategy for generating augmented sensory signals may resolve this problem. For example, instead of deviated angle from the reference, sway velocity can be used to proportionally induce skin stretch at the fingertip. The comparison between position-based and velocity-based control strategy will be discussed in Chapter 4. Different populations (e.g., the elderly or patients with balance disorders) can be examined instead of healthy young adults with simulated sensory deficits Furthermore, because applying skin stretch feedback at the fingertip may hinder the use of the hand and fingers and eventually activities of daily living, we will investigate the effects of other potential locations of skin stretch for balance control. A wrist-worn skin stretch device will be presented in Chapter 5.

3.7 Summary

A prototype of a sensory augmentation system for postural control rehabilitation has been developed using skin stretch feedback. The feasibility of the developed system for balance rehabilitation was evaluated. The results showed that the sensory augmentation due to skin stretch feedback at the fingertip can enhance balance as evidenced by several traditional postural sway parameters even though there are several improvements that can be made for better enhancement of balance. Overall, the skin stretch feedback showed great potential in balance rehabilitation. The findings in this study can also lead to development of portable balance rehabilitation devices for use in activities of daily living.

4. POSITION AND VELOCITY INFORMATION FOR BALANCE CONTROL

4.1 Introduction

In Chapter 3, the development of a portable sensory augmentation system for balance rehabilitation using skin stretch feedback has been presented. The sensory augmentation device proportionally induced skin stretch at the fingertip when subjects moved away from their neutral position. Results have shown that the sensory augmentation due to skin stretch feedback could improve postural stability as evidenced by several traditional measures of postural sway. In this study, both position and velocity information of the body sway are considered as the reference to correct postural sway during quiet standing. Most studies on wearable biofeedback device have focused on correcting the error signals based on trunk tilt angle [47] [90] [44] [91] or trunk acceleration [92] [55]. However, intuitively, *light touch* encodes the directional friction experienced by the users due to rate change of body sway, suggesting that tactile feedback from *light touch* can be related to the trunk tilt rate, instead of trunk tilt angle.

The objective of this chapter is to examine which augmented feedback of body sway, position or velocity, is a more useful sensory feedback in quiet standing balance. Quiet standing balance of healthy young adults with normal and simulated perturbed sensory inputs (vision, vestibular and proprioceptive systems) were tested in this study. It was hypothesized that i) body sway velocity is a more natural inherited form, compared to position, to perceive by subjects wearing the sensory augmentation device; and ii) augmented feedback of body sway velocity would enhance postural stability.

4.2 Background

Postural sway velocity information was found to be a more accurate form of information acquired by sensory systems when compared to its sway position and acceleration information for human standing balance [16] [93]. If the sensory modalities providing velocity information (e.g., optic flow by vision or changes of muscle length by proprioception at joints) are removed or perturbed, the velocity information could be inaccurately perceived, causing imbalance during quiet standing. Both direction and velocity information of postural sway can be provided and augmented by additional *light touch* on a stationary surface [94] [95] [7]. Moreover, it was found that postural sway was highly consistent with the driving frequency of moving surface where subjects touching on, which emphasizes the important role of cutaneous feedback in influencing the control of upright posture [95] [7].

Cutaneous feedback has been of great interest as a way of sensory augmentation to improve balance [5] [8] [64] [45]. Jeka et al. [5] introduced the concept of *light touch* for the standing balance and found that subjects with additional light touch (with the normal force less than) at the fingertip on a fixed surface had reduced postural sway in tandem Romberg posture. Skin stretch and skin deformation at the fingertip provided subjects with additional information to identify the direction of body sway. Clapp and Wing [8] showed a reduction in the COP fluctuations in the sagittal plane when subjects were making light contact with their fingertips on a fixed surface during normal bipedal stance, suggesting that additional sensory input at the fingertip provides more robust regulation of postural stability. It has been proven by partially blocking sensory afferents in human standing, which the light touch effects were not due to the mechanical support on the fingertip but due to the tactile feedback [4]. Krishnamoorthy et al. [64] observed that body sway with a light touch at the head or neck reduced more effectively than a finger touch; they also found that modulation of contact forces resulted in postural sway reduction. These two additional sources provided sensory information of a fixed reference point in space and the transient force changes related to the body sway, respectively.

To better interpret standing balance in terms of stochastic process, stabilogram diffusion analysis (SDA) [96] and invariant density analysis (IDA) [97] have been used in the postural control literature in addition to traditional postural sway measures. SDA suggests that the COP trajectories are not purely random and two postural control mechanisms (i.e., open-loop and closed-loop) are involved in maintaining quiet standing. This analysis was used to evaluate the effects of visual input [98], age and fall status [99], audio-biofeedback [92], and plantar cutaneous sensation on postural stability [100]. Hur et al. [97] proposed a stochastic Markov chain model that characterize the long term behaviors in quiet standing. This technique can provide specific information about the human postural control system and recreate the actual sway behavior. It can also be used to differentiate the effects of age [97], and vision and weight of air bottle [75] on postural control.

4.3 Sensory Augmentation System

The sensory augmentation system was developed based on the system described in [45] to study whether the augmented feedback of position and/or velocity information of body sway (in AP direction) is more effective than another. The whole system consists of a skin stretch feedback apparatus, an inertial measurement unit (MPU-9150, InvenSense Inc., USA), a motor driver (L298N, STMicroelectronics, Italy), and a microprocessor (Teensy 3.6, 32-bit 180 MHz ARM Cortex-M4 processor). The

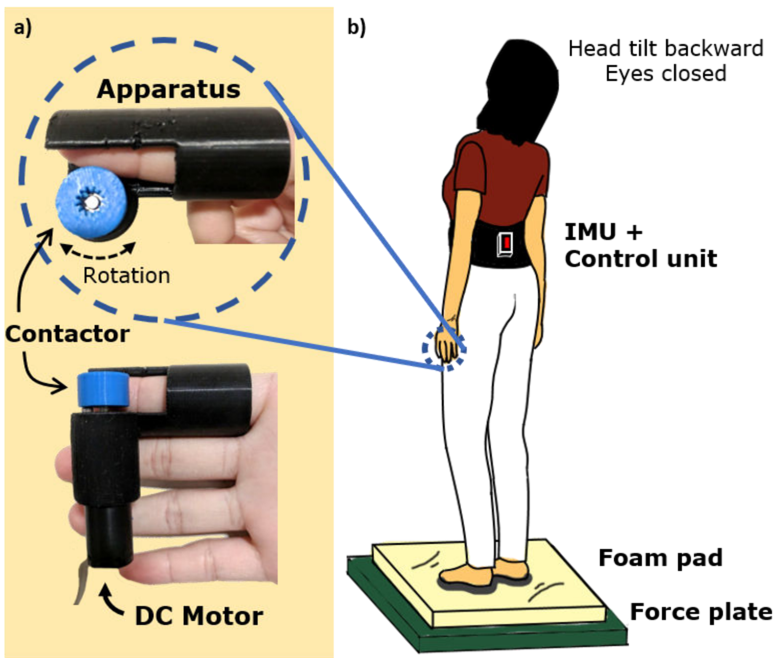

Figure 4.1: Sensory augmentation system and experiment setup. a) The sensory augmentation device induces skin stretch at an index fingertip pad with a rotating contactor. The rotating motion of the contactor is controlled by a DC motor. b) In the experiment, subjects were asked to tilt their head up with eyes closed, while standing on a foam pad on the force plate.

feedback apparatus consists of a DC motor (1524T009SR, Faulhaber, Germany) that rotates a contactor to provide the skin stretch at the fingertip (Fig. 4.1 a). Various sizes of housings and contactors were fabricated to accommodate for the difference in subject's finger sizes (Fig. 4.2). To absorb the pressure produced by the finger joints while wearing the apparatus, the housing of the DC motor was fabricated with 3D-printed flexible filaments (TPU flexible material) instead of the ABS-type plastic filament used in the previous design. The material of the contactor remains 3D-printed in plastic filament. The IMU, control unit, and power source are enclosed in a waist belt for estimating the static posture and computing the desired control output for the motor. Control strategy for the skin stretch feedback of position on stance was introduced in [45]. In the following section two other control strategies will be introduced.

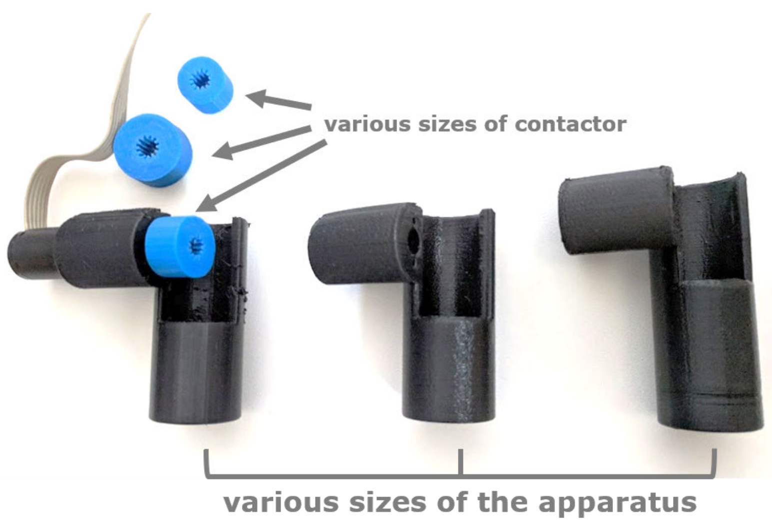

Figure 4.2: Various sizes of the feedback apparatus and contactors.

4.3.1 Augmented Feedback Strategies

Three types of feedback strategies based on different dynamical information of body sway were used: i) augmented position feedback (P), ii) augmented velocity feedback

(D), and iii) augmented position and velocity feedback (PD). For position-based control (P) (which was proposed and used in Section 3.3.2 of Chapter 3), the amount of skin stretch feedback is proportional to the body tilt deviated from the neutral position in AP direction. That is, as the subject leans further away from his/her neutral position, the contactor rotates faster in order to provide greater skin stretch feedback. The relationship between the skin stretch feedback and standing posture can be described as follows:

$$\omega_c = k_p(\theta_{AP} - \theta_{ref}) \tag{4.1}$$

where ω_c is the angular velocity of the contactor, corresponding to the amount of skin stretch feedback. θ_{AP} is the body tilt in the AP direction and θ_{ref} is the neutral position recorded in the AP direction. k_p is the control gain. The control gains can be tuned to ensure that all subjects would be able to feel light touch contact cues at their fingertips. For all subjects, the maximum ω_c was set to be 25 rad/s.

For velocity-based control (D), the body tilt deviation is ignored, whereas the body sway velocity is considered. The relationship between the skin stretch feedback and standing posture can be described as:

$$\omega_c = k_d\omega_{AP} \tag{4.2}$$

where the ω_{AP} represents the body sway angular velocity and k_d is the control gain. The amount of skin stretch feedback is proportional to the body sway velocity in the AP direction. That is, as the subject remains still, even if s/he leans forward/backward with respect to the neutral position, the contactor would not induce any skin stretch feedback; as the subject starts swaying back and forth, the contactor rotates either counterclockwise or clockwise to provide the direction and magnitude information to the subjects.

For position and velocity-based control (PD), the amount of skin stretch feedback is proportional to the magnitude of body tilt and body sway velocity, which can be defined based on the combination of aforementioned feedback strategies:

$$\omega_c = k_p(\theta_{AP} - \theta_{ref}) + k_d\omega_{AP} \tag{4.3}$$

Under such control approach, the contactor rotates faster to provide greater skin stretch feedback as the amount of postural sway displacement and/or velocity increases. All three feedback strategies were implemented in the sensory augmentation system and can be selected manually for various experimental setup.

4.4 Quiet Standing Balance Experiment

The aim of this experiment was to examine and compare the effects of different augmented feedback strategies on standing balance using our developed sensory augmentation system. Three feedback strategies described in the previous section and the feedback-off condition were tested on healthy young subjects to investigate which control strategies more effectively aids in controlling posture during quiet standing.

4.4.1 Subjects

A total of ten healthy young subjects (age $\pm$ s.d: 26.3 ± 3.26, range 22-31 years; three females) were recruited from the general university population. These individuals have neither been neurologically impaired nor had balance issues before. They were informed of the experimental procedures and had signed the consent form before the experiments started. The study was approved by the Institutional Review Board at Texas A&M University. Prior to the experiment, subjects were instructed about the functionality of the skin stretch feedback apparatus as well as three augmented feedback strategies.

4.4.2 Experimental Protocol

Participants were asked to put on the waist belt that encloses the inertia sensor at the approximated COM location (lower back) and wear the skin stretch feedback apparatus on their index fingers (Fig. **??**). During the experiment, they were asked to perform 30-s quiet standing tasks under different sensory modality conditions while standing on a force plate (OR6, AMTI, Watertown, USA). The whole experiment consisted of one practice session followed by the main session. In the practice session, subjects were given about ten minutes to familiarize themselves with the device and to understand how different augmented feedback strategies relate to their standing posture. No data was recorded in this session.

After subjects were familiarized with the use of the sensory augmentation system, they were asked to stand on a force plate and perform the quiet standing tasks. Four feedback conditions combining two sensory modality conditions were tested. For the sensory feedback conditions, three kinds of control strategies to induce the skin stretch feedback, i) P ii) D iii) PD, along with the task without feedback (OFF) were included. For sensory modality conditions, i) normal quiet stance without any perturbation and ii) perturbed stance with simulated visual, vestibular and proprioceptive systems deficits were included. In perturbed stance, subjects were asked to close their eyes, tilt their heads up at least 45° while standing on a compliant surface. Under the head extended posture, the vestibular organs are offset beyond their optimal working range which could lead to the destabilization of postural balance [67,68]. A 2-inch foam pad was placed on the top of the force plate for perturbing standing due to the altered sensory inputs to both ankle joint receptor and cutaneous mechanoreceptors in the foot [101, 102]. In the normal quiet standing condition, subjects were asked to keep theirs eyes open, look straight ahead while standing on a force plate without the foam surface.

A total of 40 trials of quiet standing task were performed in the main session (4 feedback conditions $\times$ 2 sensory modality conditions $\times$ 5 repetitions). Normal standing tasks were followed by pertubed standing tasks. The orders of these trials in two standing tasks were randomized among subjects. During each trial, subjects were instructed to *"respond to the skin stretch cues by stabilizing their posture in the sagittal plane"*. In our previous study [45], it has been shown that the passive skin stretch cues from the apparatus can be easily interpreted and responded by healthy young subjects in a mean value of 150 ms (s.d. = 22, N = 15). In this study, subjects were asked to have external focus on the postural cues at their fingertip and control body sway accordingly. Break was provided upon request and a two-minute rest was provided between every five trials to avoid muscle fatigue. The whole experiment lasted around 40 minutes. Note that in both sessions, subject wore the feedback apparatus throughout the entire experiment including the tasks without skin stretch feedback. Subjects were also unaware of which sensory augmentation condition would be performed in each trial.

After the completion of all trials, subjects were asked to complete a questionnaire to provide subjective ratings of skin stretch feedback intuitiveness, capability of discerning cue direction and cue intensity, comparison among different feedback controllers, and feasibility of the system being used for postural control enhancement. More details about questions and rating scales for subjective quality assessment can be found in the next section.

4.4.3 Assessment of Balance

Standing balance performance under different feedback conditions was assessed by the recorded COP trajectories of 30-s quiet standing. The COP data was recorded by a force plate and a data acquisition system (USB-6002, National Instruments, Austin, TX), sampled at 1kHz. The data were processed offline using MATLAB (R2018b, MathWorks, Natick, USA). To have a comprehensive view of the postural control system and to better capture the effects of sensory augmentation during quiet standing, three assessments of balance were adopted and introduced as follows. For all three methods, COP measures were computed in anterior-posterior (AP), medio-lateral (ML), and radial directions. Note that radial distance is the Euclidean distance of the AP and ML distances.

- Traditional measures

  Five traditional COP-based measures in time and frequency domains were calculated and analyzed [71]. Four time-domain measures were computed: i) the maximum distance travelled by the COP (Range), ii) root mean square (RMS) distance of the COP, iii) mean velocity of the COP (MVel), and iv) the area enclosed by the COP displacements (Sway Area). For frequency-domain measures, the centroidal frequency (CF) was computed for characterizing the frequency distribution of the COP displacements. CF is the frequency at which the spectral mass is oncentrated, and can also be referred to as the zero crossing frequency [71].

- Stabilogram diffusion analysis (SDA)

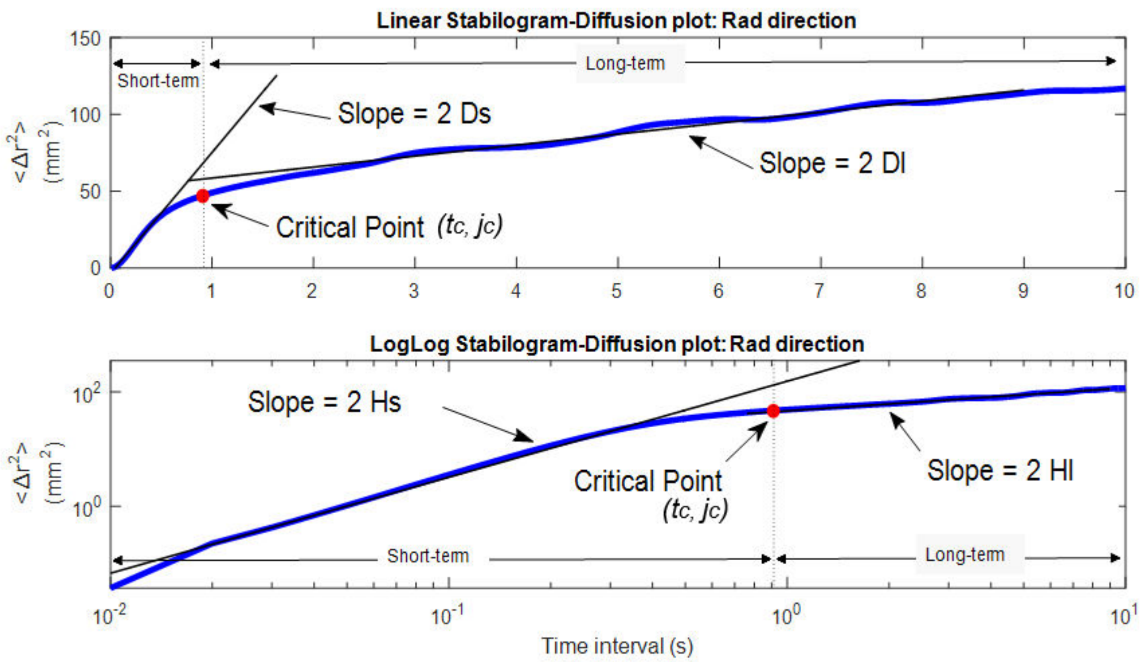

Figure 4.3: Linear and log-log stabilogram diffusion plot in radial direction. Short-term and long-term regions fitted by straight black regression lines are dominated by the open-loop and closed-loop control strategies respectively. Stabilogram diffusion parameters (D_s, D_l, H_s, and H_l) are determined by the slopes of lines fitted to short-term or long-term regions. A critical point is defined as the point where the slope changes considerably from one region to another.

Collins and DeLuca [96] proposed that SDA could, from the viewpoint of stochastic processing, provide several physiologically meaningful parameters from the stabilogram. Study has shown that postural control system during quiet standing can be categorized into two control mechanisms: the open-loop and closed-loop. Fig. 4.3 demonstrates an sample plot of a resultant planar stabiliogram-diffusion plot generated from a subject performing 30-s quiet standing. In the figure, there are two regions identified in both linear and log-log stabilogram diffusion plots, short-term and long-term regions, which are dominated by the open-loop and closed-loop control mechanisms, respectively. The following parameters are introduced for characterizing the postural control behavior.

Diffusion coefficients (D_s, D_l), which equal to one-half of the slopes of the fitting line in the linear plot (Fig. 4.3), represent the level of stochastic activities of the COP about the plane of support. The subscripts denote the short-term (s) and long-term (l) regions. A greater D_s, as compared to D_l, reflects a more random behavior in short-term interval. Scaling exponents (H_s,H_l), which equal to the slopes of the fitting line in the log-log plot (Fig. 4.3), evaluate the persistence/anti-persistence of the COP behavior. Scaling exponents suggested that COP tends to move away from the equilibrium during short-term interval ($H_s > 0.5$, persistent) whereas COP tends to return to the equilibrium over long-term interval ($H_l < 0.5$, anti-persistent). The last two parameters are critical point coordinates (t_c, j_c). A critical point is defined as the point where the slope changes considerably from one region to another, representing by the critical time interval (t_c) and critical mean square displacements (j_c).

- Invariant density analysis (IDA)

Hur et al. [97] proposed a reduced-order finite Markov chain model for analyzing the stochastic structure of postural sway, which provides insights of the long-term postural control system behavior. This model describes the states (zero-mean COP data) of the dynamical system and evolution of those states. It was found that the distribution of COP over the state space would eventually converge to a unique steady state distribution π, known as *invariant density*. Five parameters were computed as follows for describing the subject-specific COP behaviors. P_{peak} represents the largest probability of π. MDist is the average location (state) of the COP with the unit of mm. D95 is the largest state at which there is a 95% probability on containing the COP. D95 has the same unit as MDist. EV2 is the second largest eigenvalue of the transition matrix and describes the convergence rate of the system to its invariant long-term behavior. Entropy is the measure of randomness.

4.4.3.1 Subjective ratings

In addition to the quantitative measure of overall performance, qualitative scores of additional postural cues on standing balance were also collected by a post-experiment questionnaire. The questionnaire consisted of seven questions: 1) *Can you discern the direction of the skin stretch cues at your fingertip?*, 2) *Can you discern the intensity of the skin stretch cues at your fingertip?*, 3) *Can you distinguish among three types of controllers?*, 4) *What type(s) of controller was/were more intuitive for you when correcting the posture?*, 5) *Do you think the skin stretch feedback helps correcting your posture when sensory modalities were not perturbed?*, 6)*Do you think the skin stretch feedback helps correcting the posture when sensory modalities were perturbed?* and 7) *Will you recommend this device to be used by elderly or neurological disorder patients in retraining balance?*. For the questions 1 and 2, the ratings were represented by a

7-point scale from "Very difficult" (1) to "Very easy" (7); for questions 5 to 7, the ratings were represented by a 7-point scale from "Not at all" (1) to "Definitely" (7). Comments were also collected in the end of the questionnaire.

4.4.4 Statistical Analysis

An one-way repeated-measures analysis of variance (ANOVA) was conducted to examine whether the different sensory augmentation conditions have a significant effect on standing balance performance. Significant effects in the one-way ANOVA was followed by post-hoc pairwise comparisons tests applying Bonferroni correction. The pairwise multiple comparisons was applied to examine the differences between all pairs of sensory augmentation condition. Significance level was set to α =0.05 (SPSS, v21, Chicago, IL).

4.5 Experimental Results

Measures of postural sway under four feedback conditions during normal and perturbed quiet standing are presented in Table 4.1 and Table 4.2, respectively. In Table 4.1 and 4.2, value represents mean (s.d.) across 10 subjects. Bold text denotes significant dependence on sensory augmentation condition and superscript denotes statistical significance between conditions ($p < .05$). Traditional, SDA and IDA measures of postural sway in radial, AP, and ML directions are reported. Effects of sensory augmentation among four feedback conditions (i.e., OFF, P, D, and PD) were analyzed using one-way ANOVA and the posthoc pairwise comparisons were further tested for each pair of conditions. The following sections summarize the effects of sensory augmentation between OFF and all other three feedback conditions and the comparison among these three feedback conditions. Results of the questionnaire collected from all subjects are also presented.

Table 4.1: Measures of postural sway in AP, ML and Rad directions for normal quiet standing tasks

	Measure	Normal Condition				p-value
		OFF (a)	P (b)	D (c)	PD (d)	
Traditional Measures (TRAD)	Range (mm)	9.44(3.45)	13.48(7.71)	8.88(3.59)^{d}	11.89(4.84)^{c}	**0.016**
	Range-AP (mm)	16.41(6.59)	22.37(11.38)	15.56(6.22)	20.15(8.34)	**0.015**
	Range-ML (mm)	8.49(2.23)	8.34(2.31)	7.26(1.27)	7.86(2.40)	0.2
	RMS (mm)	3.99(1.44)	4.69(1.68)	3.65(1.44)	3.92(1.09)	0.27
	RMS-AP (mm)	3.56(1.48)	4.24(1.82)	3.25(1.52)	3.51(1.19)	0.34
	RMS-ML (mm)	1.66(0.48)	1.79(0.54)	1.53(0.27)	1.57(0.56)	0.62
	MVel (mm/s)	7.62 (1.68)^{b}	9.39(1.8)^{ac}	7.56 (1.45)^{bd}	9.26 (2.39)^{c}	**0.011**
	MVel-AP (mm/s)	6.12(1.3)^{bd}	7.89(1.44)^{ac}	6.09(1.23)^{bd}	7.81(2.13)^{ac}	**0.01**
	MVel-ML (mm/s)	3.36(0.99)	3.67(1.0)	3.29(0.97)	3.57(0.85)	0.19
	CF (Hz)	.61(.12)	.68(.14)	.65(.16)	.73(.09)	0.21
	CF-AP (Hz)	.42(.09)^{d}	.48(.06)	.46(.11)	.53(.06)^{a}	0.08
	CF-ML (Hz)	.21(.07)	.20(.06)	.24(.08)	.22(.06)	0.3
	Sway Area (mm^2)	192(98)^{b}	264(129)^{ac}	170(84)^{b}	215(97)	**0.004**
Stabilogram diffusion analysis (SDA)	D_s (mm^2/s)	10.23 (6.23)	16.54 (11.38)	8.21(4.46)	16(11.6)	0.06
	D_s-AP (mm^2/s)	8.04(5.4)	14.07(11)	6.46(3.68)	13.99(10.06)	0.08
	D_s-ML (mm^2/s)	2.24(1.18)	2.54(1.44)	1.74(1)	2.7(2.32)	0.38
	D_l (mm^2/s)	0.97(0.86)	1.83(1.52)^{c}	0.91 (0.89)^{b}	1.05 (1.21)	**0.043**
	D_l-AP (mm^2/s)	0.52(1.15)	1.45(1.57)	0.77(1.01)	0.76(1.01)	0.18
	D_l-ML (mm^2/s)	0.11(0.14)	0.27(0.32)	(0.14(0.1)	0.25(0.4)	0.54
	H_s	.71(.07)	.76(.06)	.71(.06)	.76(.05)	0.07
	H_s-AP	.72(.07)	.76(.06)	.71(.06)	.78(.05)	0.08
	H_s-ML	.69(.09)	.72(.1)	.69(.08)	.71(.1)	0.5
	H_l	.15(.1)	.17(.09)	.16(.04)	.11(.1)	0.76
	H_l-AP	.12(.13)	.14(.09)	.15(.06)	.1(.07)	0.67
	H_l-ML	.09(.09)	.18(.16)	.17(.1)	.16(.18)	0.35
	t_c (s)	1.59(.54)	1.72(.54)	1.21(.34)	1.41(.49)	0.17
	t_c-AP (s)	1.92(.42)	1.78(.52)	1.47(.44)	1.57(.52)	0.41
	t_c-ML (s)	1.51(.46)	1.31(.43)	1.3(.27)	1.35(.32)	0.4
	j_c (mm^2)	25.33 (13.05)^{c}	41.96(32.44)	14.84(7.25)^{a}	35.21(32.58)	**0.014**
	j_c-AP (mm^2)	25.34 (21.04)	37.67(32.61)	13.99(11.48)	31.88(30.29)	**0.027**
	j_c-ML (mm^2)	4.78(2.11)	4.56(2.39)	2.94(.8)	4.21(2.44)	**0.046**
Invariant density analysis (IDA)	P_{peak}	.11(.11)	.12(.08)	.11(.07)	.11(.09)	0.99
	P_{peak}-AP	.07(.03)	.10(.07)	.10(.07)	.12(.09)	0.25
	P_{peak}-ML	.21(.17)	.25(.18)	.18(.08)	.20(.12)	0.77
	MDist (mm)	4.30(1.98)^{c}	4.57(1.82)^{c}	3.19(0.92)^{abd}	4.07(1.09)^{c}	**0.006**
	MDist-AP (mm)	3.03(1.0)	3.91(2.15)	2.72(1.02)	3.52(1.26)	0.33
	MDist-ML (mm)	2.17(2.1)	1.75(0.36)^{c}	1.35(0.21)^{b}	1.65(0.50)	**0.024**
	D95 (mm)	4.29(1.98)	4.57(1.82)	3.19(0.92)	4.06(1.09)	**0.006**
	D95-AP (mm)	7.09(2.17)	9.69(6.31)	6.67(3.01)	7.91(2.63)	0.45
	D95-ML (mm)	3.18(1.47)	3.78(0.9)	3.04(0.71)	3.52(1.18)	0.24
	EV2	.992(.01)	.994(.01)	.985(.02)	.992(.01)	0.24
	EV2-AP	.99(.01)	.99(.01)	.987(.01)	.993 (.005)	0.72
	EV2-ML	.98(.027)	.971(.03)	.976(.015)	.969(.035)	0.12
	Entropy	4.46(1.03)	4.35(0.9)	4.28(0.96)	4.47(0.77)	0.97
	Entropy-AP	4.84(0.61)	4.54(0.84)	4.39(0.74)	4.38(0.74)	0.47
	Entropy-ML	3.44(1.01)	3.07(1.08)	3.35(0.69)	3.37(0.9)	0.83

4.5.1 Effects of sensory augmentation during normal standing

Table 4.1 shows that Range, mean velocity (MVel) of COP in both radial and AP directions and Sway Area were significantly dependent on which feedback conditions were selected ($p < 0.05$). Pairwise comparison indicated that position feedback significantly increased the mean velocity (in both radial and AP directions) and Sway Area of COP compared to the condition without feedback. In PD condition, MVel-AP and CF-AP both increased compared to OFF condition. There were no significant difference between D and OFF conditions for all TRAD measures.

For SDA measures, D_l, j_c, j_c-AP and j_c-ML revealed significant main effects of sensory augmentation conditions. Pairwise comparison indicated that velocity feedback significantly decreased j_c when compared to the condition without feedback. There were no significant difference between P and OFF or between D and OFF conditions for all SDA measures.

For IDA measures, MDist, MDist-ML and D95 revealed significant main effects of sensory augmentation conditions. Pairwise comparison indicated that velocity feedback significantly decreased MDist when compared to the condition without feedback. There were no significant difference between P and OFF or between D and OFF conditions for all IDA measures.

4.5.2 Comparison among augmented feedback strategies during normal standing

For TRAD measures, Range of COP was significantly smaller in D compared to PD condition. MVel, MVel-AP and Sway Area were significantly smaller in D condition compared to both P and PD conditions. For SDA measures, D_l was significantly smaller in D compared to P condition. For IDA measures, MDist and MDist-ML were significantly smaller in D compared to P condition and MDist was also smaller in D compared to PD condition.

For IDA measures, MDist significantly decreased in D compared to None. Decreasing trends of MDist-ML and D95 were observed in D compared to None. A similar trend can also be seen in PD compared to None. Compared P with None, trends of increasing MDist and D95 and decreasing MDist-ML can be observed.

4.5.3 Effects of sensory augmentation during perturbed standing

Table 4.2 shows that Range (in all directions), MVel (in all directions), CF (in radial and AP directions) and Sway Area were significantly dependent on which feedback conditions were selected. Pairwise comparison indicated that velocity feedback significantly decreased Range-AP compared to the condition without feedback. Both P and PD significantly increased MVel (in all directions) and CF-AP compared to the OFF condition.

For SDA measures, D_s, D_s-AP, H_s, H_s-AP revealed significant main effects of sensory augmentation conditions. Pairwise comparison indicated that position feedback significantly increased D_s, D_l-ML and H_l-ML when compared to the condition without feedback. In PD condition, H_s and H_s-AP significantly increased compared to OFF condition. For IDA measures, EV2-ML significantly increased in P compared to OFF condition.

4.5.4 Comparison among augmented feedback strategies during perturbed standing

For TRAD measures, Range, Range-AP, MVel (in all directions) and Sway Area were smaller in D compared to both P and PD conditions. Range-ML and RMS were smaller in D compared to P condition. For SDA measures, H_s and H_s-AP

Table 4.2: Measures of postural sway in AP, ML and Rad directions for perturbed quiet standing tasks

	Measure	Perturbed Condition				p-value
		OFF (a)	P (b)	D (c)	PD (d)	
Traditional Measures (TRAD)	Range (mm)	19.23(7.44)	21.40(7.95)^{c}	17.46(6.11)^{bd}	22.98(8.41)^{c}	**0.034**
	Range-AP (mm)	33.07(12.32)^{c}	35.01(11.56)^{c}	28.32(8.92)^{abd}	37.32(13.95)^{c}	**0.021**
	Range-ML (mm)	15.14(5.41)	18.96(8.54)^{c}	14.91(5.89)^{b}	18.10(7.22)	**0.02**
	RMS (mm)	7.75(2.96)	7.96(2.24)^{c}	6.77(1.99)^{b}	7.83(2.29)	0.06
	RMS-AP (mm)	7.04(2.84)	6.81(2.00)	5.89(1.86)	6.92(2.16)	0.13
	RMS-ML (mm)	3.04(1.23)	3.80(1.87)	3.16(1.33)	3.51(1.27)	0.17
	MVel (mm/s)	13.75(3.71)^{bd}	17.24(4.39)^{ac}	12.66(2.17)^{bd}	19.35(6.29)^{ac}	**0.01**
	MVel-AP (mm/s)	11.60(3.1)^{bd}	14.56(3.72)^{ac}	10.54(2.03)^{bd}	16.77(5.92)^{ac}	**0.012**
	MVel-ML (mm/s)	5.28(1.7)^{bd}	6.7 (2.16)^{ac}	5.06(1.0)^{bd}	6.77 (2.38)^{ac}	**0.037**
	CF (Hz)	.63(.14)	.72(.12)	.65(.12)	.82(.15)	**0.012**
	CF-AP (Hz)	.44(.11)^{bd}	.53(.09)^{a}	.46(.09)	.58(.17)^{a}	**0.02**
	CF-ML (Hz)	.19(.05)	.24(.09)	.22(.05)	.23(.08)	0.057
	Sway Area (mm^2)	637(354)	852(440)^{c}	543(268)^{bd}	854(437)^{c}	**0.019**
Stabilogram diffusion analysis (SDA)	D_s (mm^2/s)	33.71(19.5)^{b}	49.18(24.74)^{ac}	25.32(10.79)^{b}	66.21(56)	**0.01**
	D_s-AP (mm^2/s)	27.33(14.98)	38.98(20)^{c}	19.71(9)^{b}	56.43(52.33)	**0.022**
	D_s-ML (mm^2/s)	6.58(5.6)	10.31(7.36)^{c}	5.77(3.54)^{b}	9.87(7.59)	0.08
	D_l (mm^2/s)	2.59(3.91)	2.67(2.97)	2.01(4)	0.83(3.1)	0.45
	D_l-AP (mm^2/s)	2.5(3.77)	3.24(5.6)	1.38(2.94)	0.64(2.62)	0.51
	D_l-ML (mm^2/s)	0.03(0.52)^{b}	0.65(0.81)^{a}	0.51(1.1)	0.2(0.47)	0.06
	H_s	.78(.04)^{d}	.81(.04)^{c}	.77(.04)^{bd}	.84(.06)^{ac}	**0.01**
	H_s-AP	.78(.04)^{d}	.82(.04)^{c}	.77(.05)^{bd}	.85(.06)^{ac}	**0.019**
	H_s-ML	.77(.06)	.78(.05)	.77(.06)	.79(.05)	0.8
	H_l	.07(.1)	.11(.08)	.09(.16)	.02(.13)	0.33
	H_l-AP	.07(.14)	.13(.16)	.1(.16)	.03(.13)	0.48
	H_l-ML	.03 (.12)^{b}	.11(.08)^{a}	.07(.19)	.06(.09)	0.05
	t_c (s)	1.62(.54)	1.48(.42)	1.7(.54)	1.51(.47)	0.75
	t_c-AP (s)	1.61(.56)	1.64(.57)	1.52(.41)	1.52(.52)	0.84
	t_c-ML (s)	1.55(.27)	1.53(.37)	1.53(.28)	1.67(.41)	0.6
	j_c (mm^2)	97.84(59.30)	115.9(79.1)	72.01(38.40)^{d}	121.6(78.39)^{c}	0.084
	j_c-AP (mm^2)	75.33(55.22)	93.69(63.2)	52.29(34.1)^{d}	100.4(71.11)^{c}	0.053
	j_c-ML (mm^2)	16.66(13.26)	28.12(28.2)	16.48(13.84)	23.47(16.11)	0.059
Invariant density analysis (IDA)	P_{peak}	.08(.05)	.07(.03)	.07(.01)	.06(.03)	0.82
	P_{peak}-AP	.13(.18)	.10(.10)	.06(.02)	.10(.16)	0.35
	P_{peak}-ML	.17(.08)	.16(.11)	.12(.04)	.12(.07)	0.13
	MDist (mm)	6.31(2.46)	6.75(2.06)	6.26(2.09)	6.67(2.14)	0.63
	MDist-AP (mm)	5.28(2.29)	5.69(2.06)	5.36(2.24)	5.37(1.94)	0.88
	MDist-ML (mm)	2.57(0.96)	4.05(3.59)	2.65(1.04)	2.97(1.1)	0.22
	D95 (mm)	6.30(2.46)	6.75(2.06)	6.26(2.09)	6.66(2.14)	0.63
	D95-AP (mm)	12.72(4.54)	13.50(4.86)	12.42(5.41)	12.93(4.27)	0.62
	D95-ML (mm)	5.85 (2.04)	6.3(3.66)	6.07(2.85)	6.78(2.62)	0.27
	EV2	.995(.004)	.997(.003)	.996(.003)	.996(.002)	0.22
	EV2-AP	.995(.004)	.997(.004)	.996(.003)	.997(.002)	0.1
	EV2-ML	.989(.007)^{b}	.994(.008)^{a}	.992(.007)	.991(.008)	0.09
	Entropy	4.69(0.64)	4.87(0.61)	4.81(0.32)	4.99(0.62)	0.82
	Entropy-AP	4.7(1.09)	4.94(0.79)	5.09(0.39)	5.16(0.95)	0.64
	Entropy-ML	3.55(0.6)	3.77(0.93)	3.9(0.54)	4.13(0.92)	0.18

were smaller in D compared to both P and PD conditions. D_s and D_s-ML were smaller in D compared to P condition; j_c and j_c-AP were smaller in D compared to PD condition. For IDA measures, there were no significant difference among three augmented feedback conditions.

4.5.5 Subjective ratings

For discerning the direction (forward and backward) of the skin stretch feedback provided by the apparatus, the mean score rated from 10 subjects is 4.9 out of 7 with a standard deviation (s.d.) of 1.6. For discerning the intensity level (e.g., strong or mild) of the skin stretch feedback provided from the apparatus, the mean score rated is 5.9 out of 7 with a s.d. of 0.74. To see if the subjects were able to differentiate the effects of augmented feedback strategies, 40% of all subjects stated they were able to distinguish all three types of feedback (i.e., P, D and PD), 20% stated they were able to distinguish only between position and velocity feedback. 10% stated they were able to identify either position or velocity feedback. 20% couldn't discern which type of feedback was being used. For comparing the intuitiveness among different feedback strategies, 70% of all subjects found position feedback was intuitive to them, 20% found velocity feedback was intuitive and 30% found the combination of position and velocity feedback was intuitive. For the efficiency of the device in improving postural stability, mean scores of 5.6 ± 1.35 (out of 7) and 6.4 ± 0.7 (out of 7) were revealed for normal and perturbed standing, respectively. Subjects were also asked if they would recommend the usage of the developed system in a rehabilitation setting, mean score of 5.5 ± 0.97 (out of 7) was recorded.

4.6 Discussion

The reduced Range and RMS of sway trajectories, when the velocity-based skin stretch feedback was applied, seem to demonstrate/support the ankle stiffening strategy, which models the whole body as an inverted pendulum with an increased stiffness. This agrees with previous studies regarding the influences of secondary tasks (e.g., internal/ external focus on touching [72]; articulation task [103]) on postural control and suggests that this adaptive mechanism ensures a more active (increased frequency of sway) and more robust (reduced sway) control strategy. Though the increasing frequencies of sway were not observed in this study, present study confirms the theoretical findings in [103] that if CNS provides sufficient body sway velocity information, ankle muscle activation can be modulated in anticipation of the changes in the COM position and therefore stabilizes the body. In addition to position information, velocity information is essential in contributing to the regulation of posture balance [104]. In this study, we found that with velocity-based feedback mechanism, distance-based measures (Range and RMS) in AP or radial directions significantly enhanced due to the sensory augmentation, which may imply that subjects learned the mapping between different levels of skin stretch and body-sway movements more quickly and efficiently by the velocity-based coding scheme when compared to the position-based coding scheme.

Sketch et al. [105] investigated both position- and velocity- based haptic feedback on cursor movement control, and reported that both position-based and velocity-based controls were highly intuitively; and interestingly, some of the subjects even preferred velocity-based paradigm. Bark et al. [106] suggested that subjects seem to have better abilities in detecting changes in stimuli (skin stretch) than in the magnitude of stimulus, which may also support the finding in this study that subjects appeared to be more sensitive to body tilt rate instead of absolute tilt angle. Based on the theoretical evidences, we have practically examined the important role played by body sway velocity information in the control of quiet standing, and have found that the sensory augmentation with body tilt velocity information helps stabilize the body sway more efficiently than position-based control mechanism.

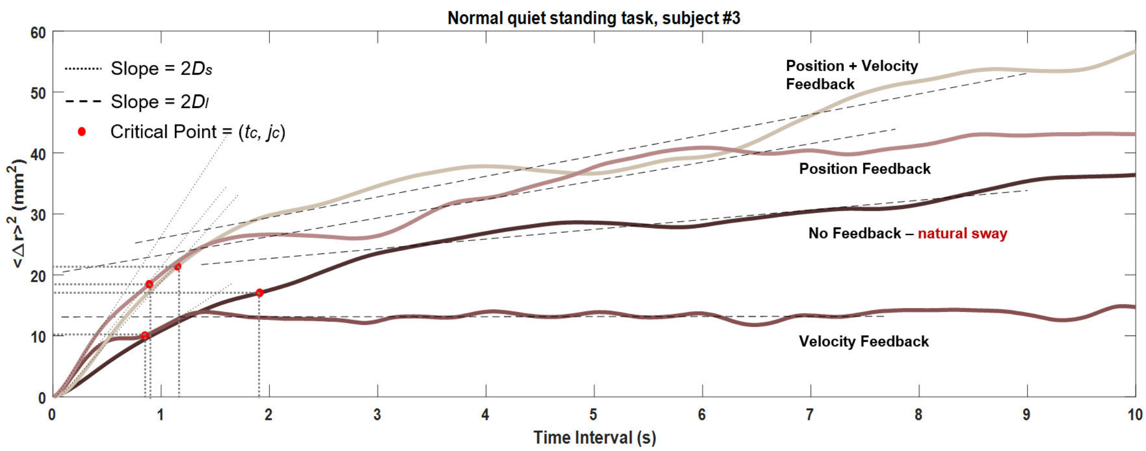

Figure 4.4: Representative stabilogram-diffusion plots of all four feedback conditions.

In addition to traditional postural sway quantification, SDA and IDA were also performed to examine the efficacy of SAD during quiet standing in terms of stochastic processes. The representative linear stabilogram diffusion plot (Fig. 4.4) shows that the slopes for both short- and long-term regions were less steep when velocity-based feedback was applied, as evidenced by smaller diffusion coefficients, compared to all other three conditions. This provides insights to the open-loop and closed-loop control mechanisms on postural stability and demonstrates a reduced random behavior of COP when velocity-based feedback was applied. COP diffuses from the equilibrium point more slowly and COP is bounded within smaller regions. A decreased D_s-ML may indicate that SAD affected more significantly on the open-loop control of posture in the ML direction. A significant reduction of j_c also implies one has a smaller feedback threshold to reach its steady-state behavior.

Another method, IDA, gives a more comprehensive view on postural sway behavior and its long-term prediction. It shows that MDist and D95 significantly decreased in the velocity-based feedback condition, suggesting that radial COP stayed closer to the centroid. In summary, examination of both SDA and IDA showed an improved postural control when the additional skin stretch based on velocity-related COM information was applied.

We state that the removed vision and perturbed vestibular and proprioceptive systems can be compensated by the additional skin stretch cues, which agreed with previous postural control systems illustrated in [15, 18]. Even though weights to visual and vestibular sensory feedbacks reduced due to the simulated sensory deficits, and thus, the corresponding velocity and acceleration information were not reliable, respectively, additional time-derivative cues from the skin stretch feedback could positively affect the sensorimotor integration to regulate the postural control system properly. In the same line of logic, it is not yet clear whether providing sensory augmentation of both body-sway position and velocity information is more beneficial than just providing body-sway velocity information. Further investigation is needed.

The sensory augmentation system in this study employs portable and wearable haptics technology to make self-rehabilitation feasible. In this study we have only performed a one-time treatment on each subject. A longitudinal study is essential to determine the feasibility of the skin stretch feedback as a long-term balance rehabilitation tool such that subjects can maintain the enhanced balance without the help of skin stretch feedback after the rehabilitation program. Some improvements for the next generation system are on-going; these include subject-specific apparatus design and control strategy. Individuals have different sizes of fingertip and their skin sensitivities can vary; it is important to determine how to make the device size-adjustable and make every user receives the haptic sensation more effectively. Elderly people or neurologically-impaired patient recruitment are required for the future work. Effect of stimulation location on the balance enhancement will also be examined.

5. SKIN STRETCH FOR WEIGHT SHIFTING*

5.1 Introduction

Physical interactions between human and machine are essential in facilitating effective physical therapy training programs. Nowadays, physical training frequently involves robotic assistive devices such as prosthetics and exoskeleton or biofeedback application. In this chapter, we present a wrist-worn skin stretch device that can deliver directional sensory cues in response to individual's weight shifting. We apply lateral skin stretch on top of the wrist to render the real-time posture information to the users. When designing such device, the mapping between skin stretch feedback and postural sway movement must be simple and natural so that the directional cues can be easily interpreted by users without increasing attentional demands. In addition, we introduce the concept of exergame combining both visual and skin stretch feedback for users to be more engaged and motivated during the physical training process. In this study, the two main objectives are

- investigating the feasibility of the developed skin stretch device in assisting dynamic weight shifting control, and
- evaluating the efficacy of such skin stretch cues in substituting for missing visual feedback

5.2 Related Work

Balance rehabilitation involving exergames has been suggested as a more sustainable home-based training approach for all age groups [39]. From a human-centric per-

*This chapter is based on the article ©2017 IEEE. Reprinted, with permission, from Yi-Tsen Pan, and Pilwon Hur, "Interactive balance rehabilitation tool with wearable skin stretch device", Proceedings of the 26th IEEE International Conference on Robot and Human Interactive Communication (RO-MAN), pp. 489 - 494, Lisbon, Portugal, August 2017.

spective, a good physical training program should not only be thorough and effective but entertaining so that users can feel motivated and are more willing to be actively involved. In conventional balance training techniques, the ability to maintain the body center of mass within the base of support while dynamically performing secondary tasks has been the common target measure. In the past few years, the game-based approach has been introduced in balance training programs. For example, the Nintendo Wii Fit balance board was used along with a desktop PC to carry out exergames for balance rehabilitation purposes [43]. A Virtual Reality (VR) system is also incorporated into balance training programs to create a more realistic and diverse environment [3]. These exergame-based interventions have demonstrated their ability to improve individual's balance performance in the framework of traditional physical training programs while offering more flexibility and greater compliance.

Positive effects of these balance training interventions are not only shown in exergames or VR trainings but also in haptic devices. Wearable haptic devices using vibrotactile instructional cues [91] [48] and skin stretch feedback [45] (as presented in Chapter 3) in response to trunk tilts have been shown to augment the impaired or unreliable sensory systems and improve standing posture. In [107], the effects of visual feedback, vibrotactile feedback, and multi-modal feedbacks on postural performance were also compared for potential home-based rehabilitation. For people with neurologic impairments such as stroke and spinal cord injury, the stimulus location is critical for the perception of cutaneous feedback. In this case, the arms, hands [45] [108], head [48] or tongue [70] seem to be more suitable than the torso [91] [107] in terms of available skin sites and wearability.

To facilitate a home-based balance training program for wider age groups and patients, we propose a novel interactive balance rehabilitation tool that combines both gaming technology and a wearable skin-stretch feedback device at the wrist. Wrist-worn devices such as a watch, or a fitness monitoring device have been widely used for tracking the user's movement. Most of these wrist-worn devices are based on sensor technologies. However, growing interests in wrist-worn "actuators" have been observed in recent studies for rehabilitation purposes. Wrist rotation guidance using vibration [109], skin stretch [110] and multiple haptic displays [111] are found to be intuitive and comfortable for achieving motor learning tasks. However, those devices are mainly for the upper limb posture guidance; there have been no studies investigating the wrist-worn device for balance training.

5.3 System Overview

The whole system consists of a wrist-worn skin stretch device operated by a DC motor, a motor driver, a microcontroller for driving the DC motor and data acquisition, a monitor displaying an interactive program, and a force plate. Fig. 5.1 shows a schematic of the system and what feedback modalities are provided to the user. Each component is described in the subsequent sections.

5.3.1 Design of the Wrist-Worn Device

The wrist-worn device comprises six major components, as shown in Fig. 5.2. All parts have been designed using Solidworks and fabricated in ABS material with a 3D printer (Replicator 2X, Makerbot, Brooklyn, NY) to develop a proof-of-concept device. A custom pinion (labeled as "A" in Fig. 5.2) and a curved rack (C) are designed to provide a one-dimensional shear force on the top of the wrist skin. A contactor with a rough surface is integrated into the curved rack (C). The design criteria of the contactor are to provide i) easily perceivable sensation to the skin, and ii) directional and intensity information of the reference inputs. Therefore, to

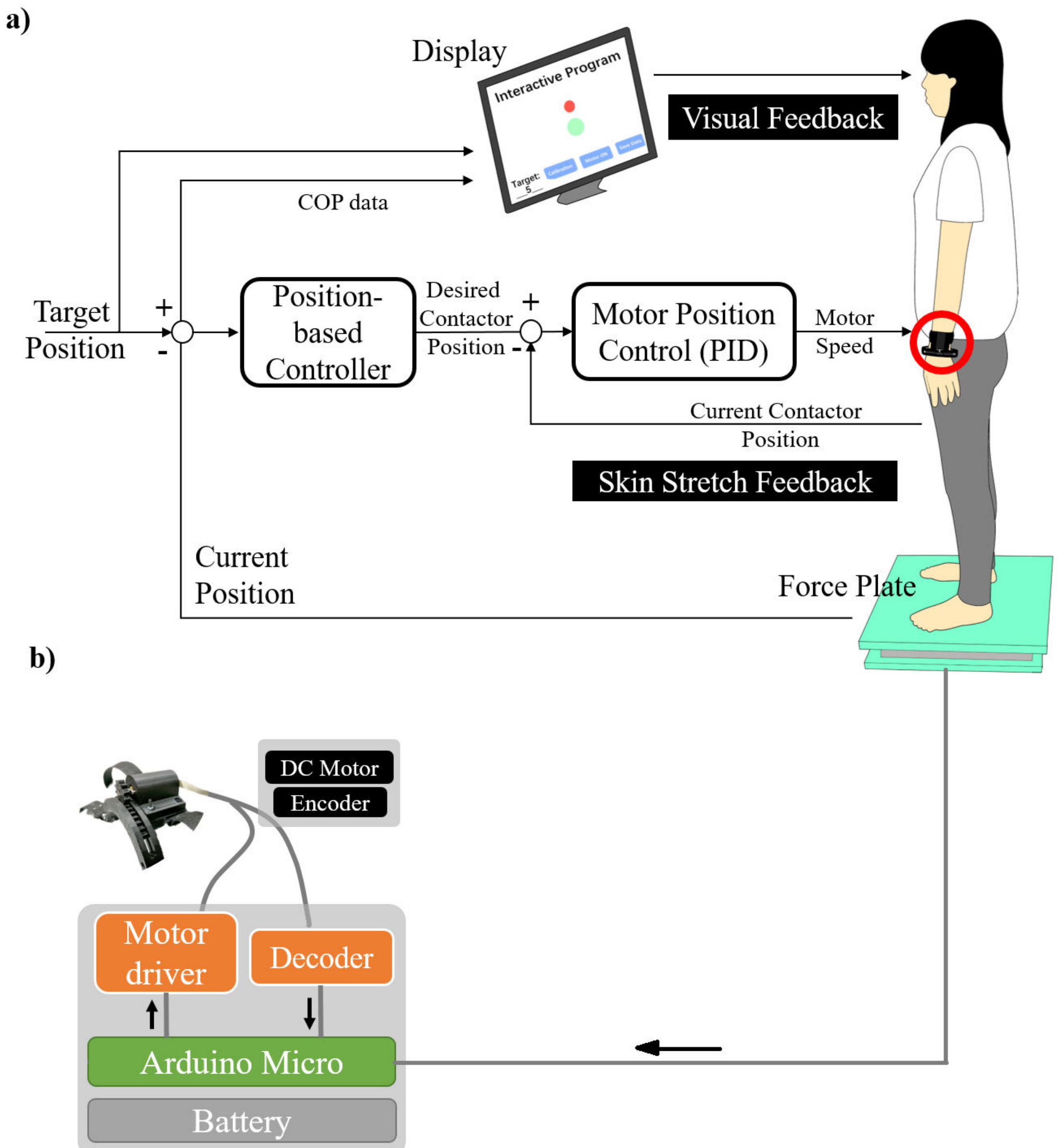

Figure 5.1: a) A schematic of the proposed system. The system consists of both visual feedback and skin stretch feedback (circled in red) of the individual's COP. A subject swaying back and forth to reach the target defined by the experimenter. For skin stretch feedback, contactor moves on the top of the wrist, providing position error cues of the current COP. The subject needs to try moving the contactor back to the wrist center point to reach the target. b) A schematic of the electrical hardware.

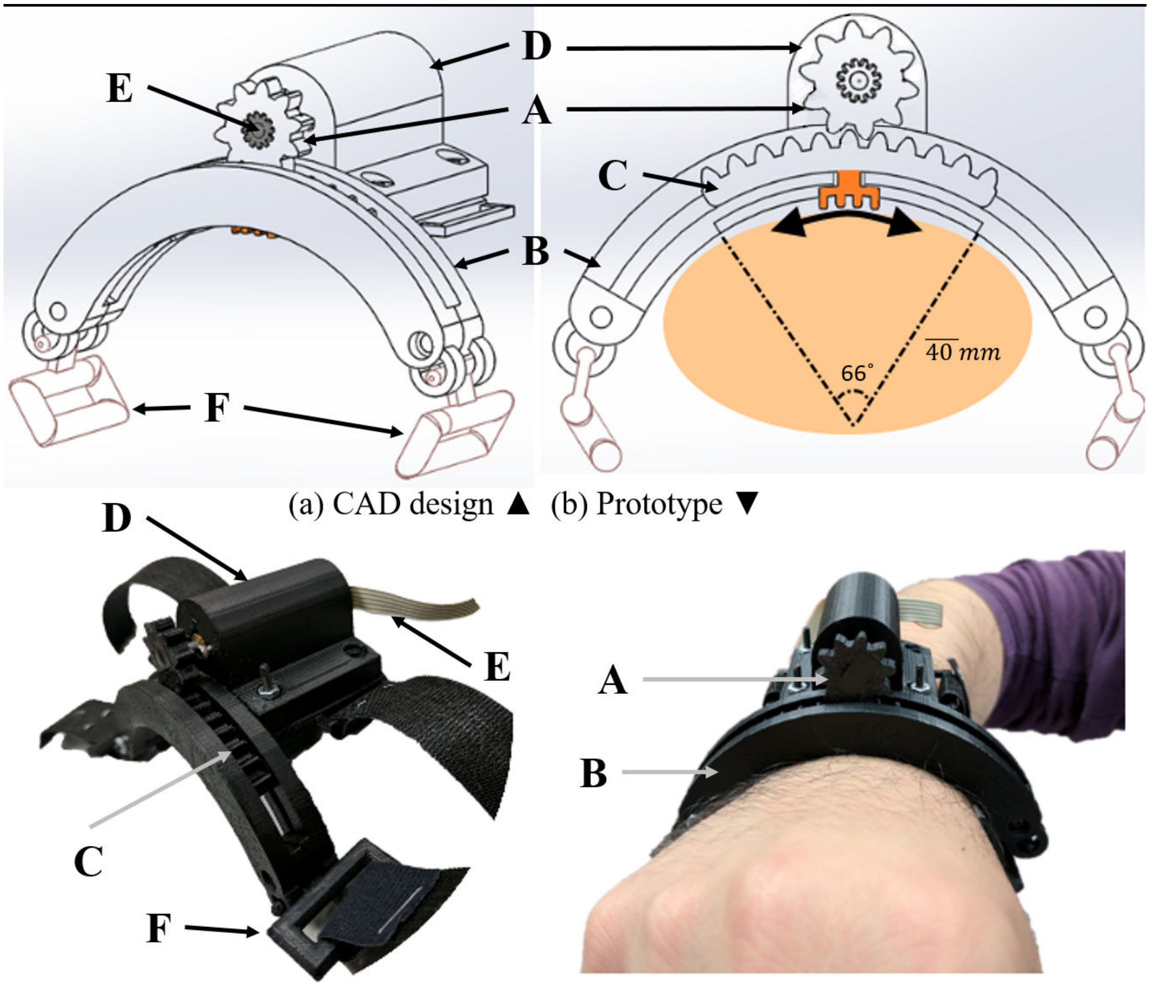

Figure 5.2: Wrist-worn skin stretch device. Skin stretch feedback is provided by the contactor connected to a curved rack (C). The rack is driven by a DC motor (E) with a custom pinion (A) attached (D: motor housing). The rack and pinion mechanism is housed inside two combined curved bands with the embedded track (B). Two movable buckles (F) are attached at each end of bands to accommodate various wrist sizes. User can wear and tighten it using two adjustable Velcro straps.

effectively convey the cutaneous feedback and to avoid the desensitization and slipping, we have designed the contactor surface to be small (8×10 mm^2), and rough (notched surface). The rack and pinion mechanism is housed inside two combined curved bands with the embedded track (B). The custom track bounds the curved movement of the contactor which defines the range of motion of the contactor. Approximately 46 mm curved displacement can be applied to the skin of the wrist. A small DC motor (1524T009SR, Faulhaber, Germany) to drive the custom pinion is mounted inside a motor housing (D). To accommodate various sizes and shapes of wrists, two movable buckles (F) are attached at the end of both bands; the device is worn and tightened using two adjustable Velcro straps to ensure that users can feel the cutaneous sensation while minimizing their discomfort levels. The weight of the entire device is approximately 75 g. More details on how to actuate the rack and pinion mechanism by the DC motor are given in Section 5.3.2.

5.3.2 Skin Stretch Feedback Actuation

A 9V DC motor that actuates the contactor is controlled by an Arduino Micro microprocessor board which is light (13 g) and small (48×18 mm^2). An h-bridge type motor driver (L298N, STMicroelectronics, Italy) was used to provide appropriate control signals to the DC motor. The unloaded maximum speed of the DC motor was about 1183 rpm (equivalent to about 1 m/s). To control the position of the motor, the angular position was measured with its embedded encoder at 9728 counts per revolution of the pinion (512 counts per revolution with 1:19 gear reduction ratio). To increase the resolution of encoder inputs for the Arduino, a 32-bit quadrature counter LFLS7366R (LSI Computer Systems, Inc., Melville, NY) was attached to the Arduino Micro through a serial port interface. The desired angular position of the motor was regulated using a PID controller. The maximum angular displacement of the custom pinion (attached on the motor) is limited to $\pm$ 150° to match the designated range of motion of the contactor. The contactor's position was always initiated at the center that corresponds to the angular position of 0°.

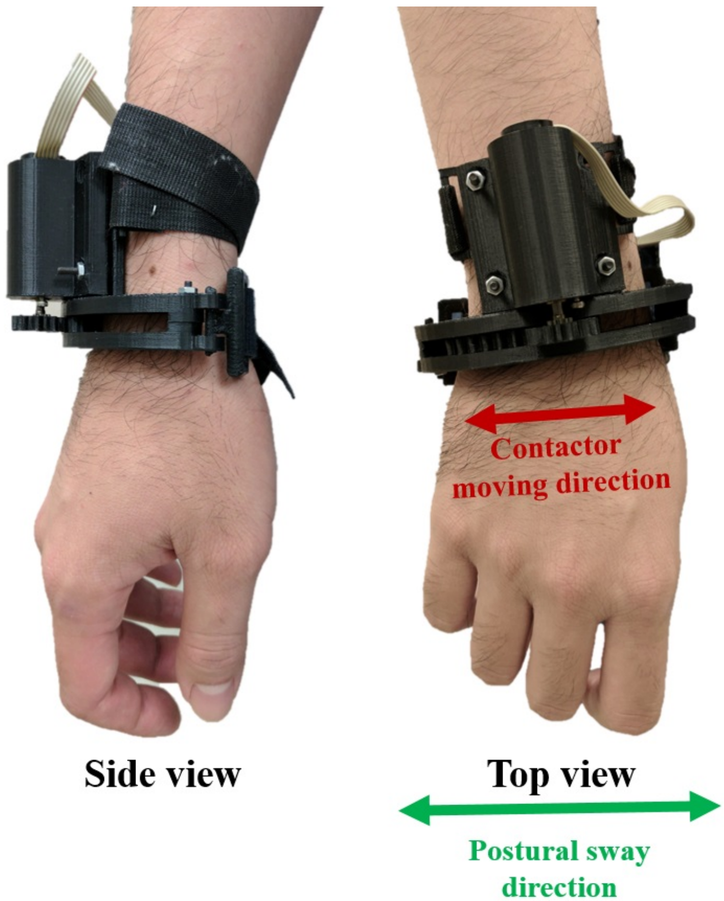

Figure 5.3: Skin stretch device worn by subject viewed from the side and the top. The contactor moves along the top of wrist surface in response to the subject's postural sway direction.

5.3.3 Tactile Coding Scheme

The amount of skin stretch rendered to the user is determined by the user's COP movement along the anterior-posterior (AP) direction. Before actuating the skin stretch device, its motion is calibrated based on each subject's COP equilibrium and the limit of stability. The limit of stability is determined by the maximum COP displacement in both forward and backward directions. The contactor location (θ_c) is therefore defined to be proportional to the user's current COP (x_{COP}) in anterior-posterior direction:

$$\theta_c = \theta_L \cdot x_{COP}/x_{FL},\ if\ x_{COP} \geq 0$$
$$\theta_c = \theta_L \cdot x_{COP}/x_{BL},\ if\ x_{COP} < 0 \tag{5.1}$$

where θ_L is the limit of the pinion angle (i.e., 150°), x_{FL} and x_{BL} are the absolute value of COP limits at front and back respectively. Since users are asked to rest their arms naturally (see Fig. 5.3), the movement of the contactor is aligned with users' COP movements as they sway back and forth. The proposed controller can be defined as position-based control, i.e. when user stands still at his/her equilibrium position, the contactor would move back to the center of the device ($\theta_c = 0°$); if s/he leans forward and reach the front limit, the contactor would move "forward" and close to the device limit at one side.

Similarly, if the COP target position is set to other than the user's equilibrium position, the contactor's initial position (i.e., the center of the device) will correspond to the target position. Therefore, by applying the same position-based control, the contactor's location is then mapped to the error between the target and current COP position within the COP range; Eq. 5.1 can be slightly modified as:

$$\theta_c = \theta_L \cdot \triangle\, x/x_{FL},\ if\ \ \triangle\, x \geq 0$$
$$\theta_c = \theta_L \cdot \triangle\, x/x_{BL},\ if\ \ \triangle\, x < 0 \tag{5.2}$$

where $\triangle\, x = x_{COP} - x_T$ and x_T is the pre-defined target position.

5.3.4 Interactive Program

We developed an interactive program using Processing, an open source software for the development of the graphic user interface (GUI). This program allows users to visually check their current COP_{AP} in an intuitive way and records their movements for each trial. Target position setup and motor actuation are also controlled by this program. Using Processing is beneficial for sending/receiving data to/from Arduino due to the built-in library for a serial communication between Arduino. The COP_{AP} data recorded from a force plate can be easily collected and displayed on a monitor. Fig. 5.4 shows a screenshot of the program. User's current COP_{AP} position is shown as a red circle, along with its absolute value recorded from the force plate. The target position is set by the experimenter and shown in the green circle in Fig. 5.4. The purpose of this GUI is to i) evaluate users' postural control performance by shifting their weight on a force plate to reach the target with visual feedback only or both visual and skin stretch feedback, and ii) provide convenient ways of data-logging and test management by clicking the custom buttons. Additionally, the calibration of the user's posture equilibrium and measurement of front/back limit are performed using the GUI.

5.4 Dynamic Standing Balance Experiment

The aims of this experiment are to i) identify the effect of skin stretch feedback on postural control when visual feedback is available, and ii) determine if subjects can still perform the same postural control task and reach the target by using only skin stretch feedback after a short learning phase. Five healthy young subjects (age $\pm$ s.d.: 25.2 ± 2.9, two females) were recruited to participate in the pilot test of the proposed wrist-worn device prototype.

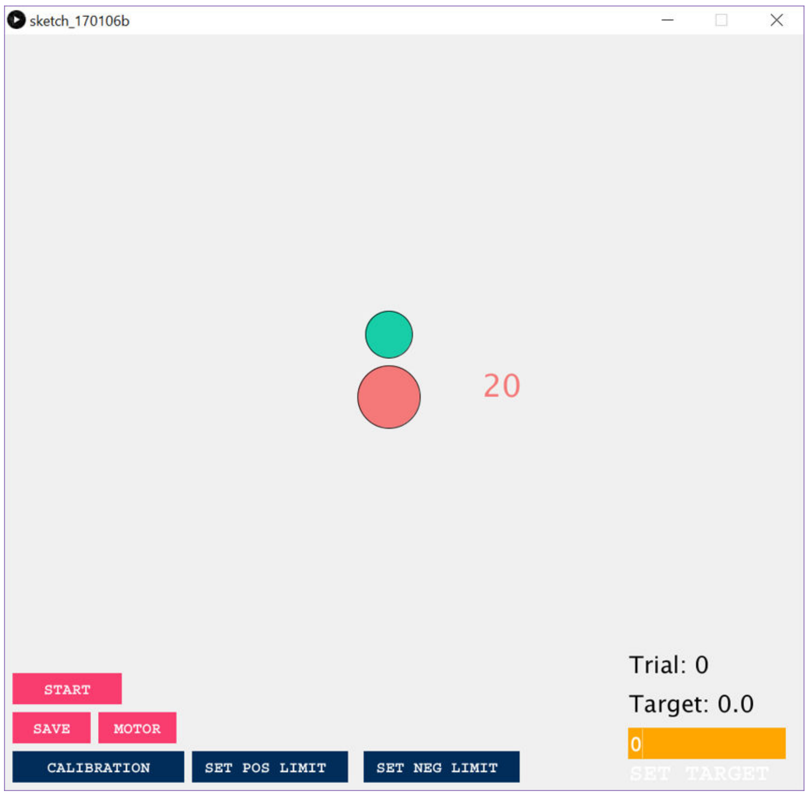

Figure 5.4: Interactive program for visual feedback. Red circle represents the subject's current position along with the text on the right. Green circle represents the target position. Target positions are entered by the experimenter in each trial. Subjects are blind to the target position indicated in the lower right corner.

5.4.1 Experimental Protocol

This pilot study is composed of three parts and conducted in the following order: visual feedback only (V), visual + skin stretch feedback (V + S), and skin stretch feedback only (S). In each part, subjects were asked to stand on a force plate in their normal stance, wear the skin stretch device on their right wrist, and let their arm hung naturally by their sides. No talking was allowed during the test. Subjects were required to perform postural control tasks by moving their body back and forth. In each part, six subtasks were performed in a randomized order (see Table 5.1). First, the experimenter instructed the subject to return to the initial position and set the target position (subjects were blind to the target position at the lower right corner of the display). When the target position was set, the experimenter double checked if the subject is in the right position, and informed the subject to start the task. Subjects were considered finishing one trial if they successfully reach the target with errors less than 5 mm (i.e., dead band) for 1 sec. When the task was completed, either the text "You have reached the target!" was displayed on the monitor or the experimenter verbally informed the subject if subject's eyes were closed. If the subject cannot reach the target within 3 mins, the trial would be considered fail and s/he would be asked to try one more trial.

Table 5.1: Experimental setting for the dynamic weight shifting tasks

	Initial position	**Target position**
1	Center	Near front limit
2	Center	Near back limit
3	Tilt forward a bit[a]	Center
4	Tilt forward a bit[a]	Near back limit
5	Tilt backward a bit[a]	Center
6	Tilt backward a bit[a]	Near front limit

a. Level of body tilt was adjusted by subjects themselves

For the last two parts that involve the skin stretch device, each subject was asked to do the calibration before activating the devices. During the calibration phase, the subject was instructed to i) stand still to calibrate for the posture equilibrium and ii) lean as far as they can in both anterior and posterior directions to calibrate the front/back limit. Each subject was given 5 to 10 mins practice session to familiarize themselves to the device and understand how the feedback relates to their body movements. After the practice session, the same procedure as in the previous paragraph was repeated.

For the last part, subjects were asked to close their eyes after they were at the right initial position and tried to complete the task based on haptic cues from the device only. A break was provided upon request and the whole experiment lasted about 30 mins.

5.4.2 Assessment of Balance and Statistical Analysis

COP_{AP} data, desired motor angular position, actual motor angular position, PWM signal and time spending for each trial were recorded and post-processed using MATLAB (R2016b, MathWorks, Natick, MA). Postural control performance for each subject was evaluated based on the movement time and the postural sway mean velocity (i.e., the ratio of total COP_{AP} excursions to movement time). Time series of COP_{AP} data and the actual motor angular position was compared, and their correlation coefficient was calculated using MATLAB.

For statistical analysis, a one-way ANOVA was performed to study the effect of skin stretch feedback and availability of sensory modality on postural control. The significance level was set to $\alpha = 0.05$ (SPSS, v21, Chicago, IL).

5.5 Experimental Results

Table 5.2 shows the mean (SD) of movement time required to complete the task and the mean velocity of trials from all five subjects under the three sensory conditions. Representative COP time series of the three sensory conditions from subject no. 2 are shown in Fig. 5.5.

Table 5.2: Dynamic balancing performance measures

Subject No.	**Sensory Modality**	**Movement time (s)** *Mean (SD)*	**Mean velocity (mm/s)** *Mean (SD)*
1	**V**	5.70 (0.80)	53.08 (10.88)
	V + S	6.25 (2.25)	50.87 (6.79)
	S	7.61 (5.16)	46.59 (16.9)
2	**V**	5.59 (1.52)	30.95 (8.36)
	V + S	4.49 (1.78)	35.02 (6.82)
	S	11.40 (4.49)	40.05 (6.27)
3	**V**	6.45 (1.09)	60.11 (16.2)
	V + S	5.10 (2.01)	68.01 (20.21)
	S	14.01 (11.51)	66.34 (19.12)
4	**V**	6.04 (2.37)	92.13 (34.81)
	V + S	5.38 (2.9)	85.48 (32.94)
	S*	13.73 (3.65)	80.29 (31.26)
5	**V**	5.93 (2.72)	68.87 (21.36)
	V + S	5.65 (1.8)	75.35 (32)
	S	10.50 (7.02)	80.94 (26.16)

$\star$one trial failed

5.5.1 Motor Skill Acquisition

All subjects could map visual and skin stretch feedback cues to their standing position and reach the desired target positions with available sensory feedback(s) (V, V + S, and S). Only one trial in S was found failed in the subject no. 4 because of a lost contact with the device that prevented the perception of haptic cues. The average time to complete the trial for all five subjects are 5.94 ± 0.34 s for V, 5.38 ± 0.65 s for V + S, and 11.45 ± 2.62 s for S, respectively. Based on Turkey HSD post-hoc test, S is significantly different from V and V + S ($p < 0.01$). The results indicated that without visual inputs, subjects needed more time to precisely move the contactor back to the center of the wrist. Fig. 5 shows that in all three trials, the subject could easily find the correct direction of the target within around 2 s, whereas in S condition (no visual feedback), more COP fluctuation was observed. The reason might be that more time was needed for locating the current contactor position and hence subjects were actively correcting their posture before they were informed the task completion. It is also known that vision dominates other senses for spatial tasks. With only tactile feedback, the training duration could also significantly affect the performance outcomes. For mean velocity of completed trials, the average for all five subjects are 61.07 ± 22.34 mm/s for V, 63.13 ± 19.77 mm/s for V + S, and 62.82 ± 18.88 mm/s for S. There are no significant differences among three sensory feedback conditions. Since the mean COP velocity may reflect the regulatory balancing activity for postural control [78], it suggested that postural stability remained similar among these sensory feedback conditions while performing weight-shifting tasks.

5.5.2 Effects of Skin Stretch Feedback

One of the goals of this research is to see if the additional skin stretch feedback can aid postural control performance while reaching the target position. From the results, even though no significant differences were found between V and V + S conditions, it could be observed in most trials that COP fluctuation seemed to decrease more in

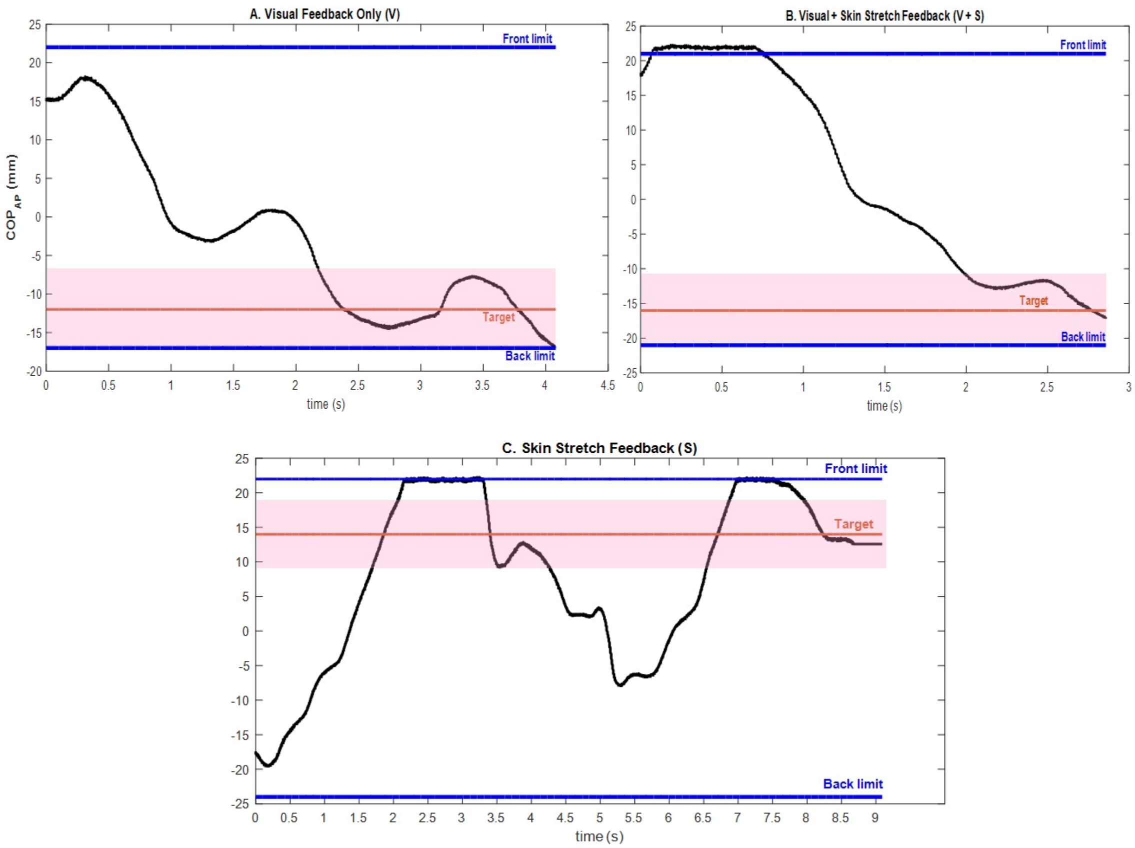

Figure 5.5: Results of COP$_{AP}$ trajectories from subject No. 2 on postural control tasks: A. Visual Feedback Only (V), B. Visual + Skin Stretch Feedback (V + S), and C. Skin Stretch Feedback (S). Front/back limits of the subject, target position of selected task are shown in blue and red lines respectively. 5 mm dead band is shaded in red. The subjects are considered to have completed the trial if they successfully reach within the dead band of the target (rectangle area) and stay within it for 1 sec.

V + S trials (for example, see Fig. 5.5 A and B). This implies that when additional skin stretch feedback was provided, it could feed the dynamical information such as relative position or rate change back to the subjects and therefore helped them stabilize their movements. However, in the self-reported questionnaire from each subject, all subjects stated they relied mainly on the visual feedback to complete the task, and it is uncertain that to what extent did the haptic feedback contribute to each task. Future work may include more complex postural control tasks to evaluate the effectiveness of additional tactile cues.

5.5.3 Skin Stretch Feedback Perception

To effectively provide skin stretch cues to users, the contact location, wearability of the device and tactile pattern have been fully considered when developing the skin stretch device. All subjects found that skin stretch cues provided by our device was easy to be perceived as the contactor moved across the surface of the wrist. The moving direction was also easily differentiated. No desensitization or uncomfortable feelings were reported throughout the whole procedure by subjects. However, one subject reported he could barely feel the contactor when it stopped moving. Therefore, it was difficult to position the contactor accurately which forced him to slightly move his body every time to move the contactor to find the current contactor position. A possible solution is to change the controller that only stops moving when the desired position is reached, instead of using position-based control only.

5.5.4 Limitations

One challenge for wrist-worn device design is to accommodate the different shapes and sizes of the human wrist. To avoid a twisted track while rotating the curved rack along with it, the prototype housing has been made using the rigid material. Using flexible material may resolve the sizing problem but also generate mechanical issues.

Further studies on device design using flexible material and different mechanisms are currently being investigated. A small number of subjects may have prevented accurate statistical results. More subjects are being recruited to have robust statistical interpretations. The force plate system we used for capturing COP data is expensive and bulky for personal use and in-home training. The potential low-cost replacement tool could be a Nintendo Wii Balance Board. Even though the lower accuracy and higher variability of COP measurements might be expected, it can still be used for the purpose of rehabilitation.

5.6 Summary

In this chapter, we have presented an interactive framework incorporating both visual and skin stretch feedback to assist users in reaching certain target positions by shifting their weights back and forth. An innovative, lightweight, and portable wrist-worn skin stretch device has been designed to provide position and directional cues for the desired position. The proposed system has been demonstrated to be easily understood that all test subjects were able to complete the tasks by the aids of the provided feedback. All subjects could complete the motor tasks by successfully interpreting the skin stretch cues at their wrists after a short-term training. This points out the potential use of wearable haptics in balance/walking rehabilitation for people with visual impairments. The wearable haptic device can also serve as an interactive tool to encourage and attract people who are in the long-term rehabilitation program.

6. SKIN STRETCH IN A WALKING AID FOR BALANCE CONTROL IN SUBJECTS WITH MULTIPLE SCLEROSIS*

6.1 Introduction

Postural imbalance and falls are commonly observed in subjects with multiple sclerosis (MS). Balance deficits are reported as one of the initial disabling symptoms of the disease [112]. Poor balance control is a significant contributor to the increased risk of falling in people with MS [28, 112, 113] and is also associated with lower engagement in activities of daily living [114]. Several studies have developed and evaluated interventions for maintaining and improving balance. It was found that the interventions related to supplementary tactile inputs are more effective at improving balance and reducing fall frequency, compared to balance rehabilitation aimed only at improving motor strategies or than nonspecific rehabilitation treatments [115].

A new functionality for a conventional walker that monitors real-time balance performance and provides this information to the user as a means of improving the postural stability is introduced in this chapter. The reasons for choosing a walker over other walking aids are the ease of use and its consistent orientation frame compared to canes or crutches. The sensory feedback for posture is augmented via a skin stretch device embedded into the handgrip of a walker. Studies have shown that skin stretch feedback about the applied forces or direction of postural sway at fingertip can be useful for balance control [6]. Significant physical stabilization is also observed with touch contact of a cane at low force levels [116]. It has been suggested that touch

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contact on those mobility aids could be used to improve balance performance at a sensory level [95, 117]. However, the effects of light touch on gripping a handle while using those mobility aids are not clear; Therefore, It was hypothesized that applying artificial skin stretch feedback could achieve similar effects to that of light touch, while persons with MS can still be provided physical supports with a mobility aid.

6.2 Related Work

6.2.1 Sensory Augmentation in Smart Mobility Aids

A crucial aspect in the development of smart mobility devices is to improve the safety for the users. Sensors are attached to monitor the surrounding environment or the states of the user. This information can then be processed for device actuation (e.g., brakes and turn) or be provided to users via available sensory systems (e.g., visual, auditory and somatosensory systems). In this section we focus on the sensory augmentation-based approach. For example, Hashimoto et al. [118] equipped a hand force feedback system on a prototype walker that allows users to perceive obstacles from the repulsive force generated as feedback on the joystick. Wang et al. [119] and Pyun et al. [120] attached an obstacle detection system on a traditional white cane to alert users via vibrotactile cues on their hands. The additional sensory cues rendered on the hand were found intuitive and easy for navigating in environments.

A fall may still occur due to the inadequate posture or impaired sense of balance of the users. Therefore, the device should continuously monitor the user's state. Martins et al. [121] present a safety feature that prevents falling in the anterior-posterior (AP) direction by detecting the distance between the user and walker and the handgrip forces. The walker will stop immediately once it detects possible falls. This feature provides some mechanical support for fall prevention but it lacks the sensorial assistance for users to learn and actively correct their posture. Uses' gait performance or health can also be recorded over time, however, these information are generally accessed by professionals only [122] [123]. Our work seeks to explore a new functionality for a conventional walker that monitors users' real-time posture and feed this information back to the users as a means of enhancing their sense of balance. Haptic feedback is chosen to convey this information as other senses are easily occupied by the environment. Furthermore, the high density of mechanoreceptors in the skin allows us to easily interpret different kinds of physical stimuli.

6.2.2 Handhold Haptic Devices

Haptic devices should be easy and comfortable to use while delivering intuitive cues. A large amount of work has been devoted to the development of handheld haptic devices because of the dexterity and rich sensory receptors of the hand. Pasquero et al. [124] present a handheld information device generating lateral skin deformation on the thumb. Spiers et al. [125] designed a compact shape-changing device that can render haptic feedback to users' entire hand. Haptic controllers for gaming are also popular in the consumer market and various devices have been designed to provide haptic feedback at the fingertips [126] [127] [128] or in the palm [129]. Most of these devices are portable and were designed from scratch which may limit the applicability of such devices in other grounded devices. Ploch et al. [130] present the design of a steering wheel with embedded haptic display. With the display directly embedded into the steering wheel, users can keep the natural driving behavior as well as perceiving feedback in the hand. Our design adopted the same concept for keeping the original structure of the walker and further explored two potential skin sites for haptic perception as gripping the handles.

6.2.3 Directional Perception of Haptic Cues

Vibrotactile feedback is commonly used for rendering the spatial information by placing multiple motors on a specific body region. For example, numbers of vibratory motors can be placed on a waist belt to specify for the same numbers of waypoint directions [131]. Different type of directional cues such as (counter-) clockwise can also be rendered by creating vibrotactile patterns using a 3 $\times$ 3 array of vibratory motors [132]. This kind of feedback is easily perceived and implemented, however, it requires more motors as the degree-of-freedom (DOF) increases which can be impractical when the skin area is small (e.g., fingertip).

To address the issues raising from vibrotactile feedback, many researchers have explored alternative haptic feedback by deforming the skin for directional perception. For example, Drewing et al. [133] present a novel multi-pin display that exerts lateral displacement on the finger pad for eight standard directions. Gleeson et al. [134] also demonstrate the efficacy of lateral skin stretch in communicating four cordial directions at the finger pad. Other studies investigated the haptic cues while touching a ground device for orientation and postural control [135] [8] [116]. Results from these studies show that sway-induced shear forces at the fingertip can be processed as an additional directional cues and aid in reducing the postural sway. Pan et al. [45] (as discussed in Chapter 3) expanded their work by providing similar frictional sensation on the finger pad with an ungrounded device. In this study, we will apply the similar directional skin stretch presented in Chapter 3 to the fingertip and palm for investigating the perception of direction and its effectiveness for balance control.

Figure 6.1: A standard front-wheel walker with the sensory augmentation system that includes: (i) a skin stretch feedback device embedded into the right handgrip and (ii) a control unit together with the power source packed at the lower part of the walker.

6.3 Initial Design of the Sensory Augmentation Walker

6.3.1 Skin Stretch Feedback

To provide intuitive and realistic cues associated with the interpretation of directions, we have chosen to employ cutaneous feedback because the sense of touch plays an important role for humans to interact with each other and their environments. It is also broadly accepted among all populations. For example, physical therapists tap on the shoulder of stroke patients to inform them the correct side for weight shifting during walking. Individuals with impaired vision use touch sensation as a sensory substitution, e.g., braille. Our skin, the largest organ in the human body, contains a variety of sensory receptors that allows human to perceive different kinds of physical stimuli. There are four different types of mechanoreceptors characterized by adaptation speed to mechanical stimuli: fast-adapting (FA) I & II and slow-adapting (SA) I & II. The density of FA I and SA I units in the skin is highly correlated with the capacity for spatial discrimination [136]. In the hand, the density of type I units at the fingertip is about five times higher than at the palm. The average two-point threshold for fingertip and palm are about 1.6 mm and 8 mm respectively [136]. This implies a minimum skin contact area at both locations.

Fig. 6.2 (A) shows the two skin contact regions while holding a handgrip which are the fingertip of the middle finger and the center of the palm (displayed as red dots in Fig. 6.2 A). To render the one degree-of-freedom (DOF) directional cues at these two locations, we have chosen to apply skin stretch feedback in which body orientation in the sagittal plane can be mapped directly from the skin stretch direction (Fig. 6.2 B). That is, when user senses a stretching of the skin from region's back to front, it represents a forward directional cue, and vice versa. An initial prototype was

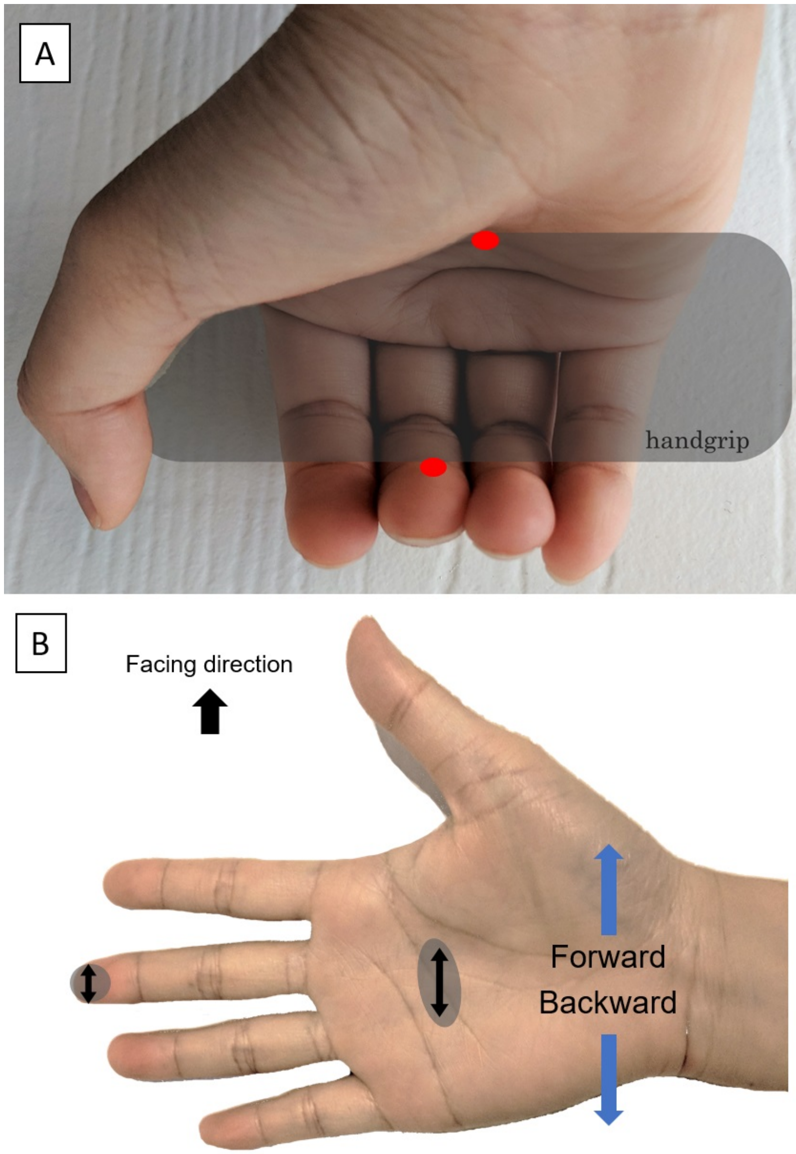

Figure 6.2: (A) The two primary skin contact areas (red dots) while holding a handgrip. (B) Mapping of the 1-DOF skin stretch direction at fingertip/ palm (gray shades) with the body orientation.

developed as a proof-of-concept. Section 6.3.2 details the device design.

6.3.2 Device Design

A conventional front-wheel walker made in aluminum was re-engineered as a fundamental structure to develop the initial proof-of-concept prototype (Fig. 6.1). The

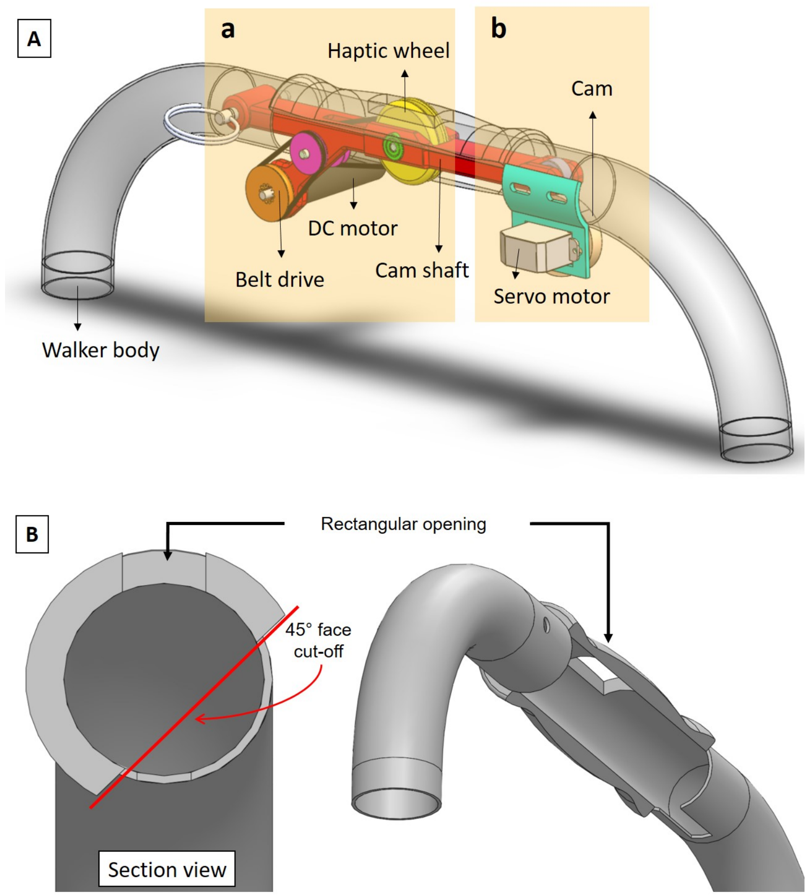

Figure 6.3: (A) CAD design of the skin stretch device embedded into the right-hand side handgrip of a walker. The mechanism consists of two parts for producing lateral skin stretch using a DC motor (a) and normal skin displacement using a micro servo (b). (B) Section view and bottom front view of handgrip tube. Two sites including a rectangular opening on the top and a 45° face cut-off along the tube were fabricated for the installation of the skin stretch device.

design comprises two parts for conveying (i) lateral skin stretch and (ii) normal skin displacement, labeled as (a) and (b) in Fig. 6.3 (A) respectively. All parts, except for the mechanical components (e.g., bearings and fasteners) that were purchased, were designed using DS SolidWorks and printed in PLA material with a 3D printer (Dreamer, Flashforge, USA, City of Industry, CA).

First, for the installation of the main part that provides lateral skin stretch, several geometrical modifications were made on the right handgrip. They include a 45° face cut-off along the tube, a rectangular opening on the top, a slot on the bottom and a M5 through-hole on the sides (Fig. 6.3 B). Skin stretch feedback is conveyed through a haptic wheel (diameter: 30.5 mm, width: 8 mm) and belt-drive systems operated by a small DC motor with a gear ratio 19:1 (1524T009SR, Faulhaber, Germany). The DC motor, pulleys, and the haptic wheel are attached to a camshaft. The camshaft is fixed on the walker body using a quick-release pin. The speed ratio of the DC motor and the haptic wheel was set to be 1:1. The pulleys and the routing of the round belts within the unit were well-positioned to ensure the handle can be gripped easily by the users. A handgrip cover printed in flexible material was made to improve comfort while gripping. For motion control of the DC motor, a Teensy 3.6 microcontroller, an h-bridge type motor driver (L298N, STMicroelectronics, Italy) and a 9V battery are used and packed in a small box on the lower part of the walker.

The normal skin displacement is controlled via a custom cam rotated by a servo motor (Futaba S3114 Micro High Torque Servo) connected to the same control unit (Teensy 3.6) and fixed on the walker body. By rotating the cam, the shaft can move vertically allowing a normal skin displacement of 5 mm at the palm. The normal skin displacement is by default set to 2.5 mm. The entire mechanism weights approximately 40 g. A close-up view of the skin stretch part and the physical prototype can be seen in Fig 6.4.

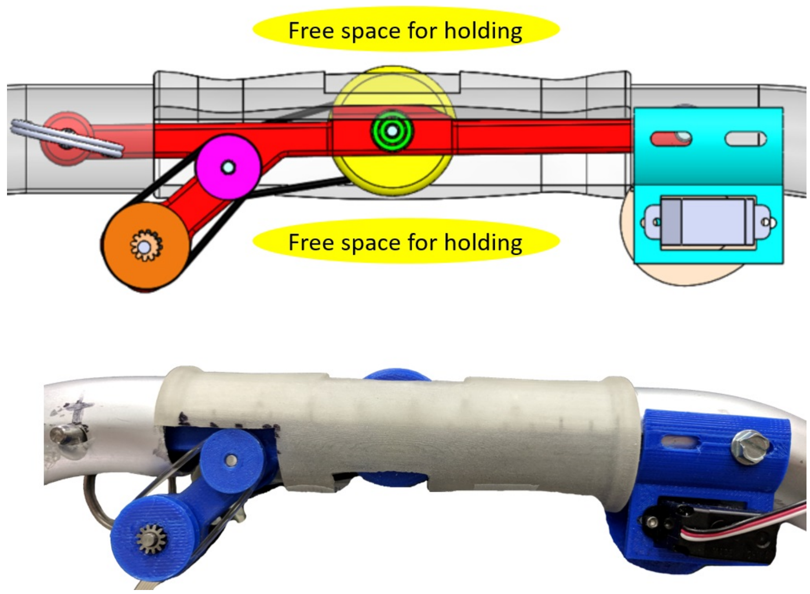

Figure 6.4: CAD design (top) and the prototype (bottom) of the skin stretch device from the left-side view.

6.4 User Perception Study

The purpose of this user study is to evaluate the functionality of our skin stretch device on rendering directional cues. Two candidate locations, i.e., palm and fingertip, were tested to assess and compare the perception of direction in the sagittal plane. The final design of the skin stretch device will be based on the preliminary results gathered from this experiment.

6.4.1 Experiment Setup

A graphical user interface (GUI) was created to control the rotation direction of the haptic wheel (referred to as the "tactor") and to record the user data. The device was connected to a PC via a USB port. Motor commands were operated using Arduino IDE and the motor driver was used to provide appropriate PWM signals to the DC motor. The DC motor rotates clockwise or counter-clockwise in order to deliver skin stretch cues in either the forward or backward direction. The output cues are combinations of multiple speeds and durations of stimulus. Four speeds: (i) 55 mm/s (ii) 85 mm/s (iii) 130 mm/s (iv) 205 mm/s and three durations: (i) 0.1 s (ii) 0.25 s (iii) 0.5 s were chosen to determine a baseline for our device. These parameter ranges were selected based on the pilot tests conducted by the researchers. The effects of speeds were examined to investigate the minimum threshold on perceiving the direction and to determine whether users are more sensitive at the slow, medium or fast speed. The effects of duration were studied to determine how quickly the users can react to a directional cue and to detect potential habituation problems from a long-duration cue. A small hand-held portable controller with two buttons ("F" and "B") was made for the users to toggle between forward or backward directions (Fig. C). The controller was held by the user's left hand while the device delivered cues at their right hand. Two front wheels on the walker were locked to provide a static standing environment.

6.4.2 Experimental Protocol

A total of eight subjects were recruited (age ± s.d.: 26.6 ± 4.57, 2 females). The experiment consisted of three sessions: (i) practice (ii) perceptual study at palm and (iii) perceptual study at fingertip. In the first session, participants were instructed on the functionality of the skin stretch feedback device and on how to perceive the cues at the two skin sites. Participants were also given time to familiarize themselves with the hand-held controller. The experimenter provided several practice trials and checked if the participants could respond to the cueing sensation and were comfortable with wearing the device. A maximum period of ten minutes was given to prevent any learning effects on one or both locations.

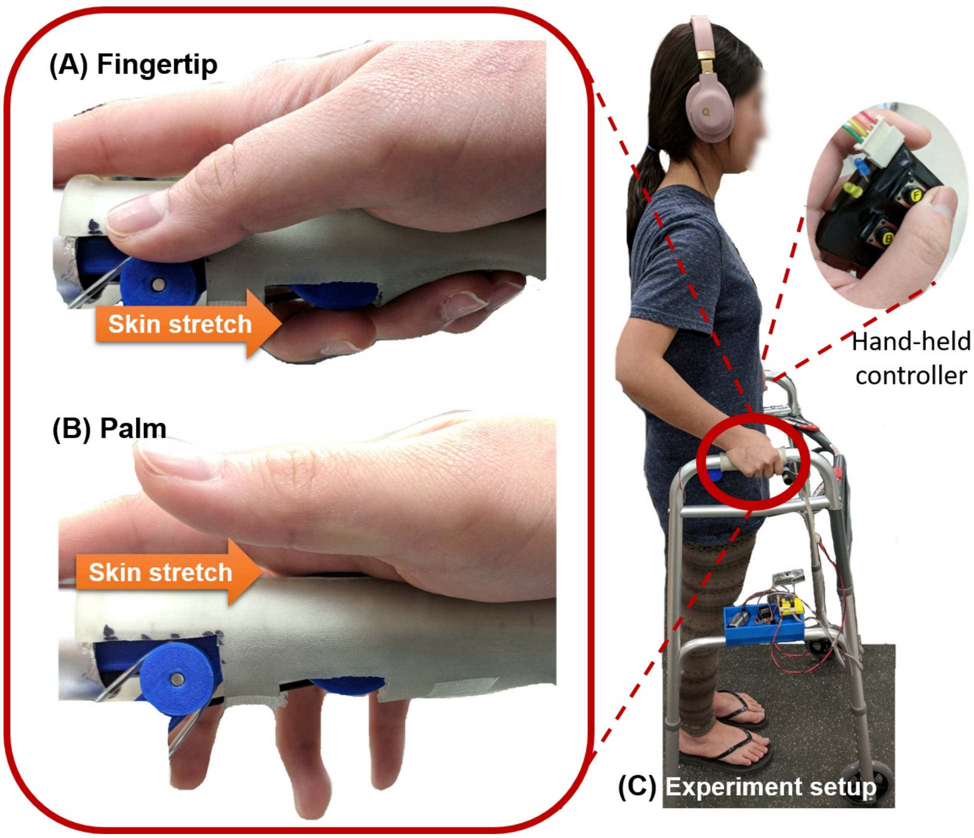

Figure 6.5: Experiment setup. During the experiment, skin stretches are applied at different locations in the fingertip session (A) and palm session (B). (C) The participant stands quietly while holding the handgrip using her right hand and a controller using the left hand to toggle between two directions.

In the main sessions, i.e., (ii) and (iii), participants were asked to put on headphones while holding the handgrip with their right hand in an upright stance. Headphones playing white noise were used to minimize distractions from the sound of the DC motor. Participants were also asked not to look down at the device and focus on the cue sensation at their hand. A series of forward and backward directional cues was given in a randomized combination of speeds and durations; for example, a 0.5 s cue was given in the forward direction with speed of 130 mm/s. The trials included 12 combinations of speed and duration with 5 repetitions in both the forward and backward directions. In total, 120 cues were tested in a randomized order in each session. Upon request, the participants were allowed to retry up to one additional trial on the same cue. If the skin stretch actuation was blocked due to normal hand gripping strength, the participant was asked to adjust the hand position and to release their hands slightly. For perceiving skin stretch at palm, the participants were instructed to touch lightly on the tactor with the palm while avoiding the fingertip contact at the opposite site of the tactor (Fig. 6.5 B). Similarly, in the fingertip session, subjects were instructed to touch the tactor lightly with one fingertip (e.g., middle finger) while avoiding skin contact between palm and the tactor (Fig. 6.5 A). The participants used the portable controller to select either forward or backward direction by pressing the "F" or "B" button respectively after receiving the cues operated by the experimenter. The entire procedure, including break, took around one hour to complete.

6.4.3 Post-Experiment Questionnaire

After completing the previous sessions, a questionnaire was provided to the participants for them to rate the overall performance using the semantic differential scales

(1 - 7 rating scales). The level of comfort (1 = very uncomfortable, 7 = very comfortable), intuitiveness (1 = very difficult to understand, 7 = very easy to understand), preferred speed (low, medium, and high) and duration (short, medium, and long) at both palm and fingertip were surveyed. They were also asked to choose a preferable location other than the palm and fingertip, and provide comments on the design of the device, haptic feedback and experimental protocol.

6.4.4 Results

6.4.4.1 Perception of Direction

Fig. 6.6 shows the mean percentage of perceiving the correct direction across all eight subjects for each of the combinations. Perception of direction at the palm yields an accuracy rate in the range from 65 - 80%. Six of the twelve conditions have accuracy rate over 75% (stippled boxes). Perception of direction at the fingertip yields a range of accuracy rate from 91% to 99%. Four out of nine conditions obtained no significant deviation from the 100% maximal result by Student's t test ($p > 0.05$).

Palm

Speed (mm/s)	0.1	0.25	0.5
205	66	76	75
130	66	68	75
85	74	65	78
55	66	80	77
	Duration (s)		

Finger

Speed (mm/s)	0.1	0.25	0.5
205	93	98	98
130	91	95	93
85	96	94	99
55	94	99	95
	Duration (s)		

Figure 6.6: Mean percentage of perceiving the correct direction at palm and fingertip under twelve speed-duration combinations. The shaded cells correspond to accuracy, with darker color representing higher accuracy.

6.4.4.2 Effects of the Speed and Duration on Discerning Direction

One-way repeated-measure analysis of variance (ANOVA) was performed to evaluate if the perception of direction changes significantly among different speeds or durations. Fig. 6.7 shows the mean accuracy and its 95% confidence interval for speeds and durations of stimuli for palm and fingertip. The results show that no significant differences were observed among different speeds, for both locations. Similarly, no significant differences were observed among different durations for both locations. While no statistical results were found, perception of 130 mm/s (medium-high) cues yield the lowest accuracy rates for both palm and fingertip. The reason for this trend is unclear since this speed profile is characterized as medium to high in this study. It will be worthwhile to investigate in a future study whether directional cues are better operated at either low or high speeds. Similar trends without statistical significance are also found in perception of 0.1 s (short) cues. A possible explanation is that the response time for each user differs, hence the pulsing duration less than 0.1 might be more difficult to be processed in time, which identify a lower bound of duration for delivering such directional cues. These results also imply that the ranges of speeds and durations chosen in the experiment can be used in our device with no significant difference in perception of direction. Further experimentation is needed to investigate whether these two factors can be used for rendering skin stretch cues of different magnitudes.

6.4.4.3 Subjective Perception

Qualitative analysis was performed using the post-experiment questionnaire. All eight participants completed the survey and commented on the device performance. The levels of comfort for the palm and the fingertip were rated both at an average score of 5.3 (out of 7). One subject suggested to design a better enclosure for the

Palm		
Speed (mm/s)	**205**	0.73 ± 0.043
	130	0.70 ± 0.031
	85	0.72 ± 0.043
	55	0.74 ± 0.046

Fingertip		
Speed (mm/s)	**205**	0.95 ± 0.003
	130	0.93 ± 0.003
	85	0.96 ± 0.003
	55	0.96 ± 0.002

Palm		
Duration (s)	**500**	0.75 ± 0.045
	250	0.71 ± 0.032
	100	0.68 ± 0.047

Fingertip		
Duration (s)	**500**	0.96 ± 0.001
	250	0.96 ± 0.001
	100	0.94 ± 0.007

Figure 6.7: Accuracy rates for discerning the correct direction at different speeds (top row) and different durations (bottom row). 95% confidence intervals are provided.

device. Since the current prototype has an open structure, extra cognitive load may be required for avoiding skin contact on other areas of the hand. This can be improved in a future study. For the level of intuitiveness, the palm was rated easier in mapping the forward and backward orientations when compared to the fingertip (average scores of 5.3 for the palm and 5 for the fingertip). This is because of the opposite skin stretch direction with respect to the spatial orientation when touching the bottom side of the handgrip. Some subjects interpreted the direction cues based on the rotational motion of the tactor while others based on the direction of skin stretch at the fingertip. Three out of eight participants were confused about the right direction even though they could precisely perceive the direction change. Two types of interpretation were observed: (i) skin stretch direction and (ii) rotating direction of the wheel. In our default setting, users were asked to interpret the directionality of fingertip feedback using the skin stretch direction. One of the subjects stated that it was not natural for the subject to respond to such a strategy. An adequate learning time is required for correctly interpreting the directionality of the rendered cues. Some users suggested that the cueing strategy should be consistent among users while some other stated that strategies could be adapted to each user as long as the instruction was clear and enough practice was provided. The latter statement was supported by the quantitative evaluation showing high accuracy rates (approximately 95+%) for perceiving directional cues at the fingertip among all users.

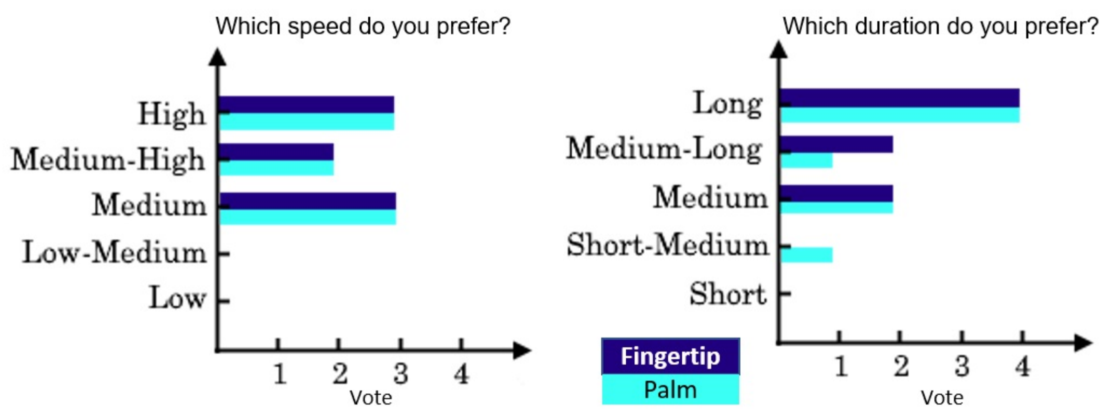

Figure 6.8: Votes of preferable speed and duration from all test subjects (n=8).

Comparing the preferable skin sites on which the feedback is applied, the fingertip is favored as six out of the eight participants chose this location while the remaining two indicated no preference, agreeing with the experimental results. For the speed and duration used to render cues (Fig. 6.8), seven out of eight participants chose medium to high speed paired with a medium to high duration for both locations. Only one participant chose a short-medium duration and mentioned a potential discomfort when perceiving strong cues at palm. All subjects stated that they can identify a set of three different durations (i.e., short, medium, long) whereas the varying speeds were not as distinguishable as durations. All subjects can identify two (low and high) out of four speed profiles used in the experiments but can hardly specify all of the speeds. This implies that stimuli of varying durations may be more suitable for representing cues of different magnitudes.

In summary, the participants were positive about the concept and believed that this device may be helpful for people needing walking aids. However, further improvement of the hardware is needed. For the elderly, it is important to provide a long and strong cue.

6.4.5 Summary of the Psychophysical Study

In the first half section of this chapter an initial proof-of-concept prototype that can provide skin stretch feedback while holding the handgrip of a walker is presented. Perceptual studies about how well users can discern the directions at two skin sites are assessed and compared. It is shown that the fingertip is an ideal location for perceiving the 1-DOF directional cues (forward and backward) supported by both quantitative and qualitative results. The accuracy rates for perceiving the correct direction at the fingertip achieved 95+% for all eight subjects whereas it fell down to around 70% for palm. No significant differences were found among stimuli speeds and among the stimuli durations with respect to perceiving the correct direction at the two hand locations. When discerning the direction, a long and strong stimulus is preferred by the subjects. To sum up, a new functionality for a walker that can provide directional cues via skin stretch feedback is introduced. Such directional cues can be used for augmenting the posture information and improving the postural stability at the sensory level.

Further details about the full closed-loop system that detects user's posture and provides feedback on balance will be introduced in the later half section of this chapter. Experimentation evaluating the efficacy of skin stretch feedback in improving the sense of balance among walker users will be presented in Section 6.6. The fingertip will be the only skin site for rendering the directional cues. An enclosure that covers the whole skin stretch mechanism will be fabricated to improve user comfort.

6.5 Full Closed-loop System of the Sensory Augmentation Walker

6.5.1 System Overview

A full closed-loop system of the sensory augmentation walker contains sensor, actuator and control subsystem, as shown in Fig. 6.9. Skin stretch feedback is provided by rotating a contactor at one's fingertip pad while gripping the handle of the walker. The rotating motion of the contactor actuated by a DC motor induces one DOF directional cues in the sagittal plane. Detailed mapping strategy and device design have been introduced in Section 6.3.2. Based on the preliminary user study, the palm side for perceiving the skin stretch feedback is removed by a rubber handle cover to prevent mixed tactile inputs. In order to examine the effect of fingertip skin stretch embedded in a walker on standing balance control, an IMU is employed to measure body tilt of a standing person. Measured data is processed by a Teensy 3.6 microcontroller (ARM Cortex-M4 at 180 Mhz) and calculate the desired motor output for skin stretch feedback. Detailed controller design for calculating desired motor output is presented in the next section.

6.5.2 Controller Design

6.5.2.1 Instantaneous Capture Point (ICP) in Standing Balance

The motor output was determined by the current body tilt position with respect to the desired extrapolated center of mass (XCOM) position. The concept of XCOM

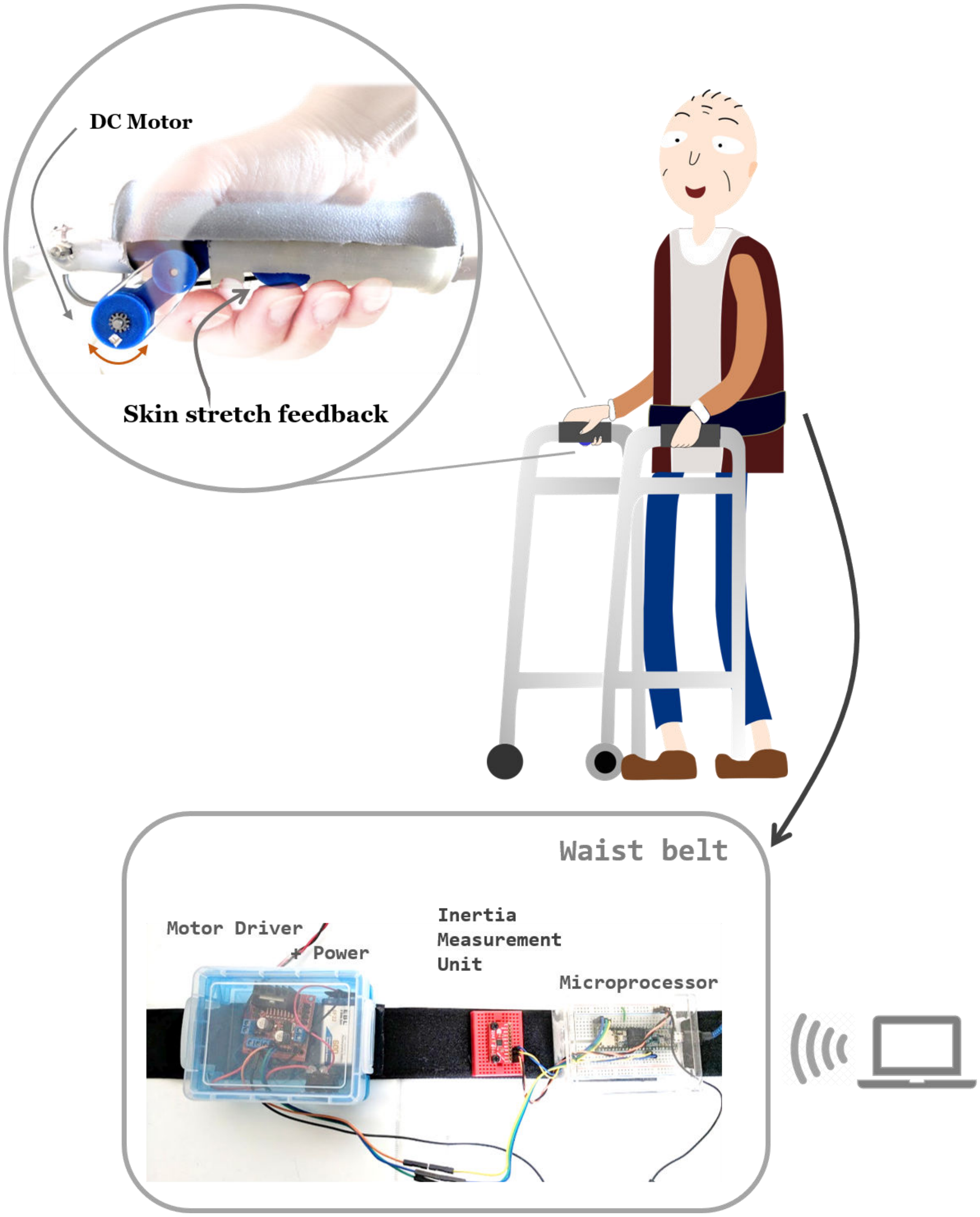

Figure 6.9: System overview of the sensory augmentation walker. The entire system consists of a waist belt that encloses the IMU, microprocessor, motor driver and power source and a conventional four-wheeled walker that integrates a skin stretch feedback device at the grip. Data is transmitted through a USB cable for real-time data logging and data collection.

is adopted for evaluating postural stability in this study. The general definition for maintaining balance is to keep the body COM within the BOS [1]. However, some studies argued that this condition is not sufficient in dynamical situation [137, 138], which velocity of COM should also be accounted for. Hof [139] stated that the position of the COM plus its velocity times a factor should be within the BOS for dynamical situation. The XCOM can be represented as $COM + \dot{COM}/\omega_0$ where ω_0 equals to $\sqrt{l/g}$, in which g is the gravitational constant and l is the COM height. By placing the COP in the XCOM, the body can come to an upright stop. The same concept was introduced by Pratt et al. [140] for studying the dynamical stability of bipedal robot. XCOM is termed "Instantaneous Capture Point" (ICP). A capture point is defined as a point where the humanoid robot can step to on the ground in order to bring itself to a complete stop. ICP is the capture point at current time step x_{ICP}:

$$x_{ICP} = x_c + \dot{x}_c/\omega_0 \tag{6.1}$$

where x_c is the COM position in AP direction, and ω_0 equals to $\sqrt{l/g}$ as described before. It should be noted that XCOM and ICP are the same variable but used in different fields of research.

The vast majority of the existing studies using augmented feedback of body sway only considered COM position as the control objective [], while the quantity including velocity $COM + \dot{COM}/\omega_0$ should be considered for formulating a stability condition, since the following holds: "for static stability, the ICP should be within the BOS" [139].

Under this framework, the controller as follows was implemented to generate the desired motor output:

$$\omega_c = k_p(x_{ICP} - x_{dICP}) \tag{6.2}$$

where x_{ICP} and x_{dICP} are current ICP and desired ICP, respectively.

6.5.3 Apparatus

There were only two minor modification of this design compared to the one introduced in the Section 6.3.2. First, the upper side of handle inducing the skin stretch feedback at palm is covered. From the preliminary study, fingertip is found to be an more ideal location for perceiving and discerning the directional cues produced by the skin stretch device, compared to palm. Second, the cam mechanism for the skin surface deformation was not applied in this study.

6.5.4 GUI Design

A graphical user interface (GUI) program was developed for real-time data logging and data collection during the experiments using Processing 3.4 (Fig. 6.10). The program allows the experimentor to visualize the IMU data that estimated postures of the subjects in real-time. The configuration of the walker system and controller types can be selected via the program. The serial communication between the GUI and Tennsy 3.6 is through USB.

6.6 Quiet Standing Balance Experiment

6.6.1 Subjects

Three MS patient were included in the study (mean age ±s.d.: 65.33 ± 2.51, 3 females; mean EDSS score: 6.5 ± 0.5, range 6 - 7). Patients were excluded from the

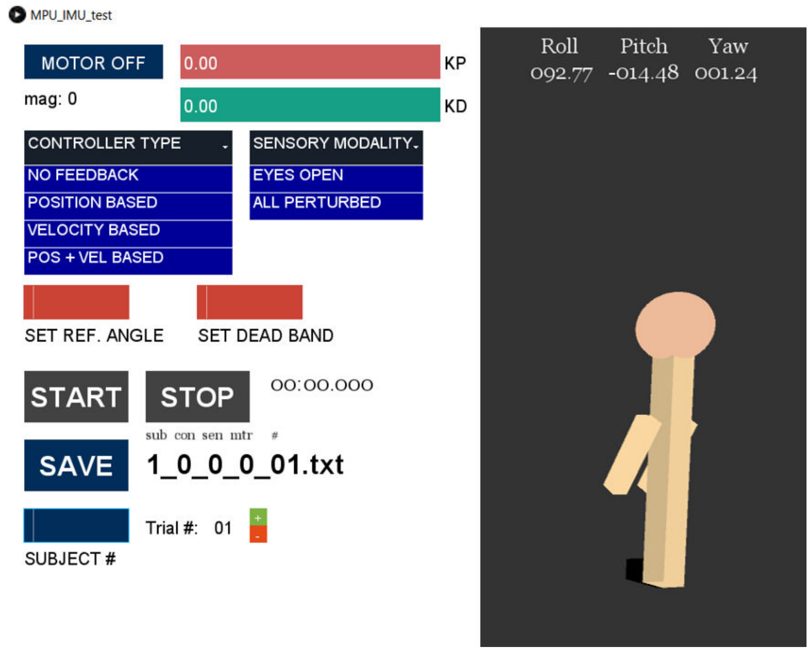

Figure 6.10: Graphical user interface for real-time posture visualization, user I/O and data acquisition.

study if they were unable to stand without walking aids, had orthopaedic problems or other diseases/disabilities than MS that could affect their balance. All participants provided written informed consent to participate before the experiment started. The study was approved by both Institutional Review Board at Texas A&M and University of Texas Health Science Center at Houston and the experiments were conducted at The Institute for Rehabilitation and Research Memorial Hermann in Houston, Texas.

6.6.2 Experimental Protocol

A physical and physiological screening evaluation were performed prior to the experiment. Subjects were asked to stand quietly on two adjacent force plates embedded in a platform (BP400600-OP-1000, AMTI, Watertown, USA) for 30 seconds. They were told to “stand quietly and not rely on the walker for balance unless it is necessary.” A total of four conditions were tested including two sensory modalities and two sensory augmentation conditions. The two sensory modality conditions included: i) Eyes open (EO), and ii) Eyes closed (EC); two sensory augmentation conditions are i) No feedback (OFF) and ii) with skin stretch feedback (ON). Subjects were given instruction on how to interpret the skin stretch feedback induced at fingertip (Fig. 6.11). To prevent falls during testing, the subjects were spotted by a physical therapist. Subjects were asked to put their hands on the walker handle and lightly touch or slightly press on the contactor with the middle finger of the right hand. The functionality of the skin stretch feedback was given to the subjects and they were instructed how to correctly interpret the feedback prior to the experiment.

The experiment consisted of two parts: i) practice and baseline calibration session and ii) main session. In the first session, subjects were instructed to stand quietly on the force plate while gripping on the handle of the walker. His/her neutral balance position was calibrated and recorded. Next, the skin stretch device was turned on, subjects were asked if they can feel the skin stretch at their fingertip and the minimum feedback threshold was tuned accordingly. After being instructed the functionality of the device, subjects were given time to practice and familiarize themselves with the device. They were asked to stand in place and move back and forth slightly to feel the skin stretch cue. The specific instruction "if you feel your fingertip is being stretched forward, lean forward, and vice versa" was provided. The main session follows if the subjects do not experience any discomfort from the device and have no questions in interpreting the skin stretch cues.

In the main session, subjects were asked to perform the quiet standing task with eyes open or eyes closed, and with feedback was on or off. Each trial lasts 30 seconds and each condition repeated five times. Total 20 trials were performed. For each

Figure 6.11: Experimental setup for the MS subject in performing static balancing tasks with the sensory augmentation walker. Subjects were spotted by a physical therapist (right) to ensure safety during the experiment.

trial, subject were asked to rely on the walker for support as little as possible, and try to *"avoid the contactor rotating by moving your body back and forth."* The purpose for this experiment is to examine the effectiveness of skin stretch feedback provided by the walker for the maintenance of upright stance.

The entire experiment took about 1 - 1.5 hours including the practice, main experiment and break. Rest breaks were provided upon request. Subject's feedback was gathered at the end of the experiment: (1) their perceived ability to interpret the intensity and direction of the skin stretch feedback, (2) how intuitive the presented feedback scheme felt, and (3) the usefulness of such device in balance rehabilitation and for people who needs walking aids. These qualitative measures, as well as other comments on the design or the experiment procedure were collected.

6.6.3 Assessment of Balance

Static balance performance under different feedback conditions was evaluated by recording COP trajectories of 30-s quiet standing tasks. Multiple measures of postural steadiness suggested in [71] were selected and evaluated in both AP and ML directions. These postural metrics are root-mean-square (RMS) distance, range and centroidal frequency (CF) of COP. The RMS distance were calculated as the square root of the mean square COP time series; range of COP (Range) was computed as the maximum distance between any two points on the trajectory. In addition to time-domain measures, frequency-domain measure, CF of COP, was also adopted from [71] and computed as the square root of the ratio of the second to the zero order spectral moments. This measure can be interpreted as the variance of the power spectral density of the COP, which is used to characterize the frequency distribution of the COP displacements.

Percent time in the desired region (PTDR) of body sway was calculated as the fraction of time that COP was in the dead zone of a neutral standing position to a 30-s period. PTDR was used to examine if subjects were able to respond to the postural cues and remove it by returning to the neutral positions. Reaction time to remove the feedback was defined as the average time period from the stimulus was first activated to the stimulus was first removed in each trial. There could be more than one reaction time segment in each trial depending on subjects' postural stability. This metric was used to determine how instantaneously can a subject remove the stimulus during quiet standing.

The force applied on the walker grips during the trial was recorded using the force plate system described before. The amount of force on the walker along with the COP trajectories measured by the force plate system were collected through an additional data acquisition system sampled at 1000 Hz.

6.7 Experimental Results

Fig. 6.12 presents various measures of COP trajectory during 30-s quiet standing under different feedback conditions. Table 6.1 presents the results for the forces applied on the walker (*walker force*), percentage time spent in the desired region (PTDR), and average reaction time to remove the stimuli (*Reaction time*), respectively, during each of the feedback condition. Reaction time were measured and reported under conditions that skin stretch feedback was on. Table 6.2 shows the qualitative results from the post-experiment questionnaire. The results of each subject are presented in the following sections.

6.7.1 Subject 1

Subject 1 was a 65-year-old with EDSS score of 7. In Table 6.12, mean value of Range of COP_{AP} decreased when feedback was ON during the eyes open tasks while the $Range_{ML}$ of COP were similar for both ON and OFF conditions. For the eyes open feedback ON conditions, RMS_{AP} and CF_{AP} of COP were smaller compared to the feedback OFF conditions; mean values of RMS_{ML} and CF_{ML} were similar in both ON and OFF conditions. Although the mean values of $Range_{ML}$ and RMS_{ML} were similar during EO condition, higher variability were observed for the feedback ON tasks. During the eyes closed conditions, opposite trends were observed for $Range_{AP}$ and RMS_{AP} where the values were greater when the feedback was ON. CF_{AP}, CF_{ML}, and RMS_{ML} were similar for ON and OFF conditions.

Walker forces were similar between feedback was OFF and ON for both eyes open and eyes closed conditions. While subjects applied around 10-N more forces on the walker for the eyes closed tasks. The values of PTDR were greater for both ON compared to OFF during eyes open and eyes closed conditions. The average reaction time was smaller in the eyes closed condition compared to the eyes open condition.

For the post-experiment survey (Table 6.2), subject gave 7 out of 7 rating in level of comfort and level of intuitiveness of the device. When considering the capability of discerning the direction and intensity of the stimuli, subject gave 4 out 7 ratings. Positive feedback were obtained which subject thought this device was helpful for correcting the posture and balancing the weight during standing. Subject also thought it would be easy to use for walker users.

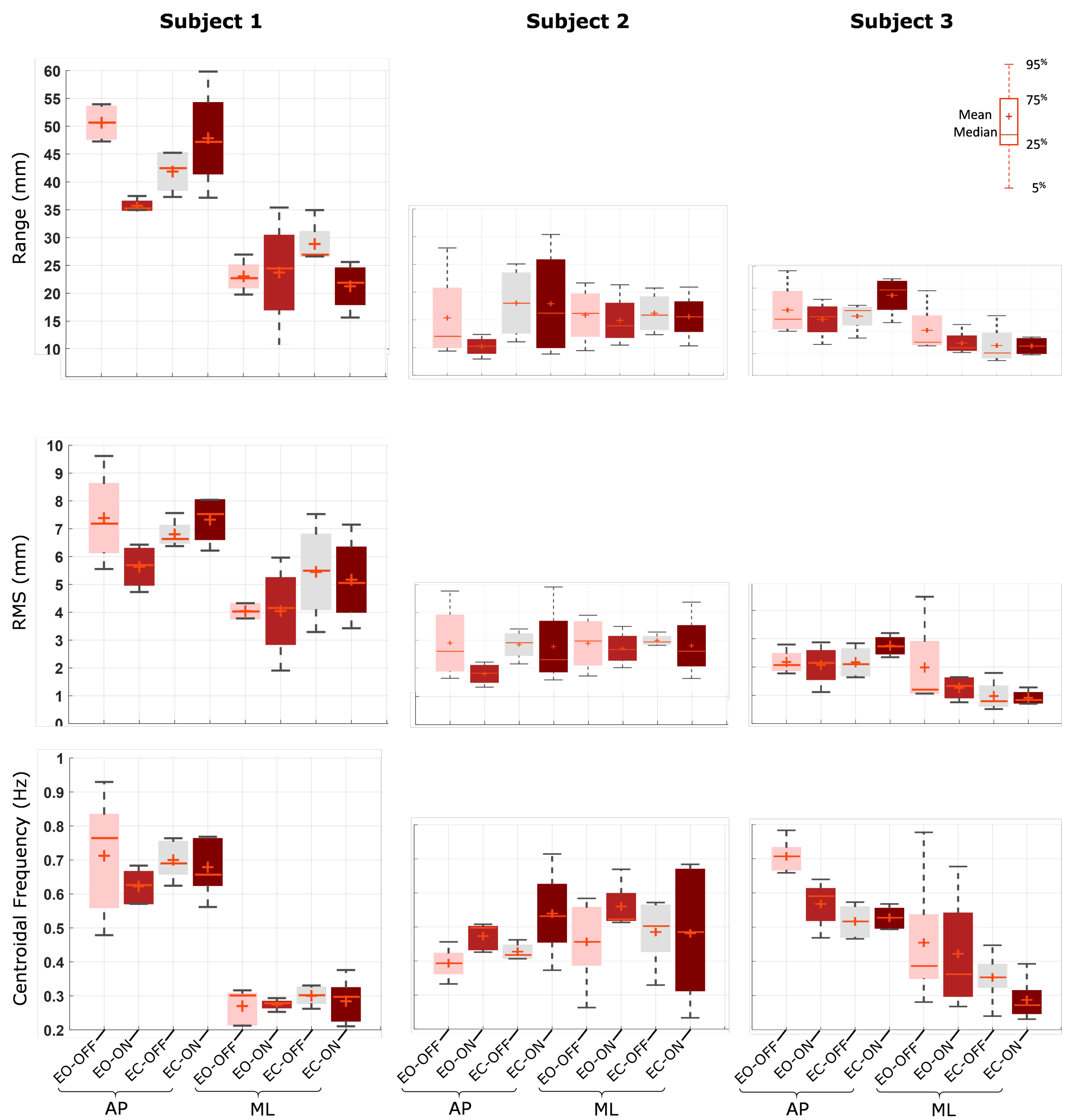

Figure 6.12: Measures of postural sway in both AP and ML directions under various feedback conditions. (EO - Eyes Open, EC - Eyes Closed, OFF - feedback was off, and ON - feedback was on)

Table 6.1: Outcome measures under sensory modality conditions (Eyes open and eyes closed) and skin stretch feedback conditions (OFF and ON). Mean values and standard deviations (s.d.) of five individual trials for each subject are reported.

	Skin stretch	**Subject 1**	**Subject 2**	**Subject 3**
	Amount of force applied on the walker (N)			
Eyes open	OFF	33.18 (9.88)	25.63(5.63)	10.57(4.38)
	ON	33.61(7.80)	25.04(5.03)	10.15(6.85)
Eyes closed	OFF	43.13(11.94)	28.78(3.30)	13.94(12.87)
	ON	45.90(16.58)	27.52(4.19)	14.99(3.84)
	Percent time in the desired region (PTDR)			
Eyes open	OFF	0.39(0.25)	0.47(0.41)	0.94(0.07)
	ON	0.52(0.29)	0.84(0.17)	0.97(0.06)
Eyes closed	OFF	0.17(0.11)	0.24(0.37)	0.51(0.68)
	ON	0.63(0.12)	0.58(0.19)	0.86(0.30)
	Average reaction time to remove the feedback (s)			
Eyes open	ON	2.38(1.86)	1.23(0.97)	1.07(0.95)
Eyes closed	ON	1.60(0.37)	2.24(1.51)	1.27(1.47)

6.7.2 Subject 2

Subject 2 was a 68-year-old with EDSS score of 6.5. Table 6.12 shows that during eyes open conditions, mean value of Range$_{AP}$ and RMS$_{AP}$ of COP both slightly decreased when feedback was ON. Range$_{ML}$ and RMS$_{ML}$ of COP were similar for both ON and OFF conditions. CF$_{AP}$ and CF$_{ML}$ both increased when feedback was ON compared to when feedback was OFF. For the eyes closed conditions, Range and RMS in both AP and ML directions were similar in OFF and ON conditions. CF$_{AP}$ increased during feedback ON condition and CF$_{ML}$ showed similar results for both ON and OFF conditions but for EC-OFF condition, there was a higher variability in CF$_{ML}$.

Table 6.1 presents that walker forces were similar between feedback was OFF

Table 6.2: Post-experiment questionnaire

Item	Subject 1	Subject 2	Subject 3
Level of comfort†	7	7	7
Level of intuitiveness ‡	7	6	6
Level of fatigue*	5	2	1
Discerning the direction$^\top$	4	6	7
Discerning the intensity$^\top$	4	5	6
Do you think it is helpful to correct your posture?‖	7	7	7
Do you think it is easy to use for people who needs walking aids?	Yes	Maybe	Yes
Comments	"I think it is a great idea to help with balance. I really thought about standing straighter and balancing my weight"	"I have MS so I can hope this might be an aide to help with balance. My balance is okay, but there are individuals who struggle more with this than I do. "	"At first it's not easy to tell the direction of the motor, it requires some time to understand the whole thing. The training time could be 10-15 mins. Was able to maintain posture to remove the feedback."

Rating scales: †1 ("not comfortable") to 7 ("very comfortable"),‡1 ("difficult to understand") to 7 ("easy to understand"), *1 ("barely tired") to 7 ("extremely tired"), $^\top$1 ("very difficult) to 7 ("very easy), ‖ 1("not at all") to 7 ("definitely").

and ON for both eyes open and eyes closed conditions. And subject applied slightly more forces on the walker under the eyes closed condition. The values of PTDR were greater for both ON compared to OFF during eyes open and eyes closed conditions. The average reaction time was greater in the eyes closed condition compared to the eyes open condition.

For the post-experiment survey (Table 6.2), subject stated the device does not cause any discomfort and barely felt tired during the experiment. Subject gave 6 out of 7 rating in the level of intuitiveness and 6 and 5 out of 7 ratings for discerning the direction and intensity of the stimuli, respectively. Positive feedback were also obtained which subject thought this device was helpful for correcting the posture. Although subject was not sure whether it would be easy to use among walker users.

6.7.3 Subject 3

Subject 3 was a 63-year-old with EDSS score of 6. Table 6.12 shows that during eyes open conditions, mean value of Range$_{AP}$ and RMS$_{AP}$ of COP both were similar for both ON and OFF conditions. Range$_{ML}$ and RMS$_{ML}$ of COP slightly decreased when feedback was ON. CF$_{AP}$ decreased when feedback was ON while CF$_{ML}$ were similar for ON and OFF conditions. For the eyes closed conditions, Range$_{AP}$ and RMS$_{AP}$ of COP both increased when feedback was ON. While Range$_{ML}$ and RMS$_{ML}$ showed similar results for both ON and OFF conditions. CF$_{ML}$ decreased during feedback ON condition while CF$_{AP}$ showed similar results for both ON and OFF conditions.

In Table 6.1, walker forces were similar between feedback was OFF and ON for both eyes open and eyes closed conditions. Subject applied slightly more forces on the walker under the eyes closed condition. The values of PTDR were similar and close to 100% for both ON and OFF during eyes open tasks. For eyes closed conditions, PTDR was greater when feedback was ON compared to OFF. The average reaction time was similar in both eyes closed and eyes open conditions.

For the post-experiment survey (Table 6.2), subject gave 7 and 6 out of 7 ratings in level of comfort and level of intuitiveness of the device, respectively. For discerning the direction and intensity of the stimuli, subject gave 7 and 6 out 7 ratings, respectively. Level of fatigue was rated 1 out 7 scale. Positive feedback were obtained which subject thought this device was helpful for correcting the posture and believed such device can be useful for people who regularly use walking aids. Subject also stated that there was a learning curve that after about 10 - 15 mins training she was able to follow the skin stretch cue.

6.8 Summary

A new functionality for a conventional walker that monitors users' real-time balance and provides this information to the user as a means of improving the postural stability. The effectiveness of the sensory augmentation walker on postural control in three MS subjects was evaluated. Quantitative and qualitative results obtained from the static balancing study demonstrated the potential benefits of incorporating such sensory feedback into mobility aids for enhancing user's control of balance.

7. CONCLUSIONS

7.1 Summary of Present Findings

The major result produced by this research is a new framework for integrating skin stretch feedback into balance rehabilitation by designing small, lightweight, and portable biofeedback devices. The novel devices that provide real time feedback of postural sway were developed for investigating the effectiveness of skin stretch feedback in improving sense of balance during standing. The feasibility and value of portable sensory augmentation systems were established by testing the balance performance though static and dynamic balancing studies on healthy subjects or subjects with neurological disorders. Experimental results have demonstrated that the haptic cues can be easily interpreted by the subjects, enhance the sensation to body movement, improve postural control in subjects. From control-theoretical point of view, skin stretch feedback induced by the developed devices enhances the quality of sensory cues of one's postural sway information and hence assists the subject to correct their posture during quiet standing.

Unlike previous approaches that require a stable surface or grounded devices for light touch contact of the finger, these devices are portable and easy to wear while applies light touch cues on the body parts. Such a small wearable device is also favored due to its flexible range of contact areas. Its portability gives it great potential for in-home rehabilitation for the target population. To the best of our knowledge, this is the first portable balance corrective system using skin stretch feedback.

The differential effects of position and velocity information of body sway for standing balance were further investigated using fingertip skin stretch feedback. The augmented velocity information of postural sway has been found to improve quiet standing balance more effectively compared to position information and the combination of the position and velocity information, which supports the findings of previous studies about the importance of body sway velocity in human postural control system.

A framework incorporating an sensory feedback device into a conventional walking aid was proposed based on the findings in Chapter 3 - Chapter 5. This device rendered users' posture information as a means of improving their sense of balance while supported by a mechanical device. Unlike the existing walking aids that primarily offer partial weight-support and force feedback of surrounding obstacles, the sensory feedback rendered by this device can encourage walker users actively engage in postural control. The experimental results and feedback obtained from the multiple sclerosis subjects further demonstrated its potential for in-home self retraining and daily use.

In summary, the novel rehabilitation methods using portable skin stretch feedback devices showed great potential in improving sense of balance and reducing fall risks among elderly or people with neuromuscular diseases. The findings in this research can also lead to development of balance rehabilitation tools for routinely used in home.

7.2 Recommendations for Future Work

Four paths can be followed based on the positive immediate effect of the sensory augmentation system via skin stretch: i) Long term effects of a portable sensory augmentation device in balance retraining, ii) Effects of a portable sensory augmentation device in elderly and other patient population, iii) Effects of a portable sensory augmentation device in dynamic balance during gait, and iv) Integration of a sensory augmentation device with robotics assistive devices, such as lower-limb exoskeleton or prosthesis. Eventually, we hope that with this multidimensional rehabilitative approach, i.e., combination of a robotics exoskeleton and a sensory feedback device, it could facilitate motor learning and more patients with various injury levels could experience an enhanced sense of balance and eventually become independent.

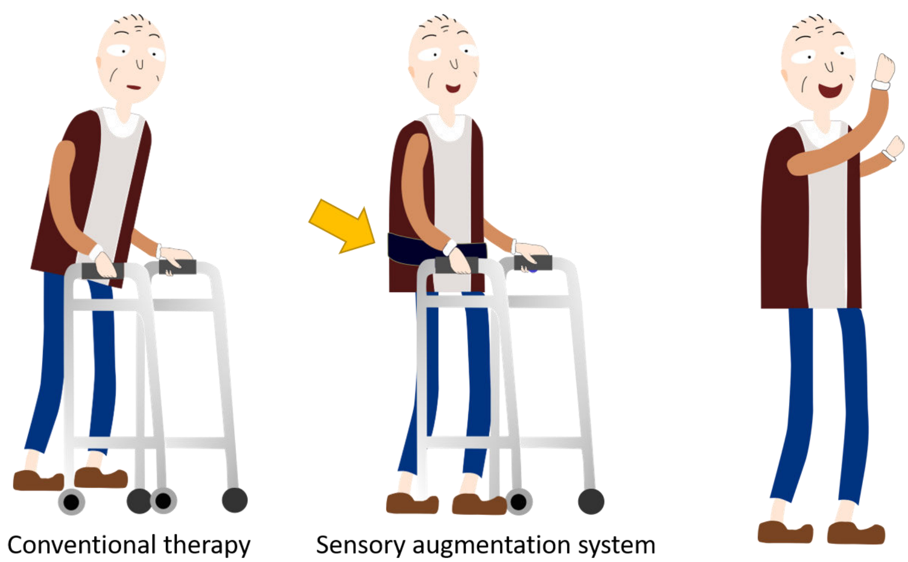

Figure 7.1: The long-term goal of this research is to accelerate the development of effective balance rehabilitation methods and eventually enhance the quality of life in people with neuromuscular diseases.

REFERENCES

[1] D. Winter, "Human balance and posture control during standing and walking," *Gait & Posture*, vol. 3, no. 4, pp. 193–214, dec 1995.

[2] K. P. Michmizos, S. Rossi, E. Castelli, P. Cappa, and H. I. Krebs, "Robot-Aided Neurorehabilitation: A Pediatric Robot for Ankle Rehabilitation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 23, no. 6, pp. 1056–1067, nov 2015.

[3] A. Kalron, I. Fonkatz, L. Frid, H. Baransi, and A. Achiron, "The effect of balance training on postural control in people with multiple sclerosis using the CAREN virtual reality system: a pilot randomized controlled trial," *Journal of neuroengineering and rehabilitation*, vol. 13, no. 1, p. 13, jan 2016.

[4] M. Kouzaki and K. Masani, "Reduced postural sway during quiet standing by light touch is due to finger tactile feedback but not mechanical support," *Experimental Brain Research*, vol. 188, no. 1, pp. 153–158, jun 2008.

[5] J. J. Jeka and J. Lackner, "Fingertip contact influences human postural control," *Experimental Brain Research*, vol. 79, no. 2, pp. 495–502, aug 1994.

[6] J. J. Jeka, "Light touch contact as a balance aid," *Physical therapy*, vol. 77, no. 5, pp. 476–487, 1997.

[7] A. M. Wing, L. Johannsen, and S. Endo, "Light touch for balance: influence of a time-varying external driving signal," *Philosophical Transactions of the Royal Society B: Biological Sciences*, vol. 366, no. 1581, pp. 3133–3141, 2011.

[8] S. Clapp and A. M. Wing, "Light touch contribution to balance in normal bipedal stance," *Experimental Brain Research*, vol. 125, no. 4, pp. 521–524, apr 1999.

[9] D. J. Reinkensmeyer and V. Dietz, *Neurorehabilitation Technology*, D. J. Reinkensmeyer and V. Dietz, Eds. Cham: Springer International Publishing, 2016.

[10] F. B. Horak and J. M. Macpherson, "Postural Orientation and Equilibrium," in *Comprehensive Physiology*. John Wiley & Sons, Inc., jan 2011, pp. 255–292.

[11] D. A. Winter, A. E. Patla, F. Prince, M. Ishac, and K. Gielo-Perczak, "Stiffness control of balance in quiet standing," *J Neurophysiol*, vol. 80, no. 3, pp. 1211–1221, sep 1998.

[12] R. J. Peterka, "Sensorimotor integration in human postural control," *J.Neurophysiol.*, vol. 88, no. 3, pp. 1097–1118, 2002.

[13] L. M. Nashner, "Analysis of Stance Posture in Humans," in *Motor Coordination*. Boston, MA: Springer US, 1981, vol. 5, pp. 527–565.

[14] D. Manchester, M. Woollacott, N. Zederbauer-Hylton, and O. Marin, "Visual, vestibular and somatosensory contributions to balance control in the older adult," *Journal of Gerontology*, vol. 44, no. 4, pp. M118–M127, jul 1989.

[15] H. van der Kooij, R. Jacobs, B. Koopman, and H. Grootenboer, "A multisensory integration model of human stance control," *Biological Cybernetics*, vol. 80, no. 5, pp. 299–308, may 1999.

[16] T. Kiemel, K. S. Oie, and J. J. Jeka, "Multisensory fusion and the stochastic structure of postural sway," *Biological Cybernetics*, vol. 87, no. 4, pp. 262–277, oct 2002.

[17] F. B. Horak, "Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls?" *Age and Ageing*, vol. 35, no. suppl_2, pp. ii7–ii11, sep 2006.

[18] T. Mergner, C. Maurer, and R. Peterka, "A multisensory posture control model of human upright stance," in *Progress in brain research*, jan 2003, vol. 142, pp. 189–201.

[19] H. Reimann, T. D. Fettrow, E. D. Thompson, P. Agada, B. J. McFadyen, and J. J. Jeka, "Complementary mechanisms for upright balance during walking," *PLOS ONE*, vol. 12, no. 2, p. e0172215, feb 2017.

[20] C. E. Bauby and A. D. Kuo, "Active control of lateral balance in human walking," *Journal of Biomechanics*, vol. 33, no. 11, pp. 1433–1440, nov 2000.

[21] H. Reimann, T. Fettrow, and J. J. Jeka, "Strategies for the Control of Balance During Locomotion," *Kinesiology Review*, vol. 7, no. 1, pp. 18–25, feb 2018.

[22] A. J. Blake, K. Morgan, M. J. Bendall, H. Dallosso, S. B. J. Ebrahim, T. H. D. Arie, P. H. Fentem, And E. J. Bassey, "Falls by Elderly People at Home: Prevalence and Associated Factors," *Age and Ageing*, vol. 17, no. 6, pp. 365–372, jan 1988.

[23] M. C. Nevitt, S. R. Cummings, S. Kidd, and D. Black, "Risk Factors for Recurrent Nonsyncopal Falls," *JAMA*, vol. 261, no. 18, p. 2663, may 1989.

[24] T. Masud and R. O. Morris, “Epidemiology of falls,” *Age and Ageing*, vol. 30, no. suppl 4, pp. 3–7, nov 2001.

[25] P. W. Overstall, A. N. Exton-Smith, F. J. Imms, and A. L. Johnson, “Falls in the elderly related to postural imbalance,” *BMJ*, vol. 1, no. 6056, pp. 261–264, jan 1977.

[26] J. Sheldon, “The Effect of Age on the Control of Sway,” *Gerontologia Clinica*, vol. 5, no. 3, pp. 129–138, 1963.

[27] V. Weerdesteyn, M. de Niet, H. J. R. van Duijnhoven, and A. C. H. Geurts, “Falls in individuals with stroke,” *Journal of rehabilitation research and development*, vol. 45, no. 8, pp. 1195–213, 2008.

[28] D. Cattaneo, C. De Nuzzo, T. Fascia, M. Macalli, I. Pisoni, and R. Cardini, “Risks of falls in subjects with multiple sclerosis,” *Archives of Physical Medicine and Rehabilitation*, vol. 83, no. 6, pp. 864–867, jun 2002.

[29] Y. Nilsagård, C. Lundholm, E. Denison, and L.-G. Gunnarsson, “Predicting accidental falls in people with multiple sclerosis - a longitudinal study,” *Clinical Rehabilitation*, vol. 23, no. 3, pp. 259–269, mar 2009.

[30] M. M. Gardner, D. M. Buchner, M. C. Robertson, and A. J. Campbell, “Practical implementation of an exercise-based falls prevention programme,” *Age and Ageing*, vol. 30, no. 1, pp. 77–83, jan 2001.

[31] M. C. Robertson, N. Devlin, M. M. Gardner, and A. J. Campbell, “Effectiveness and economic evaluation of a nurse delivered home exercise programme to prevent falls. 1: Randomised controlled trial.” *BMJ (Clinical research ed.)*, vol. 322, no. 7288, pp. 697–701, mar 2001.

[32] S. Thomas, S. Mackintosh, and J. Halbert, "Does the 'Otago exercise programme' reduce mortality and falls in older adults?: a systematic review and meta-analysis," *Age and Ageing*, vol. 39, no. 6, pp. 681–687, nov 2010.

[33] M. L. Aisen, H. I. Krebs, N. Hogan, F. McDowell, and B. T. Volpe, "The Effect of Robot-Assisted Therapy and Rehabilitative Training on Motor Recovery Following Stroke," *Archives of Neurology*, vol. 54, no. 4, pp. 443–446, apr 1997.

[34] V. Dietz, G. Colombo, L. Jensen, and L. Baumgartner, "Locomotor capacity of spinal cord in paraplegic patients," *Annals of Neurology*, vol. 37, no. 5, pp. 574–582, may 1995.

[35] Uhlenbrock and D., "Development of a servo-controlled gait trainer for the rehabilitation of non-ambulatory patients," *Biomedizinische Technik*, vol. 42, pp. 196–202, 1997.

[36] J. Hidler, D. Nichols, M. Pelliccio, K. Brady, D. D. Campbell, J. H. Kahn, and T. G. Hornby, "Multicenter Randomized Clinical Trial Evaluating the Effectiveness of the Lokomat in Subacute Stroke," *Neurorehabilitation and Neural Repair*, vol. 23, no. 1, pp. 5–13, sep 2008.

[37] E. T. Wolbrecht, V. Chan, D. J. Reinkensmeyer, and J. E. Bobrow, "Optimizing compliant, model-based robotic assistance to promote neurorehabilitation." *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 16, no. 3, pp. 286–97, jun 2008.

[38] A. J. Young and D. P. Ferris, "State of the Art and Future Directions for Lower Limb Robotic Exoskeletons," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 25, no. 2, pp. 171–182, feb 2017.

[39] J. C. Nitz, S. Kuys, R. Isles, and S. Fu, "Is the Wii Fit a new-generation tool for improving balance, health and well-being? A pilot study," *Climacteric*, vol. 13, no. 5, pp. 487–491, oct 2010.

[40] K. Bieryla and N. Dold, "Feasibility of Wii Fit training to improve clinical measures of balance in older adults," *Clinical Interventions in Aging*, vol. 8, no. 1, p. 775, jun 2013.

[41] M. Agmon, C. K. Perry, E. Phelan, G. Demiris, and H. Q. Nguyen, "A Pilot Study of Wii Fit Exergames to Improve Balance in Older Adults," *Journal of Geriatric Physical Therapy*, vol. 34, no. 4, pp. 161–167, 2011.

[42] J.-f. Esculier, J. Vaudrin, P. Bériault, K. Gagnon, and L. E. Tremblay, "Home-based Balance Training Programme Using Wii Fit with Balance Board for Parkinson's Disease : A Pilot Study," *J Rehabil Med*, vol. 44, no. 2, pp. 144–150, 2012.

[43] B. Lange, S. Flynn, R. Proffitt, C.-Y. Chang, and A. Rizzo, "Development of an Interactive Game-Based Rehabilitation Tool for Dynamic Balance Training," *Topics in Stroke Rehabilitation*, vol. 17, no. 5, pp. 345–352, sep 2010.

[44] A. A. Gopalai and S. M. Arosha Senanayake, "A wearable real-time intelligent posture corrective system using vibrotactile feedback," *IEEE/ASME Transactions on Mechatronics*, vol. 16, no. 5, pp. 827–834, 2011.

[45] Y.-T. Pan, H. U. Yoon, and P. Hur, "A Portable Sensory Augmentation Device for Balance Rehabilitation Using Fingertip Skin Stretch Feedback," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 25, no. 1, pp. 31–39, jan 2017.

[46] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, "Effects of multi-directional vibrotactile feedback on vestibular-deficient postural performance during continuous multi-directional support surface perturbations," *Journal of vestibular research : equilibrium & orientation*, vol. 18, no. 5, 6, pp. 273–285, 2008.

[47] E. Kentala, J. Vivas, and C. Wall, "Reduction of Postural Sway by Use of a Vibrotactile Balance Prosthesis Prototype in Subjects with Vestibular Deficits," *Annals of Otology, Rhinology & Laryngology*, vol. 112, no. 5, pp. 404–409, may 2003.

[48] W. Nanhoe-Mahabier, J. Allum, E. Pasman, S. Overeem, and B. Bloem, "The effects of vibrotactile biofeedback training on trunk sway in Parkinson's disease patients," *Parkinsonism & Related Disorders*, vol. 18, no. 9, pp. 1017–1021, 2012.

[49] M. S. Schwartz and F. Andrasik, *Biofeedback: A practitioner's guide*, 3rd ed. Guilford Press, 2017.

[50] A. Zijlstra, M. Mancini, L. Chiari, and W. Zijlstra, "Biofeedback for training balance and mobility tasks in older populations: a systematic review," *Journal of neuroengineering and rehabilitation*, vol. 7, no. 1, p. 58, jan 2010.

[51] A. C. H. Geurts, M. de Haart, I. J. W. van Nes, and J. Duysens, "A review of standing balance recovery from stroke," *Gait & posture*, vol. 22, no. 3, pp. 267–81, nov 2005.

[52] R. van Peppen, M. Kortsmit, E. Lindeman, and G. Kwakkel, "Effects of visual feedback therapy on postural control in bilateral standing after stroke: a

systematic review," *Journal of Rehabilitation Medicine*, vol. 38, no. 1, pp. 3–9, 2006.

[53] F. B. Horak, C. L. Shupert, and A. Mirka, "Components of postural dyscontrol in the elderly: a review," *Neurobiology of aging*, vol. 10, no. 6, pp. 727–738, 1989.

[54] M. Dozza, F. B. Horak, and L. Chiari, "Auditory biofeedback substitutes for loss of sensory information in maintaining stance," *Experimental brain research*, vol. 178, no. 1, pp. 37–48, mar 2007.

[55] L. Chiari, M. Dozza, A. Cappello, F. B. Horak, V. Macellari, and D. Giansanti, "Audio-biofeedback for balance improvement: an accelerometry-based system," *IEEE transactions on bio-medical engineering*, vol. 52, no. 12, pp. 2108–2111, dec 2005.

[56] K. H. Sienko, M. D. Balkwill, and C. Wall, "Biofeedback improves postural control recovery from multi-axis discrete perturbations," *Journal of neuroengineering and rehabilitation*, vol. 9, no. 1, p. 53, jan 2012.

[57] S. M. Kärcher, S. Fenzlaff, D. Hartmann, S. K. Nagel, and P. König, "Sensory Augmentation for the Blind," *Frontiers in Human Neuroscience*, vol. 6, p. 37, jan 2012.

[58] S. D. Novich and D. M. Eagleman, "[D79] A vibrotactile sensory substitution device for the deaf and profoundly hearing impaired," in *2014 IEEE Haptics Symposium (HAPTICS)*. IEEE, feb 2014, pp. 1–1.

[59] B. B. Edin and N. Johansson, "Skin strain patterns provide kinaesthetic information to the human central nervous system," *The Journal of Physiology*, vol. 487, no. 1, pp. 243–251, aug 1995.

[60] W. Provancher and N. Sylvester, "Fingerpad skin stretch increases the perception of virtual friction," *IEEE Transactions on Haptics*, vol. 2, no. 4, pp. 212–223, oct 2009.

[61] J. Biggs and M. Srinivasan, "Tangential versus normal displacements of skin: relative effectiveness for producing tactile sensations," in *Proceedings 10th Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems. HAPTICS 2002*. IEEE Comput. Soc, 2002, pp. 121–128.

[62] J. Jeka and J. Lackner, "The role of haptic cues from rough and slippery surfaces in human postural control," *Experimental Brain Research*, vol. 103, no. 2, mar 1995.

[63] J. Forero and J. E. Misiaszek, "The effect of light touch on the amplitude of cutaneous reflexes in the arms during treadmill walking," *Experimental Brain Research*, vol. 232, no. 9, pp. 2967–2976, sep 2014.

[64] V. Krishnamoorthy, H. Slijper, and M. L. Latash, "Effects of different types of light touch on postural sway," *Experimental Brain Research*, vol. 147, no. 1, pp. 71–79, 2002.

[65] L. R. Enders, P. Hur, M. J. Johnson, and N. Seo, "Remote vibrotactile noise improves light touch sensation in stroke survivors' fingertips via stochastic resonance," *Journal of NeuroEngineering and Rehabilitation*, vol. 10, no. 1, p. 105, jan 2013.

[66] S. O. Madgwick, A. J. L. Harrison, and R. Vaidyanathan, "Estimation of IMU and MARG orientation using a gradient descent algorithm," in *IEEE International Conference on Rehabilitation Robotics*, 2011, pp. 1–7.

[67] T. Brandt, S. Krafczyk, and I. Malsbenden, "Postural imbalance with head extension: improvement by training as a model for ataxia therapy," *Annals of the New York Academy of Sciences*, vol. 374, no. 1, pp. 636–649, nov 1981.

[68] R. T. Jackson and C. M. Epstein, "Effect of Head Extension on Equilibrium in Normal Subjects," *Annals of Otology, Rhinology & Laryngology*, vol. 100, no. 1, pp. 63–67, jan 1991.

[69] V. Anand, J. G. Buckley, A. Scally, and D. B. Elliott, "Postural stability in the elderly during sensory perturbations and dual tasking: the influence of refractive blur," *Investigative ophthalmology & visual science*, vol. 44, no. 7, pp. 2885–91, jul 2003.

[70] N. Vuillerme, O. Chenu, N. Pinsault, A. Fleury, J. Demongeot, and Y. Payan, "Can a plantar pressure-based tongue-placed electrotactile biofeedback improve postural control under altered vestibular and neck proprioceptive conditions?" *Neuroscience*, vol. 155, no. 1, pp. 291–6, jul 2008.

[71] T. E. Prieto, J. B. Myklebust, R. G. Hoffmann, E. G. Lovett, and B. M. Myklebust, "Measures of postural steadiness: Differences between healthy young and elderly adults," *IEEE Transactions on Biomedical Engineering*, vol. 43, no. 9, pp. 956–966, 1996.

[72] N. H. McNevin and G. Wulf, "Attentional focus on supra-postural tasks affects postural control," *Human Movement Science*, vol. 21, no. 2, pp. 187–202, jul 2002.

[73] C. Maurer and R. J. Peterka, "A new interpretation of spontaneous sway measures based on a simple model of human postural control," *Journal of neurophysiology*, vol. 93, no. 1, pp. 189–200, 2005.

[74] L. Nashner, F. Black, and C. Wall, "Adaptation to altered support and visual conditions during stance: patients with vestibular deficits," *The Journal of Neuroscience*, vol. 2, no. 5, pp. 536–544, may 1982.

[75] P. Hur, K. Park, K. S. Rosengren, G. P. Horn, and E. T. Hsiao-Wecksler, "Effects of air bottle design on postural control of firefighters," *Applied Ergonomics*, vol. 48, pp. 49–55, may 2015.

[76] Hur, P., K. S. Rosengren, G. P. Horn, D. L. Smith, and E. T. Hsiao-Wecksler, "Effect of protective clothing and fatigue on functional balance of firefighters," *Journal of Ergonomics*, vol. S2, no. 004, dec 2013.

[77] A. Hufschmidt, J. Dichgans, K. H. Mauritz, and M. Hufschmidt, "Some methods and parameters of body sway quantification and their neurological applications," *Archiv für Psychiatrie und Nervenkrankheiten*, vol. 228, no. 2, pp. 135–150, may 1980.

[78] B. E. Maki, P. J. Holliday, and G. R. Fernie, "Aging and postural control. A comparison of spontaneous- and induced-sway balance tests," *Journal of the American Geriatrics Society*, vol. 38, no. 1, pp. 1–9, jan 1990.

[79] D. Lin, H. Seol, M. A. Nussbaum, and M. L. Madigan, "Reliability of COP-based postural sway measures and age-related differences," *Gait & Posture*, vol. 28, no. 2, pp. 337–342, aug 2008.

[80] J. R. Davis, M. G. Carpenter, R. Tschanz, S. Meyes, D. Debrunner, J. Burger, and J. H. J. Allum, "Trunk sway reductions in young and older adults using multi-modal biofeedback," *Gait & posture*, vol. 31, no. 4, pp. 465–472, apr 2010.

[81] M. Woollacott and A. Shumway-Cook, "Attention and the control of posture and gait: a review of an emerging area of research," *Gait & Posture*, vol. 16, no. 1, pp. 1–14, aug 2002.

[82] P. Hur, Y.-H. Wan, and N. J. Seo, "Investigating the role of vibrotactile noise in early response to perturbation," *IEEE transactions on bio-medical engineering*, vol. 61, no. 6, pp. 1628–1633, jun 2014.

[83] N. J. Seo, M. L. Kosmopoulos, L. R. Enders, and P. Hur, "Effect of remote sensory noise on hand function post stroke," *Frontiers in human neuroscience*, vol. 8, no. 934, pp. 1–19, jan 2014.

[84] E. Manjarrez, G. Rojas-Piloni, I. Mendez, and A. Flores, "Stochastic resonance within the somatosensory system: effects of noise on evoked field potentials elicited by tactile stimuli," *J. Neurosci.*, vol. 23, no. 6, pp. 1997–2001, mar 2003.

[85] H. C. Diener and J. Dichgans, "Long loop reflexes and posture," in *Disorders of posture and gait*, W. Bles and T. Brandt, Eds. Elsevier North Holland, Amsterdam, 1986, pp. 41–51.

[86] H. Diener, F. Bootz, J. Dichgans, and W. Bruzek, "Variability of postural "reflexes" in humans," *Experimental Brain Research*, vol. 52, no. 3, pp. 423–428, nov 1983.

[87] L. Nashner, "Adapting reflexes controlling the human posture," *Experimental Brain Research*, vol. 26, no. 1, pp. 59–72, 1976.

[88] B. R. Bloem, D. J. Beckley, J. P. van Vugt, J. G. van Dijk, M. P. Remler, J. W. Langston, and R. A. Roos, "Long latency postural reflexes are under supraspinal dopaminergic control," *Movement disorders : official journal of the Movement Disorder Society*, vol. 10, no. 5, pp. 580–8, sep 1995.

[89] R. C. Browning, J. R. Modica, R. Kram, and A. Goswami, "The Effects of Adding Mass to the Legs on the Energetics and Biomechanics of Walking," *Medicine & Science in Sports & Exercise*, vol. 39, no. 3, pp. 515–525, mar 2007.

[90] C. Wall, M. S. Weinberg, P. B. Schmidt, and D. E. Krebs, "Balance prosthesis based on micromechanical sensors using vibrotactile feedback of tilt," *IEEE transactions on bio-medical engineering*, vol. 48, no. 10, pp. 1153–61, oct 2001.

[91] B.-C. Lee, S. Chen, and K. H. Sienko, "A wearable device for real-time motion error detection and vibrotactile instructional cuing," *IEEE transactions on neural systems and rehabilitation engineering*, vol. 19, no. 4, pp. 374–81, aug 2011.

[92] M. Dozza, L. Chiari, B. Chan, L. Rocchi, F. B. Horak, and A. Cappello, "Influence of a portable audio-biofeedback device on structural properties of postural sway," *Journal of neuroengineering and rehabilitation*, vol. 2, no. 1, p. 13, jan 2005.

[93] J. Jeka, T. Kiemel, R. Creath, F. B. Horak, and R. J. Peterka, "Controlling Human Upright Posture: Velocity Information Is More Accurate Than Position

or Acceleration," *Journal of Neurophysiology*, vol. 92, no. 4, pp. 2368–2379, may 2004.

[94] J. Jeka, K. Oie, G. Schöner, T. Dijkstra, and E. Henson, "Position and velocity coupling of postural sway to somatosensory drive," *Journal of neurophysiology*, vol. 79, no. 4, pp. 1661–74, apr 1998.

[95] R. Dickstein, R. J. Peterka, and F. B. Horak, "Effects of light fingertip touch on postural responses in subjects with diabetic neuropathy." *Journal of neurology, neurosurgery, and psychiatry*, vol. 74, no. 5, pp. 620–6, may 2003.

[96] J. J. Collins and C. J. De Luca, "Open-loop and closed-loop control of posture: a random-walk analysis of center-of-pressure trajectories," *Experimental brain research*, vol. 95, no. 2, pp. 308–18, 1993.

[97] P. Hur, K. A. Shorter, P. G. Mehta, and E. T. Hsiao-Wecksler, "Invariant density analysis: Modeling and analysis of the postural control system using Markov chains," *IEEE Transactions on Biomedical Engineering*, vol. 59, no. 4, pp. 1094–1100, 2012.

[98] J. J. Collins and C. J. De Luca, "The effects of visual input on open-loop and closed-loop postural control mechanisms," *Experimental Brain Research*, vol. 103, no. 1, pp. 151–163, jan 1995.

[99] C. A. Laughton, M. Slavin, K. Katdare, L. Nolan, J. F. Bean, D. Kerrigan, E. Phillips, L. A. Lipsitz, and J. J. Collins, "Aging, muscle activity, and balance control: physiologic changes associated with balance impairment," *Gait & Posture*, vol. 18, no. 2, pp. 101–108, oct 2003.

[100] P. F. Meyer, L. I. E. Oddsson, and C. J. De Luca, "The role of plantar cutaneous sensation in unperturbed stance," *Experimental brain research*, vol. 156, no. 4, pp. 505–12, jun 2004.

[101] J.-H. Chiang and G. Wu, "The influence of foam surfaces on biomechanical variables contributing to postural control," *Gait & Posture*, vol. 5, no. 3, pp. 239–245, jun 1997.

[102] M. Patel, P. Fransson, D. Lush, and S. Gomez, "The effect of foam surface properties on postural stability assessment while standing," *Gait & Posture*, vol. 28, no. 4, pp. 649–656, nov 2008.

[103] M. C. Dault, L. Yardley, and J. S. Frank, "Does articulation contribute to modifications of postural control during dual-task paradigms?" *Cognitive Brain Research*, vol. 16, no. 3, pp. 434–440, may 2003.

[104] K. Masani, M. R. Popovic, K. Nakazawa, M. Kouzaki, and D. Nozaki, "Importance of Body Sway Velocity Information in Controlling Ankle Extensor Activities During Quiet Stance," *Journal of Neurophysiology*, vol. 90, no. 6, pp. 3774–3782, 2003.

[105] S. M. Sketch, D. R. Deo, J. P. Menon, and A. M. Okamura, "Design and experimental evaluation of a skin-stretch haptic device for improved control of brain-computer interfaces," in *2015 IEEE International Conference on Robotics and Automation (ICRA)*. IEEE, may 2015, pp. 272–277.

[106] K. Bark, J. Wheeler, P. Shull, J. Savall, and M. Cutkosky, "Rotational Skin Stretch Feedback: A Wearable Haptic Display for Motion," *IEEE Transactions on Haptics*, vol. 3, no. 3, pp. 166–176, jul 2010.

[107] K. E. Bechly, W. J. Carender, J. D. Myles, and K. H. Sienko, "Determining the preferred modality for real-time biofeedback during balance training," *Gait & Posture*, vol. 37, no. 3, pp. 391–396, mar 2013.

[108] G. Baud-Bovy, F. Tatti, and A. Borghese, "Ability of Low-Cost Force-Feedback Device to Influence Postural Stability," *IEEE transactions on haptics*, vol. 8, no. 2, pp. 130–139, nov 2015.

[109] M. Aggravi, G. Salvietti, and D. Prattichizzo, "Haptic wrist guidance using vibrations for Human-Robot teams," in *2016 25th IEEE International Symposium on Robot and Human Interactive Communication (RO-MAN)*. IEEE, aug 2016, pp. 113–118.

[110] F. Chinello, C. Pacchierotti, N. G. Tsagarakis, and D. Prattichizzo, "Design of a wearable skin stretch cutaneous device for the upper limb," in *2016 IEEE Haptics Symposium, HAPTICS*, 2016, pp. 14–20.

[111] A. A. Stanley and K. J. Kuchenbecker, "Evaluation of Tactile Feedback Methods for Wrist Rotation Guidance," *IEEE Transactions on Haptics*, vol. 5, no. 3, pp. 240–251, 2012.

[112] M. H. Cameron and S. Lord, "Postural Control in Multiple Sclerosis: Implications for Fall Prevention," *Current Neurology and Neuroscience Reports*, vol. 10, no. 5, pp. 407–412, sep 2010.

[113] J. J. Sosnoff, M. J. Socie, M. K. Boes, B. M. Sandroff, J. H. Pula, Y. Suh, M. Weikert, S. Balantrapu, S. Morrison, and R. W. Motl, "Mobility, Balance and Falls in Persons with Multiple Sclerosis," *PLoS ONE*, vol. 6, no. 11, p. e28021, nov 2011.

[114] R. W. Motl, E. M. Snook, E. McAuley, and R. C. Gliottoni, “Symptoms, self-efficacy, and physical activity among individuals with multiple sclerosis,” *Research in Nursing & Health*, vol. 29, no. 6, pp. 597–606, dec 2006.

[115] D. Cattaneo, J. Jonsdottir, M. Zocchi, and A. Regola, “Effects of balance exercises on people with multiple sclerosis: a pilot study,” *Clinical Rehabilitation*, vol. 21, no. 9, pp. 771–781, sep 2007.

[116] J. J. Jeka, R. D. Easton, B. L. Bentzen, and J. R. Lackner, “Haptic cues for orientation and postural control,” *Perception & Psychophysics*, vol. 58, no. 3, pp. 409–423, jan 1996.

[117] R. Boonsinsukh, L. Panichareon, and P. Phansuwan-Pujito, “Light Touch Cue Through a Cane Improves Pelvic Stability During Walking in Stroke,” *Archives of Physical Medicine and Rehabilitation*, vol. 90, no. 6, pp. 919–926, jun 2009.

[118] H. Hashimoto, A. Sasaki, Y. Ohyama, and C. Ishii, “Walker with hand haptic interface for spatial recognition,” in *9th IEEE International Workshop on Advanced Motion Control*. IEEE, 2006, pp. 311–316.

[119] Y. Wang and K. J. Kuchenbecker, “HALO: Haptic Alerts for Low-hanging Obstacles in white cane navigation,” in *2012 IEEE Haptics Symposium (HAPTICS)*. IEEE, mar 2012, pp. 527–532.

[120] R. Pyun, Yeongmi Kim, P. Wespe, R. Gassert, and S. Schneller, “Advanced Augmented White Cane with obstacle height and distance feedback,” in *2013 IEEE 13th International Conference on Rehabilitation Robotics (ICORR)*. IEEE, jun 2013, pp. 1–6.

[121] M. Martins, C. Santos, A. Frizera, and R. Ceres, "Real time control of the ASBGo walker through a physical human-robot interface," *Measurement*, vol. 48, pp. 77–86, feb 2014.

[122] M. Spenko, H. Yu, and S. Dubowsky, "Robotic Personal Aids for Mobility and Monitoring for the Elderly," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 14, no. 3, pp. 344–351, sep 2006.

[123] H. Yu, M. Spenko, and S. Dubowsky, "An Adaptive Shared Control System for an Intelligent Mobility Aid for the Elderly," *Autonomous Robots*, vol. 15, no. 1, pp. 53–66, 2003.

[124] J. Pasquero, J. Luk, V. Levesque, Q. Wang, V. Hayward, and K. MacLean, "Haptically Enabled Handheld Information Display With Distributed Tactile Transducer," *IEEE Transactions on Multimedia*, vol. 9, no. 4, pp. 746–753, jun 2007.

[125] A. J. Spiers and A. M. Dollar, "Design and Evaluation of Shape-Changing Haptic Interfaces for Pedestrian Navigation Assistance," *IEEE Transactions on Haptics*, vol. 10, no. 1, pp. 17–28, jan 2017.

[126] H. Benko, C. Holz, M. Sinclair, and E. Ofek, "NormalTouch and TextureTouch: High-fidelity 3D Haptic Shape Rendering on Handheld Virtual Reality Controllers," in *Proceedings of the 29th Annual Symposium on User Interface Software and Technology - UIST '16*. New York, New York, USA: ACM Press, 2016, pp. 717–728.

[127] I. Choi, E. Ofek, H. Benko, M. Sinclair, and C. Holz, "CLAW : A Multifunctional Handheld Haptic Controller for Grasping , Touching , and Triggering in Virtual Reality," in *CHI 2018*, Montreal, QC, Canada, 2018.

[128] E. Whitmire, H. Benko, C. Holz, E. Ofek, and M. Sinclair, "Haptic Revolver : Touch , Shear , Texture , and Shape Rendering on a Reconfgurable Virtual Reality Controller," in *CHI 2018*, Montreal, QC, Canada, 2018.

[129] W. Provancher, "Creating Greater VR Immersion by Emulating Force Feedback with Ungrounded Tactile Feedback," *IQT Quarterly*, vol. 6, no. 2, pp. 18–21, 2014.

[130] C. J. Ploch, J. H. Bae, W. Ju, and M. Cutkosky, "Haptic skin stretch on a steering wheel for displaying preview information in autonomous cars," in *2016 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*. IEEE, oct 2016, pp. 60–65.

[131] J. B. F. V. Erp, H. A. H. C. V. Veen, C. Jansen, and T. Dobbins, "Waypoint navigation with a vibrotactile waist belt," *ACM Transactions on Applied Perception*, vol. 2, no. 2, pp. 106–117, apr 2005.

[132] G.-H. Yang, M.-s. Jin, Y. Jin, and S. Kang, "T-mobile: Vibrotactile display pad with spatial and directional information for hand-held device," in *2010 IEEE/RSJ International Conference on Intelligent Robots and Systems*. IEEE, oct 2010, pp. 5245–5250.

[133] K. Drewing, M. Fritschi, R. Zopf, M. O. Ernst, and M. Buss, "First evaluation of a novel tactile display exerting shear force via lateral displacement," *ACM Transactions on Applied Perception*, vol. 2, no. 2, pp. 118–131, apr 2005.

[134] B. T. Gleeson, S. K. Horschel, and W. R. Provancher, "Communication of direction through lateral skin stretch at the fingertip," in *World Haptics 2009 - Third Joint EuroHaptics conference and Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems*. IEEE, mar 2009, pp. 172–177.

[135] H. Backlund Wasling, U. Norrsell, K. Göthner, and H. Olausson, "Tactile directional sensitivity and postural control," *Experimental Brain Research*, vol. 166, no. 2, pp. 147–156, oct 2005.

[136] A. B. Vallbo and R. S. Johansson, "Properties of cutaneous mechanoreceptors in the human hand related to touch sensation," *Human neurobiology*, vol. 3, no. 1, pp. 3–14, 1984.

[137] Y.-C. Pai and J. Patton, "Center of mass velocity-position predictions for balance control," *Journal of Biomechanics*, vol. 30, no. 4, pp. 347–354, apr 1997.

[138] K. Iqbal and Y.-C. Pai, "Predicted region of stability for balance recovery:: motion at the knee joint can improve termination of forward movement," *Journal of Biomechanics*, vol. 33, no. 12, pp. 1619–1627, dec 2000.

[139] A. Hof, M. Gazendam, and W. Sinke, "The condition for dynamic stability," *Journal of Biomechanics*, vol. 38, no. 1, pp. 1–8, jan 2005.

[140] J. Pratt, J. Carff, S. Drakunov, and A. Goswami, "Capture Point: A Step toward Humanoid Push Recovery," in *2006 6th IEEE-RAS International Conference on Humanoid Robots*. IEEE, dec 2006, pp. 200–207.